Validation procedures in computerized dentistry

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CHAPTER 6

Comparative finite element stress analysis of implants with abutment and screw with different abutment materials and connections.

Keywords: Abutment, Dental Implant System, Finite Element Analysis, Stress Distribution,

Zirconia. ***6.1 Abstract***

Purpose: To evaluate by finite element analysis (FEA) the influence of the abutment material (titanium or zirconia) on the stress distribution in two implants with abutment and screw, one with an internal and one with an experimental external octagon connection

Materials and Methods: The two implants were modelled in a three-dimensional FEA program with the abutment material titanium or zirconia. The maximum principal stress distribution due to the combined influences of bite forces and the pre-load due to the tightening torque of the abutment screw was investigated.

Results: The stresses in the zirconia abutment with the internal octagon might result in failure, where the stresses in the implant with abutment and screw for the version with external octagon connection might result in unacceptable deformation of the implant for both abutment materials. For the version with internal octagon connection the higher tensile stresses in the zirconia abutment partly offset the advantage of the higher strength of this material.

Conclusions: This study indicates that to exploit the high strength of zirconia as abutment material the actual distribution of the tensile stresses and the design of the dental implant system must be taken into account. The abutment-implant combination with internal octagon connection showed to be a better design.

6.2 Introduction

Zirconia was well known in ancient civilizations as a rare gem. Its name is said to be derived from the Arabic-Persian word ”Zargon” which means gold coloured stone. It was first discovered in Germany in the seventeenth century by the chemist Martin Heinrich Klaproth. It was used in industry in areas of high chemical and mechanical stresses long before it was accepted as a biomedical material.

The introduction of 3Y-TZP zirconia as a new core material made metal free, full ceramic dental restorations possible, even in high stress areas [1] Due to its mechanical and physical properties, zirconia can replace metal taking certain design parameters into consideration [2] Yttrium stabilized zirconia is stronger than for example titanium. The tensile strength of titanium alloys is 789-1013 MPa [3] and the tensile strength of zirconia is 1074-1166 MPa [4]. Moreover, yttrium stabilized zirconia has a high fatigue resistance caused by a martensitic transformation from tetragonal to monoclinic, which is accompanied by a volume increase of 3.5%. All-ceramic restorations gained lots of attention due to their superior biocompatibility and esthetical characteristics compared to other aesthetic restorative materials which have many disadvantages as component dissolution, liquid absorption, hydrolysis, and colour change during long term service in the oral cavity [5]Although the esthetical differences between crowns on a metal or zirconia abutment are subtle [6], titanium has the disadvantage for dental implants of considerable bacterial accumulation on the supra-gingival part when compared to zirconia [7], where professional cleaning can cause damage to the relatively soft implant or supra-structure surface. Considering its (bio) material properties, zirconia has been confirmed to be a material of choice for dental prosthetic devices, and also implant-abutment systems. For "all zirconia implants” scientific studies are needed to fill the gaps concerning long-term clinical evaluation of these implants currently leading to propose an alternative use like a titanium implant with zirconia abutment [8].

However, the mechanical consequences of the introduction of zirconia to replace titanium have not been studied well. The influence on the stress distribution might be different for different connector systems between the implant and the abutment. Chun *et al* studied the stress distribution in 1-body, internal-hex and external hex implants [9]. However, they did not take the screw joint preload on the stresses into consideration.

The objective of this study was to analyze with finite element analysis (FEA) the stress distribution in two implants with abutment and screw, one with an internal and one with an experimental external octagon (Dyna Dental Engineering, Bergen op Zoom, the Netherlands) with the abutment in titanium alloy or zirconia, in order to evaluate the mechanical consequences of the change of the abutment material.

6.3 Materials and methods

FEA model design

In this study two implants with abutment and screw were analyzed. The Dyna Helix® Implant (Dyna Dental Engineering B.V., Bergen op Zoom, the Netherlands) with internal octagon connecton (A) and with an experimental external octagon connection (B) were realized as Finite Element Analysis (FEA) models with titanium alloy (Ti6Al4V) (1) and zirconia abutment (2). Fig. 6.1 showing schematic drawings of the implants with abutment and screw. The abutments were designed with the software package Cyrtina®CAD (Oratio B.V., Hoorn, Holland).

Material	**Young's modulus (GPa)**	**Poisson ratio**
Zirconia	210	0.3
Titanium alloy	109	0.31
Titanium grade IV	107	0.3
Bone	10	0.3

Table 6.1: The material properties

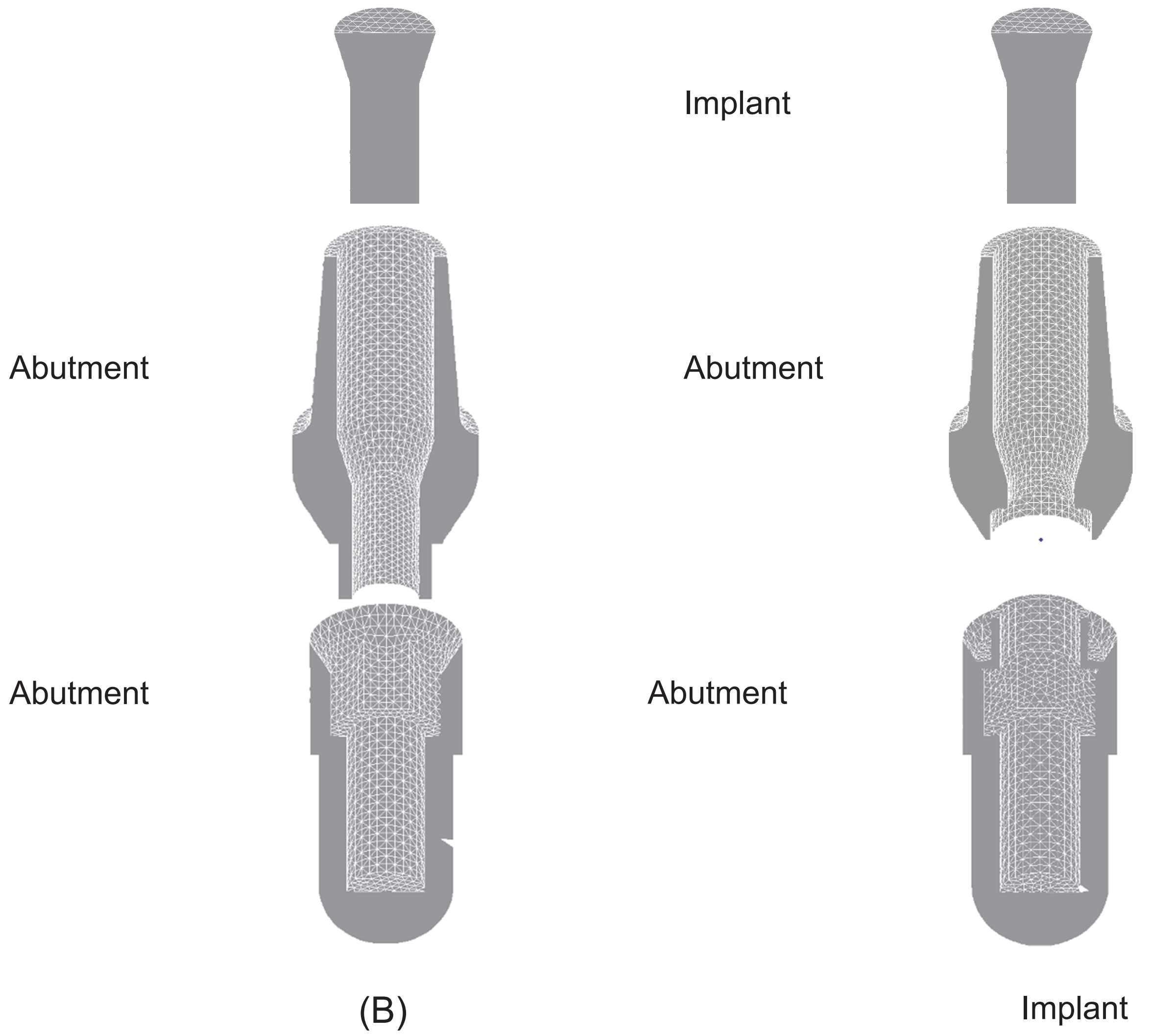

Figure 6.1: The layers composing the FEA model of the implant with abutment and screw with internal octagon connection (A) and with an experimental external octagon connection (B).

The dimensions of all components were according to the construction drawings (Fig. 6.2). The external helix of the implant for the fixation in the bone was simplified to a cylinder with the average dimensions of the thread of the implant (Ø 3.075 mm). The screw thread connection between the abutment screw and the implant was simplified by cylinders with a diameter of the average dimensions of the thread (Ø 1.78 mm). The external octagon of the system with external octagon connection had a slight wedge shape and was designed to deform the implant in the contact area over a length of 0.22 mm with a maximum deformation of 0.01 mm, when the abutment screw was fixed (Fig. 6.2). The bone surrounding the implant was simulated by a block with dimensions of 6 x 6 mm and a height of 9 mm.

The final model consisted of 55,461 parabolic tetrahedron solid elements for the system with internal octagon connection and 60,803 parabolic parabolic tetrahedron solid elements for the system with external octagon connection.

The finite element modelling and post processing was carried out with FEMAP software (FEMAP 9.3, USG Corp., Plano, Texas, USA), while the analysis was done with NX Nastran software (NX Nastran, USG Corp., Plano, Texas, USA).

The non-linear analysis was done with 10 time steps and 100 iterations per step; the convergence tolerance was set at 0.001.

In post processing, the contour options "elemental average" without use of the "corner data" were used for visualizing the results of the Maximum Principle Stress (MPS).

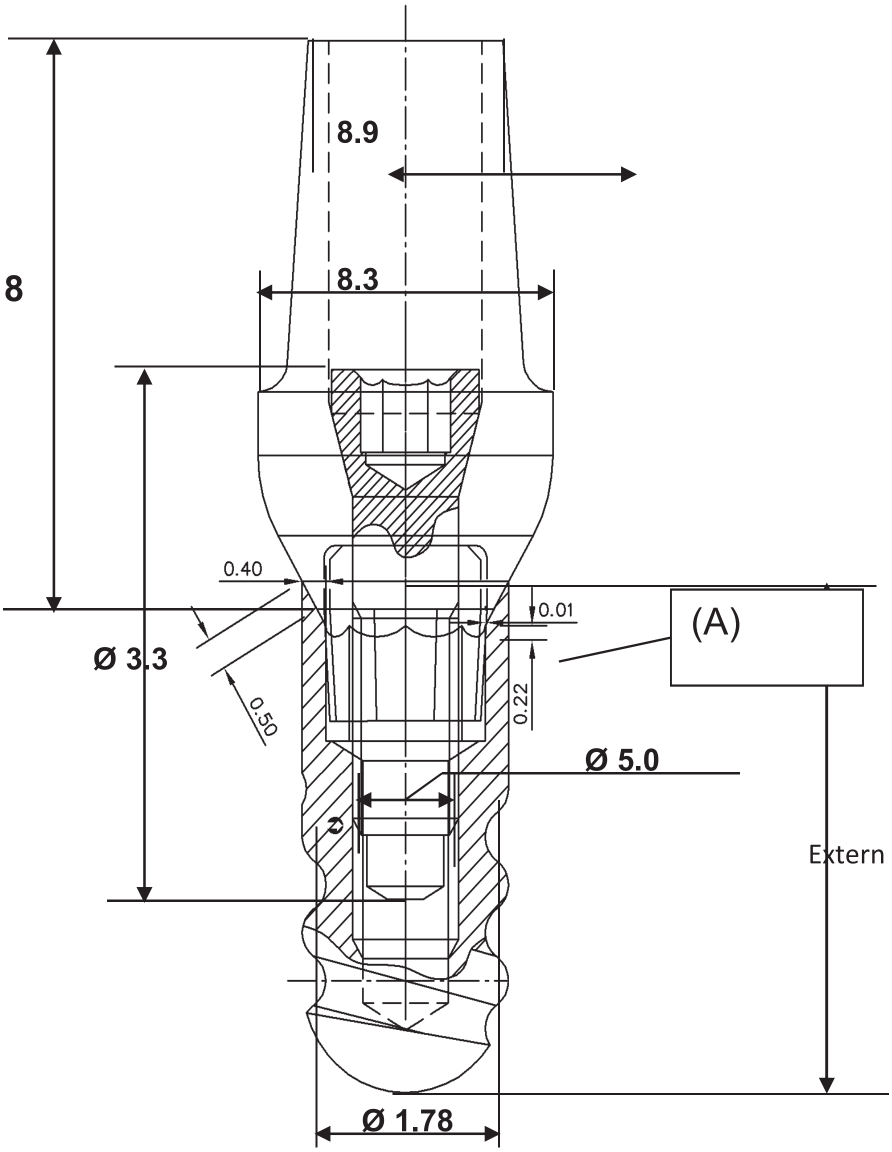

Figure 6.2: The design of the implant with abutment and screw with external octagon connection.

Zirconia is much stronger in compression than in tensile. The MPS was used because material failure will occur when the MPS exceeds the tensile strength of the material in any point.

Material properties

Both models consisted of a titanium grade IV implant, the surrounding bone, an abutment of titanium alloy and zirconia respectively, and a fixation screw of titanium alloy.

The material data used in this model are supplied by Dyna Dental Engineering for the abutment, implant and screw. Data for the bone are from literature [10]. The material data are shown in Table 1.

The interface conditions

The interface between the abutment and the implant and the interface between the conical part of the abutment screw and the abutment was designed as a contact surface. The friction coefficient between all contact surfaces was assumed to be 0.5 [11]

The interface between the external octagon and the implant was assumed bonded, taken in consideration the design of the external octagon.

The implant was assumed to be osseo-integrated with the bone and therefore the interface was assumed to be bonded.

Constraints and loads

In all models the block simulating the bone surrounding the implant was constrained at the bottom, all nodes on this surface were assumed to be fixed; no translation or rotation was allowed in any direction.

The fixation of the abutment screw over the screw thread surface in the implant in the radial direction was simulated by springs with high stiffness. In the axial direction a pre-load on the nodes on the screw thread surfaces of the abutment screw and the implant of 450 N was applied, this corresponds with an applied torque of approximately 320 N.mm. This tightening force is in line with the findings of Tan and Nicholls [12].

This study assumed a bite force on these incisors of 220 N, which is about the maximum normal bite force [13]; although it was reported by Nishigawa [14] that the maximum bite force during sleep associated bruxism can exceed this value for individuals. The bite force was applied under an angle of 45 degrees distributed evenly over the nodes in the top surface of the abutment.

6.4 Results

The highest stresses in all models are shown in Table 2, due to the simplification of the models the stresses occurring at the top part of the abutment and at the bottom of the bone are not realistic and for this reason were not taken into consideration.

Stresses in the implant with abutment and screw with internal octagon connector

Fig. 6.3 shows the mps of the stresses due to the combination of the bite forces and the forces due to the fixation screw in the implant with abutment and screw with internal octagon connection with the titanium alloy (A1) and zirconia abutment (A2). In the abutment the highest stresses occur in both models at the outside of the abutment at the sharp transition to the internal octagon, in the implant in the top at the sharp ending, in the abutment screw at the outside of the screw at the beginning of the conical part, and in the bone at the top part in contact with the implant, which is cortical bone.

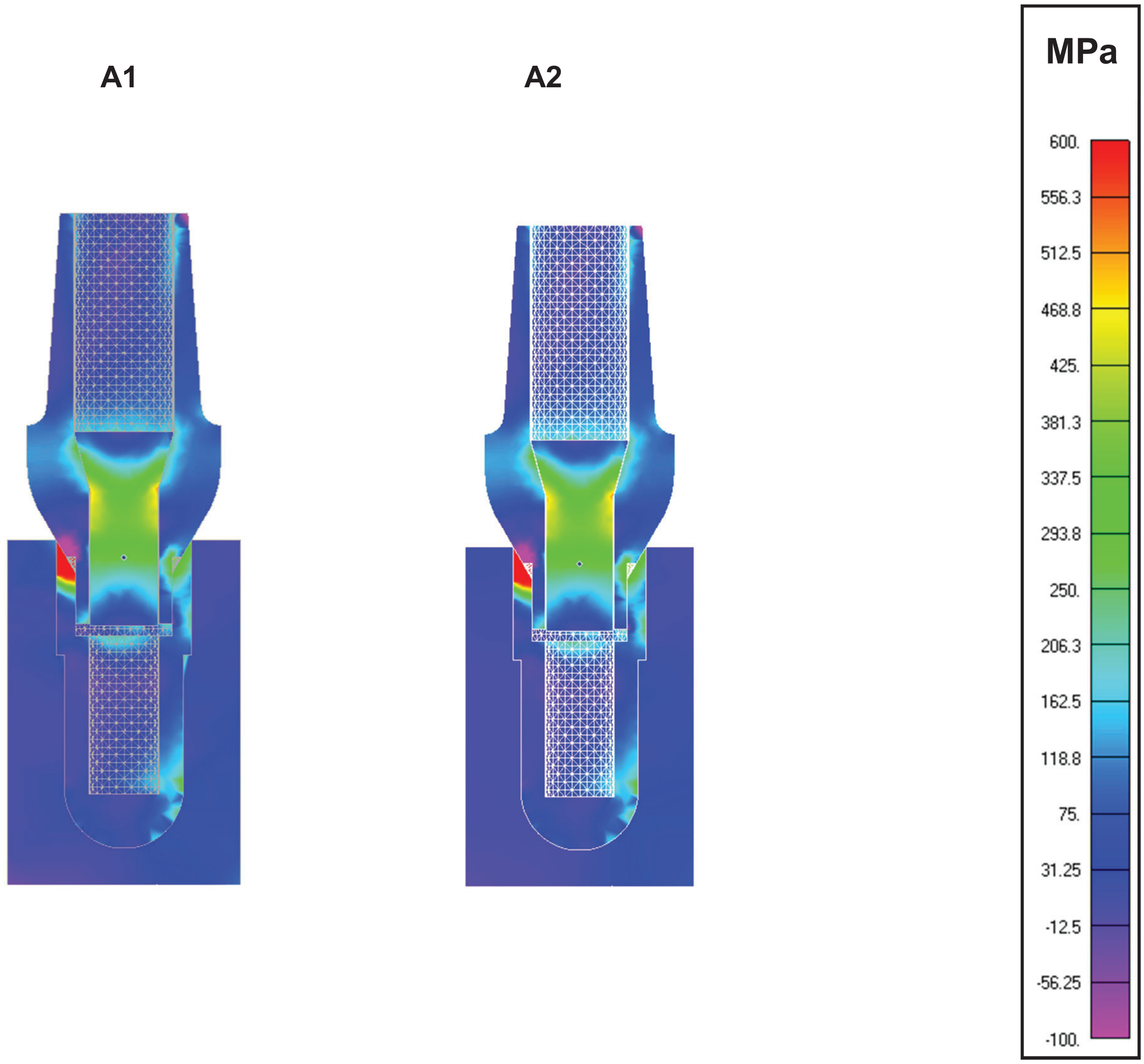

Figure 6.3: The stresses in the implant with abutment and screw with internal octagon connection with the abutment in titanium alloy (A1) and zirconia (A2).

Stresses in the implant with abutment and screw with external octagon connector

Fig. 6.4 shows the mps of the stresses due to the combination of the bite forces and the forces due to the fixation screw in the titanium alloy (B1) and zirconia abutment (B2) with external octagon connection. In the abutment the highest stresses occur in both models at the inside of the abutment just above the abutment screw, in the implant in the top at the sharp ending, in the abutment screw at the outside of the screw at the beginning of the conical part, and in the bone at the top part in contact with the implant.

Both systems

Fig. 6.5 shows the system with the internal octagon with titanium alloy abutment (A1) in the deformed mode. It can be seen that the abutment is sliding on the contact surface with the implant and the formation of a micro-gap due to the forces is shown. The abutments in all combinations were sliding due to the forces on the contact surfaces and forming a micro-gap.

6.5 Discussion

The models were realized without modelling the screw thread of the implant. Although the implant design might cause significant variations in stress distribution in the bone, the difference between cylindrical and screw-shaped implants is small [15]and the influence of this simplification on the stress distribution in the implant with abutment and screw might be negligible. Chun *et al* [9] neglected in their study the preload caused by tightening the abutment screw. However, the preload is influencing the stresses and deformation in the implant and as a consequence the stresses in the bone

The highest tensile stress in the implant with abutment and screw with the internal octagon connection (Fig. 6.3 and Table 6.2) was in the titanium alloy and zirconia abutment 448 MPa and 506 MPa respectively.. The yield strength of the titanium alloy is 789-1013 MPa [3] and the strength of the zirconia material is 1074-1166 MPa[4]. However, this strength is highly influenced by the surface roughness and can be reduced to almost half of this value [16] . In the clinical situation, when the surface finish in the corner of the octagon is not perfect, the stresses in the zirconia abutment in both executions might result in failure, especially after the fatigue effect of mastication. In the implant the highest stress was 712 MPa and 787 MPa for the titanium and zirconia abutment respectively. These stresses are close to the yield strength. In the abutment screw the stresses remained well below the yield strength. In the bone the highest stress was 34 and 36 MPa for the titanium and zirconia abutment respectively. These stresses are lower than reported by Chun *et al* [9], however, they neglected the influence of the preload caused by tightening the abutment screw. The highest stresses in the bone were in the cortical bone and are well below the strength of the bone [17].

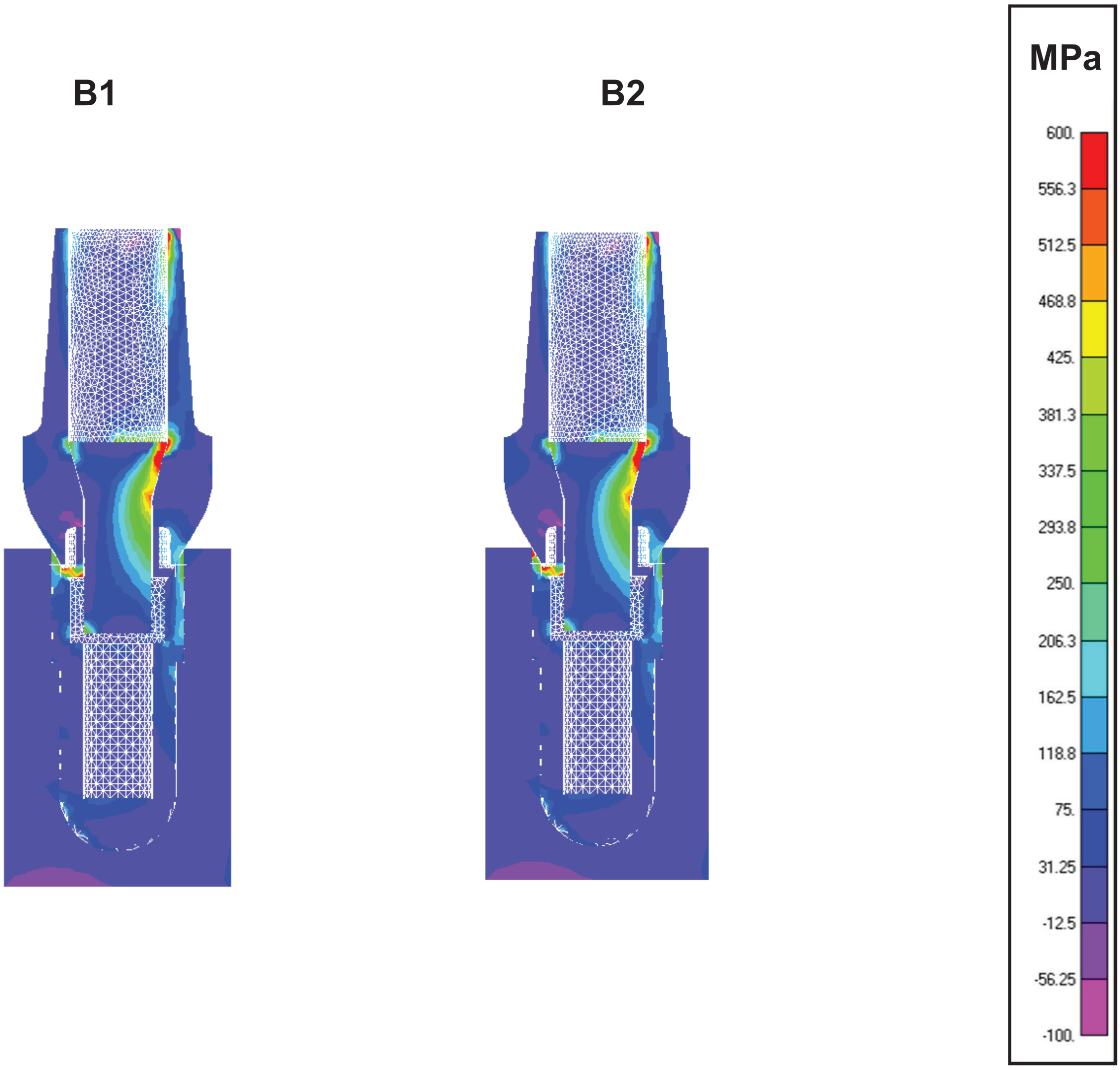

Figure 6.4: The stresses in the implant with abutment and screw with external octagon connection with the abutment in titanium alloy (B1) and zirconia (B2).

The highest tensile stress in the implant with abutment and screw with the external octagon connection (Fig. 6.4 and Table 6.2) was in the titanium alloy and zirconia abutment 278 MPa and 260 MPa respectively; these stresses are well below the strength of the material. The highest stress occurred at the inside of the abutment just above the abutment screw. The design of the abutment with external octagon shows in this respect to be better than the internal octagon design. The highest tensile stress in the implant (Table 6.2) was 1288 MPa and 1180 MPa for the titanium and zirconia abutment respectively. These stresses are just above the yield stress of the titanium alloy and might give deformation of the implant to the point where a thicker part of the implant will support more. The highest stress in the bone was in the cortical bone, 53 MPa for both abutment materials. This is below the strength of the cortical bone. However, eventual deformation of the implant might cause persistent inflammation of the tissue at the implant –abutment interface. In the abutment screw the stresses remained below the yield strength of the material.

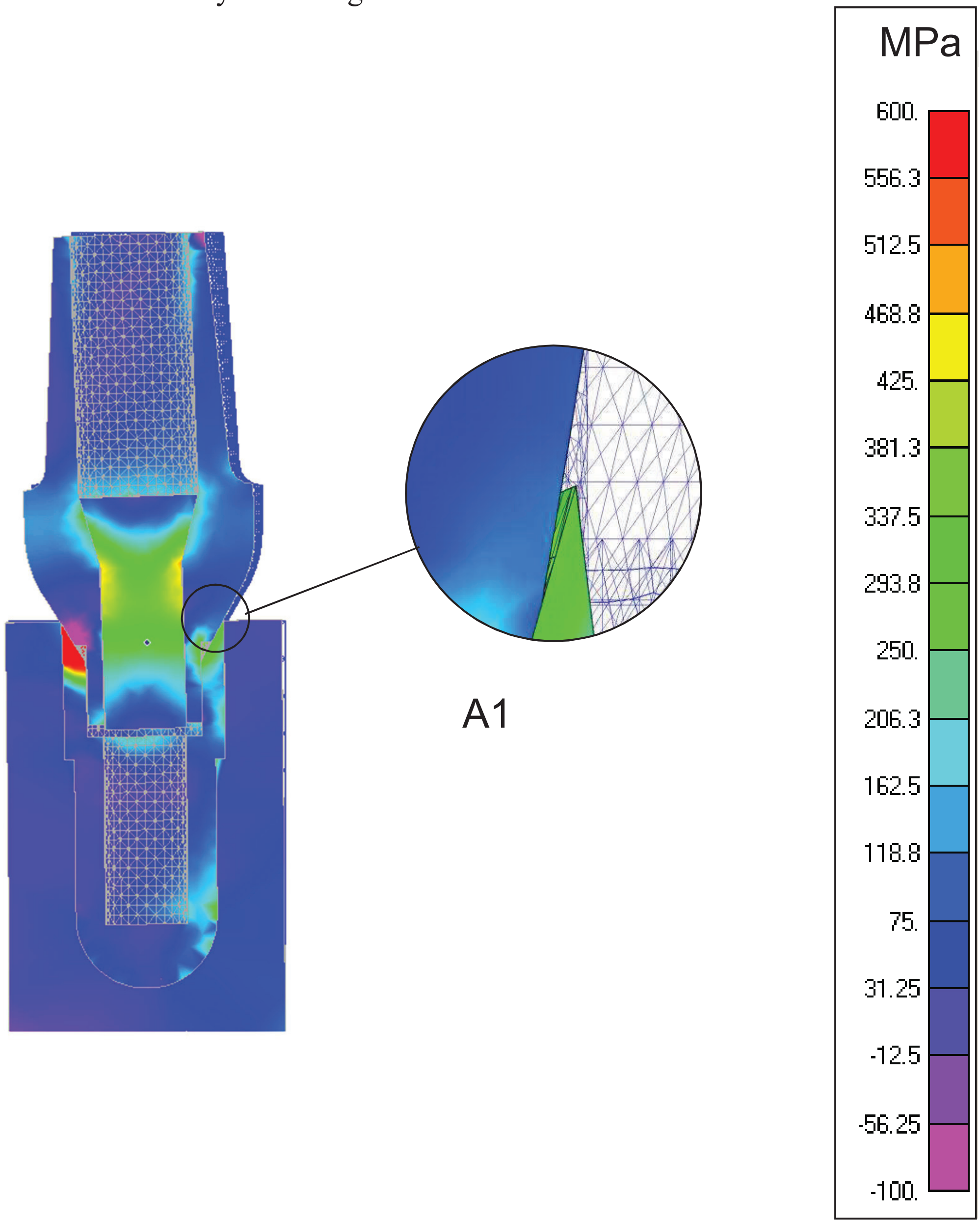

Figure 6.5: The stresses in the implant with abutment and screw with the internal octagon connection with titanium alloy abutment (A1) in the deformed mode.

The design of the implant with abutment and screw in this study for the execution with internal as well as with external octagon connection is the "one-piece" design with no micro-gap at the alveolar crest level as in the study of Boggini *et al* [18].

This “one-piece” design showed less inflammation in their study and experience with the Ankylos system with a design with no micro-gap at the alveolar crest level showed a high survival rate [19].

However, the possible deformation of the implant will lead also to micro-gap formation. Moreover, all implant-abutment combinations showed sliding of the abutment over the contact surface with the implant (Fig. 6.5). This is in line with the findings of Kitagawa *et al* [11]. This sliding caused a micro-gap, as can be seen in detail in Fig. 6.5. The inflammatory process might be reinforced by the “pumping effect” of the formation of this micro-gap under the bite forces. This “pumping effect” might explain the differences found by Broggini *et al* [18] for different designs, while micro-leakage is unavoidable among current implant systems regardless of the connection type or interface size [20]. The highest tensile stress in the abutment screw was between 586-763 MPa for the different implant-abutment combinations (see Table 6.2).

Implant –abutment combination	**MPS (MPa)**			
Internal octagon connection	**Abutment**	**Implant**	**Screw**	**Bone**
Titanium abutment A1	448	712	586	34
Zirconia abutment A2	506	787	586	36
External octagon connection	**Abutment**	**Implant**	**Screw**	**Bone**
Titanium abutment B1	278	1288	763	53
Zirconia abutment B2	260	1180	742	53

Table 6.2: The maximum principle stresses (MPS) in the models

Due to the fatigue effect during mastication, these stresses might result in screw loosening. Cibirka *et al* found lower detorque values after fatigue testing [21], although Butz *et al* did not find screw loosening in their study [22].

6.6 Conclusions

This study indicates that to exploit the high strength of zirconia as abutment material the actual distribution of the tensile stresses and the design of the dental implant system must be taken into account.

The highest tensile stress in both executions of the implant with abutment and screw with the external octolink connection was too high in the implant. Eventual deformation of the implant might cause persistent inflammation of the tissue at the implant –abutment interface. The abutment-implant combination with internal octagon showed to be a better design, although due to sliding of the abutment over the contact surface with the implant,
these type of implants with abutment and screw showed a micro-gap under the bite forces. The “pumping effect” of the formation of this micro-gap under the bite forces might cause an inflammatory process.

6.7 References

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