The Effect of Lateral Stabilization on Walking in Young and Old Adults

J. C. Dean, N. B. Alexander, and A. D. Kuo*

***Abstract*—We tested how lateral stability affects gait as a function of age. A simple computational model suggests that walking is laterally unstable and that age-related decreases in motor and sensory function may be treated as noise-like perturbations to the body. Step width variability may be affected by active control of foot placement subject to noise. We hypothesized that age-related deficits may lead to increased step width variability. A possible compensation would be to walk with wider steps to reduce the lateral instability. The addition of external stabilization, through elastic cords acting laterally on the body during treadmill walking, would be expected to yield reduced step width variability and/or reduced average step width. We measured step width, its variability (defined as standard deviation), and metabolic energy expenditure in eight adult human subjects aged less than 30 years (Young) and ten subjects aged at least 65 years (Old). Subjects walked with and without external stabilization, each at a self-selected step width as well as a prescribed step width of zero. In normal walking, Old subjects preferred 41% wider steps than Young, and expended 26% more net energy ($P < 0.05$). External stabilization caused both groups to prefer 58% narrower steps. In the prescribed zero step width condition, Old subjects walked with 52% more step width variability and at 20% higher energetic cost. External stabilization resulted in reduced step width variability and 16% decreased energetic cost. Although there was no significant statistical interaction between age group and stabilization, Old and Young subjects walked with similar energetic costs in the stabilized, prescribed step width condition. Age-related changes appear to affect lateral balance, and the resulting compensations explain much of the increased energetic cost of walking in older adults.**

***Index Terms*—Aging, elderly, energetics, gait, mobility, oxygen consumption, stability, variability.**

I. Introduction

THE RISK OF falling increases dramatically with age and has been correlated with several measures of gait kinematics and kinematic variability. Frail subjects have been shown to have significantly lower gait speeds and step frequencies and higher step widths than healthy subjects [1]. However, some mean gait variables, such as gait speed and step length, may be indicators of fear of falling more so than direct predictors of actual fall risk [2]. Others, such as step width, may be compensations for instability [3], [4]. Increased gait variability has been proposed as an indicator of fall risk, with variabilities of step time [5] and step width [6] typically correlated with age. It is, however, uncertain how these various measures are related to each other and to gait stability. Identification of these relationships may aid their application to assessment of fall risk during walking.

Mathematical models of walking and stability can yield insight regarding the changes that occur with age. In a model of passive dynamic walking, McGeer [7] found that walking can be passively stable when restricted to 2-D movement in the sagittal plane. However, a 3-D model incorporating lateral motion is unstable with a strong tendency to fall laterally while maintaining passive stability in the fore-aft direction [8]. This model can easily be stabilized most effectively with feedback control of lateral foot placement. Such control would presumably involve high-level integration of information from vision, vestibular organs, and proprioceptors. Imperfect sensing to control lateral instability would be expected to affect step width variability. In contrast, passive stability in the fore-aft direction would require far less sensing and control, resulting in less step length variability. The same model predicted that the mechanical work performed each step—termed *step-to-step transitions* [9]—will increase with average step width, due to the need to redirect the body center of mass velocity between steps. However, larger average step widths may also decrease the lateral instability, perhaps offsetting some of the energetic disadvantages.

These predictions have been tested through experiments performed on young adults. In overground walking, step width (or lateral foot placement) variability is 79% larger than step length variability [10], as expected from lateral instability. Reduction in sensory feedback through removal of visual input also causes both average step width and step width variability to increase. Another experiment applied external lateral stabilization, through elastic cords attached to subjects at the waist and pulling laterally during treadmill walking. This led to a decrease in both step width and step width variability [11], indicating that artificial stabilization reduces the need for active feedback control. Other measurements show that both mechanical work and metabolic energy expended during walking increase sharply with step width [9]. The increase in energetic cost may be due to active adjustment of lateral foot placement, as well as increased step-to-step transition costs associated with wider steps.

Similar concepts may help explain age-related changes in walking. We propose a conceptual model of balance control subject to noise [see Fig. 1(a)]. We model imperfect sensing

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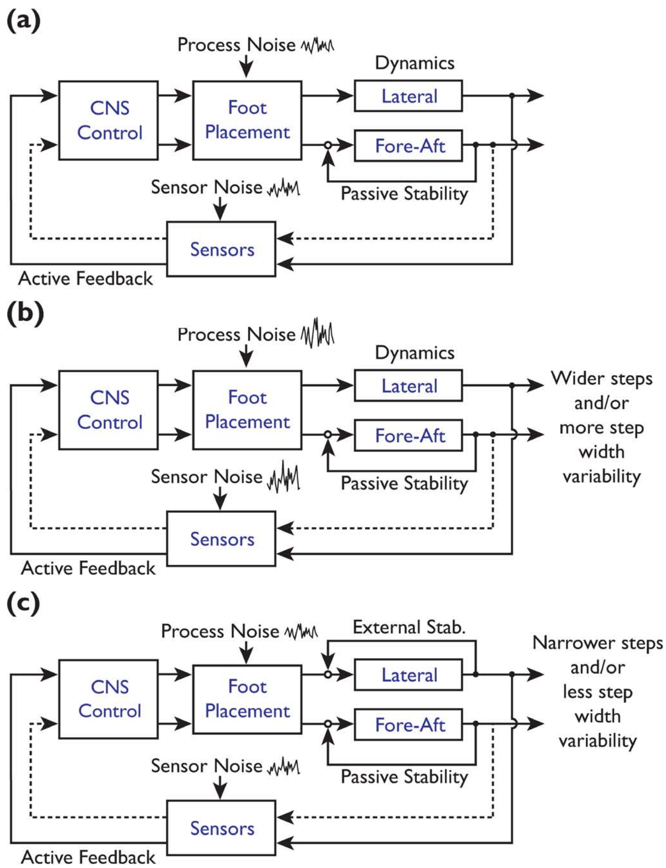

Fig. 1. During walking, lateral and fore-aft stability can be influenced through various feedback channels, the importance of which may depend on the degree of stability afforded by the passive dynamics of the body and legs. (a) In Young subjects, a hypothesized lateral instability [8] can be controlled with an active feedback strategy, in which motion is sensed and then integrated by the CNS, which can produce compensatory lateral foot placement adjustments with each step. Passive stability in the fore-aft direction implies that such integrative feedback is less critical (dashed lines) and of lower gain. Because such feedback is subject to imperfect actuation and sensing (modeled with disturbances labeled process noise and sensor noise, respectively), foot placement would be expected to depend on sensory precision and passive stability. (b) We propose a noise model for aging, in which Old subjects have noisier control and sensors, which would be expected to affect the lateral motion due to lateral instability. A possible compensation would be to walk with wider steps for greater safety, but at the expense of a higher metabolic cost for wide steps [9]. Noisy control may also result in greater lateral foot placement or step width variability at the same average step width. (c) The addition of external lateral stabilization could reduce the lateral instability, facilitating narrower steps and/or less step width variability.

and control with noisy inputs (termed sensor noise and process noise, respectively). This noise continually perturbs the body, requiring active (and imperfect) foot placement control—predominantly in the lateral direction—to maintain stability. With age, the ability to sense and actuate movement is diminished, as the number of afferent and efferent channels decreases and the quality of the signals passing through these channels degrades [12]–[14]. Modeled as increased noise, these changes may result in greater step width variability in older adults [see Fig. 1(b)]. A possible alternative or additional compensation would be to increase the average step width, taking advantage of the decreased lateral instability previously hypothesized [8]. Both increased average step width and step width variability would be expected to result in increased energetic cost in the older subjects.

Age-related changes may also be explored through the experimental manipulation of walking stability. One means of manipulation is to apply external lateral stabilization [11], which provides elastic lateral forces that counteract the hypothesized instability side-to-side [see Fig. 1(c)]. We hypothesized that improved lateral stability reduces the need for active control through lateral foot placement and found external stabilization to result in lower step width variability, average step width, and energetic cost in young healthy adults [11]. Although external stabilization has no direct effect on noise, it partially mitigates the effects, especially of sensor noise. If some age-related effects may be modeled with noise, external stabilization may reduce those effects as well.

The purpose of the present study was to test the role of lateral stability in age-related gait changes. We applied external lateral stabilization to younger and older subjects while measuring gait foot placement and energetics. Subjects walked on a treadmill at a controlled speed to standardize the system dynamics across groups. We hypothesized that older adults would walk with wider steps and/or higher step width variability than younger adults and with higher energetic expenditure. We expected the application of external lateral stabilization to reduce these measures. If lateral balance poses a greater challenge with age, we would expect external stabilization to allow older subjects to walk more like younger adults. We first tested whether external lateral stabilization caused subjects to prefer narrower steps. We next controlled for step width and tested whether external stabilization caused reductions in foot placement variability and energetic cost.

II. Methods

We examined changes in foot placement and energetic costs associated with artificially stabilizing the lateral motion of walking on a treadmill. We tested eight adult human subjects younger than 30 years (25.4 ± 3.6 years, mean ± s.d.), defined as the Young group, and ten subjects older than 65 years (73.4 ± 4.2 years), defined as the Old group. Young subjects had body mass 64.6 ± 12.9 kg and leg length $L = 0.910 \pm 0.060$ m, and Old subjects had mass 68.1 ± 21.0 kg and leg length $L = 0.847 \pm 0.046$ m. All subjects were healthy and the Old were screened by a nurse practitioner and found to have no significant abnormal cardiopulmonary, neurological, or musculoskeletal impairments on history and physical examination. This group appeared subjectively to be quite fit and active. Written consent was obtained in accordance with the Institutional Review Board for Human Subject Research at the University of Michigan Medical School.

Foot placement and energetic data were collected while subjects walked on a treadmill at 1.1 m/s. We first assessed each subject's preferred step frequency when walking at this speed and then enforced this same frequency with a metronome for data collection trials. These controls were intended to ensure that any observed gait changes were not due to differences in average walking speed, step frequency, or step length. Optotrak motion analysis markers (Northern Digital, Inc., Waterloo, Canada) were placed on the heels of the subjects' shoes to measure foot position during the trial. These data were used to calculate step length and step width. Step length variability and step width variability were calculated as the standard deviation

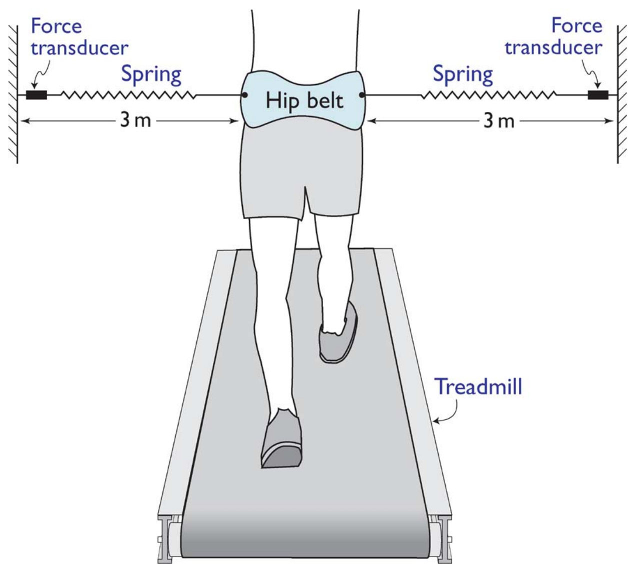

Fig. 2. Subjects were provided with artificial external lateral stabilization as they walked on a treadmill. Subjects wore a padded hip belt, to which were attached long elastic cords extending laterally to wall-mounted anchors. The cords were relatively long (3 m each), so that the elastic cords produced relatively small forces in the fore-aft and vertical directions. Force transducers were mounted at the walls to measure the lateral forces exerted by the elastic cords. Because of potential interference with the elastic cords, subjects were asked to walk with arms crossed in front of the chest in all conditions.

of the step lengths and step widths across the trial. We measured oxygen consumption using an open circuit respirometry system (VMax29, SensorMedics Corp., Yorba Linda, CA). We calculated energetic cost using the proportion 20.9 W per 1 ml/s oxygen. Each energetics trial was 7 min long with only the last 3 min of metabolic data recorded to ensure that steady state was reached. The net energetic cost of walking was calculated by subtracting out the energy consumed during a quiet standing trial.

Subjects walked under two stabilization conditions and two step width conditions. The stabilization conditions comprised normal walking (normal), and walking while laterally stabilized with an external mechanism (stabilized). External lateral stabilization was applied through two elastic cords attached to the hips, that opposed motion of the torso from side to side. Subjects wore a padded belt around their pelvis, which was attached on either side to long (2.5 m) nylon cords connecting to shorter (0.5 m) sections of rubber tubing (Fig. 2, after [11]). These spring-like elements were attached to the wall through transducers that measured the lateral forces. The stiffness was adjusted to produce good lateral stabilization as subjectively determined by one participant during preliminary tests. The mechanical properties of the lateral stabilizing elements were estimated by oscillating a known mass between them and fitting the motion to a mass-spring-damper model. We calculated the equivalent stiffness to be approximately 1200 N/m, and the equivalent damping to be about 20 N·s/m. When the lateral stabilization was first applied, subjects practiced walking for at least 10 min, until they reported feeling comfortable, before any data was recorded. In the Normal condition, subjects wore the hip belt but without the elastic cords. In both Normal and Stabilized conditions, subjects walked with their arms crossed due to interference with the elastic cords in the Stabilized condition.

For both the stabilization conditions, subjects walked with two different step width conditions: a self-selected gait with no restrictions (Preferred Step Width), and a gait in which they were instructed to walk on a line down the center of the treadmill (Zero Step Width). The Preferred Step Width condition was used to evaluate how age and artificial stabilization affect the mean step width that subjects choose and their associated energetic cost (similar to [11]). The Zero Step Width condition was used to assess age-related effects on step variability, with both stabilization and step width experimentally controlled. We expected the Zero Step Width gait to demonstrate the largest effects on foot placement variability and energetic cost, as the stabilization encourages minimal lateral motion. The Normal trial served as a control for the Stabilized trial in each step width condition.

Comparisons were made for step length, step width, step length variability, step width variability, and metabolic cost. All the foot placement measures were normalized by leg length, and metabolic cost was normalized by body weight to account for anthropometric differences. We performed two-way ANOVAs including interactions on these calculated values for both the Preferred Step Width and Zero Step Width conditions with age group and stabilization condition as independent measures. Stabilization condition was treated as a repeated measure. Tukey–Kramer post-hoc tests were performed within significant main or interaction effects. A P-value of less than 0.05 was required to indicate statistical significance.

Although all of the hypothesized effects may be stated qualitatively, they are also demonstrated quantitatively by a simple computational model. We used a model of lateral balance during walking, along with noisy disturbances (as in Fig. 1), to demonstrate how external lateral stabilization could affect step variability. Details of the model are given in the Appendix.

III. Results

When walking normally, Old subjects generally selected wider steps and expended more energy than Young. Step lengths and step frequencies were not significantly different, although step length variability was greater in the Old subject group. The absolute differences remained approximately the same when subjects were asked to walk with zero step width, although Old subjects also walked with more step width variability. The addition of external lateral stabilization caused subjects to select narrower steps. With step width prescribed, they walked with reduced step width variability. There were no significant interactions between age group and stabilization condition. The individual comparisons are summarized as follows.

In the Preferred Step Width condition, age had a significant effect on step width and energetic cost (see Fig. 3 and Table I). In normal walking without external stabilization, Old subjects walked with 41% wider steps than Young subjects $(P = 0.049)$. Age did not significantly affect step length (or step frequency), with the average value in the Old group within 3% of that in the Young. Energetic cost was 26% higher in the Old group $(P = 0.0024)$. Step width variabilities were not significantly different between age groups, but step length variability was

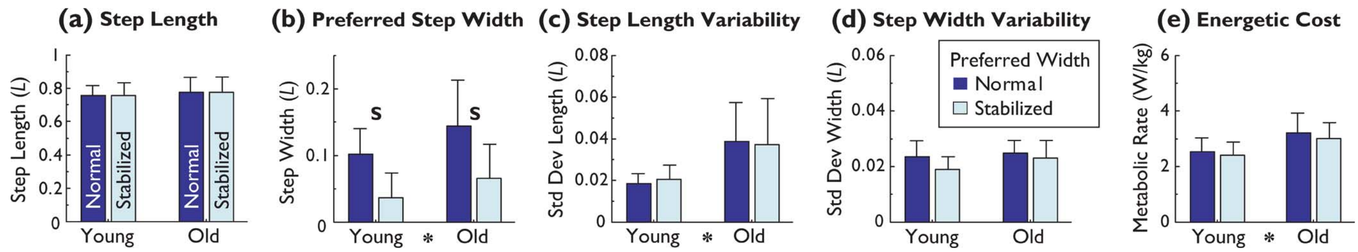

Fig. 3. During Preferred Step Width walking, subjects walked at a speed of 1.1 m/s with self-selected step widths. They walked in a Normal condition without external lateral stabilization and a Stabilized condition with external lateral stabilization. (a) Old and Young subjects selected similar step lengths (and step frequencies) in the Normal condition. The same combination of speed, step frequency, and step length was then applied in the Stabilized condition. (b) Old subjects preferred wider steps than Young ($*$ indicates significant age effect, $P < 0.05$). External lateral stabilization caused subjects to select significantly narrower steps (S indicates significant stabilization effect, $P < 0.05$). (c) Old subjects walked with greater step length variability than Young, in terms of standard deviations. (d) Old and Young walked with similar step width variability. (e) Old subjects walked with significantly greater net metabolic cost, defined as total metabolic rate subtracting that for quiet standing. Length and standard deviation measurements are given in units of leg length L.

TABLE I
FOOT PLACEMENT AND ENERGETIC COST VALUES FOR YOUNG AND OLD SUBJECTS WALKING UNDER THE PREFERRED STEP WIDTH AND ZERO STEP WIDTH CONDITIONS. DATA ARE REPORTED AS MEAN ± STANDARD DEVIATION, WITH LENGTHS GIVEN IN UNITS OF LEG LENGTH L

	Preferred Step Width				Zero Step Width			
	Young		Old		Young		Old	
	Normal	Stabilized	Normal	Stabilized	Normal	Stabilized	Normal	Stabilized
Step Length (L)	0.756 ± 0.061	0.757 ± 0.076	0.776 ± 0.089	0.776 ± 0.090	0.761 ± 0.079	0.763 ± 0.083	0.781 ± 0.100	0.802 ± 0.103
Step Width (L)	0.102 ± 0.036	0.036 ± 0.037	0.144 ± 0.069	0.066 ± 0.051	0.015 ± 0.033	0.006 ± 0.026	0.065 ± 0.046	0.028 ± 0.033
Length Var. (L)	0.019 ± 0.005	0.020 ± 0.007	0.039 ± 0.019	0.037 ± 0.022	0.018 ± 0.005	0.021 ± 0.006	0.035 ± 0.009	0.037 ± 0.022
Width Var. (L)	0.024 ± 0.006	0.019 ± 0.005	0.025 ± 0.005	0.023 ± 0.006	0.023 ± 0.007	0.018 ± 0.004	0.035 ± 0.012	0.023 ± 0.005
Ener. Cost (W/kg)	2.54 ± 0.49	2.41 ± 0.47	3.22 ± 0.71	3.01 ± 0.57	2.87 ± 0.52	2.47 ± 0.54	3.43 ± 0.70	2.85 ± 0.73

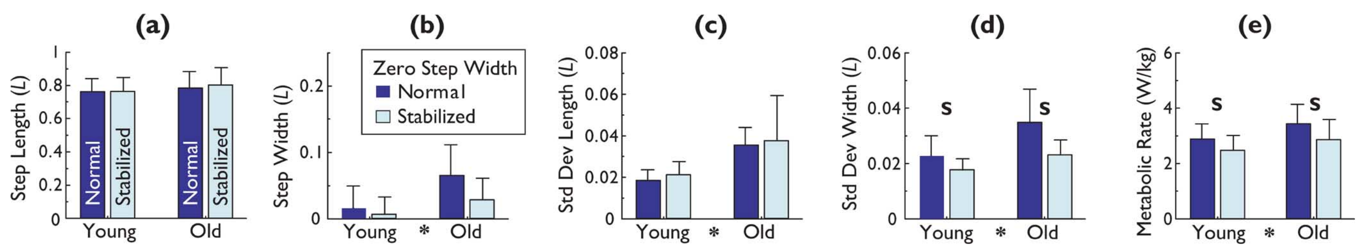

Fig. 4. During prescribed Zero Step Width walking, subjects were asked to walk on a line drawn on the treadmill belt in Normal and Stabilized conditions. (a) Old and Young subjects selected similar step lengths (and step frequencies) in the Normal condition. (b) The actual step widths of Young subjects were significantly closer to the target width of zero compared to Old ($*$ indicates significant age effect, $P < 0.05$). Old subjects had significantly greater (c) step length variability and (d) step width variability, in terms of standard deviations from average. External lateral stabilization also resulted in significantly decreased step width variability in both groups (S indicates significant stabilization effect, $P < 0.05$). (e) Old subjects walked with significantly greater net metabolic cost. External lateral stabilization also significantly reduced the energetic cost for both groups in the Zero Step Width condition. Length and standard deviation measurements are given in units of leg length L.

$0.02L$ (about 2 cm) greater in the Old subjects than the Young ($P = 0.0014$).

External lateral stabilization resulted in significantly reduced step width when subjects walked in the Preferred Step Width condition. On average, subjects selected a 58% narrower step width when externally stabilized ($P = 0.0002$). Externally stabilizing did not significantly change step length (changed less than 1%), step length variability (changed less than 1%), step width variability (changed less than 13%), or energetic cost (changed less than 6%).

There were greater age-related differences when subjects walked in the Zero Step Width condition. For walking without stabilization, age had a significant effect on step width, step length variability, step width variability, and energetic cost (see Fig. 4 and Table I). Old subjects walked with 336% wider steps than Young ($P = 0.0060$). Step variabilities were significantly higher in Old subjects, by 84% for step length ($P = 0.0006$) and 43% for step width ($P = 0.0028$). Energetic cost was 17% higher in the Old group ($P = 0.038$). Step length was not significantly different between age groups with values within 4%.

The addition of external stabilization significantly changed step width variability and energetic cost in the Zero Step Width condition. Step width variability decreased by 30% ($P = 0.0039$) when subjects were stabilized laterally. External stabilization resulted in 16% lower energetic cost ($P = 0.028$). Step length was not affected by stabilization as it remained within 2% of its initial value. Step length variability did not change after stabilization, remaining within 9%. Old subjects did not actually achieve the desired zero step width when not stabilized (average width $0.065 \pm 0.046L$), but came much closer with external stabilization (width $0.028 \pm 0.033L$). External stabilization also significantly reduced step width variability in Old subjects for Zero Step Width conditions.

Externally stabilizing the lateral motion did not affect the foot placement measures or energetics of the Old subjects more than the Young for either the Preferred Step Width or the Zero Step Width conditions. There were no significant interactions between age group and stabilization condition.

IV. Discussion

Lateral balance during walking appears to require active control and exact an energetic cost [11]. The changes in gait seen with age may be due to increased process and sensor noise or due to compensations for increased noise. Older adults have been reported to walk with wider, more variable steps and a higher energetic cost than young adults. We hypothesized that the addition of external lateral stabilization would reduce the effects of noise, allowing older adults to walk with narrower and less variable steps, and with decreased energetic cost. Our results indicate that stabilization did indeed lead to narrower steps and lower energetic cost in both younger and older adults, and with step width prescribed, to decreased step width variability as well.

The most significant effects of external stabilization were reductions in preferred step width and energetic cost. Preferred step width decreased in both Old and Young subject groups, suggesting that the choice of wide steps may be a compensation for lateral instability, made less necessary by external stabilization. However, the reduction in step width was not significantly greater with age ($0.078L$ decrease in Old with external stabilization versus $0.066L$ in Young). Even in the Zero Step Width condition, Old subjects walked with wider steps than Young and did not achieve the prescribed zero width. External stabilization did reduce actual step widths by more than half in the Old group, which we would attribute to compensations for lateral instability. There may be other factors that explain the remainder. For example, wider steps might help to avoid colliding the swing foot with the stance foot and older adults may prefer to maintain greater clearance between the feet. Wider steps may also keep the body center of mass more safely between the lateral extremes specified by foot placements [11]. Older adults may prefer a greater margin of safety, although given external stabilization we interpret this more as a function of confidence than actual need.

When step width was prescribed in the Zero Step Width condition, external stabilization resulted in 16% decreased energetic cost for both groups. Some of this decrease may be due to reduced step-to-step transition costs [9] that increase with step width, because neither group actually produced step widths of zero, and both groups reduced their step widths when stabilized. But another component of the energetic cost may be for actively controlling balance through lateral foot placement or other motions. The addition of external stabilization appears to reduce the need for such active control, and therefore, lead to reduced metabolic cost.

External lateral stabilization also led to decreased step width variability for the Old subject group in the Zero Step Width condition. It generally led to small decreases for both groups and step width conditions but the 34% decrease for Old and Zero Step Width was the largest and most significant change. Old subjects had greater step width variability than Young in the unstabilized normal condition, as would be expected from the noise model [see Fig. 1(b)]. Reductions in step variability could be partially due to improved sensory feedback from the external stabilization device. However, subjects appeared unable to identify the (left-right) direction of force produced by the device when they were walking normally. We, therefore, suspect that the major effect of external stabilization is to improve lateral stability, thereby decreasing reliance on active control and reducing the effects of noise. Conversely, the difficulty of walking with narrow steps when not externally stabilized appears to be associated with the dynamic lateral instability. The clinical task of tandem gait [15] is challenging because of this instability, which may lead to greater metabolic cost and greater step width variability.

Our results for age-related changes in gait variables were similar to those found in the literature. Customary walking speed and preferred step length typically decrease with age [16]–[18], but with controlled speed it is not unusual to observe similar step lengths [6]. We found a significant increase in energetic cost with age (again controlling for speed), consistent with earlier published findings [19]. We did, however, find differences with the literature regarding step variabilities. Others [6], [18] have reported no significant effect of age on step length variability, but significantly higher step width variability. Our results show an age-related increase in step length variability but not step width variability. We have also observed unusually high step length variabilities in younger adults in a previous external lateral stabilization study (cf. [10], [11]). These differences may be due to several particularities of our experimental protocol. For example, our subjects walked with their arms crossed (to avoid contacting the external stabilizer), which may affect both balance and the normal step pattern. Treadmill walking also presents a different visual environment from overground walking, and may affect step variability. Perhaps most significant was our use of a metronome to control for step frequency. This control was necessary to ensure that any changes in energy expenditure were associated with balance control rather than changes in step length or step frequency. However, a disadvantage is that subjects may have actively varied their step lengths in part to match the metronome. These effects make it inappropriate to compare absolute step variabilities with those for normal walking. It is more appropriate to concentrate on how step variability changed as a function of external stabilization.

The effects of external lateral stabilization were roughly comparable to those reported by Donelan *et al.* [11] for young, healthy subjects. This previous study reported decreases in preferred step width, step width variability, and energetic cost with stabilization. The most similar result in the present study was a decrease in preferred step width. The effect on energetic cost was broadly similar, with 4.9% and 14.1% decreases (only the latter statistically significant) in the Preferred Step Width and Zero Step Width conditions, respectively, compared to 5.7% and 9.2% in Donelan *et al.* [11]. One measure that did not agree with the previous results was step width variability, which did not decrease in the Preferred Step Width condition. This may be due to differences in the lateral stabilization equipment. Our lateral stabilization was performed in a different laboratory with slightly slower speed (1.1 m/s compared to 1.25 m/s), shorter elastic cords (3 m versus 8.5 m), and different stiffness and damping coefficients (1200 N/m and 20 N·s/m, respectively, versus 1700 N/m and 14 N·s/m, respectively). The shorter cords may have produced unintended fore-aft or vertical forces that disturbed our subjects and the lower stiffness may have performed less well as external stabilizers. It is preferable to use longer yet stiffer cords when possible.

The results regarding energetic cost were insensitive to the normalization procedure. The results reported here used a standard procedure, normalizing metabolic rate by body weight. This assumes that higher body mass requires a higher metabolic rate, as would be expected to sustain living tissue. In fact, it may be preferable to normalize by other factors to obtain a dimensionless metabolic rate [20], some investigators may prefer not to normalize at all. We found the type of normalization to have no substantive effect on the statistical results. This is consistent with expectation, given the similarity of the two subject groups in size and mass.

The choice of step width appears to be determined by the minimization of energetic expenditure while maintaining stability. Wider steps increase the cost of redirecting the body center of mass during step-to-step transitions [9]. Very narrow steps, however, also increase energetic cost, perhaps because of the need to move the swing foot around the stance foot and because of increased difficulty with active balance control. Given a choice, older adults appear to prefer wider steps, even when balance control is made easier through external stabilization. This may be a conservative strategy reflecting a lack of confidence or other age-related effects. When controlling for both walking speed and step width in the externally stabilized condition, age-related differences in energy expenditure were in fact statistically insignificant. Age-related changes in energetic cost of walking may therefore have a large dependency on balance.

We propose noise as a model of some of the changes in sensory and motor function that accompany aging. Physiological changes in sensors include decreases in the number of sensory nerve endings and number of nerve fibers [21], [22]. Sensors such as photoreceptors [23], vestibular organs [24], [25], and muscle spindles and other proprioceptors [26]–[28] all exhibit decreased sensitivity and/or higher thresholds with age. Motor capabilities also degrade with age [4], [15]. Motor precision decreases along with the number of motor units and the ability to finely regulate muscle force [29]. We model the loss of precision and information-carrying ability as an increase in noise, as is customary in control systems engineering. There need not be a literal addition of noise, but subtraction of information has an indistinguishable outcome [30]. Nor need the effect be purely random. Some noise-like physiological processes may ultimately be deterministic; however, from a behavioral and functional standpoint, the most parsimonious model of these processes may yet be noise, summarized conveniently by simple and understandable parameters such as standard deviation and bandwidth.

The noise model shows that walking may be laterally unstable, but can be stabilized through either active control or by artificial means as employed here. Active control is subject to imperfect motor control and sensing, modeled here as process and sensor noise, respectively. Age-related changes may be modeled with increases in noise and the resulting effects such as higher energetic cost may be mitigated by external stabilization. Similar models have been employed for standing balance and posture [31], [32] and may be useful for explaining how control challenges change with age.

APPENDIX

We devised a simple computational model of lateral balance during walking. We added noisy disturbances to the model to demonstrate how external lateral stabilization could affect stability and foot placement. The model serves as a quantitative demonstration of the noise model of aging.

The dynamics of walking were modeled by a simple, three-segment mechanism described previously [8]. This model consists of stance and swing legs attached to a pelvis and free to swing in the sagittal plane. The feet are attached to the legs at the ankles. The stance leg has a single DOF about the ankle, allowing for lateral motion in the frontal plane through eversion/inversion. The model is capable of walking down a gentle slope under the force of gravity alone, a feature of passive dynamic walking machines. We found this model to be passively stable in the fore-aft direction, but to exhibit instability in the lateral direction. This instability could be corrected through the active feedback control. The control used the state of the machine at the end of one step to make a small active adjustment to lateral foot placement in time for the next step. Fully stable walking of this model could only be achieved with this active control, or with the addition of external lateral stabilization, modeled as a lateral spring of appropriate stiffness [11].

The equations of motion for this model are integrated with respect to time, from one step to the next. The step-to-step transition equations are

$$x_{k+1} = f(x_k) + g(x_k)u_k + w_k \tag{1}$$

where x_k is the body state (angular positions and velocities of the DOF) at the beginning of each step k. Lateral foot placement is described by u_k, and f and g are nonlinear functions representing integration of differential equations of motion. External lateral stabilization results in a change to f, making the system passively stable. Process noise is added through zero-mean, normally-distributed noise w_k, disturbing the fore-aft and lateral velocity of the body center of mass, with Γ transforming that noise into the appropriate state perturbations.

The lateral motion is stabilized through feedback control of lateral foot placement. The foot placement for the next step u_{k+1} is determined from

$$u_{k+1} = -L\hat{x}_k + w'_k \tag{2}$$

where L is the state feedback gain matrix [8], itself dependent on f and g, and w'_k is process noise causing imperfect foot placement. The feedback acts on the state estimate $\hat{x}_k$, which is itself made imperfect by the addition of sensor noise v_k

$$\hat{x}_k = x_k + v_k \tag{3}$$

where sensor noise is also normally distributed and of zero mean. Not explicitly modeled but assumed present is a state estimator [32], which integrates and processes sensory information, producing a filtered estimate $\hat{x}_k$. The disturbance v_k

TABLE II
RELATIVE STANDARD DEVIATIONS FOR MODEL PROCESS AND SENSOR NOISE, FOR OLD AND YOUNG. PROCESS NOISE w PERTURBS CENTER OF MASS VELOCITY AND w' MODELS IMPERFECT LATERAL FOOT PLACEMENT. SENSOR NOISE v MODELS IMPERFECT SENSING OF ALL BODY STATES. THE NOISE MODEL OF AGING ASSUMES THAT ALL CONTROL PROCESSES BECOME LESS PRECISE. ALL STANDARD DEVIATIONS ARE REPORTED RELATIVE TO THE YOUNG PROCESS NOISE, WHICH WAS ARBITRARILY SET TO DIMENSIONLESS 3E-4, WITH STEP VARIABILITY SCALING APPROXIMATELY LINEARLY WITH THIS PARAMETER

Symbol	Young	Old
Process noise w	1	1.9
Process noise w'	1.3	2.0
Sensor noise v	0.030	0.045

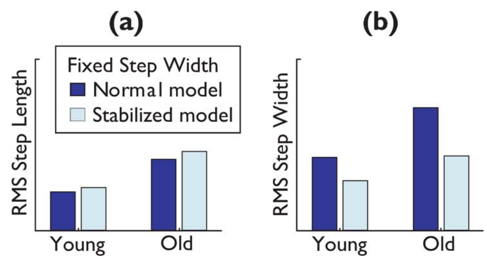

Fig. 5. Model simulation results for step variability as a function of age and external stabilization, with prescribed, fixed average step width. (a) Step length variability increases with age—modeled as increased process and sensor noise—and with external stabilization. (b) Step width variability increases by a greater amount with age, and decreases with external lateral stabilization. The model displays relative trends similar to those observed in experimental data [see Fig. 4(c) and (d)].

may, therefore, be interpreted as estimator-filtered effects of sensor noise, rather than the noise acting directly on sensors.

We performed Monte Carlo simulations of the model to predict the step variability with and without external lateral stabilization, and with and without the age-related increases in noise. All noise was formed from sequences of zero-mean, normally distributed random numbers, with standard deviation parameters chosen in an ad-hoc fashion so that noise generally increased with age (see Table II for parameter values). Step variabilities were computed from the standard deviation of foot placements over 100 steps for each condition. Without noise, the model would always step perfectly and with no variability.

The simulations produced changes in step variability as a result of external stabilization and the noise model for aging (see Fig. 5). As with experimental results, the model predicts greater step length and width variability with age. It also predicts greater step width than step length variability overall. External lateral stabilization results in decreased step width variability. Interestingly, the model also shows that stabilization can result in increased step length variability, as was observed experimentally. This is due to the external stabilizer's dynamics, causing step width variations to couple into step length to a greater degree than occurs without external stabilization. All of the relative changes occurring with age, stabilization condition, and step length versus width were also observed in experimental data.

The model suggests that the experimental results from the age comparison could largely be explained by noise alone. No other parameter changes, to either control or to body dynamics, were necessary to produce the simulation results. Similarly, the effects of stabilization can be explained largely by the dynamics of the stabilizer alone. No other parameter changes, to either control or noise, were needed to yield the model results.

The model does have some drawbacks. Neither process nor sensor noise parameters can readily be identified from experiments. They instead represent a variety of different aging effects that contribute to greater uncertainty and less precision in movements, but which are difficult to measure independently. The model is also far simpler than actual walking and cannot exclude other possible explanations for the experimental results. However, it does demonstrate that a small set of relatively simple effects is sufficient to explain these results. Moreover, the principal dependency—that step variability will increase for any unidirectional increases in noise—is largely insensitive to the specific parameter values. We would therefore expect more complex models to have similar behavior.

REFERENCES

[1] R. Moe-Nilssen and J. L. Helbostad, "Interstride trunk acceleration variability but not step width variability can differentiate between fit and frail older adults," *Gait Posture*, vol. 21, no. 2, pp. 164–170, Feb. 2005.
[2] B. E. Maki, "Gait changes in older adults: Predictors of falls or indicators of fear," *J. Amer. Geriatr. Soc.*, vol. 45, no. 3, pp. 313–320, Mar. 1997.
[3] J. S. Brach, J. E. Berlin, J. M. Van Swearingen, A. B. Newman, and S. A. Studenski, "Too much or too little step width variability is associated with a fall history in older persons who walk at or near normal speed," *J. Neuroeng. Rehab.*, vol. 26, pp. 2–21, 2005.
[4] A. Gabell and U. S. I. Nayak, "The effect of age and variability in gait," *J. Gerontol.*, vol. 39, pp. 662–666, 1984.
[5] J. M. Hausdorff, D. A. Rios, and H. K. Edelberg, "Gait variability and fall risk in community-living older adults: A 1-year prospective study," *Arch. Phys. Med. Rehab.*, vol. 82, no. 8, pp. 1050–1056, Aug. 2001.
[6] T. M. Owings and M. D. Grabiner, "Variability of step kinematics in young and older adults," *Gait Posture*, vol. 20, no. 1, pp. 26–29, Aug. 2004.
[7] T. McGeer, "Passive dynamic walking," *Int. J. Robot. Res.*, vol. 9, pp. 62–82, 1990.
[8] A. D. Kuo, "Stabilization of lateral motion in passive dynamic walking," *Int. J. Robot. Res.*, vol. 18, no. 9, pp. 917–930, 1999.
[9] J. M. Donelan, R. Kram, and A. D. Kuo, "Mechanical and metabolic determinants of the preferred step width in human walking," *Proc. R. Soc. Lond. B*, vol. 268, no. 1480, pp. 1985–1992, Oct. 2001.
[10] C. E. Bauby and A. D. Kuo, "Active control of lateral balance in human walking," *J. Biomech.*, vol. 33, no. 11, pp. 1433–1440, Nov. 2000.
[11] J. M. Donelan, D. W. Shipman, R. Kram, and A. D. Kuo, "Mechanical and metabolic requirements for active lateral stabilization in human walking," *J. Biomech.*, vol. 37, no. 6, pp. 827–835, Jun. 2004.
[12] F. B. Horak, C. L. Shupert, and A. Mirka, "Components of postural dyscontrol in the elderly: A review," *Neurobiol. Aging*, vol. 10, no. 6, pp. 727–738, 1999.
[13] D. Manchester, M. Woollacott, N. Zederbauer-Hylton, and O. Marin, "Visual, vestibular and somatosensory contributions to balance control in the older adult," *J. Gerontol.*, vol. 44, no. 4, pp. M118–M127, Jul. 1989.
[14] N. B. Alexander, "Gait disorders in older adults," *J. Amer. Geriatr. Soc.*, vol. 44, no. 4, pp. 434–451, Apr. 1996.
[15] B. L. Cho, D. Scarpace, and N. B. Alexander, "Tests of stepping as indicators of mobility, balance, and fall risk in balance-impaired older adults," *J. Amer. Geriatr. Soc.*, vol. 52, pp. 1168–1173, 2004.
[16] M. P. Murray, R. C. Kory, and B. H. Clarkson, "Walking patterns in healthy old men," *J. Gerontol.*, vol. 24, no. 2, pp. 169–178, Apr. 1969.
[17] M. M. Samson, A. Crowe, P. L. de Vreede, J. A. Dessens, S. A. Duursma, and H. J. Verhaar, "Differences in gait parameters at a preferred walking speed in healthy subjects due to age, height and body weight," *Aging (Milano)*, vol. 13, no. 1, pp. 16–21, Feb. 2001.
[18] H. Stolze, H. J. Friedrich, K. Steinauer, and P. Vieregge, "Stride parameters in healthy young and old women-measurement variability on a simple walkway," *Exp. Aging Res.*, vol. 26, no. 2, pp. 159–168, 2000.

[19] D. Malatesta, D. Simar, Y. Dauvilliers, R. Candau, F. Borrani, C. Prefaut, and C. Caillaud, "Energy cost of walking and gait instability in healthy 65- and 80-yr-olds," *J. Appl. Phys.*, vol. 95, no. 6, pp. 2248–2256, Dec. 2003.
[20] P. G. Adamczyk, S. H. Collins, and A. D. Kuo, "The advantages of a rolling foot in human walking," *J. Exp. Biol.*, vol. 209, no. 20, pp. 3953–3963, Oct. 2006.
[21] B. Bergstrom and H. Engstrom, "The vestibular sensory cells and their innervation," *Int. J. Equilib. Res.*, vol. 3, pp. 27–32, 1973.
[22] L. Hay, C. Bard, M. Fleury, and N. Teasdale, "Availability of visual and proprioceptive afferent messages and postural control in elderly adults," *Exp. Brain Res.*, vol. 108, no. 1, pp. 129–139, Feb. 1996.
[23] R. Sekuler, L. P. Hutman, and C. J. Owsley, "Human aging and spatial vision," *Science*, vol. 209, no. 4462, pp. 1255–1256, Sep. 12, 1980.
[24] U. Rosenhall, "Degenerative patterns in the aging human vestibular neuro-epithelia," *Acta Otolaryngol.*, vol. 76, no. 2, pp. 208–220, 1973.
[25] U. Rosenhall and W. Rubin, "Degenerative changes in the human vestibular sensory epithelia," *Acta Otolaryngol.*, vol. 79, no. 1–2, pp. 67–80, 1975.
[26] H. C. Diener, J. Dichgans, B. Guschlbauer, and H. Mau, "The significance of proprioception on postural stabilization as assessed by ischemia," *Brain Res.*, vol. 296, no. 1, pp. 103–109, Mar. 26, 1984.
[27] J. K. Richardson, J. A. Ashton-Miller, S. G. Lee, and K. Jacobs, "Moderate peripheral neuropathy impairs weight transfer and unipedal balance in the elderly," *Arch. Phys. Med. Rehab.*, vol. 77, no. 11, pp. 1152–1156, Nov. 1996.
[28] H. B. Skinner, R. L. Barrack, and S. D. Cook, "Age-related decline in proprioception," *Clin. Orthop. Relat. Res.*, vol. 184, pp. 208–211, Apr. 1984.
[29] Z. Erim, M. F. Beg, D. T. Burke, and C. J. de Luca, "Effects of aging on motor-unit control properties," *J. Neurophysiol.*, vol. 82, no. 5, pp. 2081–2091, Nov. 1999.
[30] T. M. Cover and J. A. Thomas, *Elements of Information Theory*, 2nd ed. New York: Wiley-Interscience, 2006.
[31] R. A. Speers, A. D. Kuo, and F. B. Horak, "Contributions of altered sensation and feedback responses to changes in coordination of postural control due to aging," *Gait Posture*, vol. 16, no. 1, pp. 20–30, Aug. 2002.
[32] A. D. Kuo, "An optimal state estimation model of sensory integration in human postural balance," *J. Neur. Eng.*, vol. 2, pp. S235–S249, 2005.

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