Design Of A Mechanism For A Robotic Transtibial Prosthesis

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Abstract: This work focuses on a part of the design of a transtibial prosthesis. The components included are a mechanism and a prosthetic foot to be used by a person with transtibial amputation. The mechanism has two degrees of freedom which is capable of performing dorsiflexion/plantarflexion and inversion/eversion. Two serial elastic actuators (comprising two universal joints, two springs, one ball screw, and one motor) were used to obtain the ankle movement. To obtain an anthropomorphic design result we used the anthropometric data of a person of 56.7 kg weight and 1.75 m height. The parts of the prosthesis were modeled on Autodesk Inventor software and the finite element analyses were made in Autodesk Inventor Nastran. The limits of ranges of movement of the ankle mechanism are from 20° to -24° in dorsiflexion/plantarflexion and from 25° to -10° in inversion/eversion and it can provide a peak torque of 1.6 Nm/kg. As a result, the safety factor of the majority of the components analyzed was approximately five, and the weight of the design was approximately 4 kg.

Index Terms: Dorsiflexion, eversion, inversion, plantarflexion, robotic transtibial prosthesis, serial elastic actuator, transtibial amputation.

1 INTRODUCTION

According to the International Diabetes Federation, in 2017 approximately 425 million adults aged 20 to 79 years old lived with diabetes, of which 4 out of 5 people lived in low- and middle-income countries. Furthermore, it is predicted that in 2045 this number will increase to 629 million [1]. Approximately 15 percent of the people with diabetes worldwide will develop the common complication of the diabetic foot ulcer that can lead to a lower limb amputation [2]. In addition, there are other causes which lead to limb amputation, like traffic accidents or war consequences, thus, it is important for humanity to find in the short term, inexpensive and efficient solutions to this kind of problems. The passive or energy storage and return prostheses are lighter, smaller and less expensive than powered transtibial prostheses, however, these prostheses do not provide the necessary torque during push-off, thus, these prostheses increment the demand of metabolic energy from the patient [3]. Whereas, people with unilateral transtibial amputation who uses a bionic prosthesis decreases their metabolic cost and the leading biological leg mechanical work, while increases their preferred walking speed diminishing the differences in biomechanical aspects compared to the non amputee population [4]. Ankle inversion/eversion control can reduce the metabolic cost of patients by diminishing the balance-related effort. Indeed, by appropriately combining this control with dorsiflexion/plantarflexion control it might be possible to improve the balance and stability from combined perturbations, lateral impulses, and sloped terrain

[5]. Furthermore, the lateral movement of people with amputation seems to be less stable than that of people without amputation, hence, the inversion moments have a strong effect in this kind of movement enhancing the balance and reducing the fear of falling and fall rate of these people [6]. Furthermore, the non-straight steps represent approximately 35-45% of all steps taken in daily life activities, arriving with some of these at 50% due to architectural constraints [7]. This kind of steps requires simultaneous dorsiflexion/plantarflexion (DP) and inversion/eversion (IE) movements, being the torque for IE larger while turning than while walking in a straight line [8]. Therefore, by adding IE control, it is possible to increase the ambulatory abilities of people with transtibial amputation. Equally important, the lack of dorsiflexion control can lead to the fall or stumbling of the person with transtibial amputation on both horizontal and inclined surfaces [9]. In the last 12 years, many prostheses (passives or actives) have been developed from 1 to 2 degrees of freedom. For instance, a prosthesis of 1 degree of freedom was patented by researchers from the Massachusetts Institute of Technology (MIT) which can control the dorsiflexion/plantarflexion movement while delivers high mechanical energy by using a Series-Elastic Actuator (SEA) and a unidirectional parallel spring [10]. Some of these prosthesis has become commercial like the BioT2 System bionic prosthetic developed by MIT researchers and by the team A Step Ahead Prosthetics of the U.S.A. which controls the dorsiflexion/plantarflexion movement by a wireless communication system enabling it to modify the ankle stiffness and energy supply in real-time while in use [11]. In addition, the ProprioFoot commercial robotic prosthesis offers active dorsiflexion to increase the distance between the fingers and the ground and thus reducing the probability of a stumble or fall [9]. Moreover, some prostheses have active dorsiflexion/plantarflexion and passive inversion/eversion such as the one developed by researchers of the University of Manchester and the University of Salford, which is composed by a SEA, two spring systems and a novel universal joint system that enables it to generate these movements [12]. Other prostheses have 2 active degrees of freedom controlling both dorsiflexion/plantarflexion and inversion/eversion, among which is the Bowden cable-controlled prosthesis developed by researchers at the Michigan Technological University, the Mayo Clinic and the Mayo University Foundation of United States, having the advantage of allowing the placement of heavy parts of the prosthesis away from the distal parts and

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near the center of gravity of the user and consequently, attaining to reduce even more the metabolic cost [8]. In another case, the prosthesis SPARKy 3 developed by researchers of the Arizona State University, also has active dorsiflexion/plantarflexion and inversion/eversion given by two twin brushless DC motors and efficient and long-lasting roller screws with which it was sought to emulate as much as possible to the human ankle to give the amputee greater agility and so permitting him to develop his athletic abilities [13]. Therefore, considering the benefits of the DP and IE movements on the patient with transtibial amputation and the scientific advances presented above, we present the design of a mechanism for a robotic transtibial prosthesis with active DP and IE movements, followed by the finite element analysis of its components which bear the most critical forces.

2. MECHANISM DESIGN

As showed in Fig. 1, the mechanical design was modeled using Autodesk Inventor Professional which is an extensively complete CAD software. The foot design is similar to the one developed by Dobson, Wei, and Ren and it consists of: An ABS plastic sole (that can be manufactured with 3D printers) with cavities for pressure sensors which measure the weight of the amputee and serve to detect the current subphase of the stance phase in which the user is. A designed universal joint (DUJ) which is divided in an upper and a lower part, containing 2 crossed stainless steel shafts for IE and DP turns, and some pieces of rubber to counteract the impacts and limit the movements. An ABS plastic plate (that can be 3D printed) which is assembled to the sole and the DUJ. The DUJ is connected to the lower support of the serial elastic actuators (SEAs). Each SEA consists of: Two springs (one for dorsiflexion and one for plantarflexion). An inner stainless steel bar which guides both springs. A stainless steel tube which encapsulates the springs and the guide above mentioned. Two commercial MiSUMi UNCA14 universal joint set pin (CUJ) with 26000 N of static tensile failure load. A block of linear movement which connects the upper CUJ and the nut. An SKF SH 12.7x12.7 R precision rolled ball screw. Two ball bearings which hold the ball screw. A Maxon motor DCX35L GB KL 12V of 77.7 mNm nominal torque and 6A nominal current. A Maxon gearhead GPX37 LN of reduction 3.9:1 and maximum intermittent torque of 2.3 Nm. A rigid coupling between the ball screw and the motor. The structure comprises the following elements: Two inner steel bars of 8 mm diameter which guides the nuts. Outer aluminum plates which hold the aluminum supports. Inner folded aluminum plates which hold the heavy elements. An adapter to connect the prosthesis to the socket. The light elements (like electronic and control components) may be held by plastic supports which might be 3D printed to diminish the weight of the prosthesis. The prosthesis might be controlled by a microcontroller located behind the battery inside a case.

![Fig. 1. The CAD model of the mechanism (the electronic components have been hidden to show completely the mechanism).](image)

Fig. 2. Conceptual diagram of motion of the prosthesis for a stance phase divided into 4 subphases: (1) collision, (2) rebound, (3) pre-load and (4) push-off, and performed in A) the sagittal plane and horizontal ground and in B) the frontal plane and sloped ground of 15°. In B) some components have been hidden to permit the visibility of the mechanism.
1 Anthropometry
The measures of the transtibial prosthesis were obtained with the body segments of a model person of 1.75 m of height and 56.7 kg of weight [14]. The mean measure of a transtibial amputation is two-thirds of the total height of the leg [12], therefore, the desired height of the device is 333 mm. The data in Table 1 is used in the design to select the measurements of the mechanical components.

| TABLE 1 |
|------------------|------------------|
|                 |                 |
| **ANTHROPOMETRIC MEASURES** |                  |
| Knee height      | 498.8            |
| Ankle height     | 268              |
| Foot length      | 266              |
| Foot width       | 96.25            |
| Ankle to heel    | 59.8             |

2.2 Biomechanics of the Ankle
The human body can be divided according to its planes of movement which pass through the human body center, the one which divides it in upper and lower body is called transversal plane, the one which divides it from right to left is called sagittal plane and lastly, the one which divides it from the front to the back is called frontal plane [15]. Regarding these planes, the foot moves in DP, IE and adduction/adduction (AA) with respect to the sagittal plane, the frontal plane and the transversal plane, respectively (Fig. 3). According to several studies the range of movement (ROM) in dorsiflexion is 10 to 20°, between 40 to 55° in plantarflexion, until 23° in inversion and until 12° in eversion [16]. Finally, using the relation between the total weight of the body and the weight of the limbs of the model person, we obtained that the weight of the leg of 56.7 kg person is about 3.46 kg [14].

![](image)

Fig. 3. Types of movement of the foot.

2.3 Biomechanics of the Ankle During Gait Cycle
The ankle angles during gait at normal speed oscillate around 18° of dorsiflexion and 22° of plantarflexion and around 10° of inversion and 8° of eversion [16]. The maximum resultant reaction force among the foot and soil during gait cycle is 612.4 N approximately and the maximum DP torque applied in the ankle is loosely 1.6 Nm/kg walking at an average speed of 1.4 m/s while performing dorsiflexion [14]. For a man of 1.77 m height and 77.8 kg which walks at a cadence of 88 to 100 steps per minute, the maximum IE torque is about 7.5 Nm while performing eversion [17]. The gait cycle is divided into 2 phases, the swing phase, and the stance phase. Following this, the single support phase is divided into 4 successive subphases which are called collision, rebound, pre-load and push-off (Fig. 2) as stated by Kuo, Donelan and Ruina [18].

2.4 Kinematic Analysis for Dorsiflexion/Plantarflexion
For the following kinematic analysis, we used the analysis of the gait cycle in the sagittal plane with a duration of 0.972 seconds of a person of 1.75 m of height and 56.7 kg of weight, which shows the force, moment, angular position, velocity and acceleration parameters acting on the ankle and foot [14]. This analysis is carried out in a similar way as done in Madusanka’s work [19], using the diagrams of Fig. 4 to find the spring stiffness which minimizes the peak power of the actuators. In Fig. 4, O is the ankle joint, A and A’ are the axis (perpendicular to the sagittal plane) of the upper and lower commercial universal joints (CUJ) respectively, F<sub>g</sub> is the force applied by the spring of each ROPOMETRIC movement and U<sub>g</sub> is the axis which is obtained while intercepting a horizontal plane which passes by O and a vertical plane which passes through A and A’. Besides, the angles θ, φ and β are constants which belong to the triangles AOA’, AA’O and UOA, respectively. The angle α is made by the initial and final positions of the segment AO (as can be seen when comparing the black and blue lines in Fig. 4b). Subsequently, R is the length of segment AO, m is the length of UO, n is the vertical distance between A and O, γ is the angle formed by the initial and final positions of the segment AA’ and the origin of the XY reference system is in O where the angles are measured from the X-axis and are positive in counter-clockwise direction. Finally, X<sub>co</sub> is the final length of the spring and L is the half of the length of the CUJ added to the spring is base height.
By applying cosine law for triangle $A'AO$ of Fig. 4a and isolating $X_G$ we obtain,

$$X_G = \sqrt{Q^2 + R^2 - 2.Q.R \cos(\theta + \alpha) - 2.L}$$  

(1)

By differentiating (1) with respect to the time and isolating $\dot{X}_G$ we obtain,

$$\dot{X}_G = \frac{Q \cos(\theta + \alpha) - R \sin(\theta + \alpha)}{\sqrt{Q^2 + R^2 - 2.Q.R \cos(\theta + \alpha) - 2.L}}$$  

(2)

By applying again cosine law in the triangle $A'AO$ of Fig. 4b and isolating $\gamma$, we obtain,

$$\gamma = \arccos\left(\frac{R^2 - Q^2 - (X_G + 2L)^2}{-2.Q.(X_G + 2L)}\right) - \phi$$  

(3)

With the moment $M$ found with Winter’s analysis and applying summation of moments with respect to $O$, we obtain the force $F_G$ (Fig. 5),

$$F_G = \frac{M}{\cos(\gamma \cos(\beta)) + \sin(\gamma \sin(\beta - \alpha))}$$  

(4)

Subsequently, we follow the procedure proposed by Hollander and Sugar [20] by drawing the diagram of our SEA in Fig. 6.

With this diagram, we define the SEA position (or displacement) $X_p$ with (5), where $X_p$ is the displacement of the ball screw nut, $X_G$ is the spring initial length and $\Delta X_G$ is the spring deformation. $X_p = X_A + X_G + \Delta X_G + 2L$  

(5)

Using Hooke’s law by considering the spring stiffness $K_g$ we obtain $F_G$ with $F_G = K_g \Delta X_G = K_g (X_g - X_G)$  

(6) Isolating $X_g$ from (6) and (5) we obtain,

$$X_g = F_G/K_g + X_G - X_p - 2L - F_G/K_g - X_G$$  

(7)

Isolating $X_g$ we obtain,

$$X_g = X_p - 2L - F_G/K_g - X_G$$  

(8)

Differentiating (8) we obtain the linear speed of the ball screw nut, $\dot{X}_g = \dot{X}_p - F_G/K_g$  

(9) Multiplying both sides of (9) by $F_G$ we obtain,

$$F_A = F_G \dot{X}_A = F_G \dot{X}_p - (F_G/K_g)$$  

(10)

In (10) we can observe that the power $P_A$ transmitted by each SEA depends on the force $F_G$ and its change with respect to the time $\dot{F}_G$, the ASE speed $\dot{X}_p$, and the spring stiffness $K_g$. All these values, except $K_g$, are already determined by the mechanism initial measurements and by the data gait cycle data [14]. Therefore, we iterate giving $K_g$ values ranging from $10^2$ N/m to $10^7$ N/m which made the peak power go from infinite values to a convergence approximately equal to 120 W (Fig. 7). Between this range of values, we choose $K_g$ equal to $3 \cdot 10^4$ N/m which releases the minimum peak power equal to 50 W. Hence, choosing this value of $K_g$, the mechanism will release a maximum power of approximately 100 W considering both actuators. To proceed with the selection components and design is necessary to know the actuator displacement (or law of motion), speed and acceleration, which depends on the torque $M$ that the SEAs must provide during swing and stance phase, hence, to achieve this, we use the procedure described by Antonelli, Alleva, Zobel, and Durante [21]. According to this method, meanwhile the DP angles during swing phase can be achieved by simply controlling the actuator displacement, the DP torque during stance phase can be accomplished by computing the differences between the torque of the SEA as a passive system (by locking the actuator movement) and the torque given out by the model person’s ankle [14] which in this case have been transformed in the force $F_G$ applied by each SEA.

Fig. 5. Force applied by the spring of each SEA.

Fig. 6. Diagram of the mechanism SEA.

Fig. 7. Actuators’ peak power as a function of spring stiffness.

The force $F_{PAS}$ which generates the torque given out by the passive system is calculated by (11),

$$\Delta X_G = \frac{F_{PAS} - \overrightarrow{AA'}}{K_g}$$  

(11)

ence, as explained, the actuator displacement is calculated by (12),

$$X_a = (F_G - F_{PAS})/K_g$$  

(12)

Following this, we differentiate the displacement two times to obtain the actuators’ speed $\dot{X}_a$ and acceleration $\ddot{X}_a$ which we use along with the force $F_G$ to select the ball screw. Nevertheless, due to the high acceleration
required by the mechanism when using springs with stiffness \(3 \times 10^4 \text{ N/m}\) and that high speed ball screws are normally large and heavy, we decided to change the value of the stiffness of the springs to \(10^5 \text{ N/m}\) overcoming the acceleration issue, with which the peak power released by each actuator increases to 89W (value calculated using the SKF catalog due to the ball screw requirements). Thanks to this, we could choose a smaller ball screw and decrease the weight and size of the mechanism.

### 2.6 Analysis of Inversion/Eversion Movement

Taking into account the measures of the mechanism design we draw a diagram including the parts of the mechanism, which are involved in the IE movement (Fig. 8) to compute the force \(F_G\) necessary to generate the IE human ankle moment \(M\) of 7.5 Nm [17], with a human ankle rotation \(\phi\) of 8° [16]. Solving the summation of moments with respect to the axis \(UO\) where \(d\) is equal to 18.5 mm we obtained \(F_G\) equal to 205 N. \(F_G = M/(2. \cos(\phi). d)\) (13) This is the necessary force to generate the maximum IE torque during the gait cycle. The 8° rotation generates a spring deformation of 1.8 mm that provides a force equal to 180 N. This force along with a force generated by a little displacement of the motors can provide 205 N, therefore, the selected springs’ with stiffness \(K_G\) can be used for IE movement as well as for DP movement.

![Diagram](image)

**Fig. 8.** (a) Parts of the CAD model of the mechanism. (b) Mechanism diagram.

### 2.7 Components Design and Selection

Due to the complexity of the CUJ movement originated by the DP and IE movements, we utilized two springs (two per each SEA) encapsulated in 2 elements which assure that the springs won't bend while compressing. Besides, with these elements only one spring of each SEA is compressed at a time, as for the case when the universal joints get nearer, only the lower springs will compress to counteract the collision impact. The procedure proposed by Norton was used to design the springs and the values considered for the calculus were \(K_G\), \(A_{X'}\) which was obtained by isolating \(A_{X'}\) in (6), the A228 wire, the minimum, and maximum forces applied to each spring which are from 0 to 558.8 N in dorsiflexion and 0 to 52.33 N in plantarflexion (Fig. 5) and also the anthropometry measures aforementioned were considered [22]. Some ball screws were selected from the SKF catalog [23], and it was found that the ball screw which fulfilled all the requirements was the SKF SH 12.7x12.7 R precision rolled ball screw, with which the peak torque required by the actuator is 1.28 Nm and the maximum speed is 1266 RPM. The ABS plastic sole was divided into two parts in accordance with the center of pressures trajectory showed in Buldt’s analysis, with the aim of avoiding that one side of the sole fails before the other by distributing the pressure equitably in each side [24]. Aluminum plates were used to support the majority of the mechanical components, although, the steel bars which guide the nut block where the ones who bore the majority of the person’s weight. Thus, these bars were analyzed against the bend effects using Telmajer’s method, resulting in a security factor of 23. Therefore, the size of these bars may be diminished to decrease the weight of the mechanism.

### 3 Finite Element Analyses and Results

In order to expose the resistance of the mechanism against the forces acting on it during the gait cycle, we developed some finite element analyses (FEA) to the elements which bear the most critical forces using Autodesk Inventor Nastran software. First, the shafts of the DUJ and the set screw which holds the shafts together were analyzed. As showed in Fig. 9, a 600 N vertical load represented by a green arrow was located in the middle of the IE shaft and a 88 N load was located along the shaft in the center of it where the set screw pulls it towards one side. These forces originate the maximum resultant reaction force in the ankle during the gait cycle.

![Diagram](image)

**Fig. 9.** The meshing, forces, and constraints of the aluminum IE shaft.

Moreover, its ends were constrained replacing the ball bearings and a mesh of 0.5 mm element size was set, and consequently, we obtained a minimum safety factor of 94.3 which assures that the IE shaft won’t fail due to static load (refer to Fig. 10).

![Diagram](image)

**Fig. 10.** Finite element analysis of the aluminum IE shaft.

Following this, the designed ball-bearing house was analyzed setting an axial load of 560 N in the direction of the axis of the ball bearing and a radial load of 270 N in the inner surface of the ball bearing house. It was constrained in the holes for the bolts which will hold it, obtaining a safety factor lower than 2,
consequently, its material was changed from aluminum to stainless steel with which the safety factor increased to 4.5 (Fig. 11).

Fig. 11. Finite element analysis of the stainless steel ball-bearing house.
Subsequently, the springs aluminum lower support assembled with the lower part of the DUJ was analyzed. We set two pulling vertical forces of 560 N located in circular zones in the lower faces near the holes for the bolts which holds the lower support to the CUs. Besides, we constrained it in the holes for the bolted connections between it and the lower part of the DUJ, and as a result, we got a reliable safety factor of 6.7 (Fig. 12).

Fig. 12. Finite element analysis of the spring aluminum lower support.

Finally, the prosthetic foot assembly was tested by setting 600 N and 88 N in the vertical and horizontal directions, respectively, and constraining the sole in the two lower parts of it, obtaining a safety factor of 6.1 which is a very reliable value (Fig. 13).

4 CONCLUSIONS
The final weight of the mechanism was 15.6% more than expected and its height was 22% more than the goal. Hence, we will have to change the size and shape of some of the designed and selected components to obtain a more compact and lightweight mechanism. This is possible thanks to the fact that the mechanical components which bare the most critical forces have a high safety factor, permitting us to make those kinds of changes without risking considerably the patient's health. The limitation in the acceleration of the selected ball screw forced us to select more powerful motors which are larger and heavier, thus, we are compelled to change it to be able to achieve our weight and height goals by changing the motors selected. Once done these changes, and due to the fact that our mechanism is able to rotate properly in 20° of dorsiflexion, -24° of plantarflexion, 25° of inversion and -10° of eversion, our mechanism will be able to be the basis of an effective 2 degrees of freedom transtibial robotic prosthesis allowing the users to walk in sloped ground, make more pronounced body turns, and have a good stability without the cost of a high consumption of metabolic energy. As can be seen in Fig. 1, the prosthetic foot has anthropomorphic form, thus, allowing the people with transtibial amputation to use a variety of shoes or sneakers, with which the ABS plastic 3D printed sole and plate can be protected against different kinds of ground. However, due to that, the upper posterior part of the mechanism sticks out of this one, it might be necessary special clothes to cover it.

5 FUTURE WORK
As future work, we plan to continue with the following steps: Improve the mechanical design by iterations to find optimal values using specialized software to find the best shapes and sizes of the components or change materials. Proceed with the design of the electronics and control systems (electronic connections, control algorithms, architecture and so forth). Construct a prototype to test the mechanic, electronic and control systems with people having a transtibial amputation and in different types of ground with different slopes.
6 Acknowledgments

The authors would like to acknowledge the members of the Laboratory of Biomechanics and Applied Robotics of Pontificia Universidad Católica del Perú.

References