A First Step Towards Adapting the 3D Reflex Based Neuromuscular Gait Model for Gait Assistive Devices

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I. INTRODUCTION

For effective control and interaction of active prosthetic and orthotic (P/O) devices with the human, understanding of human control of gait is needed. Feedback, provided by sensors and reflexes in the body, can compensate for unexpected environmental conditions or sensory noise. This was shown by a model of human gait purely based on reflexive feedback, presented by Song [3], referred to as neuromuscular model control (NMC). The NMC framework requires only basic supraspinal input of foot clearance height and foot placement location. Additionally, NMC uses local low-level muscle reflex signals (stretch, stretch rate, and force) to generate muscle activation. Eleven simulated Hill-type [2] muscles per leg generate force, resulting in a net torque around the joints.

The goal of this study is to enhance NMC to be more robust and more subject specific. This is done by re-optimizing the model parameters using perturbed human kinematic and torque data. By driving the model using subject specific input data from previous research [4], instead of optimizing by evaluating the model's forward dynamics, the model parameters can be optimized so that the output of the model follows human data more closely.

II. MATERIALS AND METHODS

To achieve a stable, human-like gait by data driven optimization, healthy human gait kinematic and torque data is used in which subjects were perturbed in the anteroposterior (AP) plane and in the mediolateral (ML) plane [4]. Perturbations were applied at the pelvis for 150 ms directly after right foot toe off. The model is re-optimized using this data. NMC uses a parameter-set of 82 reflex gains and offsets. These parameters were optimized using covariance matrix adaptation evolution strategy (CMAES [1]) to have the model walk stably and energy-efficiently in the model environment. The rigid body based forward dynamics in the model is removed. Instead, the model receives averaged joint angles and joint velocities measured from 10 healthy subjects as input and gives joint torques as output. The given input data consists of 1) four steps of unperturbed walking, 2) four steps of backward perturbed walking, 3) four steps of forward perturbed walking and 4) four steps of rightward perturbed walking. The mean absolute difference between human and model torques is minimized by reoptimizing the model parameters using CMAES. Perturbed data is provided with increasing magnitude, ranging from 0 to 0.16 times body weight. The optimization results in a *single* parameter-set, for which the model output torques are as close as possible to human torque data for *all* these four conditions.

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III. RESULTS AND DISCUSSION

The data-driven parameter optimization results in torques that resemble the human torques more than the original parameter-set (Fig. 1). On average, the original parameter set results in a mean error of 8.6 times the standard deviation of healthy human walking. After optimization, this is reduced to a mean error of 1.9 times the standard deviation. The model is able to reproduce human torque reaction to perturbations, except in the ankle. Humans counteract AP perturbations by applying a reactive torque around the ankle, providing more dorsiflexion in the case of backward perturbations. The NMC model, however, shows the opposite behavior, applying more plantar flexion. This behavior is still present after optimization. This indicates the current NMC model is unable to adjust ankle torques in reaction to pelvis perturbations in a human-like fashion.

IV. CONCLUSION

Optimizing using human data as input instead of using a model environment results in net-torques much closer to human data in all joints. However, the current, nor the re-optimized, model are able to reject perturbations on the pelvis in a human-like fashion in the ankle joint, indicating a shortcoming in the current set-up NMC model. Our current work involves extending the model with an extra module, or extra modules, activating muscles around the ankle in case of a perturbation. After this extension the model will again be re-optimized on human data. The re-optimized model will be implemented as a control strategy on P/O devices.

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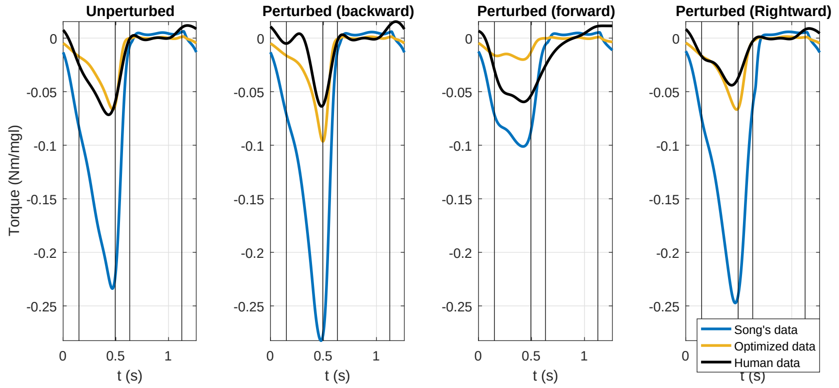
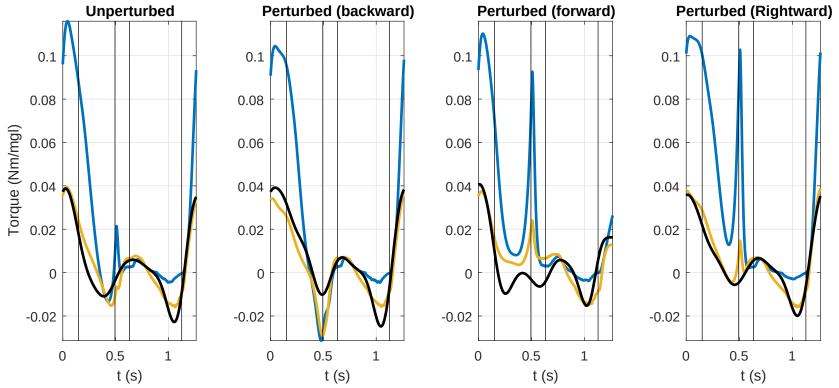
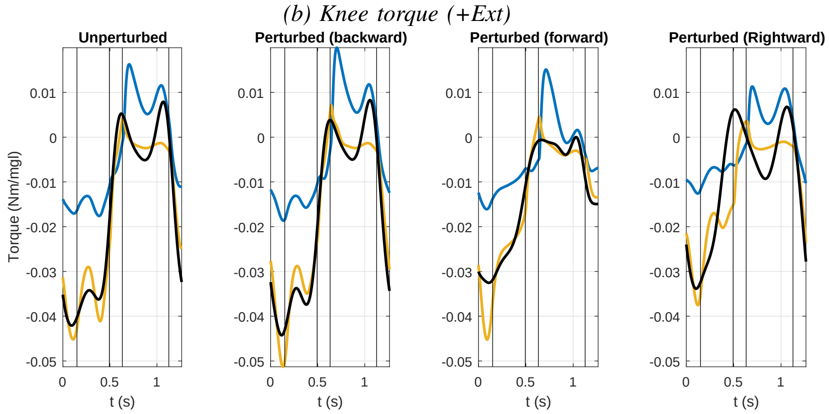


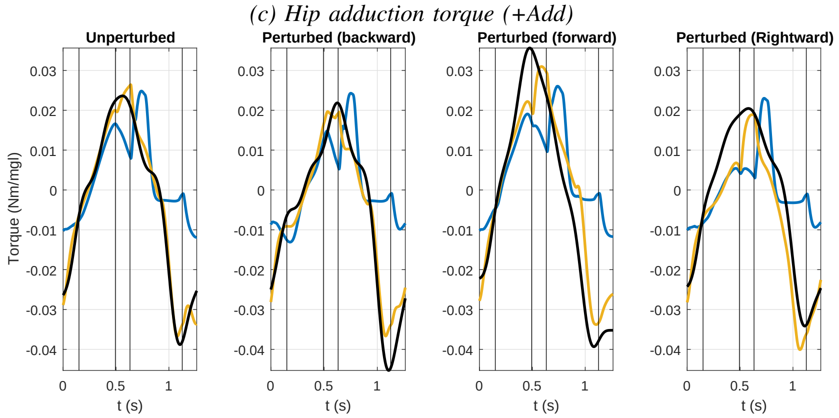
(a) Ankle torque (+DF)



(b) Knee torque (+Ext)



(c) Hip adduction torque (+Add)



(d) Hip flexion torque (+Flx)

Fig. 1: Human data of torque (black), torques resulting from human input data in the original NMC model (blue) and torques resulting from human data in a re-optimized model (yellow). Unperturbed data, backward perturbed data, forward perturbed data and rightward perturbed data are displayed for all four degrees of freedom of the model. Positive direction of movement is mentioned in subtitle. Vertical bars display events (right toe off + start perturbation, end perturbation, right heel strike, left toe off, left heel strike)