A MODEL FOR THE DESIGN OF FES STANDING-UP STRATEGIES

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Abstract - We present a physiologically based model that describes major properties of muscles and segmental dynamics during FES assisted standing-up. It accounts for relevant time-dependent and nonlinear properties (temporal and spatial summation, muscle fatigue, force-length and force-velocity relations, viscoelastic effects, nonlinear moment arms, choice of muscles stimulated, seat and upper-body interaction, etc.). The model can be implemented in model-based controllers or be used for the design of arm-supported FES standing up movements.

I. INTRODUCTION

The control of limb movements by Functional Electrical Stimulation (FES) is complex. The stimulated musculoskeletal system has multiple degrees of freedom with highly nonlinear, time-varying dynamics. Side effects such as muscle fatigue and spasticity even complicate the control task. In designing a neural prosthesis, the engineer attempts to develop a mathematical description of the system to predict its biomechanical behavior under a variety of specified stimulation inputs and external loading. Models of the musculoskeletal system can potentially assist in the design of adequate FES control strategies [1],[2].

The main objective of this paper is to present a physiologically based model that describes major properties of muscle and segmental dynamics of the lower extremities during FES. The model can assist in designing and optimizing open- and closed-loop control strategies, where effective muscle contraction and reduced fatigue are desired. The model is being used for the study of FES-supported sit-to-stand transfer in paraplegic patients.

II. MODEL

The human body is described as a three-segmental model including shanks, thighs, and upper body. Feet are fixed on the ground. We modeled nine muscle groups inducing moments about the ankle, knee, and hip joint in the sagittal plane (Fig. 1). Each group has its own activation and contraction dynamics. Input to each muscle group is the modulated pulse width and pulse frequency as provided by a stimulator (continuous-time inputs into model). Muscle activation is computed considering the effect of spatial and temporal summation by a nonlinear recruitment curve and second order calcium dynamics, respectively [3]. A fatigue/recovery model [2],[3] was incorporated which accounts for the fact that during high frequency stimulation the effect of fatigue is stronger. We implemented a constant time delay (25 ms) resulting from action potential propagation and delays in calcium release [3]. In the model a force-length and a force velocity factor are computed from muscle lengths and velocities by applying equations presented by [4] and [5], respectively. Muscle length \( l_i^M \) and velocity \( v_i^M \) of a muscle \( i \) are

\[
l_i^M = C_i + \sum_j m_a(\varphi_j) d\varphi_j, \quad v_i^M = \sum_j \dot{\varphi}_j m_a(\varphi_j). \tag{1}
\]

Index \( j \) expresses the joint number. Each joint \( j \) a muscle spans contributes to the length and velocity of the muscle. Moment arms \( m_a(\varphi) \) given as algebraic functions have been adjusted to measured moment arm curves from the literature [6]-[9]. The active moment developed by a single muscle group is calculated by multiplying its moment arm with the maximum isometric muscle force (from [10]), the actual muscle activation and the force-length and force-velocity factors.

Muscle-specific parameters were taken or derived from the literature (muscle contraction times: [11],[12]; fatigue time constants: [13],[14]; maximum muscle contraction velocities: [13],[15], optimal muscle lengths and forces: [10]). Some parameters of the force-length relation (\( C_i \) in Eq. 1, shape factor, see [4]) where adjusted so that simulated and measured maximum voluntary joint moments vs joint angle [10] showed good agreement. To keep the number of muscle parameters low, we separated the passive from the active muscle properties: all passive properties (muscle viscosity and elasticity) were assigned to the joints. Passive viscous joint moment is modeled by a nonlinear damping function [2]. Joint elasticity was modeled by a double exponential equation, where the influence of the adjacent joint angles is taken into account [16]. Total joint moment is obtained by summing active and passive moments. Equations of motion describe the segmental dynamics of the body. Interaction with the seat was modeled similar to [17]: a pair of nonlinear spring-dampers take into account the horizontal and vertical reaction force. The condition that vertical and horizontal seat reaction forces are both zero at seat-off was satisfied by introducing the effect of sliding. Shoulder forces

![Three-segmental model with mono- and bi-articular muscle groups](image-url)
represent the interacting arms. The influence of the upper extremities (arm support) is modeled by a fuzzy controller.

The model was programmed with MATLAB/SIMULINK. The computed motion is visualized by graphic animation.

III. RESULTS AND DISCUSSION

Results of the different model components showed good qualitative agreement with measured data from our own measurements and the literature. The model can describe the effect of temporal and spatial summation (frequency and recruitment curve, see [18]), muscle fatigue at different stimulation frequencies [3], [14], maximum isometric joint moments [10], passive pendulum motions, and limb movements induced by artificially stimulating different muscle groups [2]. Note that the model describes the musculoskeletal system of a generic subject. To design a controller for a specific patient, the model can be adapted to the individual by performing detailed parameter identification experiments (see [2] and [18]).

The model can be used for the design of different open- and closed-loop control strategies. In an iterative approach applied to the whole model, we found pulse width modulated input trajectories (at constant frequencies) that yielded a satisfactory open-loop sit-to-stand movement. However, we observed that the open-loop controlled movement of the stimulated human body is very sensitive to changes in model parameters and stimulation inputs.

Useful experimental kinematic and kinetic data have been obtained in motion analysis studies with healthy controls [19]. Together with the model, the experimental data is being used to tested, for example, how the stimulation of different muscle groups (especially bi-articular) contributes to the desired movement, how the arms can support the sit-to-stand transfer, or which sensors should be chosen for an effective closed-loop application. Furthermore, applying this model in a model-based control strategy showed considerable improvements in the resulting movement (similar to the results obtained in single-joint investigations [20],[21]).

ACKNOWLEDGMENT

The authors thank F. Bahrami for her advises during this study. The work was supported by the DFG (SFB 462 "Sensomotorik", project A1).

REFERENCES