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Development of 3d Printable Passive Prosthetic Feet

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ABSTRACT
There are a large number of commercially available prosthetic feet. Most of the design can be divided into two classes, articulated and non-articulated feet. One common non-articulated foot is the SACH. The solid ankle cushion heel foot referred to as the SACH foot has a rigid keel. Due to the rigidity (SACH Feet) and high cost of the feet (MPC Feet), non rigid designs are necessary to improve the performance and reduce the cost of manufacturing. This can be achieved by the use of compliant mechanisms and 3D printing. So, the objective of this thesis is to develop design approaches and models for prosthetic ankle joints using kinematic models of the human ankle and compliant mechanisms technology. Compliant mechanisms offer several potential design advantages over traditional rigid-body designs including high reliability and low cost. These design advantages are ideal for use in prosthetics and with 3D printing prosthetic legs can be manufactured at a lower cost with short production time, making them more affordable and accessible. Another benefit of 3D printed prosthetic legs is that amputees can choose various designs for their artificial limbs.
KEYWORDS: Prosthetic foot, compliant mechanisms, 3D printing

I. INTRODUCTION

Prosthetic foot-ankle systems can be placed into two categories: articulated and non-articulated systems. An articulated system prosthetic is composed of rigid or flexible links connected by joints. A non-articulated prosthetic is made of one piece or several pieces making up one segment, but has no articulating joint. In terms of function, prosthetic feet can be categorized into the following groups:

- Solid Ankle Cushioned Heel (SACH)
- Elastic (flexible) Keel Foot
- Single-Axis Foot
- Multi-Axis Foot
- Dynamic-Response Foot
- Microprocessor Foot

In summary, there are many different prosthetic feet models with varying performance and costs. The SACH foot is a common model that offers good shock absorption, but poor energy release during push-off. The dynamic response feet offer good energy release during walking, however, only the expensive models have multi- planar movement.

Due to the rigidity and high cost of the feet, non rigid designs are necessary to improve the performance and reduce the cost of manufacturing. This can be achieved by the use of compliant mechanisms.Compliant mechanisms offer several potential design advantages over traditional rigid-body designs including high reliability and low cost. These design advantages are ideal for use in prosthetics.

A compliant mechanism uses the elastic deformation of materials to create motion. Normally, these motions are accomplished with rigid-body mechanisms, rigid bars connected by pins and springs. Compliant mechanisms are often more reliable, lighter, require less maintenance and may cost less to manufacture than their rigid body counterparts. These design advantages are ideal for use in prosthetics and with 3D printing prosthetic legs can be manufactured at a lower cost with short production time, making them more affordable and accessible. Another benefit of 3D printed prosthetic legs is that amputees can choose various designs for their artificial limbs.

II. LITERATURE REVIEW

Daher [1] conducted an extensive investigation in which nine types of SACH feet were subjected to cyclic testing to assess the durability of the materials and design of the foot until breakdown occurred.

Daher found that major permanent deformation and changes in resistance at the heel occurred within only 5,000 cycles.

Wevers and Durance [2] also conducted dynamic testing on prosthetic SACH feet, but they loaded the whole trans-tibial prostheses not the foot alone. Their results were similar to Daher's and structural component failures of the feet at less than 100,000 cycles.

Toh et al [3] avoided the complex loading. They utilized a simple machine which did not mimic gait but applied cyclic vertical loads to the heel and forefoot only.

Kabra et al [4] utilized a simple, low cost machine to fatigue the Jaipur foot, similar to Toh's device, however it appears to only simulate forefoot loading. A load-deflection analysis was also performed using a sling which passes around the foot , connects to a spring balance and reads the net acting force while the degree of movement was read from a ganiometer.

Daniel Rihs and Ivan Polizzi [5] utilized the impact test. The purpose of these tests to find the shock exerted onto the residual stump of the amputee at heel strike.

Glenn K. KLute, et al [6] studied the heel region properties of prosthetic feet and shoes. To measure and model the heel in response to impact, a pendulum was constructed to mechanically simulate the conditions immediately following initial heel ground contact during walking. A pendulum mass of 6 Kg was used to duplicate the effective mass of the stance limb at instant of heel ground reaction contact.

Francis J. Trost [7] investigated different materials that store energy when compressed by the body during early stance phase. The analysis includes measurement of the determinant of gait and oxygen consumption. Fifty two juvenile amputees were studied, the energy storing feet were provided including Flex-feet, Carbon copy feet. In evaluating specific activities, most amputees responded that running, jumping, climbing stairs were easier with energy storing feet.

III. METHODOLOGY

The objective of this thesis was to lay the groundwork for a comprehensive compliant mechanism prosthetic development processes. With the positive characteristics of compliant mechanisms the approaches and models will lower the cost and improve performance of prosthetic ankles.

The criteria of the prosthesis are 3D printable, low cost, simply geometry and satisfying mechanical properties for low activity use. So, a number of CAD models is designed and finite element analysis of the designed foot prosthesis is conducted.

The methodology to turn the function of a biological ankle during walking into a compliant mechanism concept is summarized in Figure 1.

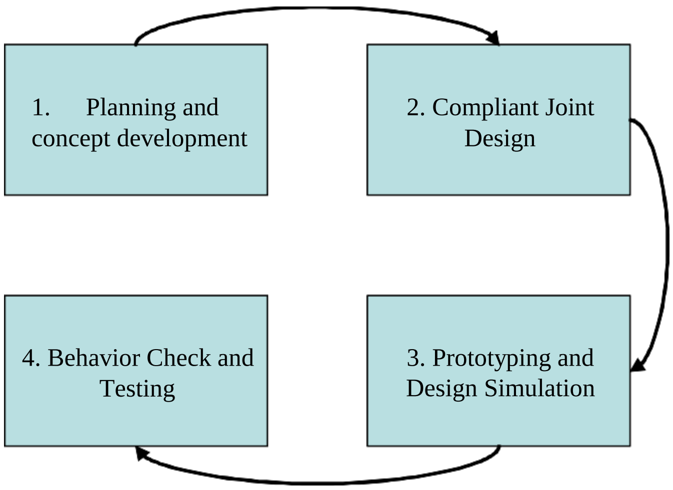
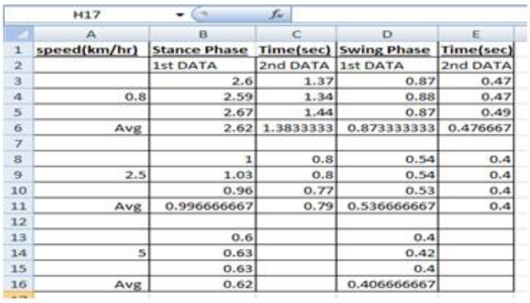


Figure 1: Development process for transforming biological ankle functions into prosthetic compliant mechanism concepts

1. Planning and concept development

In this phase, specific needs of the prosthetic ankle users are identified and biomechanics of the ankle, joint movements and various phases of gait cycle is studied. An experiment is carried out on Treadmill to get the time duration of stance and swing phase.

Figure 2: Volunteer walking on Treadmill



	A	B	C	D	E
1	speed(km/hr)	Stance Phase	Time(sec)	Swing Phase	Time(sec)
2		1st DATA	2nd DATA	1st DATA	2nd DATA
3		2.6	1.37	0.87	0.47
4	0.8	2.59	1.34	0.88	0.47
5		2.67	1.44	0.87	0.49
6	Avg	2.62	1.3833333	0.873333333	0.476667
7					
8		1	0.8	0.54	0.4
9	2.5	1.03	0.8	0.54	0.4
10		0.96	0.77	0.53	0.4
11	Avg	0.996666667	0.79	0.536666667	0.4
12					
13		0.6		0.4	
14	5	0.63		0.42	
15		0.63		0.4	
16	Avg	0.62		0.406666667	

Figure 3: Time duration of stance and swing phase.

2. Compliant joint design

The biomimetic design is mainly based on the human foot. In this phase various possible keel designs that fall within the defined design space is studied.

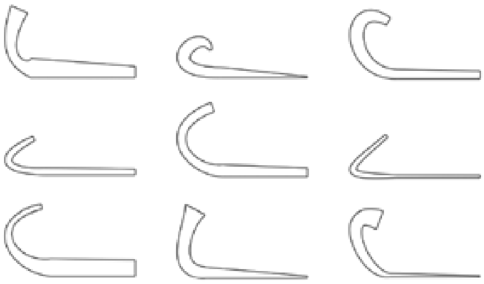


Figure 4: Various possible keel designs

3. Prototyping and Design Simulation

The modeling stage is used to verify that concepts, particularly product architecture, the function that it is designed to fulfill. This phase used finite element analysis to uncover weaknesses in product architecture.

A number of CAD models is designed, all adjustments of the model is followed by FEA, thus optimizing its strength. The split forefoot and the heel provide a 3-point support.

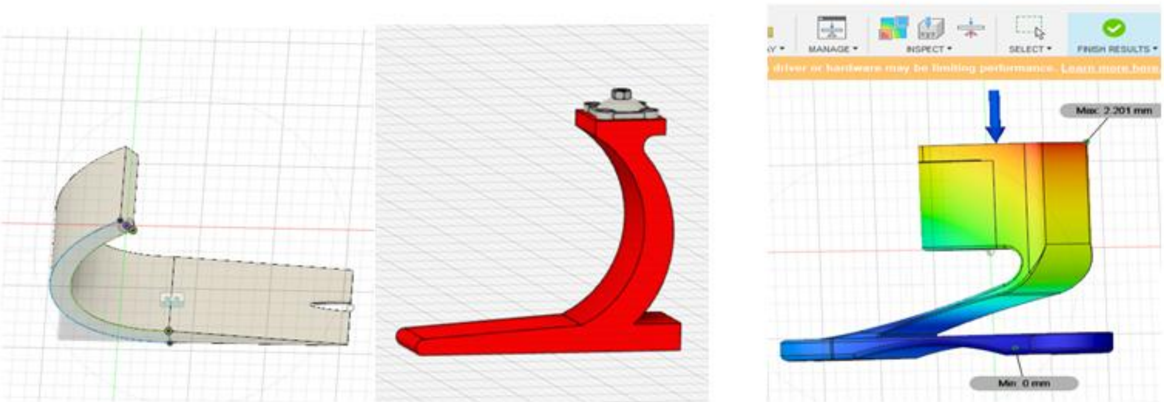


Figure 5: CAD Designs on Autodesk fusion 360

4. Behavior Check and Testing

This phase is out of the scope of the project considering the time and apparatus constraint. The prototype can be manufactured with a 3D printer. After the production, the prototype can be tested using tensile test machine and with patient.

Figure 6: Prototype testing

IV. RESULT AND DISCUSSION

Simulation is done on the various design models to know the produced stresses and displacement on the prosthetic under certain loading condition. Here, Heel strike (HS) is modelled with a fixed constraint on the sole of the heel and the 900 N vertical loads on the top surface of the mounting pyramid

Figure 7: Stress analysis

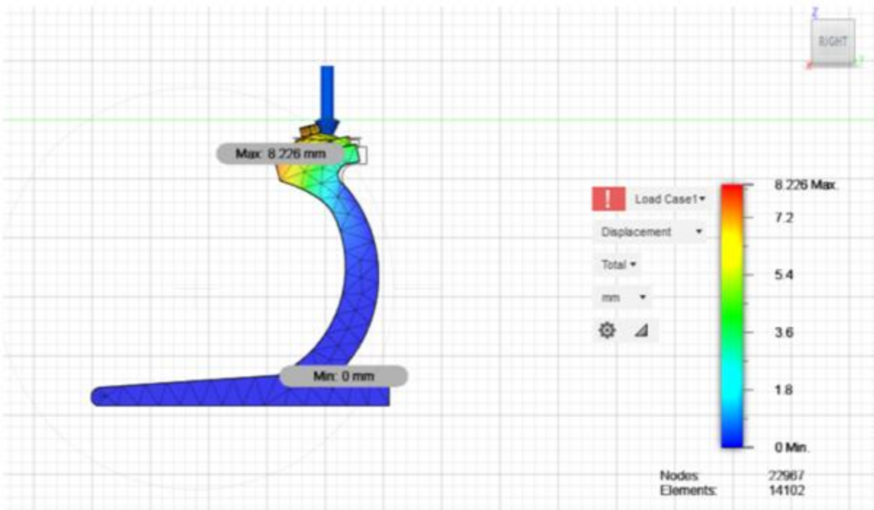


Figure 8: Displacement analysis

The maximum stress is computed at 15.31 MPa on the inner side of the ankle e.g. Fig 7 and the maximum displacement is 8.226 mm e.g. Fig 8.

In another design the FEA result shows the critical phase is the heel strike. The maximum stress is calculated at 11.54 MPa considering the auxiliary torque and the maximum lateral force besides the vertical load, e.g. Fig 9.

Shape optimization of a design shows the least material requirement under the optimum strength e.g. Fig 10.

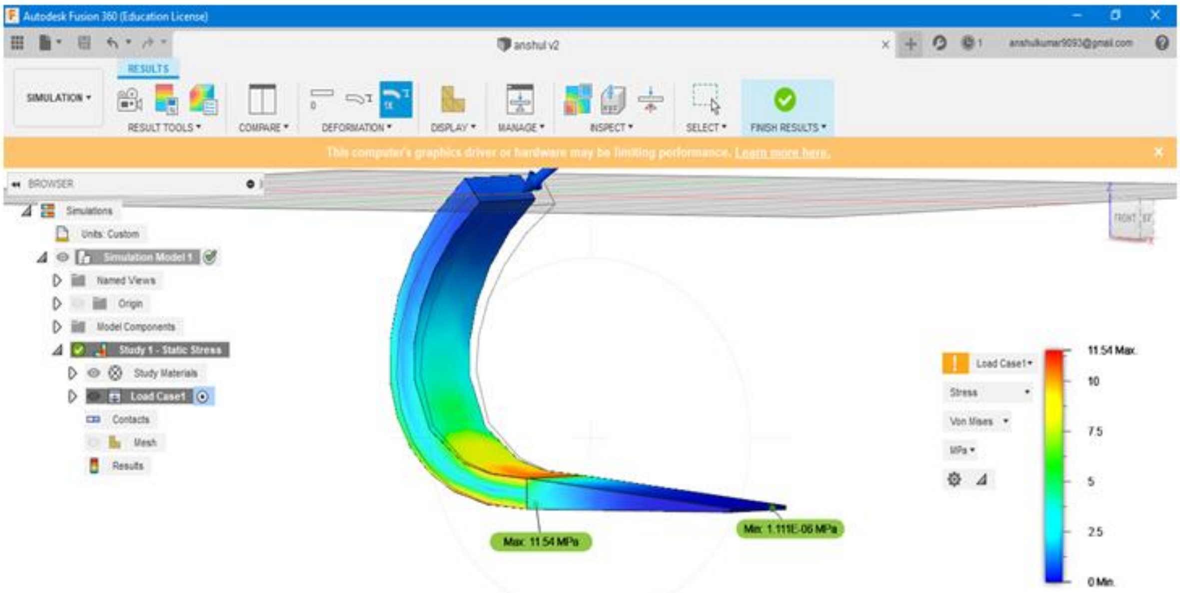


Figure 9: FEA Simulation

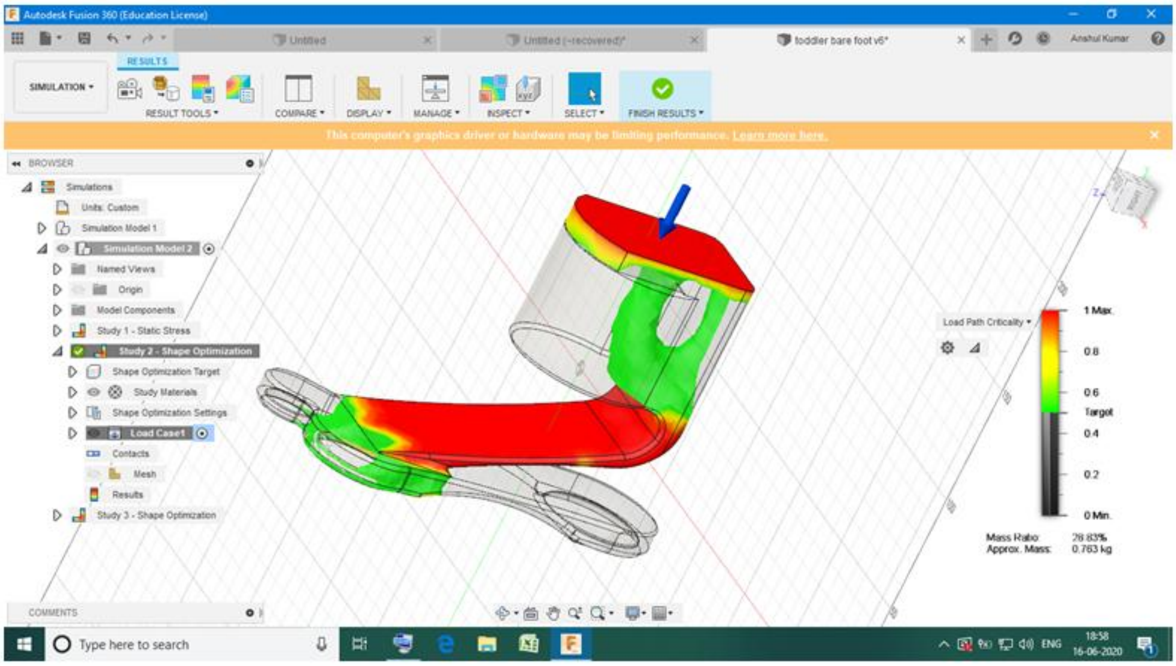


Figure 10: Shape optimizations of a CAD design

V. CONCLUSION

The FEA conducted on the prototype with 150% of presumed body weight showed that at HS the maximum stress was 15 MPa in the critical area. The FEA pointed out that the middle area of the foot does not bear enough stress compared to the critical ankle area. In conclusion, the 3D-printable prosthetic foot design presented in this paper shows that such products have a potential to be a low-cost solution for moderate activity level amputees. Considering that human feet weigh a few times more than the prototype, it is desirable to increase its mass in order to improve its strength and provide a more natural feeling for the user. The way to do so is adjusting printing settings such as wall thickness and structural infill percentage, as well as the geometry of the model, which is not the optimal yet.

Future work, based on this study will focus on testing the prosthesis and improving its strength and lifetime.

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