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Fatigue resistance of rotary endodontic files submitted to axial motion in multiplanar canals manufactured by 3D printing

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Abstract

The objective of the research herein presented was to assess the time to fracture of *Hyflex CM* files subjected to rotational bending tests, with or without back and forth motion, in an apical canal similar to those existent in a real tooth with multiplanar curvatures.

Therefore, a lower first molar was modelled and manufactured in an AISI 316L stainless steel using selective laser melting technology, featuring the required multiplanar curvatures, which were confirmed by radiography. Thereafter, endodontic files were submitted to either rotational bending tests without back and forth motion (Group A) or rotational bending tests with back and forth motion (Group B) inside an artificial root canal with primary, secondary and tertiary curvatures. Time to fracture was recorded and fragments of the endodontic files tested were observed by optical microscopy.

Instruments of Group A showed an average time to fracture of 119 seconds and an average fractured tip length of 4.9 mm, while instruments of group B displayed an average time to fracture of 194.1 seconds (+63%) and an average fractured tip length equal to 4.13 mm. All instruments fractured due to fatigue crack propagation.

Hence, it was possible to conclude that back and forth motion extended the fatigue lifetime of the files tested once allowed diminishing the number of cycles with higher stress range applied, and spread the induced stresses by an enlarged area of the instrument. Additionally, almost plane surfaces were observed at fractured cross sections of the instruments tested, allowing to infer a very low influence of torsional loading when compared with bending.

Keywords: Lower first molar (LFM); Multiplanar curvatures; Rotary endodontic files; Back and forth motion; Fatigue resistance; Selective laser melting.

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1. Introduction

Root canal morphology is commonly accepted as an important factor to consider when performing endodontic treatments. In fact, multiple curved root canals usually increase the difficulty of clinical procedures [1] and the success of a root canal debridement and shaping using rotary instrumentation is dependent on the instrument's ability to follow the original centred path of the tooth canal. Hence, flexibility is regarded as one crucial characteristic that allows the instrument to adapt to highly curved canals [2].

Current instrumentation made of Nickel-Titanium alloys (Ni-Ti) and displaying control memory effect (CM) show high flexibility, although file's fracture can still occur due to steady torsion, cyclic bending or torsional fatigue, especially when highly curved root canals are involved [3]. The lower first molars (LFM) are the type of tooth most commonly subjected to endodontic therapy [4] and conventional radiographic images show the severity of the mesiobuccal (MB) canal primary curvature; in addition, the canal is a 3D geometry with secondary and tertiary curvatures, also known as multiplanar curvatures, which are clinical challenging as they can cause file's fracture due to torsional failure and to the ever-present possibility of failure due by cyclic fatigue [1].

Several studies were made in the past years regarding the effect of root canal curvature on the fatigue life of Ni-Ti rotary instrumentation. As an example, Li et al. [5] evaluated the fatigue life of Ni-Ti files subjected to different rotation speeds for different angles of a single planar curvature. More recently, the fatigue resistance of Ni-Ti rotary instruments submitted to single and double planar curvature canals, without axial motion, was determined [5], and the experimental results confirmed that double curvature anatomies of root canal were much more stressful for Ni-Ti files than single planar curvatures. Additionally, the effect of axial motion, also known as back and forth motion, allows varying the travel distance of the endodontic file inside teeth's root canal [6].

Therefore, the objective of this study was to assess the fatigue life of Ni-Ti rotary files subjected to in-vitro rotational bending tests, with or without axial motion, in an apical canal with multiplanar curvature of a 3D stainless steel tooth model created by laser additive manufacturing technology.

2. Materials and methods

A lower first molar was manufactured through an additive technology from a CAD 3D tooth model (Fig.1), which was designed using *SolidWorks*®. For the manufacturing process, it was used the selective laser melting technology available on a SLM250 machine and the material used for fabrication was an AISI 316L stainless steel. According to the molar shape and to obtain the best surface quality, a 30 μm layer thickness manufacturing strategy with specific parameters for thin walls parts was adopted. Two X-ray images, namely bucolingual and mesiodistal views from a maxillary molar morphology study [1] were used to accurately model the artificial tooth (Figs. 1 a, b). On the bucolingual view, the canal shows a primary curvature, while a double curvature can be seen on the mesiodistal view with an upper secondary curvature and a tertiary curvature on the apical region. It's important to mention that the smallest radius of curvature, $r = 2.81$ mm, was localized in the tertiary curvature (Fig. 1c). Additionally, based on the adapted Schneider's method [7], main dimensions of the curvatures were measured from the X-ray images of the original tooth and of the modelled tooth (Table 1) (Fig. 1a, b, c), which allowed to calculate relative errors that were comprehended between 9.3 and 10.1% (Table 1). In addition, two X-ray images of the lower first molar model produced on stainless steel were taken with an industrial ANDREX Constant Potential X-Ray set (100 KV, 4,0 mA, 6 min), (Fig. 1d), respectively, allowing to confirm the similitude between the original and the tooth model fabricated just before the experimental tests had begun, in which rotational motion provided by a *WaveOne* rotary handpiece (*Dentsply*) was superimposed to axial motion allowing to simulate back and forth motion (Fig. 2a).

Therefore, as-fabricated *HyFlex CM* ref. .04/20 rotary endodontic files with tip size (diameter) equal 0.20 mm and taper equal to 0.04 were tested (Fig. 2b). All instruments were submitted to a rotational speed of 500 rpm and an input torque of 4 N.cm. Besides rotational bending loading, instruments belonging to group B were also submitted to an axial speed of 3 mm/s along a distance of 3 mm. Each test was initiated by placing the endodontic file in position (Fig.2 c) and fatigue tests were carried out until fracture occurred.

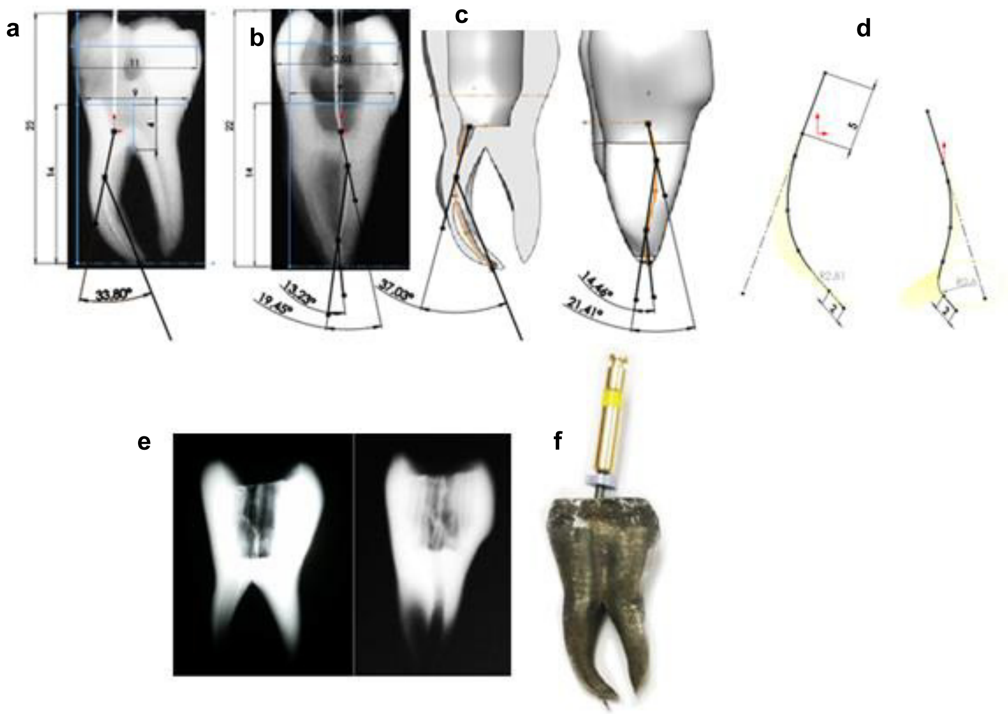

Fig. 1. (a, b) Clinical view (CV) and Proximal view (PV) of a lower first molar with primary, secondary and tertiary mesiobuccal curvatures analysed by the adapted Schneider's method; (c) (CV) and (PV) of the lower first molar - CAD model; (d) (CV) and (PV) of the lower first molar CAD model with the root path indication, as well as the point of minimum radius of curvature; (e) (CV) and (PV) radiographic images of the lower first molar produced by laser additive manufacturing technology; (f) Lower first molar model manufactured through an additive technology with an HyFlex CM rotary endodontic instrument inserted inside a root canal with multiplanar curvature.

Table 1. Measurements carried out on the mesiobuccal root canal: primary, secondary and tertiary curvatures for both the original and the CAD model: clinical and proximal views.

Lower first molar		Lower first molar: CAD model		Relative error (%)
Mesiobuccal curvatures (view)	Angle of curvature (degrees)	Mesiobuccal curvatures (view)	Angle of curvature (degrees)	
Primary (clinical)	33.8	Primary (clinical)	37.03	9.56
Secondary (proximal)	13.23	Secondary (proximal)	14.46	9.30
Tertiary (proximal)	19.45	Tertiary (proximal)	21.41	10.10

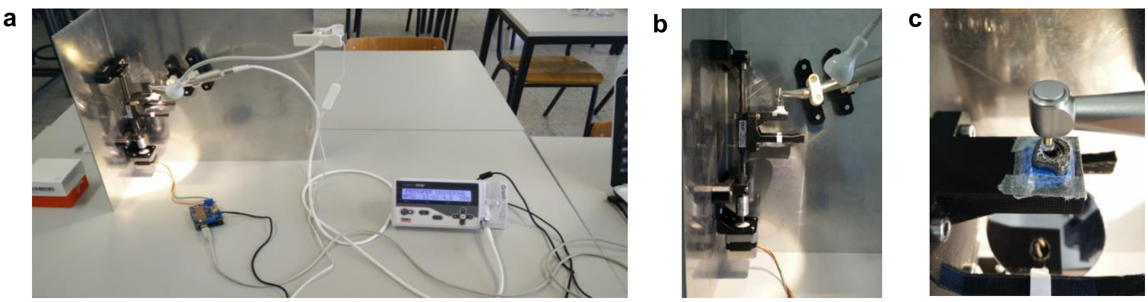

Fig. 2. (a) Overall view of the experimental *apparatus*. Axial movement was induced by a stepper motor controlled by an *Arduino* circuit board; (b) Rotational motion was provided by a *WaveOne* rotary handpiece (*Dentsply*); (c) Testing device with the artificial tooth model in-place.

Additionally, in order to calculate the induced stresses introduced by multiplanar curvatures, nonlinear finite element analyses (FEA) of *Hyflex CM* ref. .04/20 instruments were carried out using FEA software *Simulation®*. The CAD 3D model of the endodontic file under study (Fig. 3a) was created from SEM images (Fig. 3b), and the nonlinearities of the computational analyses carried out were both due to the nonlinear behaviour of the material model defined [8] and to the large geometric displacements imposed to the endodontic file simulating the multiplanar curvatures of a real tooth. As can be observed in Fig. 3c, the maximum equivalent stress induced on the endodontic file was equal to 205 MPa and it was located at a distance of about 6 mm from the instrument's tip.

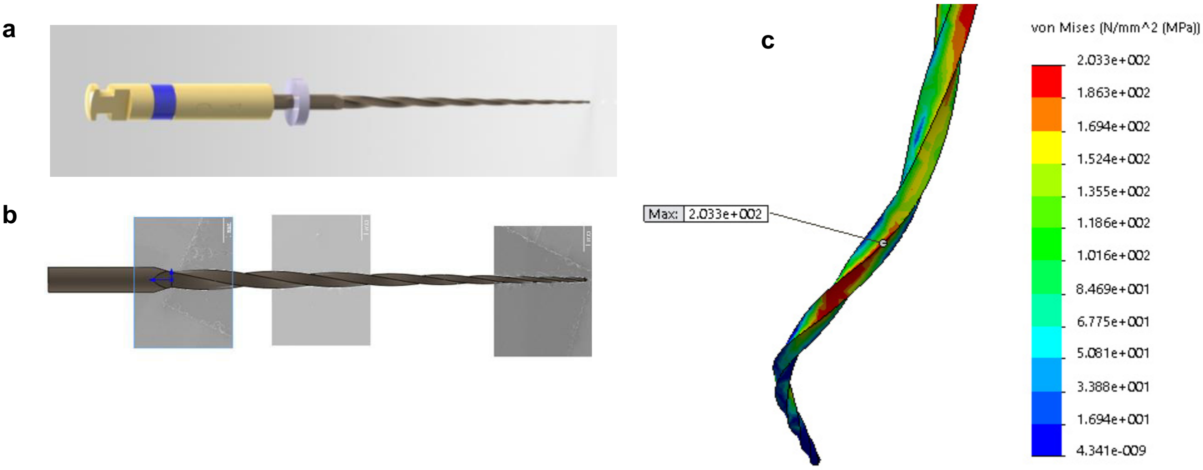

Fig. 3. (a, b) CAD 3D endodontic file ref. .04/20 modelled from SEM images; (c) Stress distribution induced at rotary instruments *Hyflex CM* ref.04/20 submitted to multiplanar curvatures.

3. Results

Experimental data from fatigue tests carried out is presented in Table 2. As can be seen, instruments from Group A, which were submitted to rotational speed, displayed an average time to fracture of 119 seconds and an average fractured tip length of 4.90 mm, while instruments belonging to group B, in which rotation and back and forth motion were superimposed, displayed an average time to fracture of 194.1 seconds and an average fractured tip length of 4.13mm.

Table 2. Experimental data from fatigue tests carried out.

Group	Spec. ref. #	Fracture region	Time until fracture [min;sec]	Average time [sec]	Standard deviation [sec]	Length of fractured file's tip [mm]	Average length [mm]	Standard deviation [mm]
A	.04/20_11	Apical	2:11	119	12	4.30	4.90	0.6
	.04/20_12		1:47			5.50		
	.04/20_2		2:57			3.30		
	.04/20_3		4:15			4.30		
	.04/20_4		4:08			4.10		
	.04/20_5		1:54			4.45		
B	.04/20_6	Apical	1:50	194.1	53.9	4.05	4.13	0.33
	.04/20_7		4:04			4.50		
	.04/20_8		2:35			4.20		
	.04/20_9		3:32			4.30		
	.04/20_10		3:52			4.0		

Fracture surfaces were analyzed with an optical microscope and representative macro photographs are presented in Fig. 4. As can be observed, fracture surfaces were composed by a main region where crack nucleation and fatigue crack propagation occurred, and simultaneously by a secondary region where the final tearing of the material was registered. In addition, cracks nucleated at instrument's flute and propagated along a direction perpendicular to the instrument's longitudinal axis, along an almost planar cross section (Fig. 4), which revealed the reduced influence of torsional loads.

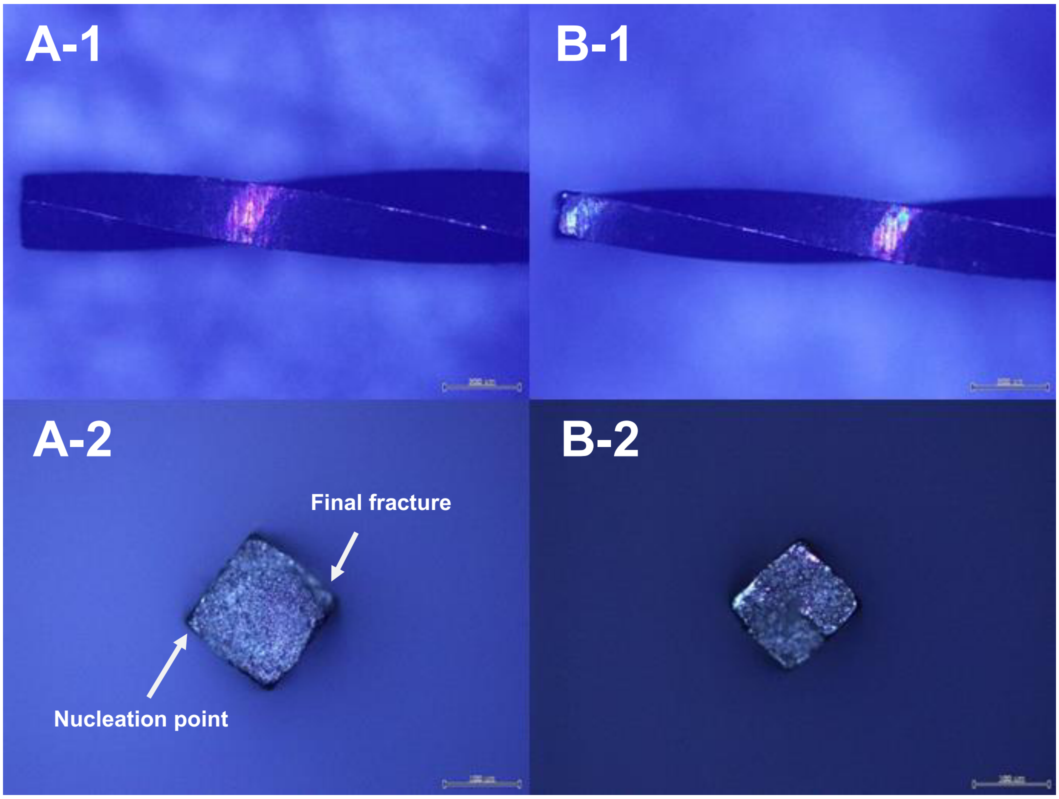

Fig. 4. Micrograph views of some fractured rotary endodontic instruments, namely: instrument ref. 04/20_12 (A-1 and A-2) and instrument ref. 04/20_9 (B-1 and B-2).

4. Discussion

The study herein presented aimed to assess the fatigue life of nickel titanium *Hyflex CM* instruments when subjected to multiple curvatures imposed by an *in-vitro* root canal. The X-ray views of LFM root canal (Fig. 1), confirmed the presence of primary, secondary and tertiary curvatures inside the metallic tooth manufactured by laser additive technology. The angles of curvature measured by the adapted Schneider's method showed good results for the primary and secondary curvatures, displaying a maximum curvature error of 10% for all curvatures analyzed.

In what the experimental *in-vitro* fatigue tests concerns, group B was constituted by nine instruments that were both subjected to rotational bending and axial motion, and exhibited an average extra time to fracture of 75 seconds (+63%) when compared with the mean time to fracture verified for the instruments belonging to Group A. This increase on the instrument's fatigue lifetime suggests the high benefit of applying axial motion with constant axial velocity to the rotary instruments during clinical treatment and can be explained by the variation over time of the maximum stress range induced at the external fibres of the rotary instruments when back and forth motion is applied. In this case, the induced stresses are spread by an enlarged area of the instrument and extended fatigue lifetime can be expected. Additionally, for the same reason, the tip's length of fractured files were in accordance with the loadings applied; in fact, endodontic files belonging to group B (also with axial movement) showed a smaller tip's average length than endodontic files of group A (0.77 mm less). In addition, the average length of fractured files' tip was about 4.9 mm for instruments belonging to Group A. This reasonably compares with the predicted location given by the FEA (6 mm), although an error of about 20% is verified. Differences between real mechanical properties of the files and those used in the FE model, or small geometrical differences between real and modelled files, or even surface finish grooves introduced during manufacturing process that were not considered during CAD 3D modelling of the files are examples of differences between real and simulated files that introduced errors in the numerical analysis. Nevertheless, and as shown by the calculations carried out, it is possible to use such numerical methods to foreseen critical regions of the files and to calculate stress concentration factors, in order to improve the design of the files and therefore to extend its fatigue resistance.

5. Conclusion

Time to fracture of endodontic files is most of the times determined by experimental *in-vitro* fatigue tests simulating single planar curvatures or double planar curvatures. However, more severe loadings could be induced on endodontic instruments when shaping and treating root canals with multiplanar severe curvatures during clinical treatments. Hence, a representative lower first molar was 3D printed in stainless steel and some *Hyflex CM* endodontic files were tested. When back and forth motion with a constant velocity of 3 mm/s was superimposed to rotational bending at 500 rpm, a mean time to fracture of 194 seconds was measured; when rotational bending at 500 rpm was the only load applied, an average time to fracture of 119 seconds was determined.

Acknowledgements

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