Thresholds for Inducing Protective Stepping Responses to External Perturbations of Human Standing

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INTRODUCTION

From a hierarchy of control strategies, the most appropriate is selected according to the size of the postural challenge. This is a commonly accepted model to describe the control of upright human standing. Corrective strategies, which maintain the base of support (BoS) (Stelmach and Worringham 1985), are used when disturbances are small. The ankle strategy is dominant for the smallest disturbances, whereas the hip strategy is more important when disturbances are larger or when the BoS is limited (Horak and Nashed 1986; Nashed and McCollum 1985; Shumway-Cook and Woollacott 2001). Protective strategies, the most common of which is stepping, are considered to be last-resort responses that are initiated when the center of mass (CoM) of the body passes outside the limits of the BoS so that balance cannot be maintained without reconfiguring the BoS by active movements such as stepping (Shumway-Cook and Woollacott 2001).

This hierarchical model of strategy selection according to the amplitude of the postural challenge is at odds with observations that subjects step more often than expected by this criterion when they are instructed to react naturally (Brown et al. 1999; Maki and McIlroy 1997; Rogers et al. 1996) and that stepping is often initiated very rapidly well before the limits of the BoS are reached (Pai et al. 2000). Thus stepping does not depend solely on the mechanical constraints of the position of the CoM exceeding the BoS. A more complex mechanical model that includes the velocity of the CoM and momentum of the body rather than just the position of the CoM is an improvement (Pai and Patton 1997) and more reliably predicts stepping during waist-pull and support-surface perturbations (Pai et al. 1998, 2000). However, subjects still step more often than predicted by this model and, on average, older individuals step more often and with smaller perturbations than younger adults (McIlroy and Maki 1993; Pai et al. 1998; Rogers et al. 2001b). This suggests that physiological factors, at this stage unknown, rather than mechanical factors are likely to determine step initiation.

Age-related changes in the sensorimotor and neuromuscular physiology critical for postural control are the likely candidates to explain the different stepping behavior in older people. Visual changes include reduced acuity, contrast sensitivity, and depth perception (Pitts 1982). Somatosensory changes include reduced proprioceptive, touch-pressure, and vibration sensitivity, particularly in the lower limbs (Kenshalo 1986; Skinner et al. 1984). Neuromuscular changes with aging include reduced muscle strength and mass (Merletti et al. 2002; Vandervoort 2002) and delayed reaction times (Welford 1988). All of these factors are associated with balance disorders and an increased risk of falls (Lord et al. 1991).

The present study investigates the displacement–velocity threshold for triggering protective stepping during waist-pull perturbations of stance. When a postural perturbation occurs, the CoM–BoS relationship is disrupted and the subject must...
perform compensatory strategies to maintain postural balance. Regardless of the exact site where the perturbation may be applied, the common key relationship among various perturbation methods is the change in orientation of the CoM relative to the BoS. It is likely that differences between subjects in their abilities to compensate for disturbances of balance reside in their control of protective stepping responses, and other strategies, rather than in the means of perturbation that precipitates the condition of falling. Waist-pull perturbations were used to alter this relationship because of the precise control over the mechanical perturbation parameters that the device provided (Pidcoe and Rogers 1998). Young and older adults are studied to identify physiological factors associated with step initiation.

**Methods**

**Subjects**

Twenty-four young subjects, 18 women and 6 men, aged 21–37 yr (mean age: 25.3 ± 4.2; weight: 68.1 ± 14.9 kg; height: 1.68 ± 0.09 m) were recruited from the staff and students of the university. Twenty-five older subjects, 15 women and 10 men, aged 60–85 yr (age 71.8 ± 7.0; weight: 71.1 ± 13.2 kg; height: 1.64 ± 0.11 m) were recruited from among community-dwelling elderly people. Subjects underwent a standardized telephone screening administered by a physical therapist and were included only if they had no significant neurological (stroke, Parkinson’s disease, peripheral neuropathy), musculoskeletal (joint replacement, leg or back pain, or any other limitations), or other medical disorders (depression, cardiac, metabolic, or respiratory problems) that affected functional movement performance. Older subjects were included only if they had no previous history of falls. The Institutional Review Board of the Northwestern University in accordance with the 1964 Helsinki Declaration approved the study, and subjects provided written informed consent.

**Protocol**

STEPPING RESPONSES. Subjects stood on two separate force platforms (AMTI, Newton, MA) that recorded forces under each foot. They took a natural and comfortable foot position that was then maintained for each trial. Stepping was induced by a constant-velocity forward pull from a linear motor, delivered by a flexible cable that was attached to a belt that subjects wore at the level of the pelvis (Fig. 1A. Pidcoe and Rogers 1998). The pulls started at an unexpected time and proceeded at the test velocity until, on reaching test displacement, the cable tension was released and subject can lean back if a step has not already been initiated. This is illustrated in profile of displacement of the pull (D) as function of time. The initial position of subject (I) is estimated from the initial position of center of pressure (CoPi) relative to the ankles. D is added to CoPi to estimate the final position of pelvis (F) at end of the pull relative to the ankle (P). This reflects how far in front of the axis of the ankles the pelvis is at end of the pull. B: velocities and displacements of 124 pulls, delivered as two sets of 62 (triangles first, then circles). It was difficult not to step for the pulls with large displacement and velocity shown as open circles. These were not delivered in the second set if, in the first set, subjects stepped for all pulls around them. The eight practice trials are shaded. Subjects stepped for some of these depending on their individual threshold.

SENSORIMOTOR FUNCTION. Six tests of sensorimotor function considered to be important for postural control and having good test-retest reliability (Lord et al. 1991; Lord and Castell 1994) were administered over 1 h. The tests were 1) vibration sense, 2) touch-pressure sensation, 3) proprioception, 4) visual acuity, 5) ankle plantar-flexion strength, and 6) foot voluntary reaction time.

Vibration sense at the tibial tuberosity was measured with an amplitude-controlled vibrator at 200 Hz. The vibrator had a 10-mm hard rubber dome that rested on the skin with a constant 1 N background force. In each trial, the vibration amplitude (0–100 μm) was either slowly increased from an imperceptible level until the subject reported the vibration or slowly decreased from a clearly perceptible level until the subject reported that the vibration had ceased. Threshold for perception of vibration was calculated as the mean intensity of six alternate trials on the dominant leg.

Touch-pressure sensation at the lateral malleolus was measured with a Semmes–Weinstein aesthesiometer. This consists of a set of 20
Ankle plantar-flexion strength was measured in the dominant leg during maximal isometric and isokinetic contraction with a Biodex System 3 Pro (Biodex Medical Systems, Shirley, NY) sampled at 100 Hz. The dominant leg was that which the subject would use to kick a ball. Subjects sat in a chair with the trunk inclined at 70°, the thigh approximately horizontal, the knee positioned at 30° of flexion and the ankle at 90°. The foot was firmly strapped to a plate. Isometric plantar-flexion strength was measured by instructing subjects to “push the plate as fast and hard as you can” for 5 s. They performed 3 trials of 5-s duration with 10 s of rest between each trial. The best of the 3 trials was chosen as the plantar flexion peak torque, and the torque rate was calculated as the peak torque divided by the time to peak. Isokinetic plantar-flexion strength was tested at 60 deg.s \(^{-1}\). Subjects were instructed to “push and pull the plate as hard and fast as you can.” Plantar-flexion strength was chosen as the mean of the three highest peak torques for the five repetitions. Because strength is critically linked with the load it has to move, the ankle strength values were normalized to the weight \( \times \) height product of each subject for statistical analysis. The assumption is that the required torque to move the body around the ankles is proportional to the mass and the height of the CoM.

Simple reaction time was measured as the mean of 10 trials in which subjects used the dominant foot to react to a bright red 10-mm LED. A variable delay period separated repeated presentation of the stimulus cue.

Data analysis

With small amplitude perturbations, the body tends to behave like a simple-inverted pendulum, whereas with larger amplitude it behaves more like a double-inverted pendulum (Nashner and McCollum 1985). Nevertheless, the final position of the pelvis at the end of the perturbation reflects the initial position of the body and the size of the perturbation. This initial position relative to the ankles (Fig. 1A, CoPi) can be estimated from the mean CoP position during the second before the pull. Because previous experiments have shown a comparable motion profile between the pelvis and the motor for this device (Piddock and Rogers 1998), the final position of the pelvis at the end of the perturbation was calculated for each trial by adding the displacement of the pull (D) to the initial position of the CoP (CoPi). This position (Fig. 1A, P) was calculated in reference to the ankle position and thus it reflects how far forward the pelvis is from the ankles at the end of the pull. This position, the velocity of the pull, and the response to step or not, characterized each trial. From these data, a total stepping record for the array of perturbations was compiled for each subject.

INDIVIDUAL THRESHOLD BOUNDARIES. For each velocity for which it was possible, a position threshold was determined as the midpoint between the largest value of position (P) (i.e., the most forward position) without a step and the smallest value of P (i.e., the closest position from the ankle) with a step. After inspection of these data, a sigmoidal relationship between position and velocity was apparent. A Boltzmann sigmoidal function was fitted to these position points (Gauss–Newton, nonlinear, least-mean-squares) to describe a position threshold for stepping as a function of velocity (Fig. 2B, equation and graph). The four parameters estimated were the threshold plateaus at low \((T_L)\) and high \((T_H)\) velocities, the velocity \((V_{so})\) that produces the midrange response, and the slope \((S)\) of the relation at midrange.

GROUP STEPPING BOUNDARIES. Ideally, a threshold, or midpoint, could be calculated for each of the 23 velocities for each subject. However, some subjects never stepped or stepped for all pulls at extreme velocities so the midpoint was undefined. Therefore the individual threshold boundary for a subject was not calculated if <12 of the 23 threshold points were defined. This excluded five young subjects from the individual analysis. In addition, the threshold boundary for a subject was also not included in the individual analysis if the range of perturbations was insufficient to capture the sigmoidal shape. The criterion for this was if \( r^2 < 0.70 \) for the Boltzmann function fit. This excluded seven older subjects from the individual analysis. To include these missed subjects in the analysis, a position boundary for producing 50% stepping for the group was calculated from the stepping frequencies for all subjects to all perturbations. For each velocity, the position at which the subjects stepped in 50% of the
trials was estimated, and these points were then fitted with the Boltzmann sigmoidal function to produce the group position boundary.

**Statistical Analysis.** To assess aging changes in stepping thresholds, each parameter of the equation was submitted to an analysis of variance with group as the between factor. Pearson correlation coefficients were calculated to assess the associations between the anthropometric, sensorimotor, and neuromuscular variables and the stepping parameters \( T_L, T_H, \) and \( V_{50} \). Partial correlations were then calculated to assess these associations while controlling for age group.

**Results**

Subjects were not restricted to stepping with a particular foot. Various patterns were observed, with 37% showing an absolute preference (100% of steps) for one foot and 53% a relative preference (70–99% of steps). These were similarly divided between the dominant (47% of subjects) and nondominant (43% of subjects) leg. Generally, subjects took a single step but some subjects took two or more. This was not quantified because it was uncontrolled and multiple steps may not necessarily be because of a postural need.

**Individual threshold boundaries**

Responses obtained during a full set of trials are shown in Fig. 2A for typical younger and older subjects. The demarcation between the position that causes subjects to step (shaded circles) or not (open circles) is obvious. Thus the position threshold points (diamonds), midway between the most anterior position without a step and the closest position from the ankle with a step, form a narrow band across the domain of pull velocities. Inspection of these data points indicates asymptotes at low and high velocities where the threshold is independent of velocity. Between these asymptotes, threshold is inversely related to the velocity.

To select a model to describe this threshold response, it is necessary to consider the physical effects of an external perturbation on stance. The fundamental requirement for standing equilibrium means that if the perturbation is large enough to move the CoM beyond the limits of the BoS, a step must be taken. This upper limit of the perturbation for stepping applies for all velocities, no matter how slow. As velocity increases, momentum becomes more significant. Thus the threshold position should move progressively closer to the ankle so that the momentum of the body produced by the forward pull can be overcome before the CoM reaches the anterior limit of the BoS. In actual subjects, another factor places a lower limit on the perturbations that will produce sufficient body movement to cause stepping. The energy of an external perturbation that is transferred to the body can take two forms: kinetic energy associated with the movement of the body; and potential energy, which may be elastic and held by deformation of the tissues and body segments, or gravitational through any elevation of the CoM of the body. Here, the gravitational energy should be trivial because the motor pulls horizontally at approximately the height of the CoM. However, as the amplitude of the displacement becomes smaller at high velocities, a greater proportion of the energy will be taken as elastic energy by the soft tissues and joints because of their lower short-range stiffness. Eventually, when all of the perturbation is inefficiently taken as potential energy, forward body movement will be insignificant so that there is no need to step. Thus this is a system with a region of transition between two limiting states that are independent of the input variable, velocity. Stepping will always occur when the perturbation pulls the CoM in front of the BoS, and stepping will never occur when the perturbation is below a critical level that does not require a response.

Both the theoretical considerations and the empirical observations illustrated in Fig. 2 indicate that a Boltzmann function of velocity would be ideal to model the threshold boundary for stepping. This sigmoidal function describes systems that have a transition between distinct response limits for high and low input values (Capaday 1997). It takes the form shown in Fig. 2B, and is described by four parameters: the high-velocity threshold \( T_{HI} \), the low-velocity threshold \( T_L \), the velocity \( V_{50} \) that produces the midrange response, and the slope \( S \) of the relation. The inverse of the slope is directly proportional to the maximal steepness of the function, which occurs at \( V_{50} \). Of the 49 subjects, 42 had an \( r^2 > 0.70 \) for the Boltzmann function fit, and the remaining seven failed to reach this criterion through insufficient data rather than an apparent different pattern of the threshold boundary.

**Group threshold boundaries**

The position and velocity data of each subject were expressed as %BoS and %BoS.s\(^{-1}\), respectively, and pooled to find a threshold boundary that would cause stepping in 50% of all trials for each group. The relative frequency of stepping for the group (0–1) is plotted against the position (%BoS) and velocity (%BoS.s\(^{-1}\)) for the young and old groups in Fig. 3. At all velocities, the group never stepped (frequency = 0) when the position of the pelvis was close to the ankles, whereas they stepped in most of the trials (frequency = 1) when the pelvis moved the pelvis beyond the limit of the BoS. On the slope between the base level (dark shading) and the plateau (white), stepping frequency increased as the final position increased anteriorly. At each velocity, the point halfway up the slope (frequency = 0.50) can be identified by fitting a Boltzmann function to the stepping frequency data with position as the independent variable. This function with its 0.50 point is illustrated for one velocity as \( F = f(v, P) \) in the top of Fig. 3. For all velocities tested, these points form across the contour of the sloped surface a curve that describes the position that produces stepping in half of the trials. Like the individual threshold boundaries, this position that produces 50% stepping for the group is also best described by a Boltzmann function of velocity (Fig. 3, “C”). These group functions, which we refer to as “position boundaries for 50% stepping,” are analogous to the individual threshold boundaries described above. The fit of Boltzmann functions to these group data were very accurate; \( r^2 = 0.995 \) for the young group and \( r^2 = 0.993 \) for the older group. In these plots, there are clearly well-defined low-velocity (S50\(_L\)) and high-velocity (S50\(_H\)) plateaus, providing further support for the choice of the Boltzmann model to describe the stepping behaviors of individual subjects.

**Age, stepping frequency, and thresholds**

The S50\(_L\), (74% versus 108% BoS) and the S50\(_H\), (52% versus 72% BoS) of the group boundaries for 50% stepping were both lower for the old than for the young. Likewise, the \( V_{50} \) of the group boundaries was a lower velocity (109% vs.
length of BoS. Similarly, $T_H$ was lower for the old than for the young, both in absolute terms [11.1 ± 1.8 vs. 13.9 ± 1.8 cm; $F_{(1,35)} = 23; P < 0.001$] and when normalized to the BoS length [59 ± 8 vs. 72 ± 8% BoS; $F_{(1,35)} = 23.3; P < 0.001$; Fig. 4, left].

The $V_{50}$, which indicates the velocity at which the transition between the low-/high-threshold boundary on the velocity scale, was lower for the older than for the younger subjects in absolute [17.8 ± 4.3 vs. 23.5 ± 5.1 cm.s$^{-1}$; $F_{(1,35)} = 13.4; P < 0.001$] and normalized units [96 ± 24 vs. 121 ± 26% BoS.s$^{-1}$; $F_{(1,35)} = 9.6; P < 0.005$, Fig. 4, right]. There was no significant difference in the slopes of the threshold boundaries between groups for the individual subject data [0.27 ± 0.20 vs. 0.36 ± 0.28, $F_{(1,35)} = 1.2; P = 0.28$].

The older subjects stepped more often than the younger subjects [69 ± 14 versus 40 ± 19% trials; $F_{(1,47)} = 38.7; P < 0.001$]. This was despite different initial positions, with the older group standing with the CoP 1.6 cm or 8% BoS further back toward the ankles [$F_{(1,47)} = 9.4; P < 0.005$]. Had they started in the same position, an even larger incidence of stepping would be expected for the older group.

Sensorimotor correlates of stepping

Table 1 shows the correlations of the sensorimotor and neuromuscular measures with $T_L$, $T_H$, and the total number of steps taken for the 124 perturbations. There were no significant correlations with the $V_{50}$. For each correlation, the Pearson $r$ (left) is the correlation between the stepping parameter and the sensorimotor measure for all subjects. The partial $r$ (right) is the partial correlation coefficient when age is included as an independent factor.

It can be seen that many of the sensorimotor measures are significantly associated with the threshold measures when the young and old group data are pooled. Of note, vibration sense and visual acuity are associated with all stepping parameters, isokinetic strength and rate of force development are associated with $T_H$, and all measures of strength are associated with the total number of steps. There is a consistent pattern in the signs of the correlation coefficients whereby the worse performances in physiological tests are associated with lower threshold pa-
rameters and a greater number of steps. However, none of these associations were present when age was included as a factor (partial r). Thus, although it is apparent that the stepping thresholds are affected by age-related factors, there is no evidence that the sensorimotor factors tested are directly causal in this.

**DISCUSSION**

By pulling on the body at the pelvis to perturb standing, we have identified a threshold that, when exceeded, induces stepping. The threshold is described by the position of the pelvis of the body in the sagittal plane relative to the BoS of the body, measured from the axis of the ankles. Faster movements had position thresholds closer to the ankles, and this threshold was measured from the axis of the ankles. Using this same concept, the stability boundary for stance and thresholds for stepping during forward perturbations were calculated by Pai and Patton (1997) from a simple pendulum-and-foot balance model combined with motion constraints (see also Patton et al. 1999). In the present study, this threshold boundary for stepping was determined experimentally. The stepping threshold boundaries of the two approaches are considered in the following discussion. They are illustrated for two subjects in Fig. 5. It is evident that at low velocities they agree.

![Figure 5](http://jn.physiology.org/)

**FIG. 5.** Threshold boundaries (solid lines) obtained in this study for a young and an older subject are compared with modeled thresholds by Pai and Patton (1997) based on a simple body and foot segment sagittal model (broken lines). For the older subject, lower dotted curve is the modeled threshold boundary assuming that limit of the base of support (BoS) is at the heads of metatarsals rather than end of the toe.
relatively well but an increasing discrepancy arises as velocity increases. These considerations point out the necessity of taking into account for future work the segmental strategies used by the subjects to keep the CoM over the BoS and their influence on the stepping threshold.

**Threshold boundary**

The threshold boundary has two identifiable position plateaus, one at low velocities (T_L) and one at high velocities (T_H). Between these plateaus, the threshold position and the velocity of the perturbation are inversely related.

**LOW-VELOCITY THRESHOLD.** T_L corresponds to the maximum amplitude that can be applied at the smallest velocities without inducing a step. In such quasi-static conditions, the position limit for equilibrium is the length of the BoS. No matter how slowly it is applied, when the pull brings the CoM to the limit of the BoS, subjects will have to step. However, it might be necessary to step before this point is reached. The external moment produced by gravity progressively increases as the CoM moves forward and approaches the limit of the BoS. If an equally large internal muscular moment cannot be produced to maintain equilibrium, a step must be taken before the CoM reaches the end of the BoS. In young subjects, T_L was not less than the BoS (Figs. 4 and 5), suggesting that this parameter reflects the anatomical constraints rather than limitation attributed to ankle strength. Similarly, the forward stability limit modeled by Pai and Patton (1997) was unaffected by ankle strengths within the normal range.

T_L in younger subjects was 25% larger than the functional BoS limits measured by having subjects lean forward without lifting the heels or bending the trunk (King et al. 1994; Schieppati et al. 1994). It is possible that subjects underestimate their limit of stability during voluntary leaning (Robinovitch and Cronin 1999). However, subjects in the present study did not have to maintain the final position after the pull, thereby removing a period of time in which superimposed naturally occurring body sway could bring them past the limit of stability. Furthermore, behavior to avoid stepping was not constrained so that subjects could raise the heels to extend the CoM forward (Endo et al. 2002), or flex the knees and extend the trunk to reduce the forward movement of the CoM (Iqbal and Pai 2000), or move their arms to stabilize the body through inertial effects (Maki and McIlroy 1997).

**VELOCITY-DEPENDENT THRESHOLD.** At higher velocities, the kinetic energy imparted by the perturbation has to be overcome. Thus the threshold at this range of velocities is now significantly affected by the velocity of the movement and the need to decelerate it before the CoM exceeds the front of the BoS. Thus it is more likely that the amount and rate that muscular torque can be generated at the ankles will play an increasing role in determining the threshold boundary as velocity increases.

**HIGH-VELOCITY THRESHOLD.** If all of the energy applied by the perturbation was used to move the whole body forward then the threshold boundary should not plateau, as we show in this experiment. Instead, it would be expected to slope downward as velocity increases. Eventually, at a finite velocity, the threshold should reach zero when the position threshold is at the ankles, or even become negative when it is behind the ankles. The model of Pai and Patton (1997) predicts a continuous inverse relationship between the threshold position to step and the velocity of the perturbation (Fig. 5). However, the prediction of the dynamic model consistently underestimated how far in front of the ankles the experimentally derived threshold was, and this discrepancy became progressively larger as velocity increased. Thus it appears that the perturbation has different effects on the forward velocity of the CoM of the modeled and real bodies. In the real situation, there comes a point where the increase in perturbation velocity has reduced the perturbation displacement required for threshold to a critically small level. Above this, further increases in perturbation velocity appear not to increase the forward velocity of the CoM of the whole body. Thus it appears that the high velocity threshold is reached when the perturbation displacement becomes so small that changes in perturbation velocity do not produce corresponding changes in the movement of the CoM.

Why does this happen? The perturbation was applied at waist level, and unlike a rigid inverted pendulum, the real body is a jointed inverted pendulum so that the trunk does not have to keep up as the hips and legs are pulled forward. Accordingly, the whole-body CoM will be behind that predicted and would more likely remain within the BoS. A strategy that accelerated the top of the body backward could compensate for the forward movement of the lower body. At very small amplitudes the compliance of the soft tissues will absorb a greater proportion of the perturbation energy so that less will be available to accelerate the body. Thus there is a relative discrepancy between the perturbation size and the movement of the CoM, which approaches zero for small perturbations. This filtering of the external perturbations by these largely passive mechanisms is therefore a means by which we normally deal with small disturbances regardless of their velocity or energy.

**Physiological correlates**

Sensory information could trigger the initiation of a step, much as inputs from hip (Grillner and Rossignol 1978) and limb extensor muscles (Duyssens and Pearson 1980) cause a transition from stance to swing phase during walking. Plantar mechanoreceptors can signal the position of the CoP under the feet during standing (Kavounoudias et al. 2001; Kennedy and Inglis 2002) and could initiate stepping from the transient associated with a perturbation. Lower-limb vibration, touch, and proprioceptive sensibilities indicate the performance of these afferent systems and we expected that they would be associated with the threshold boundary parameters. Associations were observed for vibration and touch (Table 1. Pearson r). However, the sign of the coefficients does not agree with their direct physiological reflex functions; poorer performance is associated with stepping at a lower threshold. The threshold boundary parameters were also associated with visual acuity, but again, diminished acuity was associated with a lowered threshold. A similar response of increased stepping frequency to forward perturbations is observed when the feet are cooled (Perry et al. 2000). Thus these behaviors appear to be an adaptation to diminished somatosensory input such that a step is taken in response to a smaller perturbation. Muscular strength at the ankles could limit how far forward a subject can lean before he or she loses balance. However, T_L...
was not associated with any measure of ankle strength within the younger or older groups. Therefore the observation that $T_L$ is short of the BoS limit for the older group is likely to be related to factors other than muscle strength. There were significant associations of $T_H$ with isokinetic strength and the rate of force development, parameters that should be most important for correcting these high-velocity movements. Thus the effects of lower-limb strength on stepping behavior are seen with rapid perturbations, although, like the sensory tests, muscle weakness is associated with a lowered threshold and increased stepping, again indicating a behavioral adaptation.

Simple reaction time is a global measure of sensorimotor integration, central processing, and peripheral factors, so it is surprising that it is only moderately associated with the total steps taken. This again has the same property of poorer performance being associated with more frequent stepping.

Obviously, sensorimotor elements not assessed could also affect the threshold boundary. For example, patients with bilateral vestibular loss step more often than controls with rapid perturbations (Krebs et al. 2000; Runge et al. 1998) and this implicates vestibular input in step initiation. However, here again, loss of vestibular function causes subjects to step more often and lowers the threshold boundary.

From these observations, we can state as a general principle: *Stepping is more likely when there is loss of any peripheral sensory or motor function that is used to step*. Peripheral function must be emphasized at this stage because an important variable not tested here, cognitive function, conceivably could behave in the opposite way; that is, poor cognitive function could be associated with a greater stepping threshold.

**Age-related changes**

The older group of subjects had a lower threshold boundary and stepped twice as often despite adopting a more backward initial position, which is likely to have been a compensation strategy to offset the forthcoming perturbation. Had it been possible to randomize the perturbation directions to negate this strategy, we assume that this group would have taken even more steps than seen in this case because they would have stood closer to the threshold boundary.

$T_L$ in the older group fell well short of 100% BoS, the boundary predicted by Pai and Patton (1997) for low velocities and observed in the young group (Fig. 5). For these older subjects, $T_L$ was determined by factors other than BoS limits. The associations of sensorimotor performance with $T_L$ suggest that somatosensory and visual inputs are important determinants of $T_L$.

$T_H$ in the older group was also lower, suggesting a difficulty responding to rapid perturbations without stepping. This could be expected from the slower and larger movements of older subjects in response to support-surface perturbations (Okada et al. 2001). Associations of $T_H$ with isokinetic strength and the rate of force development suggest that they are determinants of $T_H$.

Although the boundary was reduced in the older group, the fundamental relationship between amplitude and velocity was preserved, with similar shapes, slopes, and ranges of the threshold boundaries for the younger and older subjects. Thus, although the actual triggering levels adapt with age, the way that the CNS integrates the sensory information on which the decision to step is made is preserved.

Across the entire population of younger and older subjects, many sensorimotor factors were significantly associated with $T_L$, $T_H$, and the total number of steps taken but were not present when subject age was removed as a factor (Table 1: compare partial with Pearson coefficients). Therefore it is not possible to claim that the tested physiological factors are directly related to the stepping behavior, given that these factors could be age-related covariates.

**Nature of the stepping response**

When subjects arrest a forward motion to prevent stepping they allow spatial and temporal safety margins so that they regain balance well before the limit of the BoS (Patton et al., 1999, 2000). Brief training can modify these safety margins and their variability (Pavol and Pai 2002) and stepping can be delayed by 200–300 ms (Rogers et al. 2003). Thus it appears that a step is initiated before it is absolutely required to provide an adequate safety margin for the nonstep response. Only when the safety margin falls below a critical level is a corrective stepping response required.

At the higher velocities tested in the present study, the amplitude of the pull at the threshold boundary had become very small and produced only small movements of the body. Even among the younger group, but overtly apparent in the older group, many of the steps taken appeared unnecessary to correct the postural disturbance arising from the perturbation. This suggests a strategy change whereby stepping is a preprogrammed response, based on prior learning, and triggered by any number of factors. In this case, the mere existence of a perturbation triggers a step rather than the assessment of movement parameters reflecting the evolving instability that might be used to produce an appropriately timed, graded, and directed response that will efficiently arrest the imbalance. It may be for this reason that multiple stepping (Luchies et al. 1994), altered lateral stepping (Rogers et al. 2001a), and stepping that further destabilizes posture (Maki et al. 2000) are observed in many older people. This is not to say that physiological changes do not contribute to the altered behavior, but other cognitive influences such as prior experiences of poor balance, fear of falling, anxiety, and level of vitality may be more important determinants in the older group. It may be that the wait-and-see strategy is a luxury afforded only to the young and that the early step strategy adopted by older people is functionally the optimal adaptation for diminished sensorimotor performance.

**Summary**

This study identifies a threshold boundary for triggering protective stepping in response to perturbations applied to the body. This boundary shows a characteristic sigmoidal relationship between position and velocity, with distinct position thresholds for both low- and high-velocity disturbances. This shape is not as predicted from a biomechanical model, with the greatest difference at higher velocities. The differences likely reflect the real-world situation where there is variable transference of the energy of the perturbation into moving the body. A range of sensorimotor physiological variables showed significant associations with the threshold boundary across a large
range of subject ages. The nature of these associations was such that poor sensorimotor performance was associated with a position threshold boundary closer to the ankles and therefore more frequent stepping. However, accounting for the effect of age suggested that these variables could be age-related covariates and are not directly responsible for the altered stepping behavior. It appears that stepping is a triggered response based on selected strategy that may change with age.

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DISCLOSURES

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