

Article

Absorbed Organ Doses and Effective Doses in Intraoral Radiography of Left Upper Molar of Adult Patient: A Computational Study

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Highlights

What are the main findings:

- Effective doses in optimized intraoral procedures simulated using appropriate equipment are below 10 μSv .
- A substantial part of the simulated dose variability was associated with modifiable technical and operator-dependent factors.

What are the implications of the main findings?

- Special attention should be paid to structured training and periodic education programs for operators.
- Using poorly optimized imaging protocols, inappropriate equipment or inappropriate X-ray tube positioning can substantially increase patient dose.

Abstract

Objectives. To investigate effective doses and relevant absorbed organ doses for typical intraoral radiographic imaging procedures in adult patients. Variations of relevant absorbed organ doses and effective doses due to changes in imaging parameters and X-ray tube positioning were investigated. Conversion factors for estimating effective doses from incident air kerma ($K_{a,i}$) and kerma area product are proposed. **Materials and Methods.** Different imaging parameters and geometries obtained from previously published data from Serbia were simulated using Monte Carlo simulation and a computational anthropomorphic phantom. The relevant absorbed organ doses and effective doses for different exposure parameters and geometries were then compared to $K_{a,i}$ and analyzed. Since an obsolete intraoral unit with a focus-to-skin distance of 10 cm, a 50 kVp beam and circular collimator with 5 cm diameter was found in the previous investigation, these imaging parameters were also included in the simulations. **Results.** For typical $K_{a,i}$ for imaging protocols found in Serbia, effective doses for imaging of the left upper molar in an adult person were 2.9 μSv and 5.4 μSv for 60 kVp and 70 kVp beams, respectively. The highest absorbed organ doses were observed for the thyroid, eye lenses, salivary glands and brain, reaching up to 47.2 μGy , 71.4 μGy , 63.1 μGy and 26.2 μGy , respectively. For poorly optimized imaging protocols, these values were higher. Furthermore, for the obsolete intraoral units used, all doses were considerably higher and the typical effective dose was 16.8 μSv with thyroid dose of 105.3 μGy . Using rectangular instead of circular collimators



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decreased effective dose by 25% to 52.7% and absorbed organ doses by 43.3% to 84.5% in all simulated situations. Simulated positioning errors could change eye lens or thyroid absorbed organ dose by up to a factor of two when a rectangular collimator is used and up to four times when a circular collimator is used. Conclusions. Though the effective and absorbed organ doses from intraoral imaging are very low, this imaging can be further optimized by using optimal exposure parameters, rectangular collimators, and proper positioning of the X-ray tube.

Keywords: intraoral radiography; effective dose; absorbed organ dose; MC simulation

1. Introduction

Intraoral imaging techniques are used in dentistry to capture detailed images of the patient's mouth interior to aid diagnosis, treatment planning, and monitoring of oral health [1]. Common intraoral views include bitewing, periapical, and occlusal radiographs, which are essential for detecting caries, periodontal disease, and periapical pathology [2]. Dental radiography is among the most frequent medical imaging procedures in Europe with an estimated 50–150 images per 1000 population annually in EU countries. However, the low effective doses associated with these procedures mean that their contribution to the collective effective dose is very low, typically only 2–4% of the total collective effective dose from plain radiography [3].

The effective dose to the patient from intraoral radiography ranges from 1 to 8 μSv per examination [4–10], but the large number of examinations can lead to cumulative radiation exposure that may still be of concern for patient safety. The absorbed organ doses measured or calculated are higher in organs and tissues located in or adjacent to the irradiated area, so doses to the thyroid, salivary glands, eye lenses and brain are usually reported [7–10]. Most studies have been performed using measurements in phantoms [7,9–11], and Monte Carlo (MC) simulations [8,12]. MC simulations are a powerful tool for modeling photon transport and interactions within human tissue, allowing prediction of the radiation doses received by patients [13,14]. In these studies, absorbed organ doses as well as effective doses were calculated. Additionally, conversion coefficients from kerma air product (KAP) to effective dose were derived using MC simulations. These conversion coefficients ranged from 0.008 to 0.132 $\mu\text{Sv mGy}^{-1} \text{ cm}^{-2}$ for different imaging parameters [8]. By simulating various imaging techniques and parameters, MC simulations help to minimize radiation exposure and improve image quality [14,15].

As previously published study on typical patient doses and imaging parameters in Serbia showed large variations [16]; we simulated these imaging parameters and added possible positioning errors as a source of variation in relevant absorbed organ and effective doses.

The aim of this paper is to investigate typical intraoral radiographic procedures of the left upper molar in an adult patient using MC simulations. We emphasize the importance of using optimized equipment and imaging techniques, rectangular collimators, and proper positioning of the intraoral device. Additionally, conversion factors for estimating effective doses from incident air kerma ($K_{a,i}$), and KAP are proposed for selected clinical scenarios in adult patients.

2. Materials and Methods

Technical parameters of intraoral imaging units, as well as typical exposure parameters, were taken from our previously published study on the dose reference levels (DRLs) for

intraoral radiography in Serbia [16]. DRLs are defined as a level used in medical imaging to indicate whether, under routine conditions, the dose to the patient for a specified radiological procedure is unusually high or unusually low for that procedure. In that study, we determined typical imaging practice and equipment. Most of the units included used a nominal tube potential of 60 kVp or 70 kVp with total filtration equivalent to 1.5 or 2 mmAl, a focus-to-skin distance (FSD) of 20 cm, and circular collimators with a diameter of 6 cm. Although some intraoral radiography devices in that previous study were obsolete, and some techniques were not well-optimized, all imaging setups were simulated. The most important difference in equipment in this study was the use of obsolete devices produced in Serbia (Dent Ei Niš). These units use a fixed 50 kVp (60 kVp to 70 kVp is recommended for dental intraoral X-ray equipment) with a short FSD of 10 cm, whereas a minimum FSD of 20 cm has been recommended [17,18]. Due to their outdated architecture (more than 30 years old), they are expected to be replaced soon, but we modeled these units as well. DRLs in the previously published study were given for different image detectors [16]. Since dose distribution within the patient depends not on the detector type but on the imaging parameters used, we used typical dose indices for the most frequently used imaging parameters and modeled these setups, which were grouped without regard to the imaging detector used. Nevertheless, it is worth noting that detector choice influences exposure settings in clinical practice. The dose index measured in the previously published study was $K_{a,i}$ and KAP values were then obtained by multiplying $K_{a,i}$ by the area of the tube opening. All the units included in this study were equipped with circular collimators with diameters of 5 cm (outdated units Dent EI Niš) or 6 cm (all other units) [16]. Since the area of the collimators is known, the KAP value is easily calculated by multiplying $K_{a,i}$ by the collimator opening, and it is therefore not shown in this work. Typical technical factors, exposure factors, and imaging protocols from the previous study [16], with typical dose indices, are given in Table 1.

Table 1. Typical technical factors, exposure factors, and imaging protocols from a previous study [16] and simulated in this study. Simulated imaging protocol 1 was the obsolete intraoral radiography unit that was still in use (Dent EI Niš).

Imaging Protocol	Voltage (kVp)	Filtration (mmAl)	FSD (cm)	$K_{a,i}$ Median (Min–Max) (mGy)	Circular Collimator Diameter (cm)
1	50	2	10	5.6 (3–10.4)	5
2	60	2	20	1.3 (0.6–3.7)	6
3	70	2	20	1.8 (0.6–5.9)	6

Typical imaging protocols from the previous investigation are shown in Table 1, but we simulated all identified protocols, to gain better insight into variations in effective dose and relevant absorbed organ dose resulting from these changes. Although rectangular collimators are preferred for radiation protection because they allow the radiation beam to match the size of the dental X-ray detector [17,18], none of the units included in the study used them [16]. Circular collimation has been discouraged for many years and it is considered unacceptable, but this practice can still be found. Therefore, to further discourage the use of circular collimation, both collimation types were modeled for all protocols used. The field was simulated as circular with a diameter of 6 cm. Another set of simulations employed a collimated rectangular beam with dimensions 4×4.24 cm². The length of 4.24 cm was chosen because this is the maximum edge that can be inscribed in a circle of 6 cm diameter and is comparable with usual rectangular collimator dimensions in use. Additionally, we included beam angulations of 10 degrees in left-right and up-

down directions to simulate relevant absorbed organ doses and effective dose variations due to X-ray positioning variations. These simulations were used to approximate the actual positioning of the device during imaging in clinics (according to the estimate of an experienced dentist using the intraoral unit). When the direction 'down' is used, it means that the X-ray beam is directed downwards (towards the thyroid). The same labeling is used for all X-ray beam directions. Simulations were set to estimate effective dose and absorbed organ doses during imaging of the left upper molar. All simulated energies, field sizes, distances, and angles are provided in Table 2.

Table 2. Simulated energies, field sizes, distances and angles for intraoral radiography of the left upper molar. The circular collimator with diameter 6 cm is marked with 'C', and 'R' denotes the rectangular collimator dimensions $4 \times 4.24 \text{ cm}^2$. X-ray tube positions: 0—perpendicular to the left upper molar, 10U/D—10 degrees from the perpendicular direction to the left upper molar in the U (up) or D (down) direction, 10L/R—10 degrees from the perpendicular direction to the left upper molar in the right (R) or left (L) direction.

Imaging Protocol	Voltage (kVp)	FSD (cm)	Collimator (C/R)	X-Ray Tube Position (0, 10U, 10D, 10R, 10L)
4	60	20	R	0
5	70	20	R	0
6	60	20	R	10D
7	60	20	C	10D
8	60	20	R	10U
9	60	20	C	10U
10	60	20	R	10L
11	60	20	C	10L
12	60	20	R	10R
13	60	20	C	10R
14	70	20	R	10D
15	70	20	C	10D
16	70	20	R	10U
17	70	20	C	10U
18	70	20	R	10L
19	70	20	C	10L
20	70	20	R	10R
21	70	20	C	10R

MC simulations were performed using the MCNP6.3 program suite [19]. Cross-section data for photon and electron transport were sourced from the ENDF/B-VII.0 Evaluated Nuclear Data Library [20]. The simulations were run in p e mode (coupled photon–electron transport). Energy cut-off parameters were set to 1 keV for both photons and electrons. To represent human head-and-neck anatomy, the Zubal voxel phantom geometry was incorporated [21,22] as described in our previous publications [23]. The phantom consists of voxels measuring $2.2 \times 2.2 \times 1.4 \text{ mm}^3$ and natively identifies 29 distinct anatomical structures. For this study, the submandibular and sublingual glands, as well as surrounding tissues, were added manually to the predefined phantom. Organ masses used for absorbed dose calculations were determined by multiplying the assigned organ volumes by ICRU Report 46 material densities [24], yielding organ masses of (25.2 g) for the submandibular glands and (18.9 g) for sublingual glands. To ensure proper positioning of the beam within the Zubal phantom, an FMESH detector was used, and its results were superimposed on the voxelized phantom image (Figure 1).

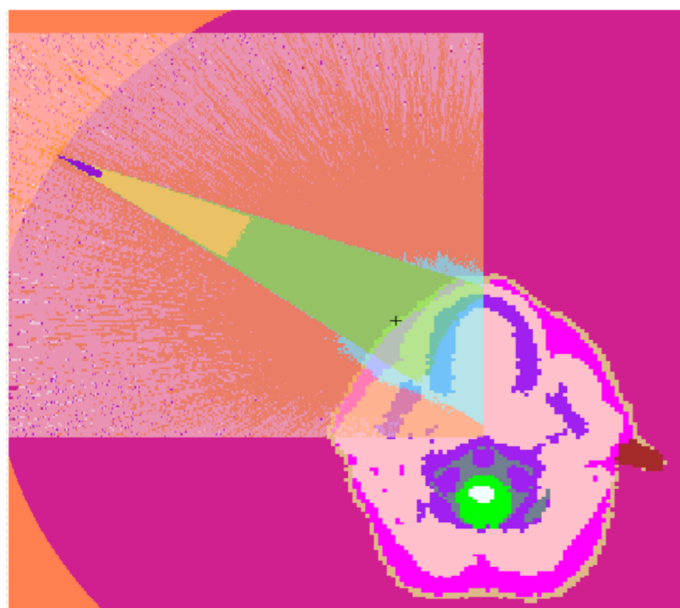


Figure 1. Positioning of the beam on left upper molar, superimposed figures of phantom setup and FMESH tally.

Diagnostic X-ray spectra typical of intraoral radiographic devices operating at tube potentials of 50, 60, and 70 kVp were generated using SpekCalc software [25,26]. The spectrum generation parameters were defined as follows: tungsten (W) anode material, anode angle of 30 degrees, and inherent filtration of 2 mm Al equivalent. The calculated Half-Value Layers (HVLs) were adjusted to fit the desired radiation quality series (RQR) 3, 4, and 5 for 50, 60 and 70 kVp, respectively [27]. To model the physical collimation of the intraoral X-ray tube, a circular beam point source emitting photons into a conical distribution was used. The cone cosine limits were set to produce a beam diameter of 6 cm at the phantom's skin surface, 20 cm from the source point. For the rectangular beam a geometric masking technique was used, namely, a bounding box of zero importance (IMP: $p = 0$) was defined around the source, out of which a 5-vertex wedge-shaped void was carved. One vertex was pinned at the focal spot, while the remaining four vertices diverged outward to project a rectangular field of $4 \times 4.24 \text{ cm}^2$ on the phantom surface.

A total of 10^8 particles were simulated per scenario to ensure high statistical precision. Results were accepted only when all 10 standard MCNP statistical checks were passed and relative errors (R) fell below 0.02 for the primary tallies.

To calculate the absorbed dose for specific organs, ICRU 46 fluence to absorbed dose factors (DE and DF cards) were used [24]. Nuclear data from the ENDF/B-VII.0 Evaluated Nuclear Data Library were used in the simulations [20].

Normalization

All the doses calculated via MC simulations were normalized to the $K_{a,i}$ measured free-in-air at the position corresponding to the phantom skin surface, since this quantity was used as a dose index measured for typical imaging parameters [16]. The F6 detector was set on the skin in front of the desired tooth, where dose was obtained in mGy/(source particle). To avoid backscatter radiation, $K_{a,i}$ was determined using simulations without the phantom in the beam. The phantom was inserted then, and all recorded absorbed organ doses required to estimate the effective dose mGy/(source particle) were normalized to the $K_{a,i}$. The normalized organ dose coefficients were obtained by dividing the phantom absorbed doses by the free-in-air incident air kerma. Although the absorbed organ doses were calculated for all 29 anatomical structures identified within the phantom, as shown

previously [23], only the highest absorbed organ doses near the primary beam path are presented here (thyroid, eye lens, brain and salivary gland absorbed organ doses) due to their primary contribution to effective dose in dental projections. Nevertheless, the effective dose was calculated according to the recommendations of the International Commission on Radiological Protection [28] by summing the weighted organ dose coefficients across all irradiated tissues. It is worth noting that eye lens absorbed organ doses are several orders of magnitude less than doses that could induce deterministic effects and are not part of effective dose calculations. The eye lens was chosen as a critical spatial indicator to evaluate scatter variations and monitor deterministic safety margins across different beams. Since the MC simulation resulted in normalized absorbed organ and effective dose coefficients, the final effective and absorbed organ doses for all simulated imaging parameters were calculated by multiplying these coefficients by the $K_{a,i}$ values found in our previous study and given in Table 1 [16].

3. Results

In the first set of simulations, the procedure for imaging the left upper molar was simulated varying field size, X-ray tube voltage and distance as described in Table 1.

For typical $K_{a,i}$ values for imaging protocols used in Serbia (imaging protocols 2 and 3) [16], effective and absorbed organ doses for imaging the left upper molar of an adult were calculated by multiplying the typical $K_{a,i}$ (reported in Table 1) by the effective or absorbed organ dose normalized to 1 mGy of $K_{a,i}$ (Table 3). Typical effective doses were 2.9 μ Sv for 60 kVp beam and 5.4 μ Sv for 70 kVp used. The highest absorbed organ doses were found for the thyroid, eye lenses, salivary glands and brain and were typically 6.6 μ Gy, 10.9 μ Gy, 3.9 μ Gy and 9.8 μ Gy, respectively, for the 60 kVp beam, and 14.4 μ Gy, 21.8 μ Gy, 7.9 μ Gy and 19.3 μ Gy, respectively, for the 70 kVp beam. For obsolete intraoral units, all doses were considerably higher, and the typical effective dose was 16.8 μ Sv, with a thyroid absorbed organ dose of 105.3 μ Gy. The values calculated above are for typically used imaging parameters (Table 1), but if effective and absorbed organ doses are calculated using the minimum or maximum $K_{a,i}$ values found (Table 1), a large variation can be observed. For example, since in imaging protocol 1 from Table 1, $K_{a,i}$ ranges from 3 mGy to 10.4 mGy, the effective dose for this imaging protocol ranges from 9 μ Sv to 31.2 μ Sv. Similarly, the absorbed organ dose to the thyroid during this imaging protocol would range from 56.4 μ Gy to 195.5 μ Gy, etc.

Table 3. Effective doses and absorbed doses to the thyroid, eye lens, brain and salivary gland doses per 1 mGy of $K_{a,i}$ for typical imaging protocols found in previously published study [16]. Values are rounded to one decimal place for clarity.

Imaging Protocol	Effective Dose (μ Sv/mGy)	Thyroid (μ Gy/mGy)	Eye Lens (μ Gy/mGy)	Brain (μ Gy/mGy)	Salivary Glands (μ Gy/mGy)
1	3.0	18.8	6.1	4.1	8.4
2	2.2	5.1	8.4	3	7.5
3	3.0	8	12.1	4.4	10.7

Effective and absorbed organ doses per 1 mGy of $K_{a,i}$ for all simulated imaging protocols are presented in Table 4.

All results presented in Table 4 are normalized to $K_{a,i}$, so these values can be used as conversion coefficients from measured $K_{a,i}$ to effective dose or absorbed organ doses. Additionally, since KAP can be easily determined from $K_{a,i}$ and tube opening area, these values can also be used as conversion coefficients from KAP to effective or absorbed

organ doses. In fact, the values shown in Table 4 should be divided by the field area (16.96 cm² for rectangular collimator and 28.27 cm² for circular collimator with 6 cm diameter). Nevertheless, here we calculated only the effective dose to KAP conversion coefficients, which ranged from 0.071 $\mu\text{Sv mGy}^{-1} \text{cm}^{-2}$ to 0.134 $\mu\text{Sv mGy}^{-1} \text{cm}^{-2}$ for different imaging parameters. Obsolete intraoral radiography units are not included here.

Table 4. Effective and absorbed organ doses, per 1 mGy of $K_{a,i}$ for all simulated imaging parameters, rectangular and circular collimators, and beam angulation when imaging left upper molar as presented in Table 2. Values are rounded to one decimal place for clarity.

Imaging Protocol	Effective Dose ($\mu\text{Sv/mGy}$)	Thyroid ($\mu\text{Gy/mGy}$)	Eye Lens ($\mu\text{Gy/mGy}$)	Brain ($\mu\text{Gy/mGy}$)	Salivary Glands ($\mu\text{Gy/mGy}$)
4	1.5	2.0	3.8	1.4	4.2
5	1.6	2.9	4.6	1.8	5.3
6	1.4	4.1	2.7	1	3
7	2.9	21.5	5.5	2.1	5.8
8	1.4	1.3	5.3	2.7	5.5
9	2.2	4	15.8	7.9	12.2
10	1.7	2.5	3.4	1.7	4.9
11	2.7	16.1	6.2	4.3	9.8
12	1.5	1.6	4.3	1.2	3.5
13	2	3.3	9.8	2.4	6.3
14	1.8	6.7	4	1.6	4.4
15	3.8	30	8.1	3.2	8.4
16	1.8	2.3	7.7	4.2	7.9
17	2.9	6.5	22	11.2	17.2
18	2.2	4.3	5.1	2.6	7.1
19	3.6	22.5	9.1	6.4	13.9
20	1.9	2.9	6.3	1.9	5.1
21	2.6	5.4	13.8	3.6	9

Comparison of effective and absorbed organ doses for 60 kVp and 70 kVp X-ray beams showed that, for the same $K_{a,i}$, the effective dose is approximately 30% higher and all relevant absorbed organ doses are approximately 50% higher when a 70 kVp beam is used.

Comparison of effective and absorbed organ doses for all simulated imaging protocols shows that reduction associated with rectangular collimation in effective doses ranges from 25% to 52.7% and in absorbed organ doses from 43.3% to 84.5%.

Figure 2 presents a comparison of effective and relevant absorbed organ doses normalized to $K_{a,i}$ for imaging protocols where the angulation of the X-ray tube is changed from the perpendicular direction to the left upper molar to 10 degrees in left/right and up/down directions for 60 kVp beams. All data are normalized to the perpendicular direction (0) to emphasize change due to direction change. Imaging protocols with rectangular (4, 6, 8, 10 and 12) and circular collimators (2, 7, 9, 11 and 13) are compared.

Figure 3 presents a comparison of effective and relevant absorbed organ doses normalized to $K_{a,i}$ for imaging protocols where the angulation of the X-ray tube is changed from the perpendicular direction to the left upper molar by 10 degrees in left/right and up/down directions for 70 kVp beams. All data are normalized to the perpendicular direction (0) to emphasize changes due to direction change. Imaging protocols with rectangular (5, 14, 16, 18 and 20) and circular collimators (3, 15, 17, 19 and 21) are compared.

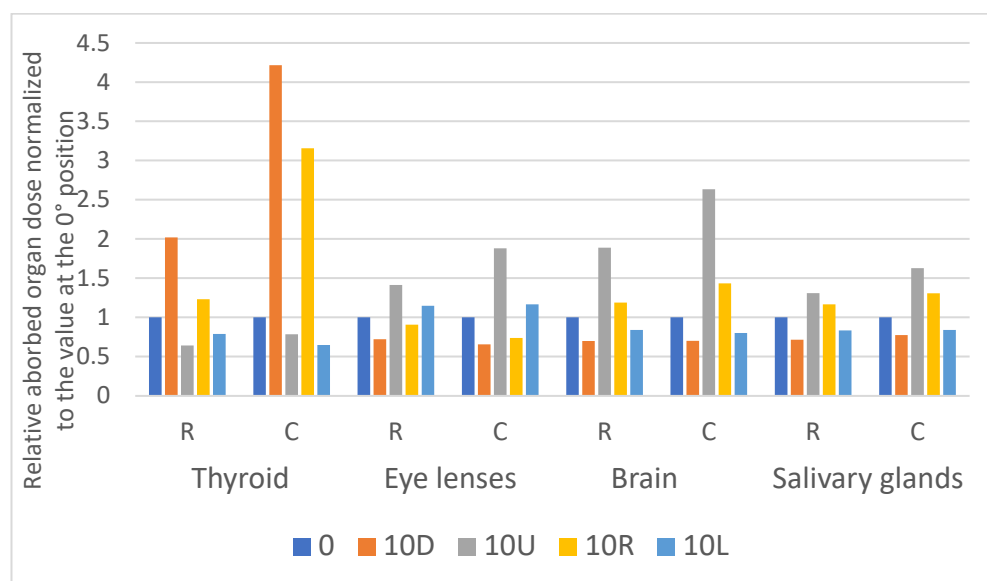


Figure 2. Comparison of relevant absorbed organ doses normalized to $K_{a,i}$ for imaging protocols where the angulation of the X-ray tube is changed from the perpendicular direction (0) to the left upper molar by 10 degrees in left/right (L/R) and up/down (U/D) directions for 60 kVp beams. All data are normalized to the perpendicular direction (0). Imaging protocols with rectangular (4, 6, 8, 10 and 12) and circular collimators (2, 7, 9, 11 and 13) are compared. R—beams using rectangular collimators and C—circular collimators. All errors are below 2%. Error bars are omitted from the figure for clarity.

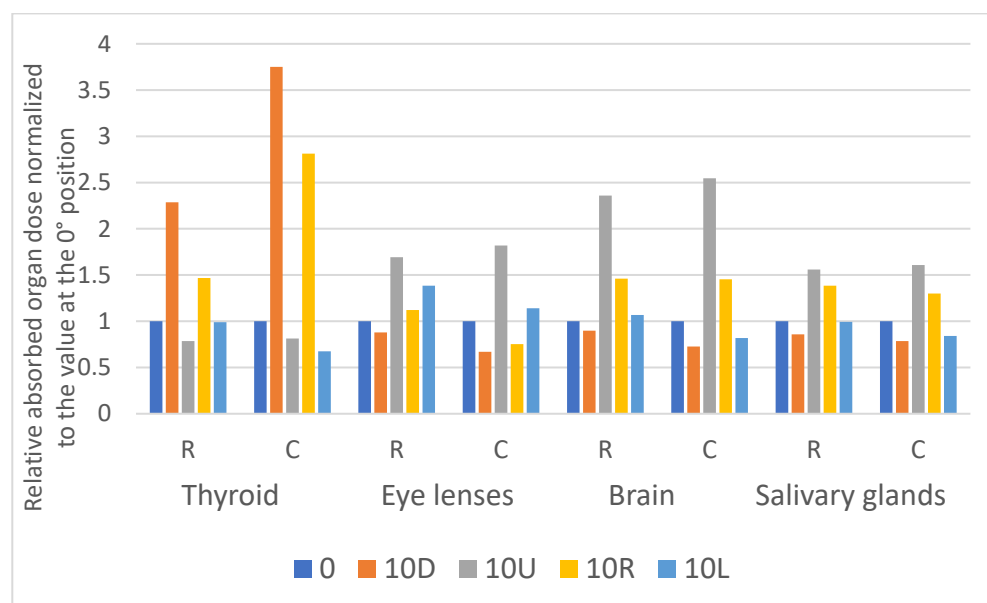


Figure 3. Comparison of relevant absorbed organ doses normalized to $K_{a,i}$ for imaging protocols where the angulation of the X-ray tube is changed from the perpendicular direction (0) to the left upper molar by 10 degrees in left/right (L/R) and up/down (U/D) directions for 70 kVp beams. All data are normalized to the perpendicular direction (0). Imaging protocols with rectangular (5, 14, 16, 18 and 20) and circular collimators (3, 15, 17, 19 and 21) are compared. R—beams using rectangular collimators and C—circular collimators. All errors are below 2%. Error bars are omitted from the figure for clarity.

4. Discussion

This work investigated changes in effective and relevant absorbed organ doses due to variations in intraoral radiography imaging parameters and technical factors found in

a previously published investigation [16] (Tables 1 and 3). Additionally, the influence of possible X-ray tube positioning variations on relevant absorbed organ doses and effective doses was also investigated. Effective and absorbed organ doses, normalized to $K_{a,i}$ for all simulated imaging parameters, rectangular and circular collimators, and angulation of the beam when imaging the left upper molar are presented in Table 4.

Results obtained in the previous investigation [16], show a large range of imaging parameters used, partly due to the use of different imaging detectors (film and digital detectors). Because of the low level of practice optimization in that study, $K_{a,i}$ was higher for 70 kVp than for 60 kVp (Table 1). This is not in line with the fact that the difference in image quality between 60 kVp and 70 kVp intraoral radiography beams is negligible when using modern detectors [29]. Beams at 60 kVp generally provide higher image contrast, but 70 kVp beams reduce patient dose because shorter exposure times can be used [16,29]. Comparison of patient doses for these beams showed that using the more penetrating, 70 kVp beam increases effective dose approximately 30% and relevant organ doses by up to 50% for the same $K_{a,i}$ used. For this reason, effective and relevant absorbed organ doses were found to be higher for 70 kVp beams than for 60 kVp (Tables 1 and 3). Nevertheless, these values are within the range of previously presented values [7–10]. Additionally, for obsolete intraoral units still in use in Serbia, all doses were considerably higher [18]. Higher patient doses for these units are partly caused by the shorter FSD of 10 cm and larger beam divergence. This confirms the need for use of FSDs longer than 20 cm [17,18].

A rectangular collimator in intraoral radiography is preferred for radiation protection because it allows the radiation beam to match the size of the dental X-ray detector. Even though the use of rectangular collimators in intraoral radiography units is recommended [17,18], and non-rectangular collimators on intraoral equipment, for which special justification has not been made, are stated as unacceptable in European Union guidelines [18], circular collimators are almost exclusively used in Serbia. It was reported that a rectangular collimator can achieve dose reductions exceeding 60% without reducing image quality in intraoral radiography [30,31]. We confirmed large dose reduction when rectangular collimation is used. These results highlight the need for increasing awareness and utilization of this dose-reducing technique in clinical practice.

Since the values in Table 4 are effective doses and absorbed organ doses normalized to $K_{a,i}$, they can be used as conversion factors from $K_{a,i}$ to effective dose or to absorbed organ doses. Furthermore, since field sizes are known, $K_{a,i}$ can be converted to KAP and these values can also be used as conversion coefficients from KAP to effective or absorbed organ doses. These conversion coefficients from KAP to effective dose were comparable to previously reported values [8].

During intraoral radiography, the beam is pointed perpendicular to the tooth. Nevertheless, it can be shifted in any direction if the X-ray tube is not positioned properly. To investigate the possible variation in doses due to incorrect positioning, we simulated a shift of the X-ray tube by 10 degrees in the up, down, left and right directions. Shifting the X-ray beam does not change effective dose considerably, but it can double the thyroid and eye lens absorbed organ doses when this shift is in the up or down direction for both 60 kVp and 70 kVp beams (Figures 2 and 3). As expected, when the X-ray tube is pointed downwards (10D in Tables 2 and 4), thyroid absorbed dose increases and, conversely, moving the beam upwards (10U in Tables 2 and 4) reduces thyroid absorbed dose, but increases the eye lens absorbed dose. This effect is more pronounced when a circular collimator is used. Thyroid absorbed dose doubles when the X-ray beam is pointed towards the thyroid (down direction with a 10° angle) and rectangular collimator is used, but almost four times when a circular collimator is used (Figures 2 and 3) for both voltages. This supports the use of rectangular collimators for patient dose optimization as already proposed [17,18].

These findings highlight the importance of precise patient positioning during intraoral imaging. Positioning errors not only compromise image quality but also lead to unnecessary radiation exposure, emphasizing the need for rigorous training and adherence to standardized protocols.

The limitations of this study include the lack of assessment of patient anatomical variability, such as differences in body size, age, sex, and craniofacial morphology, all of which may influence absorbed organ dose estimates. Also, the effective dose can be used prospectively as an indicator of radiation detriment and for comparing radiation burdens between different imaging procedures [32]. However, it is calculated for reference persons using age- and sex-averaged tissue weighting factors and therefore provides only an approximate measure of detriment rather than an estimate of individual patient risk [32]. The present study reports effective dose estimates only and does not address individual patient risk. An additional limitation is that direct physical phantom measurements for MC model validation were not performed in the present study. The general Monte Carlo methodology and computational framework used here were previously verified against experimental radiochromic film dosimetry using a cylindrical PMMA phantom, in the context of dental cone beam computed tomography [23]. However, the specific intraoral radiography geometry, beam fields source-to-object distances and exposure configurations used in this study were not independently validated through direct physical dosimetry.

5. Conclusions

Though the absorbed organ doses for the brain and salivary glands were relatively high compared to other organs and tissues, the highest impact on the effective dose was from the thyroid gland due to the higher tissue weighting factor. This was already shown before [9]. Patient effective doses from intraoral radiography found in this study were typically below 10 μSv , but reached up to 31.2 μSv for obsolete units using short FSD (10 cm), circular collimators and 50 kVp X-ray beams. Though these are low doses when compared to most of the other diagnostic radiology procedures, a large number of examinations can provide cumulative radiation exposure that may still be of concern for patient safety. Considering that a substantial part of the simulated dose variability was associated with modifiable technical and operator-dependent factors, we emphasize the importance of structured training and periodic education programs for operators. In particular, training should focus on optimizing imaging protocols, ensuring proper patient and tube positioning, maintaining quality assurance, and establishing programs to replace or upgrade outdated equipment.

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