

Effect of Strength and Speed of Torque Development on Balance Recovery With the Ankle Strategy

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Robinovitch, Stephen N., Britta Heller, Andrew Lui, and Jeffrey Cortez. Effect of strength and speed of torque development on balance recovery with the ankle strategy. *J Neurophysiol* 88: 613–620, 2002; 10.1152/jn.00080.2002. In the event of an unexpected disturbance to balance, the ability to recover a stable upright stance should depend not only on the magnitude of torque that can be generated by contraction of muscles spanning the lower extremity joints but also on how quickly these torques can be developed. In the present study, we used a combination of experimental and mathematical models of balance recovery by sway (feet in place responses) to test this hypothesis. Twenty-three young subjects participated in experiments in which they were supported in an inclined standing position by a horizontal tether and instructed to recover balance by contracting only their ankle muscles. The maximum lean angle where they could recover balance without release of the tether (static recovery limit) averaged $14.9 \pm 1.4^\circ$ (mean \pm SD). The maximum initial lean angle where they could recover balance after the tether was unexpectedly released and the ankles were initially relaxed (dynamic recovery limit) averaged $5.9 \pm 1.1^\circ$, or $60 \pm 11\%$ smaller than the static recovery limit. Peak ankle torque did not differ significantly between the two conditions (and averaged 116 ± 32 Nm), indicating the strong effect on recovery ability of latencies in the onset and subsequent rates of torque generation (which averaged 99 ± 13 ms and 372 ± 267 N \cdot m/s, respectively). Additional experiments indicated that dynamic recovery limits increased $11 \pm 14\%$ with increases in the baseline ankle torques prior to release (from an average value of 31 ± 18 to 54 ± 24 N \cdot m). These trends are in agreement with predictions from a computer simulation based on an inverted pendulum model, which illustrate the specific combinations of baseline ankle torque, rate of torque generation, and peak ankle torque that are required to attain target recovery limits.

INTRODUCTION

Falls are the number one cause of accident-related injury and number two cause of accident-related death in the elderly (Bonnie et al. 1999). Approximately 30% of community-dwelling elderly fall at least once each year, and 10–15% of falls results in a serious injury. Hip fracture is the most important fall-related injury, with approximately 300,000 annual cases in the United States and associated medical costs of nearly \$10 billion (U.S. Department of Health and Human Services 1993).

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While an individual's risk for falling is associated with a variety of sensory, motor, cognitive, and environmental variables (Gehlsen and Whaley 1990; Nevitt et al. 1989; Rubenstein et al. 1994; Studenski et al. 1991; Tinetti 1994; Whipple et al. 1987), it ultimately depends on their frequency of loss-of-balance episodes, and their ability to recover balance by stepping, grasping, or swaying (via the ankle strategy or hip strategy). Fall prevention programs therefore need to evaluate and target each of these areas.

An important prerequisite to the development of such interventions is improved understanding of the variables that govern our ability to recover balance. Laboratory studies indicate that these include the peak magnitudes of lower extremity joint torques that accompany a specific balance recovery response, and the rate of development of these torques (Chandler et al. 1990; Horak et al. 1989; Lord et al. 1999; Luchies et al. 1994; McIlroy and Maki 1996; Pai et al. 1998; Tang and Woollacott 1998; Thelen et al. 1997; Wojcik et al. 1999; Wolfson et al. 1986). This is supported by epidemiological evidence that risk for falls among older adults increases with declines in muscle strength and with increases in reaction time (Lord et al. 1994; Nevitt et al. 1991). However, tools do not exist for directly quantifying how the ability to recover balance is affected by the magnitude versus the speed of torque development (Hall et al. 1999), and this limits our ability to diagnose and target patient-specific causes of postural instability.

In the present study, we used a combination of experiments and mathematical modeling to determine how the magnitude and speed of torque development affects young, healthy individuals' ability to recover balance with the ankle strategy. In our experiments, we measured the maximum forward lean angle where subjects could recover a stable upright stance by contracting their ankle muscles and examined whether this index of recovery ability differed when subjects self-initiated their recovery versus recovered balance after being unexpectedly released from a forward leaning position. The former parameter (which we termed the "static recovery limit") should depend primarily on parameters related to muscle strength, while the latter (which we termed the "dynamic recovery limit") should depend on parameters related to both muscle

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strength and reaction time. In our mathematical modeling efforts, we determined whether dynamic recovery limits could be predicted by an inverted pendulum representation of the body having time varying ankle-torque properties. We then used this model to identify combinations of baseline ankle torques, onsets and rates of torque generation, and peak magnitudes of ankle torque required to attain target dynamic recovery limits.

METHODS

Subjects

Twenty-three subjects participated in the study, 17 males and 6 females, having a mean age of 27 ± 5 (SD) yr (range: 17–38 yr), mean body mass of 72 ± 13 kg (range: 54–106 kg), and mean height of 1.75 ± 0.1 m (range: 1.51–1.97 m). Each subject provided informed written consent, and the experiment was approved by the Committee on Human Research of the University of California, San Francisco.

Methods

During the experimental trials, we measured the maximum initial lean angle where subjects were able to recover balance by contracting the muscles spanning their ankle joint, a balancing technique often referred to as the “ankle strategy” (Horak et al. 1989; Nashner 1976). To conduct a trial, we positioned the subject (who was barefoot and wore a loose-fitting T-shirt and short pants) with their feet shoulder-width apart and arms crossed over their chest. We then inclined the subject into a stationary forward leaning position via a horizontal tether that attached at one end to an electromagnetic brake (Warner Electric model PB500, South Beloit, IL) and at the other end to a chest harness worn by the subject (Fig. 1). Finally, we instructed the subject to rise into a vertical standing position by contracting the muscles spanning the ankles, while keeping the knees and hips extended. No restriction was placed on whether or not subjects raised their heels off

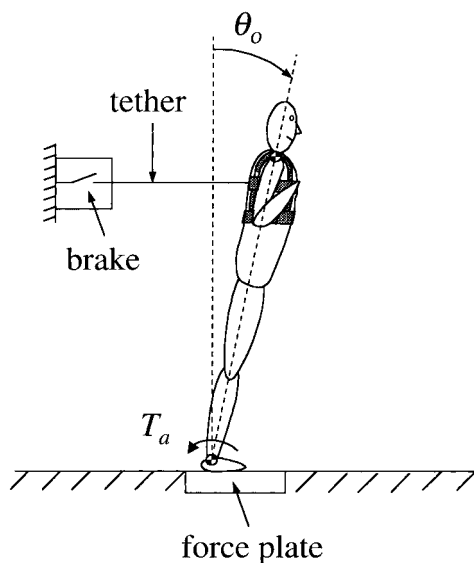


FIG. 1. Recovery limits experiment. Subjects were held by a tether in an initially inclined position and instructed to recover a stable upright position by contracting only their ankle muscles. Their “static recovery limit” was defined as the maximum angle where they could accomplish the task when the tether was not released. Their “dynamic recovery limit” was defined as the maximum angle where they could recover balance after the tether was unexpectedly released.

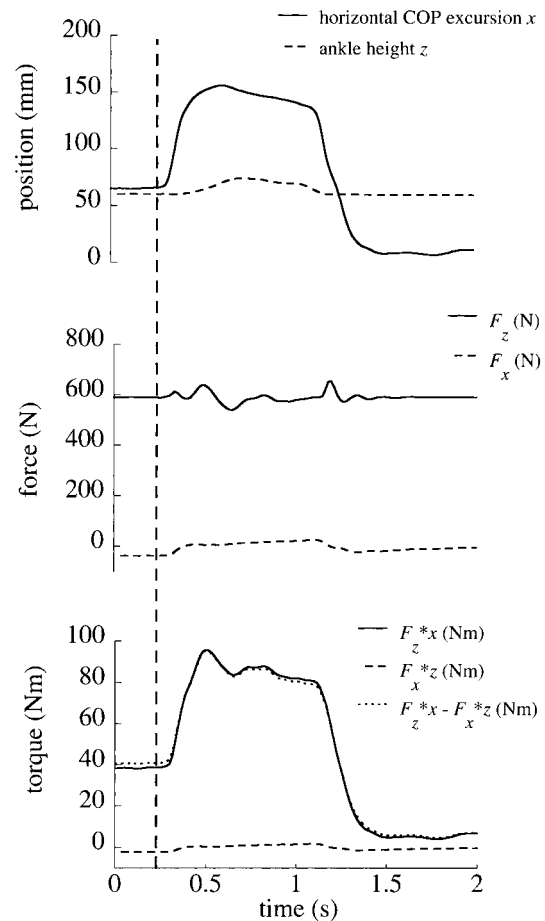


FIG. 2. Effect on ankle torque generation of magnitude and point of application of foot contact forces. Data correspond to a single, typical trial. The vertical dashed line shows the instant of tether release. Total ankle torque is given by $(F_z x + F_x z)$. However, since F_z is so much greater than F_x , total ankle torque is dominated by the product $F_z x$ (which in this trial had a peak magnitude of 95.6 Nm) instead of the product $F_x z$ (which had a peak magnitude of 2.4 Nm). Accordingly, changes in z corresponding to raising of the heels during balance recovery (which were typically about 1.5 cm) had little effect on total ankle torque generation.

the ground during the balance recovery process, and most trials involved some degree of heel rise (Fig. 2).

During each trial, we used a force plate (model 6090H, Bertec, Worthington, OH) to measure the magnitude and point of application of foot-floor reaction forces (sum from both feet) at a rate of 480 Hz. We also used a 60-Hz, six-camera motion measurement system (Qualysis, Glastonbury, CT) to measure the positions of markers secured to the skin overlying the right and left fifth toe (metatarsal), ankle (lateral malleolus), knee (lateral femoral epicondyle), hip (greater trochanter of the femur), shoulder (acromium), elbow (radial head), and wrist (junction between ulna and radius).

For each subject, the first trials involved lean angles of $\sim 2^\circ$, where they could recover balance easily. We then iteratively adjusted the length of the tether until we determined the maximum initial lean angle (with a resolution of 5 mm in tether length, and $\sim 0.2^\circ$ in lean angle) where the subject was able to recover balance in three or more of five repeated trials. Rest breaks of ≥ 30 -s duration were provided between trials to minimize muscle fatigue.

To determine the effect on recovery ability of the magnitude versus speed of torque development, we conducted first “static” and then “dynamic” trials. During static trials, the subject attempted to simply rise into a standing position without release of the tether (Fig. 3A). During dynamic trials, the subject attempted to recover balance after

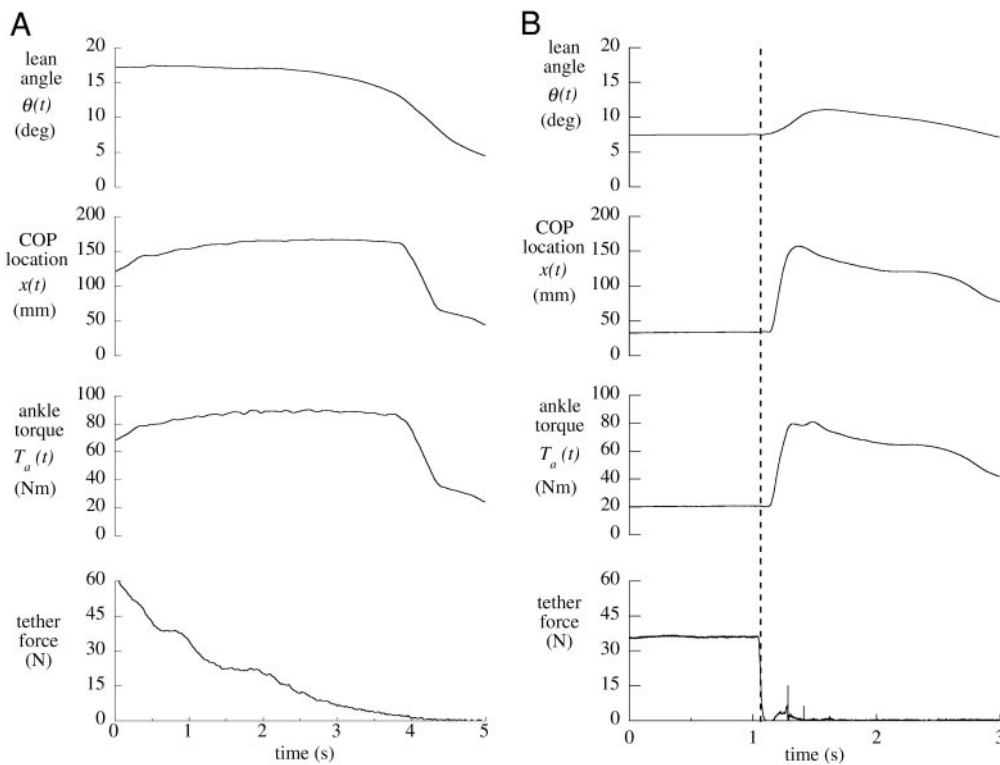


FIG. 3. Typical variations in kinematic and kinetic parameters for the same female subject in (A) static trial and (B) dynamic-relaxed trial. During the static trial, the subject self-initiates her recovery into a stable upright position by increasing her ankle torque and thereby moving her center-of-pressure (COP) forward. This, in turn, causes her body lean angle to decrease. In the dynamic-relaxed trial, the dashed line shows the instant of brake release, which is followed immediately by a sharp decline in tether force. After release, the body starts to rotate downward, and after a time delay that averaged 99 ms, ankle torque begins to increase, which allows eventually for a halting of downward rotation and return to a stable upright configuration. Note that, while the peak magnitude of ankle torque is nearly the same in the 2 trials, the initial lean angle is $\sim 17^\circ$ in the static trials and 7° in the dynamic-relaxed trial.

the tether was unexpectedly released following a random delay (Fig. 3b). Since the brake release time was small (~ 15 ms), this caused a near-step increase in the gravitational torque acting to rotate the body downward. Furthermore, we conducted dynamic trials at two levels of baseline plantar-flexor torque (T_i). In “dynamic-relaxed” trials, we instructed subjects before release to simply “relax your ankles.” In subsequent “dynamic contracted” trials, we used an oscilloscope to monitor the location of the center-of-pressure (COP) between the foot and the ground and instructed subjects to adjust the forward (i.e., anterior) excursion of their COP from the ankle to approximately one-third the peak value observed during their static trials. (Since the anterior excursion of the foot COP primarily determined ankle plantar-flexor torque, this resulted in a baseline ankle torque of about 33% of the peak value observed during static trials.) In all dynamic trials, we detected the instant of brake release as the onset of a sharp decline in the tension measured by a load cell (Sensotec, model 31) located in series with the tether.

Data analysis

We calculated the body lean angle $\theta(t)$ as the angle from the vertical to a line connecting the midpoint of the two lateral malleolus markers to the midpoint of the two acromium markers. We also calculated temporal variations in ankle plantar-flexor torque $T_a(t)$ based on the location and magnitude of vertical and horizontal components of foot reaction force

$$T_a(t) = F_z(t)x(t) - F_x(t)z(t)$$

where F_z is the resultant vertical force acting on the foot (defined positive if upward), x is the horizontal distance from the ankle joint where F_z acts (defined positive if anterior to the ankle), F_x is the resultant horizontal force in the sagittal plane acting on the foot (defined positive if directed posteriorly), and z is the vertical height of the ankle above the ground. Note that the above equation does not include components of ankle torque that balance moments associated with angular acceleration of the feet, which we found through inverse dynamics to be negligible.

For each of the three to five trials acquired at the subject’s maximum initial lean angle involving successful balance recovery, we calculated θ_{\max} as the average value of θ_i over the 500-ms interval preceding tether release. We also determined the maximum ankle torque (T_{\max}) generated during balance recovery (Fig. 4B). Furthermore, in dynamic-relaxed and dynamic-active trials we determined the following: 1) the magnitude of ankle torque before release (T_i), calculated as the average value of $T_a(t)$ over the 500 ms preceding release; 2) the ankle torque response time (Δt), calculated as the interval between release and the instant $T_a(t)$ exceeded T_i by $5 \text{ N} \cdot \text{m}$ (selected to be greater than the amplitude of fluctuations in ankle torque before the instant of release); 3) the rate of ankle torque generation following release (C), defined as the slope of a straight line joining torque-time values at the instant $T_a(t)$ exceeded T_i by $5 \text{ N} \cdot \text{m}$ to the instant $T_a(t)$ equaled T_{\max} ; and 4) the rate of ankle torque decline (D) following T_{\max} , defined as the slope of a straight line joining torque-time values at the instant of T_{\max} and 1000 ms later. We normalized C , D , T_i , and T_{\max} by the product of body mass (in kg) \times body height (in m). Values of T_i , Δt , C , T_{\max} , and θ_{\max} used in statistical analysis were averages, over the three to five repeated trials, for each subject and trial type.

Statistics

We used repeated-measures analysis of variance (ANOVA) to determine whether θ_{\max} and T_{\max} associated with trial type (static, dynamic-relaxed, and dynamic-active). If significant associations were detected, we conducted multiple comparisons with paired t -tests. We also used paired t -tests to determine whether average values of C and Δt differed between dynamic-relaxed and dynamic-active conditions. Finally, we used correlation to test for associations between continuous variables. The total number of P values we examined was 18. Based on Bonferroni considerations, to maintain a final (study-wide) level of significance of 0.05, we regarded P values from individual tests to signify significance if $P < 0.003$ ($0.05/18$).

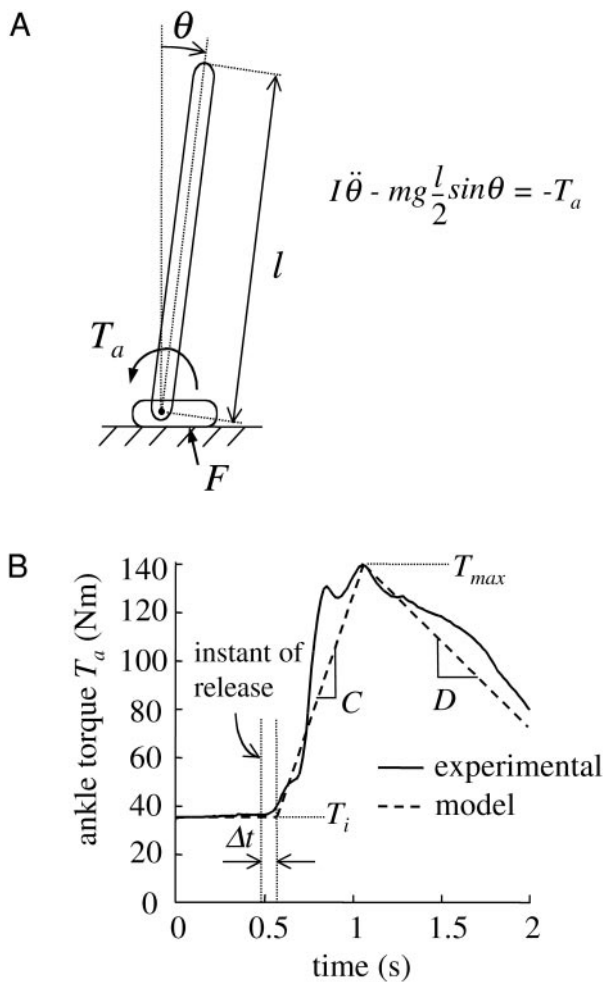


FIG. 4. Inverted pendulum model of balance recovery. A: the body is represented by an inverted pendulum of mass m and moment of inertia I . The total height of the pendulum is l , and the distance from the pivot to the center of gravity of the pendulum is $l/2$. The resultant foot reaction force is F ; the ankle torque is T_a , and the angle of the pendulum from the vertical is θ . The equation of motion for this system is shown to the right of the schematic. B: the ankle torque actuator has properties representing measured values of the baseline torque before release (T_i), the peak torque after release (T_{max}), the time delay between release and the onset of increased torque generation (Δt), the rate of torque generation following release (C), and the rate of torque decline (D) following the occurrence of T_{max} . Simulations explored how T_i , T_{max} , Δt , C , and D affected the maximum value of θ where balance recovery was possible.

Mathematical model

Our model consists of a single-link inverted pendulum “body” supported on a stationary foot, with a torque actuator at the ankle (Fig. 4A). The pendulum is released from an initial lean angle with zero initial velocity. Its downward rotation $\theta(t)$ is determined by numerically integrating the following equation of motion (using MATLAB, The MathWorks, Natick, MA)

$$I\ddot{\theta}(t) = mg \frac{l}{2} \sin \theta(t) - T_a(t)$$

where I and m are the mass moment of inertia (in kg m^2) and mass (in kg), $l/2$ is the distance (in m) from the pivot to the center of gravity of the pendulum, g is the gravitational constant (9.81 m/s^2), and $T_a(t)$ is the time-varying ankle torque (in $\text{N} \cdot \text{m}$), with properties T_i , T_{max} , Δt , C , and D (Fig. 4b).

To determine the model’s ability to predict experimental trends, we

set m , l , and I to mean experimental values (69.2 kg , 1.68 m , and 65.1 kg m^2 , respectively) and T_i , T_{max} , Δt , C , and D to mean dynamic-relaxed or dynamic-active values (Table 1). We then conducted simulations to determine the greatest initial lean angle (θ_{max}) where balance recovery was predicted to occur (defined by the occurrence of $\theta < 0$ while $\theta < 90^\circ$) within a resolution of 0.2° and compared these predicted recovery limits to those observed experimentally.

To predict the effect on θ_{max} of isolated or combined variations in strength and speed-of-response variables, we conducted simulations where T_i , Δt , C , and T_{max} (alone or in combination) were varied over the approximate range of experimentally observed values, while maintaining the remaining parameters equal to mean experimental values in dynamic-relaxed trials ($30.7 \text{ N} \cdot \text{m}$ for T_i , 99 ms for Δt , $372 \text{ N} \cdot \text{m/s}$ for C , $44.6 \text{ N} \cdot \text{m/s}$ for D , and $114 \text{ N} \cdot \text{m}$ for T_{max}). For each set of parameters, we again determined the greatest initial lean angle (θ_{max}) where balance recovery was predicted.

RESULTS

Experimental findings

We found that mean θ_{max} values were significantly larger in static trials than in dynamic trials (Table 1, Figs. 3 and 5A). The difference in mean values of θ_{max} between static and dynamic-relaxed trials was 8.9° (95% CI: 8.2 – 9.6 deg, $t = 26.3$, $df = 22$, $P < 0.001$), and the ratio of dynamic-relaxed to static θ_{max} averaged 0.40 ± 0.08 (mean \pm SD).

We also found that θ_{max} was significantly larger in dynamic-active than dynamic-relaxed trials. The difference in mean values of θ_{max} between dynamic-active and dynamic-relaxed trials was 0.9° (95% CI: 0.5 – 1.4° , $t = 4.4$, $df = 22$, $P < 0.001$), and the ratio of dynamic-relaxed to dynamic-active θ_{max} averaged 0.89 ± 0.14 .

Finally, we found that values of θ_{max} in static trials did not associate with those in dynamic-relaxed trials ($r = 0.19$; $P = 0.38$), or with those in dynamic-active trials ($r = 0.29$; $P = 0.18$).

Foot length did not associate with values of θ_{max} in static trials or in dynamic trials ($P > 0.8$). This was likely due to the relatively strong association between foot length and body height ($r = 0.88$; $P < 0.001$). In other words, while individuals with larger feet may have been able to recover from greater

TABLE 1.

| Parameter | Static | Dynamic-Relaxed | Dynamic-Active |
|----------------------|-------------------------------------|--|---|
| n | 23 | 23 | 23 |
| θ_{max} (deg) | 14.9 ± 1.4 | $5.9 \pm 1.1^\dagger$ | $6.9 \pm 1.7^\dagger, \ddagger$ |
| T_i (Nm)* | NA | 31 ± 18 (0.23 ± 0.11) | $54 \pm 24^\ddagger$ (0.41 ± 0.12) |
| Δt (ms) | NA | 99 ± 13 | 90 ± 20 |
| C (Nm/s)* | NA | 372 ± 267 (3.0 ± 2.2) | 319 ± 225 (2.7 ± 1.9) |
| D (Nm/s)* | NA | 44.6 ± 35.8 (0.36 ± 0.27) | 48.6 ± 32.8 (0.17 ± 0.13) |
| T_{max} (Nm)* | 116 ± 32 (0.92 ± 0.11) | 114 ± 32 (0.90 ± 0.13) | 118 ± 33 (0.93 ± 0.12) |

Values are mean parameters \pm SD. n is number of subjects. NA, non-applicable entries. * Cell entries display mean parameter values followed in parentheses by mean normalized values, where normalized parameter value = parameter value/[body mass (in kg) \times body height (in m)]. † Significantly different than static value ($P < 0.001$). ‡ Significantly different than dynamic-relaxed value ($P < 0.001$).

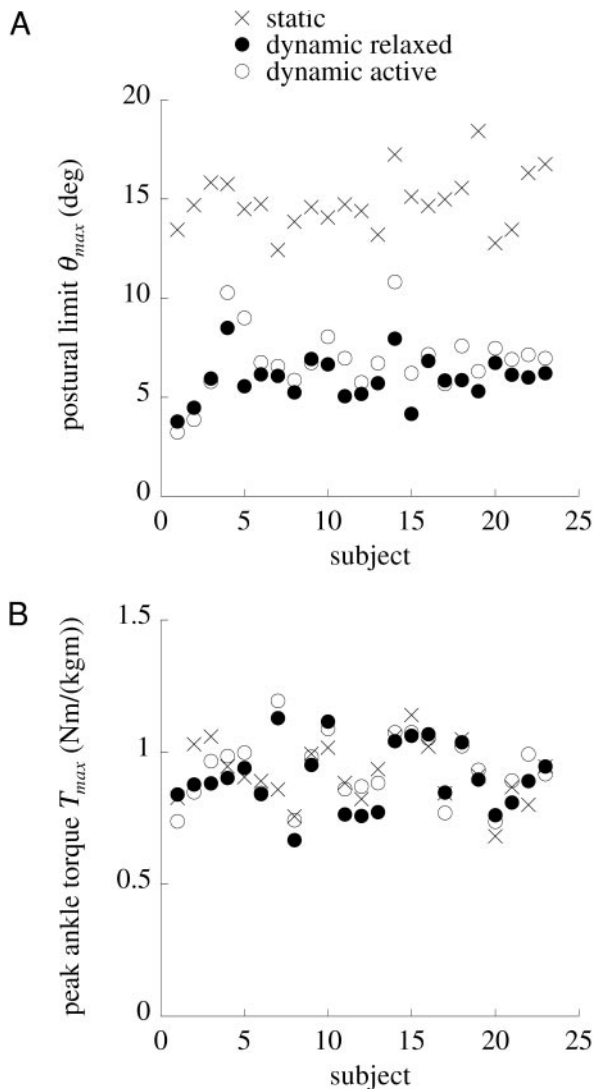


FIG. 5. Comparison between (A) maximum recovery limits and (B) peak ankle torque in static trials (X), dynamic-relaxed trials (filled circles), and dynamic-active trials (open circles). Dynamic-relaxed recovery limits averaged $40 \pm 8\%$ of static recovery limits. However, peak ankle torques were not statistically different between the series. This suggests that latencies in the onset and finite rates of torque generation strongly influence recovery ability.

horizontal excursions of the COG, they also had a corresponding increase in the height of their COG and therefore no greater θ_{max} than individuals with smaller feet.

Mean values of T_{max} were not different in the three trial types ($P = 0.27$; Fig. 5B), suggesting that differences in θ_{max} between static and dynamic trials reflect the effect on recovery ability of variables related to the speed (rather than the magnitude) of ankle torque generation. Furthermore, while T_i increased between dynamic-relaxed and dynamic-active trials [mean increase = $0.18 \text{ N} \cdot \text{m}/(\text{kg} \cdot \text{m})$, 95% CI: $0.15\text{--}0.21 \text{ N} \cdot \text{m}/(\text{kg} \cdot \text{m})$, $t = 13.8$, $\text{df} = 22$, $P < 0.001$], there was no difference between mean values of Δt in dynamic-relaxed and dynamic-active trials (mean difference = 4 ms , 95% CI: $-1\text{--}10 \text{ ms}$, $t = 1.7$, $\text{df} = 22$, $P = 0.11$) or between mean values of C in dynamic-relaxed and dynamic-active trials [mean difference = $0.36 \text{ N} \cdot \text{m}/(\text{s} \cdot \text{kg} \cdot \text{m})$, 95% CI: -0.27 to $0.99 \text{ N} \cdot \text{m}/(\text{s} \cdot \text{kg} \cdot \text{m})$, $t = 1.2$, $\text{df} = 22$, $P = 0.25$]. This

suggests that differences in θ_{max} between dynamic-relaxed and dynamic-active trials reflect the effect on recovery ability of baseline ankle torque prior to release.

In dynamic trials, there was correlation between T_{max} and C ($r = 0.69$, $P < 0.001$ in dynamic-relaxed trials; $r = 0.55$, $P = 0.007$ in dynamic-active trials), but not between T_{max} and T_i ($r = 0.09$, $P = 0.70$ in dynamic-relaxed trials; $r = 0.20$, $P = 0.36$ in dynamic-active trials) or between T_{max} and Δt ($r = -0.26$, $P = 0.23$ in dynamic-relaxed trials; $r = -0.28$, $P = 0.20$ in dynamic-active trials). Furthermore, there was no correlation between Δt and C ($r = -0.34$, $P = 0.11$ in dynamic-relaxed trials; $r = -0.11$, $P = 0.63$ in dynamic-active trials), between Δt and T_i ($r = -0.03$, $P = 0.90$ in dynamic-relaxed trials; $r = -0.40$, $P = 0.06$ in dynamic-active trials), or between T_i and C ($r = -0.33$, $P = 0.12$ in dynamic-relaxed trials; $r = -0.21$, $P = 0.34$ in dynamic-active trials).

Mathematical model predictions

There was generally good agreement between experimental and mathematical model predictions of recovery limits (Fig. 6). When all parameters were set to mean experimental values for the dynamic-relaxed case, the model predicted θ_{max} to equal 7.1° . This was within $\pm \text{SD}$ of the experimental mean of $5.9 \pm 1.1^\circ$. When all parameters were set to mean experimental values for the dynamic-active case, the model predicted θ_{max} to equal 8.5° . This was again within $\pm \text{SD}$ of the experimental mean of $6.9 \pm 1.7^\circ$.

However, experimental recovery limits exhibited considerable scatter when plotted against T_i , T_{max} , Δt , and C (Fig. 6). This may reflect subjects tendency to compensate for deficits in one parameter through enhancements in others. In contrast, model predictions of θ_{max} increased in a near-linear fashion with isolated increases in T_{max} [increasing by 64% from 5.3 deg for $T_{max} = 0.60 \text{ N} \cdot \text{m}/(\text{kg} \cdot \text{m})$ to 8.7° for $T_{max} = 1.20 \text{ N} \cdot \text{m}/(\text{kg} \cdot \text{m})$], with increases in T_i [increasing by 70% from 5.2° for $T_i = 0$ to 8.8° for $T_i = 0.48 \text{ N} \cdot \text{m}/(\text{kg} \cdot \text{m})$], and with decreases in Δt (increasing by 20% from 6.6° for $\Delta t = 150 \text{ ms}$ to 7.9° for $\Delta t = 50 \text{ ms}$). The predicted effect of C on θ_{max} was logarithmically shaped, with a strong dependency between these variables predicted for small but not large values of C [θ_{max} increased by 74% from 4.3° for $C = 0.40$ to 7.5° for $C = 4.0 \text{ N} \cdot \text{m}/(\text{kg} \cdot \text{m})$, but only 8% from 7.5° for $C = 4.0$ to 8.1° for $C = 12.0 \text{ N} \cdot \text{m}/(\text{kg} \cdot \text{m})$].

Results from additional mathematical model simulations indicate that the relationship between C and θ_{max} depends on the rate of torque decline D after the occurrence of T_{max} (Fig. 7). In particular, if ankle torque is maintained constant after T_{max} is reached (or if it declines at a relatively small rate, as observed in our experiments), then θ_{max} will always increase with increasing C . If, however, ankle torque rapidly declines to zero after T_{max} is reached, there will be an optimal value of C , above which the predicted value of θ_{max} is smaller, due to an insufficient duration where torque is large (and thus able to halt downward movement). Together, these results suggest that recovery limits depend on capacity to quickly generate and maintain high magnitudes of ankle torque.

Predicted recovery limits were affected more profoundly by combined deficits than by isolated variation in a single parameter (Fig. 8). For example, a 50% decrease in T_{max} [from 0.91 to $0.45 \text{ N}/(\text{kg}/\text{m})$] reduced θ_{max} by 39% (from 7.1 to 4.3°).

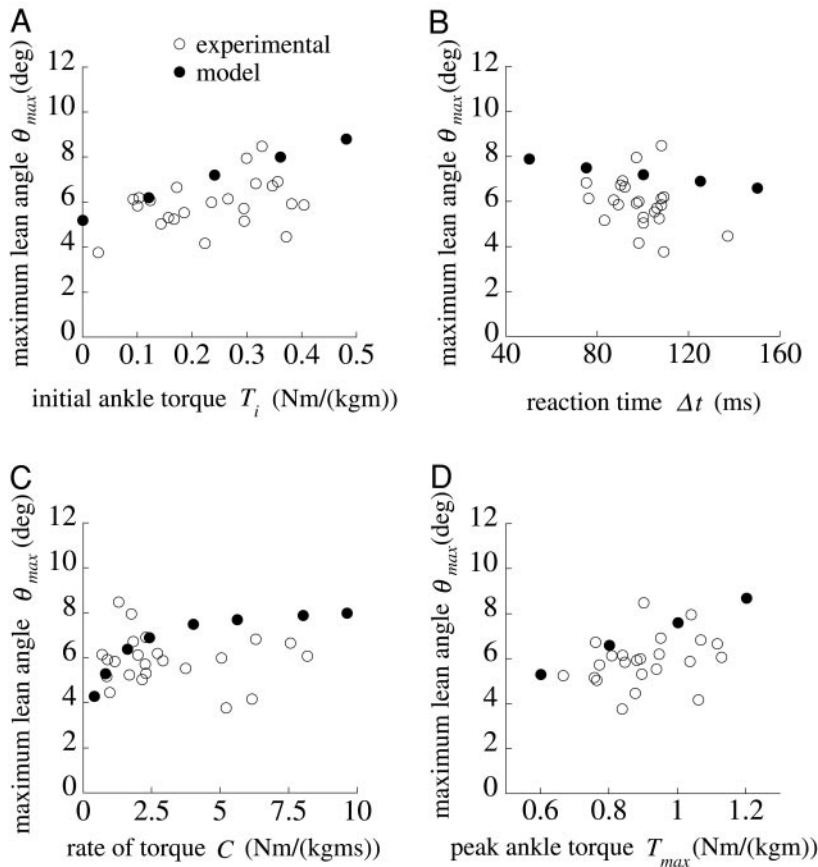


FIG. 6. Mathematical model predictions (filled circles) and experimentally observed trends (open circles) between θ_{\max} and (A) T_i , (B) T_{\max} , (C) C , and (D) Δt . Experimental values of θ_{\max} exhibited considerable scatter when plotted against T_i , T_{\max} , C , and Δt , suggesting that subjects offset deficits in 1 motor variable with enhancements in others. Model predictions indicate the theoretical effect on θ_{\max} of isolated changes in variables related to strength (T_i , T_{\max}) and speed-of-response (Δt , C).

A simultaneous decline of 50% in T_i [from 0.25 to 0.12 N · m/(kg · m)] reduced θ_{\max} by 48% (to 3.7°), and a concomitant doubling of Δt (from 99 to 198 ms) reduced θ_{\max} by 56% (to 3.1°).

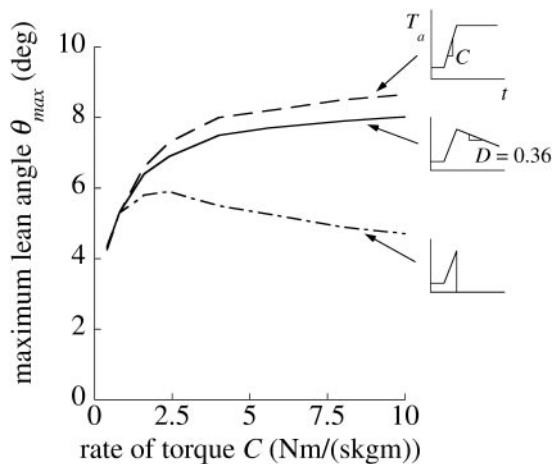


FIG. 7. Mathematical model predictions of the relationship between the rate of torque increase (C) and the maximum recovery limit (θ_{\max}), and the effect on this relationship of the subsequent rate of torque decline (D). Three cases are shown, corresponding to 1) no decline in ankle torque following the occurrence of T_{\max} (dashed line); 2) a decline in ankle torque following T_{\max} equal to the average experimental value of 0.36 Nm/skgm (solid line); and 3) an instant decline to zero ankle torque following T_{\max} (dash-dot line). Recovery ability depends on the ability to both generate and maintain high magnitudes of ankle torque. See text for further discussion.

DISCUSSION

Our results indicate that human ability to recover balance following an unexpected perturbation is limited substantially by nonzero delays in the onset and finite rates of torque generation. If, following release during dynamic trials, subjects could have instantly increased their ankle torque to its peak magnitude (T_{\max}), there should have been no difference between peak recovery angles in dynamic and static trials (since there was no difference in T_{\max} between these series). Instead, peak recovery limits in dynamic trials were less than one-half the magnitude of peak recovery limits in static trials. This suggests that parameters related to the speed of torque generation (Δt and C) reduce by about one-half the magnitude of the perturbation where one can recover balance using the ankle strategy.

We also found that peak recovery limits in dynamic trials did not associate with those in static trials. This suggests that static techniques to assess postural limits, such as Functional Reach (Duncan et al. 1990), may provide little insight on the ability to recover balance following a sudden perturbation.

We did find, however, that recovery ability increased with increases in the baseline magnitude of ankle torque (T_i) before the onset of the perturbation. The reason for this was likely twofold. First, higher T_i should have reduced the body's downward acceleration following release, and second, it should have allowed subjects to attain T_{\max} more quickly (since T_{\max} , Δt , and C did not differ between the two series). This may explain why appropriately shifting our center of pressure (and thus baseline ankle torque) enhances our ability to resist perturbations.

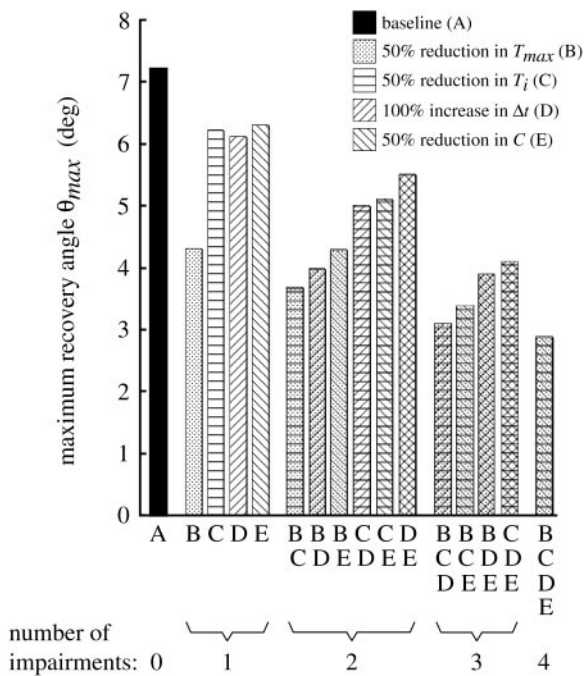


FIG. 8. Model predictions of the effect on θ_{max} of isolated and combined changes in T_i , T_{max} , C , and Δt . See text for discussion.

Predictions from our inverted-pendulum model complement experimental trends by showing that theoretical recovery limits increase in a near linear fashion with isolated increases in T_i and T_{max} and with isolated decreases in Δt . Our model also predicts that, over the range of parameter values observed experimentally, peak recovery limits in dynamic trials are affected at least as much by isolated variations in T_{max} as by isolated variations in Δt or C . Therefore exercise-based increases in ankle strength (Fatarone et al. 1994; Judge et al. 1994) should improve participants' recovery limits.

Finally, our model illustrates the cumulative effect on recover limits of simultaneous declines in both the strength and the speed of ankle torque generation. Several studies have shown that aging causes changes in each of these areas. For example, Vandervort and Hayes (1989) observed average declines of 44% in peak rates of plantar-flexor torque generation, and 71% in peak magnitudes of plantar-flexor torque, for females between mean ages of 26 and 82 yr. Similarly, Thelen et al. (1996) observed average declines of 36% in peak rates of plantar-flexor torque generation, and 32% in peak attainable magnitudes of plantar-flexor torque for females between ages 23 and 74 yr. Moreover, several studies have shown that simple reaction times increase on average by about 25% between the third and seventh decades of age (Schultz 1992; Welford 1988).

Several limitations exist to this study. First, in this preliminary study we did not directly explore how recovery limits associate with variables such as muscle architecture and fiber-type composition, muscle force-length and force-velocity properties, and the intactness of proprioceptive and vestibular afferents. Second, we released subjects from a stationary incline, and real-life loss-of-balance episodes often occur during activities such as walking or rising from a chair, where the initial velocity of the body's center of gravity is nonzero (Pai and Patton 1997). However, we can see little reason why our main

conclusions would not apply for a wide range of center of gravity velocities at the onset of imbalance. Third, the accuracy of our measured recovery limits may have been affected by constraints on the number of iterations of the initial lean angle that we could reasonably perform, or by subjects' motivation to perform to their maximum ability. Fourth, subjects' performance may have also been affected by the degree of co-contraction involved in achieving a given baseline ankle torque (which we did not control) or by partial reliance on the hip strategy to recover balance. We attempted to eliminate the latter possibility by visually inspecting recovery responses during data acquisition and repeating trials which involved obvious knee and/or hip flexions. In post-hoc analysis, we calculated peak hip flexion rotations during recovery (which for dynamic-relaxed trials ranged between 1.5 and 15.6° and averaged $8.1 \pm 3.4^\circ$) and found that these did not associate with dynamic recovery limits ($R = 0.05$, $P = 0.84$). This suggests that subjects were not relying substantially on hip flexion to recover balance. However, this does not imply that the trunk and hip muscles had no role in balance recovery. Rather, it was essential that subjects use these muscles to minimize relative motion between the trunk and the lower extremities following release. Accordingly, the ability to successfully couple the dynamic response of trunk and ankle muscles may have substantially influenced recovery limits. Finally, our inverted pendulum model does not simulate lifting of the heels off the ground during balance recovery, and this might explain some of the variability in experimental data not accounted for by the model. However, we doubt this was substantial, since the amount of heel rise observed in our experimental trials tended to be small (~ 2 cm) and appeared to have a minimal effect on ankle torque generation (Fig. 2).

A further limitation of the study is that static recovery limits may have been affected by nonzero forces in the tether during the initial period of recovery (since, even though the tether was inextensible, compliance existed in the soft tissues it contacted). Theoretically, static recovery limits should have equaled $\theta_{max} = \sin^{-1} [2T_{max}/(mgl)]$ [where T_{max} is the maximum ankle torque observed in static trials (in $N \cdot m$), m is body mass (in kg), and l is body height (in m)], or 10.9 ± 1.3 deg. Instead, they were $27 \pm 9\%$ higher, suggesting that hip rotations and/or tether forces did affect measured static recovery limits. However, these theoretical static recovery limits remain $89 \pm 44\%$ higher than dynamic-relaxed recovery limits and $68 \pm 52\%$ higher than dynamic-active recovery limits. Accordingly, this experimental limitation could not invalidate the main conclusions of our study.

Finally, we focused on the ankle strategy, which is one of several possible strategies for preventing a fall in the event of a destabilizing perturbation (Horak et al. 1989; Hsiao and Robinovitch 1999; McIlroy and Maki 1996; Pai et al. 1998; Tang and Woollacott 1998). Evidence suggests that "natural" balance recovery responses involve a combination of ankle and hip strategies and that elderly subjects rely more than the young subjects on the hip strategy to recover balance (Manchester et al. 1989; Woollacott 1993). Furthermore, the selection of a specific balance recovery response appears to depend not only on biomechanical variables (such as strength and reaction time), but also on behavioral and environmental variables. For example, Maki and McIlroy (1997) found that stepping-based strategies for balance recovery tend to be in-

voked well before recovery limits are actually reached. This may explain why Hall and co-workers (1999) found that, in the event of forward or backward displacement of the support surface, neither the magnitude nor the rate of ankle torque production associated with the use of sway versus stepping-based balance recovery responses. Instead, this was presumably dictated by fear, cautiousness, or habit.

Despite these limitations, we believe that the recovery limit experiment provides the investigator or clinician with a previously unavailable technique to determine (by measuring the ratio of static to dynamic recovery limits) how an individual's ability to recover balance is affected by strength versus speed of response. It is important to recognize that this information is distinct and complementary to measures of sway during quiet stance (Baloh et al. 1994; Nashner and Peters 1990), which may be thought of as characterizing risk for loss-of-balance more than ability to recover balance, from static measures of balance performance such as Functional Reach (Duncan et al. 1992; Wernick-Robinson et al. 1999), and from behavioral and performance-based measures of balance recovery by stepping (Luchies et al. 1994; McIlroy and Maki 1996; Wolfson et al. 1986) or grasping (Maki and McIlroy 1997). It is our hope that, by appropriately using these various assessment tools, we will be better able to identify the behavioral and neurophysiological parameters that must be targeted to reduce a given individuals' risk for falls and to design and monitor the effectiveness of programs for achieving this.

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