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Temperature rise in surrounding tissues during warm vertical compaction in c-shaped root canal systems: a finite element study

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ABSTRACT

Background:

Temperatures above 47 °C can cause irreversible damage to the periodontal ligament and surrounding tissues. This study investigated whether a 10 °C temperature rise, potentially harmful to the vascular structure of the periodontal ligament, occurs during the continuous wave compaction technique. It also assessed the effects of C-shaped root canal anatomy and device temperature settings on thermal transmission.

Methods:

Finite element analysis was used to simulate C-shaped mandibular molars, including periodontal ligament tissues with assumed perfusion rates equal to those in human gingiva. Temperature changes were recorded during warm vertical compaction and backfill phases. The temperature at the periodontal ligament adjacent to the root surface was continuously monitored, and the highest values and their timing were documented.

Results:

High-temperature settings (250 °C for down-pack, 200 °C for backfill) led to peak periodontal ligament temperatures of 66.67 °C and 53.31 °C, respectively. In contrast, lower settings (90 °C and 100 °C) kept periodontal ligament temperatures below the critical threshold, with maximum values of 44.37 °C and 43.12 °C. The greatest temperature rise occurred during the early seconds of the down-pack phase.

Conclusions:

Excessive heat generated during warm obturation techniques can elevate periodontal ligament temperature beyond safe limits, especially in anatomically complex roots. Lower temperature settings are recommended to minimize the risk of thermal injury to periradicular tissues.

Keywords: Root Canal Treatment; C-shaped Root Canal; Warm Vertical Compaction; Finite Element Analysis

Background

The primary goal of endodontic treatment is to thoroughly clean the root canal system by removing infected pulp tissue and to prevent reinfection through hermetic obturation. However, achieving this goal can be technically challenging due to the complex anatomy of root canals, particularly in teeth with C-shaped configurations (1). C-shaped canals are an anatomical variation most commonly observed in mandibular second molars, characterized by multiple interconnected canal systems that often appear as a continuous “C” shape in cross-section (2). The prevalence and morphology of these canals vary significantly depending on ethnicity, gender, age, and the diagnostic techniques used (3). Reported prevalence rates range from under 1% to over 40%, with notable variations across

populations and tooth types. Their complex and variable anatomy often leads to diagnostic and treatment challenges (4-6). Debridement and disinfection of C-shaped canals are particularly difficult because the canal system often extends perpendicular to the oval or flat coronal portion of the root canal space. Moreover, the presence of thin dentinal walls, especially in the groove areas requires clinicians to proceed cautiously to avoid procedural errors such as strip perforation during cleaning and shaping (7).

Among the various techniques available for root canal obturation, Lateral Compaction and Warm Vertical Compaction (WVC) are the two most commonly employed approaches in the context of C-shaped canals, each with distinct procedural protocols and clinical implications (8, 9). WVC is often favored for such anatomically complex cases due to its superior ability to adapt gutta-percha (GP) to the canal walls. This technique relies on heating GP to achieve plasticity and enhance the seal of the obturation, which is particularly beneficial in cases of irregular and interconnected canal systems like those found in C-shaped configurations (10). Despite these advantages, concerns have been raised regarding the thermal risks associated with WVC. The temperature rise generated during the procedure can inadvertently affect adjacent periodontal structures, such as the periodontal ligament (PDL), alveolar bone, and cementum, potentially leading to thermal injury (10, 11).Excessive heat transfer, especially in C-shaped canals with thinner dentin walls, may cause irreversible damage including local necrosis, bone resorption, and ankylosis (12).

Studies comparing different obturation systems have shown that the Elements Free system induces significantly less heat transfer and thermal flux during the down-pack phase than the Calamus Dual

system, emphasizing the impact of remaining dentin thickness on heat conduction to periradicular tissues (13).

To understand the stress and thermal dynamics in dental tissues, various methods such as finite element analysis (FEA), two-dimensional/three-dimensional photoelasticity, strain gauges, and digital moiré interferometry have been employed (14). For instance, Jacubinek et al. (15) used FEA to study the impact of light intensity, curing mode, and thermal barriers on intrapulpal temperature, revealing that light intensity and curing mode were more influential than other variables. Additionally, Er et al. (16) evaluated PDL temperature during warm GP obturation using finite element modeling, further demonstrating the relevance of this method for thermal risk assessment.

Given these findings, it is crucial to assess the limitations and potential hazards of WVC, particularly in C-shaped root systems. This study aims to analyze the thermal behavior during the WVC technique, considering the influence of root canal anatomy, device calibration, and the possibility of a ≥10 °C increase in temperature that could jeopardize the vascular integrity of the PDL.

Although WVC has shown superior obturation quality in C-shaped mandibular molars, as confirmed by micro-computed tomography (micro-CT) studies, its thermal safety profile remains a topic requiring deeper investigation.

WVC technique is widely used in C-shaped canals, yet its thermal behavior in these anatomies has not been adequately evaluated. Most earlier thermal studies relied on single-rooted teeth (16), did not account for periodontal perfusion even though blood flow influences heat dissipation, and used simplified bone–PDL geometries that do not represent the thin dentinal walls of C-shaped molars. Device temperature settings in prior work also reflect internal resistor values rather than true operative temperatures. Because rapid heat transfer through thin dentin may risk periodontal injury, a more anatomically realistic assessment is needed. This study therefore employed finite element modeling

of a micro-CT–derived C-shaped molar with simulated PDL perfusion to evaluate temperature transmission under different warm obturation settings.

METHOD

Ethical committee approval (Approval No. 2025/167) was obtained from the Clinical Research Ethics Committee of Istanbul University Faculty of Dentistry, and the study was conducted in accordance with the Declaration of Helsinki.

Establishment of an FEA Model

An access cavity was prepared, and the working length was determined using a #15 K-file. Root canal preparation was performed up to X3 using the ProTaper Next rotary system (DENTSPLY SIRONA Inc., NY, USA), following the manufacturer's instructions. The tooth was then scanned using micro-computed tomography (Skyscan 1275, Bruker Corp., Massachusetts, USA) at a resolution of 8 µm × 8 µm, generating a standard triangle language (STL) file.

Autodesk Fusion 360 (Autodesk Inc., California, USA) was used to model the tooth structures and surrounding tissues based on the STL data (Figure 1). A 250 µm-thick layer was constructed around the root, extending from the dentinoenamel junction to the apical foramen, to represent the PDL. To simulate blood flow in the periodontium, 20 vessels with a diameter of 100 µm were evenly distributed within the center of the PDL.

To design the surrounding bone, a cone beam computed tomography scan, previously obtained for unrelated dental treatment, was processed using In Vesalius 3.1 (open-source software) and exported as an STL file. This file was used to reconstruct a bone model of the relevant region. The perfusion rate in the simulated PDL was assumed to be equivalent to that of human gingival tissue (150 perfusion units (17)). Thermal parameters for the biological structures and dental materials were

obtained from the literature and are summarized in Table 1 (16, 18-21). Room temperature was assumed to be 22 °C (16, 18).

The simulation of temperature changes in the PDL during root canal filling with the continuous wave vertical compaction technique was carried out using Autodesk CFD 2024 (Autodesk Inc., California, USA).

Meshing, Nodes, and Elements

The geometrical models described above were subjected to a mesh quality assessment to ensure element shape integrity and numerical stability, followed by a convergence test for thermal distribution. The convergence criterion was defined based on variations in temperature and thermal flux, with a maximum permissible difference of 5% between successive iterations. At the interfaces between different materials, ideal thermal contact resistance was assumed, and all materials were considered isotropic, homogeneous, and perfectly bonded. The finite element meshes consisted of three-dimensional tetrahedral elements, with the total number of elements ranging from 59,252 (V/3 model) to 89,817 (V/4.5 model), and the total number of nodes ranging from 283,261 (V/3 model) to 431,980 (V/4.5 model).

Simulating the Obturation Process

The obturation process was simulated using Autodesk CFD 2024. A matched GP cone was positioned 0.5 mm short of the working length. To account for heat loss, the hand plugger's temperature was assumed to be 22 °C, equal to the ambient temperature. No sealer was included in this simulation.

The heat source used for obturation was the Eighteeth Fast Pack Pro (Changzhou Sifary Medical Technology Co., Ltd., Changzhou, China), which was applied to sear off the 3 mm apical portion of the GP cone. The plugger temperature was set to 250 °C for Model 1 and 90 °C for Model 2.

In Model 1, the plugger at 250 °C was applied to the apical GP from 0 to 3 seconds. Between 3 and 6 seconds, the device cooled to 115 °C. It was then reheated to 250 °C between 6 and 6.5 seconds to

reactivate it and remove it from the root canal. The device remained on the GP until the 7th second and was then withdrawn. From the 7th to the 12th second, a plugger at 22 °C was applied to the GP to simulate vertical compaction. The temperatures applied during the downpack process are illustrated in Figure 2 (22). The remaining canal space was then backfilled using the Eighteeth Fast Fill (Changzhou Sifary Medical Technology Co., Ltd., Changzhou, China) in approximately four equal increments, as shown in Figure 3. For this device, the warm GP temperature was set at 200 °C.

In Model 2, a plugger at 90 °C was applied to the apical GP from 0 to 3 seconds, followed by a drop to 65 °C between 3 and 6 seconds. It was reheated to 90 °C between 6 and 6.5 seconds and held in place until the 7th second before removal. As in Model 1, vertical compaction was simulated from the 7th to the 12th second using a 22 °C plugger. The remaining canal was then backfilled in four increments using the same technique, with the warm GP temperature set at 100 °C.

Temperature Measurements

In all models, temperature changes at the interface between the root surface and the PDL were continuously monitored. The time point and location of the maximum temperature rise were recorded. The initial measurement time was set at the moment of contact between the hot plugger (or heat source) and the apical GP.

RESULTS

Representative color plots depicting the temperature distribution across the PDL in Model 1 are presented in Figure 4, which also displays the peak PDL temperature recorded during the obturation process. Higher temperature settings in Model 1 resulted in a more pronounced temperature increase within the PDL.

During the down pack phase, the maximum PDL temperatures reached 66.673°C at 3.3 seconds in Model 1 and 44.366°C at 3.4 seconds in Model 2. In the backfill phase, the highest recorded

temperatures were 53.307°C at 23.44 seconds (Model 1) and 43.115°C at 23.45 seconds (Model 2). Figure 5 further illustrates the peak temperature values observed in the PDL of both models throughout the obturation process.

DISCUSSION

Root canal obturation is a critical phase in endodontic treatment, and preserving the integrity of the PDL during heat-assisted obturation is of utmost importance (23). GP remains the most commonly used core material in obturation, and while thermoplastic techniques enhance its flow and adaptation, there is ongoing concern regarding excessive heat generation. Heat transfer beyond physiological limits may inadvertently damage periodontal tissues, emphasizing the need to understand the thermal boundaries of these techniques (24, 25).

Early in vitro studies by Matthews and Hirsch (26) and Rhinelander et al. (27), demonstrated that bone alkaline phosphatase becomes inactivated at 56 °C, a threshold temperature considered critical for bone damage. Eriksson et al. (28) further confirmed that even a brief exposure (1 minute) to 53 °C may lead to irreversible injury to bone tissue. The extent of heat transfer to periradicular tissues depends on dentin thickness and its thermal properties. Due to dentin's low thermal conductivity, it serves as an effective barrier, limiting heat propagation to the PDL, which itself dissipates some of the energy (29).

According to Er et al. (16) , although GP reaches high temperatures (e.g., 200 °C) during compaction, the adjacent tissues experience only a modest rise in temperature, approximately 7.5 °C, which remains within physiological limits. The peak rise occurred about 1.265 seconds after application, and no observable damage to tissues was reported (16).

Thermal risks associated with thermoplastic obturation have long been recognized. To prevent injury, a temperature rise exceeding 10 °C on the root surface is considered unsafe (28). Although dentin

acts as an insulator, several studies have recorded external root surface temperature elevations during obturation. For instance, thermomechanical compaction has been shown to raise temperatures up to 27 °C in certain root regions. With Touch ‘n Heat, a nearly 10 °C increase was documented (30). Sweatman et al. (31) reported that System B (200 °C) caused peak temperature increases 6 mm from the apex, especially in mandibular central incisors with thin dentin, reaching nearly 10 °C (8).

Blood flow plays a crucial role in dissipating heat and maintaining tissue homeostasis. However, Ulusoy et al. (32) noted that FEA models typically omit the effect of blood circulation. Due to experimental challenges, ex vivo studies have not effectively incorporated the cooling influence of PDL perfusion (33).

Cen et al. (34) addressed this gap by simulating PDL blood flow and measuring surface temperature post-obturation. Their findings showed that both the presence of a PDL layer and active blood circulation significantly reduced temperature rise, regardless of obturation method or canal type (34). The PDL receives blood from interseptal arteries via branching networks encircling the root (35). However, their model had limitations related to undefined vessel entry/exit points and variable vessel geometry, prompting the design of an improved model in the present study, featuring defined boundaries, constant cross-sections, and a uniform three-dimensional vascular network around the root surface (33).

Monitoring temperature fluctuations during compaction revealed that elevated temperatures rarely persisted beyond five seconds. According to Floren et al. (36), brief thermal exposure under one second is unlikely to cause tissue harm. Future research may benefit from quantifying duration above threshold temperatures across various obturation protocols. Overall, a PDL layer helps attenuate peak temperatures in the alveolar bone, and blood circulation within the PDL enhances this effect. Thus, the role of PDL perfusion should be considered essential in heat transfer modeling during thermoplastic obturation (37).

Venturi et al. (38) found that System B generated only a slight increase (0.5–0.9 °C) in apical GP temperature. Buchanan (39) supported this by noting that continuous wave compaction often results in apical compaction near body temperature, attributed to gutta-percha's low thermal conductivity roughly one-fourth that of dentin. Furthermore, the depth of heat plugger insertion plays a key role in GP plasticity. Deeper penetration enhances adaptation and reproduction of canal irregularities (40, 41).

Although it is known that intracanal heat may reach external root surfaces, the specific thermal effects of warm GP on the PDL remain underexplored. Nonetheless, several studies have shown that the System B technique causes minimal thermal insult to periradicular tissues (42, 43).

The "danger zone" in multirooted teeth is characterized by thinner dentinal walls and a heightened susceptibility to both mechanical stress and thermal injury, features that cannot be adequately assessed in single-rooted tooth models. Furthermore, dentin thickness varies across different roots, influencing the pattern of heat distribution. Previous studies have established an inverse correlation between residual dentin thickness and temperature increase during obturation procedures (44). Zhou et al. (18) similarly identified a high-risk zone where elevated temperatures were consistently recorded. Notably, discrepancies in findings across studies may also stem from variations in temperature measurement methodologies. Temperature levels have been reported to reach approximately 47 °C within just three seconds of heat activation. Given the clinical difficulty in controlling such rapid increases, it is strongly advised that heat application during continuous wave compaction 0not exceed three seconds. To achieve optimal apical GP adaptation without compromising periodontal structures, further investigation is required to refine obturation protocols, including variables such as plugger penetration depth and obturation material type. Our current FEA-based findings indicate that even with a conservative 3-second activation, thermal damage may still occur in the PDL adjacent to the "danger zone" (18).

This study employed two simulation models to analyze thermoplastic obturation. In Model 1, a temperature of 250 °C was applied to the apical GP from 0 to 3 seconds. Between the 3rd and 6th second, the device cooled to 115 °C. It was reactivated at 250 °C from 6 to 6.5 seconds, maintained on the GP until the 7th second, and then removed. A plugger at room temperature (22 °C) was applied from the 7th to the 12th second to simulate vertical compaction. For the Eighteeth Fast Fill unit, the GP temperature was preset to 200 °C.

In Model 2, 90 °C was applied using the Eighteeth Fast Pack Pro from 0 to 3 seconds, then cooled to 65 °C between the 3rd and 6th second and reheated to 90 °C from 6 to 6.5 seconds for device removal. A 22 °C plugger was again used from 7 to 12 seconds. The Fast Fill unit temperature was set at 100 °C in this model.

While previous FEA studies have primarily utilized single-rooted teeth or two-rooted mandibular molars, our investigation uniquely employed a C-shaped mandibular molar, an anatomy often better filled using thermoplastic techniques. Although our study included vascular structures, unlike the study by Zhou et al. (18), a similar thermal “danger zone” was detected in the concave root region. Furthermore, while other studies have typically relied on cylindrical bone models for simplification, our simulation utilized cone beam computed tomography images from an actual patient to create a more anatomically accurate 3D bone model (16, 18, 34).

Zhou et al. (18) compared varying activation times of the System B device, and Cen et al. (18) used the Touch & Heat device to assess the influence of blood perfusion on FEA outcomes. To the best of our knowledge, this is the first FEA-based study evaluating the thermal effects of different device temperature settings on biologically vascularized periradicular tissues, aiming to inform safer and more effective clinical strategies.

Consistent with prior FEA studies involving thermoplastic GP, the cementum layer was excluded from the current model due to its indistinguishability from dentin in micro-CT imaging (18, 34). In

contrast, Er et al. (16) included cementum in their simulations but assumed its thermal properties were equivalent to dentin's, given the lack of specific data.

This finite element model required several simplifications. Only one C-shaped mandibular molar was simulated, and although this tooth was reconstructed from micro-CT data, the findings reflect the thermal response of a single morphology rather than the full spectrum of C-shaped configurations. The PDL and its vascular network were idealized. Vessel distribution, number, and diameter were standardized, and perfusion was represented as a constant heat-dissipating field rather than a dynamic circulatory system, despite the known influence of blood flow on thermal regulation (33, 34, 35). Moisture content in dentin and other environmental factors that affect heat conduction were not incorporated.

Cementum was not modeled as a separate layer due to its indistinguishability in micro-CT images, consistent with several previous FEA studies (18, 34). Root canal sealers were also excluded. This aligns with earlier work where sealer-related thermal properties were either unavailable or insufficiently reliable for modeling (45, 46). As a result, the potential insulating or heat-distributing effects of sealers were not represented.

Thermal inputs were based on manufacturer-displayed temperature settings, even though these values primarily reflect internal resistor temperatures rather than the actual heat delivered at the plugger tip or extruded GP (34, 47). This limitation applies to most published thermal simulations because empirically calibrated operative temperatures are rarely available from manufacturers. The plugger insertion depth, contact pressure, and movement were standardized for reproducibility, although clinical variability in these factors is expected.

Mesh resolution was convergence-tested but still represents an approximation of complex regions such as the concave root surface where dentin is thin. The thermal injury threshold used in this study

relied on literature-based temperature values (28) and did not incorporate time-dependent thermobiological response models.

Finally, no direct experimental validation was performed using thermocouples or in vitro measurements. Although FEA offers ethical, practical, and spatial advantages over ex vivo testing (38, 45, 48), the simulation cannot fully reproduce in vivo thermal behavior. Consequently, the present results should be interpreted as controlled, worst case thermal predictions rather than precise replications of clinical conditions.

CONCLUSION

Based on the assumptions and limitations inherent in this study's finite element model, it was observed that setting the down-pack device to 250 °C and the backfill device to 200 °C results in a temperature rise at the PDL surface, particularly adjacent to the concave root region, that exceeds the critical threshold of 47 °C. In contrast, temperature settings of 90 °C for the down-pack device and 100 °C for the backfill device did not lead to thermal levels surpassing this threshold.

In models incorporating simulated blood flow within the PDL, high-temperature obturation protocols were found to pose potential risks to the vitality of surrounding periodontal structures. Conversely, low-temperature settings appeared to preserve thermal safety and minimize the likelihood of tissue damage.

From a clinical standpoint, adopting lower temperature settings during continuous wave obturation is advisable to maintain the integrity and health of periradicular tissues. Furthermore, clinicians should be aware that not only the temperature itself, but also the duration of heat application may significantly contribute to the cumulative thermal load within the PDL.

Declarations

Abbreviations

WVC Warm Vertical Compaction

GP Gutta-percha

PDL Periodontal Ligament

FEA Finite Element Analysis

Micro-CT Micro-computed Tomography

STL Standard Triangle Language

Ethics approval and consent to participate

The study adhered to the tenets of the 1964 Helsinki Declaration and its later amendments. The Clinical Research Ethics Committee of Istanbul University Faculty of Dentistry revised and approved the study protocol (approval number: 2025/167). Informed consent was obtained from all subjects involved in the study.

Consent for publication

Not applicable

Availability for data and materials

The datasets used and/or analysed during the current study are available from the corresponding author on reasonable request.

Competing interests

The authors declare that they have no competing interests

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Authors' Contributions

TYY and SSG conceived and designed the study. TYY and AK conducted the experimental procedures and collected the data. TYY and AK performed the data analysis. TYY drafted the manuscript. All authors reviewed, revised, and approved the final version of the manuscript.

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TABLES

Table 1: Thermal parameters of dental materials used in the study

Components	Density (kg/m^3)	Specific heat (J/kg-K)	Thermal conductivity (W/m-K)
Enamel	2800	710	0,93
Dentin	1960	1590	0,58
Blood flow	1100	3470	0,50
PDL	1100	4820	0,58
Alveolar bone	1300	1840	0,58
Gutta-percha	970	940	0,15

FIGURES

Figure 1. A: Enamel, B: Dentin, C: Periodontal Ligament, D: Bone, E: Complete Model.

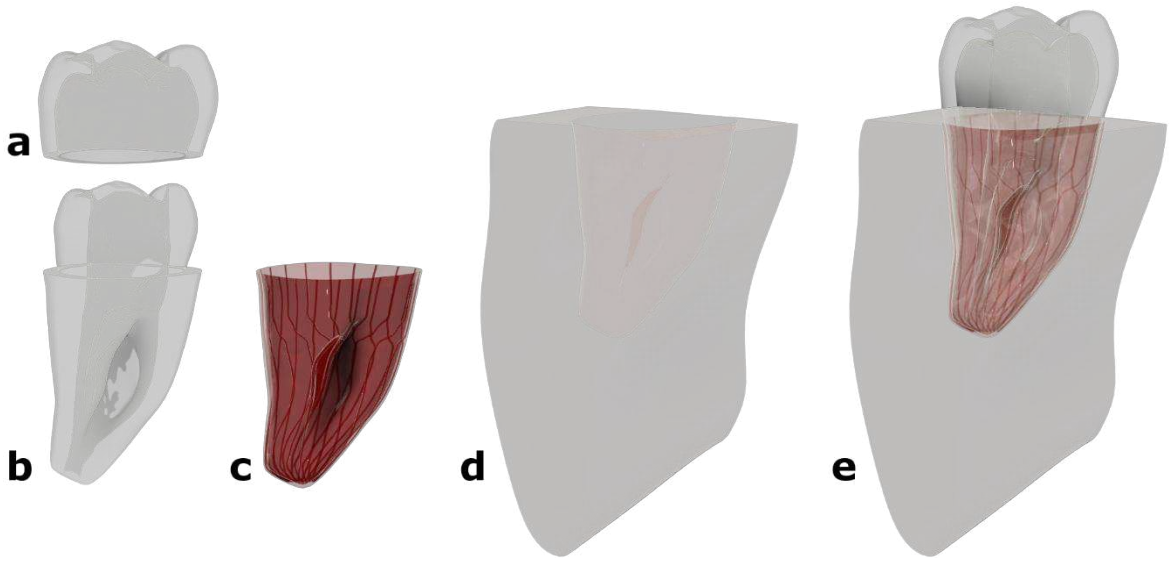


Figure 2: Temperatures applied to the apical gutta-percha surfaces during the down pack process

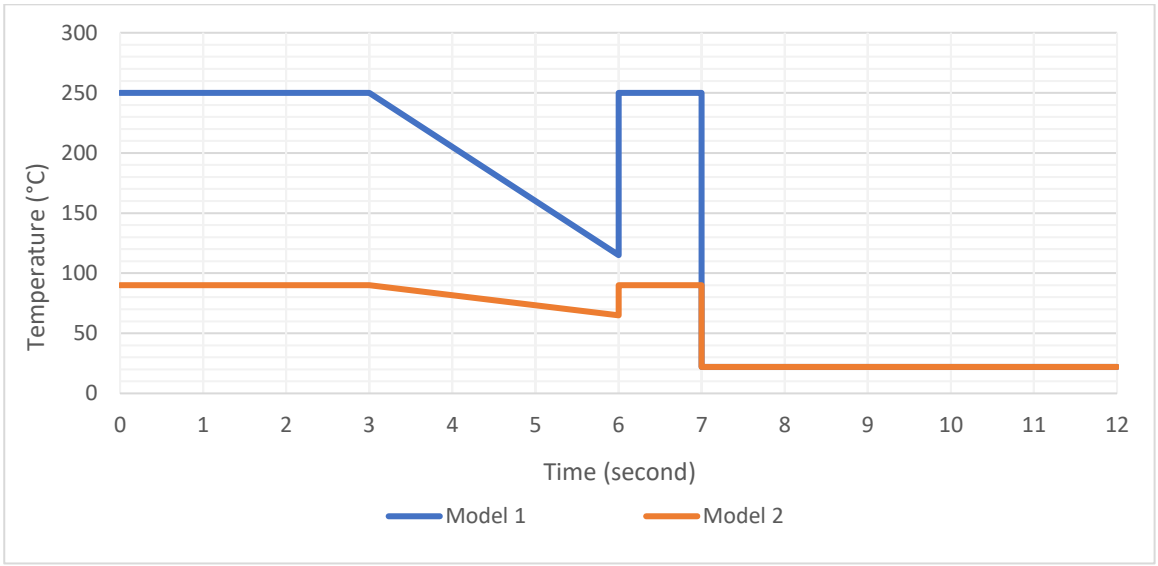


Figure 3: Gutta-percha pieces



Figure 4. A-B: Highest temperature regions during downpack process on model 1, C: Highest temperature region during Backfill 1 on model 1, D: Highest temperature region during Backfill 2 on model 1.

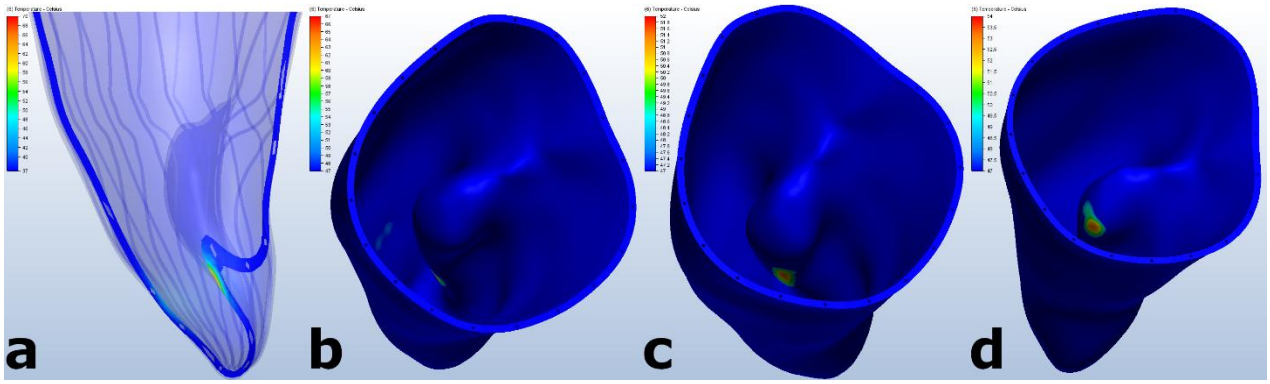


Figure 5: Highest temperatures on both models' periodontal ligaments during obturation process by seconds

