PREDICTIVE CONTROL OF PLANTARFLEXOR MUSCLE-TENDON FORCE DURING SIMULATED HUMAN HOPPING

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Abstract

Assistive devices (e.g., exoskeletons) could be used to steer a user's musculotendon unit (MTU) loading *in vivo.* A challenge is to develop a control algorithm that is simple enough to enable fast computation and adaptable enough to handle different gait patterns. Here, we propose a nonlinear model predictive control (NMPC) approach to maintain a predefined threshold tendon force within the plantarflexor muscle-tendon complex during simulated human hopping.

Introduction

Human-in-the-loop (HIL) control of exoskeletons can identify the assistance strategy that minimizes a user-related cost function during gait [1]. A downside of this approach is that the computational time required for state-of-the-art HIL approach to identify the optimal assistance pattern is approximately half an hour [2]. Alternatively, we seek direct control of muscle-tendon (MT) force via estimation using available toolboxes [3] and a personalized closed-form model [4]. In this way, it should be possible to demonstrate direct control of a user's tendon force – to say, target a given cyclic strain rate for tissue healing. Here, we use a modelling/simulation framework to demonstrate model-predictive control (MPC) that can predict future tendon dynamics and provide smooth exoskeleton assistance during simulated hopping.

Methods

Building on previous simple models of human hopping with passive-elastic exoskeletons [5] by adding a computationally-efficient closed-form equivalent model for MT force [4], a combined exoskeleton-MT model (Fig. 1a) can be obtained using the following equation:

$$\begin{aligned} \dot{F}^T &= k^T\left(\dot{L}^{MT} - \dot{L}^M \cos\alpha + L^M \sin\alpha.\dot{\alpha}\right) \\ \ddot{L}^{MT} &= \frac{-g}{W}\left(F^T + \underset{u}{F_{ac}} - W\right) \end{aligned} \qquad (1)$$

where, $F^T$and k^T are the tendon force and stiffness, respectively. The MTU length, L^{MT}, and the pennation angle, α, are derived by the limb kinematics and the muscle fibre length, L^M, is derived using the method discussed in [4]. Equation (1) is a state-space model of the combined exoskeleton-MT model:

$$\dot{x} = Ax + Bu \quad , \quad x = \begin{bmatrix} F^T \\ L^{MT} \\ \dot{L}^{MT} \end{bmatrix} \quad , \quad u = F_{ac} \,. \qquad (2)$$

Next, the derived nonlinear state-space representation is used as the inner model of an NMPC controller and the cost function of the controller is defined as:

$$J = w_1 u^2 + w_2 (\Delta u)^2 + w_3 \left(F^T\right)^2 \qquad (3)$$

where, w_i are constants and Δu is the exoskeleton actuator force increment. We extracted human-like specifications and activation dynamics from [5] (Fig. 1b) and the controller's horizon and upper bound were set to 15 steps and 1000 N, respectively. To demonstrate feasibility *in silico*, we set the goal of the controller to keep tendon force under 1000 N. An interior-point method was used for solving the optimization problem.

Results and discussion

Over a range of simulated hopping intensities (Fig. 1b), our novel NMPC controller appropriately commanded exoskeleton actuator force (Fig. 1c) to maintain tendon force under the target threshold (Fig. 1d).

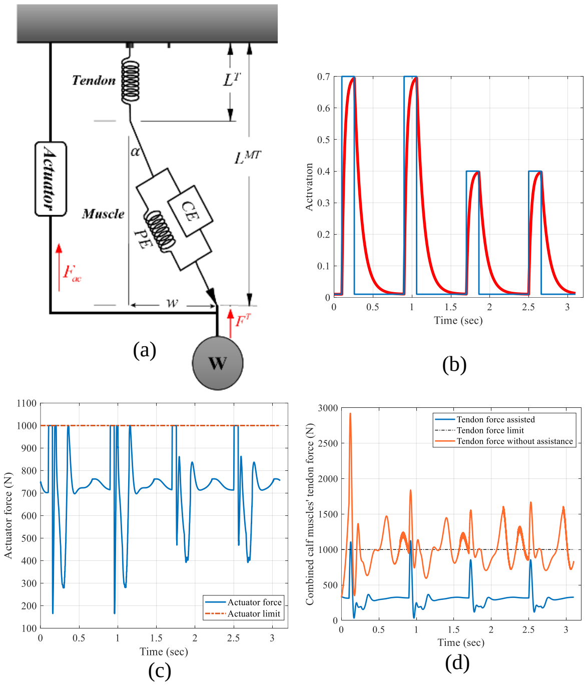

Figure 1: a) The simplified hopping and actuator model, b) activation dynamics for hopping, c) actuator assistive force, d) tendon force with and without assistance

References

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