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Effect of the Design Constraints and the Loading Model on the Geometry of Topology Optimized Acetabular Cages

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Abstract

The treatment of large acetabular bone defects is a challenging task for the clinical experts. One of the most important part is the selection or the design the most appropriate implant. The aim of the study is to explore the potential of topology optimization for the treatment of extensive pelvic bone defects. Using a finite element method, the authors investigate different design spaces and load cases. Sensitivity tests for the material characteristic of the bone and the constraint of the applied volume-fraction were performed. The results are topology-optimized acetabular cage concepts with similar designs. The conceptual designs are not sensitive to the changing of the elastic modulus of the bone and the volume-fraction constraint. The reason for the similarity between the designs is the close connections, they have a special role in maximizing the stiffness. With the use of our design space, a lot of bone grafts can be put behind the cage and it can give an idea for sheet metal conceptual designs. Due to the close connections, similar conceptual variants are generated under normal walking load, which can be used for faster calculations in similar cases.

Keywords

acetabular defect, biomechanics, finite element method, topology optimization

1 Introduction

In Section 1 the results of the literature research and the main goals of the study are included.

1.1 Clinical review

Total hip replacements have an important role in the treatment of osteoarthritis. These hip implants may need revision within 10–20 years after implantation. One of the reasons for the revision is the formation of large acetabular defects.

The Paprosky scale is suitable for classifying the extent of bone defects. This is a quality assessment. Accordingly, recommendations for the treatment of bone deficiency can be found in the clinical literature [1].

In Paprosky III type bone defects, the rim no longer provides adequate stability, and there may be no back wall.

In the case of a Paprosky III / A bone defect, component migration of up to 3 cm can be upwards. Although the anterior and posterior columns are good, the vena implanted during the revision would have less than 50% contact with the bone [2].

In Paprosky type III / B bone deficiency, the lack of the rim is greater than 50%. Due to the large component migration towards the center of the body and the lack of a wall here, the pelvis may even have a discontinuity. Less than 40% of the graft implanted during the revision would be in contact with bone marrow [2]. Fig. 1 shows an intact pelvis and a pelvis with Paprosky III defect.

Small cavitation gaps can be filled with bone graft [2]. To eliminate segmental deficiencies, a large jumbo cup and an oblong cup are used, as well as an acetabular cage, which has many variants, and reinforcement ring with allografts and trabecular metal cups with augments [2–4].

Another way to treat this is to implant custom triflange implants. Another option is the so-called "Cup - cage construct", where a baffle and a plate are located, and after the cane is ossified, the plate is removed.

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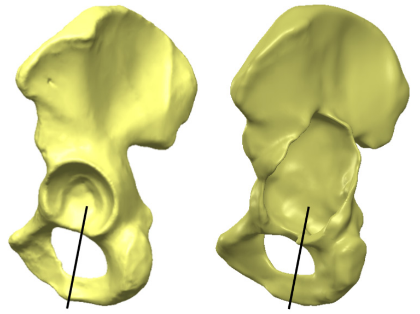

Fig. 1 The geometry of a healthy hemipelvis and a large acetabular defect

The state-of-the-art treatment option, with which the pelvis discontinuity can be successfully restored, is a patient-specific implant printed from metal powder [4].
One of the main criticisms of the treatment options, in addition to the standardized design, are the high rigidity of each implant, the high manufacturing cost, and the fact that the implanted metal parts can never be converted to living bone.

Róbert Sződy and his colleagues [5] made custom sheet metal cages (Fig. 2). These were less rigid than other variants and they have patient-specific nature. The target area of the research was to explore the development possibilities of these types of implants.

1.2 Finite element analyses and topology optimization in the literature

In the literature, a distinction must be made between finite element studies related to the pelvis and those related to implants and their development.

Fig. 2 Custom-made sheet-metal acetabular cages

Finite-element studies related to the pelvis have reached very high levels, and publications have been published on the role of bone transformation, muscles, and tendons [6–9].

In cases where an implant was also examined, the load and boundary condition models were much simpler [10, 11]. The literature uses the walking load and the load maxima of the staircase walking as the most important loads. These were described by Bergmann et al. [12] who measured the forces awakened in the hip joint with sensorized implants.

No publications were found on implants that can be created by topology optimization specifically for acetabular bone deficiency.

The closest publication found to our research is related to Iqbal et al. [13].

Here, large-scale implant variants suitable for the replacement of resected pools have been developed. In their loading model, the use of weighted loads by the frequency of daily loads was given priority.

Their load model is also used in our studies to see how well it can be used for a hemipelvis with acetabular bone deficiency and what results from it provides.

1.3 Aims of the study

The main goal of this study is to see the results of topology optimization for different loading models from the literature in the cases of large acetabular defects.

The authors will also investigate the effect of different design-space geometries and how the conceptual geometries change with a different elastic modulus of the bone and volume-fraction constraints.

To make the effect of each variable easier to detect, simplified models were also used.

2 Materials and methods

In Section 2, the topology optimization model based on the finite element method is presented.

2.1 Geometry models

Based on the level of complexity, three levels of geometric models were used. Each was based on a pelvis with a large acetabular defect. The bone-deficient pelvis was segmented with 3D Slicer based on CT data. During that, the grayscale ranges of the pelvis were highlighted in every relevant layer. After which the covering triangular mesh could be saved in STL format. The received STL mesh was smoothed and repaired using MeshMixer. The virtually reconstructed position of the hip joint center was confirmed by medical experts. For the reverse engineering process, SolidWorks 2018 was used to obtain the CAD model of the pelvis. The CAD model of the hemipelvis with acetabular defect and the position of the cage can be seen in Fig. 3.

During the preprocessing of the design space, it was taken into account that the protruding parts should not be too large and that there should be sufficient thickness everywhere.

The simplified models were made as revolved bodies. In order to create sheet metal-like conceptual designs, the part behind the acetabular cage has been removed. A total of three different geometry models were made and these can be seen in Fig. 4.

2.2 Material properties

Homogenous, linear elastic, and isotropic material properties were used everywhere. The sheet metal acetabular cages are made of stainless steel, so general steel material properties were used to model its behavior. For reasons of simplification, homogeneous material properties were used in the pelvis, but the material model is based on the literature research [9, 14]. The material properties can be seen in Table 1.

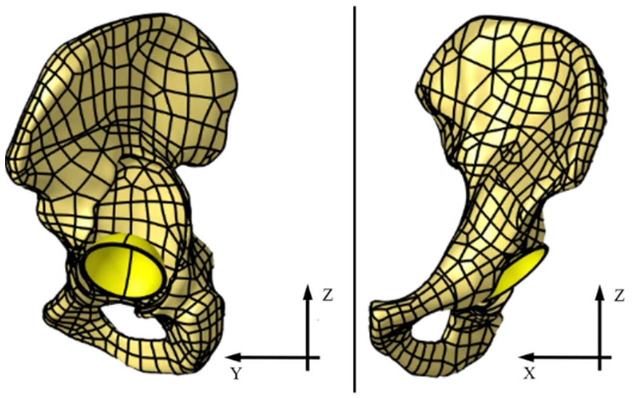

Fig. 3 The CAD model of the hemipelvis with acetabular defect and the position of the cage

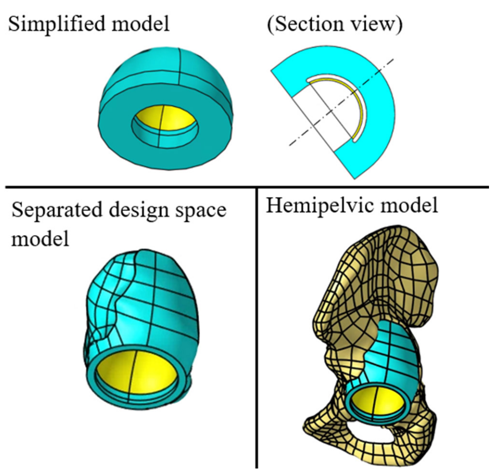

Fig. 4 The CAD models for the topology optimization (design spaces are marked with blue)

2.3 General loading model

The coordinate system is based on the publication of Bergmann et al. [12]. The loads were from the HIP98 database, from an average patient, where the loads from daily activities were measured. The maximum values of these loads can be seen in Table 2, where they were calculated by a person weighing 80 kg.

Load transfer was performed with RBE3 elements through the center of the cage to the cage's inner surface.

2.3.1 Boundary conditions of the simplified model

In the case of the simplified model, a fixed boundary condition was applied to the spherical surface. This can be seen with the loading model in Fig. 5.

Table 1 Material properties

Part	Young's modulus [MPa]	Poisson's ration [-]
Pelvis	7000	0.3
Design space	210 000	0.3
Cage	210 000	0.3

Table 2 Components of the loading model from the HIP98 database

Loading case	Component X [N]	Component Y [N]	Component Z [N]
Slow walking (SW)	141	−122	1897
Normal walking (NW)	228	−207	1803
Fast walking (FW)	292	−377	1905
Upstairs (US)	158	−468	1912
Downstairs (DS)	338	327	1989
Standing up (SU)	731	−1107	687
Sitting down (SD)	600	−900	575
Standing (S)	236	111	1799
Knee bending (KB)	534	−602	787

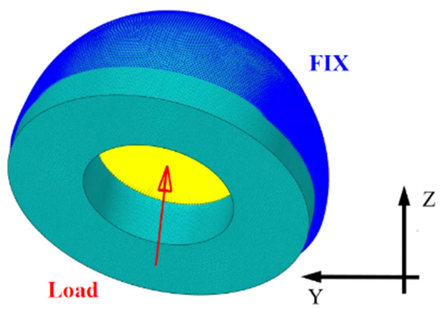

Fig. 5 Boundary conditions and loads for the simplified model

2.3.2 Boundary conditions of the separated design space model

Fixed boundary conditions were used on the parts of the design space where it would be connected virtually to the pelvis. This can be seen in Fig. 6.

2.3.3 Boundary conditions of the hemipelvis model

According to the literature research, fixed boundary conditions were used on the sacroiliac joint and the pubic symphysis [11]. The effects of the ligaments were neglected. The design space is connected to the pelvis through a mesh of common nodes. The boundary conditions and the loading model can be seen in Fig. 7.

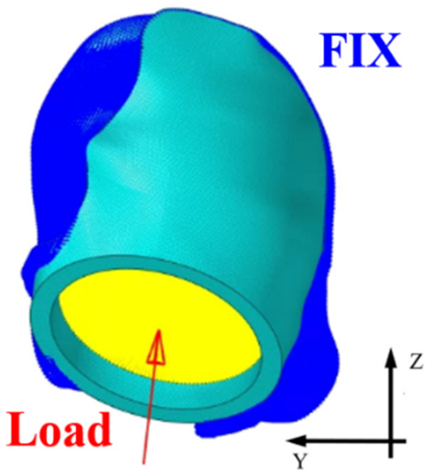

Fig. 6 Boundary conditions and loads for the separated design space model

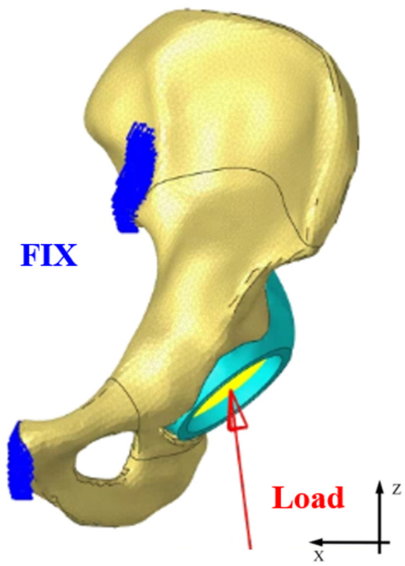

Fig. 7 Boundary conditions and loads for the hemipelvis model

2.4 Topology optimization model

During topology optimization, we look for the material distribution that best meets the given objective function in a selected space, so-called in the design space. The design variables are the relative densities of the elements, which are related to their stiffness.

The objective function can be formulated as minimization of the weighted compliance. For a design constraint, a volume-fraction constraint was prescribed. This combination results in the most rigid concept possible for a given volume-fraction. This volume fraction was 5% for the simplified model and 3% for the other two.

Five load combinations were investigated. The vector of the maximum magnitudes of each daily activity were used as listed in Table 2. The loading model combinations can be seen in Table 3, and the used weight factors by Iqbal et al. [13].

Further sensitivity analyzes were performed on the effect of volume fraction constraint and Young's modulus of the pelvis. All the mentioned load combinations were analyzed with the pelvis-containing geometry model, using a 10% permissible volume fraction instead of 3%. The modulus of elasticity of the bone was modified by +/− 30%, so the calculations for the normal walking loadcase (Loadcase 1) were performed using elastic modulus values of 9100 MPa and 4900 MPa, respectively.

The data about the finite element mesh can be seen in Table 4. Linear tetrahedron elements were used everywhere similar to the literature [13]. The use of second-order elements would have required a large amount of computation. The average element size for the design spaces was 0.8–1 mm.

Table 3 Load cases and other parameters

Name	Loads (abbreviated)	Weight factors
Loadcase 1	NW	1
Loadcase 2	NW/US/DS	1/1/1
Loadcase 3	NW/US/DS/SU/SD/S	1/1/1/1/1/1
Loadcase 4	NW/US/DS/SU/SD/S	0.88/0.027/0.027/ 0.013/0.013/0.041
Loadcase 5	SW/NW/FW/US/DS/ SU/SD/S/KB	1/1/1/1/1/ 1/1/1/1

Table 4 Mesh data

Model	Number of elements (design space)	Number of elements (other)
Simplified	3 323 077	26 042
Separated design space	1 6191 663	5 988
Hemipelvis	1 6191 663	293 228

3 Results

Section 3 presents the results, highlighting the similarities and differences between each simulation. The results are the relative densities; all of them were presented using the above 0.8 isosurfaces.

Even with the modification of the volume fraction constraint, the observed main directions remained the same; only the supports were raised. This can be seen in Fig. 8, presented with the normal walking loadcase.

The effect of changing the modulus of elasticity is shown in Fig. 9. There is no significant difference in the conceptual versions.

The concepts resulting from different load combinations are differentiated in the case of the simplified model. The other two models did not generate significantly different concepts under the effect of the examined load combinations. Moreover, these results are similar to each other, too. These results can be seen in Fig. 10.

4 Discussions

Section 4 the evaluation of results and the limitations of this study.

In the case of the simplified model, the results differ significantly.

In this case, the fixed boundary conditions are located at a considerable distance from the cage (the non-design space). It is not sufficient to attach material to the nearest parts to maximize rigidity.

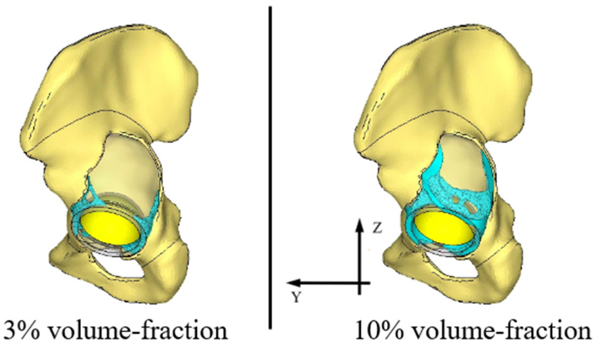

Fig. 8 Results for different volume-fraction constraints

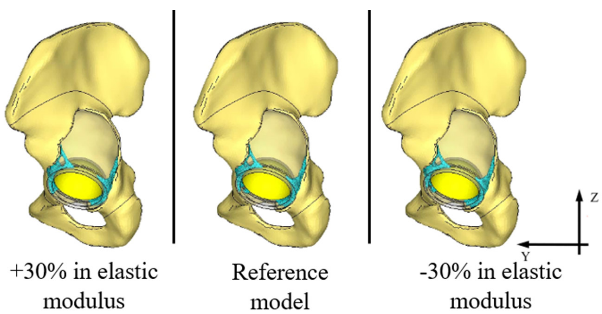

Fig. 9 Results for different elastic modulus value

In the case of the simplified model, the literature weighted load case (Loadcase 4) does not provide a significant difference compared to the normal walking load case (Loadcase 1).

The connecting parts formed at the bottom are not considered to be thick. This is because walking load plays a prominent role as the most common load when using weightings taken from the literature.

In the remaining three cases, when significant forces in other directions are already operating on the model in addition to the walking load, so the extra connection with a substantial thickness is already formed in the lower part.

The connection is typically made to nearby parts for models with more advanced design space and a hemipelvis. The results provided by these two models are very similar, with the more rigid parts starting from the same parts of the cage. Compared to the simplified model, the results differ because these variants are not recommended flanges with the farthest grip.

Among the limitations of the research, it should be mentioned that a simplified, homogeneous material model was used for the material characteristics of the bone. However, by examining its sensitivity, we found that the result is not sensitive to large changes in the elastic modulus either.

In the case of the load and boundary condition model, the effect of the muscles and tendons was neglected. This simplification of the model became necessary here. The load transfer was not performed by contacting, modeling the ball head, but as a force operated at the center of the cage.

It should be noted that these forces are still only the maximum vectors of each life load. With the same frequency, forces of similar magnitude but in different directions occur in the loads of daily activities.

Different sensitivity tests were performed for the most complex model for walking load.

5 Conclusion

During this research FEA topology optimizations were carried out. The objective functions aimed to maximize the stiffness to have a more favorable material distribution of the implant.

In the case of a near-pelvic connection, there is not much difference in the models calculated with multiple load cases from the model optimized for normal speed walking. This can be used in similar cases to generate conceptual designs.

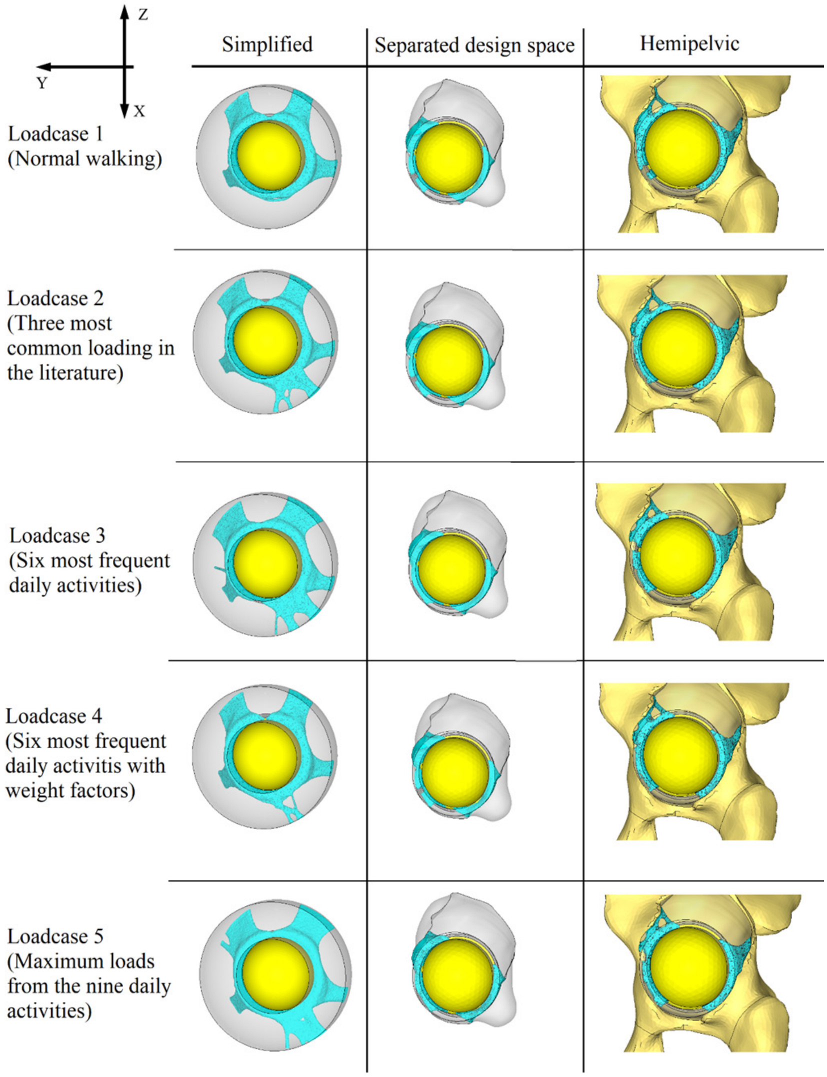

Fig. 10 Results for different load cases

The results of the constraint modification of the volume fraction show that the main conceptual material directions do not change significantly. It is important because it means, that it does not necessarily require special attention.

No significant differences were created by changing the material properties due to the large difference in modulus of elasticity between steel and bone. There is no need to specify a more specific bone material characteristic of the pelvis, which also simplifies the model.

Removing the design space behind the cage results in a concept for sheet-metal part implant because it does not start any support from the back of the cage. This can be used for the generation and design of sheet-metal acetabular cages. The authors will investigate its possibilities, but care must be taken when considering manufacturability requirements.

The use of topology optimization for stiffness maximization is most common in the literature. However, implants with maximum rigidity are not necessarily the most suitable for transforming bone grafts.

The walking load is the most common. The resulting displacement of the cage from the walking load can stimulate the transformation of bone grafts if it is adequate. For this reason, the effect of displacement constraint on the three different variants will also be examined for the normal walking loadcases in future works.

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