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INFLUENCE OF DOSEMETER POSITION FOR THE ASSESSMENT OF EYE LENS DOSE DURING INTERVENTIONAL CARDIOLOGY

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The equivalent dose limit for the eye lens for occupational exposure recommended by the ICRP has been reduced to 20 mSv y^{-1} averaged over defined periods of 5 y, with no single year exceeding 50 mSv. The compliance with this new requirement could not be easy in some workplace such as interventional radiology and cardiology. The aim of this study is to evaluate different possible approaches in order to have a good estimate of the eye lens dose during interventional procedures. Measurements were performed with an X-ray system Philips Allura FD-10, using a PMMA phantom to simulate the patient scattered radiation and a Rando phantom to simulate the cardiologist. Thermoluminescence (TL) whole-body and TL eye lens dosemeters together with Philips DoseAware active dosemeters were located on different positions of the Rando phantom to estimate the eye lens dose in typical cardiology procedures. The results show that, for the studied conditions, any of the analysed dosemeter positions are suitable for eye lens dose assessment. However, the centre of the thyroid collar and the left ear position provide a better estimate. Furthermore, in practice, improper use of the ceiling-suspended screen can produce partial protection of some parts of the body, and thus large differences between the measured doses and the actual exposure of the eye could arise if the dosemeter is not situated close to the eye.

INTRODUCTION

Recent studies[1–3] have shown that radiation produces changes in the eye lens after exposure to lower doses than would be expected. This finding highlights the fact that threshold doses seem to be lower than previously considered. In particular, there is a greater incidence of harmful effects after exposure. On this basis, the International Commission on Radiological Protection recommended in publication 118[4] an equivalent dose limit for the lens of the eyes for occupational exposure in planned exposure situations of 20 mSv y^{-1} averaged over defined periods of 5 y, and no single year was to exceed 50 mSv. This limit is significantly lower than the previous one of 150 mSv y^{-1} and could require the introduction of eye lens dose monitoring on a regular basis for some workplace. The IAEA has adopted this new limit in the IAEA Basic Safety Standards[5], and the EC has also included it in its recent EURATOM Directive[6]. These new requirements will need to be implemented in many countries in the short term.

The most accurate method for monitoring the equivalent dose to the lens of the eye, H_{lens}, is to measure the personal dose equivalent at 3 mm of depth, $H_p(3)$, with a dosemeter worn as close as possible to the eye and which has been previously calibrated on a phantom that is representative of the head. However, this procedure is often impractical as it is neither easy nor comfortable to wear a dosemeter close to the eye and also because, at present, there are few dosemeters calibrated in $H_p(3)$.

After the publication of ICRP 118, several documents and literature reviews were published in order to provide guidelines to assess the lens of the eye exposure[7–9]. ISO/DIS 15382[8] provides guidance for the design of a monitoring programme to ensure compliance with the new eye lens dose limit. Several recent studies have found that the new limit could be exceeded for interventional radiologists and interventional cardiologists with high workloads if proper radiation protection conditions are not implemented[9–12], and thus it is important to set up effective monitoring programmes or realistic and practical indirect dose assessment procedures for these workers. An additional difficulty associated with these personnel is that there is a large variation in exposure depending on the clinical procedures carried out and the use of radiation protection tools, such as ceiling suspended screen, goggles, etc.

The aim of this study is to evaluate different possible approaches in order to have a good estimate of the eye lens dose during interventional procedures. In particular, the influence of both the type and position of the dosemeter is analysed.

METHODOLOGY

Dosemeters and calibration

LiF:Mg.Cu.P chips manufactured by Conqueror Electronics Technology Co. Ltd.(Beijing. China, http://www.cet-cns.com/index.htm) under the trade name

TLD-2000C were used as detectors in passive whole-body dosemeters (WBDs) and eye lens dosemeters (ELDs). For the WBDs, the standard badge of the Universitat Politecnica de Catalunya (UPC) personal dosimetry service was used[13]. As ELDs, both an Eye-D[12] manufactured by Radcard (Krakow, Poland, http://www.radcard.pl/) and the UPC dosemeter were used. The latter consists of two thermoluminescence chips inside an 11 mg cm^{-2} opaque polyethylene film. WBDs were calibrated for $H_p(10)$ and $H_p(0.07)$ and ELDs for $H_p(3)$. Details on the read-out and calibration procedure are given in Ginjaume *et al.*[14] and Sánchez *et al.*[13]. TLDs have been type-tested according to CEI/IEC 61066:2006[15] in the UPC secondary standard calibration laboratory. For ELDs, $H_p(3)$ characterisation is based on the practical recommendations presented in ORAMED report[12]. Uncertainties associated with $H_p(d)$ were calculated following the Guide to the Expression of Uncertainty in Measurement[16]. The lower detection limit at a confidence level of 1 % is 1 μSv.

In addition, DoseAware (Philips Medical System, The Netherlands, http://www.healthcare.philips.com/) electronic dosemeters were also used. These dosemeters are specially designed to measure personal dose equivalent $H_p(10)$ within an interventional environment. The manufacturer claims that linearity is ensured from 40 μSv h^{-1} up to 300 mSv h^{-1}, and there is a 20 % variation in energy response between 33 and 100 keV and reports an angular dependence of >30 % for angles of $>50°$. The energy responses of the dosemeters were previously checked in a secondary standard calibration laboratory,[17] but no specific correction factor was used since the response was within the manufacturer's range. Previous works[17] showed differences with UPC WB TLDs within 10–15 % in cardiology scattered radiation fields.

A Thermo EPD MK2 electronic personal dosemeter (Thermo Scientific, Inc., USA, http://www.thermoscientific.com/) was also used to monitor the scatter radiation in real-time. The reading of this dosemeter (50–70 μSv) was used to decide when to stop irradiation in order to guarantee a sufficient signal in the other dosemeters.

Experimental set-up

Measurements were performed at San Carlos University Hospital in Madrid with a Philips Allura (http://www.healthcare.philips.com/) FD-10 X-ray system. Measurements were performed for low-dose fluoroscopy mode (88–114 kV; HVL 8.0- to 10-mm Al) and image acquisition mode (68–84 kV; HVL 3.5- to 4.0-mm Al) and for two projections posterior–anterior (PA) and left anterior oblique at 90° (LAO-90). The phantom entrance surface air kerma was monitored using a calibrated Radcal ionisation chamber model 20x6-60E connected to an electrometer model 20x26C (Radcal corp. Monrovia, CA, USA, http://www.radcal.com/). A $30\cdot\times\ 60\cdot\times\ 20$ cm^3 PMMA slab phantom on the treatment couch was used to simulate patient-scattered radiation. The cardiologist was simulated by an anthropomorphic phantom model Rando (The Phantom laboratory, Salem, NY, USA, http://www.phantomlab.com/), which was situated on the right-hand side of the X-ray tube at a distance of 90 cm between the Rando left eye and the beam focus. A passive WBD and a DoseAware were located on the chest on the lead apron, on the left shoulder and on the collar of the Rando phantom. A UPC ELD and the Eye-D were situated close to the left ear of the phantom. In addition, an UPC ELD was placed on the forehand and two others on the eyes. $H_p(3)$ measured with the ELD dosemeter situated on the left eye was considered the best estimate of the maximum eye lens dose, and thus the other measurements are compared with it.

Measurements were carried out in two stages, in April and July 2013, respectively. For all cases, the table shielding below the couch was correctly placed, as, in practice, it is generally used by cardiologists. In this study, all measurements were done with it in place. During the first set of measurements (experiments 1 to 4 in Table 1), the attenuation of the ceiling

Table 1. Summary of the ten set of measurements' conditions and the estimated H_{lens}.

Experiment	1	2	3	4	5	6	7	8	9	10
Mode	LD	Cine	Cine	Cine	LD	Cine	LD	Cine	LD	Cine
Projection	PA	PA	PA	PA	PA	PA	PA	PA	LAO	LAO
Protection means			CS	Go						
KAP (Gy cm^{-2})	6.2	10.7	33.7	35.0	59.7	12.7	6.5	12.7	6.9	12.7
$H_p(3) \pm$ SD (μSv)	20 ± 1	18 ± 2	3 ± 0.6	7 ± 0.6	25 ± 1	27 ± 1	34 ± 3	30 ± 1	43 ± 3	49 ± 6
$\frac{H_p(3)}{KAP}$ (μSv Gy^{-1} cm^{-2})	3.2	1.7	0.1	0.2	4.2	2.1	5.3	2.4	6.2	3.8

LD, low-dose fluoroscopy; cine, acquisition; PA, posterior–anterior; LAO, left anterior oblique at 90°; CS, ceiling suspended screen; Go, goggles; KAP, air kerma–area product calculated from the Radcal air kerma measurement and the field area; $H_p(3)$, estimate of H_{lens} using UPC ELDs situated on the left eye of the Rando phantom and combined standard uncertainty ($k = 1$).

suspended screen and that of the goggles were evaluated by doing measurements both when they were used and when they were not. In spite of long irradiations, when protection was properly used, measurements were close to the detection limit of the dosemeters, so in the second stage of the study, only the table shielding was used.

After the initial measurements in April with dosemeters in both eyes, it was decided to use only the ELD on the left eye and the two ELDs close to the left ear. Rows 1–3 in Table 1 summarise the different types of measurements undertaken.

Statistical analysis was performed using the SPSS software[18].

RESULTS

Table 1 shows the estimated eye lens equivalent dose and the associated standard uncertainty ($k = 1$) for each set of measurements. The ratio between $H_p(3)$ and kerma–area product (if the radiation beam size is the same) is higher in low-dose fluoroscopy mode than that for image acquisition mode; this is mainly due to the higher Cu filtration used for the low fluoroscopy mode. Likewise, $H_p(3)$ is higher for the LAO projection than that for the PA projection because the operator is closer to the X-ray tube.

From Experiments 1 and 2, it was found that the eye dose in the right eye and that in the forehand were of the order of 60 and 80 % of the dose in the left eye, respectively. When a ceiling suspended screen was positioned between the X-ray tube and the Rando phantom (Experiment 3), $H_p(10)$ on the thorax and the shoulder was <2 μSv and $H_p(3)$ between 1 and 3 μSv, values close to the detection limit of the dosemeters. When goggles were used, but no ceiling suspended screen was employed (Experiment 4), for the same kerma–area product, $H_p(10)$ on the thorax and the shoulder was of the order of 90 μSv and $H_p(3)$ on the left eye and on the side of the goggles was between 7 and 8 μSv, whereas on the external left side of the goggles (unprotected), $H_p(3)$ was of the order of 50 μSv. Comparing data in Experiments 3 and 4 with measurements in Experiment 2, the attenuation factor for goggles is of ~8 and that for the ceiling suspended screen 21.

It was verified that there was no significant difference ($p < 0.05$) in the WB TLD and DoseAware performance for fluoroscopy and image acquisition mode measurements. Likewise, there was no statistically significant difference between the two tested ELDs. Table 2 shows the mean value, standard deviation and range of the ratio between $H_p(10)$ measured with DoseAware and that with WB TLD considering the different positions and experiments and the ratio between the UPC ELD and the Eye-D. Experiments 3 and 4 were not included in the calculations because readings were close to the dosemeter detection limit.

Table 3 summarises the influence of dosemeter position in estimating H_{lens}. It reports the ratio of the different dose measurements in the selected body positions with $H_p(3)$ measured on the left eye. For this evaluation, only TLD data were used although similar conclusions could be derived from DoseAware measurements. The average reading of the two ELDs are used as $H_p(3)$ on the left ear, and the average of $H_p(10)$ and $H_p(0.07)$ is used for WB measurements. A paired sample t-test was used to confirm that for photon beams, such as those found in interventional cardiology, numerically $H_p(10)$ is close to $H_p(0.07)$ and both can be used to assess $H_p(3)$. The two-tailed test p-value is 0.676, and thus the mean value of the

Table 2. Comparison of $H_p(10)$ measurements with a passive TLD whole-body dosemeter and with the electronic DoseAware detector and $H_p(10)$ with the UPC holder or the Eye-D holder for the eye lens dosemeter.

	Average ± SD	Range	Sample (N)
$H_p(10)_{\text{Dose Aware}}/H_p(10)_{\text{TLD-WB}}$	1.02 ± 0.21	1.35–0.65	24
$H_p(3)_{\text{UPC ELD}}/H_p(3)_{\text{EYE-D}}$	1.07 ± 0.13	1.28–0.94	8

Table 3. Comparison of $H_p(d)$ measured at different positions and with different detectors with the best estimate of H_{lens}.

Projection	Position	$H_p(d)_{\text{position}}/H_p(3)_{\text{left eye}}$		
		Average ± SD	Range	Sample (N)
PA (Experiments 1, 2, 5, 6, 7 and 8)	Chest left side	1.3 ± 0.3	1.82–0.94	6
	Left shoulder	1.9 ± 0.3	2.45–1.61	6
	Centre of collar	0.9 ± 0.2	1.26–0.78	6
	Left ear	1.08 ± 0.05	1.14–0.98	6
LAO (Experiments 9 and 10)	Chest left side	1.8 ± 0.3	2.04–1.62	2
	Left shoulder	2.5 ± 0.6	2.91–2.10	2
	Centre of collar	1.0 ± 0.2	1.19–0.86	2
	Left ear	1.1 ± 0.1	1.18–1.04	2

two quantities $H_p(10)$ and $H_p(0.07)$ can be considered equal within a 95 % confidence interval of (−0.29 and 0.44).

DISCUSSION

The results in Table 2 and Table 3 show that there are several possible designs for the assessment of $H_p(3)$ in practice. It also confirms that the DoseAware electronic device responds satisfactorily in realistic fields in interventional cardiology scattered field[17, 19]. Although previous works reported some limitations of these devices in pulsed radiation fields[12], in the energy and dose range characteristics of the monitored scattered fields, they are proved to be useful tools for a real-time monitoring programme.

One of the most common procedures used at present to estimate the eye lens dose is the use of a second whole-body dosemeter on the apron situated on the chest or the collar. This study provides insight on the limitations and usefulness of this procedure. When the dosemeter and the eyes are similarly exposed to the radiation beam, this second WB dosemeter is a good estimator of the eye lens dose. However, if this is not the case, and the eyes are not protected in the same way as the chest dosemeter, this procedure can lead to large errors.

CONCLUSIONS

The results of this study show that to assess the equivalent dose to the lens of the eye in interventional cardiology, a passive whole-body dosemeter calibrated in $H_p(10)$ or $H_p(0.07)$ can be used satisfactorily. This finding is in agreement with previous guidance given by IAEA[7] and ISO[8]. It also confirms that the electronic DoseAware data are comparable with TLD WBD measurements with the advantage of providing real-time information, which is automatically stored in a centralised local database. Thus, it can be concluded that although there is a lack of experience in measuring $H_p(3)$, most of the available personal dosemeters can provide an appropriate estimate of the equivalent dose to the lens of the eye.

However, it is highlighted that these measurements have an intrinsic large variability due to many parameters that influence the readings. One of the most critical parameters when monitoring the lens of the eye is the position of the dosemeter and the use of the ceiling suspended screen. It has been shown that, when neither the ceiling suspended screen nor the lead goggles are used, a dosemeter placed on the left chest, left shoulder or left ear can be used to estimate the eye lens equivalent dose. However, the centre of the thyroid collar and the left ear positions are recommended because they provide a better estimate.

Dose results when using appropriate protection highlight the fact that in these cases the eye lens dose is drastically reduced and the relationship between measurements in different parts of the body changes substantially. There might be large differences between the eye dose and the monitored dose if the chest dosemeter and the eyes are not equally protected. In such situations, the eye dose could be underestimated or overestimated. In practice, it is difficult to know how and when the protection tools are really used. Thus, if the workload and monthly kerma–area product suggest routine monitoring of the eye lens, a dosemeter close to the eye would be the preferred solution. However, because of the difficulties to implement this type of monitoring in the medical practice, further work to investigate additional appropriate indicators is recommended.

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