CHARACTERISTIC CURVES OF PNEUMATIC MUSCLE FOR THE USE IN THE UFMG EXOESKELETON

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Abstract. *This work is about the development of technology for accessibility projects, with the main objective to reduce the difficulty deficiency bearer has to walk and to realize routine tasks. Nowadays, there is a great demand for the development of low cost equipments used in the treatment to recover the walk movement as well as in sessions of physiotherapy. Recently, the Laboratory of Bioengineering at the Federal University of Minas Gerais (UFMG) has developed an artificial pneumatic muscle, DGP. In this work the relationships among pressure, volume and load required for DGP positioning control were established. It has also been developed an electronic circuit capable of triggering an electro-mechanical pneumatic valve using the myoeletric signal as an activate factor. The methodology to determine the flow required in a DGP, adapted to an exoskeleton, to auxiliary in the recover of the walk pattern is also presented. The walk pattern is determined by analyzing the position, speed and acceleration of hip and knee joints. This pattern is used in the calculation of the force, pressure and volume required in the DGP to recover and control the movement.*

Keywords: *Bioengineering, artificial pneumatic muscle, walk pattern, dynamics of the walk, myoeletric signal, movement recovery, accessibility.*

1. Introduction

The walking is the most important function for the human autonomy. When the human movement is impaired, the use of mechanical equipment can aid to the functional motion and greatly improve human walking ability after neurological injuries.

Traditionally the electric actuators used to restore the walking movement are big, heavy and presents some unsafeness characteristics. Hence the DGP (NAGEM, 2002; PINOTTI *et all,* 2004*)* an artificial pneumatic muscle was developed as a specific actuator to be used in functional orthesis and prosthesis.

The use of pneumatic actuators to control the movement in the whole range is a recent technology. These actuators are usually used to control the starting and the ending positions of the movement (MCDONELL, 1997; BOBROW e MCDONELL, 1998; CHOU e HANNAFORD, 1996). The pneumatic actuators have characteristics that difficult his control, like the non-linear movement with a pressure variation, the air compressibility, and the friction in low velocities (MCDONELL, 1997; BOBROW e MCDONELL, 1998; RICHER, 2000; XIANG, 2004). However, the developments in control methods and equipments turned these pneumatic actuators capable to control robots movement with high precision (GUILHARD e GORCE, 2000; RICHARDSON, 2003; RICHARDSON, 2004).

Flexible pneumatic actuators are an alternative feasible if compare with an electrical or hydraulic actuators for some applications in prosthesis and orthosis. Because this actuator behaves like the human muscle (KLUTE, 1999;

KLUTE, 2000), has low weight and cost and produces ten times more force than a pneumatic cylinder with the same diameter using only 40% of energy (FESTO, 2004).

In the assisted walking movement it is necessary an immediately answer from the system, besides strength and high speed. These are the reasons to use the Myoeletrical signal (MES) only to start the walking movement and to use an actuator and the walking gait parameters to control the rest of the gait cycle.

The complexity of the MES means a challenge in applications to control prosthesis and orthosis. However many works have tried to solve this problem using different approaches (ENGLEHART *et all*, 1999). The main problem of the MES analysis is its utterly complex composition influenced by many factors such as body fattening, sex, age among others. The invasive methods present a signal with less complex interference in applications to control prosthetics limbs (BERTO *et all,* 1998; BURROW *et all,* 1997). Nevertheless, their use require a computational time to analyze how the signal can be transformed in strength and speed information.

In this work the relationships among pressure, volume and load required for DPG position definition were established. The force and length necessary to recover the walking gait of a patient with polio using the exoskeleton developed in UFMG (VIMIEIRO, 2004) were determinated. Beside these, has been developed an electronic circuit capable to trigger an electro-mechanical pneumatic valve using the myoeletrical signal as an activation factor.

2. Methods

2.1 DGP Muscle

The DGP, has the same characteristics of an industrial muscle (FESTO, 2004) with a better performance than conventional pneumatic actuators. This muscle consistes of an internal latex bladder surrounded by a braided nylon shell that is attached at either ends to pneumatic conectors (CHOU e HANNAFORD, 1996).

2.2. Testing system, workbench

The testing methodology used to determinate the muscle contraction among the external tensile strength and internal pressure. The variations of internal pressure originated length variation for the specific force. To make this test was developed the workbench show in Figure 1.

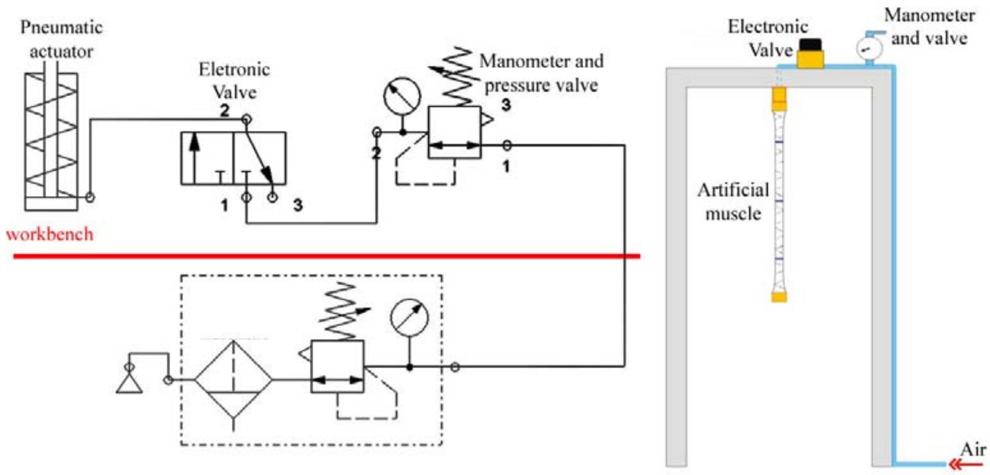


Figure 1. Schematic workbench

The workbench has an electric solenoid valve, a manometer, a valve to control the outflow, an electronic device to control the solenoid (NAGEM, 2002) and a pneumatic connection system to connect the muscle to workbench. In this workbench was performed the tests to determinate the length and volume of the muscle for known loads and pressures. With this data it is possible determinate the muscles trends.

2.3. Mioeletric signal

It’s necessary amplify the MES, to study your proprieties. The first identification and amplification is made by an electrode with a max external tension of the 2,0V. It was projected an electronic circuit (Fig. 2) capable to analyze the MES (PINOTTI *et all*, 2004), and activate an artificial muscle (Fig. 4).

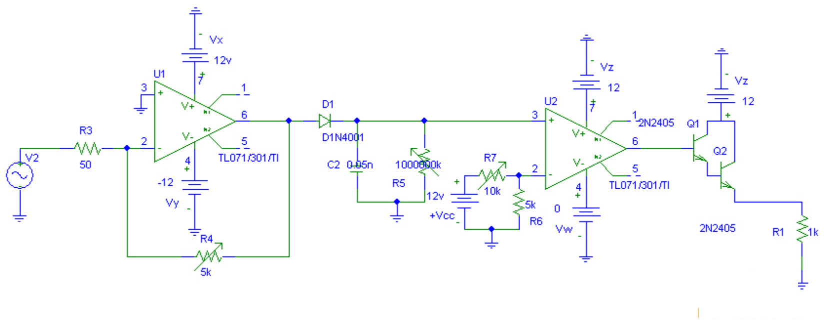


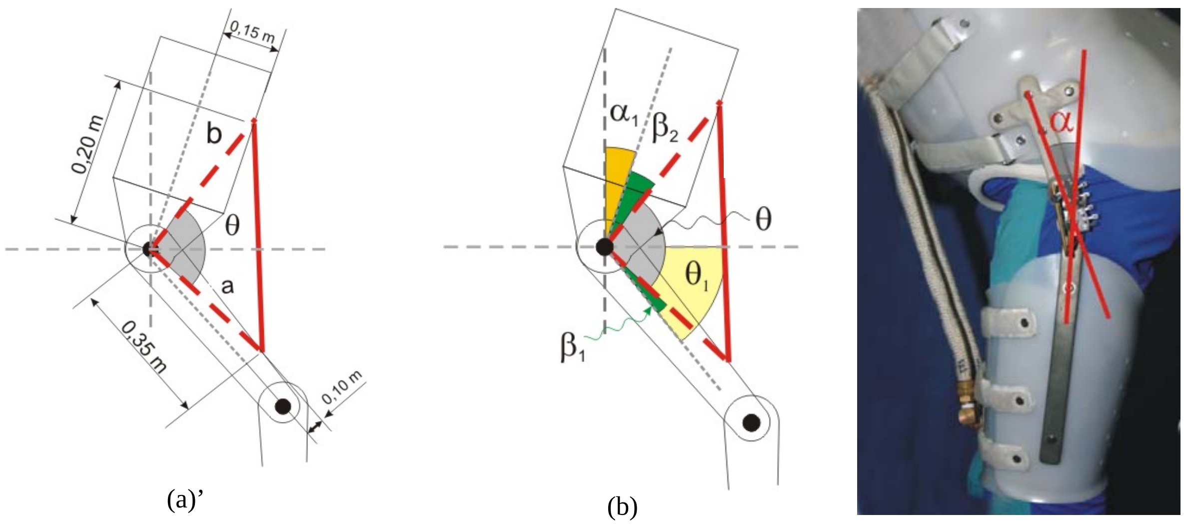
Figure 2. Electric circuit.

2.4. Modelling the walking gait

It was used an exoskeleton, a hip orthosis with artificial muscles developed in UFMG (Vimieiro, 2004) to analyze the patient gait. It was necessary to create a model, like a manipulator robot to analyze the cinematic and dynamics parameters to the patient's gait. The thigh is represented by a rigid limb, with one degree of freedom. The exoskeleton developed at the Lab-bio is able to move the hip articulation controlling only the flexion movement. The walking parameters vary according to the speed of walk, sex, age and pathology of the patient. The angle between the pelvis and the thigh was determinated using a potentiometer (Nascimento, 2005).

2.4.1 – The artificial muscle strength and length

By using the potentiometer angle and the muscle initial position values, is possible to obtain the variations of the joint position, joint speed and the artificial muscle length during the patient movement.



(a)' (b) (c)

Figure 3. (a) Size of the muscle; (b) Angle in muscle; (c) Picture of orthosis.

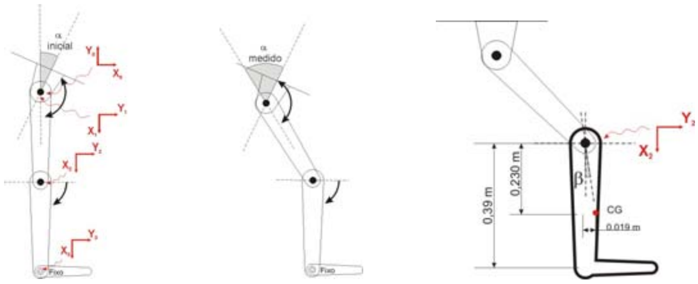
The θ angle determinates the muscle length and can be found using Figure 3 and Equation 1.

$$\theta = |\theta_1| - \beta_1 + (90 - \beta_2 - \alpha) \quad (1)$$

θ_1 angle in potentiometer
α angle initial – 23 degree
β_1 *arctan(10/35),*
. β_2 *arctan(15/20)*

2.4.2. Dynamic of walking

It is necessary to study the strengths which act on the system to promote the movement, this can be done considering the set (trunk, hip, thigh, leg and foot) as a robot manipulator. These strengths are due to the pneumatic actuator and inertial forces. By using the Euler Lagrange´s equations (CRAIG, 1989; ASADA e SLOTINE, 1986; FERREIRA, 1999; MOLINA, 2004), and working with all the system as a two degrees of freedom manipulator (Fig. 6), it is possible to calculate the necessary torque to promote the patient gait. For this analysis is necessary to determinate the rotations matrixes, the jacobians, the forces and the torques in each limb and joint.



(a) (b) (c)

Figure 4. Mechanical system with two degree of freedom.

The rotation matrix are represent by the R_1^0, R_2^1, R_3^2, in Equation 2. C_i and S_i represent Cos(θ_i) and Sen(θ_i).

$$R_1^0 = \begin{bmatrix} C_1 & -S_1 & 0 \\ S_1 & C_1 & 0 \\ 0 & 0 & 1 \end{bmatrix} \longrightarrow R_2^1 = \begin{bmatrix} C_2 & -S_2 & 0 \\ S_2 & C_2 & 0 \\ 0 & 0 & 1 \end{bmatrix} \longrightarrow R_3^2 = \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix} \qquad (2)$$

The third limb is fixed as can be observed in the Figure 4.(b) and it can be modeled together with the second limb, Figure 4.(c). With the rotation matrix, jacobians, inertial matrix, the size and weight of limbs and the speed of each joint it is possible determinate the forces and torques necessaries for the movement using the equations (3) and (4)

$$f_i = m_i a_{cgi} - m_i g_i + R_i^{i+1} f_{i+1} \qquad (3)$$

$$\tau_i = I_i \alpha_i + \omega_i \times [I_i \omega_i] - f_i \times L_{cgi} + R_i^{i+1} \tau_{i+1} + (R_i^{i+1} f_{i+1}) \times L_{cg(i+1)} \qquad (4)$$

f_i force in limb i.
m_i mass in limb i.
a_{cgi} aceleration in C.G in limb i.
g_i gravity.
R_i^{i+1} rotation matrix in limb i to i+1
τ_i torque in link i.
I_i inertial matrix of limb i
α_i aceleratiion in link i
ω_i angular speed in link i
L_{cgi} distance between link i and C.G. of limb i

The movement kinematic was determined by the hip angle position variations during patient's gait, being these the most important parameters of movements analyzed.

2.5 Establishing the muscle outflow to adjust the orthesis position

Knowing the muscle volume and inner pressure in the moment I (P_i, V_i), and the muscle volume and inner pressure in moment II (P_{i+1}, V_{i+1}). It is possible to determinate the gas volume and pressure (P_3) necessary to take the muscle to stage i to i+1. Establishing the time to reach the stage i+1 it is possible to determinate the flux. Knowing the pressure and the volume inside the muscle, and keeping constant temperature, the gas volume could be set up by Equation 5 (VAN WYLEN, 1995).

$$p\bar{v} = \bar{R}T \rightarrow ou \rightarrow pV = n\bar{R}T \qquad (5)$$

3. Results

The used artificial muscle has 2cm (DN) approximately, when pressurized. Other's studies (NAGEM, 2002) proved that this muscle is capable to support up to 980 Kpa (10 bar) of inner pressure and to carry through a force of up to 392,3 N(40,0 Kgf), although this model has a work band of 0,0 the 581 Kpa(6,0 bar). Figure 5 illustrates the curves of length in function of the pressure and the load. As can be observed, the muscle length has a nonlinear behavior in relation to the load and pressure variations. So, the muscle presents a potential curve behavior for each applied load. With the load increasing , this curve dislocates in axis X (pressure). For loads smaller then 4,9 N (0,5Kgf) this behavior does not occur because of the internal tube properties. To prevent this, it is necessary to work with the pay-pressured muscle.

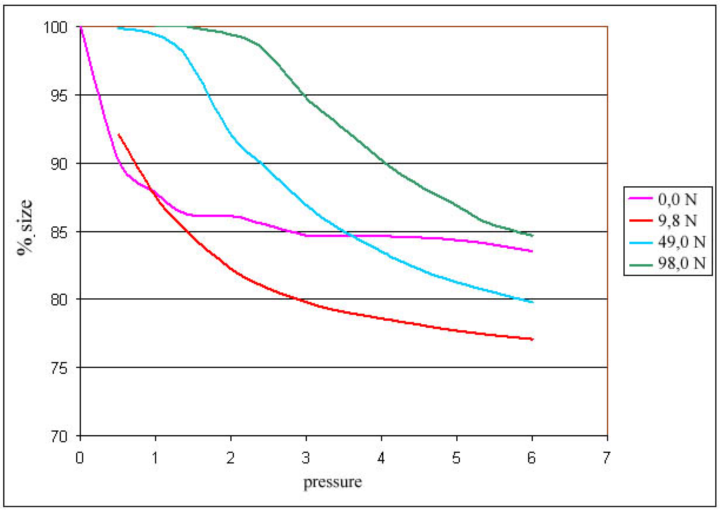


Figura 5. Size x Pressure x Load

3.1. Myoeletrical **Signal**

Using the signal of the circuit represented for Figure 4, it is possible to activate a relay to sets the pneumatic valve, thus initiating all the contraction muscle process. The signal of the circuit and the patient's signal, in healthy muscle, can be observed in Figure 6.

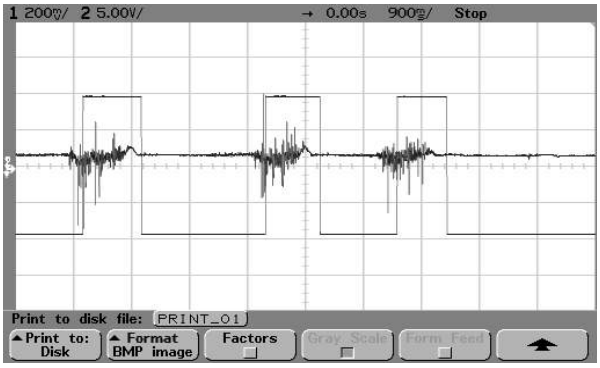


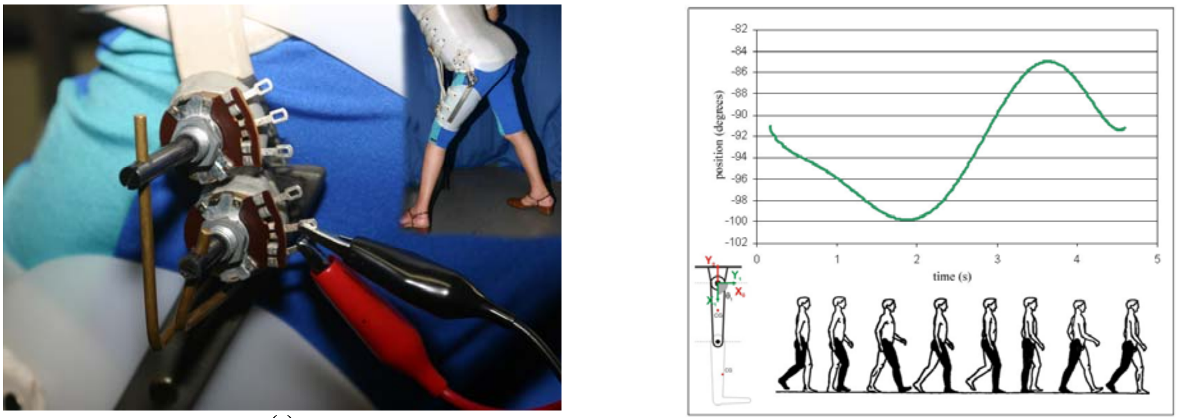
Figura 6. Myoeletrical signal and analyze signal (Oscilloscope).

It can be observed that during the muscular activity, the circuit exit signal (Fig. 2) turned into another signal with different tension (V). This proves that the circuit works to verify the presence of healthy muscular activity.

This circuit was efficient to analyse the MES of healthy muscles, but in muscles with neurological injuries like polio the system did not present a satisfactory response.

3.2. Position, Speed and Acceleration in the links

To determine the forces that act in the joints during the movement, it is necessary to know the angle, the speed, and the angular acceleration in each joint during the gait. These measures had been made using potentiometers installed in exoskeleton, as it can be observed in Figure 7 (a). With the angular position of hip (Fig. 7(b)), it was possible to determinate the speed and the acceleration of this joint and estimate the speed and acceleration of the knee joint during the movement. The ankle joint is fixed and the palm of the foot it is parallel to the floor.



(a) (b)

Figura 7. (a) Potentiometer in exoskeleton, (b) Angle measure during the walking gait.

3.3. Strength and torques during the movement

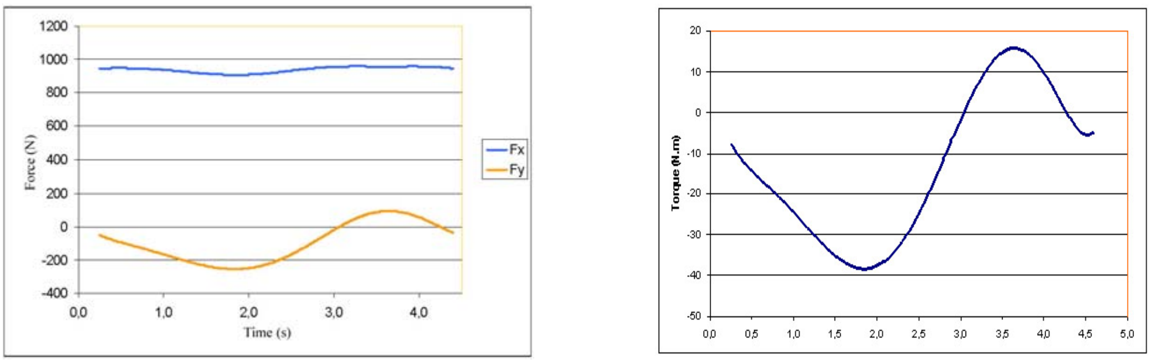
The exoskeleton and leg was described like a manipulator to calculate the torque and the strength in the joints. The base of the manipulator is defined as the trunk of the patient, being fixed during all the movement (Fig. 6 (b)). The structure of the manipulator was adapted from Vimieiro (2004), and it is physical characteristics can be observed in the Figure 6 (c) and Table 1.

Table 1. Characteristic of the manipulator.

Link	I_{zz} (kg.m^2)	r_x (m)	m (Kg)
1º	1,07E-02	1,83E-01	6,79E+00
2º	1,02E-01	2,31E-01	4,13E+00

Using the speed and acceleration of spindle and the manipulator physics characteristics, it is possible to determinate the strength and the torque in the manipulator articulations, as can be observed in figure 8.

To calculate the force generated by the artificial muscle, it is necessary to determinate the torque during the gait. The artificial muscle is able to control the movement only in positive angle (flexion), therefore the interval of analysis necessary to determinate the force, dimension and pressure is between 3 and 4,3 seconds(Fig. 8. (b)).



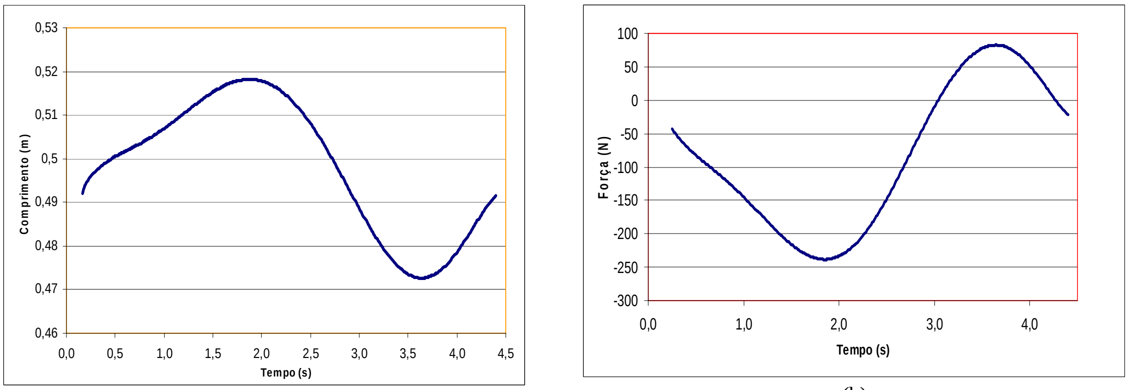
(a) (b)

Figure 8. (a) Force in C.G of spindle 1; (b) Torque in link 1.

3.4. Strength and dimension in artificial muscle

After the torque calculation in the first joint, it is necessary to determine the lenght and the strength of the artificial muscle. The maximum length of the muscle is determined by the dimensions of orthesis. The θ angle showed in the figure 5 (b) can be determined using the Equation 1.

The force generated by an artificial muscle during the movement is not greater than 100,0 N, (Fig. 9), this force is smaller than the maximum force generated by a healthy muscle, but it is enough to perform the movement in this model. In the exoskeleton assembly it is necessary to apply two muscles in parallel to distribute the forces between them.



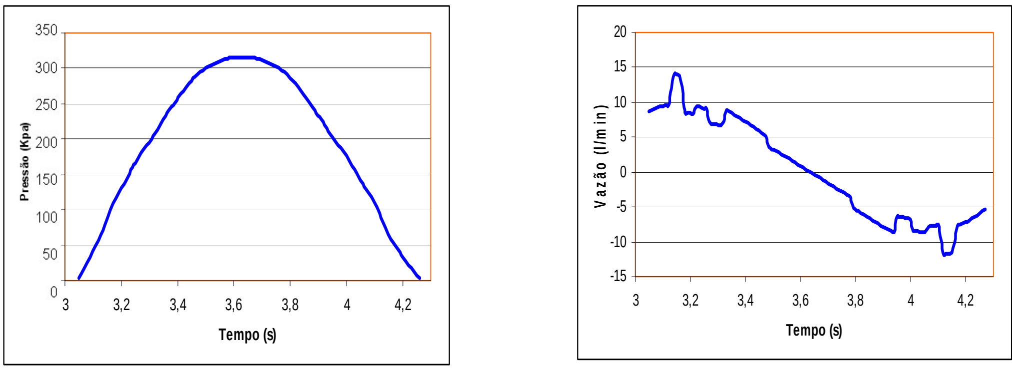
(a) (b)

Figure 9. (a) Size of muscle during the walk; (b) Force in muscle during the walk.

After the force determination, it is necessary to calculate the outflow to produce the length and the force adjusted during the gait (Fig. 9). With the outflow control it is possible control the muscle and exoskeleton movement.

3.5. Outflow determination

Using the curves of length x pressure x load and the curve of force, it is possible calculate the outflow necessary to control the muscle position (Fig 12(b)). A software developed in the LABBIO was used to generate the graphs pressure x time (Fig 12(a)) and outflow x time (Fig12 (b)) in the interval between 3 and 4,3 seconds.



(a) (b)

Figure 10 (a)Internal pressure in muscle; (b) Outflow during the walking gait

The highest pressure is present with the least length of the muscle and, consequently, the highest volume. The biggest pressure occurs in the same time that the highest torque in the joint (Fig 9,10,).

To determine the muscle position, it is necessary to determine the air outflow for pressurization, thus controlling the movement of the joint 1 (Fig. 3, 10(b)). The feeding pressure is defined in 60,0 N/cm^2 (6bar), being this the higher work pressure for the muscle. Using the Equation 5 it is possible to calculate the feeding pressure necessary to perform the movement. The main points of non-linearity in the control of the outflow occur when the load and the pressure are small, causing an instability in the system, this instability occurred in the interval between 4,1 and 4,3 seconds. By applying antagonistic muscles in exoskeleton assembly, situations of low pressures and loads can be avoided propitiating an improvement in the control the muscle position and preventing the regions where outflow changes fast.

4. Conclusions and perspectives

In agreement with the results, it is possible to conclude that with the developed methodology it is possible to calculate the outflow of pressurization in an artificial muscle by determinating the curves of length x pressure x load and the curves of force x length during the gait analysis. Due to modeling of orthesis it is necessary to determine the speed and the acceleration of the patient's joints and thus to determine the force and the length of the artificial muscle applied in the orthesis to perform the movement. The artificial muscle presents an non-linear behavior to low loads and pressures harming the controller of the othesis position. Because of this, combination of low loads and pressures must be prevented using antagonistic muscles.

The developed circuit is capable to identify, interpret and convert the signal of a healthy muscle to activate an system to the pneumatic muscle control, but it does not present a coherent reply with the MES of a muscle affected by the Polio.

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