**SCIENTIFIC PAPER**

# Evaluation of new hip prosthesis design with finite element analysis

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## Abstract

Cemented and cementless hip prostheses are used in total hip replacement surgery. Short, medium and long term success rates of these hip prostheses are controversial in the literature. Traditional cemented and cementless hip prostheses have advantages and disadvantages affecting the success of the implantation process. In this study, a new design of hip prosthesis is presented considering the advantages and disadvantages of the prostheses. Femur and prostheses were modeled and combined with each other to perform the finite element analysis (FEA). The new design of prosthesis was compared to the conventional prosthesis in terms of mechanical aspects. The evaluation criteria are the maximum von Mises stress and micro-movement of the contact between femur and prosthesis. In conclusion, the new design of prosthesis was found to provide a sufficient amount of primary stability and decreased the risk of stress shielding.

**Keywords** Hip prosthesis · Finite element analysis · New hip prosthesis · Computed tomography

## Introduction

Total hip prosthesis surgery is a successful method that frequently used for hip pain relief. An estimated one million hip prostheses are performed worldwide annually. During the first ten years of these operations, approximately 5–10% of patients is reoperated due to different reasons [1]. Considering the numbers, it is seen that there are serious loss of material and money and also reducing the quality of the patient's life. Aseptic loosening, bone resorption, pain, post-prosthetic bone fractures, and dislocations are the main causes of this revision. Generally, mechanical problems are among the causes of the revision in the short (1 year) and middle (5 years) period. These problems are insufficient implant design, stability, overload, etc.[2].

There are two methods, cemented and cementless, used in hip surgery. In the cemented method, bone cement is placed between the bone and the stem and the stem is fixed to the femur in this way. In the cementless method, it is fixed by a press-fit method which means nailing to the bone. The success rates of these methods are still being discussed. Some studies suggest the cemented method [3], while some other studies argue that the cementless method is more successful [4]. The primary mechanical problem in the cemented method is aseptic loosening. The main reason for the aseptic loosening in this method is the cement crack [5]. Because cement cracking triggers the emergence of all other problems. The load was transferred from the prosthesis to femur through the cement. For this reason, the stresses on the cement are very important in the formation of the cement crack. On the other hand, the formation of cement crack is associated with cement thickness [6]. Experimental and numerical studies emphasize that the thickness of the cement should be a minimum of 2 mm for optimum results [7]. In the cementless method, the main problem is the aseptic loosening related to the primary stability of the prosthesis. The insufficient stability causes the aseptic loosening of the prosthesis. According to the literature, the formation of osseointegration for the cementless method depends on the amount of the motion between the bone and the stem [8]. In this study, a new prosthesis design was performed to eliminate the mentioned problems and it was compared with the traditional one. The aim of this study is to reduce the risk of prosthesis failure by modifying the traditional cemented prosthesis.

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## Materials and methods

Traditional and new designed prostheses were modeled via SolidWorks as seen in Fig. 1. Femur model was generated using Computer Tomography (CT) images. The femur and prostheses models were combined in the SolidWorks program as in the surgery. The models were transferred to the Ansys Workbench software to analyze the system.

The material properties of all models were assumed to be linear, elastic and isotropic. The material properties of the femur were defined according to the density that was calculated from the CT images [9]. The material properties of Titanium Alloy (Ti6Al4V) which the most commonly used for the prosthesis were described for the prostheses. Polymethylmethacrylate (PMMA) was selected for the material properties of the bone cement (Table 1) [10]. The compressive strength of the bone cement is 93 MPa, the elasticity of the cement is 2130 MPa [11].

Mesh convergence study was performed by refining the element size from 6 to 2 mm at the 0.5 mm interval for the femur, 4 to 1 mm at 0.25 mm interval for the prosthesis and 2 to 0.5 mm at 0.25 mm interval for the cement models. The most appropriate element sizes for the optimum results were specified as 4, 1.5 and 0.75 mm for the femur, stem, and cement, respectively. Additional

mesh refinement was applied to the contact regions to get convergences. Solid 187 tetrahedron element was used for the finite element models. The contact types between cement and bone, cement and stem and bone and stem were defined bonded, debonded (frictionless), and frictional, respectively. The models were applied to static load obtained from the literature for a 700 N weight person walking normal speed. Maximum forces resulting from walking were applied to the stem head [12]. The muscle forces were performed as indicated by Duda et al. [13]. The distal femur was fixed in three directions as seen in (Fig. 2, Table 2).
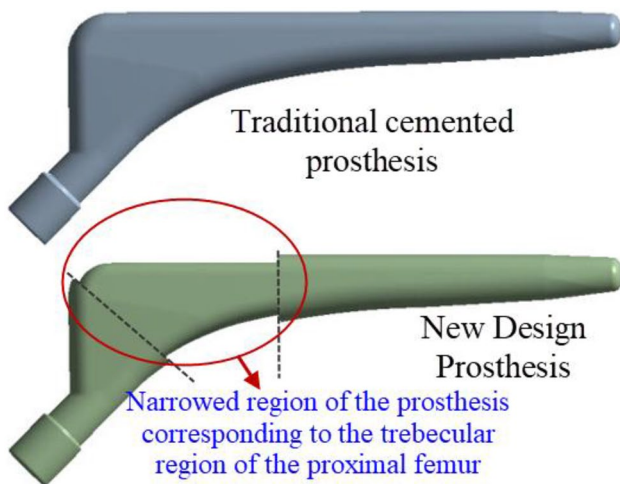


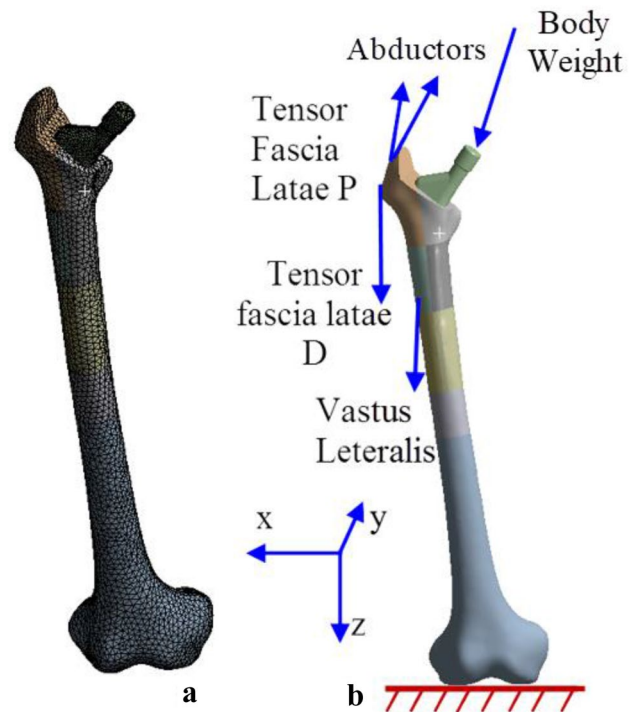**Fig. 1** Traditional and new designed prostheses models



**Fig. 2** (**a**) Finite element model of femur-stem system, and (**b**) loads and boundary condition

| | Materials | Density (g/cm$^3$) | Modulus of elasticity (MPa) | Poison ratio |
|---|---|---|---|---|
| Prosthesis | Ti6Al4V | 4.4 | 113,000 | 0.33 |
| Femur | Bone | $\varphi = 1.067 \times HU + 131$ | $E = 0.004 \times \varphi^{2.01}$ | 0.3 |
| Bone cement | PMMA | 1.18 | 2130 | 0.3 |

**Table 1** The material properties of the models [9, 10]

*HU* hounsfield unit, $\varphi$ density (g/cm$^3$)

**Table 2** Load values used in finite element analysis [12, 13]

| Force (N) | $F_x$ | Fy | $F_z$ | $F_{resultant}$ |
|---|---|---|---|---|
| Hip joint | 378 | 230 | 1603 | 1669.2 |
| Abductors | − 406 | − 30.1 | − 605.5 | 729.7 |
| Vastus leteralis | 6.3 | 129.5 | 650.3 | 663.1 |
| Tensor fascia latae lateral part | 3.5 | 4.9 | 133 | 133.2 |
| Tensor fascia latae proksimal part | − 50.4 | − 81.2 | − 92.4 | 132.9 |

## Results

### Validation of the FEA models

The strain values predicted by current FEA for femur under loading conditions are similar to the experimental study [14]. The difference rate between FEA results and experimental studies are found an average of 4.5%. Besides, FEA results from the current study are similar to the FEA studies in the literature [15, 16].

Two criteria were selected to determine the FEA results according to the literature [15]. One of them is the von Mises stresses to evaluate the strength of the prosthesis and cement. The other criterion is the axial normal strain distributions of the proximal femur to examine the stress shielding [16].

### von Mises stresses on the prostheses

The maximum von Mises stresses were obtained on the new designed prosthesis as seen in Fig. 3. The maximum stresses occurred in the distal region of the traditional prosthesis. The maximum stresses were calculated in the middle region of the new designed prosthesis due to stress concentration. The sharp edges of the new prosthesis were increased the von Mises stress on the prosthesis. However, the yield strength of the material is higher than the obtained stress values of both prostheses. For this reason, the mechanical failure of prostheses is not expected. If the sharp edges can be smoothed, the von Mises stress on the newly designed prostheses is reduced.

### von Mises stresses on the cement

The stress distributions of the cement were given in Fig. 4. The cement for traditional prosthesis has higher stress values than the new designed prosthesis. The risk of failure is higher in the cement used for the traditional prosthesis. Because the cement for the traditional prosthesis is between the bone and prosthesis. Thus, the forces acting on the prosthesis transferred to the femur throughout the cement. The cement stresses in the traditional prosthesis are concentrated in the distal part of the cement as seen in Fig. 4a. However,
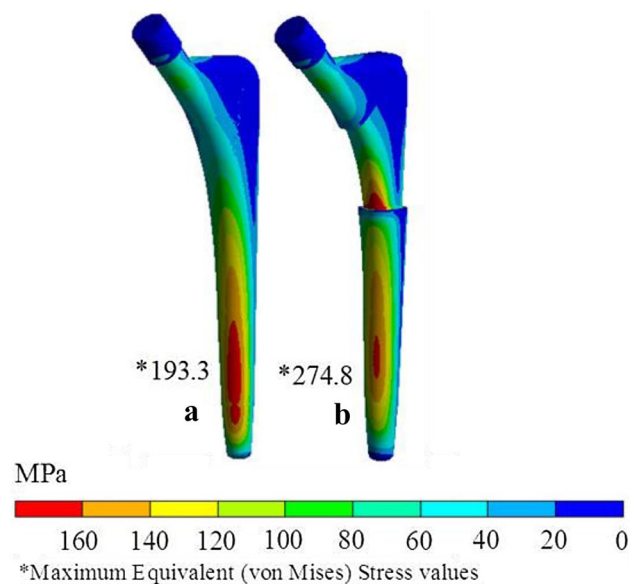


**Fig. 3** von Mises stresses on the prostheses. (**a**) Traditional cemented stem (**b**) new designed prosthesis

the maximum stress results were obtained in the proximal part of the cement for the new designed prosthesis. The maximum stress values were calculated 32.1 MPa for the traditional prosthesis and 11.1 MPa for the new designed prosthesis.

### The equivalent strain distribution on the femur

7 points were created for each region determined on the femur. The strain values were calculated at these points. These values were carried out to investigate the distribution of the load transfer in the femur after the stem placement and to observe the load transfer of these distributions in different prostheses. This criterion is an important parameter for studying the stress shielding and bone destruction in the femur after placement [17].

The highest compressive strain values were calculated on the intact (non-surgical) femur as seen in Fig. 5. The reference graph is the intact femur's graph. The higher strain value causes more bone density according to Wolff's Law [17]. Hence, more strain gives better results for a prosthesis. The strain values were decreased as a result of traditional and the new designed prostheses insertion, especially on the traditional prosthesis. The negative values in Fig. 5 demonstrate the compressive strain, the positive values demonstrate the tensile strength. The strain values on the femur that inserted the new designed prosthesis were obtained higher than the traditional one. Hence, the new designed prosthesis has more advantages compared with the traditional prosthesis. As a result, the new designed prosthesis

**Fig. 4** von Mises stresses on the cement mantles. (**a**) Cement for traditional stem (**b**) cement for new designed prosthesis
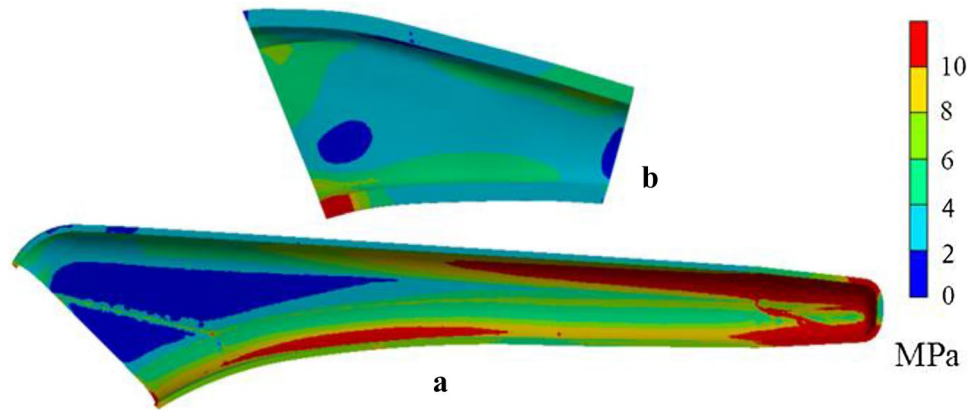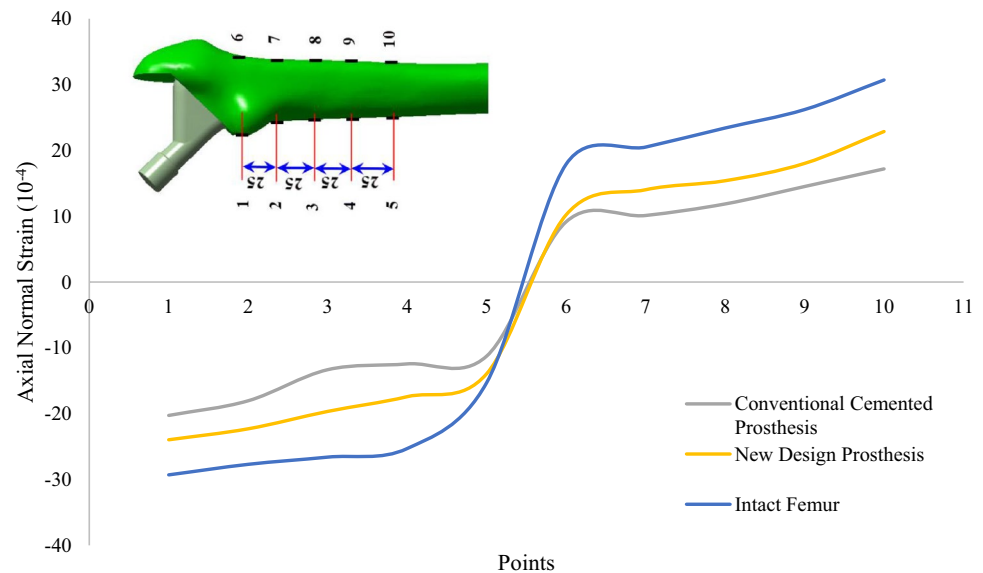


**Fig. 5** The equivalent strain values on the femur



was increased the strain values on the femur that prevent the stress shielding.

## Discussion

The determination of the proper method for the surgeon (cemented and cementless) is still unclear. The effects of the cement usage were evaluated in the previous study [15]. According to the study, the cement usage was reduced to stress shielding and aseptic loosening risk. However, the cement crack is an important problem for the cemented method. Hence, the advantages of both cemented and cementless methods were combined for the best results. In this study, the new hip prosthesis was designed according to this perspective.

Most of the problem that arises in the traditional hip prosthesis designs are due to mechanical reasons. The load transfer in the intact femur is from the outer region of the femur (cortical bone) to the distal end of the femur. However, this

load transfer changes when the hip prosthesis is inserted in the femur. The load in the proximal femur is decreased and this situation causes the bone weakening. This event, which is also called stress shielding, results in excessive weakening of the bone, loosening of the prosthesis and removal of the prosthesis. For these reasons, many design studies have been carried out on traditional prosthesis designs [18-20]. In this study, the new prosthesis design that combines the advantages of two different methods (cemented and cementless) used in the hip prosthesis placement is presented and this design is compared with traditional design.

The risk of cement cracking is the most important problem in cemented hip prostheses [1, 15]. Since the bone cement is placed between the bone and the prosthesis, the load transfer is transferred from the prosthesis to the cement and from the cement to the bone. The transferred load from the prosthesis, which has a higher elastic modulus, to the cement material that elastic modulus was lower than prosthesis material increases the risk of cement crack. The rapid integration of the bone with cement eases the insertion of the

prosthesis to the bone. Taking this advantage into account, the prosthesis design, which is presented as a new design, is intended to be created a cavity so that the bone cement can enter into the prosthesis and reduce the impact of the transferred load on the cement. Thus, the risk of cement damage is eliminated.

The aseptic loosening is the most important problem in cementless hip prostheses. This depends on the placement of the prosthesis in the bone with sufficient stability [15]. In the cementless hip prosthesis design, a cavity in which bone cement can be introduced, and the placement of bone cement in this cavity causes sufficient primary stability of the prosthesis. Another important problem in cementless hip prostheses is the high risk of dislocation. This problem will also be significantly reduced because the bone and cement can be rapidly integrated.

Some assumptions that may be limited the results were used in this study. The load values used for FEA were taking into account during only regular walking. The other loading conditions may change the results. The other assumption is the design parameters of the prosthesis. The result may be affected when the design parameter was change. Finally, the material properties of the FEA models were defines as isotropic material properties. However, bone materials are usually anisotropic [21]. In spite of that, the bone material properties can be acceptable as anisotropic according to Peng et al. study [22].

## Conclusion

Based on the new design presented in this study, sufficient mechanical stability was achieved, the risk of cement cracking and the risk of dislocation of the prosthesis were reduced, and the risk of aseptic loosening was reduced.

## Compliance with ethical standards

**Conflict of interest** The authors declare that there is no conflict of interest regarding the publication of this paper.

## References

1. Abdulkarim A, Ellanti P, Motterlini N, Fahey T, O'Byrne JM (2013) Cemented versus uncemented fixation in total hip replacement: a systematic review and meta-analysis of randomized controlled trials. Orthop Rev (Pavia) 5(1):e8. https://doi.org/10.4081/or.2013.e8

2. Smith AJ, Dieppe P, Howard PW, Blom AW, Wales NJRE (2012) Failure rates of metal-on-metal hip resurfacings: analysis of data from the National Joint Registry for England and Wales. Lancet 380(9855):1759–1766

3. Makela KT, Eskelinen A, Pulkkinen P, Paavolainen P, Remes V (2008) Total hip arthroplasty for primary osteoarthritis in patients fifty-five years of age or older. J Bone Joint Surg Am 90(10):2160–2170

4. Hailer NP, Garellick G, Karrholm J (2010) Uncemented and cemented primary total hip arthroplasty in the Swedish hip arthroplasty register evaluation of 170,413 operations. Acta Orthop 81(1):34–41

5. Katz JN, Wright EA, Wright J, Malchau H, Mahomed NN, Stedman M, Baron JA, Losina E (2012) Twelve-year risk of revision after primary total hip replacement in the US medicare population. J Bone Joint Surg 94(20):1825–1832. https://doi.org/10.2106/jbjs.k.00569

6. Jeffers JRT, Browne M, Lennon AB, Prendergast PJ, Taylor M (2007) Cement mantle fatigue failure in total hip replacement: experimental and computational testing. J Biomech 40(7):1525–1533. https://doi.org/10.1016/j.jbiomech.2006.07.029

7. Ramaniraka NA, Rakotomanana LR, Leyvraz PF (2000) The fixation of the cemented femoral component—effects of stem stiffness, cement thickness and roughness of the cement-bone surface. J Bone Joint Surg Br 82(2):297–303

8. Reggiani B, Cristofolini L, Varini E, Viceconti M (2007) Predicting the subject-specific primary stability of cementless implants during pre-operative planning: preliminary validation of subject-specific finite-element models. J Biomech 40(11):2552–2558. https://doi.org/10.1016/j.jbiomech.2006.10.042

9. Rho JY, Hobatho MC, Ashman RB (1995) Relations of mechanical-properties to density and Ct numbers in human bone. Med Eng Phys 17(5):347–355

10. Senalp AZ, Kayabasi O, Kurtaran H (2007) Static, dynamic and fatigue behavior of newly designed stem shapes for hip prosthesis using finite element analysis. Mater Des 28(5):1577–1583

11. Kühn KD (2000) Bone cements. Springer, Berlin

12. Bergmann G, Deuretzbacher G, Heller M, Graichen F, Rohlmann A, Strauss J, Duda GN (2001) Hip contact forces and gait patterns from routine activities. J Biomech 34(7):859–871

13. Duda GN, Schneider E, Chao EYS (1997) Internal forces and moments in the femur during walking. J Biomech 30(9):933–941

14. Higa M, Tanino H, Nishimura I, Mitamura Y, Matsuno T, Ito H (2015) Three-dimensional shape optimization of a cemented hip stem and experimental validations. J Artif Organs 18(1):79–85

15. Celik T, Mutlu I, Ozkan A, Kisioglu Y (2017) The effect of cement on hip stem fixation: a biomechanical study. Australas Phys Eng Sci Med 40(2):349–357. https://doi.org/10.1007/s13246-017-0539-1

16. Goshulak P, Samiezadeh S, Aziz MSR, Bougherara H, Zdero R, Schemitsch EH (2016) The biomechanical effect of anteversion and modular neck offset on stress shielding for short-stem versus conventional long-stem hip implants. Med Eng Phys 38(3):232–240

17. Ridzwan MIZ, Shuib S, Hassan AY, Shokri AA, Ibrahim MNM (2007) Problem of stress shielding and improvement to the hip implant designs: a review. J Med Sci (Faisalabad) 7(3):460–467. https://doi.org/10.3923/jms.2007.460.467

18. Kim JT, Yoo JJ (2016) Implant design in cementless hip arthroplasty. Hip Pelvis 28(2):65. https://doi.org/10.5371/hp.2016.28.2.65

19. Abdelaal O, Darwish S, El-Hofy H, Saito Y (2018) Patient-specific design process and evaluation of a hip prosthesis femoral stem. Int J Artif Organs 42(6):271–290. https://doi.org/10.1177/0391398818815479

20. Falkenberg A, Drummen P, Morlock MM, Huber G (2019) Determination of local micromotion at the stem-neck taper junction of a bi-modular total hip prosthesis design. Med Eng Phys 65:31–38. https://doi.org/10.1016/j.medengphy.2019.01.003

21. Wirtz DC, Schiffers N, Pandorf T, Radermacher K, Weichert D, Forst R (2000) Critical evaluation of known bone material properties to realize anisotropic FE-simulation of the proximal femur. J Biomech 33(10):1325–1330

22. Peng L, Bai J, Zeng XL, Zhou YX (2006) Comparison of isotropic and orthotropic material property assignments on femoral finite element models under two loading conditions. Med Eng Phys 28(3):227–233