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Discrete control of transfemoral prostheses for human walking with magnetorheological compensation

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Abstract. This abstract presents the modeling and control of a transfemoral prosthesis with magnetorheological compensation. Initially, leg modeling is shown by the formulation of the Euler-Lagrange motion equations and the inclusion of the modified Bouc-Wen model for the magnetorheological damper. Subsequently, the linealization of these models is presented using the Taylor series approximation. For the design of the controller is used the analysis of the geometric place of the roots. The simulation results of the control system indicate that the prosthesis is capable of accurately following established joint references and effectively rejects external disturbances.

1. Introduction

Biomedical uses engineering knowledge towards the field of medicine, in which it seeks the invention of equipment capable of diagnosing diseases with greater speed and accuracy, in addition, it is responsible for the design of new prostheses for disabled people that allow better mobility and prevent future damage [1]. Prostheses are used to replace a missing body member or part. There are several types of prostheses, some being cosmetic and others more functional [2]. The latter seek the implementation of electronic circuits that manage to capture signals and perform automatic joint control. Such prostheses may be pneumatic, hydraulic or bioelectric [3], and are designed to replace and fulfil the task of that member. In the human march the individual departs from a vertical position and begins to move forward alternating the weight of the body on its lower limbs [4]. When a person loses a leg, it becomes difficult to walk, so a prosthesis is used to help improve this process. In this research, a controller is designed for a magnetoreological compensation prosthesis that can be implemented in an embedded system. In Section 2, the Euler - Lagrange formalism is applied for dynamic leg modeling during the gait cycle. Section 3 presents the Bouc-Wen model modified to describe the behavior of the magnetorheological shock absorber that replaces the knee joint. Section 4 the design of a discrete control using the root geometrical location method and an analysis of the results obtained for reference monitoring and disturbance rejection. Finally, section 5 presents the conclusions.

2. Theoretical model

The leg was modeled using the Euler - Lagrange formalism that describes the equations of motion for the joint of both the hip and the knee from the energies of the system [5]-[6], in Figure 1 shows the variables of our system.

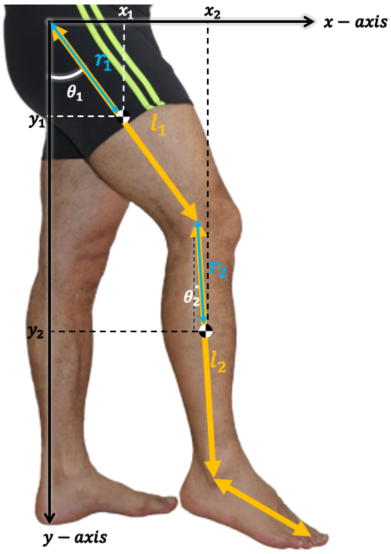

Figure 1. Variables of system

Based on the angle convention in Figure 1, we have two generalized coordinates that are (θ_1, θ_2), resulting in the following equations, resulting the motion equations,

$$\begin{aligned}\tau_1 &= \ddot{\theta}_1(m_1r_1^2 + m_2l_1^2) + m_2l_1r_2\ddot{\theta}_2\cos(\theta_1 - \theta_2) + m_2l_1r_2\dot{\theta}_2^2\sin(\theta_1 - \theta_2) + \\ &+ (m_1r_1 + m_2l_1)g\sin\theta_1 + \beta_1\dot{\theta}_1.\end{aligned} \tag{1}$$

$$\begin{aligned}\tau_2 &= \ddot{\theta}_1 m_2l_1r_2\cos(\theta_1 - \theta_2) + m_2r_2^2\ddot{\theta}_2 - m_2l_1r_2\dot{\theta}_1^2\sin(\theta_1 - \theta_2) + \\ &+ m_2r_2g\sin\theta_2 + \beta_2\dot{\theta}_2.\end{aligned} \tag{2}$$

2.1. Numerical solution of the leg

The numerical model of the leg defined by the Euler - Lagrange equations (1) and (2) is given by substituting the values of the constants presented; here, $m_1 = 8\ kg$, is thigh mass; $m_2 = 3.72\ kg$, the calf mass; $l_1 = 0.45\ m$, thigh length; $r_1 = 0.195\ m$, length to center of mass of thigh; $r_2 = 0.165\ m$, length to center of mass of calf; $g = 9.81\ m/s^2$, the gravitational acceleration; $\beta_1 = 2.288\ kg/s$, hip coefficient of friction and $\beta_2 = 0.175\ kg/s$ knee friction coefficient.

The measurements were taken for a male person 1.77 m tall and 80 kg, the mass of the thigh segment and the calf next to the center of mass of each one was taken according to the synthesis of classic anthropometry works presented by [7].

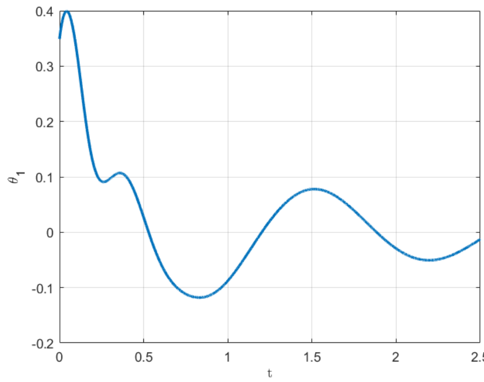

Figure 2. Angular position of the hip

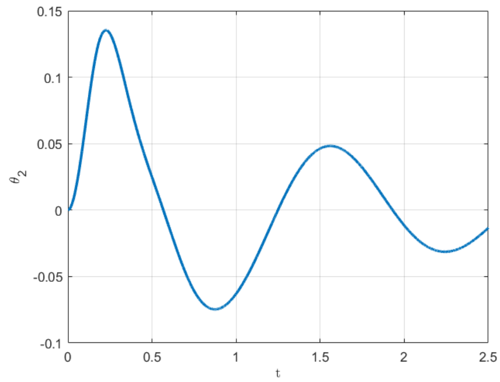

Figure 3. Angular knee position

The Figure 2, begins at 0.349 *rad*, which represents the angular position of the hip that oscillates according to the equation of motion, while the Figure 3, begins at 0° for the neutral position of the knee and continues to oscillate because the equations of motion are coupled.

3. Modeling the magnetorheological damper

The magnetoreological damper is modeled based on the parameters proposed by [8] and the modified Bouc-Wen model that describes the hysteresis phenomenon generated in the damper [9]. Its representation is shown in Figure 4.

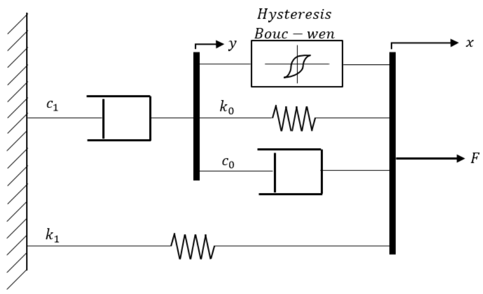

Figure 4. Bouc-Wen model

The force generated by the shock absorber is given by the expression $F = c_1\dot{y} + k_1(x - x_0)$, being k_1, are the spring stiffness constants and c_1 are the damping coefficients, and it is an internal state that reproduces the sliding effects present at low speed expressed by the following equation,

$$\dot{y} = \frac{1}{(c_0 + c_1)}[\alpha z + c_0\dot{x} + k_0(x - y)], \tag{3}$$

here, α is a parameter as a function of the hysteresis phenomenon, z is the restoring force as a function of the shock absorber displacement, the expression of z is defined in the following expression, that is, $z = -\gamma|\dot{x} - \dot{y}|z|\dot{z}|^{n-1} - \mu(\dot{x} - \dot{y})|\dot{z}|^n + A(\dot{x} - \dot{y})$, where, μ, γ and A are the constants that form the hysteresis cycle, through this modeling the error is minimized and the behavior of the damper is described.

3.1. Numerical solution of the magnetorheological damper

The equations of the Bouc-Wen model modified in Simulink of the Matlab software were constructed, the values of the parameters given by Spencer are presented in Table 1.

Table 1. Modified Bouc - Wen model parameters.

Variable	Value	Units	Variable	Value	Units
c_{0a}	2.71×10^3	Ns/m	γ	3.63×10^6	m^{-2}
c_{0b}	3.50×10^2	Ns/mV	β	3.63×10^6	m^{-2}
c_{1a}	2.83×10^4	Ns/m	A	301	–
c_{1b}	2.95×10^2	Ns/mV	x_0	0.143	m
α_a	1.40×10^4	N/m	η	190	s^{-1}
α_b	6.95×10^4	N/mV	k_1	500	N/m
k_0	4.69×10^3	N/m	η	2	-

The simulation was carried out with a sinusoidal displacement of amplitude 1.5 *cm* and frequency 2.5 *Hz*, the results of the model are shown in the Figures 5 and 6, with different voltage values at the input.

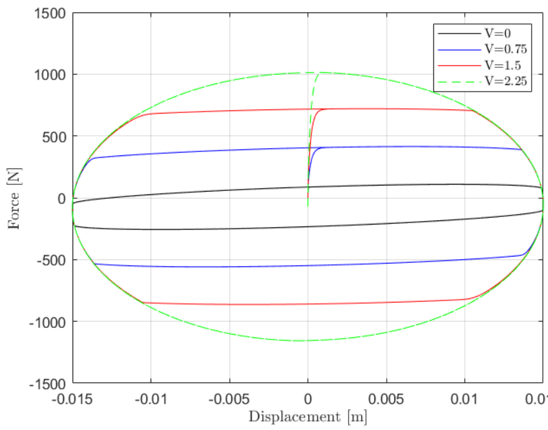

Figure 5. Force vs Displacement.

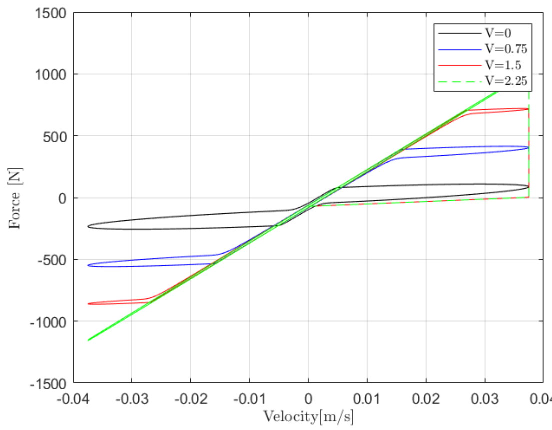

Figure 6. Force vs Velocity.

In Figure 5 we have the result of force against displacement, which describes the behavior of both extension and compression of the shock absorber, on the other hand, in Figure 6 the phenomenon of hysteresis generated when the shock absorber does not exert force is observed. disappears instantly due to the effects of materials.

4. Controller design

The controller is designed from the linearization of the nonlinear system with the equations in state space.

4.1. Linealization

Linearization is implemented using the linear approximation by Taylor series expansion [10]-[11], where the first order terms of the derivative of the differential equation evaluated at the equilibrium point are taken, which in this case is $(0,0,0,0)$ when the leg is in a vertical position, that is, when the person is standing on both feet. The linearized system in state space is written as [12]-[13], this is, $\dot{\mathbf{x}} = A\mathbf{x} + Bu$ and $y = \widetilde{C}\mathbf{x}$, where, A, B y $\widetilde{C}$, are the Jacobian matrices of the system and replacing the value of the constants of the subsection 2.1, the following linearized matrices of the system are obtained,

$$A = \begin{bmatrix} 0 & 1 & 0 & 0 \\ -104.2917 & -7.5214 & 53.9840 & 1.5689 \\ 0 & 0 & 0 & 1 \\ 284.4320 & 20.5128 & -206.6837 & -6.0069 \end{bmatrix} \qquad B = \begin{bmatrix} 0 \\ -8.9654 \\ 0 \\ 34.3250 \end{bmatrix} \qquad \widetilde{C} = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & 0 & 1 & 0 \end{bmatrix}$$

4.2. Digital PI controller design

The design of the digital PI control is done by means of the geometric place of the roots from the discrete model of the system [14], to minimize the energy consumption the controller is designed with overshoot or over-peak specifications of less than 10% and (t_s) is a settling time of less than 2 seconds for the system to stabilize quickly,

$$\xi = \frac{\ln\left(\frac{OV}{100}\right)}{\sqrt{\pi^2 + \ln\left(\frac{OV}{100}\right)^2}} = 0.5912 \qquad \omega_n = \frac{4}{\xi t_s} = 3.3832 \qquad \omega_d = \omega_n\sqrt{1-\xi^2} = 2.7288 \tag{4}$$

In the system equations (4), the variables ξ, represent the damping coefficient; ω_n, the undamped natural frequency and ω_d, the damped frequency, with the above factors and the sampling time T_s of 0.1 s, the pole of the controller that meets the following specifications is determined, which are defined as follows, $|z| = \mathrm{e}^{-T_s\xi\omega_n} = 0.8237$, indicates the magnitude of the pole and $\angle z = T_s\omega_d = 0.2647\ rad = 15.1656°$, represents the pole angle with this the controller pole can be presented in both polar and binomial form, that is, $z = 0.8237\angle 15.1656° = 0.795 + 0.2155\,i$.

The design by means of the locus of the roots allows to know the condition of both magnitude and angle to determine the controller gains [15]. With the zeros and poles of the knee transfer function we obtain the angle of each with respect to the pole point of the controller to determine the angle provided by the controller, which is obtained with the difference between the sum of both the angles of the zeros and the poles and equaled to 180°, the angle provided by the controller is defined to be $-192.5681°$, where, the Figure 7 represents the response of the system with the controller to a step input. The controlled system presents an overshoot of approximately 10% and a settling time close to 2 s. Both values match the time specifications established during design.

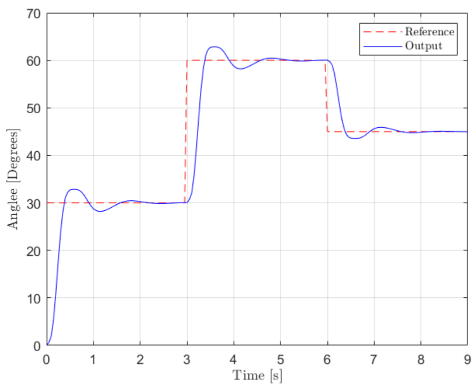

Figure 7. Knee response to step entry with engineered PI control.

In the Figure 8, shows the controller's response to a strong disturbance at the plant entrance. After tracking the reference, this disturbance is applied at 2.5 seconds. In the figure it can be seen that the control system recovers quickly.

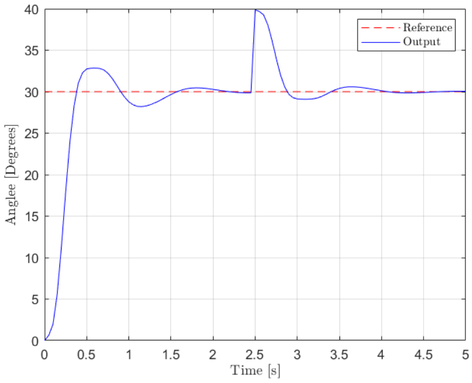

Figure 8. Knee response to step entry with engineered PI control.

5. Conclusions

In this work, the modeling and control of a lower limb prosthesis activated by a magnetoreological damper was presented. The Euler - Lagrange formalism and the implementation of the modified Bouc - Wen model for the magnetoreological damper acting as the knee in the leg model allows to obtain the equations of motion of the leg with the magnetoreological compensation. Using the Taylor series approximation, the linearized model of the open-loop system is obtained. With linear model and temporal specifications, such as settling time and overshoot, the PI controller is designed using the roots locus. The simulation results indicate that the control system is robust to external disturbances and can faithfully follow any desired positional reference.

Acknowledgments

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