Does the transversal screw design increase the risk of mechanical complications in dental implants? A Finite Elements Analysis

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Abstract

The transversal screw was introduced in order to overcome some disadvantages of the transocclusal screw. However, its mechanical risk has not been studied sufficiently. The main purpose of this research was to assess and compare stress distribution in the screws and abutment of a single-crown implant with transversal and transocclusal screw models. Two 3-D models were assembled to analyze a single implant-supported prosthesis with transversal and transocclusal screws embedded in the jawbone. The crown was subjected to a static load of value 300 N with different levels of inclination. The transversal screw model, with an axial load of 15 degrees, was the one with lowest stress values in all its components. However, the stress was greater with more inclined loads when compared to the transocclusal model. The prosthetic transversal screw showed much less stress than the rest of the components for any load inclination**.** The transversal screw design is the option with the lowest risk of mechanical complications, both in the prosthetic screw and in the abutment screw, when applying forces of lower inclination. The more oblique forces favoured a better biomechanical environment in the abutment and its screw in the transocclusal screw model.

Keywords: Transversal screw, transocclusal screw, biomechanical stress, implant-supported prosthesis, axial and oblique load.

INTRODUCTION. Dental prostheses to replace one or more teeth have a survival rate of 96.3% after 5 years and 89.4% after 10 years in single-unit protheses [1], and of 95.6% at 5 years and 93.1% at 10 years for implant-supported fixed dental prostheses [2]. Consequently, patients prefer implant-retained single-unit prostheses to a conventional bridge. The method of joining this prosthesis to the implant can be by employing dental cement (cemented prosthesis) or through the use of a screw (screwed prosthesis). Both present distinct advantages and disadvantages. The high probability of periimplantitis due to unremoved cement and the difficulty involved in retrieving the prosthesis are the main disadvantages of a cemented implant-retained prosthesis. In contrast, one of the advantages of a screwed implant-retained prosthesis is the ease of recovery when necessary. The use of a transocclusal screw, which screws from the crown to the implant or to an intermediate abutment, is the most frequently-used method of retention in screwed prostheses. To insert or access the fixing screw, this method needs a screw channel through the occlusal surface of the crown, which is an aesthetic disadvantage. Likewise, this method is not recommended when the implants/abutments are inclined with a disparallelism greater than 30 degrees, or when the access hole to the screw is situated towards the vestibular of the crown. In these cases, irrespective of any other prosthetic solution, the use of a transversal screw can overcome these two disadvantages. This screw is a component of a transversal screw prosthesis model which consists of an abutment screwed to the implant and a transversal screw (or lateral fixing screw) used to attach the crown to the aforementioned abutment.

With either screw system, during functional and parafunctional activities the implant/crown system is subjected to complex force patterns of varying intensity and direction which transmit different kinds of stress to all of its components. Different biomechanics studies show that the stress on the prosthetic components (abutment and screws) of a single-unit prosthesis is mainly found in the conical connection between the abutment and the implant [3,4] and in the neck and first threads of the abutment screw [5,6] or along the threads of the transocclusal screw [7]. However, no data is available for the stress on the implant-retained prosthesis of a transversal screw model; the information available shows only the description and method results of clinical cases [8-13]. Furthermore, determining the distribution and the location of the highest concentration of stress in the prosthetic components has clinical interest inasmuch as it can provide data to help implement designs which offer greater resistance and optimum stress distribution in order to avoid or prevent mechanical and technical risks. Regardless of any biological complications, which can occur in all screw prosthesis models, mechanical and technical complications are more commonly encountered than biological ones [1]. The loosening or loss of the abutment screw or transocclusal screw is reported to occur within the range of between 6-12.7% of screw-retained prostheses [1,14,15], while fracturing of the abutment or screws occurs in 0.35-4% [14,15] both depending on the prosthetic design. In transversal screw designs, loosening of the lateral screw was reported to occur in 15.1% of cases, fracturing of this screw in 1.4% and loosening of the abutment screw in 2.7% [8]. In other words, loosening of the lateral screw occurred more frequently than crown debonding in a single-implant-retained cemented prosthesis [9]. Insomuch as the transversal screw prosthesis represents a possible clinical option that may be chosen instead of a cemented prosthesis or transocclusal screw model, the lack of clinical and biomechanical data means that the analysis of the biomechanical behaviour of the transversal screw model needs to be clarified. In any case, more data related to the distribution and concentration of stress in the prosthetic components is needed so that the practitioner may opt for the type of implant-supported prosthesis that best avoids and prevents mechanical risks. To this end, the purpose of this research is to assess the distribution of stress in the screws and in the abutment of a single-crown implant with transversal screw subjected to occlusal loading, axial and non-axial. This information will enable practitioners to make the right decision.

METHODS:

FEM model. Stress distribution in screws (prosthetic and abutment screw) and in the abutment of a single-crown implant have been evaluated by means of two 3D finite-element models. According to the Lekholm and Zarb classification [16], a type 2 edentulous mandibular posterior bone segment was modelled to represent the section of the mandible in the second premolar region. A threaded implant was modelled using as a reference a 4.1x10 mm screw-shaped dental implant Straumann system (Straumann AG, Waldenburg, Switzerland) inserted in a mandibular bone section. Furthermore, abutments and screws were modelled taking the following structures as references: 1) abutments SynOcta TS, 4 mm in height, and SynOcta 1.5, 1.5 mm in height, both with an internal morse cone-connection of 8 degrees; 2) fixture screws from the abutment to the titanium implant (6.7 mm in height) and with a countersunk head of 15 degrees; 3) two prosthetic fixture screws from the crown to the abutment: one transocclusal (SCS), 4.4 mm in height and the other transversal, 2 mm in diameter and 3 mm in length. All of these structures came from the Straumann Company (Straumann AG, Waldenburg, Switzerland). A cobalt-chromium (Co-Cr) alloy was used as the crown framework material (8 mm high, 10 mm buccolingual and mesiodistal diameter) and veneered with feldespathic porcelain. The porcelain thickness varied from 0.5 mm to 1 mm from the cervical area to the occlusal area.

One finite element model included the abutment SynOcta TS and its fixture screw to the implant, the transversal screw for fixing the crown to the abutment (prosthetic screw), and the crown; this is the transversal screw model. The other model included the abutment SynOcta 1.5 and its fixture screw to the implant, the transocclusal screw SCS for fixing of the crown to the abutment, and the crown; this is the transocclusal screw model. **Figure 1** and **Figure 2** show real parts of both types of implants. **Figure 3** shows a section of the bone where the dental implant is inserted. The trabecular and cortical zones can be appreciated.

Interface conditions and material properties. Materials considered in this research were modelled as homogenous, isotropic and linearly elastic. The Poisson’s ratio and Young´s modulus properties were set according to data available in existing literature [17-21], **(Table 1)**.

All simulations have been executed as static linear analyses. This simplification is acceptable as long as the maximum value of the stress is under the elastic limit value of the material and if the relationship between stress and deformation is linear up to this value. Another condition for congruent static linear analysis is that all parts of the model should behave as a whole: no gaps or multipoint-constraints should be allowed. In this research, to obtain approximate stress distribution at a reasonable computational cost, the bone has been modelled as an isotropic linearly-elastic material, although this is not completely true. Also, the interface between bone and implant has been modelled with a tie to simulate a perfect osseintegration, which was in consonance with previous research of this type. Furthermore, all related structures such as abutments, framework crowns and screws were assumed to be perfectly bonded together through the contact surfaces without any loosening [3,6] or friction. A passive adjustment of all the structures was also assumed, and neither the tolerance margin nor the pre-loading of the screws was considered, in accordance with other biomechanic studies [22,23].

Boundary conditions and loads. A static axial load of 300N was applied to both models. The location of the load was at the central fossa of the surface of the occlusal crown. Bucco-lingual load was applied with different levels of inclination with respect to the longitudinal axis of the implant: 0, 15 and 30 degrees respectively. The choice of these values was not arbitrary. In clinical practice, angled implants are common, and these angle values of load inclination are the best option to analyze the biomechanical behaviour of these implants.

Stress distribution was produced numerically. In order to compare and identify biomechanical discrepancies between transocclusal and transversal screw models, colour-coded plots of von Misses stresses were rendered. Ansys 11.0 finite element package was used to model and mesh the mandibular bone segment, dental implant, abutments, abutment screws, transocclusal and transversal screws, and crown (framework and veneering material). The transversal screw model had 57,559 nodes and 32,255 elements while the transocclusal screw model had 82,402 nodes and 46,737 elements. All parts of each model were meshed with a SOLID187 element. This finite element is a high-order tetrahedral element with 10 nodes. Each node has three degrees of freedom representing displacement in the X, Y and Z axes. The shape of this element is best suited to modelling complex structures. In addition, the quadratic behaviour exhibited by this element in the displacement field is the best option to represent complex deformations, as was the case here.

RESULTS.

Implant abutment stress distribution. With respect to the inclination of the load, both models show similar stress values in the abutment. The least amount of stress is registered with the axial load (0º). This value increases as load inclination becomes greater. Nevertheless, some differences were noted. The application of inclined loads of below 30 degrees favours a better biomechanical environment, with lower stress values in the abutment of the transversal screw model (abutment SynOcta TS) when compared to the abutment SynOcta 1.5 of the transocclusal screw model. The opposite is the case with loads of higher inclination, the abutment of the transversal screw model exhibiting greater stress values with a maximum of 1,252.1 MPa at 30º compared to 1,100.9 MPa in the case of the transversal screw abutment, (**Table 2)**. In addition, the different geometry of the two abutments seems to influence the distribution and location of the stress. Whereas in the abutment SynOcta TS the stress is located in the internal cone, spreading towards the body in a lingual or vestibular direction depending on the inclination of the load, in the transocclusal screw model abutment it is found mainly on the occlusal surface of the body and on small surfaces of the body-cone transition area of the abutment with axial load. With loads of greater inclination these surfaces extend towards the neighbouring areas of the cone and body of the abutment **(Figure 3).**

Abutment screw stress distribution. The maximum von Mises stress values in the abutment screw of the transversal screw model show a similar tendency to that described for the abutment, with lower values with axial load and a progressive increase as the inclination of the load increases, up to a maximum of 309.96 MPa at 30 degrees of inclination. In contrast, the stress values in the abutment screw in the transocclusal screw model barely change with the load inclination while registering the lowest level at a load inclination of 30 degrees (190.61 MPa), and with a range between 198.3 - 201.32 MPa for the rest of the load inclination value **(Table 2)**. In any case, for axial load and 15 degrees inclination, the stress in the abutment screw is somewhat greater than that found in the transversal screw model, and with loads of greater inclination it could be anywhere from 1/3 to 2/3 less **(Table 2)**. In the transversal screw model, however, the stress in the abutment screw is distributed through areas located in the neck, the occlusal surface of the head and threads of the screw, with hardly any change when load inclination is increased. In the transocclusal screw model, the location of the stress is different, being distributed primarily over a small area of the first and fifth threads for any load inclination. With the greater load inclination of 30 degrees, stress is also registered in small areas of the neck and body of the screw, as in **Figure 4**.

Stress distribution in the prosthetic screw. Of all the structures under review (abutments and screws), the least amount of stress is recorded in the prosthetic screw of the transversal screw model (transversal screw), the values varying little when the load does not exceed 30º inclination. In contrast, the prosthetic screw of the transocclusal screw model (transocclusal screw) showed greater stress values at any load inclination when compared with the transversal screw. These values increased progressively according to the load inclination, from 365.96 MPa with axial load up to 828.81 MPa with the load of greatest inclination. At all events, the stress values in the prosthetic screws are very different to those registered in the corresponding abutment screw (see **Table 2**). Stress is located and distributed in the non-threaded area of the transversal screw closest to the threads and in most of the threads with the least inclined loads, whereas on increasing the inclination, it is distributed in a similar way in the non-threaded area and in the second- and fifth-nearest threads. In the transocclusal screw, for any load inclination, the stress is mainly distributed and located through the neck and head of the screw, with some difference in the extension in the head, depending on the exact inclination**, Figure 5**.

Stress distribution in the trabecular and cortical bones. Figure 6 shows the distribution of tensions in the peri-implant bone under an axial and non-axial load, both for the transverse screw model and the transocclusal screw model. Under conditions of axial loads, in the transverse screwed model the stress is located in the area of the cortical bone around the upper margin of the implant neck. The stress peak occurs in the lingual area. On the other hand, in the transocclusal screwed model, stress is limited to the periphery of the entire implant neck, without exceeding it laterally. When the load is not perpendicular to the axis of the dental implant, both models present a similar pattern of stress distribution, no longer appearing in the periphery of the implant, but located in the vestibular area, opposite the application of the load, and extending through the transition zone between both bones without passing the cortical bone.

DISCUSSION

Biological and Clinical Implications. This research shows that regardless of the screw system, stress on the abutment is directly related to load inclination. Stress increases constantly with increasing values of the angle between the longitudinal implant axis and the load. Although this tendency coincides with what is described in existing literature [7, 24-26], the greater level of stress registered in the abutment with loads exceeding 15 degrees of inclination confirms that an oblique load considerably worsens stress distribution; this is probably one of the worst loading cases and should be prevented whenever possible [27]. Previous studies with different biomechanical designs have shown this tendency in the transocclusal screw abutment [7,24,26], though data has not been found in the case of the transversal screw. However, results show that a greater or lesser inclination of the applied load can influence the selection of the type of screw model, given that with very inclined loads, the biomechanical environment in the transocclusal screw abutment model is better in comparison to the transversal screw model, while the opposite occurs with axial and angular forces of 15 degrees. This result may be explained by the different geometry of the abutment. Likewise, the greater level of stress registered in the abutment of both models, compared to that registered in the abutment screw, supports the importance of the abutment when extending stress through the peri-implant bone, implant and screw. Therefore, the greater stress in the abutment could have dissipated and reduced the stress transferred to the screw, protecting the latter from excessive tension and preventing it from loosening, as cited in previous articles [28,29]. On the other hand, the location in the conical connection and anti-rotational hexagon of the abutment of the transversal screw model, similar to what is related for other abutment implants [4,26,30], could make the implant more likely to fracture if, for any reason, the tension in the anti-rotational hexagon increases due to this coinciding with the weakest area of the implant [25,31]. Also, an increase in the stress distribution in the abutment cone could be related to the mechanical complication of a loosening or fracturing of the abutment screw [32-34] but also to increased deformation, which favours the appearance of implant micromovements [22] that could lead to periimplant bone loss. However, although in existing literature a similar location of stress is described in the abutments of implant-supported prostheses of transocclusal screws [4,26,30], the results obtained indicate that what is described in the aforementioned literature is not applicable to the transocclusal screw model abutment as it shows a different stress location.

In an implant-supported screw prosthesis of the crown-abutment-implant system, the weakest part is often one screw or another, based on the fact that the majority of clinical studies have revealed that the mechanical complications with the highest frequency and rate are the loosening or fracture of the abutment screw or of the prosthetic screw which screws the crown to the abutment [1,9,14,15]. This study reveals the influence of the inclined load on the possible occurrence of mechanical complications in the abutment screw of the transversal screw model but not in that of the transocclusal screw model. For any load inclination, this screw shows similar stress values which are much lower than the elastic limit of titanium [35]. In contrast, the stress level in the abutment screw of the transversal screw model, with an inclined load of 30º, is close to this value, and with loads of lesser inclination the stress values are reduced by half, thus rendering complications less likely in this screw. Likewise, the location of the areas of greatest stress in the neck and threads of the abutment screw of the transversal screw model indicates the possible fracturing/loosening areas of this screw, which in general coincides with what is described in previous biomechanics studies with reference to the abutment screw in designs without the transversal screw [5,24,26,29,36] and also in clinical studies [37-40]. On the other hand, if the stress in the screw makes it more likely to fracture due to material fatigue, the elastic limit being exceeded or deformation being caused with loss of preload-sliding and subsequent loosening [41-43], the results show that the prosthetic screw of the transversal screw model is the one in which these complications are least likely to occur. The lowest level of stress of all the structures in the study was measured in this screw, with values of only approximately 10% of the elastic limit of titanium. In contrast, the prosthetic screw of the transocclusal screw model can withstand elevated stress values; these are close to the elastic limit of titanium with axial load, and easily exceed it as the load becomes more inclined. These results might be explained by the different morphology, geometry and placement of both screws. The location of the stress in the neck of the prosthetic screw of the transocclusal screw model as well as in its head coincides with the area most prone to the loosening/fracturing described in other studies [7,43], whilst in the transversal screw the stress is primarily located in the area connecting to the abutment, indicating that this screw can bring the retainer closer towards the margin of the implant. This leads to a reduction in misfit, which in turn clinically hinders the filtration and colonisation of bacteria [3,11]. In accordance with what has been stated here, the clinical application of this is that the practitioner may choose a transversal screw model with a view to achieving fewer mechanical complications in the prosthetic screw model

- including with oblique loads - compared to those encountered with the transocclusal screw model. However, existing literature tells us that the estimated percentage of complication-free single-implant restorations with internal connection is 97.6% after 3 years [44] and the success rate for single-unit screw restorations approximately 96% [45], the most frequent cause being occlusal overload [46]. However, in the case of the single-unit transversal screw model, only 65.75% of those tested were free from complications over a period of 3 years, though in most cases pressing the screw was enough to solve the problem [8]. Furthermore, 15.0% transversal screw loss has been reported together with a 1.4% fracture rate and 2.7% abutment screw loss [8], added to which lateral screw loosening occurred more frequently than crown debonding in a single-unit implant-retained cemented prosthesis [9].

Justification of the finite element analysis and its limitations. Stress distribution in the screws and abutment has been evaluated by means of the finite element method. This numerical method has been used over last 25 years to predict and characterize stress in prostheses, prosthetic components, orthopaedics [47], peri-implant bone and implants subjected to diverse load conditions. Nevertheless, numerical methods suffer from numerical errors during calculation and from mathematical simplifications when modelling loads, material behaviour and geometry. If these limitations are known and controlled, the method will provide good approximations of stress distribution compared to real ones. Therefore, they should be interpreted with care, and whenever possible compared to in vivo studies. To this end, quantitative data should not be the main source of comparison when performing FEM analysis. Numerical values from the simulations are dependent on the sophistication of the model. Conversely, qualitative data, such as the location/distribution of the stress/strain, may be considered realistic and should be the main source of comparison.

On the other hand, the distribution of tension and deformation depends significantly on the mathematical model used to simulate the behaviour of the material. Although the materials can be modelled as anisotropic, orthotropic or isotropic, in this study, and in most existing literature, the hypothesis is that the materials are linear elastic, isotropic and homogeneous. Likewise, a single titanium alloy is assumed for all the components of the study with the same mechanical properties. Although it is similar to the designs of other studies [4,26], it may also be a limitation, in that these components are produced for clinical use with commercially pure titanium or with different titanium alloys. It is also assumed that the interfaces between all the components (abutment, screws and crown) are perfectly bonded together through the contact surfaces without any loosening. Similarly, a passive adjustment is assumed without vertical or horizontal discrepancies. The friction coefficient, tolerance margin and the pre-loading of the screws were not taken into consideration either, and this could be a limitation.

Finally, during chewing, variable and complex strength patterns of different intensity and direction are produced in the teeth which are impossible to simulate mathematically. Nevertheless, axial and oblique occlusal loads should be considered when performing FEM simulations of an implant/abutment/crown system. In this research, a 300 N force, with inclinations of 0, 15 and 30 degrees, was applied to the occlusal fossa of the premolar implant-supported crown. Although studies of the bite force show appreciable discrepancies from different areas of the mouth and among individuals, the load used is very close to that of the estimated force for the back teeth [48]. The different load inclinations were chosen because they may simulate the biomechanical behaviour of common clinical situations associated with implants and abutments. However, a variability in the inclination of the applied loads is a constant feature in existing literature, studies having been found with axial load alone or combined with two or more inclinations of 15, 30, 45 or 60 degrees.

CONCLUSIONS

According to finite elements simulations, and within the limitations of this method, the following key conclusions can be extracted. The transversal screw model may be the choice with the lowest risk of possible mechanical complications in the abutment and abutment screw, and in the prosthetic screw when loads of low inclination are applied. The behaviour of the stress levels for any of the prosthetic components is similar in both models, and these increase as the load inclination becomes greater.

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TABLES

Table 1. Material elastic constants and structures.

Material	Structure	Young's modulus (E) (GPa)	Poisson Ratio (v)	References
Titanium	Implant, abutments and screws	110.0	0.35	Lewinstein et al (1995)
Co-Cr Alloy	Crown frameworkk	218	0.33	Anusavice and Phillips (2003)
Feldspathic Porcelain	Crown veneer	82	0.35	Eskitascioglu et al (2004)
Bone	Cortical	15	0.25	Benzing et al., 1995
	Trabecular	1	0.25	Natali et al., 2006a and b

Table 2. Maximum von Mises stress (MPa) in the abutment, the abutment screw and prosthetic screw in the transversal and transocclusal screw models during the application of an axial and non-axial load of 300 N.

Structure	Transversal screw model			Transocclusal screw model		
	0º	15º	30º	0º	15º	30º
Abutment screw	143.22	168.57	309.96	198.3	201.32	190.61
Abutment	487.48	552.74	1252.1	598.74	845.28	1100.9
Prosthetic Screw	69.13	68.85	58.05	365.96	659.41	828.81