

Stability of Cementless Hole and Non-hole Hip Prosthesis

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Abstract— Lacking of stability known to be as one of the principal factors contributes to loosening of hip prosthesis. Initial stability of the hip prosthesis related with the magnitude of relative displacement at the femoral bone-prosthesis interface. The present study aimed to investigate the effect hole as additional features on the initial stability. 3D finite element model of femur and prosthesis developed based on CT dataset of a Malaysian patient. Simulations of normal walking condition performed on the models to investigate the relative displacement between the bone and prosthesis interface. The simulations results report that additional hole as a feature on the proximal region of prosthesis, produce highest relative displacement at the proximal region on the medial side.

Keyword-component; stability, hip prosthesis, cementless

I. INTRODUCTION

Hip arthroplasty is a procedure to replace the disease bone on the hip joint with an implant called hip prosthesis. However, several areas are important in determine the longevity of the hip prosthesis such as stress distribution in the femoral bone and the stability of the hip prosthesis. Many authors have agree, one of the factor contribute to the long-term of the hip arthroplasty is the stability of the hip prosthesis [1, 2, 3]. Unlike cemented prosthesis, the stability of cementless prosthesis is depending on the rate of bone growth to the prosthesis surface. There are two types of stability: Initial stability and secondary stability. Initial stability is referring to the amount of relative motion at the bone-prosthesis surface induced by the physiological loading before biological process is happen. While, secondary stability is the relative motion at the bone-prosthesis surface once the biological process is completed [4]. There are many factors influence the initial stability such as geometry and material properties of the prosthesis, quality of the bone, and the human activity. Different approaches have been used in evaluated the stability either in-vitro study or in-vivo study. These method are important in determined the long-term fixation of hip prosthesis and the successful of the hip arthroplasty. Many previous studied have analyzed and investigate the effect of cross section on the hip prosthesis. However, among of Them were more interested to investigate the stress distribution on the prosthesis surface compared to the stability[5, 6, 7]. Therefore, this study will focus to

analyses and determine the relative motion on the cementless prosthesis by investigate the effect of hole as an additional feature on the initial stability. In this study, finite element analysis performed to evaluate the relative motion at the bone-prosthesis interface for normal walking condition. Three-dimensional solid model of femur bone constructed from the Computed Tomography (CT) dataset obtained from a male patient. Then, the prosthesis developed based on the morphological data extracted from femoral bone constructed earlier.

II. METHODOLOGY

A three-dimensional (3D) model of femur bone was developed based on a computer tomography (CT) dataset obtained from a male patient. The segmentation of 2-dimensional (2D) CT dataset was compiled automatically to generate 3D triangular surface using AMIRA software. Figure 1 shows the completed 3D solid model of femoral bone. The whole model of hip joint is constructed based on the left leg of human.

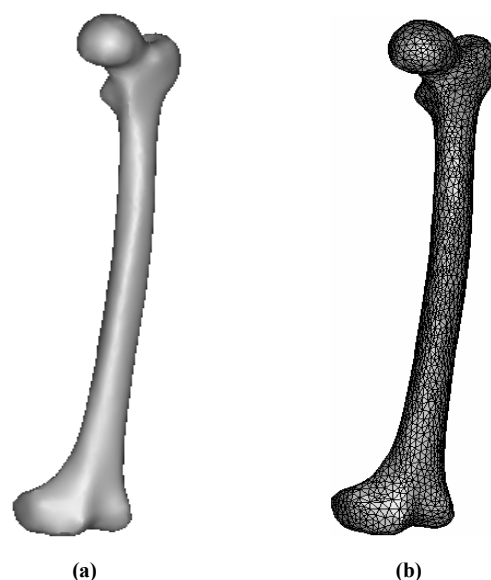


Figure 1. Construction of 3D femoral bone (a) 3D model of femoral bone (b) 3D triangular surface of femoral bone

While, hip prostheses were created based on the extracted morphological data of femoral bone model developed earlier as shown in Figure 1. There are some parameters need to be considered in designing a good hip prosthesis. Figure 2 (a) and Figure 2 (b) shows the relationship between the geometric parameters of the femoral bone and the hip prosthesis. There are various of geometric parameters need to be considered to design a hip prosthesis such as femoral neck shaft angle, femoral head diameter, femoral head offset length and isthmus size. Figure 3 shows the profile for the first design of the hip prosthesis. The profile of first prosthesis is cylindrical and straight stem, double tapered and collarless. In order to investigate the effect of additional features of hole on the stability of prosthesis, the second design was incorporated with feature of hole on the proximal region as shown in Figure 4.

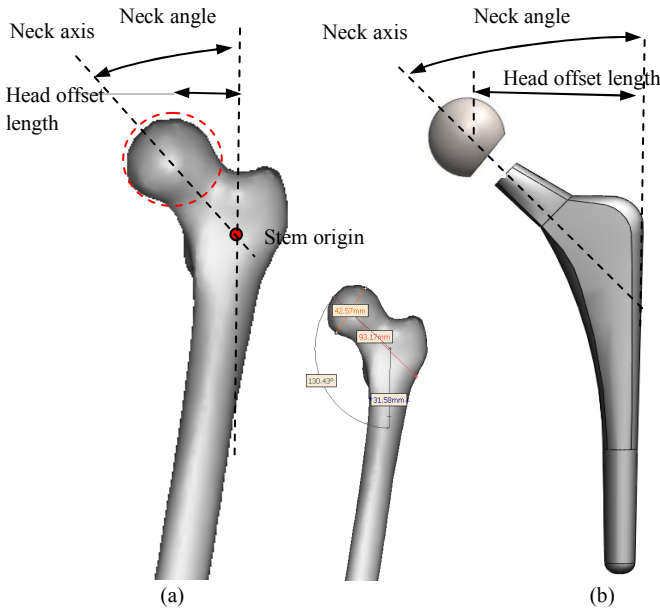


Figure 2. Relationship between the hip geometric parameters and hip prosthesis (a) femoral bone (b) hip prosthesis

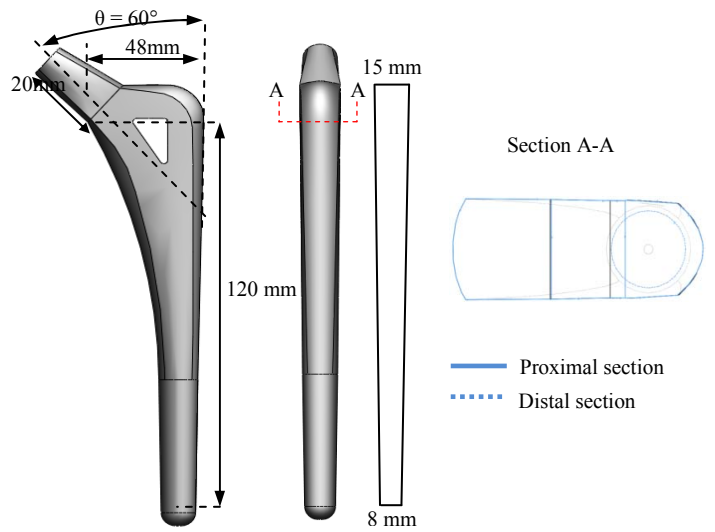


Figure 4. Profile for second design design

ABAQUS finite element software was employed in this study because it allows one to used contact analysis using linear tetrahedral elements. The total numbers of elements for all models listed in Table 1. Material properties for the bone and is adopted from Chae *et al* 2006 [1] while the prosthesis's material from Scott and Jeffrey 2006 [8] . The value of Elastic Modulus and poison ration for bone and prosthesis were shown in Table 2. All the materials were assumed elastic, homogenous and isotropic. By using contact analysis, the frictional contact prescribed at the bone-prosthesis interface as face-to-face contact elements. The surface of prosthesis was modeled as the master surface while the femoral bone surface was modeled as slave surface. As shown in Figure 5, the friction coefficient, 0.3 has applied to the interface.

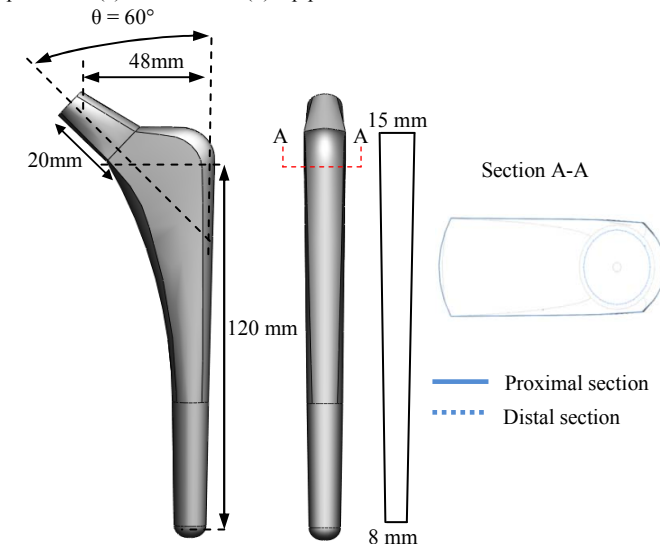


Figure 3. Profile for first design

TABLE I. NUMBER OF ELEMENT CONSTRUCTED

Component of hip joint	Number of Elements	Total (Femur and prosthesis)
Femur	46885	52522
Second design with hole	5637	

TABLE II. MECHANICAL PROPERTIES OF BONE AND PROSTHESIS

Materials	Elastic modulus (GPa)	Yield strength (MPa)	Poison ratio (ν)
Cortical bone	17.26	115	0.29
Titanium alloy	110	485	0.3

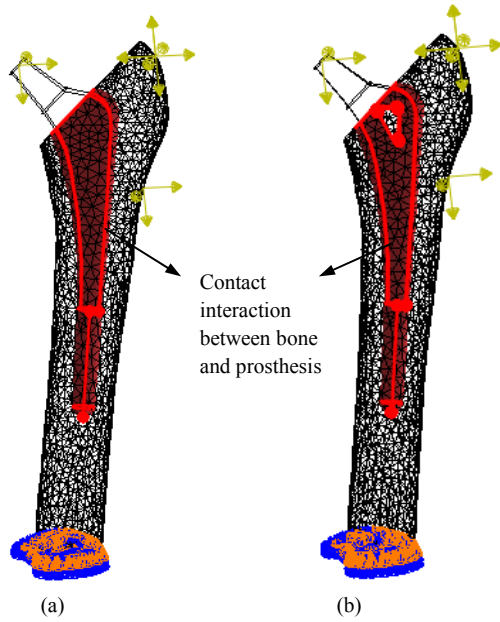


Figure 5. Contact interactions at the femoral bone-prosthesis interface (a) First design (b) Second design

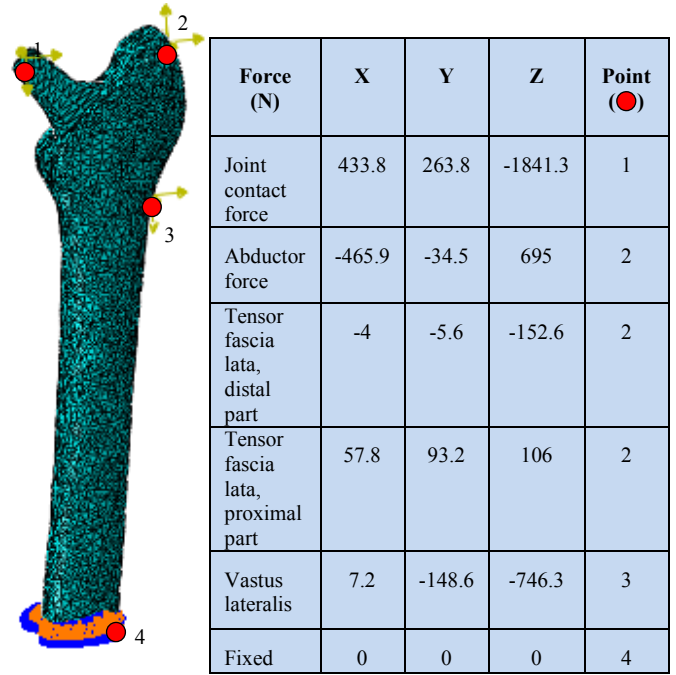


Figure 6. Load and boundary condition

As for loading and boundary condition, normal walking condition was applied on this study. In reality, the movements of leg involved the muscle forces. Unfortunately, not many studies include the value of muscle force. Based on this reason, Pancanti *et al.* (2003) [2] suggested the value of muscle forces is not necessary to determine the relative motion between the femoral bone and prosthesis surface. In this study, the value and location of force obtained from Abdul Kadir *et al.* (2007) [9], which is derived from Duda *et al.* (1997) [10] and Duda *et al.* (1998) [11]. Figure 6 shows the location of forces for normal walking condition. For normal walking condition, there are four types of force applied on the femoral bone; joint contact force, abductor force, tensor fascia lata and vastus lateralis. Joint contact force located on point 1 is depending on the weight of the patient. While, abductor force located on point 2 is the net force caused by gluteus muscle. The function is to extend and externally rotate the hip joint. Similar to abductor force, tensor fascia lata also locate on point 2. However, the function is to abducts and flexes thigh of hip. Vastus lateralis on point 3 is to extend the leg at knee. Lastly, fixed boundary condition was applied at the end of femur which is locate on point 4.

III. RESULTS AND DISCUSSION

The relative displacement for the comparison of first design and prosthesis with hole shown in Figure 7 and 8. From Figure 7, it is indicated that the highest magnitude of relative displacement on lateral side occur on the proximal region. The magnitude of relative displacement for first prosthesis is between $1\mu\text{m}$ and $56\mu\text{m}$. While for the

prosthesis with hole, the relative displacement is between $1\mu\text{m}$ and $172\mu\text{m}$. In addition, it is also indicated that the prosthesis with hole was increased the magnitude of relative displacement from $24\mu\text{m}$ to $172\mu\text{m}$ at the top of proximal region. Since the value of $172\mu\text{m}$ exceeds the threshold value for bone growth of $40\text{--}150\mu\text{m}$, it assumed that the prosthesis with hole feature will lead to loosening of prosthesis. As shown in Figure 8, the highest magnitude of relative displacement of the prosthesis with hole on medial side is $240\mu\text{m}$. Similar with lateral side, the highest magnitude on the medial side also occurs on the proximal area. When compared with first design, the relative displacement of the prosthesis with hole increased by 36 percent. In addition, the magnitude of $240\mu\text{m}$ has exceeded the maximum threshold value. As a result, it will allow the osseointegration to occur at the bone-prosthesis interface.

From the both graphs, it is show that the hole feature has causes instability of the prosthesis. Based on the results and as observed in the both graph, the relative displacement of the prosthesis with hole at the proximal region exceeds the threshold value, $150\mu\text{m}$. Harman (1993) [12] also obtained the similar result. He was conducting an experiment to investigate the relative motion of hip prosthesis. However, in his study, the hollow prosthesis was comparing with Titanium and Cobalt chromium prosthesis.

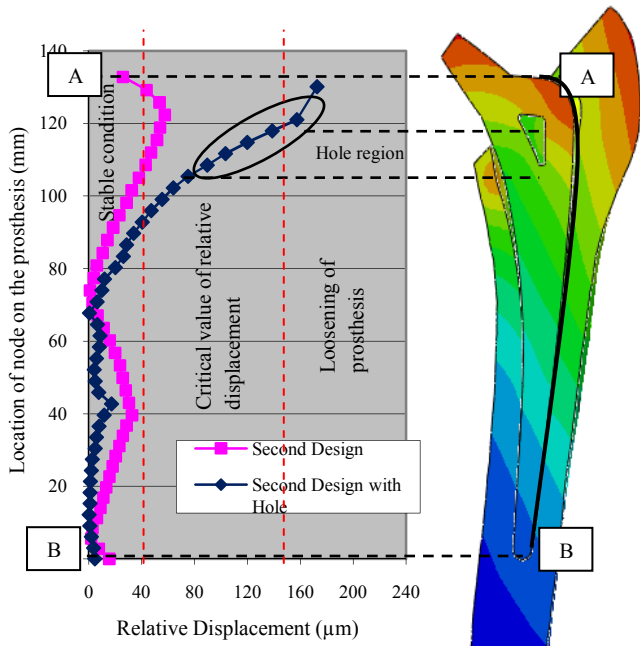


Figure 7. Relative displacement along lateral side (AB)

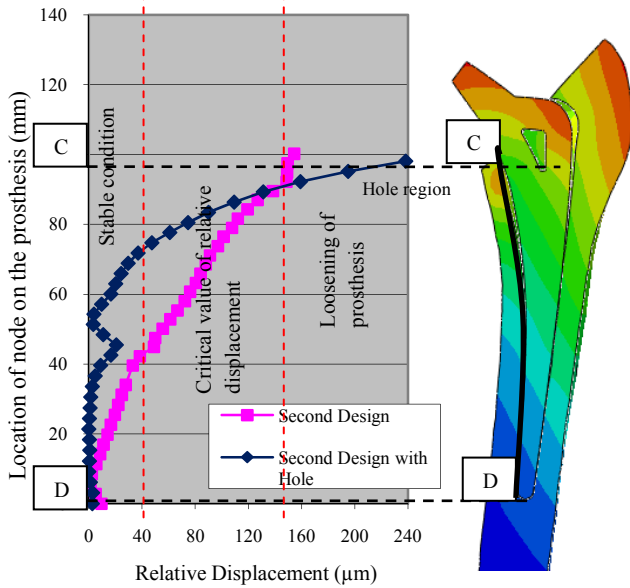


Figure 8. Relative displacement along lateral side (CD)

Generally, hollow prosthesis produces highest relative motion, which is 172 μm . This problem can be explained by the bending theory. The joint contact force producing tensile bending stress on the medial and lateral side. For a linear elastic material, the value of strain is:

$$\varepsilon = \frac{\sigma}{E} \quad (1)$$

According to this formula, the value of strain is depending on the value of stress. Due to the flexure formula as shown in Equation 2, the value of stress is depending on the internal moment, M and moment of inertia, I. As we

know the value of moment inertia is influencing by the area of cross section.

$$\sigma = \frac{My}{I} \quad (2)$$

From the Figure 9, it shows that the hole feature was reduce the area of cross section. Consequently, it was decreasing the value of moment inertia. As a result, it was producing the larger the value of stress and strain. For instance, Figure 10 shows the displacement distribution for the first design and hollow prosthesis. The hollow prosthesis is believed to produce higher relative displacement at proximal region. As expected, the magnitude of displacement at the hole area is larger when compared to areas with no hole.

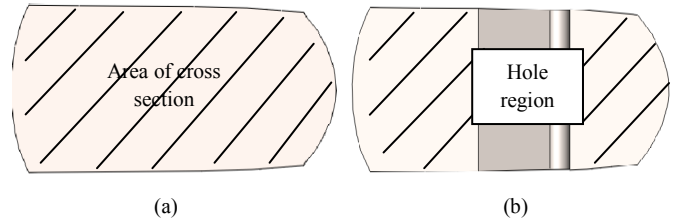


Figure 9. Cross section of prosthesis (a) First design (b) Second design

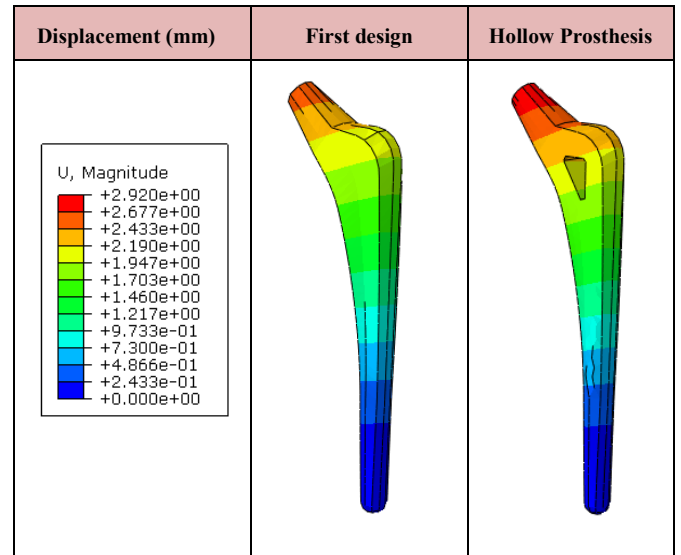


Figure 10. Distribution of displacement for first design and a hollow prosthesis

CONCLUSION

In this study, hip prostheses are designed based on the morphological data of the bone for a male patient. The design hip prostheses are focused on the collarless and cementless type. Initial stability of the hip prosthesis for total hip arthroplasty has been determined by obtaining the relative displacement of node on the master surface (prosthesis) to the node on the slave surface (femoral bone) by using finite element method, under normal walking condition. threshold value used in this study is 40 μm -

150 μ m. From the comparison between the first design and the prosthesis with hole as additional feature, prosthesis with hole feature is produced in larger relative displacement either on the lateral side or on medial side. Besides that, the highest magnitude of relative displacement occurred on the proximal part it exceeded the limit of threshold value. It was expected to lead to the loosening of prosthesis.

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