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Design and Optimization of Sinusoidal Formed Femur Prosthesis

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Keywords

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Abstract: One of the major problems in hip replacement surgery is the hip replacement loosening. Hip replacement loosening occurs over time after the surgery and it is related to the discretization between the bone cement and prosthesis. The underlying factors of this situation are the stress occurring in the bone cement and the shape of the prosthesis. In this study, cortical and trabecular layers of the femur, bone cement and prosthesis were modeled. The models of bone cement and prosthesis were constructed parametrically and two different sinusoidal formed prostheses were developed unlike the former prostheses shapes. Analyses were conducted for these two different sinusoidal forms by using finite element method and optimization was conducted to obtain the appropriate prosthesis stem shape and bone cement thickness by using parametric modeling in finite element analyses. For finite element analyses and optimization, Ansys Workbench software was used and analyses were conducted for 316LS stainless steel material. Finally, the optimum prosthesis stem shape and bone cement thickness was determined by using the results of the analyses in the first stage.

Sinüsoidal Formda Kalça Protezi Tasarımı ve Optimizasyonu

Anahtar Kelimeler

Femur
İmplant
Sap
Protez
Sinüzoidal Şekil
Sonlu Elemanlar Metodu

Özet: Kalça protezi cerrahisindeki en büyük problemlerden birisi kalça protezi gevşemesidir. Kalça protezi gevşemesi operasyon sonrasında zamanla meydana gelir ve kemik çimentosu ve protezin birbirinden ayrılması ile ilişkilidir. Bu durumun oluşmasının altında yatan sebepler kemik çimentosunda oluşan gerilme ve protezin şeklidir. Bu çalışmada, femura ait kortikal ve trabeküler kemik tabakaları, kemik çimentosu ve protez modellenmiştir. Kemik çimentosu ve protez parametrik olarak modellenmiş olup eski protezlerden farklı olarak iki farklı sinusoidal formda protez geliştirilmiştir. Sonlu elemanlar metodu ile bu iki farklı sinusoidal form için analizler yapılmıştır ve optimum protez sap şekli ve kemik çimentosu kalınlığını belirlemek amacıyla sonlu elemanlar analizlerinde parametrik modelleme yardımıyla optimizasyon yapılmıştır. Sonlu elemanlar analizleri ve optimizasyon için Ansys Workbench yazılımı kullanılmış olup analizler 316LS paslanmaz çelik malzemesi için gerçekleştirilmiştir. Son olarak, gerçekleştirilen analiz sonuçları ile optimum protez sap şekli ve kemik çimentosu kalınlığı belirlenmiştir.

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1. Introduction

In this century, biomaterials and arthroplasty have become quite important topics of interest with the development of technology. Although biomechanics looks like a new branch of science, its history dates back to ancient times. Basically, the idea of replacing a new part instead of a limb of the body is based on Egyptians. First application of hip replacement surgery was performed by an American surgeon in 1940 with a metallic prosthesis which was made of Cobalt-Chromium alloy Vitallium. Today, hip replacement applications are very common.

Total hip replacement is applied to the patients affected by hip diseases and implants are used to place in broken or cracked places or in joint renewal (Savilahti et al., 1997; Marston et al., 1996; Neumann et al., 1994; Delaunay and Kapandji, 1996). Hip replacement surgery is among the most common orthopedic procedures. While the procedure has become more successful and safer, it has not been perfected. One of the most common concerns of both patients and physicians is the problem of hip replacements becoming loose over time. Hip replacement loosening can cause important problems on the patient such as pain, loss of motion and sometimes with dislodgement of the prosthesis this situation can result in death.

Basically, there are two types of hip prosthesis which are commonly used on the patients: Hip prostheses with bone cement and cementless hip prostheses. Cementless hip prosthesis has porous structure on the surface and the bone grows into porous structure over time. Also during the surgery it is press-fit into the bone. Cemented hip prosthesis is fastened into the bone with bone cement. Either way, they are fit tightly into the bone of the femur and pelvis so that the implant cannot move. When implant loosening occurs, hip replacement begins to move small amounts at first, if it is not interfered, implant can dislodge.

There are many studies performed about hip prosthesis. Miljkovic et al. (1998) worked on an algorithm and a program for computer aided evaluation of total hip prosthesis stability in order to gain the most objective and sufficiently fast information which could greatly help the surgeon during the operation, before closing the wound. Main aim is to decide whether the prosthesis is stable or not.

In Şenalp et al.'s (2007) study four different stem shapes of varying curvatures for hip prosthesis were considered to avoid possible implant loosening. Static, dynamic and fatigue analysis for these designed stem shapes were performed. While static analyses were conducted under body load, dynamic analyses were performed under walking load.

In the study of Yang et al. (2010), loosening of model cemented joint replacement specimens were examined under cyclic loading and the fatigue strength of the bone/cement interface was characterized. Results of the evaluation suggest that the fatigue strength of the bone/cement interface in cemented total joint replacements can be estimated from simple quasi-static shear tests.

Kayabaşı (2004) designed a new prosthesis and investigated the behavior of his newly designed implant under the body weight load during stumbling by parametric modeling to avoid implant loosening. Two different implant materials were selected to obtain the appropriate material and fatigue life resistant. The probability of failure was investigated for both the initial and shape-optimized prosthesis designs by using several simple performance functions describing fatigue theory, static and dynamic failure of the cement–prosthesis interface. The results for new prosthesis were compared with Charnley's implant results.

In this paper the design of sinusoidal formed femur prosthesis was performed to avoid possible hip replacement loosening. As a first step parametric models for prosthesis were modeled different from the former prostheses, later prostheses in sinusoidal form were designed and optimum sinusoidal form was searched for. Verification analyses for each sinusoidal form were also executed in the final step (Canpolat, 2010).

By using finite element method stress variations in cortical bone, trabecular bone, bone cement, prosthesis and sinusoidal part of the prosthesis were inspected. For the finite element analyses and optimization, Ansys Workbench software was used and analyses were conducted for 316LS stainless steel material. By using parametric modeling in the finite element analysis, optimization was performed to determine the optimum prosthesis stem shape and bone cement thickness. Finally, the optimum prosthesis stem shape and bone cement thickness was determined by using the results of the analyses in the first stage.

2. Geometric and Finite Element Models of the Prosthesis

2.1. Geometric Model of the Prosthesis

Stem shape and the bone cement thickness are the most effective parameters on the performance of the prosthesis. By optimizing these parameters implant loosening can be avoided. Therefore, in this study sinusoidal waved stems were designed to achieve good bonding capability at the stem-bone cement interface. In this context, two different sinusoidal waved stems were modeled where one has two sinusoidal waves and the other has three sinusoidal waves. CAD models of the designed stems are shown in Fig. 1.

Figure 1 Prosthesis models (a) two waved (b) three waved.

The modeled parts of the whole assembly are prosthesis, bone cement, and femur where prosthesis and bone cement were modeled parametrically and femur was modeled in two parts (trabecular and cortical bone). Prosthesis model was composed of stem, neck and head (ball) parts. Stem part was designed parametrically in two different sinusoidal forms. Head diameter of the prosthesis is 22 mm. Neck of the prosthesis makes 120^{0} angle with the femur. The length of the stem is 85 mm and the diameter is 15 mm. For the femur geometry the standardized model of Viceconti et al. (1996) was used. To construct the finite element model for the analyses, CAD model of the complete assembly was imported into Ansys Workbench environment.

In Fig. 2, the parametric model of the stem and sinusoidal splines can be seen. The below equations represent the spline equations of the sinusoidal curved parts of the stem (The spline equations of the sinusoidal curved parts of the stem are given in Eqs. 2).

For two wave stem;

$$x_i = a + r \cdot \sin\theta_i,\ y_i = (h/6)\cdot i, \quad i = 0,\ldots,12$$
$$\theta_i : \text{with } \pi/6 \text{ increments in } 0,2\pi \text{ range} \tag{2.1}$$

For three wave stem;

$$x_i = a + r \cdot \sin\theta_i,\ y_i = (h/6)\cdot i, \quad i = 0,\ldots,18$$
$$\theta_i : \text{with } \pi/6 \text{ increments in } 0,3\pi \text{ range} \tag{2.2}$$

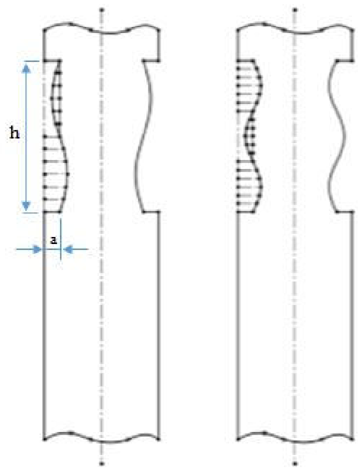

Figure 2 Prosthesis sketches (a) two waved (b) three waved.

The design parameters used in the analyses are *a*, *r*, *h* and *t*, where;
"*a*" is the parameter which specifies the starting depth of the sinusoidal form (from the outer surface of the stem) as shown in Fig. 2 (a),
"*r*" is the amplitude of the sinusoidal form,
"*h*" is the length of the sinusoidal form,
"*t*" is the thickness of the bone cement.

Table 1 Values of the parameters used in the analyses.

Parameters (mm)	First value	Second value	Third value	Fourth value	Fifth value
a	2	2.5	3	-	-
r	0.5	1	1.5	-	-
h	20	25	30	35	40
t	2.55	3	3.5	-	-

The values of the parameters are given in the Table 1. To avoid unnecessary complexity, stem head diameter, stem neck shape and diameter were taken constant. Totally 270 analyses were conducted for two and three waved forms (135 analyses for each type of stem) by using the combinations of the values presented in Table 1.

2.2. Finite element model of the prosthesis

The main idea of the finite element analysis is analyzing the system by discretizing the geometric models into smaller and simpler elements which are called meshes. Each analysis basically has six steps: Defining material, importing geometry data to the finite element analysis software, defining contacts, meshing, defining boundary conditions and loads and lastly solving the system. The three-dimensional solid model assembly of femur (cortical and trabecular bone), bone cement and prosthesis were transferred to ANSYS Workbench environment. ANSYS Workbench automatically recognizes the contacts existing between each part and establishes the contact conditions for corresponding contact surfaces (Ansys, 2010).

Following this step, each part of the assembly (cortical and trabecular bone, bone cement and prosthesis) was meshed in ANSYS Workbench environment to build the finite element model. The finite element models of the cortical and trabecular bone, bone cement and prosthesis are shown in Fig. 3.
Due to the complex geometry of the femur, tetrahedral element type (tetrahedrons in ANSYS Workbench) was used in cortical bone and trabecular bone.

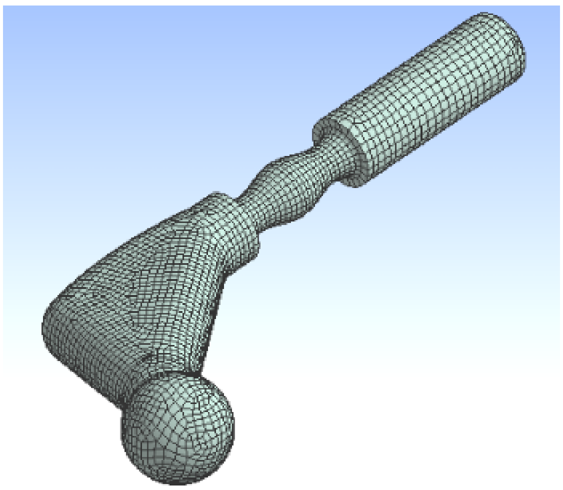

Figure 3a Finite element model of stem.

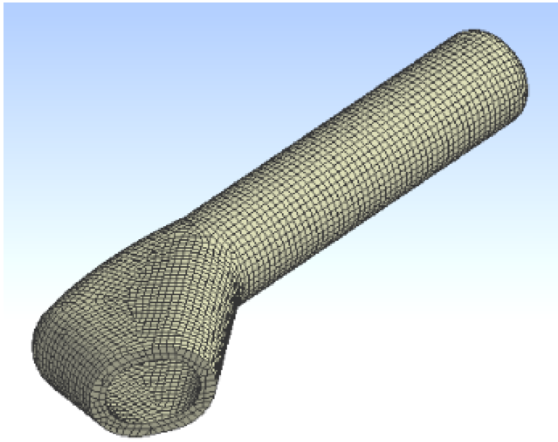

Figure 3b Finite element model of bone cement.

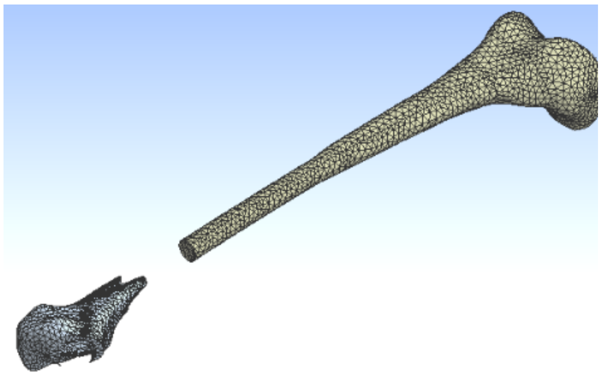

Figure 3c Finite element model of trabecular bone.

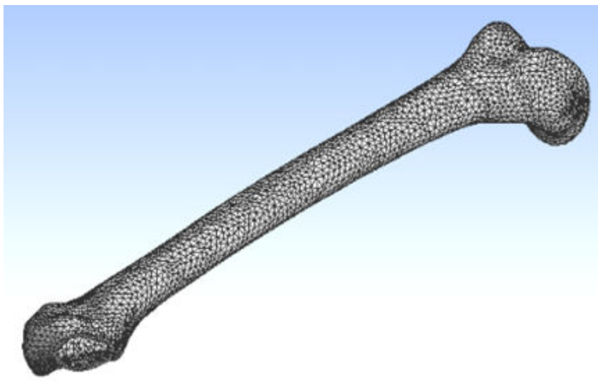

Figure 3d Finite element model of cortical bone.

Hexagonal element type (hex dominant in ANSYS Workbench) was used for prosthesis and bone cement. (Cortical bone, upper part of the trabecular bone, lower part of the trabecular bone, bone cement, prosthesis and the whole assembly were consisted of 32852, 6950, 12755, 19693, 28212 and 100462 elements, respectively.)

For contact surfaces between prosthesis-bone cement, bone cement-trabecular bone and trabecular bone-cortical bone, completely bonded contact type was chosen for the whole assembly.

Table 2 Mechanical properties of prosthesis, bone cement and trabecular bone materials

.Material	Elastic Modulus (MPa)	Poisson Ratio	Density (kg/mm^3)	Yield Stress (MPa)
316LS	$2x10^5$	0.3	$7.85x10^{-6}$	946
Bone cement	2700	0.35	$1.19x10^{-6}$	80
Trabecular Bone	2130	0.3	-	53

2.3. Material models of the prosthesis

In total hip arthroplasty (THA) stainless steel, Ti-6Al-4V and cobalt-chromium alloys can be used as prosthesis material and polymethyl methacrylate (PMMA) can be used as bone cement. Each of these materials is bio-compatible materials. In this study, 316LS stainless steel was used as the prosthesis material in the finite element simulations and polymethyl methacrylate (PMMA) material was used as the bone cement material.

Table 3 Mechanical properties of cortical bone material (Elastic modulus [E], Poissons ratio [v], Shear modulus [G]).

Mechanical Properties of Cortical Bone	
E_x	15800 MPa
E_y	15800 MPa
E_z	22000 MPa
v_{xy}	0.302
v_{yz}	0.109
v_{xz}	0.109

Inner and outer sides of the bone (trabecular bone and cortical bone) were modeled with different material properties for more accurate results. Here, cortical bone shows an orthotropic material behavior and the rest of the parts of the assembly show linear isotropic material behavior.

Mechanical properties of 316LS material, bone cement and trabecular bone material (Nuño and Avanzolini, 2002) are given in the Table 2 and mechanical properties of the cortical bone material (Joshi et al., 2000) are given in the Table 3.

2.4. Loading conditions

Mainly, two types of forces act on the prosthesis and femur: Contact force and muscle force. Contact force are due to human body-weight acts on the head part of the prosthesis and muscle forces act on the femur during walking. In this study, only statical analysis was conducted therefore muscle force values were the maximum values occurred during walking. These muscles consist of the abductor muscle, vastus lateralis muscle and tensor fasciae latae muscle. The contact and muscle force values (Huiskes and Verdonscot, 1997; Bergmann et al., 1993) are shown given in the Table 4. The application of forces is shown in the Fig. 4.

Table 4 Contact and muscle forces.

Forces	F_x *(N)*	F_y *(N)*	F_z *(N)*
Contact Force	338	-208	1462
Abductor Muscle	-484.8	35.948	-723.14
Tensor Fasciae Latae Muscle Distal	4.18	-5.85	158.84
Tensor Fasciae Latae Muscle Proximal	-60.19	96.97	-110.35
Vastus Lateralis Muscle	7.524	154.66	776.64

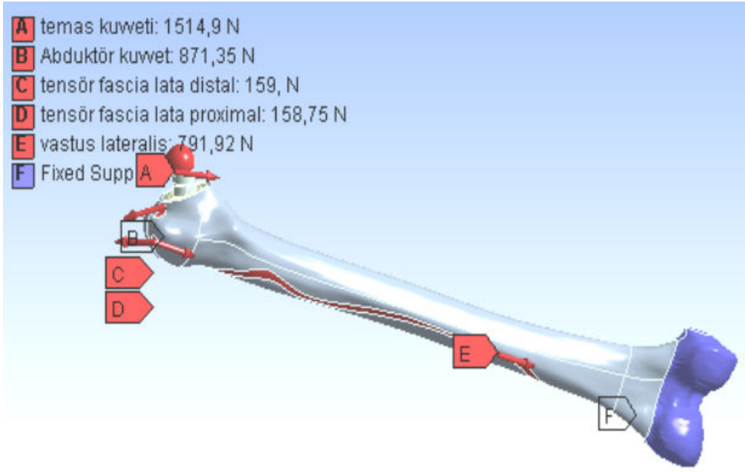

Figure 4 Surfaces on which contact and muscle forces act on.

3. Finite Element Analysis and Optimization

In this study, finite element analyses of 270 different CAD models were performed according to the parameters which were indicated in the Table 1. For each prosthesis type represented as two waved and three waved, there are 135 CAD model combinations and 135 analyses were performed for each combination set. With respect to von Mises Criteria minimum selected values of maximum stresses occurred in cortical and trabecular bone, bone cement, prosthesis and sinusoidal part of stem for two waved prosthesis are shown in Fig. 5, Fig. 6, Fig. 7 and Fig. 8. For three waved prosthesis these values are shown in Fig. 9, Fig. 10, Fig. 11, Fig. 12 and the corresponding stress values are also given in Table 5.

As seen in Fig. 5, Fig. 6, Fig. 7 and Fig. 8, the maximum von-Mises stresses occurred in cortical bone, trabecular bone, bone cement and stem are 31.087 MPa, 18.276 MPa, 13.277 MPa and 96.079 MPa, respectively for two waved prosthesis. In Fig. 9, Fig. 10, Fig. 11 and Fig. 12, the maximum von-Mises stresses occurred in cortical bone, trabecular bone, bone cement and stem are 31.094 MPa, 18.365 MPa, 14.361 MPa and 94.775 MPa, respectively for three waved prosthesis.

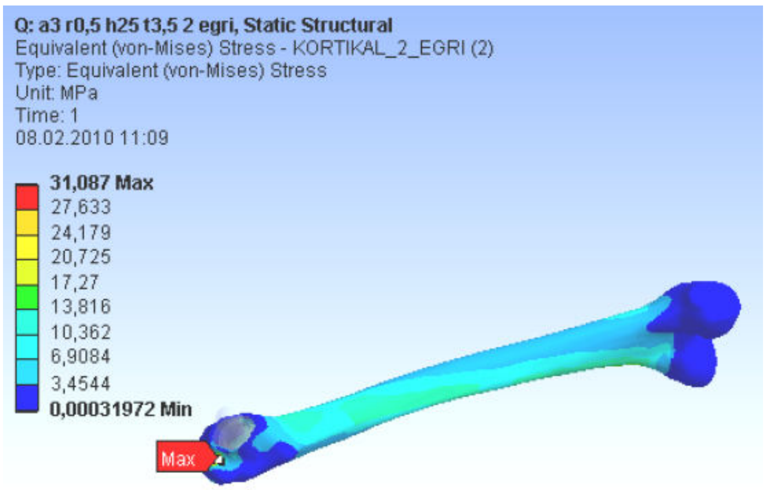

Figure 5 Analysis results for two waved prosthesis (cortical bone).

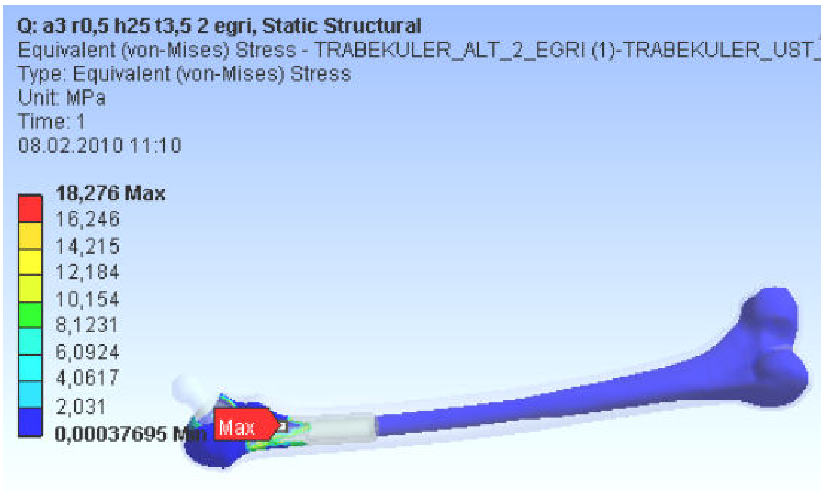

Figure 6 Analysis results for two waved prosthesis (trabecular bone).

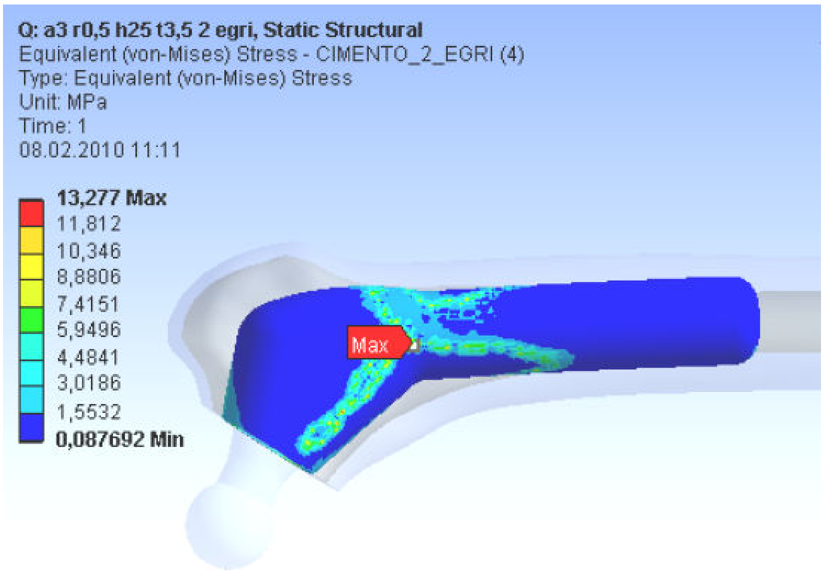

Figure 7 Analysis results for two waved prosthesis (bone cement).

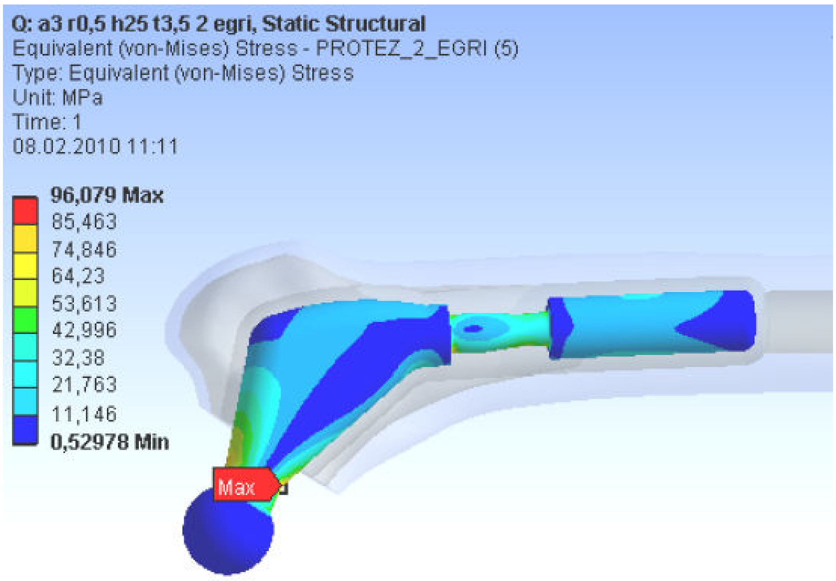

Figure 8 Analysis results for two waved prosthesis (prosthesis).

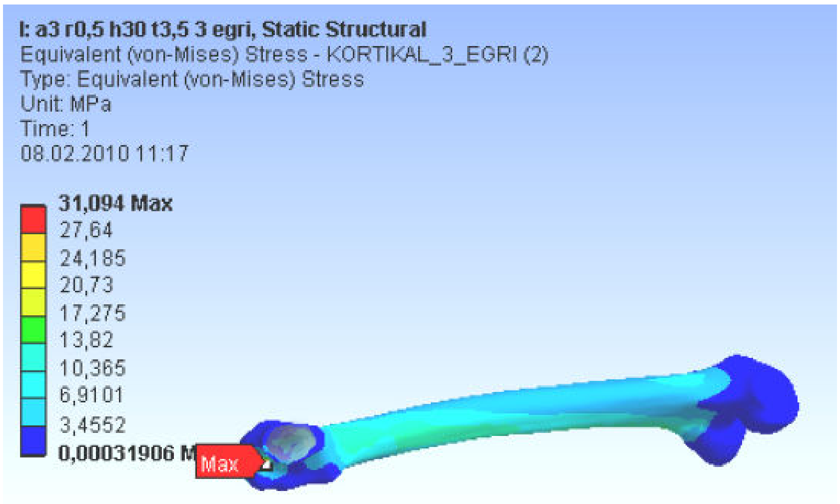

Figure 9 Analysis results for three waved prosthesis (cortical bone).

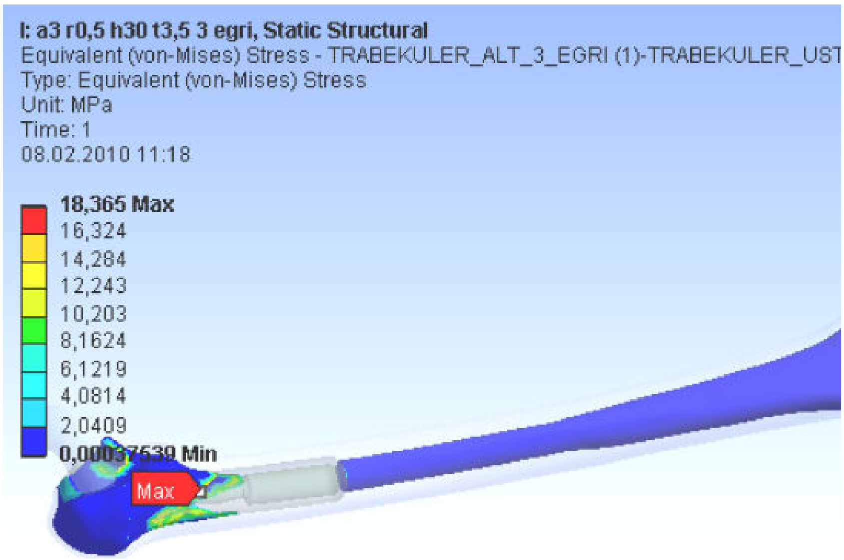

Figure 10 Analysis results for three waved prosthesis (trabecular bone).

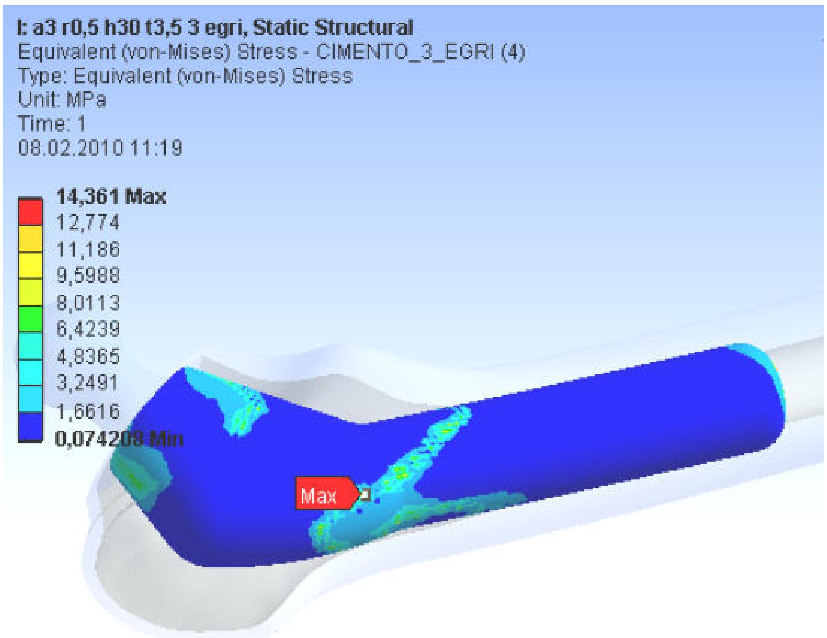

Figure 11 Analysis results for three waved prosthesis (bone cement).

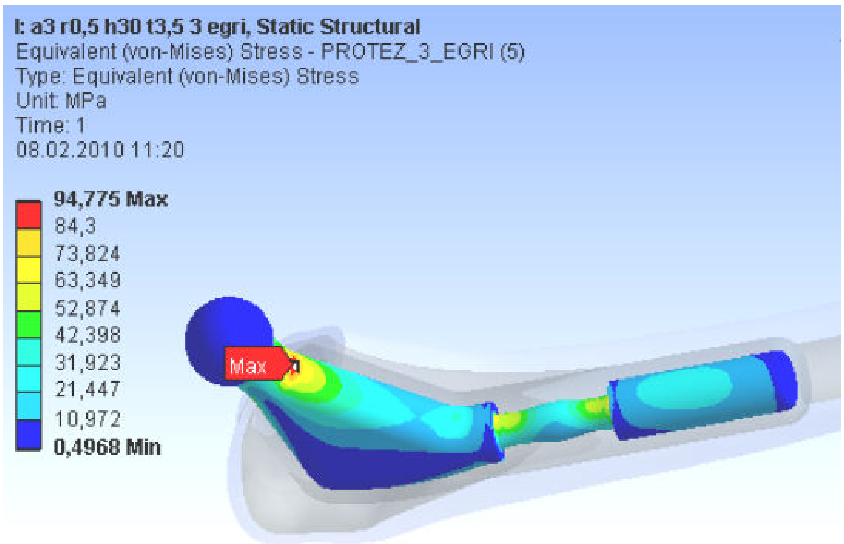

Figure 12 Analysis results for three waved prosthesis (prosthesis).

Mostly hip replacement loosening starts with separation between bone cement and stem interface over time. Therefore, the maximum von Mises stress occurring in bone cement becomes very important in this context. As seen from the results, the minimum stress in bone cement occurred in two waved prosthesis for the parameters a=3 mm, r=0.5 mm, h=25 mm, t=3.5 mm and the minimum stress in bone cement occurred in three waved prosthesis for the parameters a=3 mm, r=0.5 mm, h=30 mm, t=3.5 mm. These stress values are the minimum selected values of maximum von Mises stresses for each stem type.

4. Results and Discussion

From the analyses performed, it is prevailed that bone cement thickness is an important parameter on the equivalent von Misses stress occurring in the bone cement. The stress in bone cement reduces when bone cement thickness increases. The minimum von Misses equivalent stress value in the bone cement was obtained when bone cement thickness value is taken 3.5 mm. The optimum prosthesis shape was determined as the two waved sinusoidal type which has parameters a=3, r=0.5, h=25 and t=3.5 mm from the optimization.

Table 5 Minimum selected values of maximum von Mises stresses occurred in cortical and trabecular bone, bone cement, prosthesis and sinusoidal part of stem obtained from optimization analysis.

Stem type	Cortical bone (MPa)	Trabecular bone (MPa)	Bone cement (MPa)	Stem (MPa)	Sinusoidal part (MPa)
Two waved a=3, r=0.5, h=25, t=3.5	31.087	18.276	13.277	96.079	73.253
Three waved a=3, r=0.5, h=30, t=3.5	31.094	18.365	14.361	94.775	72.387

In Fig. 13, four plots are given to show how the equivalent stresses occurred in the cortical and trabecular bone, bone cement and prosthesis change with different amplitudes of the sinusoidal form (r). In these graphics a, h and t parameters were taken constant as a=2, h=20 and t=3.5 mm. It is clear from Fig. 13 (a) that, with increasing r values, the equivalent stress arised in cortical bone increases for two waved prosthesis and decreases for three waved prosthesis. In Fig. 13 (b), with increasing r values, the equivalent stress arised in trabecular bone has a peak point near r=1 for two waved prosthesis and stays nearly constant for three waved prosthesis. In Fig. 13 (c), with increasing r values, the equivalent stress arised in bone cement shows irregular changes for two waved prosthesis and it stays nearly constant for three waved prosthesis. Finally, in Fig. 13 (d), with increasing r values, the equivalent stress arised in prosthesis increases after the value of r=1.2 for two waved prosthesis and increases after the value of r=1 for three waved prosthesis.

a

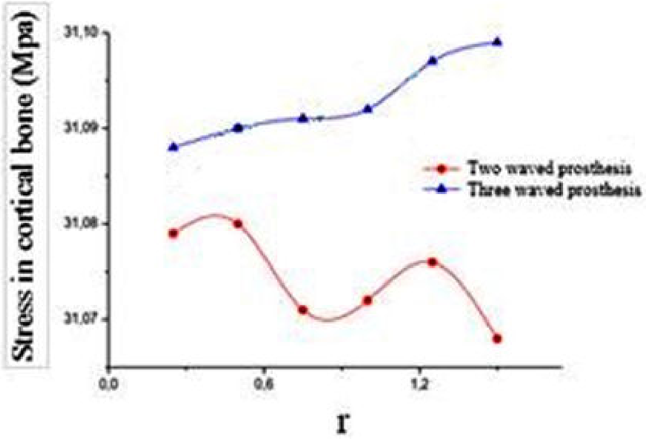

Figure 13a The effect of the amplitude of the sinusoidal form (r) on von-Mises stresses occurred in cortical bone.

b

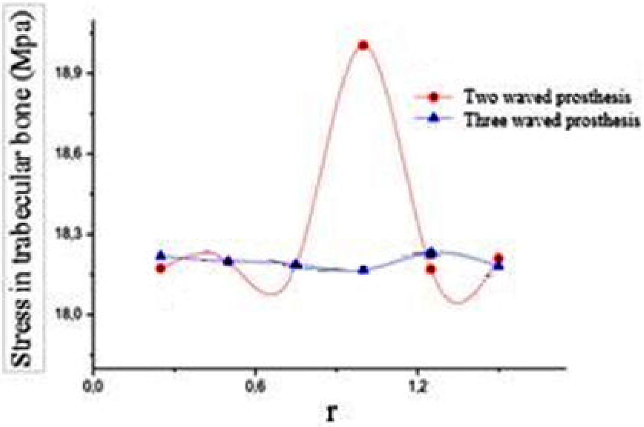

Figure 13b The effect of the amplitude of the sinusoidal form (r) on von-Mises stresses occurred in trabecular bone.

c

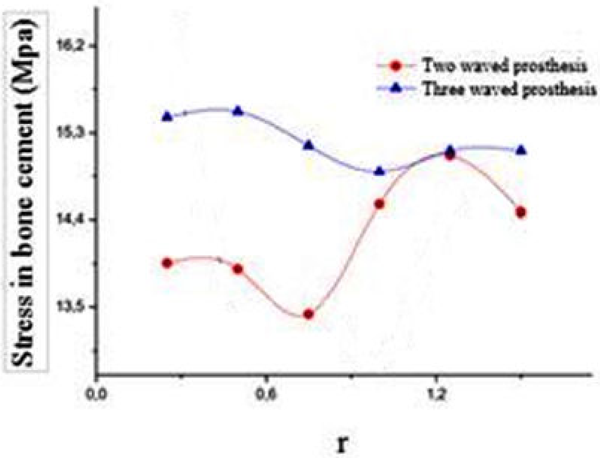

Figure 13c The effect of the amplitude of the sinusoidal form (r) on von-Mises stresses occurred in bone cement.

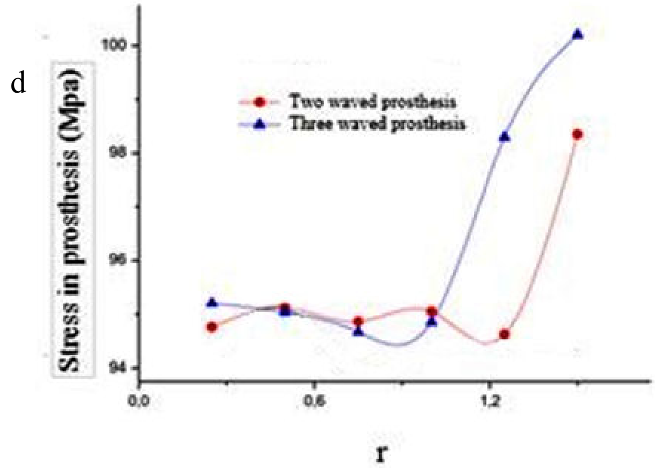

Figure 13d The effect of the amplitude of the sinusoidal form (*r*) on von-Mises stresses occurred in prosthesis.

5. Conclusion

In this study, unlike the former prostheses, sinusoidal formed prostheses were modeled and finite element analyses were conducted for two and three waved sinusoidal forms to avoid the implant loosening. This study aims the determination of optimum prosthesis shape which causes minimum stress occurrence in the bone cement among these different prosthesis models. The prostheses and bone cement were modeled parametrically (with *a*, *r*, *h* and *t* parameters). The modeled prostheses have two and three waved sinusoidal form.

It is concluded that the cement thickness (*t*) is very effective on the equivalent von Mises stresses occurring in the bone cement from the analyses. While *t* increases the stress in bone cement decreases. In this context it can be deduced that, maximum allowable bone cement thickness should be chosen for high performance.

Aseptic loosening is related to the stress between the prosthesis and bone cement therefore the stress occurred in bone cement becomes very momentous. When the minimum equivalent stress occurred in the bone cement is taken into consideration, the optimum prosthesis shape will be the two waved sinusoidal type which has the parameters a=3, r=0.5, h=25 and t=3.5 mm because the minimum value of the maximum stresses occurred in bone cement was obtained with these parameters from 270 analyses.

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