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Effect of Femoral Head Size on Contact Pressure and Wear in Total Hip Arthroplasty

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ABSTRACT: Total hip arthroplasty using artificial materials is a widely used treatment for osteoarthritis and similar disabling conditions. The hip is essentially a ball and socket joint, linking the “ball” at the head of the thigh bone (femur) with the cup-shaped “socket” in the pelvic bone. The longevity of total hip arthroplasties is significantly reduced by the wear of joint surfaces. Prosthetic implant wear and joint degeneration mechanism can be estimated for the hip joint by predicting contact area and pressure distribution during activities of daily living which provide biomechanical rationales for preoperative planning and postoperative rehabilitation. The objective of the current study was to generate a finite element model capable of predicting the influence of femoral head size and coefficient of friction between femoral head and cup on contact pressure distribution as well as wear in total hip replacement. These stress distributions represent an important factor which affects the development of hip and also determines the state of health or disease of the adult hip. Moreover, the understanding of wear behavior will aid in the evaluation of the clinical penetration measurements of patients with hip arthroplasty and design of THR devices. Furthermore, the prediction of contact pressure and wear rates can help in deciding for a treatment and in planning the operation. The finite element models were generated using ANSYS finite element software by developing an ANSYS Parametric Design Language macro for easy and quick calculations. A standardized femur was used as a basis for the FEM models. This study has demonstrated that the increasing of head diameter can reduce the maximum contact pressure between the metallic femoral head and the Ultra High Molecular Weight Poly-Ethylene cup. Furthermore, wear penetration was found to decrease asymptotically with increasing head size and the predicted wear rates were well within clinically observed ranges for each component size.

KEYWORDS: Total Hip Replacement (THR), Finite Element Analysis (FEA), Contact pressure distribution, Femoral head diameter, Coefficient of friction,Wear.

I. INTRODUCTION

The analysis technique that can study the patient's hip joint contact force/pressure distribution would be useful to assess the effect of abnormal biomechanical conditions and anatomical deformities on joint contact stress for treatment planning purpose [1]. The dome of the acetabulum, which has been considered a weight bearing area, is in fact flexible. The horns of the acetabulum can thus close and contact the femoral head when the joint is loaded. The stresses on the femoral head usually act on the anterior superior quadrant, and surface motion can be considered as sliding on the acetabulum.

The longevity of total hip arthroplasties is significantly reduced by the wear of joint surfaces. Wear is a result of the product between contact pressure and sliding distance that is generated by the patient's daily activity [2].

Yoshida et al. [3] investigated a three dimensional dynamic hip contact area and pressure distribution during activity of daily living by using Discrete Element Analysis (DEA) technique to represent the assumed spherical geometry of the femoral head. They found that, during fast, normal, and slow walking; the peak pressure of moderate magnitude was

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located at the lateral roof of the acetabulum during mid-stance. The peak pressures were located at the edge of the posterior horn and the magnitude of the peak pressure during sitting down was 2.8 times that of normal walking.
The analysis of contact mechanics in McKee-Farrar metal on metal hip implants was studied by Yew et al. [4]. They concluded that for zero clearance, although the contact pressure was significantly reduced over most of the contact area, the whole acetabular cup came into contact with the femoral head, leading to stress concentration at the edge of the cup. They added that the design optimization of the geometrical parameters, in terms of the acetabular cup thickness and the radial clearance is important not only to minimize the contact stress at the bearing surface but also to avoid equatorial and edge contact.

Ipavec et al. [5] developed a mathematical model for calculating the contact stress distribution in the hip for a known resultant hip force and characteristic geometrical parameters. They showed that not only the value of the maximum contact stress is important for the loading of the hip but also the shape of the stress distribution. The value of the maximum contact stress may be relatively low if the body weight is low and if the articular sphere radius is large. The stress distribution may in spite of this be unfavorable if the pole lies outside the weight bearing area due to the steeply descending stress distribution.

The influence of femoral head size on impingement, dislocation, and stress distribution, in total hip replacement is also studied by Kluess et al. [6]. Stress analysis showed decreased contact pressures at the egress site of the liners with the larger inner diameters during subluxation. Their analysis show that an optimal implant position and a larger head diameter can reduce the risk of dislocation induced by impingement.

Furthermore, the effect of femoral head diameter and operative approach on risk of dislocation after primary total hip arthroplasty was investigated by Daniel et al. [7]. Their results demonstrated that in total hip arthroplasty, a larger femoral head diameter was associated with a lower cumulative risk of dislocation.

Sultan et al. [8] enhanced stability of total hip replacement implants resulting from use of an elevated-rim acetabular liner and 32-mm femoral head. They compared the stability of the hip with a 32 mm femoral head with the standard 28 mm head. They found that the 32 mm head may contribute to hip stability in primary total hip arthoplasty (THA), and in instances where a posterior approach is used, an elevated-rim liner placed in the posterior.

Genda et al. [1] investigated the inter-parameter correlations of the patient's hip joint contact force/pressure distribution. They calculated the pressure distribution in the hip joint by using a three-dimensional discrete element analysis (DEA) technique. They demonstrated that the normalized peak contact pressure was correlated both with acetabular coverage and head–trochanter ratio. Also they added that the change of abductor force direction within normal variation did not affect the joint peak contact pressure. However, in simulated dysplastic conditions when the centre edge (CE) angle is small or negative, abductor muscle direction becomes very sensitive in joint contact pressure estimation.

The effect of the friction coefficient on contact stresses was studied by Suhendra [9]. Their observations showed that the surface shear stress distribution at the cup liner may help in understanding the response of cup material to the sliding contact force. They also reported that the increase in shear stress acting at the interface, arising from increased friction, has an important effect on crack formation and subsequent surface failure of UHMWPE cup.

Laurian and Tudor [2] found that with the aid of finite element method it was possible to study the behavior of hip joint prostheses having different radial clearances, different femoral head diameters and constant coefficient of friction under similar loads. It is emphasized the influence of radial clearance on the contact area, maximum contact pressure and pressure distribution. They concluded that the maximum contact pressure is located at the pole, decreasing with the increase of head diameter and with the decrease of clearance. Also, they added both extremes; very small or very large clearance; lead to dangerously high values of the contact pressure.

Nicholas et al. [10] analysed seven acetabular cup systems for metal-cup/polyethyelene liner conformity. They concluded that polyethylene/metal conformity not only decreases the potential for polyethylene deformation, but also reduces contact stress concentrations by increasing the surface area available for load transfer. An increase in contact area will reduce stress induced redefinition of the polyethylene surface as will as inhibit the catalysis of wear mechanisms.

The effects of femoral head size on the deformation of ultrahigh molecular weight polyethylene acetabular cups were studied by Hoeltzel et al [11]. They concluded that the selection of an acetabular cup size and corresponding femoral head size in a total hip arthroplasty should not be an arbitrary one, but should be based on scientific studies which indicate minimum states of stress within the cup and cement mantle, as well as clinical evidence that the combination of components shows a reduced incidence of failure. They experimentally quantified the states of stress on the surface of the acetabular cup and pointed to the possible existence of an optimum component size to minimize surface stress.

Burroughs et al. [12] investigated the range of motion (ROM) and stability in total hip arthroplasty with 28, 32, 38 and 44 mm femoral head sizes. Their work was conducted using an anatomic full-size hip model and a novel anatomic dislocation simulator with 28, 32, 38 and 44 mm diameter femoral heads within a 61 mm acetabular shell. They found that, larger femoral heads offer potential in providing greater hip ROM and joint stability.

Wear in total hip arthroplasty is frequently estimated from patient radiographs by measurement of penetration of the femoral head into the polyethylene liner. Thus Bevill et al. [13] used a finite element simulation of early creep and wear in total hip arthroplasty to determine polyethylene creep and wear penetration and volumetric wear during simulated gait loading conditions for variables of head size, liner thickness, and head–liner clearance. They concluded that the least volumetric wear but the most linear penetration was found with the smallest head size.

The analysis of the catastrophic failure of a total hip replacement based on the role of the acetabular component was studied by Maccauro et al. [14]. They found that the initiator of the failure may be considered to be the pre-operative planning: the limited thickness of the PE liner, due to the selection of socket external diameter and to the ball head diameter leads to high wear.

The present study was restricted to a cemented acetabular component with metal backing shell and CoCrMo femoral head. In the current study we aim to develope a finite element model capable of predicting the influence of femoral head size and the coefficient of friction between the femoral head and the acetabular cup on contact pressure distribution and wear in total hip replacement (THR). Determination of stress in the hip joint can be of help in deciding for a treatment and in planning the operation. Furthermore, the finite element model enables simplification of design variations compared to experimental studies since prototyping and assembling will replaced by prompt numerical simulation.

II. MATERIALS AND METHODS

The current implant design is based upon the pioneering work conducted by Sir John Charnley in one legged stance. The prosthesis for total hip replacement consists of a femoral component and an acetabular component. The femoral component (femoral head) articulates against the acetabular component (acetabular cup) as shown in Fig. 1. The prosthesis can be modular when they consist of two or more parts and require assembly during surgery. When the acetabular component is modular, it consists of a metallic shell and an Ultra-High Molecular Weight Poly-Ethylene (UHMWPE) insert. The metallic shell seeks to decrease the micro-deformation of the UHMWPE and to provide a porous surface for fixation of the cup. The metallic shell allows worn polyethylene liners to be exchanged. Furthermore, in cases of repetitive dislocation of the hip after surgery, the metallic shell allows replacing the old liner with a more constrained one to provide additional stability. Great effort has been placed on developing an effective retaining system for the insert as well as on maximizing the congruity between insert and metallic shell.

The metal backing acetabular component model of Charnley prosthesis cemented into the acetabular cavity was used. Therefore, the developed model was divided into the cortical bone, subchondral bone, trabecular bone (which was divided into three regions; interamedullary, medium and denset; varying due to the variation of moduli of elasticity), cement layer, metal backing shell, acetabular cup and the femoral head as shown by Fig. 2.

In the developed model, a femoral head from cobalt-chromium-molybdenum (CoCrMo) alloy articulates against an UHMWPE acetabular cup in a total joint prosthesis. These components are fixed in place using Poly-Methyl-Meth-Acrylate (PMMA) bone cement. The CoCrMo alloy femoral head component were introduced as low friction metallic

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substitutes as a means of reducing polyethylene wear debris in hip replacement, they are highly biocompatible and significantly smoother, harder and more scratch resistant than their metallic counterparts.
All materials are assumed to be isotropic, linearly elastic and homogeneous. The mechanical characteristics of bone, cement, metal backing shell, UHMWPE and CoCrMo were taken from literature [9,15,16]. The values of these characteristics used by the developed finite element models are shown in table 1.

Table 1. Properties of materials represented in the FE model

Material	Modulus of elasticity (MPa)	Poisson's ratio
Cortical bone	17000	0.3
Subcondral bone	17000	0.3
Trabecular bone		
a. Intramedullary	1000	0.3
b. Medium	1500	0.3
c. Denset	3000	0.3
PMMA cement	2300	0.32
Stainless steel MB	200000	0.3
UHMWPE (Cup)	1000	0.3
CoCrMo (femoral head)	208000	0.3

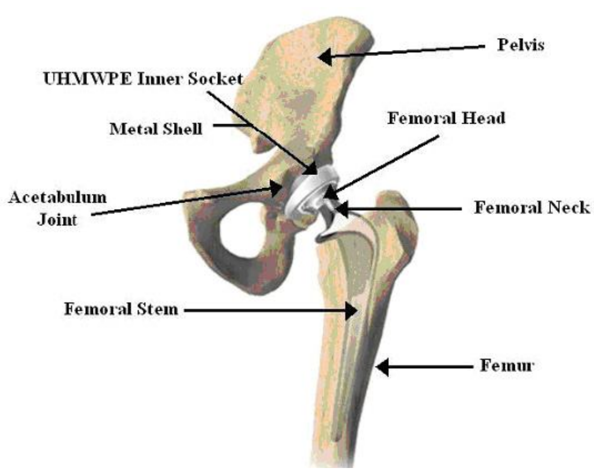

Fig. 1. The model of THR.

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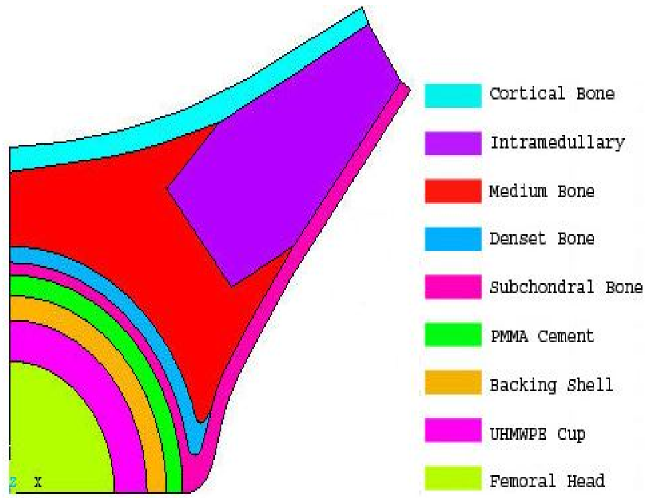

Fig. 2. Different materials used in THR model

FE modeling provides a rapid and inexpensive estimate of implant-related factors (e.g. femoral head diameter i.e. geometry of the acetabular cup) but the final judgment is based on laboratory tests and clinical data [17]. In the present study, a two dimensional (2D) axisymmetric finite element model was developed to investigate the influence of femoral head diameter and coefficient of friction between the femoral head and the acetabular cup on contact pressure developed between them. The commercially general purpose Finite Element Method (FEM) package ANSYS was used to develop the 2D axisymmetric model of the THR. The use of an axisymmetric model greatly reduces the modeling and analysis time. The model is used to incorporate the contact occurring between the femoral head and the cup bearing surfaces.

The axisymmetric finite element model simulation was conducted using the following boundary conditions :

- The model was fixed superiorly at the ilium as described early by Douglas et al [18] and as shown in Fig. 3.
- Constant loading force of 3 KN is used to establish contact pressure between the articulating components. This force is applied vertically on femoral head center line as shown by Fig. 3.
- It was presumed that no sliding motion would occur between the UHMWPE liner and the MB shell once the components has initially been seated, and therefore rigid fixation between the two components was assumed.
- Also, zero radial clearance between the femoral head and the cup liner was assumed.

At the beginning of the simulation, a small displacement was applied to establish full contact between the cup and the head. This avoids instabilities that would otherwise arise if the force was applied without contact/equilibrium having been established first.

In order to investigate the influence of femoral head diameter on contact pressure between the femoral head and the UHMWPE cup, several cases employing different head diameters and coefficients of friction were evaluated. A number of 9 cases were simulated, three femoral head diameters of 22, 28 and 32mm were studied assuming zero clearance between the femoral head and the cup liner, using three different friction coefficients of 0.02, 0.05 and 0.10. Table 2 present the characteristics of the different cases. The hip dimensions were constant for all studied cases except the inner diameter of the acetabular cup which varies by the femoral head diameter variation. The outer diameter of the UHMWPE acetabular cup was taken 60 mm for all studied cases.

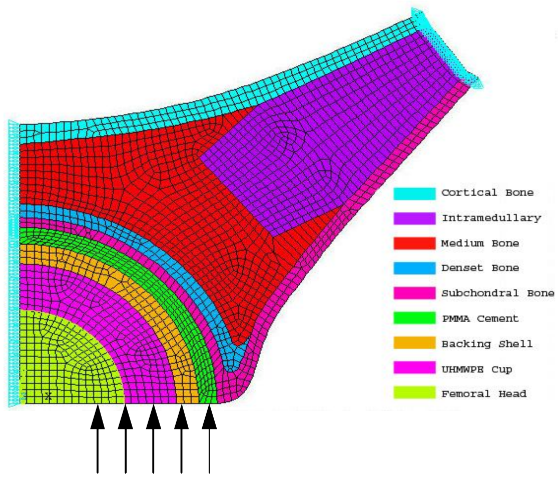

Fig. 3. The FE model (gross mesh) of total hip replacement showing the different materials, loading conditions and constraints.

The models were discretised geometrically using approximately 10000 plane 82 two-dimensional structural solid elements which defined by eight nodes having two degrees of freedom at each node (translations in the nodal x and y directions). Convergence of the FE model was assessed to justify the number of elements. A special feature in this study was to model the femoral head and the acetabulum part instead of the approximate modeling of the femoral head and the UHMWPE cup (ball and socket) as presented by many literatures. Considering the huge difference of the structural stiffness between the CoCrMo femoral head and the UHMWPE liner, the deformation of head could be neglected.

Table 2. Characteristics of the studied cases.

Case No.	Case symbol	Femoral head diameter (mm)	Coefficient of friction
1	D22 –MU0.02	22	0.02
2	D22 –MU0.05	22	0.05
3	D22 –MU0.10	22	0.10
4	D28 –MU0.02	28	0.02
5	D28 –MU0.05	28	0.05
6	D28 –MU0.10	28	0.10
7	D32 –MU0.02	32	0.02
8	D32 –MU0.05	32	0.05
9	D32 –MU0.10	32	0.10

III. WEAR EQUATIONS

It is generally recognized that wear of un-lubricated surfaces is directly related to a material wear coefficient, pressure (load), sliding distance (velocity) and time (Bartel et al., 1995) [21]. Acetabular cup linear penetration including both wear and volumetric wear were determined for femoral head sizes of 22, 28 and 32 mm.

The numerical wear simulation was performed based on the methods presented in detail originally by Maxian et al [22] and recently by Teoh et al [21] and Bevill et al [13]. Polyethylene cup wear was enforced on a nodal basis, and estimated using Archard's law (Archard, 1953): [21] as following:

$$W = K_w P S \qquad (1)$$

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Where:

- ***W*** is the wear depth for one cycle,
- ***K_w*** is a wear coefficient for metal on polyethylene, (K_w=1.0656* 10^{-7} $mm^3.N^{-1}.m^{-1)}$
- ***S*** is the sliding distance,
- ***P*** is the contact stress.

Archard's law was discretized by assuming that the surface normal contact pressure distribution and wear coefficient were constant over each increment of the gait loading. Relative sliding distances at the cup surface were retrieved from the FE results. Experiments have been carried out to derive a relationship between linear, volumetric or gravimetric wear and other variables, in particular, pressure and sliding distance. (Archard, 1953) This can be expressed in the following equation (2) [21] :

$$V = \left(\frac{K_0}{3B_H}\right).F_N\,.S \qquad (2)$$

Where:

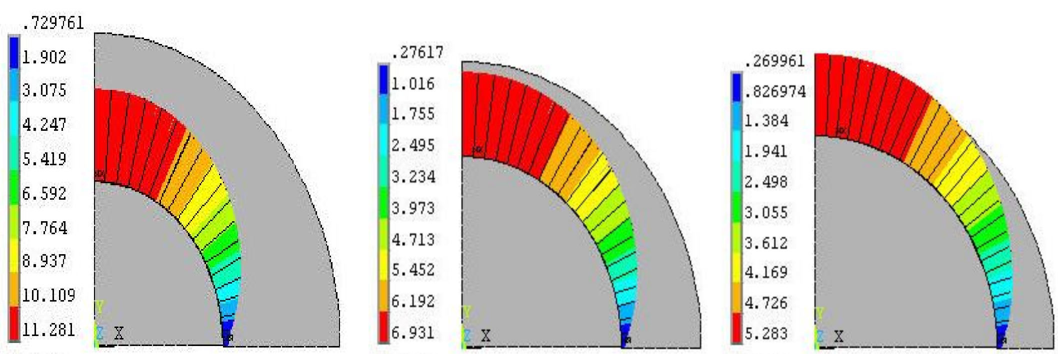
- ***V*** is the wear volume of an unlubricated surface,
- ***F_N*** is the normal load,
- ***S*** is the sliding distance,
- ***B_H*** is the yield pressure of the softer material
- ***K_0*** is a constant that depends on the type of materials, interfacial friction and the geometry of the surfaces in contact.

This equation applies to both adhesive and abrasive wear. It shows that the wear volume is proportional to the normal load and the sliding distance and inversely proportional to the yield pressure.

IV. RESULTS AND DISCUSSIONS

To examine the effect of femoral head diameter and coefficient of friction on contact pressure between the femoral head and the UHMWPE cup, the nine different cases listed above in table 2 were used in the analysis. Fig. 4 show the contact pressure distributions for case no. 1, 4 and 7, i.e. for 22, 28 and 32 mm femoral head diameters at 0.02 coefficient of friction.

While Fig. 5 present a comparison for the different values of contact pressure (Fig. 5-a) and sliding distance (Fig. 5-b) distributions for these three cases.



(a) (b) (c)

Fig. 4. Contact pressure distribution for femoral head diameter of 22mm (a), 28mm (b) and 32mm (c) at 0.02 coefficient of friction.

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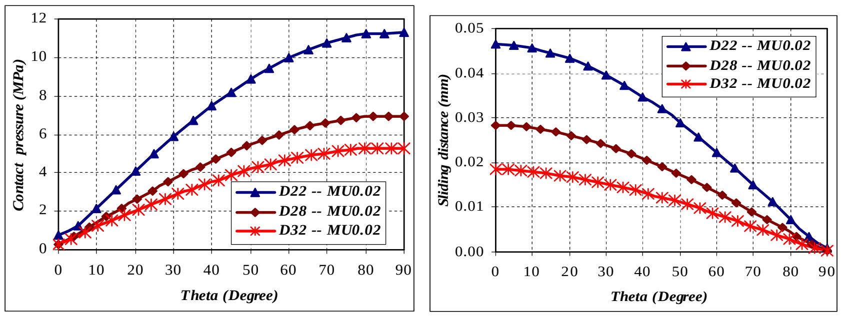


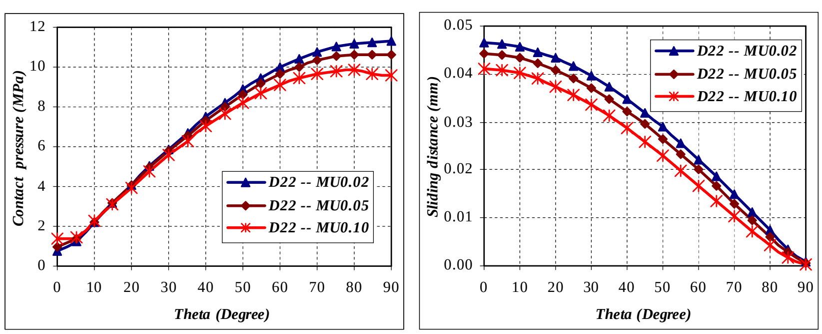
(a) **(b)**

Fig. 5. Contact pressure (a) and sliding distance (b) distributions for the different femoral head diameters at 0.02 coefficient of friction.

These results indicate that the maximum contact pressure for the 22 mm prosthesis were about 38% and 53% larger than those for the 28 and 32 mm. From this comparison we can conclude that the contact pressure and its maximal value changes as the bearing area changes due to the variation of the head diameter, i.e. when the femoral head diameter increase the maximum contact pressure decrease. Also, Fig. 5-b show that with the increasing of femoral head diameter the maximum sliding distance decreases.

Fig. 6-8 present the effect of coefficient of friction on contact pressure and sliding distance distributions between the femoral head and the UHMWPE cup for the different femoral head diameters. From the first look to these figures (i.e. Fig. 6-8) one can say that from the point of view of maximum contact pressure, the higher the coefficient of friction, the smaller the contact pressure and hence, the smaller the wear of the acetabular cup. This is true only until a certain coefficient of friction. Above it, the maximum pressure migrates towards the rim of the cup and increases exponentially with the increase of coefficient of friction. This means that both extremes (very small or very large coefficient of friction) lead to dangerously high values of the contact pressure.

Fig. 6 indicate that the maximum contact pressure for the 22 mm prosthesis at 0.02 coefficient of friction was about 6% and 13% larger than those for the 0.05 and 0.10 coefficient of friction. Similarly, Fig. 7 indicate that there is an increasing of about 8% and 12% in the maximum contact pressure value for the 28 mm prosthesis at 0.02 coefficient of friction than those for the 0.05 and 0.10 respectively. By the same manner, the maximum contact pressure for the 32 mm prosthesis at 0.02 coefficient of friction was about 4% and 10% larger than those for the 0.05 and 0.10 respectively as shown by Fig. 8.



(a) (b)

Fig. 6. Contact pressure (a) and sliding distance (b) distribution for the 22mm femoral head diameter at different coefficients of friction.

To get more evidence for the validity the results we compared them with the results reported in literature. Laurian and Tudor [2] indicated that, the maximum contact pressure is located at the pole and varies in the range of 4-11 MPa. While, Hodge et. al. [19] measured experimentally contact pressures in the human hip joint in vivo and found that, the data reveal very high local and nonuniform pressures (up to 18 MPa). Also, they show that at 2.6 body-weight force at

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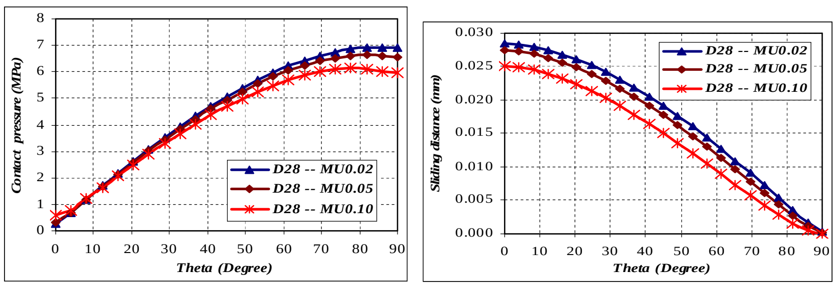
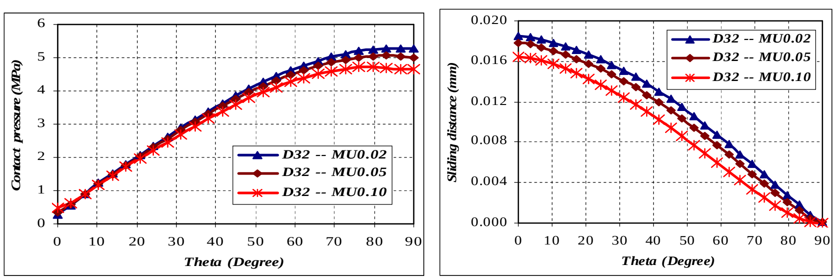
the hip (appropriate for the single-leg support phase of level walking), the peak local cartilage stress was 6.78 MPa. In addition, Afoke et. al. [20] measured the pressure distribution between the cartilaginous surfaces in the human hip joint using pressure-sensitive film. They found that the pressure distribution was not uniform and the maximum pressures recorded were about 10 MPa. Therefore, the results obtained from the developed numerical models compare well with the available literature results.



(a) (b)

Fig. 7. Contact pressure (a) and sliding distance (b) distribution for the 28mm femoral head diameter at different coefficients of friction.



(a) (b)

Fig. 8. Contact pressure (a) and sliding distance (b) distribution for the 32mm femoral head diameter at different coefficients of friction.

Sliding distances of points on the femoral head surface were obtained as shown in the above figures. Wear rates were determined by using the relationship that coupled contact stress, sliding distance (equations 1 and 2). Fig. 9 show the wear depth distributions (calculated from Eq. 1) for 22, 28 and 32 mm femoral head diameters at 0.02 coefficient of friction. The figure indicate that wear penetration was found to decrease asymptotically with increasing head size.

The effect of friction on wear has been debated. Two distinct points of view have emerged over the impact of friction on wear, one supporting and the other opposing [21]. The obtained results from the developed models indicate that the volumetric wear decreses with the increase of the coefficient of friction.

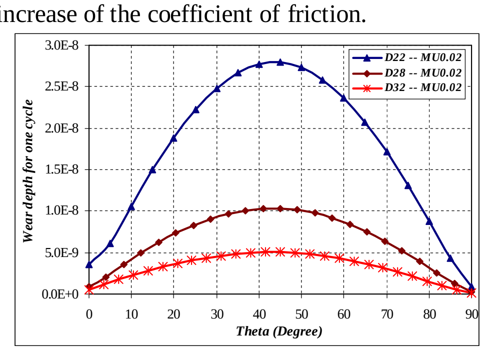


Fig. 9. Wear depth distributions for the different femoral head diameters at 0.02 coefficient of friction.

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V. CONCLUSION

The effect of contact pressure in the hip prosthesis for total hip replacement is critical in the wear of UHMWPE cup. In this paper, the finite element program ANSYS has been used in the developing numerical models to predict the effect of femoral head diameter and the coefficient of friction between the femoral head and the UHMWPE cup on the contact pressure and wear developed between them.FE analyses of the present study showed that the femoral head diameter, as well as the friction coefficient between cup and head modify the mechanical stresses experienced in the hip joint. These factors should be taken into account in the future development of THR and the FE technique is a powerful tool for this purpose because the developed FE model enables simplification of design variations compared to experimental studies since prototyping and assembling are replaced by prompt numerical simulation. Furthermore, these results are useful in selecting the optimal head diameter and may contribute to our understanding of contact conditions and overall performance of the hip joint. It can be concluded based on the developed models, the larger the femoral head diameter the smaller the contact pressure and wear depth. It is also found that with the increase of coefficient of friction between the head and cup the contact pressure developed decrease. The results obtained from the developed numerical models compare well with the available results in the literature.

REFERENCES

1. Genda, E., Iwasaki,. N., Li, G., Bruce, A., Williams, M., Barrance, P.J., Chao, E.Y.S., "Normal hip joint contact pressure distribution in single-leg standing - effect of gender and anatomic parameters", Journal of Biomechanics, Vol. 34, PP 895–905, (2001).
2. Laurian, T., Tudor, A., "Some aspects regarding the influence of the clearance on the pressure distribution in total hip joint prostheses", National Tribology Conference, ROTRIB'03 GALATI, PP 393-397, 24-26 September 2003.
3. Yoshida, H., Faust, A., Wilckens, J., Kitagawa, M., Fetto, J., Chao, E.Y.S., "Three-dimensional dynamic hip contact area and pressure distribution during activities of daily living", Journal of Biomechanics, Vol. 39, Issue 11, Pages 1996-2004, (2006).
4. Yew, A.,Jagatia, M., Ensaff, H.,Jin, Z.M., "Analysis of contact mechanics in McKee-Farrar metal-on-metal hip implants", Proceedings of the IMECHE Part H, Journal of Engineering in Medicine,Vol. 217, No.5, PP 333-340, (2003).
5. Ipavec, M., Brand, R.A., Pedersen, D.R., Mavcic, B., Kralj-Iglic, V., Iglic, A., "Mathematical modeling of stress in the hip during gait", Journal of Biomechanics, Vol. 32, PP 1229–1235, (1999).
6. Kluess, D.,Martin, H., Mittelmeier W., Schmitz, K.P., Bader R.,"Influence of femoral head size on impingement, dislocation and stress distribution in total hip replacement", Medical Engineering & Physics,Vol. 29, PP 465-471,(2007).
7. Berry, D.J., Knock, M.V., Schleck, C.D., Harmsen W.S., "Effect of femoral head diameter and operative approach on risk of dislocation after primary total hip arthroplasty", The Journal of Bone and Joint Surgery Vol. 87, PP 2456-2463, (2005).
8. Sultan, P.G., Tan, V., Lai, M., Garino, J.P.,"Enhanced stability of total hip replacement implants resulting from use of an elevated-rim acetabular liner and 32-mm femoral head", The University of Pennsylvania Orthopaedic Journal, Vol. 15, PP 17-19, (2002).
9. Suhendra, N., "Analysis of mechanical and thermal responses of total hip joint replacement acetabular components using FEM models", Prosiding Semiloka Teknologi Simulasi dan Komputasi serta Aplikasi, (2005).
10. Plaxston,N.A.,Postak, P.D., Greenwald, A.S.,"Cup/liner conformity of metal-backed acetabular designs", Orthopedic Research Laboratories, The Mt. Sinai Medical Center Cleveland, Ohio 44106, (1998).
11. Hoeltzel, D.A., Walt, M.J., Kyle, R.F., Simon, F.D.,"The effects of femoral head size on the deformation of ultrahigh molecular weight polyethylene acetabular cups", Journal of Biomechanics, Vol. 22, Issues 11-12 , PP 1163-1173, (1989).
12. Burroughs, B.R., Hallstrom, B., Golladay, G.J., Hoeffel, D., Harris, W.H., "Range of motion and stability in total hip arthroplasty with 28-, 32-, 38-, and 44-mm femoral head sizes: an in vitro study", The Journal of Arthroplasty, Vol. 20, Issue 1, PP 11-19, (2005).
13. Bevill, S.L, Bevill, G.R., Penmetsa, J.R., Petrella, A.J., Rullkoetter, P.J., " Finite element simulation of early creep and wear in total hip arthroplasty", Journal of Biomechanics, Vol. 38, PP 2365– 2374, (2005).
14. Maccauro, G., Piconi, C., Proietti, L., Timpanaro, M., De santis, V., Magliocchetti, G., De santis, E., " Analysis of the catastrophic failure of a THR: The role of the acetabular component", Hip international, Vol.11no.4, PP 201-208, (2001).
15. Hedia, H.S.,Abdl-Shafi, A.A.A., Fouda, N.,"Shape optimization of metal backing for cemented accetabular cup", Bio-Medical Materials and Engineering, PP 1-10, (2000).
16. Hedia, H.S., Abdl-Shafi, A.A.A., Fouda, N., "The effect of elastic modulus of the backing material on the fatigue notch factor and stress", Bio-Medical Materials and Engineering, PP 1-16, (2000).
17. Korhonen, R.K.,Koistinen, A.,Konttinen, Y.T., Santavirta, S.S., Lappalainen, R.,"The effect of geometry and abduction angle on the stresses in cemented UHMWPE acetabular cups- finite element simulations and experimental tests", BioMedical Engineering OnLine 2005, 4:32, http://www.biomedical-engineering-online.com/content/4/1/32 , (2005).
18. Douglas, R., Crowninsheld, R., Brand, R., Johnson, R., "An axisymmetric model of acetabular components in total hip arthoplasty", Journal of Biomechanical Engineering, Vol. 15, PP 305-315, (1982).
19. Hodge, W.A., Fijan, R.S., Carlson, K.L., Burgess, R.G., Harris, W.H., Mann, R.W., "Contact Pressures in the Human Hip Joint Measured in vivo", Proceeding of the National Academy of Sciences of the USA (PNAS), Vol. 83; PP 2879-2883, (1986)
20. Afoke, N.Y.P., Byers, P.D., Hutton, W.C., "Contact pressure in the human hip joint", The Journal of Bone and Joint Surgery, Vol. 69-B, No. 4, PP 536-541, (1987).
21. Teoh, S.H., Chan, W.H., Thampuran, R., "An elasto-plastic finite element model for polyethylene wear in total hip arthroplasty", Journal of Biomechanics, Vol. 35, PP 323–330, (2002).
22. Maxian, T.A., Brown, T.D., Pedersen, D.R., Callaghan J.J., "A sliding distance-coupled finite element formulation for polyethylene wear in total hip arthroplasty", Journal of Biomechanics,Vol. 29, PP 687–692, (1996).