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Intelligent bionic leg for transfemoral amputation

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Abstract

This paper study the design and realization of a bionic leg for transfemoral amputation and is a contribution in the development of bionic legs especially in a world where the number of amputees is increasing for several reasons, accidents or illness. The bionic leg presented in this work ensures or approaches as closely as possible the role of the amputated leg of a handicapped person. It guarantees the safety and stability of the user, the comfort as much as possible of the action of walking. It has an aesthetic appearance that approaches as much as possible to the appearance of a natural leg. It is designed under the constraints of guaranteeing the availability of all spare parts for prior maintenance, guaranteeing the adaptability of the leg to angular morphological defects at the legs and ankles of a person, guaranteeing the adaptability of the leg to the rest of the body essentially for the length of the natural leg and being of acceptable purchase cost of possession, of use and maintenance.

The design process for this project began first of all with field study with amputees and specialists who treat amputees of this type in order to get precious ideas, especially on the medical side of the reaction of the amputee's body to this types of components. In parallel with these visits, a detailed bibliographic study on the analysis of the human walking cycle and existing prosthesis technologies.

Following this study, a functional and technical analysis of a system capable of providing or best approaching walking function of an amputee was developed. This analysis led to a technological choice that meets certain criteria imposed in relation with the amputee and his environment. The chosen solution is then dimensioned and drawn up in a detailed graphic file and is after that constructed.

Keywords: transefemoral amputation, bionic leg, design, biomecanics

Introduction

Aaccidents and diseases such as arterial disease, infections, trauma, tumors or birth defects are the major causes behind the increase in the number of leg amputations worldwide. The World Health Organization (WHO) estimates that there are 40 million amputees in developing countries.

To continue to carry out his activities an amputee often has recourse to an artificial leg which approaches the functioning of his amputated leg. For a transfemoral amputation things are even more complicated given the absence of the natural knee unlike a tibial amputation.

Thus and according to studies of the existing an artificial leg for a transemoral amputation can be a source of suffering and discomfort for its wearer if it does not meet essential conditions that may vary from one amputee to another. Given that the most basic function of lower limb prostheses is to provide the structural support that would otherwise be provided by the missing or removed part of the skeletal system.

Advanced prostheses will contain energy storage and release and physical cushioning. Features to provide stability control. Amputation is one of the main causes of disability in developing countries [1].

Different amputation consists of the removal of a limb or a segment of a limb. A distinction is made between “major” and “minor.

A prosthesis is an artificial medical device used to replace a missing part of the body. First of all, it begins with the appearance of the first men, they innovated the first aids or crutches to continue to stand even with limbs amputated or crippled, for example the wooden leg which has existed for 2000 years.

The oldest "prostheses" found date from the Egyptians. Greeks and Romans then also made. In the Middle Ages with mainly prostheses for fighting. The Renaissance and the revival of the development of the sciences (medicine, surgery) have largely contributed to a new development of the apparatus. But it is above all the industrial revolution of the 19th century and the two Great Wars which allowed the development of the prosthetic world and better "care" of amputees.

Functional specifications

Table 1

Functions	Assessment criteria	Level	Flexibility
FP: Ensure or approach the function of walking in an amputee	- Angular movement -Load applied by the	180° →90° 55kg→110kg	±5° ±10kg
FC1: Be adaptable to the amputee's thigh.	-thigh diameter -socket length	100→200mm 150→200mm	±10mm ±10mm

	-normal leg length -normal leg alignment -angular deviation of the normal ankle	100 → 112mm ±168° Dorsiflexion: 20° to 30° Plantar flexion:30° to 60 °	
FC2: Be of acceptable overall weight for an amputee according to his age.	Overall weight	3 kg	±50g
FC3: Be maintainable / give access to maintenance.	- spare parts availability -maintenance level -Access to detect operating faults (for systematic or conditional control) -Maintenance cost	Availability rate of spare parts (100%)	F0
FC4: Operate with a portable power source.	-Voltage -Intensity	U [V] I [A]	F0
FC6: Guarantee comfort and please the eye.	-color -Damping level of supports and shocks	Skin color	F1
FC7: Respect the environment (do not disturb or pollute).	-humidity -dust -temperature	ISO	F0
FC8: Guarantee the safety of the operator.		ISO45000	F0

Technical analysis

Technical functional analysis (AFT)

The objective of AFT is to find the way in which the product meets the functional specifications by highlighting the relationships that exist between the various components, which allow the technical functions necessary for the performance of the service functions to be carried out.

It is therefore a matter of choosing the solutions that achieve the expected performance.

The AFT comparing the following steps:

Search for solutions

Study of solutions

Solutions evaluations (technical-economic evaluation and reliability evaluation)

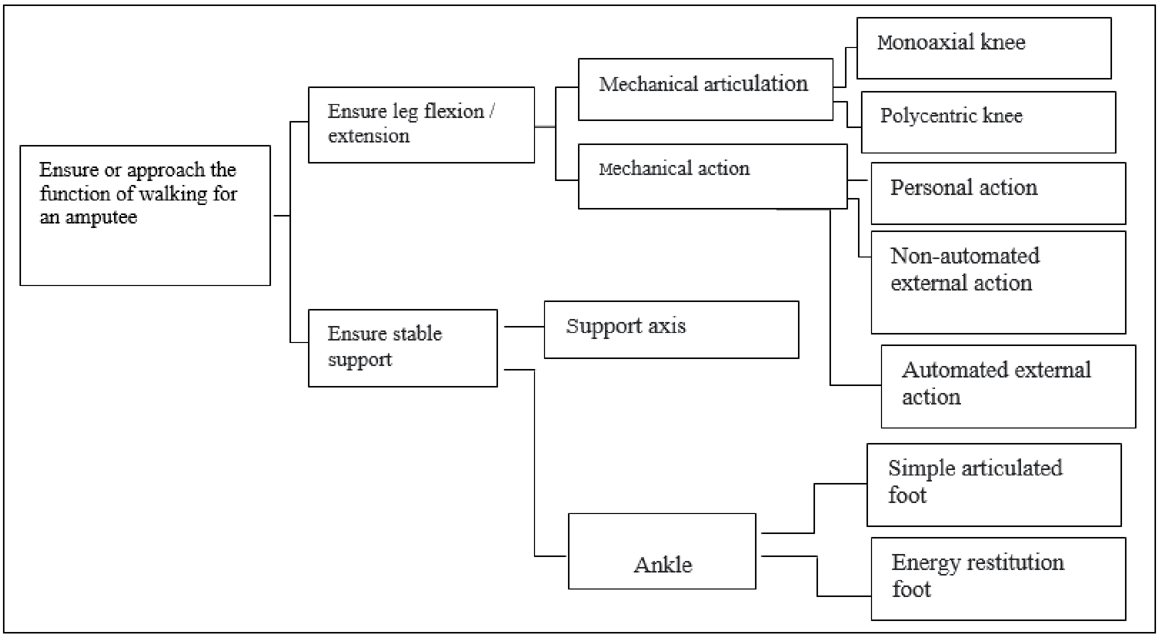


Fig 1

Calculation and sizing of the components of the leg solution retained from the previous study

Like any design project which breaks down into the pre-project phase and the study phase, the realization phase and the validation phase; our pre-project phase began with sketches of the overall bearing of the leg and mainly of the knee joint.

Below are some manual sketches which helped us to develop our first ideas.

Fig 2: Preliminary sketch

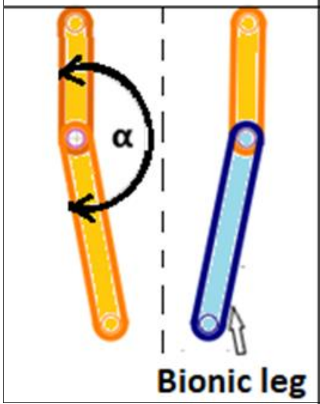
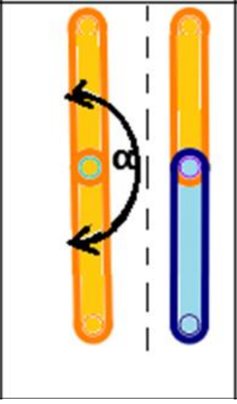
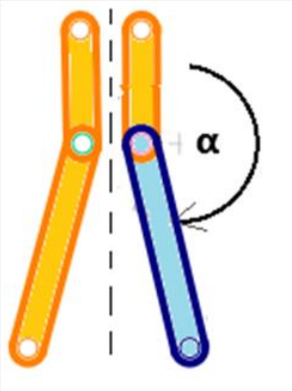
Description of the designed system

Two phases emerge when studying the walking cycle; the support phase and the oscillating phase. The cycle is defined as the period of time elapsing between two successive events of the same foot, it is therefore composed of two steps. Based on this data our system was designed.

So; our bionic leg system is made up of three essential parts. A polycentric knee system based on a four bar system, an electronic oscillation adjuster and an ankle equipped with an axis adjustment system.

This correction corresponds to the adjustment of the angle between the vertical axis of the socket and the vertical axis of the leg. The table below shows the different configurations that exist for angular morphological defects in a man. The angle α is adjusted with reference to it in the normal leg (see table below).

Table 2: Morphological defects symmetry of the legs

Legs with defect in "O"	Legs without morphological defect	Legs with defect in "X"
		

After adjusting the angle α, the amputee assumed to be standing (in a vertical support position), the cylinder rod is in the fully extended position and the spring at the ankle is in the total compression position, which represents the first phase of the walk cycle.

The system is equipped with two gyroscope sensors that indicate a zero in the upright position.

The gyroscope1 sensor is at the socket and is capable of detecting the variation in the angle of the socket from the vertical, this variation in angle being created by the amputee by making a slight pressure from the hip. As a consequence of this variation in angle, the cylinder rod begins to return to a well-defined angular position (detected by a gyroscope sensor 2) between the axis of the vertical leg. Once this angular position is reached, the order to extend the cylinder rod is given up to a maximum rod extension position which corresponds to full extension of the leg in the air.Pushing forward by the amputee causes contact with the ground at the heel of the ankle followed by full contact between the ankle mat and the ground (compressed spring); in parallel with this step, the amputee's normal leg is brought to rest on the tips of the fingers with the ground.

The next step is manifested in a normal advance of the leg, so the bionic leg is in contact at the level of the front point of the ankle, the socket will be brought forward, so the gyroscope sensor1 will detect the new angular position of the ankle. The socket in relation to the vertical and will give an order for the return of the cylinder rod up to the angle of release of the gyroscope2.

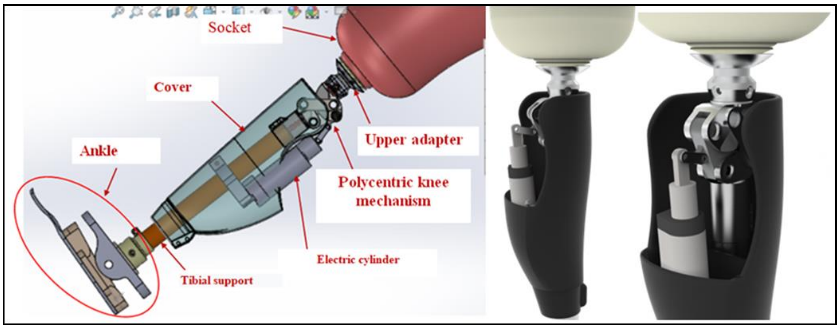


Fig 3: Leg subsets (solidworks model)

Below are real illustrations of the prototype of the intelligent bionic leg produced

Fig 4: First produced prototype

Here after a detailed description of the walking cycle with the bionic leg.

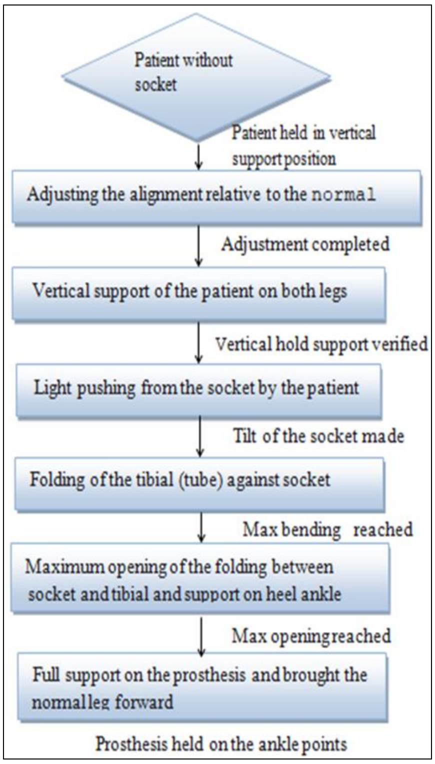


Fig 5: Scripting diagram of a walking cycle with the prosthesis

Kinematic modeling of the leg

The main design that has been used in our knee that the Jaipur knee uses is the four bar linkage system. The four bar linkage system that is used is a Type II, non Grashof, triplerocker. These linkage system is called "TypeII" because the length of the shortest and longest links, added together, greater than the added length of the two remaining links [3].

s + l> p + q

The linkage is considered a triple rocker because no link performs a full 360 degree rotation with respect to another link.

This type of four bar linkage works best because it provides the solid support that a knee should have when standing straight and it allows for the flexing motion of a real knee. This will give the user the ability to safely perform simple actions such as walking and sitting [4-5].

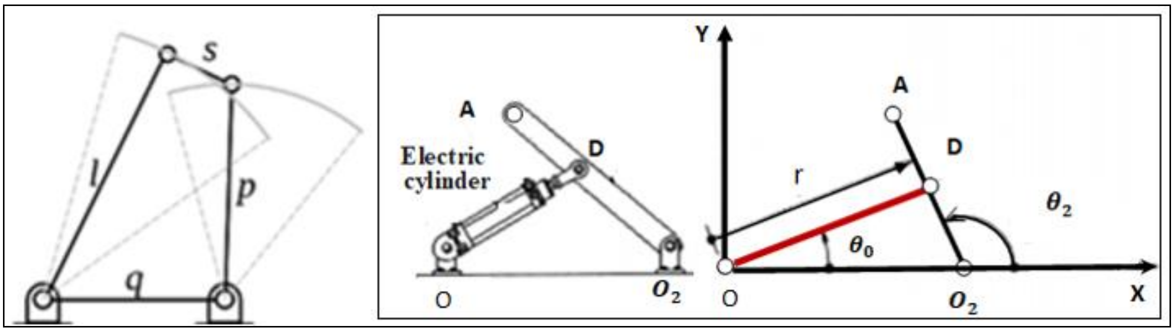


Fig 6: Double-rocker Four bar linkage and illustration of the input bar actuator (connecting rod)

The polyccentric knee system is actuated by an electric cylinder (see figure below)

The figure below shows the four-bar mechanism in the general coordinate system. The procedure for designing a four-bar linkage begins with the vector loop equation referring to the figure below. Position vectors are given as: $\overrightarrow{R_1}$, $\overrightarrow{R_2}$, $\overrightarrow{R_3}$ et $\overrightarrow{R_4}$.

The offset angle is denoted θ_0 and the entry angle is θ_2. Position vectors are used to obtain four bar linkages as in the equation:

$$\overrightarrow{R_1} + \overrightarrow{R_4} = \overrightarrow{R_2} + \overrightarrow{R_3} \qquad (1)$$

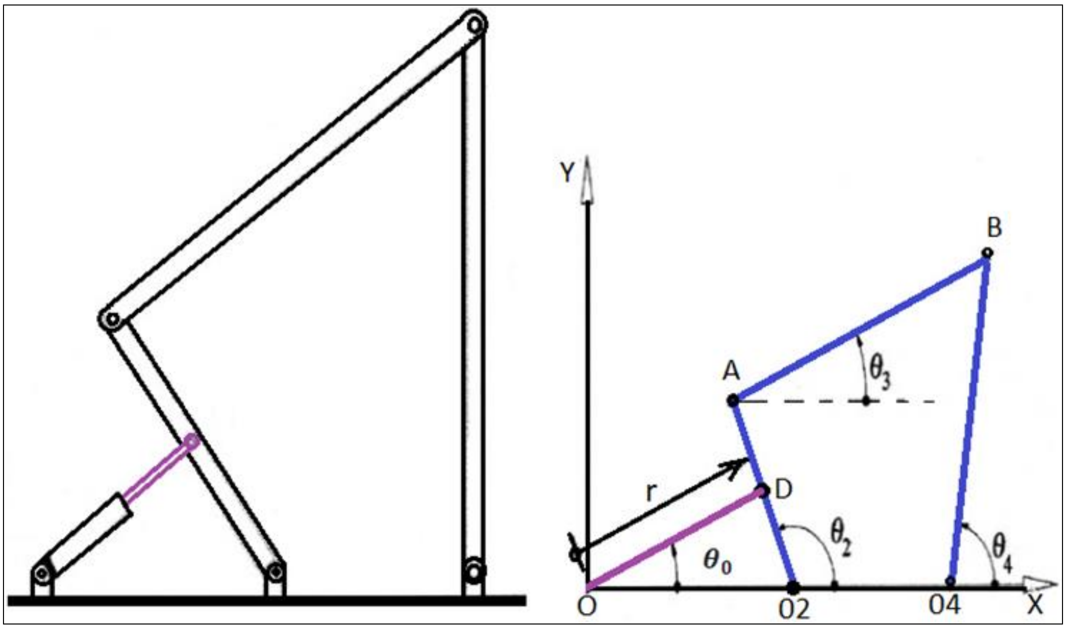


Fig 7: Polycentric knee four bar mechanism setup

The notation of complex numbers can then be used by using the scalar lengths of links such as r_1, r_2, r_3 and r_4. This relationship is expressed by the equation:

$$r_2 e^{i\theta_2} + r_3 e^{i\theta_3} = r_1 e^{i\theta_0} + r_4 e^{i\theta_4} \qquad (2)$$

Here θ_3 and θ_4 are the angles sought and they can be expressed as follows

$$\theta_3 = f(r_1, r_2, r_3, r_4, \theta_2, \theta_0) \tag{3}$$

$$\theta_4 = f(r_1, r_2, r_3, r_4, \theta_2, \theta_0) \tag{4}$$

For θ_0 = 0; (bar r1 is horizontal)
Equation (2) is expressed with its real and imaginary parts which are written as in the following equations:

$$r_2\sin\theta_2 + r_3\sin\theta_3 = r_4\sin\theta_4 \tag{5}$$

$$r_2\cos\theta_2 + r_3\cos\theta_3 = r_1 + r_4\cos\theta_4 \tag{6}$$

$$K_1\cos\theta_3 - K_4\cos\theta_2 + K_5 = \cos(\theta_2 - \theta_3) \tag{7}$$

$$K_1\cos\theta_4 - K_2\cos\theta_2 + K_3 = \cos(\theta_2 - \theta_4) \tag{8}$$

Where:

$$K_1 = \frac{r_1}{r_2}, \quad K_2 = \frac{r_1}{r_4}, \quad K_3 = \frac{r_2^2 - r_3^2 + r_4^2 + r_1^2}{2r_2r_4}, \quad K_4 = \frac{r_1}{r_3}, \quad K_5 = \frac{r_4^2 - r_1^2 - r_2^2 - r_3^2}{2r_2r_3}$$

$$\theta_3 = 2\tan^{-1}\left(\frac{-B \pm \sqrt{B^2 - 4AC}}{2A}\right) \tag{9}$$

$$\theta_4 = 2\tan^{-1}\left(\frac{-E \pm \sqrt{E^2 - 4DF}}{2D}\right) \tag{10}$$

In the equations above; The ± sign refers to two different configurations of the four-bar mechanism. The expressions A, B, C, D, E and F are then written as:

$$\begin{cases} A = \cos\theta_2 - K_1 - K_2\cos\theta_2 + K_3 \\ B = -2\sin\theta_2 \\ C = K_1 - (K_2 + 1)\cos\theta_2 + K_5 \\ D = \cos\theta_2 - K_1 + K_4\cos\theta_2 + K_5 \\ E = -2\sin\theta_2 \\ F = K_1 + (K_4 - 1)\cos\theta_2 + K_5 \end{cases} \tag{11}$$

Components resistance study

The tibial axis is a hollow tube of length l with an external diameter D and a smaller diameter = d the material is made of steel.

Condition of resistance:

$$\frac{F}{S} \text{ (Force applied)} < \sigma_u$$

$$F \leq \sigma \times S$$

$$F \leq \sigma_u \times \pi \left(\frac{D^2}{4} - \frac{d^2}{4}\right)$$

$$\frac{D^2}{4} \geq \frac{\text{F applied}}{\sigma_u \times \pi} + \frac{d^2}{4}$$

$$D \geq \sqrt{4 \times \left(\frac{750}{40 \times \pi} + \frac{20}{4}\right)}^{2} = 20{,}54 \text{ mm}$$

D = 22 mm

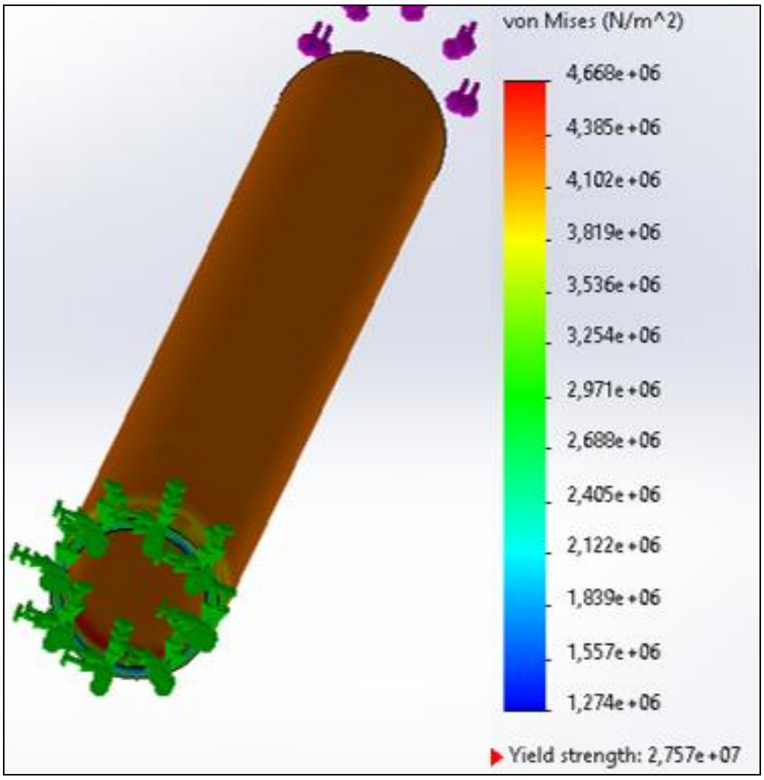


Fig 8: Checking the resistance of the tibial axis

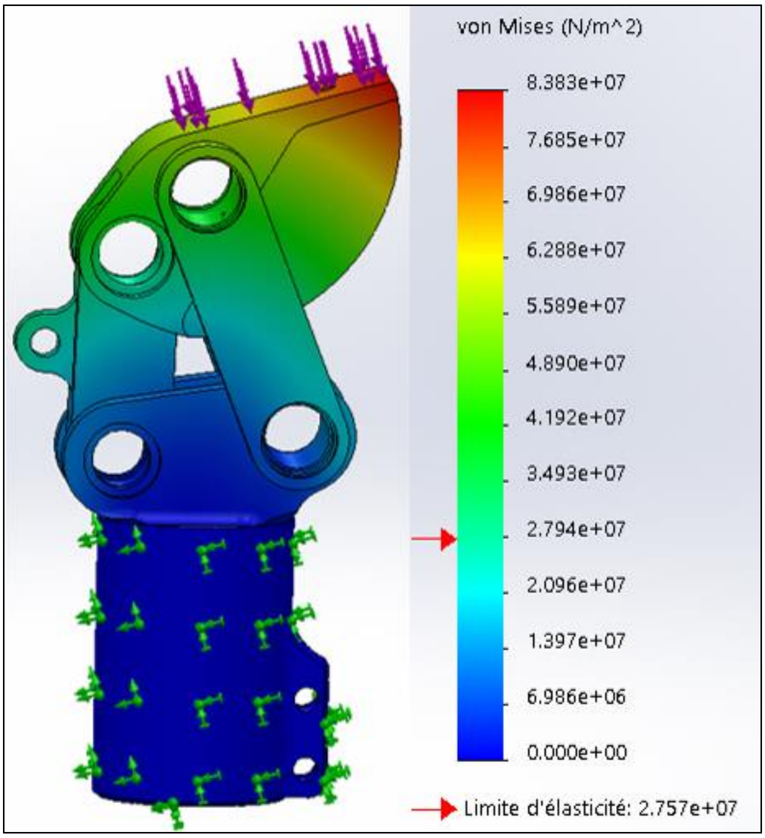


Fig 9: Checking the resistance of the group of bars

Sizing of articulation pins of the bars of the polycentric knee system At the level of the polycentric system of the knee (four bar system); the bars are articulated by rods / pins, below they are dimensioned by means of an RDM calculation.

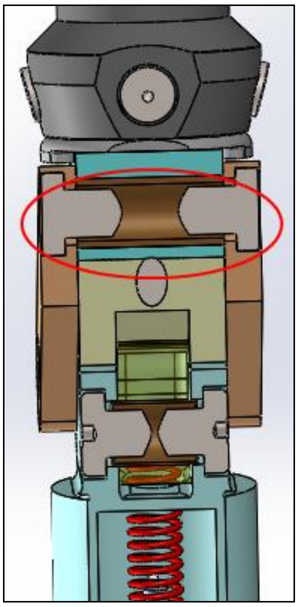


Fig 10: knee joint mechanism detail view

Below are the sizing details of the joint axis. the axis of the articulation is assimilated to an articulated beam subjected to distributed load as it is indicated in the figure below

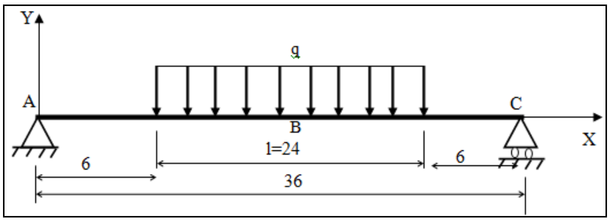


Fig 11

Static twists at different points:

$$\text{At point A: } \{\tau_A\} = \begin{Bmatrix} 0 & 0 \\ Y_A & 0 \\ Z_A & 0 \end{Bmatrix}$$

$$\text{At point B: } \{\tau_B\} = \begin{Bmatrix} 0 & 0 \\ -Q & 0 \\ 0 & 0 \end{Bmatrix}$$

$$\text{At point C: } \{\tau_C\} = \begin{Bmatrix} 0 & 0 \\ Y_C & 0 \\ 0 & 0 \end{Bmatrix}$$

Critical load is assumed for the upright position where all the body load is supported by the knees. The load Q represents the weight of the amputee.

At point A: $Q = q \times l$ and for a 75kg amputee and

$l = 24$

$Q = q \times l = 750N$

$q = 31.25\ N/mm$

Application of the fundamental principle of dynamics:

$$\sum \{\tau_i\}_A = \{0\}$$

Thus the torsors of actions and links will be expressed as follows:

At point A: $\{\tau_A\} = \begin{Bmatrix} 0 & 0 \\ 375 & 0 \\ 0 & 0 \end{Bmatrix}_A$

At point B: $\{\tau_B\} = \begin{Bmatrix} 0 & 0 \\ -750 & 0 \\ 0 & 0 \end{Bmatrix}_B$

At point C: $\{\tau_C\} = \begin{Bmatrix} 0 & 0 \\ 375 & 0 \\ 0 & 0 \end{Bmatrix}_C$

Cohesion torsors at different potitions:

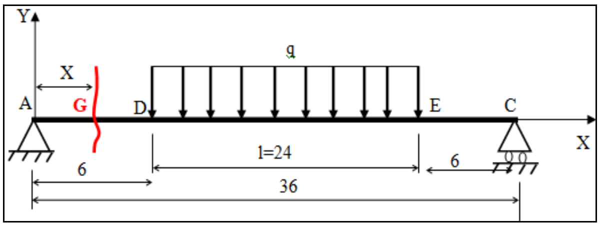


Fig 12

$\{\tau_{Coh}\} = -\{\tau_{left}\} = -\{\tau_A\}_G$

$\overrightarrow{M_G} = \overrightarrow{M_A} + \overrightarrow{GA} \wedge \overrightarrow{R_A}$

$\overrightarrow{M_G} = \begin{pmatrix} -X \\ 0 \\ 0 \end{pmatrix} \wedge \begin{pmatrix} 0 \\ 375 \\ 0 \end{pmatrix}$

$\overrightarrow{M_G} = \begin{pmatrix} 0 \\ 0 \\ -375X \end{pmatrix}$

$\{\tau_{Coh}\} = \begin{Bmatrix} 0 & 0 \\ -375 & 0 \\ 0 & 375X \end{Bmatrix}_G$

For $0 < X < 6$

$0\text{N.mm} < M_{fz} < 2250N.mm$

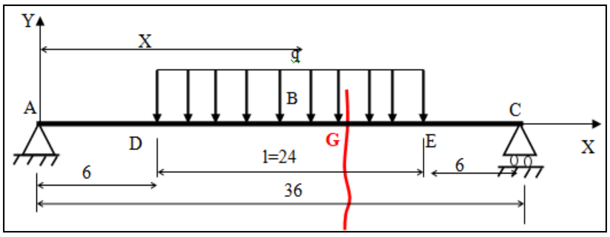


Fig 13

$\{\tau_{Coh}\} = -\{\tau_{gauche}\} = -(\{\tau_A\}_G + \{\tau_B\}_G)$

$\{\tau_A\}_G = ?$

$\overrightarrow{M_G} = \overrightarrow{M_A} + \overrightarrow{GA} \wedge \overrightarrow{R_A}$

$\overrightarrow{M_G} = \begin{pmatrix} -X \\ 0 \\ 0 \end{pmatrix} \wedge \begin{pmatrix} 0 \\ 375 \\ 0 \end{pmatrix}$

$\overrightarrow{M_G} = \begin{pmatrix} 0 \\ 0 \\ -375X \end{pmatrix}$

$\{\tau_{Coh}\} = \begin{Bmatrix} 0 & 0 \\ -375 & 0 \\ 0 & 375X \end{Bmatrix}_G$

$\{\tau_B\}_G = ?$

$\{\tau_B\} = \begin{Bmatrix} 0 & 0 \\ -q(X-6) & 0 \\ 0 & 0 \end{Bmatrix}$

$$\overrightarrow{M_G} = \begin{pmatrix} \frac{-(X-6)}{2} \\ 0 \\ 0 \end{pmatrix} \wedge \begin{pmatrix} 0 \\ -q(X-6) \\ 0 \end{pmatrix} = \begin{pmatrix} 0 \\ 0 \\ \frac{q(X-6)^2}{2} \end{pmatrix}$$

$$\{\tau_q\}_G = \left\{ \begin{matrix} 0 & 0 \\ -q(X-6) & 0 \\ 0 & \frac{q(X-6)^2}{2} \end{matrix} \right\} = \left\{ \begin{matrix} 0 & 0 \\ -48(X-6) & 0 \\ 0 & \frac{31.25(X-6)^2}{2} \end{matrix} \right\}$$

$$\{\tau_{Coh}\} = -\{\tau_{gauche}\} = -\left(\left\{ \begin{matrix} 0 & 0 \\ -375 & 0 \\ 0 & 375X \end{matrix} \right\}_G + \left\{ \begin{matrix} 0 & 0 \\ -24(X-6) & 0 \\ 0 & \frac{31.25(X-6)^2}{2} \end{matrix} \right\} \right)$$

$$\{\tau_{Coh}\} = \left\{ \begin{matrix} 0 & 0 \\ 375 & 0 \\ 0 & -375X \end{matrix} \right\}_G + \left\{ \begin{matrix} 0 & 0 \\ 31.25(X-6) & 0 \\ 0 & \frac{-31.25(X-6)^2}{2} \end{matrix} \right\}$$

$$\{\tau_{Coh}\} = \left\{ \begin{matrix} 0 & 0 \\ 375 & 0 \\ 0 & -375X \end{matrix} \right\}_G + \left\{ \begin{matrix} 0 & 0 \\ 375 + 31.25(X-6) & 0 \\ 0 & -375X - \frac{31.25(X-6)^2}{2} \end{matrix} \right\}$$

$$M_{fz=} \begin{cases} -2250 \text{ pour } X = 6 \\ 13500\text{N pour } X = 18 \end{cases}$$

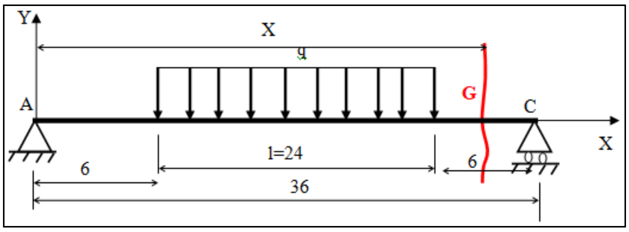


Fig 14

$$\{\tau_{Coh}\} = \{\tau_{droite}\} = \{\tau_C\}_G$$

$$\text{Au point C : } \{\tau_C\} = \left\{ \begin{matrix} 0 & 0 \\ 375 & 0 \\ 0 & 0 \end{matrix} \right\}_C$$

$$\{\tau_C\}_D = ?$$

$$\overrightarrow{M_D} = \overrightarrow{M_C} + \overrightarrow{GC} \wedge \overrightarrow{R_C}$$

$$\overrightarrow{M_D} = \begin{pmatrix} 36 - X \\ 0 \\ 0 \end{pmatrix} \wedge \begin{pmatrix} 0 \\ 375 \\ 0 \end{pmatrix}$$

$$\overrightarrow{M_G} = \begin{pmatrix} 0 \\ 0 \\ 375 \times (36 - X) \end{pmatrix}$$

$$\{\tau_{Coh}\}_G = \left\{ \begin{matrix} 0 & 0 \\ 375 & 0 \\ 0 & 13500 - 375X \end{matrix} \right\}_G$$

$$M_{fz=} \begin{cases} 13500\text{N.mm pour } X = 18 \\ 0\text{N.mm pour } X = 36 \end{cases}$$

Resistance condition

$$\frac{M_{fz}}{\frac{I_{Gz}}{\frac{d}{2}}} \leq R_{pe}$$

For a full circular section of diameter d the quadratic moment is:

$$I_{Gz} = \frac{\pi d^4}{64} \ ;$$

$$\frac{32M_{fz}}{\pi d^3} \leq R_{pe}$$

$$\sqrt[3]{\frac{32M_{fz}}{\pi R_{pe}}} \leq d_{min}$$

$$\sqrt[3]{\frac{32 \times 13500}{\pi \times 167.5}} \leq 9.36$$

d= 10mm

The same calculation procedures are followed to dimension the articulation axes at the level of the ankle

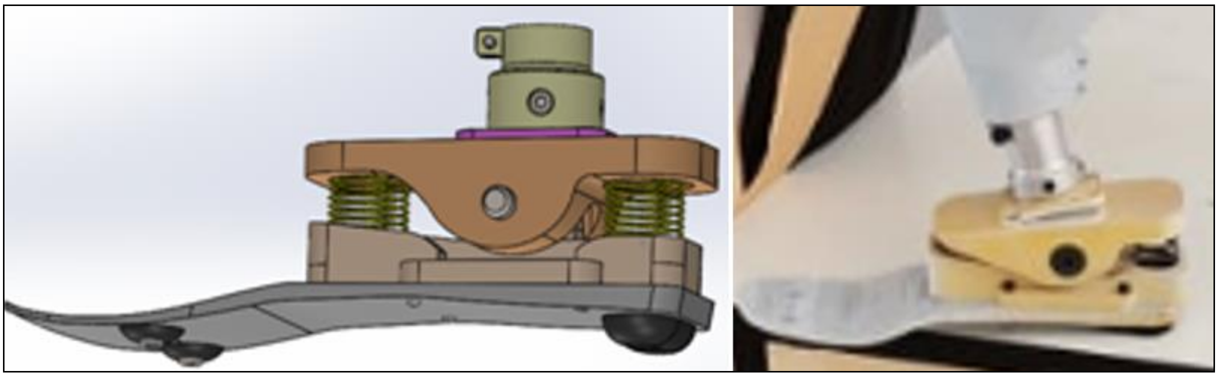


Fig 15: Ankle illustration

Conclusion

This work presents an approach for the design and realization of an intelligent bionic leg. We presented a contribution in the field of biomeanics with the aim of participating in the improvement of the comfort of a prosthesis and reducing the physical and psychic suffering of an amputee and this by presenting an automated product which by a simple action of the amputee can simulate the function of the lost leg in a walking cycle

The leg has been designed under constraints to guarantee its adaptability to motphological defects in the legs or ankles that arise with man in general. Thus adjustment mechanisms are considered and allow the adjustment from the beginning of use to approach the bionic leg to the normal one and ensure the symmetries of the both.

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