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Effects of external loads on balance control during upright stance: Experimental results and model-based predictions

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ABSTRACT

The purpose of this study was to identify the effects of external loads on balance control during upright stance, and to examine the ability of a new balance control model to predict these effects. External loads were applied to 12 young, healthy participants, and effects on balance control were characterized by center-of-pressure (COP) based measures. Several loading conditions were studied, involving combinations of load mass (10% and 20% of individual body mass) and height (at or 15% of stature above the whole-body COM). A balance control model based on an optimal control strategy was used to predict COP time series. It was assumed that a given individual would adopt the same neural optimal control mechanisms, identified in a no-load condition, under diverse external loading conditions. With the application of external loads, COP mean velocity in the anterior–posterior direction and RMS distance in the medial–lateral direction increased 8.1% and 10.4%, respectively. Predicted COP mean velocity and RMS distance in the anterior–posterior direction also increased with external loading, by 11.1% and 2.9%, respectively. Both experimental COP data and model-based predictions provided the same general conclusion, that application of larger external loads and loads more superior to the whole body center of mass lead to less effective postural control and perhaps a greater risk of loss of balance or falls. Thus, it can be concluded that the assumption about consistency in control mechanisms was partially supported, and it is the mechanical changes induced by external loads that primarily affect balance control.

1. Introduction

Falls are one of the most common incidents leading to injuries in daily activities and occupational settings. Fall-related injuries have substantial adverse impacts on functional ability and life quality. As unintentional falls often result from a ‘loss-of-balance’, an improved understanding of balance control may thus aid in understanding and preventing falls. A number of factors have been identified as influencing balance control, such as aging [1–3], localized muscle fatigue [4,5], and decrements in the quality of sensory input [6].

External loads also appear to affect balance control, and many daily and occupational activities require load carriage (e.g. material handling and military marches [7]). Hence, further investigation of how and why balance control is affected by external loads is warranted. Previous studies have suggested that external loads adversely affect balance control, since such loads resulted in increased postural sway during quiet erect stance [7–10]. Increased postural sway indicates that the whole-body center-of-mass (COM) gets closer to the limits of the base of support (BOS) and thus leads to less stability. Existing studies of external loads, however, have been somewhat narrowly focused on the effects of external load mass on balance control. The location of any external loads would also seem relevant; while this has been investigated with respect to energy costs [11], we are unaware of any evidence regarding whether load location affects balance control.

In addition to experimental studies, balance control models have been widely used to facilitate an understanding of underlying balance control mechanisms. For example, Ishida et al. [12] adopted a balance control model to identify the roles of different sensory systems, and Maurer and Peterka [13] applied a model based on a PID (proportional, integrative, and derivative) neural controller to examine the effects of aging on balance control. Evaluation of such mathematical models is often challenging. Some studies [13,14] have examined whether balance control models could accurately simulate experimental center-of-pressure (COP) based measures, and Winter et al. [15] compared simulated

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and experimental relationships between sway amplitude and effective stiffness of balance control.

We recently presented a balance control model based on an optimal control strategy to simulate spontaneous sway behaviors, and used this model to identify aging effects on balance control [16]. Our model was shown to generate reasonable simulations of measures derived from COP time series. Models are expected to be able to address "what if" questions [17], thus one can argue that a stronger test involves determining whether this (or related) model can make predictions of sway or COP behaviors under novel circumstances. In the current context, these circumstances are different configurations of an external load during quiet upright stance.

One purpose of this study was to determine the actual effects of external loads on balance control, and specifically the influences of load mass and height, by using the data obtained while participants were loaded. We hypothesized that increasing mass and/or height would challenge balance control, based on expected mechanical effects for an analogous inverted pendulum (as is frequently used to model upright stance). The effects of external loads on balance control were identified by COP-based measures, since the COP reflects the net motor control signal output necessary to keep the projection of the center-of-mass (COM) within the BOS [18,19]. Another purpose was to determine whether, or to what extent, our optimal control model could predict changes in balance control behaviors under novel conditions. Such 'predictive ability' was assessed using the same scenario involving application of external loads during quiet upright stance. This scenario was considered advantageous, since changes in mechanics due to loading could be estimated in a straightforward manner (see Section 2). Model predictions were based on an assumption that the same optimal control mechanisms, identified during unloaded conditions, are also employed during loaded conditions (see Section 2). Since experimental data were not used to generate behavioral simulations in the loaded conditions, we considered this a relatively strong test of the model's predictive ability.

2. Methods

2.1. Participants and experimental procedures

Twelve participants (five female and seven male) were involved, with mean (S.D.) age = 29(6) years, stature = 169.4(11.2) cm, and body mass = 61.9(10.6) kg. None had any current or recent self-reported injuries, illness, or musculoskeletal disorders. All participants completed an informed consent procedure approved by the Virginia Tech Institutional Review Board.

Each participant performed 15 trials involving quiet upright stance. In each trial, participants stood barefoot on a force platform (AMTI OR6-7-1000, Watertown, MA, USA), with eyes closed and arms at sides, and were requested to stand as still as possible. Trials lasted 90 s, with at least 2 min of rest between each. Triaxial ground reaction forces and moments were sampled at 60 Hz and used to derive COP time series. Noise likely exists in the obtained ground reaction force and moment signals due to the properties of analogue amplifiers in the force platform. This noise is usually at relatively high frequencies, so a low-pass filter (Butterworth, second order) was used. The cut-off frequency was set at 5 Hz, consistent with earlier approaches [20,21]. The initial 20 and last 10 s of each time series were discarded.

Effects of external loads were investigated by manipulating two aspects of the load, specifically the mass and height. This was achieved using two load packs, placed dorsally and ventrally (Fig. 1). Total load mass was set to 0 (no packs), 10%, and 20% of individual body mass, by inserting small metal cylinders into several pockets in the packs. These cylinders were placed symmetrically, to the extent possible, with respect to the frontal and sagittal planes. Load packs were placed at one of two heights, such that the COM of the external load was at height equal to the whole-body COM (=58.8% of stature [22]), or at 15% of stature above the whole-body COM. These are referred to as 'low' and 'high' conditions, respectively. Since the no-load condition was identical for both heights, there were five combinations of external load mass and height examined. After an initial familiarization period, participants performed three trials in each combination, with the order of combinations randomized to minimize confounding related to learning effects.

2.2. Model description

Our existing balance control model [16] was used to predict the effects of external loads under the conditions described above. In this model, the human body is represented as a simple single-segment inverted pendulum during quiet upright stance, and the neural controller was assumed to be an optimal controller. The optimal neural controller is defined by optimal feedback gains, which are calculated by solving the Riccati equation, and generates ankle control torques by minimizing a certain performance criterion. This performance criterion was defined by several physical quantities relevant to sway, including delayed sway angle and ankle control torque. Several model parameters, such as the weightings of relevant physical quantities, random disturbance gain, and sensory delay time, cannot be specified *a priori*. To determine the values of these unspecified model parameters, an

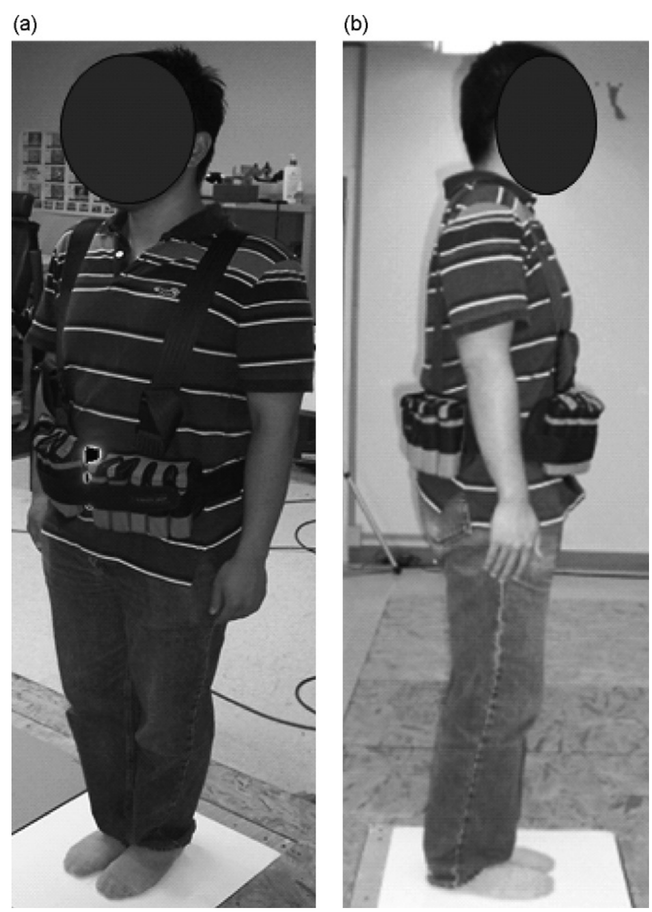

Fig. 1. Participant wearing two load packs (load height equal to whole-body COM).

optimization procedure was used with an objective defined as the scalar error between experimental and simulated COP-based measures. Specific terms in the scalar error were: mean distance (MD), root mean square distance (RMS), maximum distance (MAXD), mean velocity (MV), mean frequency (MFREQ), 50% power frequency (P50), 95% power frequency (P95), centroidal frequency (CFREQ), and frequency dispersion (FREQD). According to our earlier simulation results, this model was able to simulate COP-based measures, and helped to identify potential age-related changes in balance control.

2.3. Model-based predictions

Model-based predictions involved four sequential steps. First, several anthropometric measures were required: moment of inertia of the body about the ankle (I), body mass (M), height of whole-body COM (h), mass of the feet (m_F), height of the ankle (h_F), and anterior–posterior (A/P) distance between the ankle and the COM of the feet (d_F). Several measures were obtained directly from each participant prior to the experimental trials: M, stature (l) and foot length (l_F). The remaining measures were estimated from existing equations [22–24] as follows:

$$\begin{cases} I = M \times l^2 \times [0.3^2 + (0.588 - 0.039)^2]; \\ h = l \times (0.588 - 0.039); \\ m_F = M \times 0.0137 \times 2(\text{male}) \text{ or } = M \times 0.0129 \times 2(\text{female}); \\ h_F = l \times 0.039; \\ d_F = l_F \times 0.4415(\text{male}) \text{ or } = l_F \times 0.4014(\text{female}). \end{cases} \quad (1)$$

Second, unspecified parameters in the model were determined. Using procedures described in Qu et al. [16], COP-based measures from the no-load trials were used to determine these parameters for each participant.

Third, prior to using the model to predict the effects of external loads, anthropometric descriptors in the model were adjusted to account for the loads, using:

$$\begin{cases} M' = M + \Delta M \\ I' = I + \Delta M \times (h + \Delta h)^2 \\ h' = \dfrac{M \times h + \Delta M \times (h + \Delta h)}{M + \Delta M} \end{cases} \quad (2)$$

where M', I', and h' are the adjusted values, ΔM is the external load mass, and Δh is the external load height relative to whole-body COM. In the conditions with an external load, the five weights defined in the neural optimal controller, and the sensory time delay, were held at the values determined from the no-load trials. This was based on an assumption that a given individual would adopt the same neural optimal control mechanisms under diverse external loading conditions. In contrast, changes involving the application of external loads should affect external random disturbances, and we adjusted the random disturbance gain so that the ratio between the random disturbance gain and total mass (body + external loads) was held constant.

Fourth, the now fully specified model was used to simulate sway behaviors under the four loaded conditions (see [16] for a description of such simulation). Three sway trials with different initial random disturbance seeds were simulated for each loaded condition and each simulation trial was 90 s in duration. From the predicted COP time series, several measures were derived as described below. Note that the model employed is 2-D (sagittal plane), hence predicted COP measures are A/P only.

2.4. Dependent measures

Prieto et al. [21] systematically presented a set of COP-based measures, which in general can be classified into two groups: time-domain and frequency-domain. A number of measures in both groups have been used in investigations of balance control [25,26]. However, these measures do not account for dynamic characteristics of COP time series [27], for which measures derived from statistical mechanics have been proposed [28–30]. According to the classification of COP-based measures mentioned above, several measures were derived from the experimental and predicted COP time series, and used as dependent measures to characterize the effects of external loads on balance control. Most time-domain measures, including RMS, MD, and MAXD, appear to be highly correlated [24]. Among time-domain measures, MV has been found to be the most reliable [31]. Thus, among time-domain measures, we chose RMS and MV for analysis. The selected frequency domain measures were CFREQ and FREQD, with CFREQ reflecting central frequency tendency and FREQD the variability in frequency content. Among potential statistical mechanics approaches, one considering the trajectory of the COP as fractional Brownian motion (fBm) appears most relevant and has been widely used when analyzing upright postural control [32–36]. Thus, four statistical mechanics measures [35] derived from the fBm model were chosen (TT: transition time; TA: transition amplitude; H_S: short-term scaling exponent; H_L: long-term scaling exponent). Note that both TT and TA were defined by the transition point in the fBm model. Descriptions and units of dependent measures are given in Table 1.

2.5. Analysis

In order to compare sway behaviors across individuals, COP-based measures from the loaded conditions were normalized by the corresponding average measures in the no-load condition (Table 2). External load mass and load height, each at two levels, served as the two independent variables in subsequent analyses. Two-way analysis of variance (ANOVA) was performed to identify the effects of load mass and height on the normalized experimental COP-based measures. To compare loaded versus unloaded conditions, the normalized experimental COP-based measures under loaded conditions were compared with a value of unity using t-tests. The same statistical analyses were also performed for the model-predicted COP-based measures. A/P H_L and medial–lateral (M/L) H_L data from one participant were removed from these analyses, as they were more than 2.5 times the interquartile range away from the upper or lower quartile of the data, and thus considered outliers [37].

To evaluate the model's predictive ability, model-based predictions and experimental results were compared both qualitatively and quantitatively. Qualitative comparisons were based on changes in normalized COP-based measures versus external load mass and height. Quantitative comparisons of these measures under loaded conditions were done using unpaired t-tests. For all statistical test, significance was concluded when $p < 0.05$.

Table 1
Glossary of COP-based dependent measures

Acronym	Description
RMS (mm)	Root mean square distance
MV (mm/s)	Mean velocity
CFREQ (Hz)	Centroidal frequency
FREQD	Frequency dispersion
TT (s)	Transition time
TA (mm^2)	Transition amplitude
H_S	Short-term scaling exponent
H_L	Long-term scaling exponent

Table 2
COP-based measure means (S.D.) under the no load condition

	RMS	MV	CFREQ	FREQD	TT	TA	H_S	H_L
Experimental								
A/P	5.31(2.26)	7.64(2.29)	0.468(0.116)	0.890(0.068)	0.651(0.244)	14.3(7.0)	0.788(0.036)	0.156(0.101)
M/L	5.21(2.07)	9.30(3.00)	0.512(0.139)	0.842(0.065)	0.638(0.407)	24.0(3.1)	0.785(0.065)	0.117(0.108)
Predicted								
A/P	4.56(1.77)	7.46(2.18)	0.498(0.086)	0.915(0.041)	0.766(0.166)	13.3(7.7)	0.746(0.034)	0.164(0.093)

3. Results

Application of external loads led to significant changes in several COP-based measures (Table 3). Specifically, there were significant increases in A/P MV, A/P TA, M/L RMS, M/L TT, and M/L TA, and a significant decrease in M/L H_S, at both levels of external load mass. Several measures significantly decreased only at the higher load mass: A/P FREQD, M/L CFREQ, and M/L H_L. When external load mass changed from 10% to 20% of body mass, significant changes occurred in A/P MV, A/P TA, M/L FREQD, and M/L TA. In addition to these significant effects, there were two trends apparent, including an increase in A/P RMS at the higher load mass ($p = 0.074$), and a decrease in A/P FREQD at the lower load mass ($p = 0.069$).

Three A/P measures and one M/L measure were significantly affected by external load height (Table 3). Specifically, A/P RMS and M/L MV became significantly larger, and A/P CFREQ and A/P H_S became significantly smaller when the external load was raised above the COM. There were also two apparent trends in A/P TT ($p = 0.077$) and M/L RMS ($p = 0.089$), both of which increased at the higher load level. A significant mass × height interaction effect was found only for M/L MV. When external load mass changed from 10% to 20% of body mass, M/L MV decreased at the lower load level (from 0.974 to 0.952), but increased at the higher load level (from 0.992 to 1.068).

Except for A/P H_L, all remaining predicted measures were significantly affected by both levels of external load mass (Table 4). A/P RMS, A/P MV, A/P TT and A/P TA increased, while A/P CFREQ, A/P FREQD, and A/P H_S decreased, with the application of external loads. A significant increase in A/P H_L was also found with the higher load mass. None of the predicted measures was significantly affected by external load height (Table 4), however changes in A/P RMS ($p = 0.053$) and A/P CFREQ ($p = 0.058$) approached significance. There were no significant mass × height interaction effects on any predicted measures.

Table 3
Normalized experimental COP-based measure means (S.D.)

	External load = 10% body mass	External load = 20% body mass	Low external load level	High external load level	p-Value			
					10% vs. 20%	10% vs. 0%	20% vs. 0%	Low vs. high
A/P								
RMS	1.03(0.28)	1.05(0.30)	0.98(0.24)	1.10(0.32)	0.65	0.19	0.074	0.013*
MV	1.05(0.14)	1.11(0.14)	1.07(0.15)	1.09(0.13)	0.009*	0.001*	<0.001*	0.49
CFREQ	0.97(0.29)	1.02(0.32)	1.05(0.32)	0.94(0.27)	0.26	0.16	0.27	0.040*
FREQD	0.99(0.08)	0.98(0.09)	0.98(0.09)	0.99(0.08)	0.63	0.069	0.022*	0.39
TT	1.05(0.32)	1.04(0.30)	1.00(0.29)	1.09(0.33)	0.92	0.11	0.12	0.077
TA	1.21(0.46)	1.41(0.61)	1.24(0.54)	1.38(0.54)	0.031*	<0.001*	<0.001*	0.11
H_S	1.00(0.05)	1.01(0.05)	1.01(0.05)	0.99(0.04)	0.37	0.48	0.11	0.008*
H_L	1.05(0.62)	1.02(0.61)	1.01(0.59)	1.05(0.64)	0.59	0.28	0.42	0.50
M/L								
RMS	1.09(0.27)	1.12(0.26)	1.07(0.21)	1.14(0.31)	0.43	0.005*	<0.001*	0.089
MV	0.98(0.15)	1.01(0.16)	0.96(0.13)	1.03(0.17)	0.27	0.16	0.29	0.007*
CFREQ	0.97(0.24)	0.91(0.24)	0.93(0.26)	0.94(0.23)	0.13	0.14	<0.001*	0.73
FREQD	1.03(0.07)	1.00(0.08)	1.01(0.06)	1.02(0.08)	0.040*	<0.001*	0.39	0.60
TT	1.19(0.81)	1.28(0.72)	1.24(0.81)	1.22(0.72)	0.48	0.026*	<0.001*	0.87
TA	1.14(0.70)	1.44(0.86)	1.20(0.74)	1.38(0.85)	0.028*	0.043*	<0.001*	0.17
H_S	0.980(0.08)	0.98(0.08)	0.99(0.08)	0.98(0.08)	0.77	0.013*	0.031*	0.40
H_L	1.02(0.81)	0.82(0.88)	0.84(0.75)	1.00(0.93)	0.17	0.42	0.048*	0.30

* p-Value<0.05.

Table 4
Normalized predicted COP-based measure means (S.D.)

	External loads = 10% body mass	External loads = 20% body mass	Low external load level	High external load level	p-Value			
					10% vs. 20%	10% vs. 0%	20% vs. 0%	Low vs. high
A/P								
RMS	1.09(0.14)	1.13(0.19)	1.08(0.14)	1.14(0.19)	0.21	<0.001*	<0.001*	0.053
MV	1.02(0.04)	1.03(0.05)	1.03(0.05)	1.02(0.05)	0.15	<0.001*	<0.001*	0.27
CFREQ	0.87(0.19)	0.86(0.20)	0.89(0.19)	0.83(0.19)	0.73	<0.001*	<0.001*	0.058
FREQD	0.99(0.04)	0.99(0.03)	0.98(0.04)	0.99(0.04)	0.86	0.002*	<0.001*	0.47
TT	1.31(0.40)	1.42(0.69)	1.36(0.54)	1.36(0.59)	0.24	<0.001*	<0.001*	0.94
TA	1.37(0.55)	1.60(0.97)	1.47(0.87)	1.50(0.71)	0.086	<0.001*	<0.001*	0.83
H_S	0.97(0.06)	0.97(0.06)	0.96(0.06)	0.97(0.06)	0.91	<0.001*	<0.001*	0.68
H_L	1.13(0.71)	1.24(0.84)	1.09(0.86)	1.29(0.67)	0.40	0.058	0.008*	0.12

* p-Value < 0.05.

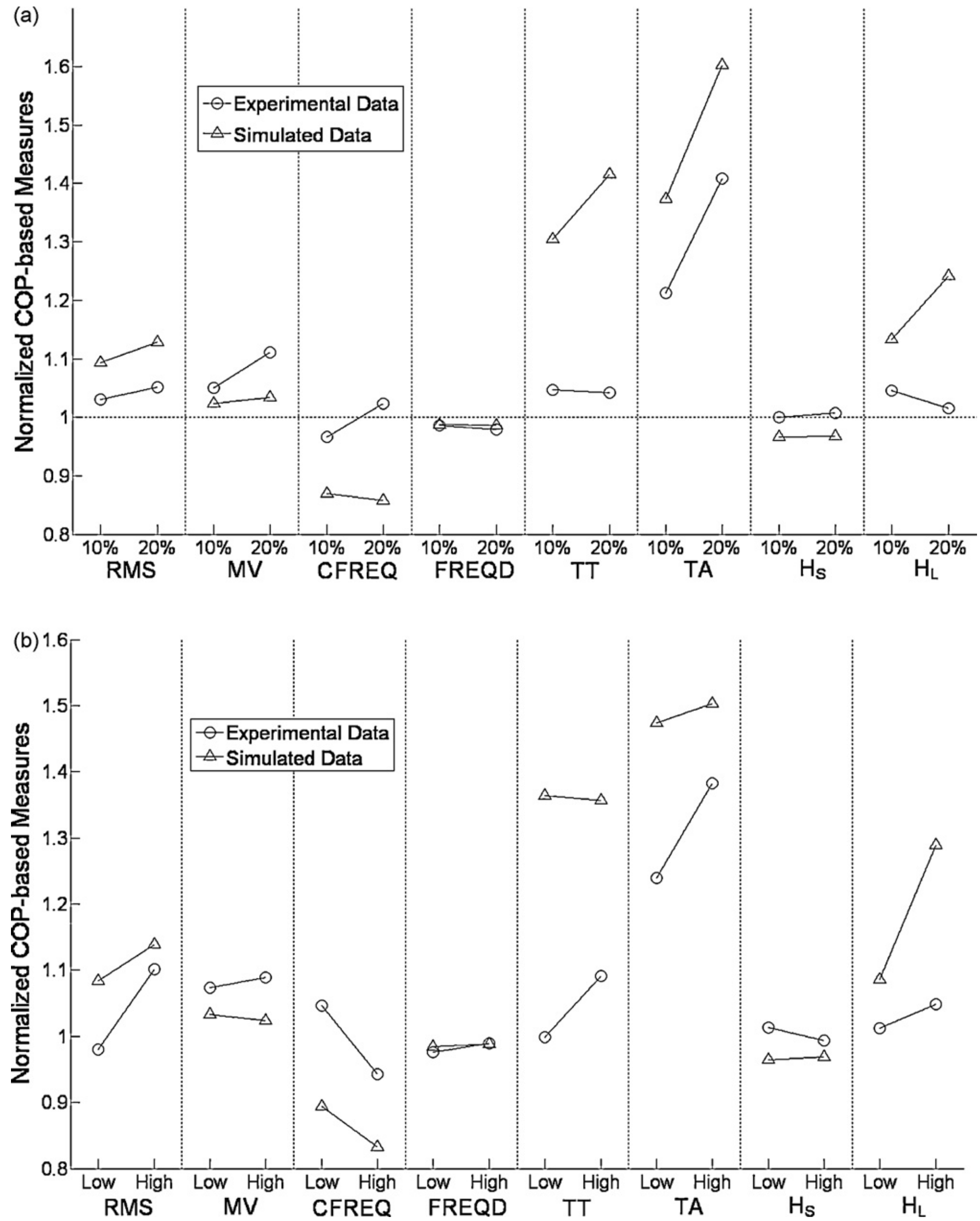

Fig. 2. (a) Average trends in experimental and predicted A/P COP-based measures vs. external load mass. (b) Average trends in experimental and predicted A/P COP-based measures vs. external load height. Standard deviation values of these COP-based measures are provided in Tables 3 and 4.

Comparison of the experimental and predicted dependent measures versus external loads (Fig. 2) indicated that most of the predicted trends were consistent with experimental findings. However, qualitative discrepancies between experimental and simulation results were found in the effects of load mass for A/P CFREQ, A/P TT, and A/P H_L, and in the effects of load height for A/P MV, A/P TT, and A/P H_S. Quantitative comparisons revealed, that except for A/P FREQD ($p = 0.626$) and A/P H_L ($p = 0.192$), significant differences between experimental and predicted outcomes existed in the remaining COP-based measures (Fig. 3).

4. Discussion

One purpose of this study was to identify whether balance control (as assessed indirectly using COP) was influenced by different external loading conditions. With the application of external loads, A/P MV and M/L RMS both increased (Table 3 and Fig. 2(a)). These findings are qualitatively consistent with previous studies [7,38], in which time-domain measures were shown to increase when carrying external loads. More quantitatively, Schiffman et al. [7] reported that 16-kg external loads increased A/P MV by 21%, while an 8.1% increase was found here across all loaded conditions. The discrepancy is likely due to a difference in load application; Schiffman et al. [7] used standard US army clothing and equipment while the current study adopted packs with a symmetric load distribution in the horizontal plane. The present results also showed that RMS and MV, in both the A/P and M/L directions, tended to increase at the higher load level (Table 3 and Fig. 2(b)). Since the COP is always in phase with the COM [18], increases in these time-domain measures indicate that the COM more closely approached the boundary of the BOS with increasing load mass or height. Thus, it can be inferred that there is a greater likelihood of fall-related incidents with heavier or higher external loads. Our initial hypothesis was also supported, that increasing mass and/or height would challenge balance control.

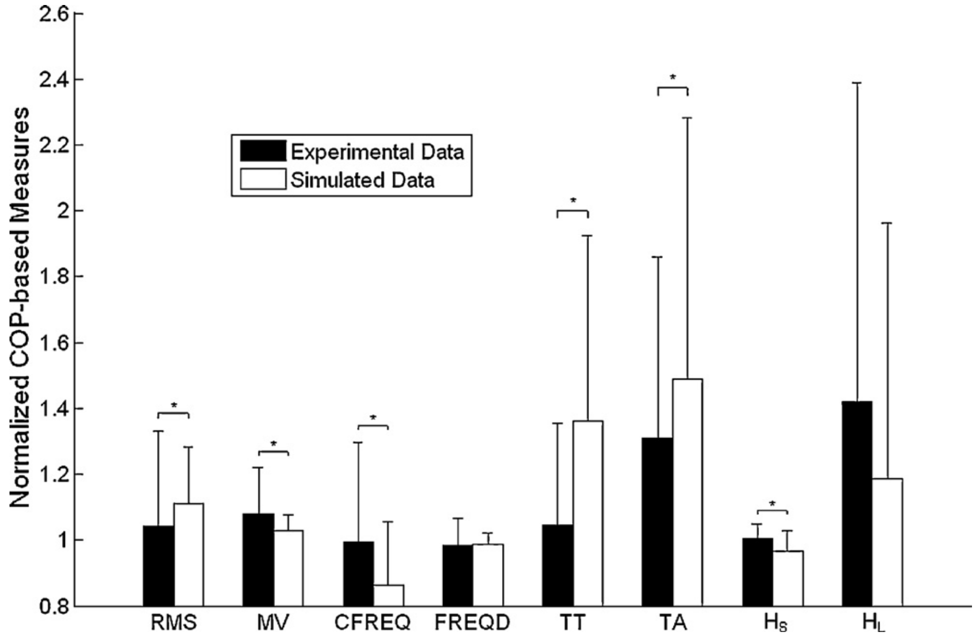

Fig. 3. Comparison of experimental and predicted A/P COP-based measures under loaded conditions. Error bars indicate 1 S.D., and symbol '*' indicates a significant difference.

A/P CFREQ decreased when the load was moved above the COM, whereas M/L CFREQ decreased at the higher level of external load mass (e.g. 20% of body mass). A decrease in CFREQ indicates an increase in amplitude of lower frequency components. A/P and M/L stabilities are primarily maintained through muscular adjustments at the ankle and at the hip, respectively [39]. Thus, changes in frequency content of ankle control forces and hip control forces may have led to the observed changes in A/P CFREQ and M/L CFREQ, respectively. Another explanation for the changes in A/P CFREQ and M/L CFREQ is that increased load mass and load height increase the moment of inertia of the body, which results in a decrease in the natural frequency of the body. In addition, it can be surmised that variability in the frequency content of ankle control forces decreased when external load mass increased since there was an inverse relationship found between A/P FREQD and load mass.

Analysis of COP time series using an fBm model reveals two types of behavior [35]. Specifically, over short-term intervals, past and future displacements are positively correlated (persistent behavior and $H_S > 0.5$), while over long-term intervals these displacements are negatively correlated (anti-persistent behavior and $H_L < 0.5$). In general, larger H_S indicates more persistent postural control, whereas larger H_L corresponds to less anti-persistent behaviors [27]. The intersection of the short-term and long-term regions is the transition point whose coordinate is [TT, TA]. Hence, TT indicates the duration of persistent behaviors, while TA is the measure used to quantify the amplitude of sway movement over short-term intervals.

As external load mass increased (i.e. from 0% to 20% of body mass), both A/P TA and M/L TA increased. Hence, sway movement in either the A/P or M/L directions led to postures further from upright equilibrium, suggesting an increase in short-term instability. External loads also led to more persistent behaviors over a longer duration in the M/L direction, as indicated by increases in M/L H_S and M/L TT in the loaded conditions, respectively. Persistent sway behaviors cause movements away from upright equilibrium [28]. Thus, it might be further concluded that M/L balance control deteriorated over short-term intervals with the application of external loads.

In contrast, M/L H_L decreased at the higher load mass, indicating more anti-persistent behaviors. In other words, in the presence of external loading M/L sway movement was more likely to return to upright equilibrium, and to be less random over long-term intervals. Thus, M/L balance control might be improved over long-term intervals with heavier external loads. A possible explanation for these findings is that after experiencing decreased stability over short-term intervals, humans may have the ability to adjust their balance control so as to improve stability over long-term intervals. Thus, it can be hypothesized that the changes of M/L H_L represent an adaptation of the postural control system to the decreased stability imposed by applied external loads.

Increasing external load height caused A/P postural sway to be less persistent or more random in the short-term (i.e. a decrease in A/P H_S). Contrasting this, A/P TT exhibited an increasing trend with external load height, suggesting that A/P persistent behaviors occurred over a longer duration. Given these contrasting results, it is unclear how A/P balance control was affected by load height over short-term intervals on the basis of these statistical mechanics measures.

A quantitative discrepancy was evident between experimental and predicted outcomes for several COP-based measures (Fig. 3). Quantitative correspondence was not actually expected, mainly due to two factors. First, intra-individual differences in behaviors exist, but only three trials were conducted by each participant under a specific loading condition. Second, our model has some limitations that might induce errors. For example, the body was described simply as a single-segment inverted pendulum and only a limited set of physical quantities relevant to sway were considered in the model of optimal neural control. Thus, when evaluating the model performance, we sought only to determine whether the model could qualitatively reflect the observed trends in the COP-based measures as external loading conditions changed, and such trends were considered sufficient to reflect how the postural control strategies are adjusted according to different external loads. At the same time, even though the neural controller was assumed to be an optimal controller, the predicted postural sway should change with diverse loading conditions since external loads affect human body dynamics in the postural control system [16].

The model duplicated most of the significant effects of external load mass (and apparent trends) on the observed A/P COP-based measures, though more predicted measures were found to be significantly affected (see Tables 3 and 4). In general, the predicted results indicated that heavier external loads challenged balance control, with A/P RMS, A/P MV, A/P TT and A/P TA all predicted to increase with the application of external loads. In addition, when external load height increased, an increase in predicted A/P RMS ($p = 0.053$) and a decrease in predicted A/P CFREQ ($p = 0.058$) were evident as trends, and these trends were consistent with experimental findings. Since the predicted A/P RMS increased with an increase in external load height, we might predict that increasing height would challenge balance control as well as load mass. Similar to the experimental data, the predicted results generally supported the initial hypothesis regarding the effects of external load mass and height. In other words, if the experimental data were unavailable, model-based prediction would lead to the same general conclusion as did the empirical data. At the same time, as noted earlier, given the average trends in the dependent measures versus external loads (Fig. 2), our simulation results also matched most of the experimental findings. Hence, it might be concluded that although some discrepancies existed between the experimental and predicted results, our model was still able to provide some useful information on how balance control was affected by external loads.

Some limitations in this study should be noted. First, learning effects may have been present. However, since practice was provided before data collection, and the order of the different loading conditions randomly presented, these effects are thought unlikely to have substantially influenced the experimental findings. Second, the sample size ($N = 12$) may have been insufficient to detect effects of external loading on some measures. Third, males and females are different in many aspects, for example, their anthropometric estimates (Eq. (1)) are different. Hence, the application of external loads may lead to different responses between genders. Future work is warranted to determine if such differences are important. Fourth, some anthropometric measures and the body COM were estimated, and thereby served as a source of errors in the model-based prediction. Such errors should not be critical, though, since effects of external loading were determined within-subjects.

Perhaps a more critical limitation is that we do not have direct evidence to support the two primary assumptions we made in this study. Specifically, the first assumption was that the same neural optimal control mechanisms would be employed under diverse conditions of external loading, based on which the model parameters defined in the neural optimal controller were held at the values determined from the no-load trials. The second was that external loads adversely affected the accuracy of control signals, and we implemented this by linearly increasing the random disturbance gain. During model simulations, only mechanical changes with external loads were taken into account.

In contrast to the current approach, we could have used methods similar to those reported earlier [16], optimally determining the weightings of sway-relevant physical parameters so that the simulated behaviors matched the observed behaviors. From this, we could make inferences regarding changes in postural control mechanisms that might occur with external loading. However, this was not our intent in this study. Instead, we wanted to determine if, or to what extent, our model was predictive. Essentially, we consider the approach used here a relatively strong test of the model's predictive ability, since no experimental postural sway data under loaded conditions are needed. The alternative approach, in which we allow the weightings to change and determine if measured behaviors can be replicated, we consider a much weaker test.

Due to the existence of discrepancies between the experimental and predicted results, the assumptions made in model-based prediction were not strongly supported. However, the predicted results did support the hypothesis that increasing mass and/or height would challenge balance control, and were consistent with the experimental findings. Hence, while there may be changes in neural optimal control mechanisms, and random disturbance gain may not change linearly with external loads, the mechanical changes induced by external loads appear to primarily affect balance control.

Although quantitative discrepancies were evident between experimental and predicted COP-based measures, qualitative comparisons were favorable (i.e. the effects of external load mass and height). Some support is thus considered evident for the predictive ability of our optimal control model. In the future, approaches that can predict postural behaviors more quantitatively should be investigated. In addition, since a loss-of-balance often leads to unintentional falls, investigating effects on balance control caused by external loads may aid in better understanding and preventing fall-related injuries, especially for those occupational and daily activities involving load carriage. Results presented in this study were used to explain how balance control was adjusted according to the application of external loads and may provide a basis for developing intervention strategies for the improvement of balance.

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Conflict of interest

We declare that both authors have no financial or personal relationships with other persons or organizations that might inappropriately influence our work presented therein.

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