SHEAR STRESS ANALYSIS BY FINITE ELEMENTS OF A METAL-CERAMIC DENTAL BRIDGE ON A CoCr ALLOY SUPPORT

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A dental bridge is a fixed prosthesis that replaces lost teeth. It can consist of three or more elements and uses the patient's natural teeth as support. Using the finite element analysis (FEA) approach, the current work seeks to evaluate, the performance of a ceramic dental bridge with a CoCr alloy insert, under the action of masticatory forces, in relation to the technique used to obtain the metal insert: sintered, CAM-milled, and cast. The results of this investigation show that the differences in mechanical strength of a dental bridge construction on a sintered, cast, or CAM-milled CoCr insert are insignificant in terms of these properties and cannot harm the prosthetic or the teeth that serve as the abutment. Also, this type of assembly can be considered an optimal economic solution for treating edentation and restoring the dental-maxillary system's functioning. However, as the connectors and holding components are the weakest parts, additional consideration must be given to their design, particularly in the connection area between the bridge pieces.

Keywords: finite element analysis, dental bridge, biomaterials, Co-Cr alloys, dental applications

1. Introduction

Calculations, models, and simulations are used in finite element analysis (FEA) to forecast and comprehend how an item might react under various physical

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conditions. In the analysis of biomechanics, finite element methods are frequently used because it is simple to identify the stress distributions and make comparisons with the others, especially in various implant designs [1–3]. Finite element analysis can be used to optimize design parameters and alternative material combinations. This approach is important for analysing the causes of prosthesis failure [4–8].

Most people over the age of 40 suffer from missing teeth because of decay, accidents, anomalies, or hard impacts. The reasons for tooth loss vary from patient to patient, some may become edentulous, while others may only lose a few teeth. The benefits of repairing the teeth are numerous, including improving the pronunciation of words, the mastication process, and, obviously, self-confidence. This work can also prevent the early leaving of the face and protects any remaining natural teeth - in the case of partial dentures. There are numerous ways to replace real teeth with artificial teeth by using various methods including fixed partial dentures (Fig. 1), removable partial dentures, full dentures, interim partial dentures, implant-supported bridges, and cantilever bridges - each with advantages and disadvantages [9,10]. Complete prostheses are used exclusively in edentulous patients (who have no teeth at all). Total dentures are prosthetic devices designed to replace all missing teeth on a single jaw and restore the reduced functioning of the dental-maxillary apparatus. When all the teeth on one jaw have been lost, a total prosthesis is used to treat this condition. A total prosthesis can be unimaxillary or bimaxillary based on the number of teeth lost on each jaw. Removable partial prostheses are used in patients who are missing a significant portion of their teeth but still have viable natural teeth. A prosthesis consists of replacement teeth attached to a pink acrylate base that is attached to a metal frame that holds the prosthesis in place within the mouth. When only a few natural teeth are remaining in the upper or lower jaw, partial dentures are used. In addition to filling the edentulous spaces, partial dentures help to maintain the position of the remaining teeth. In addition to being removable, partial prostheses appear natural. Fixed partial prostheses are pruriently devices fixed by cementing or gluing, possibly through other means of retention (screws) to natural teeth (prepared in the form of abutments), dental roots, and/or implant posts. Fixed partial prostheses restore the morphological and functional integrity of dental arches interrupted by small, interspersed, extended, or multiple gaps.

Fig. 1. Metal-ceramic dental bridge

The dental bridge is a fixed prosthetic work used in dentistry for the morphological restoration of missing teeth from the dental arches or for the protection and remodeling of teeth that have suffered significant destruction. The dental bridge is used especially in treating partial edentates, compensating the masticatory, aesthetic, and phonetic functions of the absent teeth, and preventing complications that may occur because of tooth loss. The advantages of dental bridges are: low risk of material rejection, short execution time, can replace a variable number of dental units, excellent aesthetics when covered with ceramic materials that imitate the natural tooth, and a very good lifespan if they are taken care of. One significant disadvantage of dental bridges is the aggressive reduction of healthy naturally supporting teeth. The use of dental cement, which has issues with its varying levels of solubility and comparatively low strength, is another drawback of dental bridges.

To estimate physical and chemical impacts and comprehend engineering issues better, finite element analysis (FEA) is also applied. Due to the various stresses that dental restorations are subjected to during mastication, research has shown that FEA helps to better understand how they behave. The failure analysis of prosthetics also uses finite element analysis [11]. The power of computing resources and the sophistication of data manipulation algorithms are making finite element analysis more and more important to research. In the past, three-dimensional models of complex systems were not adequately modeled using tools that had sufficient power.

Usually, biomaterials are used not just in dentistry [12,13], but in a wide range of medical specialties, including general surgery [14], cardiothoracic surgery [15], ophthalmology, and orthopedic surgery [16–19]. By coordinating the functions, qualities, and structures, biomaterials are mainly used in dentistry to enhance oral health. Base metal alloys are widely used in dentistry for dental work and instruments, as detailed below. Casting alloys Cr-Co and Cr-Ni have been used for many years to make the bases of partial dentures, almost completely replacing type IV gold alloys. Cr-Ni mixtures are used for the execution of crowns and bridges, being developed as substitutes for type III gold alloys. Although they do not have the properties of noble metals, metal-ceramic works with a "non-noble" skeleton have good resistance and aesthetics, an acceptable lifespan, and a reasonable price. The metal-ceramic crowns and the metal-ceramic bridges are made up of a supporting metal skeleton over which several layers of dental ceramic are added. Metal crowns are the most durable and are generally used for molars or teeth that are not visible during speech. The supporting metal skeleton ensures the resistance of the entire prosthetic work, being located under the ceramic layer and giving it support. It has a reduced thickness and its adaptation to polished teeth is extremely intimate. The alloys used are specific for ceramic works. The composition allows a very strong chemical connection with the ceramic layer. During the addition of layers of ceramic material, these alloys release a layer of oxides on the surface. This layer will form a very resistant chemical bond with the layer of oxides that, in turn, the ceramic releases during the same process. This is one of the most important advantages of metal-ceramic works compared to metal-acrylic and metal-composite ones. The connections between the metal skeleton and the acrylate composite are strictly mechanical, having a significantly lower strength. The quality but also the price of work largely depends on the metal alloys used to make the skeleton. Non-noble metal alloys are the most used materials for various metal-ceramic works (crowns, bridges, the dental crown on the implant) [20].

Co-Cr-based alloys have a high content of cobalt (53–67%) and chromium (25–32%) in the composition and small quantities of Mo, W, Si, and others. They were developed as an alternative to noble alloys and their properties are comparable. But they require a complex technological process, to obtain different dental prostheses because: they have a high melting range, they cannot be melted with an oxy-gas flame, they have high hardness values, and they are processed and finished very hard (that's why their use is limited to certain types of medical applications), they have a lower cost price and superior mechanical properties to noble alloys. Co-Cr alloy ductility is relatively low and is the major deficiency of these alloys when they are used for casting partial dentures. The density has values between 8-9 g/cm^3, much lower than noble alloys (19-20 g/cm^3). The modulus of elasticity is approximately twice as high as that of the noble class IV alloys. Casting shrinkage has high values due to the high casting temperature [21,22].

2. Materials and methods

2.1. The mechanical properties of the materials used in the numerical simulation.

The simulated model is divided into two geometric domains: one representing the metallic structure framing the dental bridge and another domain represented by the ceramic crowns.

Dental crowns were assigned generic properties for dental ceramics used on the market. Thus, in studying the elastic and mechanical resistance properties of several dental ceramics the Poisson's ratio was used. One of the references used was the Ivoclar Vivadent ceramic, whose properties are presented in Tables 1 and Fig. 2 respectively [23,24]. The indications of Ivoclar Vivadent ceramics include crowns, inlays, onlays, and veneers [25].

In the case of ceramic materials, as in the case of metal insert materials, the technology of obtaining the semi-manufactured products and the manufacturing technology of the finished product, strongly influences the mechanical behavior, respectively both the mechanical resistance and especially the resistance to crack generation and propagation.

Table 1

Physical properties of Ivoclar Vivadent ceramics [23,24,26]**.**

Physical properties	Value
Flexural strength (ISO 6872)	360 ± 60 MPa
Fracture toughness (SEVNB)	2.0 – 2.5 MPa m^{½}
Vickers hardness	5900 ± 100 MPa
Modulus of elasticity	95 ± 5 GPa
Coefficient of thermal expansion (100-500°C)	10.45 ± 0.25 10^{-6}/K^{-1}
Density	2.5 ± 0.1 g/cm^3
Chemical solubility	30 – 50 µg/cm^2
Poisson’s ratio υ	0.23

Table 2 presents some mechanical properties closely related to the method of obtaining the ceramic material.

Table 2

Mechanical properties of dental crowns made of two ceramic materials from Ivoclar Vivadent [27].

Material	Strength (MPa)	Obtaining method	Fracture Toughness (MPa $m^{1/2}$)	Elastic modulus (GPa)	Hardness (GPa)
E1 - ceramic material (recommended for inlays, onlays, veneers, and anterior crowns)	106	The dental restorations are fabricated by hot-pressing an ingot of the material into a mold	1.2	65	6.5
E2 - ceramic material (recommended for the fabrication of a 3-unit fixed partial denture)	306		2.9	105	5.3

Recent studies show the differences in the mechanical properties of Co-Cr metal insert about how it was obtained: casting, CAM milling, or laser sintering. Padrós et al. studied nine samples from each manufacturing group (casting, milling, and sintering) of the Co-Cr alloy [28]. The results of the experimental determinations focused on surface roughness, marginal tolerance, hardness, and mechanical properties and highlighted the distribution of properties depending on the method of obtaining the alloy. Thus, the hardness and mechanical properties are presented in Table 3, where it can be observed that strictly from a mechanical point of view, the alloy obtained by CAD/CAM milling is superior to the other two [29]. These properties were used in the numerical analysis of the dental bridge structure.

Regarding the alloy's workability, elongation describes the physical deformation of a substance before fracture when subjected to tensile stress. Yu et al. verified the mechanical properties of the Co-Cr alloy obtained by casting, CAD/CAM milling, and laser sintering [30].

Table 3

Mechanical properties of the Co-Cr alloy manufactured by casting, CAM milling, and laser sintering [29].

Properties/(Units)	Manufacturing Process/Results (mean ± STD)		
	CAD/CAM	Casting	Laser Sintering
Marginal Gap, (µm)	50.53 ± 10.30	85.76 ± 22.56	60.95 ± 20.66
Sa, (nm)	731.27 ± 19.0	796.60 ± 19.8	859.5 ± 18.9
SA Index area	2.1 ± 0.1	2.2 ± 0.1	1.8 ± 0.1
Hardness, (HV)	356 ± 20	390 ± 15	473 ± 25
Flexural load to fracture, (N)	6813 ± 169	6291 ± 105	5422 ± 302
Deflection to fracture, (mm)	4.10 ± 1.12	2.55 ± 1.21	3.75 ± 1.10

The tensile specimens were equipped with strain gauges to determine the deformation. The obtained mechanical properties are presented in Table 4, where we can observe the fact that the elongation and elastic modulus have the lowest values in the samples obtained by SLS and the highest in those obtained by CAD/CAM milling and the Yield Strength has the lowest value in the SLS samples and the highest in those obtained by milling [30]. When a dental bridge is applied, the enamel of the natural teeth and its degree of wear are factors that significantly influence the masticatory function of the oral cavity. As a result, the wear amounts of dentures and real teeth must be comparable.

Table 4

Mechanical properties of Co-Cr alloy obtained by casting, CAD/CAM milling, and laser sintering [30]

Properties/(Units)	Manufacturing Process/Results (mean ± STD)		
	CAD/CAM	Casting	Laser Sintering
0.2% Yield Strength (MPa)	438 ± 15	501 ± 33	1008 ± 59
Elongation (%)	12 ± 5	11 ± 3	9 ± 2
Elastic Modulus (GPa)	235	226	219

2.2 Dental bridge reconstruction

3D reconstruction is a Reverse Engineering procedure by which a series of 2D images obtained by X-ray or optical scanning is transformed into a virtual volume with geometry very close to the real one. This procedure lends itself well to biological structures because their complex geometries would be difficult to obtain through classical modeling. For the reconstruction and the finishing of the mesh model, two programs were used: Meshmixer and Geomagic Studio.

Meshmixer and Geomagic Studio programs were used for the reconstruction and finishing of the model mesh. Thus, we moved from 2D grayscale images to closed volumes that possess the characteristics of solids: they are finite and integral. To reduce the size of the working file, a virtual resurfacing of the model was carried out, thus moving from the triangular mesh of reduced dimensions to a surface divided by patches bounded by NURBS curves.

Fig. 2 shows a summary of several steps taken to move from 2D images to the 3D model used in the simulation. It also contains the structure of the crown made of ceramic material and the metal insert in a conjugation relationship.

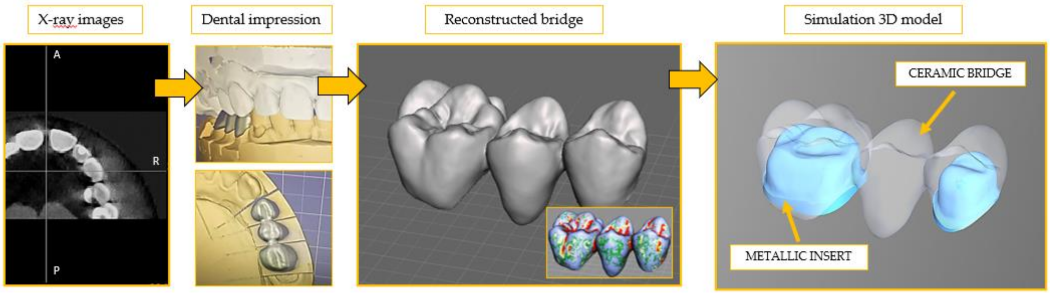


Fig. 2. Stages of the Reverse Engineering transformation of the dental bridge

2.3 FEA simulation itinerary

The numerical simulation was carried out in the Ansys 2019 R2 program, considering four types of materials, two types of geometric structure, two types of stresses, and two directions of stress. All these variables were connected according to the logic in Fig. 3.

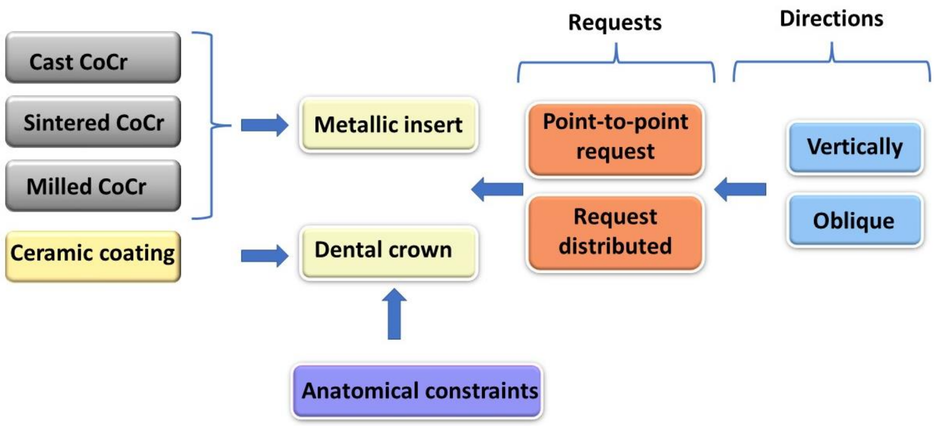


Fig. 3. FEA simulation itinerary

The two geometric models (the dental crown and the metal insert) are at the centre of the simulation itinerary. Constraints like those of anatomical type were applied to them, made concrete by the definition of contacts. The materials used were generic ceramics on the one hand and CoCr alloy obtained by 3 primary manufacturing technologies: cast, sintered and CAM processed, on the other. The simulation was performed one by one for each insert type, with the material defined for the crown remaining with constant properties and geometry. The demands applied to the structure are of two types: point and distributed, and the directions of action of the demands were also two: vertical and oblique.

The simulation was carried out in the elastic range of the materials used. Elasticity is quantified by Young's modulus and Poisson's ratio and describes the deformation mode of a body under the action of the applied stress. The tension state of the dental bridge, as well as its deformation, was monitored through simulation. Stress is a measure of the additional internal forces that arise within a body when it is subjected to stress. It is defined in the simplest way as the ratio between the locally applied force and the surface area.

2.4 Discretization, boundary conditions, and structure loading

The discretization was carried out with mechanical elements of the automatic type, with a constant size of 0.5 mm. The domain was divided into 76765 elements containing 120699 nodes. The finished network, the mesh, can be seen in Fig. 6, as a whole and in detail. Bonded contact relationships were defined between the geometric elements of the dental bridge. This type of contact simulates a perfect assembly, without separation or possibilities of mutual movement, both between the metal insert and the crown but also between the metal insert and the dental structure.

The dimensions of the surfaces in contact depend directly by the transmission of the state of tension from the stress point of application to the surface considered fixed. The distributed force to the dental bridge was applied to simulate different occlusion modes, which generate compressive and shearing stress. They were chosen two directions of stress: a vertical one along the axis of the tooth root, respectively an oblique force at 45° oriented about the same axis. Also, the manner in which forces are distributed in the 2 directions was considered in two cases: with the force applied quasi-pointwise (a very reduced surface) and with the force applied uniformly distributed on the surfaces of the dental bridge. The modules of the stress forces were 75 N in each of the simulated cases and which quantifies a unilateral value of the masticatory force under physiological conditions. The state of tension and deformation that will result from the analysis must therefore be viewed under the aspect of the stress state. Any change in the modulus of this force will decisively change the stress values in the dental bridge.

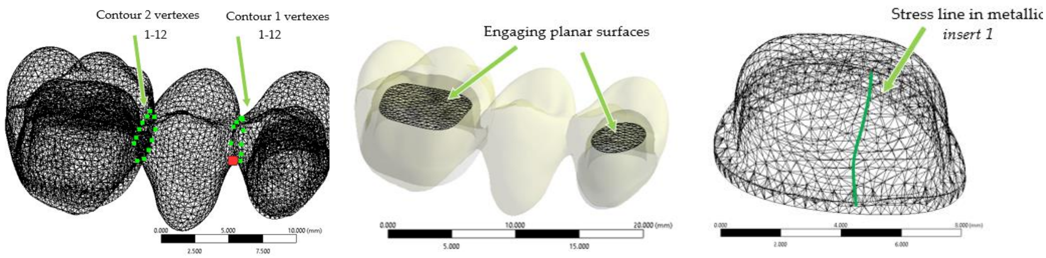
3. Results and discussion

The numerical simulation was performed on a 3D model of a dental bridge built on a CoCr metal support. The crown materializes three molars and was simulated with the generic mechanical properties of dental ceramics. The simulation conditions were of a physiologic type, both in terms of modules and loading directions, but also in terms of constraints.

The results of the analysis refer to the state of tension and deformation in the dental crown and the metal insert in the case of three different mechanical properties of the CoCr insert, corresponding to its manufacturing technology: sintered, CAM milled, and cast.

Principal, shear, and von Mises stresses were extracted from a series of nodes, elements, or surfaces as follows. On two circular outlines that stand in for the bridge's bond locations, stresses were taken out of the dental crown. These were extracted from the level of 12 nodes of each contour (Fig. 9). These two areas were chosen because they are the ones susceptible to fracture, considering the way the structure is supported, and respecting the way of loading.

From the level of the metal insert, the equivalent stresses were extracted from two areas: a surface area of engagement of the metal insert on the natural tooth structure and a linear area of the metal insert, from its most stressed area.



a) b) c)

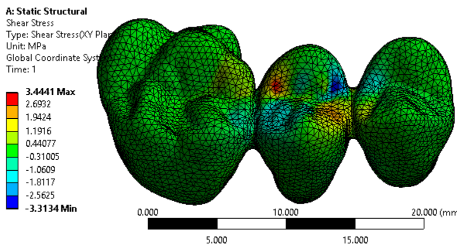
Fig. 4 Stress sampling areas: a) from the ceramic crown; b) from the engagement surfaces of the metal insert; c) from the most requested area of the metal insert

3.1 The displacement of the dental bridge under the action of mechanical stress

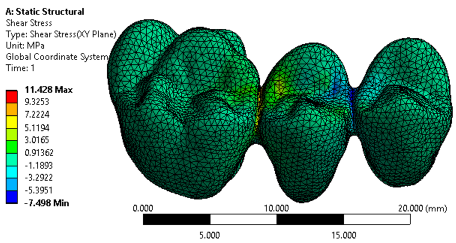
The displacement of the dental bridge highlights, qualitatively and quantitatively, how the structure loaded with stress and fixed using metal inserts will deform. Two types of structure displacements are presented: the vertical displacement (Z) and the total displacement. It can be seen from them the significantly different ways of moving the structure depending on the direction of the request and the concentrated or distributed way of its application. The maximum value of the directional displacement is in the range of 1-3x10^{-4} mm and of the total displacement of 2-9 x10^{-4} mm. It is therefore highlighted that the dominant displacement is in the direction of the stress, so mainly in the vertical direction. From the location point of view of the maximum displacement, this is recorded on the median crown because the metal inserts were considered on the two marginal crowns. Regarding the directional displacement of the insert, it is at values of the order of 10^{-5}-10^{-6} mm, namely very low. These values are found regardless of the CoCr alloy production type, because in any production method (cast, sintered, or CAM milled), its rigidity is similar and at a very high level. The fact that the metal insert is captive between the ceramic crown and the simulated surface of the tooth structure also contributes to this.

3.2 The equivalent and shear stresses of the dental bridge

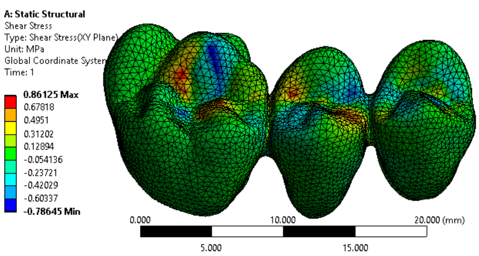
Shear stresses refer to the stresses that occur in the structure to oppose the tendency of relative rotational or sliding movement of two surfaces. Fig. 5 shows the mode of distribution of these stresses in close connection with the direction of stress and the mode of distribution. The maximum values recorded at the level of the entire structure vary between 1 and 12 MPa, which is little compared to the mechanical strength of dental ceramics and CoCr alloy. This indicates that the structure will remain intact for physiological stress values, both from the point of view of the insert and from the point of view of the ceramic.



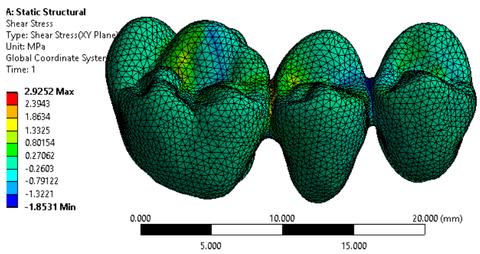
a) The concentrated force applied vertical



b) The concentrated force applied oblique



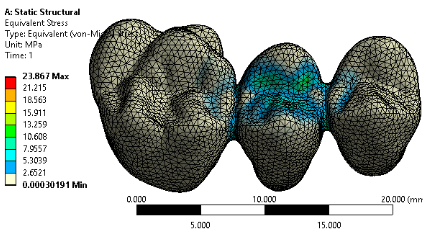
c) The distributed force applied vertical



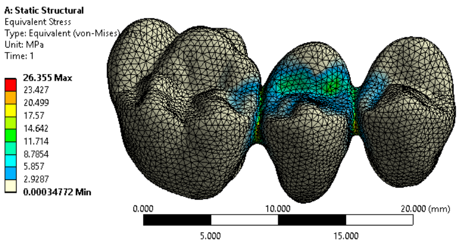
d) The distributed force applied oblique

Fig. 5. Shear stresses in the dental bridge

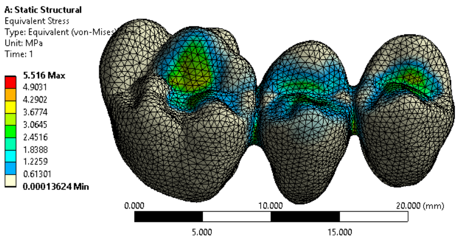
Equivalent stresses on the other hand are higher, between 2 and 25 MPa due to how they are defined (Fig. 6). In the calculation of the equivalent stress, the effect of the state of triaxiality of the stress appears, and hence the higher values. However, the values of the equivalent stresses do not exceed the values of the mechanical strengths of the CoCr alloys and the ceramic material.



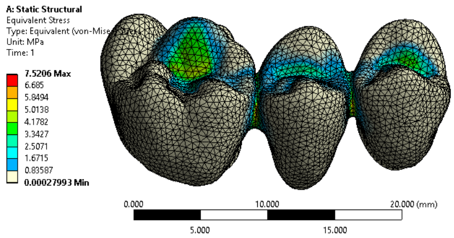
a) The concentrated force applied vertical



b) The concentrated force applied oblique



c) The distributed force vertically applied

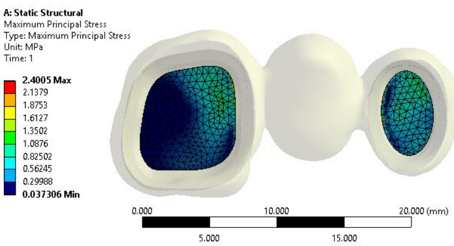


d) The distributed force applied oblique

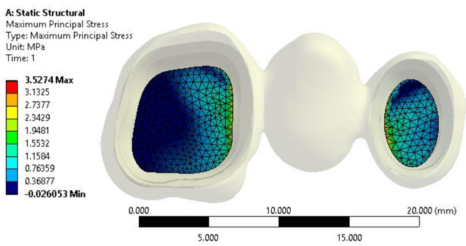
Fig. 6. The equivalent stress in the dental bridge

The maximum principal stresses are followed because they are a predictor of the surface fracture. Thus, if a model with brittle material properties is subjected to a multiaxial state of stress, then failure will occur when the maximum principal stresses exceed the local mechanical strength anywhere in the component.

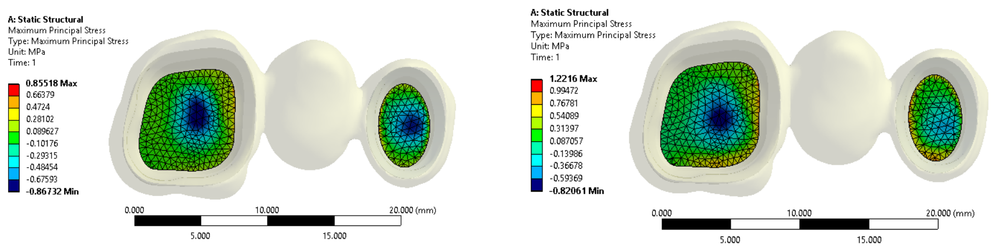
Fig. 7 shows the maximum principal stresses in the surfaces of the engaging elements (the metal insert). A circular distribution of them is observed in the case of uniformly distributed stress, with increased values towards the extremities of the sections. In the case of point forces, the bending effect leads to a loading of the inserts oriented towards the middle crown. The stress values are very low compared to the yield limit and R_m of the CoCr alloy and therefore the structure does not present any risk of mechanical failure.



a) The concentrated force applied vertical



b) The concentrated force applied oblique



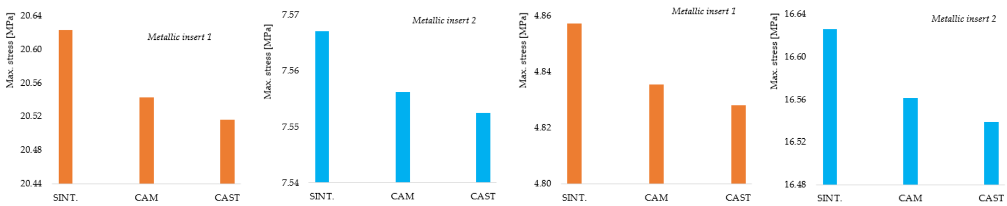
c) The distributed force applied vertical d) The distributed force applied oblique

Fig. 7. Maximum main stresses on the surface of the engagement elements

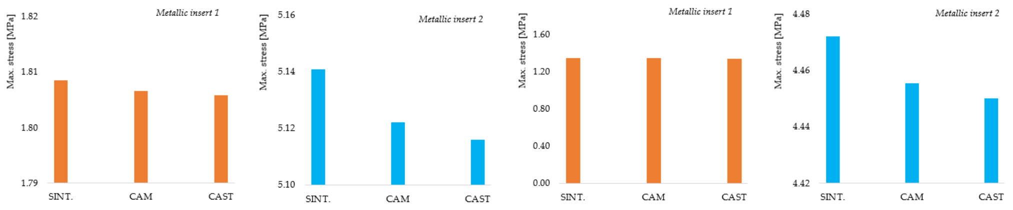
3.3 The performance of the dental bridge in relation to the technique used to obtain the CoCr insert

The performance of the dental bridge under the aspect of the type of CoCr alloy used, is presented in the following, as values of the maximum stress in the metal insert. The presentation of the chart-bar diagrams (Fig. 11) was made for each of the two metal inserts and the three types of CoCr alloy used. The stress value obtained is the most dependent on the mode of stress, with a maximum value in the vertical point case. When comparing the stress values according to the three types of materials, a similar trend can be observed regardless of the forces applied. Thus, the sintered CoCr alloy insert appears with the highest stress values, followed by the one from CAM milled alloy and finishing with the cast one. This indicates that the stiffness of the sintered alloy is superior, it takes the stress better. Anyway, the differences in stress variation are very small between the three types of CoCr alloy used, of the order of magnitude < 1MPa, which leads us to the conclusion that from the point of view of the load transfer and transmission mode, they are similar.

In Fig. 8 there are presented the variations of the stress values in the maximum stress zone of the metal elements. These are presented as values from 20 nodes and according to the direction and type of force applied. It can be seen how the stress jumps in the maximum stressed area are very strong for point-type stresses, independent of the metal insert 1 or 2 (so almost independent of the size of that area of the bridge). It is observed instead that for the uniformly distributed force an almost constant tension and obviously at lower values.



a) The concentrated force applied vertical b) The concentrated force applied oblique



c) The distributed force applied vertical

d) The distributed force applied oblique

Fig. 8. The maximum stresses in the metal insert depending on the types of CoCr alloy used

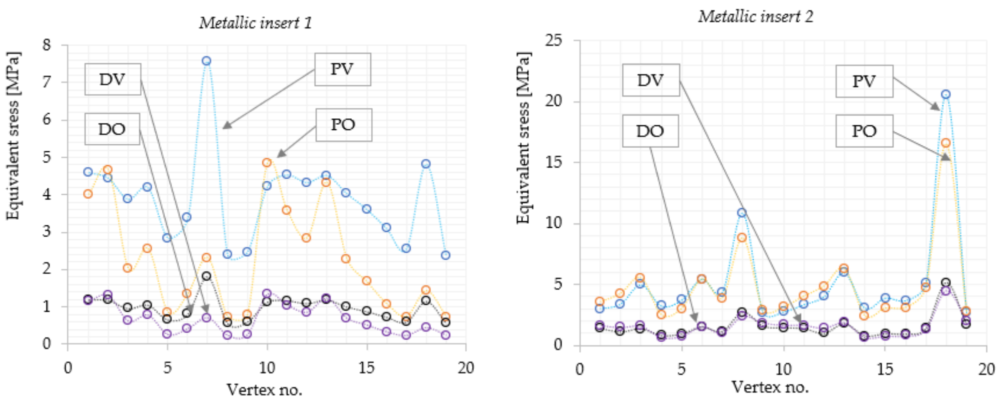


Fig. 9 Variation of stresses in the metal insert on the line of maximum stress

Fig. 10 shows, in the form of a circular representation, the values from 12 nodes sampled circularly and equidistantly on the two contours of the ceramic element of the bridge. And around ceramic sections, the decisive dependence of stresses is related to the type of stress and not to the type of material from which the insert is made. The lowest stress values are recorded for uniformly distributed stress.

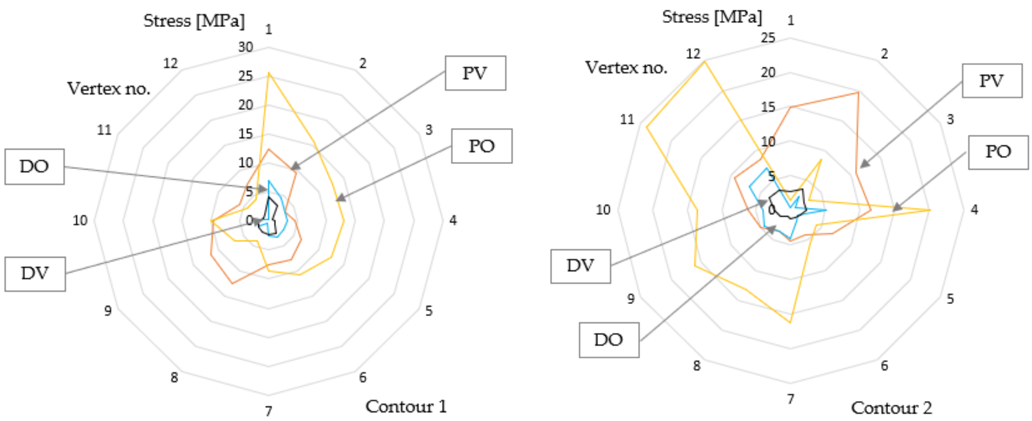


Fig. 10 Variation of stresses in the contours of the dental crown

4. Conclusion

Being a purely mathematical procedure, finite element analysis has several drawbacks because it enforces certain simplifications that result in some degree of approximation of conclusions [31]. This kind of research does not consider several significant factors that can determine whether prosthetic rehabilitation is successful or unsuccessful, including the materials used, the complexity of the masticatory forces in relation to the point of application, the mobility of the supporting teeth, and the characteristics of the mandible itself. In our investigation, the models were loaded with constant, evenly distributed vertical and oblique stresses that were applied at various locations over the dental bridge's surface [32]. However, in real life, masticatory pressures vary according to the kind and consistency of the meal as well as the person's muscle activity.

The results of the analysis refer to the state of stress and deformation in the dental crown and in the metal insert in the case of three different mechanical properties of the CoCr insert, corresponding to its manufacturing technology: sintered, CAM milled, and cast.

Following the numerical analysis, we can conclude that the metal insert plays a decisive role in obtaining the mechanical tenacity behavior of the inner deck assembly. We also note that the differences in mechanical strength of a dental bridge construction on a sintered, cast or CAM-milled CoCr insert are insignificant. We can observe that the rigidity of construction using sintered CoCr is 2-3% higher than in the case of the alloy obtained by the other two manufacturing technologies.

The method of obtaining the metal insert is particularly important in terms of crack initiation and propagation, directly influencing the fatigue behavior of the structure. The presence of the metal inset reduces the risk of fracture of the dental crown, due to the tenacity of the alloy.

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