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DESIGNING OPTIMAL CONTROLS BY PARAMETER OPTIMIZATION FOR A STANCE-CONTROL KNEE-ANKLE-FOOT ORTHOSIS

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Inverse dynamics simulation is often used in robotic and mechatronic systems to track a desired trajectory by feed-forward control. Musculoskeletal multibody systems are highly overactuated and 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 musculoskeletal system. The presented simulation approach is fully based on optimization [1].

In this work, the gait simulation of a subject wearing an active stance-control knee-ankle-foot orthosis is carried out. This device is aimed at assisting incomplete spinal cord injured (SCI) subjects that preserve motor function of the hip muscles, but have partially denervated muscles at the knee and ankle joints [2]. The considered prototype is shown in Figure 1. The ankle joint constrains the angle to be between 0 and 20° (dorsiflexion), and incorporates a spring that provides a passive torque. The knee joint consists of two independent systems: An electrical DC motor controls the swing flexion-extension motion and a controllable locking mechanism is used to prevent knee flexion during stance. The orthosis is equipped with plantar sensors and angular encoders for control purposes.

Trajectories, muscle force histories and motor controls are parameterized by using 5th order polynomials 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 patterns and of the actuating motor performance. The constraints are related to kinematics, dynamics and physiology. Moreover, some constraints must be added for the part of the cycle where the knee joint is locked. The vector of design variables presented in [1] is augmented by including the motor control history along the gait cycle and the stiffness constants related to ankle

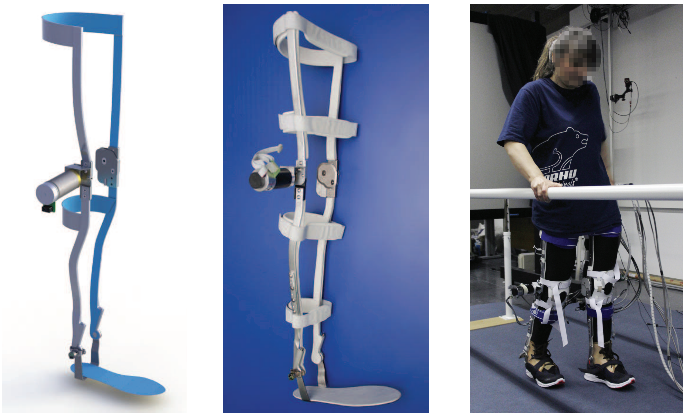

Figure 1: CAD design (left), orthosis prototype (middle) and subject wearing the orthosis (right).

joints. This vector can be written as:

$$\boldsymbol{\chi} = \left[\begin{array}{cccccccccc} \boldsymbol{y}_1^T & \dots & \boldsymbol{y}_{n_c}^T & \boldsymbol{f}_1^{mT} & \dots & \boldsymbol{f}_{N_m}^{m\ T} & \boldsymbol{T}_{kr}^T & \boldsymbol{T}_{kl}^T & \boldsymbol{t}_{ph}^T & \boldsymbol{p}_g^T & \boldsymbol{p}_o^T \end{array} \right]^T \tag{1}$$

where $\boldsymbol{y}_i$, $i = 1, 2, ...n_c$ contains all nodal values of the different generalized coordinates, $\boldsymbol{f}_j^m$, $j = 1, 2, ...N_m$ contains all nodal values of the different muscle forces, $\boldsymbol{T}_{kr}$ and $\boldsymbol{T}_{kl}$ contain all nodal values of the motor torques at the right and left knees, respectively, $\boldsymbol{t}_{ph}$ ($ph = 1, 2, ...8$) contains eight components representing the durations of the eight phases of the gait cycle, $\boldsymbol{p}_g$ contains two geometrical parameters describing the kinematic constraints due to the step lengths L_R and L_L and finally, $\boldsymbol{p}_o$ contains the design parameters associated with the flexible ankle of the orthosis. Note that the partially denervated muscles of the considered SCI subject are modelled as healthy muscles with limited activation as in [3].

The proposed cost function includes a term related to the orthosis energy consumption. In this way, the combined human-orthosis actuation can be evaluated. The effect of different walking velocities and step lengths on the human-orthosis co-actuation is investigated. We believe that designing the orthosis control system according to the minimization of metabolical cost, deviation from normal walking and orthosis energy consumption is expected to help people using such devices to have a more efficient gait.

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