Effects of femoral head diameter on the predicted acetabular cup wear volume of CP-Ti and UHMWPE hip implants

Handoko1* Suyitno2 Rini Dharmastiti1 Rahadyan Magetsari2
1Department of Mechanical and Industrial Engineering, Faculty of Engineering, Universitas Gadjah Mada, Indonesia
2Department of Orthopedics and Traumatology, Sardjito General Hospital and Faculty of Medicine, Universitas Gadjah Mada, Indonesia

*Corresponding author (E-mail): handoko.dtm@ugm.ac.id

Abstract. Numerical investigation is a method capable to study the expensive and long-time duration wear problems in biotribology. The technique has been used to predict the wear of hip implants specially those made from metal on polymer biomaterials. This research studied the effect of commercially pure titanium (CP-Ti) femoral head diameter on the wear volume of ultra-high molecular weight polyethylene (UHMWPE) acetabular cup. Diameter variations are 22 mm, 28 mm and 32 mm. Hip implant models were created and the numerical approximations conducted with the finite element method. Wear calculations used a nonlinear load and contact area mathematical model. A Python script was developed to proceed the calculations. Accurate wear predictions can be obtained with the use of fine elements those maximum contact loads are closer to the physiological load. Verification with experimental data on the 32 mm femoral head model shows a good agreement with error rates below 1%. 

1. Introduction
Total hip replacement (THR) surgery is a treatment conducted on patients with severe hip arthritis. There have been a growing number of this medical operation in many countries following the increased life expectancy. Although commercial hip implants are able to perform up to three decades, recent trends demand more. Nowadays, patients undergo THR are younger than generations before [1]. Attempts to obtain a longer life service should be the primary purpose of implant designs. Parameters strongly correlate to wear must be considered carefully. Those are material properties of the implants, dimensional aspects such as diameter and clearance and surface roughness due to processing.

Various metal, polymer or ceramic biomaterials have been used to make hip implants. One of the most popular is the metallic femoral head and polymeric acetabular cup [2]. The metals commonly used are cobalt chrome alloy, Ti6Al4V titanium alloy and 316L stainless steel. The polymer is ultra-high molecular weight polyethylene (UHMWPE). This biomaterial pair must deal with wear problem more than others. Excessive volumetric wear of the polymer leads to implant loosening and failures [3].

The most reliable way to ensure the performance of a material to resist wear is to test it experimentally. From this point, problems arise. Realistic biotribological tests should be able to mimic the in vivo conditions inside the body of patients. Cyclic tests are usually conducted up to several million
cycles duration at low frequency, i.e. one cycle per second [4] [5]. The wear tests are expensive and demand long time durations to deliver the required data. These obstacles can be reduced with computation and simulation techniques. A widely used and commercially available numerical method is the finite element analysis. This method is well perform to simulate the foundation of wear phenomena i.e. contact mechanics. Carefully determined applied loads and boundary conditions would obtain contact pressure, contact area and sliding distance of every nodes in a computed domain. These contact mechanic parameters are the inputs for the wear calculations. The results are useful for lifetime prediction of the implants without extensive experimental wear tests.

Wear prediction of the UHMWPE in metal on polymer implants are mostly use the Archard wear equations. Initially developed for wear abrasion between metals, many advancements on this model has been proved that it can be used to model wear in polymers [6]. Otherwise, several studies proved and suggested that it is not suitable for polymer materials [7]. The arguments are based on the wear behavior of polymer which has a nonlinear correlation with contact load [8]. Further development was made [9]. The result was a nonlinear load and contact area wear model developed for UHMWPE. This new model has a better curve fitting with experimental UHMWPE wear data in hip simulators. The aim of this study is to apply that model for a specific case in hip implants. It is the effects of femoral head diameter on wear of UHMWPE cup. This case arose due to the body size variations from different race and gender of the patients. The femoral head material studied and modelled is the commercially pure titanium (CP-Ti). This material is highly biocompatible. Unlike other popular metallic biomaterials, CP-Ti is expected to release nontoxic metal ions and safer for lifetime services.

2. Materials and Methods

Archard equation is a simple mathematical model to calculate sliding wear of the materials. It states that wear volume ($\Delta V$) is linearly proportional to normal load ($W$) and sliding distance ($s$) as follow:

$$\Delta V = k.W.s$$  (1)

The factor $k$ is a wear factor to accommodate the probability of a portion of materials at contact area to displace and formed into debris. Obtained experimentally, the factor is also represent other physical magnitudes those contribute to wear such as temperature and sliding speed. However, there is another wear model developed for polymer especially on the hip simulator [9]. This model is based on the assumption that wear is proportional to the product of frictional work and sliding path. It can be expressed independent from the sliding distance as follows:

$$\Delta V = K W_{\max}^{2/3} A^{1/3} N$$  (2)

with $A$ and $N$ are the contact area and number of cycles respectively. The proportionality constant $K$ is a new wear factor for circular motion paths as indeed exists in the experimental hip simulators. Equation (2) is the wear model, a mathematical basis used to predict UHMWPE polymer wear in this study.

Biaxial rocking motion (BRM) hip simulator is a machine capable to mimic the wear of hip implants [10]. Although its rotary system is not fully replicate the hip joint mechanism, wear data from the tested specimens are close to the in vivo data. Experimental wear test for the CP-Ti and untreated UHMWPE hip implants with BRM was conducted by [11]. The similar setup and geometry models were created for computational works in this research as shown on Fig 1. Three femoral head size variations with diameter from 32 mm, 28 mm up to 22 mm were studied. Radial clearance between femoral head and acetabular cup radius was set to 100 microns similar to the other published works [12] [13]. Abaqus finite element software was used to create the models and proceed the explicit dynamic contact mechanic computations. Discretization of the models performed with linear hexahedral (HL) meshes. Mesh sensitivity tests performed with element size variations of cup models from 1 mm up to 0.2 mm at the contact surfaces. The femoral head element size was set fix equal to 1 mm. The outputs i.e., $W$, contact pressure ($p$), contact area ($A$) and von Mises stress ($S$) were then extracted from computational databases for the wear calculations and physical interpretations. A flow chart in Fig 2 described the steps. Custom scripts written in Python scientific programming language were created to automate the processes.
Mechanical properties of the studied biomaterials are shown in Table 1. The softer material, UHMWPE is the worn material and its surface acted as a slave surface. The biomaterial elastoplastic
properties were set in material definition section of the finite element software. All of these numerical input data were recalculated physical dimensions consistency, i.e. length, area, force, pressure and stress.

Table 1. Mechanical properties related to contact mechanics

| Material   | Density in g/cm³ | Modulus of elasticity, E (GPa) | Poisson ratio, v | Contact mechanic setup       |
|------------|------------------|--------------------------------|------------------|-------------------------------|
| CP-Ti      | 4.5              | 110                            | 0.34             | Rigid body, master surface    |
| UHMWPE     | 0.9              | 0.8                            | 0.46             | Deformable, slave surface     |

In this study, a series of experimental wear tests for the CP-Ti and UHMWPE biomaterials was performed with a three station pin on disc tribotester. The purpose of the tests is to obtain friction coefficient data needed by Abaqus contact mechanics algorithm and the factor $K$ for equation (2). Wet polished 50 mm diameter cast CP-Ti discs with surface roughness of 0.08 micron and 9 mm diameter UHMWPE pins were prepared for the tests. The speed of the tests was set at a cycle of one Hertz multidirectional sliding. Load cells were mounted to measure the normal and friction forces. The tests used contact pressure setups of 2 MPa and 3 MPa to acquire wear similar to clinical data [14]. The lubricant was a simulated body fluid (SBF) with controlled temperatures at 37 ± 2 °C to mimic the in vivo condition inside patient’s body [15]. SBF lubricant consists of bovine serum and distilled water. It is capable to simulate the behavior of human body fluid. Experimental results from one million cycle tests are the coefficient of friction equal to 0.2 and $K = 8.17 \times 10^{-8}$ mm$^{(7/3)}$ N$^{(2/3)}$. After all of the experimental tribotests and computational works were completed, the numerical wear prediction result was verified with experimental wear data from [11].

3. Results and discussion

Models with smaller element size have larger number of elements (Fig 3) and higher computational costs expressed in CPU time (Fig. 4). All of the femoral head diameter variations (22, 28 and 32 mm) have similar trends. The CPU time increased exponentially for element size below 0.3 mm. The use of small elements reduce $L_{min}$ (smallest element dimension in the mesh) and increase mesh densities.

![Figure 3](image1.png) **Figure 3.** Relation between element size and mesh density.

![Figure 4](image2.png) **Figure 4.** Effect of element size on CPU time.

Stable time increments ($\Delta t$) and the number of increments ($n$) will be affected as shown in the equations below [16]:

\[
\Delta t = \frac{t}{n}
\]
\[ \Delta t \approx L_{\text{min}} \sqrt{\frac{\rho}{\lambda + 2\mu}} \]  
(3)

with \( \rho \) is the density of UHMWPE, \( \lambda \) and \( \mu \) are the effective Lamé constants. If \( \Delta t \) is constant, the number of increments needed to complete a \( T \) simulated time period is:

\[ n = \frac{T}{\Delta t} \]  
(4)

Small \( \Delta t \) and high \( n \) lead to expensive computational costs. In general, \( \Delta t \) will not remain constant. The time increment can be rewritten based on the element by element stability as follow:

\[ \Delta t \leq \min \left( L_e \sqrt{\frac{\rho}{\lambda + 2\mu}} \right) \]  
(5)

The parameter \( L_e \) is a characteristic length associated with an element. Applied contact loads distort the elements so then change the \( L_e \) and \( \Delta t \). Small distorted elements would have faces with even smaller \( L_e \) and \( \Delta t \). The number of increments will arise exponentially and so does the computational costs.

Although the thickness of all cups were kept equal, larger diameter models still have bigger volume with more elements to calculate simultaneously. At the same element size, the mesh density and CPU time are higher than the smaller diameter models.

Femoral head diameter affected the calculated contact pressure as shown in Fig 5. Larger diameter models have lower maximum contact pressures. This relation can be explained with the calculated contact area (Fig 6). The models have larger contact areas during mechanical contact between femoral head and acetabular cup compared to the smaller ones. The large contact areas with the same physiological load would have lower contact pressures. On the other numerical output, maximum von Mises stresses have a similar trend as in the contact pressure (Fig 7). Von Mises stress is directly proportional to the deviatoric stress tensor and hydrostatic stress which also proportional to pressure. Hip implants with smaller femoral head diameter have higher contact pressures and von Mises stresses. The stress magnitudes of all variations are below the yield stress of UHMWPE [12]. This is an indication that the materials should not have local high stress zones which could cause sub surface fatigue and severe or excessive wear. The limit of UHMWPE fatigue failure is 32 MPa [17].

**Figure 5.** Maximum contact pressures on various studied femoral head diameter.

**Figure 6.** Predicted contact area with 0.3 mm element size.
Trends shown in Fig 3 up to Fig 7 have no indications of any convergence limit to guide the UHMWPE acetabular cup wear volume prediction. Fortunately, the nonlinear model of equation (2) demands the maximum load data which is easier to obtain thru the load cycle curves (Fig 9). Acetabular cup models with coarse elements have maximum contact loads higher than the physiological load value (1.96 kN) from [11]. Verification with experimental data of the 32 mm femoral head (90.7 mm$^3$/million cycle) was successfully obtained with the 0.3 mm element size as follows:

| Element size (mm) | Max. load (kN) | Max. load error from gait cycle (%) | Predicted wear volume (mm$^3$/million cycles) | Wear volume error rate (%) |
|-------------------|----------------|-----------------------------------|-----------------------------------------------|---------------------------|
| 1.0               | 2.440          | 24.46                             | 126.387                                       | 39.35                     |
| 0.5               | 1.868          | 4.71                              | 96.355                                        | 6.23                      |
| 0.3               | 2.007          | 2.34                              | 91.207                                        | 0.56                      |

Figure 7. Maximum von Mises stresses on the studied femoral head diameter variations.

Figure 8. Predicted contact area with optimized element size for the wear volume predictions.

Figure 9. Contact load compared with physiological load up to 0.3 mm element size.

Figure 10. Calculated contact load at various element size closer to the physiological load.
This finding was then used to select element size for the other models with diameter of 28 mm and 22 mm. The selected size should deliver accurate predicted wear volume. Computation results show that the 0.3 mm element size is not suitable for the 28 mm and 22 mm diameter models (Fig 9). The maximum contact loads are too high. Numerical study was then continued to test the 0.2 mm element size. It solved the problem (Fig 10). The calculated maximum loads are closer to the physiological load even at the expense of much higher computational costs (Fig 4). Contact area were updated (Fig 8) and the wear volume for these two models were calculated. The results are:

![Figure 11. Predicted wear volume on various femoral head diameter.](image)

According to [18], the wear volume difference of the cups with diameter between 22 mm to 32 mm are not significant. However, the difference of wear volume between tested models in this study cannot be concluded as less significant (Fig 11). Recent review on the improved wear resistant UHMWPE reported total volume losses below 20 mm$^3$ per million cycles [19]. It is as much as the wear volume difference between the 22 mm and 32 mm cup diameters. Other volumetric wear study based on clinical reports of the 4-15 years follow up implants suggested an optimal low wear group of 22 mm and 28 mm femoral head diameters [20]. The group distinguishes the wear rate between small and large diameter hip implants.

Large diameter femoral heads produce more volumetric wear. However, these designs have several advancements. The range of motion is larger so then capable to support better hip mobility. The lower contact pressure should theoretically means less wear penetration depth ($h$). Smaller diameters with less contact area might experience deeper penetration from the stiff metallic femoral head to the softer UHMWPE cup. These advantages and disadvantages lead researchers to seek balance. There was a study tried to determine an ideal size with least amount of wear volume and penetration depth [21]. The result suggests that a 28 mm hip diameter is the ideal one. This result had been criticized due to the other clinical finding on head size variations [22]. Reduced cup thickness on larger head size and plastic deformation of the polymers encounter the theoretical assumption. Linear penetration depths increased as head size became larger.

4. Conclusion

Large femoral head metal on polymer hip implants have lower contact pressures. At the same physiological load, the contact areas are larger than the smaller ones. This relation influences on the predicted wear volume with a nonlinear load and contact area wear model. Larger diameters tend to have higher UHMWPE cup volumetric wear. Accurate wear prediction obtained with the use of fine element sizes those have calculated contact loads closer to the physiological load. For the 22 mm up to 32 mm femoral head diameter, it was achieved with the 0.2 mm and 0.3 mm element size. The error is below 1 % on the 32 mm diameter femoral head validated with experimental data.
References
[1] Bergmann G, Bender A, Dymke J, Duda G and Damm P 2016 Standardized Loads Acting in Hip Implants PLoS ONE, 11, 1–23
[2] Bozic K J, Lau E C, Ong K L, Vail T P, Rubash H E and Berry D 2012 Comparative Effectiveness of Metal-On-Metal and Metal-On-Polyethylene Bearings in Medicare Total Hip Arthroplasty Patients The Journal of Arthroplasty 27 8 37–40
[3] Shanbhag A, Rubash H E and Jacobs J J 2006 Joint Replacement and Bone Resorption (New York: Taylor & Francis)
[4] Cheng G and Shan X 2012 Dynamics analysis of a parallel hip joint simulator with four degree of freedoms Nonlinear Dynamics 70 2475–86
[5] Jourdan F and Samida A 2009 An implicit numerical method for wear modeling applied to a hip joint prosthesis problem Comput. Methods Appl. Mech. Eng. 198 2209–17
[6] Mattei L., Puccio F D and Cuilli E 2013 A comparative study of wear laws for soft-on-hard hip implants using a mathematical wear model Tribology International 63 66–77
[7] Wang A, Essner A and Klein R 2001 Effect of contact stress on friction and wear of ultra-high molecular weight polyethylene in total hip replacement Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine 215 133–9
[8] Wang A, Sun D, Stark C and Dumbleton, J 1995 Wear mechanisms of UHMWPE in total joint replacements Wear 181 241–9
[9] Wang A, Lee R, Herrera L and Korduba L 2013 Modeling and verification of ultra-high molecular weight polyethylene wear in multi-directional sliding Wear 301 162–16
[10] Saikko V 2008 Friction Measurement in the Biaxial Rocking Motion Hip Joint Simulator Journal of Tribology 131 011201–8
[11] Röstlund T, Albretksson B, Albretksson T and McKellop H 1989 Wear of ion-implanted pure titanium against UHMWPE Biomaterials Biomaterials 10 176–80
[12] Monif M M 2012 Finite element study on the predicted equivalent stresses in the artificial hip joint Journal of Biomedical Science and Engineering 5 44
[13] Puccio F D and Mattei L 2015 Biotribology of artificial hip joints World J Orthop. 6 77–94
[14] Saikko V 2006 Effect of contact pressure on wear and friction of ultra-high molecular weight polyethylene in multidirectional sliding Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine 220 723–31
[15] Affatato S, Spinelli M, Zavalloni M, Mazzega-Fabbro C and Viceconti M 2008 Tribology and total hip joint replacement: Current concepts in mechanical simulation Medical Engineering & Physics 30 1305–17
[16] Abaqus Inc. 2014 Abaqus Analysis User’s Guide 6.3.3: Explicit dynamics analysis
[17] Khosravipour I 2015 Contact stress analysis of Surface guided knee implant using finite element modeling (University of Manitoba)
[18] Gandhe A and Grover M 2008 Head size, does it matter? Current Orthopaedics 22 155–64
[19] Baena J C, Wu J and Peng Z 2015 Wear Performance of UHMWPE and Reinforced UHMWPE Composites in Arthroplasty Applications: A Review Lubricants 3 413–36
[20] Oparaugo P C, Clarke I C, Malchau H and Herberts P 2001 Correlation of wear debris-induced osteolysis and revision with volumetric wear-rates of polyethylene: a survey of 8 reports in the literature Acta Orthop Scand. 72 (1) 22–8
[21] Livermore J, Ilstrup D and Morrey B 1990 Effect of femoral head size on wear of the polyethylene acetabular component J Bone Joint Surg Am 72 (4) 518–28
[22] Kabo J M, Gebhard J S, Loren G and Amstutz H C 1993 In vivo wear of polyethylene acetabular components J Bone Joint Surg Br 75 (2) 254–8