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Research paper

A novel 3D-printed, patient-specific alloplastic temporomandibular joint replacement allowing enthesis reconstruction: A finite element analysis

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ABSTRACT

This study aims to evaluate the mechanical performance of a unique, state-of-the-art, patient-specific alloplastic total replacement system for the temporomandibular joint that allows for enthesis reconstruction developed by CADskills BV (Ghent, Belgium), and its influence on the remaining adjacent healthy tissue, by looking into the magnitude and location of the stresses and micromotions. Because the reattachment of the lateral pterygoid muscle is unique, having never before been used in temporomandibular joint prostheses, the loading patterns and performance may be completely different from existing devices and their analyses. Therefore, multiple finite element models were created to compare existing devices, prosthetic models of the CADskills device, and healthy situations. These were used to investigate the influence of such patient-specific prostheses through the evaluation of micromotion, loading on the opposite joint, strain on the healthy bone and condyle, and stress shielding.

The results showed that the temporomandibular joint prostheses were subject to stress considerably below their yield strength, except for the polyethylene of the fossa component, which might undergo abrasion under extreme muscular loads (larger than loads occurring during the activities of daily living). Additionally, the implant was shown to have little influence on stresses in the bone compared to a healthy model. Furthermore, the dynamic model shows how the load upon and stresses in the healthy joint increase when maximum muscle activation takes place with the mouth open.

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1. Introduction

Temporomandibular joint disorders, covering dysfunctions of the mastication muscles and temporomandibular joints (TMJs), are common in modern society. While often causing pain, these disorders are linked to joint clicking and restricted mandibular motion that can affect the patient's quality of life [1]. Approximately 25% of the world population is estimated to suffer from at least one of these disorders [2], with articular disk dislocation being one of the most common problems [3,4]. However, 5% of occurrences of this type of pathology require treatment, and of those, 3 to 5% require invasive management surgery [5]. For patients with severe pain and/or malfunction (e.g., end-stage degenerative/rheumatoid osteoarthritis, (recurrent) ankylosis, congenital hypoplasia/aplasia, iatrogenic/idiopathic condylar resorption, and some posttraumatic conditions) a TMJR (temporomandibular joint replacement/reconstruction) [6] might be inevitable to restore the functionality of one or both TMJs.

TMJ prostheses are not (yet) as prevalent as orthopedic implants (such as hip or knee implants [7]), and still present challenges like material failure (which is quite rare [8]) and jaw loading asymmetries that cause the contralateral joint to be overloaded [9]. The latter can lead to pain or other undesirable effects on the healthy side [2,10]. Moreover, the activities of daily living necessitate more than 2000 hinge and slide motions per day [11]. Therefore, it is of great importance to make sure that both the biomechanical behavior and load distribution are acceptable. TMJ prostheses are commonly classified according to their design as either patient-specific/patient-fitted [4] or stock prostheses [2,4,8,9,12–15].

The introduction of virtual surgical planning, in combination with computer aided design and additive manufacturing, has given a boost to the manufacturing of patient-specific TMJ TJR systems [16]. As a consequence, many manufacturers are trying to optimize and adapt their designs to improve and compete with others [16]. By investigating the aforementioned biomechanical behavior and load distributions, issues concerning device component stability, embodiment, and osseointegration can be avoided. This is often done by implementing FEA/FEM (finite element analyses/methods), a powerful tool

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for analyzing biomechanical behavior of the prosthesis that offers availability, few requirements, and low cost, making this virtual practice highly viable as a primary process [2,14,17,18]. In this way, researchers can simulate the structures under different loading conditions and constraints to account for variability in usage, or to test particular manufacturing criteria and objectives [2,4,6,8,10,14,20–23]. Furthermore, with its potential to perform virtual tests on the investigated joint type, this approach provides insightful information on various local effects, enabling researchers to perform experiments that would be considered unethical or even impossible *in vivo* [9]. Additionally, finite element (FE) modeling of a TMJ enables the analysis of micromotion for osseointegration [14], microstrain comparisons between healthy and prosthetic models [6], dynamic modeling of the mastication cycle [19], and so forth. Because the results of patient-specific prostheses cannot be extrapolated to others, an FEM is developed for each design. FE modeling, together with experimental testing, has proven to be the most powerful tool for diagnosing failures of the implant and jaw mechanics [2,6–8,10,13–16,23,24].

Finite element analyses can be used to numerically evaluate the behavior of the implant-bone interface and the magnitudes of the maximum stress and strain, as well as the regions where they occur in the implant and surrounding bone. The material and fixation method of the implant can be assessed and optimized, to reduce the stress concentrations and risk of failure under several loading conditions. The outcomes of the numerical models can thus be used for design purposes, because the load-bearing capacity and potential problems are determined without the need to make and mechanically test a prototype. The latter method has major drawbacks in that geometrical and load variations can occur, the results might be difficult to compare, and local results are difficult to quantify. Moreover, FE analyses facilitate the assessment of the structural integrity of a prosthesis, including the effects of multiple design iterations, without the costly and time-consuming need to manufacture different versions of the device. While mechanical experiments might provide more realistic and reliable results if performed correctly, these are often complex and costly to perform, and are thus more suited for end-stage analyses in the development of a product.

The objective of the present study was to perform a comprehensive analysis of the influence of a patient-specific prosthesis incorporating the reattachment of the lateral pterygoid muscle, through the evaluation of micromotion, loading on the opposite joint, strain on the healthy bone and condyle, and stress shielding. Additionally, the removal of two superior screws in the mandibular component (MC), with addition of a medial hooked lid, was investigated [22]. Overall, a complete analysis of the implant stability, bone ingrowth, screw stresses, and result validation was made. A state-of-the art, dynamic model from 4D CT scan data was also developed to accurately test the load distribution for a range of mouth openings which, to date, has been done only for discrete mouth positions (not more than two) using data from video recordings [8,23].

The implant considered for the study is a patient-specific total alloplastic replacement system for the TMJ (shown in Fig. 1A and B) that was developed by CADskills (Ghent, Belgium). The MC is made of Ti_6Al_4V Grade 23 ELI, with a specific HadSat® (DLC) coating on the condylar head. In addition, Ti_6Al_4V scaffold of 80% porosity is implemented at the medial side of the ramal part to promote bone ingrowth, and at the medial side of the condylar neck to promote reattachment of the lateral pterygoid muscle (see Fig. 1C). As a consequence of this muscle reattachment, the patient's range of motion can potentially be largely restored [24]. A design-specific medial lid (Fig. 1D) saddling the mandibular ramus is implemented to increase stability. The fossa consists of a UHMWPE bearing layer, which is thermally compressed into a Ti_6Al_4V scaffold belonging to the fossa backing part. The UHMWPE is highly cross-linked for better wear, and enriched with vitamin E to attain better oxidation resistance.

2. Materials and methods

The models were constructed using the geometries obtained from 3D and 4D CT scans (DICOM images), using segmentation tools in materialize Mimics Innovation Suite 22 (materialize, Leuven, Belgium). These 3D models were then imported into Autodesk Meshmixer 3.5 (Autodesk, California, USA) and materialize 3-Matic 12.0 (materialize, Leuven, Belgium) for mesh preparation and meshing using quadratic tetrahedral elements (C3D10). These remeshed geometries were then imported into Abaqus/CAE 2017 (Dassault Systèmes, Vélizy-Villacoublay, France) for the actual finite analyses and accompanying convergence studies.

Appropriate material and contact properties (following the Coulomb friction model), boundary conditions, and constraints were chosen, and masticatory forces were adopted from Korioth and Hannam [25]. The muscles included were the superficial and deep masseter (SM and DM), the medial and lateral pterygoid (MP and LP), and the anterior, middle, and posterior temporalis (AT, MT and PT). Studies vary as to the number of muscles simulated [14]. A relatively large number of muscles were incorporated in this study, and because the state-of-the-art prosthesis of interest allows reattachment of the LP, this was included as well. In addition, three clenching tasks were simulated: bilateral molar clench with maximal muscle activation (MAX), right unilateral molar clench (RMOL) and left unilateral molar clench (LMOL), where RMOL and LMOL simulated chewing bites. Implant positions were determined either by a CT scan or by design, and because no two jaws are completely identical, another healthy patient was added as benchmark for indicative comparison.

The properties of the materials used were obtained from the material providers, except that properties of the cortical bone were taken from the literature (shown in Table 1). The molar teeth were constrained to simulate tooth clenching. For the healthy side, vertical and lateral movements were constrained, because anatomical structures (ligaments and temporal bone) prevent movement of the joint in these directions. On the prosthetic side, the temporal bone would be fully constrained.

The loads determined in previous experimental research [14,25], were used to simulate molar tooth clenching. As previously mentioned, only the jaw-closing muscles (which have the greatest influence on the loads of the TMJ [6,14,19,25]) were introduced in the simulation, which included the masseter, temporalis, and pterygoid, shown in Table 2.

To evaluate stress shielding by the mandibular component and the influence of the implant on mandible behavior, two lines were defined in the model: line 1, which progresses through the center of the mandible bone from one ramus to the other, and line 2, situated underneath the implant on the surface of the ramus (not in contact with any screw, so that the strain was not influenced), as shown in

3. Results

The mandibular component reached the maximum Von Mises stress of 169 MPa in the corner situated below the 4th screw (

Fig. 2A and B). This was far below its yield strength limit of 1301 MPa. The UHMWPE of the fossa component reached its yield strength limit of 22 MPa just below the surface of contact with the mandibular component (

Fig. 2C and D). This indicated that high muscle activation during tooth clenching might produce abrasion in the fossa component. No parts of the mandibular components experienced micromotions higher than the 28 μm threshold (Fig. 2E).

Mandibular behavior was visualized by comparing a prosthetic model with a healthy mandible in the same loading situations. The microstrains, evaluated through the path traced by line 1, showed great similarities between these two models (Fig. 3). The main

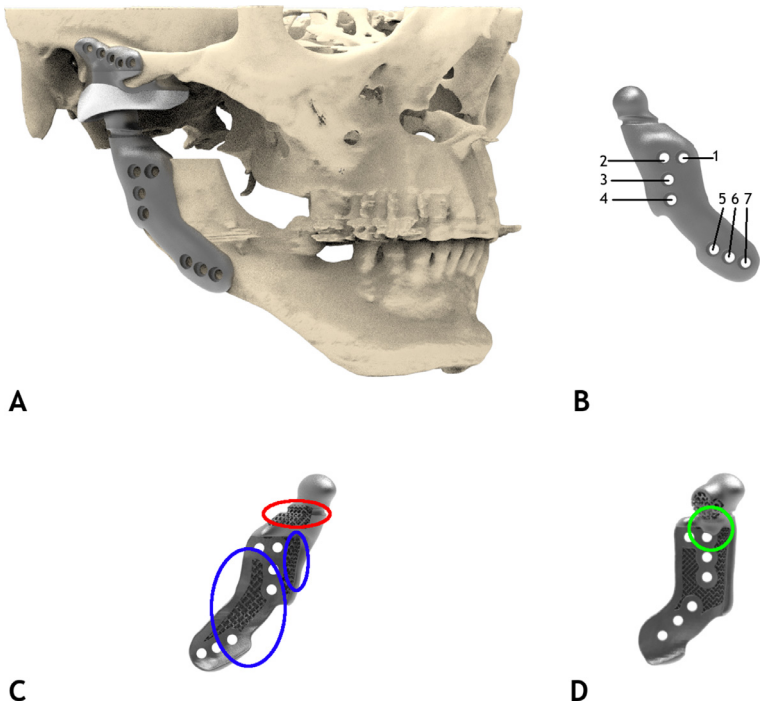

Fig. 1. The studied TMJ prosthesis. A: A render of the implant in its planned position. B: Lateral view of the mandibular component with the numbering of the screws. C: Medial view of the mandibular component where the scaffold is visible. The enthesis reconstruction scaffold is indicated by a red circle, while the scaffold for the mandibular bone ingrowth is indicated by two blue circles. D: A mandibular component with a medial lip (indicated by a green circle).

difference lay in the attachment of the superficial head of the masseter muscle. The Von Mises strains through line 2 aided in the diagnosis of stress shielding. Two cases were evaluated. The first was before osseointegration had taken place, where, as expected, strains were lower because the bone carried no load. The other case presumed bone ingrowth into the implant, and the observed strains were higher. The maximum difference between the two strains corresponded to 151.6 $\mu\varepsilon$.

The loads on both sides of the mandible are shown in Fig. 4. It can be observed how the load on the healthy condyle increased with the distance between the incisor teeth. It increased from 350 N to 432 N, which would also heighten stresses on the healthy condyle and articular disk.

4. Discussion

Although analyses of TMJ TJR systems have previously appeared in the literature, only a very few have included a fossa part or a medial lid hooking behind the mandibular ramus bone [6], but foremost, no research has combined these features with the novelty of re-insertion of the pterygoideus lateralis muscle. Also, regarding the MC fixation screws, few studies have investigated their placement to avoid the inferior alveolar nerve branch [22], and not many have investigated the inserted portion of screws and their altered stress distributions [14]. As such, a combination of these TMJ TJR features presented the opportunity for a novel study.

Table 1
Mechanical properties of the materials involved in the FE model [7,9,14].

Material Properties	E (MPa)	ν
Ti_6Al_4V	113,000	0.3
Cortical Bone	15,000	0.3
Scaffold	9000	0.1
UHMWPE	683	0.46

Among the prosthetic components modeled here, the only one to reach stress close to its yield limit was the UHMWPE part, which reached 22 MPa when simulating a static tooth-clenching situation. This has been shown to be common in patient-specific prostheses that use UHMWPE [6,14]. For similar UHMWPE fossa parts, high stress values have been reported (50 MPa and 19.61 MPa, reported by Xu et al. [7] and Chen et al. [6], respectively), such that these parts potentially operate close to their limits. This does not encourage smaller thicknesses to be investigated for the UHMWPE. Moreover, this might be indicative of abrasion for long-term usage [7]. Although, for an earlier variant of the TMJ TJR Parametro system, *in vivo* tests in sheep [26,27] have shown promising results concerning wear, further mechanical tests and long-term follow-up studies in human patients should be conducted to take into account more critical loading scenarios. Furthermore, high stresses similar to these are not expected with mouth openings, because the load on the right condyle decreases with the opening, as shown in Fig. 4. Moreover, the loads exerted by the muscles are not expected to be the same in the pathological model as in the healthy model, from which the data were extracted.

The maximum Von Mises stress of 169 MPa for the MC is in line with those found for similar devices: Chen et al. [6] and Kashi et al. [20] reported maximal values of 126 MPa and 170 MPa for a chewing bite and maximal chewing bite, respectively. The results are also consistent with a report from Huang et al. [14] that indicated that the maximal Von Mises stress in the MC was 159.1 MPa for a simulated LMOL task. Furthermore, the stresses remained below those reported by Ackland et al. [22] for the Biomet Microfixation during normal or maximal chewing bites.

Micromotion is a crucial parameter to evaluate osseointegration of the implant. There are discrepancies among reported values of this threshold. Some authors suggest that 40 μm is the highest value for osseointegration to take place [28,29]. However, the present study used a threshold of 28 μm, which is the most conservative found in the literature [14,29]. As shown in

Table 2
Muscle forces and directions during tooth clenching [15,29].

Side	Direction	Muscular force (N)						Constraint
Right	Force	190.4	81.6	146.8	154.8	91.8	71.1	Molars
	Fx	−39.4	−44.6	71.3	−23.1	−20.4	−14.8	
	Fy	−79.8	29.2	−54.8	−6.8	45.9	60.8	
	Fz	168.3	61.9	116.1	152.9	76.8	33.7	
Left	Force	190.4	81.6	146.8	154.8	91.8	71.1	
	Fx	39.4	44.6	−71.3	23.1	20.4	14.8	
	Fy	−79.8	29.2	−54.8	−6.8	45.9	60.8	
	Fz	168.3	61.9	116.1	152.9	76.8	33.7	

Fig. 2E, no point of the scaffold surpassed this threshold. Thus, it can be assumed that bone ingrowth will take place in the implant, hence improving its stability and stress shielding, because the underlying bone keeps its stimulus for continued bone remodeling.

As aforementioned, the microstrains produced in the mandibular bone were evaluated through lines 1 and 2 in Fig. 3. Line 1 serves the purpose of evaluating the influence of the implant on mandibular deformation. The spike in strain taking place near the sagittal plane can be explained by the attachment of the superficial masseter, because part of its attachment surface is taken over by the implant with no reduction in force, thus the strain produced on the mandible is greater. Furthermore, it was shown that the influence on the mandibular bone is not substantial. The microstrains displayed through line 2 have been shown to stay above 100 $\mu\varepsilon$, a threshold below which bone resorption has been stated to arise [30]. Additionally, the strains are observed to increase when osseointegration takes place, because the bone carries more load. This is obviously a positive consequence of the osseointegration of the implant. The strains are nevertheless too low to reach the threshold of 4000 $\mu\varepsilon$ to produce hypertrophy of the bone [30].

The dynamic simulation shows how the load on the prosthesis decreases while the load on the healthy side increases with the opening of the mandible. This takes place due to the reduced mobility of the mandibular component on the fossa [23].

Little research has been published regarding the influence of the prosthesis on the healthy side of the joint. However, there is consensus that a healthy mandible withstands lower stresses in the TMJ—particularly in the articular disk—than the contralateral side of the mandible that has undergone total alloplastic replacement [9], especially if the prosthesis is not custom-made [15]. However, it has been shown in numerical models that the healthy side is positively affected postoperatively by a 15% decrease in stress in the articular disk [2], because the physiological joints in a pathological model do not perform as in the healthy model. Nonetheless, the load on the contralateral side in a patient that has undergone total TMJ reconstruction with a stock prosthesis is approximately 15% greater than the load on a condyle of a healthy patient [9]. Increased mechanical loads have been shown to produce cartilage and articular disk damage [19]. Therefore, it is crucial for the implant not to overload the contralateral side [10].

The mastication cycle is composed of opening and closing movements of the mandible [19], in combination with lateral movements. However, only the closing movement, with resistance from the teeth or other structures, puts the TMJ under stress [10,19]. Consequently, only the jaw-closing muscles need to be included in the model [2,7–9,12–15]. Previous research has shown tooth clenching to be the situation putting the TMJ under the greatest loading [6,7,14], because it fully activates the jaw-closing muscles [25]. Consequently, previous studies have focused on simulating this type of static loading on the mandible [6,7,9,12,19,20], with only a few analyzing other types of clenching situations [14] or mouth openings [8,23]. The positioning of the components of the FE model were obtained from CT scans in the case of static modeling, or from video recordings [8,23].

This research could be optimized further to obtain the most accurate results by retrieving the muscle forces from the actual patient whose implanted TMJ TJR model is being analyzed, but this is hard to do, because the technology for doing so is not widely available and/or unvalidated, and because the need to obtain *in vivo* measurements would be undesirable for the recovering patient. Additionally, long-term follow-up studies of multiple patients implanted with this device, including higher loading scenarios in comparison with the tests in sheep, should be carried out.

5. Conclusion

It can be concluded that the proposed TMJ prosthesis design performs adequately for several reasons. Firstly, the mandibular

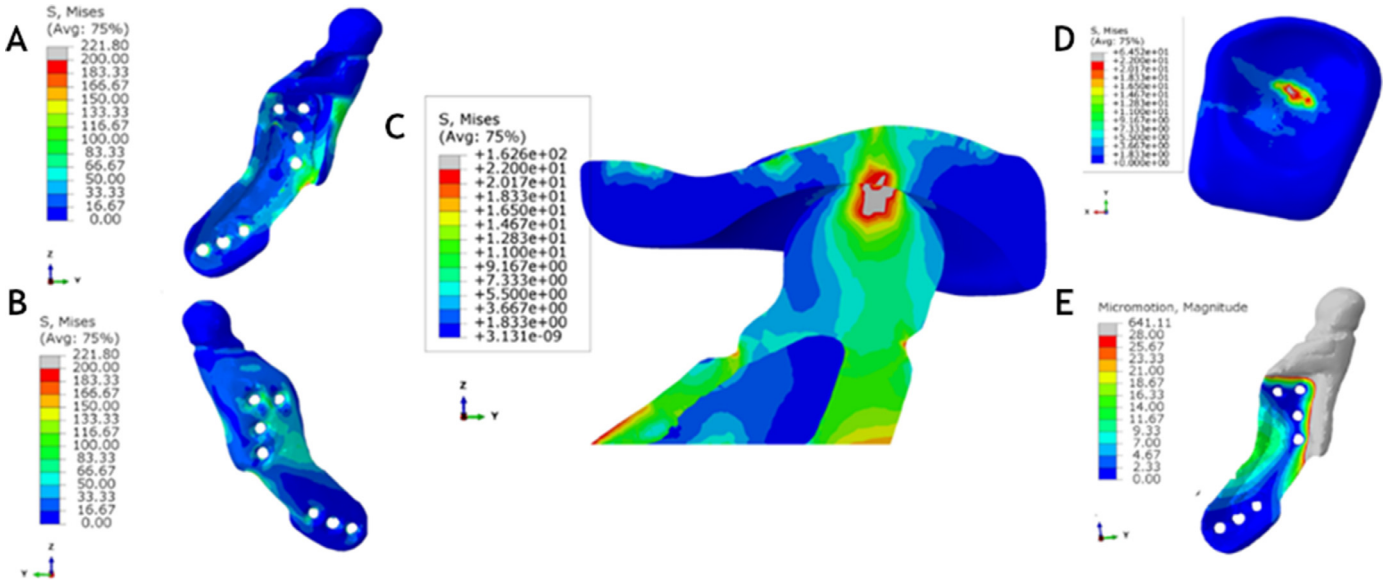

Fig. 2. Von Mises stress contours for: A: the medial side of the mandibular component; B: the lateral side of the mandibular component; C: the contact between the UHMWPE and the mandibular component; D: the fossa component; and E: the micromotions (in μm) of the mandibular component with respect to the bone.

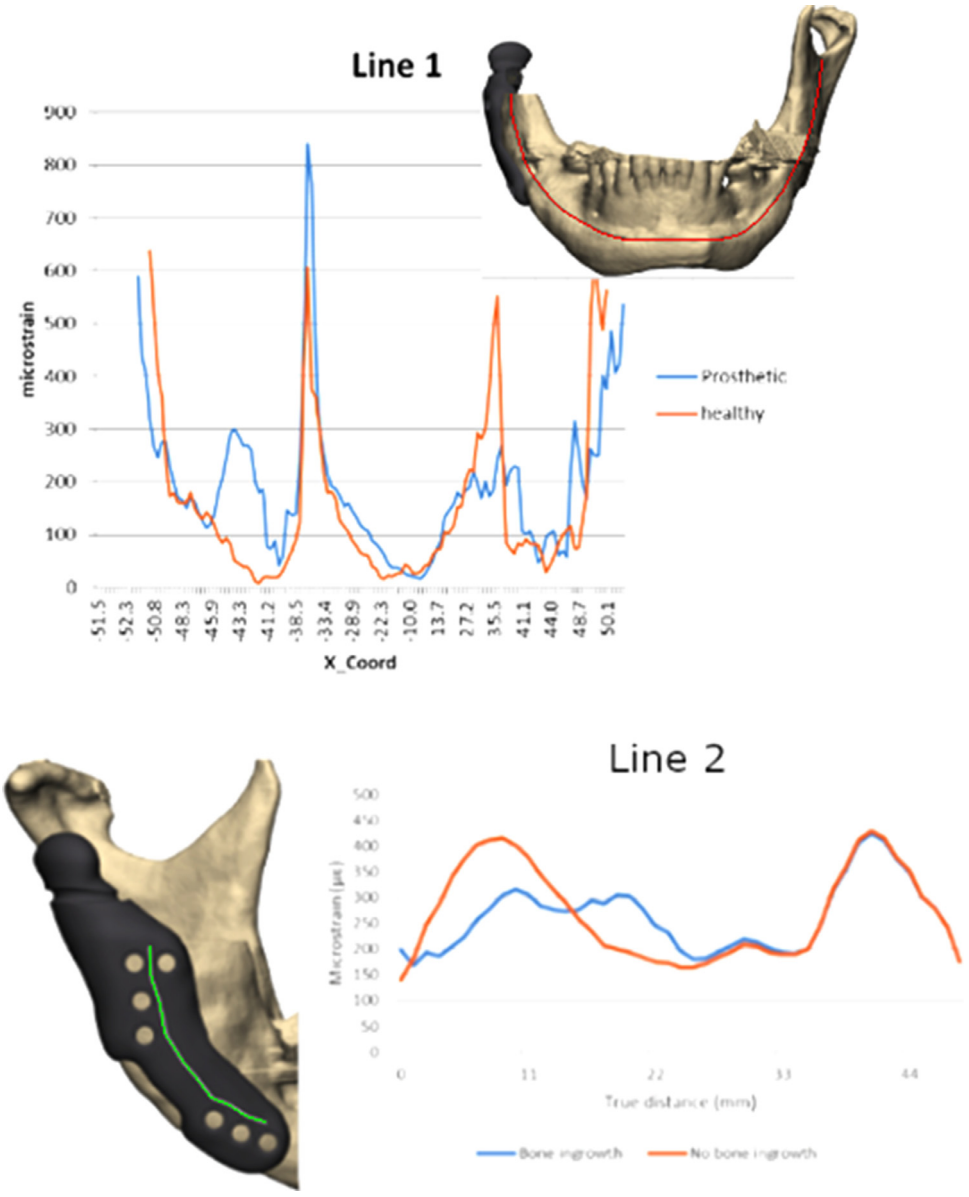

Fig. 3. Microstrains ($\mu\varepsilon$) for the comparison between the healthy and prosthetic models, evaluated through line 1 (red), and for the evaluation of stress shielding (through line 2, green).

component will not produce stress shielding on the contralateral condyle, because the loads are high enough to prevent bone resorption. The mandibular behavior is very similar to a healthy model, with the exception of the altered muscle attachments which intensify stresses on the mandible through a decrease in the surface area of attachment.

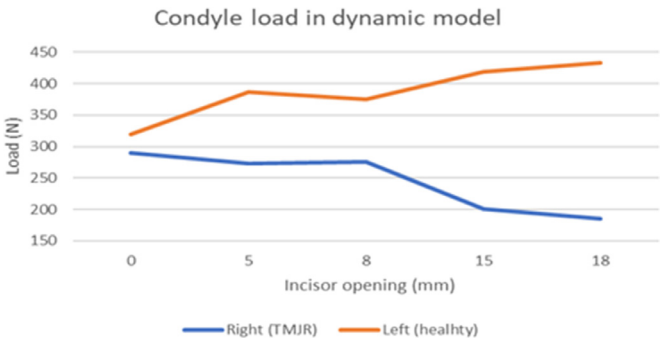

Fig. 4. Load being transmitted through the condyles as a result of muscle activation, for different mouth apertures.

All the materials (excluding the UHMWPE) perform at stresses below their yield strengths. Under the highest masticatory force, the UHMWPE is subjected to high loads, which might lead to abrasion, and possibly failure of the implant over the long term.

The dynamic model provides insightful information regarding the load on the healthy condyle, showing how, with wider mouth apertures (up to 18 mm), the load on the condyles increases (provided the masticatory forces remain at their maximum). Finally, the present results suggest against any further decrease in thickness of the fossa part, even though this would benefit easier implantability.

This proposed form of the total TMJ prosthesis offers a unique design, with the possibility of doing without the superior screws, thereby removing the risk of nerve damage complications, while retaining implant stability and ensuring osseointegration. Furthermore, the reattachment of the lateral pterygoid offers the possibility of partially restoring the patient's range of motion.

Declaration of Competing Interest

Dr. Ir. Stijn Huys is the R&D officer at CADskills, which produces the patient-specific implant that was analyzed. Prof. Maurice Mommaerts is the innovation manager and co-owner of CADskills. The other authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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References

[1] Van Bellinghen X, Idoux-Gillet Y, Pugliano M, Strub M, Bornert F, Clauss F, Schwinté P, Keller L, Benkirane-Jessel N, Kuchler-Bopp S, Lutz JC, Fioretti F. Temporomandibular joint regenerative medicine. Int J Mol Sci 2018;19. doi: 10.3390/IJMS19020446.
[2] Ramos AM, Mesnard M. The stock alloplastic temporomandibular joint implant can influence the behavior of the opposite native joint: a numerical study. J Cranio-Maxillofac Surg 2015;43:1384–91. doi: 10.1016/j.jcms.2015.06.042.
[3] Wolford LM, Dingwerth DJ, Talwar RM, Pitta MC. Comparison of 2 temporomandibular joint total joint prosthesis systems. J Oral Maxillofac Surg 2003. doi: 10.1053/joms.2003.50112.
[4] Rodrigues YL, Mathew MT, Mercuri LG, da Silva JSP, Henriques B, Souza JCM. Biomechanical simulation of temporomandibular joint replacement (TMJR) devices: a scoping review of the finite element method. Int J Oral Maxillofac Surg 2018. doi: 10.1016/j.ijom.2018.02.005.
[5] Kent JN. An important lesson about biomaterials in the TMJ. J Oral Maxillofac Surg 1991. doi: 10.1016/0278-2391(91)90401-7.
[6] Chen X, Wang Y, Mao Y, Zhou Z, Zheng J, Zhen J, Qiu Y, Zhang S, Qin H, Yang C. Biomechanical evaluation of Chinese customized three-dimensionally printed total temporomandibular joint prostheses: a finite element analysis. J. Cranio-Maxillofac Surg 2018;46:1561–8. doi: 10.1016/j.jcms.2018.06.018.
[7] Xu X, Luo D, Guo C, Rong Q. A custom-made temporomandibular joint prosthesis for fabrication by selective laser melting: finite element analysis. Med Eng Phys 2017. doi: 10.1016/j.medengphy.2017.04.012.
[8] Mesnard M, Ramos A, Ballu A, Morlier J, Cid M, Simoes JA. Biomechanical analysis comparing natural and alloplastic temporomandibular joint replacement using a finite element model. J Oral Maxillofac Surg 2011. doi: 10.1016/j.joms.2010.02.019.
[9] Bekcioglu B, Bulut E, Bas B. The effects of unilateral alloplastic temporomandibular joint replacement on the opposite-side natural joint: a finite-element analysis. J Oral Maxillofac Surg 2017;75:2316–22. doi: 10.1016/j.joms.2017.05.017.
[10] J.H.K, van Eijden TMGJ. Consequences of viscoelastic behavior in the human. J Dent Res 2007;86:1198–202.
[11] Sinno H, Tahiri Y, Gilardino M, Bobyn D. Engineering alloplastic temporomandibular joint replacements. McGill J Med 2010. doi: 10.26443/mjm.v13i1.252.
[12] Abel EW, Hilgers A, McLoughlin PM. Finite element analysis of a condylar support prosthesis to replace the temporomandibular joint. Br J Oral Maxillofac Surg 2015. doi: 10.1016/j.bjoms.2015.01.016.
[13] Ackland DC, Moskaljuk A, Hart C, Vee Sin Lee P, Dimitroulis G. Prosthesis loading after temporomandibular joint replacement surgery: a musculoskeletal modeling study. J Biomech Eng 2015. doi: 10.1115/1.4029503.
[14] Huang HL, Su KC, Fuh LJ, Chen MYC, Wu J, Tsai MT, Hsu JT. Biomechanical analysis of a temporomandibular joint condylar prosthesis during various clenching tasks. J Cranio-Maxillofac Surg. 2015. doi: 10.1016/j.jcms.2015.04.016.
[15] Ramos A, Mesnard M. Comparison of load transfers in TMJ replacement using a standard and a custom-made temporal component. J Cranio-Maxillofac Surg 2014. doi: 10.1016/j.jcms.2014.06.012.
[16] Elledge R, Mercuri LG, Attard A, Green J, Speculand B. Review of emerging temporomandibular joint total joint replacement systems. Br J Oral Maxillofac Surg 2019. doi: 10.1016/j.bjoms.2019.08.009.
[17] Mesnard M, Ramos A. Towards a rigorous approach to designing a TemporoMandibular joint prosthesis. From clinical challenge to numerical prototype. Proced CIRP 2013. doi: 10.1016/j.procir.2013.01.029.
[18] Hsu JT, Huang HL, Tsai MT, Fuh LJ, Tu MG. Effect of screw fixation on temporomandibular joint condylar prosthesis. J Oral Maxillofac Surg 2011. doi: 10.1016/j.joms.2010.05.074.
[19] Commisso MS, Martínez-Reina J, Ojeda J, Mayo J. Finite element analysis of the human mastication cycle. J Mech Behav Biomed Mater 2015;41:20–35. doi: 10.1016/j.jmbbm.2014.09.022.
[20] Kashi A, Chowdhury AR, Saha S. Finite element analysis of a TMJ implant. J Dent Res 2010. doi: 10.1177/0022034509357716.
[21] Mesnard M, Ramos A, Simões JA. Influences of implant condyle geometry on bone and screw strains in a temporomandibular implant. J Cranio-Maxillofac Surg 2014. doi: 10.1016/j.jcms.2013.04.010.
[22] Ackland DC, Robinson D, Redhead M, Lee PVS, Moskaljuk A, Dimitroulis G. A personalized 3D-printed prosthetic joint replacement for the human temporomandibular joint: from implant design to implantation. J Mech Behav Biomed Mater 2017. doi: 10.1016/j.jmbbm.2017.01.048.
[23] Ramos A, Mesnard M. Load transfer in Christensen® TMJ in alloplastic total joint replacement for two different mouth apertures. J Cranio-Maxillofac Surg 2014. doi: 10.1016/j.jcms.2014.04.008.
[24] Mommaerts MY. On the reinsertion of the lateral pterygoid tendon in total temporomandibular joint replacement surgery. J Cranio-Maxillofac Surg 2019;47:1913–7. doi: 10.1016/j.jcms.2019.11.018.
[25] Korioth TW, Hannam AG. Mandibular forces during simulated tooth clenching. J Orofac Pain 1994;8(2):178–89.
[26] De Meurechy N, Verwilghen D, De Brucker Y, Van Thielen B, Mommaerts MY. Lateral pterygoid muscle enthesis reconstruction in total temporomandibular joint replacement: an animal experiment with radiological correlation. J Cranio-Maxillofac Surg 2021. doi: 10.1016/j.jcms.2021.01.029.
[27] De Meurechy N, Merve J, Aktan K, Boeckmans B, Huys S, Denis |, Verwilghen R, Braem A, Mommaerts MY. Surface wear in a custom manufactured temporomandibular joint prosthesis. J Biomed Mater Res Part B Appl Biomater 2022. doi: 10.1002/JBM.B.35010.
[28] Harrison N, McHugh PE, Curtin W, Mc Donnell P. Micromotion and friction evaluation of a novel surface architecture for improved primary fixation of cementless orthopaedic implants. J Mech Behav Biomed Mater 2013. doi: 10.1016/j.jmbbm.2013.01.017.
[29] Kienapfel H, Sprey C, Wilke A, Griss P. Implant fixation by bone ingrowth. J Arthroplast 1999. doi: 10.1016/S0883-5403(99)90063-3.
[30] Wiskott HWAA, Botsis J, Scherrer SS, Belser UC, Mellal A, Wiskott HWAA, Botsis J, Scherrer SS, Belser UC. Stimulating effect of implant loading on surrounding bone. Comparison of three numerical models and validation by *in vivo* data. Clin Oral Implants Res 2004:1–10. doi: 10.1111/j.1600-0501.2004.01000.x.