INTERNATIONAL JOURNAL OF

ENERGY AND ENVIRONMENT

Volume 13, Issue 1, 2022 pp.45-52

Journal homepage: www.IJEE.IEEFoundation.org

Optimal design of elastic curved shank of below knee prosthesis

Ammar M. Khalaf, Mahmud. R. Ismail

Al-Nahrain University, College of Engineering, Prosthetics and Orthotics Engineering Department, Iraq.

Received 5 Jan. 2022; Received in revised form 20 Apr. 2022; Accepted 25 Apr. 2022; Available online 10 July. 2022

Abstract

Researchers attempt to gain the benefit of any energy that can be produced from the structure of prosthetic legs. Such energy can be stored and released from elastic parts of the prosthesis like the foot, shank, and joints. The advantage is to reduce patient effort and consumption of metabolic energy. Unfortunately these prosthetic parts are mostly expensive and can not be manufactured locally by Prosthetics and Orthotics centres. The purpose of this study is to design a low-cost elastic shank that can be locally manufactured and can produce equivalent performance. In this work, four elastic shank models are suggested and designed using the SOLIDWORKS program. These models are; C- shape, S-shape, single circular shape, and double circular shape. Each model is analyzed using the finite element method to evaluate stresses, vertical deformation, and safety factors. Also, the stiffness for each model is calculated to investigate the amount of kinetic energy which can be stored in the shank and then can be released to assist the patient and reduce efforts. This energy can be used to improve a patient gait or reduce its metabolic energy. During the research spine, the main limitation was that there were a very low number of studies dealing with the topic of the research. It is found that; the lowest stiffness is found in the C-shape shank which means it is the most elastic design, in the same time it has the highest vertical elastic deformation at a maximum range that can be reached at 15 mm which is preferable when compare it to conventional shock absorber Pylon (shank) while the highest stiffness and lowest deformation at double circular shank which is not preferred for this application but it is the safest design due to reducing the stress concentration. Trade between safety and elasticity can be made to reveal that using of C shape is more optimal for designing elastic shanks. The value of this research is can be a base to manufacture a local elastic shank in Prosthetic and Orthotics centres and use it instead of using high-cost foot, shanks, and adapters.

Keywords: Shank; Prosthesis; Elastic; Below knee.

1. Introduction

Comfort and reducing patient labour are the main aspects of improving the reliability of the prosthesis for walking. Researchers try to solve the problem either by using an active design of prosthesis in which external energy is added by using actuators or by making use of muscle to store energy in elastic parts of the prosthesis in a passive way. Many attempts are carried out to make use of elastic energy from prosthesis foot structures such as leaf foot or using elastic shank (pylon). In this regard; Coleman, K. L. et al. [1] investigated the effect of flexibility of pylon on the ground reaction forces. Four subjects with below-knee prostheses wear flexible (nylon) pylon and rigid (aluminum) ones. GRF was selected for the amputated limb during the gait cycle. They found that for the flexible pylon in the walking test, there was an increase in stance time, reducing the irregularities of the curve. During the step-down test, the vertical loading rate was increased and the peak of the force was greater. The subjects preferred the flexible pylon which was more comfortable. Buckley, J.G. et al. [2] try to find the energy consumption and the Comfort of subjects during walking in below-knee amputation patients using their prosthesis fitted either with or without a tele-torsion pylon. The subjects walked on a treadmill and the oxygen, and carbon dioxide consumption was selected. The results showed that consumption of O2 for subjects who walked with the prosthesis had the TT shank was 5.4 percent and 9.1 percent smaller than when walking with the prosthesis without the TT shank, at the speeds of 130% and 160% of comfortable speed, Gard, S. A. et al. [3] investigated the effects of the shock-absorbing pylon (SAP) on the gait of below-knee patients. The test was done on 10 subjects walking with and without Endolite telescopic torsion pylon. Then comparing the kinematic and kinetic results between them. The most significant difference was a reduction in force transition during the stance phase and this effect was clearer at high walking speeds (above 1.3 m/s). Berge. S. A. et al. [4] showed the magnitude of elastic proprieties for (ICON SAP and Mercury TT Pyramid Pylon). By using the static compressive testing method and least square root mathematical formula, the results showed that the spring constant was (start from 74 to 110 kN/m). In addition, the maximum displacement is 15 mm and 12.5 mm. Berge. S. A. et al. et al. [5] compared a commonly prescribed SAP with a conventional rigid pylon, using a within-subject design (n = 15 unilateral transtibial amputees), to assess the effect on gait mechanics, measure transmitted accelerations in situ, and determine functional outcomes using step counts and questionnaires. The results suggest that the SAP in this study is as effective as a rigid pylon for unilateral transtibial amputees. Lee. W. C. et.al. [6] made a monolimb prosthesis with SACH foot for 14 below-knee amputees They used two designs. ES monolimb, elliptical cross-section. And CS monolimb, circular cross-section. The results showed that the more flexible ES monolimb reduced intact limb ground reaction force GRF at heel strike and the prosthetic limb GRF at toe-off. Elizabeth Klodd et al. [7] investigated if the flexibility of the forefoot of below-knee prosthesis plays a role in the energy expenditure or oxygen consumption in addition to the preferences of 13 subjects. Five feet were manufactured for use in this research (F1, F2, F3, F4, and F5). F1 was the most flexible, and F5 was the least flexible. Partici¬pants walked at the same comfortable, freely selected speed on the treadmill for 7 min with each foot while energy expenditure was measured. They found that there is no effect on oxygen cost but the subjects prefer the most flexible forefoot. Mohsin Hamzah et al. [8] designed a prosthetic foot under the AUTOCAD program and analyzed it by the ANSYS program in different stages of the stance phase during the gait cycle to obtain equivalent von-messes stresses and determine the factor of safety in addition to total deformation and strain energy. The analysis was based on carbon-epoxy (230 GPa) wet material. Saleel H Abood et al. [9] designed and manufactured a sprint foot (prosthetic foot for running) made from carbon fiber and epoxy. Firstly, the model was designed then analysis the foot by the ANSYS program to obtain maximum principle stress and total deformation using two types of composite materials one of them was carbon-epoxy, the other was glass-epoxy. By comparing the numerical results for both the two materials, found that the carbon was better at absorbing energy than glass fiber. Mahmud R Ismail et al. [10] aimed to make enhanced pylon with internal rubber to make special stiffness and damping properties. The rubber acted as a shock absorber and an energy storage mathematical model was done to analyze the impact during the gait cycle. The modified prosthesis was tested with BK amputation patients and compared with the traditional one by using a force platform and treadmill The enhanced pylon was compared to the traditional pylon (stiff pylon) using a force plate test. The results showed there was an improvement in gait characteristics. Fariborz Tavangarian et al. [11] made a pylon through an additive manufacturing technique a using 3D printer. Which were tested by a static proof test and compression test on the pylons to evaluate their performance. The results showed that 3D printed pylons have enough strength under stress and exceed the requirements of the standards and therefore can replace the metallic pylons in lower limb prostheses. Ameer A. Kadhim et al. [12] made a design and analysis of a pylon in the SOLIDWORKS program, then manufactured the pylon by the 3D printer using different materials (PLA). The pylon was tested by fatigue test using a spe l machine designed for this purpose. The results showed that the pylon can withstand the condition of walk loading. Jenny Anne Maun et al. [13] determined the effects of walking speed and SAP stiffness on mechanical work. The results showed that faster walking increased mechanical work while reducing the SAP stiffness will increase the energy absorption during early stance.

Most of the previous studies dealt with the designing of feet made from composite material regardless of the shank. Other studies evaluate the performance of conventional shanks and their properties. Other studies investigate the stiffness effect of the conventional shanks on the patient's comfort. In this study, three designs of composite elastic shanks were analyzed using FEM to obtain the optimum design that can produce enough elasticity and strength for patients with 70 kg of mass.

2. Modeling by SOLIDWORKS

Four models are suggested for shank modification to perform elastic structure to store energy from the effect of patient boy weight. The shank model must be strong enough to prevent buckling and give an elasticity value at the same time with the limit of dimensions to be used instead of the ordinary pylon. Figure 1 shows the shapes of these designs. The detailed drawing of the shank models are shown in Figure 2.

The design was done based on the anthropometric data to obtain the correct length for the shank which was supposed to be (118 mm). The holes for the attachment with clamp adapter from the top to be connected to the prosthetic socket and with tube adapter from the bottom to be connected with the prosthetic foot. The dimensions showed in Figure 3. The hole diameter was chosen to be (6 mm) to fit with the M6 bolt according to the metric bolt standard. The distance from the hole center to the edge was chosen to be (10 mm) to avoid tearing according to the equation of minimum margin (3.3), [14],

$$M = 1.5\,D \quad (1)$$

Where M is the length from the center of the hole to the edge. D is the diameter of the hole.

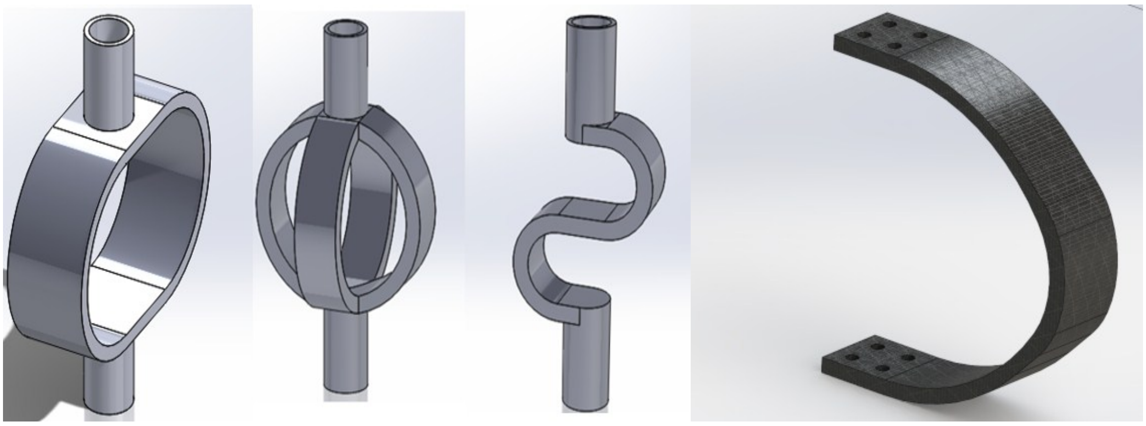


Figure 1. Some suggested designs for the shank.

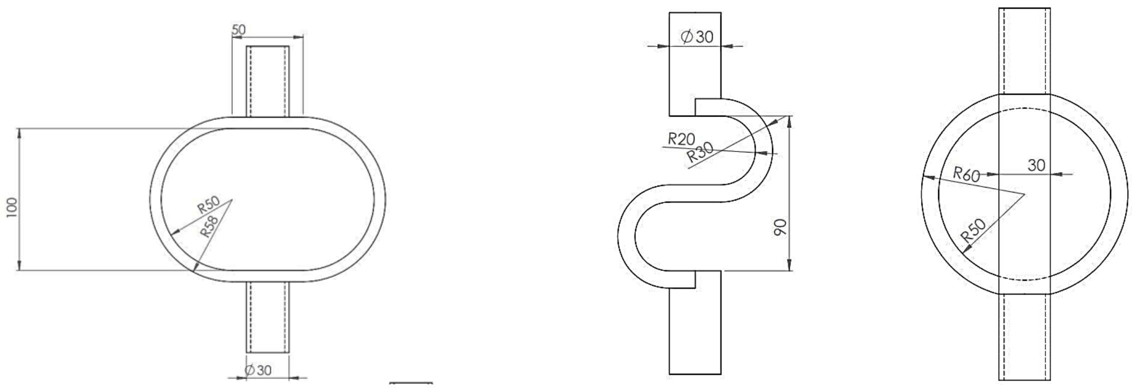


Figure 2. Detailed drawing of the suggested models.

3. Finite element analysis FEM

The numerical analysis is carried out by using ANSYS-15 software to find stress analysis, deformation, and safety factors. The following steps are performed for FEM. The shanks were analyzed through the static structure by applying a force on the top of the shank and supported at the bottom by compression only as shown in Figure 4. The shank was divided into tiny triangular elements by meshing as shown in Figure 5. For more insurance, a convergence test was done to get accurate calculations.

Also, the convergence test was carried out to select the best finite element meshing to be employed in the analysis of static structures of the elastic shank. Meshes were developed by increasing the number of elements up to (340255) and the stress remain stable approximately at (73 MPa) as shown in Figure 6.

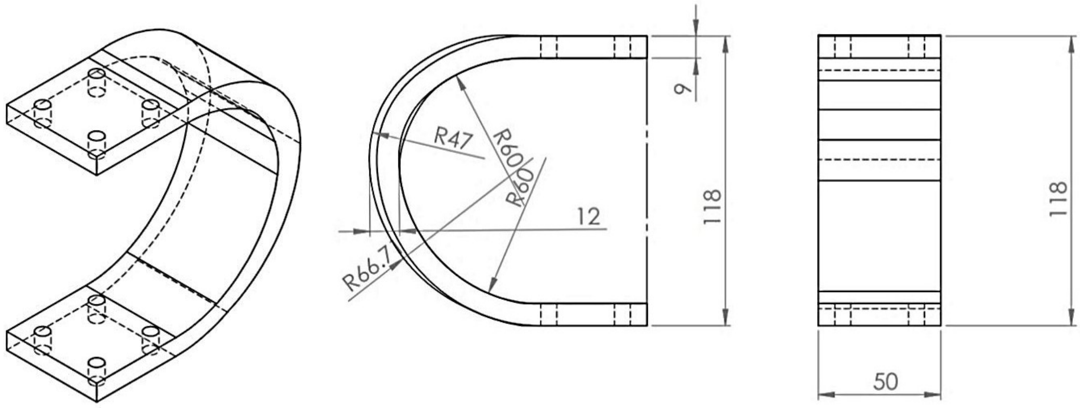


Figure 3. Isometric drawing of C shape used in SOLIDWORKS.

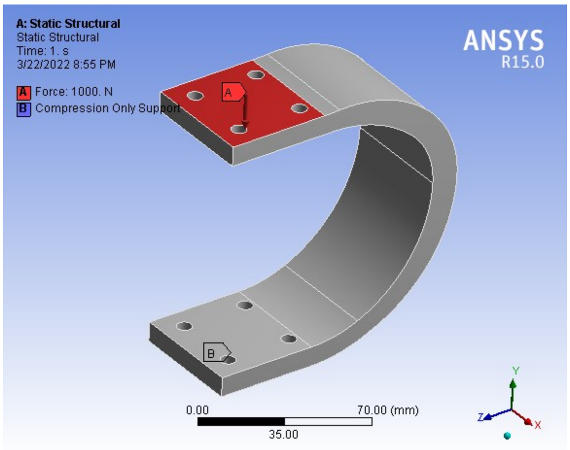


Figure 4. Load location and support.

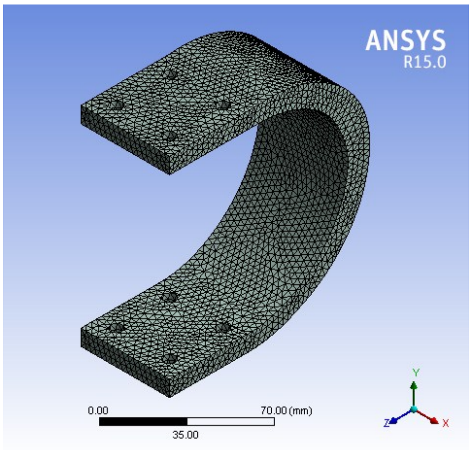
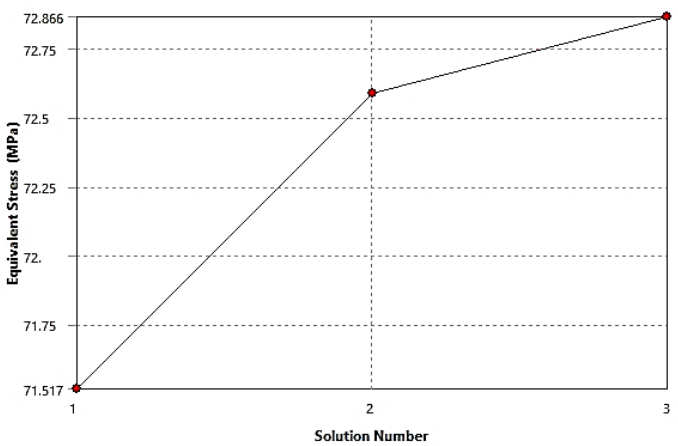


Figure 5. Meshing of model.



Model (A4) > Static Structural (A5) > Solution (A6) > Equivalent Stress > Convergence

	Equivalent Stress (MPa)	Change (%)	Nodes	Elements
1	71.517		41571	26646
2	72.59	1.4881	184866	126476
3	72.866	0.38004	484348	340255

Figure 6. Mesh convergence test.

By setting element size to 4 mm and making face sizing in the inner face of the top curvature to 1 mm in addition to edge sizing at the holes. Slow transmission of the sizing was done for a more smooth transition between large and fine elements. The tetrahedron element type was used for more accurate results as shown in Figure 7.

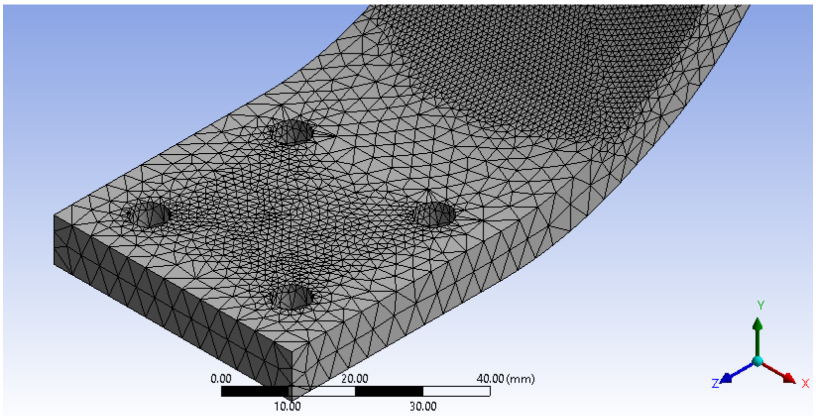


Figure 7. Mesh type, mesh sizing.

4. Results

The results of ANSYS analysis of the four shank models are shown in Figures 8 to 15. For each model, both Von-mises stress and vertical deformations are presented. The applied load is 900 N to represent the maximum bodyweight. For C and s shape shanks the maximum stress and deformation are located at the top of the curvature as shown in Figures 8 to 11 while for both circular and double circular shanks the maximum stress and deformation at the connection end with the shank adapter.

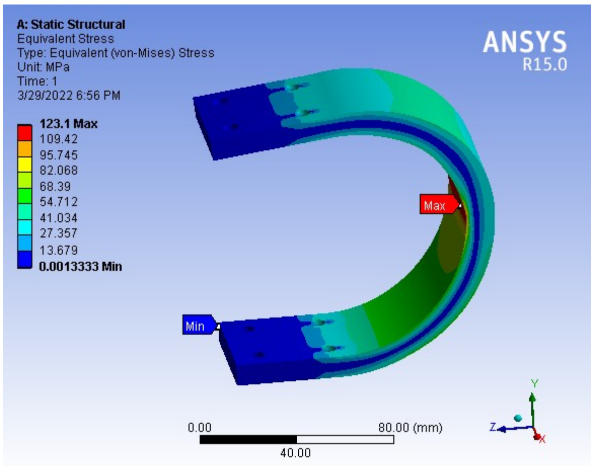


Figure 8. Deformation in Y-axis for C model.

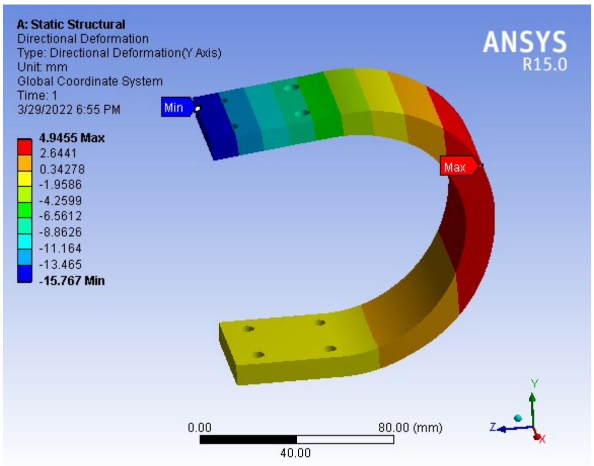


Figure 9. Equivalent Von-Mises for C model.

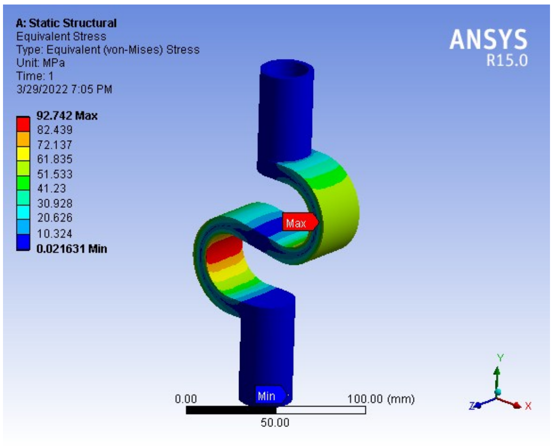


Figure 10. Deformation in Y-axis for S model.

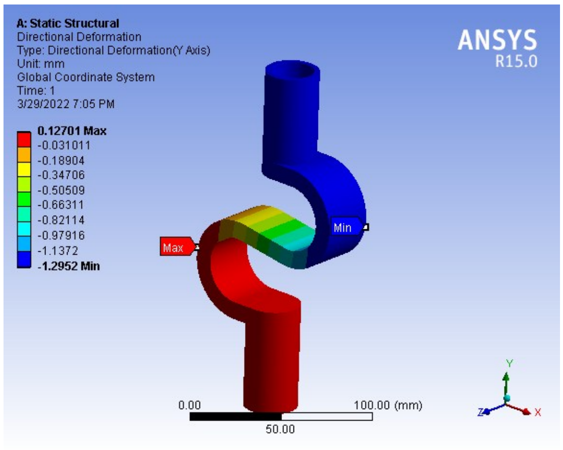


Figure 11. Equivalent Von-Mises for S model.

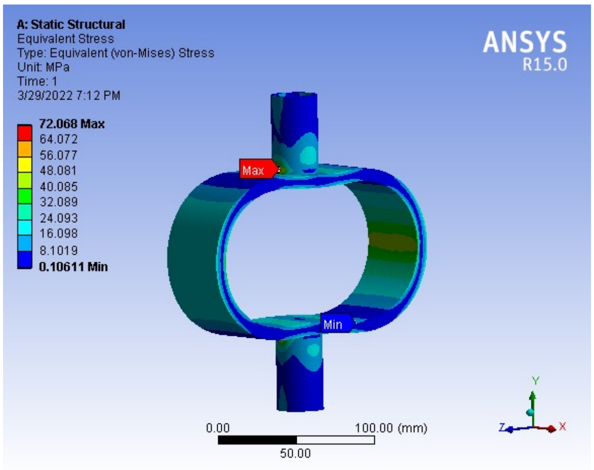


Figure 12. Deformation in Y-axis for the circular model.

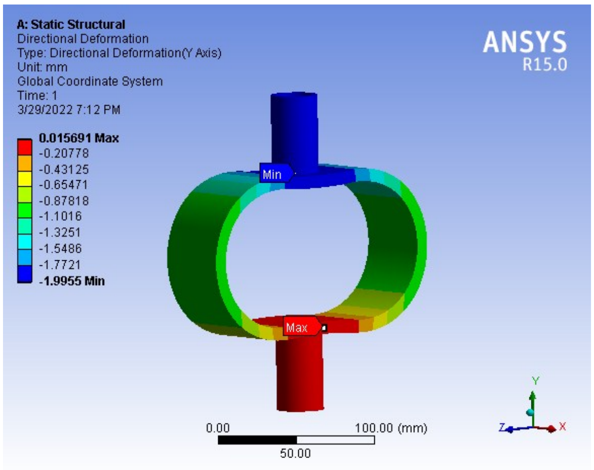


Figure 13. Equivalent Von-Mises stress for the circular model.

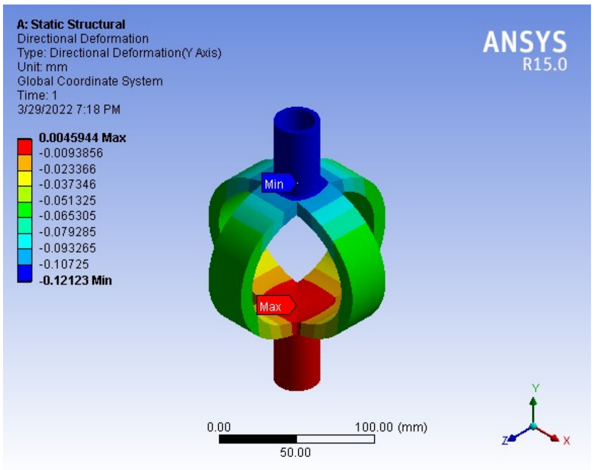


Figure 14. Deformation in Y-axis for double circle model.

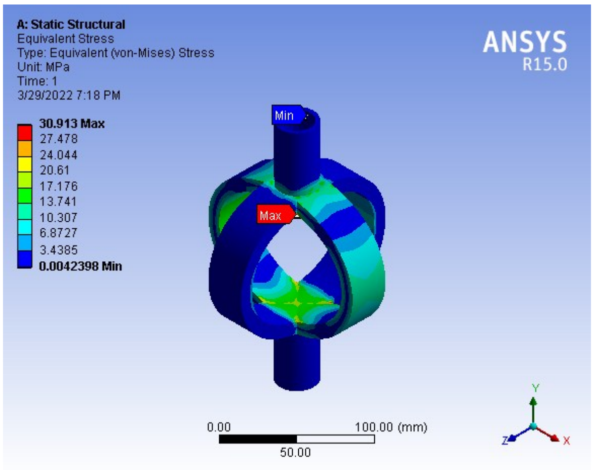


Figure 15. Equivalent Von-Mises stress for double circle model.

To compare the models, vertical maximum deformation, Von-mises stress, and stiffness are collected and displayed in Table 1. The stiffness is calculated by dividing the applied load by the maximum vertical deformation. As it is clear from the table the lowest stiffness was found in the C model while the highest in the double circular model.

Table 1. Comparison of stresses, deformations, and stiffness of Four shank models.

Model	Maximum Vertical deformation (mm)	Equivalent von-mises stress (MPa)	Stiffness (N/mm)	Minimum Safety Factor
C-shape	15.7	101.2	57.3	2.9
S-shape	1.3	92.74	692	3.1
circular shape	2	72	450	4.5
Double circular shape	0.12	30	7500	5.2

5. Conclusions

Four elastic shank shapes are proposed in the work to select the best model for storing energy. It is found that:

1- the C shape is the best since it has the highest elasticity with the lowest stiffness (57.3 N/mm), and highest deformation (15.7 mm) which is close to the conventional shock absorber pylons as mentioned in Reference [4] and good safety factor (2.9).

2- The worst is found in a double circular shape with the stiffness of 7500 N/mm and deformation of 0.12 mm which is not sufficient to save enough energy for the application.

3- This reveals that using of C shape is more optimal for designing elastic shank.

The stress was higher in the C shape model due to stress concentration at the top of the curvature therefore the C shape shank is suitable for the K3 activity level in which the maximum load applied on the shank is 1.3 times the body weight.

References

[1] K. L. Coleman, D. A. Boone, D. G. Smith, J. M. Czerniecki 'Effect of trans-tibial prosthesis pylon flexibility on ground reaction forces during gait' Prosthetics and orthotics international, Vol. 25, No. 3, pp. 195-201, 2001.

[2] J. G. Buckley, S. F. Jones, K. M. Birch 'Oxygen consumption during ambulation: comparison of using a prosthesis fitted with and without a tele-torsion device' Archives of physical medicine and rehabilitation, Vol. 83, No. 4, pp. 576-581, 2002.

[3] S. A. Gard, R. J. Konz 'The effect of a shock-absorbing pylon on the gait of persons with unilateral transtibial amputation' Journal of Rehabilitation Research and Development, Vol. 40, No. 2, pp. 109-124, 2003.

[4] J. S. Berge, G. K. Klute, J. M. Czerniecki 'Mechanical properties of shock-absorbing pylons used in transtibial prostheses' J. Biomech. Eng., Vol. 126, No. 1 pp. 120-122, 2004.

[5] J. S. Berge, J. M. Czerniecki, G. K. Klute 'Efficacy of shock-absorbing versus rigid pylons for impact reduction in transtibial amputees based on laboratory, field, and outcome metrics' Journal of rehabilitation research and development, Vol. 42, No. 6, 2005.

[6] W. C. Lee, M. Zhang, P. P. Chan, D. A. Boone ' Gait analysis of low-cost flexible-shank transtibial prostheses' IEEE Transactions on neural systems and rehabilitation engineering, Vol. 14, No. 3, pp. 370-377, 2006.

[7] Klodd, E., Hansen, A., Fatone, S., & Edwards, M. (2010). Effects of prosthetic foot forefoot flexibility on gait of unilateral transtibial prosthesis users. J Rehabil Res Dev, 47(9), 899-910.

[8] Hamzah, Mohsin & Gatta, A. (2018). Design of a Novel Carbon-Fiber Ankle-Foot Prosthetic using Finite Element Modeling. IOP Conference Series: Materials Science and Engineering. 433. 012056. 10.1088/1757-899X/433/1/012056.

[9] Abood, S. H., & Faidh-Allah, M. H. (2019). Analysis of prosthetic running blade of limb using different composite materials. Journal of Engineering, 25(12), 15-25.

[10] Ismail, Mahmud & Y. Kahtan, Yassr & Abbas, Saif M. (2020). A modified shank for below knee prosthesis. IOP Conference Series: Materials Science and Engineering. 671. 012061. 10.1088/1757-899X/671/1/012061.

[11] Tavangarian, F., Proano, C., & Zolko, C. (2019). Performance of low-cost 3D printed pylon in lower limb prosthetic device. In TMS 2019 148th Annual Meeting & Exhibition Supplemental Proceedings (pp. 1207-1215). Springer, Cham.

[12] Kadhim, A. A., Abbod, E. A., Muhammad, A. K., Resan, K. K., & Al-Waily, M. (2021). Manufacturing and analyzing of a new prosthetic shank with adapters by 3D printer. J. Mech. Eng. Res. Develop., 44(3), 383-391.

[13] Maun, J. A., Gard, S. A., Major, M. J., & Takahashi, K. Z. (2021). Reducing stiffness of shock-absorbing pylon amplifies prosthesis energy loss and redistributes joint mechanical work during walking. Journal of neuroengineering and rehabilitation, 18(1), 1-15.

[14] R. S. Khurmi, J. K. Gupta ‘A textbook of machine design’ S. Chand publishing, 2005.