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# Numerical Convergence in Wear Volume Prediction of UHMWPE Acetabular Cup Paired with cp Ti Femoral Head Hip Implants

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**Abstract.** Wear is a problem for metal on polymer (MOP) hip implants to perform lifetime endurance. Polymer excessive volumetric loss leads to implant failures. Attempts to solve this problem are usually initiated with tribological tests. The method is time-consuming because the sliding speed is low. There is a faster way to use a computational method to gather wear data. This research aims to investigate the numerical convergence of predicted wear volume with the finite element method (FEM). The model is a commercially pure titanium (cp Ti) and ultra-high molecular weight polyethylene (UHMWPE) MOP hip implant. A dynamic Paul physiological load was applied to the model. Volumetric loss of the polymer was calculated with a wear equation involved nonlinear contact load and contact area. The inputs of calculation are wear factor and the computational contact mechanic performed by FEM. The wear factor was obtained by performing biotribological experiments with a multidirectional pin on disc tribotest. Predicted wear volume was validated with hip simulator experimental data from the literature. Convergences were found at the mesh density of 1.38 elements/mm$^3$. An acceptable numerical error was obtained in the model with 1 mm element size for femoral head and 0.3 mm for acetabular cup. This model was then used for the investigation of load increment effects. The result is that load increment variations do not affect wear volume and contact mechanic numerical outputs. The calculated stresses are below the UHMWPE yield stress limit. In this elastic region, the effects of strain rate caused by load increment are negligible.

## Introduction

Lower limb diseases hampered human productivities. The patients became less mobile or immobile depend on the degree of illness. In the past, older patients were dominant due to degenerative conditions. Nowadays, the population of younger patients significantly increased [1]. It is caused by obesity, extreme sports, and various accidents. This phenomenon raised concern about the survival of prostheses. The survival rate of patients over 70 years old is high. It is 95 % for the total hip arthroplasty (THA) and total knee arthroplasty (TKA) [2]. Meanwhile, the rate is lower for younger patients. Values from various sources vary, from 63 % up to 87 % in 10 years after implantations [3-4]. Projections show that in the year 2030, 52 % primary THA and 55 % primary TKA will be implanted to the younger patients less than 65 years old [5]. This is an engineering problem to produce durable implants. The main obstacle is the wear of biomaterials. Engineers must assess the tribological performance of implant products with various wear tests. Setups and procedures have been developed to mimic the in vivo conditions. These tests are expensive and take a long time duration. The frequency of sliding is low, only one Hertz and the number of cycles in several million performed.

Progress of the desktop microprocessor computing power and related hardware has enabled researchers to develop computational methods for various physical problem simulations. These methods supported the need to gather scientific data faster than experiments, especially in biotribology. On hip implants, computation with the finite element method to predict the wear of polyethylene acetabular cup was initiated by [6]. This work inspired others to simulate various implant designs [7-8] and biomaterials such as the popular metal on polymer (MOP) [9-10] and metal on metal (MOM) [11-12]. These attempts obtained accurate results validated with experimental data. The computations need mathematical models capable of describing the relation between input variables and desired results. Most of the wear predictions employed Archard equation as described below:

$$\Delta V = k.W.L.N \tag{1}$$

with $\Delta V$, $k$, $W$, $L$, and $N$ are the volumetric loss, wear factor, applied load, sliding distance, and the number of cycles respectively. The factor $k$ is useful to accommodate the probability of a portion of materials displace and became wear debris at the contact area. This factor also represents the contributions of other physical magnitudes in tribology, such as temperature, sliding speed, and the behavior of lubricants. In MOP hip implant, the use of Eq. (1) had been criticized in several studies because it does not fit with the polymer wear behavior. There is another research that tried to formulate a suitable wear model for polymer even though it is currently limited for the hip simulator model [13]. The model uses the assumption that polymer wear is proportional to frictional work and sliding distance. Analytical formulation resulted in the equation independent of sliding distance as follows:

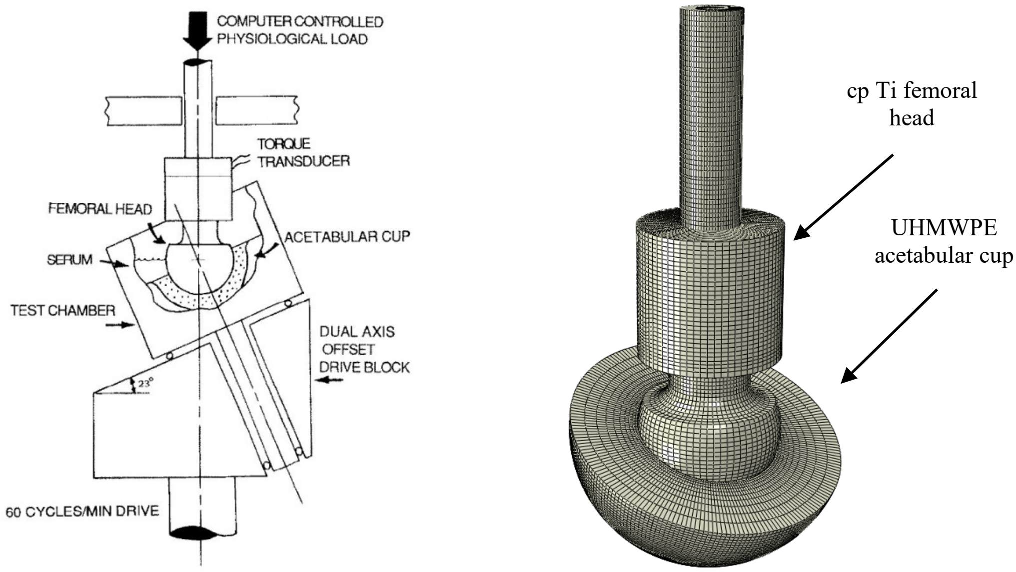
$$\Delta V = K.W^{2/3}.A^{1/3}.N \tag{2}$$

with $K$ is a proportionality constant and A is the contact area. The constant $K$ is a new wear factor applicable for circular motion paths of the hip simulator. Eq. (2) is a model that describes a nonlinear relation between wear volume, contact load, and contact area. Experimental verifications showed a good agreement with the calculated wear using this new model [14].

The purpose of this research is to investigate the numerical convergence and examine the results of computational works using Eq. (2). The model is an assembly of MOP hip implants in an inverted biaxial rocking motion (BRM) hip simulator adapted from [15]. Commercially pure titanium (cp Ti) and ultra-high molecular weight polyethylene (UHMWPE) are the biomaterials of the implants. Titanium and its alloys are rarely studied. The use of cp Ti in this study is to support the search for less toxic biomaterials. The commonly used metallic hip implants contain nickel, which is toxic to the human body [16-18]. Pure titanium has excellent biocompatibility. Its fracture toughness is high. The modulus of elasticity is low, closer to the bone stiffness than any other metals [19-20]. The UHMWPE is not harmful, but high volumetric wear induced osteolysis and implant failures. The quantity is described by [21].

## Materials and Method

Numerical investigations performed for a 32 mm cp Ti/UHMWPE hip implant assembled within a BRM hip simulator [15] with an explicit dynamic finite element method (FEM). This tribo pair used cp Ti and UHMWPE material models for the femoral head and acetabular cup (Fig. 1). The applied load to the femoral head is a simplified Paul's physiological load of a normal human walking gait cycle (Fig. 2). The work station is a quad-core Intel i7 with 16 GB random access memory. Validations proceeded by comparing the predicted results with experimental data.




(a) Experiment schematic [15] (b) CAE computation model

**Figure 1.** The inverted BRM hip simulator experiment and computation model with a 32 mm femoral head diameter.

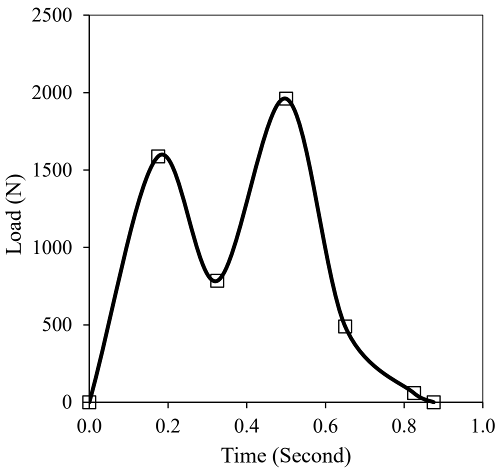



**Figure 2.** Simplified Paul's physiological load applied to the model [15].

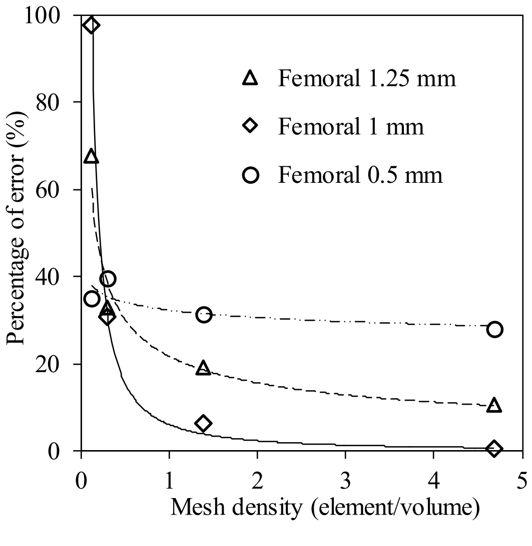
All computational models used a linear hexahedral element type. Its performance to achieve numerical convergence and accuracy is better than tetrahedral [22-23]. This linear element is also suitable for contact mechanic simulation [23]. There is a suggestion to avoid the use of elements with higher order. These elements have interpolated shape functions. Interpolations would cause nodal force oscillations between mid-side and corner nodes [24]. The acetabular cup was meshed with an automatic seeding from the outer to the inner perimeters. Computational contact mechanic setup in FEM is a surface to surface finite sliding penalty contact. Hard material (titanium) surface is the master surface. Softer material (polymer) surface is the slave surface. The femoral head model was constrained as a rigid body due to its much higher modulus of elasticity (110 GPa) compared to the polymer (0.8 GPa). The angular sliding velocity is $2\pi$ radian/second, equal to the common value of one Hertz in biotribology. Boundary conditions restrict any irrelevant linear and rotational movement.

The convergence of computational results was investigated with mesh size and load increment variations. The sensitivity analysis tested four acetabular cup element sizes, from the coarse 1.5 mm, 1 mm, 0.5 mm up to the finest 0.3 mm, and three for the femoral head model, from 1.25 mm, 1 mm down to 0.5 mm. These are the mesh sizes at contacting surfaces. The 1.25 mm is the largest element size applicable for a 32 mm femoral head diameter without any poor mesh element. A validated model with the smallest numerical error was then used for the convergence studies of load increments. Three load increment variations of 200 N, 400 N, and 2000 N similar to the magnitudes used by [24] were tested.

Equation (2) was developed specifically for the UHMWPE acetabular cup wear volume calculation in a hip simulator [14]. Therefore the term "wear" in this study is meant to the polymer. These calculations required wear factor data. The value was obtained by undertaking experimental wear tests with a multidirectional pin on disc tribotester. These tests used UHMWPE pins and cp Ti discs similar to the material models in computational setups. Other tribological parameters such as sliding speed, lubricant, and temperature were set close to the in vivo human body conditions. The detailed method was described by [25]. These experiments produced another important data, the coefficient of friction required by penalty contact computations. It is 0.2 for the cp Ti/UHMWPE tribo pair.

**Results and Discussion**

Experimental data of acetabular cup volumetric wear is 90.7 mm$^3$ in a million cycles [15]. This value validates numerical results of all mesh variations. Figure 3 (a) presents the percentage of wear volume numerical errors. These errors depend on the mesh density. Models with femoral mesh sizes of 1 mm and 1.25 mm have similar trends, convergence at a mesh density of 1.38 elements/mm$^3$. The femoral mesh size of 0.5 mm has more stable results. The errors are nearly independent of mesh density, but the values are unacceptable. Two models of the 1 mm femoral mesh size have acceptable low numerical errors at 6.23 % (mesh density equal to 1.38 elements/mm$^3$) and 0.56 % (mesh density at 4.68 elements/mm$^3$). The acetabular cup mesh sizes of those models are 0.5 mm and 0.3 mm. These accurate results can be obtained when the curve started to convergence. However, attempts to simulate wear volume should consider the computational costs. Figure 3 (b) shows linear relations between mesh densities and computational costs, represented with the central processing unit (CPU) time for all studied femoral heads. This graph showed that the CPU time of 0.3 mm acetabular cup is four times higher than the 0.5 mm. It means that the time to achieve less than 5% accuracy is too high. A recent similar method to relate numerical accuracy with computation time was conducted by [26] in biomedical research. The result showed a linear relationship between computation time and the number of elements. Biomechanical modeling is a powerful tool, but embedding into the clinical reality has to be cost-effective [27].



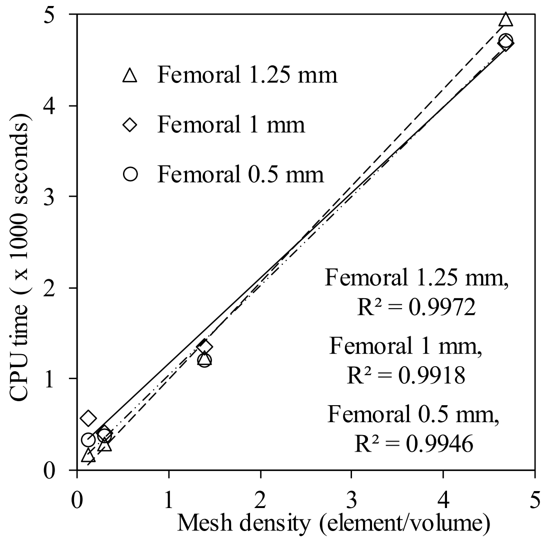




**Figure 3.** The convergence of predicted volumetric wear errors (a) and the relation with computational cost (b).

Modeling technique with femoral head and acetabular cup element size variations affect the contact mechanic results (Fig. 4). Once again, the 0.5 mm femoral models have more stable results on the maximum contact load, contact area, and contact pressure. The maximum contact load of other femoral models (Fig. 4 (a)) converges similar to the trend of volumetric wear numerical errors (Fig. 3). Meanwhile, convergence points are not seen clearly on the maximum contact area and contact pressure results (Fig 4 (b) and Fig. 4 (c)). Judgment can be made based on the main purpose of the study, i.e., volumetric wear, so the convergence point is the same as mentioned above.

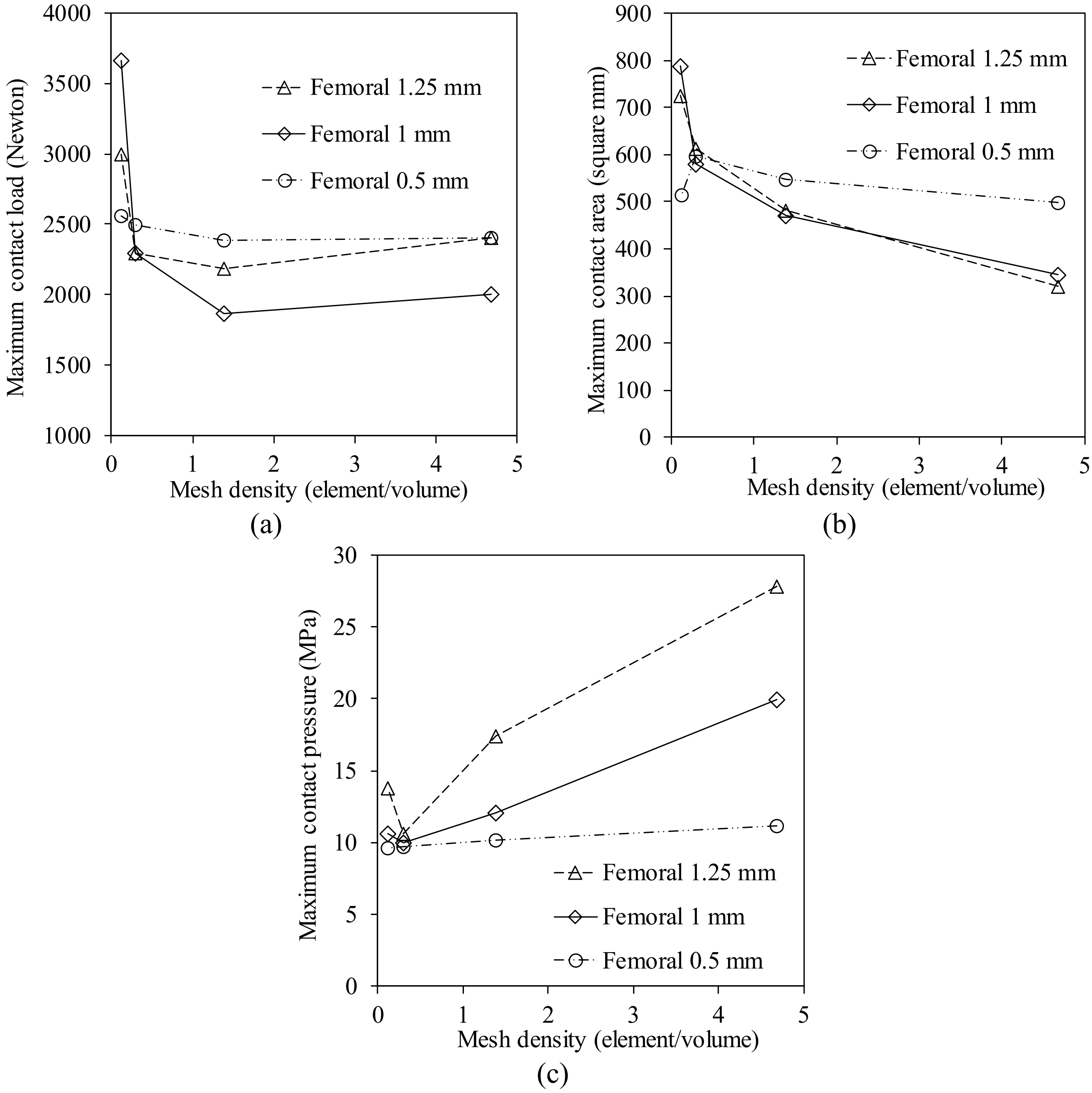



**Figure 4.** Maximum values of (a) contact load, (b) contact area, and (c) contact pressure, at various element sizes.

The dynamic of all contact loads as the function of time is presented in Fig. 5. The curves in Fig. 5 (a) show the incorrect results of 0.5 mm femoral model at 0.5 second. Although this model is numerically ideal, quasi-independent from the mesh density, the calculated contact loads are not equal to physiological load (Fig. 2). Peak loads around 2500 N are equal to 20 % error. The other 1 mm femoral models deliver more accurate results at the fine acetabular cup meshes (Fig. 5 (b)). It does not apply for a very coarse 1.5 mm mesh in the acetabular cup. This mesh size caused wildly unstable contact load numerical values. According to [26], both problems in Fig. 5 (a) and the 1.5 mm coarse acetabular cups of Fig, 5 (b), (c) may be caused by the coarse mesh of the slave surface, which enables gross penetrations of the master surface into it. The numerical solution became inaccurate. Meanwhile, the problem of 1.25 mm femoral model (Fig. 5 (c)) seems similar, but it arose from the femoral head coarse mesh. The slave nodes could get snagged around the sharp vertices [28]. The solution for the acetabular cup and femoral head coarse elements is to refine their respective surfaces.

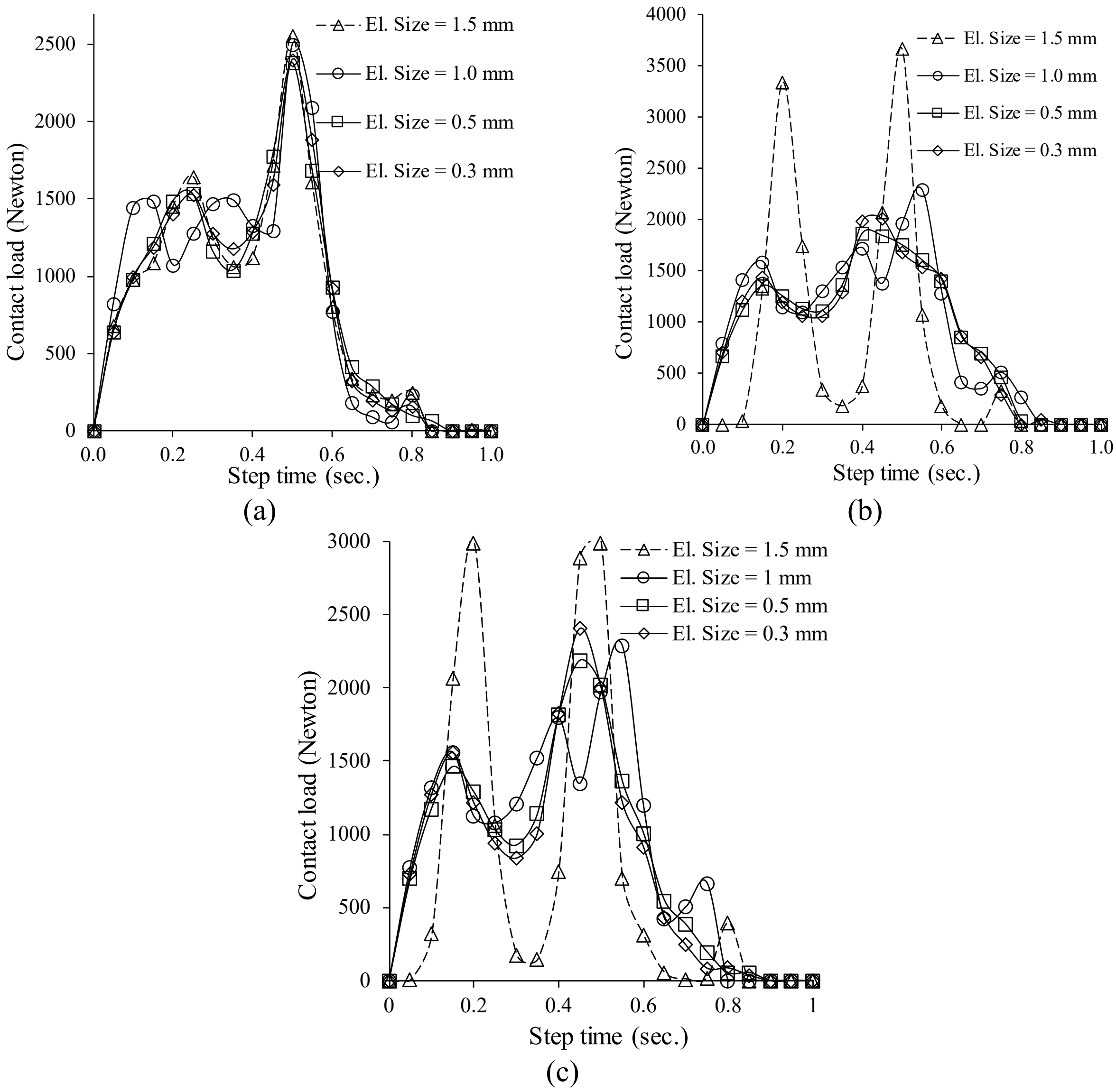



**Figure 5.** Contact loads on a femoral diameter of: (a) 0.5 mm, (b) 1 mm and (c) 1.25 mm.

The dynamics of contact area (Fig. 6) and contact pressure (Fig. 7) are similar to contact load when anticipating acetabular cup element size variations (from 1.5 mm down to 0.3 mm). However, the magnitudes are different. Femoral head model with a 0.5 mm fine mesh penetrated the acetabular cup deeper (Fig. 6 (a)) than the other two (Fig. 6 (b) and (c)). The calculated contact areas are higher. Consequently, the contact pressures are lower (Fig. 7 (a)). Based on the contact load miscalculations in Fig. 5 (a), these results are doubtful. The other 1.25 mm femoral model has curves of contact area (Fig. 6 (c)) similar to the 1 mm femoral model (Fig. 6 (b)). With its previous contact load miscalculations (Fig. 5 (c)), this 1.25 mm femoral model also has doubtful results, especially on the contact pressures (Fig. 7 (c)). Experimental contact mechanic data from [15] are not available, while the numerical results vary. Hence, the contact area and contact pressure correct value determinations are based on the model with accurate predicted wear volume. It is the 1 mm femoral model. Its contact load is correct, fit with physiological load, and the 0.3 mm acetabular cup variation has an acceptable volumetric wear loss prediction. Therefore the correct contact area and contact pressure are the curves of 0.3 mm acetabular cup in Fig. 6 (b) and Fig. 7 (b).

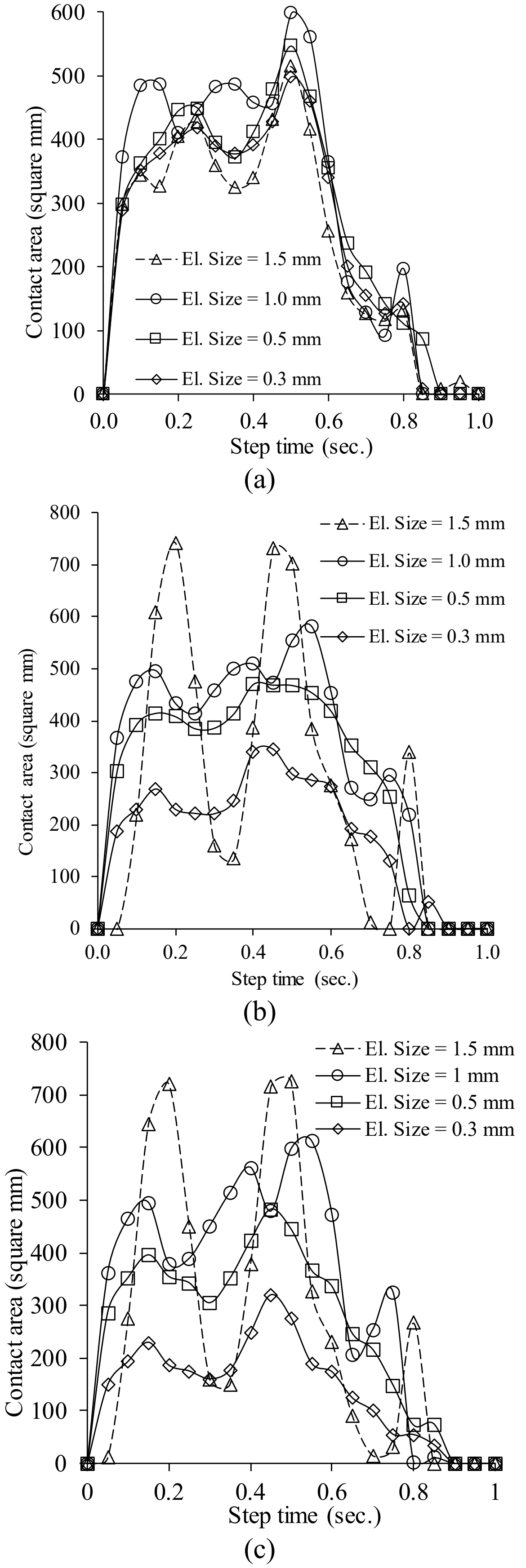



**Figure 6.** Contact areas with a femoral diameter of: (a) 0.5 mm, (b) 1 mm, and (c) 1.25 mm.

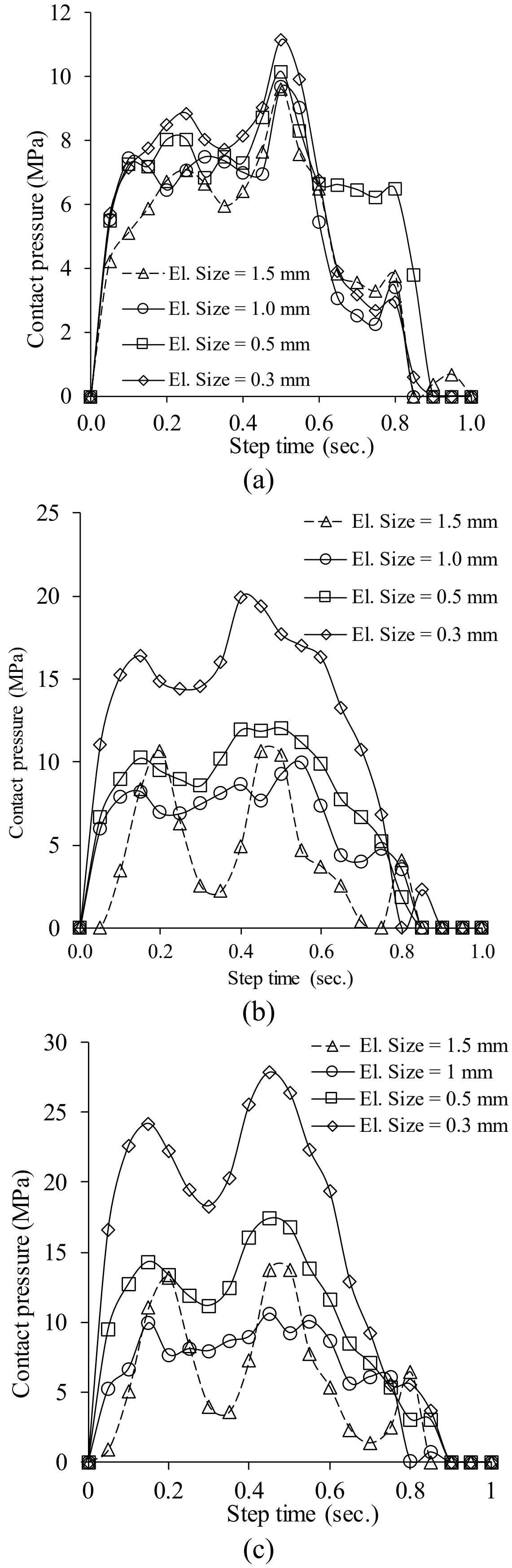



**Figure 7.** Contact pressures from a femoral diameter of: (a) 0.5 mm, (b) 1 mm, and (c) 1.25 mm.

Further investigation in this research is the effect of load step applied to the 0.3 mm acetabular cup model and 1 mm femoral head meshes (Fig. 8). Normal human walking hip gait cycle with sudden loads is a high strain rate problem for the polymer. Different load step means different strain rate. However, for this simplified Paul physiological load with a maximum force of 1.96 kN, load step does not affect all contact mechanic numerical values, i.e., contact loads (Fig. 8 (a)), areas (Fig. 8 (b)) and pressures (Fig. 8 (c)). UHMWPE is sensitive to strain rate. It is shown in the true stress versus true strain curves of [29]. Meanwhile, those data also revealed that below the plastic region (≈ 20 MPa), the effect of strain rate is negligible. Other data from [30] show similar trends. The true stress versus true strain curves at the tested three strain rate variations are close with small deviations. A study with a much higher maximum strain rate, up to five times than [29], was shown by [31]. The engineering stress versus engineering strain curves at various strain rates are also close. In computational study, the stress of polymer can be checked with the von Mises stress output (Fig. 9). The calculated stress is 9.56 MPa (solid line). It is below the limit of UHMWPE plastic region, hence the effects of load step were not found.

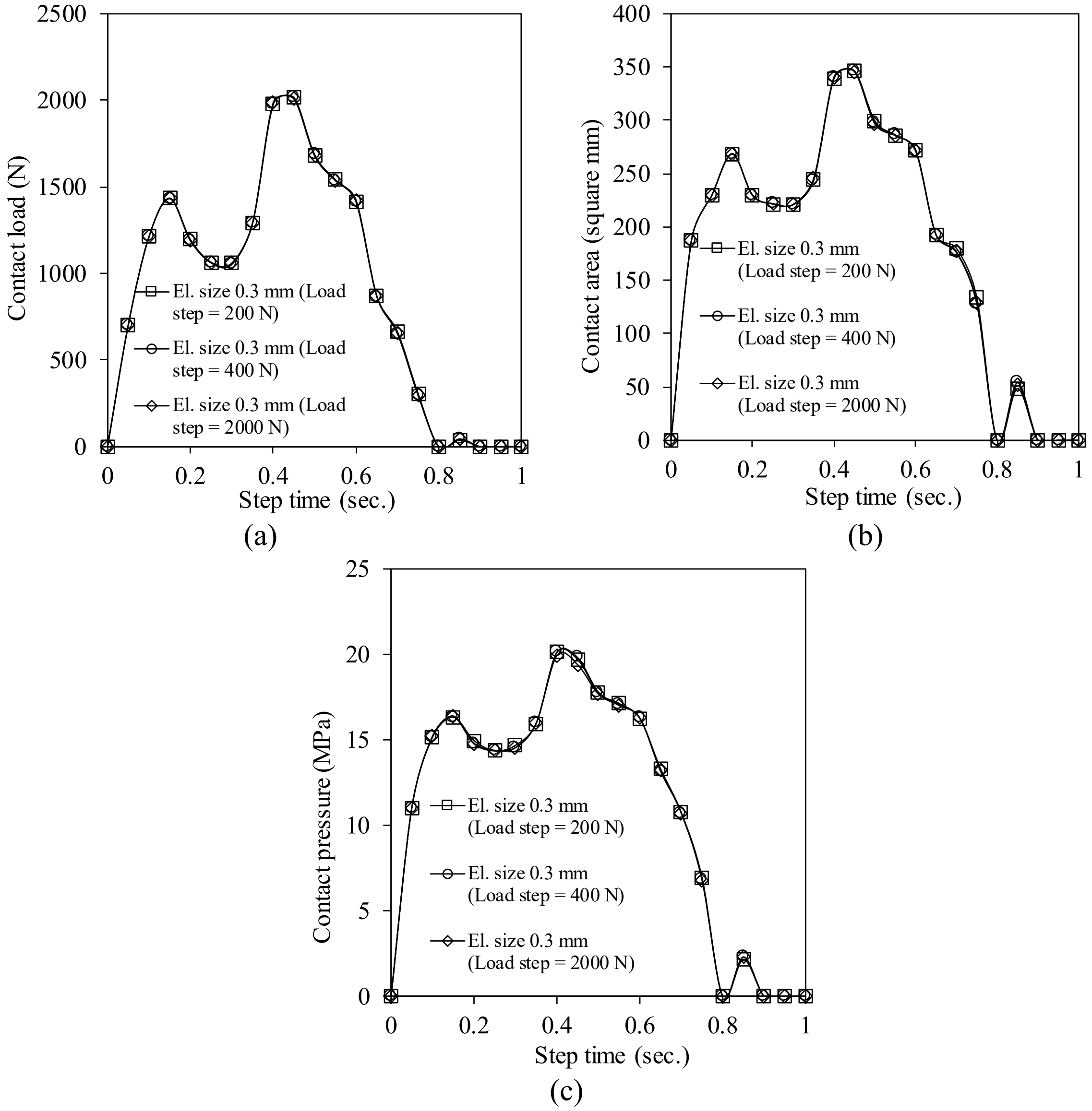



**Figure 8.** Load step effects on computed: (a) contact load, (b) contact area and (c) contact pressure.

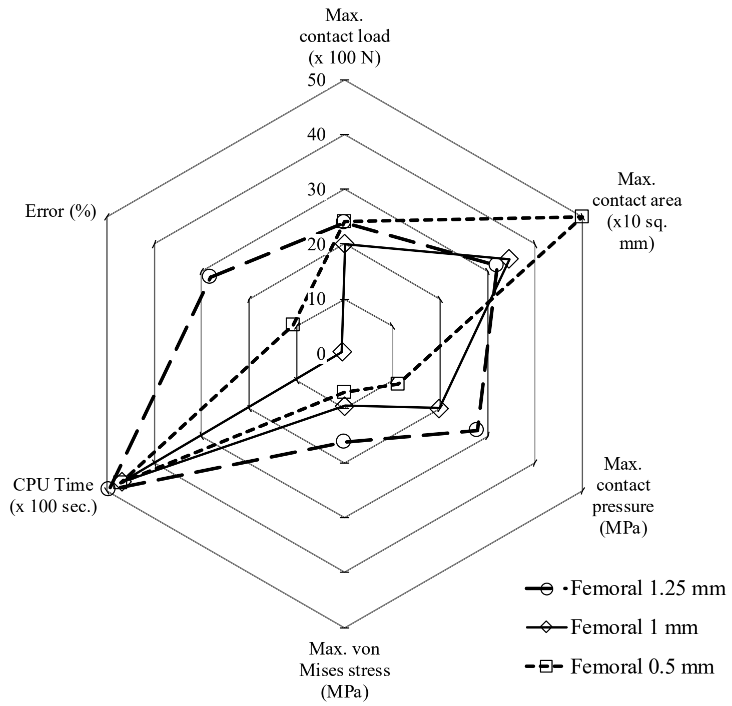



**Figure 9.** Radar chart of errors, CPU times, polymer stresses (von Mises stress) and contact mechanics (contact pressure, contact area, contact load) for all femoral head element size (1.25 mm, 1 mm, and 0.5 mm).

## Conclusions

Volumetric wear modeling and computation of the 32 mm cp Ti/UHMWPE hip implants convergences at 1.38 elements/mm$^3$ mesh density. An accurate prediction was achieved from a model with 1 mm and 0.3 mm element sizes for the femoral head and acetabular cup respectively. This setup has a calculated dynamic load closer to the applied physiological load. Other setups have larger load deviations with higher numerical errors. Load increment does not affect wear volume and contact mechanic numerical results. It is due to the calculated stress that below the UHMWPE yield stress. The effect of strain rate caused by load increment is negligible in the elastic region.

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