RESEARCH PROGRESS IN MECHANICAL AND MANUFACTURING ENGINEERING
e-ISSN: 2773-4765

RPMME

Vol. 5 No. 2 (2024) 88-100
https://publisher.uthm.edu.my/periodicals/index.php/rpmme

Simulation of Hip Prosthesis based on Dynamic Loading using Finite Element Method

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DOI: https://doi.org/10.30880/rpmme.2024.05.02.010

Article Info

Received: 15 June 2024
Accepted: 10 October 2024
Available online: 31 December 2024

Keywords

Hip prosthesis , Finite Element Method (FEM), Total Hip Arthroplasty (THA)

Abstract

Hip prosthesis is important in the surgical industry, and research on hip prostheses is crucial to providing the best surgeries for patients. This study aims to design hip prosthesis and analyze the stress distribution and displacement occurring on hip prostheses based on different types of designs under dynamic loading. The outcome of this study is to identify the best design of hip prosthesis to implement for patients. The method of analysis involves using SolidWorks software for designing the hip prosthesis and Abaqus software for simulation in the Finite Element Method (FEM). Three design of hip prosthesis has beed designed to assesed the effect of the geomtery shape on the stress distribution. Based on the analysis, it was found the lowest readings of stress distribution and displacement on the hip prosthesis, as these are the optimal characteristics for the best hip prosthesis. Results from the analysis show that Design 3 has the lowest readings of stress distribution and displacement, making this design suitable for implementation in patients.

1. Introduction

Total hip arthroplasty (THA), commonly known as hip replacement surgery, is a medical operation that involves replacing a broken or diseased hip joint with an artificial joint known as a prosthesis. Individuals with significant hip discomfort and limited mobility due to illnesses such as osteoarthritis, rheumatoid arthritis, or hip fractures are often candidates for this operation. Total hip arthroplasty (THA) is generally a successful treatment for hip-related issues, but like any medical procedure, it can face challenges leading to failure. Two critical factors contributing to this are stress and stability [1]. Stress-related failures in hip prosthetics often arise from wear and tear, especially in weight-bearing situations. This wear can release particles that cause inflammation and damage. Over time, continuous stress and micro-motions can loosen the components from the bone, leading to pain and instability. A major stability concern is hip dislocation, where the femoral ball comes out of the socket, influenced by implant positioning, soft tissue tension, and patient activity. Improper alignment can cause impingement, leading to abnormal collisions in the hip joint, resulting in pain, restricted motion, and implant failure.

Understanding stress shielding in dynamic loading conditions helps improve the overall longevity and performance of hip arthroplasty, enhancing patient outcomes and reducing the need for revision surgeries. Stress shielding can be applied to hip arthroplasty in dynamic loading by optimizing implant design and surgical techniques to minimize the reduction in mechanical stimulus to the host femur. This can be achieved by making the load transfer from the femoral stem to the host skeleton occur as proximally as possible, thereby minimizing the reduction in mechanical stimulus to the host femur. Implant design factors that affect implant stiffness, such as stem length and localization of the porous coating, have received the most attention in stress shielding research [2]. During dynamic loading conditions in hip arthroplasty (implant), the phenomenon of stress shielding can occur. This can result in bone resorption and potential implant loosening, compromising the long-term stability of the implant. Understanding of the moment and force acting on hip replacements is crucial in developing appropriate implants that can withstand dynamic loading conditions. Testing hip implants is a next step to know that higher stiffness in the implant material leads to less deformation and better mechanical stability, while lower stiffness materials can result in increased load transfer to the host bone, promoting bone ingrowth and long-term stability [3]. Continuing hip prosthesis research is crucial for improving total hip arthroplasty and enhancing patient outcomes. Identifying potential complications can lead to better surgical practices. The study found that routine follow-ups help detect and address issues like implant loosening, periprosthetic fractures, and infections, which can reduce the need for revision surgeries and lower healthcare costs. It also emphasizes the need for standardized follow-up protocols and explores the potential benefits of alternative follow-up models. [4].

Stress can arise in total hip arthroplasty (THA) for a variety of reasons, with the mechanical forces and interactions within the replacement hip joint frequently being the cause. In hip arthroplasty, comprehending both static and dynamic loading is essential for implant design and ensuring their resilience. Implants must endure the consistent forces encountered during routine activities (static loading) as well as the fluctuating, occasionally elevated forces experienced during motion (dynamic loading). Variations in stress and strain distribution on the hip implant and femoral bone were noted between dynamic and static loading conditions, with marginally higher values observed under static loading conditions [5]. Selecting hip replacement materials is vital to prevent stress issues during movement. Opting for materials similar to bone properties ensures natural load distribution. Durable, biocompatible materials like alloys or ceramics strike a balance of strength and flexibility. This careful choice enhances implant longevity, creating a healthier biomechanical environment, reducing complications, and optimizing hip prosthesis functionality [6]. In the field of hip arthroplasty, tackling clinical considerations like stress shielding, post-THA bone loss, and ensuring the enduring success of implants is of utmost importance. While stress shielding might seem advantageous initially, it can result in bone atrophy, and subsequent post-THA bone loss poses a threat to implant stability. If not meticulously addressed, these factors directly impact patient outcomes. Failures in hip arthroplasty could manifest as pain, instability, and the potential necessity for revision surgeries [7][8].

The selection of materials in hip arthroplasty plays a pivotal role in mitigating the phenomenon of stress shielding, particularly under dynamic loading conditions. Choosing materials with mechanical properties closer to those of the surrounding bone helps maintain a more natural load transfer, minimizing stress differentials. The selection of materials in hip arthroplasty plays a pivotal role in mitigating the phenomenon of stress shielding, particularly under dynamic loading conditions. Choosing materials with mechanical properties closer to those of the surrounding bone helps maintain a more natural load transfer, minimizing stress differentials. Biocompatible and durable materials, such as certain alloys or advanced ceramics, are often favored to strike a balance between strength and flexibility. This thoughtful selection not only enhances the longevity of the implant but also promotes a healthier biomechanical environment, reducing the risk of complications and optimizing the overall functionality of the hip prosthesis. Commonly on previous study, material that had been studied were CoCrMo, Ti-6Al-4V, and Ti-6Al-7Nb. These three material have thier own advantages on applying this material into the hip prosthesis body.

Certain study choose CoCrMo as th emain material for hip prosthesis based on a few reason. One of the reason is CoCrMo material demonstrates maximum stress with minimum deformation and strain compared to Ti-6Al-4V and Ti-6Al-7Nb [3]. Another reason of choosing CoCrMo material for the stem of the hip prosthesis was likely based on its well-established mechanical properties and clinical performance, as well as its compatibility with the other materials used in the implant design steel material for the hip joint implant [6]. Beside choosing CoCrMo as material for hip prosthesis, there are also studies that chose Ti-6Al-4V as the material selected. Ti-6Al-4V become favorable due to it mechanical properties, including a high Young's modulus, low density, and high ultimate tensile strength, making it well-suited to bear the wear and tear of hip joint movement. Additionally, Ti-6Al-4V is known for its biocompatibility, which is crucial for its performance within the body tissues [9]. The use of Ti-6Al-4V as the material for the hip prosthesis is due to its favourable mechanical properties, including high strength, good fatigue resistance, and biocompatibility [5] while another reason to use Ti-6Al-4V as the material for hip prosthesis is due to its high strength, low density, and excellent corrosion resistance. It has a high modulus of elasticity, which allows it to withstand high loads and stresses without deforming or breaking. Additionally, it has good biocompatibility, which means that it is well-tolerated by the human body and does not cause adverse reactions or immune responses [10].

Loading on hip prosthesis occurs in many ways. The first type of loading based on previous study is during one loading cycle of normal walking activity, the resultant force is highest in the seventh phase with a value of 2326 N, which is 2.5-3 times the average human body weight. This phase is part of the 'stance phase', which is the first 19 phases of the loading cycle. The lowest phase occurs in the 30th phase. The maximum Tresca stress value changes due to the difference in the resultant force applied during loading in the normal walking condition, where the highest maximum Tresca stress value is in the seventh phase for all metal-on-metal bearings in this study [11]. Another research suggests using cadaveric match pair femurs with physiological loading conditions to reproduce the conditions for which the implant is designed. This means that the loading conditions should mimic the natural loading conditions of the femur in the human body [12].

Another study for type of loading is an investigation of dynamic loading with two type of activities that are fast walking and stair climbing. According to the research, the location of the load for the two physiological activities, fast walking and stair climbing, was applied to a finite element model of a composite bone. The resultant forces for fast walking simulated the abductor muscle and hip joint contact forces, with applied forces of 871 N and 1948 N, respectively. For stair climbing, the applied forces simulated the abductor muscles and hip joint contact forces, with applied forces of 951N and 2107 N, respectively [13].

2. Methodology

During this investigation, three distinct hip prosthesis designs will be meticulously crafted utilizing SolidWorks software, with reference to dimensions gleaned from prior research. The research begins with a comprehensive literature review, summarizing existing knowledge. Following this, a three-dimensional (3D) bone structure is constructed. Using SolidWorks, the design for a hip prosthesis is meticulously crafted. The design is then exported to Abaqus, where material properties are defined. Subsequently, force and boundary conditions are established, along with amplitude for dynamic loading. The simulation is initiated, and upon completion, an analysis of stress distribution and relative displacement is conducted. If successful, the process concludes; otherwise, adjustments are made in the material properties before looping back to the simulation phase. Ultimately, the project concludes after a thorough examination of the stress distribution and relative displacement.

2.1 Construction of 3D bone

The three-dimensional femur bone has been constructed based on Computed Tomography scan (CT) datasets in DICOM format are greyscale images from x-ray beam dispersion. Since bone has a higher dispersion rate, the femur looks white in the CT pictures taken every 1.5 mm. Unlike genuine bones, the model was one substance. Femur bone has cortical and cancellous layers [5]. Young's modulus was 2.7 GPa and Poisson's ratio 0.35. Figure 1 shows the femur bone that will be used in the Abaqus.

Fig. 1 *Femur bone scanned by Computed Tomograpghy scan (CT)*

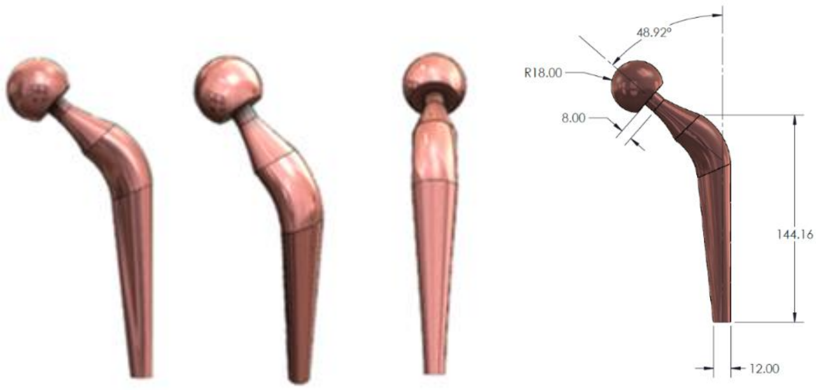
2.2 Hip Prosthesis Design

The design for the first hip prosthesis is with a hole created on the center of hip prosthesis surface. Design 1 has an elongated hole constructed at the center of hip prosthesis in a symmetry pattern as show in Figure 2. This function allows for potential bone ingrowth, promoting better integration of the prosthesis with the surrounding bone. This can enhance long-term stability and reduce the risk of loosening over time. For the body of prosthesis, the shape is rectangular and has the larger proximal area. The length of the prosthesis from the proximal area is 130 mm.

(a) **(b)**

Fig. 2 *(a) Three-dimensional model and (b) Dimension of Design 1*

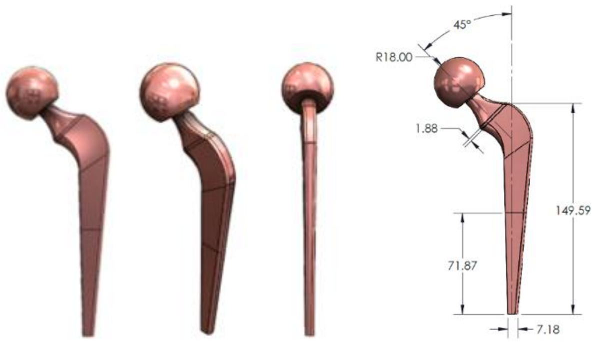
For the second design, it is designed with circular shape, where the diameter of the bottom of the hip prosthesis is same but increases the wide size as it reaches the curve part of hip prosthesis as shown in Figure 3. This shape allows for a more natural range of motion as it follows the curvature of the hip joint and the curvature reduces the potential for friction between the prosthesis and surrounding tissues, potentially minimizing wear and tear over time.



(a) **(b)**

Fig. 3 *(a) Three-dimensional model and (b) Dimension of Design 2*

Lastly, the third design has flat surface at the left and right side of the hip prosthesis with maintained size of wide as shown in Figure 4. Advantages of this design are its stability and load distribution. For stability the flat surface design provides a larger contact area with the surrounding bone, enhancing stability while for the load distribution, the flat surface distributes the load evenly, which can be advantageous for minimizing stress on specific areas of the hip joint.



(a) **(b)**

Fig. 4 *(a) Three-dimensional model and (b) Dimension of Design 3*

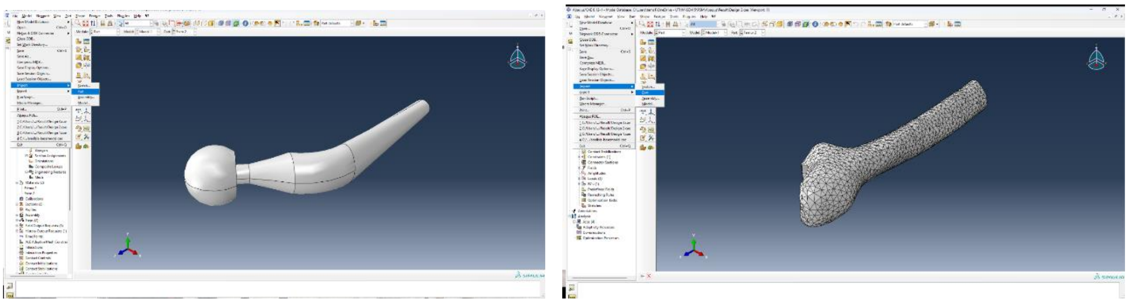
2.3 Finite Element Analysis

The hip prosthesis design created in SolidWorks was exported to Abaqus software to study the stress and displacement on the hip prosthesis and femoral bone. For the simulation, the material, loads, boundary conditions have been defined. In this study, CoCr alloy will be used as the material for the hip prosthesis as the main material while for femur bone will use PMMA as the material. Table 1 shows the mechanical properties of material of prosthesis and femoral bone used in the analysis. The design, along with the femur, was imported into Abaqus using the IGS format. The material properties were then determined to assess the applied stress.

Table 1: *Material used on THA components*

Materials	Placed to use	Young Modulus (GPa)	Poisson Ratio	Density (ton/mm3)
CoCrMo Alloy	Hip Prosthesis	230	0.29	8.17e-9
Cement PMMA	Femur Bone	2.7	0.35	1.18e-9

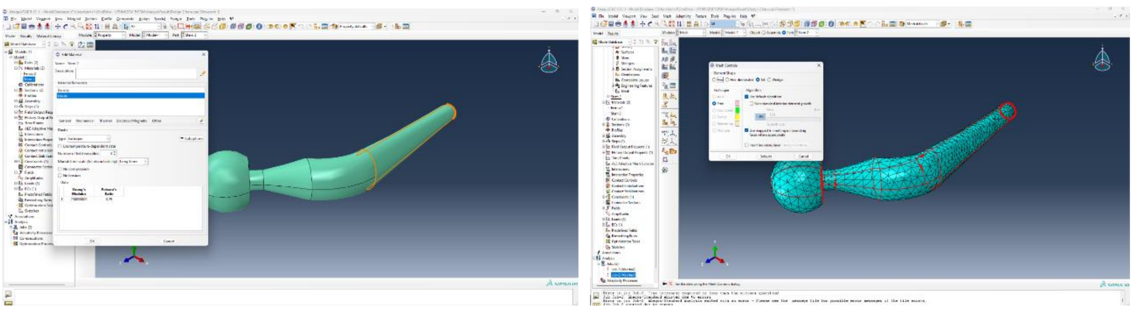
Firstly, the design of Hip Prosthesis and Femur must be imported into the Abaqus Software by using IGS format as shown in Figure 5 below.



(a) (b)

Fig. 5 *(a) Import stem into Abaqus Software and (b) Import femur into Abaqus Software*

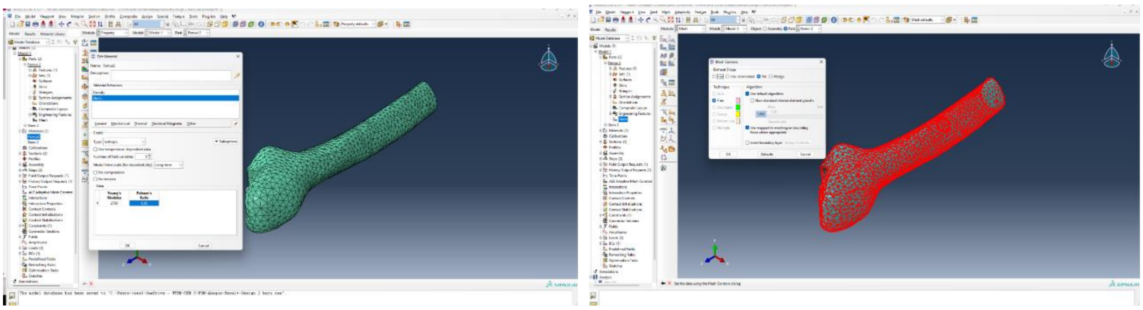
Following that, the meshing step began, with Tet Mesh being used. According to the literature review, boundary criteria were created, and a specified load was applied to the chosen area or spot. Figure 6 depicts the material parameters used in the Hip Prosthesis, highlighting a Young's Modulus of 230 GPa, a Poisson's Ratio of 0.29, and a density of 8.17 ton/mm3.



(a) (b)

Fig. 6 *(a) Apply material for stem and (b) Applied Tet Mesh to the stem*

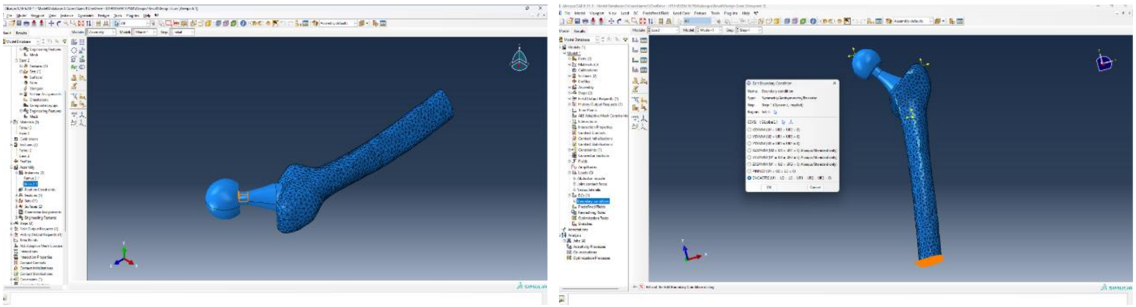
Used global size that is 5.4, and the fraction of the global size is 0.1 for defining the tet mesh on hip stem as shown in Figure 6(b). The maximum deviation factor is currently 0.1 After applying material, force, and boundary conditions to the hip stem, the Abaqus can identify it. The next stage, as indicated in Figure 7(a), was to determine the femur constructed in Solid Work and exported to the Abaqus software. Then, for the Femur bone, use the material's characteristics of 2.7 GPa and 0.35 for the Poisson's Ratio as shown in Figure 7(a) and Figure 7(b) shows how to apply Tet Mesh on the femur bone.



(a) (b)

Fig. 7 *(a) Assign material for the femur and (b) Applied tet mesh to the femur*

After the Femur and Stem components have been identified by applying the material's properties, they can be assembled into one part shown in Figure 8(a). Apply load and boundary condition at each coordinate as shown in Figure 8(b) by using the coordinates from Table 2.



(a) (b)

Fig. 8 *(a) Assemble femur with stem's part and (b) Apply load and boundary condition at each coordinate*

Table 2: *Maximum loading configurations of the major muscles from normal walking activity*

Force	Point	Fx(N)	Fy(N)	Fz(N)
Joint contact force	1	-433.8	263.8	1841.3
Abductor muscle	2	465.9	34.5	695.0
Vastus Lateralis	3	-7.2	148.6	746.3
Boundary condition	4	-	-	-

3. Result and Discussion

The simulation of the hip prosthesis analyzed stress and displacement results for four geometric positions: medial, lateral, anterior, and posterior, as shown in Figure 9.

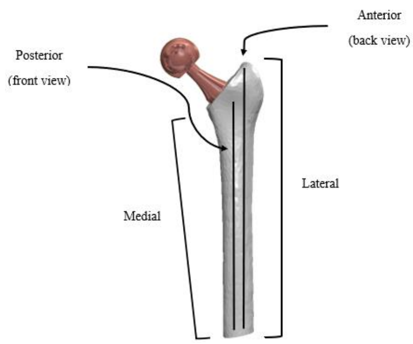


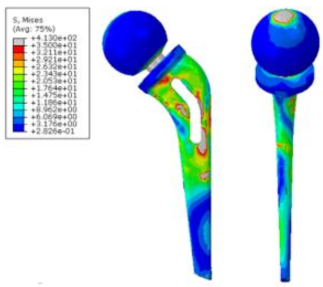
Fig. 9 *Position Geometry Types on Hip Prosthesis*

3.1 Stress Distribution on Hip Prosthesis

In hip prostheses, higher stress values indicate areas experiencing greater mechanical loading, which can lead to material failure, implant loosening, or bone resorption over time. Lower stress values suggest a more even load distribution, reducing the risk of localized stress concentrations. Lower stress levels are preferred as they indicate balanced load transfer between the prosthesis and bone, minimizing the risk of implant failure and bone complications.

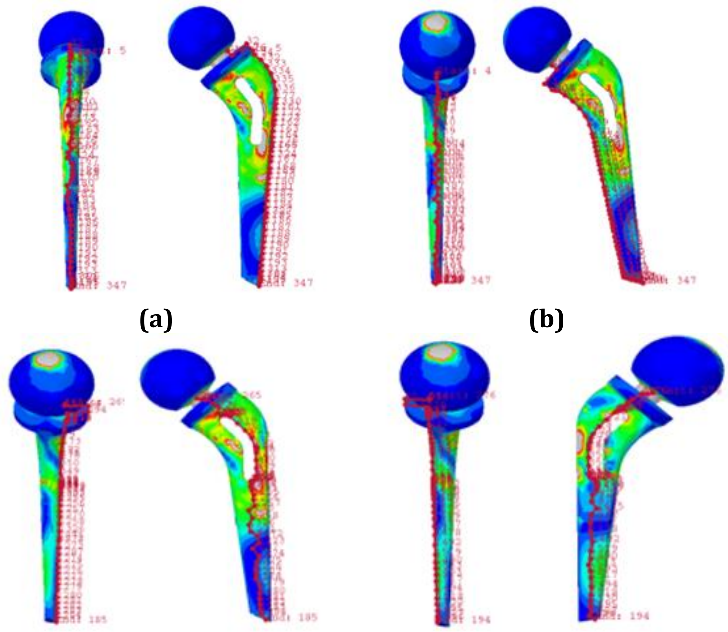
3.1.1 First Design

Figure 10 shows the stress distribution occur on the hip prosthesis for first design with each colour presented the value of stress distribution as shown at the legend. The highest stress distribution value for the first design is 413 MPa and the lowest reading of stress distribution for first design is 0.286 MPa.



(a) (b)

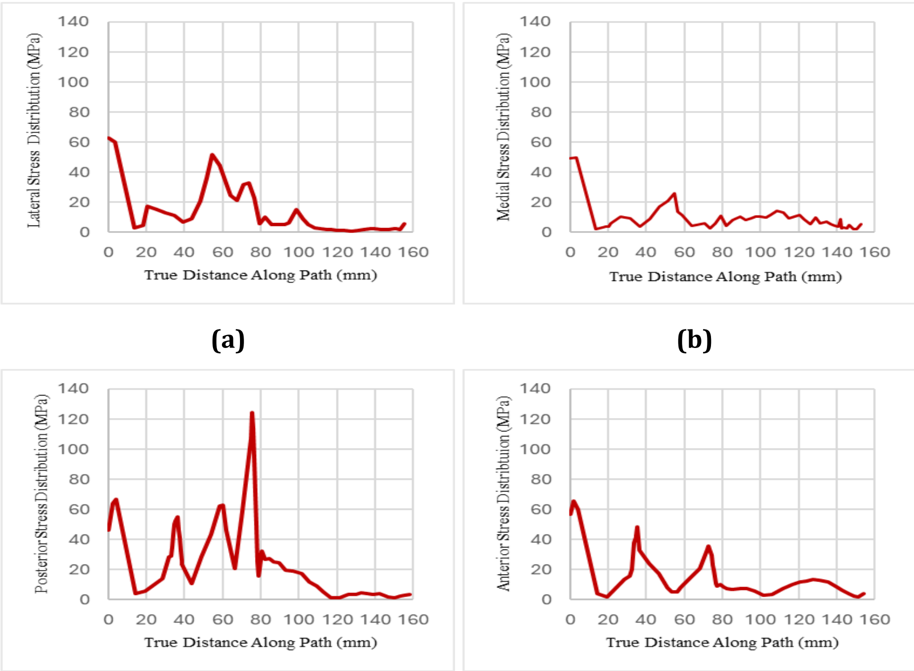
Fig. 10 *Stress distribution on hip prosthesis for first design, (a) Side view of hip prosthesis and (b) Front view of hip prosthesis*



(a) (b) (c) (d)

Fig. 11 *Stress distribution of hip prosthesis at (a) Lateral path, (b) Medial path, (c) Posterior path and (d) Anterior path for first design*

A graph has been plotted along the lateral, medial, posterior and anterior path to determine the stress values from the proximal area to the distal end along the lateral. Figure 12 show the graphs of stress distribution based on selected nodes on the lateral, medial, posterior and anterior path marked as shown in Figure 11 above for second design of hip prosthesis:



(a) **(b)**

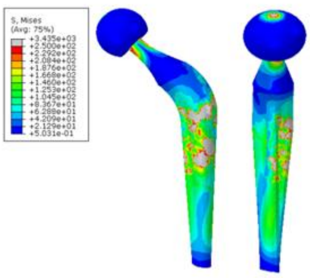
(c) **(d)**

Fig. 12 *Graphs of stress distribution at (a) Lateral path, (b) Medial path, (c) Posterior path and (d) Anterior path for first design*

Based on Figure 12, it can be observed that the posterior path has produced highest stress compared to the lateral, medial and anterior path. The value is 129 MPa and the location is in the middle of the prosthesis. The higher stress values occur due to the presence of a hollow cavity in the middle part of the prosthesis as illustrated in Figure 12(b). Similarly, for the lateral and medial paths, the stress is found to be relatively high in the middle section. Thus, it can be concluded that the creation of a hole in the middle part has caused high stress levels. Additionally, overall, the graph shows that higher stress values occur in the proximal part and that the values fluctuate, decreasing towards the distal part of the prosthesis.

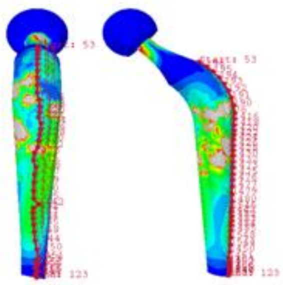
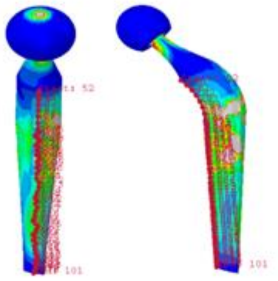
3.1.2 Second Design

Figure 13 below shows the stress distribution occur on the hip prosthesis for second design with each colour presented the value of stress distribution as shown at the legend. The highest stress distribution value for the second design is 3450MPa and the lowest reading of stress distribution for the second design is 0.503MPa.

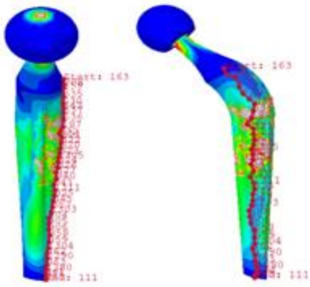
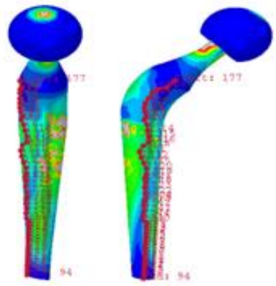


(a) **(b)**

Fig. 13 *Stress distribution on hip prosthesis for second design, (a) Side view of hip prosthesis and (b) Front view of hip prosthesis*

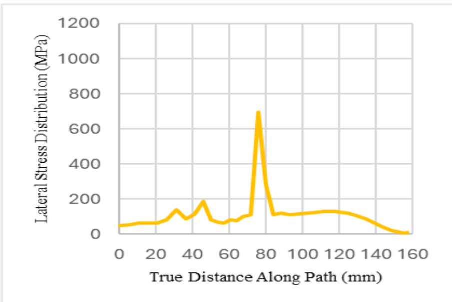
(a) (b)

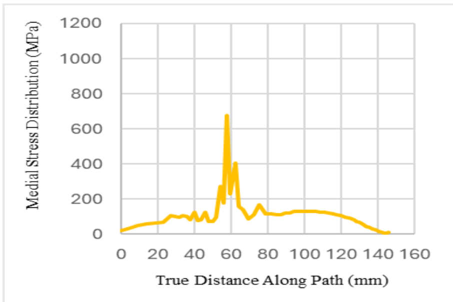



(c) (d)

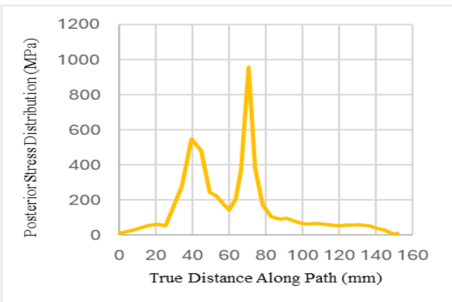
Fig. 14 *Stress distribution of hip prosthesis at (a) Lateral path, (b) Medial path, (c) Posterior path and (d) Anterior path for second design*

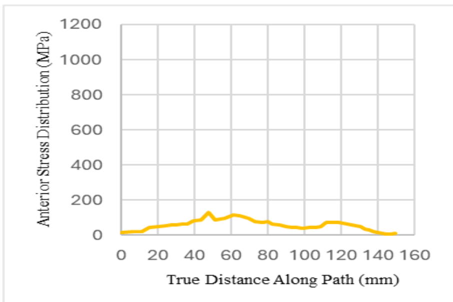
A graph has been plotted along the lateral, medial, posterior and anterior path to determine the stress values from the proximal area to the distal end along the lateral. Figure 15 below show the graphs of stress distribution based on selected nodes on the lateral, medial, posterior and anterior path marked as shown in Figure 14 above for second design of hip prosthesis:





(a) (b)





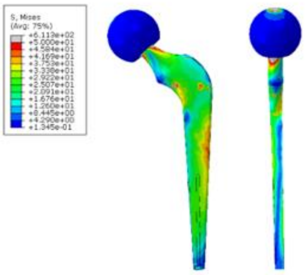
(c) (d)

Fig. 15 *Graphs of stress distribution at (a) Lateral path, (b) Medial path, (c) Posterior path and (d) Anterior path for second design*

Based on Figure 15, the highest stress distribution values for the second design of the hip prosthesis analyzed at posterior path with stress distribution recorded 956.114 MPa at 70.48 mm. Other location also recorded exhibit significant peaks such as at lateral path 694.547 MPa at 75.57 mm while at medial path recorded 678.35 MPa at 57.46 mm. Anterior path recorded lowest stress distribution due to no load applied or force at the anterior side. This means the design of this hip prosthesis that has wider circular shape at the curve side produces high stress distribution after applied force on the hip prosthesis. Additionally, overall, the graph shows that higher stress values occur in the proximal part and that the values fluctuate, decreasing towards the distal part of the prosthesis.

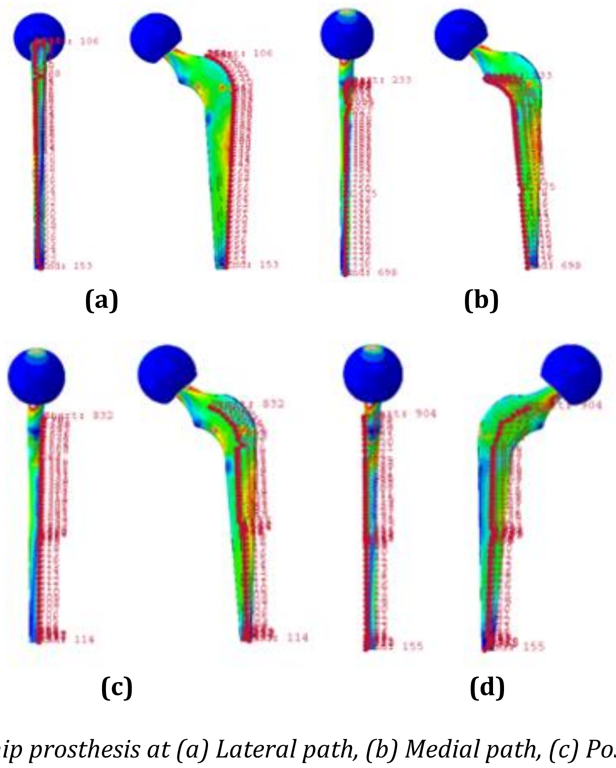
3.1.3 Third Design

Figure 416 below shows the stress distribution occur on the hip prosthesis for third design with each colour presenting the value of stress distribution as shown at the legend. The highest stress distribution value for the third design is 611MPa and the lowest reading of stress distribution for the third design is 0.135MPa.



(a) **(b)**

Fig. **16** *Stress distribution on Hip Prosthesis for third design (a) Side view of hip prosthesis and (b) Front view of hip prosthesis*

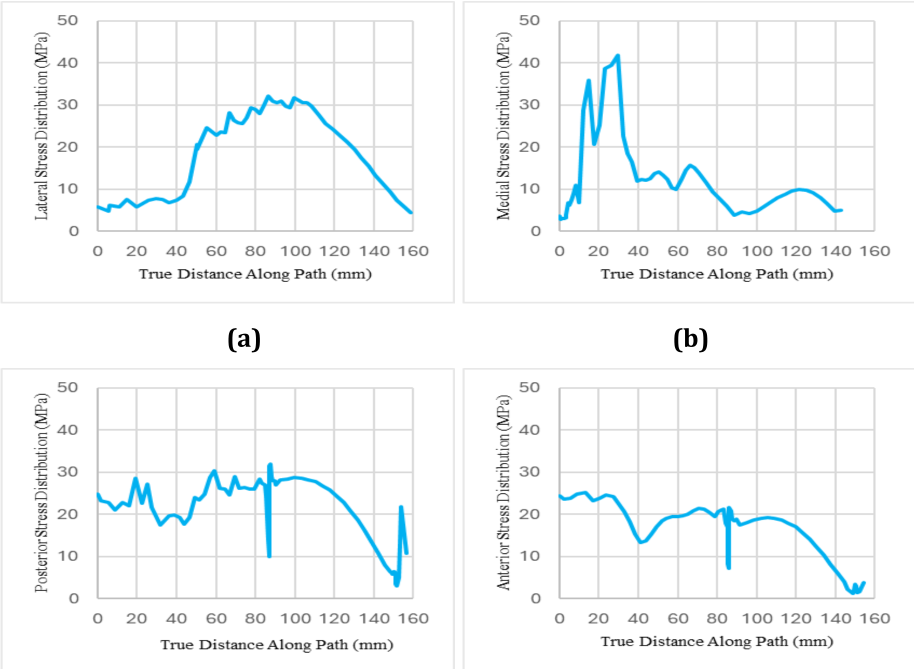


(a) **(b)**

(c) **(d)**

Fig. 17 *Stress distribution of hip prosthesis at (a) Lateral path, (b) Medial path, (c) Posterior path and (d) Anterior path for third design*

A graph has been plotted along the lateral, medial, posterior and anterior path to determine the stress values from the proximal area to the distal end along the lateral. Figure 18 show the graphs of stress distribution based on selected nodes on the lateral, medial, posterior and anterior path marked as shown in Figure 17 above for third design of hip prosthesis:



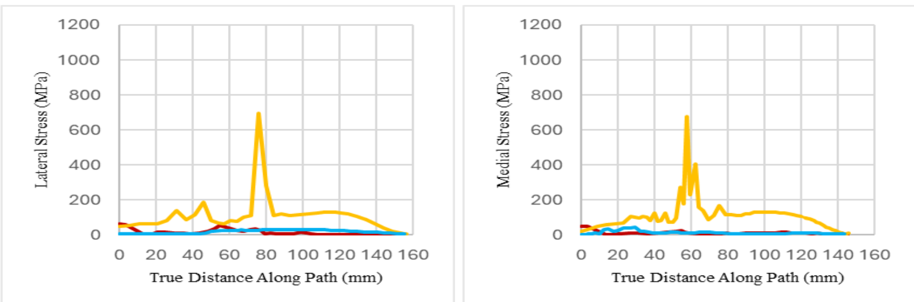
(a) **(b)** **(c)** **(d)**

Fig. 18 *Graphs of stress distribution at (a) Lateral path, (b) Medial path, (c) Posterior path and (d) Anterior path for third design*

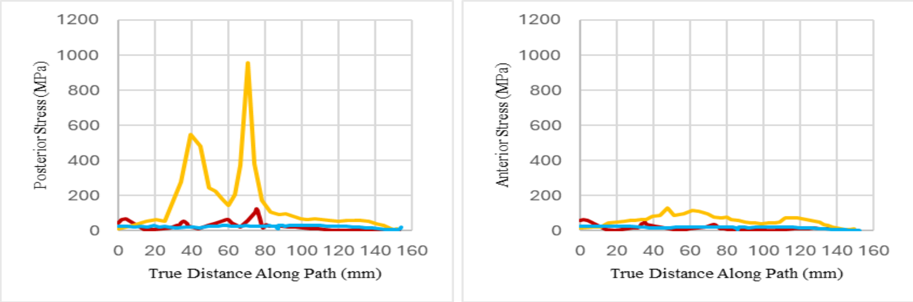
Based on Figure 18, the highest stress distribution values for the third design on the hip prosthesis showed at medial path compared to the other path. The graph shows that the medial path for third design spikes until reaches 41.8773 MPa at 29.44 mm. These peaks indicate high stress distribution values along the medial path which is along the path there are load applied on the hip prosthesis at the lateral path and force the medial path part had to absorb the load from the load applied to the hip prosthesis. Posterior and anterior path show the same pattern with both having average stress distribution that are 15-33 MPa and 12-37 MPa and suddenly drop drastically in same location that is in range of 82-90 mm. Additionally, overall, the graph shows that higher stress values occur in the proximal part and that the values fluctuate, decreasing towards the distal part of the prosthesis.

3.2 Comparison Stress Distribution on Hip Prosthesis

Based on Figure 19, Design 2 shows the highest level of stress with major fluctuations on lateral, medial and posterior path while anterior show nearest range of stress distribution with the other designs. The highest data recorded of stress for design 2 occurs at posterior path with 956 MPa at 70.48 mm. Design 1 with all the data recorded show the moderate stress distribution for all the lateral, medial, posterior and anterior path not higher than design 3 and not lower than design 2. Design 2 can be classified as not suitable to be implemented to the patient since all the paths recorded high stress distribution that can damage the hip prosthesis and then give risk to the patient. Design 1 is the suitable design to be implemented to the patient since all data recorded for all paths has the lowest stress distribution compared to the other two designs.



(a) **(b)**



(c) (d)

Fig. 19 *Comparison graphs of stress distribution at (a) Lateral path, (b) Medial path, (c) Posterior path and (d) Anterior path for all design*

4. Conclusion

The objective of this study is to design a hip prosthesis implant for hip surgery using CAD software and to study the effect of prosthesis geometry on stress distribution and displacement distribution based on dynamic loading. The method to achieve those objectives is by creating the best design for hip prosthesis, choosing suitable material to be applied on the hip prosthesis and locating type of loading at the hip prosthesis based on previous study. From the result obtained, a conclusion from those three designs has been carried out with Design 3 having the best design that has lowest reading of stress distribution along all the lateral, medial, posterior and anterior path and displacement distribution along the lateral and medial path. This makes sure the design can withstand all the obstacles such as stress distribution and displacement distribution for a long period of use when being implemented to the patient.

Acknowledgement

This research was supported by University Tun Hussein Onn Malaysia (UTHM) through Tier 1 (vot Q384).

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