

# Load displacement simulation of CP-Ti/UHMWPE hip implant

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| Received                 | 15 October 2020  |
| Revised                  | 01 December 2020 |
| Accepted for publication | 15 December 2020 |
| Published                | 31 August 2021   |

**Abstract:** Hip implant made from titanium is special due to its lower modulus of elasticity to avoid stress shielding with the human bone. One type of load the material designed to withstand is a gradually increased compressive force which happened when the patient changes his/her position from sitting to standing. This study examined the capability of a metal on polymer (MOP) implant made from commercially pure titanium (CP-Ti) and ultra-high molecular weight polyethylene (UHMWPE) loaded up to 3 kN according to ISO 14242 standard. Two CP-Ti/UHMWPE MOP models with femoral diameter of 22 mm and 32 mm were simulated with finite element. The results expressed in load displacement curves were validated with compressive load experimental tests. Both materials are capable to withstand the load. Simulation data are in good agreement with the experiments.

**Keywords:** compressive load, simulation, finite element

## 1. Introduction

Lower limb orthopedic implant is a biomedical product designed to help patients with osteoarthritis to regain their productive life. This joint disease occurs in the hip, knee and ankle, could be caused by degenerative process and trauma by accidents. Patients with acute stage would experience severe pain that reduce their mobility or became immobile at all. This problem is solved when doctors performed surgeries to put the implants to replace the defective joints [1]. One of these implants is the hip implant. This product has two main parts, an acetabular cup to be placed at the acetabulum of pelvic bone and a stem positioned at the femur. The biomaterials most commonly used for the cup and stem are polymer, i.e. Ultra High Molecular Weight Polyethylene (UHMWPE) and metals such as stainless steel, cobalt chrome (CoCr) and titanium alloys. These materials must be capable to support human body load in patient's daily activities and biocompatible with the surrounding tissues.

Current commercially available implants are successful to be used from 10 up to 20 years [2-4]. However, several problems are still challenging to be solved. One of which is the load problem. Metallic biomaterials used for stem are too stiff compared to the thigh bone. The modulus of elasticity that is far higher than bone would cause a stress shielding problem [5]. When high load take place, the bone which is mechanically inferior to the metallic implant could crack or even break. This modulus mismatch problem [5-6] could arise specially in younger active patients. Recent trends show the increased number of young patients between 45 to 65 years old [7]. Stem material should ideally not only strong enough to withstand mechanical load but also has a modulus of elasticity closer to the bone. The purpose of this research is to study a low modulus of elasticity metal, i.e. pure titanium as an alternative for metallic stem material. Its modulus is about a half compared to the other stem biomaterials (Table 1). Low modulus

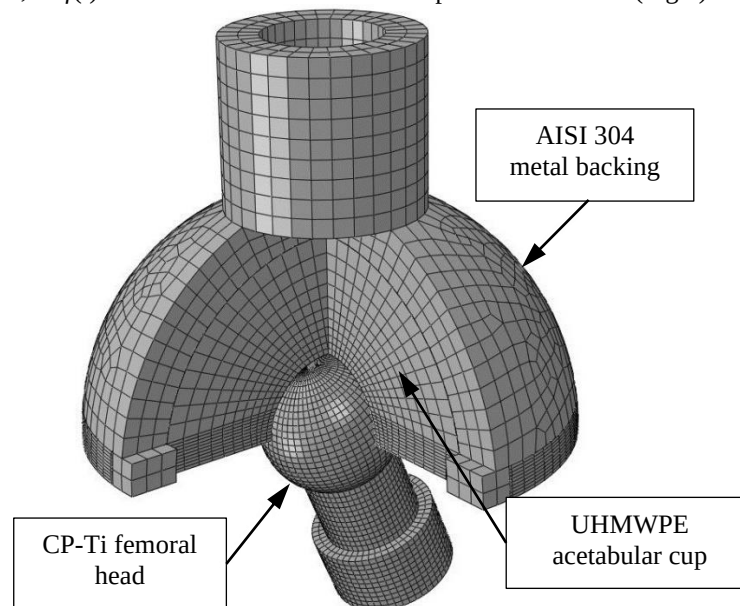
materials are able to transmit more physiological load to the femur, hence stress shielding would be avoided. These materials are also enable bone resorption into the femur [8]. Aside from that, titanium is also light and highly biocompatible. The released titanium ions to the surrounding tissues are not toxic as found in other metals which contain nickel such as stainless steel and CoCr alloys. The disadvantages are its less wear resistant and more expensive to fabricate. Hip implant computer model was created, computed, and verified with experimental data. The experiments performed to test the capability of implants to support mechanical load. Femoral head and cup components of the implants were made from cast commercially pure titanium (CP-Ti) and machined UHMWPE.

**Table 1.** Mechanical properties of metallic stems and human bone

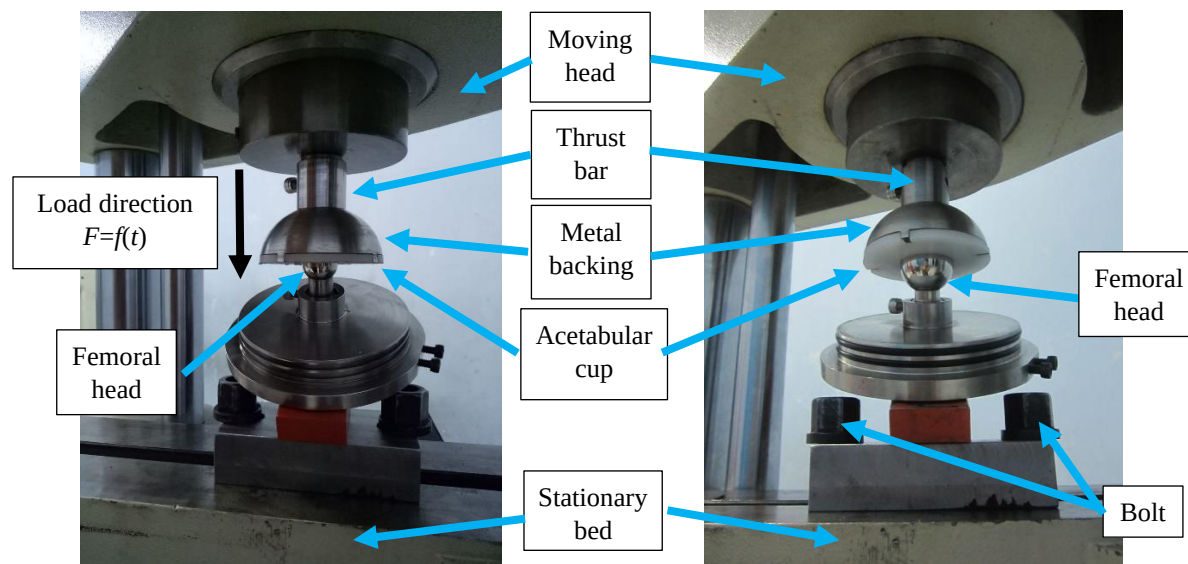
| Material   | Density in $\frac{gram}{cm^3}$ | Modulus of elasticity, $E$ (GPa) | Remark                                                                               |
|------------|--------------------------------|----------------------------------|--------------------------------------------------------------------------------------|
| AISI 316L  | 8 [9]                          | 200 [10-11]                      | Biocompatible but the nickel content is toxic.                                       |
| CoCr alloy | 9.2 [12]                       | 200-300 [8][10]                  | Biocompatible but it still contain nickel, although in less concentration than 316L. |
| CP-Ti      | 4,51 [9]                       | 110 [8][10]                      | Highly biocompatible but expensive and less wear resistant.                          |
| Bone       | 1.8-2.1 [12]                   | 1-30 [11][13]                    | Its mechanical properties depends on various factors such as age, gender and race.   |

## 2. Materials and methods

Load displacement modelling of the CP-Ti and UHMWPE hip implant were performed with two models, the 22 mm and 32 mm femoral head diameters as shown in Fig 1. The radial clearance on both models are similar. It is 100  $\mu m$  for those femoral size range [14-15]. These hip implant models were created and computed with Abaqus finite element software. The numerical results were then validated with experimental data. Fig 2 and Fig 3 show the experimental setup of the 22 mm and 32 mm femoral head size models in a universal testing machine. Femoral heads were set at a base 23° from horizontal surface as commonly used in hip simulators [16]. Compressive loads of approximately 3 kN according to ISO 14242-1 were applied gradually. The recorded force ( $F$ ) and time ( $t$ ) were then combined as dynamic load data,  $F=f(t)$ . These data were used as the input of simulations (Fig 4).

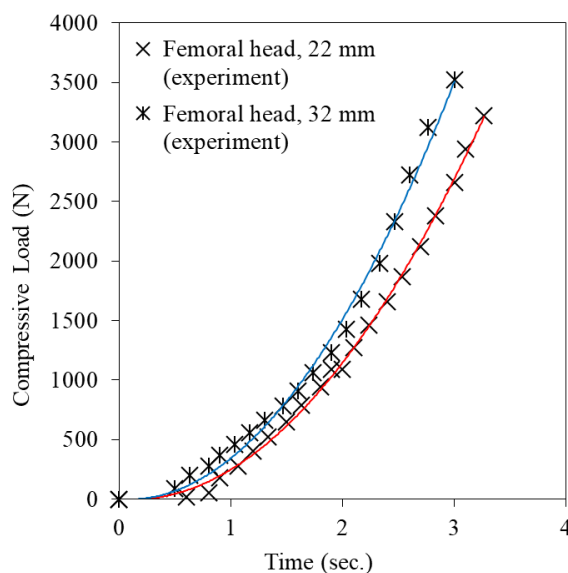


**Figure 1.** Physical model



**Figure 2.** Experimental setup for the 22 mm femoral head model.

**Figure 3.** Experimental setup for the 32 mm femoral head model.

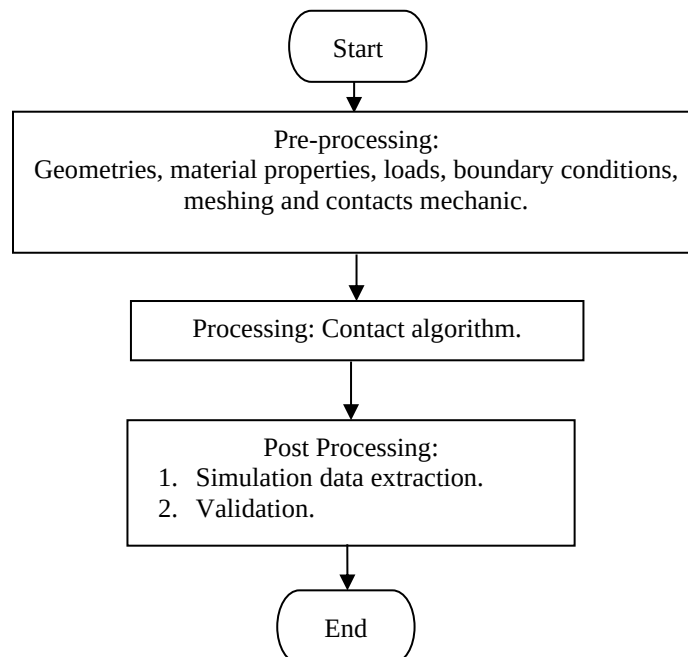


**Figure 4.** Dynamic load input data,  $F=f(t)$  for simulation.

Specimens were prepared as other hip implant studies. The femoral head material is a cast CP-Ti produced by investment casting method from a Japanese investment local industry, Itokoh Ceperindo in Klaten, Central Java, Indonesia. The purity of titanium was tested with XRF and the result is 99,7 %. Femoral head specimens were produced by finishing the cast products with a turning machine. This machining process was followed by wet polishing to meet the required maximum surface roughness of metallic femoral material [17]. On the other hand, the acetabular cup specimens were produced with the same turning process without polishing, in order to avoid adsorbed abrasive particles on the surface of polymer. Metal backing made from AISI 304 was used to support the acetabular cup. A set of three femoral head and acetabular cup pairs were tested to obtain experimental data for simulation validations.

Hip implant computer models were examined with a contact mechanic algorithm of the finite element software. Fig 5 shows the simulation steps. First, a preprocessing step was performed. Models were prepared by setting up

simulation inputs such as material properties, loads, boundary conditions (BCs), and contact mechanic. The material properties of all models are shown in Table 2. Further input data for the polymer, i.e. elastoplastic properties was also set. All metallic biomaterials model, i.e. femoral head and metal backing were determined as rigid bodies because their modulus of elasticity are far higher than polymer. Load inputs  $F=f(t)$  were taken from Fig 4 and applied in vertical direction, downward into the metal backing model. BCs were set properly with femoral component in a static position, pushed by acetabular cup with certain compressive load,  $F$ . The next input was contact mechanic setup. The master and slave surfaces are the metals and UHMWPE respectively. Normal behavior contact was set as hard contact. All models were then discretized with linear hexahedral meshes [18]. Femoral head element size was set 1 mm. Acetabular cup was treated gradually with larger element at the outer surface to the smaller element at the inner surface. Element size at this inner or contact surface was set 0.2 mm. The output of computation i.e., contact load,  $F$  and displacement were then extracted from the databases for further data analysis and interpretations. A custom script written in Python scientific programming language was created to automate the data extraction process.



**Figure 5.** Flow chart of simulation processes.

**Table 2.** Mechanical properties related to contact mechanics

| Component     | Material | $\rho \left( \frac{\text{gram}}{\text{cm}^3} \right)$ | $E \text{ (GPa)}$ | Poisson ratio, $\nu$ | Contact mechanic setup     |
|---------------|----------|-------------------------------------------------------|-------------------|----------------------|----------------------------|
| Femoral       | CP-Ti    | 4.5                                                   | 110               | 0.34                 | Rigid body, master surface |
| Cup           | UHMWPE   | 0.9                                                   | 0.8               | 0.46                 | Deformable, slave surface  |
| Metal Backing | AISI 304 | 8                                                     | 193               | 0.3                  | Rigid body                 |

### 3. Results and discussion

Computational and experimental results in this study are presented in Fig 6 and Fig 7. The experimental results are not calculated in average values because the recorded time from data acquisition are not at the same or constant intervals. Finite element contact algorithm calculated the displacements based on the compressive loads input. The combination of these data in form of  $F = f(d)$  where  $d$  is the displacement, are in good agreement with experimental data.

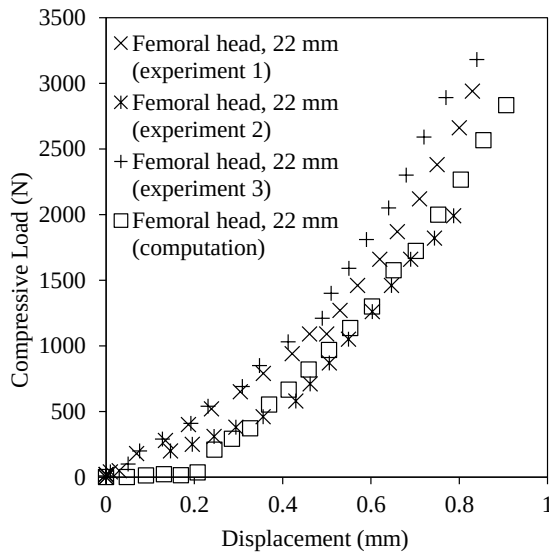


Figure 6. Load displacement relation of the 22 mm femoral head model.

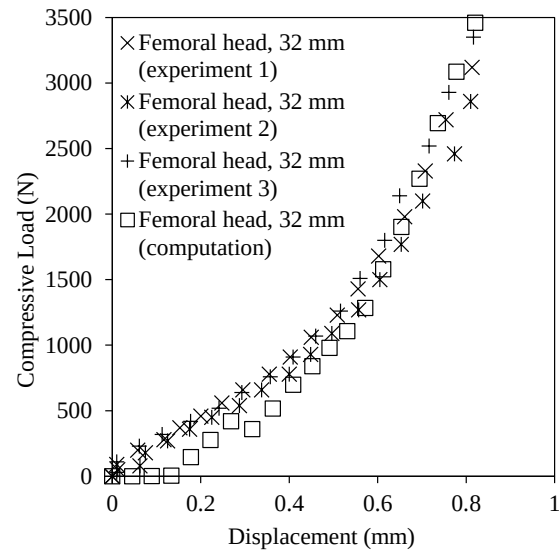


Figure 7. Load displacement relation of the 32 mm femoral head model.

The relationship between compressive load and displacement is nonlinear. This trend is similar to the experimental result of [19] on the test of CoCr and UHMWPE. Compressive force rises exponentially as the displacement increased. According to [20], the ratio between force ( $F$ ) and displacement ( $d$ ) is called stiffness of the contact ( $c$ ). Calculation results of  $c$  and made a chart of  $c = f(d)$  yield to a linear trend (Fig 8). By comparing Fig 6 and Fig 7 to Fig 8, it means that as  $d$  increase, so does the stiffness to resist contact. Further penetration is much harder and the force needed will be too high. The geometry of the femoral and acetabular cup mechanical contact is conformal. The contact state of this ball and socket conformal geometry is area contact [21]. It is different from the Hertz contact theory which assumes the state is a point contact. Finding the solution of a conformal contact is very difficult, therefore it is necessary to use the numerical method to solve this problem. The finite element contact mechanic algorithm is successfully modelled the contact phenomena between pure titanium and UHMWPE. The computed displacements are in good agreement with the experimental data.

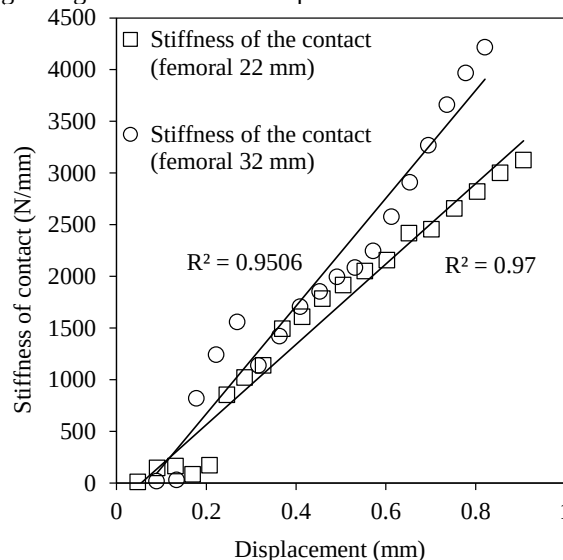


Figure 8. Stiffness of the contact.

Experimental results show that the pure titanium is capable to support a 3 kN ISO standard physiological load in the soft on hard mechanical contact setup. The position of implant components remains stable during and after the compressive normal force was applied through the metal backing. No visible damage marks were found at the surface of both titanium and UHMWPE materials.

Computational data could represent the contact mechanic situation between pure titanium femoral and UHMWPE cup. Nonlinear regressions (power law trend lines) of the data at Fig 6 and Fig 7 in form of equation  $F = a \cdot d^b$ , yield to  $b = 2.83$  ( $R^2 = 0.95$ ) and  $b = 3.13$  ( $R^2 = 0.89$ ) for the 22 mm and 32 mm femoral head models respectively. The magnitude of these exponents are higher than the Hertzian contact theory where  $b = 1.5$  [22]. It means that these contact mechanic situations are less stiff than Hertzian contact. The physical meaning is the contact areas are large and the Hertz assumptions are no longer hold. This would impact on the analysis of other phenomena, i.e. the wear of polymer. The most popular mathematical model of wear assumes that normal force ( $F_N$ ) is proportional to the contact area ( $A$ ) or  $F_N \propto A$ . This is true as long as the real contact area is small compared to the nominal contact area [23]. In the case of soft on hard contact such as this studied polymer and pure titanium pair, the deformation of UHMWPE on the surface asperities of the much harder metal yields a larger contact area. Hence the effect of this should be considered on the worn polymer models.

## 4. Conclusion

Pure titanium femoral component of hip implant is capable to withstand 3 kN mechanical load as suggested by ISO 14242-1. The contact mechanic between polymer cup and pure titanium femoral head can be modelled and simulated with finite element method. Results from the two models, i.e. 22 mm and 32 mm femoral diameter are in good agreement with experimental data. These contact mechanic situations are less stiff than theoretical Hertzian contact, hence the contact areas are large and the Hertz assumptions are no longer hold.

## Acknowledgements

Authors would like to thank to the supports of Laboratory of Computation, Production Process and Engineering Material, Department of Mechanical Engineering, Vocational School, Universitas Gadjah Mada and Engineering Material Laboratory, Department of Mechanical and Industrial Engineering, Faculty of Engineering, Universitas Gadjah Mada.

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