Acta of Bioengineering and Biomechanics
Vol. 17, No. 1, 2015

Original paper
DOI: 10.5277/ABB-00091-2014-02

Strength analysis of a three-unit dental bridge framework with the Finite Element Method

ŁUKASZ REIMANN*, JAROSŁAW ŻMUDZKI, LESZEK A. DOBRZAŃSKI

Institute of Materials Engineering and Biomaterials, Faculty of Mechanical Engineering,
Silesian University of Technology, Gliwice, Poland.

Purpose: The aim of the study was to analyse the strength of a prosthetic bridge with variable geometry in the connectors between the span and the retention elements on the pillar teeth crowns. Methods: Research was carried using the Finite Elements Method (FEM) on a model of the bridge in the anterior teeth arch in the field 21–22–23, obtained using a contact scanner and computer aided design (CAD) system, with four different cross-sectional areas of the connectors: 4.0, 5.0, 5.5, and 6.0 mm^2. For that purpose, the impact of the properties of selected metal alloys on the deflection of the prosthesis was analysed. Results: On the basis of the analyses, it was found that when the loading force acted obliquely, the stress was 19% higher compared to the stress with a loading vertical force. In the case of connectors with the smallest cross-sectional area, the stress exceeded permissible value (with safety factor $n = 2$) for one of the alloys. Conclusions: Deflection of the bridges tested changed depending on the connector cross-section and the elastic modulus of the selected material.

Key words: *dental bridge, Hubert–Misses stress, connectors, materials selection*

1. Introduction

Healthy teeth have been a concern for humans since the beginning of civilization and are also associated with the problem of solving the rebuilding of damaged and rotten teeth, leading to the development of dental prosthetics. In human history, prosthetics had already appeared around 5000 BC, for which proof can be found in a tomb from the area of Saida in Lebanon, where a bridge rebuilding two missing teeth in the mandible, connected with the rest of the teeth by means of a gold wire, was found. A prosthesis made of ivory is available in the Louvre Museum. At the present time, with the improvement of the living conditions and wealth of communities, there is a growing requirement with regard to health status for the implementation of new technology materials for dental engineering [21] and a demand for better dental prostheses [4], [6].

Most common is the construction of partial dentures on a substructure of metal alloy, which is divided into four groups according to the yield strength value (Fig. 1). For each group of materials, alloys with different chemical compositions of high noble alloys, noble alloys, and base alloys may be classified. An important factor determining the quality of a prosthetic bridge is the design and construction of the span, which should take into account the strength. The existing criteria for the selection of metal alloy dentures classification are in accordance with the yield strength [2], [20]. Durability depends not only on the material from which the dental bridge is made but also on its span and shape. Meanwhile, advanced computer aided systems for the manufacture of dental bridges provide the possibility of designing high accuracy dentures with individual anatomic features, but the displacements and stresses are unknown. The finite element method (FEM) is used for strength analysis

* Corresponding author: Łukasz Reimann, Silesian University of Technology, Institute of Materials Engineering and Biomaterials, ul. Konarskiego 18A, 44-100 Gliwice, Poland. Tel: +48 696 83 65 05, e-mail: lukasz.reimann@polsl.pl

Received: May 5th, 2014
Accepted for publication: July 10th, 2014

of dental prostheses and appliances [14], [25]. The results are dependent on the biomechanical factors and simulation conditions. These factors include, for example, the value of the occlusal load, which is assumed in the analysis to be at the level of 50 N [19], although average forces recorded on incisors are in the range of 200–300 N [3]. The factor affecting the results of the analysis is the assumed occlusal force per tooth [19] or the distribution of occlusal forces on different parts of the bridge [5]. Although previous analysis of tooth-supported dental bridges shows the general influence of stiffness of the bridges on their strength, but due to some schemes of occlusal loads they are difficult to univocal interpretation and application to the individual denture design The direction of the occlusal force has a significant impact on distribution and stress value in dentures and implants [10], [25]. Despite this fact, there is no explicit comparison of the effect of vertical or oblique occlusal loads on strength of the bridge framework, which allows the principles to be established in modeling loads in individual denture design.

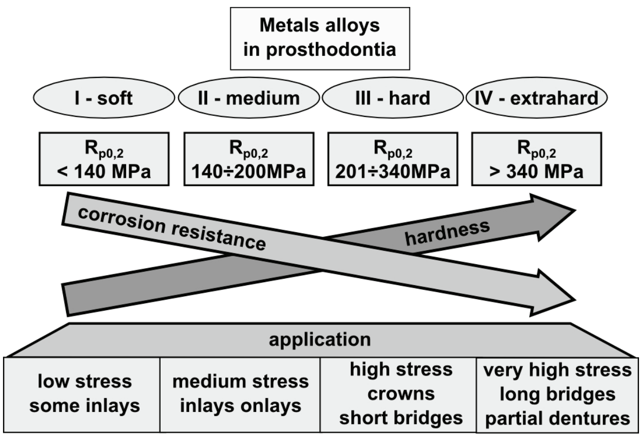

Fig. 1. Classification of metallic alloys used in dental engineering by properties [20]

The aim of the study was an FEM analysis of occlusal load transfer in a three-unit denture framework in the incisor region depending upon the metal alloy substructure and the cross-section of the connectors between the tooth span and retention elements on the abutment teeth of the dental bridge. The hypothesis of the study was that the design of a three-unit denture framework in the incisor region done on the basis of the stresses under occlusal vertical load may not provide load-bearing capacity in the real conditions under oblique loads. The individual prosthetic field was scanned and numerical models of dental bridges were designed with the professional CAD dental system.

2. Research methodology

Research was carried out on the numerical model of a bridgework prepared on the basis of a clinical case in which incisor tooth 22 was missing in the tooth arch. On the basis of the impression tray with a registered upper dental arch, a plaster model of the prosthetic field was made. Then, according to the laboratory procedure, a plaster model was prepared (Fig. 2).

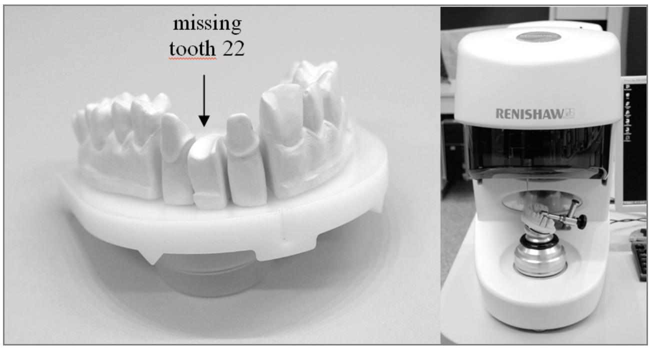

Fig. 2. Prepared plaster model with a plaster template for restoration of the missing tooth in the upper teeth arch using a Renishaw 3i Incise dental contact scanner

The simulation of the occlusal load transfer required the preparation of a numerical model of the prosthesis as a solid model, which procedure consisted in [7]:

- saving the physical model of the prosthetic foundation as a numerical form on the basis of the coordinates of the points,
- designing the dental bridge on the numerical model of the prosthetic foundation,
- creating the solid model.

The physical plaster model was transformed into numerical form by using a Renishaw contact scanner DS10 (Fig. 2), in which the working elements were two contacting probes with diameters of μ and 3 mm. The dental scanner specifications were a measuring area of 90 mm in diameter and 45 mm in height, contact probe force of 0.5 N/mm, measuring accuracy of 20 μm, scan interval of 0.1 mm, and scan speed of 600 mm/min.

The model was drawn and saved in digital form for five plaster models corresponding to teeth 11, 21, 22, 23, and 24 (Fig. 3). The registered tooth arch model was saved in the STL file format and used for the design in a CAD environment using DentCAD Delcam software. Four variants of the dental bridge were designed with a variable cross-sectional area of the connector between the pontic and the crowns on the pillar teeth. The thickness of the crowns on pillars was the same in all the cases and was 0.5 mm. The bridge dimensions are presented in Fig. 4. Bridges with pontics with smaller cross-sectional areas were not examined in the study due to the need to increase the thickness of the aesthetic porcelain, which may result in faster damage.

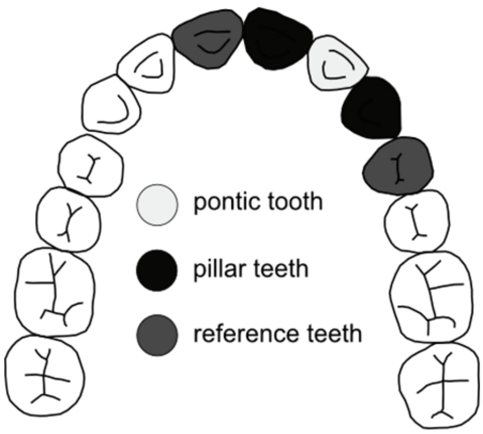

Fig. 3. Selected points for preparing the numerical model of the bridge

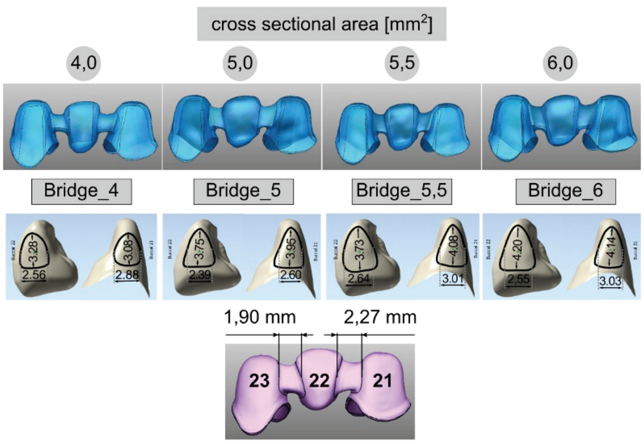

Fig. 4. Characteristic of the bridges analysed

The strength analysis of the bridges was done using FEM and ANSYS Workbench software. Bridge models were meshed with tetrahedral 10-node elements. The numerical convergence analysis was done by concentrating the mesh by decreasing the size of the finite elements. A size of about 0.2 mm of the finite elements in the bridge connector area was determined and used in the FEM analysis on the basis of the displacement and energy error norm in each element (SERR). Table 1 presents the number of elements and nodes in the models prepared for the research.

Table 1. Finite element mesh parameters of the bridge framework models

Dentures signature	Numbers of nodes in finite element mesh	Numbers of elements in finite elements net
Bridge_4	35777	20229
Bridge _5	36466	20758
Bridge _5,5	36672	20912
Bridge _6	37247	21258

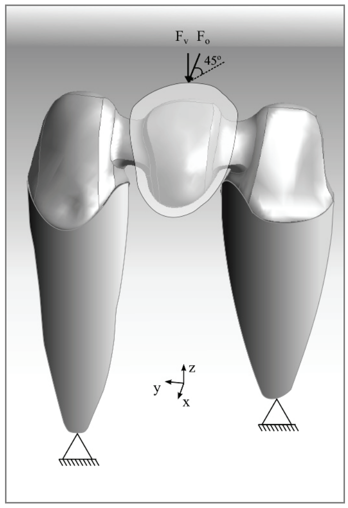

Fig. 5. Boundary conditions of the simulation: scheme of support and loads (F_v: vertical load; F_o: oblique load)

The loads of occlusal forces were taken on the basis of values developed during food mastication by healthy adult men. An average occlusal force of 250 N was assumed for tooth 22 [3]. The loading was put on the incisor edge. Two variants of the loading were analysed: a force acting perpendicularly to the occlusal surface and a force acting obliquely at 45° from tongue to lip (Fig. 5). Simplified support of the model was assumed in which the crown–tooth interface was fixed and individual tooth movement was omitted. Two alloys from each group of metallic materials: high noble, noble, and base alloys, were selected for the numerical research (Table 2); they are characterized by significant differences in yield strength and Young's modulus.

Permissible stress was determined on the basis of information about the yield stress of dental alloys with taking into account a safety factor, which was assumed to be 2.0.

Table 2. Chemical composition and material properties of the metal alloys used in the research

Alloys group	Sign.	Material	Chemical composition % [mas.]						Young's modulus [GPa]	Poisson's ratio	$R_{p0,2}$ [MPa]
High noble	W1	JELENKO Benchamark C	Au 87	Pt 11	Fe 0,2	In 0,6	Sn 0,6		69	0,30	350
	W2	ELEPHANT Orion WX	Au 84	Pd 5	Pt 8	Ag 0,9	Cu 0,4	In 1	124	0,30	650
Noble	S1	IVOCLAR W-1	Pd 53,3	Ag 37,7	Sn 8,5	In 0,3	Ru 0,2		114	0,30	450
	S2	JELENKO Jel 55	Pd 55	Ag 34	In 6	Zn 1	Sn 3	Ga 0,7	125	0,30	724
Base	N1	BEGO Wirobond 280	Co 60,2	Cr 25	W 6,2	Mo 4,8	Ga 2,9	Si 0,5	220	0,30	540
	N2	DENTAURUM Remanium 2000+	Co 61	Cr 25	W 5	Mo 7	Mn 0,5	Si 1,5	200	0,30	700

4. Results

In the strength analysis of the denture framework, the occlusal force transfer, the equivalent Huber–Misses stress (H-M), and displacement in the area of pontic teeth were taken into account.

The H-M stresses in frameworks with minimal and maximal connector cross-sectional area are shown in Fig. 6, while Fig. 7 shows a comparison of all frameworks. The highest values of stress were read from the connector area, and the displacement was read from the area of the pontic teeth close to the gingiva. A maximum H-M stress of 187 MPa was registered for the bridge with connectors with the smallest cross-sectional area of 4.0 mm^2 and an oblique load (Fig. 6b). For the bridge with connectors with the cross-sectional area of 5.0 mm^2, the H-M stress was 171 MPa; for the cross-sectional area of 5.5 mm^2, the stress value decreased to 157 MPa, and for the cross-sectional area of about 6.0 mm^2 the stress was the smallest and was 152 MPa (Fig. 6d). Comparing the stress values for all four connectors, it was found that the maximum stresses occurred in an oblique force loaded prosthesis.

Analysis of the bridge with safety factor $n = 2$ revealed that for the connector with the smallest cross-section (4.0 mm^2) the use of such a material as W1 with $R_{p0,2}$ 350 MPa is not recommended. The analysed remaining materials were characterized by higher yield strength, so their use guarantees strength reserve in relation to the maximum H-M stress for material identified as S2 almost double, even for bridges with 4 mm^2 connectors (Fig. 7).

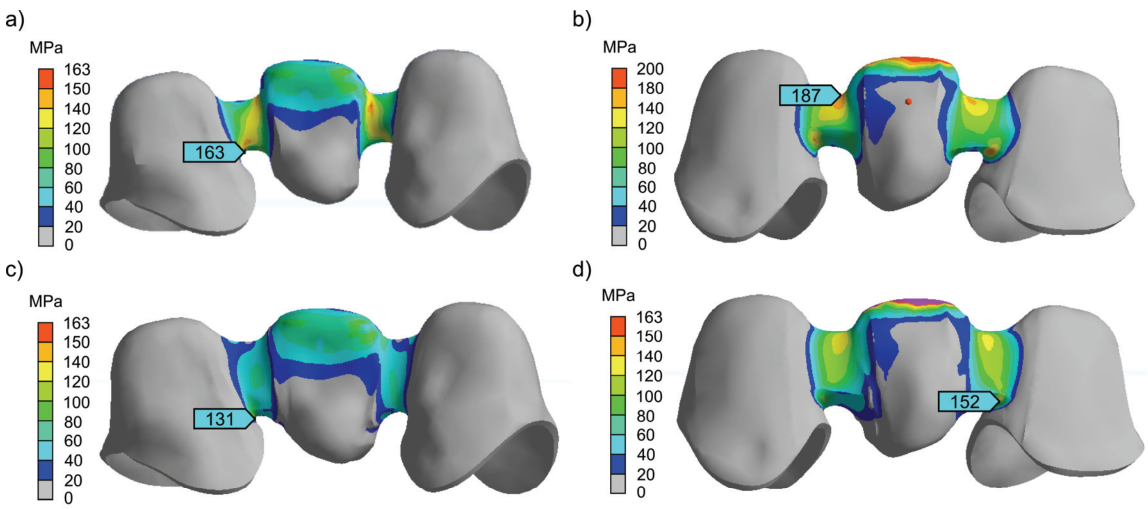

Fig. 6. Equivalent stress distribution in the dental bridge with connector cross-sections of:
(a) and (b) 4 mm^2, (c) and (d) 6 mm^2; vertical loading force: (a) and (c); oblique loading force: (b) and (d)

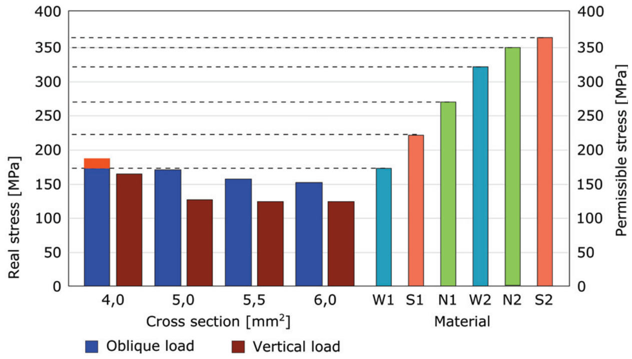

Fig. 7. Huber–Misses stresses, depending on the area of the connector cross-section with a load of force directed perpendicularly and obliquely and strength reserve obtained by applying specific materials for the construction of bridges under investigation

Based on the analysis of the bridge displacements, it was found that the connector's cross-sectional area between the span and retention elements affects the amount of displacement of the prosthesis as shown in Figs. 8 and 9. The largest displacement value, which was 11.5 μm, was found for the bridge investigated when the 4.0 mm^2 connectors and an oblique load were applied, and the smallest displacement, which was 1.0 μm, was found for connectors with a cross-sectional area of 6.0 mm^2 with a force load directed perpendicularly. The biggest difference in displacement due to the type of applied load was found when comparing the results of the calculation for the bridge with connectors with a cross-sectional area of 6.0 mm^2. When the force load directed perpendicularly to the

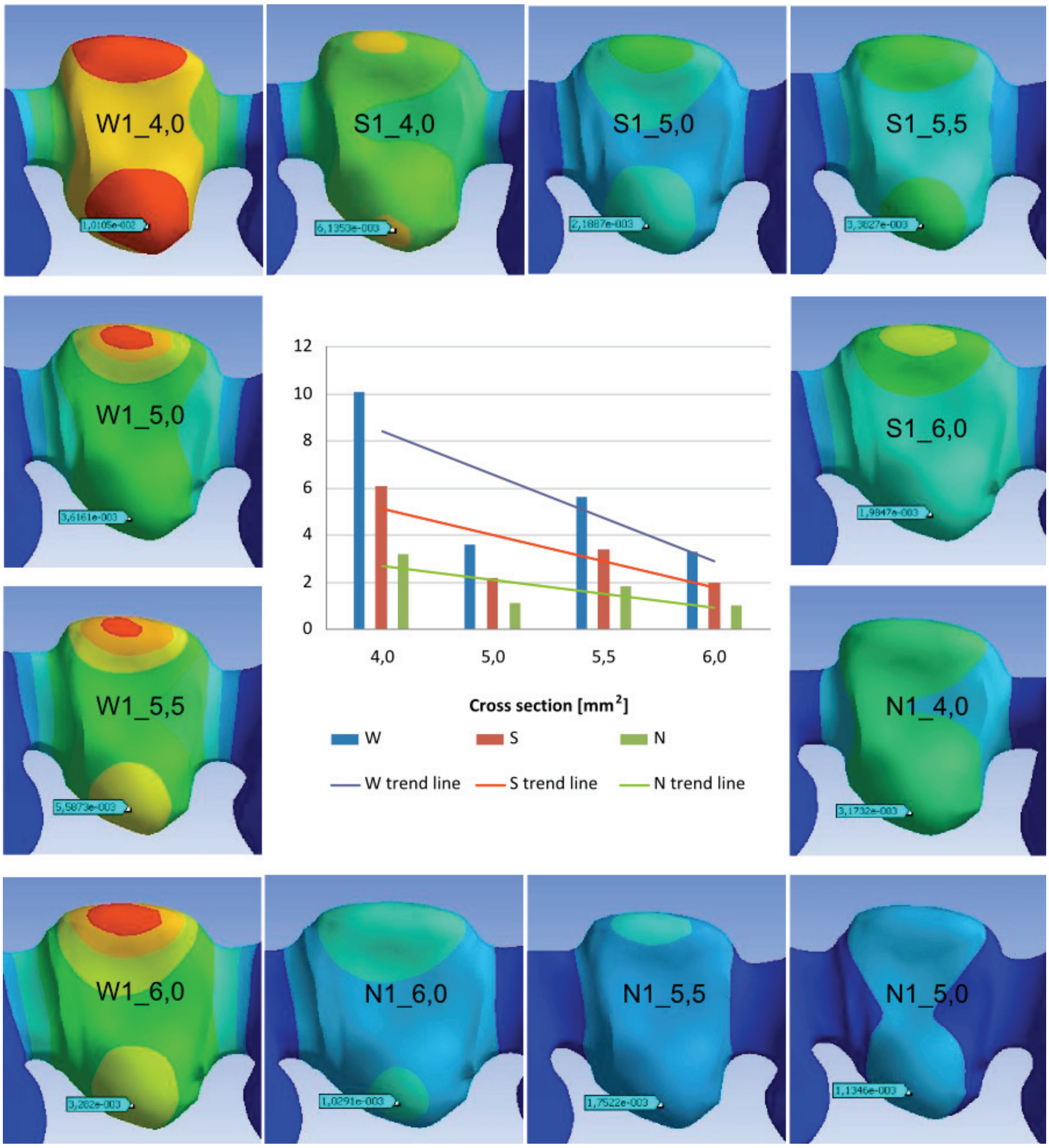

Fig. 8. The influence of the connector cross-sectional area on the value of resultant displacement of the bridge when loaded by a perpendicular force

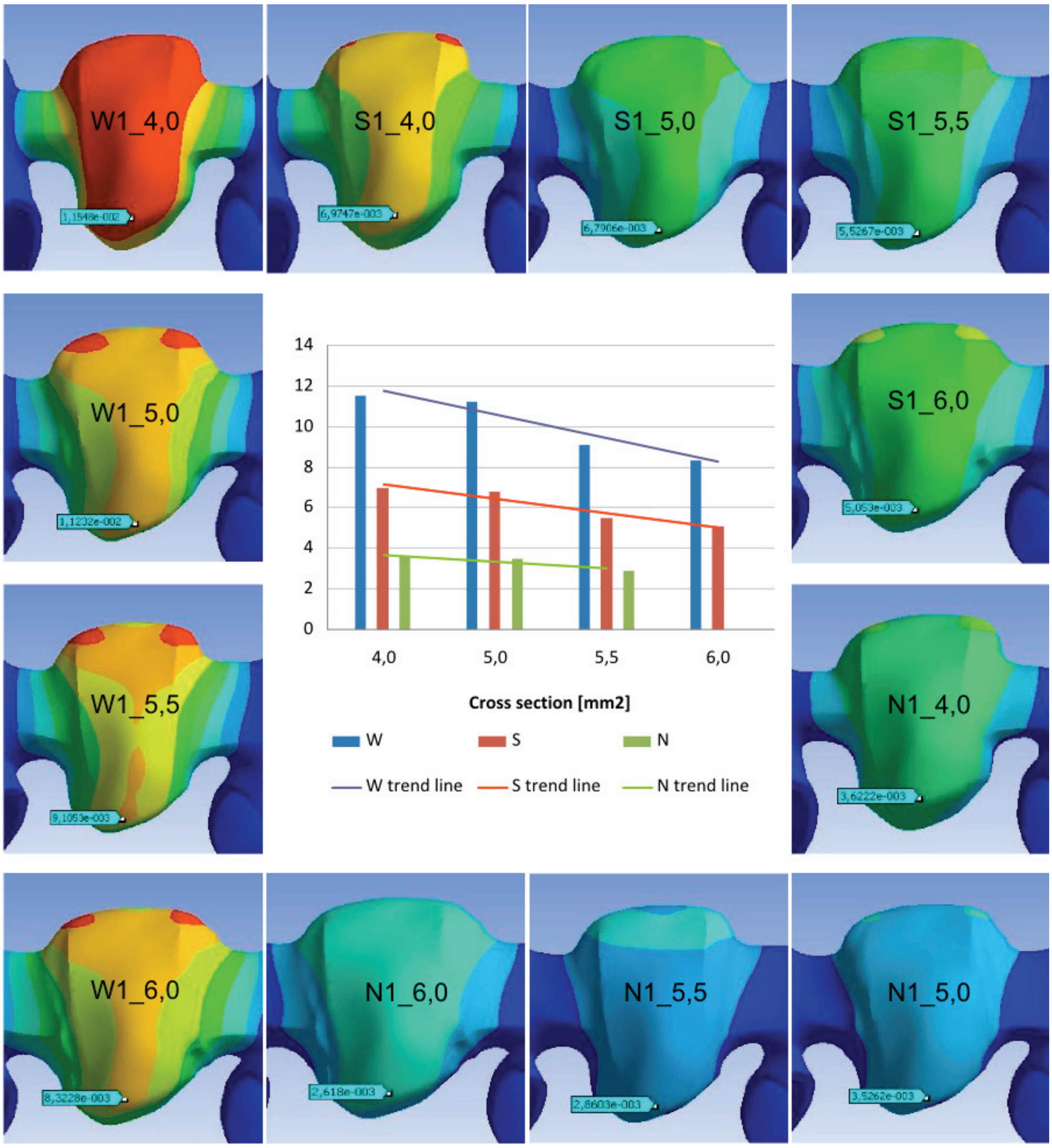

Fig. 9. The influence of the connector cross-sectional area on the value of resultant displacement of the bridge when loaded by an oblique force

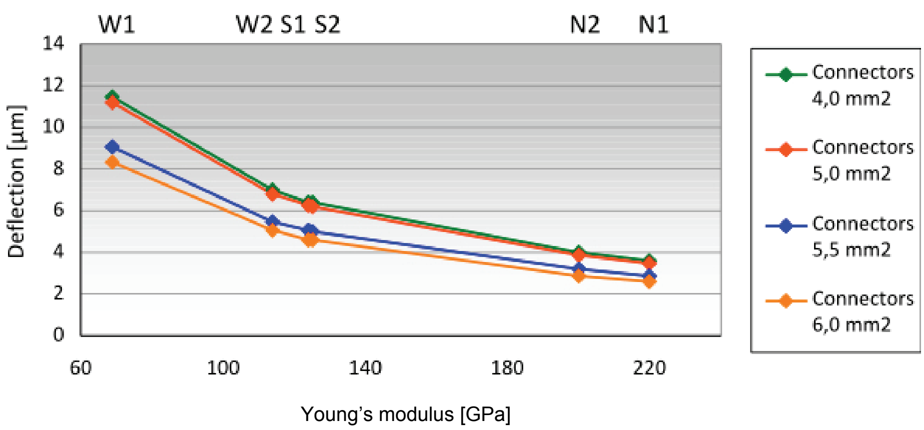

Fig. 10. Influence of Young modulus of metal alloys used in the substructure of the bridge on the value of deflection

alloy designated as N2, the recorded resultant displacement of the bridge was 1.1 μm, while the displacement of the prosthesis by an oblique load was 2.9 μm, which was more than two and a half times greater. Similarly, significant differences in the displacement values of the denture for the type of applied load were found for the other alloys. An analogous relationship was observed by the research with regard to the amount of movement relative to the *Z* axis. On comparing the relation of displacement on the *Z* axis to the accidental displacement under different load conditions, it was found that when the load on the bridge is an oblique force the values differ by more than 220%. Displacement on the *Z* axis directed toward the alveolar gingiva has been adopted as a criterion for evaluation.

The impact of the material properties of six metal alloys used in the manufacture of prosthetic bridges on the amount of displacement according to the simulation research results is shown in Fig. 10.

5. Discussion

In the dental prosthesis analysed, during transfer of the vertical force, the stress in the bridge framework decreased in relation to the value recorded for the smallest cross-section (Fig. 8) in the range from 24 to 26% for connectors with cross-sectional areas of 5.0 to 6.0 mm^2. In the case when the load on the bridge was an oblique force, the stress values of the prostheses with connectors of 5.0 and 6.0 mm^2 decreased by about 9 and 19%, respectively, compared with the bridge with connectors of 4.0 mm^2. Similar conclusions were reached by the authors of [17], who, based on the H-M stress, stated that increasing the cross-sectional area of the connectors by 4 and 8 mm^2 increased the strength of the three-point prosthetic bridge by 33 and 55%, respectively. In strength tests [5], [14], [20] of the value of stresses and displacements in dental bridges depending on the cross-sectional area of connectors between pillars, a reduction of the cross-sectional area from 15.7 to 10.1 mm^2 led to stresses in the substructure that were almost 100% higher, which is almost the same difference as that shown by the results presented, where the difference in stress as a result of reducing the connectors' cross-sectional area by 2 mm^2 caused an approximately 35% increase in stress. Taking into consideration the differences in geometry in the restoration models analysed and accepted way of supporting the model, the simulation tests performed can be considered to be correct.

Other studies [8], [14] showed the relation of stress values of the connector joint geometry with the tooth span, especially the radius of the rounding of the connection, because its increase associated with an increased connector cross-sectional area resulted in a reduction in the stress of up to 30%.

It should be noted that the direction of the loading force on the bridge has a significant impact on the criteria values of stresses. The biggest difference in stress between the case of oblique and vertical force loads was greater than 25 MPa. Larger values of H-M stress in the case of loading by an oblique force are the result of bending that occurs in the second plane and torsion. It must be emphasized that the incorrect assumptions of the method of loading the prosthetic bridge only by a vertical force lead to significant underestimation of the stress, which translates directly into an incorrect choice of the connectors' cross-sectional area or material whose strength properties are insufficient. Significant differences in the stress values arising from the direction of the applied force occurred on implants in research [9], where the stress of the oblique load was on average 35% higher than the stress of a vertical load. In studies of endodontic posts [1], [22], differences in stress for oblique and vertical loads were quite significant and ranged up to 60 MPa. The application of an oblique load to overdentures in work [25] increased the stress values more than 20 times compared to the value of the vertical force. The huge difference was due to the loss of stability of the supports of the removable complete denture, whereas the support of bridge prosthesis is stable. Nevertheless, in the light of the comparison of the stress under vertical and oblique load and reported impact on the results of torsional stiffness of the bridge it should be noted that the approach to the analysis of strength in many recent works [11], [12], [16], [23], [24] is not eligible to individual design of bridges.

Designating the stress values in the simulation research makes it possible to select the material according to the engineering criteria for permissible stress. Reducing the connectors' cross-sectional area to 4.0 mm^2 caused an increase in the real stresses level of 187 MPa, and therefore the permissible stress of the Au-Pt alloy (Jelenko Benchmark C) was exceeded by about 7%. Designing the bridge from material having the highest permissible stress of 362 MPa (Jelenko Jel 55) allows keeping the reserve to the permissible stresses of approximately 94%. Strength reserve measured for the Jelenko Benchmark C material is about 15% after increasing the connectors' area, while that for the Jelenko Jel 55 material is almost 140%

(Fig. 7). Analysis of the bridge deflection results under the influence of an oblique load showed that, relative to the bridge with the smallest connector cross-section of 4.0 mm^2, the critical vertical displacement of the bridge with a connector cross-section of 5.0 mm^2 was reduced by about 3%, while in the case of the connectors of 5.5 and 6.0 mm^2 the displacement was reduced by 21 and 28%, respectively. The values of vertical deflections are an important criterion in designing bridges. In designing a prosthetic bridge, the displacement of the span in relation to the space beneath the bridge should be considered, because excessive deflection can result in compression of the mucous membrane. The deflection of the bridge can also be controlled by Young's modulus of the framework material. Based on the numerical analysis of the bridges for the material with the lowest Young's modulus of 69 GPa, it was determined that the deflection of the bridge is between 8 and 12 µm (depending on the connector cross-section) whereas for the material with the highest modulus of 220 GPa, the deflection was two to four times smaller (Fig. 10). Compared to dentures made of composite materials [19] with Young's modulus of 22 GPa, the application of metallic materials in the bridge substructure allows the deflection of the bridge span to be reduced by one order of magnitude, that is, from about 60 µm to an average of 6 µm, using metal alloys.

6. Conclusions

Based on FEM studies of the numerical simulation of occlusal load transfer on the substructure of bridges at anterior region, it was found that in the case when only vertical occlusion forces are acting there is a significant underestimation of the maximum stress in the substructure of 13 to 27% compared to the case of an oblique force in accordance with the loads prevailing in the oral cavity.

Hypothesis has been positively verified, since one of the analyzed three-unit denture framework bore the vertical loads but lost load-bearing capacity under oblique loads.

Exceeding the range of load-bearing capacity occurred when using the alloy with the lowest yield stress of the analyzed materials after reducing the connector cross-sectional area up to 4 mm^2. The research performed on stresses and displacements of the prosthetic bridge showed that decreasing the connectors' cross-sectional area by about 1 mm^2 increases the maximum stress by an average of 15%. Increasing the connectors' cross-sectional area by about 1 mm^2 reduces the deflection of the bridge by an average of 16%, while Young's modulus that is two or three times higher causes a decrease in deflection by an average of 45 or 65%.

FEM analysis makes it possible to design individual connector cross-section for prosthetic bridge substructures and to select material on the basis of the stress and deflection criteria of the bridge designed. In the work it was documented that the torsional stiffness must be taken into account in the strength analysis, therefore there was found motivation for further study how the denture strength depends on geometry of the connectors, especially on their bending and torsional stiffness

Acknowledgements

The research was partially funded by the Polish Ministry of Science and Higher Education under a research project No. N N507 438539.

References

[1] ADANIR N., BELLI S., *Stress analysis of a maxillary central incisor restored with different posts*, Eur. J. Dent., 2007, 1(2), 67–71.

[2] ANUSAVICE K.J., *Phillips' Science of Dental Materials*, Elsevier Health Sciences, 2003.

[3] CHLADEK W., *Biomechanika inżynierska narządu żucia. Zagadnienia wybrane*, Wydawnictwo Politechniki Śląskiej, Gliwice 2008.

[4] CHLADEK G., WRZUŚ-WIELIŃSKI M., *The evaluation of selected attachment systems for implant-retained overdenture based on retention characteristics analysis*, Acta Bioeng. Biomech., 2010, 12(3), 75–83.

[5] CORREIA A.R.M., FERNANDES J.C.S., CAMPOS J.C.R., VAZ M.A.P., RAMOS N.V.M., *Stress analysis of cantilever-fixed partial denture connector design using the finite element method*, Rev. Odonto Ciênc., 2009, 24(4), 420–425.

[6] DĄBROWA T., DOBROWOLSKA A., WIELEBA W., *The role of friction in the mechanism of retaining the partial removable dentures with double crown system*, Acta Bioeng. Biomech., 2013, 15(4), 43–48.

[7] DOBRZAŃSKI L.A., REIMANN Ł., *Digitization procedure of creating 3D model of dental bridgework reconstruction*, JAMME, 2012, 55(2), 469–476.

[8] DORNHOFER R., ARNETZL G.V., KOLLER M., ARNETZL G., *Comparison of the static loading capacity of all-ceramic bridge frameworks in posterior teeth using three hard core materials*, Int. J. Comput. Dent., 2007, 10(4), 315–328.

[9] FAWZI S., *The effect of dental implant design on bone induced stress distribution and implant displacement*, Int. J. Comput. Appl., 2013, 74(17), 15–21.

[10] FEDERICK DR., CAPUTO AA., *Effects of overdenture retention designs and implant orientations on load transfer characteristics*, J. Prosthet. Dent., 1996, 76(6), 624–632.

[11] KERMANSHAH H., BITARAF T., GERAMY A., *Finite Element Analysis of IPS Empress II Ceramic Bridge Reinforced by Zirconia Bar*, J. Dent. (Tehran), 2012, 9(4), 196–203.

[12] LIN J., SHINYA A., GOMI H., SHINYA A., *Finite element analysis to compare stress distribution of connector of lithia disilicate-reinforced glass–ceramic and zirconia-based fixed partial denture*, Odontology, 2012, 100, 96–99.

[13] MILEWSKI G., HILLE A., *Experimental strength analysis of orthodontic extrusion of human anterior teeth*, Acta Bioeng. Biomech., 2012, 14(1), 15–21.

[14] MOLLERS K., PATZOLD W., PARKOT D., KIRSTEN A., GUTH J.F., EDELHOFF D., FISCHER H., *Influence of connector design and material composition andveneering on the stress distribution of all-ceramic fixed dental prostheses: A finite element study*, Dent. Mater, 2011, 27, e171–e175.

[15] OH W.S., ANUSAVICE K.J., *Effect of connector design on the fracture resistance of all-ceramic fixed partial dentures*, J. Prosthet. Dent., 2002, 87(5), 536–542.

[16] ONODERA K., SATO T., NOMOTO S., MIHO O., YOTSUYA M., *Effect of connector design on fracture resistance of zirconia all-ceramic fixed partian dentures*, Bull. Tokyo Dent. Coll., 2011, 52(2), 61–67.

[17] REZAEI A., HEIDARIFAR H., AREZODAR F., AZARY A., MOKHTARYKHOEE S., *Influence of connector width on the stress distribution of posterior bridges under loading*, J. Dent. (Tehran), 2011, 8(2), 68–74.

[18] ROMEED S.A., FOK S.L., WILSON N.H.F., *Finite element analysis of fixed partial denture replacement*, J. Oral. Rehabil., 2004, 31, 1208–1217.

[19] SAKAGUCHI R.L., POWERS J.M., *Craigs Restorative Dental Materials*, Elsevier, 2012.

[20] SHINYA A., YOKOYAMA D., *Finite Element Analysis for Dental Prosthetic Design*, Finite Element Analysis, In. Tech., 2010.

[21] SCHOENBAUM T.R., *Dentistry in the Digital Age: An Update*, Dent Today, 2012, 31(2), 112–113.

[22] SHUKRI B.M.S., AL-HASHIMI W.N., *Finite element stress analysis of endodontically treated teeth restored by prefabricated posts*, J. Bagh College Dentistry, 2005, 17(3), 27–32.

[23] TAKUMA Y., NOMOTO S., SATO T., SUGIHARA N., *Effect of framework design on fracture resistance in zirconia 4-unit all-ceramic fixed partial dentures*, Bull. Tokyo Dent. Coll., 2013, 54(3), 149–156.

[24] THOMPSON M.C., ZHANG Z., FIELD C.J., LI Q., SWAIN M.V., *The all-ceramic, inlay supported fixed partial denture. Part 5. Extended finite element analysis validation*, Aust. Dent. J., 2013, 58(4), 434–441.

[25] ŻMUDZKI J, CHLADEK G, KASPERSKI J, DOBRZAŃSKI LA., *One versus two implant-retained dentures: comparing biomechanics under oblique mastication forces*, J. Biomech. Eng., 2013, 135(5), DOI, 10.1115/1.4023985.