CONTROL SYSTEM FOR AN ACTIVE AFO TO ASSIST ANKLE DORSIFLEXION MOVEMENTS IN SWING PHASE OF GAIT

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Abstract: An inadequate dorsiflexion movement during the swing phase of gait is a common disability in patients after a stroke accident and affects the normal gait. To assist this biomechanical impairment, orthopedically doctors prescribe ankle foot orthosis (AFO), nevertheless, this mechanical support in many occasions is not successful. This paper presents a state feedback controller for an active AFO for advancing the foot in the swing phase to a neutral position with a DC motor. To determine the torque required to assist the ankle in the sagittal plane, a model of the system leg-foot-ankle is developed. Also a mechanical design of the AFO is presented. Results indicate that the control system discussed can be used for an active AFO in cases of the drop foot assistance for rehabilitation applications.

Keywords: Active orthosis, Ankle, Foot drop.

Introduction

The main musculoskeletal deformity in patients after stroke is an insufficient dorsiflexion movement in the swing phase of the gait, commonly called foot drop (FD) [1], [2]. This biomechanical impairment is caused by weakness of the leg muscles involved in this movement. This results in a steppage type gait pattern, which is dangerous to patients for the high risk of fall and because it alters the load distribution [3].

Polypropylene ankle foot orthosis (AFO) makes up the bulk of the devices prescribed by clinicians to prevent the FD, maintaining toe clearance during the swing phase [2]. AFO is part of the walking assistance for these people with disabilities, been considered as an important element in rehabilitation of patients with stroke [3], [4]. However, the passive nature of these AFOs limits the functional benefit they are capable of providing [1]. For example, it does not allow a gradual advance of dorsiflexion.

Projects to develop active orthosis to assist movement impairment represent an alternative for people with this physical disability in order to improve their quality of life [5], [6], [7], [8]. These types of active devices have an onboard or tethered source of power, one or more actuators to move the joint, sensors and a computer to control the application of torque during gait [1]. However, despite recent advances in computing, sensing and enabling technologies, there are currently no practical portable powered AFO systems [1].

This paper presents the design of a state feedback integral controller used as feedback from the ankle angle as an option of control for a portable active AFO to assist FD for rehabilitation applications.

Materials and methods

A biomechanical model was developed in Matlab-Simmechanics to determinate the torque required for correct dorsiflexion in the swing phase and to define the plant for the controller. In this instance, simulations in inverse dynamic mode were developing. Also, based on the designed controller, a proposal of portable active AFO is here presented.

Simmechanics model of the swing phase – The system has one degree of freedom, with the leg and the foot segments linked by a joint that represents the ankle (Figure 1(a)). Equations (1) and (2) express this system [7], [9].

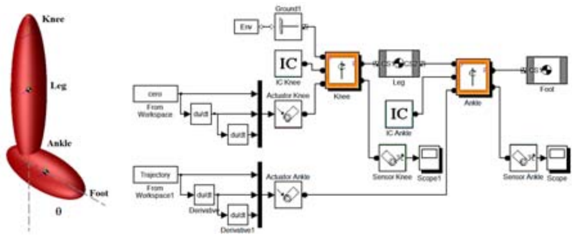
$$(J_c + md^2)\ddot{\theta} + k\dot{\theta} + mgdSen\theta = T_d \tag{1}$$

$$T_t = T_d - T_c \tag{2}$$

where: θ represents the angle between the leg and the foot long axis, J_c is the inertia of the foot, m is the mass of the body and d represents the distance between the ankle joint and the center of mass of the foot. Tt is the torque of the joint and T_c is the orthosis torque by joint friction. k represents a coefficient to adjust the angular trajectory between mathematical and biomechanical models.

Figure 1(b) shows the block diagram in simmechanics that represents the equations (1) and (2). To configure the block that represent a body, the distance, mass and inertial moments of the segments leg and foot are required [14]. Table 1 shows the values of these parameters.

In normal conditions, the torque required to keep the foot to the neutral position in the swing phase is 0.04 N·m/Kg [10]. Nevertheless, to forward the foot in the pre-swing phase of the gait a higher torque is required.



(a) (b)

Figure 1: Simmechanics model (a) leg, foot and the ankle joint. (b) block diagram.

Simulations, considering a user of 50 kg and 1.52 m, demonstrate that to advance the foot 20° of dorsiflexion the torque required is 0.31 N·m. Considering the FD case with a deficiency of 10°, to forward the foot to neutral position (90°), the maximum torque required is 0.15 Nm (figure 2). In that instance, the torque that the control actuator needs to apply at the joint is 0.16 N·m. To this case, we assume that the gait speed does not change.

Control System – The controller feedbacks the ankle angle to make corrections to keep the foot in the neutral position during the swing phase. A device to identify this phase with contact sensor is required, also a sensor to measure the ankle angle [7]. Figure 3 shows the block diagram of the system, and equations (3) and (4) represent the state continuous system.

$$\dot{x_1} = x_2 \quad (3)$$

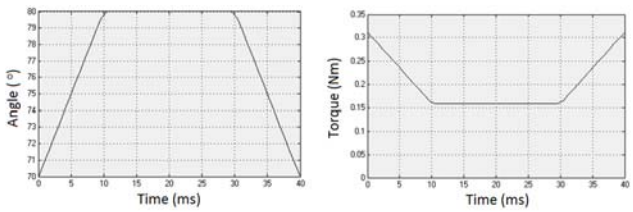
$$\dot{x_2} = \frac{T_t + T_c - kx_2 - mgdSenx_1}{J_c + md^2} \quad (4)$$

For this system, to assist the ankle sagittal movement, a DC motor with angular rotary action and torque suitable was chosen. The dynamics of the DC motor based on Kirchhoff's and Newton's laws is given by:

$$V(t) = Ri(t) + L\frac{di(t)}{d_t} + k_e\dot{\varphi}(t) \quad (5)$$

$$T(t) = J_m\,\ddot{\varphi}(t) + b\dot{\varphi}(t)\,, \quad (6)$$

where $V(t)$ is the motor supply voltage, R and L are the resistance and inductance of the motor winding respectively, $i(t)$ is the armature current, and k_e is the electromotive force constant.



(a) (b)

Figure 2: DF case. (a) trajectory; (b) torque.

Table 1: Five-percentile (5%) female anthropometric data of body for leg and foot segments [12], [13].

Parameters	Leg	Foot
Length [m]	0.2246×h	0.1520×h
Mass [kg]	0.0465×m	0.0145×m
Inertia matrices [kg·m^2]	$\begin{bmatrix} 0.0369 & 0 & 0 \\ 0 & 0.0369 & 0 \\ 0 & 0 & 0.0026 \end{bmatrix}$	$\begin{bmatrix} 0.0037 & 0 & 0 \\ 0 & 0.0037 & 0 \\ 0 & 0 & 0.0008 \end{bmatrix}$

h is the user height and m is the user mass.

$T(t)$ is the torque applied by the motor, J_m is the inertia of the actuator armature, $\dot{\varphi}(t)$ is the angular velocity of the motor shaft, and b is the coefficient of viscous friction. Assuming the constant magnetic field, the torque is proportional to the armature current (7).

$$T(t) = K_t i(t)\,, \quad (7)$$

where K_t is the constant of armor. (3), (8) and (9) describe the complete model assumed the angular velocity equal to the DC motor.

$$\dot{x_2} = \frac{K_t x_3 - kx_2 - mgdSenx_1}{J_c + md^2} \quad (8)$$

$$\dot{x_3} = \frac{V - K_e x_2 - Rx_3}{L} \quad (9)$$

The linear system representation in the state space with the balance point X_{bal} (0°) is given by:

$$\begin{bmatrix} \widehat{\dot{x_1}} \\ \widehat{\dot{x_2}} \\ \widehat{\dot{x_3}} \end{bmatrix} = \begin{bmatrix} 0 & 1 & 0 \\ \frac{-mgd}{J_c+md^2} & \frac{-k}{J_c+md^2} & \frac{K_t}{J_c+md^2} \\ 0 & -\frac{K_e}{L} & -\frac{R}{L} \end{bmatrix} \begin{bmatrix} \widehat{x_1} \\ \widehat{x_2} \\ \widehat{x_3} \end{bmatrix} + \begin{bmatrix} 0 \\ 0 \\ 1/L \end{bmatrix} u(t) \quad (10)$$

$$\hat{y} = \begin{bmatrix} 1 & 0 & 0 \end{bmatrix} \begin{bmatrix} \widehat{x_1} \\ \widehat{x_2} \\ \widehat{x_3} \end{bmatrix} \quad (11)$$

For the control law, the closed-loop system is represented as:

$$\dot{x}(t) = (A - BK)x(t) + Br(t) \quad (12)$$

$$y(t) = Cx(t)\,, \quad (13)$$

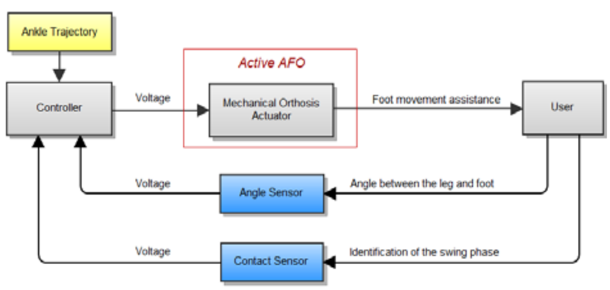
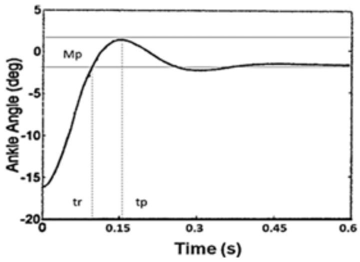


Figure 3: Block diagram of the control system.



$$M_p = e^{-\frac{\zeta\pi}{\sqrt{1-\zeta^2}}}$$

$$t_r = \frac{1}{\omega_n\sqrt{1-\zeta^2}}\tan^{-1}\left[\frac{\sqrt{1-\zeta^2}}{\zeta}\right]$$

$$t_p = \frac{\pi}{\omega_n\sqrt{1-\zeta^2}} \qquad t_s = \frac{4}{\zeta\omega_n}$$

Figure 4: Human ankle trajectory for slow gait speed (0.55m/s).

where K is the gain matrix of state feedback. This value is defined by the pole placement method. For this, the damping ratio and natural frequency for gait speed were found using (14) and (15) [14].

$$\zeta = \frac{-\ln(\%OS/100)}{\sqrt{\pi^2 + \ln^2(\%OS/100)}} \qquad (14)$$

$$\omega_n = \frac{\pi}{t_p\sqrt{1-\zeta^2}} \qquad (15)$$

where ζ is the damping ratio, %OS is the percent overshoot, ω_n is the natural frequency and t_p the peak time. The parameters that describe the temporal response performance were found using the second order model to FD gait shows in Figure 4, adapted from Blaya [14]. In this, M_p is the maximum overshoot, t_r is the rise time, and t_s in the setting time. For very slow gait speed (0.55 m/s) [11], the swing phase is 0.6 s.

The augmented system is given by (16) and (17).

$$\begin{bmatrix}\dot{x}\\ \dot{x_a}\end{bmatrix} = \begin{bmatrix}A-BK & -BK_a\\ -C & 0\end{bmatrix}\begin{bmatrix}x\\ x_a\end{bmatrix} + \begin{bmatrix}0\\ 1\end{bmatrix}r \qquad (16)$$

$$\dot{x}_a = r - y = r - Cx \qquad (17)$$

To find the control matrix K the command `K = place (A, B, P)` was used. Where P is a vector desired self-conjugate closed-loop pole locations. Figure 5 shows the Simulink block diagram of the mathematical model, and the controller (tracking controller on nonlinear model described by (1) and (2)).

In the control system of Veneva [7], a rotary voice coil actuator was employed with good results. For this reason, for our controller, the following parameters for this actuator was used: Kt = 0.113 [Nm/A], Ke = 0.11 [V/(rad/seg)], R = 13 [Ω] and L = 0.01 [H].

Results

Figure 6 shows the result obtained by the control system with integral action on nonlinear model. In this figure, the reference signal is the smooth line and the angle output is the line marked with x. The reference represents an angular trajectory of the ankle in the

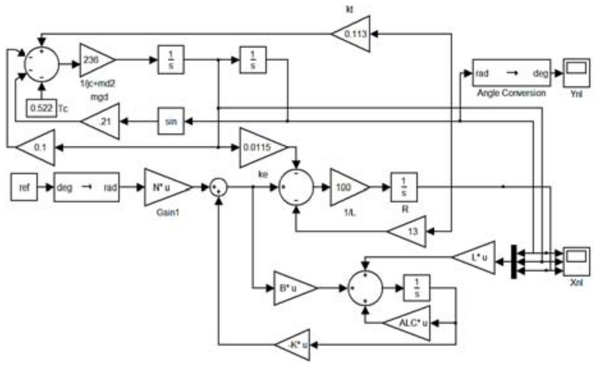


Figure 5: Controller and nonlinear plant.

swing phase. Figure 7 shows the trajectory tracking of a position of the foot in order to advance different degrees of dorsiflexion.

Figure 8 shows the portable active AFO proposed, designed using the software Solidworks. The total weight of the AFO is 1.01 kg and is within the range of other prototypes (0.95 kg [15] and 2.2 kg [6]). The total excursion angle of the motor is 8º of maximum dorsiflexion and 24º of plantar flexion.

The AFO includes a potentiometer to measure the ankle joint angle and two force sensors are attached to the sole in order to determine the beginning of the gait cycle and recognize the swing phase [10], [25]. An electric circuit with a dsPic microcontroller and 4 Ni-Cd batteries are attached at the back of the shank (Figure 8).

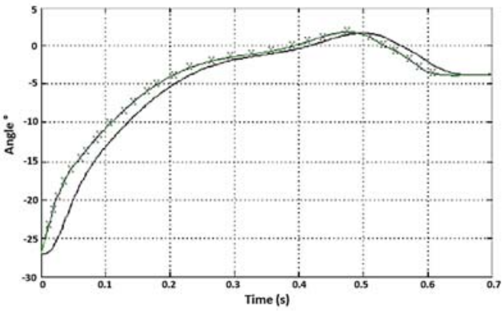


Figure 6: Trajectory tracking of the control with integral action on nonlinear model.

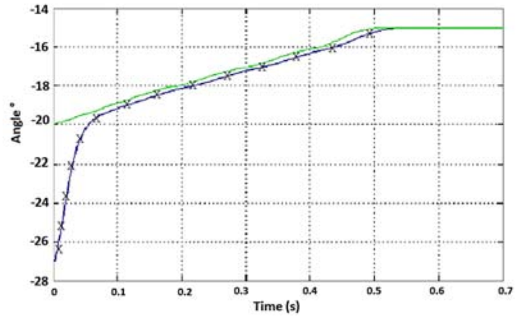


Figure 7: Trajectory tracking of a position of the foot in order to advance different degrees of dorsiflexion.

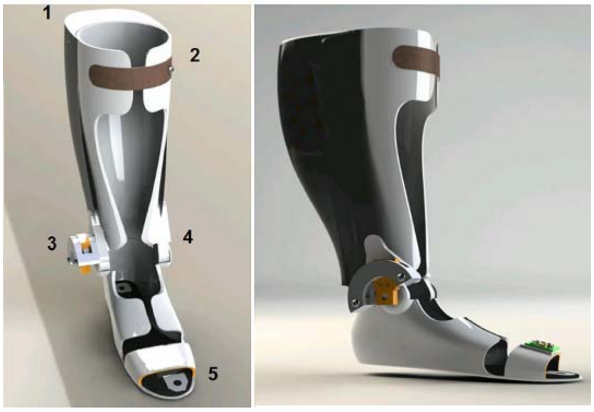


Figure 8: Portable active AFO proposed 1. Power supply and electric circuit; 2. Shank segment (it can be adjusted to different diameters); 3. Powered joint; 4. Potentiometer, 5. Foot segment and sensor sole.

Discussion

Each trajectory depends on the rehabilitation strategy recommended by the specialist in order to avoid the FD pattern. In both cases (Figures 6 and 7), the integral action of control positioned the foot near to the reference throughout the time of simulation.

From a biomechanical point of view, the angle variability during operation does not alter the gait advance, and the foot could be located on the neutral position. However, the progress of the foot in the pre-swing phase should be evaluated once the control system implemented on the prototype of active AFO is ready.

The orthosis proposed is only able to operate and maintenance the ankle joint angle on the neutral position during the swing phase. For the entire gait cycle other actuators with higher torque are necessary.

Conclusions

The results indicate the effectiveness of the tracking of the foot position for a trajectory based on the ankle angle. The control system considers the swing phase of gait in which there is no contact with the ground. For that reason, it is necessary to recognize the beginning and end of this phase. The control system here presented for an active AFO can be used in cases of FD movement assistance for possible applications in rehabilitation.

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