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In vitro evaluation of stiffness graded artificial hip joint femur head in terms of joint stresses distributions and dimensions: finite element study

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Abstract The aim of the present work is to evaluate the artificial hip joint femur head that is made of Stiffness Graded (SG) material in terms of joint stresses distributions and dimensions. In this study, 3D finite element models of femur head that is made of SG material and traditional femur heads made of Stainless Steel (SS), Cobalt Chromium alloy (Co Cr Mo) and Titanium alloy (Ti) have been developed using the ANSYS Code. The effects on the total artificial hip joint system stresses due to using the proposed SG material femur head (with low stiffness at the outer surface and high stiffness at its core) have been investigated. Also, the effects on the polymeric cup contact stresses due to the use of different sizes of femur heads, presence of metal backing shell and presence of radial clearance (gap) between cup and femur head have been investigated. The finite element results showed that using SG femur head resulted in a significant reduction in the cup contact stresses even for small femur heads compared with Ti alloy, SS and Co Cr Mo femur heads. The presence of radial clearance resulted in significant increase in the cup stresses especially for small femur heads. Finally, the presence of SS metal backing shell resulted in slight increase in the hip joint stresses especially for small femur head joints. This work analyzes successfully the usage of proposed SG material as femur head in order to reduce the predicted stresses at the total hip joint replacement due to the redistribution of strain energy in the hip prostheses. Therefore, the present results suggest that minor changes in design and geometrical parameters of the hip joint have significant consequences on the long term use of the joint and should be taken into consideration during the design of the hip joint.

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1 Introduction

The success of the artificial hip joint in increasing mobility and reducing the pain can lead to significant improvements in the quality of many people's lives. Around 80% of the total hip joint replacements are successful between 15 and 20 years [1, 2]. However, since these joints are now being implanted in younger people, they must be durable for at least more than 20 years. Currently, it is believed that the long-term behavior of the total hip joint replacement is dependent on obtaining low wear rate within the hip joint polymeric cup [3–8].

According to experimental and clinical results, there are many reasons for the artificial hip joints failure. These include: infection, dislocation, stem fracture and loosening, etc. However, the aspect loosening has been considered as one of the most common reasons for artificial hip joint failure. The hip joint loosening is linked to the presence of a large number of wear debris particles from the Ultra High Molecular Weight Polyethylene (UHMWPE) cup. The presence of such particles is considered as one of the main factors that affects the long-term stability of the total hip arthroplasty. These particles are known to induce adverse tissue reaction that leads to extensive bone loss around the implant and consequently osteolysis and implant loosening [1, 2, 8]. Also, the entering of foreign particles (third body) between the cup and femur head leads to hip joint loosening. These particles normally originate from the bone cement which the implant is fixed with.

Previous experimental results indicated that under high stress levels and with sliding distance comparable to those encountered in artificial hip joints; the UHMWPE cup subsurface fatigue contributes to the volume of wear debris [1, 9–11]. Since the fatigue process is influenced by surface stress levels, it is important to reduce contact stress at the cup with the aim of reducing the UHMWPE cup wear volume. Previous work indicated that the femur head dimensions and material, cup thickness and material, radial clearance between cup and ball and metal backing shell stiffness are identified as the most important factors that affect the contact pressure on the cup [12–15]. Therefore, over the past 50 years, different artificial hip joint designs and materials have been demonstrated to modulate implant survival [16–22].

The most common materials combinations of the artificial hip joint are metallic femoral head, polymeric cup, PMMA bone cement and sometimes metal backing shell that cover the UHMWPE layer for better fixation on the pelvis [12]. The most commonly used materials for artificial femur head are Co Cr Mo, SS, Ti, ceramic and carbon composite materials. They are used because of their high strength and good biocompatibility [9, 12, 13, 23]. One of the problems of the artificial hip joint implant is that most of the above femur head materials do not have the same strength and characteristics of acetabular component and human bone.

To the best of our knowledge, no studies have investigated the use of Stiffness Graded (SG) material (with low stiffness at the outer surface to simulate the natural hip joint cartilage and high stiffness at the core to support the joint load) in the artificial hip joint as a femur head. Also, the effects of different hip joint design parameters on resultant joint stresses are still unclear and can be important for the long term clinical performance of the prosthesis [23]. Therefore, the present study is part of a research project that is indented to fill the gap of using SG materials as a femur head and investigate their effects on the improvement of long term performance of the total hip joint replacement. This part of research aims at using the 3D finite element modelling for investigating the modification on the UHMWPE cup stresses when using SG material as femur head instead of traditional femur heads made of SS, Ti and Co Cr Mo under static loading conditions. Also, this study considers the effects of femur head dimensions and presence of radial clearance between ball and cup on the resultant contact stresses at the hip joint cup. The effects of the presence of metal backing shell on the UHMWPE contact stress are also studied. Preliminary results of the present project are previously published in [24]. Detailed analyses are considered in this study.

2 Materials and methods

2.1 Finite element modelling

Because of the difficulties in performing the in vivo tests for the artificial implants, mathematical models are developed to carry out the structural analyses of artificial implants prior to application on patients. Accordingly, the hip prostheses could be designed and examined with computer simulation prior to implantation. The finite element modeling (FEM) is a standard tool that can be used in the biomedical engineering field to accurately predict the state of stresses and strains in geometrically complex structures. The FEM is widely used in the design and analysis of total joint replacements and other orthopedic devices. The precision in the FE results can be achieved once geometry, material properties and boundary conditions are carefully specified.

In the present study, the 3D finite element models of the hip joint components are generated using the finite element code ANSYS v12 [25]. This implicit Lagrangian non-linear finite element code is used because of its efficiency for linear and non-linear quasi-static simulation. Previous work [26] indicated that the inclusion of the pelvic bone in the hip joint model can influence the resultant stresses on the UHMWPE component especially for non-metal backed model. Therefore, in the present study, the hip joint model is assumed to be composed of pelvic bone due to its remarkable effects on the results, femur head, UHMWPE cup and metal backing shell as shown in Fig. 1. The dimensions of hip joint models components are tabulated in Table 1 [27]. The pelvis is assumed to have a hemispherical shape with 42 mm thickness of cancellous bone [16].

Two different hip joint models are considered in the present study. In the first hip joint model, the artificial hip joint contains pelvic bone, femur head with different stiffness and dimensions and their corresponding UHMWPE cups. In the second model, a metal backing shell covers the UHMWPE cup as shown in Fig. 1. For both models, a radial clearance of 0.0, 0.1 and 0.2 mm between the femur head and UHMWPE cup are taken into consideration.

The geometries of femur heads, UHMWPE cups, pelvic bone, and metal backing shell are all constructed with CATIA software V5R19 [28] and then imported into the ANSYS code as IGES files. A 3D 20 node solid element (tetrahedron Solid 95) is chosen for modeling the hip joint assembly. This element is defined by 20 nodes having three degrees of freedom per node: translations in the nodal x, y, and z directions. Also, this element has the capability to accurately model the creep, the large deflection, and the large strain that may occur during the simulation. The contact interface between UHMWPE cup, metal backing

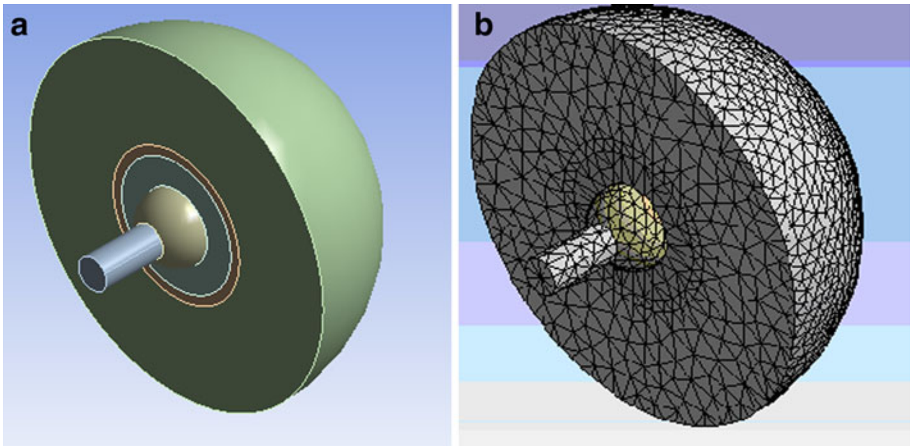

Fig. 1 Hip joint model

Table 1 Dimensions of hip joint

Femur head diameter (mm)	22	28	32	42
UHMWPE cup inner diameter (mm)	22, 22.1, 22.2	28, 28.1, 28.2	32, 32.1, 32.2	42, 42.1, 42.2
UHMWPE cup outer diameter (mm)	39	41	51	59
Metal backing thickness (mm)	0, 3	0, 3	0, 3	0, 3

shell and pelvic bone is represented as completely bonded surfaces. In order to simulate the low friction sliding between the metallic femur head and polymeric cup, surface to surface contact elements (target 170 and contact 174) are created between the femur head and UHMWPE cup. These 8 node elements support large deformation, with significant amount of sliding and friction efficiently.

The effects of mesh size and density on the predicted results are examined by increasing the number of elements until the predicted results become constant with increasing the mesh density. A static load of 3KN, which corresponds to 3–6 times body weight [13], is applied at the femur head neck as shown in Fig. 2. This resultant load is based on the assumption that the body weight is 70 kg. The other initial conditions like sex, age, activity, etc. are neglected [13].

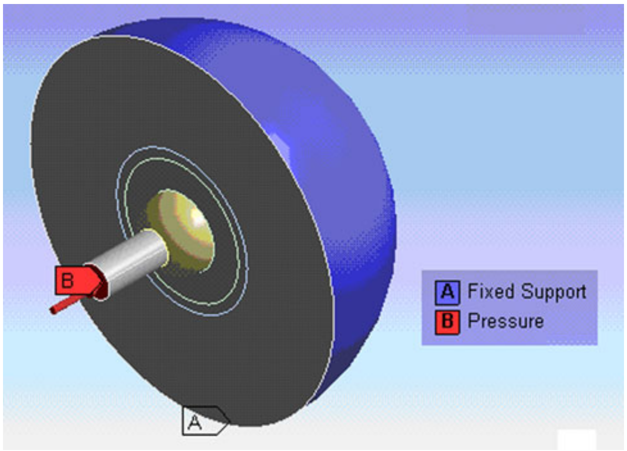

Fig. 2 Boundary conditions of the hip joint

2.2 Material properties

Four different materials are used in the present study to simulate the femur head behaviour using finite element simulation. Theses materials are Ti alloy, Co Cr Mo, SS and SG material. They are used to simulate the femur heads due to their high strength and sufficient biocompatibility in clinical conditions. The SS is also used as the metal backing shell material due to its high strength. Finally, the pelvis which is assumed to have a hemispherical shape is represented as a cancellous bone [16]. The femur head materials and the cancellous bone are assumed to be homogenous, isotropic and linearly elastic for simplification. The values of elastic modulus and Poisson's ratio of these materials are summarized in Table 2 [12, 13, 16, 22]. The SG material which is assumed to have an elastic modulus of 110GPa at the femur head core varies according to Eq. 1 to 50 GPa at the femur head outer surface. The SG material is assumed to have a constant Poisson's ratio of 0.3.

$$E_f = E_{\max}\, e^{-\ln\phi} \tag{1}$$

where $\phi = \left(\frac{E_{\max}}{E_{\min}}\right)^{\frac{d}{D_o}}$, E_f is the Young's modulus of the SG femur head at any position, E_{min} is the value of femur head modulus at the outer surface (equals 50 GPa), E_{max} is the

Table 2 Mechanical properties of the hip joint components

Material	Ti alloy	Co Cr Mo	FG material	Cancellous bone	SS
Elastic modulus GPa	110	230	Changes from (50–110)	0.33	210
Poisson's ratio	0.3	0.3	0.3	0.3	0.3

value of femur head modulus at the inner surface (core which equals 110 GPa), D_o is the maximum diameter of femur head and d is the variable diameter of femur head.

Finally, the UHMWPE material is used to simulate the polymeric cup behavior [29–34]. The UHMWPE true stress/strain data which is previously obtained by the author [31] has been used to simulate the behaviour of the polymeric cup in the FEM. The UHMWPE cup behaviour is modeled in the ANSYS Code using a Multi-linear Isotropic hardening material model that is recommended for large strain analysis. The behaviour of UHMWPE cup is represented by a piece-wise linear stress–strain curve, starting at the origin with positive stress–strain values through a certain number of points. The tensile stress/strain curve of UHMWPE is fed into the ANSYS Code as two regions. The first is the linear deformation region which is modeled by the knowledge of the tested material Young's modulus (900 MPa) and the corresponding Yield strength (23.5 MPa). The second region which represents the tested material plastic deformation (non-linear strain hardening) is modeled by the knowledge of the flow stresses and their corresponding total strains. The UHMWPE is assumed to have a constant Poisson's ratio of 0.4 [12].

3 Results and discussions

3.1 Effects of femur head dimension on the UHMWPE contact stresses

Figure 3 shows the variation on the maximum Von Mises stresses at the UHMWPE cup for different femur head geometries and stiffness for metal backed joint. It can be noticed that the maximum Von Mises stresses at the UHMWPE cup decreased significantly with increasing the femur head dimension for all types of femur head materials. For the Co Cr Mo femur head, the stress at the cup decreased by 39% when a femur head of 32 mm diameter is used instead of 22 mm femur head. Also, for SG femur head, the stress at the cup decreased by 45% when a femur head of 32 mm diameter is used instead of 22 mm. The significant reduction in UHMWPE cup stresses due to using large femur heads can be referred to the increase in the contact area between the femur head and the cup. The contact stresses at the cup are inversely dependent on the femur dimension as shown in the following equation [35].

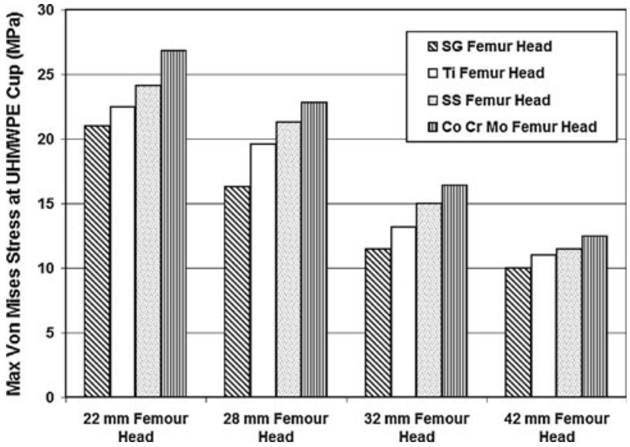

Fig. 3 Variation of UHMWPE cup Von Mises stress with femur head diameter for different femur head stiffness for metal backed joint

$$\sigma = \frac{12\,F}{5\pi\,R^2_{\text{femur}}} \tag{2}$$

The larger contact area between the cup and head induced lower contact stress at cup.

For the smallest femur head (22 mm diameter), the values of maximum Von Mises stresses on the UHMWPE cup are 26.8, 24.1, 22.5 and 21 MPa for Co Cr Mo, SS, Ti and SG femur heads respectively. These values (except for SG femur heads) are higher than the yield strength of UHMWPE material which is around 23.5 ± 1 MPa [31–34]. The presence of such local higher stress zones will affect the subsurface fatigue of UHMWPE cup, which contributes to aspect joint loosening. As the femur head dimension increased to 32 mm, the maximum values of Von Mises stresses in the UHMWPE cup became 16.4, 15, 13.2 and 11.5 MPa for Co Cr Mo, SS, Ti and SG femur heads respectively. In this case, the contact stresses in the UHMWPE cup are lower than its yield strength. This reduction in the UHMWPE contact stresses will lead to an improvement in the short and long term performance of the hip joint implant [1, 19]. As noticed from the above results, it can be concluded that, for the same femur head size, the use of SG femur heads results in a reduction in the UHMWPE stresses when compared with the other femur heads at the same conditions. The reduction in the predicted cup contact stresses will improve the long term performance of the total hip joint replacement.

Previous results indicated that the use of larger femur heads may induce larger wear rate but give a smaller wear depth [36]. Therefore, the advantage of using femur heads of larger diameter not only includes the possibility for a greater range of motion, greater intrinsic stability of the implant and reduction in incidence dislocation but also results in a reduction of the cup contact stress and wear depth [9, 36].

3.2 Effects of femur head stiffness on the UHMWPE stresses

The effects of femur head stiffness on the resultant Von Mises stress at the UHMWPE cup are also indicated in Fig. 3. From this figure it can be noticed that the resultant stresses on the UHMWPE cup decreased significantly when using SG material instead of Co Cr Mo, SS and Ti. For example, for 22 mm femur head, the resultant Von Mises stresses on the UHMWPE cup decreased from 26.8 to 21 MPa when using SG femur head instead of Co Cr Mo. For 32 femur head, the resultant Von Mises stress on the UHMWPE cup decreased from 16.4 to 11.5 MPa when using SG femur head instead of Co Cr Mo. The proposed SG femur head resulted in the reduction of the UHMWPE cup Von Mises stresses compared with SS, Ti and Co Cr Mo femur heads. It is important to mention that when using SG material as femur head, the values of the resultant stresses on the UHMWPE cup are always below its yield strength even for the smallest femur head size (22 mm).

The reduction in the cup contact stresses can be referred to that the outer layer of SG femur head has relatively low stiffness that allows more deformation and more contact area between the SG femur head and UHMWPE cup. Therefore, the hip joint load is distributed onto larger contact area which results in the reduction of the joint peak stresses. Similar trends are obtained by Jiang [12], where the UHMWPE cup stresses decreased due to the insertion of soft artificial cartilage between the metallic femur head and polymeric cup. Also, similar trends are obtained by Lan Chen et al. [37] where the stresses at the UHMWPE cup decreased by 32% when using Carbon femur head with low modulus instead of ceramic femur head.

From the above results it can concluded that the usage of proposed SG material as femur head successfully reduces the predicted stresses at the total hip joint replacement. This reduction can be regarded to the redistribution and absorption of strain energy in the hip prostheses due to the high flexibility of the SG femur head outer layer compared to other femur heads. The reduction in the predicted contact stresses at the UHMWPE cup will decrease the subsurface fatigue that may occur at the polymeric cup. This will lead to a reduction of the artificial hip joint loosening rate and improving its short and long term performance.

3.3 Effects of metal backing on the UHMWPE stresses

Figure 4 shows the effects on the resultant UHMWPE cup Von Mises stresses with femur head diameters for metal backed and non backed implants. The results indicated that for small femur heads, the presence of metal backing resulted in a reduction in the UHMWPE cup Von Mises stress for all femur head materials. For 22 mm femur head, the UHMWPE cup stresses decreased by 7, 11, 8 and 9% when using SS metal backing for Co Cr Mo, SS, Ti and SG femur head joint respectively. These results are similar to the results obtained by Hia-bo [12] where the UHMWPE cup stresses decreased by about 10% when using SS as a metal backing for Ti and Co Cr Mo femur heads. On the other side, Barreto et. al. [26] indicate that the presence of metal backing in the artificial hip joint results in a remarkable increase in the UHMWPE cup stresses under dynamic loading conditions.

For the implants with SG femur head more than 28 mm in diameter, the resultant contact stresses at the UHMWPE cup is unchanged due to the presence of SS metal backing. There is no need for metal backing when using SG femur head with a diameter of more than 28 mm. This result can be referred to in this case; the UHMWPE cup has sufficient dimensions to carry out the implant load. In contrast, Hio-bo [12] indicates that the presence of metal backing thickness results in a slight increase in the UHMWPE cup stresses for the artificial hip joint with a soft artificial cartilage between the 28 mm metallic femur head and polymeric cup.

Finally, the presence of metal backing results in a uniform distribution of Von Mises stresses at the UHMWPE cup. This uniform stress distribution at the cup theoretically decreases the loosening rate that may be caused by creep of the cup and improves the artificial hip joint longevity.

3.4 Effects of the presence of gap in the hip joint design on the cup stress

Figure 5 shows the effects on the resultant UHMWPE cup Von Mises stresses due to presence of radial clearance between cup and femur head for different femur head dimensions and stiffness. The results of Fig. 5 indicate that the presence of radial clearance results in a significant increase in the cup stresses especially for small femur heads for all femur head materials. The increase in UHMWPE cup Von Mises stress due to the presence of clearance can be attributed to the reduction in the contact area that carries the joint load. Similar results have been obtained experimentally and numerically by Korhonen et al. [9]. Their results indicate that the presence of 0.25 mm clearance between cup and femur head elevates peak contact pressure at cup-head interface by 27% and at the cup-cement interface by 65%. Therefore, the reduction of radial clearance between the bearing surfaces in the joint replacements would be useful in reducing the contact stress at the cup and hence preventing the onset of wear [23, 38].

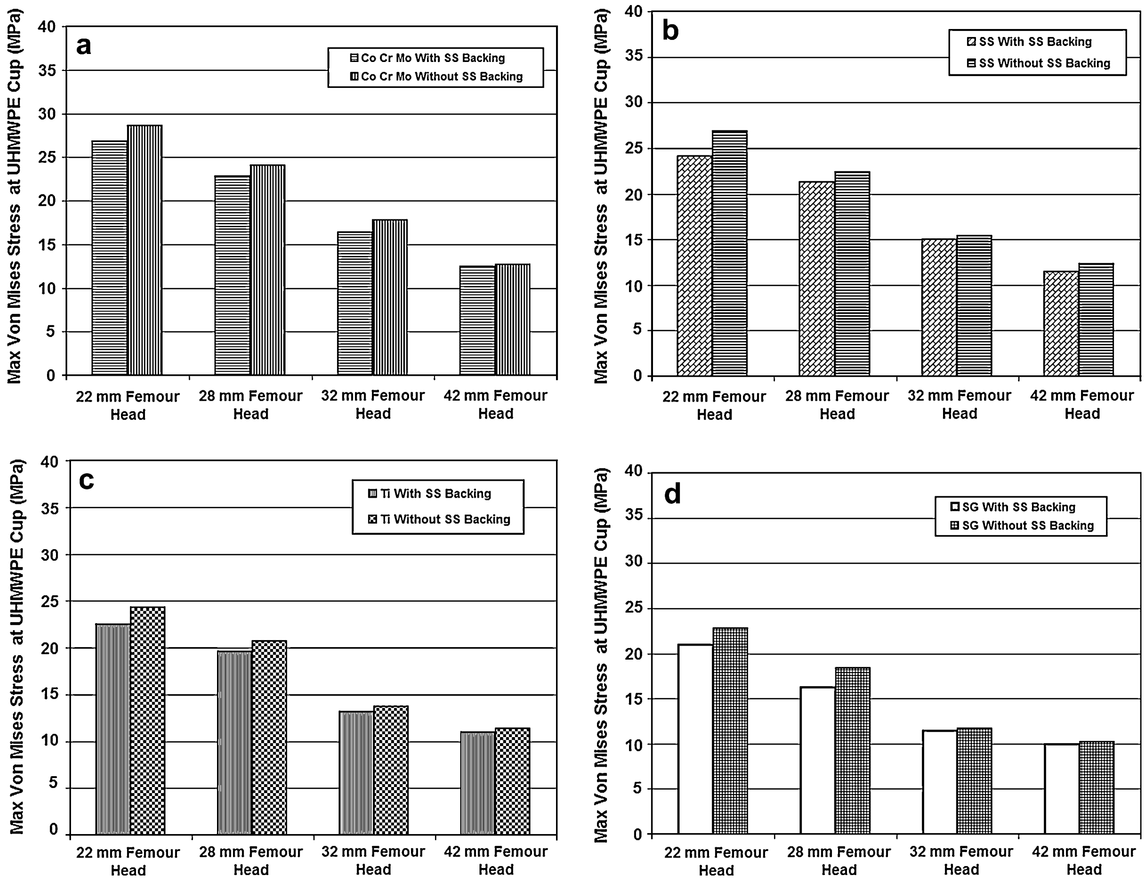

Fig. 4 Variation of UHMWPE cup stress with femur head diameter and stiffness for different hip designs

Also, Teoh et al. [3] indicate that the predicted wear rate at the UHMWPE cup is sensitive to radial clearance between the femur head and UHMWPE cup. They have found that severe wear occurs when the clearance is close to 0 and <0.5 mm. The best clearance range is between 0.1 and 0.15 mm where the average linear wear rate is 0.1 mm/year and the volumetric wear is 55 mm^3/year. Therefore, it is important to avoid too tight or too loose diametrical clearance between the cup and the femur head.

From Fig. 5, it can be noticed that for Co Cr Mo femur head with 22 mm diameter, the cup stresses increase by 83% due to the presence of 0.2 mm gap while for SG femur head, the cup stresses only increase by 53% due to the same reason. In this case, the predicted stresses at the UHMWPE cup exceed its yield point even for SG femur head. For larger femur heads (32 mm), the predicted stresses at the cup become 26, 24, 23 and 20 MPa due to the presence of 0.2 mm gap for Co Cr Mo, SS, Ti and SG femur heads respectively. Theses values (except for SG femur head) exceed the yield point of the UHMWPE cup. From the above results, it can be concluded that the use of SG femur head (with low stiffness at its outer surface and higher stiffness at the core) in the artificial hip joint with gap or without gap results in lower stresses at the cup compared to Co Cr Mo, SS and Ti femur heads. This reduction can refer to that the outer layer of SG femur head has relatively low stiffness that allows more deformation and more contact area between the SG femur head and UHMWPE cup. The hip joint load is then dispersed onto larger contact area which results in the reduction of the joint peak stresses. So, the use of SG femur head results in a remarkable reduction of the overall hip joint stresses and will improve the long term performance of the total joint prostheses.

3.5 Stresses at bone

The variations of the resultant stresses at the bone with femur head dimensions and stiffness are shown in Fig. 6. The results indicate that the stresses at the bone decrease when the femur head dimensions increase. Also, the results indicate that the use of SG femur head results in a slight reduction of the stress at the bone especially for large femur heads. The presence of metal backing also results in the reduction of stress at the bone especially for small femur heads. This reduction can be attributed to the

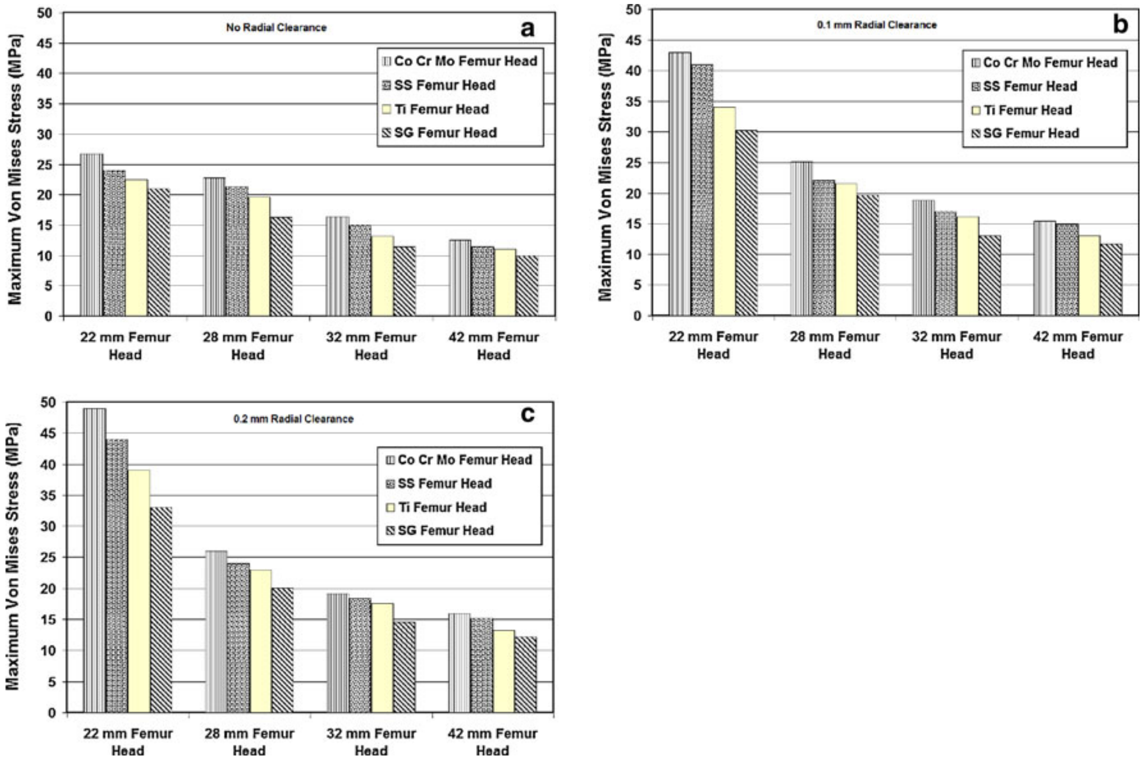

Fig. 5 Variation of UHMWPE cup stress with radial clearance for different femur head diameters and stiffness for 3 mm SS metal backed implant

absorption of load by the metal backing shell. In the present study, the resultant stresses at the pelvic bone range from 3.5 to 5.5 MPa. Other finite element studies on the artificial hip joint have reported predicted stresses of 6.7 MPa in the pelvic [23]. Generally, for all femur head materials and dimensions, the resultant stress at the cancellous bone has acceptable values (see Fig. 7) [39]. The use of SG femur head and presence of metal backing shell has insignificant effects on the predicted stresses at the human bone.

3.6 3D Empirical relation for stresses at the cup

From the FE results of polymeric cup stresses, a 3D empirical relation has been established using Mathmatica software to describe the variation of cup stresses (σ in MPa) as a function of femur head stiffness, dimensions and radial clearance as the following form:

$$\sigma(D,E,z) = \sum_{i=0}^{3} c_i D^i + lE + \gamma_1 \ln(D) + \mu_1 \ln(E); \tag{3}$$

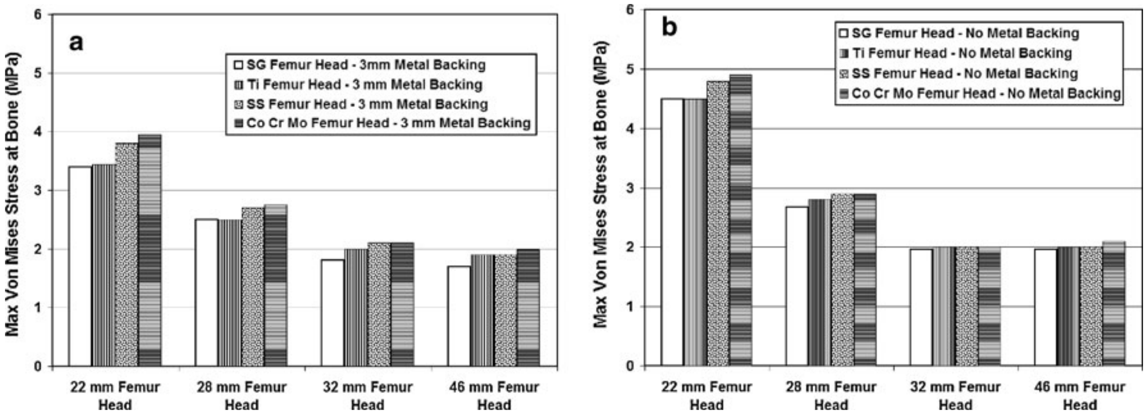

Fig. 6 Variation of bone Von Mises stress with femur head diameter for different femur head stiffness

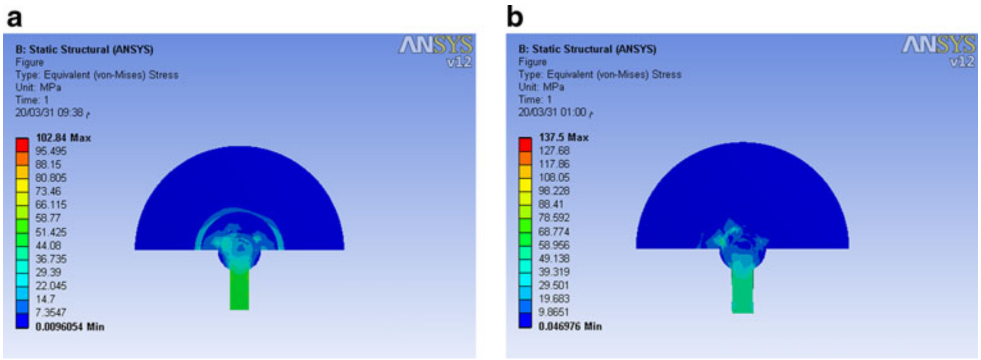

Fig. 7 Variation of Von Mises stress on the hip joint of 22 mm SG femur **a** metal backed design **b** non-metal backed design

$$\sigma(D,E,z)=\sum_{i=0}^{3}c_i D^i+lE+\gamma_2\ \ln(D)+\mu_2\ \ln(E); \qquad (4)$$

$$\sigma(D,E,z)=\sum_{i=0}^{3}e_i D^i+fE+\gamma_3\ \ln(D)+\mu_3\ \ln(E); \qquad (5)$$

Equation 3 represents the variation of cup stresses with cup stiffness and dimensions for zero radial clearance hip joint models, where D is the femur diameter in mm and E is the femur stiffness in GPa. The constants of Eq. 3 are:

$$a_0=-17.424,\quad a_1=25.8,\quad a_2=-0.8,\quad a_3=0$$

$$b=1.54,\qquad \eta_1=0,\quad \gamma_1=-25.23,\quad \mu_1=-51$$

Eq. 4 represents the variation of cup stresses with cup stiffness and dimensions for 0.1 mm radial clearance hip joint models. The constants of Eq. 4 are:

$$c_0=14.34,\quad c_1=-12.4,\quad c_2=0.24,\quad c_3=0$$

$$l=-0.8,\qquad \eta_2=0,\quad \gamma_2=21.1,\quad \mu_2=40.8$$

Equation 5 represents the variation of cup stresses with cup stiffness and dimensions for 0.2 mm radial clearance hip joint models. The constants of Eq. 5 are:

$$e_0=20.7,\quad e_1=-20.7,\quad e_2=0.5,\quad e_3=0$$

$$f=-1.1\quad \eta_3=0,\quad \gamma_3=30.1,\quad \mu_3=59$$

Figure 8 shows the variation of cup stresses as a function of femur head stiffness and dimensions obtained from the 3D empirical relation established using Mathmatica software. From Fig. 8, it can be concluded that the predicted stresses at UHMWPE cup can be reduced significantly when using larger femur head regardless its stiffness. Also, this figure indicates that the use of low stiffness (SG) femur head results in a remarkable reduction in the cup stresses even for smaller femur heads. Therefore, the present results suggest that rather minor changes in design and geometrical parameters of the hip joint have significant consequences on the long term use of the joint and should be taken into consideration during the design of the hip joint.

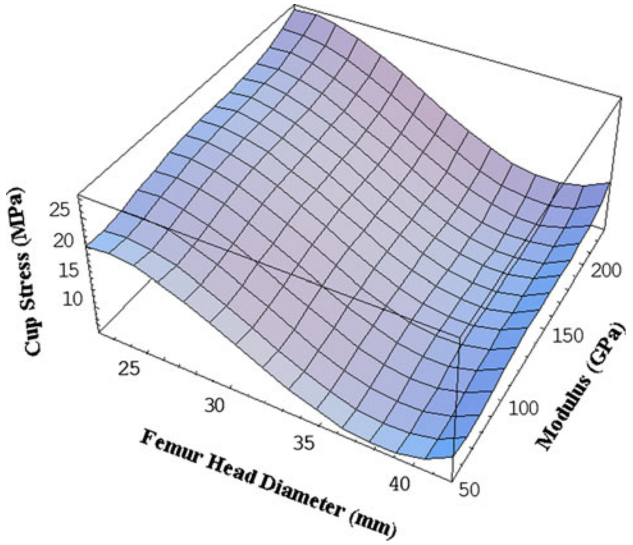

Fig. 8 Variation of cup stresses as a function of femur head stiffness and dimensions obtained that obtained using Mathmatica software

4 Conclusion

This study analyzes successfully the usage of proposed SG material as femur head to reduce the predicted stresses at the total hip joint replacement due to the redistribution and absorption of strain energy in the hip prostheses. The FE results indicate that the use of SG femur head with low stiffness at the outer surface and high stiffness at the core results in significant reduction in the UHMWPE cup contact stress even for small femur heads compared with Ti, SS and Co Cr Mo femur heads. For example, for 3 mm metal backed prosthesis, the use of SG material as a femur head instead of Co Cr Mo results in a reduction in the UHMWPE cup Von Mises stress with 22, 29, 30, 21% for 22, 28,32 and 42 mm femur diameters respectively. From the FE results it can also be remarked that for 22 mm Co Cr Mo femur head, the cup stresses increase by 83% due to the presence of 0.2 mm gap while for SG femur head, the cup stresses only increase by 53% due to the same reason. Also, for larger femur heads (42 mm), the stresses at the cup increase by 28 and 20% due to the presence of 0.2 mm gap for Co Cr Mo and SG femur heads respectively. Therefore, the use of SG femur head (with low stiffness at its outer surface and higher stiffness at the core) in the artificial hip joint with gap or without gap results in lower stresses at the cup compared to Co Cr Mo, SS and Ti femur heads. Accordingly, the author believes that the use of SG material as a femur head will be better for the reduction of the resultant stresses at the joint and will improve the artificial hip joint performance.

References

1. Barbour PSM, Barton DC, Fisher J. The Influence of contact stress on the wear of UHMWPE for total replacement hip prostheses. Wear. 1995;181:250–7.
2. McNie CM, Barton DC, Ingham E, Tipper JL, Fisher J, Stone MH. The prediction of polyethylene wear rate and debris morphology produced by microscopic asperities on femoral heads. J Mater Sci: Mater Med. 2000;11:163–74.
3. Teoh SH, Chan WH, Thampuran R. An elasto-plastic finite element model for polyethylene wear in total hip arthroplasty. J Biomech. 2002;35:323–30.
4. Saikko V, Shen M. Wear comparison between a dual mobility total hip prosthesis and a typical modular design using a hip joint simulator. Wear. 2010;268:617–21.
5. Wilches LV, Uribe JA, Toro A. Wear of materials used for artificial joints in total hip replacements. Wear. 2008;265:143–9.
6. Ilchmann T, Reimold M, Müller-Schauenburg W. Estimation of the wear volume after total hip replacement. A simple access to geometrical concepts. Med Eng Phys. 2008;30:373–9.
7. Barnes CL, DeBoer D, Corpe RS, Nambu S, Carroll M, Timmerman I. Wear performance of large-diameter differential-hardness hip bearings. J Arthroplast. 2008;23:56–60.
8. Affatato S, Bersaglia G, Rocchi M, Taddei P, Fagnano C, Toni A. Wear behaviour of cross-linked polyethylene assessed in vitro under severe conditions. Biomaterials. 2005;26:3259–67.
9. Korhonen RK, Koistinen A, Konttinen YT, Santavirta SS, Lappalainen R. The effect of geometry, abduction angle on the stresses in cemented UHMWPE acetabular cups-finite element simulations, experimental tests. BioMed Eng Online. 2005;4:32.
10. Rixrath E, Wendling-Mansuy S, Flecher X, Chabrand P, Argenson JN. Design parameters dependences on contact stress distribution in gait and jogging phases after total hip arthroplasty. J Biomech. 2008;41:1137–42.
11. Bertram T, Anton H, Johan K, Veronika K, Gunnar F, Nico V, Ron D. Association between contact hip stress and RSA-measured wear rates in total hip arthroplasties of 31 patients. J Biomech. 2008;41:100–5.
12. Jiang Hai B. Static and dynamic mechanics analysis on artificial hip joints with different interface designs by the finite element method. J Bionic Eng. 2007;4:123–31.
13. David B, Tarun G. Finite element analysis of hip stem designs. Mater Des. 2008;29:45–60.
14. Schmid DM, Wullschleger L, Derler S, Schmitt KU. Development of a new design of hip protectors using finite element analysis and mechanical tests. Med Eng Phys. 2008;30:1186–92.
15. Kang L, Galvin AL, Jin ZM, Fisher J. A simple fully integrated contact-coupled wear prediction for ultra-high molecular weight polyethylene hip implants: Proceedings of the Institution of Mechanical Engineers. Part H J Eng Med. 2006;220:33–46.
16. Griza S, Cê AN, Silva EP, Bertoni F, Reguly A, Strohaecker TR. Acetabular metal backed fatigue due to severe wear before revision. Eng Fail Anal. 2009;16:2036–42.
17. Wasielewski RC, Jacobs JJ, Arthurs B, Rubash HE. The acetabular insert-metal backing interface: an additional source of polyethylene wear debris. J Arthroplast. 2005;20:914–22.
18. Chen FS, Di Cesare PE, Kale AA, Lee JF, Frankel VH, Stuchin SA, Zuckerman JD. Results of cemented metal-backed acetabular components: a 10-year-average follow-up study. J Arthroplast. 1998;13:867–73.
19. Burger NDL, De Vaal PL, Meyer JP. Failure analysis on retrieved ultra high molecular weight polyethylene (UHMWPE) acetabular cups. Eng Fail Anal. 2007;14:1329–45.
20. John GB, Thomas KD, Paul AW, Ian CC. Surface damage after multiple dislocations of a 38-mm-diameter, metal-on-metal hip prosthesis. J Arthroplast. 2008;23:1090–6.
21. Alistair PDE, Richard MH, Ian MP, Anthony U. Wear in retrieved acetabular components: effect of femoral head radius and patient parameters. J Arthroplast. 1998;13:291–5.
22. Darwish SM, Al-Samhan AM. Optimization of artificial hip joint parameters. Mater Sci Eng Technol. 2009;40:218–23.
23. Cilingir AC. Finite element analysis of the contact mechanics of ceramic-on-ceramic hip resurfacing prostheses. J Bionic Eng. 2010;7:244–53.
24. Fouad H, Darwish SM. Effects of femur design parameters on the resultant contact stress on the UHMWPE Cup. In: 4th international conference on advanced computational engineering and experimenting. Paris; 2010.
25. ANSYS Version 12. ANSYS Inc., Canonsburg; 2009.
26. Barreto S, Folgado J, Fernandes PR, Monteiro J. The influence of the pelvic bone on the computational results of the acetabular component of a total hip prosthesis. J Biomech Eng. 2010;132(5):054503.
27. Muratoglu OK, Bragdon CR, O'Connor D, Perinchief RS, Estok DM 2nd, Jasty M, Harris WH. Larger diameter femoral heads used in conjunction with a highly cross-linked ultra-high molecular weight polyethylene: a new concept. J Arthroplast. 2001;16:24–30.
28. CATIA V5. Dassault Systemes, Mississauga; 2008.
29. Qingliang W, Jinlong L, Shirong G. Study on biotribological behavior of the combined joint of CoCrMo and UHMWPE/BHA composite in a hip joint simulator. J Bionic Eng. 2009;6:378–86.
30. Furmanski J, Anderson M, Bal S, Greenwald AS, Halley D, Penenberg B, Ries M, Pruitt L. Clinical fracture of cross-linked UHMWPE acetabular liners. Biomaterials. 2009;30:5572–82.
31. Fouad H. Experimental and numerical studies of the notch strengthening behaviour of semi-crystalline ultra-high molecular weight polyethylene. Mater Des. 2010;31:1117–29.
32. Mourad A-H I, Fouad H, Rabeh E. Impact of some environmental conditions on the tensile, creep-recovery, relaxation, melting and crystallinity behaviour of UHMWPE-GUR 410-medical grade. Mater Des. 2009;30:4112–9.
33. Fouad H, Mourad A-H I, Barton DC. Effect of pre-heat treatment on the static and dynamic thermo-mechanical properties of ultra-high molecular weight polyethylene. Polym Testing. 2005;24:549–56.
34. Fouad H. Effect of long-term natural aging on the thermal, mechanical, and viscoelastic behavior of biomedical grade of ultra high molecular weight polyethylene. J Appl Polym Sci. 2010;118:17–24.
35. Hedia HS, Abdel-Shafi AA, Fouda N. Shape optimization of metal backing for cemented acetabular cups. Biomed Mater Eng. 2000;10:73–82.
36. Wu JS, Hung JP, Shu CS, Chen JH. The computer simulation of wear behavior appearing in total hip prosthesis. Comput Method Program Biomed. 2003;70:81–91.

37. Chen L, Du X, Zheng Y, Bao Y, Chen M, Hu M, Yu D. Finite element analysis of LTI carbon hip joint head. Bioinformatics Biomed Eng. ICBBE 3rd International Conference; 2009. 11–13 June. p. 1–4.
38. Kosak R, Antolic V, Pavlovcic V, Kralj-Iglic V, Milosev I, Vidmar G, Iglic A. Polyethylene wear in total hip prostheses: the influence of direction of linear wear on volumetric wear determined from radiographic data. Skeletal Radiol. 2003;32:679–86.
39. Currey JD. What determines the bending strength of compact bone? J Exp Biol. 1999;202:2495–503.