PROBABILISTIC DESIGN OF A NEWLY DESIGNED CEMENTED HIP PROSTHESIS USING FINITE ELEMENT METHOD

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The finite element method is an important tool used in the design of orthopedic prosthesis. One of the important orthopedic applications is hip prosthesis replacement. This operation is so complex that it requires close co-operation between engineers and surgeons. They have to work together in order to produce durable and reliable hip joint prosthesis. The reason for this is that the nature of bone strongly affects the design.

In reality, uncertainties exist in the system and environment that may make the application of a deterministic design decision unreliable. That is, the values of the variables that are acting on the system cannot be predicted with certainty. For instance, probabilistic approach was applied to the model after deterministic design results. Thus, using probabilistic approach reliability of newly design cemented hip prosthesis was quantified.

The new design is modeled parametrically to investigate the effects of different geometrical parameters on the relative displacement. These parameters are then optimized. Using the results of this investigation, the probability of failure was investigated for both the initial and shape-optimized prosthesis designs using several simple performance functions describing fatigue theory (Goodman, Gerber, Soderberg), static and dynamic failure of the cement–prosthesis interface. The optimum geometry and material properties are then compared with Charnley's implant results.

Keywords: Probabilistic Design, Reliability, Fatigue, Finite Element Method, Hip Prosthesis

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1. Introduction

When a new design is desired to be checked for these criteria the conventional design and analysis methods are not functional because conventional design and analysis of bone-implant hip prosthesis highly rely on expert’s knowledge, experience and ability to avoid any irreversible damage on bones of patients. Due to the difficulties of implant tests on human body and vital importance, some mathematical models have been developed to further investigate structural analysis of implant before application on a patient. Bone-implant hip prosthesis could be designed and studied in computer environment before implementing on a patient. Finite Element Method (FEM) as one of the most advanced simulation technique has been used in orthopedic biomechanics for many decades. It is an important tool used in the design and analysis of total joint replacements and other orthopedic devices. Finite element modeling and analysis present a non-destructive design approach for bone-implant hip prosthesis. It allows many complex what-if scenarios to be studied in computer environment before the prosthesis is actually applied on the patient. This will save time for the design and prevent any permanent damage caused by mis-implementation of bone-implant hip prosthesis.

The orthopedic implants were studied in literature in many different ways. Each study considered its own topic or some of the topics. Since the success of the implant depends on the integrity of implant with the skeleton, the static bone – implant interface strength and the fatigue strength in whole life of patient, the study must explain all of these criteria.

The design of stem shape in total hip is very crucial to keep the long term attachment of the prosthesis to the bone. If the stem shape design leads to high stresses in fixation areas of the prosthesis, fracture in short term or fatigue failure in long term of the prosthesis quite likely occurs. One of the most important factors in the implant design is to reduce stress and prosthesis. The other factor is to minimize movement at the both bone-cement-stem interface and bone-cement femur interface.

Some investigators have postulated that the failure mode of the bone-cement or bone implant interface is a result of the shear stress distribution and shear strength of the interface. A fracture mechanics approach using strain energy density and stress intensity factors, combined with the assumption of an interface layer with gradual variation of physical properties, has also been used to describe interface failure modes. Such criteria can be used to define an objective function for the optimization problem [1]. References [2,3,4,5] used theoretical optimization models to find best overall external shape of the stem, achieving similar shapes. Reference [6] applied optimization methods to select the elastic modulus of the stem. References [7,8], used a parallel walled hollow stem while [9,10] lessened the stiffness of the stem by using transverse holes.

The long-term reliability of today's orthopedic implants is a major concern. As a result, much of today's research is aimed at improving the overall reliability (or some specific aspect of reliability) of orthopedic implants. During current analyses, the actual probability of failure is not computed per se; rather a relative measure of performance is inferred. Improvement of the long-term reliability of orthopedic implants by using a design optimization approach that integrates probabilistic analysis methods is also investigated [11].

In reality, uncertainties exist in the system and environment that may make the application of a deterministic design decision unreliable. That is, the values of the variables that are acting on the system cannot be predicted with certainty. For instance, probabilistic approach was applied to the model after deterministic design results. Thus, using probabilistic approach reliability of newly design cemented hip prosthesis was quantified.

In this study, an efficient, effective and automated design strategy is proposed to design a hip prosthesis. In this strategy, Finite Element Analysis, Approximate model, a numerical optimization algorithm and probabilistic design method Monte Carlo Simulation are integrated to create an automated design tool. Using this approach, shape design of the prosthesis is formulated in the form of an optimization problem that can be solved easily by a conventional numerical optimization algorithm. Computationally expensive objective and constraint functions in the optimization problem, which come from prosthesis dynamic Finite Element analysis results, are replaced with their approximations before the optimization problem is solved. Solution of the optimization problem leads to the optimum design. Using the results of this investigation, the probability of failure was investigated for both the initial and shape-optimized prosthesis designs using several simple performance functions describing fatigue theory (Goodman, Gerber, Soderberg), static and dynamic failure of the cement–prosthesis interface. The reliability of the hip prosthesis is predicted using NESSUS linked with the ANSYS finite element analysis program. Performance of the optimum prosthesis is compared with that of the commonly used Charnley type design.

2. Methods

2.1 Probabilistic Methods and Monte Carlo Simulation

The probabilistic response of the cemented hip prosthesis system was modeled as

$$Z(X)=Z(X_1,X_2,X_3,...,X_n), \qquad (1)$$

Z(X) is a random variable describing the response of the system (e.g. stress, strain) at a node or element), X_i (i=1,n) are random variables describing the model input variables. In this study, Z is obtained finite element model. For the probabilistic response Monte Carlo sampling techniques was used. The Monte Carlo method provides approximate solutions to a variety of mathematical problems by performing statistical sampling experiments on a computer.

In traditional engineering calculations mean value of each random variable is used. However random variables are not constant and change in a range. If the ranges of the results are important for the design, some other values should be used more than mean value such as standard deviation. In standard calculations, the range factors in the results are eliminated by using safety factors. In probabilistic design, probability distributions of the design parameters are calculated. Probability distribution results show the reliability change of the design. Designer uses the reliability values according to customer desires. This process leads to maximum safety and quality with minimum cost. The distributions of design parameters or independent random variables show a type. Most common used distribution types are normal, lognormal, uniform, exponential, triangular, beta, gamma, and weibull.

In a probabilistic analysis, the probability of failure can be quantified by formulating a performance function, which is a function of random variables, by comparing the probabilistic response of the structure to the probabilistic strength of the material [11].

Risk of failure was calculated based on a performance function of the type

$$g(\boldsymbol{X})=R(\boldsymbol{X})-S(\boldsymbol{X}) \tag{2}$$

Where $R(\boldsymbol{X})$ is a random function describing the "resistance" or strength of the component or constituent, $S(\boldsymbol{X})$ is the response of the structure (e.g. stress), also a random variable, and $\boldsymbol{X}$ is the vector of random variables. $g(X)$ represents a failure event.

The probability of failure defined as

$$p_f=P(g(X)\leq 0) \tag{3}$$

Performance functions were formulated for stem shape, bone-cement and femur. Eleven performance functions are investigated

$g_1=R_{UCS}-S\sigma_{cement}(X)$ Compressive failure (4)
$g_2=R_\tau-S_{Trcement}(X)$ Shear failure (5)
$g_3=R_{FL}-S\sigma_{cement}(X)$ Fatigue failure (6)
$g_4=R_{UCS}-S\sigma_{prosthesis}(X)$ Compressive failure (7)
$g_5=R_\tau-S_{Trprosthesis}(X)$ Shear failure (8)
$g_6=R_{FL}-S\sigma_{prosthesis}(X)$ Fatigue failure (9)
$g_4=R_{UCS}-S\sigma_{femur}(X)$ Compressive failure (10)
$g_5=R_\tau-S_{Trfemur}(X)$ Shear failure (11)
$g_6=R_{FL}-S\sigma_{femur}(X)$ Fatigue failure (12)
$g_{10}=R_T-S\sigma_{cement\text{-}stem\ interface}(X)$ Tensile failure (13)
$g_{11}=R_\tau-S\sigma_{cement\text{-}stem\ interface}(X)$ Shear failure (14)

R_{UCS}, R_τ, R_{FL} are the ultimate compressive strength, shear strength, and fatigue limit strength of the bone cement and prosthesis and femur respectively. S(X) is the computed response measure determined from a three dimensional finite element model. The probabilistic response and probability of failure were quantified for both the initial and the deterministically optimized prosthesis design. The random variable statistics are defined in the random variable definition window in NESSUS [12]. The numerical model definition allows the use of finite element solver ANSYS, or a user-defined model. The regression model definition allows input of function coefficients or data sets that can be fit using linear regression to standard linear or quadratic functions. [12]

Flow chart of Monte Carlo Method which is shown in Figure 1 is one of the techniques that edit the values of the parameters at each loop according to their probability of occurring. In Monte Carlo Method, uniform distributions of random numbers are generated. These values are transformed to numbers by using one of the parameter's cumulative distribution functions. Number transforming process is done according to chosen parameter's probability density function. To reach a similar distribution type, large numbers of samples are required.

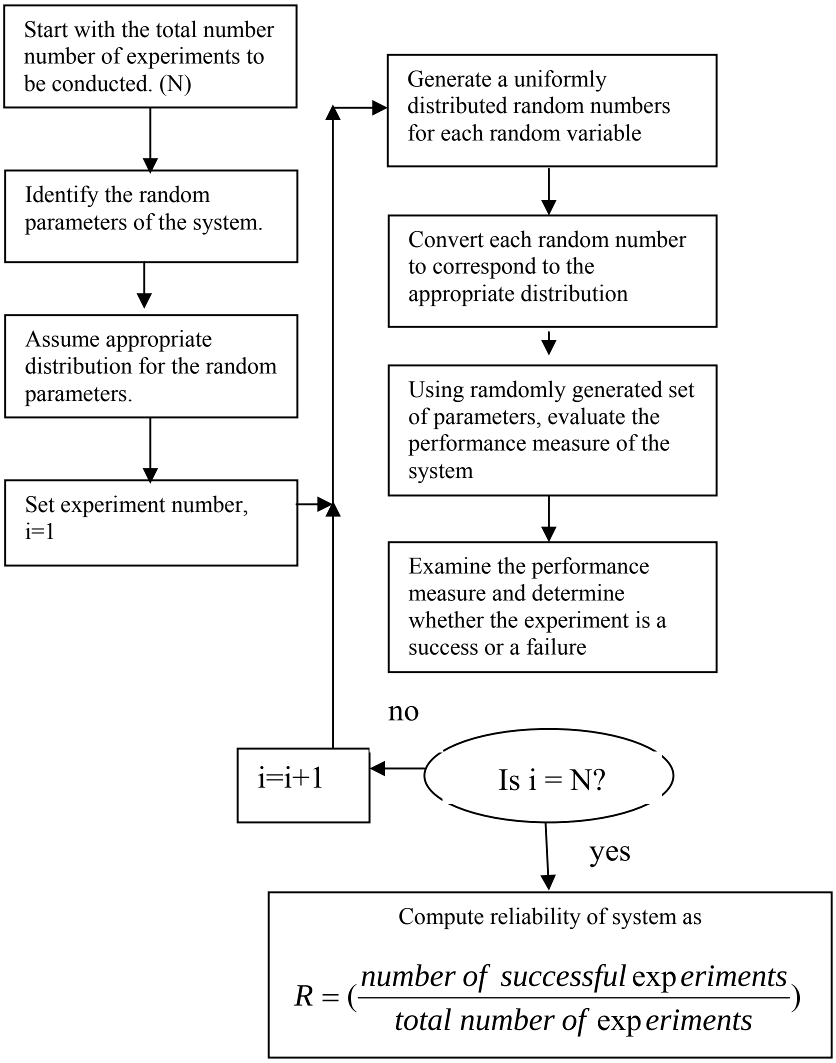


Fig. 1. Algorithm of Monte Carlo Simulation

2.2 Parametric Modeling

Search process for best (optimum) prosthesis geometry or shape can be automated through a numerical optimization algorithm if the goal and the requirements are expressed in the form of an optimization problem definition as following [13]. Parametric model is shown in Figure 1.

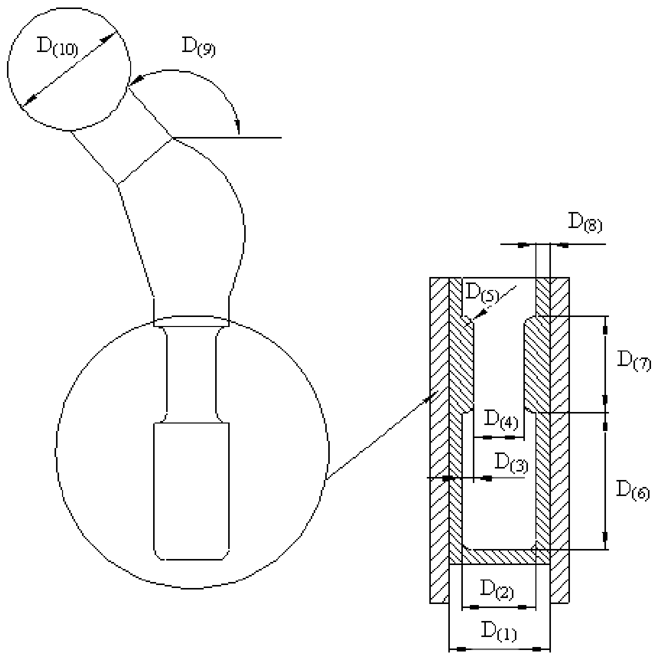


Fig. 2. Geometry of the hip prosthesis and design parameters selected.

Find design *parameters*:

$$D_{(1)}, D_{(2)}, D_{(3)}, D_{(4)}, D_{(5)}, D_{(6)}, D_{(7)}, D_{(8)}, D_{(9)} \text{ and } D_{(10)} \quad (15)$$

to minimize *design objective*:

Maximum Stress (von Mises stress) (16)

subjected to *design constraints*:

$$\sigma_{stemshape} \le \sigma_{stemshape}^{Yield} \quad (17)$$

$$\sigma_{cement} \le \sigma_{cement}^{Yield} \quad (18)$$

$$\sigma_{femur} \le \sigma_{femur}^{Yield} \quad (19)$$

$$\tau_{cement-femur} \le \tau_{cement-femur}^{failure} \quad (20)$$

$$\tau_{cement-stemshape} \le \tau_{cement-stemshape}^{failure} \quad (21)$$

$$N_{cement} \geq 1 \tag{22}$$

within the *design space*:

$$18\text{ mm} \leq D_{(1)} \leq 20\text{ mm} \tag{23}$$
$$10\text{ mm} \leq D_{(2)} \leq 16\text{ mm} \tag{24}$$
$$0\text{ mm} \leq D_{(3)} \leq 3\text{ mm} \tag{25}$$
$$10\text{ mm} \leq D_{(4)} \leq 16\text{ mm} \tag{26}$$
$$0\text{ mm} \leq D_{(5)} \leq 3\text{ mm} \tag{27}$$
$$40\text{ mm} \leq D_{(6)} \leq 65\text{ mm} \tag{28}$$
$$20\text{ mm} \leq D_{(7)} \leq 40\text{ mm} \tag{29}$$
$$1\text{ mm} \leq D_{(8)} \leq 4\text{ mm} \tag{30}$$
$$120^{o} \leq D_{(9)} \leq 150^{o} \tag{31}$$
$$32\text{ mm} \leq D_{(10)} \leq 38\text{ mm} \tag{32}$$

Equation 15 indicates design parameters describing the geometry/shape of the prosthesis. Equation 16 represents design objective which is the minimization of the maximum stress on the whole prosthesis. Equations 17-22 correspond to design constraints on structural and fatigue strength limits on components of the prosthesis (i.e. stem shape, femur, cement, cement-femur interface, cement-stem shape interface). N_{cement} in Equation 22 indicates fatigue safety factor for the bone cement. Fatigue safety factor is calculated using Goodman theory. Only fatigue safety of the bone cement is considered in constraint definitions since cement is much more suceptible to fatigue failure than the prosthesis. Prosthesis geometry is generally assumed not to fail by fatigue. Equations 23-32 represent design space where parameters are to be searched. Limits of the design space are determined based on the experience and literature.

Objective and constraint functions in Equations 17-22 are not explicitly (analytically) known before the solution of the optimization problem. They are created applying the least-square fitting to Finite element analysis results for certain number of randomly selected design parameter values that correspond to different prosthesis shapes. Since approximations of the objective and constraint functions are utilized, the optimization method is sometimes referred to as approximate optimization method and approximations are called as response surface approximations or response surface models. Linear and quadratic polynomial functions are often used to create the response surface approximations of the objective and constraint functions. In this study quadratic polynomial response surface functions, $\tilde{y}(\mathbf{x})$, as given by the following equation are employed:

$$\tilde{y}(\mathbf{x}) = \underbrace{\underbrace{\underbrace{a_0 + \sum_{i=1}^{n} a_i x_i}_{linear} + \sum_{i=1}^{n} b_i x_i^2}_{quadratic} + \sum_{j=1}^{n-1} \sum_{i=j+1}^{n} c_{ji} x_j x_i}_{quadratic+crossterms} \tag{33}$$

where a, b, c are tuning coefficients to be determined and n is the number of design parameters.

Approximate optimization method is applied to the optimization of the prosthesis through Design Optimization (DO) module of ANSYS [14]. Flow chart of the approximate optimization method implemented in this paper is shown in Figure 3. In design optimization process, ANSYS DO first creates n+2 design sets to construct a linear approximation. Here set indicates values of design parameters for a specific prosthesis geometry. ANSYS DO will either generate design sets randomly within the design space or use the existing ones in the optimization database. Shape optimization analysis is carried out at available design sets. Analysis results are then used to create linear approximations of objective and constraints. Higher order approximations such as quadratic and quadratic with cross terms RS approximations are created using least square method when there are enough design sets in the database. The optimum design is predicted by solving Equations 15-32 with a numerical optimization algorithm based on penalty functions. The predicted optimum is verified by Finite Element analysis with ANSYS. If the predicted objective and constraints are identical with the results from ANSYS, or the estimated optimum design is satisfactory enough, the optimization loop is stopped. Otherwise, the newly calculated results are added to the existing design sets and new approximations are created followed by the solution of the optimization problem.

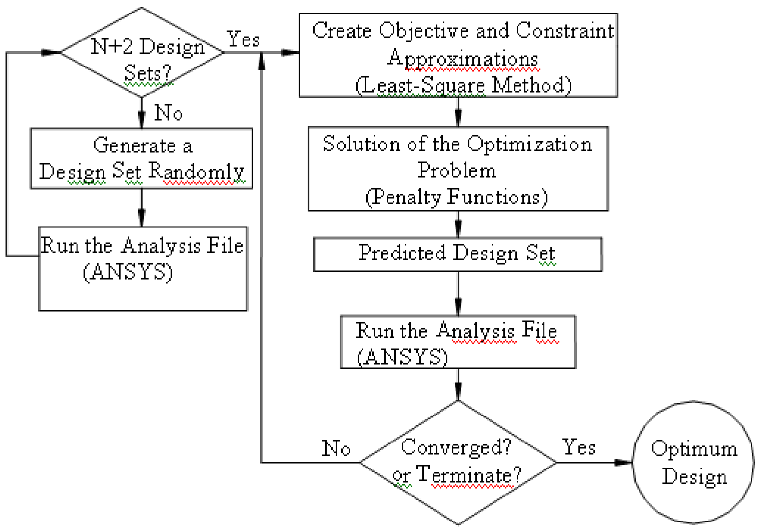
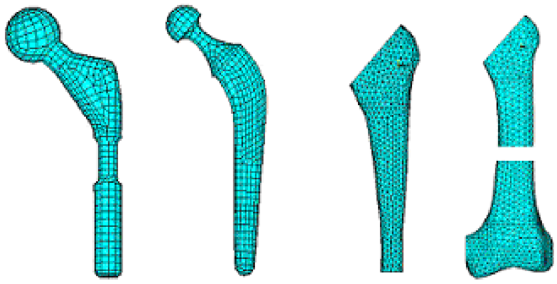


Fig. 3 Flow chart of the approximate optimization process with ANSYS DO module.

2.3 Finite Element Method

Finite element model is obtained by discritizing the geometric (CAD) model of the prosthesis with simpler and smaller elements (i.e. finite elements) using a meshing procedure. In this work, femur (cancellous bone, cortical shell), bone-cement and prosthesis are meshed using four node tetrahedral elements. The complete model consisted of 72,458 elements. The femur consists of 48,205 elements, the bone-cement consists of 13,760 elements and the prosthesis consists of 10,493 elements [15]. The femur, the bone cement and the prosthesis are modeled with the element type of SOLID45 in ANSYS. SOLID45 elements have four nodes each having three degrees of freedom. The finite element model of the femur, the bone cement and the prosthesis are shown Figure 4.



(a) (b) (c) (d)

Fig. 4 Finite element model of (a) new design, (b) Charnley design, (c) Bone cement and (d) femur.

The physical interactions among joints, which are involved in the prosthesis, are simulated by using contact algorithms. Surface to surface contact algorithms of ANSYS are used for the femur-bone cement interface and the bone cement- prosthesis interface. Three different the bone-cement and the implant interface conditions were used [15].

1-Completely bonded contact type
2-Debonded contact type with the coefficient of friction of μ=0.22
3-Debonded contact type with the coefficient of friction of μ=0

Two different material models for prosthesis are used in finite element analyses. These materials are titanium alloy (Ti-6Al-4V) and cobalt-chromium alloy. Behaviors of each of these materials are represented with linear isotropic material model. To investigate the effect of the bone behaviour on the prosthesis accurately, inner and outer sides of the bone (cortical bone and cancellous bone) were modelled with different material properties. Inner side of the bone (cortical bone) is represented with transversely isotropic material model. Outer side of the bone (cancellous bone) is modelled with linear isotropic material model. Cement (Polymethyl methacrylate) is modelled with linear isotropic material model. Random variable descriptions of model input variables were determined from the literature and actual experimental test data (Table 1) [11].

Table 1 Random variable for finite element model

Variable	Mean	S.D	Coefficent of variation (%)	Distrubition Type
Cortical bone Young's Modulus (GPa)	20.3	2.3	11.3	Lognormal
Cancellous bone Young's Modulus (GPa)	2.13	0.12	10	Lognormal
Bone Cement Young's Modulus (GPa)	2.7	0.2	8.2	Lognormal
Ti-6Al-4V Young's Modulus (GPa)	110	3.5	12	Lognormal
Cobalt-chromium alloy Young's Modulus (GPa)	207	2.75	10.75	Lognormal
Cobalt-chromium alloy Young's Modulus (GPa)	207	2.75	10.75	Lognormal
Bone Cement Shear Strength (MPa)	30	2.7	9	Normal
Bone Cement Compression Strength (MPa)	81.4	2.14	2.6	Normal
Bone Cement Fatigue Limit (MPa)	7.98	4.47	41.3	Lognormal
Prosthesis (Ti-6Al-4V) Shear Strength (MPa)	650	2.45	12.3	Normal
Prosthesis (Ti-6Al-4V) Compression Strength (MPa)	850	0.19	11.8	Normal
Prostesis (Ti-6Al-4V) Fatigue Limit (MPa)	350	0.29	8.9	Lognormal
Prosthesis (cobalt-chromium) Shear Strength (MPa)	550	2.98	9.3	Normal
Prosthesis (cobalt-chromium) Compression Strength (MPa)	750	0.26	13	Normal
Prostesis (cobalt-chromium) Fatigue Limit (MPa)	250	0.31	9.2	Lognormal
Femur Shear Strength (MPa)	20	2.46	10.75	Normal
Femur Compression Strength (MPa)	45	0.21	9.98	Normal
Femur Fatigue Limit (MPa)	15	0.26	9.2	Lognormal

Viscoelastic behavior of bone cement is modeled with a time hardening implicit creep equation [16,17]

$$\acute{\varepsilon}_{cr} = 5.168 \times 10^{-6} \sigma^{1.858} t^{-0.717} \qquad (34)$$

where $\acute{\varepsilon}_{cr}$ is the creep strain rate, σ is equivalent von Mises stress and t is time.

Fatigue life of the bone cement and prosthesis are calculated using ANSYS/Workbench [14]. S-N curve shown in Figure 5 and Figure 6 are employed in fatigue life calculation [15].

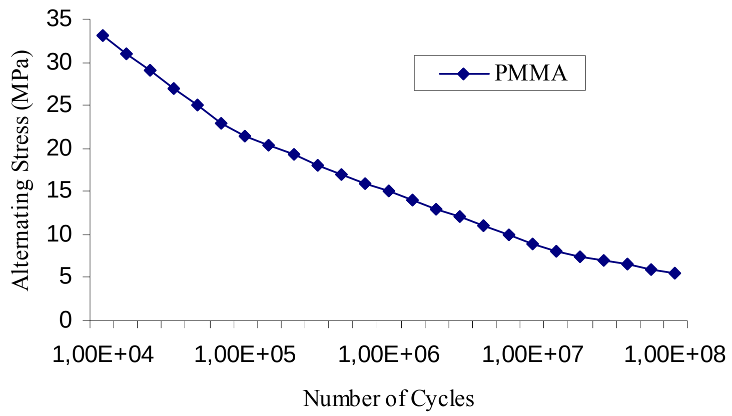


Fig. 5 Fatigue curve (S-N curve) for bone cement material (PMMA).

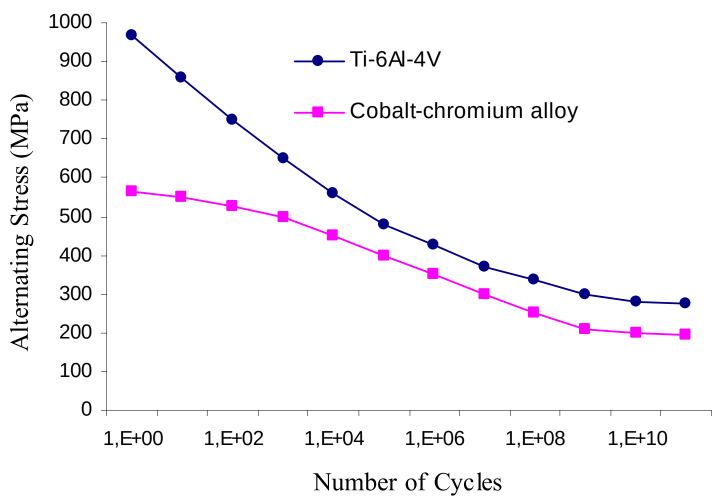


Fig. 6 Fatigue curves (S-N Curve) for Ti-6Al-4V and cobalt-chromium alloy materials.

Goodman Gerber and Soderberg theories are used in this study in order to take into account the mean stress effect on fatigue life. Goodman theory is expressed as

$$(\frac{\sigma_a}{S_e}) + (\frac{\sigma_m}{S_u}) = \frac{1}{N_s} \qquad (35)$$

Gerber theory is expressed as

$$(\frac{N\sigma_a}{S_e}) + (\frac{N\sigma_m}{S_u})^2 = 1 \qquad (36)$$

Soderberg theory is expressed as

$$(\frac{\sigma_a}{S_e}) + (\frac{\sigma_m}{S_y}) = \frac{1}{N} \qquad (37)$$

where N_s indicates fatigue safety factor, S_e for endurance limit and S_u for ultimate tensile strength of the material. Mean stress σ_m and alternating stress σ_a are defined respectively as:

$$\sigma_m = \frac{(\sigma_{max} + \sigma_{min})}{2} \qquad (38)$$

$$\sigma_a = \frac{(\sigma_{max} - \sigma_{min})}{2} \qquad (39)$$

Fatigue safety factor using Goodman theory is calculated based on the infinite fatigue life of 10^9 cycles in this study. Alternating and mean stresses are based on Von Mises stresses.

The third step is to apply loading and boundary condition. Random variable descriptions of model input variables were determined from the literature and actual experimental test data (Table 2) [11].

Table 2 Random variable for finite element model

Variable	Mean	S.D	Coefficent of variation (%)	Distrubition Type
Joint Load X direction (N)	1492	237.23	15.9	Lognormal
Joint Load Y direction (N)	915	408.09	44.6	Lognormal
Joint Load Z direction (N)	2925	731.35	25.0	Lognormal
Muscle Load X direction (N)	1342	335.50	25.0	Lognormal
Muscle Load Y direction (N)	832	208.00	25.0	Lognormal
Muscle Load Z direction (N)	2055	513.75	25.0	Lognormal

An ilio tibial-tract load of 250 N ($F_{ilio\ tibial\text{-}tract}$) is applied to the bottom of the femur in the longitudinal femur direction. Distal end of the femur is constrained not to move in horizontal direction. Applied loads are shown in Figure 7.

Finite element analyses of the prosthesis are carried out using ANSYS on a P4 2.0 GHz Intel processor PC. Each analysis takes about 48 hours of CPU time.

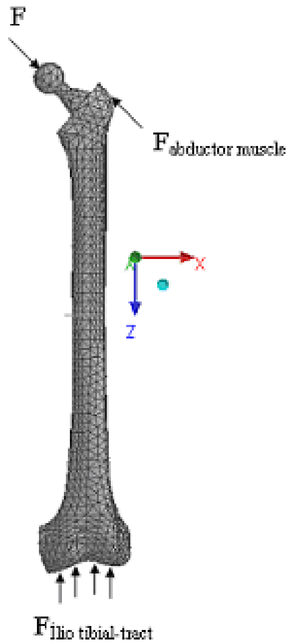


Fig. 7. Applied forces on the bone-cement-prosthesis assembly.

3. Results

In this paper, design optimization methodology was applied to evaluate the failure probability of the bone-cement, bone-cement hip prosthesis interface, bone-cement femur interface, hip prosthesis and femur under different loading, different material properties for hip prosthesis and different interface conditions. Performance function of probabilistic analysis reveals finite probabilities of failure in Table 3. All performance functions were examined. The newly designed hip prosthesis resulted in a decrease in the probability of failure compared to Charnley designed (Table 3).

Table 3. Performance function results

Failure Criteria	Newly Design p_f%	Charnley Design p_f%	Improvement %
Compressive failure (Cement)	3.86	6.45	40
Shear failure (Cement)	2.26	5.23	57
Fatigue failure (Cement)	1.54	3.84	60
Compressive failure (Prosthesis)	2.69	4.78	44
Shear failure (Prosthesis)	3.39	5.71	41
Fatigue failure (Prosthesis)	2.46	3.98	38
Compressive failure (Femur)	1.32	2.56	48
Shear failure (Femur)	1.69	3.09	45
Fatigue failure (Femur)	1.05	2.68	61
Tensile failure (Cement-stem interface)	2.56	4.96	48
Shear failure (Cement-stem interface)	1.78	4.21	58

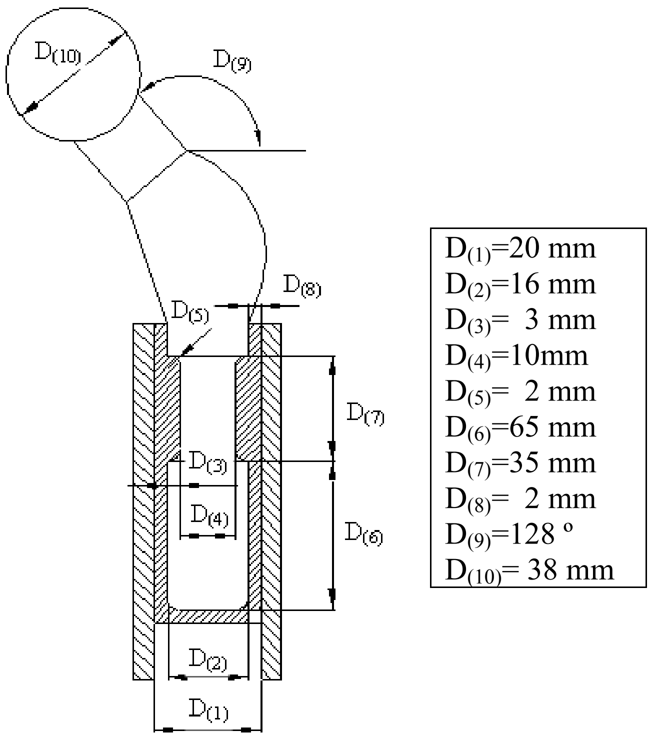


Fig. 8. Optimum prosthesis shape.

The deterministic optimization resulted in a final prosthesis shape is shown in Figure 8. This final shape, driven by the deterministic minimization of Von mises stress in the hip prosthesis, resulted in a decrease in the maximum value of the Von mises stress in the hip prosthesis compared to Charnley results. These values are 23.54% with Ti-6Al-4V and 28.92% with cobalt-chromium in Table 4. At the bone-cement-femur and bone-cement–prosthesis interface, the prosthesis shape optimization resulted in a reduction of the Von mises stress respectively (54.33%, 31.45%). This shape also resulted in decreases in the maximum value of the Von mises stress in the bone-cement. The shape optimization resulted in a decrease in both the mean value and the standard deviation of each structural response Table 4.

Table 4. Compared results

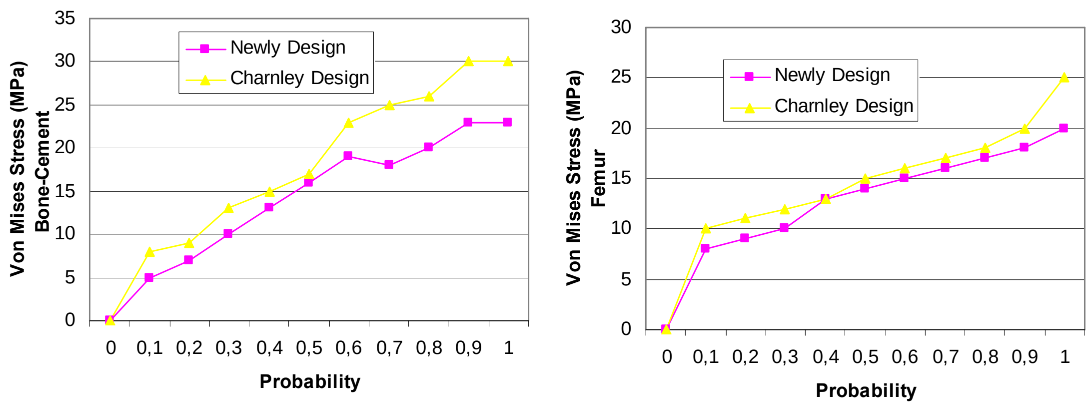
	Newly Design		Charnley Design		Improvement %	
	Mean	S.D	Mean	S.D	Mean	S.D
Hip Prosthesis Von mises stress (MPa) Ti-6Al-4V	85.65	10.63	112.02	16.87	23.54	36.98
Hip Prosthesis Von mises stress (MPa) Cobalt-chromium	96.47	12.48	135.73	20.26	28.92	38.40
Bone-Cement Von mises stress (MPa)	22.91	8.71	30.09	13.39	23.87	34.95
Interface Von mises stress (MPa) Bone-cement-femur	9.07	3.63	19.86	6.75	54.33	46.22
Interface Von mises stress (MPa) Hip Prosthesis-Bone-cement	18.17	5.41	26.51	9.16	31.45	40.93

For all performance functions examined, the deterministic shape optimization resulted in a decrease in the probability of failure in Table 5. Compressive failure of the bone-cement, prosthesis and femur were reduced 75.78%, 32.59% and 7.78% respectively compared to Charnley design. Shear failure of the bone-cement, prosthesis, femur and bone-cement-prosthesis interface shear failure were reduced 48.50%, 34.00%, 28.87% and 44.98% respectively compared to Charnly design. Most likely failure mode remained bone-cement fatigue failure. The final prosthesis shape resulted in a decrease in the probability of bone-cement fatigue failure of 81.63%. This result is quite important. Because bone-cement fatigue life is very important factor in the implant design. Designer works reduce stress in the bone cement and increase fatigue life of bone-cement.

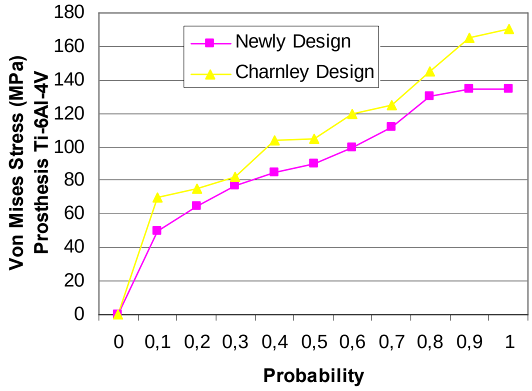
Table 5. Performance function results

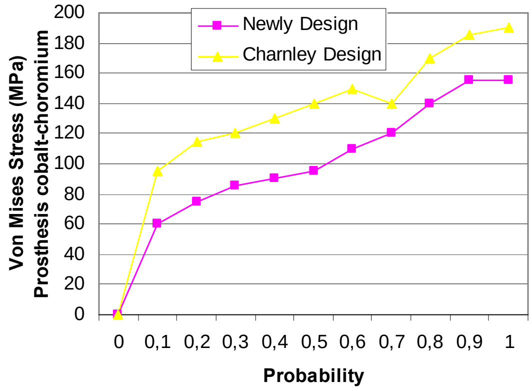
Performance Function	Failure Criteria	Newly Design p_f (%)	Charnley Design p_f (%)	Change (%)
1	Bone-cement compressive failure	5.21	21.51	75.78
2	Bone-cement shear failure	6.72	13.05	48.50
3	Bone-cement fatigue failure	2.32	12.63	81.63
4	Prosthesis compressive failure	32.23	47.81	32.59
5	Prosthesis shear failure	21.28	32.28	34.00
6	Prosthesis fatigue failure	32.41	45.98	29.51
7	Femur compressive failure	9.71	10.53	7.78
8	Femur shear failure	3.45	4.85	28.87
9	Femur fatigue failure	1.52	1.74	12.65
10	Bone-cement-prosthesis interface tensile failure	56.95	86.97	34.52
11	Bone-cement-prosthesis interface shear failure	17.84	32.42	44.98

A reduction of 28.92% of the deterministic response (Von mises stress in the hip prosthesis with Ti-6Al-4V) resulted in reduction in the probability of compressive failure of 32.59 %, shear failure 34.00% and fatigue failure 29.51%. A reduction of 23.87% of the deterministic response (Von mises stress in the bone cement) resulted in reduction in the probability of compressive failure of 75.78%, shear failure 48.50% and fatigue failure 81.63%. The shape optimization resulted in a reduction of the Von mises stress on the bone cement %23.33, on the femur 23.07% on the hip prosthesis with Ti-6Al-4V 22.35%, with cobalt-chromium 18.91%, on the hip prosthesis-femur interface 18.22%, on the bone-cement-hip prosthesis interface 17.85 %. Results are shown in Figure 9.

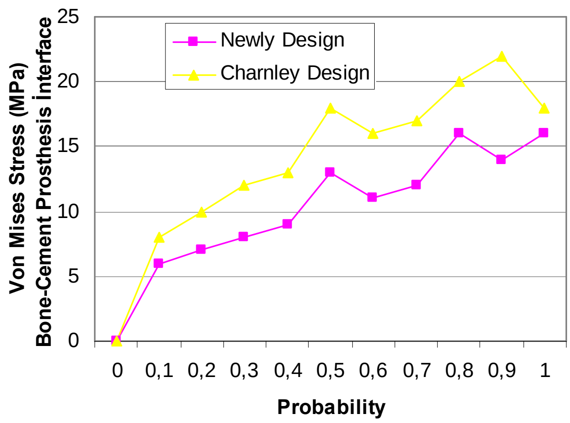


(a) (b)





(c) (d)



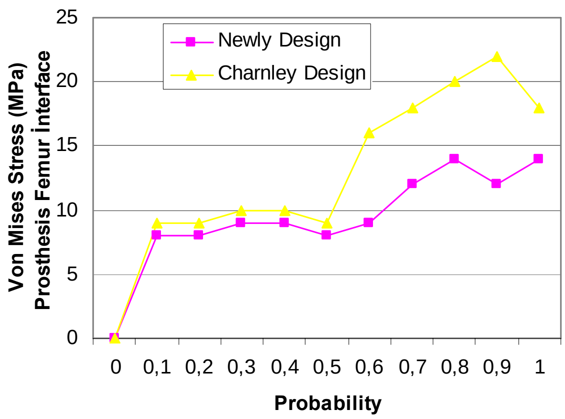


Fig.9 Effect of deterministic design optimization (a) Von mises stress bone-cement, (b) Von mises stress femur, (c) Von mises stress hip prosthesis (Ti-6Al-4V), (d) Von mises stress hip prosthesis (cobalt-chromium), (e) Von mises stress bone-cement-hip prosthesis interface, (f) Von mises stress hip prosthesis-femur interface

In this study we also investigated effects of design parameters. We calculated affects of stress distribution and fatigue life that design parameters how much affect on the bone-cement, hip prosthesis and femur from optimization history. Results are shown in Table 5. If we arrange design parameters we show that most affect design parameters are $D_{(8)}$, $D_{(2)}$, $D_{(9)}$, $D_{(10)}$, $D_{(5)}$, $D_{(4)}$, $D_{(1)}$, $D_{(7)}$, $D_{(6)}$, $D_{(3)}$ respectively stress distribution on the bone-cement, femur and hip prosthesis and also affect of fatigue.

Table 5. Affects of design parameters

Design Parameters	Affect of Stress Distribution on Bone Cement (%)	Affect of Stress Distribution on Femur (%)	Affect of Stress Distribution on Prosthesis (%)
$D_{(1)}$	46	21	2
$D_{(2)}$	79	7	81
$D_{(3)}$	37	8	40
$D_{(4)}$	51	12	58
$D_{(5)}$	67	8	65
$D_{(6)}$	38	32	34
$D_{(7)}$	44	38	26
$D_{(8)}$	85	42	21
$D_{(9)}$	76	72	81
$D_{(10)}$	71	76	83
Design Parameters	Affect of Fatigue Life Distribution on Bone Cement (%)	Affect of Fatigue Life Distribution on Femur (%)	Affect of Fatigue Life on Prosthesis (%)
$D_{(1)}$	46	21	2
$D_{(2)}$	79	7	81
$D_{(3)}$	37	8	40
$D_{(4)}$	51	12	58
$D_{(5)}$	67	8	65
$D_{(6)}$	38	32	34
$D_{(7)}$	44	38	26
$D_{(8)}$	85	42	21
$D_{(9)}$	76	72	81
$D_{(10)}$	71	76	83

4. Conclusion

Probabilistic approach was applied to current work in order to effect of three-dimensional deterministic shape optimization of a newly design cemented hip prosthesis on the predicted probability of failure of the prosthesis system. A major limitation of numerical modeling in orthopedics is the inability to specifically account for uncertainty and variability associated with biological structures. In this work, uncertainty and variability are accounted for in a probabilistic finite element analysis of newly designed cemented hip prosthesis. Random variables are used to describe joint and muscle loading, bone and cement material properties, and cement and interface strengths. The risk of failure is explicitly quantified in terms of two failure modes: bulk cement failure and cement-implant interface failure. Each failure mode, various failure criteria are investigated. Newly designed prosthesis has low probability of all performance functions failure than Charnly design. Cement stress optimized prosthesis significantly reduces risk of cement fatigue failure. Newly designed cemented hip prosthesis results have better behavioral characteristic than Charnly design. A general conclusion is that implant hip prostheses can be designed and studied with computer models before implementation on the patient. This procedure reduces design time while helping to prevent permanent damage caused by miss-implementation.

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