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RESEARCH ARTICLE - MECHANICAL ENGINEERING

Optimal Material Selection for Total Hip Implant: A Finite Element Case Study

Abdullah Tahir Şensoy[1] · Murat Çolak[2] · Irfan Kaymaz[3] · Fehim Findik[4,5]

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Abstract

The selection of most proper materials in engineering design is known as an important stage of the design process. In order to successfully complete this stage, it is necessary to have sufficient knowledge about the structure of materials, density, melting point, thermal expansion coefficient, tensile and yield strength, elongation, modulus of elasticity, hardness and many other properties. There are several selection systems that help the design engineer to choose most suitable material that meet the required properties. In the field of bioengineering, the selection of materials and the development of new materials for the clinical needs are increasingly important. In this study, the cases of optimal implant stabilization were investigated, material alternatives for hip prosthesis were evaluated, and optimal materials were determined. Using computerized tomography data with MIMICS software, virtual surgery was applied the hip bone and the implant was attached to bone. Boundary conditions and material properties have been defined, and finite element model has been created. FEA investigation of the mechanical behavior of the hip implant for various material alternatives determined by the CES software showed that the best material candidate is austenitic, annealed and biodurable stainless steel in terms of the micromotions at the implant–bone cement interface regarding osseointegration. This candidate showed 20.69% less strain value than the most commercially used hip implant material, Ti6Al4V. Therefore, the findings of this study suggest that the use of some specific stainless steel materials for implants may reduce the operation cost and increase the operation success for the total hip arthroplasty.

Keywords Hip prosthesis · Optimal material · Selection · CES · Finite element analysis · Implant

1 Introduction

Material selection is one of the most fundamental issues in engineering design. It is very important and difficult to choose among the more than 200,000 materials that can meet different design requirements and have the characteristics to be used by the designer. It has been investigated by many researchers for more than 20 years that to find the most proper material in various applications. The selection of the most proper materials in engineering design is known as an important stage of the design process. The use of each material requires a selection methodology and this can be considered as a problem-solving activity [1–3]. This activity requires a decision-making process using a deep know-how and engineering methods. Considering the importance and difficulty of solving the problem can be better understood with the daily life examples. For instance, a tractor has 15–20 thousand, a car has 25–30 thousand, a tank has 40 thousand, a submarine has 120 thousand and a plane has 2–6 million parts made from different materials [4–11].

✉ Abdullah Tahir Şensoy
tsensoy@bayburt.edu.tr

Murat Çolak
mcolak@bayburt.edu.tr

Irfan Kaymaz
irfan.kaymaz@erzurum.edu.tr

Fehim Findik
findik@sakarya.edu.tr

1 Technical Sciences Vocational School, Bayburt University, Bayburt, Turkey

2 Mechanical Engineering Department, Bayburt University, Bayburt, Turkey

3 Engineering Faculty, Erzurum Technical University, Erzurum, Turkey

4 Technology Faculty, Sakarya Applied Sciences University, Sakarya, Turkey

5 BIMAS-RC Research Center, Sakarya University, Sakarya, Turkey

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The oldest and simplest method of material selection methods is the conventional selection method which is getting the required information such as the physical properties and performance characteristics of the materials from material manufacturers, suppliers and standards. In addition to conventional design procedures, computer-based material selection methods have been developed [9, 12–15]. The first of these methods was done by Ashby [16, 17] and Dargie [18]. They have introduced the Ashby and Granta design graphical interface programs for the elimination method [19]. It is possible to see where the material groups are located for the two existing features by means of Ashby diagrams. Information-based systems are available as the second method, but the complexity and intensive knowledge necessity of these methods limit their use [20]. Shanian and Savadogo have developed various methods, but these methods do not allow that materials to be relatively weighted [20–22].

In recent years, some researchers have presented different methods in order to offer a better material selection strategy in terms of desired properties. Findik and Turan [23] used the weight property index method (WPIM) for optimal material selection. Another approach, expert system-based material selection, was offered by some other researchers to provide a good opportunity for filtering of alternatives [24]. To sum up, the significance of the related topic has attracted great attention from many scientists studying in a wide range of disciplines. On the other hand, finite element analysis (FEA) is a well-known, reliable and very useful numerical solution technique used by many researchers [25]. Therefore, in this study we aimed to offer an insight into FEA-based optimal material selection strategy for patient-specific biomedical applications.

In recent years, material selection and also developing “new materials” in biomedical engineering has gained more popularity in order to better simulate the mechanobiological properties of the living tissues. Replacing the damaged or injured tissue with an artificial product may lead several complications or disfunctionalities. Therefore, materials used as medical replacements should have been optimally selected in order to meet the physiological, morphological and the mechanobiological properties of the original one. The prosthesis can be considered as a replacement for missing body parts. Unhealthy organ should be removed from the body. Organ damage resulting from different facts such as injuries, tumor surgeries, gangrene or inflammation causes a necessity to remove the unhealthy part in order to protect the rest of the body health. These implants are located into the problematic region using auxiliary material such as bone cement and surgery equipment [21, 26]. The optimal material selection for prosthetic organs has four main steps. First one is to analyze the material requirements. This step has several goal parameters such as high strength, high corrosion resistance, low cost, biocompatibility, malleability and product reliability. The second step is to select the candidate materials regarding the goal parameters mentioned above. Third is ranking the candidates according to the goal parameters. The final step is selecting the optimal material for the specific case.

Since a great need for high amounts of artificial replacements in healthcare industry, the design solutions should have both enough mechanobiological performance and low manufacturing cost.

Total hip arthroplasty (THA), first introduced in 1938, has had a wide use with over 300,000 operations carried out world-wide every year [27]. THA is one of the most popular research areas in bioengineering and the patient population in need of hip joint replacement is constantly increasing. On the other hand, the proportion of the young patients in this population is also increasing day by day because of lifestyle factors [28].

The literature review in the field of biomedical engineering showed that different methods were used for optimal material selection. However, there is not a gold standard in terms of material selection method and there is still need for further investigations in order to offer a better material selection approach. Although, some researchers offered different methods for material selection of hip implant such as analytic hierachy process (AHP) method [29], AHP-integrated Vise Kriterijumska Optimizacija Kompromisno Resenjemeaning (VIKOR) method [30] and the multi-objective optimization on the basis of ratio analysis (MULTIMOORA) [31], none of them considered the complex and patient-specific geometry of the femur.

To the best of our knowledge, no study has considered to integrate a material selection method with a patient-specific FEA model of a cemented hip implant. In this study, since the patient-specific anatomy has a great importance on the biomechanical performance of the implant, the femur of the patient was accurately modeled using grayscale values of the computed tomography (CT) images. The image processing steps were done using MIMICS 17.0 software. Simultaneously, a computer-aided material selection software, CES selector 2014, developed by Ashby and Granta, was used in order to determine the material alternatives for hip prosthesis in terms of boundary limits such as biocompatibility, biomechanical performance and cost. The material alternatives determined by the CES software were identified to ANSYS software, and the mechanical behavior of a hip implant was investigated for each model using FEA. The best material candidate for THA was determined in terms of the micromotions at the implant–bone cement interface regarding osseointegration.

2 Methods

This section includes biomedical modeling for virtual surgery, optimal material selection with CES software and FEA simulations regarding candidate materials selected.

2.1 Biomedical Modeling

In this study, 3D bone model of femur was created by using computed tomography (CT) images in database of a modeling software [Mimics 17.0 software (Materialise, Leuven, Belgium)]. Cortical and trabecular segments of the femur are separately modeled in order to better reflect the mechanical properties of the bone under physiological loading. In the following step, a well-known hip implant is modeled and placed to the femur. To better represent the real surgery, bone cement modeled was assembled to the bone–implant interface. The components of the total hip implant are repositioned according to anatomical posture of the human body. Modeling steps performed in Mimics 17.0 and 3-Matic 9.0 (Materialise, Leuven, Belgium) softwares are illustrated in Fig. 1. The steps were importing CT images, segmentation, creating 3D bone model, virtual surgery and creating finite element volume mesh in 3-Matic, respectively.

Generated mesh structures of the femoral components are then exported as *.cdb files to create a finite element (FE) assembly model using ANSYS Workbench 16.0 (ANSYS, Inc., Canonsburg, PA, USA)

2.2 Optimal Material Selection

In this study, all material selection procedures were conducted using a computer-aided material selection program (CES selector 2014) to define the material candidates for hip prosthesis. Firstly, background information such as mechanical properties and manufacturing process about the candidate materials was reviewed using the database of the software. In the following step, the material selection process has been continued by defining the boundary limits in accordance with the desired biomechanical performance of the hip prosthesis [5, 19]. The material database of the software and the boundary condition values used for material selection process are given in Table 1.

After boundary limits applied, it has been obtained that 10 material candidates were met the required specifications. Comparison between the material candidates obtained from CES software is given in Table 2. For each material option, "M" notation was used in order to provide a better understanding for the following sections of the study.

According to the boundary condition values given in Table 1, a computer-aided material selection program, CES selector, was run and 10 different materials which are suitable for hip prosthesis were determined. In order to select the most suitable material between these candidates in terms of biocompatibility and biomechanical performance, ANSYS

Table 1 The material database of the software and the boundary conditions

Attribute	Constraints
Material universe	Ceramics and glasses
	Fibers and particulates
	Hybrids: composites, foams, honeycombs, natural materials
	Metals and alloys
	Polymers: plastics, elastomers
Price (USD/kg)	≤ 20
Density (kg/m^3)	8×10^3
Tensile strength (MPa)	≥200
RoHS (EU) compliant grades	✓
Medical grades? (USP Class VI, ISO 10993)	✓

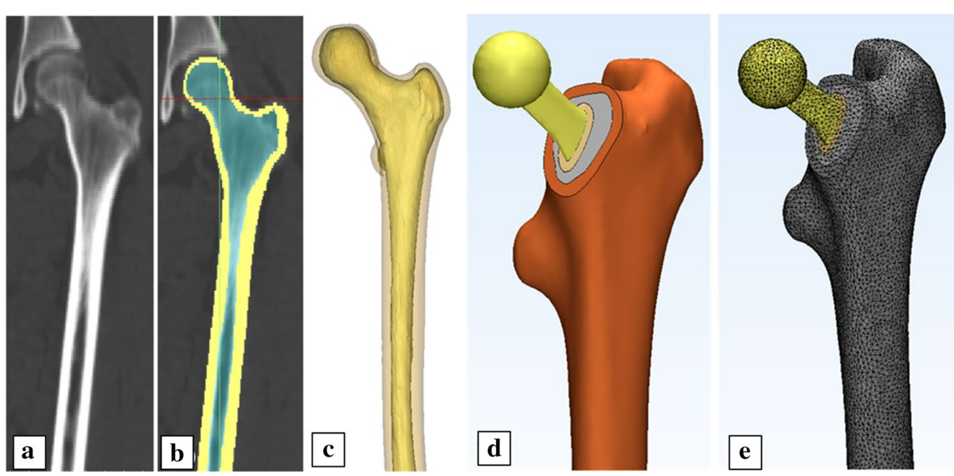


Fig. 1 **a** Importing CT images, **b** segmentation, **c** creating 3D bone model, **d** virtual surgery and **e** creating finite element volume mesh in 3-Matic

Table 2 The mechanical properties of the candidate materials

Materials	Notation	density (kg/m^3)	Young's modulus (GPa)	Yield strength (MPa)	Tensile strength (MPa)	Elongation (%strain)	Poisson's ratio
Nickel–titanium alloy wire, annealed, austenitic	M1	6430 6560	– 41 to 75 (58)	324–472	1030–1290	10–17.5	0.32–0.33 (0.325)
Nickel–titanium alloy, austenitic	M2	6410 6540	– 41 to 83 (62)	195–690	895–1900	5–50	0.32–0.34 (0.33)
Polyarylamide (50% glass ftber)	M3	1630 1660	– 17.8 to 22.2 (20)	231–289	234–286	1.8–2.6	0.32–0.33 (0.325)
PPS (40% glass ftber)	M4	1600 1670	– 7.58 to 14.5 (11.04)	138–145	121–201	0.9–4	0.34–0.36 (0.35)
Stainless steel, austenitic, AISI 301, wrought, annealed	M5	7880 7960	– 200 to 210 (205)	179–207	503–556	30–40	0.27–0.28 (0.275)
Stainless steel, austenitic, Bio-Dur 108, wrought, annealed	M6	7560 7720	– 198 to 202 (200)	580–592	922–940	51.5–52.5	0.29–0.30 (0.295)
Stainless steel, martensitic, AISI 410, wrought, annealed	M7	7650 7850	– 190 to 210 (200)	276–310	483–600	20–35	0.27–0.28 (0.275)
Titanium, commercial purity, R50700	M8	4490 4530	– 107 to 112 (109.5)	172–483	241–552	10–25	0.34–0.35 (0.345)
Zirconia bioceramic	M9	5850 5900	– 205 to 212 (209.5)	750–850	750–850	0.35–0.41	0.24–0.26 (0.25)
Ti6AlV	M10	4430 4450	– 113 to 115 (114)	768–898	869–996	8–10.1	0.34–0.36 (0.35)

modeling was performed. The FEA models were considered as homogenous and linear isotropic. The mean values of elastic modulus (E) and Poisson ratio (ν) for each material candidate were calculated using the upper and lower limits given in Table 2 and used as material properties for ANSYS modeling steps.

2.3 Finite Element Case Study

First of all, the candidate materials obtained using CES software were defined to the database of ANSYS workbench. All candidate materials and bone segments were assumed as linear elastic. For the material properties of the femur, the Young's modulus was defined as 20.3 GPa for the cortical bone and 2.13 GPa for the trabecular bone [32]. Considering the dynamic conditions, the E value of bone cement was taken as 9.8 GPa [33]. The Poisson's ratio values for all segments were taken as 0.3 [32]. The material data and the FE assembly model were transferred to the ANSYS Workbench static structural module. The boundary conditions are defined according to real anatomical loads. Table 3 shows the loading conditions used in numerical models.

Figure 2 shows the boundary conditions and the mesh structure of proximal femur where the virtual surgery applied. As seen in the picture, resultant hip joint reaction and muscle force values are 2872.9 N and 1239.4 N, respectively.

3 Results

Considering the maximum von Mises strains and displacements of the implant, cement and the bone layers, zirconia bioceramic (M9) shows the lowest values. Different types of stainless steel (M5, M6 and M7) also offer a good biomechanical stability. However, PPS (M4) shows worst performance as shown in Table 4.

The results given in Table 4 are graphically illustrated in Fig. 3 in order to better compare the femoral regions in terms of biomechanical responses. It can be seen from the figure that M10 has the least von Mises strain value

Table 3 Loading conditions innumerical models [34]

	Hip joint force (N)	Muscle force (N)	Distal femur condyles
X-direction (medial–lateral)	320	−310	Fully restrained
Y-direction (anterior–posterior)	170	0	Fully restrained
Z-direction (inferior–superior)	−2850	1200	Fully restrained

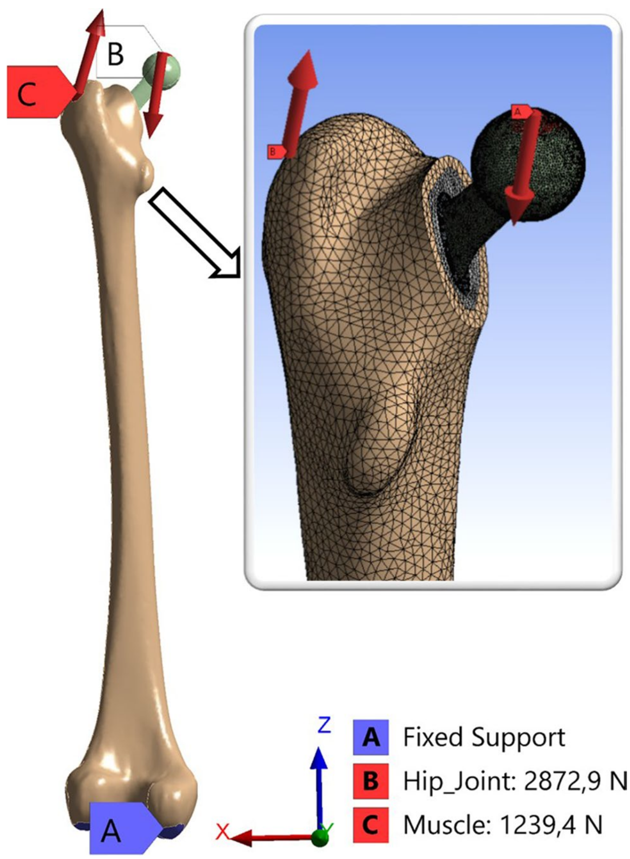


Fig. 2 Boundary conditions and ANSYS mesh structure of the virtual surgery FE model

for the cortical bone in comparison with the other models. However, same model has the greatest strain value for trabecular bone.

Considering the biological compatibility, osseointegration and the marketshare of the materials, M4 cannot pass the comparison criteria even it has the highest stability rate. M5, M6 and M7 have similar responses; but in terms of biocompatibility, M6 eliminates the others. Therefore, M6 and M10 show promising results for a detailed comparison.

Figures 4 and 5 show the total deformation distributions for femoral regions and von Mises elastic strain distribution for bone–implant complex for M6 and M10, respectively. From medial to the lateral regions, implant, bone cement, trabecular bone and cortical bone were investigated. In all components of FE model, displacement values obtained in M6 are much lower than M10. Similarly, maximum von Mises elastic strain values are much lower for M6. For instance, the most critical strain value obtained for M10 was decreased by 20.69% from 0.087 to 0.069.

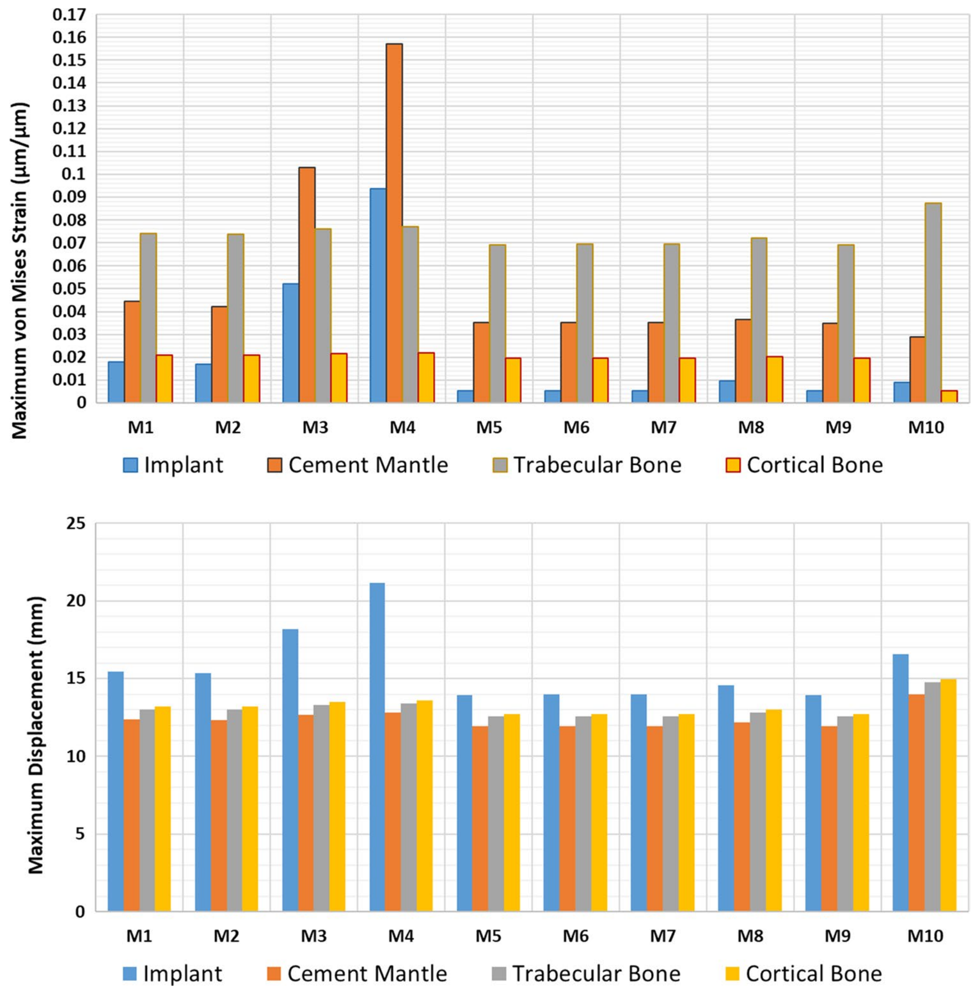
4 Discussion

Stainless steel was used in the early prosthetic hip designs for the femoral stem and ball. Presently, the femoral stem is constructed from either Co–Cr–Mo or Ti metals. Some models employ low sulfured 316 L stainless steel. Due to lower crevice and pitting corrosion resistance, and lower fatigue strength, employing of this material is gradually reduced [35]. Titanium-based alloys such as Ti–6Al–4V and Ti–6Al–7Nb are the most common materials preferred for hip implants [36], the reason why their good biomechanical properties such as high biocompatibility, high corrosion resistance and low density/strength ratio. Even so, the surface of stainless steel may be coated with Ti–6Al–4V in order to enhance the weaker biomechanical properties of this material such as biocompatibility and osseointegration [37, 38]. On the other hand, the release of aluminum and particularly vanadium ions from Ti–6Al–4V alloy can generate long-term health problems such as peripheral neuropathy, osteomalacia and Alzheimers disease. Therefore, recently instead of using aluminum, niobium is used and rather than using vanadium,

Table 4 Maximum von Mises strain and displacement values of the models

FE simulations	Maximum von Mises strain values (μm/μm)				Maximum displacement values (mm)			
	Implant	Cement mantle	Trabecular bone	Cortical bone	Implant	Cement mantle	Trabecular bones	Cortical bone
M1	0.01811	0.0445	0.0741	0.0209	15.45	12.36	13.02	13.20
M2	0.01695	0.0421	0.0739	0.0208	15.35	12.34	13.00	13.18
M3	0.05203	0.1028	0.0762	0.0216	18.17	12.64	13.29	13.48
M4	0.09360	0.1570	0.0770	0.0219	21.18	12.82	13.40	13.59
M5	0.00526	0.0351	0.0692	0.0195	13.94	11.92	12.55	12.72
M6	0.005286	0.0352	0.0694	0.0196	13.96	11.93	12.56	12.73
M7	0.00534	0.0352	0.0694	0.0196	13.96	11.93	12.56	12.73
M8	0.00963	0.0364	0.0720	0.0203	14.58	12.16	12.81	12.98
M9	0.00528	0.0350	0.0691	0.0195	13.92	11.91	12.54	12.71
M10	0.008999	0.0288	0.0872	0.0054	16.59	13.99	14.75	14.94

Fig. 3 Comparison of the models in terms of biomechanical response



zirconium and tantalum can be preferred in order to solve the possible long-term health problems in the body [39, 40, 42]. In addition, the use of 316L as the hip implant material may reduce not only the operational cost, but also eliminate the aforementioned problems.

Due to higher wear resistance and lower frictional stress at the joint, aluminum or zirconium oxide can be used for the ceramics choice for prosthetic device. Nevertheless, the Young's moduli of these ceramics are large and the fracture toughness of alumina is fairly low. Therefore, the femoral stem is still produced from one of the above alloys and is then attached to the ceramic ball; this femoral stem–ball component thus becomes a two-piece unit.

Preferably, the biomaterial should be biocompatible and also have enough mechanical properties with the replaced bone. Nevertheless, none of artificial material supplies the biocompatibility as well as suitable mechanical properties such as low elastic modulus, high fracture toughness, low friction coefficient and noble wear resistance. As a result, material property should be approval for the replacement of the bone. Unfortunately, artificial materials that are both biocompatible and fairly strong also have high modulus of elasticity [35]. Therefore, it is better to make porosities inside the implant materials [39].

Bioactive coatings should also be applied as a means to accomplish stability of fixation [40]. Lately, a group of scientists has shown the potential of a hydroxyapatite coating on a wire mesh surface to yield 30% bone ingrowth in a sheep knee model.

After 24 months, a substantial amount of resorption of the hydroxyapatite coating was detected, but still a high amount of interface stiffness and bone ingrowth was perceived. Seemingly mechanical interlocking overhauls the task of chemical bonding in offering stability of fixation [41].

Thanks to the porosity production, the mechanical incompatibility between the implant and bone was removed and the density value was close to the bone. In addition, porosities enable implant permeability for vascularization and mineralization, and support mechanical fixation. The optimum isolated and distributed pore size is 50–100 microns for biocompatibility and corrosion resistance. However, porosity reduced the corrosion resistance. For this purpose, bioactive composite coating containing hydroxyapatite–graphene oxide–collagen was carried out by anodization and

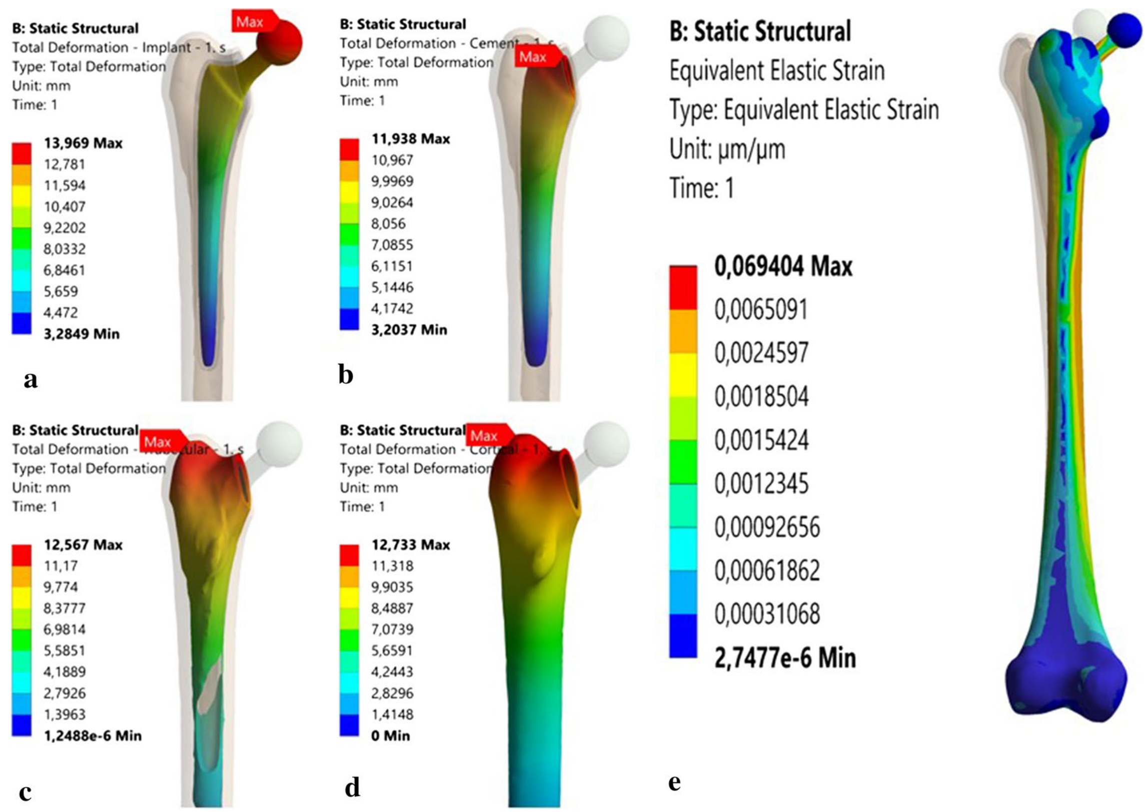


Fig. 4 Total deformation distributions of **a** hip implant, **b** cement mantle, **c** trabecular bone, **d** cortical bone and **e** equivalent von Mises elastic strain distribution in bone–implant complex for M6

electrodeposition processes. As a result of these surface modification processes, corrosion resistance was increased and osteoblast-like cell growth and viability were increased, and protein adsorption was improved by forming a hydrophilic surface [42]. The main motivation of this paper was to compare the possible implant materials while all other factors are fixed prior to the surface modification process.

There are many factors affecting the implant stability such as design of implant, bone quality, implant size, weight of the patient, implant material and surgical method (cemented and cementless surgeries) [43]. These surgical techniques are the methods to provide bonding between bone and implant in clinical applications. However, the cemented method is superior to the other method in terms of bioactivity and biocompatibility [44, 45]. Therefore, the cemented case was considered for all FE simulations in this study. FE results obtained in this study showed that the best material option for hip prosthesis is M6 in terms of maximum displacement values. On the other hand, according to equivalent von Mises strain values, the best-fitted material to Neo-Hookean model is Ti–6Al–4V [46]. The magnitude of smallest strain for cemented model, as 0.0253 stated in the literature [46], is a good agreement with the results obtained for M10.

Moreover, strain values determined along the implant from metaphysis to diaphysis region also show same trend with the literature for all FE models [47, 48].

Although, the variations in the force vectors were not considered within the scope of this paper; for future studies, probabilistic FEA approach may be used to consider not only the variations in force vectors related with the dynamic loading conditions, but also to represent the anthropometric differences. Thus, we may better understand the effects of material selection in different critical cases.

5 Conclusions

In this study, computer-aided material selection strategy for hip prosthesis was applied. Design parameters for hip prosthesis were limited as strength, malleability, corrosion resistance and biocompatibility. According to the boundary conditions defined in the CES software database, 10 different alternative materials which are nickel–titanium (2 kind), polyarylamide, PPS, stainless steel (3 types), pure titanium, zirconia bioceramic and Ti6Al4V were found to meet the specified boundary conditions.

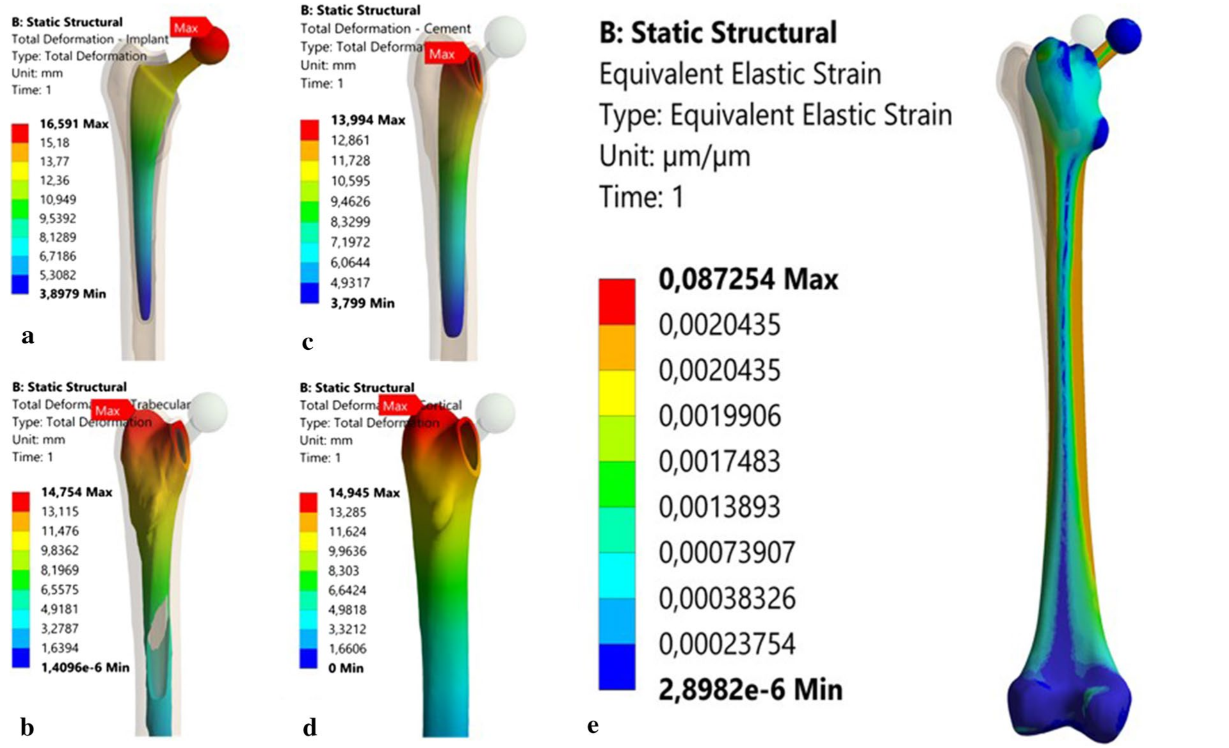


Fig. 5 Total deformation distributions of **a** hip implant, **b** cement mantle, **c** trabecular bone, **d** cortical bone and **e** equivalent von Mises elastic strain distribution in bone–implant complex for M10

The findings of this study suggest that the optimum material option for hip prosthesis is austenitic, annealed and biodurable stainless steel (M6) while limiting the material candidates in terms of biomechanical response under physiological loading and also biocompatibility. Although titanium alloys have much more interest for hip implant technology in recent years, austenitic, annealed and biodurable stainless steel may be a good alternative for hip implants in terms of biomechanical stability and operational cost.

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