Wear Prediction of UHMWPE Acetabular Cup against CP Titanium Femoral Head in a Hip Joint Simulator

H Handoko1*, S Suyitno1, R Dharmastiti1, R Magetsari2

1 Department of Mechanical and Industrial Engineering, Faculty of Engineering, Universitas Gadjah Mada, Indonesia
2 Department of Orthopedics and Traumatology, Sardjito General Hospital and Faculty of Medicine, Universitas Gadjah Mada, Indonesia

*handoko.dtm@ugm.ac.id

Abstract. Ultra high molecular weight polyethylene (UHMWPE) is a biomaterial used for the human hip joint prosthesis bearings for decades. The main disadvantage is the wear resistant. The products should be in service in the lifetime of the patients. Tribological assessments by experiments and computations should be made to solve that problem. The aim of this research is to predict the UHMWPE acetabular cup wear volume sliding against commercially pure titanium (cp Ti) femoral head in a hip simulator. The cp Ti has the highest biocompatibility compared with other femoral head metals such as cobalt chrome, titanium alloy and 316L stainless steel. Computation processes use the Abaqus contact mechanics algorithm followed by a custom made Python script for the data extractions and Archard wear model calculations. Models were an assembly of the UHMWPE acetabular cup and 32 mm cp Ti femoral head in a biaxial rocking motion hip simulator. Sliding between the models was a 60 cycles per minute rotation. Calculations conducted for one million cycles cumulative polymer wear volume in five steps of 200,000 cycles. The geometries of the models were updated in every step. Experimental pin on disc tests conducted to obtain the wear factor and coefficient of friction. Those data support the input of computations. Wear volume verification use the experimental data from Röstlund. Results show a numerical error at 31.42 percents.

1. Introduction

Ultra High Molecular Weight Polyethylene (UHMWPE) is a polymer used as bearing materials of the hip joint implants for decades. It is biocompatible and durable for up to 30 years in clinical use [1]. The material is usually paired with other metallic biomaterials known as metal on polymer (MOP) hip implants. The metals are 316L stainless steels, cobalt chromium alloys and titanium alloys. A common problem with the use of MOPs is the wear of polymers. Excessive UHMWPE wear debris would cause osteolysis and inflammation at the tissues around the hip joint implants. In osteolysis, wear debris will trigger the autoimmune of the body to clean it. Macrophages will phagocyte the wear debris. The process absorbs bone minerals. It would release the inflammatory mediators and leads to the bone resorption [2]. Implant loosening, bone breakage and inflammation will follow. The effects are severe pain for the patients and the need for costly revision surgeries. Based on these facts, it is important to study the wear volume of the UHMWPE used as hip bearing materials.
Wear especially in hip implant is a complex phenomenon. The metallic femoral head implants slide along the bearing surface materials when the patient moves their hips. This repeated process combined with the pressure at the implants due to the bodyweight of the patients creates a combined adhesive abrasive wear. Many parameters are involved such as sliding velocity, mechanical properties of biomaterials, lubricant from the body fluid of the patients etc. It is also nearly impossible to accurately measure the volumetric wear in vivo. These problems had been solved with experiments in vitro either by pin on disc tribotesters or hip simulators. A simple Archard wear equation is commonly used to calculate the wear volume. The complexity of the parameters other than contact pressure and sliding distance is represented by a wear factor. The main disadvantage of this method is the time needed to obtain wear data. The tests were usually conducted at a low frequency of one Hertz to mimic the in vivo conditions. Wear data from two to five million cycles would take months to complete with procedures for example as described in [3].

Significant progress on the capabilities of computer hardware and software in the recent decades support the numerical computation methods. Researchers have been using it to study the wear of the hip implants ever since. Maxian et al [4] are the first to use Archard wear law for the hip implant models. The researchers use the finite element method to calculate the contact pressures at incremental sliding distances. Others followed the similar procedures to study the various situations of hip implant models. The most recent published works are the improved solution of hip implant contact problem [5], motion inputs [6], additional device such as hip resurfacing [7] and the type of biomaterial pairs [8]. The outputs of the computation are the wear prediction data. Most of the predicted data are in good agreements or fit well with the experimental data. The method requires lesser time duration to complete. A two days experimental test can be replaced with a two hours numerical computation [9].

Wear predictions on the MOP hip prosthesis biomaterials are dominated by the Co-Cr alloy and UHMWPE pairs. Other femoral implant material rarely studied is the commercially pure titanium (cp Ti). This metal is the most biocompatible material even compared to the titanium alloys. Co-Cr alloys and 316L stainless steels release Ni ions in vivo. This process would lead to inflammation problems for the sensitive patients. The famous Ti-6Al-4V titanium alloys were suspected to release a combined Al and V ions. It would cause diseases such as Alzheimer, neuropathy and osteomalacia in the long time period of clinical services [10]. Based on those comparisons, attempts can be made to use cp Ti as the femoral implant biomaterial. The aim of this research is to assess the wear volume of UHMWPE acetabular cup paired with cp Ti femoral head. The expectation was that cp Ti could reduce long term clinical effects on the THR (total hip replacement) patients. Numerical investigations predict the wear volume of the UHMWPE cups. The output poses the effectiveness of commonly used Archard wear model to mimic the wear of UHMWPE sliding against cp Ti. It also found out the appropriate pressure of a physiological hip load which contact pressure closes to the in vivo data.

2. Materials and methods
Wear volume prediction in this research relies on the Archard wear law. The law can be derived into a linear wear depth \((h)\) equation as follows:

\[
h = k \cdot p \cdot s
\]

The equation requires \(k\) as a wear factor, \(p\) is the contact pressure and \(s\) is the sliding distance. Wear factor, \(k\) is obtained with the biotribological experiments. Wear volume can be calculated by multiplying the linear wear depth with the contact area at every contact nodes of the models. The calculation of the contact mechanic parameters, i.e. contact pressure and contact area use a commercial finite element software, Abaqus. Its algorithm is mature and used by previous hip modeling and computation studies [11-14]. The hip joint model used was a 46° total excursion biaxial hip simulator model studied experimentally by [15]. Fig 1 shows the model schematically. UHMWPE cup wear volume calculations conducted in repeated steps as shown in Fig 3.
Figure 1. The hip simulator model [15]

Figure 2. Paul type physiological hip load [15]

Figure 3. The wear volume calculation steps
The calculations need a coefficient of friction data between the contact materials. It was obtained by experiments with a multi directional pin on disc tribotester. The biomaterials were 50 mm diameter cp Ti discs and 9 mm diameter UHMWPE pins. Contact pressure was set at 3 MPa. Specimens were lubricated with simulated body fluids (SBF). The SBF consists of mainly distilled water with 25 % v/v of bovine serum [16-17]. The temperature of the fluid in the wear test was set between 36 to 38 degrees centigrade to simulate the physiological human body temperature. Polymer weight reductions were measured at every 250,000 cycles.

3. Results and Discussion
The first step to study wear phenomena in the hip joint simulator is to evaluate the contact pressures. This numerical study use the Paul type physiological hip load curve (Fig 2) as experimentally by [15]. The curve has a peak pressure up to 3.5 MPa. It was evaluated with the static three dimensional contact mechanic computations. Contact pressures on the UHMWPE cup was calculated with Abaqus at three physiological pressure variations from 3 MPa close to the curve, 6 MPa and 9 MPa (Fig 4). Results show that the realistic physiological pressure is 3 MPa due to the contact pressure values. High contact pressure (CPRESS) at 42.5 MPa and 40.56 MPa were found from two contact nodes but the other 1869 nodes are below 34 MPa (Fig 5). These numbers were relatively too high compared to the measured in vivo hip joint contacts [18-19]. Other calculations suggest higher values for the peak hip contact pressure of 22.4 MPa at the mid stance of normal walking [20] up to 31.06 MPa [21]. Hence the Paul type physiological hip load curve is realistic, close to the recent results. It can be used for the loads on the hip implant wear prediction.

![Figure 4](image1.png)

**Figure 4.** Effect of physiological pressures on the contact pressures at the surface of the UHMWPE cup (denoted as CPRESS by Abaqus), (a) 3 MPa, (b) 6 MPa and (c) 9 MPa

![Figure 5](image2.png)

**Figure 5.** Frequency histogram of contact pressure distributions from the 3 MPa physiological pressure. The acetabular cup model has 1871 contact nodes.
The next step was the contact mechanics computation and wear volume calculations. Experimental wear tests conducted with a three stations multi directional tribotester to support necessary data for the numerical calculations. Results from a million cycles were the coefficient of friction equal to 0.2 and the wear factor, \( k = 1.657 \times 10^{-6} \text{ mm}^3 (\text{N.m})^{-1} \). The coefficient of friction was then fed into the contact mechanics computations. The results were stored in the Abaqus output databases. Contact pressure, contact area and sliding distance data extractions use a custom Python script. Linear wear depth \( h \) calculations use those data and the wear factor \( k \). The magnitude of \( h \) was used for the wear volume calculations and to update the contact nodes geometries. The result of numerical prediction was a volumetric wear data at the rate of 119.2 mm\(^3\) per million cycles. Experimental data from [15] was 90.7 mm\(^3\) UHMWPE wear per one million cycles. It was a 31.42 % numerical error.

![Figure 6](image1)

**Figure 6.** Contour of the acetabular cup wear volume after one million cycles, (a) overall acetabular plane, (b) detailed contour of the large worn area and (c) protuberance found at the surface of UHMWPE pin.
The cause of numerical error in this study is the use of a constant wear factor for the all contact nodes. Different contact pressures at specific contact points could have different wear rates. The points can be treated with different wear factor values. An equation of wear factor as a function of contact pressure is needed and should be obtained experimentally. The equation is available from [23-24] but it was for the UHMWPE and cobalt chromium pairs. The next possible factor affects to the wear calculation errors is the Archard wear model itself. Recent studies began to critically evaluate the use of the law. The polymer deformation remains elastic hence the linear relationship between the load and the contact area might not valid anymore [25]. The quest to modify or even replace Archard wear law for the metal on polymer or the soft on hard biomaterials will support the future numerical wear prediction. Sophisticated future numerical models will reduce the costly and time consuming biotribological experiments.

The contours of the worn acetabular cup (Fig 6a) are similar to the contours of the contact pressure. It is because of the use of Equation (1) relies numerically to the contact pressure values. Local and small pockets of worn surface indicate the potential of protuberance. Experimental pin on disc tests confirm that indication. Protuberance was found at the surface of the UHMWPE pin (Fig. 6c). Surface heterogeneities and micro defects or micro fissures formation lead protuberances in the space between contact surfaces. Elastic properties of the polymer and friction force determine the deformation of protuberances [22]. The repeated deformations due to sliding contact would lead to the plastic deformation and surface strain. From that point, the protuberances turn into wavelets. As the number of wavelets increased, the surface became more defective. Finally the wear becomes catastrophic.

4. Conclusions
This study predicted the wear of UHMWPE cups paired with cp Ti femoral heads in a hip joint simulator model. The results can be summarized as follows:

1. Paul type physiological load applied to the model obtained contact pressures close to the recent calculation results.
2. The numerical wear prediction error is 31.42% compared to the experimental data.
3. Accurate wear factor equations or wear models should be obtained to increase the accuracy of the wear predictions.

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