Optimization methods for identifying muscle forces in spinal cord injury subject during crutch gait

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Abstract

Some patients with spinal cord injury (SCI) can walk with the help of orthoses and crutches. However, this assisted gait often causes the patient to suffer from pain in the active part of his body, as for example in the upper extremities. Although many musculoskeletal models have been proposed to estimate muscle forces during the gait of healthy subjects, the same cannot be said for SCI subjects. Therefore, the purpose of this work is to propose a full body musculoskeletal model of a SCI subject for assisted gait and to validate it. Both the kinematics and the ground reactions were experimentally obtained from a female subject with SCI at T11 and applied to a three-dimensional human model featuring 53 muscles in the right leg, trunk and right arm. An inverse dynamic analysis provided the histories of the joint drive torques. Then, four static and one physiological criteria were used to estimate the muscle forces at hip, trunk, shoulder and elbow levels (since the subject did not possess muscular activity under hip level). Finally, the electromyographic signals from 15 out of the 53 modeled muscles were recorded and compared with the corresponding estimated histories.

Keywords: biomechanics, assisted gait, SCI subject, muscle forces, muscle recruitment criteria

1. Introduction

Advances in the care of individuals with spinal cord injury (SCI) have resulted in an increased life expectancy in this population. As an individual with SCI ages, an increase in the prevalence of problems like pain in the upper extremity might be expected. The structures of the upper extremity are designed primarily for prehensile activities. Because SCI patients also need them for daily functions such as mobility, they are used more frequently and strenuously and subject to increased stresses compared to those of an able-bodied individual [1]. Moreover, recent advances in passive and active orthoses allow people with neuromuscular disorders to move with crutches instead of wheelchairs. In the same way as wheelchair activities, walking with crutches produces important joints loads at the upper extremities: Westerhoff [2] reported maximum loads of up 170% of the bodyweight at shoulders.

Crutch-assisted gait was studied at skeletal level by various authors to determine joint loads [2-4], but not at musculoskeletal level, like Morrow did for wheelchair activities [5]. Additionally, people with SCI or other neuromuscular diseases, have a different gait pattern which requires a specific analysis [6-7].

Determination of muscle forces during gait (or any other exercise) is of great interest to extract the principles of the central nervous system (CNS) control [8] (assessment of pathological gait from muscular activation abnormalities, diagnosis of neuromuscular disorders), or to estimate the loads on bones and joints [9] (prevention of injuries in sports, surgical planning to reconstruct diseased joints). The invasive character of in vivo experimental measurements, and the uncertain relation between muscle force and electromyography (EMG), makes computer modeling and simulation a useful substitutive approach [10].

The fundamental problem is that there are more muscles serving each degree of freedom of the system than those strictly necessary from the mechanical point of view, which implies that, in principle, an infinite number of recruitment patterns are acceptable. This problem is often referred to as the redundancy problem of the muscle recruitment [11] or the force-sharing problem [12]. Experimental studies [13] and EMG collections [14] suggest that a specific strategy of muscle coordination is chosen by the CNS to perform a given motor task.

A popular mathematical approach for solving the muscle recruitment problem is the optimization method, which can be associated to inverse or forward dynamics [15]. These methods minimize or maximize some criterion (objective function or cost function) which reflects the mechanism used by the CNS to recruit muscles for the movement considered. The proper cost function is not known a priori, so the adequacy of the chosen function must be validated according to the obtained results [16]. Many criteria have been proposed in the literature to predict muscle forces.

In previous papers, the authors presented a comparison among four muscle recruitment criteria working on a static optimization scheme, and an additional criterion applied within a physiological optimization approach based either on an inverse dynamic analysis of the acquired motion [17] or on a forward dynamic one [18]. However, all methods were applied to healthy subjects and normal gait, while, in this work, they are applied to the assisted gait of an adult SCI female. The objective is to propose a musculoskeletal model of the SCI subject walking with the help of orthoses and crutches that provides an estimation of muscular forces and to validate it.

2. Experiment and model

2.1. Subject

The subject was an adult female of age 45, mass 65 kg and height 1.52 m, with SCI at T11. Her injury allowed her a normal motion of the upper extremities and trunk, while partially limiting the actuation at the hips due to partial or no muscular innervation. Therefore, in order to walk she required the assistance of a pair of passive knee-ankle-foot orthoses (KAFO) and two crutches. In daily life she mainly used a wheelchair to move and resorted to the mentioned assisted gait only occasionally and during short periods of time. In order to assess muscle activity at hip level, surface EMG was used (equipment to measure deep muscles was not available). Based on the results, it was considered that pelvic and biceps femoris muscles were active, while other muscles going down the leg were inactive.

2.2. Instrumentation and data collection

The subject walked over two embedded force plates (AMTI, AccuGait sampling at 100 Hz) with the help of two crutches instrumented for ground contact force measurement [19], as shown in Figure 1a, and completed the 4-point crutch-assisted gait cycle shown in Figure 2. Her motion was captured by 12 optical infrared cameras (Natural Point, OptiTrack FLEX:V100 also sampling at 100 Hz) that computed the position of 43 optical markers. Position data were filtered using an algorithm based on Singular Spectrum Analysis (SSA). Additionally, 15 surface EMG muscle signals were recorded at 1 KHz: 4 at the right hip, 4 at trunk, 4 at the right shoulder and 3 at the right arm; they were normalized and then filtered by SSA with a window length of 250.

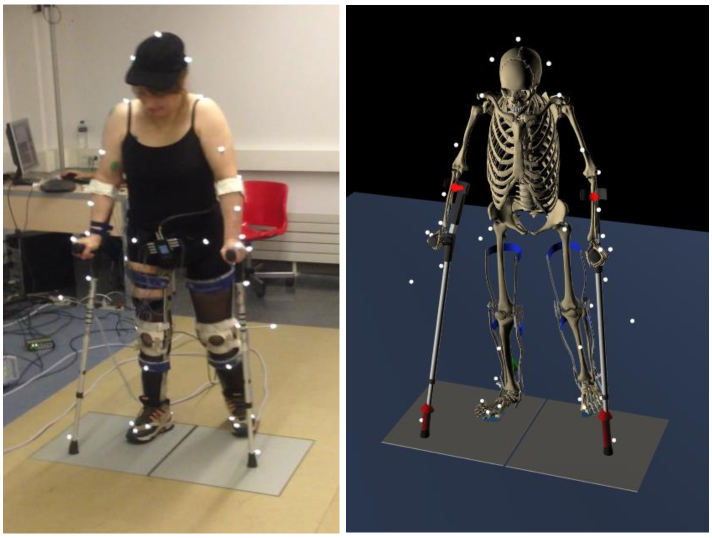

Figure 1: Gait of spinal cord injured subject assisted by passive orthoses and crutches:
a) acquired motion; b) computational model.

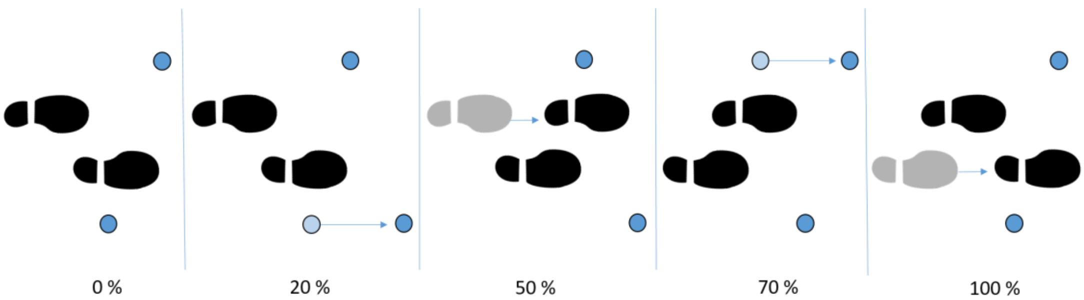

Figure 2: 4-point crutch-assisted gait cycle.

2.3. Model description

The 3D human model (Figure 3a) consisted of 18 anatomical segments: pelvis, torso, neck, head, and two hind feet, forefeet, shanks, thighs, arms, forearms and hands, with the crutches rigidly connected to the hands and the orthoses embedded in the corresponding body links (thighs, claves and feet). The segments were linked by ideal spherical joints, thus defining a model with 57 degrees of freedom (6 of the base body plus 17x3 of the joints). The geometric and inertial parameters of the model were obtained, for the lower limbs, by applying correlation equations from a reduced set of measurements taken on the subject, following the procedures described in [20]. For the upper part of the body, data from standard tables [15] was scaled according to the mass and height of the subject. In order to adjust the total mass of the subject, a second scaling was applied to the inertial parameters of the upper part of the body. Assistive devices were taken into account by altering the inertia properties of hands (crutches) and thighs, calves and feet (orthoses).

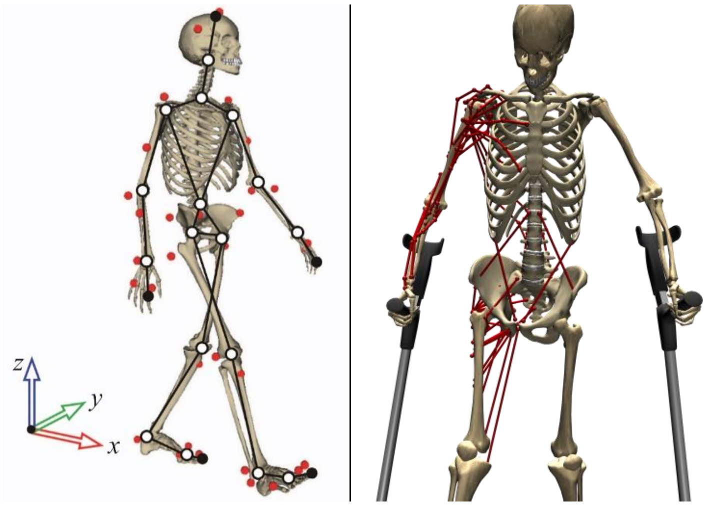

Figure 3: a) Human multibody model; b) musculoskeletal model details.

Inverse dynamic analysis [21] was applied to obtain the ground reactions and joint drive torques along the motion. Measurements from the force plates and instrumented crutches were just used to overcome the indeterminacy in the distribution of ground reactions during the multiple-support phases [19]. Therefore, the obtained joint drive torques and external reactions were consistent with the corresponding motion and no residuals were generated.

The musculoskeletal model (Figure 3b) was composed of 53 muscles: 21 muscles at the right hip, 6 at trunk, 15 at the right shoulder and 11 at the right elbow, their properties taken from OpenSim [22].

3. Optimization methods

3.1. Static optimization

The first approach considered was static optimization. Joint drive torques at the right hip, shoulder and elbow, and at the lumbar joint, should be reproduced by muscle forces. The following optimization problem was stated,

$$
\begin{aligned}
&\min_{\mathbf{F}} \; C(\mathbf{F}) \\
&\text{subject to} \quad \mathbf{J}^{\mathrm{T}}\mathbf{F} = \mathbf{Q} \\
&\qquad\qquad\quad 0 \le F_i \le F_{i,0} \quad i = 1, 2, \ldots, m
\end{aligned}
\tag{1}
$$

where C is the cost function, $\mathbf{Q}$ is the vector of joint drive torques at the mentioned joints (where the force-sharing problem is addressed), $\mathbf{F}$ is the vector of muscle forces, $\mathbf{J}$ is the Jacobian whose transpose projects the muscle forces into the joint drive torques space, and $F_{i,0}$ is the maximum isometric force of muscle i, with m the number of muscles (in this case, m=53).

Regarding the cost function C, four cases were considered, shown in Table 1:
I) Sum of the squares of muscle forces.
II) Sum of the squares of normalized muscle forces.
III) Sum of muscle stresses, with A_i the cross sectional area of muscle i.
IV) Largest normalized muscle force.

Table 1: The four muscle recruitment criteria compared.

I	II	III	IV
$\sum_{i=1}^{m} F_i^2$	$\sum_{i=1}^{m} \left(\frac{F_i}{F_{i,0}} \right)^2$	$\sum_{i=1}^{m} \left(\frac{F_i}{A_i} \right)^2$	$\max \left\{ \frac{F_i}{F_{i,0}} \right\} \quad i = 1, 2, \ldots, m$

Before showing the results, there is an issue which deserves to be mentioned. It has been said when describing the human model that spherical kinematic pairs have been considered for all the joints. This means that three joint drive torques are obtained at each joint from the inverse dynamic analysis. However, not all of them are due to the actuation of muscles. For example, the abduction/adduction torque at the elbow is not provided by muscles, but rather by other joint structures as trochlea and ligaments, being more a reaction moment than a drive torque. Therefore, the following joint drive torques were selected to form vector **Q**: the three torque components at the hip, lumbar joint and shoulder, and the flexion/extension torque at the elbow. A discussion on how the modeling of the joints and the considered torques affect to the results can be found in [23].

3.2. Physiological optimization

In constrast to the static optimization, the so-called physiological static optimization takes muscle dynamics into account by introducing dynamic muscle force constraints [24]. This method applies static optimization techniques at each time point but prescribes minimal and maximal constraints for the muscle forces by extrapolating the force values from the previous time point using feasible muscle activation values. In this way, the optimization process remains efficient, but muscle dynamics are considered. The following optimization problem was stated,

$$
\begin{aligned}
&\min_{\mathbf{F}} \sum_{i=1}^{m} \left(\frac{F_i}{F_{i,\max}} \right)^2 \\
&\text{subject to} \quad \mathbf{J}^{\mathrm{T}} \mathbf{F} = \mathbf{Q} \\
&\qquad\qquad\quad F_{i,\min} \le F_i \le F_{i,\max} \quad i = 1, 2, \ldots, m
\end{aligned}
\tag{2}
$$

where all the symbols have the same meaning as in (1), and $F_{i,\min}$, $F_{i,\max}$ are, respectively, the minimum and maximum admissible forces for muscle *i* at the corresponding time point. In what follows, the way to determine such force limits is explained.

If the Hill's muscle model is used [25], the states of muscle *i* are denoted by (index *i* is dropped for simplicity),

$$
\mathbf{x} = \begin{Bmatrix} a \\ F \end{Bmatrix}
\tag{3}
$$

where *a* is the muscle activation and *F* is the musculotendon force. The Hill's muscle first-order differential equations are,

$$
\dot{\mathbf{x}} = \begin{Bmatrix} \dot{a} \\ \dot{F} \end{Bmatrix} = \begin{Bmatrix} f_1(a,u) \\ f_2(a,F,l,v) \end{Bmatrix} = \mathbf{f}(\mathbf{x},u,l,v)
\tag{4}
$$

being *u* the muscle excitation, *l* the musculotendon length and *v* the musculotendon velocity.

If the states are given at a certain time *t*, the minimum and maximum state variables at time $t + \Delta t$ can be computed by setting the neural input *u* to its minimum ($u = 0$) and maximum ($u = 1$) possible values during the time interval *Δt*, and integrating the muscle equations,

$$
\begin{aligned}
\mathbf{x}_{\min}(t+\Delta t) &= \mathbf{x}(t) + \int_{t}^{t+\Delta t} \mathbf{f}(\mathbf{x}, u=0, l, v)\,dt \\
\mathbf{x}_{\max}(t+\Delta t) &= \mathbf{x}(t) + \int_{t}^{t+\Delta t} \mathbf{f}(\mathbf{x}, u=1, l, v)\,dt
\end{aligned}
\tag{5}
$$

The two integrations in (5) were performed by using numerical integrator *ode23* from Matlab. Values of *l* and *v* inside the time interval *Δt* were obtained by linear interpolation of their values at times *t* and $t + \Delta t$. The solution of (5) provided the limits $F_{i,\min}$, $F_{i,\max}$ for muscle *i*. This process was repeated for all the muscles.

It must be noted that the lowest activation at $t + \Delta t$ is not always obtained for $u = 0$. In the long term, the activation converges to the excitation value if the latter remains constant. However, for small *Δt* values, an excitation higher than 0 can lead to a lower activation at $t + \Delta t$. Therefore, the $F_{i,\min}$ used for the optimization is not always guaranteed to be the smallest possible, but the error remains under 2.5% of the maximum activation.

Once the force limits for all the muscles were determined, the optimization problem (2) could be solved, thus yielding the muscle forces F_i, $i = 1, 2, ..., m$ for time $t + \Delta t$. At this point, an iteration process for each muscle was run in order to find out the (assumed constant) excitation value u during the time interval Δt that led to the obtained muscle force F_i at time $t + \Delta t$. To that end, different values of u (index i is dropped again) were tried until the bottom part (that affecting the force; see (3)) of the following equation was satisfied,

$$\mathbf{x}(t+\Delta t) - \mathbf{x}(t) - \int_t^{t+\Delta t} \mathbf{f}(\mathbf{x}, u, l, v)dt = 0 \tag{6}$$

Function *fsolve* from Matlab was used for the iteration process, starting with initial guess u=1. The bottom part of (6) was integrated for each value of u provided by *fsolve*, until the resulting muscle force fell within a certain tolerance of the force obtained in the optimization (2). The companion muscle activation was then obtained from the upper part of (6), being the activation at time $t + \Delta t$.

So far, it had been assumed that the muscle states were known at time t in order to move to time $t + \Delta t$. Therefore, a particular procedure had to be followed for the initial conditions, i.e. at time $t = 0$. For that time, it was supposed that muscle velocity was zero, $v_{\mathrm{M}} = 0$, for all the muscles, which implied that the force-velocity relationship of the Hill's muscle model was equal to one, $f_v(v_{\mathrm{M}} = 0) = 1$. To determine the initial muscle forces, the optimization problem (2) had to be solved, being the force limits $F_{i,\min}$ and $F_{i,\max}$ the ones obtained by considering the minimum and maximum muscle activations, respectively, $a = 0$, $a = 1$. Since the velocity term is equal to 1, the muscle force can be obtained for a given activation by solving a nonlinear equation, instead of requiring the integration of an ODE. A more detailed explanation of this method can be found in [26].

3.3. Results

Comparison of the estimated muscle forces obtained through the different described criteria (four static and one physiological) for some representative muscles of each joint is shown in Figure 4. As it can be seen, no significant differences were observed among the excitation patterns obtained with the five compared methods.

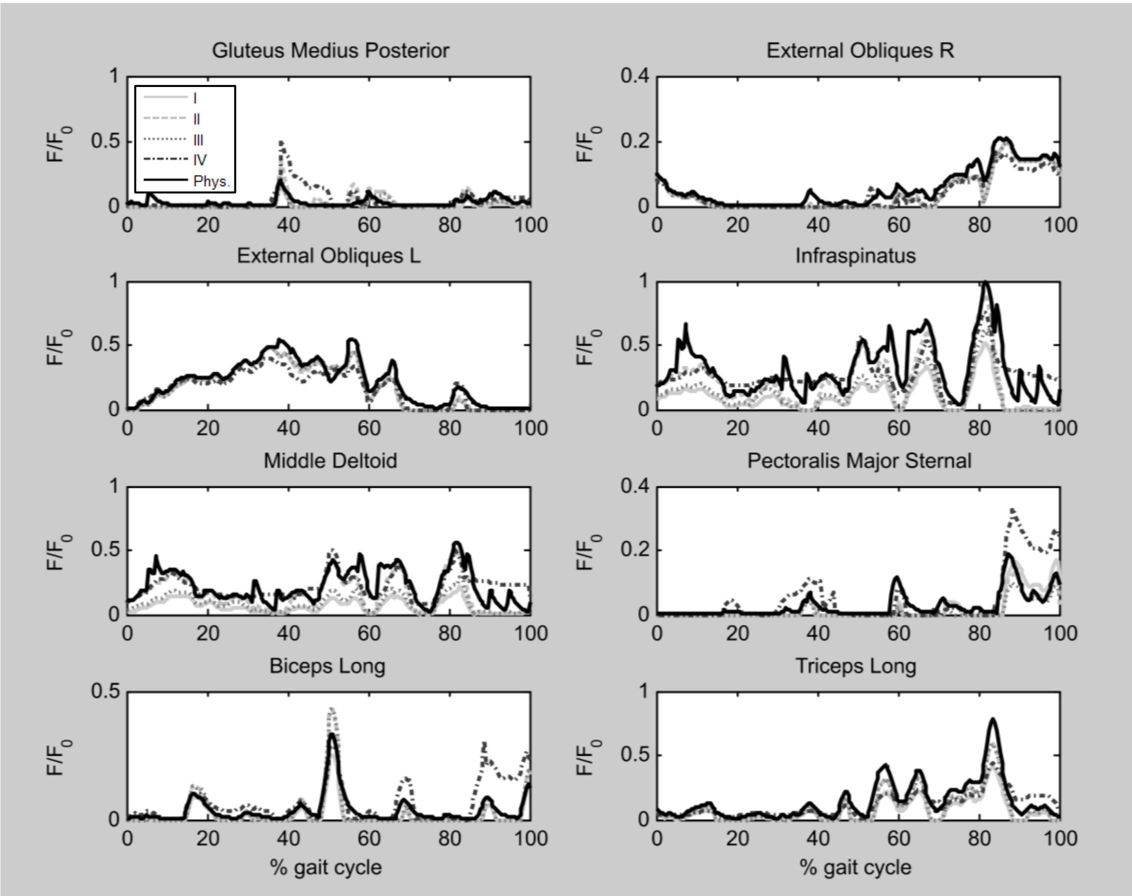

Figure 4: Comparison of estimated muscle forces obtained with different criteria.

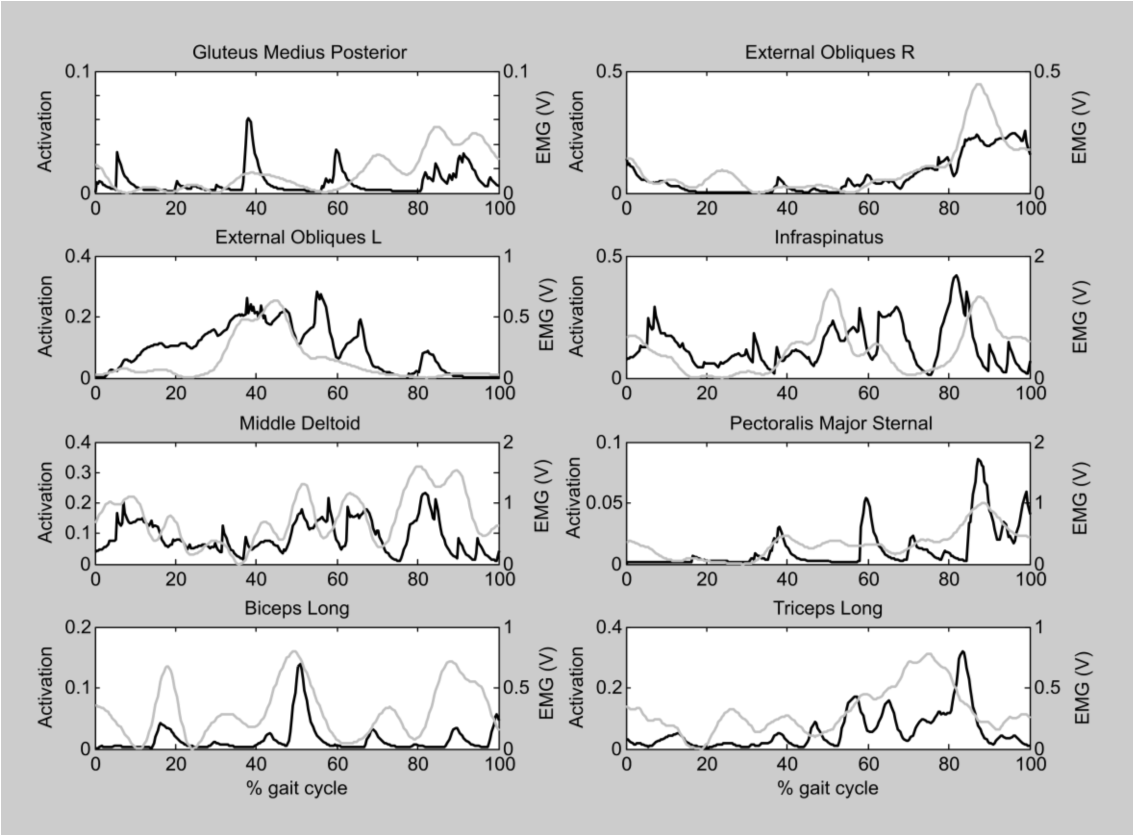

Figure 5: Muscle activations obtained from physiological optimization (black) and EMG measurements (grey).

Furthermore, the activation obtained with the physiological approach is compared with EMG measurements in Figure 5. Knowing the uncertain relation between muscle force and EMG, especially the difficulties in scaling the EMG magnitude of the filtered and normalized signals, it was decided not to scale the EMG processed data in order to focus more on the activity and coordination of muscles. Results show a reasonable correlation between calculated and experimental data.

Unlike the normal gait of healthy people, which is smooth, crutch gait is noisy due to the numerous phases of the cycle (Figure 2) and their corresponding load distributions. On the other hand, muscular activity is higher in the upper extremities, as almost no phase of rest is observed. This results explain why the subject cannot walk for a long time. During the experiments, the patient complained of arm pain after few gait cycles and needed to rest many times. The partial actuation at hip level could be the reason for this permanent effort of the upper extremities, required to keep stability.

4. Conclusions

The acquired gait of an adult female with spinal cord injury assisted by a pair of passive knee-ankle-foot orthoses and two crutches was analyzed through a three-dimensional personalized human model featuring 53 muscles in the lumbar joint and the right hip, shoulder and elbow. Both static optimization with four muscle force-sharing criteria and static-physiological optimization were applied to estimate the histories of muscle forces. All the methods compared showed similar results. Furthermore, comparison of the activations provided by the physiological approach with EMG measurements was established for several muscles, and a reasonable correlation was obtained. The validated model could be helpful in the design of assistive gait devices in the context a predictive framework.

It should be remarked that this model has to be adapted to each subject, especially the lower extremity muscular modeling. Indeed, another spinal cord injured subject will not necessary have the same muscle limitations. Furthermore, muscle activity will strongly depend on the subject’s general physical condition.

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References

[1] M. Dalyan, D. Cardenas, and B. Gerard, “Upper extremity pain after spinal cord injury.,” *Spinal Cord*, vol. 37, no. 3, pp. 191–5, 1999.

[2] P. Westerhoff *et al.*, “In vivo measurement of shoulder joint loads during walking with crutches,” *Clin. Biomech.*, vol. 27, no. 7, pp. 711–718, 2012.

[3] B. A. Slavens, P. F. Sturm, and G. F. Harris, “Upper extremity inverse dynamics model for crutch-assisted gait assessment,” *J. Biomech.*, vol. 43, no. 10, pp. 2026–2031, 2010.

[4] B. A. Slavens, N. Bhagchandani, M. Wang, P. A. Smith, and G. F. Harris, “An upper extremity inverse dynamics model for pediatric Lofstrand crutch-assisted gait,” *J. Biomech.*, vol. 44, no. 11, pp. 2162–2167, 2011.

[5] M. M. B. Morrow, K. R. Kaufman, and K. N. An, “Shoulder model validation and joint contact forces during wheelchair activities,” *J. Biomech.*, vol. 43, no. 13, pp. 2487–2492, 2010.

[6] J. L. Hicks, M. H. Schwartz, A. S. Arnold, and S. L. Delp, “Crouched postures reduce the capacity of muscles to extend the hip and knee during the single-limb stance phase of gait,” *J. Biomech.*, vol. 41, no. 5, pp. 960–967, 2008.

[7] K. M. Steele, A. Seth, J. L. Hicks, M. S. Schwartz, and S. L. Delp, “Muscle contributions to support and progression during single-limb stance in crouch gait,” *J. Biomech.*, vol. 43, no. 11, pp. 2099–2105, 2010.

[8] M.R. Pierrynowski and J. B. Morrison, “A physiological model for the evaluation of muscular forces in human locomotion: theoretical aspects,” *Math. Biosci.*, vol. 75, pp. 69–101, 1985.

[9] D. E. Hardt, “Determining muscle forces in the leg during normal human walking - An application and evaluation of optimization methods,” *Journal of Biomechanical Engineering*, vol. 100. pp. 72–78, 1978.

[10] A. Nagano, T. Komura, S. Fukashiro, and R. Himeno, “Force, work and power output of lower limb muscles during human maximal-effort countermovement jumping,” *J. Electromyogr. Kinesiol.*, vol. 15, no. 4, pp. 367–376, 2005.

[11] M. Damsgaard, J. Rasmussen, S. T. Christensen, E. Surma, and M. de Zee, “Analysis of musculoskeletal systems in the AnyBody Modeling System,” *Simul. Model. Pract. Theory*, vol. 14, no. 8, pp. 1100–1111, 2006.

[12] J. Dul, M. A. Townsend, R. Shiavi, and G. E. Johnson, “Muscular synergism – I. On criteria for load sharing between synergistic muscles,” *J. Biomech.*, vol. 17, pp. 663–673, 1984.

[13] W. Herzog and P. Binding, “Cocontraction of pairs of antagonistic muscles: analytical solution for planar static nonlinear optimization approaches,” *Math. Biosci.*, vol. 118, pp. 83–95, 1993.

[14] C. A. M. Doorenbosch and G. J. Schenau van Ingen, “The role of mono- and bi-articular muscles during contact control leg tasks in man,” *Hum. Mov. Sci.*, vol. 14, pp. 279–300, 1995.

[15] J. A. C. Ambrosio and A. Kecskemethy, “Multibody dynamics of biomechanical models for human motion via optimization.,” *J.C. Garcia Orden, J.M. Goicolea, J. Cuadrado Multibody Dyn. – Comput. Methods Appl.*, pp. 245–270, 2007.

[16] R. D. Crowninshield, R. C. Johnston, J. G. Andrews, and R. A. Brand, “A biomechanical investigation of the human hip,” *J. Biomech.*, vol. 11, pp. 75–85, 1978.

[17] F. Michaud, U. Lugris, Y. Ou, J. Cuadrado, and A. Kecskemethy, “Influence of muscle recruitment criteria on joint reaction forces during human gait,” in *Proceedings of the ECCOMAS Thematic Conf. on Multibody Dynamics*, pp. 1024–1031, Barcelona, 2015.

[18] F. Michaud, U. Lugris, Y. Ou, J. Cuadrado, and A. Kecskemethy, “Comparison of several forward dynamics based approaches to estimate muscular forces in human gait,” in *Proceedings of the 4th Joint Int. Conf. on Multibody System Dynamics*, Montreal, 2016.

[19] U. Lugris, J. Carlin, A. Luaces, and J. Cuadrado, “Gait analysis system for spinal cord injured subjects assited by active orthoses and crutches,” *J. Multi-body Dyn.*, vol. 227, No. 4, pp. 363–374, 2013.

[20] C. L. Vaughan, B. L. Davis, and J. C. O’Connor, *Dynamics of Human Gait (2nd ed.)*. Kiboho Publishers, 1999.

[21] J. G. de Jalon and E. Bayo, *Kinematic and Dynamic Simulation of Multibody Systems*. Springer-Verlag, 1994.

[22] S. L. Delp *et al.*, “OpenSim: Open-source software to create and analyze dynamic simulations of

movement," *IEEE Trans. Biomed. Eng.*, vol. 54, No. 11, pp. 1940–1950, 2007.

[23] H. Mokhtarzadeh *et al.*, "A comparison of optimisation methods and knee joint degrees of freedom on muscle force predictions during single-leg hop landings," *J. Biomech.*, vol. 47, no. 12, pp. 2863–2868, 2014.

[24] G. Pipeleers *et al.*, "Dynamic simulation of human motion: Numerically efficient inclusion of muscle physiology by convex optimization," *Optim. Eng.*, vol. 9, no. 3, pp. 213–238, 2008.

[25] F. E. Zajac, "Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control," *Crit. Rev. Biomed. Eng.*, vol. 17, pp. 359–411, 1989.

[26] Y. Ou, "An analysis of optimization methods for identifying muscle forces in human gait," University of Duisburg-Essen, 2012.