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Thermal effects of various drill materials during implant site preparation—Ceramic vs. stainless steel drills: A comparative in vitro study in a standardised bovine bone model

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Abstract

Objectives: The aim of this study was to evaluate thermal effects of ceramic and metal implant drills during implant site preparation using a standardised bovine model.

Material and Methods: A total of 320 automated intermittent osteotomies of 10- and 16-mm drilling depths were performed using zirconium dioxide-based and stainless steel drills. Various drill diameters (2.0/ 2.2, 2.8, 3.5, 4.2 mm ⌀) and different cooling methods (without/ with external saline irrigation) were investigated at room temperature (21 ± 1°C). Temperature changes were recorded in real time using two custom-built multichannel thermoprobes in 1- and 2-mm distance to the osteotomy site. For comparisons, a linear mixed model was estimated.

Results: Comparing thermal effects, significantly lower temperatures could be detected with steel-based drills in various drill diameters, regardless of drilling depth or irrigation method. Recorded temperatures for metal drills of all diameters and drilling depths using external irrigation were below the defined critical temperature threshold of 47°C, whereas ceramic drills of smaller diameters reached or exceeded the harmful temperature threshold at 16-mm drilling depths, regardless of whether irrigation was applied or not. The results of this study suggest that the highest temperature changes were not found at the deepest point of the osteotomy site but were observed at subcortical and deeper layers of bone, depending on drill material, drill diameter, drilling depth and irrigation method.

Conclusions: This standardised investigation revealed drill material and geometry to have a substantial impact on heat generation, as well as external irrigation, drilling depth and drill diameter.

KEYWORDS
ceramic drills, dental implants, heat generation, implant osteotomies, irrigation methods, multiple temperature sensors, standardised testing specimen, thermal osteonecrosis

1 | INTRODUCTION

During the past decades, implant-supported treatment solutions have become a predictable and viable option for prosthetic rehabilitation of toothless and partially edentulous patients. In this respect, titanium dental implants have been well documented over a long period of time and are widely accepted as the gold standard in contemporary dental implantology (Buser et al., 2012; Chiapasco et al., 2020; Chrcanovic et al., 2020; Ducommun et al., 2019; Jung et al., 2012; Pjetursson et al., 2012). Regardless of the reported success rates for dental implants, the implications of implant failure can be both medically and financially challenging (Mardinger et al., 2008). Hence, a variety of factors (biological, iatrogenic, mechanical and patient-associated complications) resulting in peri-implant diseases and therefore affecting implant success have been investigated (Esposito et al., 1998a; Lang et al., 2000; Schwarz, 2000). Biological failures have been described as early versus late implant losses, depending on whether failure to achieve or to maintain already established osseointegration was observed (Esposito et al., 1998a). While clinical and diagnostic criteria for peri-implant diseases have been described in detail in the past (Heitz-Mayfield, 2008; Renvert et al., 2018), the contribution of factors to early and late implant failures (such as surgical trauma, chronic marginal infection, implant overload and poor bone quality) remains controversial (Esposito et al., 1998b; Piattelli et al., 2003). Recently, discussion on titanium-based failures due to hypersensitivities and allergies became a focus of attention (Fage et al., 2016; Pigatto et al., 2009; Sicilia et al., 2008; Siddiqi et al., 2011), significance supporting this theory remains still unproven (Javed et al., 2013).

However, greyish peri-implant soft tissue discolouration with titanium implants may pose a challenge in aesthetically sensitive areas, especially in combination with a mucosal thickness of less than 2 mm (Benic et al., 2017; Cosgarea et al., 2015; Ioannidis et al., 2017; Jung et al., 2008). The rising criticism of titanium and the fact that an increasing number of patients is requesting entirely metal-free dental reconstructions have led to ceramic oral implants being considered as promising alternative to titanium (Andreiotelli et al., 2009; Sivaraman et al., 2018; Spies et al., 2019). Aluminium oxide (Al_2O_3, alumina), a ceramic material for dental implants introduced at a similar time as titanium implants (Sandhaus, 1968, 1971; Schulte & Heimke, 1976), was eventually withdrawn from the market due to an increased risk of implant fractures (Ananth et al., 2015; Andreiotelli et al., 2009; Hobkirk & Wiskott, 2009; Kohal et al., 2004). Zirconium dioxide (ZrO_2, zirconia), however, shows favourable physicochemical properties (high bending strength and fracture toughness) depending upon its composition and processing (Piconi & Maccauro, 1999; Sivaraman et al., 2018) as well as high biocompatibility similar to titanium (Benic, Thoma, et al., 2017; Bormann et al., 2012; Gahlert et al., 2012; Janner et al., 2018; Roehling et al., 2019). Although long-term results for zirconia implants are still missing and elevated heat generation was reported in vitro during implant insertion with zirconia implants when compared to titanium implants (Zipprich et al., 2019), promising clinical data investigating the outcome of one-piece zirconia implants after an observation period of 5 years have been recently obtained (Balmer et al., 2020). Newly developed two-piece zirconia implants, which are supposed to overcome problems caused by challenging abutment angulations (Wenz et al., 2008), showed a similar screw-retained stability in vitro (artificial ageing process) when compared to a conventional titanium-based connection (Joos et al., 2020).

Regardless of the dental implant used, an atraumatic and delicate surgical preparation technique is considered to be a key prerequisite for successful osseointegration (Albrektsson et al., 1981; Eriksson & Adell, 1986). Therefore, thermal bone injury has been discussed in-depth as a cause for bone tissue necrosis, followed by impaired microcirculation, activation of bone marrow macrophages and consequently implant failure (Augustin et al., 2012; Eriksson & Albrektsson, 1983, 1984; Eriksson et al., 1982; Esposito et al., 1998b; Piattelli et al., 1998; Trisi et al., 2014; Yoshida et al., 2009). Eriksson & Albrektsson identified the temperature threshold for bone survival to be between 44 and 47°C with an exposure time of less than 1 min (Eriksson & Albrektsson, 1983, 1984); however, an exact threshold for osteonecrosis remains still unclear (Augustin et al., 2012; Oliveira et al., 2012; Yoshida et al., 2009). Numerous factors have been reported to influence heat generation during implant bed preparation, such as drill diameter (Strbac, Giannis, et al., 2014; Strbac, Unger, et al., 2014), drill design and geometry (Cordioli & Majzoub, 1997; Oh et al., 2011; Sannino et al., 2015), drill load (Abouzgia & James, 1997), sharpness and drill wear (Salimov et al., 2020; Scarano et al., 2007), surgical technique used (Frösch et al., 2019; Lajolo et al., 2018; Lucchiari et al., 2016), irrigation (Gehrke et al., 2018; Harder et al., 2013; Strbac et al., 2015) and bone density (Yacker & Klein, 1996). Moreover, drill material has been suggested to affect temperature increase during osteotomy (Hochscheidt et al., 2017; Oliveira et al., 2012; Scarano et al., 2020; Sumer et al., 2011). Conventional rotary preparation with metal burs is still the most commonly used procedure in dental implantology.

However, as a result of the above-mentioned focus on freedom from metal, the fact that metallic contamination of bone due to drilling procedures (Hobkirk & Rusiniak, 1978) and surface corrosion of steel drills after contact with disinfecting liquids has been reported (Scarano et al., 2019) and the introduction of mixed ceramics with improved mechanical properties (Gaertner et al., 2005; Piconi & Maccauro, 1999) may lead to an increasing use of ceramic burs for implant site osteotomies.

The clinical use of ceramic implant drills is still a topic of discussion, and scientific evidence regarding thermal performance is scarce and controversial. This study aims to investigate temperature changes with different implant drill materials and various irrigation methods in standardised bovine specimens in order to expand knowledge regarding future implant procedures. Our investigation was based on the working hypothesis that ceramic implant drills would provide for beneficial temperature effects when compared to stainless steel drills.

2 | MATERIAL AND METHODS

2.1 | Bone specimens and implant drills

In the present in vitro study, temperature measurements during drilling procedures were recorded using artificially manufactured bone specimens (BoneSim™, 1800.35/1300.14 Composite, BoneSim™, Newaygo, MI, USA) with distinctive cortical (3 mm) and cancellous (15 mm) bone sections (Figures 1 and 2). By simulating human mandibular bone density (type 2 according to Lekholm and Zarb classification), this testing specimen represents a novel standardised bone model for thermal evaluation in bone osteotomies (Abboud et al., 2015; Delgado-Ruiz et al., 2016, 2018; Lekholm & Zarb, 1985; Strbac et al., 2015; Strbac, Giannis, et al., 2014). A thermal conductivity of 0.3–0.4 W m^{-1} K^{-1} ensures a similar thermal conductivity as human bone, thus providing comparable clinical testing conditions (Clattenburg et al., 1975; Davidson & James, 2000). Commercially available surgical twist drills made of stainless steel (stainless martensitic steel DIN Code: 1.4108; 2.2, 2.8, 3.5, 4.2 mm ∅; Straumann PRO™, Straumann®, Basel, Switzerland) and alumina-toughened zirconia (2.0, 2.8, 3.5, 4.2 mm ∅; Komet Ceradrill™, Komet®, Gebr. Brasseler, Lemgo, Germany) for graduated preparation were used (Figures 3 and 4). Ethics approval was not required for this in vitro study.

2.2 | Experimental set-up

A customised surgical system (SH-Surgical Drilling-Sequence-Simulator System, Center for Medical Physics and Biomedical Engineering, Medical University of Vienna, Austria) with a stepper motor for automated and therefore reproducible drilling cycles was manufactured. Its software program (SSH-Surgical Drilling-Sequence-Software 1.0; Center for Medical Physics and Biomedical Engineering, Medical University of Vienna, Austria) provided for a precise intermittent vertical dislocation of a surgical handpiece (WS-75 E/KM 20:1, W&H, Bürmoos, Austria) mounted in a computer-milled clamp (Figure 5). Parameters (drilling/withdrawing feed rate, depth control and dwell time) for two different drilling sequences (10 and 16 mm) were predetermined according to related previous investigations (Strbac et al., 2015; Strbac, Giannis, et al., 2014;

FIGURE 1 Standardised bovine bone specimen (BoneSim, 1800.35/1300.14 Composite, BoneSim, Newaygo, MI, USA) embedded in polystyrene test box

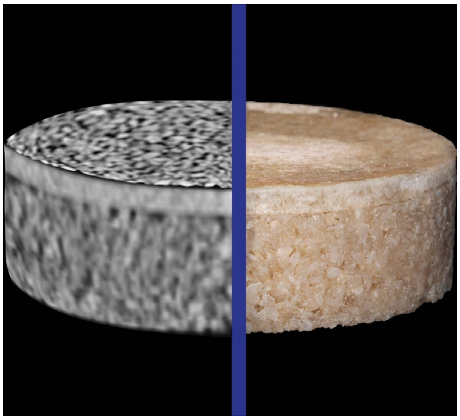

FIGURE 2 Comparison bone specimen in computed tomography presenting 3 mm cortical and 15 mm cancellous bone sections according to type 2 Lekholm and Zarb classification

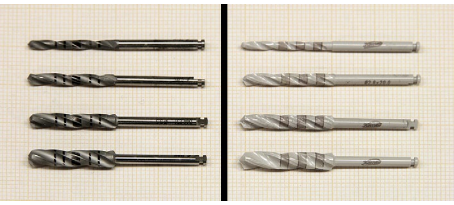

FIGURE 3 Implant twist drills with respective drill diameters used for the investigation, left image: metal drills (2.2, 2.8, 3.5, 4.2 mm ∅, Straumann PRO™, Straumann®, Basel, Switzerland), right image: ceramic drills (2.0, 2.8, 3.5, 4.2 mm ∅, Komet Ceradrill™, Komet®, Gebr. Brasseler, Lemgo, Germany)

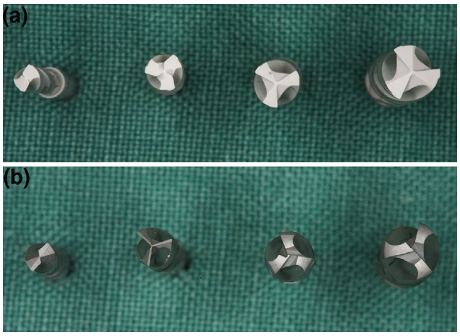

FIGURE 4 Comparison of drill geometries and drill shapes with increasing drill diameter due to different material properties ((a): ceramic drills, (b): metal drills)

Strbac, Unger, et al., 2014). For the 10-mm drilling depth, the automated osteotomies lasted 27.6 s (drilling 17.3 s, withdrawing 10.3 s) and for the 16-mm drilling depth 43.5 s (drilling 27.1 s, withdrawing 16.4 s). The drilling and withdrawing feed rate in cortical bone was 0.5 mm/s, in cancellous bone 1 and 5 mm/s during final withdrawing.

2.3 | Temperature measurement set-up

Two custom-built thermoprobes (SHT-Thermoprobe, Center for Medical Physics and Biomedical Engineering, Medical University of

FIGURE 5 Automated intermittent and graduated preparation sequence (e.g. 4.2 mm ∅ ceramic implant drill, 16-mm drilling depth, without irrigation) and temperature measurement system (2 thermoprobes with injected heat-transfer compound)

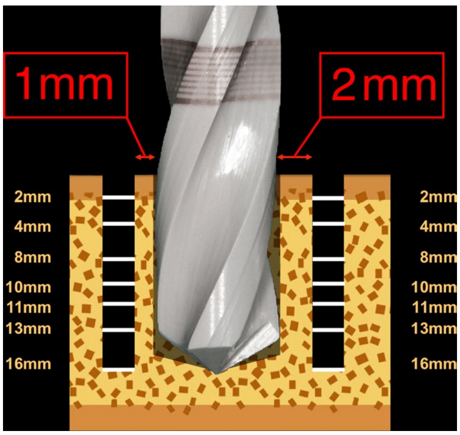

FIGURE 6 Schematic illustration of real-time temperature measurement system, 2 thermoprobes in 1- and 2-mm distance to the drilling site, 14 individual temperature sensors and their respective measurement depths

Vienna, Austria) with 1.5-mm diameter and 18-mm length were used to obtain real-time temperature changes. These multichannel measuring devices consisted of a 3D printed body with 14 individual temperature sensors (7 sensors per thermoprobe, 0.4 mm ∅, response time >0.2 s, 10 KΩ at 25°C, temperature range −40 to +250°C). The thermoprobes were software designed (NX 5.0.3.2 Unigraphics, PLM Software, Siemens, Cologne, Germany) and manufactured by a rapid prototyping system (Eden350, Objet Ltd., Rehovot, Israel) using photopolymer resin (Objet FullCure720™, Objet Ltd., Rehovot, Israel). They were planned with predefined notches for the NTC sensors (negative temperature coefficient sensors; Miniature Axial Glass Thermistor, No. GA10KM3499J15, Measurement Specialties™, Hampton, VA, USA) at depths of 2, 4, 8 and 10 mm for 10-mm drilling sequence and additionally 11, 13 and 16 mm for 16-mm drilling sequence (Figure 6). A computer-aided temperature measurement system (SHTM-Temperature Measurement System, Center for Medical Physics and Biomedical Engineering, Medical University of Vienna, Austria), a measurement amplifier (SHU-Measurement Amplifier, Center for Medical Physics and Biomedical Engineering, Medical University of Vienna, Austria), an ADC-converter (Analogue-to-digital-converter; NI DAQCard™-6062E, National Instruments™, Austin, TX, USA) and a software-controlled program (DASYLab® Software 5.0; Measurement Computing Corporation, Norton, MA, USA) detected electrical resistance of the NTC sensors, allowing real-time measurement and recording of temperature after initial calibration against traceable standards (Strbac et al., 2015; Strbac, Giannis, et al., 2014; Strbac, Unger, et al., 2014).

2.4 | Experimental protocol

A total of 50 bovine bone specimens were embedded in rectangular polystyrene test boxes (No. 34160-0101; Bock, Lauterbach, Germany) and bonded with a two-component epoxy resin adhesive (Loctite® Double Bubble™, Henkel AG & Co. KGaA, Düsseldorf, Germany), ensuring a stable position throughout the experimental drilling procedure. Individual metal templates for positioning the thermoprobes were CNC-milled in order to precisely position them 1 and 2 mm from the final preparation site using a twist drill (2 mm ∅, drilling depth 18 mm, 210L20.205.020, Komet®, Gebr. Brasseler, Lemgo, Germany) (Oliveira et al., 2012; Rashad et al., 2011; Strbac, Giannis, et al., 2014; Strbac, Unger, et al., 2014) (Figure 6). Before inserting the two thermoprobes, a heat-transfer compound (HTCP20S 20 ml, Electrolube®, Leicestershire, UK) was injected in each canal for optimal thermal conductivity (Ercoli et al., 2004; Harder et al., 2013; Strbac et al., 2015). The experimental protocol consisted of two drill materials (ceramic and stainless steel) with 4 diameters each (2.0/ 2.2, 2.8, 3.5 and 4.2 mm ∅), 2 different cooling methods (without/with external saline irrigation), two drilling depths (10 mm/16 mm) and was conducted with 10 repetitions ($n = 320$ preparations in total) under constant room temperature (21 ± 1°C). A computer-milled table for horizontal dislocation of the embedded specimen ensured precise automated preparations by the surgical handpiece connected to a surgical motor unit (Implantmed SI-923; Surgical control S-N1, W&H, Bürmoos, Austria) with a new and unused drill for each osteotomy (Figure 5). Graduated predrilling according to clinical recommendations was performed prior to measurements. The real-time temperature at 1- and 2-mm distance was recorded during all osteotomies starting 10 s before drilling and ending 25 s after withdrawing (Figure 7). In case of external irrigation sequences, constant saline cooling of 50 ml/min at room temperature (Ecobag® click, 0.9% NaCl, 5,000 ml, B. Braun Melsungen AG, Melsungen, Germany) was applied throughout the whole preparation period (drilling start to withdrawing end) using an irrigation tubing set (Irrigation set for machinery—80 mm, 32.F0139, Omnia®, Fidenza, Italy) and surgical suction was applied at 1.5 cm from preparation site. All osteotomies were performed according to a standard protocol for an atraumatic preparation at 800 rpm (Gaspar et al., 2013; Koopaie et al., 2019; Oliveira et al., 2012; Scarano et al., 2020; Strbac et al., 2015; Strbac, Giannis, et al., 2014; Strbac, Unger, et al., 2014).

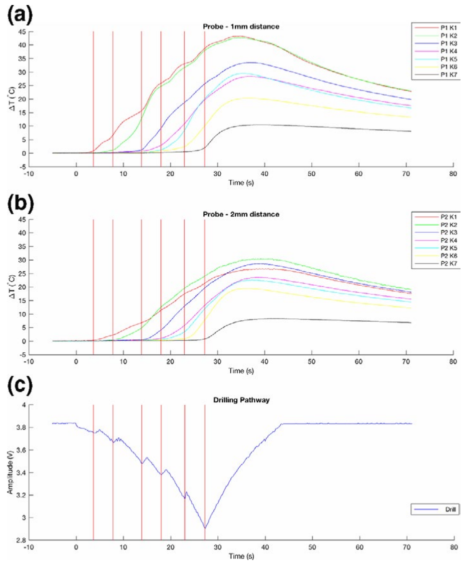

FIGURE 7 Illustration of multichannel real-time measurement of temperature changes (ΔT) at 16-mm drilling depth in (a) 1- and (b) 2-mm distance to the osteotomy site, (c) corresponding drilling pathway recorded by external linear motion potentiometer

2.5 | Statistical analysis

The obtained experimental data were recorded (ASC file format) for each osteotomy and included real-time recordings of 14 temperature sensors, a time signature (δ = 0.001 s) and a record of an external linear motion potentiometer (Linear Potentiometer 600 Series, Type 9615R5.1KL2.0, BEI Sensors, Goleta, CA, USA). In order to match temperature variations with each osteotomy and to process the data, a custom analysis software for descriptive statistics was used (MATLAB®, R2016a, MathWorks®, Natick, MA, USA). Temperature changes were calculated [$\Delta T(°C) = T_x - T_0$] by subtracting the recorded temperature [T_x] with the bone specimen baseline temperature [T_0] before each osteotomy (Abboud et al., 2015; Calvo-Guirado et al., 2015; Gehrke et al., 2015; Oliveira et al., 2012; Rashad et al., 2011; Sannino et al., 2015; Strbac, Giannis, et al., 2014; Strbac, Unger, et al., 2014). For statistical analysis, temperature and temperature differences were normally distributed and described and tabulated with mean ± standard deviation. Depth of sensor channel with maximum temperature increase was described and tabulated with median, minimum and maximum. For comparison of ceramic and stainless steel drills, a linear mixed model was fitted including the variables material (ceramic/metal), drilling diameter (2.0/2.2, 2.8, 3.5, 4.2 mm), drilling depth (10 mm/16 mm) and irrigation (with/without). Compound symmetry was assumed for repeated measurements. A four-way interaction analysis of the four explanatory variables was conducted. In case of significant interactions, subgroup analyses to test for differences in drill materials were performed and corresponding *p*-values are presented. Statistical calculations were performed with the statistical software SAS® (Version 9.4, SAS Institute Inc., Cary, NC, USA). All *p*-values are two-sided, and $p \leq .05$ was considered statistically significant.

3 | RESULTS

Temperature changes during implant site preparation of 320 osteotomies with different drill materials, drill diameters, drilling depths, irrigation methods and 10 repetitions were investigated. The mean bone specimen baseline temperature [T_0] before osteotomy procedures was 21.74 ± 1.18°C for the 10-mm drilling depth and 21.80 ± 1.26°C for the 16-mm drilling depth. Bone temperatures increased with all the implant drills tested; the distributions of mean differences between drilling and baseline temperatures over all temperature measurement sensors (10-mm drilling sequence: 2×4 sensors, 16-mm drilling sequence: 2×7 sensors) are shown in Figure 8 and Table 1.

3.1 | Highest mean temperature increase

The maximum mean temperature increases for 10-mm drilling sequences without irrigation were as follows [ΔT°C mean (*SD*)]: 18.70 (1.14) for the 2.0 mm drill (CER) and 13.93 (3.05) for the 2.2 mm drill (MET); 13.15 (2.35) for the 2.8 mm drill (CER) and 12.90 (2.93) for the 2.8 mm drill (MET); 14.33 (1.83) for the 3.5 mm drill (CER) and 15.16 (2.08) for the 3.5 mm drill (MET); and 12.61 (1.80) for the 4.2 mm drill (CER) and 10.17 (5.17) for the 4.2 mm drill (MET; Figure 8, Table 1).

The highest mean temperature increases for 10-mm drilling sequences with external cooling were as follows [ΔT°C mean (*SD*)]: 12.13 (5.68) for the 2.0 mm drill (CER) and 3.73 (0.73) for the 2.2 mm drill (MET); 7.47 (1.89) for the 2.8 mm drill (CER) and 2.59 (0.68) for the 2.8 mm drill (MET); 3.94 (0.61) for the 3.5 mm drill (CER) and 3.16 (0.79) for the 3.5 mm drill (MET); and 2.73 (0.31) for the 4.2 mm drill (CER) and 3.06 (0.76) for the 4.2 mm drill (MET; Figure 8, Table 1).

The maximum mean temperature increases for 16-mm drilling sequences without irrigation were as follows [ΔT°C mean (*SD*)]: 27.20 (2.81) for the 2.0 mm drill (CER) and 23.85 (4.75) for the

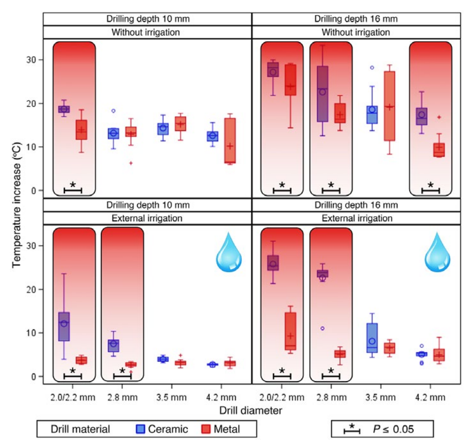

FIGURE 8 Temperature increase in investigated drill materials (blue colour = ceramic, red colour = metal, *$p \leq .05$) of different drill diameters (2.0/2.2, 2.8, 3.5, 4.2 mm ∅), irrigation methods (without/external irrigation) and drilling depths (10/16 mm)

TABLE 1 Maximum temperature increase [ΔT°C mean (*SD*)] over all temperature sensors (10-mm drilling sequence: 2×4 sensors, 16-mm drilling sequence: 2×7 sensors) at different drilling depths with various drill diameters and irrigation methods, testing drill material ceramic (CER) versus metal (MET); *P*-values for drill material comparisons are calculated by a linear mixed model after significant four-way interaction

		Drilling depth 10 mm		Drilling depth 16 mm	
		Irrigation method		Irrigation method	
Drill diameter	Drill material	Without	External	Without	External
2.0 mm	CER	18.70 (1.14)	12.13 (5.68)	27.20 (2.81)	25.79 (2.86)
2.2 mm	MET	13.93 (3.05)	3.73 (0.73)	23.85 (4.75)	9.29 (4.40)
	p-value	.002*	<.001*	.025*	<.001*
2.8 mm	CER	13.15 (2.35)	7.47 (1.89)	22.63 (7.20)	22.65 (4.23)
2.8 mm	MET	12.90 (2.93)	2.59 (0.68)	17.41 (2.82)	5.02 (1.17)
	p-value	.824	.002*	<.001*	<.001*
3.5 mm	CER	14.33 (1.83)	3.94 (0.61)	18.58 (4.50)	8.09 (3.59)
3.5 mm	MET	15.16 (2.08)	3.16 (0.79)	19.14 (8.15)	6.55 (1.36)
	p-value	.539	.596	.708	.312
4.2 mm	CER	12.61 (1.80)	2.73 (0.31)	17.41 (3.13)	5.01 (1.19)
4.2 mm	MET	10.17 (5.17)	3.06 (0.76)	9.90 (2.99)	4.97 (1.91)
	p-value	.095	.803	<.001*	.984

*$p \le .05$.

2.2 mm drill (MET); 22.63 (7.20) for the 2.8 mm drill (CER) and 17.41 (2.82) for the 2.8 mm drill (MET); 18.58 (4.50) for the 3.5 mm drill (CER) and 19.14 (8.15) for the 3.5 mm drill (MET); and 17.41 (3.13) for the 4.2 mm drill (CER) and 9.90 (2.99) for the 4.2 mm drill (MET; Figure 8, Table 1).

The highest mean temperature increases for 16-mm drilling sequences with external cooling were as follows [ΔT°C mean (*SD*)]: 25.79 (2.86) for the 2.0 mm drill (CER) and 9.29 (4.40) for the 2.2 mm drill (MET); 22.65 (4.23) for the 2.8 mm drill (CER) and 5.02 (1.17) for the 2.8 mm drill (MET); 8.09 (3.59) for the 3.5 mm drill (CER) and 6.55 (1.36) for the 3.5 mm drill (MET); and 5.01 (1.19) for the 4.2 mm drill (CER) and 4.97 (1.91) for the 4.2 mm drill (MET; Figure 8, Table 1).

3.2 | Temperature increase and drill material

With regard to implant material, significant temperature differences ($p \le .05$) during implant preparations at drilling depths of 10 and 16 mm with different drill diameters and irrigation methods were observed. Drill materials were compared with respect to corresponding diameter, cooling method and osteotomy depth. The differences in heat generation between the drill materials differed significantly and, whenever significant results were found, ceramic drills invariably showed higher mean temperature increases [ΔT°C (*SD*)] compared with metal drills (Figure 8, Table 1).

3.2.1 | Drilling osteotomies of 10-mm depth

During drilling sequences of 10-mm depth, significantly higher temperatures were observed using ceramic drills of 2.0-mm diameter without irrigation ($p = .002$), with external irrigation ($p < .001$), as well as using 2.8-mm ceramic drills with external irrigation ($p = .002$) (Figure 8, Table 1).

3.2.2 | Drilling osteotomies of 16-mm depth

During drilling osteotomies of 16-mm depth, significantly higher temperatures were found using ceramic drills of 2.0-mm diameter without cooling ($p = .025$), with external cooling ($p < .001$), using 2.8-mm ceramic drills without cooling ($p < .001$), with external cooling ($p < .001$), as well as using 4.2-mm ceramic drills without cooling ($p < .001$) (Figure 8, Table 1).

3.3 | Temperature increase and sensor location/corresponding depth

In order to analyse bone areas affected by the temperature increase, the occurrence of maximum temperature changes at median sensor channel depth [ch (minimum–maximum)] (sensor channel depths: 2, 4, 8, 10 mm for 10-mm drilling sequence and additionally 11, 13, 16 for 16-mm drilling sequence) was calculated for 1- and 2-mm measuring distance (Table 2).

3.3.1 | Drill diameter 2.0/2.2 mm

Highest temperature changes using 2.0-/2.2-mm drills during 10-mm drilling osteotomies were found between median sensor channels [ch] of 3 and 4 mm without irrigation and 4 and 8 mm with external irrigation.

TABLE 2 Location of maximum temperature increase: median sensor channel location [ch (minimum–maximum)] (sensor channel depths: 2, 4, 8, 10 mm for 10-mm drilling sequence and additionally 11, 13, 16 for 16-mm drilling sequence) in 1- and 2-mm measuring distance (MD = measuring distance, CER = ceramic, MET = metal)

			Drilling depth 10 mm		Drilling depth 16 mm	
			Irrigation method		Irrigation method	
Drill diameter	Drill material	MD	Without	External	Without	External
2.0 mm	CER	1 mm	4 mm (2–4)	4 mm (4–8)	4 mm (2–4)	4 mm (4–11)
		2 mm	4 mm (4–4)	8 mm (4–8)	6 mm (4–8)	8 mm (8–11)
2.2 mm	MET	1 mm	3 mm (2–4)	8 mm (4–8)	4 mm (2–11)	11 mm (11–13)
		2 mm	4 mm (4–4)	8 mm (4–8)	8 mm (4–8)	13 mm (8–16)
2.8 mm	CER	1 mm	4 mm (2–4)	8 mm (4–8)	6 mm (4–11)	11 mm (4–11)
		2 mm	4 mm (4–8)	8 mm (8–8)	8 mm (8–13)	10 mm (8–13)
2.8 mm	MET	1 mm	4 mm (2–4)	8 mm (8–8)	4 mm (4–4)	11 mm (11–11)
		2 mm	4 mm (4–4)	8 mm (8–8)	8 mm (4–11)	13 mm (13–13)
3.5 mm	CER	1 mm	4 mm (4–4)	8 mm (4–8)	9 mm (2–11)	11 mm (11–13)
		2 mm	4 mm (4–4)	8 mm (8–8)	8 mm (4–8)	13 mm (8–13)
3.5 mm	MET	1 mm	4 mm (2–4)	8 mm (4–8)	4 mm (2–11)	11 mm (11–11)
		2 mm	4 mm (4–4)	8 mm (8–8)	8 mm (4–8)	13 mm (10–13)
4.2 mm	CER	1 mm	4 mm (4–4)	8 mm (4–8)	4 mm (2–11)	11 mm (11–11)
		2 mm	4 mm (4–8)	8 mm (4–8)	8 mm (8–8)	13 mm (8–13)
4.2 mm	MET	1 mm	4 mm (2–4)	8 mm (4–8)	4 mm (2–11)	11 mm (11–11)
		2 mm	4 mm (4–4)	8 mm (4–8)	8 mm (8–8)	13 mm (8–13)

During 16-mm drilling depths, maximum temperature changes using 2.0-/2.2-mm drills were observed between median sensor channels [ch] of 4 and 8 mm without irrigation and 4 and 13 mm when external irrigation was used (Table 2).

3.3.2 | Drill diameter 2.8 mm

Maximum temperature changes using 2.8-mm drills during 10-mm drilling sequences were observed at median sensor channel [ch] of 4 mm without irrigation and 8 mm when external irrigation was used.

During 16-mm drilling osteotomies, highest temperature changes using 2.8-mm drills were found between median sensor channels [ch] of 4 and 8 mm without irrigation and 10 and 13 mm with external irrigation (Table 2).

3.3.3 | Drill diameter 3.5 mm

Highest temperature changes using 3.5-mm implant drills during 10-mm drilling depths were found at median sensor channel [ch] of 4 mm without irrigation and 8 mm when external irrigation was applied.

During 16-mm drilling sequences, maximum temperature changes using 3.5-mm drills were observed between median sensor channels [ch] of 4 and 9 mm without irrigation and 11 and 13 mm with external cooling (Table 2).

3.3.4 | Drill diameter 4.2 mm

Maximum temperature changes using 4.2-mm drills during 10-mm drilling osteotomies were observed at median sensor channel [ch] of 4 mm without irrigation and 8 mm when external cooling was used.

During 16-mm drilling depths, highest temperature changes using 4.2-mm drills were found between median sensor channels [ch] of 4 and 8 mm without irrigation and 11 and 13 mm with external irrigation (Table 2).

4 | DISCUSSION

As yet, there have been only a few published investigations exploring the performance of ceramic and metal implant drills with regard to intrabony thermal effects. These scientific studies have been performed on a variety of osseous bone models using different temperature measurement systems (various thermocouples or infrared thermography devices) and in vitro study designs. In addition, they have been mainly focused on the temperature correlation between drill material and drill wear (Hochscheidt et al., 2017; Koo et al., 2015; Koopaie et al., 2019; Oliveira et al., 2012; Pires et al., 2012; Scarano et al., 2020).

The purpose of this investigation was to examine metal and ceramic implant twist drills with identical or similar diameters during automated and reproducible drilling osteotomies by using a highly sensitive real-time multichannel temperature measurement system and a standardised bovine bone model, previously introduced for temperature testing of surgical instruments (Abboud et al., 2015; Delgado-Ruiz et al., 2016, 2018; Strbac et al., 2015; Strbac, Giannis, et al., 2014). Infrared thermography poses a valid alternative to thermocouple technology and has been successfully used in temperature investigations in the past (Augustin et al., 2012; Benington et al., 2002; Frösch et al., 2019). Doubts concerning its accuracy when recording temperatures at irrigated preparation sites (Benington et al., 1996; Tehemar, 1999) led to further development of real-time multichannel thermoprobes by the authors in the past (Strbac et al., 2015; Strbac, Giannis, et al., 2014; Strbac, Unger, et al., 2014). This established temperature measurement system was applied in the present investigation as well, taking into account the fact that the comparability of results may be limited when it comes to different study designs (such as varying number of thermocouples, distance to preparation site, number and configuration of temperature sensors).

For overcoming limitations of previous temperature studies and as a result of the fact that temperature increase during implant osteotomies is considered to be a complex interaction of multiple factors (Augustin et al., 2012; Möhlhenrich et al., 2015; Tehemar, 1999), drill wear was consciously excluded as a contributing factor in this investigation by only testing new and unused drills.

The mean temperature increase for metal drills with external irrigation was below the defined critical temperature threshold of 47°C in all drill diameters and drilling depths, thereby confirming the cooling effect of external irrigation on bone temperature using metal implant drills (Augustin et al., 2008; Harder et al., 2013; Oliveira et al., 2012; Rashad et al., 2011; Sener et al., 2009; Strbac, Unger, et al., 2014). One of the most important findings of this investigation was the statistically significant difference in temperature performance between metal and ceramic drills, especially with small drill diameters (2.0/2.2 and 2.8 mm ∅) (Figure 8, Table 1). Recorded temperatures with ceramic drills at 16-mm drilling depth reached or exceeded the harmful temperature threshold, regardless of whether irrigation was applied or not. Local temperatures at the drilling site can be presumed to be even higher due to the technical measuring distance of 1 and 2 mm to the osteotomy site (Oliveira et al., 2012; Strbac, Giannis, et al., 2014; Yacker & Klein, 1996).

Internal or combined internal and external irrigation may be considered a convenient alternative for overcoming cooling problems with external irrigation alone (Gehrke et al., 2018; Harder et al., 2013; Lavelle & Wedgwood, 1980; Strbac et al., 2015; Strbac, Giannis, et al., 2014; Strbac, Unger, et al., 2014; Tehemar, 1999). However, ceramic drills are not manufactured with internal cooling channels due to the risk of fracture (Pires et al., 2012). Even though our present findings suggest that cooling efficiency of external irrigation using ceramic drills of smaller diameter (2.0/2.8 mm ∅) compared with metal drills should be considered as less effective, external irrigation itself was confirmed to be one of the most influential factors on heat generation (Augustin et al., 2008; Ercoli et al., 2004; Kerawala et al., 1999; Koo et al., 2015; Sener et al., 2009; Strbac, Giannis, et al., 2014; Strbac, Unger, et al., 2014).

Graduated osteotomy technique is believed to reduce friction and consequently temperature by removing a smaller quantity of bone material in each drilling step when compared to single drilling procedures. The observation that predrilling with pilot drills of both drill materials (2.0/2.2 mm ∅), especially in deeper osteotomies, was associated with considerably higher temperatures when compared to expansion drilling procedures (2.8, 3.5, 4.2 mm ∅) and the fact that predominantly lower temperatures both in metal and ceramic drills were recorded with increasing diameter, confirm the beneficial effect of graduated drilling technique (Augustin et al., 2012; Eriksson & Adell, 1986; Lucchiari et al., 2016; Oh et al., 2011; Strbac et al., 2015; Strbac, Giannis, et al., 2014).

As demonstrated by Eriksson & Albrektsson, intrabony thermal effects are not only influenced by temperature generated during preparation, but also influenced by exposure time (Eriksson & Albrektsson, 1983). Previous investigations confirmed that the induced amount of frictional heat correlates with the drilling time (Abouzgia & James, 1995; Grunder & Strub, 1986; Iyer et al., 1997; Sener et al., 2009) and that drilling depth is considered to be a factor influencing temperature generation (Augustin et al., 2012; Cordioli & Majzoub, 1997; Lee et al., 2012; Oliveira et al., 2012; Strbac, Giannis, et al., 2014; Strbac, Unger, et al., 2014). In accordance with these findings, the present study was able to demonstrate that greater drilling depths (10- vs. 16-mm drilling depth) and thus a prolonged exposure time to frictional forces (27.6 s vs. 43.5 s) were almost invariably associated with higher temperatures.

When evaluating temperatures in terms of the respective bone level, temperature increase was recorded at all temperature sensor depths (both in cortical and deeper cancellous layers of bone) for all drill materials, drill diameters, drilling depths and cooling methods. The distribution of maximum temperature for 10-mm drilling osteotomies indicates that highest temperatures for all drill materials and diameters were mainly observed at subcortical levels (4 mm) without irrigation and in deeper bone sections (8 mm) when external irrigation was used. In 16-mm drilling depth, maximum temperatures were mostly recorded between 4 and 8 mm for all drill materials and diameters without irrigation and between 8 and 13 mm with external irrigation when using larger drill diameters (2.8, 3.5, 4.2 mm ∅) of both drill materials. However, when comparing ceramic and metal pilot drills (2.0/2.2 mm ∅) of 16-mm drilling sequence with external irrigation, ceramic drills were found to cause temperature changes in more superficial layers of bone than metal pilot drills (Table 2). These results support earlier investigations, which reported twist drills of small diameters to be associated with higher temperatures (Cordioli & Majzoub, 1997; Strbac et al., 2015; Strbac, Giannis, et al., 2014) and identified maximum temperatures to be located in subcortical or deeper bone sections (Cordioli & Majzoub, 1997; Harder et al., 2013; Misic et al., 2011; Misir et al., 2009; Strbac et al., 2015; Strbac, Unger, et al., 2014; Sumer et al., 2011). Our findings additionally confirmed that external irrigation seems to be associated with temperature increase in deeper layers of bone, thereby verifying the superficial efficiency of external cooling (Cordioli & Majzoub, 1997; Harder et al., 2013; Lavelle & Wedgwood, 1980; Misir et al., 2009; Moshiri et al., 2013).

Comparing the temperature performance of metal and ceramic implant drills, our findings clearly seem to contradict the majority of similar previous investigations, which observed lower temperatures with ceramic drills (Koopaie et al., 2019; Oliveira et al., 2012; Scarano et al., 2020) or did not find any statistically significant differences between the two drill materials (Harder et al., 2013; Koo et al., 2015; Moshiri et al., 2013; Pires et al., 2012). Our investigation was able to confirm findings of Sumer et al., who observed significantly higher temperatures in 3-mm depth using ceramic drills when compared to stainless steel (Sumer et al., 2011). The observed differences between the two drill materials in our present investigation could mainly be explained by deviating material properties (in particular thermal conductivity) of alumina-toughened zirconia and stainless steel drills (Gaertner et al., 2005), but also by drill geometry. With regard to the latter, earlier findings suggested drill geometry to be a key factor associated with temperature generation (Ali Akhbar & Yusoff, 2019; Chacon et al., 2006; Oh et al., 2011; Oliveira et al., 2012; Sannino et al., 2015; Scarano et al., 2011; Strbac, Giannis, et al., 2014), given that sharpness and geometry are influencing friction and heat production by having an impact on pressure exerted on the drill bit (Pirjamalineisiani et al., 2016).

However, the present investigation had some limitations. The results of this in vitro study have been obtained using artificial bovine discs with no vital bone experiments or direct simulation of in vivo conditions (such as body temperature or blood flow). Consequently, heat generation of the used instruments and techniques may vary in vivo from the present experimental set-up. Previous investigations may be regarded as non-standardised and diverse due to major differences in their study designs (such as bone model, experimental set-up, drilling speed, temperature measurement system, focus on durability/ drill wear), therefore making comparison between these previously published findings rather difficult.

The aim of this study was, despite of its limitations, to further refine and establish standardised instrument performance testing and to contribute to safety of medical components by avoidance of human or animal experiments on ethical grounds. Future studies could pursue more uniform in vitro testing conditions, facilitating comparison of testing results, especially when it comes to new and hardly investigated treatment options.

In summary, the present study could confirm preceding investigations and reveal drill material and geometry as significant factors for temperature generation, even though recommended saline irrigation, intermittent and graduated drilling were performed. Zirconia and mixed ceramics can be recognised as innovative and promising new dental materials due to their physicochemical and biological characteristics, although further modifications in terms of drill geometry, sharpness, material thickness and properties should be considered and investigated, especially when ceramic drills are being used in bone osteotomies.

To the best knowledge of the authors, this study can be considered as the first fully standardised in vitro investigation using an established uniform bone model for performance testing of ceramic and metal drills under recommended atraumatic clinical conditions.

5 | CONCLUSION

This standardised comparative investigation revealed the significant impact of drill material and geometry on intrabony heat generation. Previously published results were mainly obtained using a huge variety of non-standardised experimental set-ups including focus on drill wear, impairing the comparability of results. Our findings suggest a beneficial effect of graduated, intermittent preparation technique using external saline irrigation, although major differences in temperature performance between metal and ceramic implant drills could be observed. Furthermore, the results of this study may contribute towards technical modifications of ceramic drills in the future and thus further improve the long-term clinical results of dental implants, especially considering the use of new upcoming ceramic dental implants.

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AUTHOR CONTRIBUTION

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AUTHOR CONTRIBUTIONS

D.T. involved in concept/design, led the experimental investigation and the writing of the manuscript. K.G. and X.R. revised the manuscript critically. M.M. performed statistical analysis of the collected data. E.U. involved in concept/design and supported the investigation technically. GD.S. conceived the ideas, concept/design, revised the manuscript critically, involved in approval of article and funding secured.

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