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 reflexove feedback, presented by Geyer [1] and Song [4], 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 [3] muscles per leg generate force, resulting in a net torque around the joints.

In this work, a first step is done towards adjusting the neuromuscular model for the use in P/O devices. Predefined trajectory-based impedance controllers are currently the golden standard of control of gait in P/O devices [5]. These controllers require switching to different trajectory sets when in different environments, like rough or sloped terrain. In simulation, NMC has shown to be able to generate stable human-like gait for different environments without adjusting parameters nor switching to different trajectories [1].

Safety and stability are an important part of controller design for P/O devices. A robust gait controller should at least not decrease, but preferably increase the stability against slipping, tripping or perturbations on the patient wearing the P/O device. A safe and robust device will minimize injury, allow walking in more challenging environments and increase the patient's confidence in his/her own mobility. However, this part is often neglected in research.

The goal of this study is to enhance NMC to be more robust and more subject specific. The robustness against perturbations of the NMC is investigated as a first step. The next step is to make NMC more subject specific by re-optimizing the model parameters using human kinematic and torque data. By driving the model using subject specific input data from previous research [6], 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.

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II. MATERIALS AND METHODS

To achieve a stable, human-like gait by data driven optimization, we first investigate the robustness of the original NMC model [4] against perturbation forces at the pelvis. The perturbations consist of a forward or backward force applied in the anterioposterior (AP) plane, acting on the central point between the hips for a duration of 150 ms directly after right foot toe off. The model data is compared to collected healthy human data [6] in which subjects were perturbed in the same way.

Next, the model is re-optimized using healthy human gait kinematics and torques. NMC uses a parameter-set of 82 reflex gains and offsets. These parameters were optimized using covariance matrix adaptation evolution strategy (CMAES [2]) 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 joint angles and joint velocities measured from healthy subjects as input and gives joint torques as output. The mean absolute difference between human and model torques is minimized by reoptimizing the model parameters using CMAES. As input, human data from *one* subject, instead of mean subject data, is used. The given input data consists of 1) two steps of unperturbed walking, 2) two steps of backward perturbed walking and 3) two steps of forward perturbed walking. The optimization results in *single* parameter-set, for which the model output torques are as close as possible to human torque data for *all* these three conditions.

III. RESULTS AND DISCUSSION

Fig. 1 shows model and human torques around the ankle joint in response to AP force perturbations at the pelvis. The model torques show a different response to force perturbations than human data. Humans counteract AP perturbations by applying a reactive torque around the ankle, providing more dorsiflexion in the case of backward perturbations. They keep the step length after the perturbations constant. The NMC model, however, shows the opposite behavior, applying more plantar flexion and taking smaller steps to reject the perturbation.

For unperturbed walking, the data-driven parameter optimization results in ankle torques that resemble the human torques much better than the original parameter-set (Fig. 2a). Although not shown here, this effect is present in all joints.

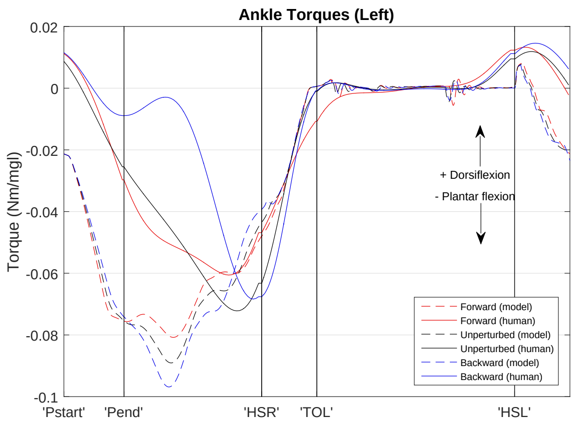

Fig. 1. Human data of ankle torque (solid lines) compared to Song's model data (dotted lines) in case of perturbations at t = 0 s. Perturbations are applied right after toe-off of the right leg in AP direction. PStart = start of perturbation , Pend = end of perturbation, HSR = heel strike right, TOL = toe-off Left, HSL = heel strike left. Humans compensate a backward perturbation (blue) by applying more dorsiflexion in the stance leg (Pstart until HSR) to counter perturbations. Model data shows the opposite effect (more plantar flexion).

However, the effect shown in Fig. 1 is still present after re-optimization. While human data shows more dorsiflexion after backward perturbation and more plantar flexion after forward pertubation, the model shows an opposite effect. This indicates the current NMC model is unable to adjust ankle torques in reaction to pelvis perturbations in a human-like fashion.

IV. CONCLUSION

The current NMC model is not able to provide human-like ankle torques reacting to AP perturbations. The reaction to a perturbation on the original un-optimized model is opposite to the human reaction in this joint. 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, 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. As an indication for perturbations the velocity of the center of mass will be used, to stay within the current bounds of the model and not use information unknown to the human body. 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.

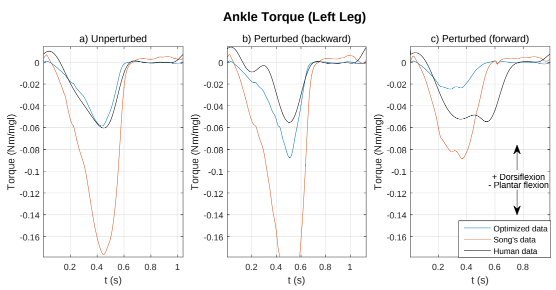

Fig. 2. Human data of ankle torque (black), torques resulting from human input data the un-optimized original NMC model (red) and torques resulting from human data in a re-optimized model (blue). A) unperturbed data, b) backward perturbed data, c) forward perturbed data.

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