ORIGINAL ARTICLE

Influence of Dimple Depth on Lubricant Thickness in Elastohydrodynamic Lubrication for Metallic Hip Implants Using Fluid Structure Interaction (FSI) Approach

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ABSTRACT

Introduction: The lubricant thickness in clearance between bearing surfaces for metallic hip implants are currently incapable of accommodating the motion experienced (high load and low entraining motion) in hip walking cycle. Thus, micro-dimpled surfaces were introduced onto surfaces of metallic acetabular cups to improve lubricant thickness. Micro-dimpled surface is a method of advanced surface improvement to increase the lubricant thickness in various tribological applications, such as hip implants. However, the application of micro-dimpled surfaces in hip implants has not yet been explored adequately. Therefore, this study aims to identify the influence of micro-dimpled depth on lubricant thickness elastohydrodynamically for metallic hip implants using Fluid-Structure Interaction (FSI) approach. **Methods:** Fluid-Structure Interaction (FSI) approach is an alternative method for analysing characteristics of lubrication in hip implant. Dimples of radius 0.25 mm and various depths of 5μm, 45μm and 100μm were applied on the cup surfaces. The vertical load in z-direction and rotation velocity around y-axes representing the average load and flexion-extension (FE) velocity of hip joint in normal walking were applied on Elastohydrodynamic lubrication (EHL) model. **Results:** The metallic hip implants with micro-dimpled surfaces provided enhanced lubricant thickness, namely by 6%, compared to non-dimpled surfaces. Furthermore, it was suggested that the shallow depth of micro-dimpled surfaces contributed to the enhancement of lubricant thickness. **Conclusion:** Micro-dimpled surfaces application was effective to improve tribological performances, especially in increasing lubricant thickness for metallic hip implants.

Keywords: Micro-dimpled Surfaces, Fluid-Structure Interaction (FSI), Elastohydrodynamic Lubrication (EHL), Metallic Hip Implants, Lubricant Thickness

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INTRODUCTION

Metallic hip implants referred to metallic surfaces for both femoral head and acetabular cup has become more popular than metal-on-polyethylene (MoP) due to their low wear rates, namely 2-20μm/year compared to 100-300μm/year, respectively. As the metallic wear debris has nanometers in size (1,2), it could cause higher levels of cobalt and chromium ions in serum, urine, and red blood cells of patients (3,4), leading to hypersensitivity, tissue toxicity and carcinogenesis (5,6). Thus, decreasing metallic wear debris as much as possible is crucial to improve the lifetime of hip joints implants (7,8).

Micro-dimpled surfaces is an approach for advanced surface improvement, which has been shown experimentally and theoretically to enhance the tribological performance including reducing friction and increasing the lubricant thickness (9–11). It acts as a lubricant reservoir to prevent direct surface-to-surface contact (12–14). D. Choudhury et al (15,16) observed in their experimental study that honed surface on metallic cup produced lowest friction coefficient. Furthermore, the experimental study by Hui Zhang et al (17) on the effect of petaloid surfaces for cobalt chromium alloy (Co-Cr) artificial joint showed that petaloid surfaces significantly improved the tribological performance by minimizing the friction and wear of Co-Cr hip implant. It also produced micro-hydrodynamic pressure under high load and low entraining motion. L. Gao et al (18) used an advance numerical model to solve the problem of mixed elastohydrodynamic lubrication (EHL) for metallic hip implants with dimpled surface. The study indicated that the lubricant thickness in dimpled surfaces models were improved compared to non-dimpled surfaces. However, their investigations were limited to analysis of only one size of micro-dimple surface. Limitations remained in investigation of various sizes of micro-dimples such as depth on optimum tribological performances particularly lubricant thickness.

Full fluid lubrication in hip implants are crucial to improve tribological performances for metallic bearing configurations. It is an ideal lubrication regime in hip joint capsules which minimises the number of wear debris, because optimum lubricant thickness between the bearing contact between surfaces can retain the high load to prevent direct contact (19). However, both experimental and theoretical investigations showed that metallic hip implants perform in a mixed lubrication regime (20–22). Due to high load and low entraining velocity of hip joint in normal walking, the lubrication mode is potentially changing to become boundary lubrication or direct contact. Therefore, the idea of micro-dimpled surfaces was introduced to improve the EHL lubrication performance by increasing the lubricant thickness. The micro-dimpled surfaces are capable of acting as secondary lubrication due to squeezing action under load and the sliding motion (23). However, a conclusive solution remains devoid due to lack of research regarding efficiency of the micro-dimpled surfaces, especially for hip joint replacements. In this study, the influence of micro-dimple depth on lubricant thickness in EHL for metallic hip implants was identified using Fluid-Structure Interaction (FSI) approach.

MATERIALS AND METHODS

Governing equations

The primary objective of governing equations is assessing hydrodynamic pressure in the lubricant area. Thus, primary governing equations were used namely the Continuity and Navier-Stokes equations. The equation of Navier-Stokes governed the lubricant flow and can be seen as Newton's second law of lubricant flow:

$$\frac{\partial \rho}{\partial t} + \nabla(\rho \mathrm{u}_{fluid}) = 0 \tag{1}$$

$$\rho \frac{\partial \mathrm{u}_{fluid}}{\partial t} + \nabla(\rho \mathrm{u}_{fluid} \cdot \mathrm{u}_{fluid} - \left[-pI + \mu\left(\nabla \mathrm{u}_{fluid} + \left(\nabla \mathrm{u}_{fluid}\right)^{T}\right)\right]) = Fv \tag{2}$$

where, ρ is density of lubricant, u_{fluid} is velocity of lubricant, and Fv is body force.

Pressure, which produced in the lubricant area, is deforms the elastic of cup surfaces based on the equation of elasticity as shown in Equation (3):

$$\rho \frac{\partial^2 u_{solid}}{\partial t^2} - \nabla . \sigma = Fv \tag{3}$$

where u_solid is cup surfaces deformation, Fv is force of body, and σ is tensor of stress calculated from Hooke's law.

$$\sigma = C.\varepsilon \tag{4}$$

where C is fourth-order stiffness tensor and ε is tensor of strain calculated from:

$$\varepsilon = \frac{[(\nabla \mathrm{u}_{solid})^T + \nabla \mathrm{u}_{solid} + (\nabla \mathrm{u}_{solid})^T \nabla \mathrm{u}_{solid}]}{2} \tag{5}$$

However, for thin-film cases, the Reynolds equation was utilized to analyse the hydrodynamics lubrication between moving surfaces in FSI method. This equation correlates between hydrodynamic pressure and lubricant thickness, velocity of lubricant and surfaces. Thus, Reynolds equation was used in FSI simulation to analyse the lubrication problem and was separated by two parts, which are lubricant and solid as shown in Equation (6)

$$\nabla_T \rho \left(\frac{-h^3}{12\eta} \nabla_T p_{fluid} + \frac{h}{2} v_{fluid} + f . v_{fluid}\right) = \nabla_T \rho \left(\left(s . v_{solid}\right) - \frac{h}{2} v_{solid}\right) \tag{6}$$

where h is the lubricant thickness, ρ is lubricant density, p is the pressure of lubricant, fis location of lubricant region, s is location of solid wall, v_fluid is the tangential speed (m/s) of the lubricant area, and v_solid is the tangential speed (m/s) of the solid wall. Because the pressure is constant throughout the lubricant thickness, Comsol uses the tangential projection of the gradient operator, ∇T to calculate the pressure distribution on the lubricant surface. Then, this equation was combined to form the Reynolds equation for FSI simulation as shown in Equation (7).

$$\nabla_T \left(\frac{-\rho h^3}{12\eta} \nabla_T p_{fluid} + \frac{\rho h}{2}\left(v_{fluid} + v_{solid}\right)\right) - \rho\left(\left(\nabla_T s . v_{solid}\right) - \left(\nabla_T f . v_{fluid}\right)\right) = 0 \tag{7}$$

The lubricant thickness, h is defined as:

$$h = c(1 - \varepsilon_x \sin\theta\cos\phi - \varepsilon_y \sin\theta\sin\phi - \varepsilon_z \cos\theta) \tag{8}$$

and it is stated that $\varepsilon_x = \varepsilon_y = 0$ in this study. Therefore,

$$h = c(1 - \varepsilon_z \cos\theta) \tag{9}$$

where c is radial clearance, ε_z is eccentricity ratio and θ is polar coordinate of a point on the lubricant.

The total lubricant thickness (Σh) included the clearance between femoral head and acetabular cup surfaces (h) and the deformation (δ) produced by the lubricant pressure was evaluated in Equation (10) :

$$\sum h = h + \delta \tag{10}$$

In EHL for metallic hip implants with dimpled surfaces, the total lubricant thickness (*Σh*) included the clearance between femoral head and acetabular cup surfaces (*h*), deformation produced by lubricant pressure and the dimple depth (*d*), as evaluated in Equation (11):

$$\sum h = h + \delta + d \tag{11}$$

The external force, F is counterbalanced by the pressure in the lubricant. This is imposed as a constraint:

$$\int p_f \, dS - F = 0 \quad (12)$$

The hydrodynamic pressure generated by the lubricant caused elastic deformation of two surfaces containing the lubricant. Thus, the hydrodynamic pressure was used as mechanical load on the elastic wall to evaluate the deformation of the acetabular cup surfaces.

Lubrication model

A model of metallic hip implant was assumed as cobalt-chromium (CrCo) in this study (Fig. 1(a)). The 14mm femoral head radius (R1) and the 30µm radial clearances, with 9.5mm acetabular cup thickness (t) were adopted in this study (18,24,25). The material properties of metal (CoCr) bearing had a Young's modulus of 210MPa and a Poisson's ratio of 0.3. The equivalent support for metallic acetabular cup and femoral head, were considered linear elastic. The model was analysed using commercial software of Fluid-Structure Interaction (FSI) (Comsol Multiphysics) to solve the EHL problem of two articulating surfaces.

The lubricant filled in the clearance between the femoral head and acetabular cup configurations in hip joint is synovial fluid. Synovial fluid is non-Newtonian fluid and has shear thinning characteristic, especially under relatively low shear rates. Nevertheless, at very high shear rates, as experienced in steady-state walking conditions, synovial fluid acts as a Newtonian fluid (26). Thus, the lubricant used in this study was considered to be a Newtonian and isoviscous in nature (27) with viscosity of 0.001 Pa.s (28,29).

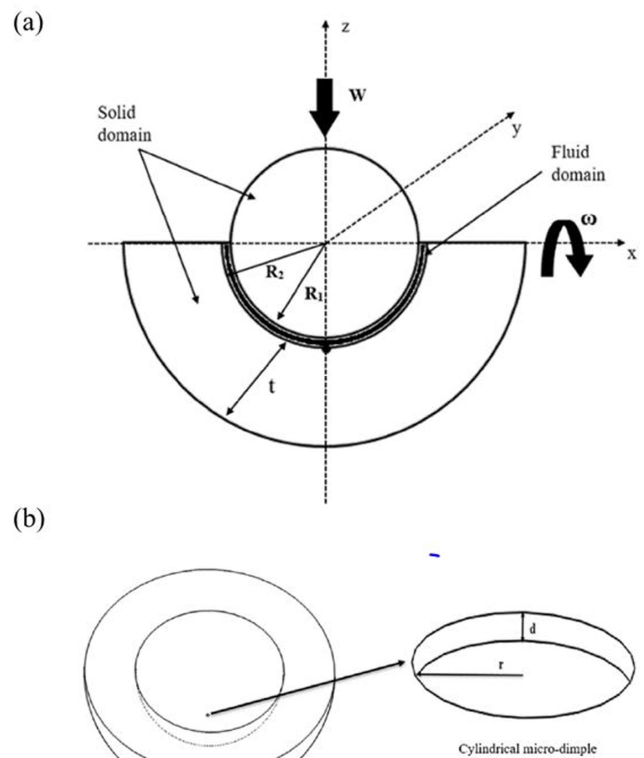

Figure 1: (a) Metallic hip implant model in EHL analysis (b) Cylindrical dimple profile on the centre of metallic acetabular cup surfaces

Geometry of surface dimple

A cylindrical micro-dimpled profile was constructed on the centre of the acetabular cup surfaces. Radius (r) of the micro-dimple was assumed to be 0.25mm while depth size (d) was considered to be 45µm to analyse the influence of micro-dimpled surfaces on EHL for metallic hip implants and the result was compared to non-dimpled surfaces. To assess the influence of dimple depth on EHL, micro-dimples with various depths (d) of 100µm, 45µm and 5µm were used in the present study. The illustration of micro-dimpled surfaces is shown in Fig. 1(b).

Mesh convergence analysis of FSI simulation indicated different sensitivity with different size of mesh elements (Fig. 2(a)). It was carried out to ascertain the accuracy of simulation. From Fig. 3, the selected number of mesh element was 80,440 elements. The elements were assumed to discretize all regions, including solid and fluid domain (70,628 elements of solid and 9,812 elements of fluid) and are displayed in (Fig. 2(b)). The elements of fluid domain (FSI boundary) were smaller than the solid domain. Thus, a higher mesh density on the FSI boundary benefits to attain a more accurate answer. The mesh sensitivity of elements was carried out to ascertain the accuracy of simulation. The accuracy of

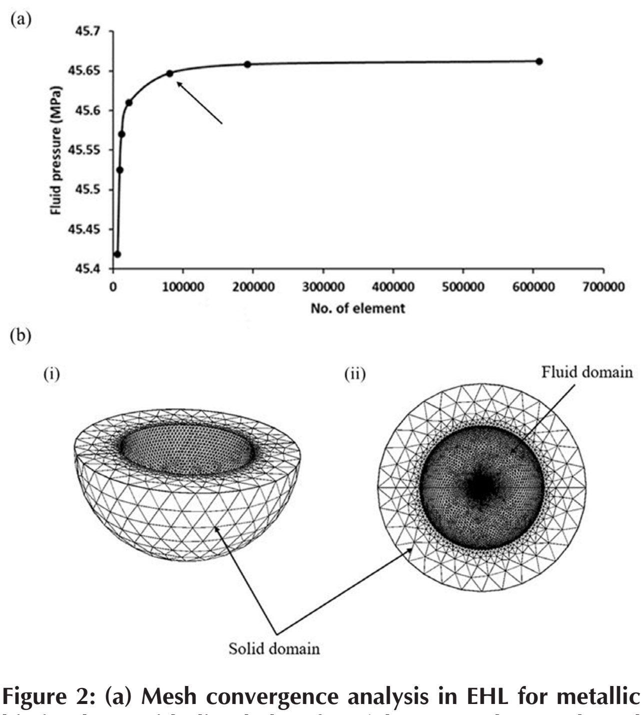

Figure 2: (a) Mesh convergence analysis in EHL for metallic hip implants with dimpled surface (The arrow shows selected the number of elements) (b) EHL model with 80,440 elements involving solid and fluid region in different view (i) isometric view (ii) top view

simulation based on percentage error was less than 5%.

Loading and motion conditions

In the simulation, flexion-extension (FE) motion, ω of 2 rad/s was considered around the y-axis of hip joint in normal walking under steady-state condition (18,25). The position of acetabular cup was considered to be placed horizontally under the average load, w of 1500N applied in vertical direction (Fig. 1a). The outer surface of the acetabular cup was rigidly fixed. The initial lubricant pressure on the edge of the acetabular cup was assumed to be zero.

RESULTS

Comparison between non-dimpled and micro-dimpled surfaces

Fig. 3(a) and 3(b) shows the lubricant pressure and thickness for metallic hip implants with non-dimpled and micro-dimpled surfaces in EHL under steady-state condition. As observed, the lubricant pressure for metallic hip implants with micro-dimpled surfaces were lower compared to that of non-dimpled surfaces. The maximum lubricant pressure for micro-dimpled surfaces was 45.772MPa while for non-dimpled surfaces was 49.549MPa. However, the lubricant thickness for micro-dimpled surfaces increased on contact point, namely by 6%, compared to non-dimpled surfaces.

Lubricant Behavior of Micro-Dimpled Surfaces in Elastohydrodynamic Lubrication (EHL)

The contour profile of micro-lubricant velocity and lubricant flow for non-dimpled and micro-dimpled surfaces (depth=45µm) are illustrated (Fig. 4). It shows the behaviour of micro-lubricant with micro-dimpled surfaces at the centre of contact zone under steady-state condition of EHL for metallic hip implants. The lubricant velocity for micro-dimpled surfaces was lower, 28.87µm/s compared to non-dimpled surfaces, which demonstrated a velocity of 93.84µm/s. In Fig. 4(b), the lubricant flowed toward the centre of micro-dimpled surfaces.

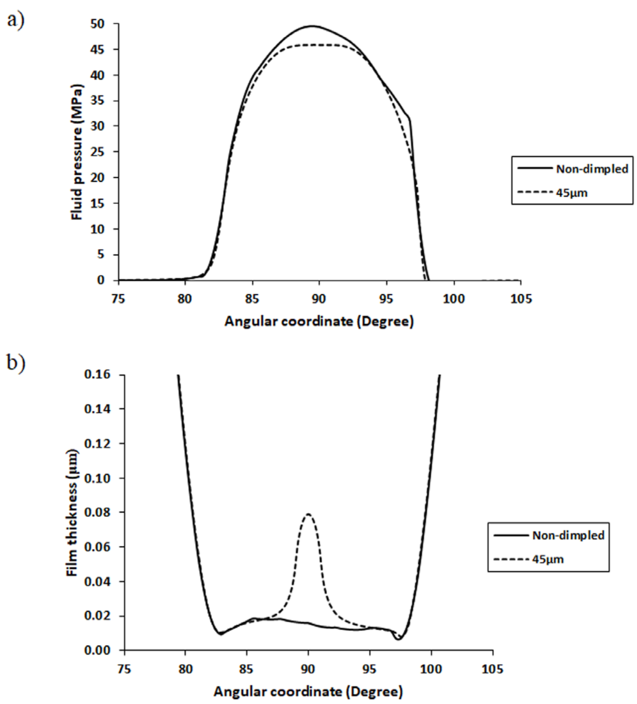

Figure 3: Comparison between non-dimpled and micro-dimpled surfaces for metallic hip implants in EHL (a) Lubricant pressure (b) Lubricant thickness

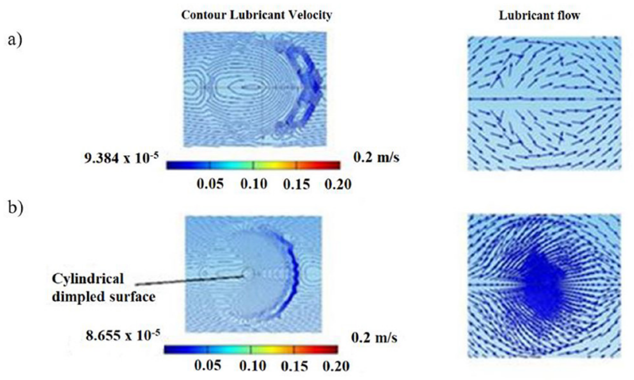

Figure 4: Contour profile of micro-lubricant velocity for (a) non-dimple surface (b) micro-dimple surface (depth = 45µm)

Effect of dimple depth on lubricant thickness

The lubricant pressure and thickness of micro-dimpled surfaces with various dimple depth of 100µm, 45µm and 5µm for metallic hip implants were compared Fig. 5(a) and Fig. 5(b), respectively. It was demonstrated that the lubricant pressure for a shallow dimple (depth = 5µm) was lower compared to a dimple depth of 100µm and 45µm. Likewise, the lubricant thickness also displayed similar results. It was observed that the shallow dimples

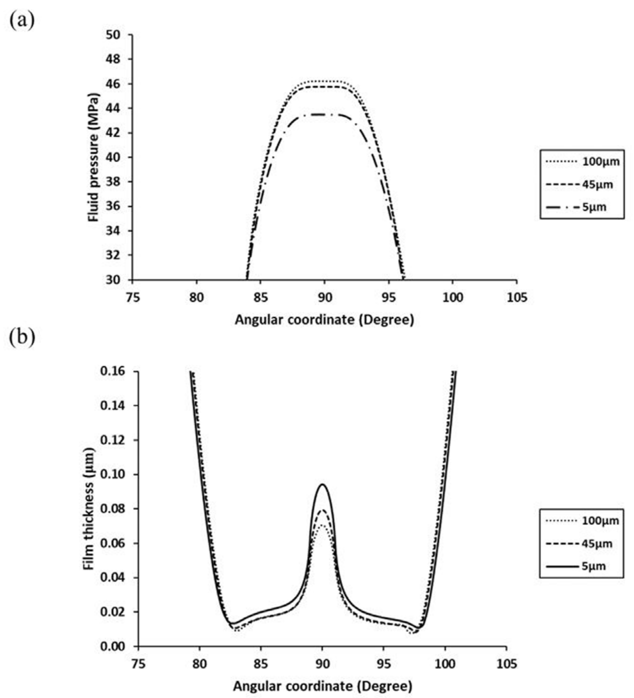

Figure 5: (a) Lubricant pressure and (b) Lubricant thickness with micro-dimple surface of 100µm, 45µm and 5µm

(depth = 5μm) provided increment of lubricant thickness, which declined as the dimple depth increased (30–32). The lubricant thickness for shallow dimples (depth = 5μm) was 0.094μm compared to deep dimples (depth = 100μm), which demonstrated lubricant thickness of 0.071μm.

Fig. 6 shows the deformation of acetabular cup between a non-dimpled surface and three surfaces with different depths of micro-dimples. It was observed that the shallow dimple (depth = 5μm) had increased the deformation size at the centre of the acetabular cup's surface by 21% compared to non-dimpled surfaces, and by 8% compared to surfaces with micro-dimple depth of 45μm.

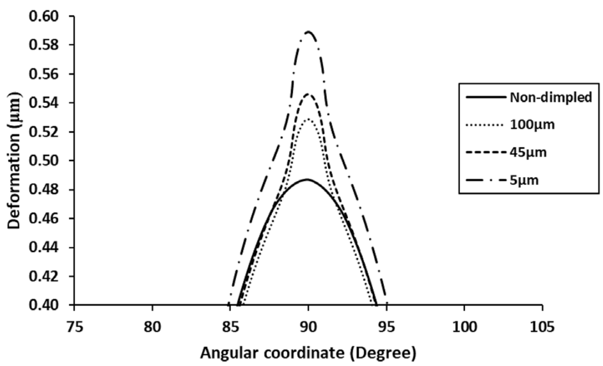

Figure 6: Deformation of solid structure with micro-dimpled surfaces of various depths and non-dimpled surfaces at the centre of metal acetabular cup for metallic hip implants

DISCUSSION

Comparison between non-dimpled and micro-dimpled surfaces

The presence of micro-dimple on the surfaces of acetabular cups can produce lower lubricant pressure, and thereby enhance lubricant thickness. This occurrence was mostly due to the lubricant trapped within micro-dimple (29,33). The volume of lubricant trapped inside the micro-dimple was sufficient to completely fill the gap between contact surfaces of bearing and thus, provide protection from direct contact.

Furthermore, micro-dimpled surfaces acted as secondary lubrication (23,34). Fundamentally, in EHL with micro-dimpled surfaces, the lubricant thickness between the femoral head and the micro-dimpled acetabular cup forms reservoirs to sustain lubricant. This reservoir acts as an amplifier of lubrication. When the surface of femoral head squeezes and slides against the micro-dimpled surface of acetabular cup, the pressure and friction shear deforms reservoirs elastically. This phenomenon generates pressure to squeeze lubricant out from the micro-dimple and consequently, resist the contact between the two lubricated bearing surfaces (Fig. 7). Therefore, the use of a micro-dimpled surface for metallic hip implants in EHL has significant positive

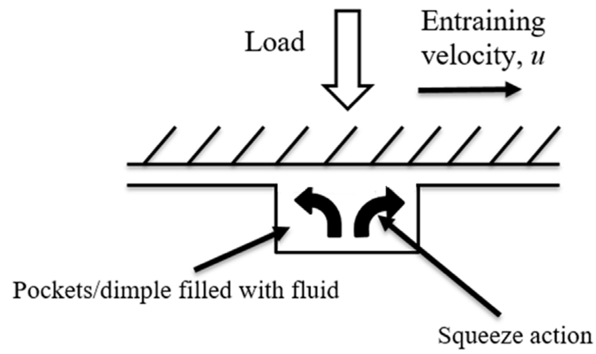

Figure 7: Secondary lubrication effect of micro-dimpled surfaces under steady-state condition

value, especially in increasing lubricant thickness to prevent direct contact between two lubricated surfaces.

Lubricant Behavior of Micro-Dimpled Surfaces in Elastohydrodynamic Lubrication (EHL)

In Fig. 4(b), the velocity of lubricant for micro-dimpled surfaces was low based on low entraining velocity, which is experienced in walking cycle. Consequently, lubricant flow through the micro-dimpled surfaces became very slow compared to non-dimpled surfaces. Furthermore, the shape of micro-lubricant velocity contour profile at the centre of contact zone for both surfaces appeared differently under steady-state condition. The micro-lubricant velocity contour profile for micro-dimpled surfaces resembled the shape of a horseshoe while that of non-dimple surfaces resembled a hexagon. The significant difference was due to the trapped lubricant inside the micro-dimple, which precedes or fall behind that dimple and elastically deforms the contact surfaces; thus forming a horseshoe shape (35). Furthermore, the lubricant on micro-dimpled surfaces flowed toward the centre of dimple due to the lower pressure inside the dimple compared to the outside.

Effect of dimple depth on lubricant thickness

Based on Fig. 5, the shallow dimples (depth = 5μm) provided increment of lubricant thickness because it had positive effect on lubrication efficiency compared to the surface of deep dimples. This occurs due to disturbance in distribution of lubricant thickness by the movement of a micro-dimple in sliding motion (31). When the lubricant entered the contact zone area, rapid enhancement of lubricant thickness occurs near to the shallow micro-dimple area under the effect of squeeze and hydrodynamic motion.

In Fig. 6, the deformation of the acetabular cup with deep micro-dimple depth generated higher lubricant pressure compared to that of 45μm and 5μm dimple depth. However, this pressure was incapable of forcing more the acetabular cup surfaces. This occurred due to lubricant of high pressure attempting to thrust lubricant out from inside the dimple and consequently, this pressure was inadequate to force the deformation of acetabular cup surface. The lubricant trapped inside the micro-dimple caused deformation on the acetabular cup surfaces within the micro-dimple. Therefore, the solid structure of acetabular cup surface decreased as the depth of the dimple increased.

There were some limitations to the current study. For instance, the effect of dimple diameter on lubricant thickness was not considered in the current study. Furthermore, the current study did not cover the transient motion in EHL. Moreover, only micro-dimples of cylindrical shape was considered in the current study. Therefore, various shapes of micro-dimples must be considered in further studies to investigate the suitable micro-dimple shape capable of increasing lubricant thickness optimally for metallic hip implants in EHL using FSI method.

CONCLUSION

The study filled the gap in identifying the influence of micro-dimple depth on lubricant thickness in EHL for metallic hip implants using Fluid-Structure Interaction (FSI) approach. The application of micro-dimpled surfaces is an effective approach to improve tribological performances, especially in increasing the lubricant thickness. Steady-state condition was applied in this study with vertical load (w) and rotation velocity around y-axes (ω) representing the average load and flexion-extension (FE) velocity of hip joint in normal walking. The results indicated that metallic hip implants with micro-dimpled surfaces enhanced lubricant thickness compared to non-dimpled surfaces. The following conclusions can be taken from this study:

a) In comparison between non-dimpled and dimpled surfaces, the dimpled surface was provided the highest lubricant thickness by 6% compared to non-dimpled surface.

b) Increment of lubricant thickness for micro-dimpled surfaces was due to entrapment of lubricant within the micro-dimple. The lubricant trapped inside the micro-dimple was sufficient to support high load.

c) Furthermore, micro-dimpled surfaces acted as secondary lubrication. This effect generated pressure to squeeze lubricant out from the micro-dimple and consequently prevented contact between two lubricated bearing surfaces.

d) The shape of micro-lubricant velocity contour profile for micro-dimpled surfaces resembled a horseshoe due to the lubricant trapped within the micro-dimple, which precedes or fall behind that dimple and elastically deforms the contact surfaces.

e) In investigating the influence of dimple depth on lubricant thickness, it was observed that the shallow dimple (depth = 5μm) increased lubricant thickness, which declined as the dimple depth increased.

f) The presence of micro-dimple had facilitated the deformation of acetabular cup surface to be higher than non-dimpled surface.

g) The shallow dimple (depth = 5μm) had increased the deformation size at the centre of the acetabular cup's surface by 21% compared to non-dimpled surfaces, and by 8% compared to surfaces with micro-dimple depth of 45μm.

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REFERENCES

1. Gao LM, Meng QE, Liu F, Fisher J, Jin ZM. The effect of aspherical geometry and surface texturing on the elastohydrodynamic lubrication of metal-on-metal hip prostheses under physiological loading and motions. Proc Inst Mech Eng Part C J Mech Eng Sci. 2010;224(12):2627–36.
2. Ingham E, Fisher J. Biological reactions to wear debris in total joint replacement. Proc Inst Mech Eng H. 1999;214.
3. Savarino L, Granchi D, Ciapetti G, Cenni E, Nardi Pantoli A, Rotini R, et al. Ion release in patients with metal-on-metal hip bearings in total joint replacement: A comparison with metal-on-polyethylene bearings. J Biomed Mater Res. 2002;63(5):467–74.
4. Delaunay C, Petit I, Learmonth ID, Oger P, Vendittoli PA. Metal-on-metal bearings total hip arthroplasty: The cobalt and chromium ions release concern. Orthop Traumatol Surg Res [Internet]. 2010;96(8):894–904. Available from: http://dx.doi.org/10.1016/j.otsr.2010.05.008
5. Drummond J, Tran P, Fary C. Metal-on-Metal Hip Arthroplasty: A Review of Adverse Reactions and Patient Management. J Funct Biomater [Internet]. 2015;6:486–99. Available from: http://www.mdpi.com/2079-4983/6/3/486/
6. Reza H, Hosseinzadeh S, Eajazi A, Shahi AS. The Bearing Surfaces in Total Hip Arthroplasty – Options, Material Characteristics and Selection. Recent Adv Arthroplast. 2012;163–210.
7. Uddin MS, Zhang LC. Predicting the wear of hard-on-hard hip joint prostheses. Wear. 2013;301(1–2):192–200.
8. Mattei L, Di Puccio F, Piccigallo B, Ciulli E. Lubrication and wear modelling of artificial hip joints: A review. Tribol Int. 2011;44(5):532–49.
9. Tiainen L, Abreu P, Buciumeanu M, Silva F, Gasik M, Serna R, et al. Novel laser surface texturing for improved primary stability of titanium implants. J Mech Behav Biomed Mater [Internet]. 2019;98(May):26–39. Available from: https://doi.org/10.1016/j.jmbbm.2019.04.052
10. Kaur S, Ghadirinejad K, Oskouei RH. An Overview on the Tribological Performance of Titanium Alloys with Surface Modifications for Biomedical

Applications. Lubricants. 2019;(V).

11. Galda L, Sep J, Olszewski A, Zochowski T. Experimental investigation into surface texture effect on journal bearings performance. Tribiology Int [Internet]. 2019;136(March):372–84. Available from: https://doi.org/10.1016/j.triboint.2019.03.073
12. Ibatan T, Uddin M, Chowdhury M. Recent development on surface texturing in enhancing tribological performance of bearing sliders. Surf Coat Technol. 2015;
13. Gupta KK, Kumar R, Kumar H, Sharma M. Study on Effect of Surface Texture on the Performance of Hydrodynamic Journal Bearing. Int J Eng Adv Technol. 2013;(31):2249–8958.
14. Zhang B, Huang W, Wang X. Biomimetic surface design for ultrahigh molecular weight polyethylene to improve the tribological properties. 2015;226(8):705–13.
15. Choudhury D, Walker R, Roy T, Paul S, Mootanah R. Performance of honed surface profiles to artificial hip joints: An experimental investigation. Int J Precis Eng Manuf. 2013;
16. Choudhury D, Walker R, Shirvani A, Mootanah R. The Influence of Honed Surfaces on Metal-on-Metal Hip Joints. Tribol Online [Internet]. 2013;8(3):195–202. Available from: http://jlc.jst.go.jp/DN/JST.JSTAGE/trol/8.195?lang=en&from=CrossRef&type=abstract
17. Zhang H, Qin LG, Hua M, Dong GN, Chin KS. A tribological study of the petaloid surface texturing for Co-Cr-Mo alloy artificial joints. Appl Surf Sci. 2015;
18. Gao L, Yang P, Dymond I, Fisher J, Jin Z. Effect of surface texturing on the elastohydrodynamic lubrication analysis of metal-on-metal hip implants. Tribol Int. 2010;43(10):1851–60.
19. Meng Q, Liu F, Fisher J, Jin Z. Contact mechanics and lubrication analyses of ceramic-on-metal total hip replacements. Tribol Int. 2013;63:51–60.
20. Stewart TD. Tribology of artificial joints. Orthop Trauma [Internet]. 2010 [cited 2017 Dec 27];24:435–40. Available from: http://www.orthopaedicsandtraumajournal.co.uk/article/S1877-1327(10)00100-4/pdf
21. Syahrullail S, Sapawe N, Razak MD, Azli Y. Effect of Surface Modification of Acetabular Cup with Embedded Micro-Pits on Friction Properties. Am J Mech Eng. 2014;2(5):125–9.
22. Al-Hajjar M, Fisher J, Williams S, Tipper JL, Jennings LM. Effect of femoral head size on the wear of metal on metal bearings in total hip replacements under adverse edge-loading conditions. J Biomed Mater Res - Part B Appl Biomater. 2013;
23. Zhang H, Qin LG, Hua M, Dong GN, Chin KS. A tribological study of the petaloid surface texturing for Co-Cr-Mo alloy artificial joints. Appl Surf Sci. 2015;332:557–64.
24. Gao L, Wang F, Yang P, Jin Z. Effect of 3D physiological loading and motion on elastohydrodynamic lubrication of metal-on-metal total hip replacements. Med Eng Phys. 2009;31(6):720–9.
25. Meng Q, Gao L, Liu F, Yang P, Fisher J, Jin Z. Contact mechanics and elastohydrodynamic lubrication in a novel metal-on-metal hip implant with an aspherical bearing surface. J Biomech. 2010;43(5):849–57.
26. Jin ZM, Dowson D, Fisher J. Analysis of fluid film lubrication in artificial hip joint replacements with surfaces of high elastic modulus. Proc Inst Mech Eng Part H-Journal Eng Med. 1997;211:247–56.
27. Jin ZM, Dowson D. A full numerical analysis of hydrodynamic lubrication in artificial hip joint replacements constructed from hard materials. Proc Inst Mech Eng Part C J Mech Eng Sci. 1999;213(4):355–70.
28. Gao LM, Meng QE, Wang FC, Yang PR, Jin ZM. Numerical solutions for the elastic deformation of spherical bearing surfaces of metal-on-metalhip joint implants. Proc Inst Mech Eng Part J J Eng Tribol. 2010;224(8):797–805.
29. Gao L, Yang P, Dymond I, Fisher J, Jin Z. Effect of surface texturing on the elastohydrodynamic lubrication analysis of metal-on-metal hip implants. In: Tribology International. 2010.
30. Mourier L, Mazuyer D, Ninove FP, Lubrecht AA. Lubrication mechanisms with laser-surface-textured surfaces in elastohydrodynamic regime. Proc Inst Mech Eng Part J J Eng Tribol. 2010;224(8):697–711.
31. Mourier L, Mazuyer D, Lubrecht a. a., Donnet C. Transient increase of film thickness in micro-textured EHL contacts. Tribol Int. 2006;39(12):1745–56.
32. Galda L, Sep J, Prucnal S. The effect of dimples geometry in the sliding surface on the tribological properties under starved lubrication conditions. Tribol Int. 2016;99:77–84.
33. Choudhury D, Rebenda D, Sasaki S, Hekrle P, Vrbka M, Zou M. Enhanced lubricant film formation through micro-dimpled hard-on-hard artificial hip joint: An in-situ observation of dimple shape effects. J Mech Behav Biomed Mater. 2018;81.
34. Taee M, Torabi A, Akbarzadeh S, Khonsari MM, Badrossamay M. On the Performance of EHL Contacts with Textured Surfaces. Tribol Lett. 2017;65(3).
35. Krupka I, Hartl M. Effect of surface texturing on very thin film EHD lubricated contacts. Tribol Trans. 2009;52(1):21–8.