Low Cost Mechatronics Prototype Prosthesis for Transfemoral Amputation Controled by Myolectric Signals

Bravo Xavier, Comina Mayra, Tobar Johanna, Danni De La Cruz, David Loza, Jonathan Corella

Abstract— This article presents the development of low cost prototype prosthesis for transfemoral amputation leg, the prototype has not only low weight but also emulates the natural movement of a leg. It is activated by myoelectric signals obtained through the rectus femoris muscle of the amputated leg stump. Additionally, the prosthesis has a constant monitoring system of physiological parameters of temperature and humidity, this monitoring can be seen through an application for mobile phones. For this, the biomechanical analysis of gait was performed, after that the mechanical and electronic design, and finally test and results.

Keywords—Prosthesis, Transfemoral, Myoelectric, Physiological, Monitoring.

I. Introduction

The importance of this work lies in providing a solution to the problem faced by much people in Ecuador who have been affected by a transfemoral amputation, since at present people with physical disabilities in Pichincha province represent the 2.51% of the total population. The vast majority of these people has lack of economic resources so it

A transfemoral amputation prosthesis consists of three main parts: the socket, the knee, and the foot. The Socket is the only part of the prosthesis that is in direct contact with human skin and allows the prosthesis to attach to the body Research is currently underway on the development of new styles of sockets in order to improve the people comfort during the use of this coupling part; one of these innovations is detailed in the article named "Development of an immediate automatic lace system for artificial leg and modeling of a socket with Fuzzy Control "developed by Kohuei Najamura, Kazuhiko Sasaki and Kinya Fujita Tokyo University of Agriculture and Technologies members.

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is difficult to access a prosthesis that helps them walk, for this reason the social impact of the project, due to be a functional and economic prosthesis, is highly beneficial [1]

The socket has a coupling system by pneumatic compression whose rigidity can be controlled by the Fuzzy system that allows the real-time automation of pressure in specific areas of the socket. [2]

Another of the main parts is the knee, although the mechanical knees are capable of performing a quite adequate cycle, they present a deficiency adapting to these changes. The article "A Robotic Leg Prosthesis" by Brian E. Lawson integrates various electronic components, adds and coordinates the actions of a knee and ankle motor with the intention of replicating biomechanics of the healthy limb. [3]

Regarding the feet designs, that have position sensors to regulate the angles of foot movement according to the running cycle, are implemented; a clear example of this is developed in the title article: "Design of foot and ankle powered prosthesis By parallel elastic actuators "developed by Fei Gao, Wei-Hsin Liao, Bing Cehn, Hao Ma, Lai-Yin Qin. Its design improves the storage efficiency and release of mechanical energy by reducing the dimensions of the device. [4]

II. Biomechanical Analysis of the Gait

A. Analysis of parameters during the gait cycle.

Slow-cycle analysis is performed by placing markers at four reference points along the leg:

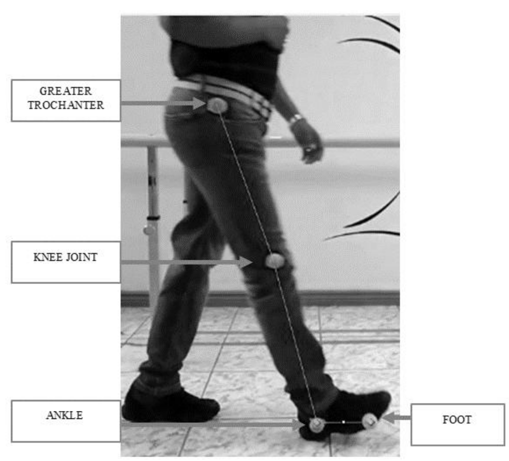

Fig 1. Reference points for gait cycle analysis

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These reference points are captured by a camera to generate a series of frames that after being analyzed allow to obtain the data of: speed, walking time, number of steps, step length, and acceleration; so they can be used in the design of parts and pieces.

Table 1 shows the results of the variables analyzed during the slow gait cycle.

TABLE I
GAIT CYLLE VARIABLES

Variables	Results
Time slow gait	3 seconds
Number steps slow gait	5 steps
Step length slow gait	510 mm
Speed slow gait	1.67 m/s
Acceleration slow gait	0.55 m/s^2
Maximum knee angle	65^0
Minimum knee angle	25^0

B. Force analysis

The forces analysis in the knee was performed considering the gait where the person can displaced by a horizontal plane with a slight inclination, taking into account the angle of variation that happens in the knee movement.

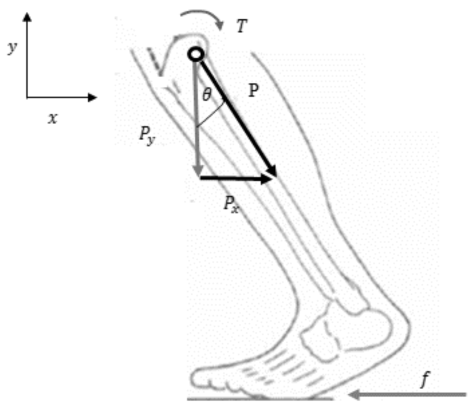

Fig 2. Load-displacement of the human leg

In Figure. 2 we can be observed the decomposition of the P force in the corresponding axes. P_xy P_y are the components of the prosthesis weight in the x and y axes respectively.

This analysis is necessary to find the torque that the motor should have to simulates the knee movement. For this, the following parameters are considered: person mass (m), the knee radius (R), acceleration (a), gravity (g) , the angle variation that occurs during the gait (θ), the motor efficiency (e) and the safety factor(f_s).

As a result of that previous process the necessary torque for the motor is obtained considering a safety factor of 2 and an efficiency motor of 90% :

$$T_m = f_s \left(\frac{100}{e}\right) \cdot T \qquad (1)$$

In the analysis of the movement intervals of human gait was observed that the maximum angle reaching the knee is 65^0 and the minimum angle is 25^0, considering 65^0at most, calculate torque shown below:

$$T_{fm} = 1.18 \text{ Nm}$$

C. Angular analysis of the leg during the gait cycle

A cycle starts when one foot touches the ground and ends when the same foot touches the ground again. The cycle is formed by two main phases: first, the support phase that starts with the initial contact of the heel and ends with the takeoff of the fingers after that the swing phase begins; that phase is defined as the time from takeoff to when the foot touches the ground again. (Sanz, 2017) [4]

Next, the angular analysis of the leg without amputation during the support phase is shown in Figure.3, considering the following steps:

- Heel contact: Refers to the moment when the heel of the reference leg touches the ground.
- Plantar support: refers to the contact of the forefoot with the floor.
- Medium Support: occurs when the greater trochanter is aligned vertically with the center of the foot, seen from a sagittal plane.
- Heel lift: occurs when the heel rises from the floor,
- Foot peeling: occurs when fingers rise from the floor. (Sanz, 2017)

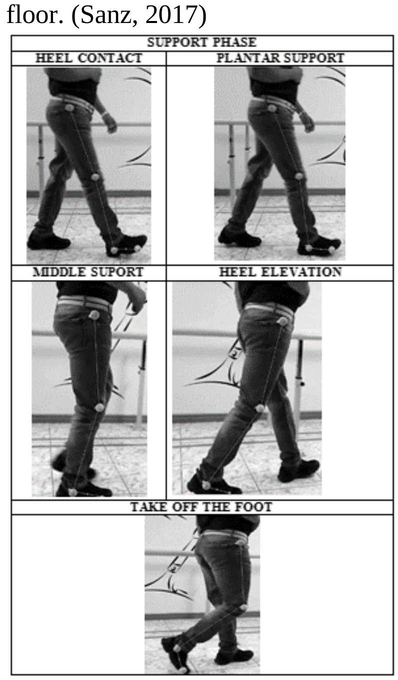

Fig 3. Leg support phase without amputation

Then, Figure. 4 shows the angular analysis of the leg without amputation during the swing phase considering its next steps:

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-Acceleration: it is characterized by the rapid acceleration of the end of the leg immediately after the fingers leave the ground.
-Medium swing: the balanced leg passes to the other leg, moving forward of the same leg, since it is in the support phase
-Deceleration: the leg stops quickly when it approaches the end of the interval. (Sanz, 2017)

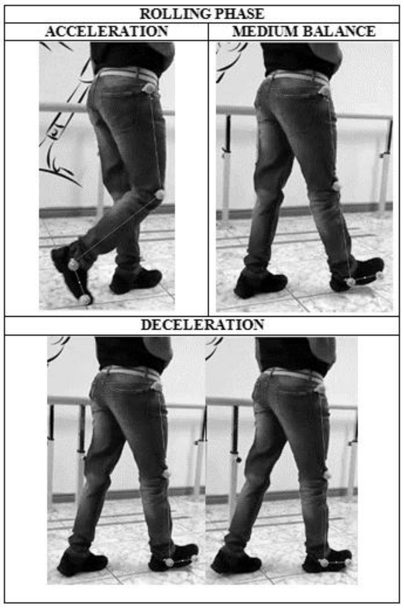

Fig 4. Swing phase of the leg without amputation.

III. Diseño Mecánico

A. Traction Test – Material Compression

The material used in the prosthesis is Aluminum. This material was subjected to tensile and compression tests, in order to determine if the material will support the 80 kg of the person, who will use the prosthesis.
The tensile and compression tests were performed in a universal test machine, in which the following results were obtained: maximum compression value of 9000 kg and traction of 120 kg. With these data, it was possible to conclude that the 120 kg that the molten aluminum specimen supported in the tensile test is the critical value in relation to the 9000 kg that are supported in the compression test. So the ultimate strain is then calculated as shown in equation 2.

$$\sigma_u = \frac{P}{\pi\, {d_o}^2/4} \tag{2}$$

Then the values are replaced assuming that the specimen had an initial diameter of 6.40 mm.

$$\sigma_u = \frac{P}{\pi\, {d_o}^2/4}$$

$$\sigma_u = \frac{1176\ \mathrm{N}}{\pi \cdot (6.40\mathrm{mm})^2/4}$$

$$\sigma_u = 36.55\ \mathrm{MPa}$$

As observed the ultimate strain is 36.55 MPa, value which was higher in relation to Von Mises' efforts than was obtained in the finite element analysis performed on each of the elements of the prosthesis. Which allowed to verify using a real data that the prototype will support the efforts for which it was designed

B. CAD Model

The knee and the foot are the main elements of the prototype. On the knee a mechanical brake was incorporated to a servomotor that allowed to control the return of the prosthesis, this because the inertia that is generated during its operation emulating the movement of a human leg. The foot is formed by a simple articulation that allows to regulate the angles with which it moves, improving stability when walking. Figure 5 shows the CAD model of the prototype

Fig 5. CAD Model

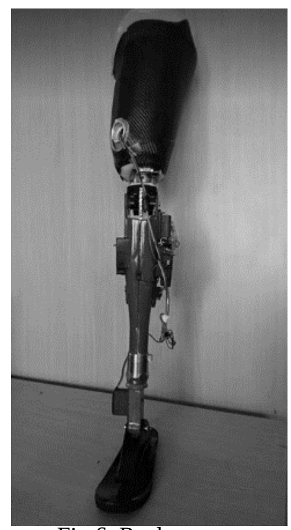

Fig 6. Real prototype.

C. CAE Analysis

The finite element analysis was performed in a software to observe the behavior of the parts under static and fatigue loads as seen in Figure. 7

The safety factor against fatigue based on finite life was calculated by applying Modified Goodman's equation.

$$n_f = \frac{1}{\frac{\sigma_a}{S_e} + \frac{\sigma_m}{S_{ut}}} \tag{3}$$

These values were used in the mechanical design phase for static load and fatigue analysis, obtaining the safety factors specified in the Table II.

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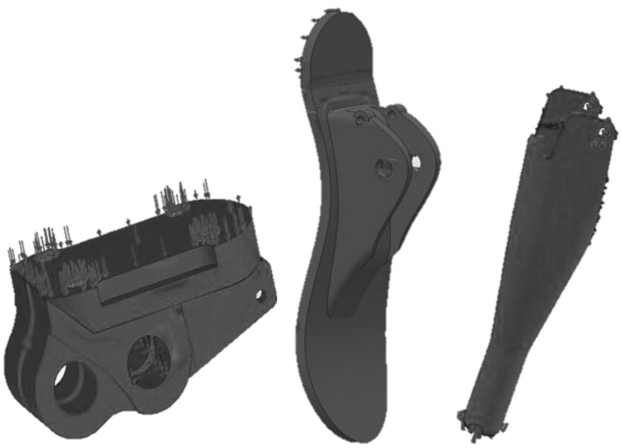

Fig 7. FEA Analysis

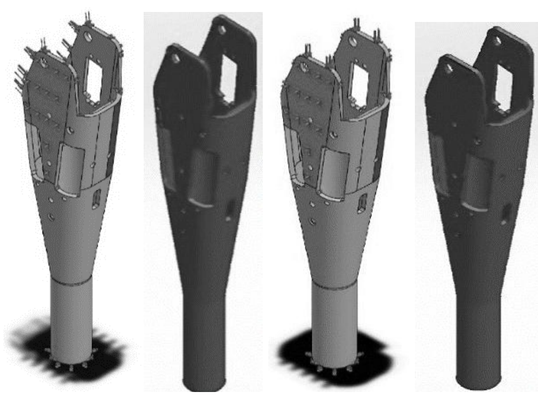

Fig 8. Stem FEA Analysis

TABLE II
CAE ANALYSIS RESULTS

Parts	Tension (GPa)	Displacement (mm)	Unitary Strain	Minimum Safety Factor
Socket coupling joint	14.5	1.097 x 10-2	1.635 x 10-4	1.9
Knee coupling joint	16.2	1.445 x 10-3	1.741 x 10-4	1.7
Knee	2.55	1.906 x 10-3	2.823 x 10-5	1.7
Steam	11.3	6.973 x 10-4	1.133 x 10-4	2.5
Ankle	9.58	1.058 x 10-2	9.042 x 10-5	2.9
Foot	19.9	1.066 x 10-2	2.085 x 10-4	1.4
Knee axle	10.6	6.356 x 10-3	3.832 x 10-4	1.62
Foot axle	0.37	1.083 x 10-5	1.44 x 10-6	1.62

IV. ELECTRONIC DESIGN

In order to activate the prosthesis, we obtained the signals from the anterior rectus muscles and the vastus external, with the purpose of selecting the muscle that generates better biopotentials, as can be seen in Figure. 9, there is greater repeatability in the measurements of the anterior rectus muscle, whereby this was selected to give the signal of activation to the prosthesis

The prosthesis was activated by the myoelectric signals obtained through the anterior rectus muscle of the amputated leg stump, thus this way the microcontroller can send the signals to the knee and hell servomotors to start the movement when it detects a potential difference when contracting the muscle.

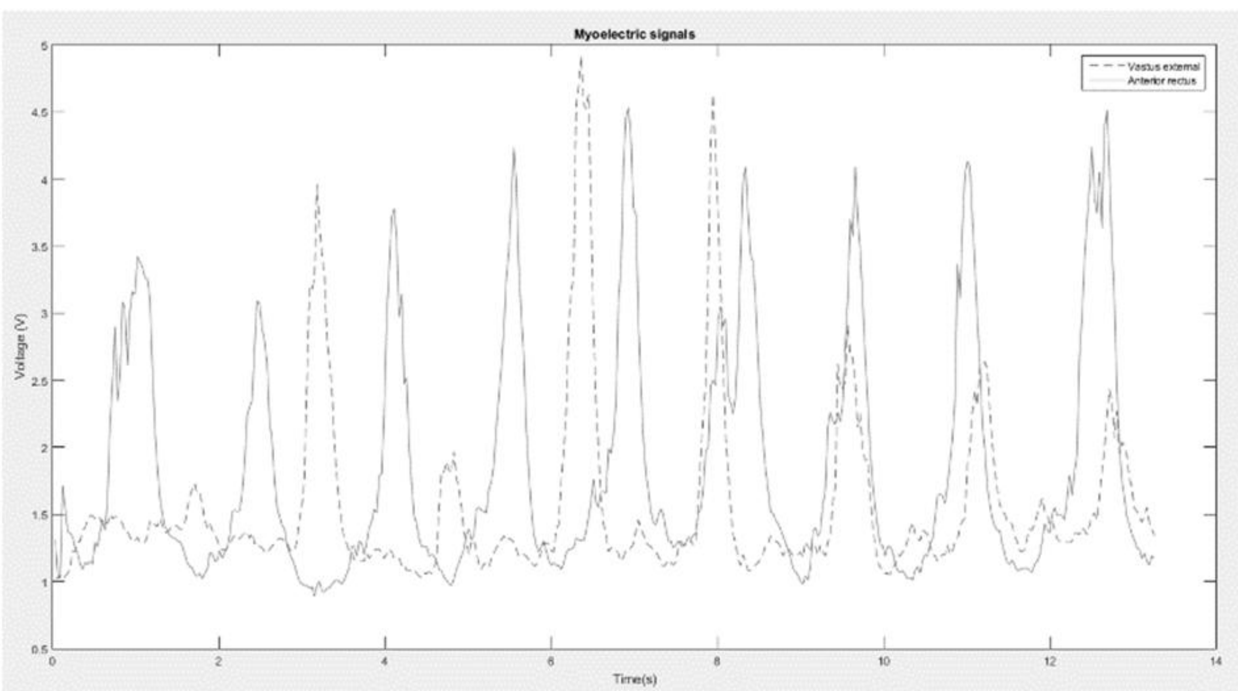

Fig 9. Comparison of vastus external and rectus muscle signals during gait cycle.

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V. Test and results

The protocol of tests is made with the collaboration of the “prosthetist”, establishing the guidelines and requirements of a standard prosthesis.

A. Bank alignment

In the test, the prosthesis parts were assembled on a work station without the patient, the dimensions, alignment and stability of the prosthesis were checked and the foot was aligned horizontally. The dimensions were verified comparing them with those of the healthy limb; it was achieved alignment and stability in the prosthesis when checking that the load line is below the axis, with this it was possible to determine that the prosthesis is perfectly armed for patient use.

B. Static alignment

In the test, the patient is standing with the prosthesis in which heights, rotations, inclinations, and others parameters were verified to obtain a stable standing and a level support on the surface of the floor. It was also verified that the patient was aligned, So that it distributes its weight symmetrically, 50% to its affected extremity and 50% to its healthy extremity as shown in Figure.10.

Fig 10. Standing patient.

C. Dynamic alignment

In the test the patient performed his first steps and started walking, after that the following aspects were verified: that the passage length is correct, that the patient is treading on the surface of the floor as flat as possible, that it does not have many inclinations. Two main views were valued:

- Front view or back view: In this view the stump inclination was determined, in which two main cases can be observed: the adduction which is an internal deviation of the stump and the abduction is an external deviation of the stump relative to the neutral line.

- Side view: In this test the patient is standing sideways in the sagittal plane, in this view the flexion or extension of the stump was evaluated since these parameters modify the alignment of the prosthesis. Ideally, the flexion degrees should be equal to the extension degrees during the extension of the leg. Figure. 11 and Figure. 12 shows the extension and flexion of the stump, respectively

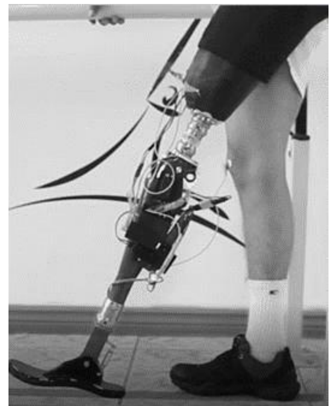

Fig 11. Extension of the prosthesis.

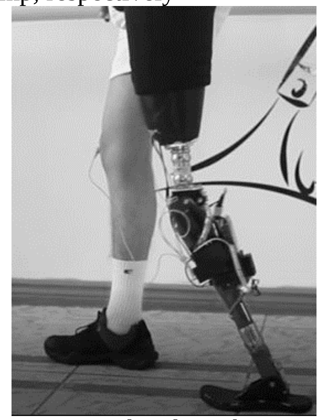

Fig 12. Flexed prosthesis

D. Gait angular analysis

Fig. 13 shows the walking cycle analysis during the support phase and the swing phase in which angular displacements of the knee and foot were verified.

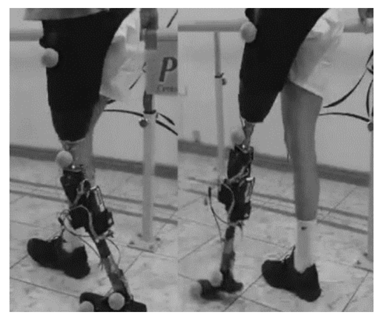

Fig 13. Angular analysis of the prosthesis during the gait cycle

VI. Results

The angles were obtained from the analysis of the gait cycle of the leg without amputation and the prosthesis, as presented in Figure 14 y 15

The Table V below shows a comparison between measurements taken on the leg without amputation and the prosthesis. The angles correspond to the knee, ankle and foot.

As can be seen in the table the error rate do not exceed 5%, allowing to affirm that the operation of the prosthesis resembles the natural movement of a human leg and that can be verified in the development of the patient's gait. Results guarantee the high efficiency of the prosthesis that improves stability when walking allowing you to have greater confidence and safety to carry out your activities.

Regarding the acquisition of temperature and humidity signals, different conditioning circuits were performed, based on the results shown in Table VI and VII, The most suitable circuits for its implementation were selected.

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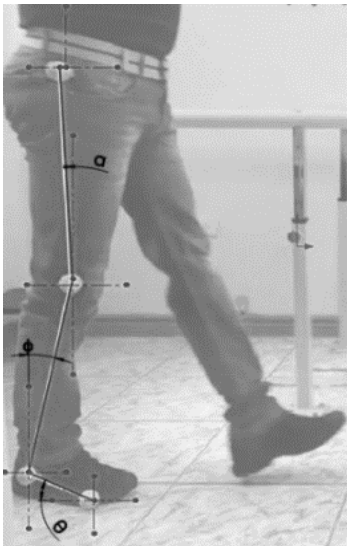

Fig14. Leg angle analysis

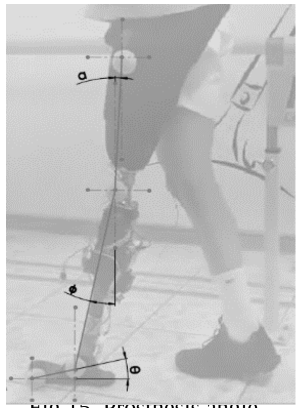

Fig 15. Prosthesis angle analysis

TABLE V
COMPARISON OF LEG ANGLES WITHOUT AMPUTATION AND PROSTHESIS

α (⁰)			Ø(⁰)			θ (⁰)		
Without Amputation.	Prosthesis	% Error	Without Amputation.	Prosthesis	% Error	Without Amputation.	Prosthesis	% Error
3,11	2,89	1,93%	12,92	10,87	2,79%	23,37	18,55	2,91%
4,06	3,75	1,72%	59,90	57,68	0,05%	132,50	129,47	0,94%
21,83	23,50	1,01%	11,04	13,54	1,36%	0,64	0	4,69%
17,60	16,96	3,64%	3,77	2,88	2,39%	4,93	3,96	1,01%
1,30	0,96	3,08%	18,62	16,96	0,81%	9,34	8,53	1,39%
14,81	12,25	1,55%	31,55	30,45	3,49%	6,14	7,74	2,61%
11,67	10,97	1,46%	52,09	51,12	1,86%	85,36	88,92	2,45%
6,24	4,24	2,24%	7,05	6,16	1,28%	13,16	11,23	1,37%
1,42	1,82	3,52%	1,42	1,96	7,04%	1,76	1,74	1,14%
14,62	12,62	0,82%	20,09	18,46	0,70%	5,46	7,53	4,58%
18,43	17,55	1,25%	33,38	31,71	1,26%	30,85	28,69	3,57%

TABLE VI
ERROR RATE: TEMPERATURA SENSOR CIRCUIT

	CALCULATED	MEASURED	SIMULATED	C/M	C/S	M/S
VOLTAGE DIVIDER	2,51 %	2,56 %	2,49 %	1,99 %	0,80 %	2,73 %
WHEATSTONE BRIDGE	2,04 %	2,089 %	2,04 %	2,40 %	0,00 %	2,35 %
VOLTAGE AMPLIFIER	3,87 %	3,79 %	3,88 %	2,07 %	0,26 %	2,37 %

TABLE VII
ERROR CIRCUITO DE ACONDICIONAMIENTO SENSOR DE HUMEDAD

	CALCULATED	MEASURED	SIMULATED	C/M	C/S	M/S
VOLTAGE DIVIDER	3,83 %	3,839 %	3,75 %	0,23 %	2,09 %	2,32 %
WHEATSTONE BRIDGE	1,33 %	1,309 %	1,31 %	1,58 %	1,50 %	0,08 %
VOLTAGE AMPLIFIER	2,75 %	2,766 %	2,71 %	0,58 %	1,45 %	2,02 %

VII. CONCLUSION

A prototype of leg prosthesis for transfemoral amputation was made using low cost mechanical and electronic components. The design requirements were obtained from the gait cycle analysis, from which the angular displacements that were produced when walking were acquired and compared with the angles of the leg without amputation obtaining an error of between 0% and 5%, allowing having a natural movement in the gait. The temperature and humidity monitoring allows the prosthetics to be kept under constant control, so as not to exceed 37 ° C and 70% humidity, parameters that will be displayed in a mobile phone application and keep the patient informed, avoiding alterations in the skin of the stump.

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