Modulation of Ankle Stiffness during Postural Sway

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Abstract— **Ankle stiffness is a nonlinear, time-varying system which contributes to the control of human upright stance. This study sought to examine the nature of the contribution of stiffness to postural control by determining how intrinsic and reflex stiffnesses varied with sway. Subjects were instructed to stand quietly on a bilateral electro-hydraulic actuator while perturbations were applied about the ankle. Subjects performed three types of trials: normal stance, forward lean, and backward lean. Position, torque, and EMGs from the tibialis anterior and triceps surae were recorded. Background torque, intrinsic stiffness and reflex stiffness were calculated for each perturbation. Intrinsic and reflex stiffnesses were heavily modulated by postural sway. Moreover, they were modulated in a complimentary manner; intrinsic stiffness was lowest when reflex gain was highest, and vice versa. These findings suggest that intrinsic stiffness is modulated simultaneously with reflex stiffness to optimize the control of balance.**

I. INTRODUCTION

Controlling and maintaining upright stance is an inherently difficult task. Humans must regulate the position of their center of mass (COM) to keep their center of pressure (COP) within the base of support. This involves controlling a large mass at a significant height above the ankle over a relatively small base of support; this is often modeled as an inverted pendulum. An ankle strategy for postural stabilization is typically used when instabilities are minimal [1]. When this is the case, the majority of the experimentally observed stabilizing activity occurs about the ankle. Unperturbed stance is characterized by low-frequency, quasi-random motion of the COM, termed postural sway. This instability is controlled by various passive and active mechanisms which generate torque about the ankle joint to maintain the upright position of the body. However, knowledge of the functionality of these mechanisms across the broad spectrum of human movement is incomplete.

Many studies of upright stance have used measures of the COM and COP to study the dynamics of balance, and have shown the importance of several pathways to postural control. Reflex responses can be induced and modulated depending on what is required to maintain stability in any given situation [2]. Feed-forward dynamics have also been observed, as EMG activity may correlate with body sway in an anticipatory fashion [3], although this may be the result of the use of a specific controller. However, many studies examine overall COP-COM dynamics and do not distinguish between active and passive mechanisms, thus limiting insight into the underlying mechanisms.

Joint stiffness defines the relationship between the position of a joint and the torque produced about it. Intrinsic and reflex stiffnesses change with the postural state. Both increase as ankle position moves away from the neutral position [4][5] and both change with activation level [4][6]. Both ankle position and contraction level vary continuously during standing. Therefore, the contributions of intrinsic and reflex mechanisms are likely to change constantly with postural sway.

This study examined the intrinsic and reflex stiffnesses as the postural state changed. This was performed by analyzing responses to position perturbations applied bilaterally by a hydraulic actuator. The following sections describe the experiments performed and analysis procedures used to analyze the data on a response-by-response basis. These data are then summarized to characterize the modulation of the intrinsic and reflex responses during posture.

II. METHODS

A. Experimental Apparatus

Subjects stood on the foot pedals of a bilateral hydraulic actuator [7], shown in Figure 1. Rotac 26R-2 1V actuators were used to apply position perturbations to the ankle. The angular position of the foot pedals was measured using Maurey Instruments 112-P19 potentiometers. A positive position indicates ankle dorsiflexion. Torque was collected using Lebow 2100-5k torque transducers. More positive torques indicate a COP position nearer the heel. EMG data for medial and lateral gastrocnemius (GS), soleus, and tibialis anterior (TA) were measured from Delsys Bagnoli DE-2.1 surface electrodes. The actuator was equipped with redundant safety mechanisms to ensure subject safety.

B. Trials

Nine subjects (6 male, 3 female) aged 23-29 with no history of injury or neuromuscular disease participated in this study. Each subject performed three types of trials: normal, backward leaning, and forward leaning. Subjects were instructed to stand normally with feet shoulder width apart, looking straight ahead for normal trials. In the backward leaning trials, subjects were instructed to maintain their weight slightly backward from normal. In forward leaning trials, subjects were instructed to maintain their weight slightly forward from normal. Each subject performed 6 trials: 2 normal, 2 forward-leaning, and 2 backward-leaning.

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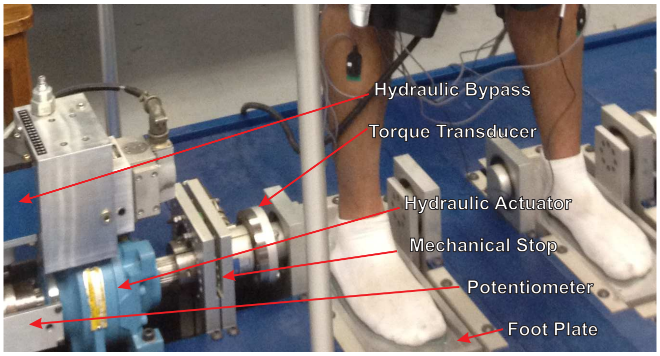

Fig. 1: Experimental set-up used to apply perturbations to the ankle with major components labeled.

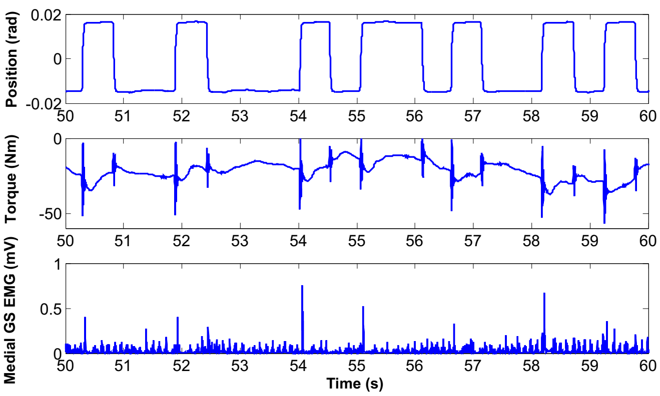

Fig. 2: Ten seconds of sample data from one ankle during a normal stance experiment. The intrinsic response is evident for all perturbations and reflexes can be observed for dorsiflexing perturbations.

A pseudo-random binary sequence (PRBS) input with a peak-to-peak amplitude of 0.03 radians and switching rate of 500ms was applied to each foot pedal. There was minimal cross-correlation between the inputs to each leg. Subjects stood for 120 seconds while perturbations were applied bilaterally. Position, torque and load cell data were collected. All signals were anti-aliased and sampled at 1 kHz. Fig. 2 shows sample position, torque, and GS EMG data from a segment of a typical trial.

C. Data Processing

Each trial was divided into a collection of individual responses, comprising a period of 50 ms before to 450 ms after the peak velocity associated with a PRBS transition. Each response was divided into four periods as shown in Fig. 3, and analyzed as follows:

1) A linear model was fitted to the first 25 ms of torque to model the background torque trend. The estimate of background torque was not confounded by previous reflexes, since reflex responses were separated by a minimum of 500ms and lasted no more than 300ms after a perturbation. This trend was extrapolated to 100ms and removed from the torque response. The average of the EMG signal for period A was defined as the background EMG.

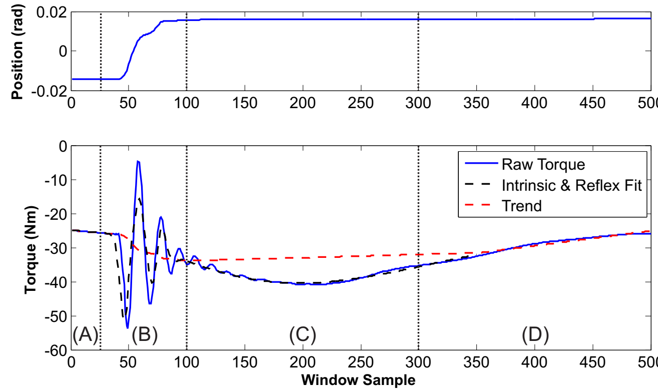

Fig. 3: Sample position (top panel), torque, and intrinsic/reflex fits for one response (bottom panel). The dotted red line shows the summed estimated background and intrinsic torque for the response. The response is subdivided into four periods: (A) pre-response (0-25ms), (B) intrinsic response (25-100ms), (C) reflex response (100-350ms), (D) post-reflex (350-500ms).

2) A second order IBK model was estimated from the residual torque in period B. The elastic stiffness value, K, was estimated for each response and normalized by the subject's critical stiffness [8], the stiffness required to maintain the stability of an equivalent inverted pendulum model, computed from (1).
3) The background torque in period C was estimated and removed by fitting a line between the sum of background and intrinsic torque at 100ms and total torque at 350ms. Reflex stiffness dynamics were estimated by fitting a second order, time-domain model (2) to the residual torque in period C.

$$K_{crit} = m \cdot g \cdot h_{COM} \tag{1}$$

$$TQ_{ref} = \frac{G\omega_0^2}{\omega_0\sqrt{1-\zeta^2}} e^{-\zeta\omega_0 t} \sin(\omega_0\sqrt{1-\zeta^2 t}) \tag{2}$$

In (1), m was subject mass, g was the gravitational constant, and h_{COM} was the height of the COM (based on anthropometric data). The parameters estimated in (2) were gain G, natural frequency ω_0, and damping parameter ζ. The reflex gain was defined as the resulting estimate of 'G'. The reflex gain was set to zero for trials with no significant reflex response.

A MATLAB-integrated fitting algorithm (the 'fit' function with the fit type set to 'smoothingspline') was used to summarize relationships between the intrinsic stiffness and postural sway. The smoothing parameter p was set to p=0.01. Since the reflex gains were generally non-Gaussian for any range of background torque, a moving window median of 10 Nm width was used to create a trendline.

III. Results

A. Intrinsic Stiffness

Fig. 4 shows the relation between intrinsic stiffness and background torque for a typical subject. Data are shown from

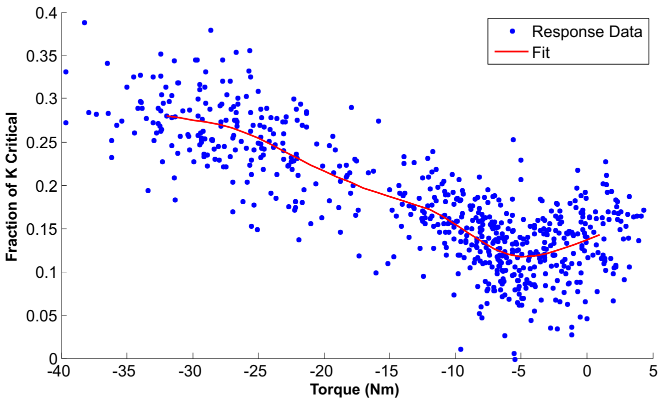

Fig. 4: Sample intrinsic stiffness for the left leg of subject 4. The data includes two trials of each type. Each point represents a single response.

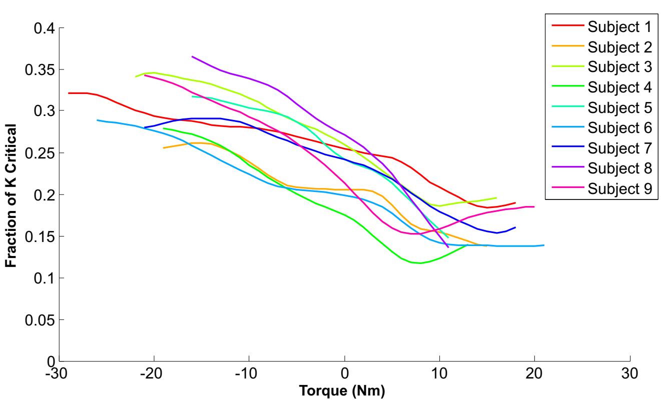

Fig. 5: Left ankle trends for intrinsic stiffnesses.

two trials of each type. Individual responses are indicated by the blue dots while the smooth red line is the spline fitted to the data. The intrinsic stiffness was smallest near -5 Nm and increased with background torque in either direction. The fits for other subjects were comparable, as demonstrated by Fig. 5, which shows the trends for all subjects after removing the mean background torque. For all subjects, intrinsic stiffness increased as background torque became more negative (i.e. as the COP moved toward the toes). Additionally, most subjects had a local minimum in intrinsic stiffness or a change in the slope of the relationship at approximately 10 Nm.

For a majority of postural states, intrinsic stiffness alone was insufficient to stabilize the inverted pendulum model of upright stance. The maximum contribution of each leg to the critical value ranged from 26-45% between subjects. When adding the maximum observed values of stiffness in the trends for each leg, the summed intrinsic stiffness values ranged from 52-80% of the critical value.

Linear fits to the intrinsic stiffness responses for the linear range of stiffnesses (approx. -10 to 6 Nm) are shown in Table I. This corresponded to the range of torques observed for each subject in the non-leaning experiments. For all subjects, the intrinsic torque decreased as background torque became more positive (as the COP moved toward the heels). The narrow 95% confidence intervals show that the true

TABLE I: Linear fit for intrinsic stiffness calculated from response data for the left ankle, using fit: $K_{crit} = a \cdot TQ_0 + b$. The slope column includes 95% confidence bounds.

Subject	**a** $(1/rad) \times 10^{-3}$	$\mathbf{r^2}$
1	-1.5 $\pm$ 0.2	0.24
2	-3.8 $\pm$ 0.5	0.24
3	-4.5 $\pm$ 0.4	0.48
4	-3.3 $\pm$ 0.4	0.30
5	-5.4 $\pm$ 0.5	0.43
6	-3.0 $\pm$ 0.2	0.45
7	-3.4 $\pm$ 0.4	0.29
8	-5.2 $\pm$ 0.5	0.42
9	-4.2 $\pm$ 0.3	0.62

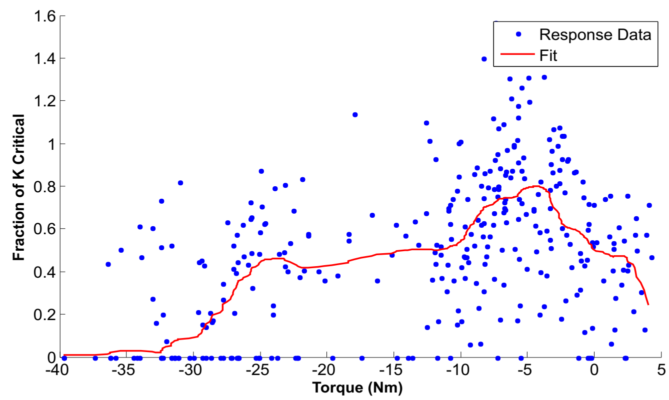

Fig. 6: Sample reflex gains for the left leg of subject 4. The data includes two trials of each type. Each point represents a single response.

relationship between these parameters is likely to be different from zero. This was also observed in the right ankle.

B. Reflex Stiffness

Fig. 6 shows reflex gain as a function of background torque for the same trials shown in Fig. 4. It was greatest near -5 Nm, and decreased to either side. This pattern was characteristic of all subjects, with intrinsic stiffness minima located at approximately the same background torque value as the reflex maxima.

Fig. 7 shows the reflex trends, after removing the mean background torque, excluding four subjects who had little or no reflex behavior. While the variability of the data is large, the trend reflects the size and number of reflexes occurring. Reflex gain increased as torque became more positive (i.e. as the COP moved toward the heels). For most subjects, the maximum reflex gain occurred at a background torque of approximately 10 Nm. The gain decreased for both greater and smaller torques.

For each subject, the collection of responses was divided into five groups sorted by increasing background torque, each containing 20% of the total number of responses. Wilcoxon signed-rank tests were performed between the distributions containing the largest and smallest median reflex gains. For all subjects and all legs, the statistical test indicated that these two distributions were different for a significance level $\alpha = 0.05$.

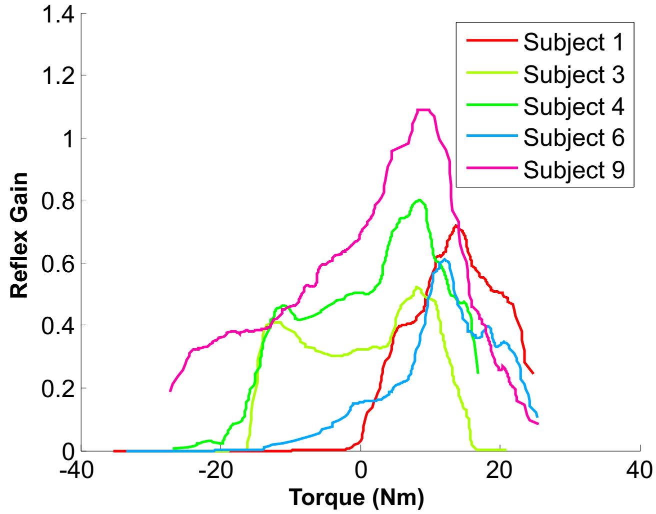

Fig. 7: Reflex trends for the left ankle.

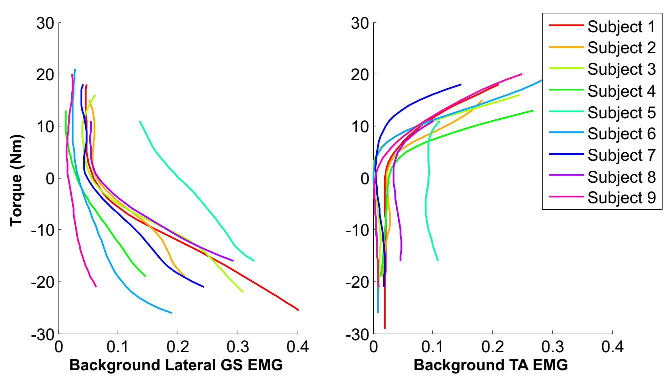

Fig. 8: Mean torque levels for differing levels of EMG activation during leaning experiments fit using the smoothing spline algorithm. Background EMG is normalized to a range of zero to one, corresponding to the minimum and maximum background activation observed from any individual response from each subject.

C. *Background Muscle Activation*

Fig. 8 shows EMG activation patterns and the resultant background torque. Considering Fig. 8 in conjunction with Figs. 5 and 7, when intrinsic and reflex stiffnesses reached local extremes, lateral GS and TA were minimally active. As background torque became more negative/positive, lateral GS/TA activity increased, respectively; intrinsic and reflex stiffnesses changed simultaneously as well. Soleus and medial GS activity patterns were similar to the lateral GS.

IV. Discussion

The results presented in this paper demonstrate that intrinsic and reflex stiffnesses are strongly modulated by postural sway. By asking subjects to lean forward and backward, we were able to observe intrinsic and reflex dynamics that would not be evident for a normal range of torques (these showed only the linear portion of each trend). Additionally, the modulation of intrinsic and reflex stiffnesses are complementary. Intrinsic stiffness was lowest when reflex gain was highest, and vice versa. These maxima and minima correlate with a net minimum level of background activation of the triceps surae and TA muscles.

These observations suggest a linkage between the control of intrinsic stiffness and reflex gain, and support the claim for the existence of hybrid control strategies. An explanation of this phenomenon can be related to the inverted pendulum model of upright posture. At the most stable point, the mass of the body is directly above the base of support. No torque would be theoretically required and any reactive stabilizing ankle torque would have a maximum effect on the motion of the COM. This agrees with the observation that intrinsic and reflex stiffnesses reach local extrema at a background torque slightly greater than the mean value (i.e. - COP further toward heels). Subjects additionally tend to prefer to sway about a mean COP location which is slightly in front of the ankle joint [9], which suggests that these extrema correspond to a COM position located directly above the ankle. Both results are in agreement with the hypothesis that the postural control system chooses a strategy which minimizes energy expenditure, as a minimum stiffness at the most stable point would minimize unnecessary muscle co-contraction.

The large variability of the reflex gain across responses suggests that other variables may also contribute to its modulation. The smaller variability of the intrinsic stiffness suggests that it is better predicted by background torque alone. Additionally, sway velocity has been proposed as a modulator of reflex gain [10], however no clear relationship between the two was observed from subjects in this study.

References

[1] R. Creath, T. Kiemel, F. Horak, R. Peterka, and J. Jeka. A unified view of quiet and perturbed stance: simultaneous co-existing excitable modes. *Neuroscience letters*, 377(2):7580, 2005.

[2] E. Toft, T. Sinkjaer, S. Andreassen, and K. Larsen. Mechanical and electromyographic responses to stretch of the human ankle extensors. *Journal of neurophysiology*, 65(6):14021410, 1991.

[3] K. Masani, M. Popovic, K. Nakazawa, M. Kouzaki, and D. Nozaki. Importance of body sway velocity information in controlling ankle extensor activities during quiet stance. *Journal of Neurophysiology*, 90(6):3774 3782, 2003.

[4] M. M. Mirbagheri, H. Barbeau, and R. E. Kearney. Intrinsic and reflex contributions to human ankle stiffness: variation with activation level and position. *Experimental Brain Research*, 135(4):423436, 2000.

[5] P. L. Weiss, R. E. Kearney, and I. W. Hunter. Position dependence of ankle joint dynamicsi. passive mechanics. *Journal of biomechanics*, 19(9):727735, 1986.

[6] P. L. Weiss, I. W. Hunter, and R. E. Kearney. Human ankle joint stiffness over the full range of muscle activation levels. *Journal of biomechanics*, 21(7):539544, 1988.

[7] S. M. Forster, R. Wagner, and R. E. Kearney. A bilateral electro-hydraulic actuator system to measure dynamic ankle joint stiffness during upright human stance. In *Engineering in Medicine and Biology Society, 2003. Proceedings of the 25th Annual International Conference of the IEEE*, volume 2, pages 15071510. IEEE, 2003.

[8] M. Casadio, P. G. Morasso, and V. Sanguineti. Direct measurement of ankle stiffness during quiet standing: implications for control modelling and clinical application. *Gait & posture*, 21(4):410424, 2005.

[9] D. A. Winter, A. E. Patla, S. Rietdyk, and M. G. Ishac. Ankle muscle stiffness in the control of balance during quiet standing. *Journal of Neurophysiology*, 85(6):26302633, 2001.

[10] P. J. Bock and R. E. Kearney. Modulation of stretch reflex excitability during quiet human standing. *Proceedings of the 26th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, Vols 1-7, 26(1):46844687, 2004.