A Literature Review on the Linear Elastic Material Properties Assigned in Finite Element Analyses in Dental Research

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ABSTRACT

Introduction: Finite element analysis (FEA) is a numerical procedure utilised in the engineering analysis of structures and is one of the most common numerical methods utilised in many research activities in dentistry such as implantology, prosthodontics and restoration. FEA can be considered a useful tool in order to describe the deformation aspects of dental components that cannot be measured easily by *in vivo* models. The geometry, material properties, finite element model (mesh structure) and boundary conditions defined for a particular FEA setup are the factors affecting the accuracy of the results of a FEA. Most especially, material models employed in FEA play a critical role, however, the literature cannot provide standard material models and data in agreement to be defined in the FEA studies handled specifically for human teeth. The aim of this study is reviewing the most utilised data related to material properties (limited to linear homogeneous isotropic material model) of the tooth components, evaluate the sources and reasons for the different values defined in dental research and provide filtered material data which can be utilised in related FEA studies.

Material and Methods: Electronic databases (PubMed and Web of Science) were reviewed for publications on FEA utilised in dentistry research. 155 research publications in total were considered in this paper. The search keywords of "finite element analysis", "finite element study", "mechanical properties" and "teeth" were combined through Boolean operators. The primary question under review was: "How were the material properties of the tooth components and numerical ranges, which are assigned in a FEA utilised in dental research, obtained and verified?".

Results: It was possible to determine sixteen different elastic modulus (EM) and seven Poisson' ratio (PR) values for enamel, eighteen EM and five PR values for dentin, sixteen EM and four PR values for periodontal ligament, eight EM and one PR values for pulp, ten EM and five PR values for cementum, twelve EM and four PR values for cortical bone, and eleven EM and four PR values for cancellous bone. As a result, it was seen that various EM, PR, density and strength values were considered and these were obtained from a limited number of FEA studies.

Conclusion: Average ranges for the core material properties such as EM, PR, density and strength values to be utilised in a FEA set up were presented. Further studies, specifically on determination of the mechanical properties of tooth components are still needed in order to successfully utilise them and confirm the accuracy of the FEA studies related to dental research.

KEYWORDS: Finite Element Analysis, Finite Element Study, Mechanical Properties, Tooth, Teeth, Dental.

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INTRODUCTION

Due to the complex biomechanical system of the oral cavity and its limited accessibility, most of the research of this type has been evaluated through *in vitro* models. Mostly, fracture resistance, deflection magnitude under respective load and mechanical properties of human tooth structures have been physically studied through destructive or non-destructive test methods in order to describe the deformation behaviour of the tooth components. In a structural context, it is commonly known that, if the stress magnitude under loading exceeds the elastic limit of the component's material, structural failure (plastic deformation) would initiate. However, physical and direct measurements of the structural stress distribution, determination of the location of the failure initialisation zones, threshold of the material strength and life of the components belonging to a complex oral system under loading would become a very difficult phenomenon to comprehend (1).

In the scope of computer aided engineering (CAE), finite element method-based analysis (FEM/FEA) is a numerical procedure that aims to describe/simulate the physical phenomena in a dimensional (1, 2 or 3 dimensions) virtual environment through a mathematical approach. This provides a solution by generating a finite element discretisation which enables simulation of the behaviour of a materials deformation, from an elementary level to full component level, under pre-defined loading conditions (2). Related literature indicates that FEA was first utilised in the aerospace industry for solving structural, heat transfer and fluid flow based engineering problems in the 1960s and then spread to other research and engineering fields (2). The first use of this method in dentistry was published by Weinstein *et al*. in the field of implantology in 1979 (3). Since then, FEA has been frequently utilised in many research branches in dentistry such as prosthodontics, restorative dentistry and especially implantology in order to evaluate the deformation behaviour and to map structural stress distributions across teeth.

Although FEA may provide a good understanding of the physical phenomena of a situation, special care and scientific experience are needed during setup of the physical model in the digital environment in order to obtain accurate results and which is considered one of the most difficult aspects of the method (4,5). In principle, four parameters affect the solution and accuracy of a FEA. These are the geometric feature of the object to be modelled, the element type and count (mesh structure), the material properties and the boundary conditions applied (6).

Since the method became popular in academia and industry, many software platforms, which promise FEM based solutions, are now commercially available. In fact, the fundamental application steps of the method consist of the same application algorithm steps for all such types of analysis. These steps are pre-processing, solution and post-processing stages. Pre-processing is the main setup procedure and consists of modelling, defining the material model and its properties, a description of the boundary conditions (loads and constraints) and creating the finite element model (mesh structure) (Figure 1).

(**Figure 1.** FEA application procedure)

The material properties assigned in a FEA is one of the primary calculation parameters in the pre-processing procedure. Specific to the dental research domain, based on experimental evidence related to real-life applications, the structure of teeth show similarities with human bone which exhibit viscoelastic behaviour under deformation phenomenon as the material behaviour of the teeth is also highly nonlinear and viscoelastic (7). This type of nonlinear viscoelastic behaviour can be considered as a time-dependent plasticity phenomenon, which is called viscoplasticity (8). It is also known that the structures of the tooth components are anisotropic and non-homogeneous (9). As such, a description of nonlinear viscoelastic behaviour of the teeth structure in a FEA would become a very complex phenomenon. Therefore, in order to explain the viscoelastic behaviour of the teeth, researchers are mostly forced to make simplifying assumptions and apply the theories of linear viscoelasticity or Hookean elasticity in dental FEA studies. Initial damage occurrence is considered in this type of simplification operation, as the damage occurrence in a solid structure can be easily determined by considering the critical stress point (mostly yield stress points) defined for the FEA model as the damage behaviour of teeth corresponding to the generation of microcracks over the yield stress point, like bones (10). However, any real material shows deviation from the ideal material models and numerical method-based simulation tools still have some limitations in modelling real-life responses in this manner. In this context, the scientific literature supports that in the static loading cases, homogeneous isotropic material model assumptions provide acceptable results when compared to an inhomogeneous anisotropic material model (11). These findings indicate that assigning isotropic homogenous linear elastic material model definitions for the tooth components utilised in the FEA studies would satisfactorily serve the major aim of a deformation analysis. This approach is commonly seen in dental research, however, appropriate assumptions should be made with respect to the material properties and the purpose of the simulation study in dental applications.

In describing an isotropic homogenous linear elastic material model in a FEA setup, the material properties of modulus of Elasticity (EM), Poisson's Ratio (PR) and material density have to be defined. Additionally, threshold deformation or critical stress magnitudes are given for damage evaluation, however, the difficulty is seen specifically for this point: there are a wide range of differences between the magnitudes of such specific material properties of EM, PR and the damage threshold points for the tooth components of enamel, dentin, periodontal ligament (PDL), pulp, cementum, cortical bone and cancellous bone (age factor has also been considered). In relation to medical applications, a detailed review on bone properties by Novitskaya *et al*. report a disagreement (changing values) on the material properties (most especially on EM) of human bone structures given in the literature (12–14). The components of a tooth would exhibit similar structural features with bone as a result of the nature of solid-like organic materials. It was also reported that different material testing methods (such as tensile, compression and bending tests) might provide different ranges for the material properties of bone structures (15). As such, the literature search related to FEA of tooth components was carefully conducted and the material properties and associated data utilised for the tooth components were carefully evaluated. Although some experimental studies did provide clear information related to material properties to be utilised in the numerical analysis, an agreement on specific material properties for tooth components could not be found (1). In addition, there was much replication reported in some publications for the selection of material properties which prove problematic in finding the main source of the data used. This was the main motivation for this paper. Consequently, it is acknowledged that although the application procedure for the FEA is the same, the data for the material properties of the same tooth components utilised in the dental FEA studies is highly variable. The literature has also indicated that a comprehensive and specific review work on material properties utilised in the FEA studies in dental research was limited.

The aim of this study is to review the most utilised data related to the material properties of teeth components and to evaluate the sources and reasons for the different values assigned in dental research and provide filtered material data which can be trusted when utilised in related FEA studies in the dental research area.

MATERIAL AND METHODS

This literature review was conducted in accordance with the preferred reporting items for systematic reviews and meta-analyses (PRISMA) guidelines (Includes 155 suitable studies) (Figure 2). The research questions were described according to PICO (P: Population/Patient/Problem; I: Intervention; C: Comparison; O: Outcome) question: 'Related material properties of the tooth and numerical ranges, which are assigned in FEA studies in dentistry: how were they all obtained and what are the variances?' (in consideration of the isotropic homogenous linear elastic material model).

(**Figure 2.** The systematic flow chart of the study selection process)

Literature Search Strategy

An electronic search was limited to publications in the English language and performed in PubMed and Web of Science (WoS) databases using a series of search terms combined with the Boolean Operators "AND" and "OR", prior to 1st December 2020. The keywords used in the electronic search were "finite element analysis", 'finite element study', 'mechanical properties' and 'teeth'. The following search string was developed with the combination of relevant keywords: (((finite element analysis) OR (finite element study)) AND (mechanical properties)) AND (teeth))). The core references of the material properties of teeth used in the studies obtained by an electronic search were also considered and these studies were also included in this review work.

Inclusion and Exclusion Criteria

Studies were included in this systematic review if they met the following inclusion criteria: (a) Finite element studies made in any field of dentistry such as implantology, prosthodontics, orthodontics and restorative dentistry related to the material properties of teeth; (b) The studies that include the value of EM and PR of at least two of the components of teeth (enamel, dentin, cementum, pulp, periodontal ligament, cortical, cancellous and/or compact bone); (c) Studies in other languages, studies that have no information about both EM and PR of the components of teeth and studies that include information about material properties of teeth but use different terminology were not included.

Evaluation of Selected Studies

Related to the publication search during the review, both titles and abstracts were carefully assessed. Full-text evaluation of the relevant articles was performed and the articles which were not considered eligible to the inclusion criteria were excluded from the study. Disagreements concerning the inclusion of a study were discussed by the authors until a decision was obtained by consensus. The following information was specified for each study and recorded on a data extraction form: EM, PR, density and the strength properties of the components of teeth (enamel, dentin, cementum, pulp, periodontal ligament and cortical cancellous and/or compact bone). A sample tooth model (mandibular left central incisor) and its component cross section is illustrated in Figure 3.

(**Figure 3.** A sample tooth model (mandibular left central incisor) and its component cross section)

RESULTS

Study Selection

Using the present electronic search strategy, 871 records were collected from the PubMed and WoS databases. After removal of duplicates, 818 records remained for assessment of title and abstract and 316 of these studies were considered as eligible for full text analysis. 155 studies were considered as suitable for this review (Figure 2). The data of the related material properties of teeth obtained from the studies included are shown in detail in Table 1.

(**Table 1.** The data of the material properties of tooth components extracted from the studies included)

Scoping Synthesis of Parameters

There were sixteen different EM and seven PR values for enamel, eighteen EM and five PR values for dentin, sixteen EM and four PR values for periodontal ligament, eight EM and one PR values for pulp, ten EM and five PR values for cementum, twelve EM and four PR values for cortical bone, and eleven EM and four PR values for cancellous bone.

For enamel, the minimum value of EM was 41 GPa and the maximum was 100 GPa. The most frequently used value of EM was 84.1 GPa in 46 studies (56%). PR was set at 0.002, 0.4 and 0.23-0.30 in only one study, 0.2 in two studies, 0.3 in thirty-one studies, 0.33 in forty studies. The minimum value of tensile strength was 10 MPa and the maximum was 48 MPa. The minimum value of compressive strength was 95 MPa and the maximum was 400 MPa. The data obtained for density of enamel ranged from 2.14 to 4 g cm^{-3}.

For dentin, the minimum value of EM was 10.2 GPa and the maximum was 21 GPa. The most frequently used value of EM was 18.6 GPa in 88 studies (70.4%). PR was set at -0.11-0.07 and 0.33 in only one study, 0.3 in eleven studies, 0.31 in one hundred and three studies, 0.32 in eight studies. The minimum value of tensile strength was 10 MPa and the maximum was 234MPa. The minimum value of compressive strength was 232 MPa and the maximum was 315 MPa. The data obtained for density of dentin ranged from 2 to 2.97 g cm^{-3}.

For PDL, the minimum value of EM was 0.01 MPa and the maximum was 175 MPa. The most frequently used value of EM was 68.9 MPa in 48 studies (55.1%). PR was set at 0.3 in four studies, 0.45 in fifty-six studies, 0.46 in only one study, and 0.49 in eight studies. The data obtained for density of PDL ranged from 0.95 to 1.10 g cm^{-3}. There was no data found for tensile strength or compressive strength.

For pulp tissue (pulp chamber), the minimum value of EM was 0.003 MPa and the maximum was 20 MPa. The most frequently used value of EM was 2 MPa in 18 studies (40%). PR was set at 0.45 in 46 studies. The data obtained for density of the pulp tissue ranged from 1 to 1.1 g cm^{-3}. There was no data found about tensile strength or compressive strength.

For cementum, ten different values were used in twelve studies for EM. The minimum value of EM was 2.7 GPa and the maximum was 22.4 GPa. PR was set at 0.27, 0.322 and 0.35 in only one study, 0.30 in three studies and 0.31 in five studies. The value of density was 2.03 g cm^{-3}, the tensile strength was 29 MPa and the compressive strength was 32.1 MPa in one study.

For cortical bone, the minimum value of EM was 10 GPa and the maximum was 340 GPa. The most frequently used value of EM was 13.7 GPa in forty-nine studies (65.3%). PR was set at 0.26 in twelve studies, 0.3 in sixty-one studies, 0.32 in two studies, and 0.33 in one study. The tensile strength was 133 MPa in one study. The data obtained for cortical bone ranged from 1.3 to 2 g cm^{-3}. There was no data found about compressive strength.

For cancellous bone, the minimum value of EM was 0.056 GPa and the maximum was 13.4 GPa. The most frequently used value of EM was 1.37 GPa in fifty-two studies (68.4%). PR was set at 0.22 in one study, 0.3 in sixty-seven studies, 0.31 in four studies, 0.38 in eight studies. The density data obtained for cancellous bone ranged from 0.7 to 1.87 g cm^{-3}. The tensile strength was 75 MPa in one study. There was no data found compressive strength.

Additionally, in consideration of the material data compiled above, the maximum, minimum and mean value of the material properties of teeth were calculated and are represented in Table 2. The average values were obtained by summing and averaging all the different values collated from the literature.

DISCUSSION

FEA enables the undertaking of repeatable experiments that do not require ethical approval and design studies can be altered with minimal effort in accordance with the specific requirements (16). However, there are some limitations because it is a computerised approximation of an *in vitro* study. Furthermore, stress analysis is generally executed under static loading scenarios, and the mechanical properties of teeth are set as isotropic, homogeneous and linearly elastic, even if it does not correspond to *in vivo* conditions. In addition, over-simplification in geometry could affect the accuracy of the results (16,17).

It is well known that material properties, the functional loading, applied boundary conditions and the geometric detail of the object are the most important factors influencing the predicted accuracy of FEA (4). In this review, the importance of the assignment of proper material properties of teeth, which lay the foundation of the stress and strain measurement, was emphasised.

The reported material property index values vary between different research groups. The EM value for enamel ranged from 41 to 100 GPa in these studies. Zhang *et al*. explained this variability with possible influence of external factors such as the measuring system used, applied load, nature of samples and direction of enamel rod (18). In another study, it was indicated that EM and hardness of enamel in younger individuals are lower than those of older teeth (19).

Dentin has a more complex structure than enamel. The direction of the dentinal tubules and the collagen fibres and the average density of the mineral phase are the internal factors that affect the mechanical properties of dentin (18). Moreover, it was shown that the mechanical properties of dentin are affected by external factors such as hydration of the environment, as well as internal factors (20). However, no standard technique and measuring environment were discussed.

Measuring the distribution of stress and strain occurring within teeth and PDL is very difficult phenomenon to understand with *in vivo* experiments. Because of this reason, there are various methods in use, for example laser holography, optoelectronic set-ups, and photoelastic models. It would be true to say that the most commonly utilised and effective method in order to evaluate the stress and strain distribution during active movements is FEA. The different measuring methods, the type of human teeth, the length and shape of the root which influences the formation of PDL, and the humidity of test specimens could lead to obtaining different values for the material properties of a tooth component (21).

Additionally, in a FEA setup for a tooth deformation, all of the tooth components should be considered, for instance, the exclusion of PDL in a 3D model (because of its relatively lower elasticity properties), where it may not be the main research focus and may seem to complicate the model, it could cause inaccurate results in deformation behaviour and stress distribution of the full model (16,21).

The development of FEA models needs sufficient knowledge of the material properties of the human oral system, most especially the jawbone (mandible). However, determination of appropriate biomechanical properties of living tissues, especially bone tissue in FEA, is still challenging (22). As shown in this study, the wide range of values in the literature for EM also proves this situation. There are many factors that affect the measurement of the mechanical properties of human bony structures including teeth. The structure and scale of human bones are some of these factors, and this can make sampling cubic shape of bone from trabecular regions larger than 5 mm for compression tests difficult (23). The size and shape of the sample, the storage and loading conditions, the mechanical tests used, the age, sex and systemic disease of the donor, porosity density and mineral content of the sample could also affect the measurements (22,23). In addition, it was shown that mechanical properties of cortical bone changes in an edentulous human mandible (24).

In light of all this information, the most appropriate values for the purpose and methods of the study should be preferred when evaluating the material properties of dental tissues in order to obtain the accurate results that reflect the clinical situation in the FEA to be conducted.

CONCLUSION

Average ranges for the core material properties for a linear material model to be utilised in a FEA setup such as EM, PR, density and strength values were reviewed and presented. This review revealed that the values of the material parameters varied. The reason for this may be explained with different testing methods, age factor and experience based untested assumptions which are considered by some researchers. Additionally, a specific value for a specific tooth component may not be obtained due to the organic structure of human teeth components. It was also shown that the material properties of dental tissues in some of the research papers related to FEA studies were usually quoted from previous FEA studies, and there were very few studies in which *in vitro* evaluation of the mechanical properties of dental tissues was performed. The data reviewed in this paper related to linear material models to be utilised in a FEA would provide an understanding of the range in magnitudes of the specific material properties of the tooth components which should be carefully evaluated and adopted by researchers in dentistry. The average ranges were calculated in this paper in order to frame a general perspective for the values for specific tooth components. The literature search also indicated that experience in FEA setup in order to simulate real-life responses in deformation analysis is essential. Finally, it was specifically concluded that comprehensive and up-to-date physical material testing studies are required in order to obtain reliable material data for realisation of accurate finite element studies in dentistry.

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Figure Captions

Figure 1. FEA application procedure.

Figure 2. The systematic flow chart of the study selection process.

Figure 3. A sample tooth model (mandibular left central incisor) and its component cross section

Table Captions

Table 1. The data of the material properties of tooth components extracted from the studies included

Table 2. The maximum, minimum and mean value of the mechanical properties of teeth.

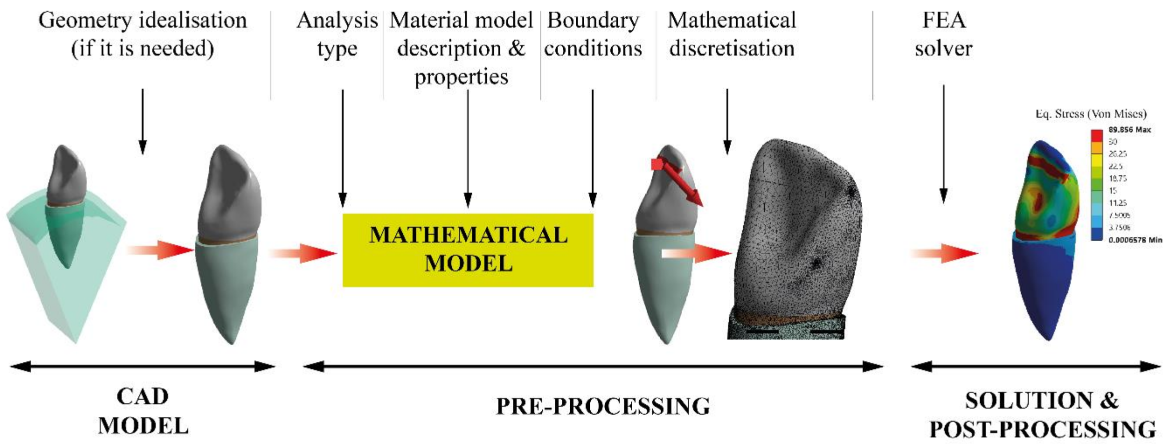

Figure 1. FEA application procedure.

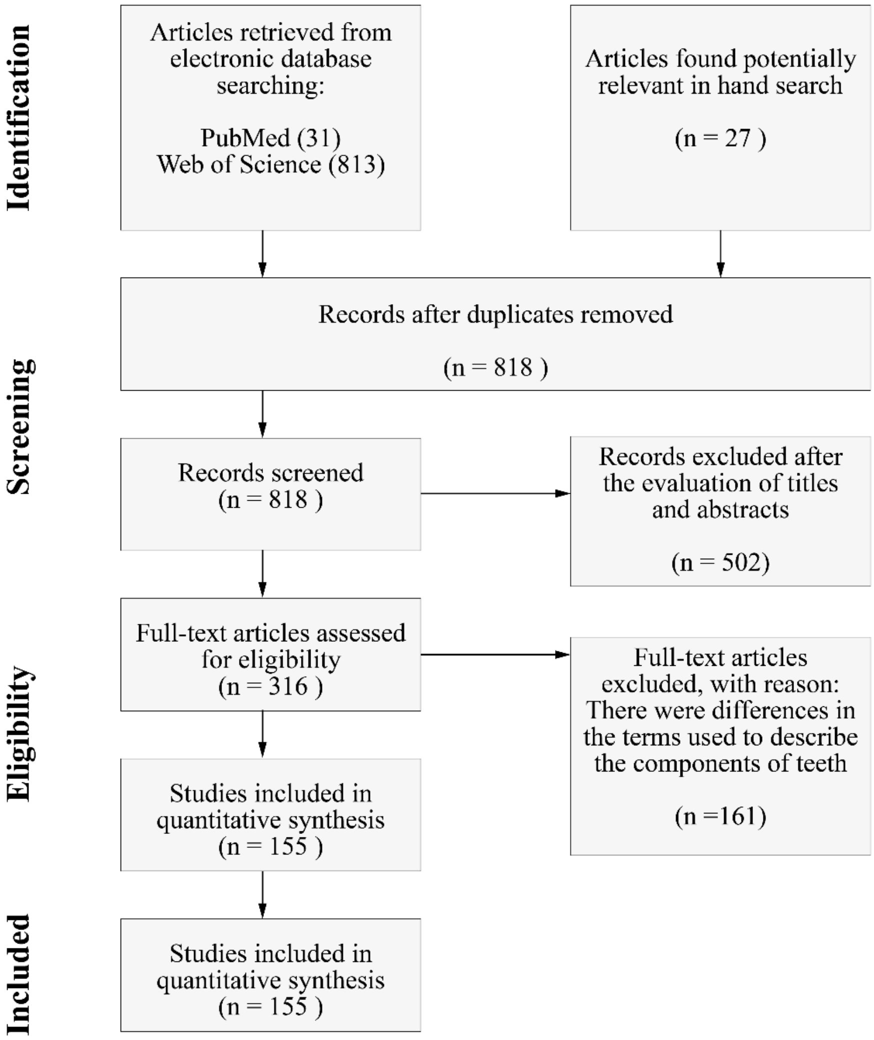

Figure 2. The systematic flow chart of the study selection process

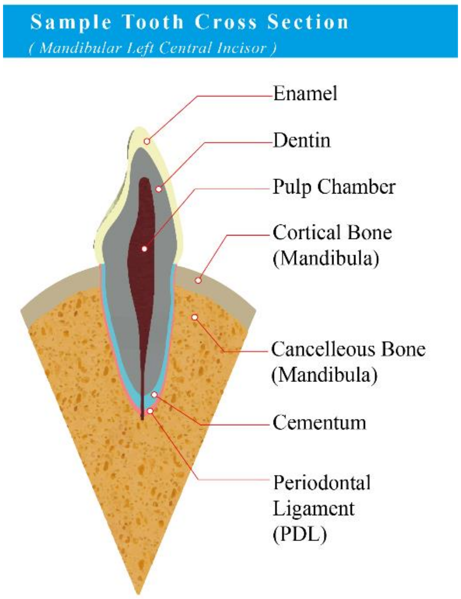

Figure 3. A sample tooth model (mandibular left central incisor) and its component cross section

Table 1. The data of the material properties of tooth components extracted from the studies included

Material	Elastic Module	Poisson's Ratio	Density	Tensile Strength	Compressive Strength
Enamel	**41.0 GPa** (25–35), **41.4 GPa** (36,37), **46.8 GPa** (38), **48.0 GPa** (39), **50.0 GPa** (40), **60.0 GPa** (41), **72.7 GPa** (42–44) **75-100 GPa** (45), **77.9 GPa** (46–48), **80.0 GPa** (49–56), **82.0 GPa** (41), **82.5 GPa** (57), **83.0 GPa** (58) **84.0 GPa** (59–64), **84.1 GPa** (5,46,61,65–106), **85.0 GPa** (107),	**0.002** (77) **0.2** (75,76,78,85,86,96,97,102), **0.23-0.30** (108), **0.30** (25–27,29,30,32,34–38,40,49–51,53,56,61,62,73,74,79,87,88,93,95,98,99,103,109,110), **0.31** (28,52,89,111,112), **0.33** (5,33,39,42–44,46,48,54,55,57–60,63–68,70–72,80–84,90–92,94,100,101,104–107,113,114), **0.4** (45),	**2.1 g cm^{-3}** (110), **2.5 g cm^{-3}** (18), **2.8 g cm^{-3}** (81,84,103), **3 g cm^{-3}** (46,48), **4 g cm^{-3}** (29)	**10 MPa** (37) **10.3 MPa** (50,74,103) , **11.5 MPa** (90,105), **30-35 MPa** (51)(108), **48 MPa** (39),	**95-386 MPa** (51,108), **384 MPa** (50,74,90,103,105), **288-400 MPa** (37)
Dentin	**10.2-15.6 GPa** (108), **11.7 GPa** (109), **12 GPa** (40), **13 GPa** (39), **13.8 GPa** (36), **14.7 GPa** (115,116), **15 GPa** (52), **16 GPa** (46), **16.6 GPa** (47,48), **16.7 GPa** (117), **18 GPa** (5,26,55,59,63,113,118–122), **18.3 GPa** (65,94,103), **18.5 GPa** (93), **18.6 GPa** (25,29–35,37,38,42–44,49,51,54,56–58,60–62,66–68,71–88,90–92,96–98,100–102,104–106,111,112,122–152), **19 GPa** (27,41,153), **20 GPa**(50,64,95,99,154) **20.7 GPa** (155), **21 GPa** (107),	**−0.11–0.07** (108), **0.30** (29,36,42,73,109,110,127,153–156), **0.31** (5,25–27,30–35,38,39,43,44,46,48–52,54–68,71,74–78,80,81,83–88,90–100,102–107,111–113,115,116,118,119,121–123,125–129,131,132,134–136,138–152,157,158) **0.32** (37,72,79,82,101,104,130,137), **0.33** (120)	**2 g cm^{-3}** (81,84,103), **2.20 g cm^{-3}** (46–48), **2.9 g cm^{-3}** (108), **2.97 g cm^{-3}** (110)	**10 MPa** (39), **40-276 MPa** (51,108), **48 MPa** (37) **51.7 MPa** (103), **98.7 MPa** (50,74,124), **105.5 MPa** (90,105), **234 MPa** (117),	**232-297 MPa** (37) **249-315 MPa** (51,108), **288 MPa** (39), **297 MPa** (50,74,90,103,105,124),

Periodontal Ligament (PDL)	**0.01-100 MPa** (159), **0.15 MPa** (160), **0.5 MPa** (133), **0.75-1.5 MPa** (96), **11.76 MPa** (116), **12 MPa** (75,76), **20 MPa** (154,161), **50 MPa** (32,33,36,46–48,74,85,95,99,110,122,124,140), **50-100 MPa** (65), **66.7 MPa** (162), **68 MPa** (86,87), **68.9 MPa** (2,25,28,30,34,35,42,49,51,54,60,66,68,71,77,82,88,91,92,94,106,111,112,114,115,119,121,123,125–128,131,132,136,141–143,145–147,151,155,157,163–166), **69 MPa** (59,84,104,113,118,120,138,139,144,152), **70 MPa** (78,97), **170 MPa** (153), **175 MPa** (107),	**0.30** (36,154,161,167), **0.45** (2,5,25,28,30,34,35,42,46,48,49,51,54,59,60,65,66,68,71,74–78,82,85,87,88,91,92,94,96,97,104,106,107,110,111,113,115,118–128,131–133,136,138–143,145–147,151–153,155,157,159,160,163,165,166,168), **0.46** (114), **0.49** (32,33,86,95,99,116,162,164)	**0.95 g cm^{-3}** (110) **1.10 g cm^{-3}** (46–48),	**X**	**X**
Pulp Chamber	**3x10^{-3} MPa** (36,37), **2 MPa** (5,25,30,35,44,56,61,62,76,77,81,86,88,93,96,97,102,119), **2.03 MPa** (65,67), **2.07 MPa** (27,33,50,55,64,92,94,106), **2.1 MPa** (49,51,54), **3.0 MPa** (28,31,32,38,111,120), **6.89 MPa** (46–48,68,87), **20 MPa** (153),	**0.45** (5,25,28,30–33,35–38,44,46,48–51,54–56,61,62,64,65,67,68,76,77,81,84,86–88,92–94,96,97,102,106,111,112,119,120,144,153),	**1 g cm^{-3}** (46–48), **1.1 g cm^{-3}** (84)	**X**	**X**
Cementum	**2.7 GPa** (26), **3.58 GPa** (169), **6 GPa** (91), **7 GPa** (134), **7.18 GPa** (114), **8.2 GPa** (170), **15 GPa** (67), **18 GPa** (59,113), **18.6 GPa** (128,136), **22.4 GPa** (129),	**0.27** (91) **0.30** (26,134,170), **0.31** (59,67,113,128,136), **0.322** (114), **0.35** (129),	**2.03 g cm^{-3}** (171)	**29 MPa** (129),	**32.1 MPa** (169),

Cortical Bone	**10 GPa** (46,48,59,94,113,118,160)**,** **10.7 GPa** (70)**,** **13 GPa** (133,172)**,** **13.7 GPa** (25,30,32–35,42,43,60,66,72,77,80–84,89,91,92,104,110–112,114,115,119–121,124,125,128,131,136,138,141,143,145–148,150,151,155,156,166,173–175)**,** **13.8 GPa** (85,86,96)**,** **14 GPa** (142)**,** **14.5 GPa** (49,54)**,** **17 GPa** (159)**,** **15 GPa** (62,75,76,97,149,152)**,** **21.4 GPa** (176) **34 GPa** (162)**,** **340 GPa** (65)**,**	**0.26** (46,48,65,85,86,96,115,138,139,156,162,173)**,** **0.3** (25,30,32–35,42,43,59,60,62,66,70,72,75–77,80–84,89,91,92,94,97,104,111–114,118–121,124,125,128,131,133,136,141–143,145–153,155,159,160,166,172,175,176)**,** **0.323** (49,54)**,** **0.33** (110)	**1.3 g cm^{-3}** (81,84)**,** **1.4 g cm^{-3}** (46,48)**,** **1.99 g cm^{-3}** (176)**,** **2 g cm^{-3}** (110)	**133 MPa** (176)**,**	**X**
Cancellous Bone	**0.056 GPa** (159) **0.25 GPa** (59,94,113,118) **0.345 GPa** (85,86,96) **0.5 GPa** (46,48,160) **0.508 -11.2 GPa** (155) **0.91 GPa** (70,177) **1 GPa** (133,172) **1.37 GPa** (25,30,32–35,42,43,49,54,60,65,66,72,77,81,82,84,89,91,92,104,106,111,114,115,119–122,124,125,128,131,136,138–140,142,143,145–148,150,151,153,156,166,173,175,176)**,** **1.4 GPa** (110) **1.5 GPa** (62,75,76,97,149,152) **13.4 GPa** (162)**,**	**0.22** (177) **0.3** (30,32–35,42,43,49,54,59,60,62,66,70,72,74–77,81,82,84,85,89,91,92,94,96,97,104,106,111–115,118–122,124,125,128,133,136,140–143,145–153,155,159,160,166,172,175,176), **0.31** (25,86,110,131)**,** **0.38** (46,48,65,138,139,156,162,173)**,**	**0.70 g cm^{-3}** (110) **1.3 g cm^{-3}** (81,84)**,** **1.4 g cm^{-3}** (46,48)**,** **1.87 g cm^{-3}** (176)**,**	**75 MPa** (176)	**X**

Table 2. The maximum, minimum and mean value of the mechanical properties of teeth.

Material*	Elastic Modulus	Poisson's Ratio (-)	Density (g cm^{-3})	Tensile Strength (MPa)	Compressive Strength (MPa)
Enamel	Min : 40 GPa Max : 100 GPa Average :70 GPa	Min :0.002 Max :0.4 Average :0.2	Min :2.5 Max :4 Average :2.8	Min :10 Max :48 Average :24	Min :95 Max :400 Average :310
Dentin	Min : 12 GPa Max :21 GPa Average :17 GPa	Min :0 Max :0.3 Average :0,2	Min:2 Max:2.9 Average:2.5	Min :10 Max :276 Average :108	Min :232 Max :315 Average :279
Periodontal Ligament (PDL)	Min :0.5 MPa Max :175 MPa Average :50.8 MPa	Min :0.3 Max : 0.5 Average :0.4	Min : 1.1 Max : - Average : -	-	-
Pulp Chamber	Min : 0.003 MPa Max : 7 MPa Average : 4.75 MPa	Min :0.45 Max : - Average : -	Min :1 Max :1 Average :1	-	-
Cementum	Min :2.7 GPa Max :22.4 GPa Average :10.6 GPa	Min : 0.2 Max : 0.3 Average : 0.2	Min :2.03 Max : - Average : -	Min :29 Max : - Average : -	Min :32 Max : - Average : -
Cortical Bone	Min :10 GPa Max :34 GPa Average :16 GPa	Min :0.2 Max : 0.3 Average :0.3	Min :1.3 Max :2 Average :1.6	Min :133 Max : - Average : -	-
Cancellous Bone	Min :0.05 GPa Max :13.4 GPa Average :2.65 GPa	Min : 0.2 Max : 0.3 Average : 0.3	Min :0.7 Max :1.8 Average :1.3	Min : 75 Max : - Average : -	-