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Spasticity model based on muscular activations using genetic algorithm

S. Otmani^{a,b}, G. Michon^b and B. Watier^a

^aLAAS-CNRS, Université de Toulouse, CNRS, UPS, Toulouse, France; ^bUniversité de Toulouse, CNRS, ICA, ISAE-SUPAERO, Toulouse, France

1. Introduction

Cerebral palsy (CP) is a 'group of non-progressive, but often changing, motor impairment syndromes secondary to lesions or anomalies of the brain arising in the early stages of development' (Rosenbaum 2003). The severity of the CP can be quantified using Growth Motor Function Classification System (GMFCS). It is decomposed in 5 levels from the lowest severity to the highest. CP has different subtypes like spastic cerebral palsy which is the case studied in this paper. Spasticity is a condition in which there is an abnormal increase in muscle tone or muscles stiffness, which might interfere with movement, gait and speech and can also generate discomfort or pain. It can create pathological and involuntary reflexes (Lance 1990). Spasticity is described as a 'hypersensitive, velocity-dependent response to passive muscle stretch' (Fee and Foulds 2004). To better understand the spasticity effect on the mechanical behavior of the musculoskeletal system, the pendulum drop test was first developed by Wartenberg (1951). It was used then used in Fee and Foulds (2004) to compute and compare the knees angular displacement of three brothers with two of them with cerebral palsy. With the same idea, this study aims at analyzing the spasticity effect at the knee level and how an exoskeleton can restore normal joint behavior.

2. Methods

2.1 Subject information

To model the spasticity, a pendulum drop test was done on 9 years old child called C. who has CP with a GMFCS level of 2. During the experiment, anthropometric information of C. was collected: the height (123cm), body weight (22kg), shank length (28cm), shank mass (1.1kg), foot mass (0.46kg). The masses of the foot and the shank were computed using regression equations of Jensen (1986) for children of 9 years old. The foot is considered as a point mass at the end of the shank.

Table 1. Values to determine the model of both pendulum (activations and GA parameters).

Muscular activations parameters						
Flexor			Extensor			
Extension	Onsets	Delays	Onsets	Delays		
	60	600	159	559		
	680	300	736	323		
	1427	443	1059	599		
Flexion	Onsets	Delays	Onsets	Delays		
	323	267	0	907		
	1078	465	1015	590		
	1772	623	2002	436		
GA Parameters						
Extension						
K	K ₁	B _e	B ₁	KI _e	K2 _e	KI _r
3.44	2.35	0.08	0.06	1.62	1.49	1.83
K ₂	B _{r,1}	B _{r,2}	B _{r,3}	s _{o,1}	s _{o,2}	s _{o,3}
1.16	0.17	0.05	0.01	0.06	0.11	0.37
Flexion						
K	K ₁	B	B ₁	KI _e	K2	KI _r
3.79	1.96	0.08	0.09	1.94	1.71	0.6
K ₂	B _{r,1}	B _{r,2}	B _{r,3}	s _{o,1}	s _{o,2}	s _{o,3}
1.59	5.53e-7	0.009	0.097	0.06	4.57e-6	0.024

Onsets/delays in ms, Ks in N.m.rad⁻¹, Bs in N.m.s.rad⁻¹

2.2. Experiment devices

The EduExo system was used to perform the pendulum drop test. This device, firstly used for the upper limb, was divert to be applied on the child lower limb and is composed of thigh and calf 3D printed interfaces, angle and force sensors. To determine the muscular activations, two electromyograms (EMG) sensors were added and applied on the vastus lateralis and medial gastrocnemius following the SENIAM recommendations. Those muscles are agonist and antagonist muscles for the knee joint. EMG signals and knee's angular displacement were measured at a sampling frequency of 70Hz. Angular displacement values and EMG signals were filtered using a 4th order low pass Butterworth filter with a cutoff frequency of 10Hz.

2.3. Mechanical mode/ and GA

The model of the pendulum was done using mechanical equations and the parameters were determined using Genetic Algorithm (GA). Two initial conditions were considered: an extension case and a flexion case both with zero velocity in the initial condition and initial position of $\pm 25^\circ$.

The knee's angular displacements were computed using equations from (Fee and Foulds 2004) described in the following Eq. (1). The spasticity was modeled using a velocity dependent torque.

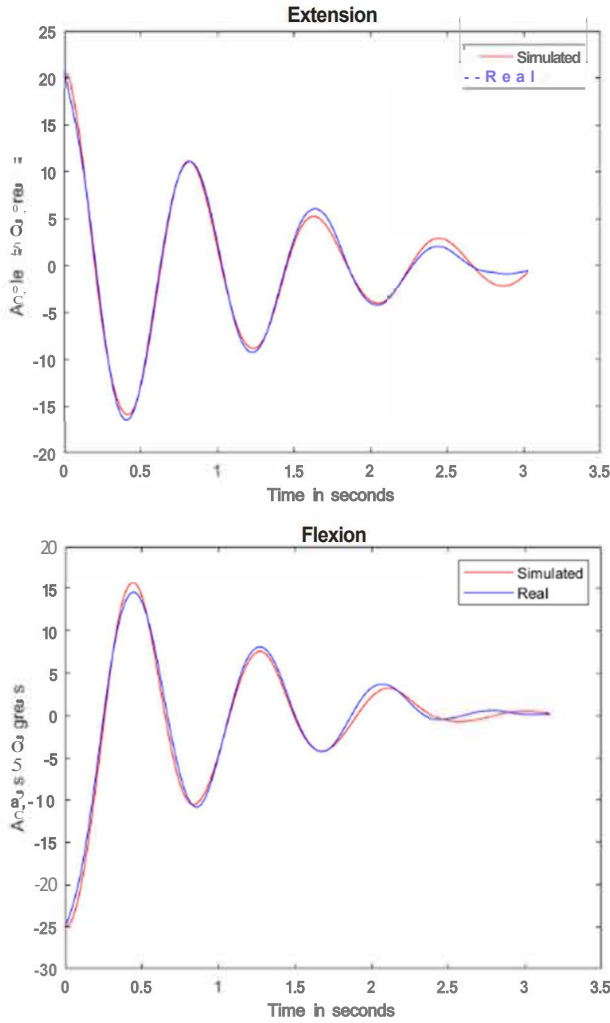


Figure 1. Experimental (real) and modeled (simulated) knee's angular displacement for initial condition in extension and in flexion.

$$I\ddot{\theta} + B_e \dot{\theta} + K_e \theta + T_e f + B^e f \dot{\theta} = mgL \sin(\theta) \quad (1)$$

with:

- $\theta, \dot{\theta}, \ddot{\theta}$: angle, velocity and acceleration.
- I : moment of inertia of the shank and foot
- B_e, B_f : damping coefficients
- K_e, K_f : stiffness coefficients
- T_e, T_f : nonlinear stiffness torques defined as $Kl_e f(e^{-K_f J^2} - 1)$
- m : mass of the shank + foot
- g : gravity acceleration
- L : distance from the joint to the shank's center of mass + distance from the joint to the foot's center of mass
- B^e, B^f : velocity feedback gains

I, m, g and L were fixed, as presented previously, using anthropometric table. The other constants of

the model were determined using a GA. It allows finding the best parameters which best fit the experimental data by minimizing an objective function (R^2 in this paper). To modelize the spasticity, a torque was defined by the velocity feedback gains multiplied by the velocity as spasticity is known to be velocity dependent. B^e and B^f are gains which are applied at a given time (onset) and during a certain amount of time (delay). The onsets and delays correspond to the moment where a muscular activation begins and its duration for extensors and flexors.

3. Results and discussion

Onsets and delays of the different activations extracted from experimental data for both extensor and flexor are presented in Table 1. Parameters obtained thanks to the GA were also reported in Table 1.

Using the parameters found by the GA and the onsets/delays of the activations described in Table 1 for each pendulum, simulated knee's angular displacements were obtained. Figure 1 present the comparison between experiment and model with $R^2 = 99.3\%$ (mean error: $-0.11^\circ \pm 1.31^\circ$) for extension and $R^2 = 99.1\%$ (mean error: $0.31^\circ \pm 0.66^\circ$) for flexion.

4. Conclusions

In this paper, we modeled the mechanical behavior of the knee of a child with CP in both flexion and extension. GA succeeds in simulating the knee's angular displacement in both cases by determining the physical constant of our model (Eq. (1)). Our results demonstrate the relevance of the method. Values obtained thanks to the GA. are very closed to the ones obtained in Fee and Foulds (2004). We also confirmed the mechanical differences between the flexion and the extension. Ongoing works will validate our method with other spastic children. This research models the knee as 1DoF joint with 2 antagonists muscles. It can, however, be enhanced adding new muscles with different onsets and delays. Additional work carried out on the healthy twin of C. without spasticity, not presented here, demonstrated the capacity of the model to correctly fit the behavior of healthy subjects. Ongoing works will focus on kinematic and dynamic data during gait of C. and her twin.

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