Exp Brain Res (2010) 206:359–370
DOI 10.1007/s00221-010-2409-x

RESEARCH ARTICLE

The effects of error augmentation on learning to walk on a narrow balance beam

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Received: 5 March 2010 / Accepted: 31 August 2010 / Published online: 19 September 2010

Abstract Error augmentation during training has been proposed as a means to facilitate motor learning due to the human nervous system's reliance on performance errors to shape motor commands. We studied the effects of error augmentation on short-term learning of walking on a balance beam to determine whether it had beneficial effects on motor performance. Four groups of able-bodied subjects walked on a treadmill-mounted balance beam (2.5-cm wide) before and after 30 min of training. During training, two groups walked on the beam with a destabilization device that augmented error (Medium and High Destabilization groups). A third group walked on a narrower beam (1.27-cm) to augment error (Narrow). The fourth group practiced walking on the 2.5-cm balance beam (Wide). Subjects in the Wide group had significantly greater improvements after training than the error augmentation groups. The High Destabilization group had significantly less performance gains than the Narrow group in spite of similar failures per minute during training. In a follow-up experiment, a fifth group of subjects (Assisted) practiced with a device that greatly reduced catastrophic errors (i.e., stepping off the beam) but maintained similar pelvic movement variability. Performance gains were significantly greater in the Wide group than the Assisted group, indicating that catastrophic errors were important for short-term learning. We conclude that increasing errors during practice via destabilization and a narrower balance beam did not improve short-term learning of beam walking. In addition, the presence of qualitatively catastrophic errors seems to improve short-term learning of walking balance.

Keywords Gait · Task-specificity · Rehabilitation · Movement variability

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Introduction

Physical guidance is often given in rehabilitation settings via the hands of a therapist. More recently, robotic devices have been developed to provide physical guidance in rehabilitation settings. The use of robotics has much potential in rehabilitation because of their ease of use, reliable measurement of performance and their capability to deliver a high intensity and dosage of therapy (Reinkensmeyer et al. 2004; Huang and Krakauer 2009). However, to maximize rehabilitation outcomes, it is important to first understand how best to use physical guidance, robotic or otherwise (Marchal-Crespo and Reinkensmeyer 2009; Reinkensmeyer and Patton 2009).

Several studies have shown that physical guidance during practice hinders motor learning. For upper limb movements, guidance given frequently during practice improved performance, but once the guidance was removed, the improvements were not present (Armstrong

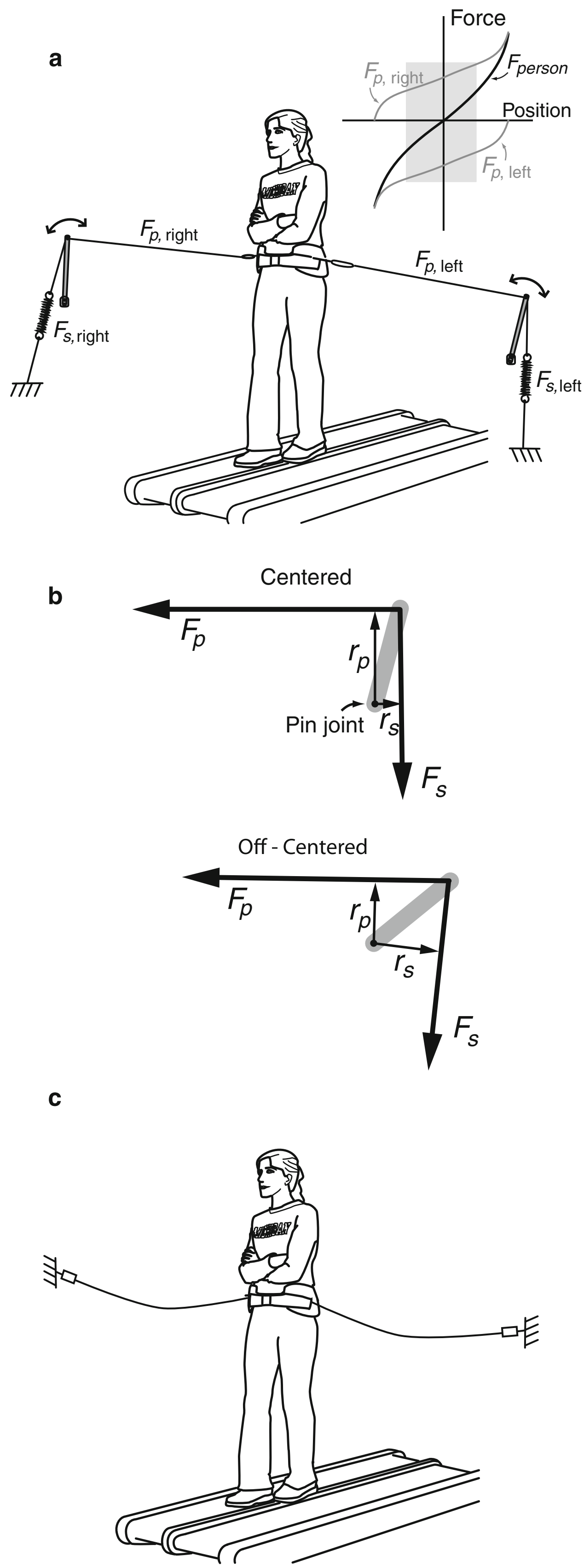
1970; Winstein et al. 1994). Similarly, we showed in a recent study that error-reducing physical assistance given during practice was detrimental to short-term learning unassisted walking on a narrow balance beam (Domingo and Ferris 2009). These findings are consistent with the theory that error detection and correction are needed for forming and updating internal models during motor learning (Wolpert and Ghahramani 2000). Internal models, or neural representations of the body, task and environment, are used to compare the expected movement to the actual movement produced (Kawato 1999). When errors occur (differences between the expected and actual movement), the internal model is updated, and motor output is modified to produce the correct movement. Over time, these errors drive learning of a new internal model for new limb dynamics or environment. Previous studies have shown that motor learning is proportional to motor errors experienced in upper limb tasks (Thoroughman and Shadmehr 2000; Scheidt et al. 2001). From this evidence, it could be inferred that magnifying errors, rather than reducing errors, that the subject experiences may increase the rate of motor learning.

Previous motor adaptation studies have already shown that amplifying errors improves motor learning of a new task. Error augmentation of trajectory errors can enhance learning of visuomotor rotations in an upper extremity reaching task in healthy subjects (Wei et al. 2005). In another study, robot-generated forces were applied to the arm of individuals with stroke while the moving their arm through a plane. After training, the individuals had improved movement trajectories in directions where error was amplified more than when error was reduced or was zero (Patton et al. 2006). For motor learning in the lower limb, Emken and Reinkensmeyer (2005) showed that error amplification lead to faster formation of the internal model in a novel walking task. No study as of yet has tested whether error augmentation could be used to improve motor learning of walking balance. This is an important question because dynamic balance is a critical component of gait control necessary for patients to safely practice walking.

The purpose of this study was to determine whether augmenting error during training affects short-term learning of walking balance. We studied able-bodied subjects learning to walk on a treadmill-mounted balance beam (beam-mill). Beam walking is similar to over ground walking, but is more challenging to dynamic balance because it exploits the lateral instability of walking (Donelan et al. 2004; Schrager et al. 2008; Domingo and Ferris 2009). We hypothesized that using error augmentation during training would improve motor short-term learning of walking on the beam-mill more than practice without error augmentation.

Previous error augmentation studies involved enhancing motor adaptation to an altered environment by using amplification of continuous trajectory errors. Unlike these studies, we sought to enhance motor learning of an unaltered environment by augmenting the discrete error of stepping off of the beam during practice. Two groups of subjects practiced walking on the wide beam (2.5 cm) with a destabilization device applied at the hips (Fig. 1a). The destabilization device had the properties of a spring with negative stiffness and augmented error by increasing the number of times subjects stepped off of the beam. There were two levels of spring stiffness used (Medium Destabilization and High Destabilization groups). We tested a third group that walked on the narrow beam as a form of error augmentation. This group had also experienced error augmentation during training (i.e., had stepped off the beam more often) but the training was more similar to the evaluation task than using the destabilization device. We then compared these results to a group that practiced walking on the wide beam without the destabilization device (Wide group). All subjects from the four groups were evaluated on the wide beam without the device pre- and post-training. We hypothesized that subjects using error augmentation during practice would have greater performance gains than subjects that did not use error augmentation. We based this hypothesis on the rationale that error drives motor learning (Rumelhart et al. 1986; Lisberger 1988; Dancause et al. 2002); therefore, augmenting error would lead to a faster rate of learning. We also hypothesized that subjects walking on the narrow beam during practice would have greater performance gains than those that practiced with the destabilization device on the wide beam. Walking on the narrow beam has more similar task dynamics to the evaluation task of walking on the wide beam because moments are still generated at the foot to help maintain balance and no additional external forces are introduced anywhere else in the body.

We also wanted to investigate the relative roles of making smaller control errors in movement (movement variability of the pelvis) and larger catastrophic errors (stepping off the beam) on learning to walk on the beam-mill. In a second, follow-up experiment, we tested another group of subjects that practiced walking on the beam-mill while wearing a “kinematic channel” device (Assisted group) that greatly decreased the number of catastrophic errors experienced during practice but still allowed a similar amount of movement variability as walking on the wide beam without assistance (Wide group). We hypothesized that the Assisted and Wide groups would have similar performance gains, based on results from the first experiment that showed performance gains were better correlated to movement



◀ **Fig. 1** Walking balance devices. **a** Destabilization device. A subject walking on the beam-mill with the destabilization device used to apply forces on the subject with springs that appeared to have negative stiffness. This was accomplished by varying the moment arms via an over-center linkage placed between the springs and the subject. When the subject's pelvis moved away from the center of the beam, the device applied a proportional force onto the subject in the same direction that the subject was moving. The *inset graph* shows a simplified representation of the properties of the device. The *thin gray lines* represent the forces due to the device at each side of the person, where the *heavy black line* shows the net force due to both sides of the device as a function of the subject's pelvis position. The *shaded* area represents the operating range of the device. Physical blocks were set so that the device was would stop applying additional force soon after the subject stepped off of the beam. **b** Detail of over-center linkage used in destabilization device. The linkage provided a "moment balance" between the moment produced by the spring force (F_s) and the moment produced from the cable tension that applied force to the person (F_p). When the linkage rotated as the subject moved from being "centered" over the beam to "off-centered", the moment arms changed greatly, resulting in an increase in F_p. **c** A subject walking on the treadmill-mounted balance beam with the kinematic channel device. The assist device had straps that were set so that each subject would have maximal space to move in the frontal plane but not so much space that the subjects would be unable to right themselves as they were beam walking

variability than to catastrophic errors experienced during practice.

Methods

Subjects

We tested 50 able-bodied subjects (see Table 1 for subject characteristics). Subjects were medically stable and had no history of major leg injuries. The University of Michigan Institutional Review Board approved this study. All subjects gave written informed consent in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki prior to participating.

Table 1 Subject demographics

Group	Gender		Body mass (kg)	Leg length (m)
	M	F		
Narrow	2	8	60.3 ± 10.5	0.88 ± 0.032
Medium Destabilization	3	7	59.1 ± 8.3	0.88 ± 0.054
High Destabilization	2	8	60.7 ± 8.9	0.87 ± 0.048
Wide	4	6	64.6 ± 14.7	0.89 ± 0.063
Assisted	4	6	65.5 ± 7.2	0.91 ± 0.019

Procedures

Five groups of 10 subjects walked on the beam-mill for a 3-min pre-training evaluation, a 30-min training period (with rest breaks every 10 min) and a 3-min post-training evaluation. During the pre- and post-evaluation periods, all subjects walked on the wide beam (2.5 cm wide) to test for performance gains and were made aware of this at the beginning of the experiment. The first two groups walked with the destabilization device with medium spring stiffness or high spring stiffness during the training period (Medium Destabilization or High Destabilization groups, respectively). A third group walked on the narrow beam (1.27 cm wide) without a device during the training period (Narrow group). A fourth group walked on the wide beam without a device during the training period (Wide group). The last group walked on the wide beam while wearing the kinematic channel device (Assisted group). Data presented in this paper from the Wide group were collected and published in a previous study (Domingo and Ferris 2009) but are used here to compare to the data from the other three groups.

Treadmill speed was set at 0.22 m/s. This speed was determined during pilot testing. Subjects were instructed to walk on the beam for as long as possible without stepping off. Instructions were given to all subjects by the same experimenter. They had to walk heel-to-toe with arms crossed over their torso. They were also instructed not to lean forward, twist their trunk, angle their feet away from the longitudinal direction of the beam, or look down at their feet. View of the walking surface was obscured by wearing dribble goggles. Subjects were allowed to move their pelvis and trunk in the frontal plane to help maintain balance. All subjects wore standardized orthopedic shoes. Subjects had to wait 5 s after stepping off the beam before attempting to walk on it again.

Equipment

The equipment for this experiment consisted of a treadmill-mounted balance beam (beam-mill), a destabilization device, a kinematic channel device, force plates and a motion capture system. The beam-mill was made of interchangeable small wooden blocks attached to the treadmill belt that lined up to make a continuous balance beam. One beam was 2.5 cm wide by 2.5 cm tall (Wide) and the other was 1.27 cm wide by 2.5 cm tall (Narrow). Smaller wooden blocks were added to either side of the bases of both beams to make them more stable in the frontal plane.

Destabilization device

The destabilization device was made up of latex tubing springs, an over-center linkage and cables that attached to the subject via a padded hip belt (Fig. 1a). This device applied forces onto the subject with springs with an effective negative stiffness. The negative spring stiffness was achieved by placing an over-center linkage between the subject and the spring. As the person's pelvis moved away from the center of the beam, a proportional force was applied to the subject in the same direction, but if the person was centered over the beam, the device applied approximately zero net force onto the subject.

The linkage on each side of the destabilization device provided a "moment balance" between the moment produced by the spring (F_s) and the moment produced from cable tension that applied a force to the person (F_p) (Fig. 1b). Therefore, the force applied to the person by the left or right side of the device is described by the following equation:

$$F_p = F_s\left(\frac{r_s}{r_p}\right)$$

where F = force, r = moment arm, p = person, and s = spring. The springs were pre-tensioned so that the spring force on each side was non-zero at each side when the person was centered over the beam, but the net force ($F_{p,\ \mathrm{right}} + F_{p,\ \mathrm{left}}$) summed to approximately zero and the person felt no pull to either side. When the subject moved their pelvis toward the left, the linkage on the left side of the device rotated away from the person. The spring on the same side then shortened and the spring force, $F_{s,\ \mathrm{left}}$, decreased. However, the moment arms also changed, where the spring moment arm ($r_{s,\ \mathrm{left}}$) length increased and the moment arm length of the person ($r_{p,\ \mathrm{left}}$) decreased. The relatively larger changes in moment arm, and relatively smaller decrease in $F_{s,\ \mathrm{left}}$, resulted in $F_{p,\ \mathrm{left}}$ increasing overall, and pulled the person with greater force in same direction as they originally had moved. On the right half of the device, movement of the subject's pelvis to the left caused the linkage on the right side to rotate toward the person. On the right side, $F_{p,\ \mathrm{right}}$ then decreased ($F_{s,\ \mathrm{right}}$ was greater, but $r_{s,\ \mathrm{right}}$ was shorter and $r_{p,\ \mathrm{right}}$ was longer). The net force ($F_{p,\ \mathrm{right}} + F_{p,\ \mathrm{left}}$) on the subject resulted in the device pulling the subject to the left when the subject's pelvis moved to the left.

The device made it difficult to stay on the beam if the hips moved away from the center of the beam. The device also gave subjects feedback about their position relative to the beam. The subjects were made aware of the function of the device and were encouraged not to translate anteriorly or posteriorly on the treadmill. When the subject's pelvis was centered over the beam, there was approximately zero net force applied to the subject. We had 8 pairs of springs of different stiffnesses. For each subject, we chose the spring that would provide the stiffness closest to the non-dimensionalized spring stiffness of 0.2978 for the Medium Destabilization group and 0.4404 for the High

Destabilization group. To determine the desired spring stiffness, we used the following equation:

$$k = \bar{k} \cdot \frac{l}{mg}$$

where $k =$ dimensionalized stiffness, $\bar{k} =$ non-dimensionalized stiffness, $l =$ leg length and $mg =$ bodyweight. The non-dimensionalized spring stiffnesses of 0.2978 and 0.4404 were based on springs used during pilot testing. The average total stiffness of the device was 192.5 N/m for the Medium Destabilization group and 298.4 N/m for the High Destabilization group.

Kinematic channel device

For the second part of the experiment, we tested a group of subjects that walked on a beam using a device that allowed normal frontal plane movement variability but helped to minimize stepping off of the beam. This training device was made up of lightweight cables and adjustable straps that attached to the subject via a padded hip belt (Fig. 1c). The straps were set so that each subject would have maximal space to move their pelvis in the frontal plane but not so much space that the subjects would be unable to right themselves as they were beam walking. We placed single-axis tension/compression load cells (1,200 Hz; Omega Engineering, Stamford, CT, USA) in series with the cables on both sides of the subject to measure the tension in the cables produced by the subjects during walking. Subjects were instructed not to depend on the device because they would not be able to use it during the post-training period.

Recording procedures

The treadmill was placed above two force plates (sampling rate 1,200 Hz; Advanced Mechanical Technology Inc., Watertown, MA, USA) so that we could calculate center of pressure from the forces and moments produced by the subject while walking. The center of pressure helped us determine when the subject was on or off the beam.

We used an 8-camera video system (frame rate 120 Hz; Motion Analysis Corporation, Santa Rosa, CA, USA) to record the positions of 4 reflective markers placed on the subject's pelvis, neck and shoulders during walking. We calculated the standard deviation of the medio-lateral movement of the marker placed at the sacrum and neck to determine movement variability.

Performance measures

We recorded the number of times the subject stepped off the beam per minute. We then divided this quantity by the fraction of time the subject was on the beam (not touching the treadmill surface with either foot). This quotient, failures per minute, was our primary performance metric because it took into account the number of errors with respect to the amount of time the subject successfully walked on the beam. We also calculated the standard deviation (SD) of the medio-lateral movement of markers placed at the sacrum and the neck (Motion Analysis Corporation, Santa Rosa, CA; 120 Hz). We calculated percent change of the performance variables by subtracting the pre-training value from the post-training value and dividing by the pre-training value for each subject to normalize to pre-training performance.

For the pre- and post-training periods, we recorded data for the duration of the 3-min trial. For the 30-min training period, we collected only 20 s of data per each minute of training. We used a 4th order low-pass zero-lag Butterworth filter with a cutoff frequency of 6 Hz to smooth raw marker data. Values for SD of markers were calculated only using the data from when subjects were on the beam. We used a 4th order low-pass zero-lag Butterworth filter with a cutoff frequency of 25 Hz to smooth raw force data, then a 4th order low-pass zero-lag Butterworth filter with a cutoff frequency of 6 Hz to smooth center of pressure data. Data were processed using custom software written in MATLAB (The MathWorks, Inc., Natick, MA).

Statistical analysis

We first performed an analysis of variance (ANOVA) to determine whether groups evaluated on the same beams had similar failures per minute during pre-training (JMP IN software, SAS Institute, Inc., Cary, NC).

We then performed an ANOVA to test for differences between the groups for each of the following dependent variables: percent change for failures per minute, failures per minute during training and sacral marker SD. For post hoc analysis, we performed Tukey's honestly significant difference (THSD) test to compare results between groups as needed to delineate the differences between groups.

For the error augmentation groups and the Wide group, we calculated the correlation coefficient between sacral marker standard deviation and percent change in failures per minute. This would help to determine whether there was a relationship between movement variability while walking on the beam and the performance gains.

To analyze the load cell data from the assist device, we calculated net force and then normalized the data to each subject's body weight. We calculated the root mean square (RMS) of the data only from when the subject was on the beam. The force RMS data for the training period was then averaged into six 5-min blocks. We then performed a repeated measures ANOVA as an omnibus test to find differences in force RMS between the 5-min blocks. We performed a paired t test to find statistical difference between the first and last 5-min blocks of force RMS data.

We also compared the sacral marker SD data during training between groups to further examine whether movement variability was similar between groups throughout the training period. We performed t tests to compare data between groups during the first 5-min of training and the last 5-min of training.

Results

All groups had the similar pre-training values for failures per minute ($P = 0.087$) and sacral marker standard deviation ($P = 0.364$).

Error augmentation groups versus Wide group

The groups that practiced with error augmentation experienced more failures per minute (Medium Destabilization: 27.3 ± 2.0, High Destabilization: 29.6 ± 1.4, Narrow: 26.5 ± 2.8, mean $\pm$ SEM) during the training period than the Wide group (12.6 ± 1.3) (Fig. 2a) (ANOVA, $P < 0.0001$, power $= 0.99$, THSD, $P < 0.05$). All three error augmentation groups had similar amounts of failures per minute during training (THSD, $P > 0.05$).

Although more error was experienced during practice in the error augmentation groups than in the Wide group, the Wide group had significantly greater performance gains than all other groups ($-61.2 \pm 6.0\%$) (ANOVA, $P < 0.0001$, power $= 0.99$, THSD, $P < 0.05$) (Fig. 2b). The High Destabilization group had a smaller percent change in failures per minute ($-8.1 \pm 5.3\%$) than the Medium Destabilization group ($-23.6 \pm 6.2\%$), but the difference was not significant (THSD, $P > 0.05$). The performance gains were significantly higher in the Narrow group than in the High Destabilization group (THSD, $P < 0.05$) (Fig. 2b). The Narrow group had a $-34.6 \pm 7.9\%$ change in failures per minute.

Sacral marker movement variability versus performance gains

The relative trend in performance gains (the additive inverse of percent change in failures per minute) for all groups was similar to that in the movement variability of the sacral marker (Fig. 3a, b). The correlation coefficient, ρ, between these variables was 0.4281 ($P = 0.0059$) and $R^2 = 0.1833$. The relative trend in failures per minute was opposite that of the performance gains (Fig. 3a, c).

Assisted group versus Wide group

Subjects in the Assisted group decreased use of the device as the training period progressed. Root-mean-square

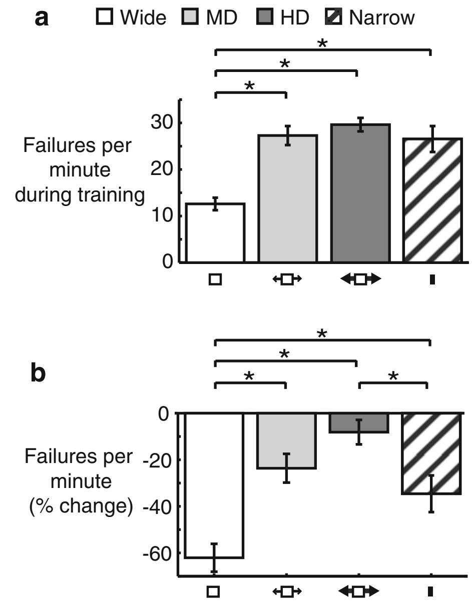


Fig. 2 Averaged failures per minute and percent change. **a**. Averaged failures per minute during training across subjects for each group. Error bars are ±1 Standard Error of the Mean (SEM). The Medium Destabilization (MD), High Destabilization (HD) and Narrow groups had significantly greater failures per minute during training than the Wide group (ANOVA, $P < 0.0001$, THSD*, $P < 005$). There was no statistical difference between the error augmentation groups (MD, HD, Narrow) (THSD*, $P > 0.05$). **b** Averaged percent change ((post-training−pre-training)/pre-training values) for failures per minute across subjects for each group. Error bars are ±1 SEM. The Wide group had greater performance gains after training than both Destabilization groups and Narrow group (ANOVA, $P < 0.0001$, THSD*, $P < 0.05$). The Narrow group had significantly greater performance gains than the High Destabilization group (THSD*, $P > 0.05$)

(RMS) of the net force (normalized to bodyweight) per minute was calculated for each subject. Force data were only included in calculations from when the subject was on the beam. Figure 4a shows the averaged force RMS for each minute of data across subjects that used assistance during the training period. We averaged the RMS across 5-min intervals and then performed a repeated measures ANOVA to find if there were differences in force RMS across the 30-min training period. The analysis showed that there was a statistically significant difference between the different 5-min blocks (ANOVA, $P < 0.0001$). Post hoc analysis showed that there the force RMS for the first 5 min ($1.2 \pm 0.24\%$ bodyweight) was significantly greater than for the last 5 min of the training period ($0.72 \pm 0.15\%$ bodyweight) (paired t test, $P = 0.0265$).

Using the assist device greatly reduced the number of failures during training, but during post-training, the number of errors returned to pre-training values. Figure 4b shows the averaged failures per minute for both groups during pre- and post-training and during each minute of training. Figure 4c shows the averaged sacral marker SD

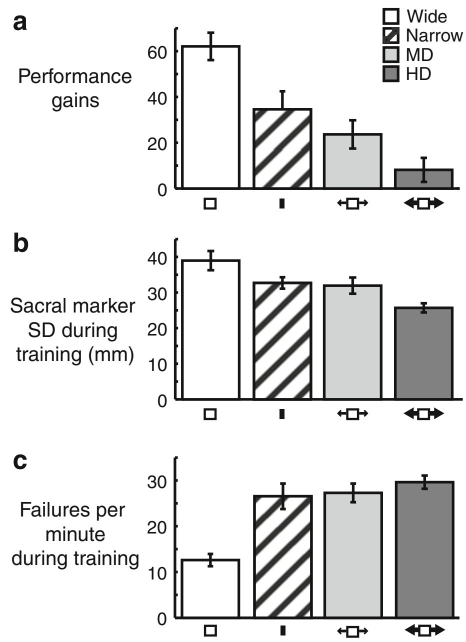


Fig. 3 Performance gains versus sacral marker movement variability and failures per minute during training. Error bars are ±1 SEM. **a** Performance gains are the additive inverse of the percent change in failures per minute for each group. The relative performance gains between groups were similar to the **b** relative sacral marker movement variability during training between groups and had an inverse relationship with **c** failures per minute during training

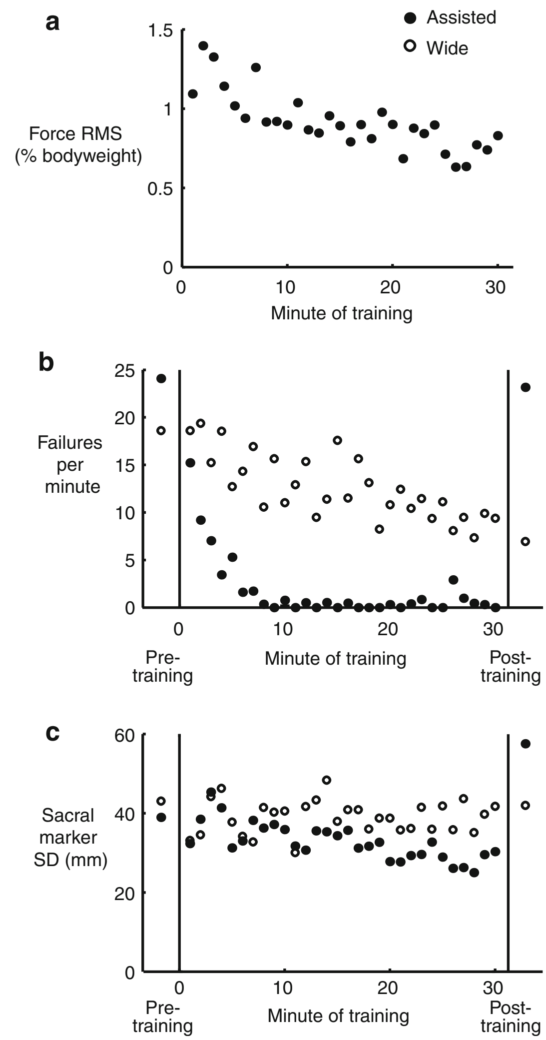


Fig. 4 Averaged time series data from the training period. **a** Averaged root-mean-square (RMS) of net force from the assist device as a percent of bodyweight. Data are taken only from when subjects were walking on the beam. **b** Averaged number of failures per minute for each minute across subjects for each group. The Assisted group had very few failures per minute after 10 min of the training period. **c** Averaged standard deviations (SD) for the sacral marker in the frontal plane as a measure of movement variability across subjects for each group. Data included in the calculation was only from when subjects were on the beam. Averaged data from the first 5 min of training showed that there were no differences in movement variability (SD) between groups. Averaged data from the last 5 min of training showed that movement variability was higher in the Wide group (ANOVA, $P = 0.0143$)

for both groups during pre-and post-training and during each minute of training.

We wanted to verify that both groups had similar amounts of movement variability over the whole training period. Sacral marker movement variability was slightly greater in the Wide group (39.0 ± 2.7 mm, mean ± SEM) than the Assisted group (32.7 ± 4.7 mm), but the difference was not significant (ANOVA, $P = 0.2626$) (Fig. 5a). When comparing 5-min blocks of data during the training period, we found that there were no differences in sacral marker SD (t test, $P = 0.8760$) between groups during the first 5 min of training. During the last 5 min of training, movement variability was greater in the Wide group (t test, $P = 0.0143$) by 30%.

We also wanted to ensure that the kinematic channel device was effective at preventing subjects from stepping off the beam. We compared the number of failures per minute during training for both groups and found that they were significantly different (ANOVA, $P < 0.0001$) (Fig. 5b). The Assisted group had an average of 1.7 ± 0.44 failures per minute during the training period, while the Wide group had an average of 12.6 ± 1.3 failures per minute during training.

Practicing with the assist device clearly hindered learning (Fig. 5c). The Assisted group had −1.7 ± 11.7% change in failures per minute, while the Wide group had −61.2 ± 6.0% change in failures per minute. There were much greater performance gains in the Wide group (ANOVA, $P = 0.0003$, power = 0.99).

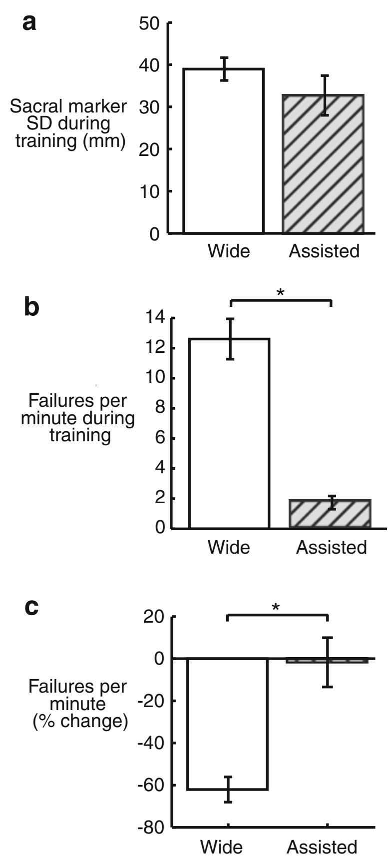


Fig. 5 Averaged sacral marker SD during training, failures per minute during training, and percent change in failures per minute across subjects for each group. **a** Sacral marker SD calculated from marker position in the medio-lateral direction when subjects were on the beam. Data is averaged over the entire training period across all subjects. The difference in movement variability between groups was not significant (ANOVA, $P = 0.2626$). **b** Averaged failures per minute over the entire training period across all subjects. The Assisted group had significantly less failures per minute during training (ANOVA, $P < 0.0001$). **c** Averaged percent change in failures per minute from pre- to post-training. Using the assist device during practice clearly hindered learning, as there were significantly greater performance gains in the Unassisted group than the Assisted group (ANOVA, $P = 0.0003$)

Discussion

Error augmentation

The main result of this study showed that augmented error training with either the destabilizing device (Medium Destabilization and High Destabilization) or with a narrower balance beam was actually worse for short-term learning of walking balance than unaltered practice. This was contrary to our hypothesis that subjects using error augmentation during practice would have greater performance gains than subjects that did not use error augmentation. We also found that when the error augmentation has more similar task dynamics to the desired task (narrow beam training), it led to greater performance gains compared to error augmentation with less similar task dynamics compared to the desired task (destabilization device training).

One explanation for why practicing with the destabilization device led to poorer performance gains compared to unaltered practice is the role of internal models in motor learning. Considerable research supports the theory that the nervous system forms internal models of movement dynamics during motor learning (Kawato 1999; Wolpert et al. 2001). Recent studies have provided specific evidence that humans use internal models during walking (Emken and Reinkensmeyer 2005; Lam et al. 2006) and stationary balance (Ahmed and Ashton-Miller 2007). When using the destabilization device, the dynamics of the task were changed. As a result, the learner may have formed an internal model for walking on the beam that included the device dynamics. Once the device was removed, the subjects had not developed the internal model for beam walking without the device and exhibited minimal learning during the post-training period. Detecting and correcting errors are important for motor learning, but the errors must be specific to the dynamics of the desired task.

The importance of task dynamics on internal models could also explain why subjects in the Narrow group had greater performance gains than the High Destabilization group (Fig. 2b). Walking on the narrow beam during practice likely has more similar task dynamics than walking with the destabilization device because using the destabilization device applies additional external forces to the pelvis and walking on the narrow beam does not. As a result, the internal model formed during narrow beam walking was more transferable to wide beam walking than the internal model formed during walking with the destabilizing device.

Another possible reason why the Wide beam group may have had the greatest performance gains is that practicing on the wide beam unassisted may have provided optimal level of error experience (i.e., stepping off the beam) during practice [i.e., at the “optimal challenge point” (Guadagnoli and Lee 2004)]. Too many errors experienced during practice may not allow for an appropriate example of the task (Sanger 2004) and may lead to decreased motivation because of frustration. In contrast, too few errors experienced during practice may not provide enough feedback to refine the internal model of task dynamics (Scheidt et al. 2000; Patton et al. 2006). The error augmentation groups in our study had experienced more errors during practice than the Wide group. The increased task difficulty may have been too high to stimulate motor learning.

Our findings were different than previous studies that found error augmentation to be beneficial for motor learning. There are several reasons why this may be the case. The present study is different than previous studies involving error augmentation because of (1) the type of error that was amplified and (2) the type of task goal. Previous studies involved augmenting continuous trajectory errors to adapt to a novel task [e.g., with a force field (Patton et al. 2006; Emken and Reinkensmeyer 2005) or visual feedback (Wei et al. 2005)]. In our study, we augmented the discrete error of stepping off of the beam with the destabilization device or with the narrower beam in order to enhance learning of the unaltered task.

Stepping off the beam is a discrete and qualitatively catastrophic error (i.e., losing balance so that beam walking is no longer possible). We chose to augment this type of error because this error was directly related to the subjects' task goal: walking on the balance beam for as long as possible without stepping off. Other parameters of walking dynamics, such as center of mass amplitude or movement variability, were not a primary measurement of error in this study because these parameters were not part of the stated task goal. Subjects could have learned better control of their center of mass, allowing for greater center of mass movement without stepping off of the beam with extended practice and therefore this parameter would not be an accurate measure of error.

The second reason our results may have been different than those of other studies is because previous studies utilized error augmentation to enhance motor adaptation to a novel environment. That is, subjects learned a new sensorimotor calibration in order to adapt to a different dynamic or visual environment, resulting in the desired trajectory. The motor learning paradigm in the present study is different than that of motor adaptation. In our study, subjects practiced with a variation of the goal task (one that was similar, but with increased catastrophic errors) to test if this variation would enhance motor learning of the unaltered task goal. Although there are similarities between motor adaptation and motor learning, these are distinct processes that likely have different neural and behavioral mechanisms (Huang and Krakauer 2009).

Our findings may also have differed from previous studies because we specifically tested learning of walking balance, while others examined learning of discrete arm movements in a plane (Patton and Mussa-Ivaldi 2004; Patton et al. 2006) or learning to step through a viscous force field (Emken and Reinkensmeyer 2005). These types of movements may be less complex than the task of maintaining walking balance, which involves multiple sensory inputs (visual, vestibular, and proprioceptive) and a high degree of coordination among multiple body segments in the upper and lower body. Perhaps, the complexity and higher degree of difficulty of our task would not be aided by error augmentation, especially in the earlier stages of learning for our naïve subjects.

There may be some instances when error augmentation for learning walking balance may be useful. A common issue in rehabilitation is preparing patients for the "real world." Walking does not always occur in a straight line and over smooth surfaces. Practicing with error augmentation may help patients respond to perturbations or changes in the environment. If the unaltered task can be performed proficiently, augmenting error with different task dynamics may be beneficial. By having diverse practice conditions, individuals can generalize learning so that learning of a new task happens at a faster rate (Seidler 2004). Similarly, a recent study showed that humans use an internal model for the dynamics of an environment and task to create appropriate feedback responses for unanticipated errors, even if they had never previously experienced the error (Wagner and Smith 2008).

Sacral marker movement variability in the frontal plane correlated well with performance gains (the additive inverse of the percent change in failures per minute) (Fig. 3a, b). The destabilization device in this study increased catastrophic error (i.e., stepping off the beam) based on the subject's movements (Fig. 3c), but it also limited the amount of movement variability that the subject was able to experience while walking on the beam (Fig. 3b). Movement variability at the pelvis may reflect the number of smaller errors in control that are made, allowing for updates to the internal model. This may be an alternative indicator of learning compared to catastrophic errors experienced during practice. The destabilization device may have increased catastrophic errors, but it also decreased the smaller errors experienced while walking on the beam that are evidenced by movement variability. There was a significant correlation between movement variability and performance gains ($\rho = 0.4281$, $P = 0.0059$), but only 18% of the variance was explained by this relationship due to high inter-subject variability.

Greater movement variability may be indicative of greater learning for this walking balance task. This is supported by the observation that humans seem to detect a loss of balance with a "control error signal anomaly" (CEA) during standing balance (Ahmed and Ashton-Miller 2004, 2007). To determine motor output for a desired movement, the central nervous system creates an internal model of limb dynamics based on previous sensorimotor experiences. The expected sensory feedback is then compared to the actual sensory feedback. If a sufficiently large difference between the two is detected, or CEA, then a compensatory response will occur. Subjects that successfully learned to walk on the balance beam had experienced greater movement variability, better explored the movement space, and used the movement errors to update the internal model.

Movement variability versus stepping off the beam

Based on these results, we wanted to more specifically examine the role of movement variability and smaller control errors to delineate its effects on learning relative to larger error experience. For example, learning to ride a bike with training wheels that do not touch the ground while the bicycle is vertical should provide a means for riders to explore the task space of balancing without falling over. We built a similar type of stabilization device for walking on the beam-mill. It provided a channel of very low forces on the torso during task space exploration while providing high forces if the torso moves too far to one side or the other to prevent failure. This could be seen as similar to the type of kinematic channel in hindlimb movement used during robotic locomotor training in spinalized mice (Cai et al. 2006).

The main result of our follow-up experiment showed that experiencing catastrophic errors (stepping off the beam) during practice may be important for learning this beam walking task, contrary to our hypothesis. Subjects that used the kinematic channel device experienced movement variability similar to unassisted subjects (Fig. 5a) and had a reduced number of failures during practice (Fig. 5b), but had very small performance gains (Fig. 5c). This suggests that giving assistance that reduced qualitatively catastrophic errors hindered motor learning of a walking balance task.

It is likely that catastrophic balance errors involve higher level cortical processes. This has been demonstrated in a number of EEG studies on humans performing balance tasks (reviewed in Maki and McIlroy 2007). Results from follow-up studies in our laboratory have found that the anterior cingulate displays an error-related negativity (ERN) event related potential when subjects step off of the balance beam. Our working hypothesis is that this ERN detected in the anterior cingulate is important for motor learning during walking balance. Future studies will continue to explore this relationship.

There are several reasons why "kinematic channel" assistance may have hindered learning of narrow beam walking. A learner's ability to recognize and correct their errors increases as movement skill improves (Liu and Wrisberg 1997). Although using the assist device allowed for a similar amount of movement variability as the Wide group, it also greatly reduced opportunities for detection and correction of the larger errors (stepping off the beam). After about 10 min of training, the Assisted group rarely stepped off the beam, while the Wide group continued to step off the beam throughout the training period. It is possible that stepping off the beam actually provides higher level information about performance compared to the smaller control errors, contributing to the learning process (Wei and Kording 2009). Learning occurs in response to both small and large errors and with distinct neural pathways (Criscimagna-Hemminger et al. 2010).

The assist device also changed task dynamics by applying forces to stop lateral translation of the pelvis once the subject reached a predetermined distance away from center. The presence of these forces could affect how subjects learn to maintain balance on the beam-mill. Strategies formed to balance while using the assist device are likely very different than those used without the device.

Subjects in the Assisted group had about the same amount of movement variability at the pelvis as the Wide group throughout most of the training period (Fig. 5a). This shows that the assist device did allow enough space for normal movement variability. However, the subjects in the Assisted group were less variable with their movements by the end of training. These subjects were told at the beginning of the experiment not to become dependent on the device because they would be evaluated on Wide beam walking. They may have concentrated too much on avoiding using the device and as a result, ended up with reduced movement variability. Alternatively, subjects may have been able to use very low forces from the device toward the end of training for feedback to limit their movement variability.

Although the Assisted group used the device minimally during training, especially toward the end of training (Fig. 4a), even very small forces may have helped to maintain balance. Several studies have shown that light touch (less than 1 Newton of force) at the fingertip can greatly reduce postural sway during standing with eyes closed due to the augmented sensory feedback rather than physical stabilization (Holden et al. 1994; Jeka and Lackner 1994; Kouzaki and Masani 2008). In our study, most subjects in the Assisted group reported that they felt they had greatly decreased the use of the device at the end of the training period. However, it is possible that subjects unknowingly became dependent on the very low forces during practice. These forces may have been able to give some cues to their position in space. Perhaps, these low forces from the device within the kinematic channel could be eliminated by placing physical blocks a small distance away from each side of the pelvis. Even so, the restriction in movement provided by these blocks would likely change task dynamics enough to hinder learning.

There is another possible reason why the Assisted group had lower performance gains. There are two separate parts of this task that require different dynamics: getting on the beam initially and then taking steps to stay on the beam. Because this group spent most of their time walking on the beam, they had fewer opportunities to learn the act of successfully getting back on the beam after a failure. Without this skill, subjects were more likely to step off the beam soon after stepping on, greatly increasing the number of failures per minute. Using the kinematic channel device made practice less similar to the walking on the beam without any assistance.

A recent study comparing the effectiveness of locomotor training using the Lokomat (a robotic exoskeleton used for automated treadmill stepping) versus conventional gait training in patients with subacute stroke (Hidler et al. 2009) supported the results of our study. They found that subjects that received conventional gait training had greater improvements in gait speed and walking distance than those that trained in the Lokomat. They attributed these results in part to how the Lokomat provides guidance of the lower extremities and greatly restricts motion at the trunk and pelvis. If motion is limited at the trunk and pelvis, the patients are unable to sense and correct for movement errors during walking and would greatly limit learning of balance.

Conclusions

This study showed that (1) error augmentation achieved with destabilization or a narrower balance beam is not better than practicing the unaltered beam-walking task, (2) task-specific dynamics are important considerations for practice, (3) movement variability of the pelvis correlates well with performance gains for beam walking, and (4) making qualitatively catastrophic errors may be important for short-term learning of walking on a narrow beam. This suggests that rehabilitation strategies should be devised so that assistance allows patients to make catastrophic errors (so that the goal movement is no longer possible) during practice but still maintain safety and prevent falls. Future studies should be conducted to further examine the relationship between catastrophic errors and motor learning.

Acknowledgments The authors would like to thank Daniela Weiss, Sarah Weiss, Evelyn Anaka and other members of the Human Neuromechanics Lab for help with data collection and processing. We would also like to thank Shawn O'Connor and Peter Adamczyk for help with data analysis and Steve Collins for help with the negative-stiffness spring design. This work was supported by the Rackham Graduate Student Research Grant, the Foundation for Physical Therapy PODS II Scholarship, and National Institutes of Health F31 HD056588-01.

Conflict of interest The authors declare that they have no conflict of interest.

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