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SIMULATION OF AN ACTIVE ANKLE PROSTHESIS OF ONE DEGREE OF FREEDOM

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ABSTRACT

In this paper is presented the design of an active ankle prosthesis with a single degree of freedom (DoF) actuated by a linear actuator. The analysis is focused in the plantar base that shall allow the impacts absorption in gait cycle with the aim to replace the human lower limb of a person who has suffered ankle amputation. For the design it has been simulated the natural motion of a human ankle which trajectory is known for previous studies based in a biomechanical analysis of the same. For the static structural analysis has been used finite elements software ANSYS, obtaining data of stress, deformation and security factor which allow choose and couple properly the ideal linear actuator for the kind of patient and, the properly thickness of the plantar base with which the prosthesis will be built to avoid mechanic failures.

1 INTRODUCTION

Amputation of a limb is a permanent and traumatic event that changes people's lives. Amputations occur for different reasons, can be caused by diseases such as cancer, diabetes, vascular diseases, injuries caused by accidents. Diabetes is one of the diseases with the highest amputation rate in patients due to the damage caused by the disease in the arteries (Niemann 2016). Currently, there are about 1.9 million amputees according to the Amputee Coalition of Washington, DC and it is estimated that 185,000 new amputations take place every year. In base of the statistical data of the European Union, it is estimated that there are about 3 million amputees and 290,000 new amputations every year in these countries (Micera 2016). In Ecuador between 2007 and 2015 there were 1061lower limb amputations, according to the National Institute of Statistics and Censuses (INEC 2017). Through the kinematic study of the prosthesis, is determinate that a spherical joint allows realize rotational movements of the ankle both in the sagittal plane as in the frontal plane (Masum, Bjaumik, and Ray 2014). Kinetic and kinematic properties of human limbs can be analyzed through software to determinate ankle motion and compare with the foot orientation and the tibial coordinate system (Sinitski, Hansen, and Wilken 2012). Other authors propose improve performance of an ankle prosthesis using a stiffness analysis of the forefoot and hindfoot during gait cycle (Adamczyk, Roland, and Hahn 2017). In context with the analysis previously made by the authors mentioned above, the study of an ankle prosthesis is based in gait cycle, stiffness and torques that are generated in the ankle joint but it's not considered the deformation and stiffness of the planar base, for this reason this study propose the design of an ankle prosthesis with an linear actuator, obtaining data of stress, deformation, security factor and convergence analysis through a static structural simulation realized using ANSYS software for finite elements analysis, with the aim to maintain the natural trajectory generated by a human ankle to avoid failures in healthy joints, due to the gait cycle also depends of the motion of plantar base.

2 BACKGROUND

The biomechanical design of the human body allows a representation of each of the limbs by means of a diagram, in which it is taken as the origin of cartesian coordinates or systems of reference to the torso, since it constitutes the largest segment of the body, in addition of its central location that allows determining the orientation and position of the other segments of the body in relation to this point. To calculate the effort, anthropometric data of the person should be taken into account, such as age, weight, height, nutrition and level of disability in the joint (Paine and Sentis 2014). In the design of this type of prosthesis, it is feasible to use aluminum, titanium or polymers materials able to withstand elasticity, corrosion and be light to avoid wear of other joints.

In mechanical transfemoral prostheses, the ankle joint allows a slight flexion of approximately 5 ° to 15 °, obtaining a displacement of between 4 and 6 mm, with a bending moment at around 22 Nm (0,26 Nm / kg) (Fey, Simon, Young, Hargrove 2014). Where the actuator must possess the necessary characteristics to guarantee the stability and a natural movement of the ankle joint, one of the most important characteristics is the torque that varies according to the type of surface and the inclination of the same, that torque can reach up to a maximum of 1N.m / kg, See Figure 1 (Wise, Hobbelen, Schawab 2007).

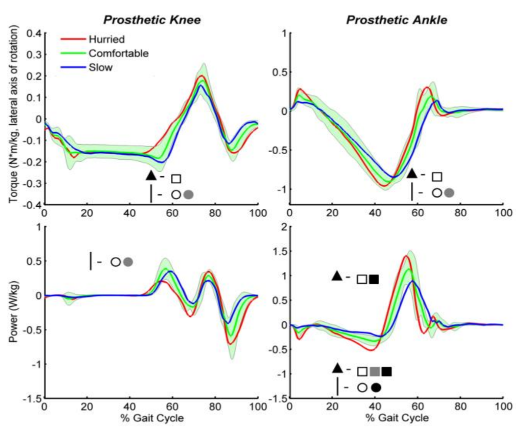

Figure 1: Torques and powers of knee and ankle during the walk on a slope. The shaded portion represents the standard deviation of the comfortable walking condition. The torque of the knee in terminal position and a maximum torque of plantar flexion of the ankle are compared as well as the power of the knee and ankle.

Different materials can be used for the design, such as stainless steel, which is ideal for the construction of gears, aluminum to construct the joint shaft and carbon fiber, which due to its resistance and flexibility properties is used for prosthesis plants (Calle, Cordero and Zambrano 2015).

3 DESIGN OF LOWER MEMBER PROSTHESIS OF A DEGREE OF FREEDOM.

3.1 Design specifications:

- The prototype design will be targeted to an adult weighing between 150 and 200 pounds, with a minimum height of 1.70 m and a shoe size of 40.
- The total height of the prosthesis must not exceed 12% of the height of the patient.
- The prosthesis will be constructed of carbon fiber for the foot and aluminum for the body of the foot, will have a 50 ° flexion restriction and 20 ° dorsiflexion, withstand flexion fatigue of approximately 1000 cycles per day.
- The actuator must be capable of generating torque at the ankle joint up to a maximum of 1N.m / kg.
- The type of plant to be designed will be focused so that the person can wear all types of footwear and be used in any terrain or ground.

3.2 Preliminary designs of one-degree lower limb prosthesis.

Figure 2 shows the initial designs of the ankle prostheses, which are based on concepts such as those of main components to give mobility of flexion angles and dorsiflexion.

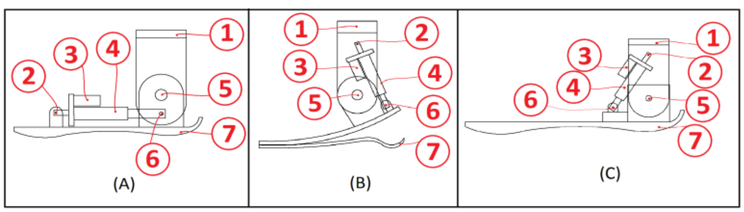

Figure 2: Sketches of prosthesis of lower limb, (1) Support for socket, (2) Upper joint of linear actuator, (3) Electric motor, (4) Recirculating ball screw, (5) Ankle joint, (6) Linear actuator lower joint, (7) Plantar base.

The considerations that have been taken into account to choose the design are based on the biomechanical analysis of the human ankle, as well as on the economy necessary for the patient's comfort in the use of daily footwear.

3.3 Analysis of design selection.

In Figure 2 (A) allows the actuator to be housed in the plantar base, allowing the upper part of the prosthesis to be a smaller size, which leaves the use of the device to persons of shorter stature.
The disadvantage of this design is given by the location of the actuator, which would generate a greater torque due to the weight that is located in a greater length with respect to the knee, as indicated by equation (1), which would cause a severe injury (Calle 2014).

$$T = \theta * I + m * g * l * \sin\theta \qquad (1)$$

Another disadvantage presented in Figure 2 (A) and 2 (C) is that these do not allow adjustment of traditional footwear, because the plantar base forms a single body which does not meet one of the main design features which is mimicry. The heel is one of the most important parts of the human foot because it absorbs the impacts during the cycle of the march due to its spongy physiology, as well as the bones and ligaments of the foot conform an elastic vault that allows to adapt to the irregularities of the ground and acts as a shock absorber on the fly, so sketches 2 (A) and 2 (C) do not meet these characteristics, the considerations to be taken into account are: a two-part plantar base for cushioning during take-off of the foot and first contact during the walking cycle, as well as the weight distributed at the top to generate less torque due to the length of the actuator with respect to the knee. Taking these concepts and the sketch of Figure 2 (B), the prosthesis is designed in the Inventor Software for its later simulation in the Ansys Software, as shown in Figure 3.

Figure 3: Final assembly of lower limb replacement prosthesis.

4 PREPROCESSING

4.1 Material

For the design of the prosthesis plant carbon fiber will be used for its properties such as strength, flexibility and weight, also based on concepts of structure and functionality of the real human lower limb (Carbon Fibers 2010), the characteristics are shown in the Table 1.

Table 1: Characteristics of the carbon fiber.

Property	Value	Unit
Density	1.9	g cm^-3
Young's modulus	241	MPa
Poisson's Ratio	0.25	-
Bulk Modulus	160.67	MPa
Shear Modulus	9.64	MPa
Tensile Yield Strength	2000	MPa
Compressive Yield Strength	2000	MPa

4.2 Restrictions

For the simulation, ANSYS software 17 has been used, in which three static structural type analyzes will be performed based on the critical conditions to which the prosthesis will be exposed in the development of the march cycle (Fatehi, Akbarimajd, Majdi 2010), see Figure 4.

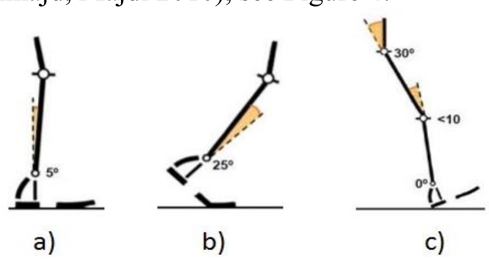

Figure 4: Critical conditions of gait cycle, a) total support of the plantar base, b) frontal support of the plantar base, c) posterior support of the plantar base.

Each position of the march cycle indicates the constraint that must be taken into account for each analysis, as well as the force that the actuator will exert in that position.

4.2.1 First Critical Condition Restrictions

The total support of the base is in the first critical condition (Figure 4a), exists for both the simulation of the constraints that are placed on the two supports as indicated in Figure 5:

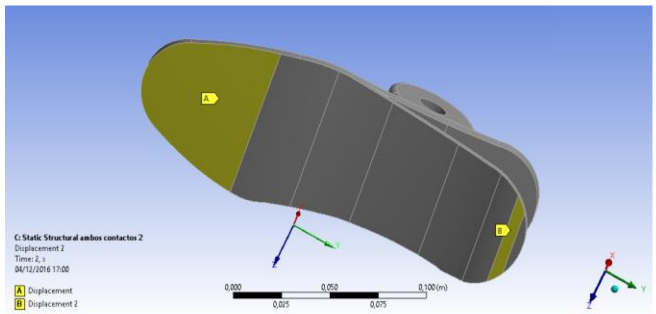

Figure 5: Displacement constraints in critical case 1.

The conditions of the restrictions are listed in Table 2.

Table 2: Conditions of travel restrictions in critical situation.

Axis	A(cm)	B(cm)
X	0	Free
Y	0	0
Z	0	0

When this critical point occurs, A and B points are in contact with the floor, for what is not possible to perform a side-to-side motion or a top to bottom movement, but it is possible to perform a displacement along the x-axis that can be forward and backward to point B; this is possible due to this region is designed to absorb impacts during gait cycle, such as performed by the human heel.

4.2.2 Restrictions of second critical condition

The frontal support of the plantar base is the second critical condition (Figure 4 b). Therefore, for the simulation the restrictions will be placed on the frontal support as indicated in Figure 6:

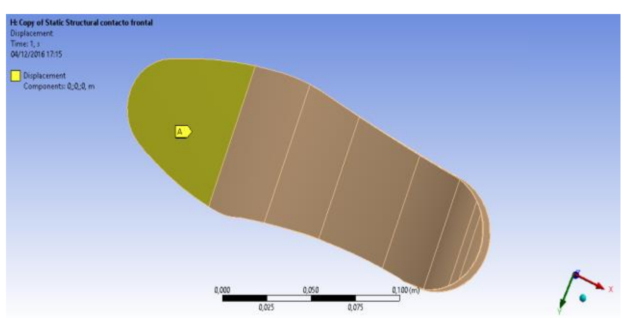

Figure 6: Displacement constraints in critical case 2.

The conditions of the restrictions are listed in the Table 3.

Table 3: Conditions of travel restrictions in critical case.

Axis	A(cm)
X	0
Y	0
Z	0

At point B, the x-axis remains free to allow deformation or displacement due to the loads.

4.2.3 Restrictions third critical condition

The posterior support of the plantar base is the third critical condition (Figure 4c). Therefore, for the simulation the restrictions will be placed on the posterior support as indicated in Figure 7:

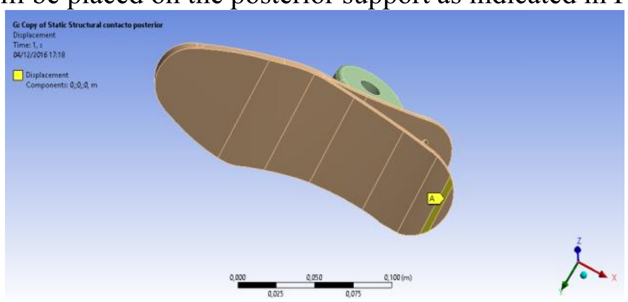

Figure 7: Displacement constraints in critical case 3.

The conditions of the restrictions are listed in chart 4.

Table 4: Conditions of travel restrictions in critical case.

Axis	A(cm)
X	0
Y	0
Z	0

At point B, the x-axis remains free to allow deformation or displacement due to the loads.

4.3 Loads on the system

For loads acting on the lower limb prosthesis, the weight of the person and the maximum value of the force exerted by the actuator will be considered, taking into account that the person's weight will be present in the three analyzes while the force of the actuator is only present in the second analysis because only in that phase of the human march generates a force in the actuator to complete the step during the walk. (Figure 8)

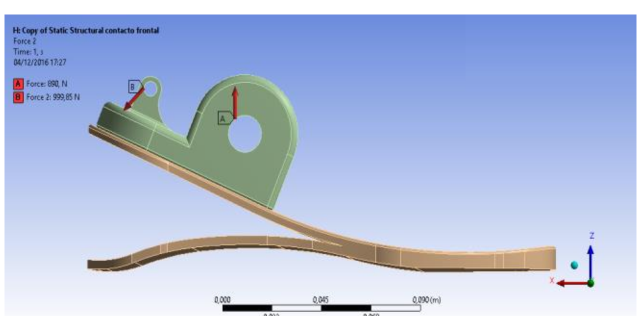

Figure 8: Loads acting on the prosthesis.

Figure 8 shows the load A of 890 N, which represents the force value exerted by the person's weight on the main axis that replaces the ankle, the load B of 1000N is the maximum force value exerted by the actuator in critical condition 2.

5 RESULTS OF SIMULATION WITH PRIMARY GEOMETRIES

5.1 Results of simulation with primary geometries

The simulation was performed, where the von Mises equivalent stresses were obtained. (Boldeiu, Vasilache, Moagar, Stefanescu, Ciuprina, 2015), which are the maximum and minimum efforts or stress that can be suffered by the material, as well as the value of the safety factor. In addition, convergence was applied to determine the real value of stresses by refinement of the meshing at critical points.

5.1.1 Critical case simulation 1

For the simulation of critical case 1, in Figure 9 we can see the location of the maximum and minimum values of von Mises pressure. It is also observed that the zone containing the maximum value fulfills the function of the human foot, since it is the place where it absorbs impacts and delivers energy during the cycle of the march.

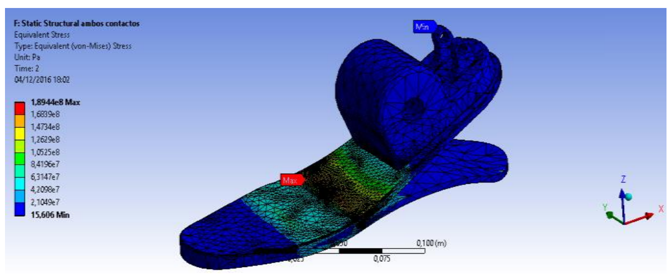

Figure 9: Maximum and minimum pressure values of von Mises.

For the verification of the results an analysis of mesh refinement and convergence was performed less than 5%. For the safety analysis and guarantee that the element does not fail during its operation, the safety factor analysis was performed, which resulted in a minimum value of 3.5. (Figure 10)

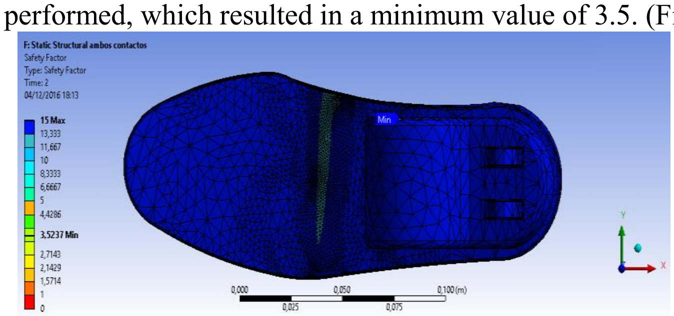

Figure 10: Safety factor case 1.

5.1.2 Critical case simulation 2

In critical case 2, the von Mises pressure values are shown in Figure 11, where the maximum and minimum values are detailed, these values are observed at the junction of the aluminum moving body and the plantar base of carbon fiber so the values are probably not true and are only referred to the union.

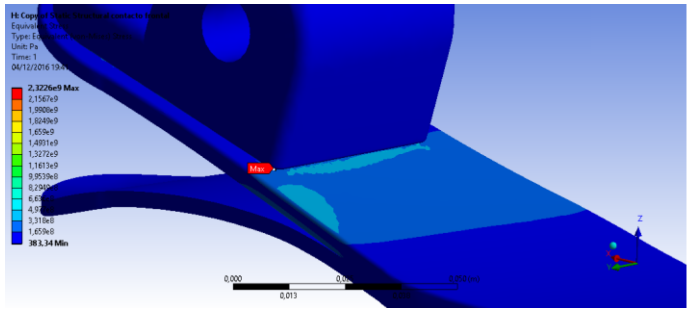

Figure 11: Maximum and minimum pressure values of von Mises.

An analysis of mesh refinement and convergence was performed with the objective of obtaining a value less than 5%. The safety analysis was performed using the safety factor, which resulted in a minimum value of 0.12. (Figure 12)

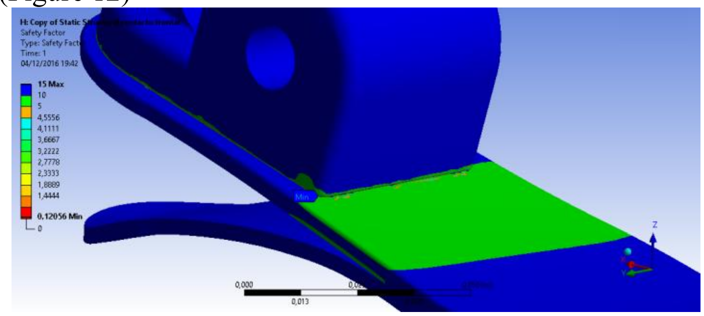

Figure 12: Safety factor case 2.

As it is appreciated the value of the factor of security is smaller than one. Therefore, the prosthesis plant in this conditions could not work and would probably fail.

5.1.3 Critical case simulation 3

For the simulation of critical case 3, we obtained values of von Mises pressure, as can be seen in Figure 13. In the same way as in case 2 the maximum values are in the union of two pieces, so they are not conclusive values of the simulation for this case:

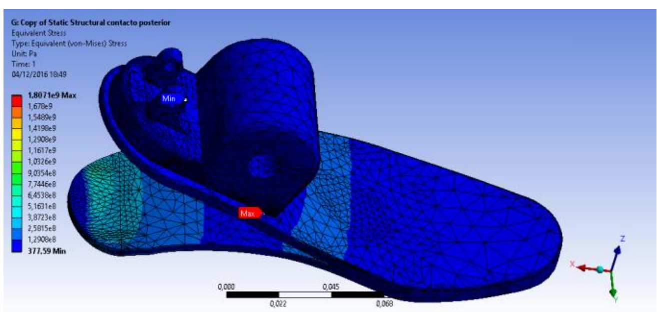

Figure 13: Maximum and minimum pressure value of von Mises.

A 5% mesh and convergence refinement was performed. For the safety analysis and guarantee that the element does not fail during its operation, the safety factor analysis was performed, which resulted in a minimum value of 0.15 (Figure 14).

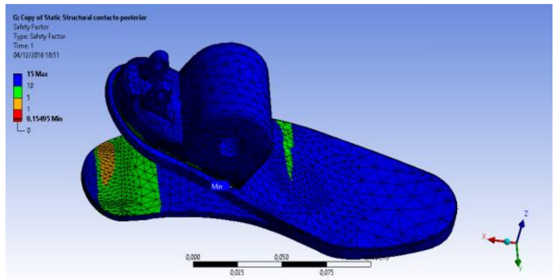

Figure 14: Safety factor case 3.

The safety factor for this case is less than 1, so the design features are not in the capacity to withstand the maximum loads to which it would be exposed.

6 PAGES SIMULATION RESULTS WITH MODIFIED GEOMETRIES

Given the values of safety factors less than 1 in cases 2 and 3, the thickness of the plantar base (Figure 5) was increased, which is coupled to the mobile base so that it can withstand the pressure during the cycle of the march.

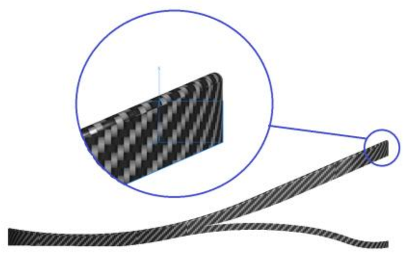

Figure 15: Increased thickness of 5 mm in the plantar base.

6.1 Critical case simulation 1

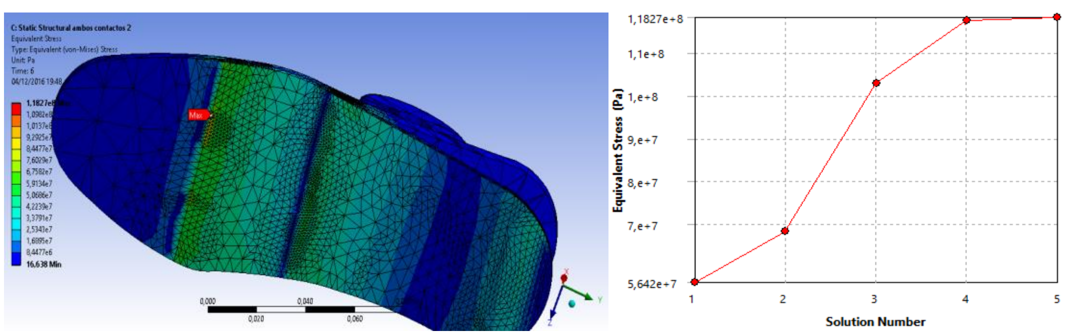

Figure 16: von Mises pressure and convergence analysis in critical case 1.

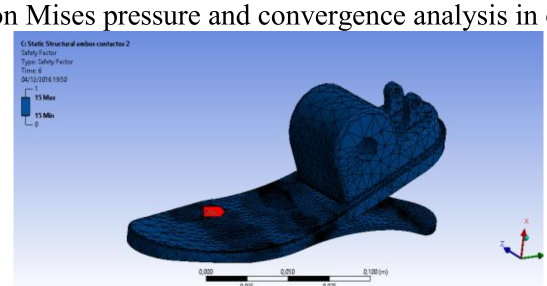

Figure 17: Safety factor value in critical case 1.

The safety factor that can be observed in Figure 17 is 15 so it is a very high value under design conditions, but because the study is based on other cases, we must observe the value of them to consider whether the geometry should be changed.

6.2 Critical case simulation 2

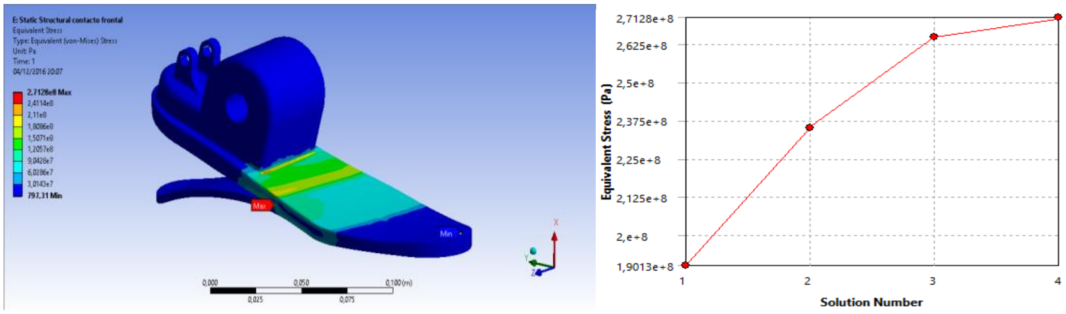

Figure 18: von Mises pressure and convergence analysis in critical case 2.

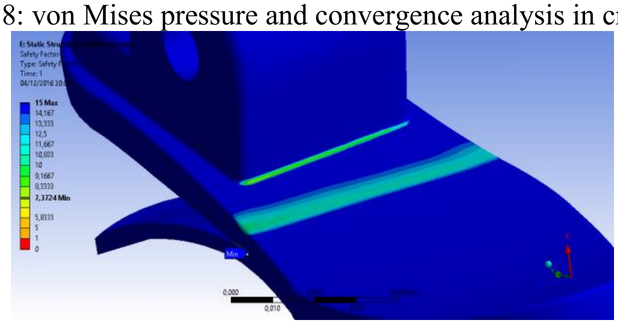

Figure 19: Safety factor value in critical case 2

Figure 19: Shows a minimum safety factor of 7.3 that meets the safety requirements.

6.3 Critical case simulation 3

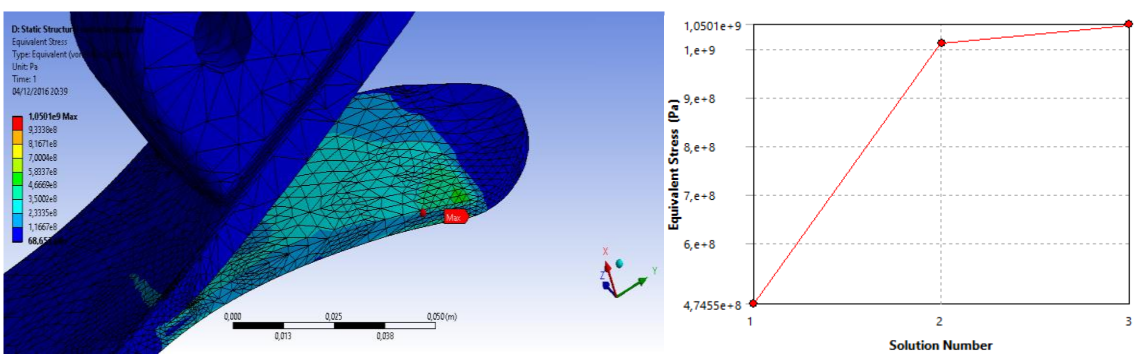

Figure 20: von Mises pressure and convergence analysis in critical case 3.

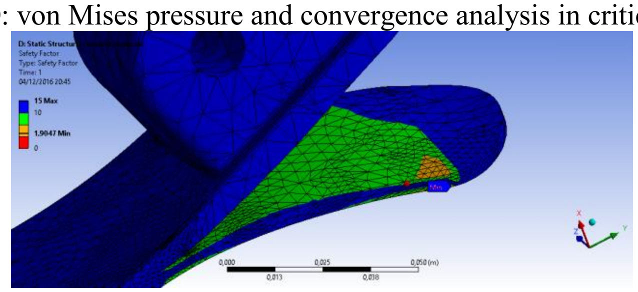

Figure 21: Safety factor value in critical case 3.

In Figure 21 the safety factor has a value of 1.9 which is much greater than the 0.15 obtained with the primary geometry

CONCLUSIONS

Static structural simulation has been able to verify the validity of the proposed geometries because it is observed that the most stress seen in the model is generated in the same regions that a human foot. Through the selection of three critical positions that the human foot experimented during the gait cycle it was possible to determinate mechanic failures in the plantar base. The analysis through stiffness measurement of von Mises determined that main geometries didn't comply with the necessary security characteristics for the performance in the critical conditions to which were subjected. As seen in the analysis of the critical case 1 with the modified geometry, highest security factors are in a value of 15, this is not conclusive to change the geometry for an excess of value because a security factor of 1.9 is obtained with the same geometry in a different critical case. Increasing the thickness of the plate of the plantar base with the mobile base to 5 mm , the more critical security factor was improved from 0.15 to 1.9 which guarantees the correct performance of the lower limb prosthesis.
Future works consist in implement the improved plantar base in a lower limb prosthesis, in addition, it will seek to build a multilayer plantar base based in carbon fiber and 3D print to reduce cost of manufacture.

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