

BALANCING THE USE OF RADIATION IN ENDOVASCULAR BRACHYTHERAPY

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ABSTRACT

Endovascular brachytherapy to inhibit restenosis after percutaneous transluminal coronary angioplasty (PTCA) is a hot topic. Various clinical trials have been conducted and many new ones are being set up. In the context of justification and optimisation of such trials we have made a comparison of relevant characteristics for patient dosimetry and radiation protection, for a variety of potentially suitable radioactive sources. A summary of results is presented in this paper.

INTRODUCTION

Restenosis is a generic term for several mechanisms that cause obstructive lesion recurrence in vessels. It remains the main complication of angioplasty procedures. Although PTCA has been a successful practice for more than two decades now, restenosis has not dropped below 40%. Animal experiments and the first few clinical trials indicate that irradiation of the dilated vessel wall can prevent, or at least postpone, restenosis⁽¹⁾. Large and systematic clinical trials are currently being conducted and designed, employing a wide variety of delivery systems and radiation sources.

Basic questions still remain on dose prescription and on radiobiological issues concerning the radiation side effects to healthy tissue. For comparison of alternative treatment methods, there is a need for more precise information on the dose distribution in the target tissue and on the absorbed dose to non-target tissues. From the point of view of radiation protection for the patient, information is needed on the resulting effective dose under normal conditions and on dose consequences in the event of an incident during treatment. Additionally, characteristic information is necessary regarding radiation protection of workers and shielding requirements in working areas.

MATERIALS AND METHODS

Using the EGS4 Monte Carlo code, we performed dosimetric calculations for a variety of sources and radionuclides⁽²⁾. Calculation of the depth dose distribution provides insight in the accuracy of prescribed dose and the gradient of absorbed dose in target tissue, and in the absorbed dose in non-target tissue. For modelling purposes the sources and coronary vessels were approximated by concentric cylinders. The source is assumed to be exactly in the centre of the lumen. In our model the vessel wall, the vessel content and other tissues consist of ICRU-tissue. No attempt was made to model vessel curvature or attenuation by stents or plaque material on vessel walls. Dose calculations apply to homogenous wire sources and commercially available ribbons and balloon sources. The wire sources include three beta emitters ($^{90}\text{Sr}/^{90}\text{Y}$, ^{32}P , $^{188}\text{W}/^{188}\text{Re}$), two gamma sources (^{192}Ir , ^{169}Yb) and two X-ray emitters (^{125}I , ^{103}Pd). The wire sources were modelled as solid cylinders of length 3 cm and diameter 0.3 mm. Table 1, section A shows the basic physical properties of these sources. The commercial treatment systems are the $^{90}\text{Sr}/^{90}\text{Y}$ system by Novoste[®], and the ^{192}Ir source by Best Medical[®]. Both systems consist of an array of small seed sources embedded in a nylon ribbon. The Novoste system uses an array of twelve seeds of total length 3 cm. The Best system provides arrays of 6 to 22

seeds corresponding with lengths of 2.3 cm up to 8.7 cm. These sources were modelled using the manufacturer's specifications.

We checked the validity of our model by comparing our results with the outcome of calculations with recently published software for the calculation of absorbed dose as function of depth⁽³⁾. We found good agreement between results.

THERAPEUTIC DOSIMETRY

The reference value for dose prescription in endovascular brachytherapy is an absorbed dose of 15-25 Gy at a reference point in the first one or two millimetres of the lumen wall. Lumen diameters range typically from 1.5 to 5.5 mm, which implies that the distance between source and target tissue is only a few millimetres. The accuracy and the reproducibility of dose delivery are difficult because of these small distances and because of the irregular shape and fast movement of vessels. In practice, there is a need for source configurations that guarantee predictable and consistent depth dose distributions in target tissue. In this context it is highly important to note that sources with different geometries and/or different radionuclides cause a great variety of dose distributions. An additional requirement for sources is that they must yield a small dose gradient in the target tissue, because of treatment efficacy in combination with better dose accuracy and lower doses to healthy proximal tissue. In order to compare the characteristic dosimetric properties of different sources, we calculated depth dose distributions for a variety of sources.

Figure 1 shows percentage depth dose distributions normalised at 2 mm, for seven wire sources and two ribbon sources. The slopes of the depth dose curves for different radionuclides clearly correlate with emitted radiation types and energies. In Table 1, section B, a summary is given of the calculated dose gradients corresponding with a prescribed absorbed dose of 20 Gy at three reference distances from the source centre: 1 mm (line a), 2 mm (line b) and 3 mm (line c), respectively. These data show that different sources score differently in utility where it regards the dose gradient. Both β^- - and photon emitters show relatively large dose gradients at the extremely short distance of 1 mm; on the other hand the dose gradient for X-ray and gamma emitters rapidly decreases for source to target distances above 2 mm.

Another key parameter for comparison of different sources is the source activity. The product of source activity and treatment time that is needed for a prescribed dose at the point of reference, is the characteristic parameter for describing the source strength. In mathematical terms this is the time-integral of source activity during treatment time, or equivalently, the number of radioactive decays in source material during the therapeutic exposure. This parameter is shortly called U_s , and is expressed in units of GBq·minute. In section C of Table 1 we show typical values of U_s yielding an absorbed dose of 20 Gy at points of interest. It is important to notice that the required source strength for photon emitters is drastically higher than for β^- -emitters. It is easy to conclude widely different scores in utility where it regards source activity. This is important for radiation protection, as discussed in the next section.

RADIATION PROTECTION OF THE PATIENT

On the basis of the typical U_s -values for the time-integral of source activity, we assessed the 'truncated' effective dose to the patient. The calculation model is based upon the well-known MIRD-scheme for internal radiation dosimetry⁽⁴⁾. As a first approximation the heart wall was taken as the source organ containing the radioactivity. For each photon source we applied the appropriate energy spectrum. For the beta-sources we used the bremsstrahlung characteristics obtained by from Monte Carlo calculations. With equivalent doses per target organ, the effective dose was calculated according to

the latest ICRP recommendations⁽⁵⁾. We excluded the heart wall as target organ; this is indicated by the term 'truncated'. Typical values for the effective dose, as summarised in section D of Table 1, refer to 3 cm source length. Note that actual values depend on the actual lineal activity and source length. Outcomes are very low for beta sources, but significant for gamma sources. Values range from 5 to 25 mSv. Note that this is of the same magnitude as the effective dose from a cardiovascular X-ray intervention procedure, such as a PTCA or a stent implant.

Serious attention is needed for dose consequences of mishaps during treatment. Conceivable mishaps are for example: failure of source withdrawal from the patient and rupture of a balloon source filled with radioactive liquid. In the case of withdrawal failure the resulting dose to the patient is proportional with exposure time. Dose consequences can be assessed by using the data in section D in Table 1. Practice has learned there is a serious probability of such an incident. Therefore, procedures for trials must include incident preparedness measures. Calculations with the LUDEP model show that balloon rupture can lead to several grays of absorbed dose commitment in critical organs⁽⁶⁾. Typical values for the resulting effective dose commitment from such an incident are around 500 mSv. The use of balloon sources is a high risk procedure, knowing that the overall occurrence of balloon rupture in cardiology is about 0.1%

RADIATION PROTECTION OF WORKERS

The personal dose $H_p(10)$ for workers around the patient for a single procedure, can be assessed on the basis of the ambient dose equivalent at 1 metre from the patient. Typical values for this characteristic quantity are summarised in section E of Table 1. From this data it is easy to conclude that shielding is a necessity when photon emitters are used. However, when using low energy X-ray sources, adequate protection is implicitly guaranteed by the same shielding provisions as are good practice in interventional cardiology. The use of gamma sources necessitates additional lead shielding and keeping distance from the patient. Procedures with ¹⁹²Ir sources typically require 3 cm of lead shielding for adequate reduction of radiation levels. For ¹⁶⁹Yb, shielding with 1 cm of lead is sufficient.

The need for special care during handling of gamma sources is clearly demonstrated by applying the inverse square law to the typical values for the ambient dose equivalent at 1 metre. Exposure rates at short distances are very high. Handheld delivery systems for high dose rate/high activity gamma sources must be regarded as a bad practice.

It is obvious that manipulation of beta sources requires special attention for radiation protection. This is demonstrated by typical values for the skin dose at 30 cm distance (1-5 mGy/min), and by the contact skin dose rate of about 0.1 Gy/s.

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A. Source properties: radionuclide, type & encapsulation, radiations and half-life									
³² P	¹⁸⁸ W/ ¹⁸⁸ Re	⁹⁰ Sr/ ⁹⁰ Y	¹⁰³ Pd	¹²⁵ I	¹⁹² Ir	¹⁶⁹ Yb			
wire steel	wire W/Re	wire Y Novoste	wire steel	wire tissue	wire Pt/Ir	Best	wire Pt		
β ⁻	β ⁻ (γ)	β ⁻	X	X	γ/(β ⁻)		γ/X		
14 days	69 days	29 years	17 days	59 days	74 days		32 days		
B. Absorbed dose gradient, in Gy per 0.1 mm									
a.	2.9	2.7	2.5	2.5	2.5	2.6	2.4	1.9	2.1
b.	2.5	2.0	1.8	1.8	1.2	1.4	1.2	1.1	1.1
c.	2.3	1.8	1.6	1.6	0.8	1.0	0.7	0.8	0.7
C. Time-integrated source activity, in GBq×min									
a.	2.1	6.9	1.8	1.6	2.4×10 ³	2.2×10 ²	66	96	8.5×10 ²
b.	7.2	21	5.1	4.7	5.4×10 ³	5.5×10 ²	2.7×10 ²	2.4×10 ²	1.8×10 ³
c.	24	57	12	11	8.8×10 ³	1.0×10 ³	4.2×10 ²	3.9×10 ²	2.8×10 ³
D. Effective dose patient (truncated), in mSv									
a.	0.001	0.02	0.002	0.002	1.2	0.7	3.5	5.1	6.9
b.	0.002	0.07	0.005	0.004	2.6	1.8	14	12	15
c.	0.008	0.19	0.011	0.010	4.3	3.3	22	21	23
E. Ambient dose equivalent at 1 metre from patient, in mSv									
a.	2×10 ⁻⁵	0.002	5×10 ⁻⁵	5×10 ⁻⁵	0.08	0.09	0.2	0.2	0.3
b.	9×10 ⁻⁵	0.005	1×10 ⁻⁴	1×10 ⁻⁴	0.2	0.2	0.6	0.6	0.6
c.	3×10 ⁻⁴	0.01	3×10 ⁻⁴	3×10 ⁻⁴	0.3	0.4	1	1	0.9

Table 1 Summary of dosimetric characteristics for a single procedure

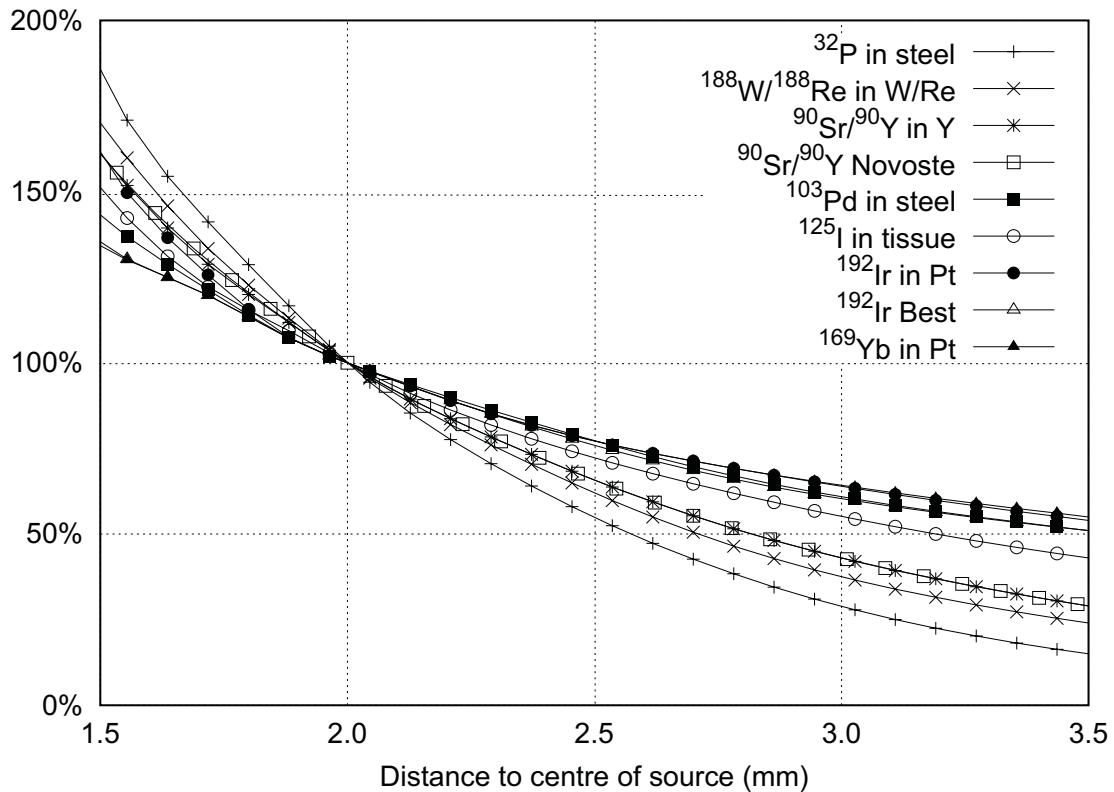


Figure 1 Percentage depth dose distribution, relative to unit dose at 2 millimetres from the source centre (2 mm = 100 %)