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Cervical Fusion Cage Lattice Structure Optimization using Computational Biomechanics

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ABSTRACT

Implants are instruments that typically inserted into a host tissue to restore any damaged physical function. In the case of a spinal implant, it usually consists of a spinal cage, pedicle screw and spinal rod, which they act together as a medical device that is implemented in the surgical treatment of patients with spinal diseases. Titanium alloy such as Ti- 6Al-4V is a biomaterial that commonly used in the spinal implants. However, the alloy is non-degradable and may cause stress shielding effect. Hence, magnesium alloy such as ZK60 is used as a substitute for the Ti-6Al-4V due to its biodegradable and bioabsorbable characteristics. The study aims to optimize the spinal cages of Ti-6Al-4V and ZK60 by incorporating lattice alteration to the implants and to evaluate the stress distribution of optimized spinal cages under sitting condition. The spinal cages of Ti-6Al-4V and ZK60 were optimized using nTop software with Diamond lattice structures, which can provide great cell growth rates and high energy absorption capacity compared to the other lattice structure types. After that, the finite element analysis (FEA) model of the solid and optimized spinal cages *were established to evaluate their stress distribution and total deformation under sitting condition. The stress in the current study showed comparable trend with the previous biomechanical and finite element analysis (FEA) study. Both current and previous study showed lower maximum stress in optimized group of spinal cages compared to the solid group, and lower maximum stress in ZK60 group than the Ti-6Al-4V group. The maximum von Mises stress in solid cages decreased by 25.63% and 26.32% for Ti-6Al-4V and ZK60, respectively, after the lattice optimization. As for the total deformation, a larger deformation occurred in the optimized group of spinal cages than that of solid group. Maximum deformation of optimized ZK60 spinal cage was 2.09×10^{-6} mm while the maximum deformation of solid ZK60 was 1.86×10^{-6} mm. In summary, the lower maximum stress value of ZK60 in the findings showed better mechanical properties of ZK60, as the magnesium alloy has elastic modulus value that is much closer to the value of normal bone tissue than Ti-6Al-4V. Besides, the findings showed that the optimized group of spinal cages further reduced their stiffness with the application of porous structure, based on the greater total deformation in the cages.*

Keywords: Spinal Cage; Stress Shielding; Lattice Alteration; Finite Element Analysis (FEA)

INTRODUCTION

The spine can be divided into five distinct regions, which are the cervical, thoracic, lumbar, sacrum, and coccyx. The main role of cervical spine is to support, and cushion loads to the head and neck, while allowing the head to move and rotate. The cervical spine consists of seven vertebrae that are C1 to C7, which they extend from the base of the skull to the top of the trunk. There are six intervertebral discs, which each one of them sits between every two vertebrae. The discs play important roles as it acts as a shock absorber and allows dispersion of weight and movement of individual vertebrae. The cervical is the second most common spinal section for disc injury due to the mobility of the cervical spine, which causes it intervertebral discs to be higher risk of damage from bending and torsion [1].

Implants are instruments that typically inserted into a host tissue to restore any damaged physical function [2]. In the case of a spinal implant, it usually consists of a spinal cage, pedicle screw and spinal rod, which they act together as a medical device that is implemented in the surgical treatment of patients with spinal diseases [3]. Spinal fusion surgery is the surgical treatment that can be performed to cure spinal diseases like degenerative disc disease and scoliosis. This treatment requires the removal of the damaged intervertebral disc from the spine, then replacing it with a titanium cage that sits between two vertebrae. The titanium plates will be attached to the vertebrae using pedicle screws as fasteners, above and below the titanium cage to give additional support to the spine after surgery [1]. Meanwhile, the spinal rod is used together with the cage and pedicle screw, to add stability to the spinal implant [4].

The most widely used biomaterial for spinal implants right now is Ti–6Al–4V [3]. Ti-6Al-4V is a titanium alloy that consists of six percent of aluminium and four percent of vanadium [5]. The alloy has good resistance to corrosion, wear, and fatigue as well as excellent load-bearing capabilities. However, the Ti-6Al-4V may cause stress shielding [1]. One of the requirements when selecting a material for bone treatment is that it must have excellent mechanical properties, in which it must have values that approximately equal to the healthy bone to avoid the stress shielding [6]. The titanium alloy has elastic modulus value of 110 GPa, which is much higher than the value of normal bone tissue, which is 18 GPa. The significant difference in the elastic modulus value contributes to the formation of stress shielding layer, which is not conducive to the growth of new bone [7].

Another disadvantage of Ti-6Al-4V is that it is a non-biodegradable implants material, in which it will not degrade and not become absorbed into the surrounding tissue after implantation [8]. Hence, ZK60 is used as a substitute for the titanium alloy due to its biodegradable and bioabsorbable characteristics. ZK60 is a magnesium alloy that consists of six percent of zinc and 0.5 percent of zirconium [9], that are both biologically safe to human body. The alloy has elastic modulus value of 43 GPa, which is much closer to the value of normal bone tissue, which is 18 GPa, hence reducing the stress shielding effect [7]. In addition, the use of biodegradable implants eliminates the need for secondary surgeries for implant removal, consequently reducing the costs of medical [10]. Furthermore, the ZK60 is relatively better than the Ti-6Al-4V as it has closer stiffness modulus with bone (E = 12,000 MPa) and good bone conduction activity [7].

The previous study has shown that any spinal implant, in which having a porous structure can improve motion of spinal, reduce damage to adjacent vertebral tissues, and enhance fusion of bone [3]. The porous structure can be implemented to the implants by using 3D printing technology [11]. Triply periodic minimal surface (TPMS) lattice structures such as Gyroid, Diamond, and Split-P are more preferred than the traditional strut-based lattice structures such as body-centered cubic (BCC) and face-centered cubic (FCC). The excellent pore connectivity and large surface of TPMS lattice structures make them more suitable for mass transport and growth of bone tissue. Meanwhile, the traditional lattice structures have stress concentrations and fractures occur at their nodes, making them unsuitable for the design of bone implants [12].

Lehder *et al.* [13] have studied about the method to optimize the geometry of six TPMS-based lattice structures to maximize rate of cell growth, while maintaining an elastic modulus that is equivalent to human bone. They found that Diamond provided the highest rate of cell growth after Lidinoid and Split-P because of its huge size of pore, large SA/VR, and high local curvature, making them the best candidate for bone scaffolds. Meanwhile, Ali *et al.* [14] proved that Diamond was first ranked in capacity of energy absorption based on their study about the mechanical behaviour of seven different polymeric lattices.

Based on the previous research, the analysis on the biomechanical behaviour of the cervical spine after implanting different interbody fusion cages, which are the titanium alloy (Ti-6Al-4V) cage and magnesium alloy (ZK60) cage has successfully done. Then, based on the biomechanical analysis results, the microporous structure of the ZK60 cage was improved by the lattice topology optimization technology and validation of static structure [7]. However, the Ti-6Al-4V cage has not been optimized with lattice topology to see its performance under spinal motion. Therefore, the limitation on the previous study has led to this study that aimed to optimize the spinal cages of Ti-6Al-4V and ZK60 by incorporating lattice alteration to the implants and to evaluate the stress distribution of optimized spinal cages under sitting condition.

METHODOLOGY

Construction of the three-dimensional (3D) model of spinal cage

The three-dimensional (3D) model of spinal cage was constructed using CATIA V5R20 software as shown in Figure 1. Then, the model was exported as an STP file to prepare for lattice alteration.

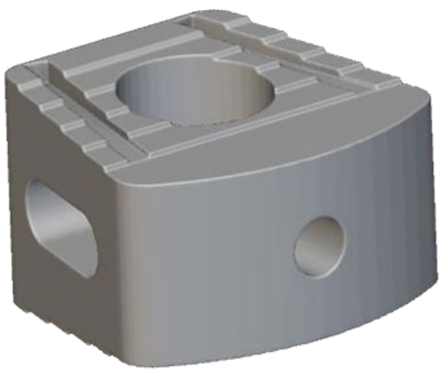
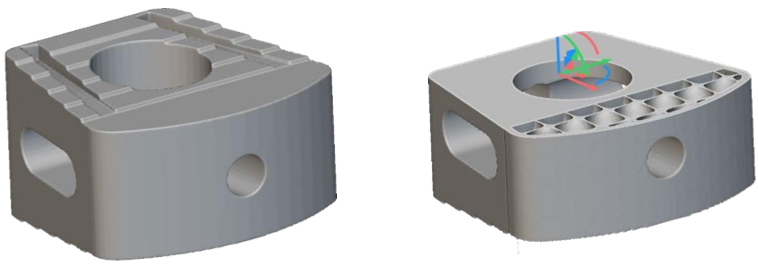


Figure 1: 3D model of the spinal cage

Spinal cage optimization with lattice alteration

Contents The spinal cage was optimized with Diamond sheet structures, which is a triply periodic minimal surface (TPMS) lattice structure. The excellent pore connectivity and large surface of the TPMS lattice structure makes it more suitable for growth of bone tissue. Meanwhile, the sheet-networks type of TPMS lattice structure have better mechanical properties than solid-networks at a certain relative density, and they have a larger specific surface, which is conducive for cell adhesion [12]. For instant, Young's modulus of sheet-networks TPMS is 15.9 GPa, much closer to bone properties as compared to solid-networks TPMS (34.6 GPa) for the same structure [15]. The optimization of spinal cage was done using nTop software. The software has a powerful topology optimization (TO) module that allows user to get topology optimized parts through a robust process and validate them. Furthermore, the nTop workflow allows to make changes in all variables of the TO process and can quickly get different possible results afterwards [16].

Firstly, the 3D model of spinal cage was imported into nTop in the type of STP file. The workflow of the spinal cage optimization by incorporating lattice alteration is shown in Figure 2. Figure 2(a) shows the solid spinal cage, while Figure 2(b) illustrates the cross section of optimized spinal cage that contains Diamond sheet structures.



(a) (b)

Figure 2: Workflow of spinal cage optimization: (a): solid spinal cage, (b) cross section of optimized spinal cage

The spinal cage was optimized with uniform lattice structures, having the size of unit cell of 4mm and a constant sheet thickness of 200 μm. The porosity, ϕ of the lattice structures is 21%, calculated by Eq. (1).

$$\emptyset = 1 - \frac{Volume\ of\ optimized\ model}{Volume\ of\ solid\ model} \quad (1)$$

Finite Element Analysis (FEA) model establishment

Equations finite element analysis (FEA) model of the solid and optimized spinal cages were established to evaluate their stress distribution and total deformation under sitting condition. The spinal cages were meshed with triangle elements in 0.25 mm size as shown in Figure 3. This high-quality element size was selected based on the convergence analysis conducted by Bin et al. [17]. The number of elements and nodes of the spinal cages are 212,791 and 62,329, respectively.

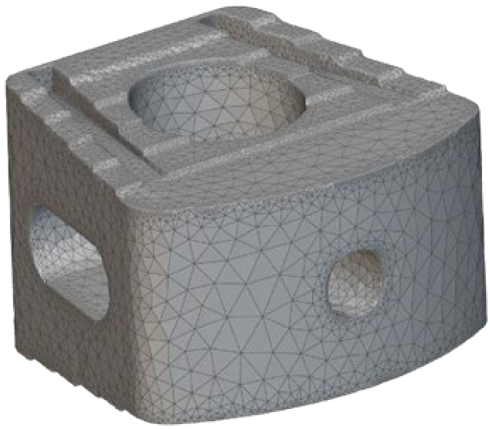


Figure 3: 3D finite element model of the spinal cage

Both spinal cages, solid and optimized, were set under boundary conditions to simulate them under sitting condition. An axial force of 300 N was applied to the upper surface of spinal cages, considering the weight and inertial load from the head under sitting condition [18]. Then, a fixed support condition was applied at the bottom of spinal cages to restrain any rotation and movement [19]. Figure 4 shows the boundary conditions of the spinal cage.

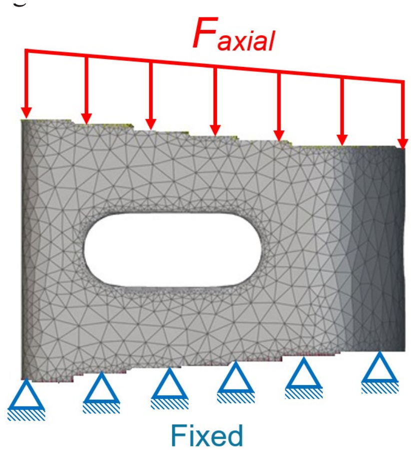


Figure 4: Boundary conditions of the spinal cage

The solid and optimized spinal cages were defined as Ti-6Al-4V and ZK60, producing two spinal cages for each material. The material properties of Ti-6Al-4V and ZK60 are shown in Table 1.

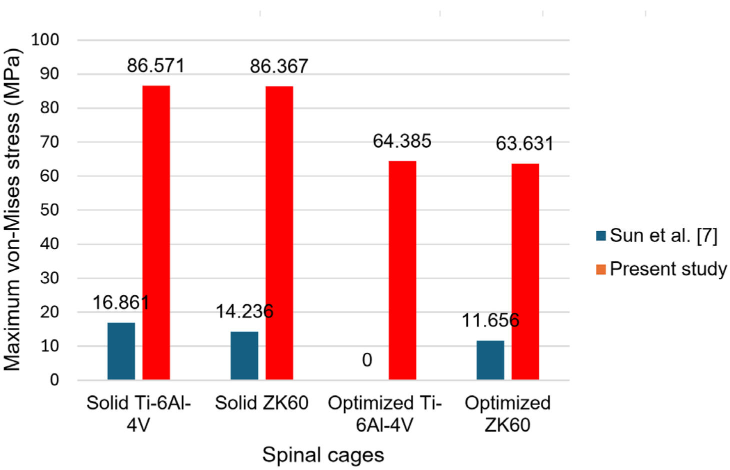
Table 1: Material properties of Ti-6Al-4V and ZK60 [7]

Material Properties	Ti-6Al-4V	ZK60
Density (kg/m^3)	4429	1835
Young's Modulus (MPa)	113,800	44,660
Poisson's Ratio	0.339	0.305

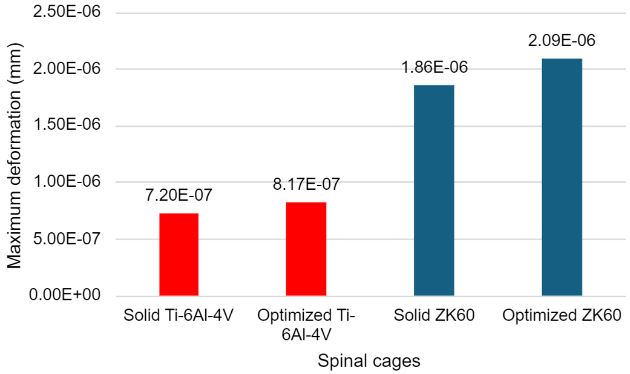
RESULTS AND DISCUSSION

The most widely used biomaterial for spinal implants right now is Ti–6Al–4V [3]. The titanium alloy has good resistance to corrosion, wear, and fatigue as well as excellent load-bearing capabilities. However, the Ti-6Al-4V may cause stress shielding [1]. Stress shielding occurs when the unfractured bone beneath the implant is not under significant loads, due to the transfer of loads is onto the implant itself. This condition causes the intact parts of the bone to lose its density, so it is important to consider an implant material that has the lowest possible stiffness at the ends of the implant to let the bone to carry loads [20]. The elastic modulus measures the stiffness of a material, where the stiffness refers to the ratio of generalised force to the generalised displacement [20]. The lower the modulus of elasticity, the lower the material stiffness. The titanium alloy has elastic modulus value of 110 GPa, which is much higher than the value of normal bone tissue, which is 18 GPa. Meanwhile, magnesium alloy of ZK60 has elastic modulus value of 43 GPa, which is much closer to the value of normal bone tissue, hence reducing the stress shielding effect [7]. In addition, the use of biodegradable implants such as ZK60 eliminates the need for secondary surgeries for implant removal, consequently reducing further the costs of medical [10].

Figure 5 shows the comparison of maximum von-Mises stress between present study and prior study by Sun *et al.* [7], and comparison of maximum deformation between solid and optimized spinal cages. The stress in the current study showed comparable trend with the previous biomechanical and finite element analysis (FEA) study by Sun *et al.* [7] as shown in Figure 5(a). Both current and previous study showed lower maximum stress in optimized group of spinal cages compared to the solid group, and lower maximum stress in ZK60 group than the Ti- 6Al-4V group. The maximum stress in solid cages of Ti-6Al-4V and ZK60 were reduced by 25.63% and 26.32%, respectively after the cage optimization. On the other hand, according to the study conducted by Jun *et al*., the maximum stress in solid cage of ZK60 was reduced by 18.12% after the optimization. Based on the results of static analysis, the maximum stress of solid spinal cages of Ti-6Al-4V and ZK60 were 86.571MPa and 86.367 MPa, respectively. The maximum stress of optimized cages of Ti-6Al-4V and ZK60 were 64.385 MPa and 63.631 MPa, respectively. Meanwhile, according to the results of the previous study, the maximum stress of solid spinal cages of Ti-6Al-4V and ZK60 were 16.861 MPa and 14.236 MPa, respectively. The maximum stress of optimized cage of ZK60 was 11.656 MPa. The lower maximum stress value of ZK60 group shows better mechanical properties of ZK60 than Ti-6Al-4V. Furthermore, the maximum stress of solid group and optimized group of spinal cages for both Ti-6Al-4V and ZK60 was far lower than their yield strength, which are 880 MPa and 365 MPa, respectively.



(a)



(b)

Figure 5: The comparison of (a) Maximum von-Mises stress between present study and previous study, and (b) Maximum deformation between solid and optimized spinal cages

Although the elastic modulus value of ZK60 is much lower than Ti-6Al-4V, the value is still higher than that of normal bone tissue. Hence, the optimization of spinal cage with porous structure was done to reduce its stiffness. The spinal cages of both Ti-6Al-4V and ZK60 were optimized with non-uniform Diamond sheet structures, having the size of unit cell of 5mm and a constant sheet thickness of 200 μm. Based on a previous study, Diamond can provide the highest rate of cell growth after Lidinoid and Split-P because of its huge size of pore, large SA/VR, and high local curvature, making them the best candidate for bone scaffolds [13]. Besides, Diamond was also first ranked in capacity of energy absorption based on the study about the mechanical behavior of seven different polymeric lattices by Ali *et al*. [14]. The lattice structures were only distributed in the front of the cage as shown in Figure 2(b) as the stress at the rear of the cage is relatively large than the stress at the front [7]. The volume of the solid and optimized cages was 1173.41 mm^3 and 926.85 mm^3, respectively, producing porosity of 21%.

The finite element analysis (FEA) model of the solid and optimized spinal cages were established to evaluate their stress distribution and total deformation under sitting condition. A force of 300 N was applied to the upper surface of spinal cages as shown in Figure 4, considering the weight and inertial load from the head under sitting condition [18]. Figure 6 shows the stress distribution in the spinal cages for solid Ti-6Al-4V, solid ZK60, optimized Ti-6Al-4V and optimized ZK60. A larger stress concentration area can be seen through solid group cages than that of optimized group. Based on the previous study, the stress shielding occurs when the load is asymmetrically distributed [20]. Porous structures ensure a more uniform distribution of load than a solid [21], resulting in a smaller stress concentration area that can be seen through the optimized group cages. However, there was no significant difference in the stress distribution between spinal cages of Ti-6Al-4V group and ZK60 group. The finite element analysis (FEA) model of the cages was established under sitting condition, in which sitting is considered as not to meet the guidelines of a physical activity [22]. Hence, Zhang *et al.* suggested that more active activities will give a greater difference in the stress distribution of cages.

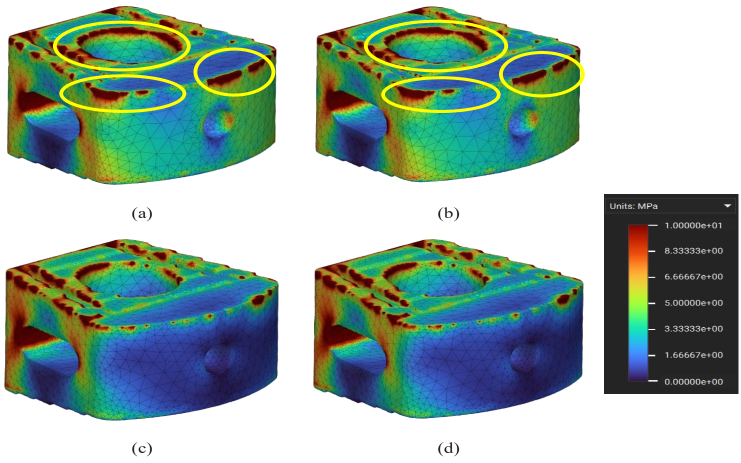


Figure 6: Von-Mises stress of spinal cages under sitting condition: (a) solid Ti-6Al-4V, (b) solid ZK60, (c) optimized Ti-6Al-4V, and (d) optimized ZK60

As for the total deformation, a larger deformation occurred in the optimized group of spinal cages than that of solid group. Figure 7 shows the total deformation in the spinal cages for solid Ti-6Al-4V, solid ZK60, optimized Ti- 6Al-4V and optimized ZK60. The result shows that the optimized group of spinal cages further reduced their stiffness with the application of porous structure, based on the greater total deformation in the cages as illustrated in Figure 7(c) and Figure 7(d). Besides, a larger deformation still occurred in the spinal cages of ZK60 group than the Ti-6Al-4V group as shown in Figure 5(b). This result is due to the difference in stiffness between Ti-6Al-4V and ZK60, as the elastic modulus of ZK60 is much closer to the value of normal bone tissue than that of titanium alloy. Since the elastic modulus measures the stiffness of a material, so the lower stiffness will result in a lower ratio of generalized force to the generalized displacement [20], producing larger deformation. Furthermore, based on the Wolff's law, an intact bone will modify itself to adapt with the force applied onto it, hence it needs to carry load during any time to stay healthy in the body [21]. The effect of stress shielding will curb this mechanism, hence it is important to select a material for bone treatment that has excellent mechanical properties, in which it must have value that approximately equal to the healthy bone to avoid the stress shielding [6] that can reduce bone mineral density, simultaneously increasing the risk of periprosthetic fracture [22].

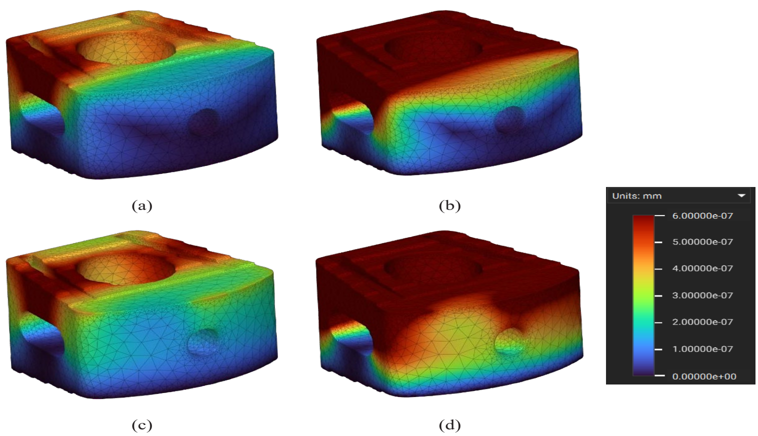


Figure 7: Deformation of spinal cages under sitting condition: (a) solid Ti-6Al-4V, (b) solid ZK60, (c) optimized Ti-6Al-4V, and (d) optimized ZK60

However, the present study has several limitations. Firstly, the finite element model was simplified by only considering the spinal cage, and neglecting the bone properties, ligament and muscle of the cervical spine. Hence, the model cannot fully imitate the state of cervical spine during sitting condition. Secondly, the finite element analysis (FEA) model was only established under static condition, so further studies need to be done to analyze the stress distribution of the spinal cages under dynamic condition.

Lastly, the effect of various porosity was not considered, so it should be considered in future study.

CONCLUSION

In summary, the lower maximum stress value of ZK60 in the findings showed better mechanical properties of ZK60, as the magnesium alloy has elastic modulus value that is much closer to the value of normal bone tissue (E = 12,000 MPa) than Ti-6Al-4V (Refer Table 1). Hence, it is important to select a material for bone treatment that has excellent mechanical properties, in which its value is approximately equal to the healthy bone to avoid the stress shielding [6]. Besides, the findings showed an increase of 13% deformation occurred in the optimized group of spinal cages than that of solid group. The result satisfied that the optimized group can further reduce its stiffness with the application of porous structure, consequently reducing the effect of stress shielding. In addition, a larger deformation also occurred in the spinal cages of ZK60 group than the Ti-6Al- 4V group due to its lower elastic modulus. As for future work, the optimization of spinal cages using different types of lattice structures can be considered to compare their reduction of cages stiffness.

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