KINEMATICS AND KINETICS PLANNING OF A TRANS-FEMORAL PROSTHESIS BEHAVIOR

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Abstract. *The study of auxiliary devices for the locomotion of amputees is usually based on the behavior of the musculoskeletal system responsible by human locomotion. Usually, this devices are modeled to restore the gait of amputee based on the Multibody System Theory, where the structure is modeled as a rigid body system in addition to elements like springs, dampers and actuators that are assembled and compounds a controllable device used to substitute the part of the lower limb that was lost through an amputation. The purpose of this work is to determine the behavior that these devices need to simulate for an approximation of the natural human gait. The methods of modeling can be Newton-Euler, Euler-Lagrange, Generalized D'Lambert and their variations that present advantages and disadvantages, depending on the application of the model obtained. In this work the Newton-Euler modeling method was used because it allows many numerical approaches and accurate control applications. According with the data obtained from normal human gait measured with a VICON system and a set of force platforms, the adequate behavior could be presented and the necessary behavior for trans-femoral amputee prosthesis can be estimated. To enable a good result it is necessary a dynamic model of the human locomotion system in the main plane of the movement in question. In that case, this modeling can be done adopting the joint that is responsible for movements that mainly influence the behavior, the knee joint, which could be described in the sagittal plane. Other hypothesis that was adopted is the determination of a relative position vector between two points, each one in a linkage of the prosthesis where the device will be connected. The variation of this relative vector is used to model a desired behavior for the controlled device. Using the data measured and the anthropometric parameters obtained from different methods of estimation, a rigid body system is modeled and some curves are obtained with a relative precision. These curves describe the behavior of a generic controlled device, active or semi-active. As a result of this study some curves about displacement and force reactions will be presented.*

Keywords*: human locomotion model, gait analysis, Trans-femoral Amputee Prosthesis, kinematics, kinetics planning*

1. INTRODUCTION

The human locomotor's system is a dynamic system whose complex configuration delivers versatile movements under the functional aspect, transforming relative rotational movements between segments into a translational movement of the full system characterized by a rhythmic and plastic perfection known as *human gait*. However, several situations can cause lacks in this perfection through the occurrence of many pathologies, congenital diseases and also total or partial lower limb's amputation. In the case of amputations, in which occur physical and functional lost of part of the system, is necessary to compensate these lost introducing new segments trying to reproduce the natural dynamics. That is, introducing prosthesis into the handicapped system.

Amputations are scissors done through medial portions of the limb segments, or less frequents, disarticulations. The position of these scissors define the type of amputation, for instance, trans-femoral amputees are individuals that suffer amputations above the knee, while trans-tibial amputees are ones that suffer amputation bellow the knee. These two cases are the more common, being the trans-femoral amputation the case with major functional costs, because knee and ankle articulations are lost, and as consequence, the functionality and versatility of the system is affected more effectively than in the trans-tibial case. Based on this observation, a conclusion about the importance of the knee joint can be taken; this articulation presents a much more significant influence in the behavior of the whole system than the ankle joint.

Several studies have being developed focusing the gait restoration of amputees. Many proposals for devices have being presented with the goal of reproduce the rhythmic and plastic perfection of the human gait for them (Popovic, 1994). An able device to reproduce the functional features of lower limb must be designed taking in account plastic and functional skills of the system. Recently, researchers have proposed the application of active control and semi-active control devices aiming to substitute the complex behavior of the set of articulations by a controllable device that permits to follow a desired behavior according with the natural amputee's day-by-day activities (Herr, 2003).

The Multibody Systems Theory allied to powerful numerical methods of differential equations solution allows a full knowledge from the human locomotor's system if also allied to a set of kinematic and dynamic experimental data and inertial parameters for elements of system. This information can be obtained using measurement devices as the computer-aided video motion analysis system (VICON), force platforms (Bertec Co.), and load cells for amputees (Raptopoulos, 2005), whose provide the measurement of kinematic features of locomotor's system as trajectories of several strategic points of human body, and also, dynamic features as ground reactions forces and moments of forces to the action of human body during gait. About inertial parameters of each element of lower limb, they are useful to afford inertial considerations to the mathematical model, conferring more fidelity to it (Winter, 1983), (Vaughan, 1999) and (Raptopoulos, 2003). Based on this fully defined universe in terms of kinematics and dynamics, many studies have being realized to develop lower limb prostheses (Popovic, *et. al*, 1988, 1991a, 1991b, 1994; Kim, 2001 and Kapti, 2006).

Regarding the prostheses design, several studies need to be done to complete the necessary information's universe to make an accurate evaluation of them. Amongst the disciplines needed to make it possible are found the theory of mechanisms synthesis, the design of control devices and the development of control strategies. These related fields have presented several advances in the last years, for instance, in the materials field (application of materials as electro and magneto-rheological fluids, electro and magneto-strictive polymers as actuators (Steffen Jr., *et. al*, 2004) and intelligent control strategies, like fuzzy logic, neural-networks, and genetic algorithms, amongst others.

To design prosthesis is essential to establish a reference behavior for the set of mechanical and electromechanical elements that will be used. Thus, this behavior is obtained from the observational gait analysis in normal individuals. Therefore, using the kinematics and dynamic models obtained from theory and the experimental data measured during the gait, becomes possible to plane the prosthesis' behavior in terms of kinematics and dynamics. Based on the results obtained from the current study, a discussion about the application of some actuators and control strategies, amongst other aspects, becomes possible.

This work aims to provide some fundamentals to aid in the development of a device able to restore the gait of individuals with trans-femoral amputations. As known, a trans-femoral prosthesis must be prepared to work restoring the functions of the knee joint, ankle joint and metatarsophalangeal joints in the maximal way, as it is mechanically impossible because of the complexity in the configuration of this set of joints, several controllable devices had been used to make that prosthesis' behavior gives mimics at almost a part of this complex behavior. Focusing on the way of control and maintaining the simplicity of mechanisms, a mono-axis prosthetic knee, a SACH foot and tubular linkages are considered as de basic system in which a control device will be coupled, being it an active or semi-active one.

2. METHODS

2.1. Bipedal Model

A model of seven segments and six joints was adopted to represent the human locomotor's system (Figure (1)). Several foot joints were not considered because of the impossibility to observe the motion of these segments using the VICON 140 system. In this modeling, all segments were considered rigid bodies. The joints that were considered in the model were modeled as spherical joints and were used to estimate the movement and the forces involved in the sagittal, frontal and transverse planes. This assumption admitted that all translations in joints were neglected and a uniquely rotation centre were considered. To the adopted model there were 12 degrees of freedom in the support phase (closed chain) and 18 degrees of freedom in the swing phase (opened chain). During the support phase two feet are in contact with the ground while in the swing phase only one foot is in contact with the ground.

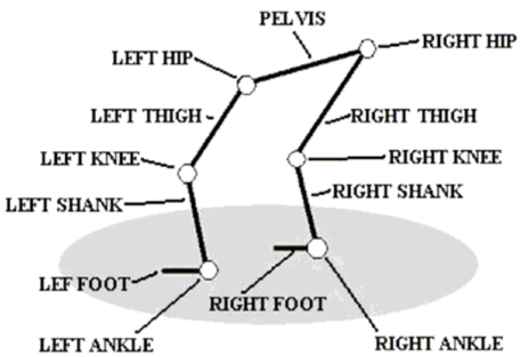

Figure 1. Scheme of the Mechanical model of human locomotor's system.

2.2. Kinematic Theory

The kinematic model of the human locomotor's system was built considering all body segments as rigid bodies and their relative rotations were assumed to take place about a fixed point in the center of joint (Kadaba et al., 1990; Wu et al., 1995; Raptopoulos et al., 2001, 2002, 2003a, 2003b). The relative rotations between adjacent segments were described by a matrix representing a three-dimensional space. This matrix is the product of three rotations around the axis x, y and z. The final rotation matrix is defined in the Eq. (1),

$$R_{xyz}=\begin{bmatrix} \cos\phi_y\cos\phi_z & -\cos\phi_y sen\phi_z & sen\phi_z \\ \cos\phi_x sen\phi_z + sen\phi_x sen\phi_y\cos\phi_z & \cos\phi_x\cos\phi_z - sen\phi_x sen\phi_y sen\phi_z & -sen\phi_x\cos\phi_z \\ sen\phi_x sen\phi_z - \cos\phi_x sen\phi_y\cos\phi_z & sen\phi_x\cos\phi_z + \cos\phi_x sen\phi_y sen\phi_z & \cos\phi_x\cos\phi_z \end{bmatrix} \tag{1}$$

where ϕ_x is the rotation angle around x axis, ϕ_y around y axis, and ϕ_z around z axis.

Based on this matrix (Eq.(1)) two of these angles can be calculated, ϕ_x and ϕ_z .

$$\phi_x = arctg\left(-R_{xyz}(2,3)\big/R_{xyz}(3,3)\right) \tag{2}$$

$$\phi_z = arctg\left(-R_{xyz}(1,2)\big/R_{xyz}(1,1)\right) \tag{3}$$

To calculate the angle ϕ_y it is necessary to manipulate the Eq. (1). From the Pos-multiplication of the matrix in Eq. (1) by the inverse of the rotation matrix around z axis, is constructed a new matrix that represents the product of the rotation around x and y axis (Eq. 4).

$$R_{xy} = R_{xyz}.(R_z)^{-1} = \begin{bmatrix} \cos\phi_y & 0 & sen\phi_y \\ sen\phi_x sen\phi_y & \cos\phi_x & -sen\phi_x\cos\phi_y \\ -\cos\phi_x sen\phi_y & sen\phi_x & \cos\phi_x\cos\phi_y \end{bmatrix} \tag{4}$$

So, the angle ϕ_y will be calculated through Eq. (5).

$$\phi_y = arctg\left(R_{xy}(1,3)\big/R_{xy}(1,1)\right) \tag{5}$$

2.2.1. Inertial and Local Frames

In this work, two types of local frames were considered, the external and internal. The external reference system was used to define the position of the removed marks at the transformation from the static for the dynamic protocol. The internal reference system was used to calculate the joint relative angles and the net forces and moments in each articulation. For each joint there were two adjacent local frames to calculate the rotation around each axis making up six local frames near each joint.

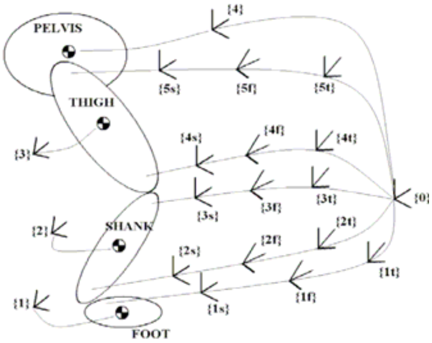

Figure 2. External and internal frames of the lower limb.

The exception is in the hip, where there were three references in the femur and only one in the pelvis. In the Figure (2) is presented the external and internal frames of one lower limb, where {0} is the inertial reference system, {k} is the center of mass system of the segment k (k=1..4), {ps} is the local frame used to calculated the sagittal rotation of the joint (p=1..5), {pf} is the local frame used to calculated the frontal rotation of the joint (p=1..5) and {pt} is the local frame used to calculated the transverse rotation of the joint (p=1..5). These systems are used to represent the different rotation axis in this approach.

2.3. Kinetic Theory

The Figure (3) shows the free-body diagram of body i, where ${}^{i}F_{i-1,i}$ is the force acting from the body $i-1$ in the body i and described in the center of mass reference system, ${}^{i}M_{i-1,i}$, is the moment of forces acting from the body $i-1$ in the body i, ${}^{i}F_{i+1,i}$ the force from the body $i+1$ in the body i, ${}^{i}M_{i+1,i}$ is the moment of force from the body $i+1$ in the body i, ${}^{i}r_{i-1}^{i}$ is the position of joint $i-1$ relative to the centre of mass of segment i, ${}^{i}r_{i}^{i}$ is the position of joint i relative to the centre of mass of segment i, ${}^{i}a_{i}$ is the acceleration of the centre of mass of segment i, ${}^{i}\omega_{i}$ is the angular velocity of segment i, ${}^{i}\dot{\omega}_{i}$ is the angular acceleration of segment i and m is the mass of segment i. All these variables are described in the local reference system of centre of mass { i }.

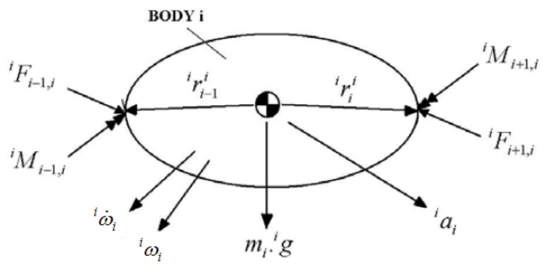

Figure 3. General diagram of loads for each body.

Adding all forces that act on the body and equaling them to the net force, is defined the Eq. (6), which calculates the force by the body i+1 on the body i. This equation describes de force in the center of mass reference system.

$$ {}^{i}\vec{F}_{i+1,i} = m_i({}^{i}\vec{a} - {}^{i}\vec{g}) - {}^{i}\vec{F}_{i-1,i} \tag{6} $$

Adding all moments that act on the body and equaling them to the net moment of force, is defined the Eq. (7), which calculates the moment of force exercised by the body i+1 on the body i. This equation describes de moments around the center of mass reference system.

$$ {}^{i}\vec{M}_{i+1,i} = {}^{i}\underline{I}_{i}.{}^{i}\dot{\vec{\omega}}_{i} + {}^{i}\vec{\omega}_{i} \times {}^{i}\underline{I}_{i}.{}^{i}\vec{\omega}_{i} - ({}^{i}\vec{r}_{i-1}^{i} \times {}^{i}\vec{F}_{i-1,j}) - {}^{i}\vec{M}_{i-1,i} - ({}^{i}\vec{r}_{i+1}^{i} \times {}^{i}\vec{F}_{i+1,j}) \tag{7} $$

The anthropometrical parameters (mass, inertia, etc.) were estimated through regression equations obtained from the gamma scanner method (Zatsiorsky et al., 1990). Based on the knowledge of the transformation matrixes between the references become easy to transform these loads from the center of mass system for another system and it will be used to present the results.

2.4. Experimental Protocol

In this work were used the same static and dynamic protocol used in Raptopoulos 2002, 2003, 2004). The static protocol was used to define the external local frame and relative position of some points that are removed during the walking recording; the markers in the dynamic protocol were similar to the static besides the removed points. The hip's center of rotation (HCR) was estimated through a set of proposed regression equations (Bell et al., 1989, 1990; Seidel et al., 1995; Leardini et al., 1999). Some authors used a cluster design criteria for optimization and dynamics estimation (Cappozzo et al., 1997; Andriacchi et al., 1998; Leardini et al., 1999) but this method cannot be employed in the most abnormal gaits, such: cerebral palsy, total hip arthroplasty and others patients that cannot do movements with great width.

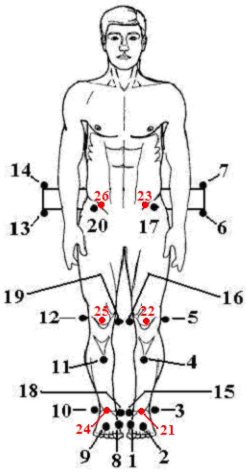

Figure 4. Experimental Protocol.

The Figure (4) presents the anatomical points used in these protocols, where: (1) is the medial face of the halux in the left foot, (2) is a point on the long axis of left foot, (3) is the lateral malleolus of left ankle, (4) is a point in the anterior surface of left shank, (5) is the lateral epicondyle of the left knee, (6) and (7) are the left vertexes of pelvis' reference structure, (8) is the medial area of is the medial face of the hálux in the right foot, (9) is a point on the long axis of right foot, (10) is the lateral malleolus of right ankle, (11) is a point on the anterior surface of right shank, (12) is the lateral epicondyle of the right knee, (13) and (14) are the right vertexes of pelvis' reference structure, (15) is the medial malleolus of left ankle, (16) is the medial epicondyle of left knee, (17) is the left anterior-superior iliac spine, (18) is the medial malleolus of right ankle, (19) is the medial epicondyle of right knee, and (20) is the right anterior superior iliac spine. The points (21) and (24) are sagittal-symmetric and are determined as internal points in the middle distance between the medial face of the hálux (point (3) in the left, and (10) in the right side, and the lateral malleolus of the ankle (point (15) in the left and (18) in the right side), respectively. According with the same plane of symmetry, the points (22) and (25), are determined by the middle distance between the medial epicondyle of knee (points (16) in the left and (19) in the right side) and the lateral epicondyle of knee (points (5) in the left and (12) in the right side). As the intern points mentioned before, the points (23) and (26) are the HCR (Hip's Center of Rotation).

The motion analysis for normal individuals was performed using a computer-aided video motion analysis system with three infrared cameras (VICON 140) and two platforms of force (Bertec Co.).

2.5. Group of Control

To accomplish a good trans-femoral prosthesis is necessary to analyze the human gait patterns. In this work, this analysis was done observing the results presented by Raptopoulos (2003). From the gait of a group of control composed by 57 normal and healthy individuals, in which 24 are males and 33 are female, were presented kinematic and dynamic patterns for each joint in the three anatomic planes of movement. Besides that, are presented forces and moments of forces in each joint, as well as the ground reactions. The features presented by the group of control are presented in Table 1.

Table 1. Characteristics of the normal's group of control

Characteristic	Unit	Medium	Standard deviation
Age	[years]	22.27	2.16
Mass	[kg]	66.500	14.888
Height	[m]	1.689	0.091
Cycle	[s]	1.241	0.127
Support phase	[s] / [% cycle]	0.824 / 66.37	0.092 / 2.48
Swing phase	[s] / [%cycle]	0.417 / 33.63	0.051 / 2.48
Stride	[m]	1.165	0.092
Step	[m] / [% stride]	0.584 / 50.10	0.061 / 4.55
Width of the base	[m]	0.123	0.034
Progression speed	[m/s]	0.949	0.116
Cadence	[steps/min]	98.021	9.508

2.6. Human Locomotion

Presenting the human gait as a composition of happenings that repeat themselves in cycles according with the gait cadency becomes possible a better knowledge from each mechanism of movement that this complex system presents. A gait cycle should be divided into two main phases, the stance and swing phase. During stance phase the foot is on the ground and the system could be interpreted as an inverted double pendulum, whereas in swing phase the same foot is no longer in contact with the ground and the lower limb is swinging through in preparation for the next foot strike, as an ahead double pendulum. The stance phase may be subdivided into three separate phases: first double support, single limb stance and second double support.

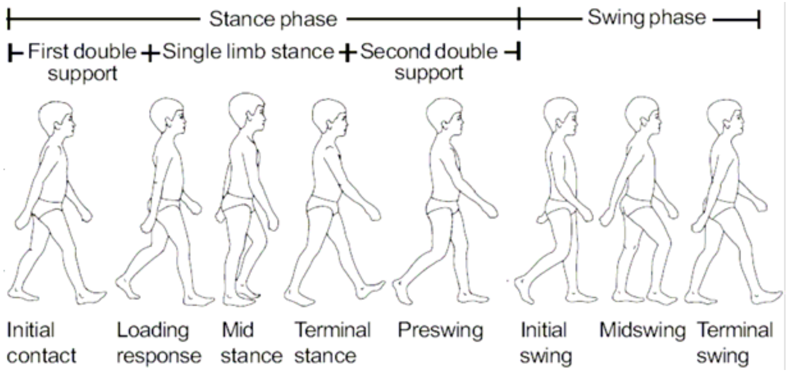

Figure 5. Phases of human gait (Vaughan, 1999)

Another division of the gait is useful to describe the functions of each articulation in the movement (Figure 5). The Initial contact is better defined as an instant and it is adopted as the initial phase of gait, because represents the instant in which the foot contacts the ground, after that begins the phase of Loading response responsible for the weight acceptance. This phase is characterized by the impact absorption, and the knee joint is the main articulation in this phase, absorbing much more impact then the hip and ankle. Their actions avoid the structure collapse and store energy. The next phase is the Midstance phase and it is characterized by the bearing of shank in the ankle, progressing the body forward, in this phase the knee joint is responsible for a stabilization of the system. The next phase is the Terminal stance, this phase is characterized by the end of the single stance and consists on the shank bearing in the toes progressing the body forward and heel rising. In that phase the knee joint needs to flex a little to prepare the transition to swing phase. Following the sequence to complete the gait phases, the pre-swing phase begins and represents the second period of double support in the stance phase. In that phase the knee flexes preparing to the swing phase. Defining the start of the swing phase is the Initial swing, in which the hip and knee joints flex to ensure that foot passes upward ground safely during swing (*foot clearance*). The Midswing phase is the next one and the only difference to the Initial swing phase is that the knee extends by the action of inertial forces and gravity action. The last but not least gait phase is the Terminal swing that represents the phase in which the limb prepares itself to return to the ground.

Regarding to the knee behavior, the necessities of this mechanism to yield the desired behavior in amputees gait restoration are: to give stability to the lower limb to accept weight in early stance, give some shock absorption and allow that body continue forward movement smoothly during weight acceptance. After these two first functions, the knee needs to support the body weight with knee flexed, then, it needs to begin the flexion during single limb weight bearing at end of terminal stance. In the other main phase, swing phase, the knee must respond instantaneously to cadence variations and increases in the pace velocity, making the prosthetic behavior to mimic the natural lower limb.

3. RESULTS AND DISCUSSION

Regarding to techniques and methods related to human gait analysis presented, it is possible to define the full behavior of lower limb during walking. From the forward and inverse kinematics becomes possible the description of translational movements and also rotational relative movements in each joint of lower limb. On the other hand, the internal forces and moments of forces acting on each joint of the lower limb are described with basis on the ground reaction forces and moments of forces to the individual movement obtained from force platforms.

The aim of designing a trans-femoral prosthesis is to yield the gait restoration of amputees. Thus, the methodology adopted to determine the desired behavior to auxiliary device goals to establish curves of normal behavior to be used as reference. These curves are obtained from experimental data and application of multibody theory for a normal and healthy group of control. This result allowed some observations about the amplitudes involved on the human locomotor's system. The anatomic plane in which the lower limb presents more significant behavior in terms of movement's amplitude is the sagittal plane and more, the behavior of each joint and how forces and moments of forces act in the joints during the human gait.

The reference curves can be obtained accounting some hypotheses. Firstly, the dynamic model adopted to compute forces and moments of forces was established under the consideration of bodies that compose the system are thin elements with concentrated mass in the center of gravity. Moreover, in respect to the control system, the inertia and mass properties of its elements were not considered. Another consideration, about actuators, was about their possible ways of work, in that aspect, a linear actuator was considered. The same adopted procedure could be applied to a hypothetic rotational device considering in addition some algebraic manipulation.

To guarantee a better agreement of results that will be presented, some observations are necessary and some considerations must be explained. The reference curves to the desired behavior of the prosthesis was determined from the average among all individuals in the group of control composed of 57 normal and healthy young people, what means that, after the kinematics' measurement of each individual from the group of control, an average of these values was calculated and applied to the kinematics equations showed in a Section 2.2. The inertia and mass features, as well as the height of the individual, had been also gotten from the average values of each individual.

Another relevant consideration that needs to be well explained is the prosthesis configuration. In this work was considered a prosthesis configuration with only one degree of freedom composed by thin metal segments, a mono-axis joint as the prosthetic knee and a SACH foot. About control device, the actuator functions in linear displacement and its fixation is accomplished, in the up segment, very close to the knee joint, while in the down segment the actuator was connected close to the center of mass of the segment. The considered actuator presents a stroke of 5.3 cm, and the position of each attachment point was determined based on the maximum value that it can range.

From the knowledge of knee's behavior it becomes easy to conclude that the sagittal plane is the anatomical plane with major significance in terms of amplitude of movements, and therefore, the simplification of the three-dimensional behavior of lower limb while walking to a plain movement occurring in the sagittal plane is adequate for the modeling in question.

3.1. Kinematics

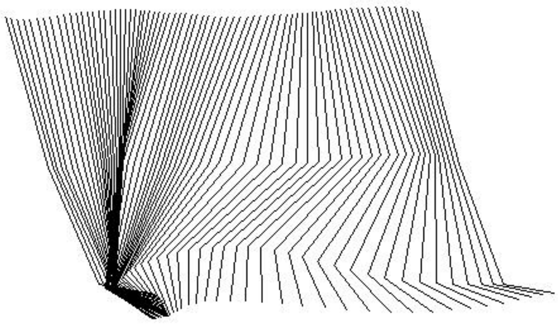

Figure 6. Plotting of segments of lower limb during a gait cycle.

In terms of kinematics, the desired behavior can be obtained through the determination of the relative position between two points established in the system as the attachment points of the control device presented in Figure 7. A point established in the segment of the thigh (point D) and another point established on the segment of the shank (point E).

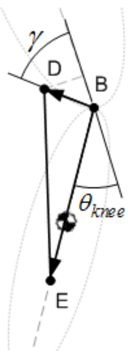

Figure 7. Scheme for the kinematic planning of the prosthesis.

The procedure is simple and can be described by the following steps: first, the variation of the angular position of the knee in the sagittal plane is determined, based on the experimental data, from the inverse kinematics (Figure 8a), then, the points are established according with the restrictions of the device, as the range of displacement and the way in which the device actuates, and finally, the vector is established by the known position between the two attachment points. The relative position vector between the points of segment attachment of the device of control in the leg with respect to the segment of the thigh and its derivative, supplied the velocity of the displacement.

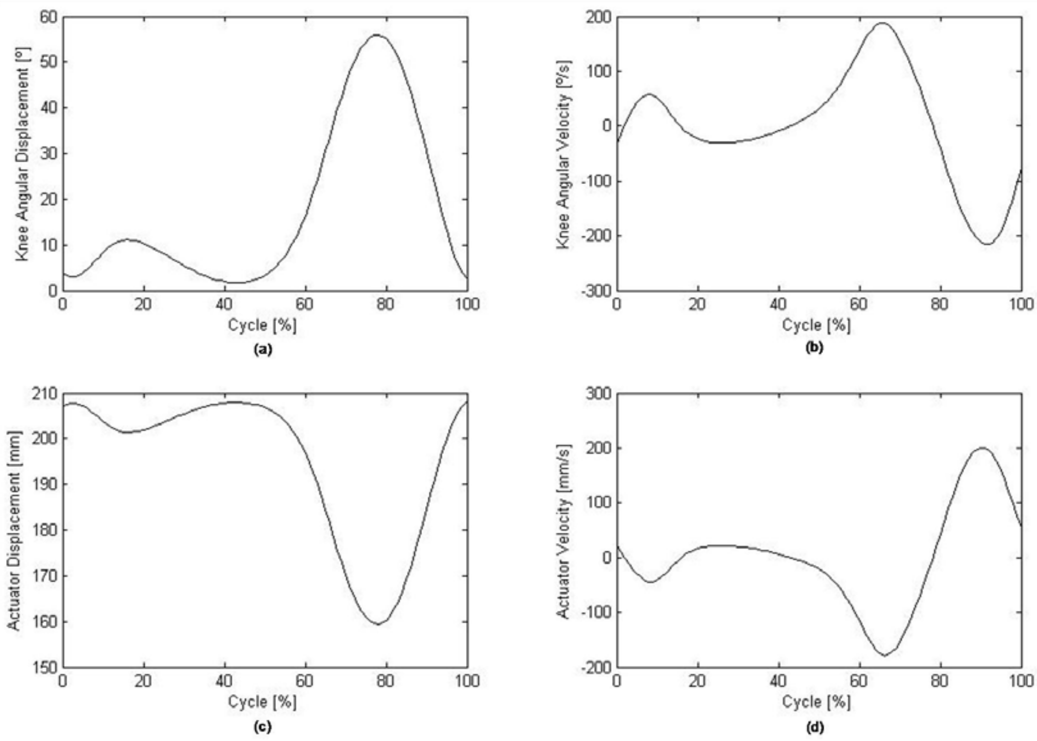

Figure 8. Curves for the kinematics planning of the Prosthesis behavior. (a) Angular displacement of knee joint in a cycle; (b) ratio of the angular displacement of knee in a cycle; (c) desired curve to the actuator behavior; and (d) desired ratio for the actuator displacement.

3.2. Kinetics

The kinetic behavior of prosthesis is determined feeding the dynamic model of prosthesis with the experimental data. At the cost of some manipulation it is possible to determine the positions, velocities and accelerations of the main points of the system, as joints and center of mass in each segment. Moreover, the forces and moments of the forces, applied in each extremity of segments, are determined with basis on the reaction forces caused in the ground by the individual during its locomotion, also obtained from experiments while walking, through force platforms installed in a gait lab. Appling these experimental data to the movements' equation, the determination of desired forces to make prosthesis behavior as smooth as the normal lower limb becomes possible.

The observation of the curve of forces (Figure 9) that the control device must lead to the system, in comparison with the resultant moments in the knee joint that was measured permit to perceive two picks of force. The first one occurs in the end of the Load acceptance phase and little before the knee joint experiments the maximum flexor moment. The second pick occurs in the Preswing phase, when the impulsion of body forward happens and the actuator needs to extend. During the Midswing the actuator needs only to maintain the knee flexed for the foot passage upward the ground. And finally, in the Terminal swing the control device must permit the knee flexion, and consequently, prepare the shank to the next ground contact.

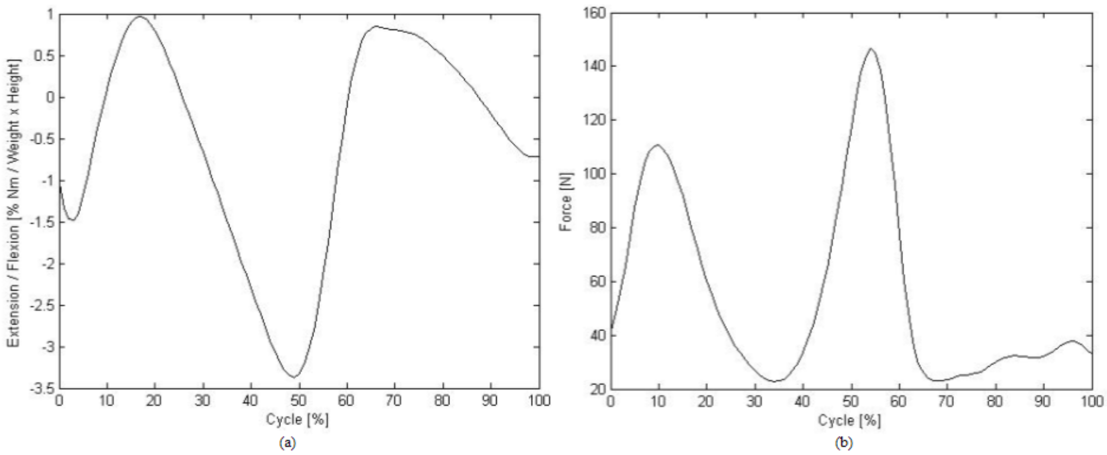

Figure 9. Kinetic Behavior of knee (a) Measured moment of force in the knee joint during gait; (b) Curve of force desired for the kinetic behavior of the control system.

4. CONCLUSION

Many kinds of devices have been tested looking for a system that can follow the desired curves with relative easiness. The choice of the ideal control system depends on many different aspects. The power consumption, computational cost and weight are problems that are faced in that case. Other difficulty that is presented as a big problem is the control strategy. Many different strategies have been used because of the complex behavior that system needs to mimic.

The present work permitted the determination of the curves that are desired as the behavior of a control device designed to give a restored gait to amputees. Motivated by the search for new control devices, being active or semi-active ones, a discussion regarding to the kinematics and dynamics planning of prosthesis behavior can be a useful approach to discover a better actuator to be used in that application. This study permitted to plan the ranges of displacements and the amplitudes of forces that are needed to make the prosthesis behavior mimic the natural lower limb.

5. ACKNOWLEDGEMENTS

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