

Analysis and design of an active orthosis by using parameter optimization to simulate assisted gait dynamics performance

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Abstract

Inverse dynamics simulation is often used in robotic and mechatronic systems to track a desired trajectory by feed-forward control. Musculoskeletal multibody systems found in biomechanics are highly overactuated due to the many muscles, and they show a switching number of closed kinematical loops. The method of inverse dynamics is also successfully applied to overactuated systems by parameter optimization for two- and three-dimensional models of the human musculo-skeletal system. The simulation approach used in this research is fully based on optimization, see Ref. [1].

In this work, the gait simulation of a subject wearing an active stance-control knee-ankle-foot orthosis is carried out. This device was presented by Font-Llagunes et al. [2] and it is aimed at assisting incomplete spinal cord injured (SCI) subjects that preserve motor function at hip muscles, but have partially denervated muscles at the knee and ankle joints. The orthosis is shown in Figure 1. The ankle joint mechanically constrains the angle to be between 0 and 20° (dorsiflexion), thus avoiding drop-foot gait, and incorporates a spring that provides a passive dorsiflexion torque within this range of motion. The knee joint consists of two independent systems to assist during swing and stance. The swing flexion-extension motion is controlled by means of an electrical DC motor and a commercial controllable locking mechanism is used to prevent knee flexion during stance. It is important to say that the orthosis is equipped with plantar sensors and angular encoders for control.

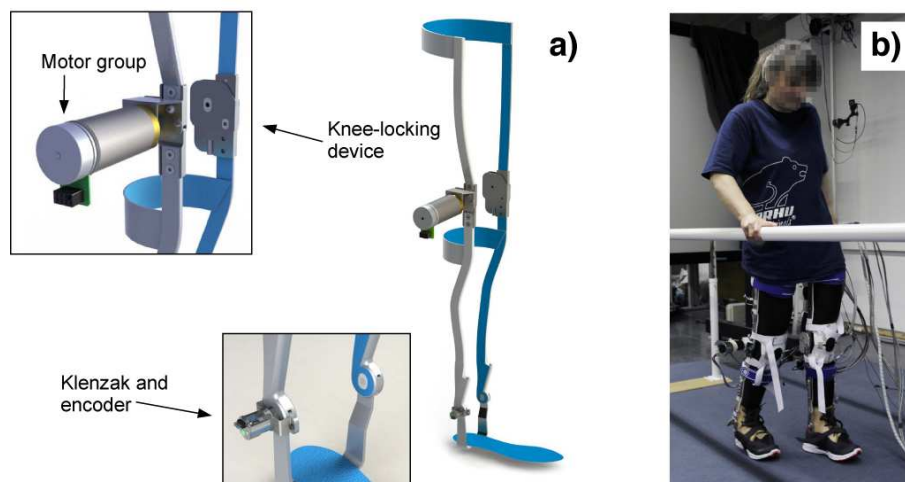


Figure 1: a) CAD design of the active orthosis. b) SCI subject wearing the orthosis during experimental tests.

The operation of the orthosis during the gait cycle is as follows: At initial stance, the plantar sensors detect contact and then the knee joint is locked. During the stance phase, the motor does not exert any torque on the joint, and the plantar sensors and ankle encoder data give information on the evolution of the gait cycle. Once contact is over, the knee joint is unlocked and the swing phase begins. During this phase, the motor controls the knee flexion-extension motion. Then, the foot makes contact again with the ground and a new step begins.

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 in [1] is augmented by including the motor control history along the walking cycle. Also, some constraints must be added for the part of the cycle where the knee joint is locked. In addition, stiffness constants of the orthosis ankle joints are included as design variables. Then, the full vector of design variables can be written as:

$$\chi = \left[y_1^T \quad \dots \quad y_{n_c}^T \quad f_1^{mT} \quad \dots \quad f_{N_m}^{mT} \quad T_{kr}^T \quad T_{kl}^T \quad t_{ph}^T \quad p_g^T \quad p_o^T \right]^T \quad (1)$$

where y_i , $i = 1, 2, \dots, n_c$ contains all nodal values for each generalized coordinate, f_j^m , $j = 1, 2, \dots, N_m$ contains all nodal values for each muscle force, T_{kr} and T_{kl} contain all nodal values of the motor torques at the right and left knees, respectively, t_{ph} ($ph = 1, 2, \dots, 8$) contains eight components representing 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 the design parameters associated with the compliant orthosis ankle joint. Note that the partially denervated muscles of the considered SCI subject are modelled as healthy muscles with limited activation as in [3].

As an example of the results provided by the optimization procedure, Figure 2 shows the optimal values obtained for the torque in neutral position and for the rotational stiffness of the compliant orthosis ankle as a function of a denervation parameter α that represents the level of injury of the subject. In this simulation, the level of injury has been represented by limiting the maximum neural excitation of the main lower limb muscles according to a function of parameter α , so that the lower the value of α the higher the subject injury level is.

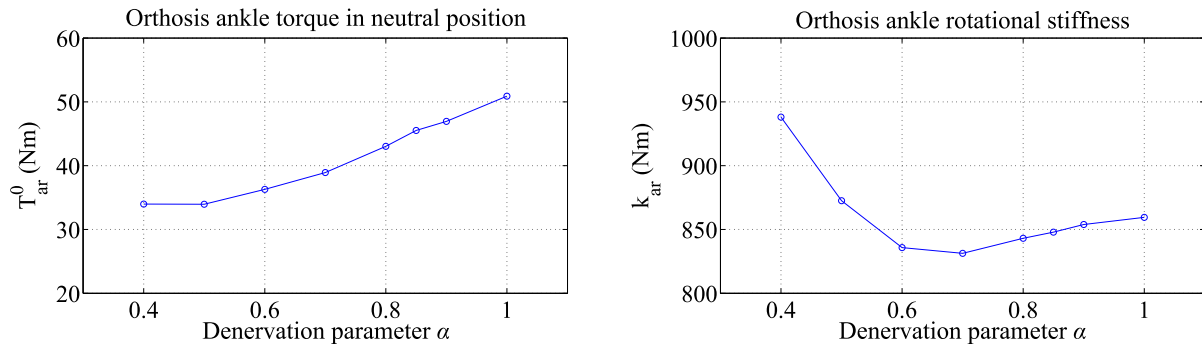


Figure 2: Optimum ankle torque in neutral position and optimum ankle rotational stiffness as a function of the denervation parameter α .

References

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