MODULATION OF PROPORTIONAL AND DERIVATIVE (PD) CONTROL COMMAND CAN MATCH PHYSIOLOGICAL ANKLE TORQUE MODULATION DURING QUIET STANDING

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INTRODUCTION

It has been shown that the lost ability of individuals with spinal cord injury (SCI) to stand can be regained by artificially stimulating the skeletal muscles in the lower limbs that cannot be controlled voluntarily [1]. Recently, *arm-free* standing of individuals with SCI using functional electrical stimulation (FES) has drawn much attention in the field as it might allow individuals with SCI to stand and at the same time use both arms to perform activities of daily living (ADL) [2,3]. However, current FES systems do not open up this possibility yet. Instead, they require that the individual actively regulates balance using one or both arms, thus limiting the use of the FES systems during ADL [4,5].

This lack of knowledge of an effective as well as efficient closed-loop control strategy to regulate FES induced muscle contractions might also be due to an insufficient understanding of the control strategy that healthy individuals apply to compensate for a long sensory-motor time delay. In this context, it has been shown by our team that a feedback system regulated by a proportional and derivative (PD) position controller has the potential to represent the control scheme that the CNS applies to generate an active torque component and regulate body sway despite the sensory-motor time delay [6]. In a series of studies, we have 1) identified appropriate PD gains that fulfill stability as well physiological requirements [7]; and 2) verified experimentally that the implemented PD controller can, in fact, improve balance during standing and generate quiet standing dynamics that exhibit physiological characteristics [8].

In order to validate our previous results and to provide further evidence that a PD control strategy can mimic the physiological control task, the purpose of the present study was to investigate whether the modulation of the PD control command can match the ankle torque modulation as observed in healthy individuals during quiet standing.

METHODS

PD Controlled Feedback Model

Since the dynamics of quiet standing can be approximated by an inverted pendulum [9], the primary purpose of the feedback control system during quiet stance is to generate an appropriate motor command for the ankle joint muscles (*controlling* ankle torque) based on sensory feedback information about the body kinematics (*controlled* body angle). A model of the feedback loop, which assumes that a PD controller with particular gains (Kp and Kd) can mimic the physiological control task at the level of the central nervous system (CNS), is shown in Fig. 1.

In order to stabilize the body, the PD controller has to compensate for the following sensory-motor time delay (Fig. 1):

- a feedback time delay that represents cumulative time loss due to neural-transmission from the ankle somatosensory system to the brain (τ_F = 40 ms)[10];
- a motor time delay, which represents cumulative time loss due to the sensory-motor information process in the CNS and the neural transmission from the CNS to the plantar flexors (τ_M = 40 ms) [11]; and

- an electromechanical time delay between the neural command and the torque generation that is due to the critically damped, second-order dynamics of the muscles (τ_E) [12].

Note that τ_E is a varying quantity that depends on the frequency spectrum of the neural input and the physical properties of the muscles, i.e., the plantar flexors (characterized by the contraction time T).

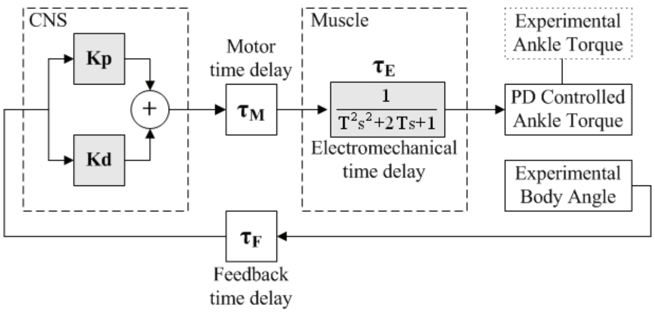

Figure 1: Sensory-motor feedback model of quiet standing when controlled by a PD controller. The controller with gains Kp and Kd takes advantage of the position and velocity information of the body to regulate balance in spite of the sensory-motor time delay (τ_F, τ_M, and τ_E).

Using body angle data from quiet standing experiments with able-bodied subjects, the feedback model shown in Fig. 1 was used to generate PD controlled ankle torque data. The data were then compared with experimental ankle torque data and optimized by tuning the variables Kp, Kd, and T (grey boxes in Fig. 1).

Experimental Data Acquisition and Processing

In order to acquire experimental body angle and ankle torque data, quiet standing experiments with ten healthy male subjects (age: 30.7 ± 4.1 yr; height: 178.1 ± 4.5 cm; weight: 72.6 ± 4.1 kg) were performed. Each subject was asked to maintain a quiet stance posture standing barefoot with eyes open (EO) or eyes closed (EC). For each visual condition, two trials of 130 seconds each were performed, and data from the latter 120 seconds were subjected to subsequent analyses.

During each trial, the subject was standing on a force platform (9281B, Kistler, Switzerland) that measured the fluctuation of the subject's total ankle torque. Additionally, a laser displacement sensor (LK2500, Keyence, Japan) was placed behind the subject recording the anterior-posterior body sway fluctuation at the midpoint between the right ankle and knee joints.

The time series were logged at a sampling frequency of 1 kHz and low-pass filtered using a fourth order, zero phase-lag Butterworth filter with a cut-off frequency of 2 Hz. After calibrating the DC offset of the body sway fluctuation against the ankle joint position, the body sway recordings were converted into body angle data using the horizontal height of the laser device.

Optimization Procedure

For each trial of each subject, a segment of 30 seconds of body angle data was used to calculate the PD controlled ankle torque by means of the sensory-motor feedback model shown in Fig. 1. In order to investigate whether a good matching between the PD controlled and experimental ankle torque was possible, the gains of the PD controller (Kp and Kd) as well as the time constant of the second-order dynamics of the plantar flexors (T) were optimized by means of the *DIRECT optimization technique* (grey boxes in Fig. 1) [13]. This technique is a sampling algorithm that requires no knowledge of the objective function gradient. Instead, the algorithm samples points in the domain and uses the information it has obtained to decide where to search next.

First, a PD controller with initial gains of Kp^* = 750 Nm/rad and Kd^* = 350 Nm·s/rad [7] and a time constant of T^* = 116 ms for the plantar flexor dynamics [14] were used to estimate the open-loop ankle torque fluctuation. Subsequently, the PD gains as well as the time constant of the second-order muscle dynamics were optimized over a wide range of values. For each trial, the optimization procedure was terminated once the error difference between two subsequent iterations was smaller than 10^{-5} Nm (within 30 iterations).

RESULTS

In Fig. 2, examples of the experimental ankle torque (bold grey line) and the PD controlled

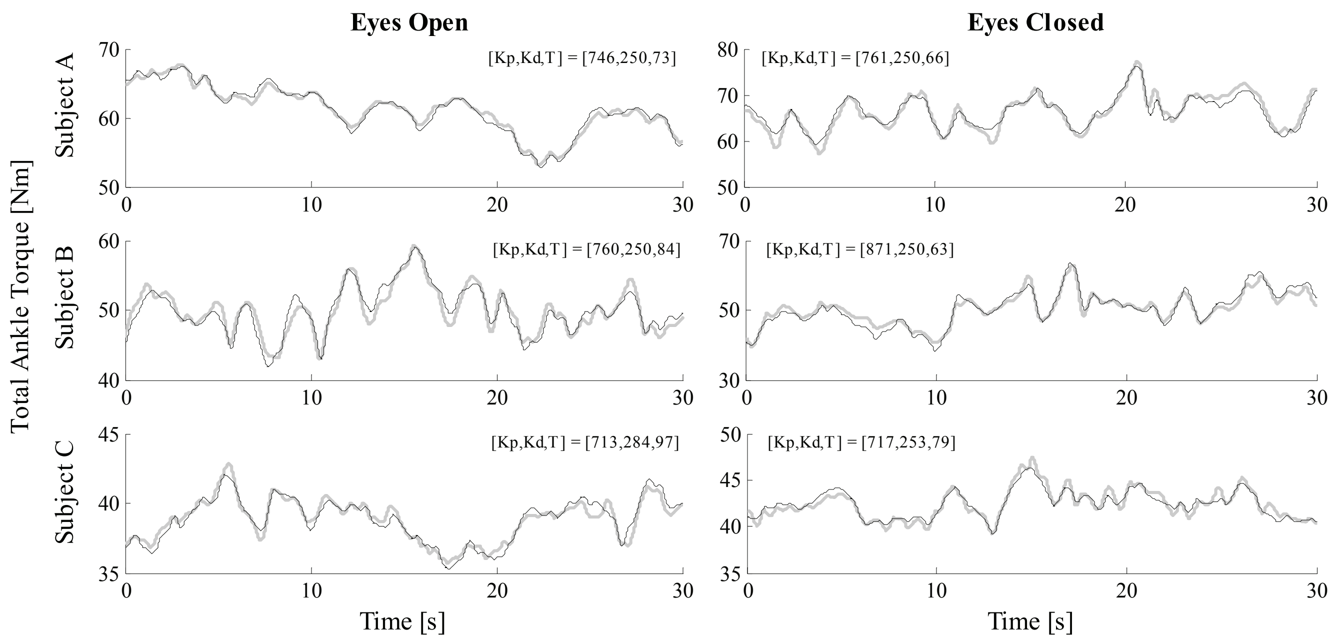

Figure 2: Examples of the experimental ankle torque (bold grey line) and PD controlled ankle torque (thin black line) after optimization of Kp [Nm/rad], Kd [Nm·s/rad] and T [ms]. Shown are three subjects for the two eye conditions (EO and EC).

ankle torque (thin black line) are shown for three different subjects and both eye conditions, EO (left panel) and EC (right panel). Note that the graphs show the PD controlled ankle torque fluctuation after optimization of the parameters Kp [Nm/rad], Kd [Nm·s/rad], and T [ms]. A simple visual inspection suggests that the fluctuation of the PD controlled ankle torque closely matched the one of the experimental ankle torque as observed in the healthy subjects during quiet stance.

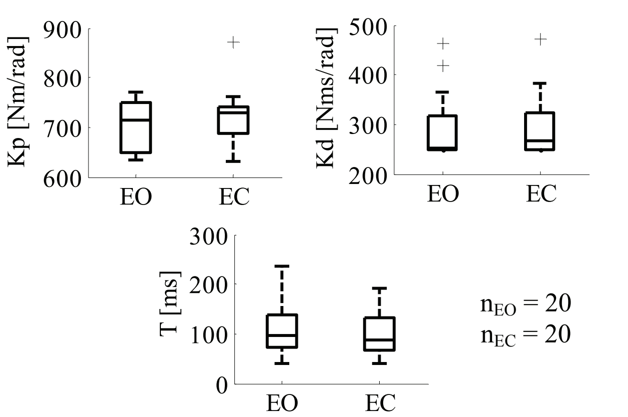

Figure 3: Group values of the optimized parameters Kp, Kd, and T for both eye conditions (medians, quartiles, and extrema).

In Fig. 3, the group values of the optimized parameters are shown as box plots. Note that all three sets of optimized parameters included the initial values Kp^*, Kd^*, and T^*. Moreover, the overall error torque and matching percentage as indicators for the matching capability of the PD controller are depicted in Fig. 4. For each of the 40 trials, the overall error torque was less than 1.5 Nm, resulting in a matching percentage of over 95%.

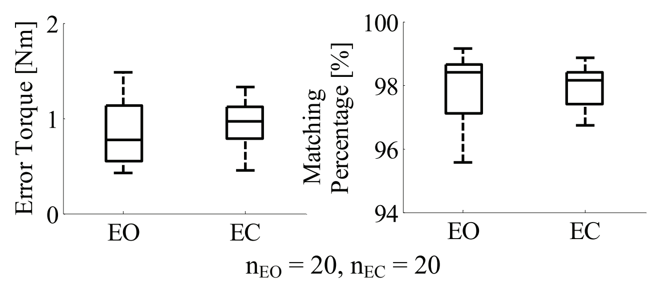

Figure 4: Group values of the overall error torque and matching percentage (medians, quartiles, and extrema).

DISCUSSION

The most significant result obtained from the torque matching procedure is that a PD controller with particular gains is, in fact, capable of successfully matching the actual ankle torque of healthy individuals during quiet standing. This is true even when considering that the PD controller has to compensate for a sensory-motor time delay of 80 + τ_E ms, where τ_E is as large as or even larger than 100 ms (not shown). Recall that τ_E depended on the frequency components of the neural input and the contraction time T of the plantar flexors.

It should also be noted that the final Kp, Kd, and T values did not only include the initial values that were based on previous studies [7,14], but are also in agreement with other studies that have been performed [15,16]. This primarily shows that the PD gains were optimized against a muscle contraction time T that is physiologically reasonable.

Taking our previous findings into account, we conclude that a PD control strategy is a legitimate model for the strategy that the CNS of healthy individuals applies in order to regulate balance during quiet standing. The future goal is to implement this control strategy in combination with an FES system to help individuals with SCI to perform ADL while standing.

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