Occlusal Loading Effect on Stress Distribution of Endodontically Treated Teeth: Finite Element Analysis Study

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ABSTRACT

Aim: Evaluate the influence of occlusal loading on the stress distribution of endodontically treated teeth after root canal preparation with different file's sizes and tapers by means of finite element analysis. Methodology: Seven three-dimensional models of a single-rooted, single-canal lower second premolar were established, one healthy control and six endodontically treated and restored models. The shape of root canal preparations followed file configurations 30/.05, 30/.09, 35/.04, 35/.06, 40/.04, and 40/.06. Von-Mises equivalent stresses were calculated by applying 30 N, 90 N and 270 N loads to the buccal cusp tip, each one at 90º, 45º and 20º angles from the occlusal plane simulating occlusion, dental interference and laterality, respectively. Results: 45º loading was more prone to formation of higher stress values. The simulation of occlusion and laterality resulted in maximum stress areas located at the inner side of the root curvature, while under occlusal interference they were on the lingual surface over the tooth's long axis. Conclusions: The angulation of occlusal loading and magnitude were determinants for stress distribution on dental structure. Both variations of size and taper were not determinants for the increase in the maximum stress areas.

INTRODUCTION

Nickel-titanium (NiTi) rotary files have become an important adjunct in endodontic therapy due to their easier and faster preparation, thus reducing operator's fatigue and patient's discomfort.[1,2] However, the overpreparation can result in dentin weakening, defects and complications such as apical transportation, ledge formation, perforation, loss of working length, craze lines, cracks, and vertical root fractures.[1–5]

Despite of the high success rates of endodontic therapy, the decrease in stiffness and fracture resistance of endodontically treated teeth can be related to the loss of structural integrity associated with caries, trauma, and extensive cavity preparation.[6] Likewise, during functional mandibular movements the teeth are subjected to forces that influence the maintenance of their position and integrity.[7,8] Many different factors may affect occlusal loading (OL) such as craniofacial morphology, age, race, gender, muscle size and thickness, temporomandibular disorders, periodontal support of teeth, posture of the subject's head, occlusal contact area and location of the measurement on the dentition.[9] In order to maintain balance within periodontal structures, occlusal forces must be directed at a tooth's long axis.[10] Oblique forces cause higher strain concentrations, resulting in tensile, compressive and shear areas which can suffer plastic deformation.[11,12] Thus, the stress distribution (SD) in teeth is an important factor to the overall success of dental treatments.[10]

The interaction between OL and endodontic procedures may determine a tooth's susceptibility to fracture.[10] Finite element analysis (FEA) is a very efficient and reliable method used to simulate clinical conditions in many different applications of dentistry.[7,8,11–15] It has been successfully applied in endodontics for the prediction of fracture patterns and evaluation of SD under OL.[16,17] Therefore, this research aims to elucidate the influence of OL at different loads and angles on the SD of endodontically treated teeth after root canal preparation with different file's sizes and tapers by means of FEA.

METHODS

THREE-DIMENSIONAL FINITE ELEMENT MODELING PROTOCOL

A generic three-dimensional (3D) model (https://sketchfab.com/)[18] of a single-rooted, single-canal lower second premolar was adapted in this study. Average values of tooth's structure were based on previous literature and adapted using Computer Aided Design (CAD) software SpaceClaim (SpaceClaim Corporation, Concord, MA, USA): the tooth was 22 mm length,[19] the apical constriction was established at 1 mm from the apex with a diameter of 0.26 mm,[4] and the periodontal ligament thickness of 0.2 mm was homogeneous over the root surface.[13] Also, mechanical properties of the tooth *(Tables 1 and 2)*[15] were applied using Ansys 19 (Ansys Inc., Canonsburg, PA, USA) software. The surrounding bone and periodontal ligament were not represented as a 3D solid as the premolar. Instead, the boundary conditions were assumed as Elastic Support and the constant applied to this contact (foundation stiffness) were approximate by the Periodontal Ligament's Young Module. The Elastic Support was applied over all the external region of dentin (immediately above the enamel external surface). The enamel and dentin were considered orthotropic while pulp and periodontal ligament were isotropic.

ROOT CANAL PREPARATION

Six other models were created with the same basic anatomy of the control model (sound). Each model had its main canal shape matching the taper and tip diameter of the respectively selected file *(Figure 1)*. The files selected for the current study had sizes #30, #35 and #40, with two different tapers for each one. Therefore, we used #30/.05 MTWO (VDW, Munich, Germany) and #30/.09 Protaper Universal (Dentsply Tulsa Dental Specialties, Tulsa, OK, USA); #35/.04 MTWO and #35/.06 WaveOne Gold (Denstply Maillefer, Ballaigues, Switzerland); #40/.04 MTWO and #40/.06 Reciproc Blue (Vdw, Munich, Germany).

Table 1. Young's modulus (MPa) and Poisson's ratio of isotropic structures.

Structures	Young's modulus (MPa)	Poisson's ratio
Pulp	2,07	0,45
Periodontal ligament	68,9	0,45
Cortical bone	13700	0,30

Table 2. Young's modulus (MPa), Shear's coefficient (MPa) and Poisson's ratio of orthotropic structures.

Structures	Longitudinal (L)	Transversal (T)	Z
	Young's modulus (MPa)		
Enamel	73720	63270	63270
Dentin	17070	5610	5610
	Shear's coefficient (MPa)		
Enamel	20890	24070	20890
Dentin	1700	6000	1700
	Poisson's ratio		
Enamel	0,23	0,45	0,23
Dentin	0,30	0,33	0,30

SEALING AND RESTORATIVE MATERIALS

The mechanical properties of gutta-percha were inserted within the length between the apical constriction and cementoenamel junction (CEJ) (Young's modulus: 0.69 MPa and Poisson's coefficient: 0.45),[20] while Filtek Z250 (3M/ESPE, St Paul, MN, USA) was located from the CEJ to the occlusal surface (Young's modulus: 187000 MPa and Poisson's coefficient: 0,308).[21,22]

FINITE ELEMENT ANALYSIS

In all seven 3D models the Mesh was obtained by the Ansys mesh generator using an overall element order of 0.5 mm and forced a tetrahedron element due to the geometry complexity. Additionally, was defined a refinement in the dentin using tetrahedron even smaller (in order of 0.2 mm) to capture the high stress gradient in the bottom curvature of the model. Due to the dentin region refinement, the overall quality of the mesh was improved resulting in an average skewness of 0.22 (representing low distortions in the tetrahedrons). Altogether, the mean of elements and number of nodes were 169,544 and 244,173, respectively. The boundary condition consisted of elastic support

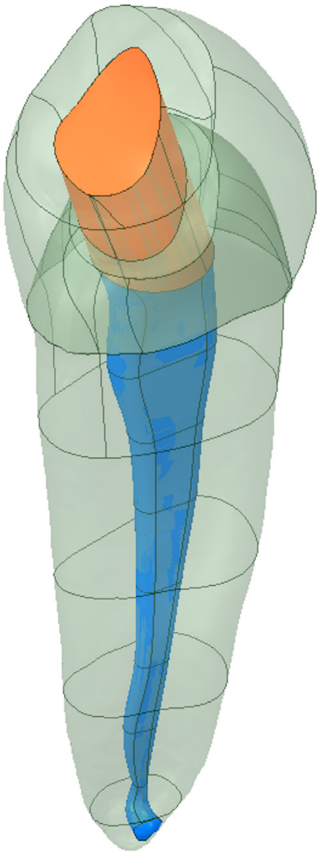

Figure 1: Three-dimensional (3D) finite element model of the lower second premolar with root canal preparation.

fixation offered by Ansys 19 software in order to reproduce a realistic clinical condition by simulating the physiological tooth mobility through the periodontal ligament. All the models were analyzed in terms of static structural (all the forces and moments equal to zero) in a linear elastic study using constant properties of the tooth's materials describe previously. Von Mises equivalent stresses were calculated by applying 30 N, 90 N and 270 N loads to the flat buccal surface of the buccal cusp tip on all models. Each load was applied over three different angles from the occlusal plane in order to simulate conditions of occlusion (90º), occlusal interference (45º), and jaw laterality (20º).

RESULTS

Overall, sixty three FEA simulations were performed. The numerical variations of maximum resultant forces at different angles are represented in Table 3. We observed different distributions of maximum stress from each loading angulation. For the 30 N load, we found a greater maximum SD at a 45º angle, while smaller stress areas were generated on the root surface at a 20º angle (45º> 90º> 20º) *(Figure 2)*. The results for the 90 N load were similar to 30 N, represented by 45º> 90º> 20º *(Figure 3)*. However, at 270 N loading, we observed a decrease in the maximum stress area at 90º angle and an increase at 20º, while at 45º the SD was similar to the 30 N loading (45º> 20º> 90º) *(Figure 4)*.

We also found patterns of maximum tension locations from OL at different angulations. With 20º and 90º angulations the maximum stress was generated at the internal curvature area (on the distal surface) while at the 45º angle it was located on the lingual root surface along the long axis of the tooth.

Ultimately, forces applied in different angles were evaluated to determine the influence of dentin loss in the maximum stress area formation. Both variations of size and taper were not determinants for the increase in the maximum stress area. However, 45º angle generated higher stress values in all loads applied.

DISCUSSION

Based on the results found in this study, greater sizes and taper of instruments did not produce an increase of generated stress in the dental element, as also reported by Adorno *et al.*[23] Many studies have shown a higher susceptibility to fracture of endodontically treated teeth because of enlargement and shaping of canals.[3,4,22] These authors claim that the excessive dentin removal using different NiTi instrument designs, such as greater taper, sizes and cross-section, could weaken it and cause dentinal defects. For this reason it has been assumed that vertical fractures in endodontically treated teeth may not

Table 3. Maximum stress values (MPa).

Loads	Angles	Control	30/.05	30/.09	35/.04	35/.06	40/.04	40/.06	Mean
30N	20°	19,82	19,13	19,19	19,28	19,13	19,20	19,21	19,28
	45°	26,49	26,47	70,58	26,05	26,07	26,50	26,89	32,72
	90°	23,42	23,39	23,41	23,43	23,52	23,46	23,57	23,46
90N	20°	59,44	57,40	57,56	57,84	57,39	57,59	57,62	57,83
	45°	116,13	152,33	127,68	136,61	140,86	97,57	126,77	128,28
	90°	70,26	70,18	70,23	70,30	70,56	70,38	70,70	70,37
270N	20°	178,32	172,20	172,68	173,50	172,17	172,76	172,86	173,50
	45°	348,39	457,00	383,03	409,83	422,58	380,32	292,71	384,84
	90°	210,78	210,53	210,70	210,89	211,68	211,13	212,11	211,12

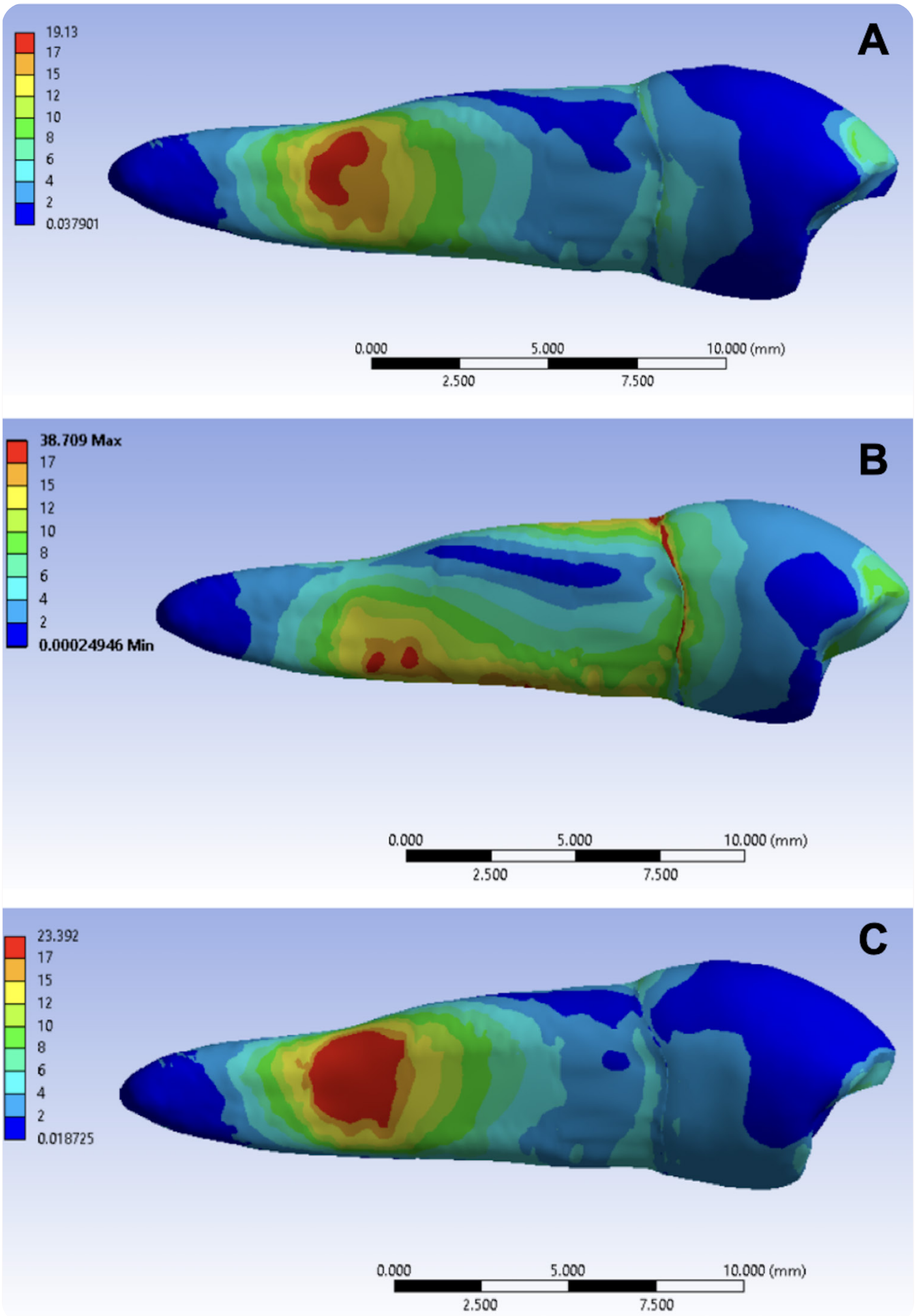

Figure 2: Maximum stress distribution at 30 N load: A- 20º, B- 45º and C- 90º.

occur immediately after root canal instrumentation, since they can be diagnosed a few years after finishing endodontic and restorative treatments.[3] Contrarily, Munari *et al.*[24] found that increasing the diameter of the root canal may decrease the maximum tensile stress. Therefore, fracture susceptibility may not be only linked to dentin thickness by the amount of dentin removed, but also root morphology, main canal shape and placement of posts.[16,25]

Occlusal interferences are defined as any tooth contact that inhibits the remaining occluding surfaces from achieving stable and harmonious contacts.[26] In this study, oblique loads showed higher prevalence of areas of maximum tensile stress and greater distribution compared to vertical loads.[7,11] We also found that models loaded at 20º required greater forces to generate areas of highest stress, while at 45º the lowest load (30 N) was able to produce maximum stress on the root surface. The study of Eken *et al.*[11] showed similar results, in which oblique loads caused greater areas of stress compared to vertical loading. It was also found by Tanaka *et al.*[12] that forces at 20º only caused plastic deformation of dental elements at 55 N/mm, while at 45º plastic deformation occurred with a load of 8.6 N/mm. Then, we assume that loads from occlusal interferences may generate compression and tensile stress in the dental structure that increase maximum tensile areas, which are considered as the initial point of dental cracks.[16,17] Our

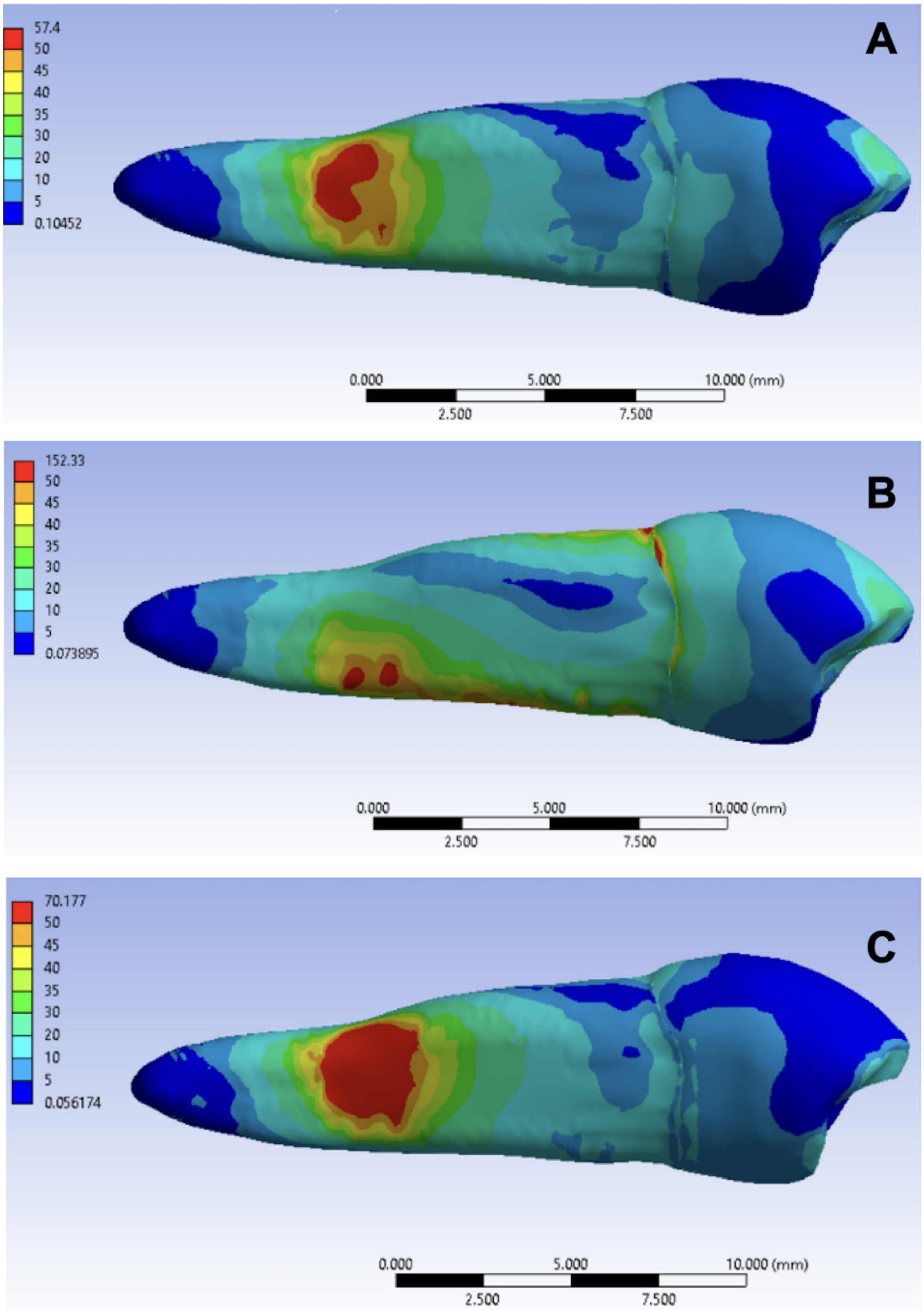

Figure 3: Maximum stress distribution at 90 N load: A- 20º, B- 45º and C- 90º.

results showed that the maximum stresses generated at 20º and 90º loading were located in the inner side of the root curvature, while at 45º loading they were in the lingual surface. These findings may suggest tendencies to horizontal fractures under both laterality and vertical loading, and to vertical fractures under occlusal interference.

FEA is a computational method used to quantify the effects of external forces on objects and predict fracture patterns.[13,16] One of its main advantages is the possibility to evaluate and improve the mechanical behavior of complex structural problems in dentistry by simulating clinical conditions, which would be difficult to do with laboratory or in vivo studies.[16,27]

Another benefit of FEA is that it allows successive testing within the same model without changing the original one.[15] However, it is not possible to incorporate small structural defects as cracks or canal irregularities in finite element modeling technique, thus describing one of our model's limitations. Defects within the dentin may affect its tensile strength, which could have a great influence on the generation and propagation of cracks.[16]

The boundary conditions are the restrictions applied over the external surface of the object in order to represent its degree of freedom.[15] In this research, elastic support fixation was applied over the entire root surface to reproduce the physiological mobility provided by the periodontal ligament.

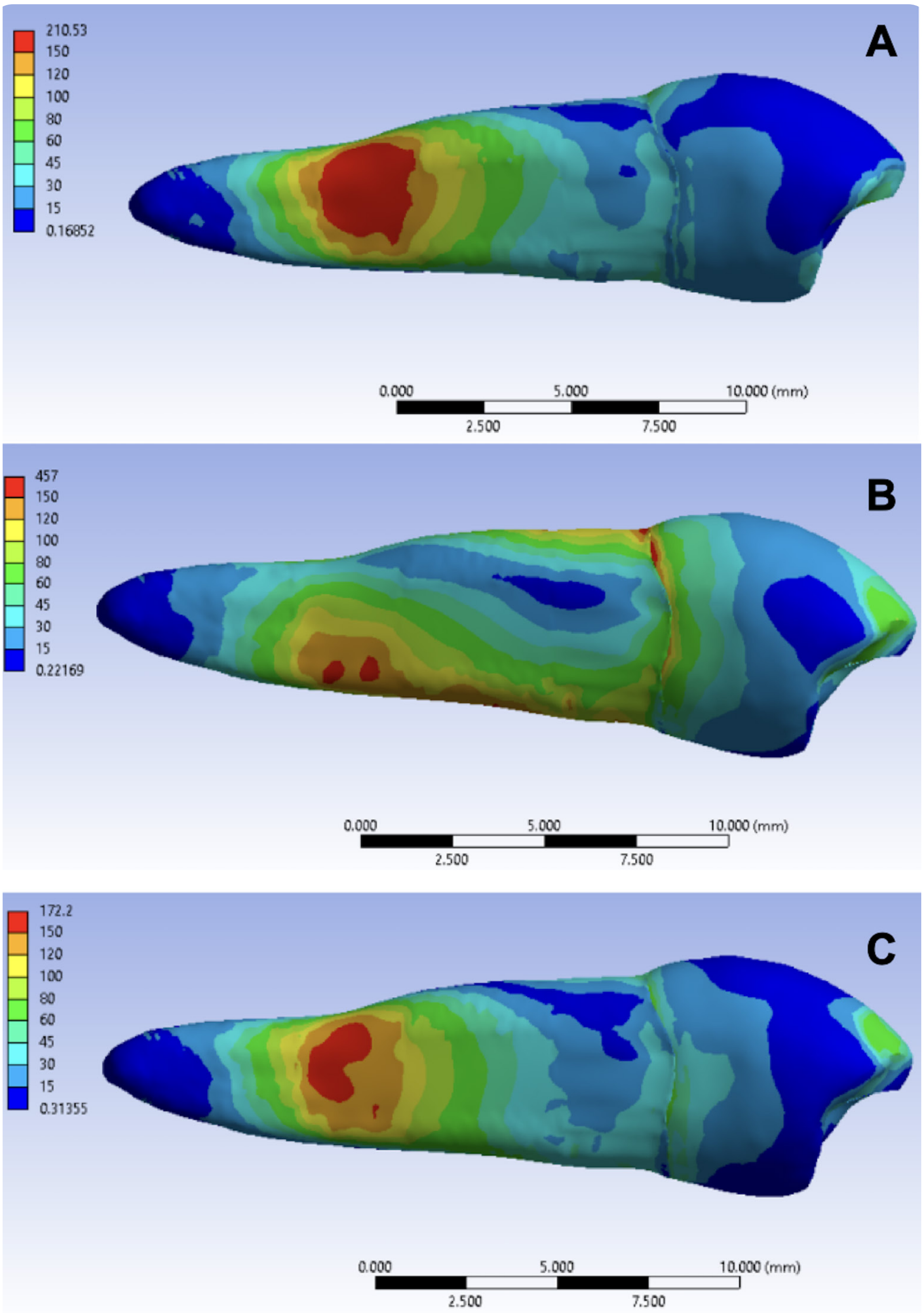

Figure 4: Maximum stress distribution at 270 N load: A- 20º, B- 45º and C- 90º.

Differently from our model, Eken *et al.*[11] assumed the boundary condition as fixed, while Du *et al.*[20] and Gomes de Oliveira *et al.*[8] disregarded the periodontal ligament in their model. In addition, most studies considered dentin as isotropic and uniform.[10–12,17,20,24] Since dentin areas have different properties such as Poisson's ratio and elasticity modulus, it must be considered orthotropic for more precise simulations.[16]

Within the limitations of this study, we considered constant values of 0.2 mm over the entire root surface corresponding to the periodontal ligament, which in reality may vary for each tooth type between 0.15 and 0.38 mm in cervical, medium and apical thirds.[28] Despite the known importance of a natural tooth's environment,[27] we applied the mechanical concept of elastic support fixation in order to simplify the role of the surrounding bone and periodontal ligament. Also, our study doesn't consider the influence of friction and stresses generated on canal walls during instrumentation.[3] Therefore, our results should be interpreted with caution.

For our tests we used a healthy tooth with a single low curvature degree, and only static loads were applied to the dental models. Currently, we are continuing this research *in vitro* and we intend to apply these concepts to other dental groups in the future. Thus, we suggest future similar studies to be conducted improving periodontal ligament structure, considering different levels of crown destruction or restorations, varying root curvature, and simulating cyclic loads for a more accurate model of the masticatory performance.

CONCLUSIONS

From the results of this study we found no influence of endodontic instrument's design on tooth weakening after root canal preparation. Both variations of size and taper were not determinants for the increase in the maximum stress area. Therefore, the magnitude and angulation of OL were determining factors to the SD on the dental structure. Nonetheless, other factors must be considered to influence on the fracture susceptibility of endodontically treated teeth, such as occlusal pattern, root morphology, sodium hypochlorite concentration, coronary preservation, lateral compaction force, and working length. The OL simulation of occlusion (90º) and jaw laterality (20º) resulted in maximum stress areas located in the inner side of the root curvature, while under occlusal interference (45º) they were on the lingual surface over the long axis of the tooth.

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