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PREDICTIVE SIMULATION OF THE SWING-THROUGH CRUTCH GAIT

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Abstract. *Predictive simulations have been employed in human gait analysis as a tool to elucidate and investigate neuro-musculoskeletal system coordination. However, there are few models available to apply this kind of simulation framework to the analysis of crutch gait, limiting studies in this field. These simulations could provide a broader comprehension of the gait with crutches by allowing non-intrusive observations of how the whole neuromusculoskeletal system coordinates the movement, which is important in the development and prescription of walking strategies, custom orthosis and crutches and clinical research focused on crutch users. The swing-through gait type is preferred by many crutch users because it provides greater agility and speed, including to some users with spinal cord injury that present certain levels of paraplegia. This work proposes a musculoskeletal model of the swing-through crutch gait and a simulation framework with potential use in studies of crutch gait of individuals with spinal cord injury, covering from partial to total impairment of the muscles of the hip.*

Keywords: *Biomechanics, Crutch gait, Gait, Simulation, Optimization*

1. INTRODUCTION

According to Rogers (2014) the crutch gait patterns are chosen based on type of the device, cause of the injury and personal choice. Individuals with SCI on low thoracic or lumbar vertebrae or other lower limb severe disability usually choose the swing-through pattern while other patterns are mode adequate for those who need the crutches for support and stability.

Paraplegics may become effective users of crutches and are encouraged to use them when possible as they provide great agility and an advantageous posture to overcome typical physical barriers (Rovick and Childress, 1988) Childs (1964). Furthermore, being able to move in an upright position and interact with others at equivalent height has been shown to provide physiological and psychological benefits to crutch users (Jaspers *et al.*, 1997). Nonetheless, when compared to wheelchair locomotion the crutch gait has higher metabolic energy consumption and is, therefore, only effective when used for covering shorter distances and by users with full capacity of the upper limbs.

Another factor contributing to the selection of a crutch gait type is the SCI level as it directly affects the muscles available for activation during the movement. Table 1 relates the SCIs at the lumbar and sacral levels with the muscles they impair.

Table 1: SCI injury level and the muscles impaired. SCI impairs the muscle at the level plus all the ones at lower levels.

Vertebra	Muscles
L2	Hip flexors (iliopsoas)
L3	Knee extensors (quadriceps)
L4	Ankle dorsiflexors (tibialis anterior)
L5	Toes extensors (extesor hallucis longus)
S1	Ankle platarflexors (gastronemius and soleus)

The use of dynamic optimization to generate predictive simulations of human movement increases as the computational capacity advances. It can be applied to the study of normal gait (Shourijeh *et al.*, 2017)(Ackermann and van den Bogert, 2010), jumping and sports in general (Zignoli *et al.*, 2017)(Porsa *et al.*, 2015), and neurological and orthopedic conditions (Handford and Srinivasan, 2018)(Cuerva *et al.*, 2016)(Bobbert *et al.*, 2016). These simulations allow, for example, the investigation of the effects of alterations in neural and musculoskeletal systems on gait patterns due to diseases, constituting a powerful tool for understanding cause and effect relationships between impairment and typical pattern changes in certain diseases. The gait with crutches was studied by several authors to determine joint loads (Sardini *et al.*, 2015)(Westerhoff *et al.*, 2012)(Haubert *et al.*, 2006), but few at the musculoskeletal level (Michaud *et al.*, 2017), since this type of study requires the solution of the muscle force redundancy problem. There are still few studies on crutch gait using dynamic or predictive simulations (Ackermann and Taissun, 2012)(Liu *et al.*, 2011), consequently the number of suitable models for this type of analysis is limited. In this scenario, this work proposes a musculoskeletal model of the swing-through crutch gait with potential for use in gait studies of individuals with spinal cord injury between (SCI) the T11 and L3 vertebrae, which can simulate conditions of total or partial impairment of the hip muscles.

The model herein proposed is featured with some of the muscle groups presented in Tab. 1 which can be disabled from the model to simulate SCI conditions at different levels.

2. METHODOLOGY

2.1 Skeletal System model

The proposed musculoskeletal model has five degrees of freedom in the sagittal plane and is valid for gait strategies with bilateral symmetry. It assumes the individual is using a Knee-Ankle-Foot Orthosis (KAFO) with restrained knee flexion (Rovick and Childress, 1988) (Childs, 1964). The model is composed of three rigid segments: trunk, legs with KAFO and arms with crutches, Fig. 1. The knee and shoulder joint kinematics are simplified by revolute joints.

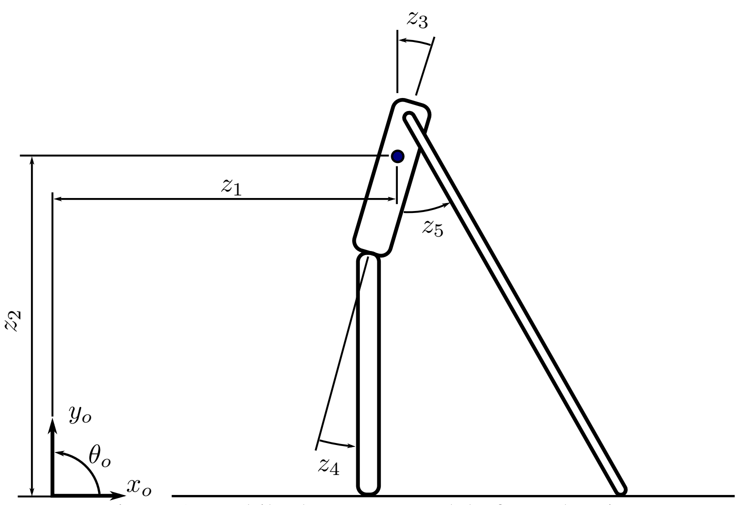


Figure 1: Multibody system model of crutch gait.

The anthropometric parameters where obtained from Rovick and Childress (1988), who implemented a similar model for forward dynamics simulations. The equations of motion were derived with the assistance of the package Symbs, an objected oriented library for automatic generation of equations of motion to test different formulations. Symbs is written in Octave (Eaton *et al.*, 2015) and is fully compatible with MATLAB (MATLAB and Symbolic Toolbox 2017a, The MathWorks, Inc, Natick, Massachusetts, United States). The configuration of the constrained multibody system is described by a minimal set of generalized coordinates $\boldsymbol{z} = [z_1 \;\; z_2 \;\; z_3 \;\; z_4 \;\; z_5]^T$, Fig. 1. The contact between the crutch tip and the ground and between the foot, considered a point at the leg's lower extremity, and the ground were modeled by nonlinear spring-damper elements as in van den Bogert *et al.* (2011), with slight changes to the parameters to facilitate convergence of the optimization process.

2.2 Model of the hip muscles

In the hip joint, four muscle groups were considered, *Illiopsoas*, *Glutei*, *Ischiotibial* and *Rectus Femoris*, all modeled as Hill-type muscles with a rigid series elastic element. This simplification eliminates de contraction dynamics, favoring the convergence of the Nonlinear Programming Problem associated to the optimal control problem (Millard *et al.*, 2013). The parameters of the muscle groups and muscle-skeleton couplings were extracted from van den Bogert *et al.* (2011). The activation dynamics is modeled using a first-order dynamics according to He *et al.* (1991).

2.3 Lumped model of the shoulder muscles

Given the complexity of the shoulder joint, current musculoskeletal models of the shoulder are three dimensional (Chadwick *et al.*, 2014) (Holzbaur *et al.*, 2005) with numerous muscles and degrees of freedom, which would be impracticable to apply in predictive simulation studies to date. Furthermore, the available musculoskeletal models usually cannot guarantee the continuity necessary to efficiently solve the large-scale Nonlinear Programming Problem associated with the Optimal Control Problem. To overcome this limitation, we use the 3D model DAS3 (Chadwick *et al.*, 2014) to adjust continuous, polynomial surface functions of the torque-angular position-angular velocity relationship for the shoulder flexors and extensors using the OpenSim program. This allows for indirectly accounting for the strength and the force-length-velocity relationships of all muscles spanning the shoulder joint in a lumped, continuous and computationally efficient joint-level model.

The DAS3 model contains seven segments (thorax, clavicle, scapula, humerus, ulna, radius and hand), 138 muscle elements and 11 degrees of freedom. For this application, we only considered the glenohumeral articulation in the local Z axis (GH_z) of the model as it has the most prominent contribution to the flexion/extension of the shoulder in the sagittal plane. While the rotation in the sagittal plane about the medio-lateral axis of the glenohumeral articulation was allowed, Fig.2, the other degrees of freedom were locked in the positions shown in Tab. 2.

Table 2: Locked positions of the DAS3 model.

Degree of freedom	Locked position
sterno-clavicular in Y SC_y	$-33.486°$
sterno-clavicular in Z SC_z	$5.002°$
sterno-clavicular in X SC_x	$32.914°$
acromio-clavicular in Y AC_y	$45.571°$
acromio-clavicular in Z AC_z	$0.458°$
acromio-clavicular in X AC_x	$-12.062°$
glenohumeral in Y GH_y	$43.864°$
glenohumeral YY GH_{yy}	$-34.659°$
elbow flexion-extension EL_x	$5.000°$
forearm pronation-supination PS_y	$5.000°$

In order to obtain the maximum torque-angular position-angular velocity relationship, a series of static optimizations was performed at constant angular velocities of the GH_z degree of freedom, from $-500\deg/s$ to $500\deg/s$ at steps of $10\deg/s$, forcing maximum activation of the agonist muscles with an external torque actuator resisting the motion. The same procedure was repeated for the antagonistic muscles with an external torque opposing the motion.

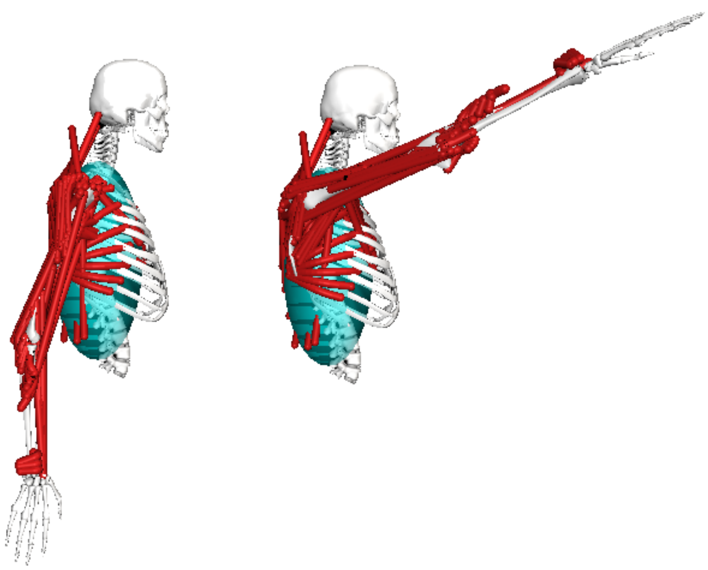


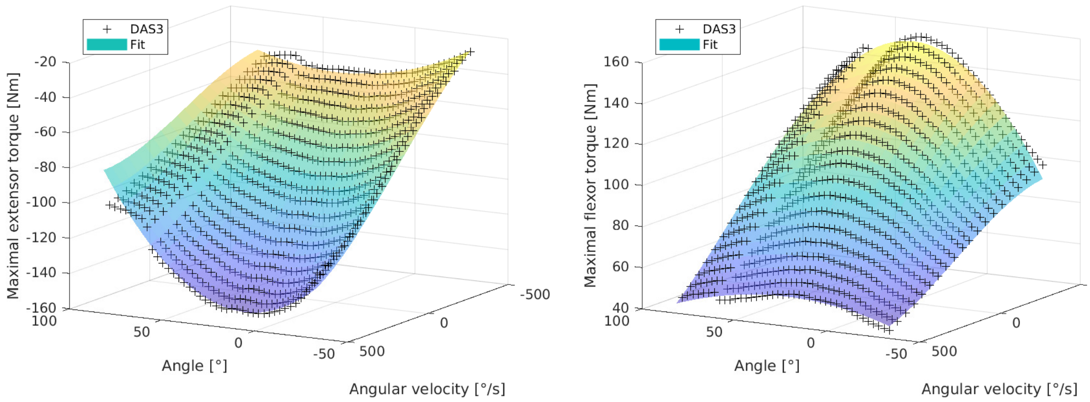
Figure 2: DAS3 simulation bounds of the shoulder extension/flexion. Adapted from Chadwick *et al.* (2014)

Continuous surfaces were fitted to the simulation results as shown in Fig. 3a and Fig. 3b for the shoulder extensors and flexors, respectively. The surfaces are defined by the general two dimensional polynomial in Eq. (1) whose coefficients are given in Tab. 3

$$\mathcal{T}(z_5, \dot{z}_5) = p_{00} + p_{10}z_5 + p_{01}\dot{z}_5 + p_{20}z_5^2 + p_{11}z_5\dot{z}_5 + p_{02}\dot{z}_5^2 + p_{21}z_5^2\dot{z}_5 + p_{12}z_5\dot{z}_5^2 + p_{03}\dot{z}_5^3 \quad (1)$$

Coefficient	Unit	Flexors	Extensors
p_{00}	Nm	9,12584e+01	-1,02188e+02
p_{10}	$Nm/(°)$	5,57984e-01	-2,86101e-01
p_{01}	$Nms/(°)$	-9,53369e-02	-1,28396e-01
p_{20}	$Nm/(°)^2$	-7,31144e-03	8,79501e-03
p_{11}	$Nms/(°)^2$	-8,62739e-04	1,09402e-04
p_{02}	$Nms^2/(°)^2$	-8,28730e-06	1,40830e-05
p_{21}	$Nms/(°)^3$	9,72236e-06	9,16030e-06
p_{12}	$Nms^2/(°)^3$	-1,30303e-07	9,481595e-09
p_{03}	$Nms^3/(°)^3$	5,49080e-08	7,315675e-08

Table 3: Coefficients of the maximum torque-angular position-angular velocity relationship. Authors.



(a) Maximal shoulder extensors torque. (b) Maximal shoulder flexors torque.

Figure 3: Surface models for maximal shoulder torque.

Therefore, the total shoulder torque is computed in Eq. (2)

$$\mathcal{T}_{sh} = a_{flex}\mathcal{T}_{flex}(z_5, \dot{z}_5) + a_{ext}\mathcal{T}_{ext}(z_5, \dot{z}_5) \tag{2}$$

Where $\mathcal{T}_{flex}$ and $\mathcal{T}_{ext}$ are the maximal torque for the flexors and extensors actuators, respectively, a_{flex} and a_{ext} are their activation levels, raging from 0 to 1. The activation dynamics is modeled according to Eq. 3 (He *et al.*, 1991).

$$\dot{a} = (u - a)\left(\frac{u}{\tau_a} + \frac{1-u}{\tau_d}\right) \tag{3}$$

Where u is the neural excitation, τ_a and τ_d are the activation and deactivation time constants.

2.4 Optimal control problem

An optimal control approach was used to predict the crutch gait patterns, which consists of searching for the states $\boldsymbol{x}(t)$ and controls $\boldsymbol{u}(t)$ so as to

$$\begin{aligned}
&\text{minimize} \quad \boldsymbol{J} = \frac{1}{t_f}\int_0^{t_f} \sum a^p dt \\
&\text{subject to} \\
&\qquad \dot{\boldsymbol{x}} = f(\boldsymbol{x}(t), \boldsymbol{u}(t), t) \\
&\qquad \boldsymbol{h}_L \leq \boldsymbol{h}(\boldsymbol{x}(t), \boldsymbol{u}(t), t) \leq \boldsymbol{h}_U \\
&\qquad \boldsymbol{\epsilon}_L \leq \boldsymbol{\epsilon}(\boldsymbol{x}_0(t), \boldsymbol{u}_0(t), t_0, \boldsymbol{x}_f(t), \boldsymbol{u}_f(t), t_f) \leq \boldsymbol{\epsilon}_U \\
&\qquad \boldsymbol{\psi}(\boldsymbol{x}_0(t), \boldsymbol{u}_0(t), t_0, \boldsymbol{x}_f(t), \boldsymbol{u}_f(t), t_f) = 0 \\
&\qquad \boldsymbol{x}_L \leq \boldsymbol{x} \leq \boldsymbol{x}_U \\
&\qquad \boldsymbol{u}_L \leq \boldsymbol{u} \leq \boldsymbol{u}_U \\
&\qquad t_f - t_0 \geq 0
\end{aligned} \tag{4}$$

In Eq. (4) the differential constraints is a set of equations which includes the state space representation of the equations of motion and the first order activation dynamics, Eq. (3), of the six muscles modeled. Thus, the state vector $\boldsymbol{x}$ is formed by the generalized coordinates, generalized velocities and activation, the controls in $\boldsymbol{u}$ are the six neural excitation of each muscle, $\boldsymbol{J}$ is an effort- or fatigue-based cost function depending on the choice of the exponent p, $\boldsymbol{h}$ imposes vertical position constraints to the crutch tip and foot to characterize the crutch gait phases, $\boldsymbol{\epsilon}$ imposes crutch strike initial conditions, $\boldsymbol{\psi}$ are the mean velocity, periodicity and phase continuity conditions. The optimal control problem was assembled using the package PSOPT (Becerra, 2010), making use of its trapezoidal collocation method, and solved with IPOPT (Wächter, 2002) using numerical differentiation for the derivatives of the constraints and cost function.

3. RESULTS

For a mesh with 100 nodes, the model converged to a feasible solution with objective function valued 4.055. The solution was obtained in an Intel i5-4690K 3.5GHz using a single processor in 71 minutes. Two models with SCI at different levels were simulate. The first model, results in Fig. 4, represents a person with SCI bellow L3 who is capable of full activation of the hip muscles (model$_1$). The second model, results in Fig. 5, simulates a person with SCI above L1 with full capacity on the upper limbs but no control of the hip muscles (model$_2$).

Figure 6 compares the models generalized coordinates. In model$_1$ the crutch stance phase correspond to 61% of the cycle while in model$_2$ it is 53% (vertical dashed line). At approximately 50% of the cycle, model$_1$ predicts a shoulder extension bellow $-50\,\mathrm{deg}$ with trunk leaning forward. This over extension was not observed in the works of Santos (2017) and Rovick and Childress (1988). This magnitude of extension is not observed in model$_2$, which takes more advantage of the pendulum movement by flexing the shoulder and elevating the trunk center of gravity at the beginning of the cycle thus releasing the accumulated potential into forward movement.

Activations are shown in Fig. 7. It is possible to observe that for model$_1$ the hip extensors (*Glutei* and *Ischiotibial*) are little activated in the swing phase of the crutch while the flexors (*Iliopsoas* and *Rectus Femoris*) are activated at the beginning of the cycle to propel the trunk and the legs through the crutch. For model$_2$ the model does not present any neural activation in the hip muscles.

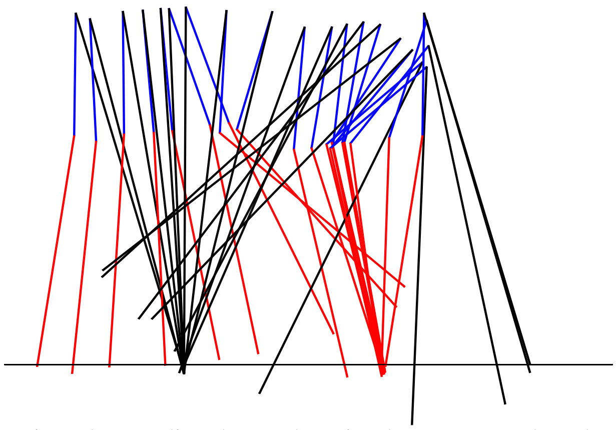


Figure 4: Stick figure representing the predicted crutch gait along a complete locomotion cycle for a low SCI person(model$_1$). Author

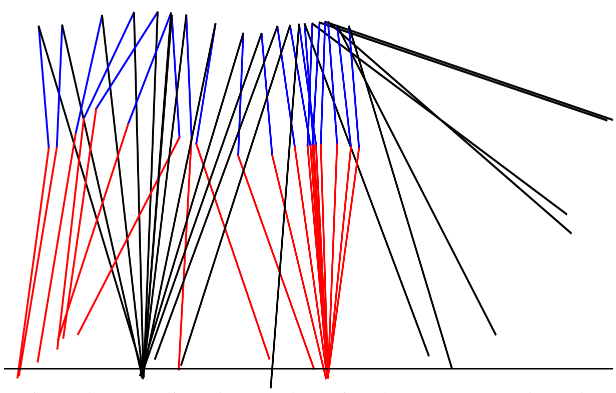


Figure 5: Stick figure representing the predicted crutch gait along a complete locomotion cycle for a high SCI person(model$_2$). Author

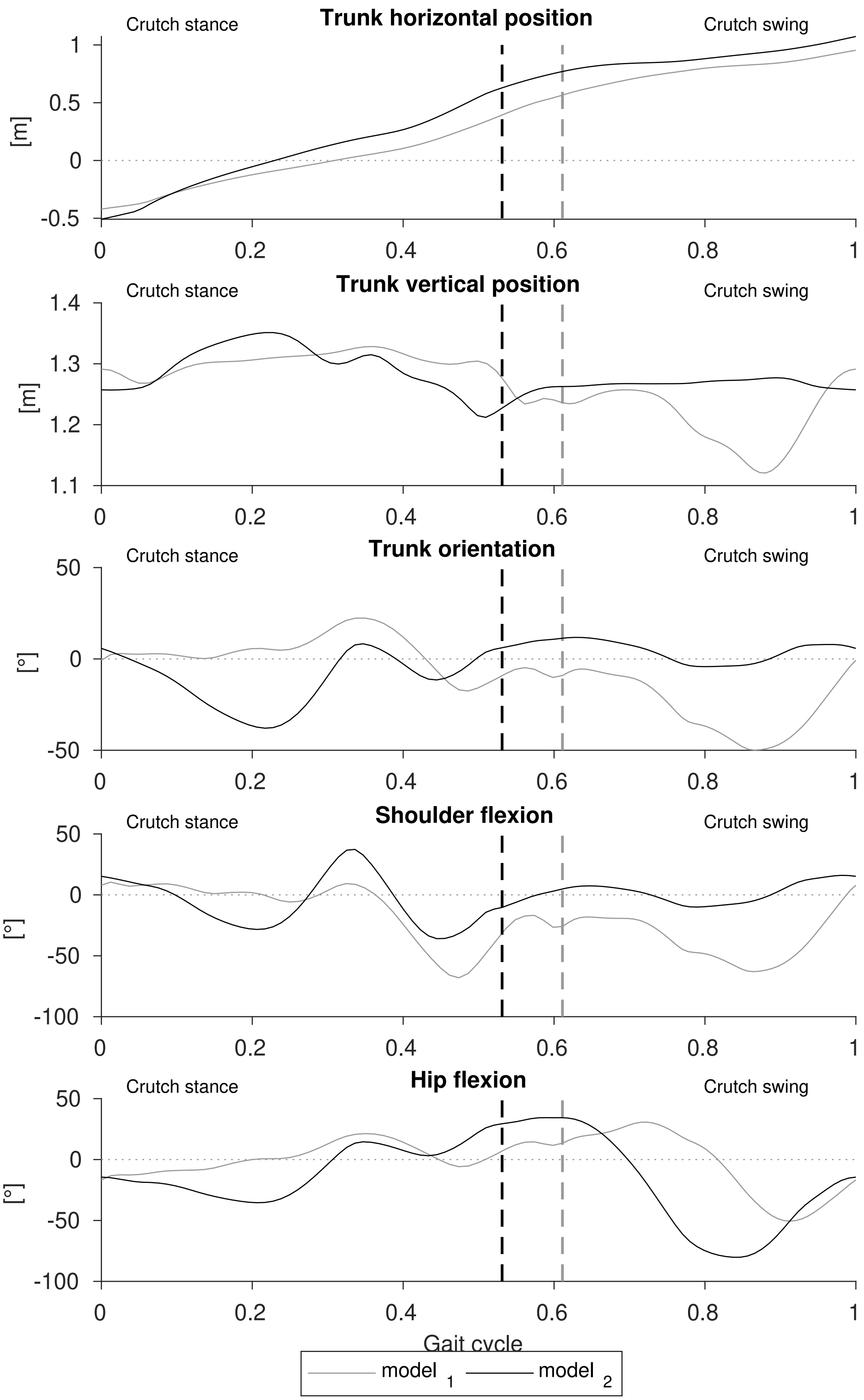


Figure 6: Predicted generalized coordinate trajectories along a complete gait cycle for the two considered models. Author.

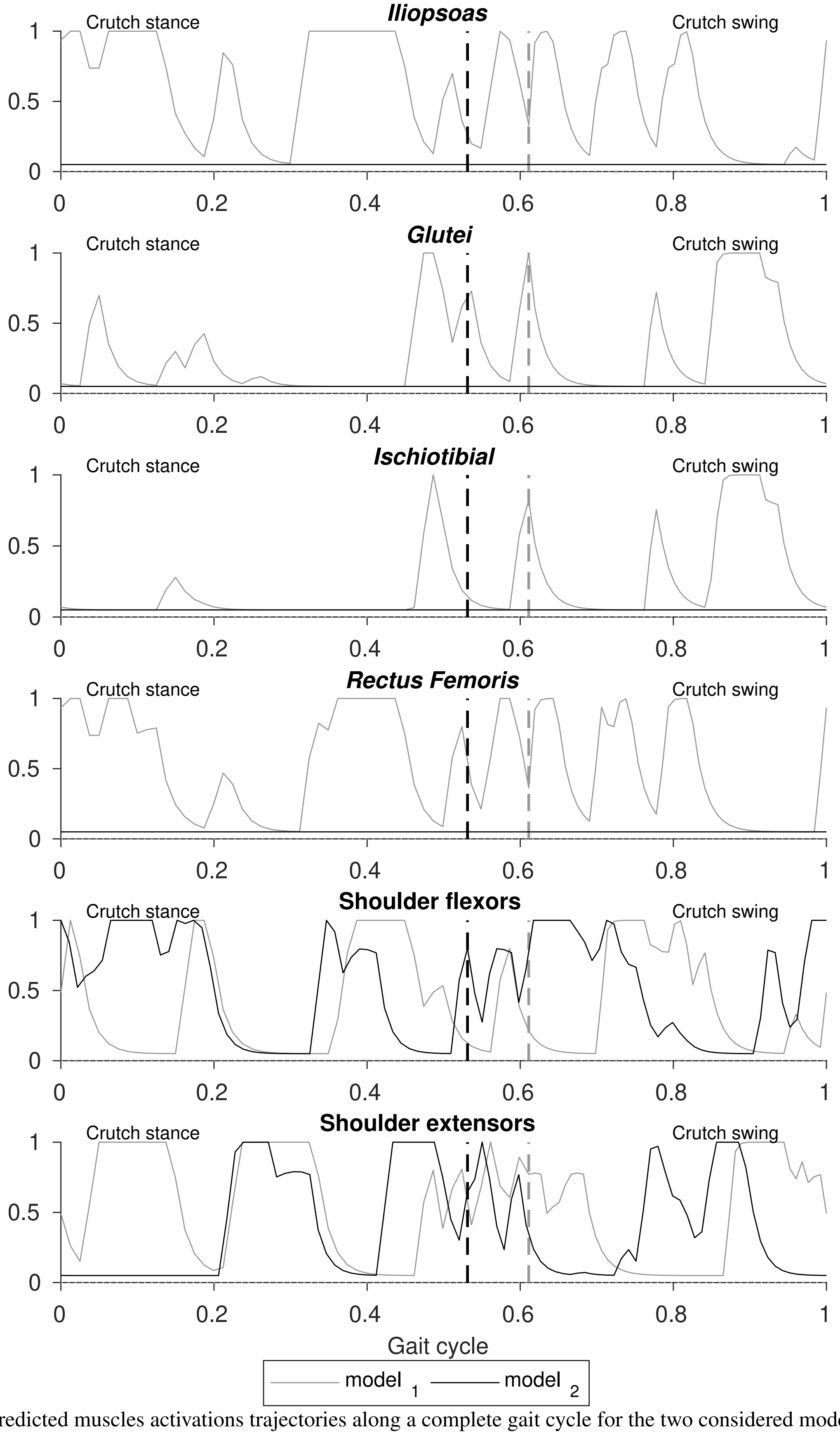


Figure 7: Predicted muscles activations trajectories along a complete gait cycle for the two considered models. Author.

4. CONCLUSIONS

The model and computational approach are able to predict symmetric swing-through crutch gait patterns that satisfy the imposed constraints and can be used to gain insights into the coordination process during locomotion with crutches. However, we observed that the patterns found deviate to some extent from data reported in the literature. The model should, therefore, be improved to properly match expected trajectories.

The automatic generation of motion equations is efficient in simplifying the assembly process of multibody systems and provides full access to movement equations, in symbolic or numerical manner, in different formulations, which is beneficial to optimal control approaches once it allows the application of efficient techniques not explored in this article such as automatic differentiation which can contribute to improve the OCP solution process. The shoulder joint musculoskeletal model is useful from a computational and physiological point of view because it eliminates the muscular redundancy problem in this joint presenting amplitude and speed compatible with the expected normal functioning.

As future work, we propose to include a foot segment and the ankle joint, which should lead to a more realistic model. The evaluation of different contact models is also proposed to reduce the observed difficulties in the converge of the OCP .

5. ACKNOWLEDGEMENTS

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