Proceedings of the 28th IEEE
EMBS Annual International Conference
New York City, USA, Aug 30-Sept 3, 2006

Instability Prediction by Monitoring Center of Pressure During Standing

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Abstract—**Incorporating an instability predictor into a portable sensor has a number of clinically relevant applications. This study investigated the feasibility of developing a real-time assessment tool to predict stepping during standing by monitoring Center of Pressure (COP) measurements. Forward and backward perturbations were performed on 16 able-bodied subjects using a pulley system attached to the subjects' waist. A linear relationship was found between the peak COP velocity (COPv) and the peak COP position caused by the perturbations. As the peak COPv occurs considerably before the peak COP, the peak COP estimated using a regression equation from the peak COPv may serve as an instability predictor. By comparing stepping thresholds with the estimated peak COP, we found that the stepping predictor successfully predicted instability (stepping) earlier than those predictors using actual COP. Results show that the proposed model is a viable solution to predict stepping, and the feasibility of incorporating the model into a neuroprosthesis system for standing.**

I. Introduction

NEURO-musculo-skeletal disorders often result in degradation of the human balance control system. Balance control deterioration may occur suddenly, for example after spinal cord injury (SCI), or gradually, as is often the case with the elderly. Our long-term goal is to develop a neuroprosthesis, using functional electrical stimulation (FES) techniques, which can be used by people with balance impairment in their daily life. We have developed an ideal theoretical framework for a neuroprosthesis for keeping quiet stance [1]. However, in daily life it is common to take steps to prevent instability during standing [2], [3]. Therefore, to develop a neuroprosthesis for standing to be used in daily life it is necessary to predict instability to allow different FES modes to be switched and therefore prevent the predicted instability.

Manuscript received April 24, 2006. This work was supported in part by grants from Consejo Nacional de Ciencia y Tecnologia, Mexico; Canadian Institutes of Health Research, Canada; Natural Sciences and Engineering Research Council of Canada, Canada; Canadian Found for Innovation, Canada; Ontario Innovation Trust, Canada; Tateishi Technology Foundation #1041019, Japan.

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To evaluate stability during standing, Popovic *et al.* [4] proposed center of pressure (COP) *stability regions* based on a subject's COP measurements. Depending on the position of the COP, they defined the different COP areas in which the subject is forced to step forward, backward or sideways to maintain stability. Therefore, it is possible to detect the moment when a subject steps by determining the unstable zone using COP measurements.

Nevertheless, it was hypothesized that if we use COP velocity (COPv) measurements we can predict stepping earlier than if using COP alone since the peak COPv occurs before the peak COP. Thus, the purpose of the present study was to investigate the feasibility of developing a real-time assessment tool to predict stepping during standing by monitoring COP measurements.

II. Methods

A. Subjects

Sixteen healthy adults (8 male and 8 female) were recruited. Each subject signed written informed consent to comply with ethics approval granted by the Health Sciences Ethics Review Officer of the University of Toronto.

B. Experimental Setup

We used a dropped-weight pulley as a perturbation system. When the release mechanism was pulled, the pulley was able to generate instantaneous, horizontal, adjustable pulling force.

Two force plates (Type 9366AB05, Kistler, Switzerland) were used to measure the subject's COP, and a camera system (Optotrak 3020 3D, Northern Digital Inc., Canada) was used to determine the movements of toe and ankle caused by the applied perturbation. The optical markers were placed on the points of the lateral malleolus and the head of the metatarsal II to examine the movements of the ankle and toe, respectively. In this study, we considered only the anterior-posterior direction of sway.

The net COP was obtained according to Winter [5]. The COPv was obtained by differentiating the COP. COP was normalized by dividing the COP coordinates by the length of the subject's feet. COPv was normalized to the subject's height [2], [6]. The force plates and the kinematic measurements were synchronized and the sampling frequency for both systems was 140 Hz. For noise removal the data were filtered using a 4th order, low pass, zero-lag Butterworth digital filter [7] with a low pass cutoff frequency of 10 Hz.

C. *Protocol*

The subjects were asked to stand still on the force plates, with bare feet, and in an upright position. Forward and backward perturbations were applied at the waist. For each subject, and for each direction, weights were used from 0.454 kg and up to the maximum weight required to produce a step (the maximum weight used among the subjects group was 6.8 kg). The weight applied was incremented by 0.454 kg until the subject made a step, and then one more weight of 0.454 kg was added, to ensure that at least two stepping trials were recorded per subject. The instance when a step occurred was defined by the moment when the vertical ground reaction force from one of the two force plates was reduced to zero. The order of the perturbations (i.e. weight used to cause the perturbation) was randomly selected.

D. *Data Analysis*

1) Estimation of COP Peak from the COPv Peak: The peaks of the COPv and COP signals were determined for all the forward and backward trials. A linear regression analysis using a least-squares method was applied to obtain an equation by which the COP peak could be estimated from the COPv peak for each trial. These equations predict the subject's COP peak based on the COPv peak.

2) Stepping Predictors: The linear regression equation described above will predict the COP peak that a subject will experience from a perturbation. Therefore, if the output of the equation is compared with a COP threshold for instability that will cause stepping, a *stepping predictor* is obtained. Five different anterior thresholds were chosen:

--Th.1F, the anterior limit of the undesirable zone proposed by Popovic *et al.* [4].

--Th.2F, the mean value of the maximum COP among trials in which no ankle movement was detected.

--Th.3F, the minimum value of the maximum COP among trials in which ankle movement was detected.

--Th.4F, the mean value of the maximum COP among trials in which ankle movement was detected.

--Th.5F, the maximum value of the maximum COP among trials in which ankle movement was detected.

These anterior stepping thresholds were determined using all the forward-perturbation non-stepping trials.

Five different posterior stepping thresholds were chosen:

--Th.1B, the posterior limit of the undesirable zone proposed by Popovic *et al.* [4].

--Th.2B, the mean value of the minimum COP among trials in which no toe movement was detected.

--Th.3B, the maximum value of the minimum COP among trials in which toe movement was detected.

--Th.4B, the mean value of the minimum COP among trials in which toe movement was detected.

--Th.5B, the minimum value of the minimum COP among trials in which toe movement was detected.

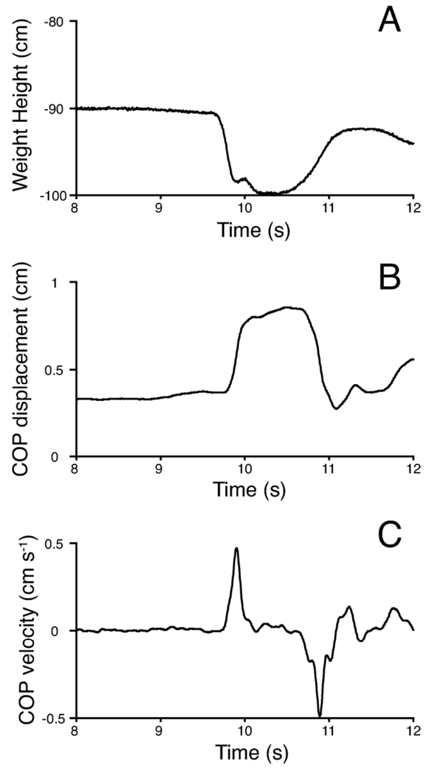

Fig. 1. Typical examples of weight height (A), COP (B), and COPv (C) traces.

These posterior stepping thresholds were determined using all the backward-perturbation non-stepping trials.

We refer to the abovementioned stepping predictors as *dynamic predictors*. We compared these dynamic predictors with the stepping predictor proposed by Popovic *et al.* [4], which used only COP information and therefore is referred to as a *static predictor*. A binary test was performed for each stepping predictor with all the perturbation trials [8], [9]. Differences in time between the moment when stepping was predicted and the moment when the subject actually made a step (predicting time) were calculated for the most suitable dynamic predictor and the static predictor. Only the trials in which the predictor performed satisfactorily (i.e. the predictor predicted stepping before the subject made a step) were used for calculating the prediction time.

III. Results

A. *Estimation of COP Terminus from COPv*

Trials from one male subject were excluded from the results due to his undesirable reaction to the perturbations. The subject tried consciously to maintain balance such that he did not need to make a step by plantar flexing after every perturbation. Therefore, the total number of the considered subjects became 15. For these considered subjects, 222 trials (137 forward perturbations, 85 backward perturbations) were

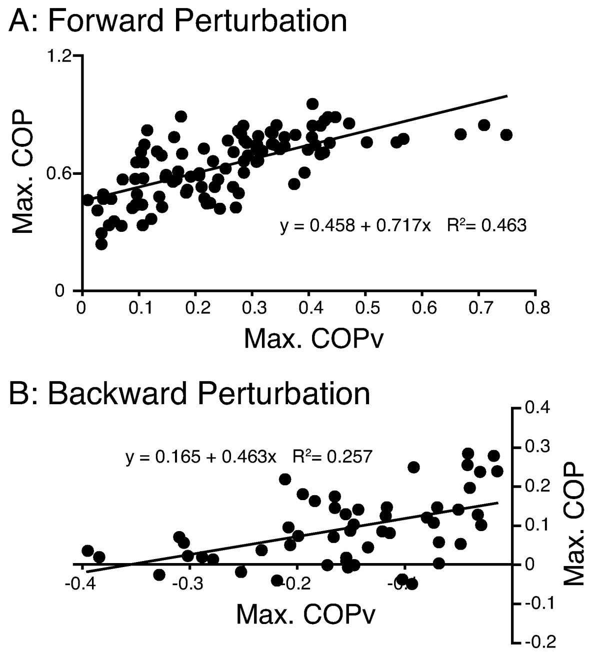

Fig. 2. COP peak vs. COPv peak from all the forward perturbation trials (A) and from all the backward perturbation trials (B).

completed successfully in total. The number of trials differed for each subject, because each individual participant required a different amount of weight to be perturbed sufficiently to make a step to recover balance.

Fig. 1 shows the time-history data from a single non-stepping trial induced by a forward perturbation. Forward COPv peak occurred before the COP peak. All perturbation trials showed the same behavior. The first peaks of each variable after the perturbation were obtained to be used in the subsequent analysis.

1) Forward Perturbations: Fig. 2A shows the group result of the relationship between the COP peak and the COPv peak from forward perturbation trials. A least-squares regression estimation resulted in the following equation:

$$COP = 0.717\, COPv + 0.458 \ . \tag{1}$$

The R^2 of the estimate was 0.463. 95% confidence bounds were 0.565 < slope < 0.869, and 0.413 < intercept < 0.503 for the slope and the intercept, respectively. The Pearson's correlation coefficient was 0.680 (n = 137, p = 0.001).

2) Backward Perturbations: Fig. 2B shows the group result of the relationship between the COP peak and the COPv peak from backward perturbation trials. A least-squares regression estimation resulted in the following equation:

$$COP = 0.463\, COPv + 0.165 \ . \tag{2}$$

The R^2 of the estimate was 0.257. 95% confidence bounds were 0.245 < slope < 0.681, and 0.126 < intercept < 0.204 for the slope and the intercept, respectively. The Pearson's correlation coefficient was 0.507 (n = 85, p = 0.001).

B. Stepping Predictor

Binary tests were performed for each stepping predictor using 2 x 2 contingency tables. The *sensitivity* of either the dynamic predictors or the static predictors is the probability that stepping was predicted in a trial in which the subject made a step. The *specificity* is the probability that stepping was not detected in a trial in which the subject did not have to step. The stepping predictors' sensitivity and specificity are plotted against the stepping threshold for forward movements in Fig. 3A. As the stepping threshold increases, the sensitivity decreases linearly whereas the specificity increases linearly.

The stepping predictors' sensitivity and specificity are plotted against the stepping threshold for backward movements in Fig. 3B. As the stepping threshold increases, the sensitivity decreases linearly whereas the specificity increases linearly, although the linearity was not clear compared with the forward perturbation case.

The dynamic predictors with sensitivity similar to that of the static predictors were the one with Th.2F for forward perturbations, and the one with Th.2B for backward perturbations. We examined how early these dynamic predictors were able to predict stepping compared with the static predictors. The paired t-test revealed that there was a significant difference in the prediction times between the dynamic predictor and the static predictor for forward perturbations (p < 0.001, n = 23), and for backward perturbations (p < 0.001, n = 21).

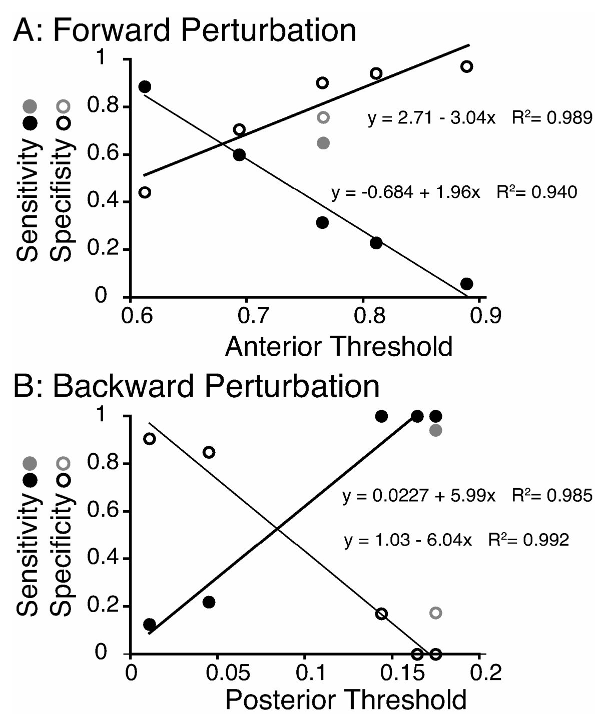

Fig. 3. Sensitivity and specificity of each step predictor vs. corresponding step threshold for forward (A) and backward (B) perturbations. The grey plots represent the sensitivity and specificity of the static predictors.

IV. Discussion

We demonstrated that there is a sufficient linear relationship between the COPv peak and the resulting COP peak involved in a perturbation (Fig. 2). The results suggest that it is possible to estimate the COP peak using COPv measurements according to (1) and (2). Introducing stepping thresholds, we predicted instability that results in stepping using the estimated COP peak. When using the stepping thresholds previously established by Popovic *et al.* (4), stepping was successfully predicted (Fig. 3). The dynamic predictors that had closer sensitivity and specificity to those from the static predictor were: the one with Th.2F for forward perturbations, and the one with Th.2B for backward perturbations. We found that the prediction times of these dynamic predictors were statistically longer than those of the static predictors. The results indicate that if these dynamic predictors are incorporated into an FES system for standing, it is possible to switch between different FES modes (according to the subject's stability situation) about 200 ms earlier than using the static predictors. This would present a considerable advantage for an FES standing system.

In practice, there is always a trade-off between sensitivity and specificity; higher sensitivity predictors have a lower specificity, and vice versa. However, the forward static predictor presented higher values for both sensitivity and specificity compared with the forward dynamic predictors (Fig. 3A). The forward dynamic predictor with sensitivity and specificity similar to the forward static predictor was the one with Th.2F. For backward perturbations, the static predictor showed good sensitivity but poor specificity (Fig. 3B). Most of the backward dynamic predictors showed poorer sensitivities than the backward static predictor. The only backward dynamic predictor that presented equivalent sensitivity to that of the static predictor was the one with Th.2B.

Because the dynamic predictors showed longer prediction times, we suggest that these predictors are suitable for implementation with an FES standing system. However, the fact that both the dynamic and the static predictors for backward perturbations showed poor specificity remains to be improved in the future.

A sensor for balance using these dynamic predictors could be integrated with a neuroprosthesis for standing. When the predictor detects a step is about to happen, the neuroprosthesis controller would switch to the appropriate FES mode to avoid falling. The dynamic predictor will provide sufficient time for switching modes.

As this study involved only young subjects, it would be valuable to examine the relationship between the COP and COPv for the elderly, and whether the predictor coefficients are affected by age, neurological and musculo-skeletal problems. Also, different thresholds may be found for different populations.

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