Controlling Balance During Quiet Standing: Proportional and Derivative Controller Generates Preceding Motor Command to Body Sway Position Observed in Experiments

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Abstract

To compensate for significant time delays in the control of human bipedal stance, it was suggested that a feed-forward control mechanism is needed to generate a preceding motor command to the body sway position observed in quiet standing. In this article, we present evidence that a feedback proportional-derivative (PD) controller can effectively generate a desired preceding motor command. We also discuss the following characteristics of the proposed PD controller: 1) the level of robustness of the controller with respect to neurological time delays, and 2) how well the controller replicates the system's dynamics observed in experiments with able bodied subjects, i.e. how well the controller generates the observed preceding motor command. Human quiet stance was simulated using an inverted pendulum model regulated by a PD controller. The simulations were used to calculate the center of mass (COM) position and velocity data, and the motor command (ankle joint torque) data as a function of time. These data and the data obtained in the experiments were compared using cross-correlation functions (CCFs). The results presented herein imply that a PD feedback controller is capable of ensuring balance during human bipedal quiet stance, even if the neurological time delays are considerable. The proposed feedback controller can generate the preceding motor command that was observed in the experiments. Therefore, we conclude that a feed-forward mechanism is not necessary to compensate for the long closed-loop time delay in human bipedal stance as suggested in recent literature, and that the PD controller is a good approximation of the control strategy applied by able bodied subjects during quiet stance.

Key Words: Equilibrium Control, Posturography, Cross Correlation Analysis, Simulation, Feedback Control, PD Control

Introduction

Human bipedal stance is inherently unstable since it requires that a large body consisting of multiple flexible segments is kept in an erect posture with its center of mass (COM) located high above a relatively small base of support. The complexity of this system and its ability to maintain stable stance, despite various perturbations, have attracted the attention of many researchers in the field and have inspired various theories that try to explain the control mechanism of bipedal quiet stance. However, the true nature of this control mechanism is still an object of discussion and controversy.

The ankle joint torque needed to control the body during quiet stance can be evoked actively and passively. *Passive torque* components are the result of the intrinsic mechanical property, i.e. stiffness and/or viscosity, produced by muscle and surrounding tissue, such as ligaments and tendons. We can refer to the additional torque as *active torque*, which is generated by active muscle contraction. Since the COM is located in front of the ankle joint, plantar flexing torque is continuously required to prevent the body from falling forward [1]. However, the passive torque by itself is not sufficient to ensure this required plantar flexing torque [2, 3, 4]. Therefore, an additional active torque, regulated by the central nervous system (CNS) and produced by the plantar flexors, is needed [2, 3, 4, 5, 6, 7, 8].

Gatev et al. [5] reported that there is a significant statistical correlation between lateral gastrocnemius muscle activity and the position of spontaneous body sway, which was measured as the COM position. This finding supports the notion that the active torque is provided by lateral gastrocnemius muscle contractions in response to body sway. They also discovered that the muscle contractions preceded changes in the COM position by approximately 200 ms. Since the motor command appears to be generated in anticipation of future positions of the COM, these findings were later used to suggest that a feed-forward control mechanism is responsible for ensuring stable balance during quiet standing. Morasso and Schieppati [2] supported the feed-forward control theory and suggested that, in order to compensate for the long neuro-transmission delay, this control mechanism is required to generate the preceding motor command.

Similar to the study by Gatev et al. [5], Masani et al. [8] also found preceding muscle activities in plantar flexors during standing. However, they demonstrated that this preceding motor command could be accomplished by applying an appropriate feedback control system. In this context, a high gain PD (proportional-derivative) controller that uses the position and the velocity information of the COM was shown to be an effective method to regulate balance during standing even when long neurotransmission delays are present [8]. In their study two PD controllers were compared; one with a high derivative gain and one with a low derivative gain. Although both controllers could stabilize the body during quiet standing with a closed-loop time delay of 100 ms, only the PD controller with the high derivative gain was able to generate a long preceding motor command (121 ms) similar to the one obtained in experiments with able bodied subjects (155 ms). In addition to the preceding time, they discovered that the shape of the cross-correlation function (CCF) between the COM position and the muscle activity was similar to the CCF between the COM position and the joint torque obtained in simulations using the high derivative gain controller. These findings suggested that the control mechanism, which is responsible for the active torque generation, adopts a control strategy that relies notably on velocity information. However, to explain the experimental results in their study, only two PD controllers with arbitrarily selected gains were compared. Since the results were favorable, we realized that a systematic investigation was needed to determine the true capability of a PD controller that regulates balance during quiet stance with a preceding motor command observed in the experiments.

The purpose of the present study was to carefully examine various proportional and derivative gain pairs in simulations and provide answers to the following questions: 1) Is the PD controller capable of facilitating robust balance during quiet standing, despite long neurological time delays? and 2) Can the PD controller generate the system behavior observed in the experiments, including the long preceding motor command? By answering these questions, we have provided strong evidence that the feedback mechanism is capable of effectively regulating balance during quiet stance in the manner observed in experiments with able bodied subjects.

A preliminary report pertaining to this study was published as an abstract in [9].

Materials and Methods

Experiments

To identify appropriate PD controller gains, we compared the CCFs between the body kinematics (COM position and velocity) and the motor command of the modeled system with the body kinematics and muscle activity obtained experimentally. As a criterion of comparison, we determined whether the lag times of the model's CCFs were in the range of the lag times of the physiological CCFs obtained in the previous study [8]. Here, we briefly present the results of the previous study [8] which was used to evaluate the PD controller in this article.

Sixteen healthy men (mean±SD age, 23.8±3.9 years; mean±SD height, 169±6.6 cm) participated in the study. Each subject was requested to keep a quiet stance posture for 30 s in five trials, standing barefoot with eyes closed. The horizontal position of the waist point was measured as an approximation for the COM position in the anteroposterior direction using a laser displacement sensor. We adopted this approach since it was confirmed that the dynamics of quiet standing can be approximated by an inverted pendulum rotating around the ankle joint [5, 10]. However, we should note that this approximation might result in an relatively small error in the measurement. Electromyograms were recorded from the right plantar flexors. In this study, we examined the behavior of the medial gastrocnemius muscle, which showed the highest correlation with the body sway compared to other plantar flexors. The rectified and smoothed (4th-ordered, zero-phase-lag Butterworth low-pass filter with cutoff frequency of 4 Hz) electromyogram (EMG) was considered to represent the level of muscle activity.

Next, using the experimental results, we calculated two CCFs: 1) CCF between COM position and EMG; and 2) CCF between COM velocity and EMG. This allowed us to determine two objective time shift ranges for the CCF comparison. The time shift from COM position to EMG is defined as the lag time of the peak of the CCF between COM position and EMG. The time shift for each subject was calculated as the average of five trials, and the group average value $\pm$ standard deviation of the time shift was -155 ± 46 ms. It should be noted that the negative value indicates that the EMG precedes the COM position. Therefore, we defined this time shift range as the objective range for the CCF between COM position and the motor command in the simulations, which is referred to as TSpos.

The CCF between COM velocity and EMG had two peaks; one with a positive time shift, and the other one with a negative time shift. According to the experimental results, the positive time shift was 121 ± 134 ms and the negative time shift was -620 ± 134 ms. Therefore, we defined these two time shift ranges as the objective ranges for the CCF between COM velocity and the motor command in the simulations, which is referred to as TSvel.

The comparison of the time shifts obtained in simulations with TSpos and TSvel finally allowed us to identify PD control gain pairs for which the simulated system has the same performance as the actual physiological control system.

Model

Fig. 1 shows the schematic representation of the used model in which the plant/body is regulated by a PD controller. While the model was adopted from Masani et al. [8], the components of the neurological time delay (system or closed-loop time delay) were chosen according to recent findings available in literature as discussed below. The body dynamics and kinematics during quiet stance were described using an inverted pendulum model with the parameters of a typical adult male as found in [8] ($m = 76\ kg$, $I = 66\ kgm^2$ and $h = 0.87\ m$). The input to the body model, the command torque Tc, was the total torque exerted about the ankle joint.

For the value of the feedback time delay (τ_F) that represented cumulative time loss due to neural-transmission from the ankle somatosensory system to the brain, we selected 40 ms. This time delay represents the latency recorded from the instant the sensory stimulation is provided to the foot, to the instant the sensory evoked potential is recorded in the somatosensory area I of the brain. In literature, this time delay was reported to be in the range of 35.1 to 40.1 ms [11]. The electromechanical response time (τ_E), which represents the time difference between the moment when the EMG signal is generated to the moment when the force occurs, was set to 10 ms. This constant was chosen according to a measured value of 10.54 ms, as suggested by Isabelle et al. [12], and of 11.5 ms, as suggested by Winter and Brooks [13].

The motor command time delay (τ_M), 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, was introduced as a variable. This was done because the exact value of the motor command time delay, i.e. the time needed for the sensory-motor information process in the CNS, is not known.

The PD controller that was used to simulate the regulation of balance performed by the CNS was defined by the proportional and derivative gains, Kp and Kd, respectively. The motor command (Mc) was calculated using the COM position and velocity information according to the following equation:

$$\mathrm{Mc}(t) = -\mathrm{Kp}\theta(t - \tau_F - \tau_M) - \mathrm{Kd}\dot{\theta}(t - \tau_F - \tau_M) \tag{1}$$

where θ is the inclination angle of the inverted pendulum with respect to the vertical axis and $\dot{\theta}$ the time derivative of θ (both θ and $\dot{\theta}$ are positive in the forward direction). In the model, Mc was assumed to correspond to the EMG activity of the medial gastrocnemius muscle.

Please note that the Kp and Kd gains do not correspond to mechanical stiffness and viscosity of the ankle joint. They are gains used by the CNS to calculate Mc based on the body angle

(θ) and the rate of change of the angle with respect to time, i.e. the angular velocity ($\dot{\theta}$). Hence, in our simulations we decided to ignore the passive torque that contributed to balance control, and assumed that the balance was regulated by active torque alone. This was a less favorable condition compared to the actual system. Our reasoning for using solely an active torque component was that if the modeled system was able to compensate for disturbances with this restriction, it would be much easier to cope with the perturbations if the model was assisted by passive torque components that were present in the system as reported by [3, 4]. Therefore, by using only an active ankle joint torque component, we were able to investigate the capacity of the proposed controller that compensates for disturbances in the worst case scenario.

The analysis was carried out by assigning the following values for the variables Kp, Kd, and τ_M:

- Kp values from 50 to 3000 $N \cdot m \cdot rad^{-1}$ (step size of 50)
- Kd values from 50 to 2000 $N \cdot m \cdot s \cdot rad^{-1}$ (step size of 50)
- τ_M values from 25 to 215 ms (step size of 10)

Consequently, the considered closed-loop time delay went from 75 to 265 ms, which represents the sum of τ_M, τ_F (40 ms), and τ_E (10 ms). The total number of tested variable sets was 48000. All calculations were performed using the Matlab software, version 6.5 (MathWorks Inc, USA).

Robust Space

The robust space is a set of Kp-Kd-τ_M combinations for which the proposed PD controller is stable and robust in the sense of linear control theory, which is defined as follows:

1) The system is stable according to Nyquist's stability criterion [14].

2) The Kp-Kd-τ_M sets that satisfied the Nyquist's stability criterion were tested for robustness. This was necessary, since the inverted pendulum model represented a simplified description of the body during quiet standing, and parameters, such as the COM location and body segment lengths, are always modeled with a certain degree of inaccuracy. Therefore, Kp-Kd-τ_M sets that do not produce an open-loop frequency response with a phase margin of at least 20 degree, as well as a gain margin of -1 dB, were not considered as sufficiently robust. These choices for the minimum phase and gain margin were used extensively in the analysis of feedback systems, where the reference signal is constant, and the dynamical behavior of the system is dominantly defined by a noise signal between the controller and the plant [14].

The Kp-Kd-τ_M sets that met the conditions of both applied criteria constituted the robust space.

Objective Space

The objective space is a subset of the Kp-Kd-τ_M sets describing the robust space, for which the modeled systems had the response that was also observed in experiments with able-bodied subjects.

We carried out simulation studies in order to obtain CCFs for the Kp-Kd-τ_M sets that defined the robust space. In the simulations, we introduced a random disturbance torque (Td) to the ankle joint, which corresponded to the summation of all internal noise inducing spontaneous body sway. Td was produced as a low-pass filtered, uniform random number with zero mean and unity variance. The random number was generated with a sample time of 0.1 s, and was filtered by a first-order filter with a cutoff frequency of 5 Hz. The maximum amplitude of Td was about ±2.0 Nm. This noise had similar amplitude and frequency components as the one used for weak external perturbations in the study by Fitzpatrick et al. [15]. Their experiments showed that the applied noise facilitated the natural spontaneous sway, but did not threaten the stability of the system. In our simulations, we used a fixed step size of 0.001 s, and the Dormand-Prince solver algorithm. The system's dynamics were simulated for 8.192 s, which equals 2^{13} data points. For each robust variable set, five simulations were executed. The average CCF between COM position and Mc, and the average CCF between COM velocity and Mc were calculated from the data obtained in simulations. The final step in the analysis was a comparison of the time shifts from COM position to Mc, and from COM velocity to Mc with TSpos and TSvel, respectively. Those Kp-Kd-τ_M sets for which the obtained time shifts were in the ranges TSpos and TSvel were used to define the Kp-Kd-τ_M 'objective space'.

Results

Robust Space

Fig. 2 shows the robust Kp-Kd-τ_M space. Seven hundred and twenty-nine different robust Kp-Kd-τ_M sets were obtained for a closed-loop time delay greater than 75 ms. As shown in the figure Kp, Kd, and τ_M ranged from 750 to 1950 $N{\cdot}m{\cdot}rad^{-1}$, from 250 to 900 $N{\cdot}m{\cdot}s{\cdot}rad^{-1}$, and from 25 to 135 ms, respectively. Further in the text, the units for Kp and Kd will be omitted. In general, smaller values for Kp were paired with larger values for Kd. Smaller Kp gains ensured stability when τ_M values were larger. The Kp/Kd ratio went from 1.00 to 5.57. Kp gains that were smaller than 750, and Kd gains that were smaller than 250 caused the system to become unstable.

The largest τ_M value for which the system was stable and robust was 135 ms. Kp and Kd gains which were able to stabilize the system with τ_M = 135 ms were 750 and 350, respectively. Further in the text, this gain pair was called '(Kp,Kd)=(750,350)'. As a consequence for the largest value for τ_M, the largest closed-loop time delay for which the system was robust appeared to be 185 ms, which was obtained as a sum of 135 ms for τ_M, 40 ms for τ_F, and 10 ms for τ_E.

Objective Space

Fig. 3 illustrates the time series of Td, COM position, and Mc for three different Kp-Kd sets with τ_M = 45 ms. Although all three Kp-Kd-τ_M sets lie in the robust space, the shapes of the respective signals were different. As expected, this indicates that different controllers generated different outputs, and that not all robust sets had similar outputs as the physiological controller. To identify the Kp-Kd-τ_M sets for which the system had a similar performance as the actual able bodied subject during quiet standing, CCF analysis was carried out.

Fig. 4A illustrates the CCFs of the COM position and the Mc for the time series in Fig. 3. The CCF peaks for (Kp,Kd)=(750,350) (thick line) and (Kp,Kd)=(850,550) (thin line) were in TSpos (shaded area), while the CCF peak for (Kp,Kd)=(1350,300) (dotted line) was not. We could also distinguish the former two CCFs. The (Kp,Kd)=(750,350) pair had a steeper left side slope and a flatter right side slope when compared to the (Kp,Kd)=(850,550) pair. The

CCF between COM velocity and Mc was able to evaluate these fine differences quantitatively. Fig. 4B illustrates the CCFs between COM velocity and Mc for the (Kp,Kd)=(750,350) and (Kp,Kd)=(850,550) pairs. The abovementioned difference of CCF shapes in Fig. 4A affected the lag time of the CCF peaks in Fig. 4B. Both of the positive peak lag times appeared in the positive TSvel (shaded area in the right semiplane), while the negative peak lag time for (Kp,Kd)=(850,550) pair was outside of the negative TSvel (shaded area in the left semiplane). Therefore, we could identify the (Kp,Kd)=(750,350) pair as the only realistic gain set among those three examples. This procedure was applied to construct the objective space.

Fig. 5 illustrates the objective space. The total number of Kp-Kd-τ_M sets was reduced from 729 that define the robust space to 89. The values for Kp, Kd, and τ_M ranged from 750 to 1150, from 300 to 550, and from 25 to 85 ms, respectively. The Kp/Kd ratio went from 1.45 (Kp,Kd)=(800,550) to 3.83 (Kp,Kd)=(1150,300), while the peak of the objective space, similar to the robust space, appeared to be around (Kp,Kd)=(750,350). The longest realistic τ_M was limited to 85 ms resulting in a maximum closed-loop time delay of 135 ms.

Fig. 6 shows the time shift from Mc to COM position, i.e. the preceding time of the motor command, as a function of the closed-loop time delay of the Kp-Kd-τ_M sets in the objective space. The longest preceding time, which was obtained with the closed-loop time of 75 ms using the (Kp,Kd)=(750,350) gain pair, was 168 ms.

Discussion

Can PD controllers stabilize the system?

In literature, it has been suggested that a PD controller can theoretically stabilize an inverted pendulum of human body size when the system's closed-loop time delay is not too long [16]. In an extreme case where the closed-loop time delay is zero, the PD controller acts as a regulator of mechanical stiffness and viscosity [16]. It has also been pointed out that a long closed-loop time delay in the neural control system can destabilize the intrinsically unstable human bipedal stance: Morasso and Schieppati [2] applied a PD controller to regulate the balance of an inverted pendulum in simulations and discovered that a closed-loop time delay of about 50 ms was sufficient to destabilize the system. However, our results proposed a very different situation. As shown in the section Results, a large number of Kp and Kd gain pair sets were found that could facilitate stable standing even with closed-loop time delays longer than 75 ms (Fig. 2). The longest closed-loop time delay that still allowed the system to maintain a stable and robust behavior was 185 ms (gain pair (Kp,Kd)=(750,350)). The difference in results can be easily explained by the fact that the Kp/Kd ratio in the study by Morasso and Schieppati [2] was approximately 15. This implies that their PD controller was strongly position-based, ignoring the benefits of using velocity information. When the gain pairs in Fig. 2 are scrutinized, it is observed that the Kp/Kd ratio of our robust space is between 1.00 and 5.57, which is much smaller than the ratio in Morasso and Schieppati [2]. This indicates that a relatively higher derivative gain is required to facilitate stable standing when long closed-loop time delays are present.

Several studies have suggested that the control mechanism of quiet stance primarily depends on a sensory-motor integration that occurs in the higher neural centers [17] and that the primary source of sensory information are proprioceptors of the leg and foot [18]. If this distribution/arrangement of the sensory and control components of the system is adopted, one is faced with a much longer closed-loop time delay compared to the situation where balance is regulated by simple reflexes. As discussed in the section Model, to estimate the closed-loop time delay we divided it into three components: 1) feedback time delay (τ_F), 2) electromechanical response time (τ_E), and 3) motor command time delay (τ_M). In literature, it has been suggested that the realistic estimates for τ_F and τ_E are 40 ms and 10 ms, respectively (as discussed in the section Model). If we assumed that $\tau_F = 40$ ms and $\tau_E = 10$ ms, our analysis suggests that the maximum value for τ_M was 135 ms. In literature, it was suggested that the neural-transmission time between the cortex and soleus muscle during quiet standing was between 27 and 36.5 ms [19, 20]. Therefore, if one assumed that the neural-transmission time was approximately 30 ms (part of the motor command time delay (τ_M)), the amount of time left for the CNS to calculate and integrate the motor command was approximately 100 ms. We believe that 100 ms may be a reasonable amount of time for the CNS to perform the necessary signal processing and the motor command generation-integration process.

In summary, the PD controller can facilitate stable quiet standing despite significant physiological time delays.

Can PD controllers generate a preceding motor command?

All Kp-Kd-τ_M sets in the objective space are capable of providing the experimentally observed preceding motor command of 155 ± 46 ms (Fig. 6). This result strongly suggests that a feedback mechanism can generate a sufficient preceding motor command despite a long closed-loop time delay.

Several authors have proposed a feed-forward control mechanism for quiet stance regulation since a preceding motor command was observed that appears to anticipate the body sway position in order to compensate for the long transmission time delay in the neural circuit [2, 5]. However, our results indicate that a simple linear feedback mechanism is able to compensate for the long closed-loop time delay and to generate a preceding motor command. Recently, Peterka [21] carried out the system identification analysis and concluded that a PID (proportional-integral-derivative) feedback mechanism can account for postural control behavior. Our results essentially confirm Peterka's findings. However, we proposed a controller that did not have an integral gain. The reason for that decision was the fact that during quiet standing COM and COP constantly oscillated about the set point. This behavior is common for PD controllers, while PID controllers have a tendency to eliminate/attenuate these oscillations. Therefore, it is not surprising that the integral gain (Ki) in Peterka's study had a relatively lower value compared to the proportional and derivative gains (Kp and Kd) [21].

We discussed the reason why the system appears to have an anticipatory behavior as follows. If we do not consider the closed-loop time delay, the output of the derivative channel will lead the input signal (body position) by a phase of $\pi/2$. Since the frequency of the body oscillation was reported to be around 0.57 Hz (Winter et al. 1998), the phase lead corresponds to a time of about 440 ms. Because the output of the proportional channel coincides with the input signal (body position) without phase lead, the summation of both channels, i.e. the controller output, would be less than the preceding time of the derivative channel of 440 ms. As a result, considering the closed-loop time delay, the preceding time of the controller output could be within the range of the present result of around 150 ms. Therefore, the phase lead of the derivative channel is the decisive factor for the anticipatory behavior. In this context, we are sure that the phase lead of the controller output will become larger when the derivative gain is being increased.

We would like to draw attention to the fact that, in our simulations, we neglected the contribution of passive stiffness and damping which are typically present in this system [4]. This was done on purpose, because our intention was to determine the capacity of the CNS-PD controller that copes with perturbations and a long closed-loop time delay. Our simulations suggest that the PD controller alone is capable of regulating balance during quiet standing in the same manner as able bodied subjects do, for the closed-loop time delays up to 135 ms. Consequently, if the passive components were added to our model, the simulations would show that the system is able to cope with even longer closed-loop time delays than 135 ms. In his study, Peterka [21], through a system identification approach, calculated optimal Kp, Kd, and Ki gains of the PID controller. Since the gain values depend on the subject's body size and since the time shift may be affected by the noise properties, it might be erroneous to compare the results in his study with the present results. However, we venture to do this here to consider the effect of the passive torque component. According to his estimate, Kp, Kd and Ki gains were equal to 856 $N \cdot m \cdot rad^{-1}$, 303 $N \cdot m \cdot s \cdot rad^{-1}$, and 115 $N \cdot m \cdot rad^{-1} \cdot s^{-1}$, respectively. Please note that the (Kp,Kd) = (856,303) pair belonged to our objective space, in spite of the following methodological differences: 1) Peterka introduced a Ki gain, 2) he used a larger perturbation stimulus, and 3) he included the passive torque component. However, in our study, Peterka's gains (Kp,Kd) = (856,303) were able to cope with a closed-loop time delay up to 125 ms, while in his study, the system was stable for a closed-loop time delay of up to 200 ms. This result further strengthens our hypothesis that the inclusion of passive stiffness and damping, as performed by Peterka, would allow for larger closed-loop time delays, i.e. a longer time for the CNS to perform the necessary signal processing and the motor command generation-integration process.

The controllers in the objective space emphasize the importance of the information provided by the body's velocity, in addition to the position information. The Kp/Kd ratio of the controllers in the objective space was in the range of 1.45 to 3.83. Thus, the CNS adopts a control strategy that relies considerably on the velocity information to compensate for a relatively long closed-loop time delay. Our findings support several previous studies that emphasized the importance of the body velocity information in the control system of quiet standing. For example, a strong coupling of motor and sensory information on body velocity via visual [22, 23] and tactile [24] sensation was reported. Jeka et al. [24] reported that the information on the body's position, as well as its velocity via haptic sensation of the fingertip, showed a strong coupling to sway control. Also, Morasso and Schieppati [2] suggested the advantage of obtaining both position and velocity parameters for the purpose of balance control.

It should be noted that we are not questioning the potential existence of a control strategy that applies a feed-forward mechanism. Our intention is to demonstrate that another viable control solution to this problem exists, which in our opinion should be easier to implement in the given physiological framework compared to the feed-forward control strategy. Our results strongly suggest that a feedback mechanism using a PD controller is capable of controlling human quiet stance even for longer closed-loop time delays. Furthermore, the feedback mechanism can generate the observed preceding motor command. The final point is that the proposed PD controller is robust and has a large space of Kp and Kd gains for which the system is stable and behaves in the same manner as able bodied subjects during quiet standing. This is an important feature, because it suggests that significant variations in the gains will still produce the same system response/performance, which is an inherent characteristic of many biological systems.

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Figure Legends

Fig. 1 An inverted pendulum model and closed-loop control scheme of quiet stance, where y is the COM position, θ is the sway angle, g is the acceleration of gravity, Tc is the total torque about the ankle, and h is the distance of the COM to the ankle.

Fig. 2 The robust space: 3D visualization (A), Kp-τ_M projection (B), Kd-τ_M projection (C), and Kp-Kd projection (D). It should be noted that the space consists of dots that were interpolated to allow the volume visualization.

Fig. 3 An example of Td and corresponding example time series of Mc and COM position, generated by three different PD controllers: Kp1350-Kd300, Kp850-Kd550, and Kp750-Kd350. All three simulations were performed with $\tau_M = 45$ ms.

Fig. 4 CCFs between COM position and Mc (A) and between COM velocity and Mc (B) for the example time series in Fig. 3. (A) shows three CCFs generated by Kp750-Kd350 (thick line), Kp850-Kd550 (thin line), and Kp1350-Kd300 (dashed line). (B) shows two CCFs generated by Kp750-Kd350 (thick line) and Kp850-Kd550 (thin line). The shaded area in (A) is the objective time shift range for the CCF between COM position and Mc (TSpos), and the shaded areas in (B) are the objective time shift ranges for the CCF between COM velocity and Mc (TSvel). The peaks of CCFs are indicated using short bars.

Fig. 5 The objective space: 3D visualization (A), Kp-τ_M projection (B), Kd-τ_M projection (C), and Kp-Kd projection (D). It should be noted that the space consists of dots that were interpolated to allow the volume visualization.

Fig. 6 Time shift from Mc to COM position, i.e. the preceding time of the motor command, as a function of the closed-loop time delay of the Kp-Kd-τ_M sets in the objective space. The interval between the dashed lines indicates the time shift range obtained experimentally (TSpos).

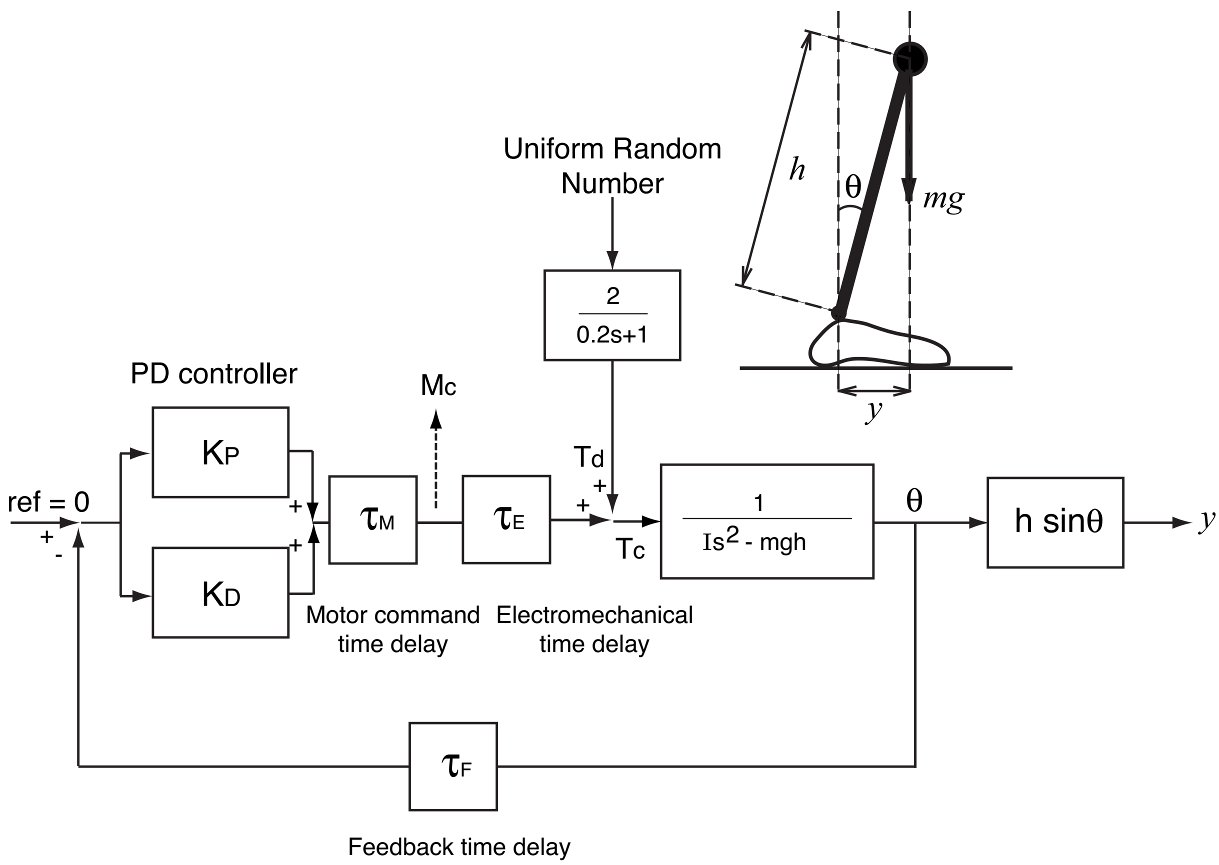
Uniform Random Number
2
0.2s+1
h
θ
mg
y
PD controller
KP
KD
ref = 0
+
-
+
+
τM
Mc
τE
Td
+
+
Tc
1
Is2 - mgh
θ
h sinθ
y
Motor command time delay
Electromechanical time delay
τF
Feedback time delay

Figure 1:

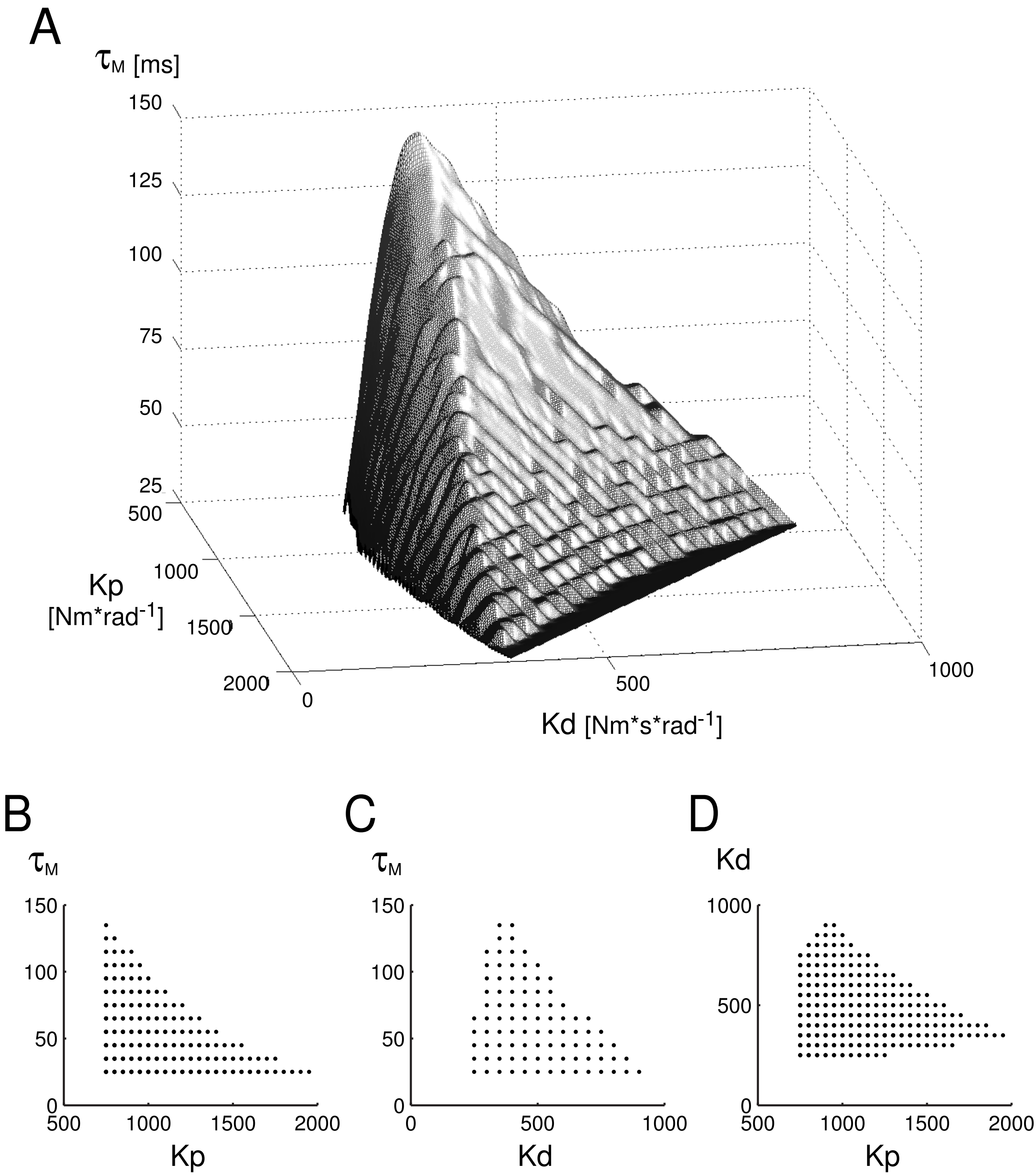
A
τM [ms]
150
125
100
75
50
25
500
1000
1500
2000
Kp
[Nm*rad-1]
0
500
1000
Kd [Nm*s*rad-1]
B
τM
150
100
50
0
500
1000
1500
2000
Kp
C
τM
150
100
50
0
0
500
1000
Kd
D
Kd
1000
500
0
500
1000
1500
2000
Kp

Figure 2:

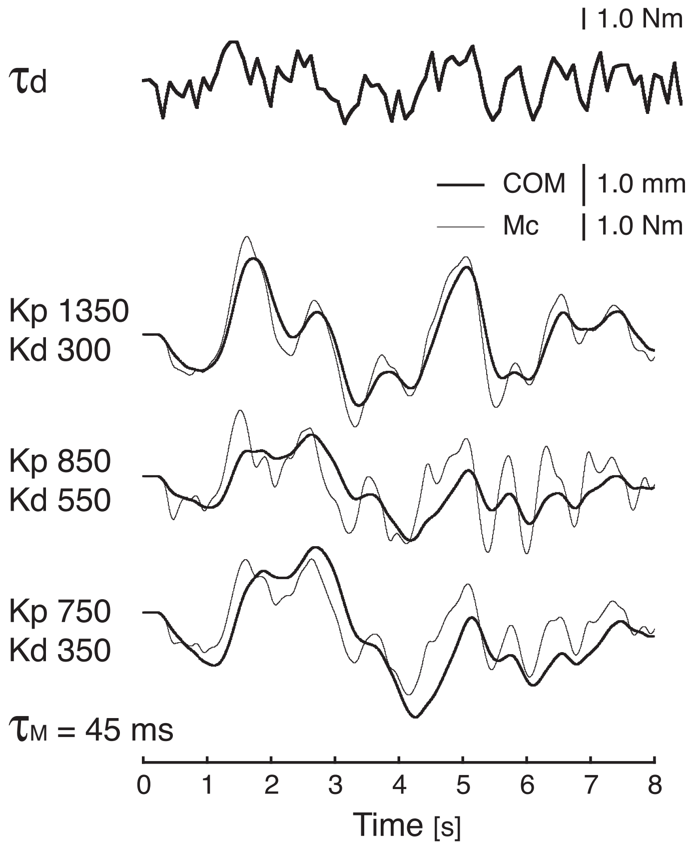
1.0 Nm
τd
COM 1.0 mm
Mc 1.0 Nm
Kp 1350
Kd 300
Kp 850
Kd 550
Kp 750
Kd 350
τM = 45 ms
0 1 2 3 4 5 6 7 8
Time [s]

Figure 3:

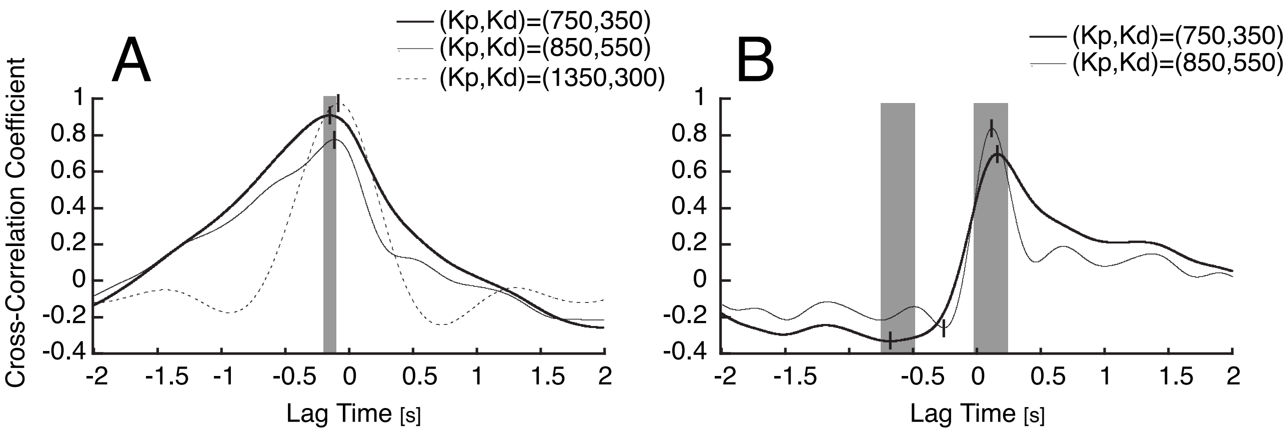
A
B
(Kp,Kd)=(750,350)
(Kp,Kd)=(850,550)
(Kp,Kd)=(1350,300)
Cross-Correlation Coefficient
Lag Time [s]

Figure 4:

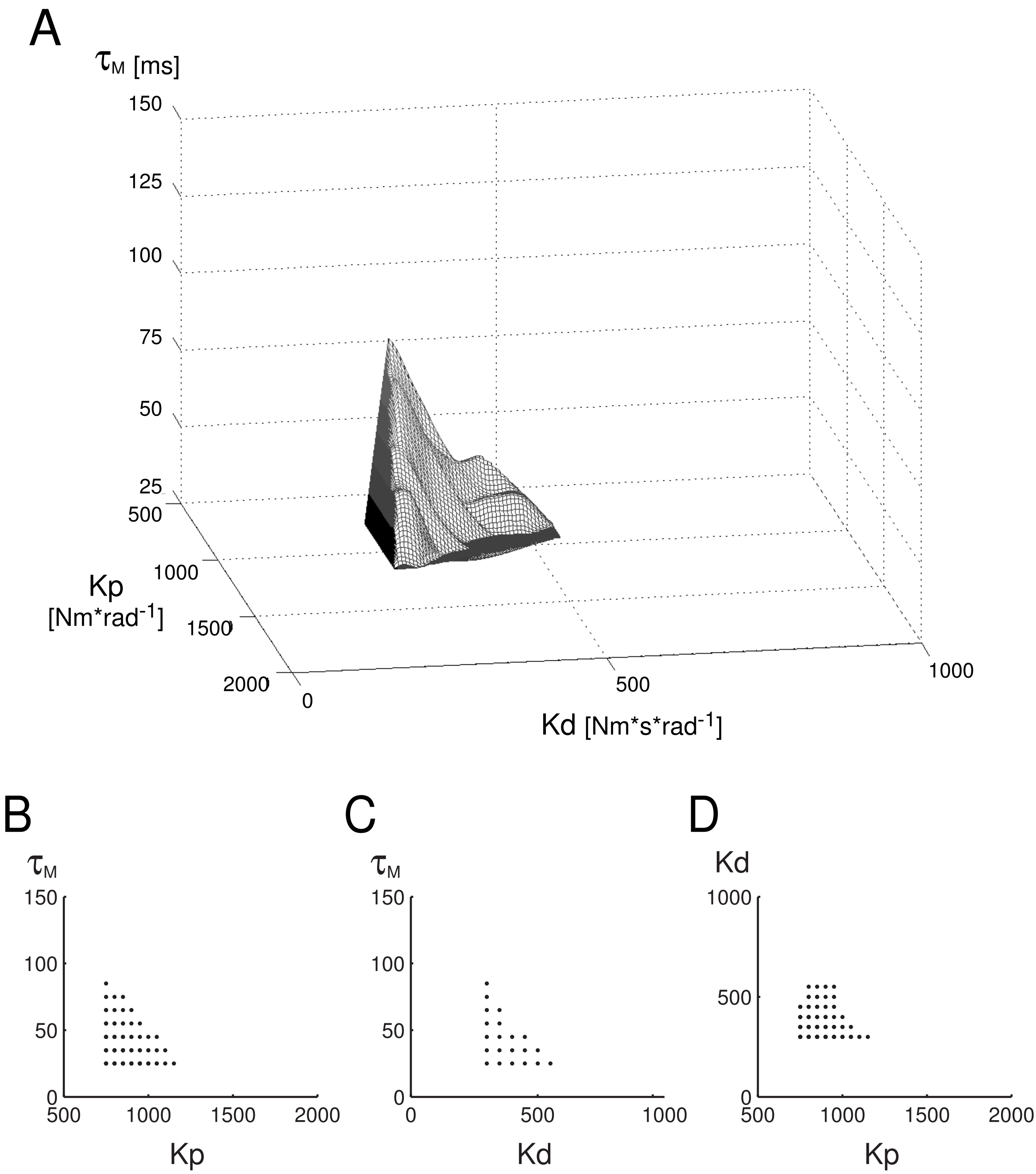
A
τ_M [ms]
150
125
100
75
50
25
500
1000
1500
2000
Kp
[Nm*rad-1]
0
500
1000
Kd [Nm*s*rad-1]
B
τ_M
150
100
50
0
500
1000
1500
2000
Kp
C
τ_M
150
100
50
0
0
500
1000
Kd
D
Kd
1000
500
0
500
1000
1500
2000
Kp

Figure 5:

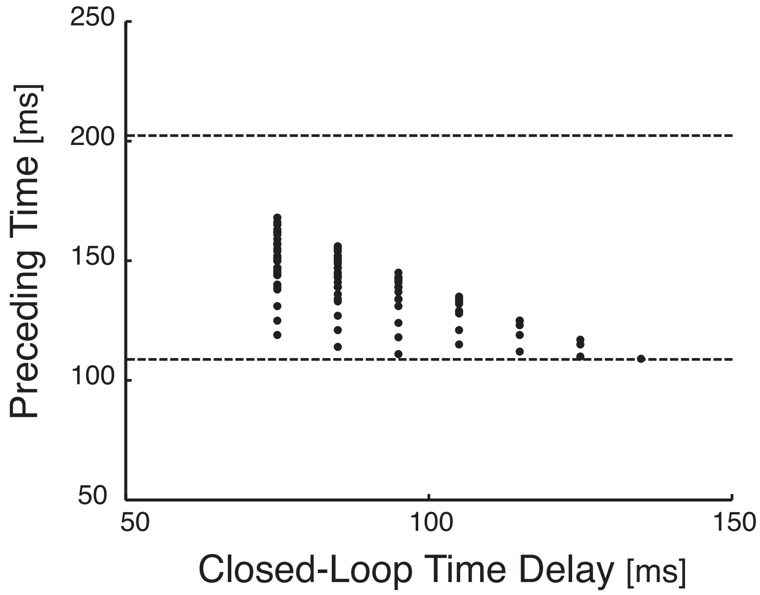
250
200
150
100
50
50
100
150
Preceding Time [ms]
Closed-Loop Time Delay [ms]

Figure 6: