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Dynamics of a Running Below-Knee Prosthesis Compared to Those of a Normal Subject

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ABSTRACT

The normal human running has been simulated by two-dimensional biped model with 7 segments. Series of normal running experiments were performed and data of ground reaction forces measured by force plate was analyzed and was fitted to some Fourier series. The model is capable to simulate running for different ages and weights at different running speeds. A proportional derivative control algorithm was employed to grant stabilization during each running step. For calculation of control algorithm coefficients, an optimization method was used which minimized cinematic differences between normal running model and that of the experimentally obtained from running cycle data. This yielded the estimated torque coefficients of the different joints. The estimated torques and the torque coefficients were then applied to specific below-knee prosthesis (a SACH foot) to simulate healthy-running motion of joints. Presently the SACH foot is designed for amputee's walking; our data was used to modify such construct for running purposes. The goal was to minimize the differences between normal human model and a subject wearing a SACH foot during running. Kinematical curves of models for the obtained optimum mechanical properties indicated that prosthetic leg can reasonably produce the kinematics of normal running under normal joint driving torques.

Keywords: Dynamic simulation; Human running; Below-knee prosthesis; Mathematical modeling; Passive controller; Optimization; SACH Foot

1 INTRODUCTION

RESEARCHES in the field of human gait have a lot of applications in medicine, ergonomics, sport science and technology. Lower limb prostheses design is one of the fields in which human gait is important.

One of the best methods of gait analysis is to use analytical models. A desirable prosthetic leg needs to provide reliable stability as well as an acceptable controlled motion. The prosthesis should reproduce normal kinematics of the gait and running cycle when subjected to normal driving torques at the amputee's healthy joints. Several experimental studies have been conducted to examine the performance of different types of below-knee prostheses [1-3, 5, 9, and 10]. Mathematical modeling techniques have been also employed to analyze the effect of the prosthetic design parameters on the kinematics, dynamics and other characteristics of amputee locomotion. The previous studies on the transtibial amputee gait and running have been limited to a single limb or to the swing phase of the gait cycle [4-7].

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Iida (2008) has simulated a model with running capability. In this model an active motor exists in hip joint and passive joints with which a spring with inherent damping property. To simulate ground forces which contain vertical force and horizontal force, Coulomb friction which is contrary to contact point motion, in tangent direction is used. Vertical force is simulated using nonlinear spring-damper model [10].

In Peter (2009) study existing torque in knee and ankle are measured using inverse dynamics. By considering adjustable torsional spring in joints and ankle and muscles, Peter has studied human running. He tried to study biomechanics of human running by solving the equations relating torques and joints and angles to the end of minimizing calculated and measured angles, mechanical work and variations in spring values. Gerrit et al (2010) designed a 2D model and 3D one with 25 degrees of freedom. In this research springs, dampers and operators with unknown values in joints, corresponding to produced torques in tendons and muscles are used. The values for spring and damper properties are extracted from physiological tables of human body. In Newton-Euler dynamic equations air friction is considered. Finally to minimize model motion energy in a cycle, operator values are evaluated using Lagrange optimization method. Akbari (2009) modeled body with prostheses in top of the knee regardless of muscles. And a series of springs and dampers are used in joints to simulate muscle torque. For calculating torque coefficients of these parameters in joints, with the purpose of minimizing the difference between cinematic model and normal gate data, optimization method was used. Instead of using normal human torque in an artificial leg, elastic, hydraulic and Coulomb control unit equations with torsional spring and damper was used and the results were compared [13].

The purpose of the present study was to employ the mathematical modeling approach to analyze the dynamics of a 7-link biped human with a below-knee prosthesis during the complete running cycle. This model was then used to determine the optimal design parameters of SACH foot controller mechanisms for running, as well as the prosthetic ankle, in order to achieve the closest kinematics to the normal running.

2 Method

2.1 Simulation of normal human running

Simulation of motion by dynamic models is one of the best ways to study walking and running. In these methods, model is considered as a multi-link rigid body. Practically determination of dynamic system variables exact values make the problem too complicated. So some assumptions are made to reduce the degree of complexity:

Bust section including head, trunk and arms are modeled by a link, such as Wojtyra [9]. Modeling of all parts are two dimensional. As running motion is often in sagittal anatomical plane one can consider that two dimensional analyses can predict three dimensional results in a good mood. As a result in this study, simulations and all links are in two dimension and joints are considered as a hinge. In the other word, without using spherical joints in ankle, knee and thigh model degrees of freedom is reduced. Friction is considered negligible in joints. [3, 9, 11]. To more idealize and smoothing the motion and due to lack of data in normal running cycle, cinematic symmetry for right and left legs are considered.

The normal human was simulated using a two-dimensional dynamic model with 7 segments, i.e., a HAT segment representing head, arms and trunk, and 6 segments representing thighs, shanks and feet of the two legs (Fig. 1). The rigid segments were connected via revolute joints at the hip, knee and ankle, providing a total of 9 degrees of freedom. The anthropometric properties of the body segments were adapted from the literature [9].

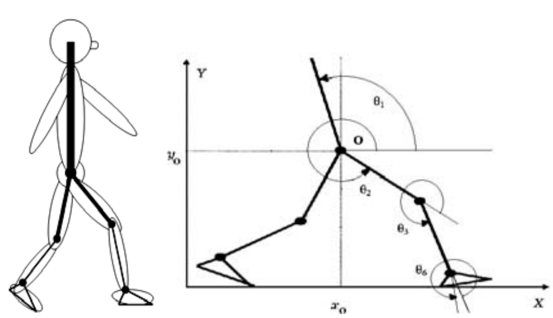

Fig. 1
The 7-link biped model of normal human body.

Table 1
Anthropometric values

parameter	unit	value
M_{HAT}	kg	51
M_{thigh}	kg	96.7
M_{shank}	kg	85.3
M_{ankle}	kg	18.1
I_{HAT}	$kg.m^2$	1.05
I_{thigh}	$kg.m^2$	0.13
I_{shank}	$kg.m^2$	0.0571
I_{ankle}	$kg.m^2$	0.0064
L_{HAT}	m	0.528
$L_{G(HAT)}$	m	0.264
L_{thigh}	m	0.435
$L_{G(thigh)}$	m	0.2
L_{shank}	m	0.425
$L_{G(shank)}$	m	0.2
L_{1ankle}	m	0.0886
L_{2ankle}	m	0.176
$L_{G(ankle)}$	m	0.065

Several normal running experiments were performed and the data of foot-ground contact, gathered by a force plate, was analyzed. Furthermore, the 3-D motion of the markers used during the normal running experiments was captured and were analyzed via our motion analysis facilities. This information was then used to calculate the relative rotations of the links. In this process, the curve fitting tool of MATLAB software was used to create a continuous data from the experiment distinct data. The data were fitted to a Fourier series of order six according to Eq. (1) below.

$$\begin{aligned} \theta_i(t) &= C_0 + \sum_{n=1}^{6} (A_n \cos(nwt) + B_n \sin(nwt)) \\ \dot{\theta}_i(t) &= \sum_{n=1}^{6} nw \times (B_n \cos(nwt) - A_n \sin(nwt)) \end{aligned} \tag{1}$$

The equations of motion of the model were derived using Lagrange's method, Eq. (2), in Maple software.

$$\frac{d}{dt}\left(\frac{\partial L}{\partial \dot{q}_k}\right) + \frac{\partial L}{\partial q_k} = Q_k \tag{2}$$

The effect of muscles in producing the driving torques was simulated using actuators located at the joints. The driving joint torques was then found so that the normal kinematical pattern of the human running cycle could be reproduced. The normal trajectories of the joints flexion angles were determined using normal running subjects. The torque generated by the actuator located in a joint was considered to be a function of the current tracking error from the reference trajectory:

$$\tau_i = K_i(\theta_i^m - \theta_i^d) + C_i(\dot{\theta}_i^m - \dot{\theta}_i^d) \qquad i = 1-7 \tag{3}$$

where Ki and Ci are constant coefficients, θ_i^m & $\dot{\theta}_i^m$ represent, respectively, the joints current angular position and velocity, and θ_i^d & $\dot{\theta}_i^d$ are, respectively, the joints reference angular position and velocity. An optimization algorithm based on the genetic optimization method was employed to find the constant coefficients of Eq.(3).

Using the Lagrange method, the position of the system is identified by 7 links angles and the coordinate of a reference point (here, the hip joint), Fig.1.

The non-linear equations of motion were numerically solved using the fourth order Runge-Kutta method. A pattern search optimization algorithm, Fig. 3, was used to optimize the following objective function

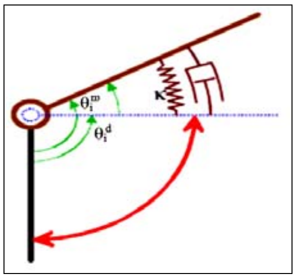

Fig. 2
Driving torque in i-th joint.

$$\int \sum_{i=1}^{7} (\theta_{normal}(i) - \theta_{model}(i))^2 \tag{4}$$

The simulation was repeated until the optimal values of the constant coefficients were found so that the differences from the reference trajectories were minimized.

2.2 Simulation of human amputee running with a prosthetic leg

The running pattern of an amputee with a below-knee prosthetic leg was simulated using a model similar to that of the normal human running. Here, the transtibial foot is modeled by a three rigid links simulating the thigh, the remaining shank and stump, and the foot. The ankle joint of the prosthetic leg was, similar to the SACH foot - equipped with a non-linear torsional spring acting as a passive motion controller, Fig. 4. An extension stop unit was also considered for the prosthetic knee to prevent it from hyper plantar flexion and hyper dorsiflexion during the running cycle. This stop unit is modeled by a torsional spring which becomes active at the final stages of ankle rotation. The locking torque is then given by Eq.(5) below.

$$T_{lock} = -E \times (\theta_6 - \theta_{6final})^2 \tag{5}$$

The ankle moments produced by a non-linear torsinal spring element controller was described as:

$$\begin{aligned} \tau &= \mathrm{K}_a \times \theta_6 \\ \mathrm{K}_a &= a \times \theta_6^2 + b \times \theta_6 + c \end{aligned} \tag{6}$$

where K_a is the magnitude of the non-linear torsinal spring's constant element and θ_6 is the amputee ankle's angle, Fig. 4.

Following formulation of the model, the optimal design parameters of the prosthetic leg were sought so that a kinematical pattern similar to that of the normal running was achieved. The joint driving torques obtained during simulation of the normal human running based on the optimized joints constant coefficients were applied to the healthy joints of the amputee. A genetic optimization algorithm was employed to repeatedly run the simulation with different values for the design parameters in order to obtain the optimal values that result in a kinematical pattern close to that of the normal human running. The objective function was defined as the integral of the absolute difference between the curves under dorsi flexion and plantar flexion angles of the normal and the simulated ankles and is shown by the optimization algorithm used was similar to that of Fig. 3

$$\int \sum_{i=1}^{7} (\theta_{ankle-dorsi/plantarflexion-normal}(i) - \theta_{ankle-dorsi/plantarflexion-prosthesis}(i))^2 \tag{7}$$

Finally, the effect of the spring stiffness variation of the prosthetic ankle on the running performance was studies.

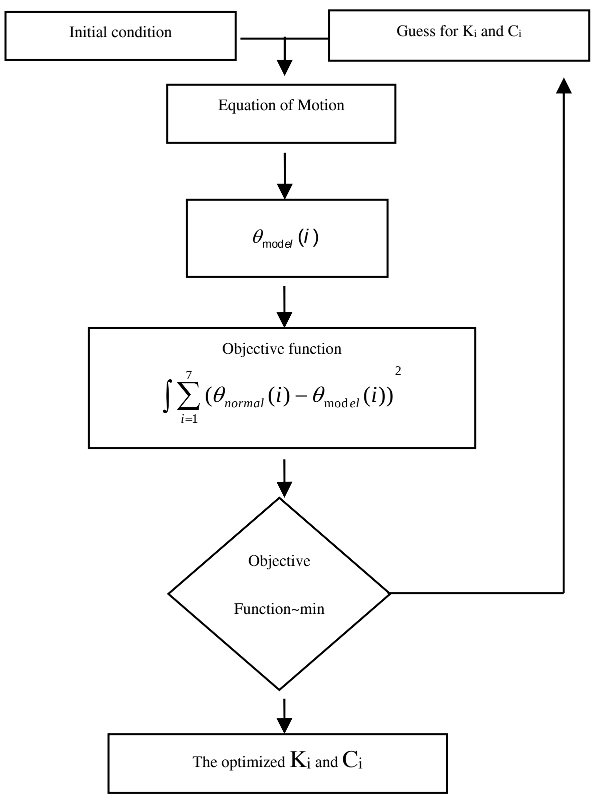

Fig. 3
The optimization algorithm to find the constant coefficients of the model.

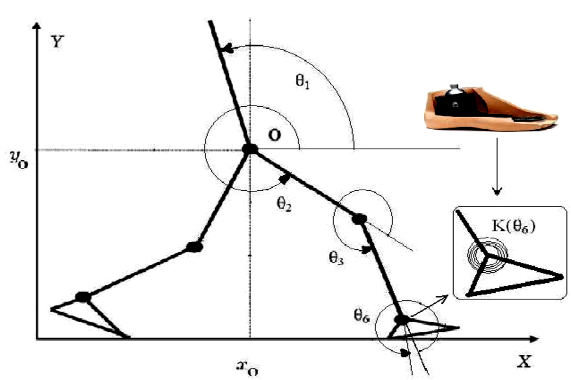

Fig. 4
The locking unit of ankle joint of the prosthetic leg.

3 RESULTS

The results of the simulation of the normal human running, based on the optimized joints constant coefficients, are shown in Fig. 5 and Fig. 6. A stick illustration of the normal human running simulation (Fig. 5) indicated a steady kinematical pattern during successive running cycles. The joint angles resulting from the model were also close to those of the reference data (Fig. 6). The joint velocity angles resulting from the model were also close to those of the reference data (Fig. 7).

The results of the simulation of the amputee running with the optimized design parameters for the prosthetic ankle are shown in Fig. 8. Results indicated a relatively good correlation between the prosthetic and normal kinematical data.

As the results shows, there is a good accordance between cinematic diagrams of this model running cycle data for trunk, legs, stalk and toes. This similarity expresses the capabilities of the model.

The effect of variation of the spring stiffness of the SACH ankle, Eq.(6), is given in Fig.9. The Figure shows the result of ±20 % stiffness change.

As we can see from the figure, there is not much variation for the diagram of knees flections angle versus gate cycle percent along oscillation phase relative to variations made in stiffness coefficient of ankle.

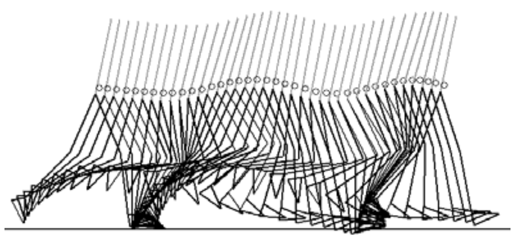

Fig. 5
Stick illustration of the normal human gait simulation.

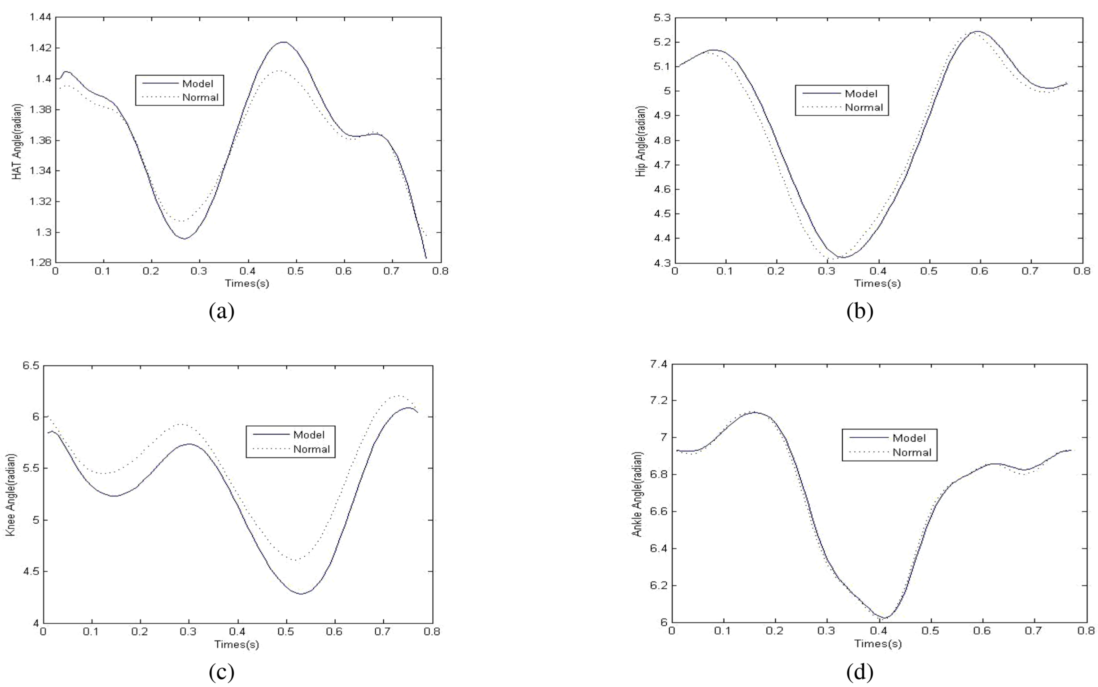

Fig. 6
Comparison of the results of modeling the normal human running and the reference data indicating the joint angles of (a) HAT, (b) hip, (c) knee, (d) ankle.

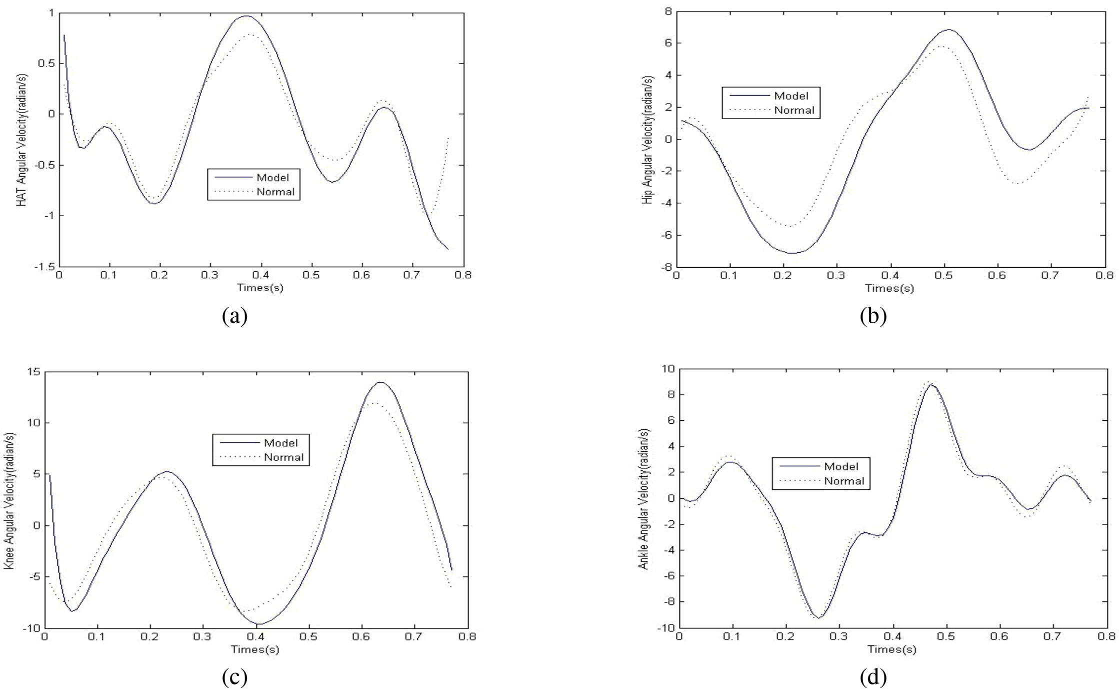

Fig. 7
Comparison of the results of modeling the normal human running and the reference data indicating the joint velocity angles of (a) HAT, (b) hip, (c) knee, (d) ankle.

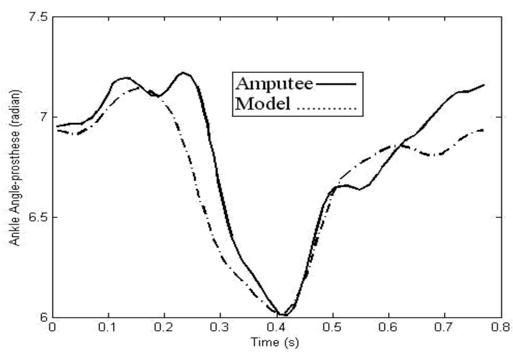

Fig. 8
The ankle plantar flexion patterns of the prosthetic during the running cycle in comparison with the normal running.

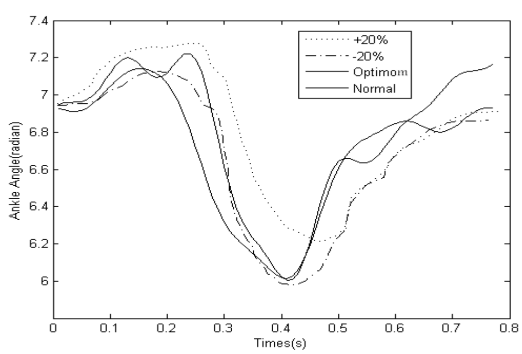

Fig. 9
The effect of spring stiffness variation on the cinematic of plantar flexion of ankle joint.

4 DISCUSSION

The results of the simulation of the normal running cycle indicate that our model could effectively mimic the general characteristics of the healthy human running. The small differences are due to the simplifications of the model. A major simplification was due to the fact that the foot was modeled as a rigid body and its interaction with the ground was simulated using a simple penetration contact model. Increasing the number of contact points [6], obviously would improve our results.

Simulation results of the amputee running cycle indicate that our model of the prosthetic leg could reasonably mimic the kinematics of the normal running cycle under normal joint driving torques, if the controller units are designed appropriately. As indicated in the diagrams of Figs. 8, 9 the major differences occur during plantar flexion due to insufficient drive for push off phase compared to that of normal running. In our model, the joints were represented by the Kelvin Voigt model which is compatible with the behavior of musculoskeletal joints. This is preferable to the inverse dynamic model since it can be used for different conditions of running and it could simulate the different contact conditions by variation of the spring and damper coefficients. This model is also capable of energy absorption during the heal contact with ground. Furthermore, we have included a proportional derivative (PD) controller to grant the stability of the HAT relative to the hip. This system helps to bring the motion of the HAT of our model close to those data of healthy subjects. Also inclusion of the stop unit in our model of ankle helps not to exceed the physiological limits of the ankle rotation. This stop unit was modeled by a torsional spring which becomes active at the final stages of ankle rotation. From Fig. 9 we understood that cinematic of SACH Foot was highly dependent on the choice of spring stiffness used in our design of the SACH Foot.

5 CONCLUSION

Some of the shortcomings of our model are as follows:

By reviewing the results, it is observed that the diagrams extracted from the model have some clear differences from normal one. These differences could be due to omitting hip in two dimensional model or errors made in estimating gait data from Fourier series. Torque diagrams in knee joint obey the normal data pattern most of the time. But in stagnation point shows a big difference. This error is due to some simplifications made in the model such as replacing knee with a hinge joint. Torque diagram in ankle joint is in a good accordance with normal running cycle data and obeys the pattern very well. Existing differences especially in stagnation phase are also because of assuming toes ass one member, removing the joint in front of the toes and limiting the contact section of foot and ground to only a point. At the end, it should be considered that in this study model only simulates the cinematic part of running. To obtain more reliable and accurate results, model should simulate both dynamic and cinematic behavior of running. Added to this, in this model torques are only applied to joints which is not happening in reality.

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