Determination of the accuracy of implant reconstruction and dose delivery in brachytherapy in The Netherlands and Belgium

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Abstract

Purpose: To gain insight into the accuracy of brachytherapy treatments, the accuracy of implant reconstruction and dose delivery was investigated in 33 radiotherapy institutions in The Netherlands and Belgium.

Materials and methods: The accuracy of the implant reconstruction method was determined using a cubic phantom containing 25 spheres at well-known positions. Reconstruction measurements were obtained on 41 brachytherapy localizers, 33 of which were simulators. The reconstructed distances between the spheres were compared with the true distances. The accuracy of the dose delivery was determined for high dose rate (HDR), pulsed dose rate (PDR) and low dose rate (LDR) afterloading systems using a polymethyl methacrylate cylindrical phantom containing a NE 2571 ionization chamber in its centre. The institutions were asked to deliver a prescribed dose at the centre of the phantom. The measured dose was compared with the prescribed dose.

Results: The average reconstruction accuracy was -0.07 mm (± 0.4 mm, 1 SD) for 41 localizers. The average deviation of the measured dose from the prescribed dose was $+0.9\%$ ($\pm 1.3\%$, 1 SD) for 21 HDR afterloading systems, $+1.0\%$ ($\pm 2.3\%$, 1 SD) for 12 PDR afterloaders, and $+1.8\%$ ($\pm 2.5\%$, 1 SD) for 15 LDR afterloaders.

Conclusions: This comparison showed a good accuracy of brachytherapy implant reconstruction and dose delivery in The Netherlands and Belgium. © 2001 Elsevier Science Ireland Ltd. All rights reserved.

Keywords: Brachytherapy; Dose delivery; Reconstruction accuracy; Quality assurance

1. Introduction

Brachytherapy is performed in 39 different institutions in The Netherlands and Belgium. The remote afterloading equipment that is currently installed in these institutions is shown in Table 1. Each institution has its own quality assurance (QA) programme for checking the accuracy of relevant physical parameters, guided by the many directives published on this subject [3–5,7,8,11,12,14,21,25]. Because of this, the test frequency, test methodology and accuracy criterion of the QA tests differ significantly between the institutions.

A taskgroup on QA of brachytherapy systems of The Netherlands Commission on Radiation Dosimetry (NCS) has developed guidelines for QA in brachytherapy [17], in the same way as was done before for medical accelerators and simulators and computed tomography (CT) scanners [15,16]. To achieve this goal, four stages have been distinguished by the task group.

1. Gain insight into the current practice of QA of brachytherapy.
2. Determine the accuracy of implant reconstruction and dose delivery in all institutions.
3. Compare the current QA practice with existing recommendations on QA.
4. Formulate a set of minimum requirements on QA.

To gain insight into the current QA practice, an extensive

Table 1
Remote afterloading high dose rate (HDR), pulsed dose rate (PDR), and low dose rate (LDR) systems installed in The Netherlands and Belgium

Brachytherapy	Afterloading system	Source(s)	No. of installed systems
HDR	Nucletron[a] microSelectron	^{192}Ir	17
HDR	Sauerwein[b] Gammamed	^{192}Ir	3
HDR	Varian[c] Varisource	^{192}Ir	1
PDR	Nucletron microSelectron	^{192}Ir	12
PDR	Sauerwein Gammamed	^{192}Ir	3
LDR	Nucletron Selectron	^{137}Cs	16
LDR	Nucletron microSelectron	^{192}Ir/^{137}Cs	13
LDR	Arplay[d] Telegyn	^{137}Cs	1
LDR	CIS bio[e] Curietron	^{137}Cs	2

[a] Nucletron B.V., The Netherlands.
[b] Isotopen-Technik Dr. Sauerwein GmbH, Germany.
[c] Varian Medical Systems, Inc., USA.
[d] Arplay Medical, France.
[e] CIS bio International, France.

questionnaire on the QA of brachytherapy systems was completed by the radiotherapy institutions in The Netherlands and Belgium in 1998. The accuracy of the implant reconstruction and dose delivery was determined by performing on-site measurements with two dedicated phantoms in the institutions. The current QA practice and the results of the on-site measurements were compared with international QA recommendations. From this, a set of minimum requirements on QA, suitable for the situation in The Netherlands and Belgium, was formulated and published as an NCS report [17].

The accuracy of a brachytherapy treatment depends on many parameters, such as source activity, source position, irradiation time, implant reconstruction and dose calculated by the treatment planning software. In practice, an accuracy of physical dose delivery of 5–10% is thought to be achievable [12].

Clearly, source calibration is the physical parameter in brachytherapy dose delivery that is most frequently studied [1,2,18,20,23]. However, the dose delivery accuracy depends on many physical parameters. Although a check of the individual parameters is part of most QA programmes, the delivered dose is generally not directly determined. It was decided to gain more insight into the accuracy of implant reconstruction and dose delivery in brachytherapy. Therefore, on-site measurements with two dedicated phantoms were performed in the institutions using brachytherapy in The Netherlands and Belgium. The methods and results are presented in this paper.

2. Materials and methods

2.1. Reconstruction measurements

To check the reconstruction methods used with brachytherapy localizers, a cubic polymethyl methacrylate (PMMA) phantom (further referred to as reconstruction phantom) was used consisting of six identical 20 mm thick slabs [3]. At each interface between the slabs, five 2 mm spheres are inserted (see Fig. 1). The positions of the spheres are known with an accuracy of ±0.10 mm (1 SD).

The institutions were asked to reconstruct the phantom using the equipment and procedures that are routinely used for reconstruction of brachytherapy implants. For this purpose, the phantom was positioned on the treatment table with the central marker close to the isocentre of the localizer, in the case of an isocentric localizer. For non-isocentric localizers, a reconstruction box was used. The co-ordinates of the spheres were determined by reconstruction from two X-ray films, except for a CT-based reconstruction. From these co-ordinates, 300 inter-sphere distances were calculated, ranging from 20 to 140 mm, with an average of 66 mm. The reconstructed distances were compared with the true distances leading to an average deviation of the 300 inter-sphere distances. In this way, the final result reflects the geometrical accuracy of the localizer, the digitizer and the reconstruction algorithm.

Reconstruction of brachytherapy implants is commonly performed on a simulator (29 institutions). Some institutions have more than one localizer available for brachytherapy implant reconstruction. The reconstruction measurements were performed on 41 localizers (33 simulators, five C-arms, one mobile X-ray unit, one integrated brachytherapy unit localizer and one CT scanner) in 33 institutions. The magnification factor used as input in case of isocentric localization varied between 1.35 and 1.55. Orthogonal reconstruction was used on all simulators, except for one case, where a variable angle reconstruction was applied using gantry angles of $-30°$ and $+30°$.

2.2. Dose measurements

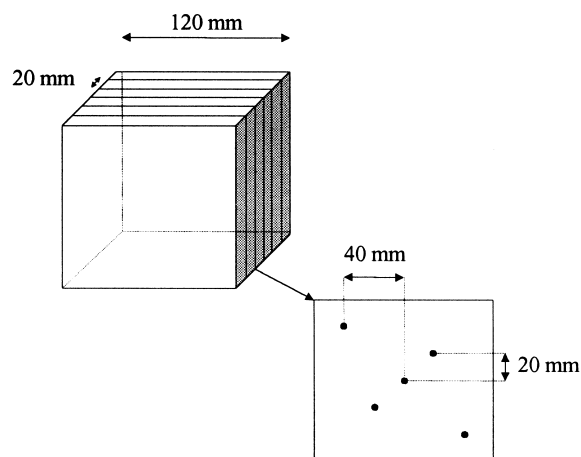To determine the accuracy of the dose delivery in the

Fig. 1. The geometry of the reconstruction phantom. The phantom contains 25 spheres at well-known positions, from which 300 inter-sphere distances can be calculated, varying between 20 and 140 mm.
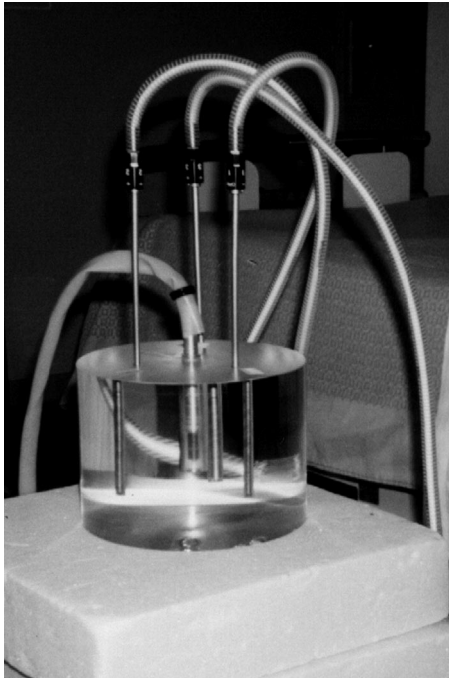
Fig. 2. Cylindrical dose phantom. Three applicators are positioned at 5.0 cm from the centrally placed ionization chamber.

institutions, a solid phantom (further referred to as dose phantom) was used, as designed by Meertens [10]. The phantom is a PMMA cylinder with a diameter of 20.0 cm and a height of 15.0 cm. An NE 2571 (Nuclear Enterprises Ltd, UK) ionization chamber is positioned centrally in the phantom and three brachytherapy applicators are placed at 5.0 cm from the ionization chamber, equally spaced at 120° angles (see Fig. 2). The dose delivered during the treatment is measured by the ionization chamber surrounded by its build-up cap, in combination with a Precitron (Precitron AB, Sweden) Janus electrometer. The chamber was calibrated in combination with the electrometer in September 1998 at The Netherlands Measurements Institute, the National Standards Laboratory for Ionizing Radiation.

2.2.1. Measurement procedure

The institutions were asked to prepare a treatment plan to deliver a prescribed dose in the centre of the ionization chamber using fixed source positions. The source positions used during the measurements are based on the source calibration protocols for ^{192}Ir HDR sources [14] and ^{137}Cs pellets [13] for this phantom. For HDR and PDR afterloaders, one source position per catheter is used in the same plane as the effective point of measurement (the centre of the ionization chamber). For Selectron LDR afterloaders, six sources per catheter are used, three on both sides of the central plane of the phantom at distances of 17.5, 20.0 and 22.5 mm from that plane (Fig. 3).

Because the reconstruction accuracy is measured separately, the source positions are entered in the treatment planning system (TPS) using co-ordinates and not by

reconstruction of the phantom. In this way, the error in dose delivery caused by an error in reconstruction of the set-up is avoided. However, in some TPSs, source position co-ordinates cannot be entered directly. In the two institutions using such a TPS, the geometry was reconstructed from a drawing of the experimental set-up as accurately as possible.

The value for the prescribed dose was determined as a compromise between the accuracy of the electrometer and practical considerations, such as the required measuring time. A prescribed dose of 75 cGy for HDR, 40 cGy for PDR, and 20 cGy for LDR afterloaders was used. These doses resulted in treatment times (depending on the exact source strength) of approximately 5 min for a 2 cGym^2/h HDR ^{192}Ir source, 25 min for a 0.22 cGym^2/h ^{192}Ir PDR source, and 45 min for 18 ^{137}Cs LDR sources with a source strength of 0.004 cGym^2/h.

After installation of the phantom, air temperature and pressure were recorded. The dose delivery measurement is performed three times, using the dwell times calculated by the TPS. The measurement is performed with plastic catheters or needles, corresponding to the normal use for treatments in the institution.

2.2.2. Conversion of electrometer readings to dose in water

The electrometer reading is converted to a dose to water value using the equation [14]:

$$D_{\mathrm{w}} = M N_{\mathrm{K}} \Pi k_i \Pi p_i \Pi f_i S(d) \left(\frac{\mu}{\rho} \right)_{\mathrm{air}}^{\mathrm{water}} (1 - g) \qquad (1)$$

where $M = M_{\mathrm{uncorr}} p_t p_p p_{\mathrm{hum}} p_{\mathrm{ion}} p_{\mathrm{pol}}$, $\Pi p_i = p_r p_{\mathrm{ce}}$, $\Pi k_i = k_w k_{\mathrm{st}} k_{\mathrm{ce}}$, and $\Pi f_i = f_{\mathrm{tr}} f_{\mathrm{ph}} f_{\mathrm{geo}} f_{\mathrm{cath}}$.

The meaning and value of these factors are described in Table 2. Most of these factors were determined in previous studies [10,13,14,22]. The air kerma calibration factor for ^{192}Ir and ^{137}Cs of the ionization chamber with build-up cap in combination with the electrometer was derived from N_{K} factors obtained during calibration at the National Standards Laboratory. Values for the mass-energy absorption coefficient $(\mu/\rho)_{\mathrm{air}}^{\mathrm{water}}$ for ^{192}Ir and ^{137}Cs were taken from the literature [4,19]. In TPSs, either the value 1.10 or 1.11 is used. Here, the value 1.11 was used in the calculations. The influence of the transit dose (f_{tr}) and the applicator attenuation (f_{cath}) had to be determined separately and will be discussed in the next paragraphs.

2.2.3. Determination of the transit dose correction factor

The clinical treatment planning generally neglects the transit dose, i.e. the dose delivered during transport of the source from the afterloader to the patient. The transit dose depends on the source strength, the velocity of the source transport and the geometry of the set-up. To compare the dose measured in the phantom with the dose calculated by the planning system, and to compare measurements in different institutions, the measured dose was corrected for the transit dose using the factor f_{tr} [14]. For a fixed geome-
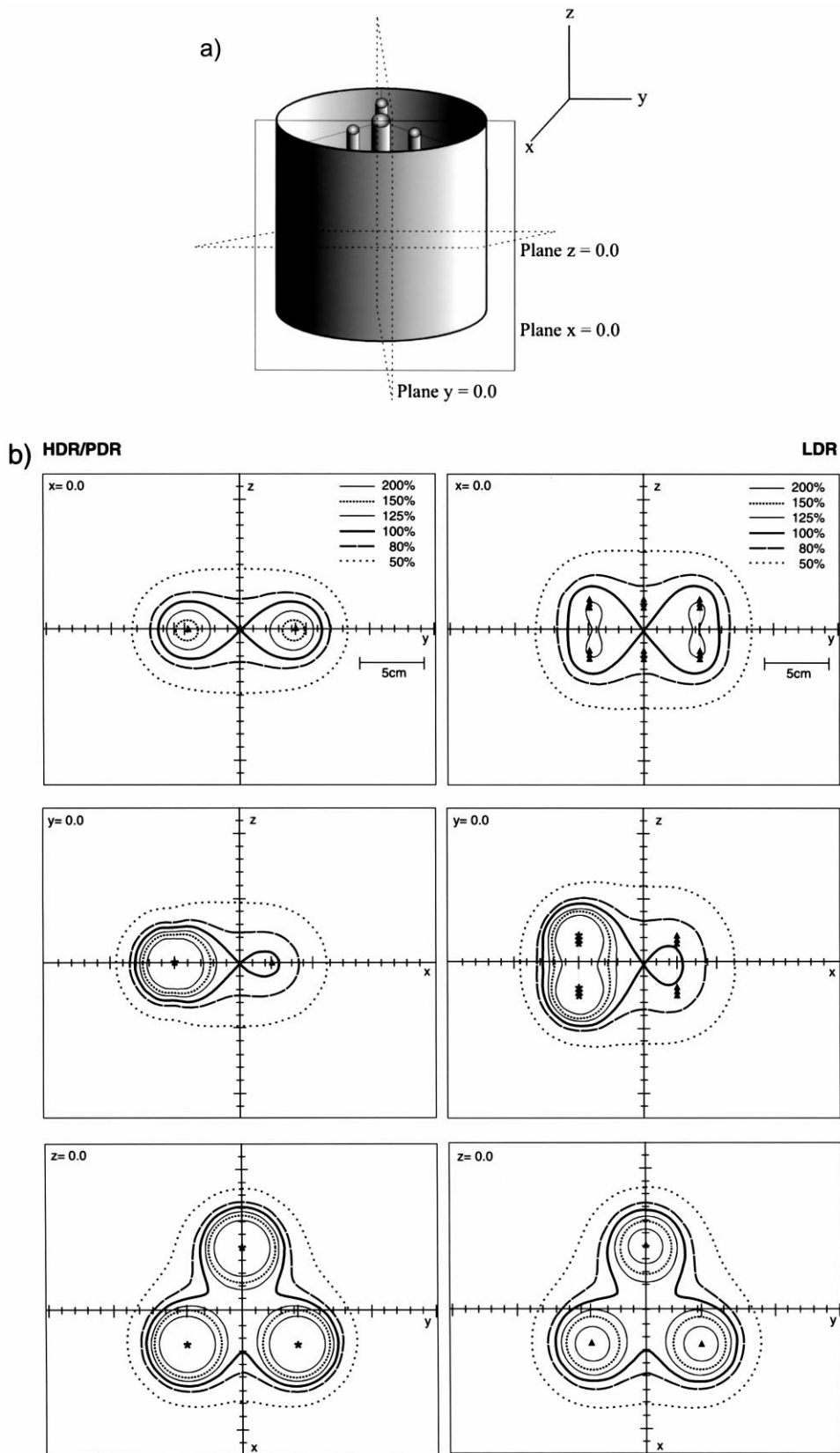
Fig. 3. The dose distribution for the cylindrical dose phantom (a) for HDR/PDR and Selectron LDR afterloaders in three orthogonal planes passing through the effective measurement point of the ionization chamber (b).

Table 2
Factors for conversion of the electrometer reading to dose in water for ^{192}Ir and ^{137}Cs sources measured in the PMMA cylindrical phantom (see also Refs. [10,13,14,22])

Factor	Description	Value for ^{192}Ir in the phantom	Value for ^{137}Cs in the phantom
M_{uncorr}	Uncorrected instrument reading		
p_t	Temperature correction factor	$(T_{meas} + 273.15/T_{calibration})$	$(T_{meas} + 273.15/T_{calibration})$
p_p	Air pressure correction factor	$(p_{calibration}/p_{meas})$	$(p_{calibration}/p_{meas})$
p_{hum}	Humidity correction factor	1.000	1.000
p_{ion}	Ion recombination correction factor	1.000	1.000
p_{pol}	Correction factor for polarity effects	1.000	1.000
N_K	Air kerma calibration factor	From calibration	From calibration
p_r	Correction factor for replacement of PMMA by the ionization chamber	1.016	0.997
p_{ce}	Correction factor for the effect of the central electrode during the measurement	1.000	1.000
k_w	Correction factor for attenuation and scatter in the chamber wall	0.984	0.9904
k_{st}	Correction factor for the stem effect during calibration	1.000	1.000
k_{ce}	Correction factor for the effect of the central electrode during calibration	1.000	1.000
f_{tr}	Correction factor for source transport time	Variable	1.000
f_{ph}	Conversion factor from the specified PMMA phantom to a full-scatter water phantom	1.033	1.041
f_{geo}	Correction factor for absorption and scatter in water	$1/S(d)$	$1/S(d)$
f_{cath}	Correction factor for attenuation in plastic catheter/needles	1.000/1.009	Included in f_{ph}
$S(d)$	Correction factor for scattering and absorption in the phantom material	See Refs. [10,13,14,22]	See Refs. [10,13,14,22]
$(\mu/\rho)_{air}^{water}$	Mass-energy absorption coefficient	1.11	1.11
g	Fraction of kinetic energy of secondary particles converted to bremsstrahlung	0.000	0.003

try, such as the dose phantom, the value for this factor can be derived from:

$$f_{tr} = 1 - \frac{M_{t0}}{M_t} \qquad (2)$$

where t is the dwell time, M_{t0} is the electrometer reading at $t = 0$ (zero dwell time, only dose contribution during source transport) and M_t is the electrometer reading for dwell time t. The value for $t = 0$, M_{t0}, is determined for the specific geometry by programming dwell times in the range of 5–120 s per channel and by linear extrapolation of the measured doses to $t = 0$. Since the transit dose linearly depends on the source strength, and experience has shown that the source transport velocities on identical machines are comparable, its value for identical machines in different institutions can be calculated from the value measured on a single afterloader with reasonable accuracy.

2.2.4. Attenuation in the applicator wall

To determine the difference in attenuation between plastic catheters and metal needles during HDR/PDR measurements, a correction factor f_{cath} was determined. The value for f_{cath} was determined as the ratio of the reading obtained when using plastic catheters and the reading obtained when using needles, and was found to be 1.009 (±0.003, 1 SD). To compare the dose measured in the phantom with the dose calculated by the TPS, and to compare measurements in different institutions, the measured dose was corrected for the attenuation in institutions where needles were used during the measurements.

For the Selectron LDR afterloader, the factor f_{cath} for the replacement of water-equivalent catheters by the stainless steel standard catheters is included in the factor f_{ph} [10].

2.3. Source positioning accuracy for HDR and PDR afterloaders

In the solid phantom dose delivery measurements, the combined accuracy of source calibration, source positioning, timer and dose calculation of the treatment planning software is determined. An error in source calibration, timer or dose calculation will influence the measured dose proportionally. For example, an error in source calibration of 2% will be measured as an error in the delivered dose of 2%. However, an error in source positioning along the catheter will hardly influence the measured dose. This is caused by the relatively uniform dose distribution around the ionization chamber, which is the result of the applied source configuration. To illustrate this, the electrometer reading is plotted as a function of the source position along a single catheter for a HDR unit in Fig. 4. The maximum reading is obtained at $z = 0$ mm, when the source is opposite to the ionization chamber. An error of 2 mm in the source position around this optimum will result in a deviation in the measured dose of only 0.2% (Fig. 4).

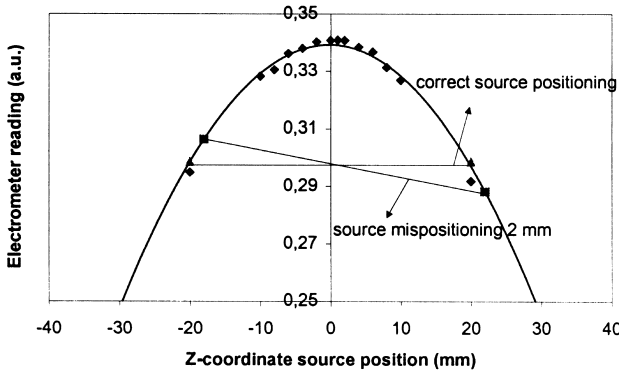In clinical treatments, however, a systematic error in

Fig. 4. Electrometer reading as a function of the source position along the catheter (z-direction) in the dose phantom (for the geometry see Fig. 3). The maximum reading is obtained at $z = 0$ mm, when the source is opposite to the ionization chamber. The difference in reading between $z = -20$ mm and $z = +20$ mm is used to estimate the error in source positioning.

source positioning of 2 mm for the entire implant can result in a significant error in dose delivery because of the possible geometrical miss of the target volume. It would therefore be useful to also obtain information on the source positioning accuracy. Because the effect of source positioning on the electrometer reading is much larger on the slopes of the curve, it was decided for HDR and PDR afterloaders to use two extra measurements at positions $+20$ and -20 mm from the optimum source position ($z = 0$ mm). The difference in electrometer reading between a source at $+20$ mm and a source at -20 mm is used to calculate the error in the source position. For example, if in an institution an error in source positioning of 2 mm exists, the source is positioned at $z = +2$ mm for the measurements; the difference in electrometer reading between source position $z = -18$ mm and $z = +22$ mm is then approximately 6% (Fig. 4), which can easily be measured. For Cs LDR afterloaders, a systematic error in source positioning is less probable because of the fixed source positions. This extra measurement was therefore not performed for the LDR afterloaders.
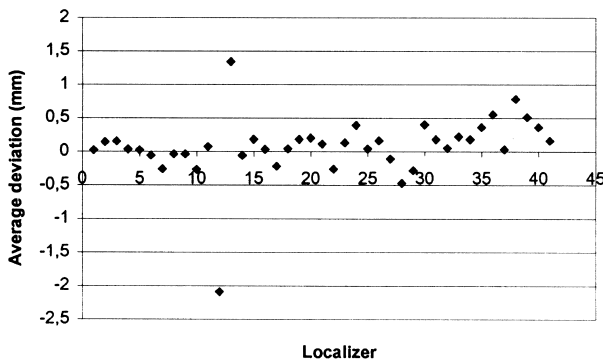

Fig. 5. Results of the reconstruction accuracy measurements on 41 brachytherapy localizers: average deviation of 300 reconstructed distances (ranging from 20 to 140 mm) from the true distances between 25 spheres in the reconstruction phantom. A positive deviation indicates a reconstructed distance larger than the true distance.

3. Results

3.1. Reconstruction measurements

The results of the measurements with the reconstruction phantom for the 41 localizers are shown in Fig. 5 as the average reconstruction error of 300 reconstructed distances. The deviation of the reconstructed distances was smaller than 0.5 mm for 36 localizers, and even smaller than 0.25 mm for 27 localizers. For one C-arm, the error was -2.1 mm, and for another C-arm (with reconstruction box) the deviation was $+1.3$ mm. For the 41 localizers, an average deviation of 0.1 mm (± 0.5 mm, 1 SD) was found.

3.2. Dose measurements

3.2.1. Transit dose correction factor

The transit dose correction factor for the specific phantom has been determined for microSelectron HDR and PDR afterloaders (Nucletron, The Netherlands) by measuring the electrometer reading with the source position opposite to the ionization chamber at different dwell times (see Fig. 6). The measurements are repeated for four afterloaders.

The y-axis cut-off of these curves determines the transit dose delivered, which is expected to vary linearly with source strength. The axis cut-off is converted into dose units and is plotted as a function of the source strength in Fig. 7. The measurements were performed on two HDR and two PDR afterloaders. Since the transit dose linearly depends on the source strength, and because the source transport velocity on identical machines is comparable, its value for identical machines in different institutions was calculated from the value measured on the four afterloaders. For non-Nucletron machines, the transit dose was determined on each individual machine using the described method.

As can be seen from this figure, the transit dose amounts for this specific experimental set-up (i.e. measurements in the solid phantom with the source positioned at 5.0 cm from the applicators and a dwell time range of 0–120 s on Nucle-
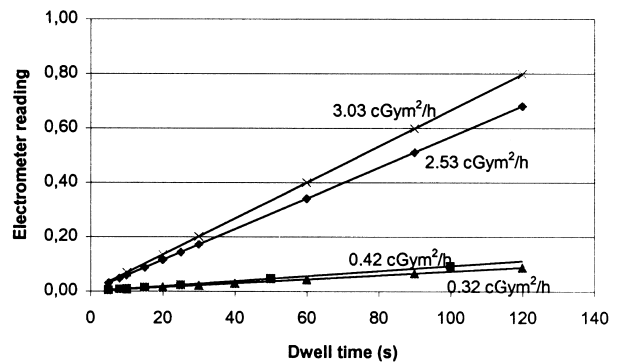

Fig. 6. Determination of the transit dose for microSelectron HDR and PDR afterloaders in four different institutions. The electrometer reading is measured for dwell times varying from 5 to 120 s. The y-axis cut-off represents the transit dose for this specific experimental set-up.
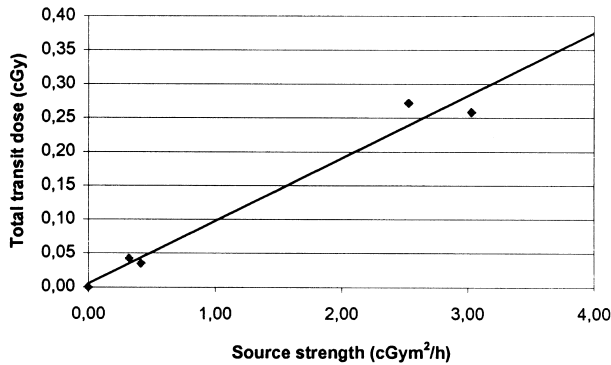
Fig. 7. Results of measurements of the transit dose for the experimental set-up in different institutions as a function of the source strength for micro-Selectron HDR and PDR afterloaders.

tron afterloaders) were approximately 0.4 cGy for a 4 $cGym^2/h$ HDR source and 0.04 cGy for a 0.4 $cGym^2/h$ PDR source. Relative to the prescribed dose of 75 cGy for HDR and 40 cGy for PDR afterloaders, the maximum contribution of the transit dose to the measured dose is 0.5% for HDR and 0.1% for PDR afterloaders. For LDR afterloaders the relative contribution of the transit dose to the prescribed dose has not been determined, but will be lower than for the PDR afterloaders because of the longer measuring times. The measured doses on HDR afterloaders were corrected according to Fig. 7. For PDR and LDR afterloaders, the effect of the transit dose was neglected.

3.2.2. Dose measurements

Dose measurements were performed on 21 HDR afterloaders, 12 PDR afterloaders and 15 LDR afterloaders. The results of the dose measurements are summarized in Fig. 8. In the histogram, the percentage difference is shown between the measured dose and the prescribed dose. For the 21 HDR afterloaders, an average deviation of +0.9% ($\pm$1.3%, 1 SD) was measured, with two afterloaders having a deviation of more than 2%. For the 12 PDR afterloaders, the average deviation was +1.0% ($\pm$2.3%, 1 SD), with five afterloaders having a deviation of more than 2%. For measurements on 15 Selectron LDR afterloaders, an average deviation of +1.8% ($\pm$2.5%, 1 SD) was found, with five afterloaders having a deviation of more than 2% and a range of −3.3 to +6.8%.

3.3. Source positioning accuracy

The accuracy of the source positioning was measured for HDR and PDR afterloaders by the method described in Section 2.3. The average error on source positioning was measured on 16 HDR afterloaders and was found to be 0.2 mm ($\pm$1.0 mm, 1 SD). In three institutions, a deviation larger than 1 mm was observed, with a maximum deviation of 2.3 mm. For 11 PDR afterloaders, the average error in source positioning was 0.2 mm ($\pm$1.1 mm, 1 SD), with four

institutions having deviations larger than 1 mm, and a maximum deviation of 2.0 mm.

4. Discussion

4.1. Reconstruction measurements

The results of the reconstruction measurements (Fig. 5) reflect the geometrical accuracy of the localizer, the digitizer and the reconstruction algorithm. For two C-arms, an average deviation of larger than 1 mm was observed. The reason for this deviation can be found when plotting the deviation of each reconstructed distance for this C-arm as a function of the reconstructed distance (Fig. 9). For comparison a similar graph is shown for a simulator with a very small average deviation.

As can be seen from Fig. 9, the deviation of the reconstructed distance increases with the distance between the spheres for the C-arm, whereas for the simulator there is no correlation between deviation and distance. This shows that the deviation for the C-arm is most likely caused by an incorrect magnification factor of the localizer.

In a comparison of computer planning systems performed in 1987, Visser [24] found deviations of up to 5 mm when reconstructing gynaecological applicators from orthogonal radiographs, leading to discrepancies of 10% in the dose rate. Kolkman-Deurloo et al. [9] found an average reconstruction error of 0.04 mm for a dedicated brachytherapy localizer using a similar phantom and orthogonal reconstruction. This value increased to 0.4 mm when reconstructing from fluoroscopy images instead of using film. It was found that these deviations did not influence the treatment time when considering clinical implant geometries.

The results in the present work indicate that the reconstruction accuracy errors in The Netherlands and Belgium are comparable to those found by Kolkman-Deurloo et al. [9] and that these errors will have only a minor effect on the
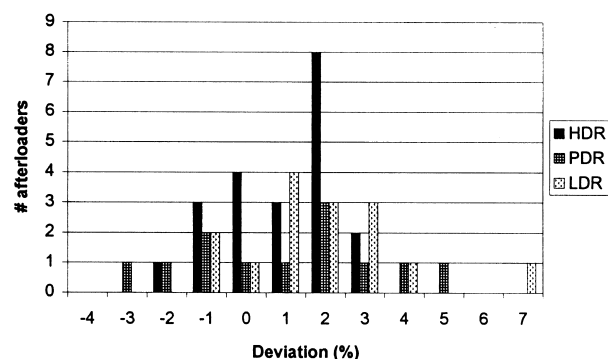

Fig. 8. Results of dose delivery measurements on 21 HDR, 12 PDR and 15 LDR afterloaders in The Netherlands and Belgium: percentage difference between the measured dose in the phantom and the prescribed dose in the planning system. The average deviation is +0.9% for HDR, +1.0% for PDR, and +1.8% for LDR afterloaders. All results were within a range of −3.3 to +6.8%.
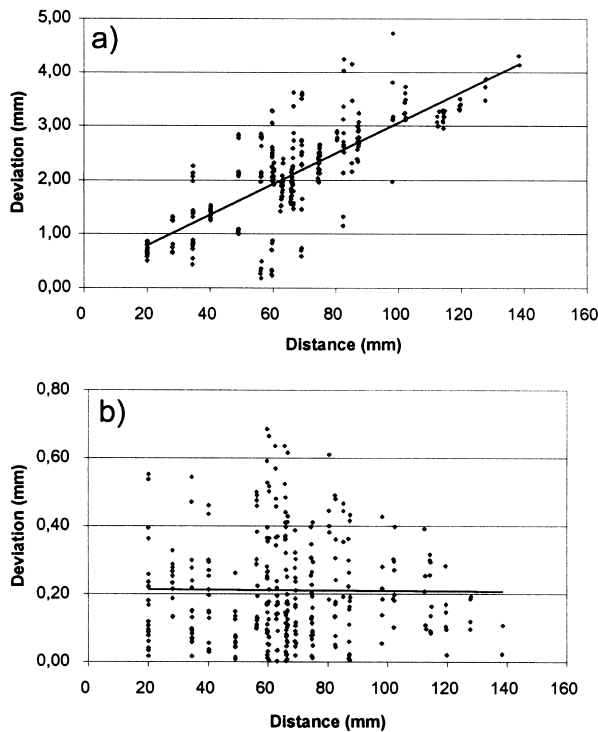
Fig. 9. Absolute value of the reconstruction deviation as a function of the reconstructed distance for a C-arm with an average deviation of 2.09 mm (a) and a simulator with an average deviation of 0.03 mm (b). The line shows that the deviation increases with reconstructed distance for the C-arm, while this is not the case for the simulator.

overall treatment time. The reconstruction accuracy has improved enormously since the 1987 investigation of Visser [24]. Based on the results of these measurements, the task group recommends an annual check of the reconstruction accuracy by reconstruction of a phantom with a well-known geometry. It is emphasized that the reconstruction error measured in this work represents the physical error in an ideal situation. In practice, the error can be larger due to, for example, patient movement.

4.2. Dose measurements

The difference between the measured dose and the prescribed dose is the result of the sum of errors in source calibration, dose calculation of the TPS, source positioning and irradiation timer.

Clearly, source activity is one of the most important physical parameters in brachytherapy dose delivery. In a comparison of source calibration procedures for ^{192}Ir HDR sources in The Netherlands and Belgium performed in 1992, Venselaar et al. [23] found deviations from the certificate's value of up to 6.8%. Baltas et al. [2] reported deviations from the certificate in the range of -13.0 to $+6.0\%$ for ^{192}Ir HDR sources. In a recent comparison of calibration procedures for ^{192}Ir HDR sources in Brazil, de Almeida et al. [1] found deviations in the range of -4.6 to $+2.4\%$.

For ^{137}Cs LDR sources, Meertens [10] found deviations from the source strength certificate in the range of -2.3 to $+3.9\%$ in nine institutions.

It is recommended by the NCS that the user of ^{192}Ir after-loading equipment performs his own calibration for each new source [14]. In the current investigation, it was found that the strength of each ^{192}Ir HDR source is independently determined in 20 out of 21 institutions using HDR and in all institutions using PDR, mostly using the in-air method with a calibration jig. In one institution, the calibration of the source is not yet performed, but is planned for the near future. The current practice of in-house source calibration in The Netherlands and Belgium was investigated by means of a questionnaire on the QA of brachytherapy and is summarized in Table 3.

Apart from source calibration, the dose calculation by the TPS is expected to be a significant cause of deviations in the measured dose. For example, Feroldi et al. [6] showed in a comparison of TPSs that dose rate errors larger than 5% may occur. To investigate the effect of the different TPSs on the total treatment time, the treatment time as calculated by the TPS in each institution for the specific experimental set-up and prescribed dose was normalized to a fixed source strength. The treatment time calculated by the TPS of an institution is then compared with the average normalized treatment time of all institutions. For HDR and PDR brachytherapy the normalized treatment times for the different TPSs varied between -0.7 and $+1.3\%$ around the average calculated treatment time. For TPSs in use for Selectron LDR brachytherapy, the effect of the TPS on the measured

Table 3
Source calibration practice of HDR, PDR and LDR brachytherapy in The Netherlands and Belgium[a]

	In-house calibration method				Value in TPS	
	In-air	In-phantom	Well chamber	No calibration	In-house measured	Certificate
HDR	15	2	3	1	14	7
PDR	11	1	–	–	8	4
Cs LDR	2	7	4	2	5	10

[a] The number of institutions performing an in-house calibration method as well as the type of calibration method are listed. Furthermore, it is listed whether the in-house measured source activity value or the value from the certificate is used in the TPS. The different methods available for in-house calibration of brachytherapy sources are described in Refs. [10,13,14].

dose was larger: the calculated treatment times ranged from -2.9 to $+3.7\%$ around the average time. The effect of applicator attenuation is ignored in most TPSs. The use of needles instead of plastic catheters results in an attenuation of approximately 0.9%. The effect of the TPS on the measured dose is thus found to be small ($<2\%$) for TPSs in use for HDR and PDR brachytherapy and larger (up to 4%) for TPSs in use for LDR brachytherapy.

In our study, for one PDR afterloader, a deviation in dose of $+5.0\%$ was measured. The reason for this deviation was found to be an error in source positioning within the calibration jig. After repair of the jig, the deviation was reduced to $+2.4\%$. For one Cs LDR afterloader, a deviation in dose of $+6.8\%$ was measured. Repeating the dose measurement several weeks later confirmed the initial measurement. The reason for the deviation is most probably an error in the source strength value as specified on the certificate of the manufacturer. This was never checked independently and was used in the TPS.

It should be emphasized that the measured values represent the physical accuracy of the treatment. In practice, one should keep in mind that these ideal conditions are not always fulfilled due to, for example, patient movement and the lack of full scatter conditions. Therefore, the accuracy of the dose delivered to a patient will be lower.

4.3. Source positioning accuracy

Because of the high dose gradient near brachytherapy sources, accurate source positioning is required to obtain accurate dose delivery. In international reports, tolerance levels of 1 mm [12,25] or 2 mm [7,17] are recommended. The results of source positioning measurements on 16 HDR and 11 PDR afterloaders showed that seven institutions had a deviation in source positioning larger than 1 mm. Three of these exceeded the recommended tolerance level of 2 mm.

4.4. Uncertainties

Uncertainties are distinguished as random (type A) and systematic (type B) uncertainties. Type A uncertainties are caused by small deviations in the readings of the electrometer, in the reading of the air pressure and temperature and statistical deviations in the distance between the source and ionization chamber. Since the inner diameter of the (Nucletron) catheters is 1.8 mm and the outer diameter of the ^{192}Ir source is 1.1 mm, there is always an uncertainty in the position of the source within the applicator in the radial direction, with a maximum margin of 0.35 mm. At a 5.0 cm distance from the ionization chamber, this can result in an error of $\pm1.4\%$ in the measured dose, when using a single catheter. In practice, differences as large as 1.4% were observed when comparing the readings from the three different channels of the phantom. However, the reproducibility of the electrometer reading from all three channels was found to be better than 0.3%, so the overall effect of the radial source positioning error on the electrometer reading is small. The random error in the reading of the barometer and thermometer is estimated to be 0.2%.

Type B uncertainties are determined by the uncertainties in the physical quantities as given in Eq. (1) and Table 1. The largest contribution is expected from the uncertainty of the N_K value (0.7%) and from the uncertainty in the replacement factor p_r (1.0%, see Ref. [22]). The uncertainties in the other parameters are shown in Table 4. The estimated combined uncertainty resulting from type A and type B uncertainties for the dose measurements is 1.5%.

5. Conclusions

The accuracy of the reconstruction method and the dose delivery in HDR, PDR and LDR brachytherapy was determined in 33 institutions in The Netherlands and Belgium.

The accuracy of the reconstruction method was determined using a cubic PMMA phantom containing 25 spheres at well-known positions. The reconstruction accuracy was found to be smaller than 0.5 mm for 36 of 41 localizers. For two localizers, a deviation larger than 1.0 mm was observed, which is most probably caused by an error in the magnification factor applied in the TPS.

The dose delivery accuracy was measured using a cylindrical PMMA phantom containing a NE 2571 ionization chamber and three applicators at 5.0 cm from the ionization chamber. The accuracy of the dose delivery was found to be $+0.9\%$ ($\pm1.3\%$, 1 SD) for 21 HDR afterloaders, $+1.0\%$ ($\pm2.3\%$, 1 SD) for 12 PDR afterloaders, and $+1.8\%$ ($\pm2.5\%$, 1 SD) for 15 Selectron LDR afterloaders. For one PDR afterloader, a deviation in dose of $+5.0\%$ was measured. The cause for this deviation was a problem with source positioning within the calibration jig. For one Cs LDR afterloader, a deviation in dose of $+6.8\%$ was measured. The cause for the deviation is most probably an error in the source strength value on the certificate of the manufacturer.

The results of the measurements showed that an accuracy of physical dose delivery better than 5% was achieved in all but one institution.

Table 4
Estimated uncertainties, expressed as one standard deviation, resulting from type A (statistical) and type B (systematical) uncertainties for the dose measurements in the cylindrical phantom[a]

Parameter	Uncertainty type A (%)	Parameter	Uncertainty type B (%)
M_{uncorr}	0.3	N_K	0.7
p_{meas}	0.2	p_r	1.0
T_{meas}	0.2	f_{ph}	0.3
		$(\mu/\rho)_{air}^{water}$	0.3
		$p_{hum}, p_{ion}, p_{pol}, p_{ce}, k_{st}, k_{ce}$	<0.1
		f_{cath}	0.3
		f_{tr}	0.3

[a] For an explanation of the parameters, see Table 2.

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