ALGORITHM FOR THE PREDICTION OF THE REACTIVE FORCES DEVELOPED IN THE SOCKET OF TRANSFEMORAL AMPUTEES

ALGORITMO PARA LA PREDICCIÓN DE FUERZAS REACTIVAS EN SOCKETS DE AMPUTADOS TRANSFEMORALES

JUAN FERNANDO RAMÍREZ
Ph.D. Universidad Nacional de Colombia, Facultad de Minas, jframirp@unal.edu.co

EMMANUEL JARAMILLO MUÑOZ
Est .Ing. Universidad Nacional de Colombia, Facultad de Minas, ejaramillom@unal.edu.co

JAIME ANDRÉS VÉLEZ
Ing. Universidad Nacional de Colombia, Facultad de Minas, jaavelezze@unal.edu.co

Received for review: September 12th, 2011, accepted: January 27th, 2012, final version: April 10th, 2012

ABSTRACT: Based on a mathematical model of the human gait, a Matlab 2010a algorithm is presented to predict the reaction forces and moments in a particular point along the socket linked to the lower limb of a transfemoral amputee. The model takes the inertia developed due the swing of the limb during the gait into consideration. A validation of the results is made with the data obtained in a gait lab, and the model results are consistent with those obtained in the gait lab.

KEY WORDS: numerical model, transfemoral amputee, dynamic analysis, gait, algorithm.

RESUMEN: Se presenta un algoritmo en Matlab 2010a basado en un modelo matemático para predecir los momentos y las fuerzas de reacción en un punto particular del socket vinculado al miembro inferior de un amputado transfemoral. El modelo tiene en cuenta las inercias desarrolladas durante el ciclo de la marcha. Se realiza una validación del modelo comparando los resultados con los datos obtenidos en un laboratorio de análisis de la marcha y se encuentra una buena correspondencia con los datos experimentales en las fases de apoyo y balanceo.

PALABRAS CLAVE: modelo numérico, amputado transfemoral, análisis dinámico, marcha, algoritmo.

1. INTRODUCTION

The study of the human gait is important not only from a medical point of view, but also for the construction of bipedal motion robots [1], the design of automated movement control systems [2], the creation of realistic animations for the entertainment industry or even in biometrics [3,4], among many other fields of application.

It is possible to say that the dynamics of the human gait is widely understood [5], most studies consist in experimental data measured for individual patients, using methods such as computerized movement capture [5], electro goniometry [6], or dynamometric platforms [7], all of them usually gathered together and used in gait analysis laboratories.

The analysis of the human gait is an invaluable diagnostic tool for those requiring precise data about the gait cycle of a particular individual, either as a pathology diagnosis instrument or in biometrics; however, this is always a fully experimental process. A mathematical description of gait exists as a sequence of static events, and dynamic effects are introduced as a correction factor [8]. These models can be used as a predictive method in specific moments along the gait, such as the stance phase.

To develop a full model of the human gait, it is necessary to consider the inertia of the whole body, as well as the fact that the swinging of the limbs is not a passive movement, as it was previously considered [9]. Currently, mathematical models of the gait are created specifically for a particular person [5] and there are no generalized models; nevertheless, there are current investigations focused on the creation of pattern generators of the gait cycle, using a normalized parameter formula and a database that contains multiple patterns of gait. Nowadays, these pattern generators only predict geometrical and displacement patterns, but not the forces developed in the limbs during the gait cycle [10]. In the case of a pathological gait, there is a lack of reference models, due to the variety of pathologies that affects the behavior of a limb during the gait cycle.

The dynamic analysis of the forces that act in the gait cycle of a particular individual is a fundamental instrument in the development of scientific and engineering applications that require a comprehensive understanding of the human gait cycle.

Mathematical models have been widely used for the development of equations that define the dynamics of the gait cycle; however, a basic algorithm for parameterization in terms of height and weight of a person would offer a generalized tool to find approximate values of forces and moments at any point along the lower limbs, with no need for experimental measures.

As was previously stated, there is no tool that allows for one to obtain the forces and moments developed in the socket of a transfemoral amputee without using direct experimental measurements in a human gait cycle laboratory, so the objective of this paper is to present the creation of a graphic interface, developed in Matlab 2010a, where the dynamic equations are based on few biometrical parameters such as the length of the stump (or residual limb) and the height and weight of the individual. The remaining parameters are taken from anthropometrical measures, a common practice in prosthesis design [11]. The results, presented as forces and moments in specific locations of the lower limbs, might be useful in the development of finite element models (FEMs) used to obtain the stress-strain state of soft and hard tissues involved in the interaction with the prosthetic system.

2. MATERIALS AND METHODS

As a first approach to the model, it is necessary to do a basic simplification, that is the development of a mathematical model restricted to the sagittal plane, because the highest percentage of movement during the gait cycle takes place in this plane [12], and the forces developed in the medial-lateral plane do not present significant changes when the limb is moved between random positions of the gait cycle. The simplification aligns the center of gravity of the individual with the limb's step plane.

Figure 1 shows the entire set of components that integrate the prosthetic system used by a transfemoral amputee. Length and size of the different constitutive elements are established according to the height of the amputee and his/her residual limb (stump) length.

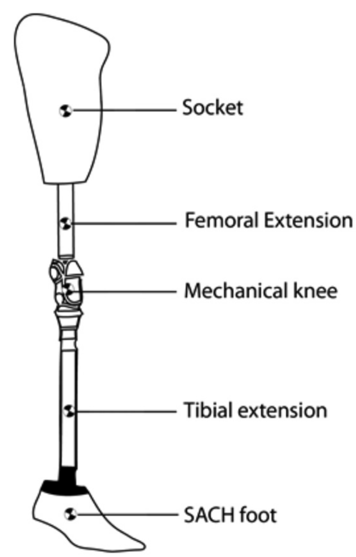

Figure 1. Prosthesis components

The socket and SACH foot are considered to be a polymeric matrix of polyvinylchloride or PVC; femoral and tibial extensions are cylindrical hollow pieces of aluminum 6061 T6 (commercial). The mechanical knee is simplified as a solid sphere made of the same aluminum alloy as the extensions. It is always considered that all parts used in the prosthetic model have a weight similar to the corresponding part of the lost limb.

Figure 2 shows ***Rx***, ***Ry***, and ***Rz*** which represent the forces acting in each coordinate axis in the individual's socket. ***Mx***, ***My***, and ***Mz*** are reactive moments acting in the individual's socket. ***Ci*** represents the mass center of each section of the prosthesis, where $\boldsymbol{i}$ = 1, 2, 3, 4 and 5. ***Fgx***, ***Fgy***, and ***Fgz*** are reactive forces on the ground, acting over the SACH foot, for each of the coordinate axis. $\boldsymbol{m_i}$ represents the mass of each section, and ***Li*** is the position of the mass center of every stretch. For both ($\boldsymbol{m_i}$, ***Li***), $\boldsymbol{i}$ = 1, 2, 3, 4, 5. The weights $\boldsymbol{m_i}$, ($\boldsymbol{i}$ = 1, 2, 3, 4, 5) vary due to characteristics of the individual's prosthesis for the model; masses are taken from the work presented by Vaughan [13].

The angular position of the prosthesis is denominated α, as well as the angular velocity ω and angular acceleration ε. Figure 2 shows all the variables as being measured from vertical axis z to an imaginary axis described by a line that joins the mass center of each part of the prosthesis and the point that represents the hip joint.

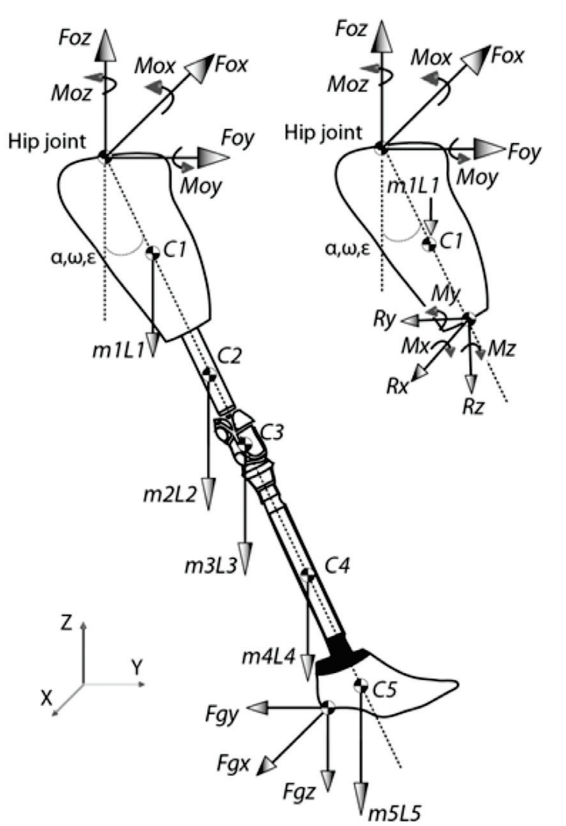

Figure 2. Free body diagram of a transfemoral prosthesis

The model includes 5 different sections representing the basic components of prosthesis for a transfemoral amputee, those are (along the proximal-distal direction): socket, femoral extension, mechanical knee, tibial extension, and prosthetic foot. For the development of the model, it is necessary to measure or quantify each one of the variables involved and presented (Fig. 2). From Table 1, ***m1***, ***m2***, ***m3***, ***m4***, and ***m5*** represent socket mass, femoral extension mass, knee mass, tibial extension mass, and prosthetic foot SACH mass, respectively. It is necessary to measure the amputee's weight ***Wt*** without the prosthesis.

Table 1. Model parameters

PARAMETER	VARIABLE
Prosthesis mass	*mt* (Kg)
Individual weight	*Wt* (Kg)
Individual height	*H* (m)
Stump mass	*m1* (Kg)
Femoral extension mass	*m2* (Kg)
Knee mass	*m3* (Kg)
Tibial extension mass	*m4* (Kg)
SACH mass	*m5* (Kg)

2.1. Prosthesis modeling

For each subsection corresponding to the parts that compose the prosthesis, a parameterization was done in terms of the individual's height ***H***, as is presented in Table 2.

Table 2. Geometric parameters

PARAMETER	VARIABLE
Stump radius	*r1* (m)
Femoral extension radius	*r2* (m)
Knee radius	*r3* (m)
Tibial extension radius	*r4* (m)
SACH radius	*r5* (m)
Stump length	*H1* (m)
Femur extension length	*H2* (m)
Tibial extension length	*H4* (m)
SACH length	*H5* (m)

Where:

r1 = 0.05 ***H***

r2 = 0.033 ***H***

r3 = 0.035 ***H***

r4 = 0.033 ***H***

r5 = 0.02 ***H***

Stump length ***H1***, as an input parameter in the model, should be measured from the greater trochanter to its distal end. Therefore [14]:

H2 = 0.245 ***H*** – ***H1***

H4 = 0.245 ***H***

H5 = 0.152 ***H***

Given the radius and lengths of all segments, as well as the density of each segment, it is possible to calculate approximately every section mass in terms of the total mass of the prosthesis ***mt.***

Biometrical data can be obtained directly from each patient; however, it is not always available; so, for the model, anthropometric standard data was used [14] from which masses and lengths of every component of the prosthesis-stump was set. Next, the masses of every section of the prosthesis, related to its total weight, are [13]

m1 = 0.3 ***mt***

m2 = 0.2 ***mt***

m3 = 0.1 ***mt***

m4 = 0.3 ***mt***

m5 = 0.1 ***mt***

2.2. Inertia moments, velocities, and angular acceleration

The masses of every segment used during the inertial analysis approximate the amputee's prosthesis weight with the lost limb's weight. The considerations that are made during this modeling come from standard models that relate individual weight to the weight of each segment of the prosthesis.

Besides this, gait laboratories have global positioning systems that allow for location and position of each part that constitutes the prosthetic system at any given instant ***t***. For the effects of this model, it is necessary to know the values of the angular acceleration; the angular velocity; and the position in the support, bipedal stance, and swing phases of the gait. Those parameters have been taken from an experimental database of forces and angular positions for a standard gait used in the medical laboratory of the *Fundación Universitaria María Cano.*

2.3. Dynamic formulation

Resultant equations for the model are presented next.

$$M_{oz} - [g\ \text{Sen}\ (\alpha)]\ [m_1L_1 + m_2L_2 + m_3L_3 + m_4L_4 + m_5L_5] + F_{gx}Y_g + F_{gy}X_g = I_o\ \varepsilon \qquad (1)$$

$$M_{ox} + F_{gy}Z_g + F_{gz}Y_g = 0 \qquad (2)$$

$$M_{oy} + F_{gz}X_g + F_{gx}Z_g = 0 \qquad (3)$$

$$F_{ox} + F_{gx} = [m_1 + m_2 + m_3 + m_4 + m_5]\ [r\varepsilon\ \text{Cos}\ (\alpha) - r\omega^2\ \text{Sen}\ (\alpha)] \qquad (4)$$

$$F_{oy} + F_{gy} - [m_1 + m_2 + m_3 + m_4 + m_5]\ g = [m_1 + m_2 + m_3 + m_4 + m_5]\ [r\varepsilon\ \text{Sen}\ (\alpha) + r\omega^2\ \text{Cos}\ (\alpha)] \qquad (5)$$

$$F_{oz} + F_{gz} = 0 \qquad (6)$$

Being Io, in Eq. (1), the total inertia of the prosthetic system,

$$Io = I1 + I2 + I3 + I4 + I5 \qquad (7)$$

2.4. Graphic interface

The mathematical model was used to develop an algorithm using Matlab 2010a with a simple interface in which the researcher (user) only needs to introduce three general parameters: patient weight, patient height, and stump length; and to select the desired phase of gait (initial contact, double support, or swinging) to study the forces and the moments. Using this data, the program defines all parameters in the model and calculates moments and forces that act in the point where the socket joins with the femoral extension (Fig. 3 shows the developed interface).

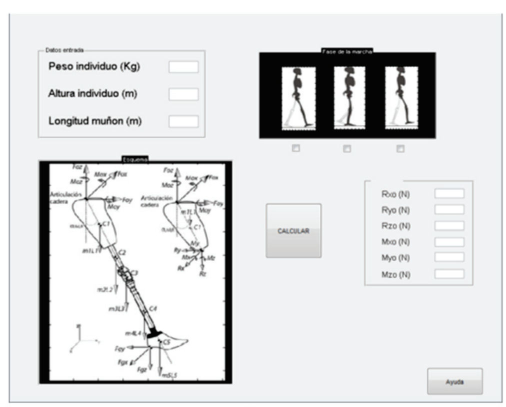

Figure 3. Graphic interface

3. RESULTS

A sample of the typical results achieved with the developed algorithm is presented in Tables 3 to 5. There, it is possible to visualize that results are differentiated in each rectangular axis—not only the forces, but also reactive moments in the joining point of the femoral extension and the socket.

It is observed how the algorithm gives different values for each of the selected phases of the gait.

Table 3. Forces and moments in the initial contact phase

		Initial contact phase
Entry parameters		
Individual height	H (m)	1.76
Individual weight	Wt (kg)	72
Stump length	H1 (m)	0.28
Algorithm results		
Force Y axis	Ry (N)	-312.283
Force Z axis	Rz (N)	-675.655
Force X axis	Rx (N)	-310.781
Moment Y axis	My (N.m)	-8.167
Moment Z axis	Mz (N.m)	-28.4
Moment X axis	Mx (N.m)	452.583
Experimental results		
Force Z axis	Rz (N)	-675.632
Moment X axis	Mx (N.m)	490

Table 4. Forces and moments in the double-support phase

		Double support phase
Entry parameters		
Individual height	H (m)	1.76
Individual weight	Wt (kg)	72
Stump length	H1 (m)	0.28
Algorithm results		
Force Y axis	Ry (N)	-301.75
Force Z axis	Rz (N)	-656.898
Force X axis	Rx (N)	-316.768
Moment Y axis	My (N.m)	-0.945
Moment Z axis	Mz (N.m)	-29.536
Moment X axis	Mx (N.m)	316.768
Experimental results		
Force Z axis	Rz (N)	-512.025
Moment X axis	Mx (N.m)	-181

Table 5. Forces and moments in the swinging phase

		Swinging phase
Entry parameters		
Individual height	H (m)	1.76
Individual weight	Wt (kg)	72
Stump length	H1 (m)	0.28
Algorithm results		
Force Y axis	Ry (N)	-310.025
Force Z axis	Rz (N)	-657.452
Force X axis	Rx (N)	-310.781
Moment Y axis	My (N.m)	6.996
Moment Z axis	Mz (N.m)	-28.711
Moment X axis	Mx (N.m)	131.982
Experimental results		
Force Z axis	Rz (N)	-614.638
Moment X axis	Mx (N.m)	110

Comparing the experimental values given by the database obtained from the gait analysis laboratory with the results obtained by the analytical model, it is possible to appreciate the following:

- In Table 3, for the initial contact phase, there is a difference-in-the-forces value in the ***proximal-distal (z)*** axis of 2.73%; while for moments along the ***medial-lateral (x)*** axis, the difference is 4.89%.

- For the double-support phase, it is observed from Table 4 that, in the moments through the ***medial-lateral (x)*** axis, the difference is 275.59%. In the forces value through the ***proximal-distal (z)*** axis in the dual support phase, the difference obtained is 28.3%.

- In Table 5, a difference of 6.25% in the force value is observed through the ***proximal-distal (z)*** axis during the swinging phase, with a difference of 19.9% existing in the moment through the ***medial-lateral (x)*** axis.

4. DISCUSSION

The experimental values have been compared with the values returned by the algorithm in Tables 3–5. However, it must be stated that the experimental data correspond to information from a single individual, due to the lack of a generalized pattern in the literature that can be used to establish a comparison of the forces and moments developed in the socket.

The differences between the experimental data and those obtained using the algorithm can be due to a number of factors, such as particular geometries of the limbs, variations in the masses in relation to the anthropometric data, variations in the lower limb tissues' composition (the ratio among muscle, fat, cartilage, etc.), and different types of prosthetic systems, among others.

Due to the particular set of data used in this study, it is likely that the model must be corrected, taking changes in the height of the amputee into account.

In this case, it is assumed that all the individuals studied must present a maximum angle, between the vertical axis Z and the imaginary line that joins the hip joint and the centers of mass of the different sections of the prosthesis, of 0.25 radians in the initial contact phase.

The great difference of the moment in the **x** axis during the double support phase must be taken into consideration. The difference can be due to the action of the lower limb muscles that creates a reaction force which is not present in the prosthetic system.

5. CONCLUSION

The model proposed gives accurate values in the phases of initial contact and swinging, which is a useful starting point in the development of a continuous dynamic model for transfemoral amputees.

Further studies should take the full body into consideration; that is, they should include parameters such as the acceleration, velocities, and position of all parts of the body during the complete gait cycle, not only the lower limbs,.

Other lines of research include the development of a comprehensive database of standard individuals categorized by their grade of amputation, the type of prosthesis, a list of common pathologies associated with the amputation, etc. All those factors can affect the results of future algorithms and must be taken into consideration.

The gait cycle varies due phenotypical differences between individuals; nonetheless, in this particular case the values given by the algorithm are close to the experimental values in the initial contact and swinging phases of the gait. Therefore, it can be stated that within the restrictions described, this algorithm can be used as preliminary tool in cases in which the reactive forces and moments in the socket-femoral extension joint are required but it is not possible to obtain experimental data.

6. ACKNOWLEDGEMENTS

The authors would like to thank the following organizations for their collaboration and involvement in this investigation: the *Fundación Universitaria María Cano, Ortopraxis S.A.S.,* and the *Instituto Neurológico de Antioquia.*

REFERENCES

[1] Zhicheng, H., Hongshuai, C., Jianle, S., Zhen S., Xiangji, C. and Yantao, T. «Design and implementation of high efficiency biped robot system», Chinese Control and Decision Conference. pp 2433-2438. 2010

[2] Collette, C., Micaelli, A., Andriot, C., and Lemerle, P. «Robust balance optimization control of humanoid robots with multiple non coplanar grasps and frictional contacts», IEEE International Conference on Robotics and Automation . pp 3187-3193. 2008

[3] Cheng-Chang, L., Chih-Chiang, T., and Jia-Ming S. «Human Gait Recogniti*on for Arbitrary View Angles»*, Second International Conference on Innovative Computing, Information and Control - ICICIC. pp 303-303. 2007

[4] Kim, D. and Paik, J. «Gait recognition using active shape model and motion prediction», IET Computer Vision, vol. 4, nº. 1, pp 25-36. 2010.

[5] Culik, J., Szabo, Z., y Krupicka, R., «Biomechanics of Human Gait Simulation», World Congress on Medical Physics and Biomedical Engineering, pp 1329 - 1332. 2009.

[6] Vanitchatchavan, P., «Termination of human gait», IEEE International Conference on Systems, Man and Cybernetics, pp 3169 -3174. 2009.

[7] Ferreira, J., Crisostomo, M. and Coimbra, A., «Human Gait Acquisition and Characterization», IEEE Transactions on Instrumentation and Measurement, vol. 58, pp 2979-2988, 2009.

[8] Perry, J., «*Gait Analysis: Normal and Pathological Function*», 1o ed. Slack Incorporated, 1992.

[9] Whittlesey, S., van Emmerik, R., and Hamill, J. «The swing phase of human walking is not a passive movement», Motor Control, vol. 4, nº. 3, pp 273-292, 2000.

[10] Lim, H., Hoon, K., Soh, Y., Tow, A., and Low, K., «Gait planning for effective rehabilitation - From gait study to application in clinical rehabilitation», IEEE International Conference on Rehabilitation Robotics, pp 271-276. 2009.

[11] Loaiza, J. and Arzola, N., «Evolución y tendencias en el desarrollo de prótesis de mano», DYNA, vol. 169, pp 191-200, 2011.

[12] Kapandji, I., «*Fisiología articular: esquemas comentados de mecánica humana. Tomo II: Miembro Inferior*». Editorial Médica Panamericana, 2002.

[13] Vaughan, C., Davis, B. and O›Connor J. «*Dynamics of Human Gait*». Human Kinetics Pub, 1992.

[14] Sánchez, J., «*Biomecánica de la marcha humana normal y patológica*». Instituto de Biomecánica de Valencia, 2005.