DYNAMIC SIMULATION OF THE BIPED NORMAL AND AMPUTEE HUMAN GAIT

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A two-dimensional seven link biped dynamic model was developed to investigate the mechanical characteristics of the normal and amputee locomotion during the complete gait cycle. The foot-ground contact was simulated using a five-point penetration contact model. The equations of motion were derived using Lagrange method. Optimization of the normal human walking model provided constant coefficients for the driving torque equations that could reasonably reproduce the normal kinematical pattern. The resulting torques were then applied to the intact joints of the amputee model with a prosthetic leg equipped with a kinematical driver controller for the ankle and either a hydraulic, elastic or constant friction controller for the knee joint. Design optimization of the prosthetic joints, to achieve the closet knee flexion and ankle plantar flexion pattern to that of the normal gait, resulted in a good correlation. The average differences were 5.9º for hydraulic knee, 11.6º for elastic knee and 13.2º for constant friction knee, and 11º for the prosthetic ankle. It was concluded that a hydraulic knee controller could provide a better performance in reproducing the normal gait kinematics.

1. Introduction

A prosthetic leg needs to provide reliable stability as well as an acceptable control of motion to substitute for a missing limb effectively. In particular, it is of major importance that the prosthesis reproduces the normal kinematics of the gait cycle when subjected to the normal driving torques at the amputee's healthy joints. Several experimental studies have been conducted to examine the performance of different types of above-knee prostheses in search for designs with superior features [1, 2, 3]. 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 modeling studies on the transfemoral amputee gait, however, have been often limited to a single limb or a part of the gait cycle, e.g., the swing phase [4, 5, 6, 7].

The purpose of the present study was to employ the mathematical modeling approach to analyze the dynamics of a 7-link biped with an above knee prosthesis during the complete gait cycle, including both the swing and stance phases. This model was then used to determine the optimal design parameters of three different knee controller mechanisms, as well as the prosthetic ankle, in order to achieve the closest kinematics to the normal gait.

2. Method

2.1. *Simulation of normal human gait*

The normal human gait 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 hip, knee and ankle, providing a total of 9 degrees of freedom. The anthropometric properties of the body segments were adapted from the literature [8].

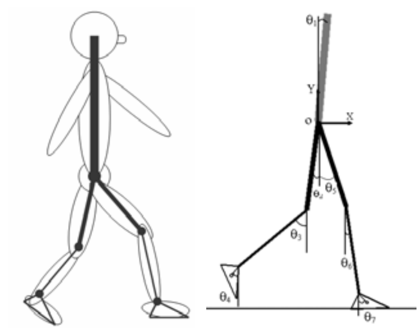


Figure 1. The 7-link biped model of human gait.

The effect of muscles in producing the driving torques was simulated using actuators located at the joints. The foot-ground contact was simulated using a penetration contact model [3], considering five contact points (Fig. 2).

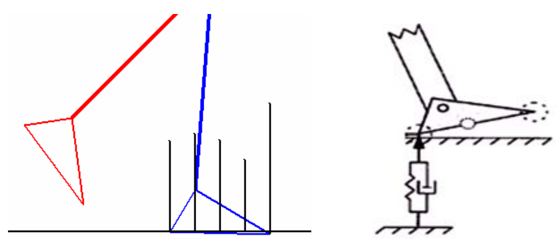


Figure 2. The foot-ground contact model included 5 distributed contact points (left) and a spring-damper system at each point to represent the vertical ground reaction force (right).

The vertical ground reaction force was formulated using non-linear spring-damper elements attached perpendicularly to the flat rigid ground surface (Fig

2) with a damping coefficient assumed to be a non-linear function of the ground penetration [3]:

$$F_N = K_G\ (d_y) + C_G\ (\dot{d}_y) \tag{1}$$

$$C_G = \begin{cases} c_{max}\left|\dfrac{3}{h^2} d_y{}^2 - \dfrac{2}{h^3} d_y{}^3\right| & d_y \langle\ h \\ c_{max} & d_y \langle\ h \end{cases} \tag{2}$$

Where d_y, $\dot{d}_y$ are the ground penetration and penetration rate at contact points.

The horizontal contact force was formulated using a Coulomb friction model [9]:

$$F_H = \mu\ F_N\ sign(\dot{x}) \tag{3}$$

where $\dot{x}$ is the horizontal velocity of the contact point relative to the floor and μ is the coefficient of friction; the $\mathrm{sign}(\dot{x})$ term ensures that the friction force acts in a direction opposing the relative motion. Two different coefficients of friction were used for static and dynamic conditions, μ_{static} when $\dot{x}$ is sufficiently small (bellow 0.05 m/s) and $\mu_{dynamic}$ otherwise. The constant force parameters used in the model are summarized in Table 1 [8, 9].

Table 1. Contact model parameters.

Parameter	value
K_G	1.5×10^4 N/m
C_{max}	1500 Ns/m
h	0.01 m
μ_{static}	0.8
$\mu_{dynamic}$	0.2

The equations of motion of the model were derived using Lagrange method. The driving joint torques were then found so that the normal kinematical pattern of the human gait could be reproduced. The normal trajectories of the joints flexion angles were adapted from the literature [10]. 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:

$$\tau_i = K_i\ (\theta_i^m - \theta_i^d) + C_i\ (\dot{\theta}_i^m - \dot{\theta}_i^d) \qquad \text{i= 1 to 7} \tag{4}$$

where K_i and C_i are constant coefficients, θ_i^m $\dot{\theta}_i^m$ represent the joints current angular position and velocity, and $\theta_i^d, \dot{\theta}_i^d$ are the joints reference angular position and velocity. An optimization algorithm based on the genetic optimization method was employed to find the constant coefficients of Equ. (1). The nonlinear equations of motion were numerically solved using the fourth order Runge-Kutta method. In a pattern search optimization algorithm (Fig 3) the simulation was run repeatedly to find the optimal values of the constant coefficients so that the differences from the reference trajectories were minimized.

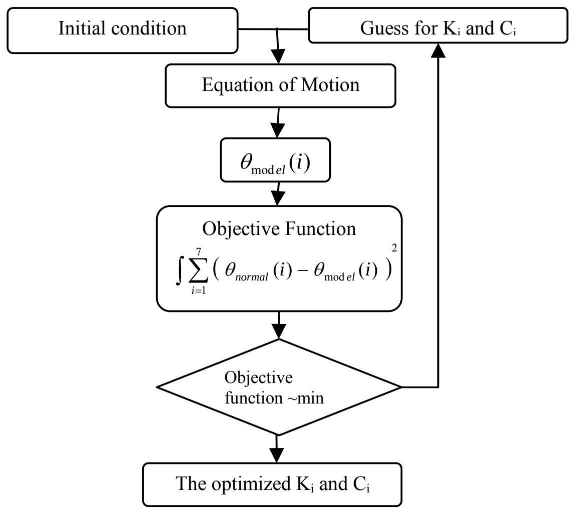


Figure 3. The optimization algorithm to find the constant coefficients of the model of the normal human gait.

2.2. *Simulation of amputee human gait with a prosthetic leg*

The walking pattern of an amputee with an above-knee prosthetic leg was simulated using a model similar to that of the normal gait (Sec 2.1). The knee and ankle joints of the prosthetic leg, however, were equipped with passive motion controllers, instead of actuators (Fig. 4.a). Three types of controllers, i.e., elastic (Fig 4.b), hydraulic (Fig 4.c) and coulomb friction (Fig 4.d), were considered for the prosthetic knee and kinematical driver controller, i.e., a torsinal spring-damper element, for the prosthetic ankle. An extension stop unit was also considered for the prosthetic knee to prevent it from hyper extension during the gait cycle.

Each of the knee mechanisms was modeled separately and its governing equations were embedded into the main equations of the model subsequently. The knee moments produced by the Coulomb friction controller was described as:

$$\tau_{fric} = D\,\mathrm{sgn}(\dot{\theta}_4 - \dot{\theta}_5) \qquad (5)$$

where D is the magnitude of the friction moment and sgn is a switching function which yields the sign of $\dot{\theta}_4 - \dot{\theta}_5$ and ensures that the friction moment acts in a direction opposing the relative motion of the knee. For the hydraulic controller:

$$\tau_{hyd}=C\left|b\dot{\theta}_5\sin(\psi-\theta_5+\gamma)+a\dot{\theta}_4\sin(\psi-\theta_4-\alpha-\beta)\right|\times \tag{6}$$
$$[a^2\sin^2(\psi-\theta_4-\alpha-\beta)\dot{\theta}_4+ab\sin(\psi-\theta_5+\gamma)\sin(\psi-\theta_4-\alpha-\beta)\dot{\theta}_5]$$

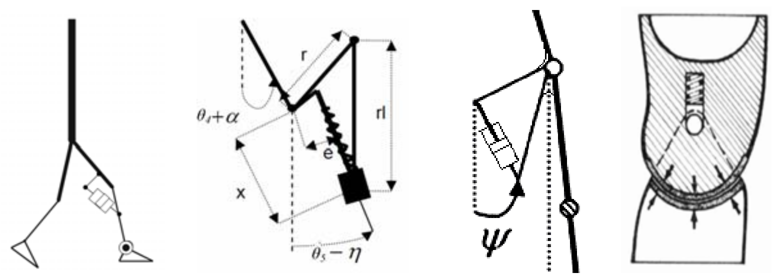
where C is the damping coefficient of hydraulic controller. Finally for the elastic controller:

$$\tau_{elas}=\{r_r\sin(\xi-\theta_4-\alpha+\theta_5-\eta)-[e-r_r\sin(\xi-\theta_4-\alpha+\theta_5-\eta)]r_r \tag{7}$$
$$\cos(\xi-\theta_4-\alpha+\theta_5-\eta)/R\}K\,x$$

where K is the spring constant and other parameters are defined as the following:

$$R=[r_l^2-[e-r_r\sin(\xi-\theta_4-\alpha+\theta_5-\eta)]^2$$
$$x=-r_r\cos(\xi-\theta_4-\alpha+\theta_5-\eta)R-l_0$$
$$l_0=l+\text{ustretchrd-length}$$

All other constants for the knee controllers are given in Table 2 [5].



(a) (b) (c) (d)

Figure 4. (a) The 7-link biped model of human gait while wearing an above knee prosthetic leg with (b) elastic, (c) hydraulic, and (d) coulomb friction knee mechanisms.

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 gait is achieved. For each knee controller, two different coefficients were obtained for each of the stance and swing phases of gait. The joint driving torques obtained during simulation of the normal human gait based on the optimized joints constant coefficients (Sec 2.1) 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 and obtain the optimal values that result in a kinematical pattern close to that of the normal gait.

Table 2. Knee controllers' parameters.

Parameter	Value
Angle between Hip-knee and crank of spring mechanism : ξ	80 deg
Length of the spring mechanism coupler: r_l	0.18
Length of the spring mechanism crank: r_r	0.4

Offset of the spring mechanism : e	0.04
Horizontal distance from upper attachment point of the controller of the knee joint: a	0.031
Vertical distance from upper attachment point of the controller of the knee joint: b	0.21
Angle between knee-ankle and knee- shank cg: γ	0
Angle between hip-knee and knee- upper attachment point: β	$\pi/2$
Distance from upper attachment point to cm of controller: r_c	0.102

3. Results

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

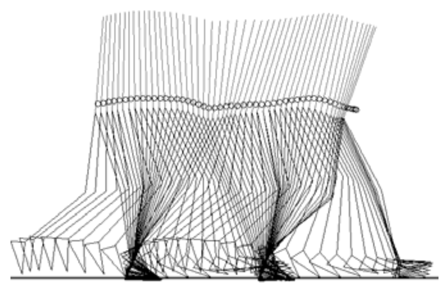
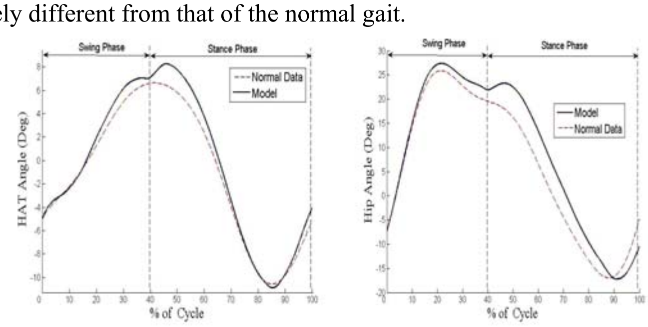
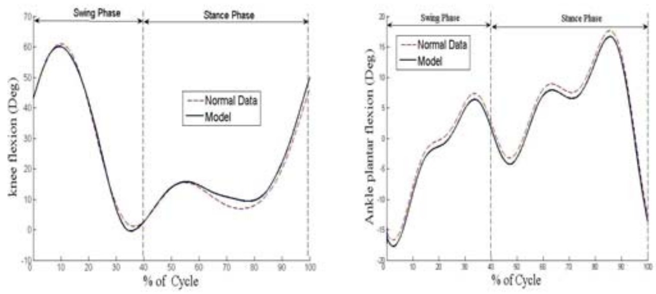


Figure 5. Stick illustration of the normal human gait simulation.

The results of the simulation of the amputee gait with the optimized design parameters for the prosthetic ankle and the three controllers of the prosthetic knee are shown in Fig 7. Results indicated a relatively good correlation between the prosthetic and normal kinematical data. For knee flexion, the average differences were 5.9º with hydraulic controller, 11.6º with elastic controller and 13.2º with constant friction controller. For the ankle joint the range of motion of the prosthetic and intact joints were similar, however, the general pattern was relatively different from that of the normal gait.

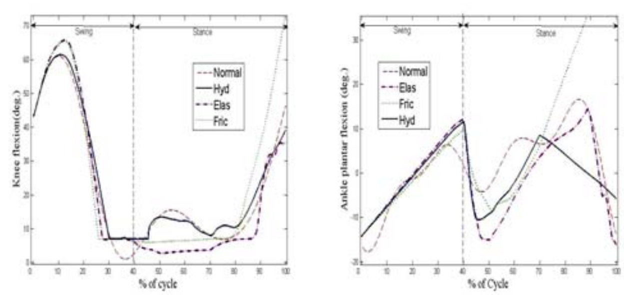


(a) (b)



(c) (d)

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



(a) (b)

Figure 8. The flexion patterns of the prosthetic joints during the gait cycle in comparison with those of the normal gait (a) knee flexion, (b) ankle plantar flexion.

4. Discussion

The results of the simulation of the normal gait cycle indicate that the biped model of the present study could effectively mimic the general characteristics of the human gait. The small differences are thought to be due to the simplifications of the model. A major simplification was 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. Although an increased number of the contact points, in comparison with the previous studies [5], provided a more realistic presentation of the foot-ground interaction, a more sophisticated flexible model in needed to improve the results significantly.

The results of the simulation of the amputee gait indicate that the prosthetic leg can reasonably reproduce the kinematics of the normal gait under normal joint driving torques, if the controller units are designed appropriately. In particular, the hydraulic controller provided a kinematical pattern highly correlated with that of the normal gait. The prosthetic ankle, however, could not reproduce the plantar flexion motion to drive the push-off phase of the normal gait cycle, due to its passive characteristics.

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