Dynamical analysis and design of active orthoses by aesthetic and energetic optimization

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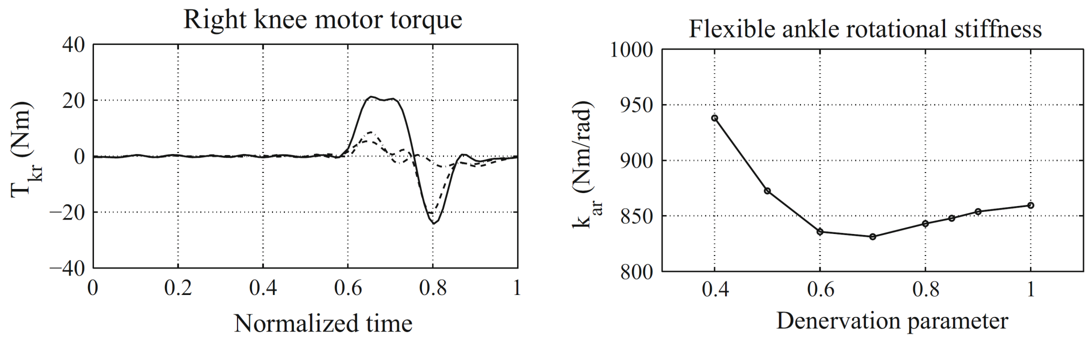
Gait analysis by computational mechanics techniques has been a major area of research interest for many years. Multibody system dynamics (MSD) techniques are potentially very powerful in this field and, in the last years, there have been many contributions from the MSD community to this challenging problem too. Among other approaches, parameter optimization techniques have been frequently used for motion synthesis of robotic and mechatronic systems. These optimization techniques have been proven to be also a powerful tool in human walking dynamics research [1, 2]. In these works, muscle forces and generalized coordinates are described in terms of a certain set of parameters, whose optimal values are found by minimizing cost functions that include an energy expenditure estimation and a measure of deviation from normal gait patterns. This method is mainly based on inverse dynamics since at each iteration of the optimization algorithm an inverse dynamics problem is solved by using the motion reconstructed from the design parameters. The main advantage of this approach is the complete elimination of the forward time integrations of the equations of motion, which significantly reduces the computational cost of simulation.

Although several studies have been performed in the last decades related to human walking dynamics, not much attention has been paid thus far to the analysis and simulation of human gait assisted by active orthoses or exoskeletons. In fact, most of the current active orthoses are designed and built without taking into account the coupled dynamic behaviour of the human-orthosis system. The aim of this work is to present a general methodology for the dynamical analysis of lower limb active orthoses using the parameter optimization approach presented in [2].

In a previous work, the authors simulated the gait of a subject wearing a pair of active stance-control knee-ankle-foot orthoses through a parameter optimization approach [3]. These orthoses were modeled based on a concept prototype presented in [4], which is aimed at assisting subjects with incomplete spinal cord injury (SCI) that preserve motor function at hip muscles, but have partially denervated muscles at the knee and ankle joints. This device incorporates a passive compliant ankle joint that constrains plantar flexion and a knee joint with two systems acting in parallel: a locking mechanism to constrain knee rotation during stance and a motor to assist knee flexion-extension during swing. The work by García-Vallejo et al. [3] allowed to simulate the orthosis-assisted gait taking into account the subject's SCI and the main design parameters into the optimization framework in order to account for different injury levels as well as to find the best set of design parameters for the orthosis. Moreover, the proposed optimization formulation allowed to obtain human-orthosis co-actuation strategies, where different device actuation was obtained depending on how the orthosis performance was included into the cost function. In addition, the simulations allowed to obtain optimal values of some orthosis design parameters as the ankle joint stiffness and the ankle torque in neutral position. Fig. 1 shows the optimal motor torque history obtained by using different performance criteria and the optimal ankle stiffness obtained for different injury levels. Thus, the results of the optimization approach allowed the design of active orthoses considering energy expenditure, aesthetics of gait and orthosis mechanical performance.

Based on the study above, the aim of this work is to apply the optimization framework to different situations where more actuation may be needed. For instance, models of a subject wearing fully actuated lower limb orthoses (active lower limb exoskeleton) including controlled motors at the hip, knee and ankle joints are simulated in order to find optimal simultaneous actuation histories. In addition, fully passive orthoses (passive lower limb exoskeleton) including compliance at the hip, knee and ankle joints are also simulated to investigate the influence of joint stiffness in the resulting gait motion.



(a) (b)

Fig. 1: (a) Optimal torque actuation at the knee of the active orthosis analyzed in [3] during a gait cycle, obtained based on three different performance criteria: cost function considering the metabolical cost of transportation and the deviation from normal walking pattern (solid line); cost function considering the metabolical cost of transportation, the deviation from normal walking pattern and the RMS value of the mechanical power of the orthosis actuators (dashed line); and cost function considering the metabolical cost of transportation, the deviation from normal walking pattern and the RMS value of the torques of the orthosis actuators (dash-dotted line). (b) Optimal ankle stiffness of the flexible ankle of the orthosis analyzed in [3] for different levels of injury parameterized in terms of the level of denervation. According to [3], the severity of the injury increases when the denervation parameter decreases.

In the applied optimization framework, trajectories, muscle force histories and motor controls are parameterized by using polynomials of 5th order and are found as a solution of a large scale nonlinear constrained optimization problem. The cost function used includes measures of the metabolical cost of transportation, of the deviation from normal walking pattern and of the motor performance. Moreover, the constraints are related to kinematics, dynamics and physiology. In the optimization problem, the vector of design variables used in [2] is augmented by including the motor control history along the walking cycle. Also, some constraints can be added if a particular joint should be locked during a part of the gait cycle. In addition, stiffness constants of the orthosis joints may also be included as design variables associated with the orthosis physical model. Then, the full vector of design variables can be written as:

$$\chi = \left[\begin{array}{ccccccccccccccc} y_1^T & \dots & y_{n_c}^T & f_1^{mT} & \dots & f_{N_m}^{m\,T} & T_{hr}^T & T_{kr}^T & T_{ar}^T & T_{hl}^T & T_{kl}^T & T_{al}^T & t_{ph}^T & p_g^T & p_o^T \end{array} \right]^T \quad (1)$$

where y_i $(i = 1, 2, \dots n_c)$ contains nodal values for generalized coordinate i; f_j^m, $(j = 1, 2, \dots N_m)$ contains nodal values for muscle force j; T_{hr} and T_{hl} contain nodal values of the motor torques at the right and left hip joints, respectively; T_{kr} and T_{kl} contain nodal values of the motor torques at the right and left knee joints, respectively; T_{ar} and T_{al} contain nodal values of the motor torques at the right and left ankle joints, respectively; t_{ph} $(ph = 1, 2, \dots 8)$ contains the durations of the eight phases of a walking cycle, p_g contains two geometrical parameters describing the kinematic constraints due to the step lengths L_R and L_L and finally, p_o contains design parameters associated with the orthosis physical model.

References

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