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Impact of source position and transit time on HDR skin surface applicator dosimetry

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Abstract

Purpose—Skin surface dosimetric discrepancies between measured and treatment planning system (TPS) predicted values were traced to source position sag inside the applicator and to source transit time. We quantified their dosimetric impact and propose corrections for clinical use.

Material and methods—We measured the dose profiles from the Varian Leipzig-style HDR skin applicator, using EBT3 film, photon diode, and optically stimulated luminescence dosimeter for three different GammaMed*plus*TM HDR afterloaders. The measured dose profiles at several depths were compared with BrachyVision AcurosTM calculated profiles. To assess the impact of the source sag, two different applicator orientations were considered. The dose contribution during source transit was assessed by comparing diode measurements using a HDR timer and an electrometer timer.

Results—Depth doses measured using the three dosimeters were in good agreement, but were consistently higher than the Acuros dose calculations. Measurements with the applicator face up were significantly (exceeding 10%) lower than those in the face down position, due to source sag inside the applicator. Based on the inverse square law, the effective source sag was evaluated to be about 0.5 mm from the planned position. The additional dose during source transit was evaluated to be about 2.8% for 30 seconds of treatment with a 40700 U (10 Ci) source.

Conclusion—With a very short source-to-surface distance, the small source sag inside the applicator has a significant dosimetric impact. This effect is unaccounted for in the vendor's treatment planning template, and should be considered before the clinical use of the applicator. Further investigation of other applicators with large source lumen diameter may be warranted.

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Keywords

Skin cancer; Leipzig skin-applicator; HDR brachytherapy; dosimetry; dwell position uncertainty; source transit dose

1. Introduction

Skin cancer is the most common type of malignancy in the United States. According to the American Cancer Society, there are more than 3.5 million skin cancer cases each year [1,2]. This is equivalent to a 1-in-5 overall lifetime risk for all Americans. Most skin cancers are basal cell carcinoma (BCC) and squamous cell carcinoma (SCC), and are relatively easy to treat when diagnosed early [3].

Skin cancer treatment options are determined by considering several factors including type and stage of the tumor, age and condition of the patient, and the cosmetic and functional impact of the treatment [4,5]. The standard of care is surgery; many skin cancers can be easily removed with good oncologic and cosmetic results. When surgery is not suitable, RT has been shown to be a good alternative, especially for lesions in areas where complete resection is difficult [6,7].

Several radiotherapy modalities have been used. These include superficial x-ray with a typical peak energy range of 70–100 kV, electron beam of 6–12 MeV from a linear accelerator, electronic HDR brachytherapy, and radioisotopic HDR brachytherapy with various skin applicators and custom molds [4,8–11].

We report on our experience in commissioning the Varian Leipzig-style (horizontal) applicator for clinical use, specifically, on discrepancies we noticed between measured and treatment planning predicted dosimetry that, to our knowledge, has not been previously reported. Published data and manufacturer instructions refer to “in-phantom” measurements [12,13], wherein treatment time calculations are typically done using a lookup table of depth-dose profiles along the applicator central axis, and assume the applicator is face down, replicating phantom measurements.

For the skin surface applicators, Varian provides applicator templates for use in treatment planning with Acuros™ inhomogeneity corrections [14]. We sought to compare the measured and TPS calculated data. The dosimetric discrepancy was traced to gravitational source sag inside the applicator source lumen. We also found that, for short treatment times used in commissioning measurements (as short as 30 seconds), extra dose, due to source transit, added to the dosimetric discrepancy. Here, we quantify the dosimetric impact of these two factors and propose corrections for clinical use.

2. Material and methods

2.1. Leipzig-style skin applicator

In this work, we used the horizontal skin applicator (model #: GM11010080; Varian Medical Systems, Palo Alto, CA), shown in Figure 1. The applicator is comprised of the Leipzig-

style cone with a permanently mounted horizontal source transit tube and four apertures of 30–45 mm in 5 mm increments for various target sizes. The applicator and collimators are made of a stainless tungsten alloy, which provides directional treatment geometry and shields the non-treatment surface. The nominal source-to-surface distance (SSD) is 12.5 mm and each collimator has a thin polycarbonate window at the distal end to ensure a flat target surface and consistent SSD. The integrated source lumen is positioned parallel to the polycarbonate window. Versions of this applicator are available for both GammaMed*plus*TM and VariSource[®] afterloaders.

2.2. HDR afterloader

Varian GammaMed*plus* iX afterloaders (Varian Medical Systems, Palo Alto, CA) were used with the skin applicator. This afterloader employs an Ir-192 source ($E_{\text{avg}} = 397 \text{ keV}$; $T_{1/2} = 74 \text{ d}$) with 0.9 mm diameter (source model: GMP Ir-192 HDR) [15,16], which is soldered onto the tip of a flexible source cable. The afterloader treatment resolution is 1 mm in source positioning along the treatment lumen, and 0.1 second in dwell time.

2.3. EBT3 film measurement

The dose distribution from the skin applicator was measured with the EBT3 radiochromic film (Ashland Inc., Wayne, NJ), since it can provide a two-dimensional (2D) dose map of the whole field with high spatial resolution. The EBT3 is self-developing film in real time, almost energy independent, and near tissue-equivalent [17–20]. Film calibration, to correlate the film's optical density with the absorbed dose, was performed with a 6MV beam from a linear accelerator in the reference conditions (field size of $10 \times 10 \text{ cm}^2$ at the depth of maximum dose (d_{max}) of 1.4 cm) for 13 different dose levels ranging from 0 to 464 cGy. Considering the energy independent nature of the EBT3 film, energy correction was not carried out.

Films used were from the same lot, and care was taken to wait 24 hours between exposure and measurement. The films were scanned one at a time, keeping the same orientation, using EPSON[®] expression[®] 11000XL scanner (EPSON America Inc., Long Beach, CA). Scanner warm-up procedure was followed to ensure electronic stability of the scanner and $\pm 1\%$ scanning reproducibility. The scan resolution was set to 72 dpi for all films, which provides a pixel size of about $0.35 \times 0.35 \text{ mm}^2$. For the calibration and further dose analysis, FilmQATM Pro software (Ashland Inc., Wayne, New Jersey) was used. The red channel was chosen for both calibration and measurements for its wider dynamic range. The calibration curve was generated in the software as a reciprocal function between red channel color transmission and dose. Dose maps were measured at the surface, 2, 3, 5 and 10 mm depth. Figure 2(a) shows the dose measurement setup at the surface of the applicator.

2.4. Additional measurement with photon diode and optically stimulated luminescence dosimeter (OSLD)

To confirm the EBT3 film results, additional dose measurements were performed with a photon diode detector (IBA PFD^{3G}, IBA Dosimetry America, Bartlett, TN) and OSLDs (nanoDotTM, Landauer Inc., Glenwood, IL). The active volumes of photon diode detector and OSLDs are very small (0.2 mm^3 and 4 mm^3 , respectively) and are thin enough (0.06

mm and 0.2 mm, respectively) to measure the depth-dose along the central axis of the applicator with the desired resolution [21,22].

Each dosimeter was independently calibrated. The diode detector was mounted on a polystyrene phantom with a hole that fits the HDR source inside. The distance between the center of the source and the active volume of the detector was set to 2.37 cm (2.3 cm of the phantom and 0.07 cm of active depth of the detector). The measured currents were correlated with the calculated doses based on TG-43 [23]. OSLDs were irradiated for six different known doses (25, 50, 100, 150, 200, and 250 cGy; two measurements for each dose level) of a 6MV beam in the reference condition and the calibration curve was acquired. The effective depth of the OSLD active volume was estimated to be 0.9 mm based on the micro-computed tomography (micro-CT) image, as shown in Figure 2(c) lower-right inset, and was applied in depth-dose measurements.

The doses were measured at the central axis of the applicator for various depths, as shown in Figure 2(b) and 2(c). The depth doses for the photon diode were measured three times for 7 different depths (0.7, 1.2, 1.7, 2.7, 3.7, 5.7, and 10.7 mm) and measurements were performed twice for 4 different depths (0.9, 3.9, 5.9, and 10.9 mm) for OSLD. For both dosimeters, the measurements were performed for 30 seconds dwell times.

2.5. Comparison with Acuros dose calculation

Different from the TG-43 formalism, the Acuros dose calculation algorithm accounts for the heterogeneity effect and therefore, it can provide more accurate dose distribution over the irradiation volume [24,25]. It is based on the Linear Boltzmann transport equation (LBTE) and a rapid alternative to Monte Carlo simulations. The Varian treatment planning system, BrachyVision, includes an applicator library in which the solid applicator model for the Leipzig skin applicator is defined. This enables Acuros calculation for the applicator. For the calculation of dose and comparison with measurement, the same geometry as measurement was defined using a CT image of a solid water phantom, with a calculation grid size of 0.2 cm. For each aperture, the 2D dose map was acquired at the depths of measurement. The calculated 2D dose maps of Acuros were compared with EBT3 film measurements.

2.6. Source position effect

During the measurement, a discrepancy was observed between the measured dose and dose calculation with Acuros. It was suspected that the discrepancy might result from source sag inside the applicator lumen. As shown in Figure 3, the inner diameter of the source tube is 2.4 mm, while the source diameter is only 0.9 mm. Because the source will sag due to gravity, the SSD is decreased, resulting in a higher dose.

The dose increase due to the SSD change can be estimated based on the inverse square law as shown in the following equation:

$$D_{\text{sag}}(d, \text{sag}) = D(d) \times \left(\frac{\text{SSD} + d}{\text{SSD} + d - \text{sag}} \right)^2 \quad (1)$$

where, $D_{sag}(d,sag)$ is the depth-dose with source sag as a function of depth (d) and extent of source sag (sag) and $D(d)$ is the dose without source sag. For the maximum source sag of 0.75 mm (Figure 3), the dose is increased by 13% at the surface.

To test the hypothesis, the dose measurements were repeated with the applicator in the opposite orientation (upside-down), while all the other conditions remained the same. The dose difference between the two applicator orientations was analyzed and the effective source sag was estimated as described in the later section.

2.7. Source transit time effect

The added dose delivered during the source transit is ignored in the TPS. It might be negligible for most clinical situations where the dwell time is usually several minutes. For a short measurement, however, the extra dose may be significant [26]. To evaluate the extent of the extra dose during the source transit, we compared the doses measured with and without source transit using the diode detector. First, the total dose was measured including the contribution during the source transit. With the electrometer in continuous collection mode, the source was deployed to the dwell position and retracted after 30 seconds of dwell time. Then, the dose contribution from the fully deployed source only was measured excluding the contribution of the transit dose. After the source was completely deployed to the dwell position, 30 second measurements were done using the electrometer timer. After comparing the doses measured in two different ways, the degree of the source transit effect was evaluated. The measurements were performed six times for two different aperture sizes (30 and 45 mm) at several different depths (0.7, 3.7 and 10.7 mm depth).

2.8. Estimation of effective source sag

The effective source sag was found by comparing the measurements of two different applicator orientations (face up or face down). For varying extents of source sag, the measured doses were corrected based on the inverse-square law, and the best agreement (where the difference of the two measurements is minimized) was found. Since the source transit effect is the same for the measurements of the two applicator orientations, there was no need to include the effect in the correction. For comparison with the TPS, however, the correction was performed for both source position and transit effects.

2.9. Estimation of dose leakage

Additional measurements were performed to estimate the dose leakage outside the applicator. Using the OSLD, the dose rate on the outside surface of the applicator cone was measured as shown in Figure 4.

3. Results

The central axis depth-dose profiles were measured with three different dosimeters and compared for normal (face down) applicator orientation. As shown in Figure 5, all the measurements were in good agreement. As expected, OSLD measurements were slightly higher, which is consistent with the observation of Jursinic *et al.*, due to the higher sensitivity of OSLD to Ir-192 compared to 6MV calibration beam [27]. When compared

with the Acuros calculation, however, the measurements were consistently higher. Figure 6 shows the difference between the Acuros calculation and film measurement in 2D isodose lines and a dose difference map for a 40 mm aperture at 2 mm depth. The horizontal dose profiles extracted from the 2D film dose map are shown for all depths, in comparison with the Acuros calculation, in Figure 7. Although the shapes of calculated and measured dose profiles agree, the dose difference is close to 10% at the surface and it decreases with depth (Table 1).

To test the hypothesis that the source position inside the applicator causes the dose discrepancy, the depth-dose measurements were repeated with the applicator upside down. As shown in Figure 8, the measurements with face down direction delivered a higher dose due to $SSD < 12.5$ mm and the measurements in the face up direction resulted in a lower dose due to $SSD > 12.5$ mm, while the Acuros calculation results were in between those measurements with $SSD = 12.5$ mm. At 5 mm depth, about 10% of dose difference was observed between the two measurements and the difference increased for shallower depths.

The additional dose during the source transit was measured to be about $2.8 \pm 0.3\%$ for 30 seconds dwell time with a 40700 U (10 Ci) source. The contribution of transit dose increased with increased source activity, and decreased with increased dwell time. The total dose can be expressed through the following equation, from which the effect can be accounted for.

$$D_{\text{total}}(S_K, t) = D_{\text{dwell}}(S_K, t) \times \left(1 + 0.028 \left(\frac{S_K}{40700} \right) \left(\frac{30}{t} \right) \right) \quad (2)$$

where, $D_{\text{total}}(S_K, t)$ is the total dose as a function of source strength (S_K) and dwell time (t) and $D_{\text{dwell}}(S_K, t)$ is the dose delivered at the dwell position. The transit dose effect can also be expressed in terms of the product of dose rate and effective treatment time by rewriting equation 2 as:

$$D_{\text{total}}(S_K, t) = \dot{D}_{\text{dwell}}(S_K) \times (t + 2.064 \times 10^{-5} \cdot S_K) \quad (3)$$

where, $\dot{D}_{\text{dwell}}(S_K)$ is the dose rate at dwell position as a function of source strength (S_K), and the quantity, in parentheses, is the effective treatment time. Therefore, the transit dose component is equivalent to the additional time of $2.064 \times 10^{-5} \cdot S_K$ seconds. For a 40700 U (10 Ci) source, the additional effective time is 0.84 ± 0.09 seconds.

Comparing the two film measurements with different applicator orientations, the best agreement was found with the effective source sag of 0.47 mm. With the corrections for the effective source sag and transit time, all the film measurements and calculations agreed within 5% difference, as shown in Table 1. In Figure 9, all the measurements from the three detectors are combined together after the corrections for the effective source sag and transit time. The 95% confidence interval (CI) of the fit indicates very small uncertainty of the measurement. At 3 mm depth, the 95% CIs are less than 2.5 % of the mean values for all apertures. When the uncertainty in dose (y-axis) was converted into the uncertainty in depth

(x-axis), the depth uncertainty was estimated to be in the range of 0.10–0.14 mm for all apertures.

The average leakage dose rate on the outside surface of the cone was estimated to be 23.8 ± 2.6 cGy/min, which corresponds to about 8% of the dose rate at 3 mm. For a typical treatment fraction of 7 Gy prescribed at 3 mm depth, the maximum outside dose at the surface of the cone becomes about 53 cGy.

4. Discussion

In this work, three independently calibrated dosimeters (EBT3 film, diode, and OSLD) were used to measure radiation dose. The depth doses along the central axis of the applicator were in good agreement. The active volumes of the diode detector and OSLD were small enough to measure the peak dose of the applicator. The EBT3 film was used to map the 2D dose. The spatial resolution of the EBT3 film is dependent on the scan setting and, for the 72 dpi used in this work, the point dose of every 0.35 mm could be measured, which shows the advantage of the film measurement in small field dosimetry [28].

The observed dose discrepancy between measurement and the Acuros calculation was traced to the uncertainty of the source position within the applicator. The inner diameter of the applicator (2.4 mm) was much wider than the diameter of the GammaMed*plus* HDR source (0.9 mm) and the source sags due to gravity. The effective source sag was estimated to be about 0.47 mm however there might be uncertainty of source sag in the applicator. The extent of source sag may differ between HDR machines due to bends in the source wire and different applicator setup geometries. It may also vary for different setups for the same machine, as previously shown in the literature [29–31].

It should be noted that in this work three identical afterloaders were used interchangeably. Although all three units are the same model and go through the identical source calibration and QA protocols, varying degrees of source sag between the afterloaders might increase the uncertainty of the measurement. This is a realistic representation of our clinical workflow, and simulates the more common scenario of a single afterloader with multiple sources installed over time.

The corrected measurements and Acuros calculation in this work were compared with previous work [12,32]. As shown in Figure 10, the depth doses and dose profiles are in good agreement.

The observed source sag may affect the delivered dose using other applicators as well. Many brachytherapy applicators have an inner diameter of 2 mm or larger, and the source sag inside the other applicator should also be a cause for concern, when the SSD is short. For example, the “Bougie” applicator used in esophagus brachytherapy has an inner diameter of about 3 mm, and the SSD is even smaller than that of the skin applicator with a range of only 4–7 mm. Considering the patient is in a supine position during treatment, the posterior esophageal wall gets a much higher dose than planned, while the anterior wall is underdosed. Similarly, other applicators with large inner source lumen diameters and short SSD should be evaluated for the source sag effect. Further, the effect is expected to be more

pronounced for sources with smaller diameters, such as the VariSource afterloader, which has a source with a diameter of only 0.59 mm.

The dose discrepancy between the two different orientations was more than 10% for 5 mm and increased for shallower depths. Considering the typical fractionation of 7 Gy prescribed at 3 mm depth (total 42 Gy with six twice per week fractions) [33,34], the observed dose variation at 3 mm depth is larger than 10% and would be clinically unacceptable. The skin surface dose is about 1.5 times larger than the dose at 3 mm depth for the applicator, which yields an even larger skin surface dose uncertainty (more than 15%) at this high curative dose. To ensure consistency of RT outcome, it is known that the dose accuracy of 5% or better is required, based on radiobiological considerations [35,36]. In modern radiotherapy, a lot of effort is made to increase dose accuracy and corrections for small factors are routinely applied, as described in the AAPM TG-43 report [23].

To minimize the effect of the source sag and the resulting dose discrepancy, the best solution is minimizing the inner diameter of the applicators. However, this may be impractical for a clinic that has the devices on hand. In that case, we recommend adopting a single applicator orientation protocol, based on careful dose measurement in the orientation used for treatment.

In this work, the extra dose during the source transit was also evaluated and quantified by measuring the dose with and without the source transit, which is currently ignored in TPS dose calculation. Various afterloading systems may attempt to compensate for the transit dose in different ways. The GammaMed*plus* afterloader does so by subtracting the transit time between two consecutive dwell positions from the time in the second dwell position. However, most of the transit dose contribution in this case is not corrected for [37]. HDR source transit dose has long been recognized and explored by many researchers using various methods [26,38–42]. Among them, Fonseca, *et al.* have extensively evaluated the transit dose of the ^{192}Ir HDR afterloader, with respect to the source speed profiles, for a more realistic dose correction [38–40]. For the skin applicator however only one dwell position is used and the absolute transit dose contribution is a function of source activity only. Therefore, the proposed simplified correction based on equations 2 or 3 would suffice. The contribution of the source transit dose will be negligible in most clinical applications with dwell time of a few minutes or longer, although the effect may become significant during short QA and commissioning measurements, or in cases of hypofractionated treatments.

To ensure applicator integrity, the leakage dose at the surface of the applicator shield was also evaluated. Through the measurements using OSLD, the leakage dose was estimated to be about 8% of the prescribed dose. Although this is relatively large amount of leakage dose, it wouldn't be clinically significant considering the patient setup and the shielding of HDR vault.

Finally, while the observed effects of source sag and source transit were demonstrated for a single applicator-afterloader combination, the effects may also affect the dose evaluation for the other vendors' applicators and afterloaders, which demands careful investigation. As

modern treatment planning algorithms and high resolution imaging systems become available, there may be an increased desire to overlay dose distributions of even “simple” plans with the patient’s anatomy. This study shows that caution must be exercised in implementing such protocols, and that vendor supplied applicator templates must be carefully scrutinized prior to clinical use.

5. Conclusions

In this work, the dosimetric effects of source position uncertainty were evaluated. With very short SSD, the source sag inside the applicator has significant dosimetric impact and should be considered before the clinical use of the applicator, especially when relying on the TPS for treatment time calculations. Further investigation for other applicators with large inner diameter may be warranted.

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List of abbreviations

HDR	high-dose rate
TPS	treatment planning system
BCC	basal cell carcinoma
SCC	squamous cell carcinoma
RT	radiation therapy
SSD	source-to-surface distance
2D	two-dimensional
d_{max}	depth of maximum dose

dpi	dots per inch
OSLD	optically stimulated luminescence dosimeter
micro-CT	micro-computed tomography
LBTE	linear Boltzmann transport equation
95% CI	95% confidence interval
AAPM	American association of physicists in medicine
TG	task group
QA	quality assurance

Highlights

- Source sag inside the HDR skin applicator affects the dose profiles
- Source sag and transit time are unaccounted for in current treatment planning
- Need to investigate other applicators for source sag effects

Figure 1.
Varian Leipzig-style HDR skin applicator (horizontal) with four apertures of different sizes.

Figure 2.

Dose measurement for three different dosimeters: (a) EBT3 film at the surface of the applicator with a solid water phantom; (b) photon diode measurement with diode detector under the applicator at the central axis of the beam; (c) OSLD measurement setup with the dedicated OSLD holder (the lower right inset image shows the cross-sectional view of OSLD taken with micro-CT to find the effective depth of the active volume).

Figure 3. Source position inside the applicator. Considering the inner diameter of the applicator (2.4 mm), the HDR source of the GammaMed*plus* (0.9 mm) might sag due to the gravity and the theoretical maximum source sag is 0.75 mm.

Figure 4.
Leakage dose measurement on the outer surface of the applicator cone using OSLD.

Figure 5. Comparison of dose rate at the central axis of the applicator measured with three different dosimeters (EBP3, diode, and OSLD). All measurements are in good agreement within 5% difference. Error bars indicate $\pm 1\sigma$ of the measurements.

Figure 6.

Comparison between Acuros calculated and EBT3 film measured dose maps for 40 mm aperture at 2 mm depth: (a) 2D Isodose lines (solid lines for Acuros and dotted lines for EBT3 film) and (b) dose difference map. The HDR source travels to dwell position from the bottom of the plot (+y-direction).

Figure 7.

The horizontal dose rate profiles at the center of the applicator measured with EBT3 film for four different aperture sizes, which are consistently higher than the Acuros dose calculation.

Figure 8.

The measured dose rates with two different applicator directions (face up and face down) with the percent differences between the two measurements. The Acuros calculation results lie in between the two measurements. Error bars indicate $\pm 1\sigma$ of the measurements.

Figure 9.

The depth dose rates measured from all three detectors after correction for the effective source sag of 0.47mm and transit time effects. Error bars indicate $\pm 1\sigma$ of the measurements and 95% confidence interval of the measurements is shown in green dashed lines.

Figure 10.

Comparison of measured depth dose and dose profile with the data reported in updated IFU based on Fulkerson *et al.* (Med. Phys. 2014;41:022104).

Dose rate comparison between Acuros calculation and EBT3 film measurement at various depths in the central axis of the applicator for four different aperture sizes. After correction of the measurements for the source sag of 0.47 mm and transit dose, the corrected dose rates were also compared with the Acuros calculation in the last two columns.

Table 1

Aperture	Depth (mm)	Dose rate (cGy/min)			Corrected dose rate (cGy/min)		
		Acuros	EBT3 film ^a	% Difference	EBT3 film ^b	% Difference	% Difference
30 mm	0	477.1	517.7	8.5%	475.3	-0.4%	-0.4%
	2	354.0	378.0	6.8%	349.0 ^c	-1.4%	-1.4%
	3	314.1	322.5	2.7%	302.8	-3.6%	-3.6%
	5	245.2	258.6	5.5%	242.9	-0.9%	-0.9%
	10	147.4	153.5	4.1%	144.4	-2.0%	-2.0%
35 mm	0	478.5	521.8	9.0%	477.6	-0.2%	-0.2%
	2	355.4	385.2	8.4%	355.6 ^c	0.1%	0.1%
	3	315.0	339.7	7.8%	314.4	-0.2%	-0.2%
	5	245.8	253.5	3.1%	240.0	-2.4%	-2.4%
	10	147.7	153.9	4.2%	145.2	-1.7%	-1.7%
40 mm	0	483.8	528.3	9.2%	479.5	-0.9%	-0.9%
	2	358.3	383.6	7.1%	354.2 ^c	-1.2%	-1.2%
	3	317.7	331.5	4.3%	308.0	-3.1%	-3.1%
	5	247.8	260.7	5.2%	244.4	-1.4%	-1.4%
	10	148.6	152.4	2.5%	144.2	-3.0%	-3.0%
45 mm	0	484.0	525.8	8.6%	477.4	-1.4%	-1.4%
	2	358.5	374.7	4.5%	345.9 ^c	-3.5%	-3.5%
	3	317.9	331.0	4.1%	307.2	-3.4%	-3.4%
	5	248.0	256.2	3.3%	240.4	-3.0%	-3.0%
	10	148.7	149.5	0.5%	142.3	-4.3%	-4.3%

^a mean values of two face down measurements

^b mean values of two face down measurement and one face up measurement after corrections

mean values of two face down measurements (face up measurements were not available for 2 mm depth)

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