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Reactive balance adjustments to unexpected perturbations during human walking

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Abstract

The purpose of this investigation was to determine the effect of unexpected forward perturbations (FP) during gait on lower extremity joint mechanics and muscle Electromyographic (EMG) patterns in healthy adults. The muscles surrounding the hip were found to be most important in maintaining control of the trunk and preventing collapse in response to the FP. Distinct lower extremity joint moment and power patterns were observed in response to the FP but an overall positive moment of support (M_s) was maintained. Therefore, reactive balance control was a synchronized effort of the lower extremity joints to prevent collapse during the FP.

Keywords: Human; Locomotion; Perturbation; Balance

1. Introduction

Human gait is one of the most common forms of human movement, yet the underlying neurological and biomechanical processes by which this movement occurs are complex and individualistic. The inherent instability involved in human locomotion results from a relatively small base of support, long single support phase, and the observation that 2/3 of the body's mass is located in the head–arm–trunk (HAT) segments [1]. Successful locomotion requires three essential elements, (a) the ability to generate and maintain fundamental locomotor patterns appropriate for moving toward an intended destination; (b) maintenance of basic dynamic equilibrium between a shifting center of mass (COM) and a constantly changing base of support; and (c) the ability to change locomotor patterns in response to external or internal inertial changes that threaten dynamic equilibrium [2]. While the first element is concerned with the generation of complex locomotor patterns, the latter two are critical in the detection of potential threats to balance and the subsequent reaction to either foreseen or unexpected perturbations during normal gait.

Often, individuals cannot anticipate external threats to dynamic equilibrium during gait and reactive mechanisms are required to act after the person experiences an unexpected perturbation [3]. Relatively few investigations have studied reactive postural adjustments during gait in response to unexpected perturbations [4–9]. Nashner [4] incorporated a moveable platform into a walkway to simulate unexpected perturbations during gait. Electromyographic (EMG) recordings from the gastrocnemius (GAS) and tibialis anterior (TA) muscles were measured along with lower extremity joint angles. Unexpected forward translation or downward rotation perturbations applied at HS produced increased TA EMG activity in the perturbed leg and unexpected backward translation or upward rotation produced increased GAS EMG activity. These recordings were similar to those obtained during standing perturbations [4,10]. Based on the observation that platform perturbations resulted in altered foot trajectory and stretching of the muscle, Nashner [4] hypothesized that alterations in

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distal limb trajectories provided the principal sensory feedback input to the central nervous system to elicit reactive balance control strategies.

Tang et al. [5] hypothesized that proximal muscles (hip and trunk) may play as important a role as do distal muscles (TA and GAS) in balance recovery. Using a paradigm similar to Nashner [4], EMG recordings were analyzed from the TA, GAS, rectus femoris (RF), biceps femoris (BF), rectus abdominis (RA), and erector spinae (ES). Their results indicated that hip and trunk muscles (RA, ES) did not play a significant role in reactive balance adjustments during perturbed gait at HS since these muscles did not demonstrate more consistent activation, earlier onset latency, longer burst duration, or larger burst magnitude compared with distal leg and thigh muscles. Leg and thigh muscles (TA, GAS, RF, BF) did, however, demonstrate earlier onset, higher magnitude, and relatively longer duration of activity as compared with normal walking. It was, therefore, concluded that activity from leg and thigh muscles was the primary contributor to reactive balance control.

Winter [1,11] suggested that hip muscle activity is critical during gait for control of the HAT segment. It was estimated that an ankle moment about eight times greater than that of the hip was needed to control the HAT segment due to the combined moments of inertia of the rest of the body. Furthermore, it was suggested that during early stance, as the HAT segment accelerates anteriorly, the ankle does not intervene but undergoes a small dorsiflexor moment, and it is the large hip extensor moment that serves to directly control the displacement of the HAT segment to maintain dynamic equilibrium. Winter [11] also documented high intra-subject and inter-subject variability of the hip moments across trials and testing times. It was postulated that the high variability in hip moments was necessary on a stride-to-stride basis in an attempt to control the HAT segment and that changes in hip moment patterns were equally matched by alterations in knee moment patterns. It was hypothesized that such a deterministic trade-off between the hip and the knee indicated a stride dependent control of the HAT segment to maintain the total moment of support (M_s) to prevent the body from collapsing due to gravitational forces [11].

There is a controversy surrounding the postural control mechanisms in response to changes in dynamic equilibrium during gait and a paucity of literature serving to explain reactive balance mechanisms to unexpected perturbations. Of those investigations, only EMG and kinematic data have been presented and no studies have calculated the joint moments necessary for understanding the relative joint contributions in maintaining dynamic equilibrium. Therefore, the purpose of this investigation was to determine the effect of unexpected forward perturbations (FP) during gait on lower extremity joint moments and muscle EMG patterns in healthy subjects.

It was hypothesized that the forward perturbation would result in a greater knee and hip joint flexion and greater ankle plantarflexion. As a result of the alterations in lower extremity joint positions, it was also hypothesized that the perturbed limb would demonstrate a greater knee and hip extensor moment and a reduced ankle plantarflexion moment and increased ankle, knee, and hip joint extensor muscle EMG activity.

2. Methods

2.1. Participants

Ten (five males and five females) healthy young adults participated in the study. The mean age, body mass, and body height of subjects were 24.4 year (±3.1 year), 67.2 kg (±10.7 kg), and 170.1 cm (±9.3 cm), respectively. All subjects were physically active, participating in regular activity at least three times per week. No subject had a prior history of lower extremity infirmity or pathology, or was suffering from any osteoarthritic or musculoskeletal disease at the time of testing that may have affected the ability to perform the experiment. Prior to participation, each subject signed a consent form approved by the University's Human Subjects Compliance Committee.

2.2. Protocol

Unexpected perturbations were induced as subjects walked along a 5 m wooden walkway in which a force plate, capable of translational movement, was embedded. When preset, the force plate moved anteriorly or posteriorly a distance of 10 cm at a velocity of 40 cm/s upon heel contact. The selected velocity was based on previous literature reporting heel velocities during realistic slip movements when a person is walking on a slippery surface [12].

The subjects walked at a self-selected comfortable pace that was maintained throughout data collection via a metronome. Each subject began walking at a sufficient distance from the force plate so that the self-selected pace was attained prior to the foot of the test limb making contact with the center of the force plate. Muscle EMG, joint kinematic and kinetic data were collected while the subjects walked along the walkway for a 5 s period, which included the step prior to and following contact with the force plate.

Data were recorded from 48 trials using the subject's right limb. The first 12 trials consisted of 'true control' non-perturbations (NP) trials to establish normal walking gait and muscle activation patterns. Following the

NP true control trials, 36 additional trials were performed consisting of 12 FP, 12 NP 'catch' trials, and 12 backward perturbations (BP). The BP condition was not analyzed and, in addition to the random order of trials, was used to help prevent possible accommodation and anticipation of the FP condition. Subjects were not allowed to practice the perturbation trials.

There was a small risk that the subjects could fall when their balance was perturbed. To minimize the risk, the subjects wore a harness attached to an overhead track and were provided a handrail to grasp if needed.

2.3. Instrumentation

EMG data were collected using bipolar surface electrodes (DE-02, Delsys, Boston, MA, USA). The electrodes were placed on the skin overlying the muscle belly of the TA, medial head of the GAS, BF, and vastus lateralis (VL) of the test limbs. To achieve an optimal EMG signal and low impedance (<5 kΩ), three, 4 cm^2 areas of skin were sanded and cleaned, and electrode gel applied between the skin and electrodes in accordance to procedures outlined by De Luca et al. [13]. All raw EMG analog signals were on-line pre-amplified ($\times$ 7000), analog filtered (20–7000 Hz), and then converted into digital signals sampled at 1200 Hz for a 5 s duration via the Associated Measurement Laboratory (AMLAB) data acquisition system (AMLAB Inc., Sydney, Australia). Prior to data analysis, EMG signals were full-wave rectified and low-pass filtered at 6 Hz using a 4th order dual-pass Butterworth filter. EMG data for the FP and catch NP conditions were normalized to maximum EMG activity produced during the true control NP condition and expressed as the NP:FP ratio.

A six-degree of freedom custom-built force plate (Institute of Neuroscience Technical Service Group, University of Oregon, Eugene, OR, USA) equipped with strain gauges mounted underneath the four corners was used to measure the vertical (F_z), horizontal antero-posterior (F_x), and medio-lateral (F_y) ground reaction forces. Using a feedback electric circuit, the F_z forces also served as trigger signals to initiate the force plate movement when the signal registered approximately 40 N ($\sim$8% of body weight). During the forward perturbation condition, onset of force plate movement occurred at 3.1 ± 0.2% of stance (approximately 29.10 ± 0.19 ms after heel strike) and ended at 59.8 ± 2.5% of stance (approximately 543.21 ± 0.24 ms after onset). Kinetic data were recorded at 1200 Hz for a 5 s duration via the AMLAB system. Prior to analysis, kinetic data were low-pass filtered between 4 and 10 Hz using a 4th order dual-pass Butterworth filter. Selected filter frequencies were determined for each force signal based on specifications from the manufacturer.

Kinematic data were collected using a PEAK Performance Technologies Real-Time Data Acquisition System (Peak Performance Inc., Denver, CO, USA). Four cameras were positioned 4 m from the sagittal plane along the progression plane of the subject's gait path. The pre-determined criterion for tolerable error in space calibration was set at 0.2% (2 mm maximum error for a 1 m-long object). Five kinematic reflective markers were placed on the skin overlying the base of the fifth metatarsal, lateral malleolus, lateral condyle of the femur, greater trochanter of the femur, and acromion process of the scapula. A reflective marker was also placed on the force plate to register plate movement and serve as the point of reference for transformation of local center of pressure (COP) coordinates to global kinematic coordinates. Kinematic data were collected at 120 Hz for a 5 s duration with each of the four cameras synchronized with the AMLAB system. Each marker was then digitized for the entire collection period including the stride before and after the stance phase on the force plate. The digitized position data for all markers were then low-pass filtered between 4 and 8 Hz using a 4th order dual-pass Butterworth filter. Optimal filter frequencies were determined for each force signal based on power spectral analyses wherein 80% of the raw signal was retained after the filtering process. Linear and angular position, velocity and acceleration data were then calculated and exported for further analysis.

2.4. Inverse dynamics calculations

The magnitude of the segmental masses along with their moments of inertia were estimated using data reported by Dempster [14] and individual subject anthropometric data. COP was calculated from the ground reaction force data within the force plate local coordinate system. Joint moments were calculated through an inverse dynamics analysis using a custom written MATLAB (The MathWorks, Inc., Natick, MA, USA) computer program combining the anthropometric, kinematic, and kinetic data. Ankle, knee, and hip joint moments were expressed as a reaction moment to all external moments and represent the internal moments normalized to subject mass. All joint moments were expressed as positive values for extensor and plantarflexor moments. Extensor angular impulse (EAI) was calculated from the positive area under the joint moment curve. Joint powers were calculated as the product of the joint moments and angular velocities and normalized to subject mass.

2.5. Data analysis

Prior to analysis, each trial was partitioned for the stance phase of the gait cycle (heel strike to toe off), interpolated as a percent of stance, and an ensemble average was created by averaging the 12 trials for each condition. For the purpose of analyzing the temporal relationship between the two time-series curves, each ensemble average curve was divided into five phases (P) and five discrete points (Pt) that were selected according to discrete kinetic events determined from vertical and anterior/posterior ground reaction forces (Fig. 1). Phase 1 (P1) ranged from heel strike to initial loading (Pt1), phase 2 (P2) from Pt1 to first acceptance of full body weight (Pt3), phase 3 (P3) from Pt3 to mid-stance (MS), phase 4 (P4) from MS to second acceptance of full body weight (Pt5), and phase 5 (P5) from Pt5 to toe off. Two other discrete points (Pt2, Pt4) denoted the troughs between Pt1and Pt3 and between Pt3 and Pt5, respectively. Comparisons were made for between condition differences, if any, in average joint moments, powers, positions, and muscle EMG activity for each of the five phases and five discrete points.

2.6. Statistical analysis

Two-way repeated measures Analysis of Variance (ANOVAs) (10×3; $\alpha = 0.01$) and a priori post-hoc tests were used to determine differences, if any, between the three conditions. The independent variables were, (1) the five phases and five discrete points of stance; and (2) condition (true control NP, catch NP, and FP).

3. Results

No significant ($P > 0.01$) differences were found between the true control blocked NP trials and the randomized catch NP trials for any lower extremity variable. The total time of stance was significantly ($P <$

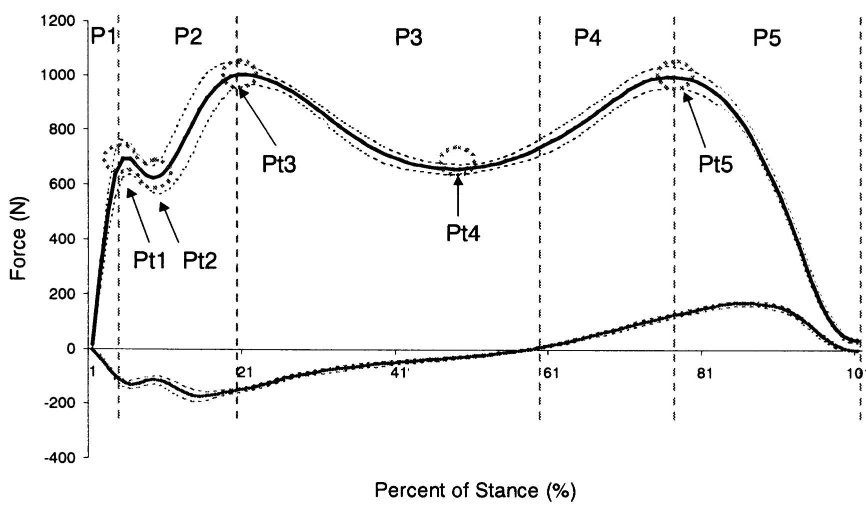

Fig. 1. Selection of five discrete points (dashed circles) and partitioning the stance phase into five phases (dashed vertical lines) according to discrete vertical and anterior/posterior kinetic events.

Table 1
Total time and percent of stance of the five phases and location of the five discrete points during stance phase of non-perturbation (NP) and forward perturbation (FP) conditions ($n = 10$)

Stance partition	NP		FP	
Phase (P)/Point(Pt)	Time (ms)	Percent of stance (%)	Time (ms)	Percent of stance (%)
P1	51.78 ± 10.44	6.34 ± 1.23	48.85 ± 9.06	5.02 ± 1.27
Pt1	51.78 ± 9.84	6.34 ± 1.21	48.85 ± 11.10	5.02 ± 1.33
Pt2	86.30 ± 11.39	10.22 ± 1.32	87.93 ± 15.04	9.64 ± 1.54
P2	129.45 ± 24.43	15.30 ± 3.33	185.63 ± 20.91	19.00 ± 2.14
Pt3	181.23 ± 14.07	21.64 ± 1.63	234.48 ± 21.10	24.02 ± 2.16
P3	284.64 ± 32.97	32.34 ± 3.82	341.95 ± 38.79	34.93 ± 3.97
Pt4	422.87 ± 12.69	49.22 ± 1.47	513.65 ± 18.26	52.61 ± 1.87
P4	207.27 ± 23.66	24.36 ± 3.07	185.63 ± 18.95	19.82 ± 1.94
Pt5	673.14 ± 28.22	78.34 ± 3.38	762.06 ± 34.68	78.77 ± 3.55
P5	190.07 ± 22.18	21.66 ± 2.65	215.76 ± 25.50	21.23 ± 2.27

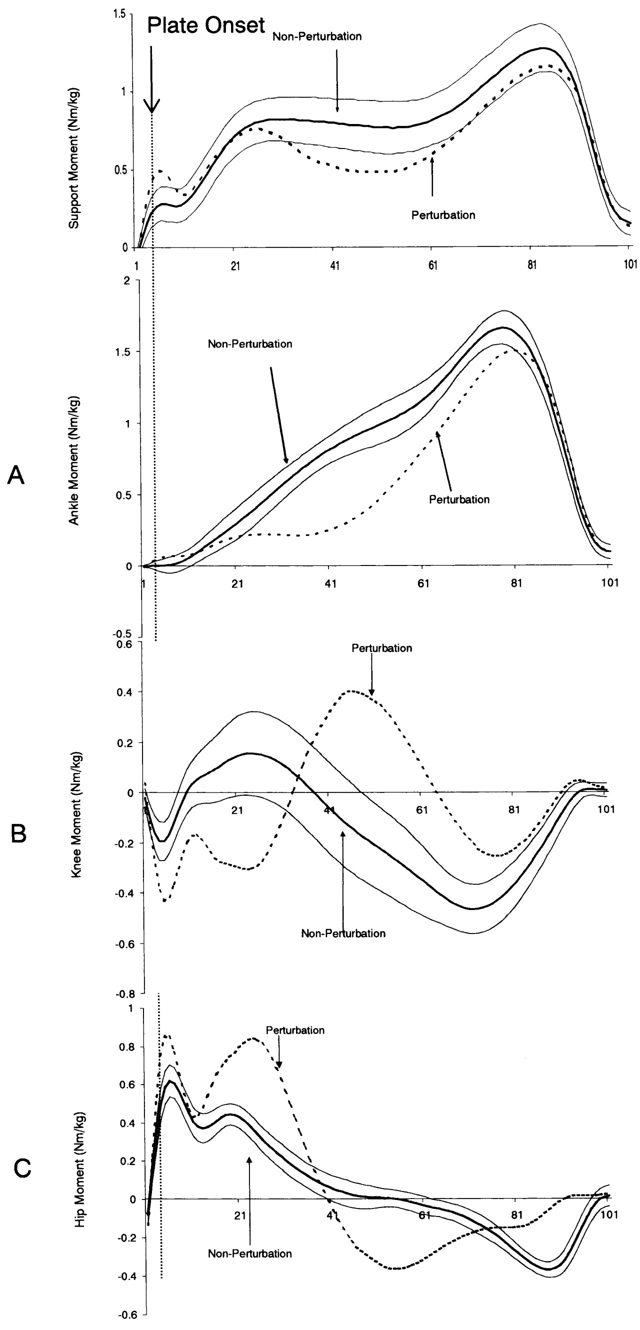

Fig. 2. Ankle (A), knee (B), and hip (C) joint moments and overall moment of support (M_s). Positive values indicate extensor and plantarflexor moments, negative values indicate flexor and dorsiflexor moments. Solid thick and thin lines are mean ± 1 S.D. for NP gait. Dashed thick line is mean of FP condition.

0.01) less for the NP condition (863.32 ± 77.27 ms) than the FP condition (977.59 ± 58.33 ms) and NP mid-stance occurred significantly ($P < 0.01$) earlier (49.6 ± 2.4% of total stance) than FP (59.8 ± 2.6% of total stance). Total time of the five phases and location of the five discrete points is summarized in Table 1.

3.1. Joint moments

The results revealed an overall positive M_s and a significantly ($P < 0.01$) reduced EAI (65.91 ± 5.61 N m/kg) for the FP condition compared with the NP (75.5 ± 6.78 N m/kg) condition (Fig. 2).

Table 2 presents lower extremity joint moments for each of the five phases (P) and five discrete points (Pt) of total stance during the NP and FP conditions as well as the total joint EAI for stance. The ankle exhibited significantly less EAI while the knee and hip exhibited significantly more EAI during the FP condition compared with NP (Table 2; Fig. 2A–C; $P < 0.01$).

The ankle NP plantarflexor moment rose steadily from heel strike through mid-stance to Pt5 before declining rapidly during the latter half of P5 (Table 2; Fig. 2A). The ankle FP plantarflexor moment remained relatively flat and was significantly smaller in magnitude that NP through the latter part of early stance (P2) and the first half of mid-stance (Pt3, P3, Pt4; Table 2; Fig. 2A; $P < 0.01$). The ankle FP moment paralleled but was significantly less than NP through the latter half of stance (P4, Pt5; Table 2; Fig. 2A; $P < 0.01$).

The knee NP generated an initial flexor moment in early stance and then followed a biphasic extensor–flexor–extensor moment pattern for early, mid-, and late stance periods, respectively (Table 2; Fig. 2B). The knee FP produced a significantly greater initial flexor moment than NP (P1, Pt1, Pt2) and, in contrast to NP, produced a flexor–extensor moment pattern during early (P2) and mid-stance (P2–Pt4) periods (Fig. 2B; $P < 0.01$). The knee FP moment paralleled NP for the remainder of stance although the knee produced a significantly smaller flexor moment during the latter half of mid-stance (P4) and a significantly greater extensor moment during late stance (Pt5, P5; Table 2; Fig. 2B; $P < 0.01$).

The hip NP extensor moment rose sharply in early stance and then decreased steadily until mid-stance after which a flexor moment was observed (Table 2; Fig. 2C). The hip FP moment paralleled but was significantly ($P < 0.01$) greater than NP immediately following the onset of force plate movement (P1, Pt1, Pt2, Table 2; Fig. 2C). In contrast to hip NP, the hip FP then produced an abrupt and significantly greater extensor moment (P2, Pt3), which rapidly decreased and became a large flexor moment through mid-stance (P3, Pt4, P4) that steadily declined through late stance (Pt5, P5; Table 2; Fig. 2C; $P < 0.01$).

3.2. Joint kinematics

Table 3 presents lower extremity joint position values during the NP and FP condition for P1–5 and Pt1–5 of total stance. The ankle NP position curve followed a plantarflexion–dorsiflexion–plantarflexion pattern over early, mid-, and late stance periods, respectively (Table 3; Fig. 3A). The mean ankle FP position curve generally paralleled the NP condition curve through the stance phase (Table 3; Fig. 3A). However, the FP ankle position was significantly more plantarflexed than NP during the early (Pt2, P2) and mid-stance (Pt3, P3) periods (Table 3; Fig. 3A; $P < 0.01$).

The knee NP position curve followed a flexion–extension–flexion pattern over early, mid- and late stance periods, respectively (Table 3; Fig. 3B). The knee FP position curve paralleled NP prior to and immediately following force plate translation after which, in contrast to NP, the knee position remained in a relatively static position ($\sim 15°$ flexion) until late stance (P2–P4; Table 3; Fig. 3B).

The hip NP position curve declined steadily from a flexed to extended position from early to mid-stance after which it followed a flexion–extension pattern from the latter half of mid-stance to late stance (Table 3; Fig. 3C). The hip FP position curve generally paralleled the NP condition curve throughout stance except the FP hip position was significantly more flexed during mid-stance (Pt3–P4) compared with NP (Table 3; Fig. 3C; $P < 0.01$).

3.3. Joint power

Table 4 presents lower extremity joint powers during the NP and FP conditions for P1–5 and Pt1–5 of total stance. The mean ankle NP ankle joint power curve revealed that small amounts of power were absorbed by the ankle during early stance and the first half of mid-stance after which the ankle sharply increased power generation until late stance (Table 4; Fig. 4A). The mean ankle FP joint power curve generally paralleled NP over the course of stance. However, ankle FP absorbed significantly less power during mid-stance (P2–P4) and produced significantly less power during late stance than ankle NP (Pt5, P5; Table 4; Fig. 4A; $P < 0.01$).

Table 2
Mean (±S.D.) of ankle, knee, and hip joint moments for non-perturbed (NP) and forward perturbation (FP) conditions ($n = 10$)

Stance partition	Ankle		Knee		Hip	
Phase (P)/Point(Pt)	FP	NP	FP	NP	FP	NP
P1	0.03±0.02	0.01±0.02	−0.26±0.06*	−0.13±0.06	0.49±0.11*	0.30±0.11
Pt1	0.01±0.04	0.01±0.04	−0.38±0.09*	−0.12±0.09	0.80±0.15*	0.39±0.17
Pt2	0.01±0.04	0.01±0.01	−0.27±0.07*	−0.09±0.08	0.59±0.14*	0.38±0.18
P2	0.17±0.06*	0.22±0.14	−0.14±0.12*	0.09±0.06	0.55±0.17*	0.24±0.13
Pt3	0.26±0.13*	0.50±1.16	0.22±0.23	0.17±0.11	0.14±0.31	0.16±0.14
P3	0.57±0.24*	0.83±0.09	0.02±0.14*	−0.08±0.12	−0.21±0.19*	0.04±0.13
Pt4	0.77±0.21*	0.97±0.11	0.09±0.12*	−0.23±0.08	−0.28±0.19*	−0.01±0.12
P4	1.28±0.16*	1.36±0.09	−0.22±0.11*	−0.41±0.09	−0.14±0.16*	−0.31±0.10
Pt5	1.51±0.12*	1.65±0.08	−0.27±0.13*	−0.44±0.07	−0.12±0.21*	−0.34±0.11
P5	0.91±0.08	0.98±0.01	−0.09±.009*	−0.18±0.04	−0.02±0.16*	−0.07±0.01
EAI	58.51±9.634*	78.88±4.49	8.39±2.74*	3.44±2.23	22.64±7.26*	12.93±7.19

Positive values indicate extensor and plantarflexor moments, negative values indicate flexor and dorsiflexor moments (N m/kg). *, Significantly different than corresponding NP condition ($P < 0.01$).

Table 3
Mean (±S.D.) of ankle, knee, and hip joint positions for non-perturbed (NP) and forward perturbation (FP) conditions ($n = 10$)

Stance partition	Ankle		Knee		Hip	
Phase (P)/Point(Pt)	FP	NP	FP	NP	FP	NP
P1	−3.79±3.61	−4.66±1.32	8.53±1.86	8.55±2.23	18.74±2.15	18.6±1.95
Pt1	−3.95±3.31	−5.47±1.85	9.46±1.65	10.31±2.35	18.15±2.33	17.94±1.92
Pt2	−3.46±1.01*	−5.45±1.65	10.47±2.24	10.86±2.68	17.55±2.25	17.77±1.97
P2	−2.03±1.32*	−1.08±1.74	11.76±2.02*	14.54±2.24	14.02±2.04	14.68±1.22
Pt3	2.17±1.60*	5.99±1.33	13.67±2.49*	16.13±2.18	12.84±2.03*	10.06±0.84
P3	6.04±1.15*	9.88±2.65	12.92±2.21	12.41±2.44	9.31±1.48*	7.45±1.35
Pt4	10.91±1.48	8.71±2.78	11.9±2.36*	9.52±2.01	7.96±1.44*	6.36±1.37
P4	11.35±2.77	10.48±3.20	10.74±2.34*	8.21±1.92	10.09±1.29*	8.46±1.88
Pt5	10.28±2.75	9.18±0.19	11.61±2.54	9.51±2.17	11.21±1.55	10.64±1.92
P5	1.05±2.92	−1.18±3.74	22.23±2.77	21.26±1.96	11.31±5.42	10.95±3.84

Positive values indicate flexion and dorsiflexion, negative values indicate extension and plantarflexion (°). *, Significantly different than corresponding NP condition ($P < 0.01$).

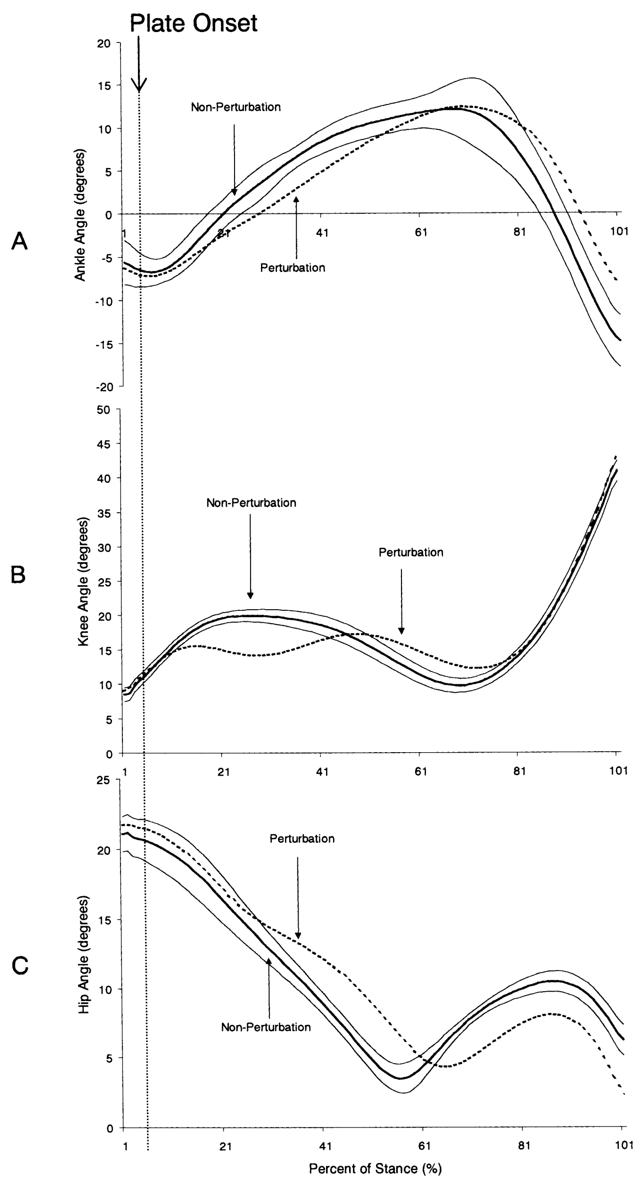

Fig. 3. Ankle (A), knee (B), and hip (C) joint positions. Positive values indicate flexion and dorsiflexion, negative values indicate extension and plantarflexion. Solid thick and thin lines are mean ± 1 S.D. for NP gait. Dashed thick line is mean of FP condition.

Knee NP and FP power curves differed markedly from one another. Both power curves were undulating in nature, with knee FP absorbing significantly more power following onset of force plate movement (Pt1; Table 4; Fig. 4B; $P<0.01$). During mid-stance, knee NP and FP power curves were prominently opposed to one another with knee FP generating power early in mid-stance (P3) while knee NP absorbed power (Table 4; Fig. 4B; $P<0.01$). Late in stance (Pt4) knee FP absorbed power while knee NP generated power (Table 4; Fig. 4B; $P<0.01$). During late stance (Pt5, P5), knee FP power absorption was significantly less than knee NP (Table 4; Fig. 4B; $P<0.01$).

The hip NP power curve exhibited power generation during early stance, power absorption during the first half of mid-stance followed by power generation for the latter half of mid-stance and late stance (Table 4; Fig. 4C). The hip FP power curve differed markedly from the hip NP power curve as significantly more power was generated by FP immediately following the onset of force plate movement (P1–Pt2) and throughout most of mid-stance (Pt3–P4; Table 4; Fig. 4C; $P<0.01$). During late FP stance, (P5), the hip absorbed power in contrast to power generation demonstrated in the late NP condition (Table 4; Fig. 4C; $P<0.01$).

3.4. Muscle EMG

Table 5 presents lower extremity muscle EMG values during the NP and FP conditions for P1–5 and Pt1–5 of total stance. Values expressed are the FP:NP ratio for the corresponding phase or discrete point of stance. The NP–TA EMG muscle response was characterized by strong activation during early stance followed by a rapid decrease to a low level for all of mid-stance and the first part of late stance, and then another surge of activity prior to toe-off (Fig. 5A). Compared with NP, FP–TA activity was significantly less during the first part of early stance (P1, Pt2) and significantly greater during the latter half of early stance (P2) and most of mid-stance (Pt3, P3; Table 5; Fig. 5A; $P<0.01$). FP–TA muscle activity then paralleled NP–TA activity for the remainder of stance.

The NP–GAS EMG muscle response was characterized by a steady rise in activity from heel strike through mid-stance and then a rapid decrease during late stance (Fig. 5B). Compared with NP, FP–GAS produced significantly more EMG activity during early stance (P1, Pt1) followed by significantly ($P<0.01$) less activity for the remainder of stance (P2–P5; Table 5; Fig. 5B; $P<0.01$).

The NP–VL EMG muscle response was characterized by a burst of activity during early stance that steadily drops and remains relatively low throughout the remainder of stance (Fig. 5C). Compared with NP, FP–VL activity produced significantly less EMG activity during most of early stance (P1–Pt2), followed by significantly greater activity for the remainder of stance (P2–P5; Table 5; Fig. 5C; $P<0.01$).

The NP–BF EMG muscle exhibited strong activation during early stance followed by a steady decrease for all of mid-stance and the first part of late stance and generated a surge of activity prior to toe-off (Fig. 5D). Compared with NP, FP–BF EMG produced significantly more EMG activity during early stance (P1–P2) and most of mid-stance (P2–Pt4) followed by reduced activity for the remainder of stance (P4–P5; Table 5; Fig. 5D; $P<0.01$).

4. Discussion

The purpose of this investigation was to determine the effect of unexpected FP during gait on lower extremity joint moments and muscle EMG patterns in healthy subjects. Tang et al. [5] postulated that TA EMG activity served to restore the disrupted ankle joint trajectory and realign the leg segment of the perturbed limb. However, Tang et al. [5] also reported that subjects exhibited GAS EMG inhibition in response to the FP but the reason for this inhibition was not addressed in that study. The results of this investigation also demonstrated significantly reduced GAS EMG muscle activity during early stance in response to the FP. It is possible that suppressed GAS activation may attenuate the effect of the FP-induced ankle plantarflexion and help maintain balance. Another possible explanation for the increase in TA and suppression of GAS activity may come from examination of the moments and powers produced at the ankle joint during the FP. A small eccentric ankle dorsiflexor moment (power absorption) followed by a large eccentric ankle plantarflexor moment was observed for early and mid-NP stance, respectively (Fig. 2A, Fig. 4A). During FP, no ankle dorsiflexor moment was observed early in stance, in contrast to NP. Instead, the ankle produced a sustained, but significantly reduced, ankle plantarflexor moment throughout stance as compared with NP. Since the knee angle is under the control of moments of force at the hip and ankle, as well as the knee due to action of bi-articular muscles [15], a stronger than normal ankle plantarflexor moment can serve to slow down, or even reverse, forward rotation of the leg segment, resulting in a reduction in knee flexion [16,17]. The sustained reduction in the ankle plantarflexor moment during early FP stance, as observed in this study, may have contributed to the static knee flexion position during the FP as a possible reactive balance strategy necessary during early FP stance.

The motor patterns of the knee during early and mid-stance FP consisted of a large flexor moment during early stance which did not switch to an extensor moment until significantly later in mid-stance compared with NP (Fig. 2B). Significant fluctuations in power production (Fig. 4B) and a static knee flexion position (Fig. 3B) were also observed during early and mid-stance of FP. An initial suppression of VL EMG activity, relative to NP, was observed early in FP stance followed by a strong VL activation coincident with a large BF EMG burst (Fig. 5C and D). The EMG activity demonstrated by these two antagonistic muscles is indicative of co-contraction, possibly to maintain knee joint stability during early and mid-FP stance [5].

In the present study a reciprocal trade-off between the knee and hip was demonstrated during FP mid-stance when the knee exhibited an extensor moment in contrast to the knee flexor moment, observed during NP (Fig. 2B). At this same time, the FP hip produced a large flexor moment in contrast to the extensor moment during NP (Fig. 2C). The reciprocal trade-off between the knee and hip moments may be necessary to maintain a positive M_s and dynamic equilibrium in response to an unexpected FP. It has been shown that there is a reciprocal trade-off between the hip and knee joints such that dynamic balance and control of the HAT segment occurs via a coordination between posterior muscles (hip extensors/knee flexors) and anterior muscles (hip flexors/knee extensors) acting at either joint [1,11].

During the late phase of FP stance, a reduction in the peak ankle plantarflexor moment and ankle power generation was observed as compared with NP (Fig. 2A, Fig. 3A). Since the ankle power absorption is reduced during the first half of FP stance, the subsequent drop in the ankle plantarflexor moment and

Table 4
Mean (±S.D.) of ankle, knee, and hip joint powers for non-perturbed (NP) and forward perturbation (FP) conditions ($n = 10$)

Stance Partition	Ankle		Knee		Hip	
Phase (P)/Point(Pt)	FP	NP	FP	NP	FP	NP
P1	0.06 ± 0.23	0.01 ± 0.01	−0.42 ± 0.24	−0.40 ± 0.19	0.56 ± 0.22*	0.36 ± 0.22
Pt1	0.08 ± 0.42	0.002 ± 0.19	−0.66 ± 0.45*	−0.37 ± 0.28	0.90 ± 0.27*	0.51 ± 0.32
Pt2	0.06 ± 0.44	−0.02 ± 0.18	−0.30 ± 0.17	−0.26 ± 0.20	0.79 ± 0.26*	0.49 ± 0.32
P2	−0.31 ± 0.17*	−0.53 ± 0.19	−0.10 ± 0.07*	0.06 ± 0.06	0.61 ± 0.21	0.46 ± 0.25
Pt3	−0.45 ± 0.22*	−0.93 ± 0.19	0.15 ± 0.27	−0.03 ± 0.11	0.01 ± 0.20*	−0.33 ± .022
P3	−0.38 ± 0.19*	−0.84 ± 0.12	−0.07 ± 0.11*	0.06 ± 0.01	0.08 ± 0.31*	−0.09 ± 0.14
Pt4	−0.39 ± 0.22*	0.94 ± 0.51	−0.13 ± 0.15*	0.22 ± 0.01	0.33 ± 0.03*	0.12 ± 0.07
P4	0.16 ± 0.81*	0.85 ± 0.55	−0.19 ± 0.12	−0.23 ± 0.11	0.25 ± 0.17*	0.04 ± 0.09
Pt5	2.96 ± 0.69	3.96 ± 1.64	−0.49 ± 0.26*	−0.92 ± 0.07	0.11 ± 0.27	0.01 ± 0.13
P5	3.42 ± 0.43*	4.14 ± 0.81	−0.26 ± 0.15*	−0.69 ± 0.14	−0.01 ± 0.01*	0.39 ± 0.08

Positive values indicate power generation, negative values indicate power absorption (W/kg) *, Significantly different than corresponding NP condition ($P < 0.01$).

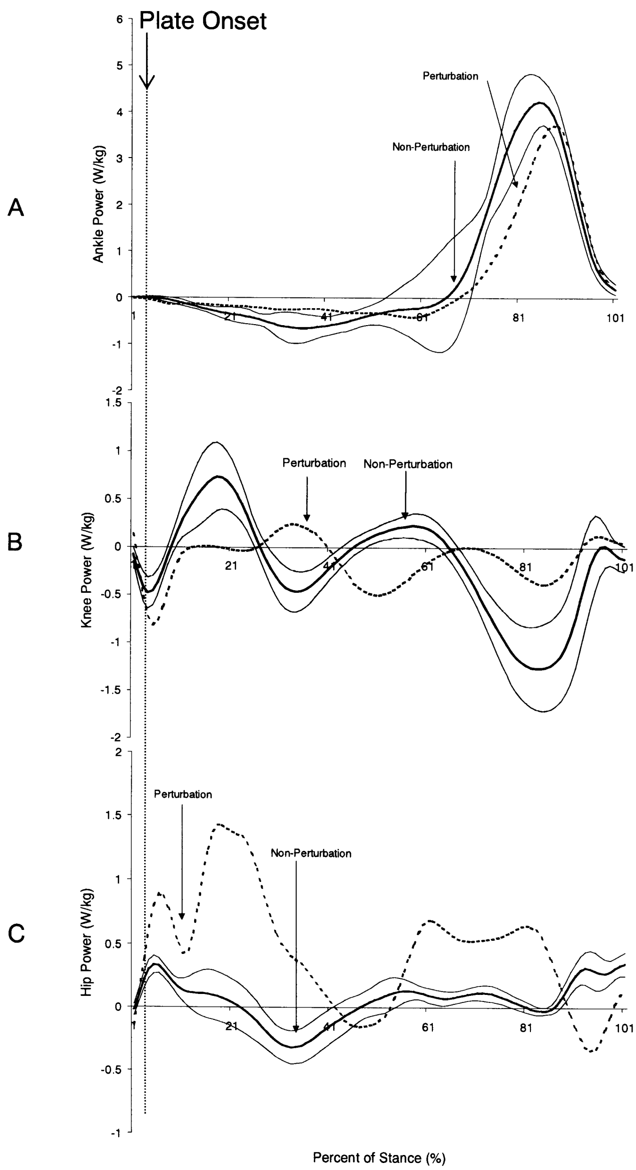

Fig. 4. Ankle (A), knee (B), and hip (C) joint powers. Positive and negative values are energy generation and absorption by the muscles. Solid thick and thin lines are mean ± 1 S.D. for NP gait. Dashed thick line is mean of FP condition.

power in late FP stance could possibly be attributed to a reduced storage of elastic energy in the plantarflexor muscles during the first half of stance. Elastic energy storage, however, was not directly measured in this study.

In the present investigation, postural adjustment responses observed during the stance phase of FP can be divided into three sections that are representative of the type of postural adjustment responses thought to occur within each. Within this conceptual model, postural responses observed from the time immediately following the onset of plate movement (< 70 ms; P1, Pt1, Pt2) are considered to be mechanical responses and can be solely attributed to viscoelastic changes in the tissues surrounding the lower extremity joints due to alteration of the body's COM relative to the base of support. It should be noted, however, that COM was not measured in this study. Responses observed between 70 and 250 ms (P2, Pt3) are considered to be a combination of mechanical and spinally-mediated (stretch reflex) neuromuscular responses, whereas responses observed after 250 ms (P3, Pt4, P4, Pt5, P5) are a combination of mechanical, spinally-mediated, and cortically-mediated neuromuscular responses [7,9,10, 18–20]. The results from this investigation demonstrate that an increase in the hip extensor moment as a result of the FP begins approximately 9.7 ms (~ 1% after plate onset) after initiation of the force plate movement, whereas the reactive TA EMG burst follows after approximately 97 ms (~ 10% after plate onset). Tang et al. [5] reported TA EMG onset latencies of 91.2 ms concomitant with suppression of GAS EMG activity and postulated that the control of these occurrences resulted from polysynaptic spinal reflexes or supraspinal loops. Other investigations have also demonstrated spinally-mediated muscle onset latencies of 90–100 ms during standing perturbations [7,9,10,19] and suggest that postural reactions to unexpected FP prior to 70 ms are due to the mechanical viscoelastic stretching of muscle, tendon, and joint capsule [10,20]. Herman et al. [20] studied the gait initiation torque patterns of humans and reported that changes in muscle stiffness (torque development) immediately following gait initiation may be attributed to the inherent mechanical properties of muscle relative to lengthening of series-elastic tissue rather than changes in motor discharge patterns (EMG). Nashner [10] described a significant stabilizing effect due primarily to ankle joint stiffness prior to leg muscle EMG activation approximately 90 ms after plate onset. It was suggested that cortically-mediated responses would not begin until at least 200 ms following plate onset and that muscle EMG responses prior to 200 ms

Table 5
Mean (±S.D.) FP:NP ratio of muscle EMG activity ($n = 10$)

Stance Partition	TA	GAS	BF	VL
Phase (P)/Point(Pt)				
P1	0.87 ± 0.09*	1.15 ± 0.09*	1.24 ± 0.12*	0.75 ± 0.18*
Pt1	0.96 ± 0.10	1.14 ± 0.08*	1.26 ± 0.08*	0.78 ± 0.17*
Pt2	0.70 ± 0.01*	1.01 ± 0.04	1.17 ± 0.06*	0.76 ± 0.12*
P2	1.71 ± 0.03*	0.92 ± 0.03*	1.51 ± 0.21*	1.23 ± 0.07*
Pt3	1.63 ± 0.15*	0.12 ± 0.22*	1.39 ± 0.01*	1.55 ± 0.14*
P3	1.17 ± 0.03*	0.80 ± 0.03*	1.41 ± 0.13*	1.51 ± 0.14*
Pt4	0.97 ± 0.03	1.10 ± 0.12	1.28 ± 0.07*	1.34 ± 0.14*
P4	1.06 ± 0.10	0.87 ± 0.03*	0.66 ± 0.01*	1.23 ± 0.07*
Pt5	0.99 ± 0.09	0.68 ± 0.08*	0.74 ± 0.08*	1.12 ± 0.13
P5	0.98 ± 0.03	0.85 ± 0.06*	0.22 ± 0.09*	1.04 ± 0.01

Values greater than 1.0 indicate FP EMG activity greater than NP condition, values less than 1.0 indicate FP EMG activity less than NP condition *, Significantly different than NP condition ($P < 0.01$).

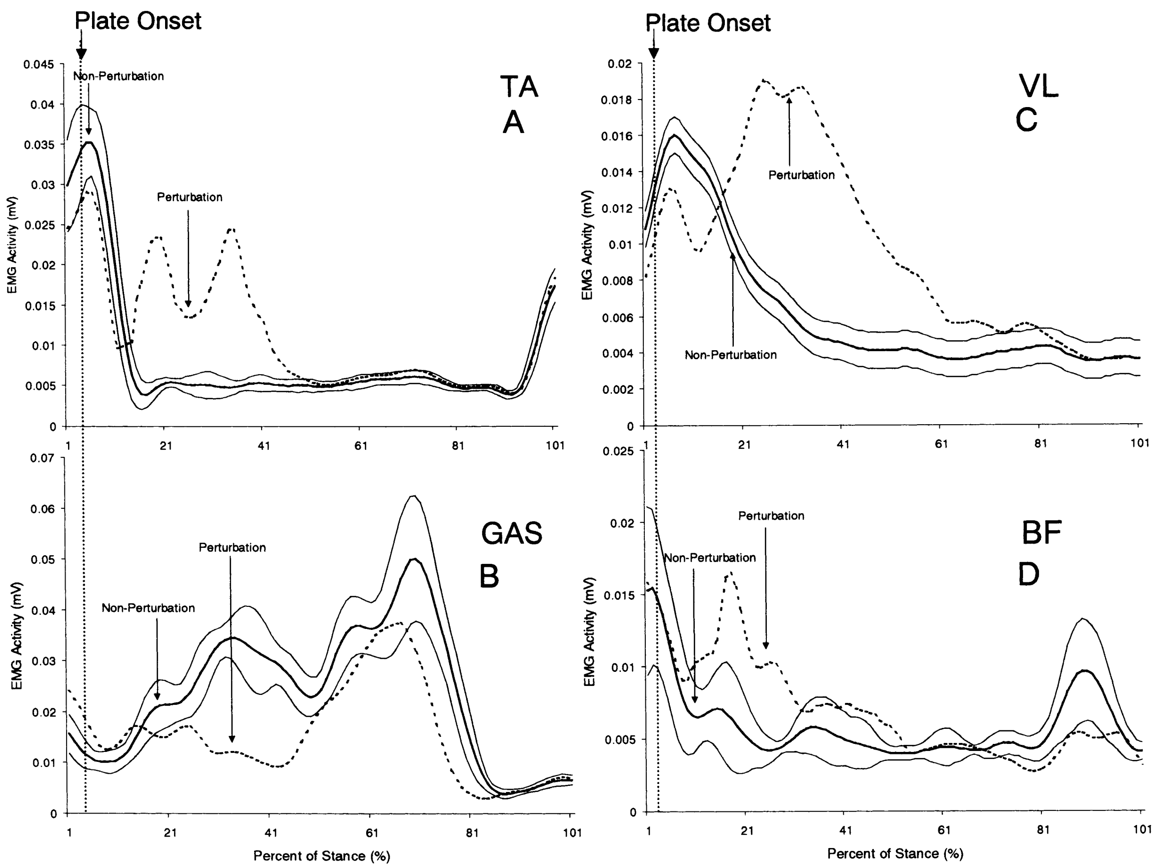

Fig. 5. Representative example of muscle EMG activity during FP (dashed thick line) and NP (solid thick and thin lines are mean ± 1 S.D.) conditions for the tibialis anterior (TA: A), gastrocnemius (GAS: B), vastus lateralis (VL: C), and biceps femoris (BF: D).

are spinally-mediated. More recent studies have suggested that cortically-mediated EMG responses begin as early as 160 ms following treadmill-induced gait perturbations [21,22].

The results of this investigation suggest that the initial responses to plate movement are mechanical in nature, rather than a neuromuscular response, and can be attributed to a viscoelastic stretching of pre-loaded joint muscles. For example, prior to heel strike, the muscles surrounding the hip are programmed to produce a certain amount of tension to provide vertical support to the body and prevent forward acceleration of the HAT segment upon heel strike and early in stance. In response to the FP, a large extensor hip moment is observed almost immediately after force plate movement and is possibly due to the mechanical stretching of pre-loaded hip muscles. Since vertical support of the body is maintained until muscle EMG responses are observed approximately 90 ms after force plate movement, postural adjustment responses prior to 70 ms are likely mechanical in nature.

The increase in TA EMG activity and reduction of GAS EMG activity exhibited approximately 97 ms after onset of FP is consistent with previous literature [5,10] and has been attributed to spinally-mediated neurological motor reflexes as a result of mechanical muscle stretch and alterations in lower extremity joint trajectories. Although the ankle is the first joint to undergo alterations in trajectory as a result of force plate movement, it seems unlikely that the ankle is the only input to the CNS to initiate postural adjustments since the hip and knee joints also undergo significant mechanical postural adjustments. EMG responses were observed in the BF and VL approximately 120–190 ms after force plate movement and appear to correspond with the delayed alteration of knee and hip joint trajectories, relative to the ankle joint, and subsequent spinally-mediated neuromuscular responses.

Afferent input from mechanical postural responses may provide input to the CNS to initiate spinally-mediated postural adjustments first observed at approximately 70 ms, and cortically-mediated neurological responses observed 160–200 ms after plate onset. Mechanical postural responses that occur later in FP stance may serve to provide continued afferent feedback to the CNS to promote continued spinal and higher level motor responses in an effort to maintain dynamic equilibrium, a positive M_s, and forward propulsion of the body during FP gait.

5. Conclusion

The findings of the present research suggest that response to perturbed gait is a synchronized effort of the lower extremity joints to maintain dynamic equilibrium and the overall M_s during the stance phase of gait. The muscles surrounding the hip were found to be most important in maintaining control of the HAT segment and preventing collapse of the lower extremity as an initial response to the FP. Muscle EMG activity from the leg and thigh segments and lower extremity joint kinematics demonstrated similar patterns compared with previous investigations. These results indicate that healthy subjects, in response to an unexpected FP, demonstrate joint moment and power patterns that are distinct from NP gait in order to maintain dynamic equilibrium during locomotion.

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2

The effects of unexpected mechanical perturbations during treadmill walking on spatiotemporal gait parameters, and the dynamic stability measures by which to quantify postural response

RESEARCH ARTICLE

The effects of unexpected mechanical perturbations during treadmill walking on spatiotemporal gait parameters, and the dynamic stability measures by which to quantify postural response

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Abstract

Most falls occur after a loss of balance following an unexpected perturbation such as a slip or a trip. Greater understanding of how humans control and maintain stability during perturbed walking may help to develop appropriate fall prevention programs. The aim of this study was to examine changes in spatiotemporal gait and stability parameters in response to sudden mechanical perturbations in medio-lateral (ML) and anterior-posterior (AP) direction during treadmill walking. Moreover, we aimed to evaluate which parameters are most representative to quantify postural recovery responses. Ten healthy adults (mean = 26.4, SD = 4.1 years) walked on a treadmill that provided unexpected discrete ML and AP surface horizontal perturbations. Participants walked under no perturbation (normal walking), and under left, right, forward, and backward sudden mechanical perturbation conditions. Gait parameters were computed including stride length (SL), step width (SW), and cadence, as well as dynamic stability in AP- (MoS-AP) and ML- (MoS-ML) directions. Gait and stability parameters were quantified by means, variability, and extreme values. Overall, participants walked with a shorter stride length, a wider step width, and a higher cadence during perturbed walking, but despite this, the effect of perturbations on means of SW and MoS-ML was not statistically significant. These effects were found to be significantly greater when the perturbations were applied toward the ML-direction. Variabilities, as well as extremes of gait-related parameters, showed strong responses to the perturbations. The higher variability as a response to perturbations might be an indicator of instability and fall risk, on the same note, an adaptation strategy and beneficial to recover balance. Parameters identified in this study may represent useful indicators of locomotor adaptation to successfully compensate sudden mechanical perturbation during

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walking. The potential association of the extracted parameters with fall risk needs to be determined in fall-prone populations.

Introduction

Falls are a serious clinical problem and often lead to injuries, the decline in mobility, and self-imposed limitations on daily activities, especially in older adults. Fall-related injuries increase costs for health care and rehabilitation and diminish the quality of life [1–3]. Most falls occur after a loss of balance while walking, which is the most common activity in daily life, and following an unexpected perturbation such as a slip or trip [4]. Therefore, understanding of how humans control balance and maintain stability during unexpected perturbed walking can help with assessment of balance recovery ability and thus may help to reduce the incidence of falls.

In order to enhance understanding of falls caused by perturbations, recent studies have examined changes in spatiotemporal gait parameters and dynamic stability (i.e., the margins of stability [5,6]) following perturbations. Evidence has demonstrated adaptations of spatiotemporal gait parameters to challenged walking by taking faster, shorter, and wider steps [7–11]. Consequently, an alteration in gait parameters led to increased margins of stability (MoS) and to enhanced stability during challenging walking [8,9]. While these alterations in spatiotemporal gait parameters and dynamic stability occurred during different types of perturbations, such as continuous mechanical and visual perturbations [9–14], it remains inconclusive whether these observable adaptations also occur during sudden mechanical surface perturbations in different directions.

The majority of perturbation studies has included perturbations only in the anterior-posterior (AP) [7,15–17] or in the medio-lateral (ML) direction [9,11,13,14,18,19]. However, each of these perturbations affects gait and stability in different ways, depending not only on the type but also on the direction of the perturbations. Exposure to the continuous support surface [10,12] and visual field [10,20,21] in both AP- and ML-directions produced anisotropic changes in gait variabilities. The effects of perturbations were also found to be significantly greater when perturbations were applied in the ML-direction [10,12,21]. Also, the unidirectionality (AP or ML) of the perturbation may help the subjects in developing a volitional plan for a stepping response thus lack's the ecological validity since falls in the real world are multidirectional and always unexpected [22,23]. Therefore, further studies on the effect of perturbations on gait-related parameters and dynamic stability, which include sudden mechanical surface perturbation in both AP- and ML-directions may reveal valuable information.

The means of gait characteristic appeared resistant to the effect of challenging walking depending on the challenge [18,24]. Alternatively, the response of variability to perturbations was stronger than the response of means during the continuous platform and visual perturbations [12]. This indicated an increased challenge in stability that was not captured by means but by the variability of parameters [12]. Thus, gait variability, which is defined as fluctuation in gait parameters from one step to the next, might be an important indicator of gait stability [25,26], and more responsive than the mean differences of the gait parameters.

Prior studies have used gait variability to characterize balance during walking [10,11,18,21,27]. However, studies on the response of variability of the gait parameters to perturbations provided contradictory results. Continuous support surface perturbations during walking in a static visual environment induced increased step width variability [14]. On the other hand, Francis et al. reported no significant increase in gait variability in young adults in response to visual ML perturbation [18]. These differences might appear due to different types of perturbations applied in these studies. In a recent work, Punt et al. explored the effects of multidirectional sudden mechanical perturbations in stroke survivors who prospectively experienced falls or no falls [28]. By comparing the gait characteristics and dynamic stability in both fallers and non-fallers group over every step after the perturbation, they observed no difference in individual's ability to cope with the perturbations. Although their study provided interesting insight into the response strategy in stroke survivors, the variability of the parameters which might reveal helpful information in discriminations between fallers and non-fallers was not included. There is a need for studies which examine the effect of sudden multidirectional unexpected mechanical perturbations on the variability of gait-related parameters.

Additionally, extremes of gait-related parameters may be a better representative estimate of the parameters in a challenging condition, such as perturbed walking compared with the mean values that traditionally being used in research [29]. Rispens et al. found a strong association between extremes relating to high gait quality and fall risk during daily life walking. During perturbed treadmill walking, extremes may better capture pronounced postural responses after perturbations, and in turn may be more sensitive indicators of gait stability [29]. To the best of our knowledge, there have been no studies to evaluate the response of extremes of gait-related parameters to quantify postural stability during perturbed walking.

The first aim of this study was to examine the changes in a candidate set of spatiotemporal gait and stability parameters in response to sudden unexpected multidirectional mechanical perturbations. Secondly, we aimed to evaluate the most affected parameters of this set for measuring the effect of perturbations on postural recovery responses. Means, variability, and extremes of gait-related parameters were used to specify responses during perturbed treadmill walking. We hypothesized that participants would exhibit: (1) alterations in spatiotemporal gait parameters to enhance dynamic stability and (2) a greater effect of perturbations on extremes and variability of measures, as compared to means.

Methods

Participants and experimental protocol

Ten healthy young adults (age: 26.4 ± 4.1 years, height: 1.7 ± 0.08 m, mass: 64.4 ± 12.5 kg, 7 females) participated in this study. All participants provided written informed consent and the study was approved by the ethical committee of the Medical Faculty, Tübingen University. Recruited subjects had no experience of walking on the perturbation treadmill.

Participants walked on a perturbation treadmill (Balance Tutor, MediTouch, Netanya, Israel) at the fixed speed of 1.11 ms^{-1} and were subjected to unexpected surface perturbations in left, right, forward, and backward directions (Fig 1). The system has been described in detail previously [30]. The treadmill platform is mounted on linear slides, which allow to translate it in the lateral direction. Left and right perturbations were induced by automatically moving the treadmill surface in ML-direction (12.8 cm and 1.5 ms^{-2}). Forward and backward perturbations were induced by acceleration and deceleration of the belt. To present the forward perturbation, the belt speed accelerated toward 2.5 ms^{-1} and subsequently decelerated toward 1.1 ms^{-1}. The backward perturbation was presented by deceleration of the belt speed toward 0 ms^{-1} and subsequent acceleration toward 1.1 ms^{-1}. First, the subjects completed 5 minutes (min) of normal walking on the perturbation treadmill without perturbations to become familiar with treadmill walking. The last min of the treadmill walking trial was used for data analysis (Normal) in order to measure the subject's normal walking pattern. Afterwards, 4 trials of 1 min perturbation treadmill walking were recorded. During each trial, participants were exposed to a single perturbation in a specific direction in order to become familiar with perturbed

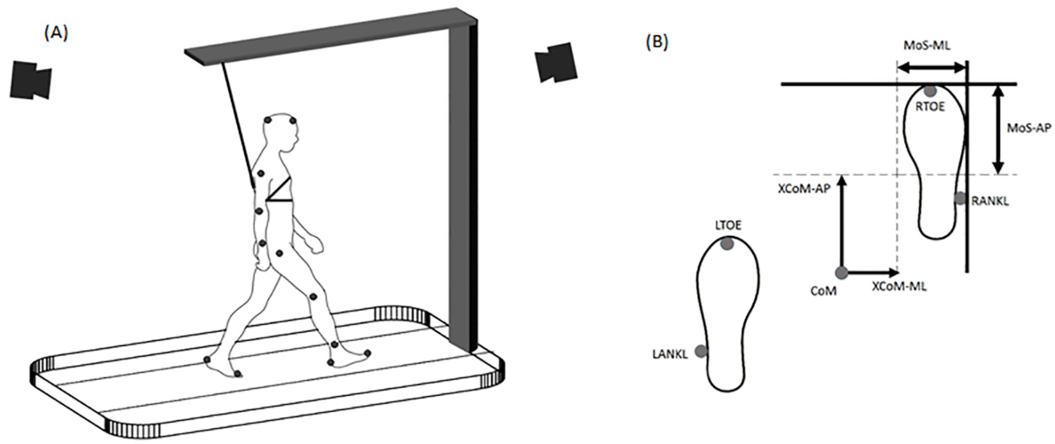

Fig 1. (A) A schematic drawing of the experimental setup. Forward and backward perturbations were induced by acceleration and deceleration of the treadmill's belt. Left and right perturbations were induced by moving the treadmill surface in the ML-direction. Reflective markers were placed at specific anatomical locations in accordance with the plug-in-gait marker set. (B) MoS-AP was defined as the AP distance between the XCoM-AP and the anterior boundary of the BoS, defined by the leading toe marker (either RTOE or LTOE for the right and the left foot, respectively). MoS-ML was defined as the ML distance between the XCoM-ML and the lateral boundary of the BoS, defined by the ankle marker (RANKL and LANKL for the right and the left foot, respectively).

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walking. Subsequently, 4 trials of 5 min perturbation treadmill walking including a series of 16 perturbations towards a specific direction were recorded. The moment of all perturbations was unpredictable. The time interval between perturbations ranged from 15–25 sec. All participants walked in their comfortable sport shoes. Subjects always wore a loss safety harness to prevent falls that prevented falls but did not restrict their gait.

Measurements and data analysis

Kinematic data were recorded at 200 Hz with an eight cameras motion capture system (Vicon Motion System, Oxford, UK). A total of 39 reflective markers were placed at specific anatomical locations in accordance with the Plug-In-Gait marker set (Bodybuilder, Plug in Gait model, Vicon Motion Systems, Oxford, UK). Motion data was analyzed using the Vicon Nexus software (Version 2.5). The time frame of interest was 15 sec including 5 sec before and 10 sec after the perturbation.

Spatiotemporal gait parameters including step length, step width, and cadence were measured at the instant of the heel strike. Heel strike was identified as the local maxima of the position of the heel markers in the AP-direction [31]. Stride length was defined as the AP-distance between heel markers at the instant of heel strike plus the treadmill translation during the stride. Step width was measured as the ML-distance between ankle markers at the moment of heel strike. Cadence was calculated as the number of steps per minute.

Dynamic margins of stability were adapted from Hof et al. [5]. In this study, the extrapolated center of mass (XCoM) was calculated as the position of the center of mass (CoM), plus its velocity multiplied by the factor $\sqrt{lg^{-1}}$, where g was the acceleration of gravity and l was the distance from the ankle marker of the trailing foot to the CoM at the instant of heel strike. The margins of stability in the anterior-posterior direction (MoS-AP) were calculated as the AP distance between the XCoM and the toe marker of the leading foot. The margins of stability in the ML-direction (MoS-ML) were calculated as the lateral distance between the XCoM and the ankle marker of the leading foot (Fig 1). MoS was calculated at heel strike for every step during each time frame (~ 24 steps per each 15 sec time frame). All processing and analyses were performed with custom MATLAB R2015a programs (Mathworks, Inc., Natic, USA). Measured values were visually checked regarding plausibility and wrong values resulted from an error in the calculations due to the disturbed trajectory of markers were removed for further analyzing.

For each time frame of 15 sec treadmill walking, the mean from all steps performed was calculated for each spatiotemporal gait parameter and MoS. Additionally, variability characterized as the standard deviation was calculated for each spatiotemporal gait parameter and MoS. Thus, gait characteristics were measured as the mean (mn) and standard deviation (sd) of the spatiotemporal gait parameters including stride length (SL_{mn} and SL_{sd}), step width (SW_{mn} and SW_{sd}), and cadence ($cadence_{mn}$ and $cadence_{sd}$). Dynamic stability was calculated as the mean and standard deviation of MoS in AP- ($MoS\text{-}AP_{mn}$ and $MoS\text{-}AP_{sd}$) and ML- ($MoS\text{-}ML_{mn}$ and $MoS\text{-}ML_{sd}$) directions.

In addition, extremes were estimated as the 10th and 90th percentiles of the stride length (SL_{P10} and SL_{P90}), step width (SW_{P10} and SW_{P90}), and cadence ($cadence_{P10}$ and $cadence_{P90}$), as well as MoS in AP- ($MoS\text{-}AP_{P10}$ and $MoS\text{-}AP_{P90}$) and ML- ($MoS\text{-}ML_{P10}$ and $MoS\text{-}ML_{P90}$) directions.

Statistical analysis

Multiple measures of variable including the mean, variability, and extremes of the spatiotemporal gait parameters as well as MoS in ML- and AP-directions were reduced to the mean values for each walking condition. Paired t-test and corresponding confidence interval (CI) was used to examine differences between normal walking and perturbed walking conditions. In addition, the effect size of responses was calculated using Cohen's *d* statistic (*d*) to describe the strength of the effect of perturbation conditions on each measurement. Cohen's *d* statistic was defined as the mean difference between normal and perturbed walking conditions divided by the standard deviation of changes between conditions.

All statistical analyses were performed using SAS software, version 9.4 (SAS Institute Inc., Cary, NC, USA) with a confidence interval of 95% for all comparisons.

Results

All subjects completed the experiment with no fall into the harness system during the perturbation trials. In total, 116 left, 130 right, 141 forward, and 144 backward perturbations were analyzed. The results for means, variabilities, and extremes of normal walking, as well as perturbed walking, are presented in Table 1. Also, results of statistical analyses including mean differences of perturbed walking conditions relative to normal walking, as well as the associated CI and effect sizes (i.e., Cohen's *d* statistic) are presented in Figs 2 and 3.

Means of gait parameters and dynamic stability

Overall, compared with unperturbed treadmill walking, participants walked with shorter stride length, wider step width, and higher cadence during ML perturbations. However, the effect of perturbations on SW_{mn} was not statistically significant (Fig 2A, 2B and 2C). Exposure to the right perturbation resulted in a significantly shorter stride length (Est. = -3.478, 95% CI [-5.302, -1.652], d = -1.363). In left perturbation, participants tended to decrease their stride length (Est. = -2.448, 95% CI [-5.101, 0.206], d = -0.66). However, there were no significant differences in SL_{mn}, SW_{mn}, and $Cadence_{mn}$ during forward and backward perturbations compared with unperturbed walking (Fig 2A, 2B and 2C).

Table 1. Results for spatiotemporal gait parameters and margins of stability during different walking conditions (mean and SD; n = 10).

	Condition				
	Normal	**Left**	**Right**	**Backward**	**Forward**
Stride length [cm]					
Mean	128.83 ± 8.68	126.38 ± 7.56	125.35 ± 7.98	129.59 ± 7.54	127.61 ± 7.56
Variability	2.08 ± 0.48	6.43 ± 1.75	7.86 ± 1.98	4.03 ± 0.78	5.41 ± 1.09
P10	126.32 ± 8.55	121.65 ± 6.61	121.52 ± 7.53	125.82 ± 7.25	122.62 ± 7.76
P90	131.54 ± 8.78	131.34 ± 8.00	130.53 ± 8.40	133.08 ± 8.09	132.18 ± 8.14
Step width [cm]					
Mean	20.97 ± 2.92	21.71 ± 3.30	21.69 ± 3.51	21.74 ± 3.22	21.14 ± 3.32
Variability	1.57 ± 0.39	3.18 ± 0.53	2.87 ± 0.38	2.02 ± 0.60	2.06 ± 0.72
P10	19.06 ± 2.86	18.56 ± 3.29	18.62 ± 3.70	19.29 ± 3.42	18.53 ± 3.46
P90	22.96 ± 3.03	25.19 ± 3.29	24.87 ± 3.34	24.49 ± 3.46	23.78 ± 3.52
Cadence [steps/min]					
Mean	103.96 ± 5.49	106.26 ± 6.26	107.14 ± 6.67	103.70 ± 5.72	105.08 ± 5.94
Variability	1.45 ± 0.40	4.83 ± 2.28	5.81 ± 1.76	2.50 ± 0.44	4.87 ± 1.35
P10	102.14 ± 6.63	102.81 ± 6.27	103.28 ± 6.33	101.06 ± 5.73	101.64 ± 5.89
P90	105.85 ± 6.64	110.96 ± 6.87	112.19 ± 7.09	106.25 ± 5.62	108.39 ± 5.81
MoS-ML [cm]					
Mean	8.89 ± 1.24	9.17 ± 1.41	9.07 ± 1.48	9.19 ± 1.38	8.92 ± 1.51
Variability	0.67 ± 0.16	1.43 ± 0.27	1.42 ± 0.18	0.97 ± 0.24	1.03 ± 0.20
P10	8.01 ± 1.30	7.73 ± 1.25	7.75 ± 1.58	8.05 ± 1.38	7.83 ± 1.39
P90	9.76 ± 1.26	10.62 ± 1.61	10.48 ± 1.46	10.42 ± 1.53	10.13 ± 1.62
MoS-AP [cm]					
Mean	9.38 ± 2.86	8.11 ± 2.39	7.61 ± 2.35	9.67 ± 2.64	8.81 ± 2.66
Variability	0.96 ± 0.25	3.37 ± 1.01	2.89 ± 0.55	1.62 ± 0.51	3.94 ± 0.48
P10	8.17 ± 3.00	4.78 ± 2.55	3.67 ± 2.57	7.78 ± 2.99	6.41 ± 2.41
P90	10.62 ± 2.69	11.01 ± 2.34	10.07 ± 2.17	11.32 ± 2.46	11.33 ± 2.74

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Similar to SL_{mn}, exposure to right perturbation resulted in significantly shorter MoS-AP_{mn} compared with unperturbed walking (Est. = -1.776, 95% CI [-2.665, -0.887], d = -1.429, Fig 3A). Also, MoS-AP_{mn} tended to decrease during left perturbation, however, the effect did not reach to the significant level (Est. = -1.269, 95% CI [-3.093, 0.555], d = -0.498). The perturbations had no significant effect on MoS-ML_{mn} (Fig 3B).

Variability of gait parameters and dynamic stability

During all perturbation conditions, the variability of stride length, step width, and cadence was significantly higher than during unperturbed walking (Fig 2D, 2E and 2F). Lateral perturbations resulted in an increase in the variability of stride length and step width than forward and backward perturbations. However, the strength of the effect on stride length variability appeared high during all perturbation conditions (Left: Est. = 4.352, 95% CI [3.091, 5.613], d = 2.468; Right: Est. = 5.784, 95% CI [4.271, 7.298], d = 2.733; Backward: Est. = 1.955, 95% CI [1.278, 2.632], d = 2.066; Forward: Est. = 3.331, 95% CI [2.488, 4.175], d = 2.826, Fig 2D). On the other hand, the results of SW_{sd} exhibited stronger effect of lateral perturbations than forward and backward perturbations (Left: Est. = 1.609, 95% CI [1.261, 1.958], d = 3.307; Right: Est. = 1.299, 95% CI [1.073, 1.526], d = 4.109; Backward: Est. = 0.448, 95% CI [0.142, 0.754], d = 1.048; Forward: Est. = 0.495, 95% CI [0.053, 0.937], d = 0.801, Fig 2E).

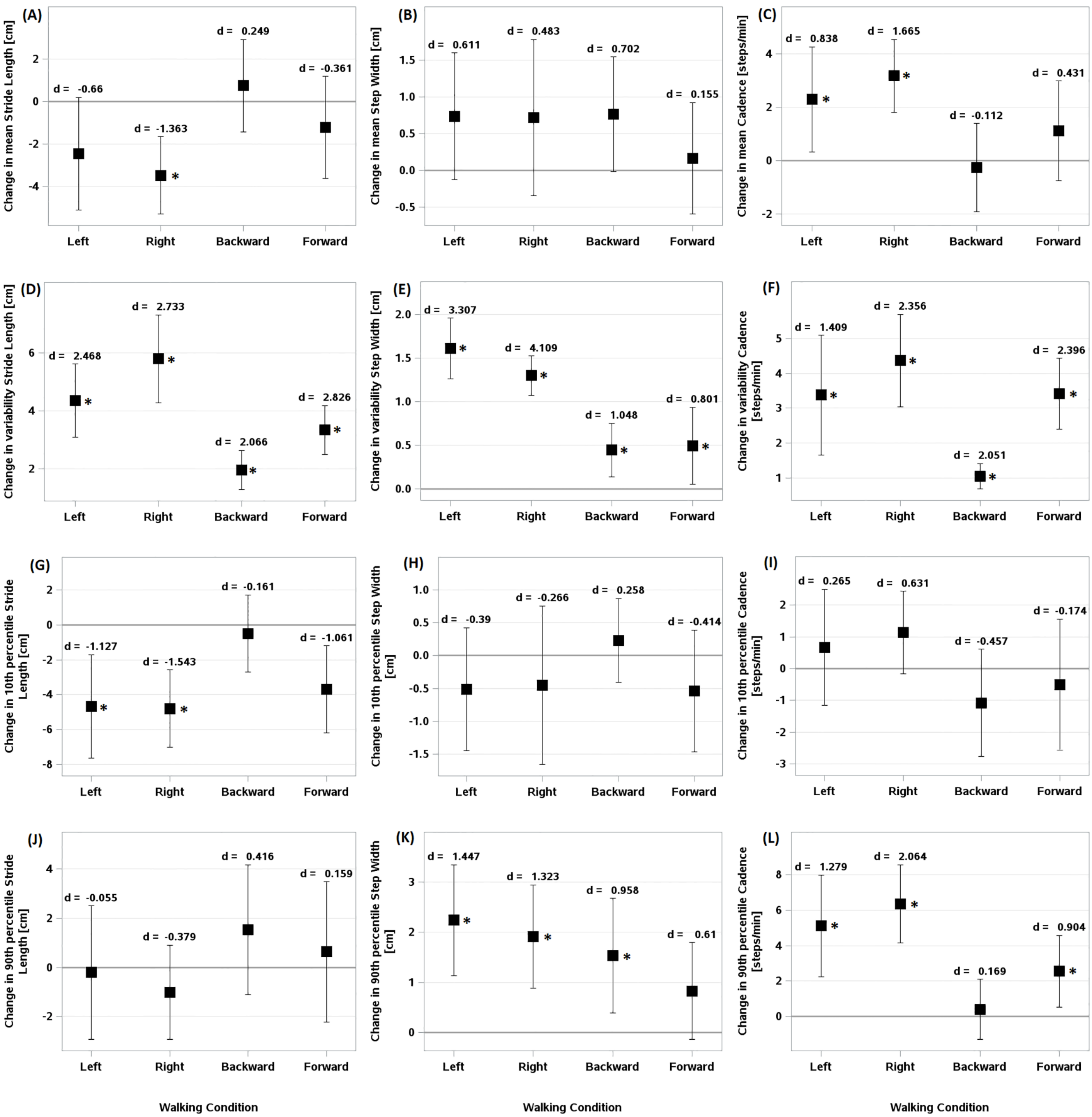

Fig 2. Difference of means, variability, and extremes of spatiotemporal gait parameters during perturbed walking conditions relative to normal walking. Difference of means of (A) stride length, (B) step width, and (C) cadence. Difference of variability of (D) stride length, (E) step width, and (F) cadence. Difference of 10th percentile of (G) stride length, (H) step width, and (I) cadence. Difference of 90th percentile of (J) stride length, (K) step width, and (L) cadence. *d* indicates Cohen's *d* statistic effect size. Error bars indicate confidence intervals. (*) indicates statistically significant differences from Normal walking.

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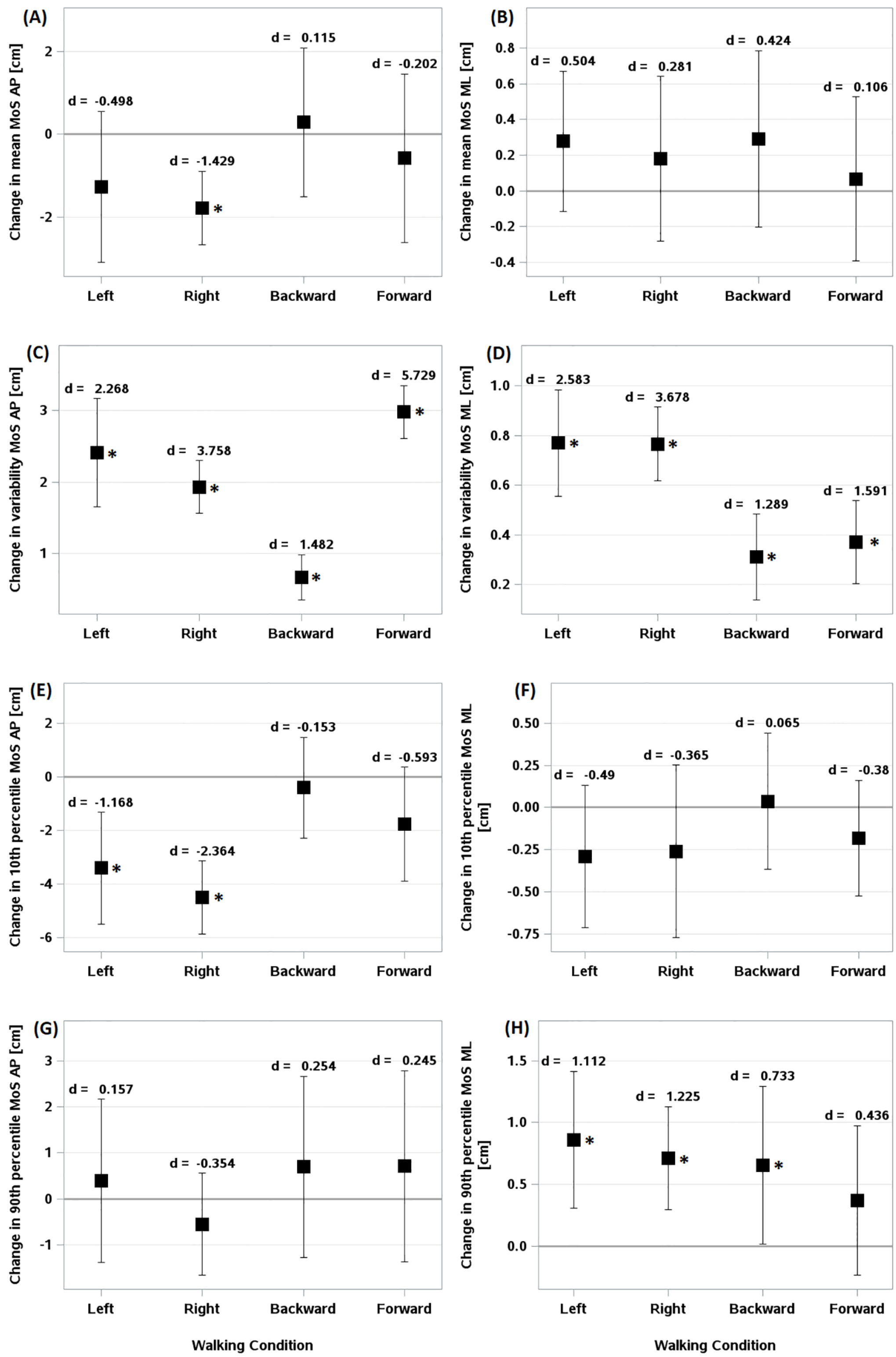

Fig 3. Difference of means, variability, and extremes of dynamic stability during perturbed walking conditions relative to normal walking. Difference of means of (A) MoS-AP and (B) MoS-ML. difference of variability of (C) MoS-AP and (D) MoS-ML. difference of 10th percentile of (E) MoS-AP and (F) MoS-ML. difference of 90th percentile of (G) MoS-AP and (H) MoS-ML. *d* indicates Cohen's *d* statistic effect size. Error bars indicate confidence intervals. (*) indicates statistically significant differences from Normal walking.

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Similar to the results of gait parameters, the dynamic stability exhibited significantly greater variability during all perturbation conditions relative to unperturbed treadmill walking (Fig 3C and 3D). However, forward perturbation had greater effect on MoS-AP_{sd} than on MoS-ML_{sd} (Est. = 2.979, 95% CI [2.607, 3.351], d = 5.729 and Est. = 0.371, 95% CI [0.204, 0.537], d = 1.591, respectively).

Extreme values

The results for extremes of spatiotemporal gait parameters showed no significant differences between SL_{P90}, SW_{P10}, and $Cadence_{P10}$ during perturbation walking conditions compared with unperturbed treadmill walking (Fig 2J, 2H and 2I). SL_{P10} during lateral and forward perturbations was significantly shorter than during unperturbed walking (Left: Est. = -4.663, 95% CI [-7.624, -1.702], d = -1.127; Right: Est. = -4.794, 95% CI [-7.017, -2.572], d = -1.543; Forward: Est. = -3.699, 95% CI [-6.192, -1.205], d = -1.061, Fig 2G). Also, SW_{P90} significantly increased during lateral and backward perturbations (Left: Est. = 2.239, 95% CI [1.132, 3.347], d = 1.447; Right: Est. = 1.913, 95% CI [0.879, 2.948], d = 1.323; Backward: Est. = 1.534, 95% CI [0.389, 2.679], d = 0.958, Fig 2K). In addition, $cadence_{P90}$ during sideway and forward perturbations was significantly greater than during unperturbed walking, however the effect of lateral perturbations was stronger compared with backward perturbation (Left: Est. = 5.11, 95% CI [2.253, 7.968], d = 1.279; Right: Est. = 6.349, 95% CI [4.148, 8.549], d = 2.064; Forward: Est. = 2.549, 95% CI [0.531, 4.568], d = 0.904, Fig 2L).

Similar to the results of step width, MoS-ML_{P90} during lateral and backward perturbations was significantly larger than during unperturbed walking (Left: Est. = 0.861, 95% CI [0.307, 1.414], d = 1.112; Right: Est. = 0.714, 95% CI [0.297, 1.131], d = 1.225; Backward: Est. = 0.656, 95% CI [0.016, 1.297], d = 0.733, Fig 3H). However, the results of MoS-ML_{P10} showed no significant change between perturbed and unperturbed gait (Fig 3F). Also, MoS-AP_{P90} was not significantly different between perturbed and unperturbed treadmill walking (Fig 3G), whereas MoS-AP_{P10} during ML perturbation was significantly greater than during unperturbed walking (Left: Est. = -3.401, 95% CI [-5.484, -1.318], d = -1.168; Right: Est. = -4.505, 95% CI [-5.868, -3.142], d = -2.364, Fig 3E).

Discussion

In this study, we found that spatiotemporal gait parameters, as well as MoS, were affected during exposure to AP- and ML- perturbations depending on the direction of the perturbations. Participants took shorter, wider, and faster steps in order to increase their dynamic stability in balance recovery during walking. More noteworthy was the increase in variability of these parameters relative to unperturbed walking. These effects were also found to be significantly greater when the perturbations were applied in the ML-direction.

Interestingly and as one might have expected by theory, the response of stride length (i.e. AP response of spatial gait parameters) and MoS-AP (i.e. AP response of dynamic stability) exhibited the same pattern of response to perturbations. Similarly, the response pattern of step width (i.e. ML response of spatial gait parameters) and MoS-ML (i.e. ML response of dynamic stability) appeared comparable. In addition, the response pattern of cadence (i.e., temporal gait parameter) was reversely the same as that for stride length. Based on the theoretical models, in which the human body during walking is modeled as a simple inverted pendulum, cadence, stride length, and walking speed cannot be adapted independently from each other [5,6,8,32]. In the present study, subjects walked on the treadmill with a fixed walking speed, therefore cadence was adapted according to the stride length.

Previous studies showed decreases in stride length, increases in step width and cadence with increasing perturbation intensity [9–11,33]. In this study, subjects exhibited shorter, larger, and faster steps during ML than AP perturbations, suggesting that ML perturbations were more challenging than AP perturbations, which is consistent with McIntosh et al. who used ML and AP overground platform perturbations during walking [34]. However, they quantified responses by CoM displacement and velocity, thus it remained unknown to what extent the stability of gait was affected by perturbations.

In line with previous studies, MoS-AP significantly decreased in response to ML perturbations [12]. MoS-AP is defined as the distance between the AP boundaries of the base of support (BoS) and XCoM. Shorter and faster steps, which bring the CoM closer to the moving BoS, improved stability in AP-direction [7,9,32,33]. Conversely, MoS-ML slightly increased in response to applied perturbations implies a decrease in risk of falling [9,12]. Similar to the previous studies, our results show that lateral dynamic stability was controlled by taking slightly wider steps to maintain stable walking during the perturbed walking [6,9,12]. The MoS in ML direction is defined as the distance between the ML borders of the BoS and XCoM. Thus, increased step width resulted in an increase in MoS-ML [9,20].

Perturbations had a strong effect on variabilities, indicating that step irregularity is a specific characteristic of walking adaptability during perturbed walking [10,11,13,21]. Our results suggest that looking at the variability of parameters over a series of steps is a responsive measure of gait adaptations happening during perturbed walking. Importantly, it should be noted that in this method, the effect of the perturbations on the mean of the parameters could be smeared out, since it was measured over a series of steps and not over every single step after the perturbation. Despite limited responsiveness for measuring the effects on means, the presented approach of capturing the variability may represent a useful measure in future studies estimating fall risk in fall-prone populations. For instance, in a recent study, Punt et al. reported no difference between fallers and non-fallers ability to cope with perturbation when measuring mean of the parameters over every single step following the perturbation [28]. In their study, the effect of the perturbations on gait variability over series of steps (i.e. fluctuations) was not investigated, which might be helpful in providing additional information to discriminate between fallers and non-fallers. Our findings of high responsiveness of variability parameters are in agreement with Terry et al. who reported variabilities of CoM position and step width as the most sensitive parameters in response to continuous visual and mechanical perturbations toward ML-direction [13]. Also, in a recent study, Stokes et al. reported a more profound effect of continuous visual ML perturbations on variabilities of step width, step length, and trunk sway [11].

Significantly greater variability in response to ML perturbations indicates that to maintain stability, participants needed to exert greater control in response to ML perturbations [10,21,35]. The variability of SL was strongly affected by both ML and AP perturbations, whereas the effect of ML perturbations on the variability of SW was much greater than the effect of AP perturbations. MoS variability increased during all perturbed walking conditions. However, similar to variabilities of gait parameters, the variability of MoS was also greater for ML perturbations, as reported previously [12], reflecting the increased fluctuations in the placement of protective stepping after the onset of the perturbation in order to enhance stability [27]. Additionally, the variability of MoS-AP during the forward perturbation increased which was also reported by Young et al., demonstrating higher fluctuations of MoS-AP in the forward direction [12]. In the present study, gait instability was analyzed using an approach similar to that used by Lipsitz et al. [36] measuring heart rate variability and by Hausdorff et al. [37] measuring gait variability. The higher variability (i.e., more fluctuations) during and immediately after recovery stepping may be referred to as unsteadiness. In this sense, the variability of gait and stability parameters may be used as a marker of unsteadiness, instability, and fall risk. This should be further explored by applying this method in older adults and impaired population since not all variability is a mark of poor locomotor control. As in heart rate variability, some variability may reflect adaptability and be beneficial especially after an unexpected loss of balance. Indeed, the ability to adapt gait when negotiating unexpected hazards is crucial to maintain stability and avoid falling [38]. In the present study, the healthy young participants experienced no difficulty and no fall during perturbed walking. Thus, the high variability may show the ability of the young subjects to adapt the gait pattern which may be a healthy behavior to respond to unexpected perturbation. This initial work suggests that just as there is much to be gained by investigating gait and heart rate dynamics, above and beyond the study of the average heart rate and gait dynamics, similar investigations of step dynamics after an unexpected loss of balance may provide insight into postural stability and may also have clinical applications.

ML perturbations resulted in a deviation from the straight walking trajectory. Consequently, a lateral step or a crossing step was necessary to prevent sideward fall. Probably, increasing the step width causing increased MoS-ML which results in decreasing the risk of a sideward fall was prioritized above the stability in AP-direction. Therefore, participants in this study increased the variability of AP responses as well as ML responses to compensate for the higher risk of fall following the ML perturbations by taking wider and shorter steps. But AP perturbations resulted in an interruption of the forward progression. In this case, the risk of fall in backward and forward direction could decrease, respectively, by taking a backward or a fast and short forward step which resulted in the higher effect on the variability of AP responses than on ML responses. This observation suggests that presenting the ML perturbations affected stability in both ML- and AP-directions with a stronger effect in sideway fall than AP falls, and AP perturbations resulted in a stronger effect in the direction of the presented perturbation.

Backward perturbation reduced the distance between the anterior border of the BoS and the XCoM thus increased MoS-AP. It should be noted that increase in MoS-AP simultaneously might have the disadvantage increasing the risk of a backward loss of balance. Consequently, subjects took wider steps to recover stability. The increased step width during backward perturbation resulted in a greater MoS in ML-direction. However, the results of backward perturbation in this study should be interpreted with some cautions. Backward perturbations were presented by deceleration of the treadmill belt, which was accompanied by a sudden stop in the belt movement. Thus, gait cycles included in the backward perturbation consisted of gait cycles before and after the belt stop, and motion's frames related to the stop of the belt were excluded from the analysis.

Extremes related to 'high gait quality' (HGQ) contain information about the best possible performance in the high-risk situation, whereas extremes related to 'low gait quality' (LGQ) contain information about responding to the risk which is related to the more demanding situations [29]. Therefore, together with the findings of this study, HGQ parameters are related to responses which show larger stride length (SL_{P90}), shorter step width (SW_{P10}), lower cadence ($cadence_{P10}$), higher MoS-AP ($MoS\text{-}AP_{P90}$), and lower MoS-ML ($MoS\text{-}ML_{P10}$). While, LGQ parameters are expected to represent subject's responses in the high-risk situations (i.e. during perturbations) which show shorter stride length (SL_{P10}), larger step width (SW_{P90}), higher cadence ($cadence_{P90}$), lower MoS-AP ($MoS\text{-}AP_{P10}$), and higher MoS-ML ($MoS\text{-}ML_{P90}$).

HGQ parameters during perturbed walking exhibited no difference with that of normal walking. Thus, they showed no sensitivity to perturbations. As suggested by Rispens et al., perhaps the HGQ extremes are an accurate estimation of the individual's capacities and do not capture the effect of perturbations [29]. Therefore, they showed the capacity and the best performance of young healthy adults in response to perturbations which exhibited no difference with normal walking.

Interestingly, the results of LGQ for all parameters were similar with the results of means and showed the same response pattern. However, the effect of LGQ of parameters was somewhat more significant and stronger compared to means. Thus, it seems that LGQ were more responsive and might be representative of the effect of unexpected perturbations.

There are some limitations in this study. First, due to technical limitations of the treadmill, all expected numbers of perturbations were not presented. Second, trials were not presented in a randomized order, therefore, the results of each condition could be influenced by learning of the previous condition. However, this fact does not interfere with the findings of this study since the main goal of this exploratory experiment was to find the effect of perturbations on spatiotemporal gait and dynamic stability parameters in order to evaluate the most sensitive measures which can better represent the effect of perturbations. Third, the data came from a fairly small sample of relatively healthy young adults. Thus there is a need to investigate larger sample sizes and explore older and "weaker" populations. Forth, there was no reflective markers attached to the treadmill. Consequently, the exact frame in which the perturbation was presented was undetectable. To address this limitation, all parameters were measured over a series of recovery steps and not over every single step after the perturbation. In this study, the extreme of the parameters may have captured the immediate effect of the perturbations on the parameters. Therefore, the present approach may potentially capture both the local effects (extremes) and the fluctuations over a series of steps (variability), although this needs to be validated in future studies. The detected information on extremes and variability of the parameters should be clinically validated as a fall risk assessment by applying this method on fall-prone populations. We acknowledge that the method of measurement over series of steps from a perturbation trial arose some limitations such as missing the subtleties that happen around the single steps following the perturbation. While the approach of analyzing a series of steps provided interesting information about the variability, it may smear out the effects of means. Therefore, the effect of the perturbations on the immediate steps after the perturbations should be investigated in future studies. In addition, the moment of the perturbation was adjusted to mid-stance of the left foot. However, there was a delay in triggering of the perturbations due to limitations in the setup of the treadmill device and since we could not detect the frame in which the perturbation was presented, the exact moment of the perturbations could not be determined. Thus, some cautions in interpreting the results should be taken into account, considering that depending on the moment of the perturbation within the gait cycle the response is different.

Conclusions

The results show that the increase in cadence and step width, as well as the decrease in stride length, are strategies to increase MoS, and thus to decrease the probability of falling in the presence of perturbations. The present study also suggests that frontal plane fluctuations (ML variability) are more variable compared with AP variability. Thus, the variability of responses depends not only on the status of the individuals but also depends on the type and direction of the perturbation. The participants were more sensitive to ML perturbations than to AP perturbations which shows the importance of including ML perturbations in assessment protocols. Variabilities, as well as extremes of gait-related parameters, showed strong responses for measuring the effects of perturbations. Therefore, measuring variabilities and extremes of the parameters in addition to means can help to better understand balance control strategies and may be used as a marker of unsteadiness, instability, and fall risk. Further studies need to evaluate whether similar postural responses exist in older adults with different balance control abilities, such as between fallers and non-fallers. In this context, this study can be useful for designing advanced stability and gait evaluation and for introducing novel assessment protocols for estimating fall risk.

Supporting information

S1 Data. Data of the gait characteristics and dynamic stability. Parameters including SL, SW, cadence, MoS-ML, and MoS-AP were measured over each gait cycle during the time frames of interest in each walking condition.
(XLSX)

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Unexpected perturbations training improves balance control and voluntary stepping times in older adults - a double blind randomized control trial

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Unexpected perturbations training improves balance control and voluntary stepping times in older adults - a double blind randomized control trial

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Abstract

Background: Falls are common among elderly, most of them occur while slipping or tripping during walking. We aimed to explore whether a training program that incorporates unexpected loss of balance during walking able to improve risk factors for falls.

Methods: In a double-blind randomized controlled trial 53 community dwelling older adults (age 80.1±5.6 years), were recruited and randomly allocated to an intervention group ($n = 27$) or a control group ($n = 26$). The intervention group received 24 training sessions over 3 months that included unexpected perturbation of balance exercises during treadmill walking. The control group performed treadmill walking with no perturbations. The primary outcome measures were the voluntary step execution times, traditional postural sway parameters and Stabilogram-Diffusion Analysis. The secondary outcome measures were the fall efficacy Scale (FES), self-reported late life function (LLFDI), and Performance-Oriented Mobility Assessment (POMA).

Results: Compared to control, participation in intervention program that includes unexpected loss of balance during walking led to faster Voluntary Step Execution Times under single ($p = 0.002$; effect size [ES] =0.75) and dual task ($p = 0.003$; [ES] = 0.89) conditions; intervention group subjects showed improvement in Short-term Effective diffusion coefficients in the mediolateral direction of the Stabilogram-Diffusion Analysis under eyes closed conditions ($p = 0.012$, [ES] = 0.92). Compared to control there were no significant changes in FES, LLFDI, and POMA.

Conclusions: An intervention program that includes unexpected loss of balance during walking can improve voluntary stepping times and balance control, both previously reported as risk factors for falls. This however, did not transferred to a change self-reported function and FES.

Trial registration: ClinicalTrials.gov Registration number: NCT01439451.

Keywords: Aging, Falls, Unexpected perturbation of balance, Step execution, Postural stability

Background

Falls are a major problem among elderly population; they are the leading cause of injury above the age of 65 [1]. Of those who fall in the U.S., 20 to 30 % suffer moderate to severe injuries that reduce mobility and independence, and increase the risk of death [2]. The financial impact in 2000, adjusted for inflation, was $30 billion and is expected to reach $67.7 billion by 2020 [1].

Walking is the major activity in which large proportion of falls in older adults occurs [3]. Sixty percent of outdoor falls among older adults resulted from slips or trips [4]. Even among older adults capable of independent walking, there could be a substantial decline in balance performance, which does not become evident until a slip or a trip happens [5]. In fact, the inability to step rapidly in response to unexpected loss of balance ultimately

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determines whether a fall occurs [6, 7]. Thus, a better way to improve balance, improve stepping and reduce risk of falls may be to direct preventive efforts towards older adults who have not yet fallen. Until recently these balance recovery responses were considered hardwired postural reflexes that could not be influenced by training. However, [8–10] it was showed that older adults were able to adapt in a reactive manner after participation in a perturbation exercises that challenged the mechanisms responsible for dynamic stability (i.e., increase in base of support and counter-rotating segments around the center of mass).

A number of studies have begun to examine the effect of perturbation training on balance of older adults. Shimada et al. [11] found improvement in mobility and a trend to fall reduction after split treadmill training. This training method is a very unnatural condition, given that most people walk with the same velocity in each leg. Other studies have other issues. Pai et al. [12] showed a rapid decrease in loss of balance in response to multiple presentations of a slip perturbation after rising from sit to stand. Mansfield et al [13] found that older adults with a history of falls or instability reduced the frequency of multi-step reactions and foot collisions after perturbation training while standing or walking in place. The training methods perturb the balance of their participants from sit to stand or during standing or walking in place, which may not be as relevant in a natural setting as might be a perturbation while walking. Melzer and Oddsson [14] found improvement in voluntary stepping, and balance control, in an exercises program that incorporate mild external balance perturbation exercises applied by the instructors; and Halvarsson et al. [15] found that old fallers that suffered from fear of falling, decreased their fear of falling, and voluntary stepping times during dual-task performance and increased velocity of walking post perturbation training.. However, in these program the perturbations of posture were expected, and not random. Recently it was showed that unidirectional translational treadmill training (i.e., a laboratory-induced trip) reduced falls [16, 17]. Bhatt et al. [18] found that inducing unannounced right-leg slips, participants significantly reduced fall and balance loss incidence. Pai, et al. [19] found that a single session of repeated-slip exposure reduced older adults' annual risk of falls from 34 to 15 % ($p < 0.05$) especially among those who had history of falls. The above protocol provided an anterior perturbation, causing a backwards "slip" initiated always by on the right foot. Participants might have learned and expected the right-leg slips perturbations. In a recent meta-analysis [20] that include 8 perturbation-based balance training studies ($n = 404$) participants reported fewer falls than those in the control groups.

Motivated by the above perturbations training studies and trying to accommodate for some of the issues mentioned above (i.e., perturbation training while standing or walking in place; highly predictable repeated-right leg slip exposure; a very unnatural walking on split treadmill), we aimed to explore whether unexpected multidirectional perturbation training while walking on a treadmill [21] can reduce risks of falls in independent older adults. A perturbation exercise while walking provides a more realistic balance training that is sufficiently task-specific so that responses on this training regime will be more likely to be transferred to other measures of balance control and voluntary stepping measures.

Our hypotheses were that following exposure to a gait training program that includes unexpected perturbations exercises during walking will significantly improve voluntary stepping times as well as balance control in older adults, two factors that are associated with falls and injuries related with falls [22–27]. We believe that perturbation exercises that will challenge both balance control during walking as well as trigger a quick stepping responses to avoid fall during walking. These postural response following an external perturbation receives a higher priority than a voluntary action thus can be incorporated into centrally programmed voluntary movements [28]. This concept should be of importance for balance training and it further supports the notion that postural perturbations should be incorporated into balance training programs.

Methods

Participants

Community dwelling older adults were recruited from two protected housing institutes. Eligibility criteria were: 70 years or older; walking independently; Mini-Mental Score higher than 24; no severe focal muscle weakness or blindness; no known neurological disorders; no metastatic cancer. Out of 72 seniors who were assessed for eligibility, 19 were excluded (see Fig. 1). All subjects provided a medical waiver signed by their primary care physician clearing them to participate in moderate physical exercise. The study was approved by the Helsinki committee of Barzilai University medical center, Ashkelon, Israel (ClinicalTrials.gov Registration number #NCT01439451). All subjects signed an informed consent statement.

Study design

After eligibility and baseline assessments subjects were randomized to two blocks (27 and 26 subjects, respectively). In the first site, 28 subjects were randomly allocated to 2 intervention groups and in the second site 25 subjects were randomly allocated to 2 intervention groups. The subjects random allocation was made by an investigator not involved in the assessments using computer random allocation software (Random allocation software version 1.1, Isfahan Iran). Performance-based and laboratory balance functions were tested before and after the training

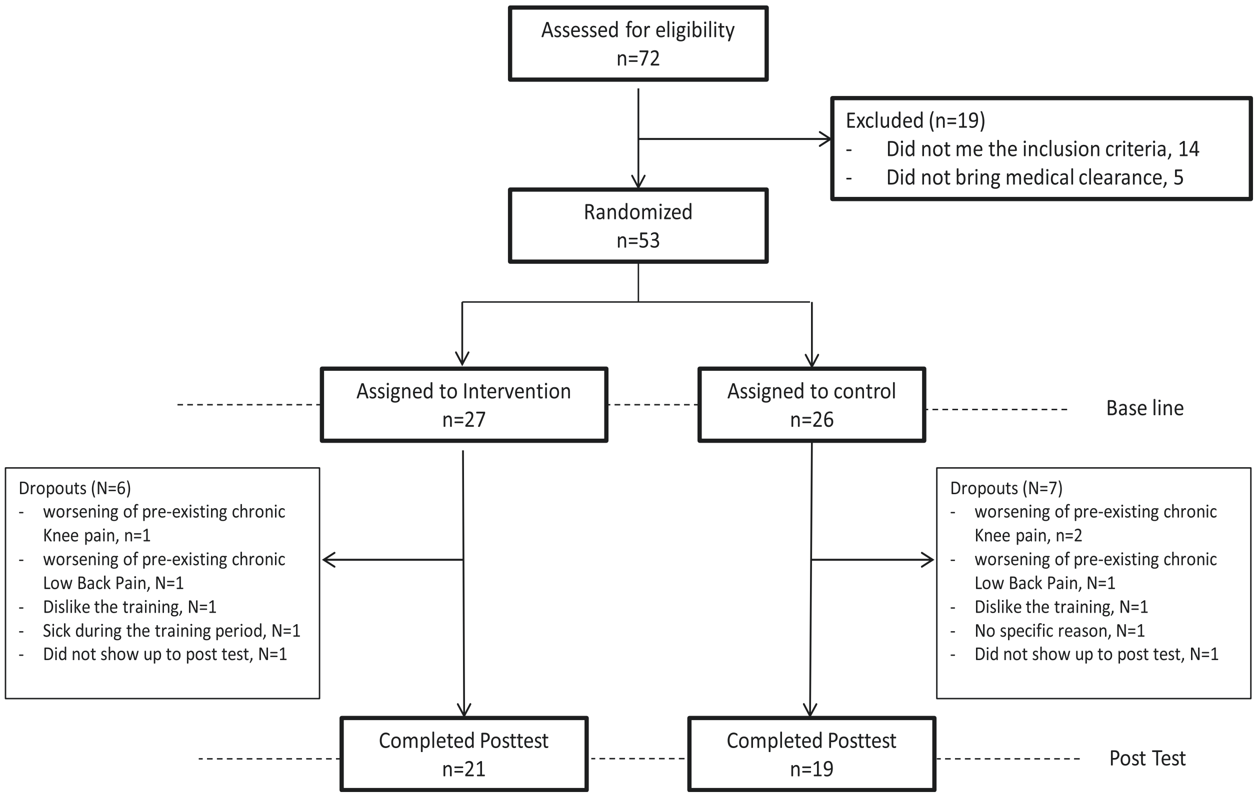

Fig. 1 Study flow chart

period by a blinded investigator. All assessment sessions were performed at the same time of day, and in the same order.

Training programs

We used a mechatronic device that provides controlled and unexpected anterior-posterior and Medio-lateral platform translations during a single belt, treadmill walking (details in reference [21] and Fig. 2a-c). The intervention group received 24 training sessions, twice a week for 12 weeks. Each session lasted for about 20 min and included 3 min warm-up walking in subject own preferred pace, 14 min of unannounced perturbations exercises, given in random direction order, during walking (every 20–40 s) and 3 min of cool down walking. During the training sessions the subjects were instructed to walk on a treadmill, wearing their own walking shoes, with their hands free to swing; there were no handrails on the treadmill. To prevent injury if loss of balance occurred during the treadmill walking, the subject wore a loose safety harness that could arrest the fall, but that allowed the subject to walk comfortably as well as freedom to execute recovery reactions without suspension (Fig. 2a). The instructions given to the subjects were: "Walk as naturally as possible at your preferred stride frequency". The treadmill's walking speed was adjusted to the subjects own preferred speed.

The perturbations were in anterior-posterior direction (i.e., sudden acceleration or stop of treadmills belt) and sudden medio-lateral horizontal translation of the treadmill that challenges the medio-lateral dynamic control. During all sessions, 400 ms horizontal surface translations were delivered as the subject walked on the treadmill. The velocity profiles were triangular waveforms with peak velocities of 0.1–3.2 m/s, resulting in displacements of 1–18 cm and peak accelerations of 0.5–16.0 m/s^2. Perturbation timing was preset and therefore was not given in a specific phase of the gait cycle or to a specific leg. The perturbation training program had 24 levels of difficulty with increasing levels of perturbations (i.e., increased displacement, velocity and accelerations of the horizontal translations, see Table 1). The difficulty level was adjusted according to the subject abilities, starting from the lowest level of 1 cm displacement at 0.1 m/s velocity and 0.5 m/s^2 acceleration at the first training session. If the subject was able to recover from all perturbations during the session (i.e., did not fell during the session) and felt that he can be further challenged, a higher level of perturbation was introduced in the next session. If not, the same level of perturbation was introduced again

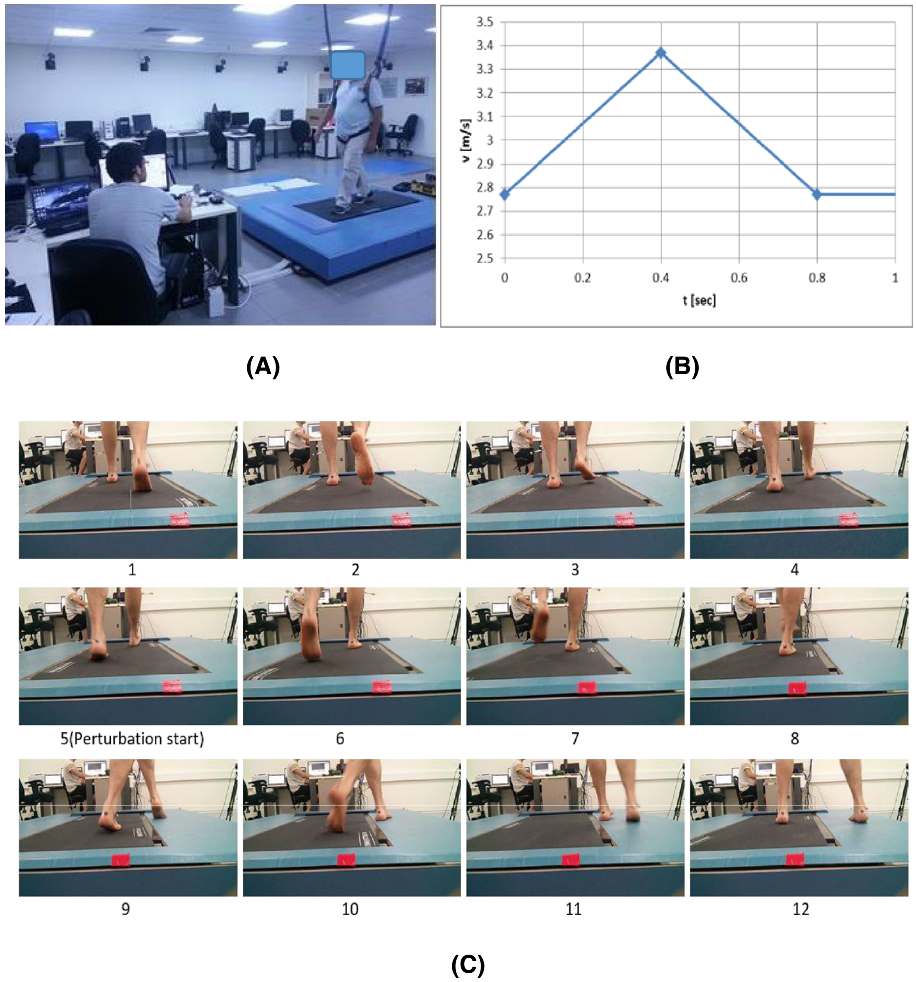

Fig. 2 The perturbation treadmill system used **a** Photo of the perturbation system during balance training. The system is compose of a motor-driven treadmill, mounted on a moving platform, motion controller, safety harness and an operator station; **b** the perturbations velocity control diagram during training delivered unpredictably in forward, backward, left, and right directions. Note those are actual measurements taken during perturbation training. **c** Example of the perturbation applied during the treadmill walking training (*c1–c12*). The perturbation applied unpredictably (*c5*) by horizontal movement of the platform towards the left side during the right foot initial contact-loading response phases of gait cycle. The participant right foot was slipped unpredictably to the left while walking in the center of the platform. The participants performed a cross over stepping response by his left foot (*c6–c9*), than additional side step was performed by the right foot stepping outside the treadmill (*c10–c12*)

until successfully dealt with. Fall during the training session was defined as load cell sensors detected 30 % or more body weight suspended by the safety harness.

The control group received 24 sessions, twice a week for 12 weeks, 20 min treadmill walking on the same treadmill but without unexpected perturbations. similar to the intervention group the control group subjects walked at their own preferred speed and in their own walking shoes, with their hands free to swing; there were no handrails on the treadmill, thus they wore a loose safety harnesses that allowed comfortable walking. Since both group trained on the same system, they were blinded to the allocation to intervention or control group.

Assessments

For the balance control testing the subjects were instructed to stand barefoot as still as possible and on a force platform in a standardized stance, their feet close together. Ten 30-s quiet-standing trials with eyes blindfolded. Center of pressure and ground reaction force data were collected with a Kistler 9287 force platform (Kistler Instrument Corp, Amherst, NY, USA), sampled at a frequency of 100 Hz. Evaluation of balance control was made using both traditional measure of postural sway in eyes closed condition (e.g. ML-sway, AP-sway, mean sway velocity, and Mean sway area), we also calculated the Stabilogram-Diffusion Analysis parameters from

Table 1 Details of Protocol Used in the perturbation intervention training program

The training sessions #	platform displacement (cm)	platform peak velocity (cm/s)	platform peak acceleration (cm/s^2)	Number of unannounced random Perturbations per minute (forward, backward, left, right)
1	1–2 cm	0.1–0.5 m/s	0.5–3.0 m/s^2	1
2	2–3 cm	0.2–0.6 m/s	0.7–5.0 m/s^2	2
3	3–4 cm	0.4–0.6 m/s	0.9–7.0 m/s^2	1
4	3–5 cm	0.5–0.7 m/s	1.1–7.0 m/s^2	2
5	4–6 cm	0.5–0.8 m/s	1.5–8.0 m/s^2	2
6	5–6 cm	0.5–1.0 m/s	2.0–10.0 m/s^2	2
7	5–7 cm	0.7–1.0 m/s	2.5–12.0 m/s^2	2
8	6–7 cm	0.8–1.2 m/s	3.0–14.0 m/s^2	3
9	6–8 cm	1.0–1.5 m/s	3.5–16.0 m/s^2	2
10	7–8 cm	1.2–1.8 m/s	4.0–16.0 m/s^2	3
11	7–9 cm	1.5–2.0 m/s	5.0–16.0 m/s^2	2
12	8–9 cm	1.6–2.2 m/s	6.0–16.0 m/s^2	3
13	8–10 cm	1.8–2.5 m/s	7.0–16.0 m/s^2	2
14	9–10 cm	2.0–2.6 m/s	8.0–16.0 m/s^2	3
15	9–11 cm	2.0–2.8 m/s	9.0–16.0 m/s^2	2
16	10–11 cm	2.2–3.0 m/s	10.0–16.0 m/s^2	3
17	11–14 cm	2.4–3.0 m/s	11.0–16.0 m/s^2	2
18	12–14 cm	2.5–3.0 m/s	12.0–16.0 m/s^2	3
19	13–15 cm	2.6–3.0 m/s	13.0–16.0 m/s^2	2
20	14–15 cm	2.6–3.2 m/s	14.0–16.0 m/s^2	3
21	14–16 cm	2.8–3.2 m/s	14.0–16.0 m/s^2	2
22	15–17 cm	2.8–3.2 m/s	14.0–16.0 m/s^2	3
23	16–18 cm	2.8–3.2 m/s	14.0–16.0 m/s^2	2
24	17–18 cm	2.8–3.2 m/s	14.0–16.0 m/s^2	3

The 24 training sessions. Each session lasted 20 min, included 3 min warm-up treadmill walking, 14 min of perturbations during comfortable treadmill walking, given in random direction (right, left, forward and backwards), and 3 min of cool down walking. The perturbation training program had 24 levels of difficulty with increasing levels of perturbations (i.e., increased displacement, velocity and accelerations of the horizontal translations). During each session, the listed platform translation unannounced perturbations were delivered in an unpredictable randomized sequence, in the directions indicated (forward, backward, left, and right). Perturbations for the treadmill walking were occur randomly (i.e., occurs in all phases of gait cycle) in order to increase the ecological validity. The perturbation was delivered after 20–30 s approximately every 20 strides and was triggered randomly

Center of pressure data. The Stabilogram-Diffusion Analysis plots (SDA) of the mean square center of pressure displacement (Critical Displacement, Cd) vs. time interval (Critical Time, Ct) parameters were extracted from the center of pressure trajectories. The SDA plots derived from COP trajectories during standing indicate the presence of two different behaviors depending on the time interval of interest. For shorter time intervals (less than 1 s) the COP tend to drift away from a relative equilibrium point while longer time intervals (more than 1 s) the COP tends to return to a relative equilibrium point [29]. It has been suggested that long-term region is governed by closed-loop control mechanisms whereas the postural control systems operate with sensory feedback, while during the short-term region the postural control system is governed by open-loop control mechanisms whereas the postural control systems operate without sensory feedback. The transition point between the short-term and long-term behavior has been termed the Critical Time (Ct) and sway displacement has been termed the Critical Displacement (Cd) at which closed-loop control begins to dominate sway behavior. It was described in detail by Collins and De Luca [29, 30]. The SDA method has been adopted by our research group, we found that SDA parameters (e.g., Critical Displacement (*Cd*), and Short-term Effective diffusion coefficients (*Ds*) were able to predict falls [26] and injury from fall [27].

For the Voluntary Step Execution Test participants were instructed to stand with both feet on a single force platform, they were instructed to voluntary step as quickly as possible following a somatosensory cue, given randomly on one of their feet [24, 31, 32]. Center of pressure movement and ground reaction force data were collected from the force platform, sampled at a frequency of 100 Hz. A total of

8 trials were conducted in single task condition, 4 forward and 4 backward as well as in dual task conditions. The average result across task condition was used for statistical analysis. For the single task, subjects viewed an "X" displayed on a screen in front of them. During the dual task they conducted the same test while performing the modified Stroop task [33, 34]. Specific temporal events were extracted from the step execution data: (a) Reaction Time; (b) Foot Contact Time; (c) Preparation Time; (d) Swing Time; as previously described in detail [24, 31, 32]. The foot contact times (i.e., stepping time) and reaction time duration especially in dual task condition were able to predict future fall [23] and injury from fall [22], thus both were selected as the primary outcome measure in the present study.

The secondary outcome measures were the self-reported function (Late Life Function and Disability Instrument (LLFDI)) [35], Fall Efficacy Scale (FES) [36] and performed the Performance-Oriented Mobility Assessment (POMA) [37].

Sample size

Sample size requirements were calculated based on AP postural sway in eyes closed condition and voluntary step execution times, both were found to predict injury from fall in older adults ([22, 27] respectively). For both calculations, the probability of type I error was 0.05, and probability of type II error was 0.2. Based on data presented by Kurz et al. [27] that found that the traditional AP postural sway in eyes closed condition was 42.3 mm in older adults who were injured as a result of falling compared with 32.6 mm in non-fallers older adults; and Melzer et al. [22] found that the step execution times (i.e., foot contact time) in dual task condition of older adults who fell and as a result injured were 217 ms longer than those of non-fallers (1,394 ms vs. 1,177 ms). Using net reduction values (9.7 mm and 217 ms, respectively) in combination with the initial variance estimates (standard deviations of 11 mm and 250 ms, respectively), it was determined that 21 and 22 participants per group would be required, respectively. To account for reported attrition rates of about 25 % in studies involving older adults [38], we decided to include about 27 participants in each group for a total of 54 ($22 \times 1.25 = 27$).

Data and statistical analysis

PASW Statistics version 18.0 was used for statistical calculations (Somers, NY, USA, version 18). Baseline characteristics were compared using Independent *t*-test and Mann–Whitney U-tests for continuous and ordinal variables, respectively. To analyze the effect of the intervention program a two-way repeated-measures ANOVA for within subjects (pre vs. post-test) and between group (perturbation intervention vs. control group) was performed. Since age was significantly different between intervention and control group subjects, we included age as a covariate in the analyses. The primary outcome variables were the parameters that we previously found to be related to falls and injury from fall: the step execution times in single and dual task conditions (i.e., foot contact-time) and step reaction time, traditional postural sway in eyes closed condition (ML- and AP- sway, sway velocity and mean sway area) as well as Stabilogram diffusion Analysis parameters (Critical Displacement (*Cd*), and Short-term Effective diffusion coefficients (*Ds*) in eyes closed condition. The secondary outcome measures were LLFDI, FES and POMA. An intention to treat analysis was conducted by carrying the last obtained measurements forward for those subjects who did not complete all aspects of the study. Adjustment of level of significance for multiple comparisons were made. For each testing procedure (e.g., single task voluntary stepping, dual task voluntary stepping, postural sway and SDA), a full Bonferroni correction was used to achieve an overall significance level of 0.05.

For the significant improvement the Effect Size (ES) of Hedge's g was calculated. The ES of g was calculated by taking the difference between the means of both groups divided by the average population standard deviation (SD). To estimate the SD for g, baseline estimated SDs of both groups was pooled. When interpreting correlation magnitudes: 0.0–0.2 is considered small, 0.2–0.5 is considered moderate and 0.5–0.8 is considered large [39].

Results

Other than age and height there were no significant differences between the groups at baseline (Table 2). During the training period we had 6 drop outs in the intervention group and 7 in the control group (Fig. 1).

Table 3 show that the perturbation training resulted in a significant group-by-time decrease in AP-sway ($p = 0.012$, [ES] = 0.59). There were also a trend towards a group-by-time decrease with a large effect size in sway velocity, sway area and ML-sway post-training ($p = 0.115$, [ES] =0.72; $p = 0.119$, [ES] =0.74; and $p = 0.142$, [ES] = 0.58, respectively); although these differences were not statistically significant, the effect size is considered to be moderate. The Stabilogram-Diffusion analysis in eyes closed condition showed a significant group-by-time decrease in Dxs, ($p = 0.012$, [ES] =0.78) and a trend towards significance in Cdy and Dys ($p = 0.028$, [ES] = 0.92; and $p = 0.095$, [ES] =0.65, respectively) (Table 4).

Table 5 shows that the perturbation training resulted a significant group-by-time interaction for foot contact time of the voluntary step execution in both the single and dual task conditions ($p = 0.002$, effect size [ES] = 0.75 and $p = 0.003$ effect size [ES] = 0.89 respectively). In addition, a significant group-by-time

Table 2 Baseline characteristics of intervention and reference group subjects: descriptive statistics and group comparisons. Values are means ±SD (95 % confidence interval for means)

	Intervention Group ($N = 27$)	Control Group ($N = 26$)	*p*-value
Age (year)	78.2 ± 5.6	81.4 ± 4.3	0.05
% Female	62 %	79 %	0.25
Number drugs/day	3.3 ± 1.7	4.2 ± 2.4	0.18
Height (cm)	161.5 ± 10.9	154.9 ± 6.9	0.03
Weight (Kg)	70.9 ± 14.9	65.5 ± 12.9	0.23
BMI (Kg /m^2)	27.1 ± 4.4	27.8 ± 5.1	0.64
Mini-Mental State Examination	29.1 ± 1.4	28.7 ± 1.4	0.49
Fall efficacy scale	20.5 ± 4.3	22.8 ± 10.3	0.36
POMA score	14.8 ± 1.3	14.7 ± 1.4	0.87
Late life function			
- Overall Function	66.8 ± 9.6	66.5 ± 6.7	0.90
- Upper Extremity Function	82.9 ± 11.7	79.2 ± 8.0	0.26
- Basic Lower Extremity Function	82.4 ± 12.4	81.3 ± 13.6	0.79
- Advanced Lower Extremity Function	59.6 ± 12.4	61.2 ± 10.4	0.66

Note: *p*-value compares baselines means in the two groups and, unless otherwise indicated, are based on *t*-test or chi-square. * *P*-value based on Wilcoxon signed rank test and Mann–Whitney *U* test. Abbreviations: *cm* centimeters, *Kg* Kilograms, *Kg/m^2* kg per meter squared

interaction for the voluntary step reaction time in dual task condition ($p = 0.010$), and a trend towards a significant group-by-time interaction for single task condition ($p = 0.057$).

We found no significant group-by-time interaction effect for all components of the LLFDI as well as for FES and POMA.

With respect to side effects and adverse events, during the exercise training program, muscle soreness was experienced by some subjects, especially in the early stage of the training. Those effects were managed by adjusting the training intensity and the symptoms disappeared during training.

Discussion

The results support in part our main hypotheses, unexpected perturbations training while walking can improve the ability to voluntarily step rapidly and standing balance control in older adults. These parameters have been shown in the past to predict injury from falls [19, 24]. This shows that the benefits of unexpected perturbation during walking were generalized to other aspects of balance. Our findings are supported by Pai and Bahtt [40] that suggested that the central nervous system makes adaptive improvements in proactive and reactive control of stability as a result of trial and error perturbation practice (i.e., a forward slip). They suggested that the central nervous system

Table 3 The effect of balance training on traditional sway Parameters under eyes closed condition. Values are means±1 SD (95 % confidence interval for means). A full Bonferroni correction (α-level 0.05/4 = 0.0125) was used for each of the four tests to achieve an overall significance level of 0.05

	Group	Baseline	post-test	ANOVA (Baseline to post-test) T	ANOVA (Baseline to post-test) T x G
ML-sway (mm)	Experimental	47.9 ± 15.3	44.6 ± 15.5	F = 0.215	F = 2.247
	Control	41.4 ± 8.8	40.5 ± 8.6	$p = 0.792$	$p = 0.142$
AP-sway (mm)	Experimental	38.6 ± 13	35.8 ± 10	F = 0.694	F = 5.315
	Control	33.8 ± 8	33.9 ± 7	$p = 0.\ 711$	$p = 0.012$
Velocity(mm^2/sec)	Experimental	37.7 ± 12	34.2 ± 10	F = 0.086	F = 2.609
	Control	35.1 ± 8	35.0 ± 9	$p = 0.770$	$p = 0.115$
Sway Area (mm^2)	Experimental	169.3 ± 9	148 ± 76	F = 0.056	F = 2.549
	Control	136.6 ± 48	135 ± 5	$p = 0.815$	$p = 0.119$

Note: Comparison of baseline and post-intervention between the two groups based on repeated measures ANOVA (Test × Group). Abbreviations: *G* group, *T* time, *mm* millimeters, *s* seconds

Table 4 The effect of balance training on Stabilogram Diffusion Parameters in eyes closed condition. Values are means±1 SD (95 % confidence interval for means). A full Bonferroni correction (α-level 0.05/4 = 0.0125) was used for each of the four tests during the two phases to achieve an overall significance level of 0.05

	Group	Baseline	post-test	ANOVA (Baseline to post-test) T	ANOVA (Baseline to post-test) T x G
Short-term Effective diffusion coefficients in $mm^2\ s^{-1}$ (*Dxs*)	Experimental	97.9 ± 71.3	84.7 ± 58.6	F = 0.856	F = 5.822
	Control	79.18 ± 38.2	87.4 ± 48.8	$p = 0.391$	$p = 0.012$
Short-term Effective diffusion coefficients in $mm^2\ s^{-1}$ (*Dys*)	Experimental	62.9 ± 54.8	47.3 ± 34.6	F = 0.009	F = 2.928
	Control	37.2 ± 29.4	38.4 ± 27.5	$p = 0.925$	$p = 0.095$
Critical (Mean-Squared) Displacement in mm^2 (*Cdx*)	Experimental	131.2 ± 105	112 ± 82.7	F = 0.038	F = 1.916
	Control	98.9 ± 48.7	96.6 ± 42.7	$p = 0.847$	$p = 0.157$
Critical (Mean-Squared) Displacement in mm^2 (*Cdy*)	Experimental	90.3 ± 72	69.0 ± 52.7	F = 1.360	F = 5.266
	Control	62.7 ± 30	65.9 ± 30.4	$p = 0.251$	$p = 0.028$

Note: Comparison of baseline and post-intervention between the two groups based on repeated measures ANOVA (Test × Group). Abbreviations: *G* group, *T* time, *mm* millimeters, *sec* seconds

probably decreases the reliance on feedback corrective mechanisms for successful recovery and builds an internal representations to improve its feedforward control while walking. Pai and Bahtt training [40] used forward slip practice, the training program, while in the present study the direction of the perturbation was highly unpredictable (forward, backward, right and left perturbations), the intervention group subjects were unable to predict the direction of perturbation during the training. Thus it is unlikely that the improvement was due to feedforward control. Our results support the notion that the central nervous system probably increased the reliance on feedback corrective mechanisms for successful recovery. This is supported by the results of the SDA.

The significant improvement in the intervention group and the large effect size in the SDF parameters (ES = 0.65–0.92) in eyes closed conditions are promising. Age-related decrease in SDF parameters are well documented [30, 33] and indicates that a COP tends to drift away from the equilibrium point is a predictor to falls in elderly persons [26, 27, 29, 30, 41–43]. The improvement in the *Dxs* and *Cdy* as well as tendency towards improvement in *Dys* of SDA in eyes closed condition showed in this study indicates that the deterioration of balance control could be reversed by perturbation balance training. AP balance control in eyes close condition (*Dys* and *Cdy*) were found to be an important risk indicator of falls and injurious falls [26, 27]. Laughton et al. [41] found a greater *Dys* in elderly fallers compared with the young's, and greater AP and ML sway in older adults who demonstrated lower scores in the POMA. Kurz et al. [27] found that a deterioration of AP postural control present higher risk of serious injury. Most Studies usually used traditional balance measures, which provide descriptive information on the postural sway, however it is very difficult to understand the mechanism of postural

Table 5 Voluntary Step Execution Test times and the preparation phase times during single task and dual task conditions (mean ± SD). Values are means ± SD (95 % confidence interval for means). A full Bonferroni correction (α-level 0.05/2 = 0.025) was used for each of the two different task conditions (single and dual task condition) to achieve an overall significance level of 0.05

	Group	Baseline	post-test	ANOVA (Baseline to post-test) T	ANOVA (Baseline to post-test) T x G
Single task condition					
Reaction Time (ms)	Intervention	215 ± 40	194 ± 36	F = 0.002	F = 2.187
	control	219 ± 70	206 ± 57	$p = 0.968$	$p = 0.057$
Foot Contact Time (ms)	Intervention	1065 ± 16	993 ± 138	F = 0.474	F = 11.325
	control	1027 ± 147	1010 ± 143	$p = 0.495$	$p = 0.002$
Dual Task condition					
Reaction Time (ms)	Intervention	412 ± 174	346 ± 99	F = 1.881	F = 7.322
	control	354 ± 98	339 ± 100	$p = 0.179$	$p = 0.010$
Foot Contact Time (ms)	Intervention	1355 ± 243	1224 ± 172	F = 0.439	F = 9.857
	control	1250 ± 165	1240 ± 171	$p = 0.512$	$p = 0.003$

Note: Comparison of baseline and post-intervention between the two groups based on repeated measures ANOVA (Test × Group). Abbreviations: *G* group, *T* time, *ms* milliseconds control using traditional COP statistics, this we performed SDA. We found training effects on open- and close-loop control mechanism when vision was occluded as indicted by a significant decrease in Critical Displacement in AP direction (*Cdy*). . Improvements in and Short-term Effective diffusion coefficients in ML direction (*Dxs*) as well as a trend towards a decreased in Short-term Effective diffusion coefficients in AP direction (*Dys*) further support this notion. These parameters reflect the degree of sway in the short term region. A decreased tendency to continue sway in an ongoing direction (Table 4) indicates a behavior that reflects a more stable balance control system. This indicates that the intervention group was able to better detect COP movement under their feet and initiate a more effectively close loop balance control. These suggest that the improved closed-loop balance control when vision was occluded would likely be from proprioceptive and/or maybe vestibular sources.

The improvement in step execution in the intervention group was seen during the step execution times i.e., foot contact times, in both ST and DT (72 ms in ST and 131 ms in DT, see Table 5). A shorter step execution indicates improvement in the ability to prevent a fall if balance is lost, consequently, the risk of fall and injury maybe reduced [22–24]. The improvement in the step execution in the intervention group was accompanied with improvement in step reaction times, in particular under dual task conditions. The step reaction time duration is mainly dependent on sensory detection thresholds, nerve conduction velocities and central neural processing times. A shortened step reaction time in dual task condition suggests that the central neural processing time was improved as a result of perturbation training. This indicate that the executive functioning was improved, subject were able to step quickly while their attention was allocated elsewhere. The executive functioning is related to the ability to rapidly shift attention from a cognitive task to the stepping task. This could be interpreted also in terms of automaticity of the stepping behavior as an essential characteristic of the central reorganization process. Therefore, we could assume that the interference effect during dual task stepping was reduced as a result of the treadmill perturbation training.

Halvarsson et al. [15] also found significant improvements in stepping performance, in a group of elderly fallers that performed perturbation exercises. Melzer and Oddsson [14], Mansfield et al. [13] and Rogers et al. [44] also found that specific step training that includes perturbation of balance improves stepping abilities in older adults. Pai et al. [12] have shown reductions in falls and balance loss following a repeated-slip during walking exposure. The same group had further showed that those gains could be retained for 6 months [18] and cut older adults' annual risk of falls by 50 % especially among elderlies who had history of falls [19]. Our results and the results above suggests that specificity-of-training principle is a major factor in achieving treatment goals and therapists need to tailor balance perturbation training programs to target functional aspects of balance control such as the ability to step rapidly [45].

It is still unknown how the improvement seen in balance and stepping carries over to real-life falls. It may be that carryover to fear of falls and physical performance as measure with POMA did exist but was not detected by these outcomes measures due to insufficient sample size or due to ceiling effects, subjects were close to score the highest score in both measures. We also did not find significant carryover effects on self-reported and performance based function. This suggests that function is not derived only by the ability to perform balance tasks, but is influenced by environmental and behavioral factors. Most perturbation training programs did not measure self-reported physical function [16–19, 40, 44]. However other studies found improvement in self-reported function [14] and fear of falls [15]. This two studies trained balance in a group setting adding also behavioral factors. This may suggest that physical intervention per-se without behavioral intervention cannot change the levels of physical function. The fact that Performance based measure are weakly associated with self-reported mobility in healthy elderly persons ($r = 0.21–0.29$) [46] support this notion.

This study has several limitations. First, the data came from a fairly small sample of independent older adults thus the results cannot be generalized to frail or institutionalized elderly persons. Second, the training program was done on a treadmill and not on leveled ground. Treadmill walking is somewhat different then over ground walking and might pose different demands on the trainee. However, it has been showed that fall-resisting skills acquired from such training can be transferred to over-ground walking [47]. Third, to provide stronger evidence of the clinical efficacy of this training procedure, future studies should compare the current intervention with an alternative treatment such as Tai Chi, or strength training as this was found beneficial in a recent review [48]. Also there is a need to determine whether this type of training improves the ability to cope with falls that resulted from slips and trips only, or can be transferred to any kind of loss of balance. Fourth, further research is needed in order to determine whether randomly ordered perturbations exercises (Random practice) would render better results than repeating the same perturbation exercises (blocked practice) in terms of acquisition, adaptation and retention.

Conclusion

Participation in a balance training program that includes unexpected perturbation of balance during treadmill walking reduced the risk factors for falls as presented by biomechanical markers that have been shown in the past to identify fallers and predict falls. However levels of physical functioning were not improved, this suggests that preventing sedentary lifestyle for the elderly require additional behavioral intervention.

Competing interests
IM and AS have developed and built the BaMPer perturbation system that was used in this project. All other authors declare that they have no competing interests.

Authors' contributions
IK was involved in planning the experiments and conducting the intervention as well as data analysis and interpretation and drafting of the manuscript. YG was involved in conducting the Pre and Post intervention tests as well as data interpretation and drafting of the manuscript. AS was involved in experimental design, data interpretation and drafting of the manuscript. RD was involved in Subject recruitment, medical screening as well as drafting and revising of the manuscript. YS was involved in Subject recruitment, medical screening as well as drafting and revising of the manuscript. IM was involved in planning the experiments and conducting the Pre and Post intervention tests as well as data analysis and interpretation and drafting of the manuscript. All authors read and approved the final manuscript.

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4

Effect of balance training on recovery of stability in children with cerebral palsy

Effect of balance training on recovery of stability in children with cerebral palsy

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This study examined the effect of massed practice in balance recovery of stability in six children (four males, two females; mean age 9 years 2 months, SD 2 years, range 7 years 5 months to 12 years 11 months) with cerebral palsy (CP). Four children were diagnosed with spastic diplegia (Gross Motor Function Classification System [GMFCS] level II) and two with spastic hemiplegia (GMFCS level I). A single-subject, multiple-baseline experimental design involving three pairs of children matched for diagnosis was used. A moveable forceplate system was used to test and train reactive balance control. Area per second (i.e. area covered by the center of pressure over a one second period) and time to stabilization from center of pressure measures were calculated following perturbations. The intervention phase consisted of massed practice on the moving platform (100 perturbations/day for 5 days). Analysis included hierarchical linear modeling and a repeated measures ANOVA. All children demonstrated a significant improvement in their ability to recover stability as demonstrated by reduced center of pressure area and time to stabilization following training. These improvements were still present 30 days following completion of training. Results suggest that postural control mechanisms in school-age children (7 to 13 years) with CP are modifiable.

See page 602 for list of abbreviations.

Cerebral palsy (CP) is a developmental disability characterized by delayed motor milestones and impaired motor control. It is one of the most common movement disorders in infancy, occurring in 2 of every 1000 children in the USA (Paneth and Kiely 1984, Gage 1991). Neuromuscular deficits noted in CP include the loss of selective motor control, abnormal muscle tone leading to an imbalance between agonist and antagonists muscles, impaired coordination, sensory deficits, and weakness (Gage 1991). A major factor contributing to functional deficits is poor postural control.

Impaired postural control limits a child's ability to recover from unexpected threats to stability, referred to as reactive balance control. Impaired postural control in children with CP has been shown to result from multiple factors: musculoskeletal problems, including contractures, reduced range of motion, and shifts in initial alignment, all affect reactive balance control in children with CP. Other motor components include the disruption of the spatial and temporal aspects of postural muscle responses during the recovery of stability following an unexpected external perturbation (Nashner et al. 1983, Burtner et al. 1998). The onset of postural muscle activity in children with CP is delayed compared with typically developing children. In addition, the sequencing of multiple muscle action is impaired and there is a high level of coactivation of agonist and antagonist muscles at a joint (Nashner et al. 1983, Burtner et al 1998). Difficulty in organizing redundant sensory cues for posture control is another source of instability in children with CP (Nashner et al. 1983, Cherng et al. 1999).

Finally, researchers have shown a reduced ability in children with CP to adapt the sensory and motor components of postural control to changing task and environmental demands (Nashner et al. 1983, Burtner et al. 1998). These postural control impairments affect the ability of children with CP to respond to threats to balance efficiently and effectively .

The relation between impaired balance and functional limitations has meant that a major focus of many intervention programs in children with CP is to improve postural control, thereby enabling the child to recover stability more effectively (Pape et al. 1993, Hur 1995, Butler 1998, Park et al. 2001). However, there is limited evidence available to show that specific training targeting the postural control system improves the efficiency and effectiveness of reactive balance control in this population.

There is evidence to suggest that in typically-developing children massed practice on a moveable platform improves the ability to recover stability in response to external threats. This ability has been shown to be important in the development of postural responses in children just beginning to stand. Sveistrup and Woollacott (1997) examined the effects of 3 days of intense reactive balance training on the ability of typically-developing children to activate three groups of muscles in the leg and trunk that work together to restore balance in response to platform perturbations. Children in the training group were given extensive reactive balance training on the moveable platform consisting of 100 perturbations per day over a period of 3 days. Results indicated a significant increase in the probability of activating the three-muscle postural response synergy; and the response was better organized following training.

Research by Sveistrup and Woollacott (1997) provides quantitative evidence for the effect of massed practice in reactive balance control on the orgzanization and efficiency of postural responses to balance threats in children who are typically-developing. However, the applications of these findings to children with CP is unknown. Thus the purpose of this research was to examine the effect of massed practice using a moving platform on recovery of stability in school-age children with CP. We hypothesized that the amount of time required to recover stability following a balance threat and the mean area of center of pressure movement per second during balance recovery would be reduced in these children as a result of training. We also hypothesized that these improvements in reactive balance control would be maintained at one month following the end of training and that the type of CP would predict the level of performance at baseline and level of improvement. Lastly we proposed that the children with spastic hemiplegia would have high baseline performances in the dependent measures and would show greater improvements following training than the children with spastic diplegia because the intact side would facilitate motor learning.

Method

PARTICIPANTS

Six children with CP (four males, two females; mean age 9 years 2 months, SD 2 years, range 7 years 5 months to 12 years 11 months) were recruited from local schools and children's therapy programs in the Seattle area of Washington State, USA. Four children had a diagnosis of spastic diplegia and two were diagnosed with spastic hemiplegia. Inclusion criteria were: (1) to stand independently for at least 30 seconds; (2) to walk independently with or without an assistive device; (3) no uncorrected vision or hearing impairments; and (4) to understand experimental procedures and give informed assent.

The four children with spastic diplegia were rated as level II on the Gross Motor Function Classification System for children with cerebral palsy (GMFCS; Palisano et al. 1997). They were able to walk independently in and outdoors. They experienced limitations in walking on uneven or inclined terrain and in distracting environments and had limitations in their ability to run or jump. The two children with spastic hemiplegia were rated GMFCS-CP level I. While they experienced some limitation in speed and coordination of movement and balance their gross motor abilities were above those children with diplegia.

Table I provides a summary of demographic information for all six participants including sex, gestational age, age at testing, type and severity of CP, and range of motion limitations in the lower extremity. Also shown is a list of the previous medical procedures and current schedule of therapies (both provided by parental report).

Approval for this study was obtained from the University of Washington Human Subjects Review Committee. Written informed consent was obtained from the parent(s) of each participant. Informed assent was obtained from the children.

INSTRUMENTATION

A moveable forceplate system (NeuroCom® International, Inc, Oregon, USA) was used to test and train reactive balance control. The system consists of two platforms that were embedded in a larger stationary platform. Each plate was controlled by an electric motor and forceplates were coupled together. Each plate contained four strain gauges, one in each corner of the plate surface to measure vertical (Fz), horizontal (Fx), and medial lateral (Fy) ground reaction forces on each plate. Analog force platform data were collected using a data acquisition system custom designed by a team member (RP). Measurement of sway was performed on unfiltered force platform data collected at a sample rate of 100Hz over a period of 10 seconds. Using additional software (custom created by RP) the area of sway per second was calculated by summing the incremental area enclosed by the points bounded by the centroid of the center of pressure distribution (Eq 1) and two sequential instantaneous center of pressure points (Eq 2 below):

$$\text{Eq 1. Centroid Location} \quad X = 1/n \sum_{i=1}^{n} x_i \quad Y = 1/n \sum_{i=1}^{n} y_i$$

$$\text{Eq 2. Area/sec (relative to X and Y)} \quad A' = f/2n \sum_{i=1}^{n-1} |x_{i+1}y_i - x_iy_{i+1}|$$

Table I: Participant demographics

	Sex M/F	*GA (wk)*	*Age (y:mo)*	*Diagnosis*	*GMFCS*	*ROM limitations procedures*	*Prior medical intervention*	*Current therapy*
Pair 1								
Dip1	M	28	12:11	Spastic diplegia	Level II	Plantarflexors, hamstrings bilaterally	Serial casting	Private PT: 1h/wk
Dip2	M	26	7:5	Spastic diplegia	Level II	Hamstring tightness L>R	Baclofen, botox, serial casting, AFOs	Private PT: 1h/wk School OT: 30min 2×wk
Pair 2								
Dip3	F	25	9:8	Spastic diplegia	Level II	Plantarflexors, hamstrings bilaterally	Bilateral shunts and revisions	Private PT, OT, SLP 30min/wk each. School OT: 45min/wk
Dip4	M	28	7:8	Spastic diplegia	Level II	Left plantarflexors	Botox, serial casting, AFOs	School OT: 45min/wk
Pair 3								
Hem1	F	41	7:10	Right hemiplegia	Level I	Right plantarflexors	Right tendon transfer, heel cord, lengthening, botox	School OT: 45min/wk
Hem2	M	39	10:4	Left hemiplegia	Level I	Left plantarflexors	Serial casting, AFOs	No therapy

GA, gestational age; GMFCS, Gross Motor Function Classification System (Palisano et al. 1997); ROM, range of motion; PT, physical therapy; Dip, spastic diplegia; Hem, spastic hemiplegia; AFO, ankle–foot orthosis; OT, occupational therapy; SLP, speech language pathology.

A perturbation to balance was given to the standing participant by a horizontal forward or backward translation to the forceplates (3cm amplitude, 12cm/s velocity). Custom software was designed to calculate time to stabilization and center of force (area per second). Time to stabilization was determined by visually examining the movement of the center of pressure at 250ms increments. Time to stabilization was defined as a constant position of the center of pressure within a 5×5mm square for a period of 500ms.

MEASUREMENT

Laboratory measures

Laboratory measures of reactive balance control included a calculation of area and time to stabilization from center of pressure measures following a perturbation on the force platform. Throughout the study, force platform data (center of pressure area and time to stabilization) were used to collect 'probe data' (data collected on a regular basis in order to examine change over time) which established repeated measures of the dependent variable across time (Horner and Baer 1978). Probe sessions were conducted before training (baseline probes), before daily training sessions (intervention probes), and following training (posttest probes). A probe session consisted of measurement of the dependent measures (center of pressure area and time to stabilization) during five forward and five backward perturbations (3cm, 12cm/s).

During testing and training sessions participants stood barefoot on the balance platform, were secured to an overhead harness, and were guarded by an observer for safety. Children were instructed to try and remain standing in one place without reaching for the grab bar or the observer during the platform movements.

Clinical measure

Dimension D (Standing) of the Gross Motor Function Measure (GMFM; Russell et al. 1993, 2000) was used as a clinical measure of motor function. Scoring for the GMFM is based on a 4-point Likert scale. The dimensions each have equal weight of 20% and a subscore for each dimension can also be calculated. The total points possible for dimension D of the GMFM is 39.

PROCEDURE

A single-subject, replicated-multiple-baseline experimental design was used with three pairs of diagnosis-matched children. With this design each child was measured repeatedly within three phases of the design: baseline, intervention, and maintenance. Data were analyzed for changes in trend, level, and variability to determine if a relation existed between the intervention (massed practice) and the measured behavior (recovery of stability as measured by center of pressure measures including area/s [mm] and time to stabilization [s]). In this design, each child served as their own control (Horner and Baer 1978). By matching the children according to similar age and diagnosis and then staggering the training phase of the experiment, internal validity was strengthened (Kazdin 1982).

Before training, children participated in two or three sessions in order to collect baseline data related to the dependent measures. Two children matched by similar age and diagnosis began the study simultaneously. In the first week of the experiment, the first child underwent 2 days of baseline assessment and then in the second week began 5 days of training. The second child in the matched pair participated in a delayed intervention protocol. This included two baseline assessments during the first week and a third baseline assessment on the day in which the first child concluded training. Child 2 then began the 5 days of training. Thus the start of the intervention in the two pairs of children was staggered in time. This resulted in a longer baseline for the second child, documenting that the dependent measures were not changing over time and with repeated testing. Changes in dependent measures when the intervention was introduced could then be attributed to the intervention rather than extraneous results, such as repeated testing or development. Both children received posttest evaluations (laboratory and clinical) immediately following training and then again 30 days following training in order to examine the retention of changes. This design was then replicated with two more pairs of children.

The intervention phase consisted of 5 consecutive days of platform training. The children were exposed to 100 perturbations at approximately four to six perturbations per minute. The perturbations consisted of forward and backward translations (3 to 6cm) at velocities that ranged randomly between 12 and 24cm/s. The children stood on the platform wearing a safety harness and watched an age-appropriate videotape during the training. Rest breaks were taken approximately every 20 to 25 perturbations. The training sessions were run by one of the coinvestigators (SH).

ANALYSIS

Visual analysis

Probe data points for both sway area per second and time to stabilization were recorded on a graph. Means (SDs) were plotted for each session. Using Kazdin (1982) and Franklin et al. (1996) criteria for visual inspection, these data were inspected for dramatic intervention effects by examining each phase for changes in magnitude and in the rate of change.

Changes in magnitude are seen as changes in the mean and in the level of graphed data points across phases. A change in the mean per phase is indicative of 'shifts in average rate of performance'. A change in level appears as a shift of performance from the end of one phase to the beginning of the next (Kazdin 1982). The latency of change refers to the period between the onset or termination of one condition and changes in performance (Kazdin 1982). In addition, the variability of probe data points, both within each phase and between phases, the stability of the baselines, and the overlap of probe data points between phases were analyzed through visual inspection.

Percentage change in mean probe data from one phase to another was calculated. Overall improvement in performance was determined by calculating the percentage change in dependent measures from the baseline phase to the first posttest assessment. Initial effects of the intervention were determined by comparing the percentage change in the dependent measures from baseline to the second day of training. The percentage of change from training day 1 to training day 5 was used as an indicator of a within training effect. Finally, retention was determined by calculating the percentage change from the immediate posttest to the 1 month posttest assessment.

Statistical analysis
Statistical analysis of single-subject data was used as an adjunct to the visual analysis. Two types of analyses that are complimentary to each other and appropriate for the analysis of longitudinal probe data were applied.

First, hierarchical linear modeling was used to analyze individual trajectories of change for both the center of pressure area and time to stabilization data at each measurement occasion over the time spanning the three phases of the design (baseline, intervention, and maintenance). Separate hierarchical linear modeling analyses were performed for each of our dependent variables (center of pressure area and time to stabilization). For the current study, hierarchical linear modeling allowed us to predict how values for our quantitative dependent variables might change over time and phase as a consequence of balance training. In addition, it allowed us to associate that change with variance in multiple factors measured on two levels simultaneously. On the first level the single within-subjects factor time (or phase of the study) modeled change in scores for the dependent variables much as a repeated measures analysis of variance (ANOVA) or linear regression would. On the second level, unlike an ANOVA or regression analysis, hierarchical linear modeling allowed us to consider simultaneously the influence of between-subjects factors (i.e. number of days in baseline conditions, age, and severity of motor problems) to model variability in the slope and intercepts of the first level, i.e. within-subjects factor. These analyses were performed with hierarchical linear modeling for Windows (version 4.01.01).

Second, two repeated measures ANOVAs (one for each dependent variable) were used to confirm the results from the first level analysis of the hierarchical linear modeling procedure as well as to estimate the effect size associated with the factor 'time' or phase of the study. This ANOVA was performed with SPSS for Windows, (version 11.0). The alpha level for the determination of statistical significance was $p<0.05$.

Results
This research examined changes in center of pressure area and time to stabilization before, during, and following 5 days of intensive balance training on a moveable platform in six school-aged children with CP in order to investigate the effects of massed practice on the ability to recover from an external threat to standing balance.

The first issue to be addressed was whether a functional relation exists between reactive balance training and improvements in balance. In a multiple baseline design the effects of the intervention are examined and then reproduced by subsequent participants. The experimental criterion for visual analysis is met by demonstrating a minimum of three participants' shift in performance at three points in time. Data presented below reveal shifts in balance performance in all six children participating from baseline to intervention and maintenance.

CHANGES IN CENTER OF PRESSURE AREA
Figure 1 plots the center of pressure at (a) baseline, (b) training, and (c) posttest session in one of the children with spastic diplegia. Figure 2 plots the center of pressure area in all six children for each session in the three phases of the study. Data points (filled diamonds) indicate the mean area for each session in the baseline, intervention, and posttest phases. Vertical lines indicate variability (SD) within the session. Figures 2a and 2b represent data from the pairs of children with spastic diplegia; Figure 2c represents data from the pair of children with spastic hemiplegia.

For simplicity, the data are also shown collapsed across phases for each participant in Figure 3. Data shown were calculated by taking the mean of the area/s measures for each session and then calculating the mean area/s score for the entire phase. Thus each child has three points representing mean area/s during baseline, intervention, and posttest phases associated with recovery of stability following a forward sway perturbation (Fig. 3a) and a backward sway perturbation (Fig. 3b).

Baseline phase
Examination of the baseline data from Figures 2a, 2b, and 2c illustrate the center of pressure area associated with recovery from a forward sway perturbation before training. The relative stability of the area measures over the two (three in the delayed start children) baseline sessions indicates the stability of this measure before the training. Figures 3a and 3b illustrate the

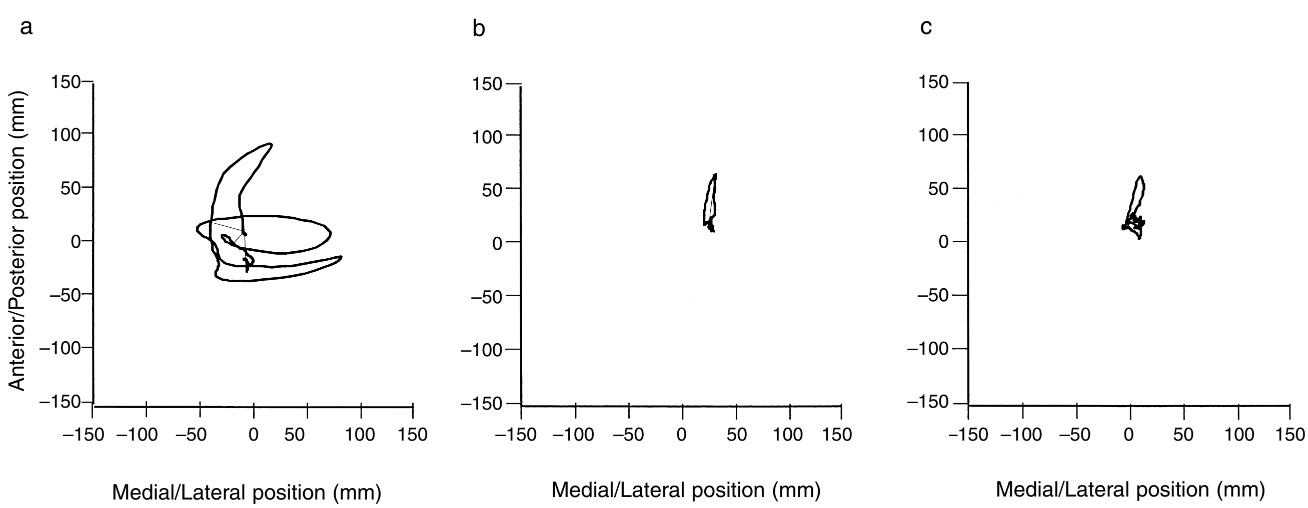


Figure 1: *A plot of center of pressure at (a) baseline (b) training, and (c) posttest session in one child with spastic diplegia.*

variability of performance among the six children before training. Thus the mean baseline measure varied from a high of 1250mm for Dip4 (child 4 with spastic diplegia) to a low of 705mm in Dip2 (child 2 with spastic diplegia), with a group mean of 936mm SD 245mm).

Intensive massed practice – intervention phase

Changes in center of pressure area associated with the intensive massed practice on the moving platform can be visually analyzed in relationship to the overall magnitude of effect, the initial effect, and the within-training effect. The magnitude of the effect of training on center of pressure area can be seen in Figures 2 and 3. The area/s measure is much higher in the baseline sessions compared with the intervention sessions. All children had a substantially reduced center of pressure area associated with recovery from an external perturbation.

Initial, within, and total training effects, expressed as percent change in center of pressure for each child are presented in Table II. An initial effect of training was determined by calculating the percentage reduction in center of pressure area from baseline to training day 2. All children showed an initial effect although the magnitude of the effect varied greatly among the children. The average improvement was a reduction of 37% (SD 29%) and the range for initial effect decreased from 8 to 83%. The greatest initial effect was seen in the two children with spastic hemiplegia.

A within-training effect was determined by calculating the percentage reduction in center of pressure area from training day 1 to day 5. Mean within-training effect was a decrease in center of pressure area of 42% (SD 19%); the range was 16 to 65%. The children with spastic hemiplegia, who had the highest initial effects, had the lowest within-training effects. The four children diagnosed with spastic diplegia had the lowest initial effects and the highest within-training effects.

Patterns of change

Two patterns of change in response to training emerged during the study. The patterns are graphically illustrated in Figures 4a and 4b. The first pattern was one of dramatic, immediate

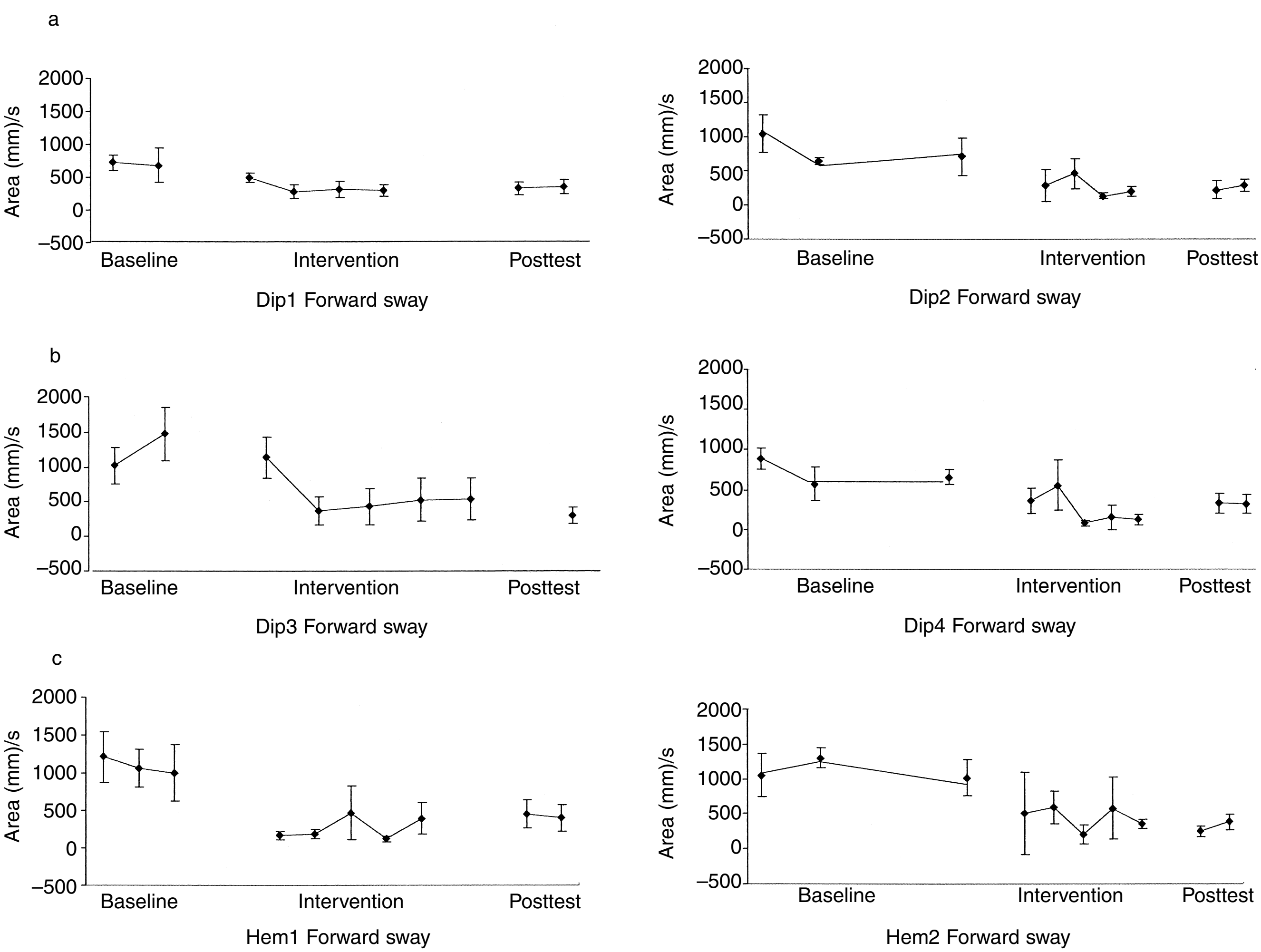


Figure 2: *A comparison of center of pressure area in all six children for each session in three phases of study. Data points (filled diamonds: (◆) indicate mean area for each session in baseline, intervention, and posttest phases. Vertical lines indicate variability (SD) within session. Figures 2a and 2b show data from pairs of children with spastic diplegia; Figure 2c shows data from a pair of children with spastic hemiplegia. Filled diamond data point represents mean of trials for that day (range 2–5 trials); whiskers represent standard deviation for same trials for that day (range 2–5 trials). Dip, spastic diplegia; Hem, spastic hemiplegia.*

change at the beginning of training (large initial effect), followed by little additional improvement throughout the training phase (no within-training effect). The second pattern was a small immediate change (small initial effect) followed by gradual or continual change throughout the training phase (continued within-training effect). The two children with spastic hemiplegia demonstrated the immediate change pattern. In contrast, change in the four children with spastic diplegia was characterized by a more gradual pattern.

Table II: Percentage change in center of pressure area associated with total, initial, and within-training effect in individual children with CP

Child	*Diagnosis*	*Total effect (B to P1)*	*Initial effect (B to T2) %*	*Within-training (T1 to T5) %*
Pair 1				
Dip1	Diplegia	−73.86	−12	−52
Dip2	Diplegia	−34.22	−43	−38
Pair 2				
Dip3	Diplegia	−70.29	−8	−57
Dip4	Diplegia	−73.91	−22	−65
Pair 3				
Hem1	Hemiplegia	−32.87	−83	−26
Hem2	Hemiplegia	−74.25	−53	−16

B, baseline; P, immediately after training; T, training session.

Retention of effects

Effects of training were examined immediately (P1) and again 30 days (P30) after the completion of training in order to determine retention. A comparison of center of pressure area (mm) at baseline, P1, and P30 for forward and backward sway perturbations are shown in Figures 5a and b. Data for individual pairs of children are shown followed by group data on the far right of the graph; all are expressed as mean change in center of pressure. In the immediate posttest session the children with CP decreased their sway area in response to a forward sway perturbation by a mean of 65% (936mm to 313mm) in reference to the baseline value. Decreases ranged from 77 to 53%. In response to backward sway perturbations, center of pressure area decreased by a mean 64% (1032mm to 374mm), with a range of 33 to 74%.

At the 30-day posttest assessment one child was unavailable for retesting. Of the remaining five all had retained improvements gained from training. In response to the forward sway perturbations there was a 60% (936mm to 347mm) mean decrease in center of pressure area 30 days after training compared with baseline with a range of 49 to 65% decrease. Retention of improvements 30 days following training was seen in the response to backward sway perturbations as well, with a mean decrease of 57% (1032mm to 372mm) and a range of 23 to 84%.

Results of the hierarchical linear modeling analysis

Hierarchical linear modeling analysis revealed several relevant findings related to the outcome variable center of

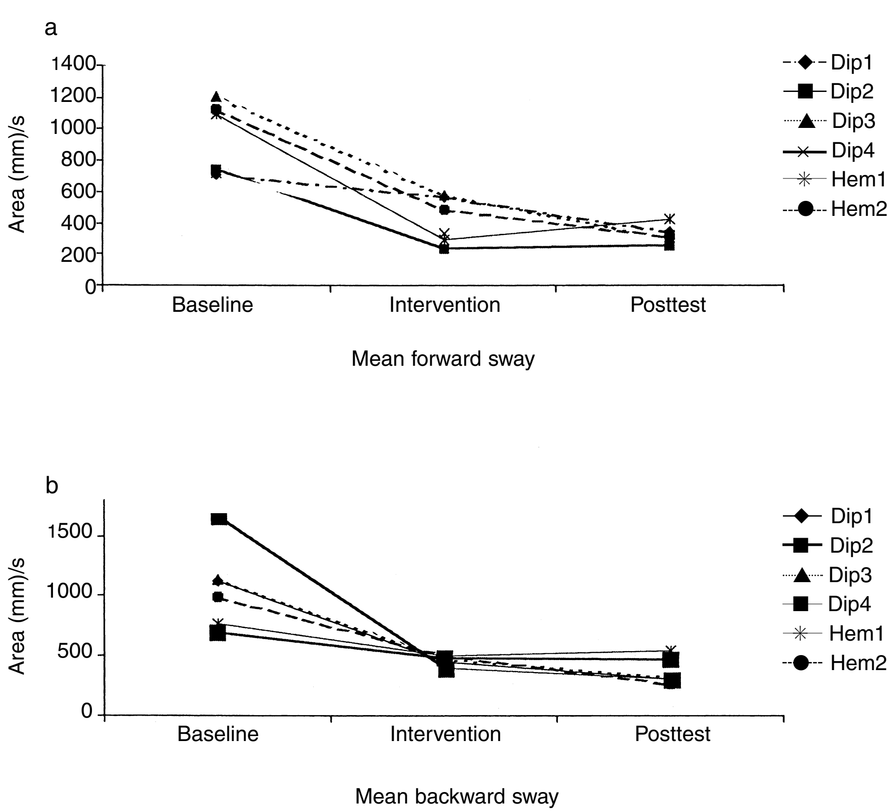


Figure 3: *A comparison of center of pressure area during baseline, intervention, and posttest phases associated with recovery of stability following a forward sway perturbation (Fig. 3a) and a backward sway perturbation (Fig. 3b).*

pressure area. The following is a regression equation that represents main and interaction effects explored with the Level 1 hierarchical linear modeling model: Y=b_0 + b_1(baseline day) + b_2 (intervention)+b_3 (maintenance) +b_4 (interaction 1) + b_5 (interaction 2) +R, where Y is the predicted area at the end of the intervention, b_0 is the mean center of pressure area at the first baseline measurement, b_1 (1=1.5) is the slope of the corresponding term, and R is random error. Results of this Level 1 analysis are shown in Table III. The reliability of coefficients for intercept and slope for all five Level 1 variables (baseline days, treatment, maintenance, day/treatment interaction, day/maintenance interaction were strong (reliability estimate >0.7). In addition, the Level 1 analysis revealed there was no effect of the baseline day variable indicating that there was no change in center of pressure area associated with the passage of time during the baseline phase. However, the analysis revealed an effect for the intervention on area measures in the maintenance phase (b=–681.67, p=0.095). Thus training had a significant effect on the outcome variable (center of pressure area) during the maintenance phase of the study. An estimation of inter-participant variability in all components of the Level 1 analysis indicated statistically significant variability in individual slopes for all main effects and interactions with the exception of the training slope. This is consistent with the finding that there were distinctly different patterns of improvement that emerged during most phases of the project.

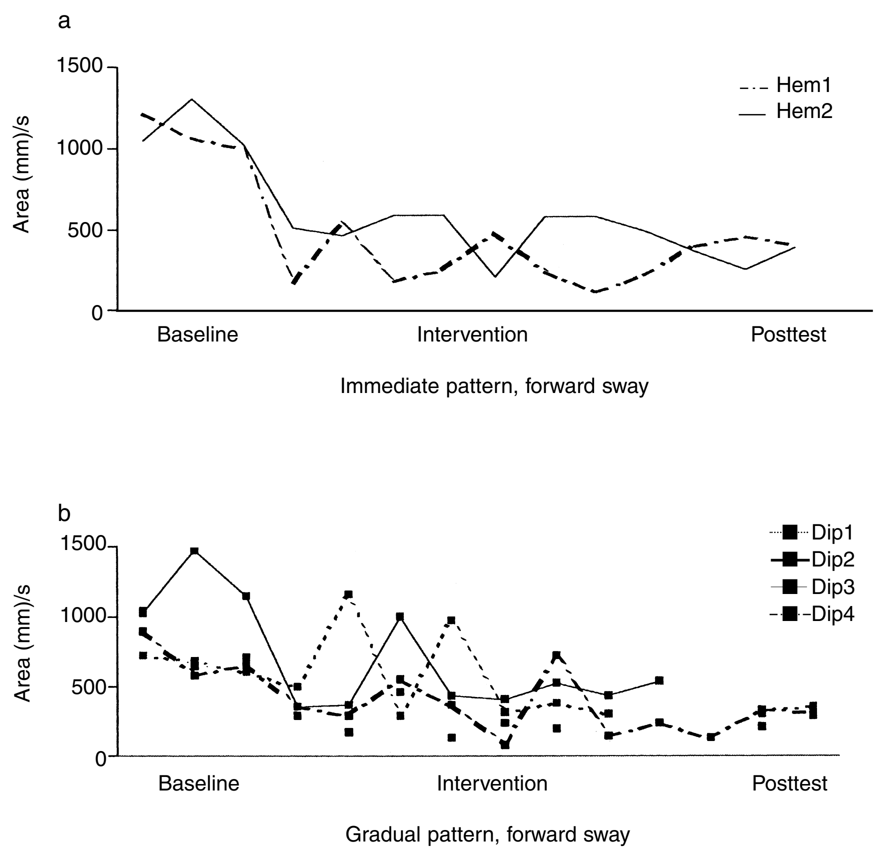
Level 2 analysis examined whether the number of days in baseline, age of the child, or level or severity (based on GMFCS-CP classification) had an effect on the Level 1 intercept and slopes. None of the Level 2 factors had significant influence on the Level 1 intercept or slopes (data not shown). In other words, the number of days spent in baseline, age, and severity of motor problems had no influence on the mean center of pressure area at baseline nor on the effects of baseline, training, maintenance phases, and the day × training and day × maintenance interactions.

Results from the hierarchical linear modeling analyses were supported by results from the repeated measures ANOVA. Two planned comparisons of differences in center of pressure area scores were performed. The first comparison examined change in area between pre-intervention measures (the mean of multiple measures – referred to as 'pre') and the post-intervention measure at P1. The second examined change in area between pre- and the post-intervention measures at P30. For center of pressure area, the comparison of pre (mean 934, SD 248) with P1 (mean 313, SD 82) was statistically significant ($F_{(1,5)}$=38.78, p<0.01), and the proportion of variance explained was η^2=0.89. Also for area, the comparison of pre (mean 871, SD 216) with P30 (mean 348, SD 48) was statistically significant ($F_{(1,4)}$=43.26, p<0.01), and the proportion of variance explained was η^2=0.92. These analyses confirmed the effects of training on center of pressure area/s and support the visual analysis suggesting that improvements in the ability to recover from an external perturbation were maintained 1 month after training was completed.

TIME TO STABILIZATION

The effects of training on the second outcome variable, time to recover stability following a perturbation to stance balance as

Figure 4: *A graphic representation of two patterns for recovery of stability using center of pressure area data. Immediate change pattern is shown in Figure 4a, while gradual change pattern is shown in Figure 4b. Hem, spastic hemiplegia; Dip, spastic diplegia.*



indicated by time to stabilization of the center of pressure, were similar to those related to center of pressure area. Figure 6 graphically displays the mean for the time taken to recover stability following a forward perturbation in the baseline, intervention, and maintenance phases of the study for each child. As was true for the area data, all of the children demonstrated a substantial decrease in time taken to recover from an external perturbation to stance balance following training.

BASELINE PHASE

Before training the mean time taken to recover from an external perturbation was 4.08 seconds (SD 0.52), with a range 3.55 to 4.89 seconds. As was true for the area measure, time to stabilization did not significantly change during the baseline phase for any child.

INTERVENTION PHASE

During the intervention phase the overall mean of time to stabilization decreased to 3.21 seconds (SD 0.27), with a range of 2.95 to 3.54 seconds. All children showed at least a slight decrease in time taken to recover stability following a perturbation by the second training session, an indicator of the initial effects of training. Mean per cent improvement was 27%; the range was 9 to 38%. The child (Hem2) with the smallest initial effect (9%) had the greatest (36%) change for the within-training effect.

Throughout the training phase five of the six children continued to show reductions in the time required to regain stability in response to a platform perturbation. This was indicated by a continued decrease in time to stabilization from the first training session (T1) to the final training session (T5): mean change was 17% (SD 16%); range of improvement was from 0 to 36%. Two children had relatively minor within-training effects with improvement of 3% (Hem1) and 6% (Dip2). The child (Hem1) with the 3% change during the training had the highest improvement (38%) for initial effect. One child (Dip4) had no within-training effect. His initial effect demonstrated improvement of 23% and he made no additional improvement in time to stabilization throughout the training. In contrast to the changes in area/s data, analysis of changes in time to stabilization did not reveal distinct patterns of recovery.

Retention

In the immediate posttest session the five children with CP who were tested (one was unavailable) decreased the time to stabilization in response to a forward sway perturbation by a mean of 34% (4.08 to 2.7 seconds); range was from 24 to

Table III: Results of Level 1 hierarchical linear modeling analysis for center of pressure area

Fixed effect	*Coefficient*	*SE*	*t-ratio*	*p*
B_0	520.86	40.53	12.86	0.0001
$B_{1\ (baseline)}$	18.48	51.6	0.36	0.75
$B_{2\ (intervention)}$	–496.15	249	–1.98	0.16
$B_{3\ (maintenance)}$	–681.67	288	–2.36	0.01
$B_{4\ (interaction\ 1)}$	–56.37	70	–0.8	0.51
$B_{5\ (interaction\ 2)}$	–18.33	53	–0.35	0.76

Table IV: Results of Level 1 hierarchical linear modeling analysis for time to stabilization

Fixed effect	*Coefficient*	*SE*	*t-ratio*	*p*
B_0	3.37	0.12	27.81	0.0001
$B_{1\ (baseline)}$	0.06	0.07	0.91	0.46
$B_{2\ (intervention)}$	–0.55	0.40	–1.37	0.30
$B_{3\ (maintenance)}$	–1.72	0.38	–4.56	0.07
$B_{4\ (interaction\ 1)}$	–0.27	0.08	–3.12	0.14
$B_{5\ (interaction\ 2)}$	–0.05	0.07	–0.75	0.53

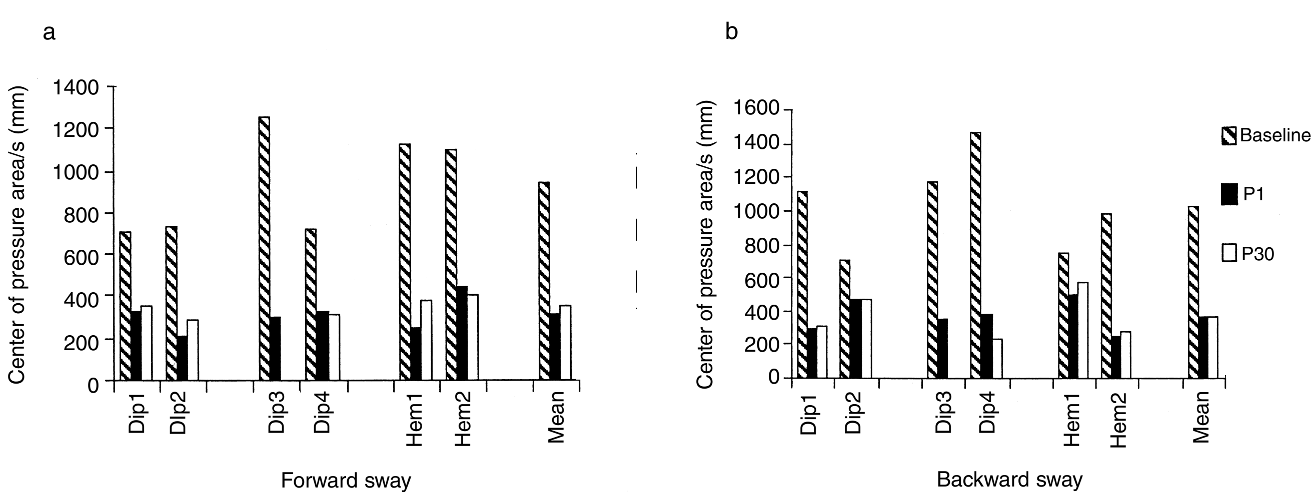


Figure 5: *A comparison of center of pressure area (mm) at baseline, Posttest 1 (P1) and 30-day Posttest (P30) for forward (Fig. 5a) and backward sway (Fig. 5b) perturbations. Data for individual pairs of children are shown; group data are shown on far right of graph. Dip, spastic diplegia; Hem, spastic hemiplegia.*

45%. As was true for center of pressure area, improvements in the time to stabilization were still present 30 days after the completion of training. In response to the forward sway perturbations there was a 24% mean decrease in time to stabilization compared with the baseline (4.08 to 3.01 seconds) with a range of 18 to 39%. Retention of improvements 30 days following training was also seen in the backward sway perturbations (data not shown).

Hierarchical linear modeling analysis

Results of the hierarchical linear modeling analysis for the time to stabilization measures were consistent with the area analysis (Table IV). Reliability estimates for the intercept and slope for all five Level 1 variables was moderate to strong (0.53 to 0.65). The Level 1 analysis revealed no significant effect of time between baseline measurements on the time to stabilization measure. The analysis revealed an effect for the intervention (b=–1.72, p=0.07) on time to stabilization measures in the maintenance phase. Thus training had a significant effect on the outcome variable time to stabilization during the maintenance phase of the study. In contrast to the hierarchical linear modeling analysis for center of pressure area, estimation of variability in all components of the Level 1 analysis for time to stabilization indicated little variability in slopes for main effects and interactions with one exception: there was significant variability ($\chi^2_{(1,6)}$=3.41, p=0.06) in individual slopes during the training phase. Consistent with the hierarchical linear modeling analysis for the area outcome, there was no effect of age nor severity on the time to stabilization outcome.

Results from the repeated measures ANOVA on time to stabilization were also consistent with those from the area analysis. For time to stabilization, the comparison of pre (mean 4.08, SD 0.52) with P1 (mean 2.7, SD 0.39) was significant at $F_{(1,5)}$=48.35, p<0.01, and the proportion of variance explained by this comparison was η^2=0.91. Also for time to stabilization, the comparison of pre (mean 4, SD 0.54) with P30 (mean 3.01, SD 0.19) was significant at $F_{(1,4)}$=19.08, p<0.05, and the proportion of variance explained was η^2=0.83.

RELATION BETWEEN CENTER OF PRESSURE AREA AND TIME TO STABILIZTION MEASURES

During each of the individual phases of the study (baseline, intervention, and maintenance) there was essentially no correlation between center of pressure area and time to stabilization measures (r=–0.11; p=0.694). During recovery of stability following an external perturbation, a slow time to stabilization was not always related to a large center of pressure area.

CLINICAL MEASURE: DIMENSION D – GROSS MOTOR FUNCTION MEASURE

Dimension D of the GMFM was administered at baseline testing and at the immediate post-training testing session for five of the six children participating. The number of trials given on individual items varied but did not exceed three attempts, in accordance with the administrative guidelines of the GMFM (Russell et al. 1993). Performance on Dimension D of the GMFM was quite varied and data are summarized in Table V.

Four of the five children showed improvements or no change in their performance on the GMFM Dimension D, from baseline to P1. One child (Hem1) demonstrated no change in the GMFM, scoring 97.44% of the total points possible for both administrations of the dimension. Another child, with spastic diplegia (Dip4), demonstrated a decline in performance from baseline to P1. The per cent score at baseline was 89.74%, and at P1, the score was 87.18%, a per cent change of 2.87%. Two children (Hem2 and Dip3) demonstrated improvements in their GMFM scores of 11.76% and 11.11% respectively. One child (Dip2) showed an improvement in the score on the GMFM from baseline of 87.18% to posttest score of 89.74% of the total for Dimension D, an increase of 2.94%.

Discussion

The purpose of this research was to examine the effect of massed practice, using a moving platform to create balance threats, on recovery of stability in school age children with CP. We hypothesized that the amount of time required to recover stability (time to stabilization) following a balance threat and the mean area of center of pressure movement per second during balance recovery would be reduced in these children as a result of training and that these improvements in reactive balance control would be maintained at 1 month following the end of training.

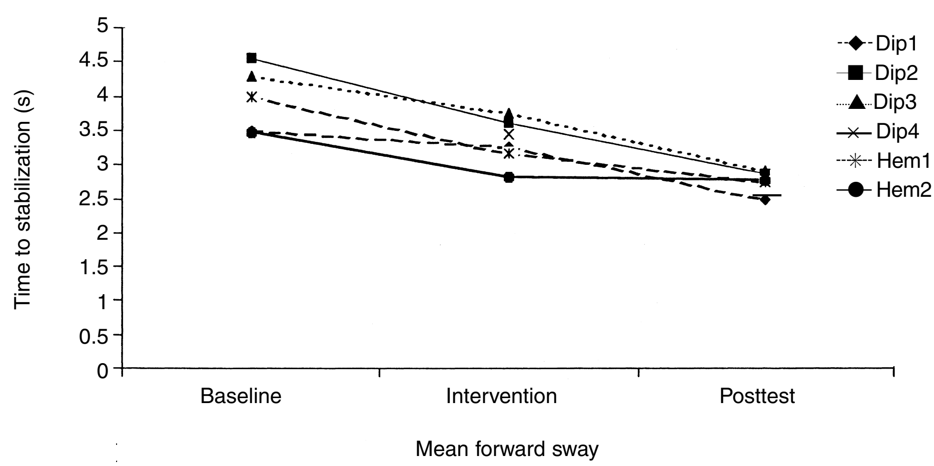


Figure 6: *A comparison of center of pressure time to stabilization during baseline, intervention, and maintenance (posttest) phases associated with recovery of stability following a forward sway perturbation in six children with CP.*

Results of the study supported our hypotheses. We found significant improvements in both time to stabilization of balance and in the mean area of center of pressure movement per second during balance recovery. In addition, this improvement was maintained for 1 month following the end of training for most of the children. A functional relation was established between the massed practice in balance recovery and our dependent variables, time to stabilization, and center of pressure mean area of movement during recovery. According to single-subject design principles analysis, the experimental criterion was achieved by demonstrating six performance shifts at three points in time related to the intervention.

All six children showed a convincing improvement in mean performance level in response to the massed practice in reactive balance control as measured by time to stabilization and mean center of pressure area per second during stabilization. Five of the six children were available for 1-month testing, and all had maintained improvements in time to stabilization and center of pressure area. Hierarchical linear modeling analysis provided statistical confirmation of the significant decrease in time to stabilization and mean center of pressure area. In addition, it was demonstrated that there was no change occurring in these variables during the baseline before intervention. A repeated measures ANOVA provided further confirmation that the changes in time to stabilization and mean center of pressure area were significant.

The changes we found in efficiency of reactive balance control, as measured by time to stabilization and mean center of pressure displacement, were supported by improvements on Dimension D of the GMFM (Russell 1993, 1997) in three of the five children. This subtest measures steady state and/or proactive balance control in standing. Currently there are no tests within the GMFM that measure reactive balance control and, therefore, this could not be measured directly in a functional test of balance control.

Data support two major findings in these school-age children with CP. First, they suggest that postural control mechanisms in school-age children (7 to 13 years) with CP could be modifiable. Second, they indicate that these modifications are maintained at least 1 month following the completion of the training protocol.

Results also suggest that two different patterns of change occurred in the sway area data during the training phase and that these patterns of change were associated with the level of the child's disability. The two children with spastic hemiplegia (both rated level I on the GMFCS-CP) showed a rapid reduction in sway area during training, while those with spastic diplegia (rated level II on the GMFCS-CP) showed a gradual improvement in this variable across training.

The emergence of these two distinct patterns raises the question as to what differences in neurological function between children with spastic hemiplegia and diplegia create the different patterns of training. One might hypothesize that it is the difference in level of involvement that influences the ability of the children with the spastic hemiplegia to respond quickly to training compared with the children with diplegia. When examining the performance level of the children with spastic diplegia and hemiplegia, however, one notes that their baseline and post-training time to stabilization and area of center of pressure during balance recovery are similar. An alternative hypothesis for the differing patterns of change in response to training may lie in the differing levels of sensory motor function in the children with hemiplegia in comparison to the children with spastic diplegia. Research by Gordon and Duff (1999) showed a similar effect in a study on anticipatory control of grip and lift forces in children with spastic hemiplegia. They found that sensory information from the non-involved hand could transfer to improved anticipatory scaling of forces by the contralateral involved hand in subsequent trials. This result must be viewed with caution, however, as the two distinct patterns for recovery were not present in the time to stabilization measure.

Changes in scores on Dimension D of the GMFM were not as robust as those for the platform parameters. All of the children in this study, children with hemiplegia and spastic diplegia, scored at approximately 70% or higher on this one dimension (D). The children with diplegia had small improvements of 5.7 to 11.11%. In addition, both children with hemiplegia scored at 97.4% at follow-up testing. This was an improvement of 11.76% for the second child (Hem2) and no change for the first child (Hem1). Thus there may have been a ceiling effect for the second child with hemiplegia as the original scores were 97.4%.

A study by Trahan and Malouin (1999) found that over a period of 8 months and three administrations of the GMFM (baseline, 4, and 8 months from baseline), the GMFM provided information on changes in the gross motor performance of children with different types of CP. They reported that the children scoring the lowest per dimension at baseline (<65%) made the most significant improvements over the two follow-up tests. These were the children with spastic diplegia and quadriplegia demonstrating improvements of approximately 10% for dimensions D. The children with the highest scores at baseline (>95%) showed improvements at follow-up of 5 to 7% for dimension C, D, and E (<1% on dimensions A and B). These were the children with spastic

Table V: Summary of Gross Motor Function Measure (GMFM; Russell 1993, 1997) scores

Child[a]	*GMFM*			
	Baseline	*Percentage score*	*Posttest 1*	*% score*
Dip2	34	87.17	35	89.74
Dip3	27	69.23	30	79.92
Dip4	35	89.74	34	87.18
Hem1	38	97.44	38	97.44
Hem2	34	87.18	38	97.44

[a]GMFM data not available for posttest in child Dip1.

hemiplegia. In their study, the children with diplegia had the highest total score gains of 7%, children with quadriplegia showed gains of 6.2%, while the children with hemiplegia had lower gains of 4.2%.

However, Russell and coworkers (1993) reported that the largest changes occurred in the children with hemiplegia (6.3%), followed by children with diplegia (4%), and children with quadriplegia (2.1%). The findings from the current center of pressure study concur with Trahan and Malouin (1999), in that the children's baseline scores were above 70% and the gains were relatively small and for the one dimension. Given that the largest changes reported above by Russell and colleagues (1993) and Trahan and Malouin (1999) were over periods of 6 and 8 months respectively, the changes demonstrated by the children in the current training study over a period of 1 week were substantial.

Results of this study have important clinical implications. The most obvious is that the ability to recover stability following perturbations is modifiable in school-age children with CP. For therapists working with children with these diagnoses and this age range the indication would be that continued practice in balance recovery and stability can help children to make changes in the their ability to maintain and recover balance. This is an important finding for children in the upper elementary/middle school ages as this is a time when therapy services may be reduced or eliminated. The results of this study support continued therapy opportunities for children with CP. Children who are typically developing may have attained adult type patterns of balance and stability by the age of 7 to 10 years (Shumway-Cook and Woollacott 1985). Children with CP, however, are delayed in this attainment and, according to our findings, are still able to make modifications even into the pre-teen years.

Specificity of training in our study and, therefore, the small but noteworthy degree of carry over we saw in terms of functional mobility also has implications for therapists working with this population of children. This may mean that a wide range of challenging posture and balance work may be necessary to facilitate functional changes for the children, including training in reactive balance control at high intensities.

Differences in the pattern of response to training in the children with spastic diplegia versus hemiplegia also indicates possible differences in training requirements for children with the two types of diagnoses. Our findings suggest that children with spastic diplegia may need prolonged training in reactive balance control to show similar improvements to those with spastic hemiplegia.

Strengths of this research include the fact that the study was designed to include provisions to strengthen the internal validity. The replicated multiple baseline design provided check points for onset of changes compared with the delivery of the intervention such that the effect or influence of other extraneous factors could be ruled out. This study paired children with similar diagnoses to begin baseline data collection together but spread out the training protocol such that when one child finished training, the second child started. Therefore, any changes seen during training can be attributed to the training as opposed to other outside factors that may have occurred during that week. Dramatic intervention effects were demonstrated in terms of both magnitude (mean and level) and rate of change (latency).

A limitation often given of single-subject research is the small number of participants in the study (n=6). Single-subject research attempts to offset the small numbers by building in protection from outside influences such as using replicated multiple baselines, as well as matching the pairs of children by diagnosis and as closely in age as possible. Despite these measures, a limitation of this study remains the small number of participants, making results preliminary in nature.

A second limitation in the study involves the lack of control of children's activity outside of training sessions. Families were not asked to alter their school, family, or therapy schedules during the study. Therefore, the extent of the balance or posture control work done during therapy sessions was not known. However, due to the specificity of the platform training it is doubtful that outside practice would have resulted in improved performance on the platform.

The type of training performed in this study was quite specific to postural control underlying recovery from an external threat to stability, thus the extent to which results from this study generalize to other forms of balance training in children with CP is unknown.

In conclusion, one question addressed by this study was whether school-age children with CP can alter their postural control systems in response to intense reactive balance training. Evidence from this study suggests that postural control in school-age children is modifiable, and does improve in response to intense balance training using a moveable force platform. It is not clear whether these findings extend to more traditional forms of therapy targeting balance control in this population of children. More research is necessary to determine the type and frequency of intervention needed to impact postural control in school-age children with CP.

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List of abbreviations

P1	Period immediately after training
P30	Period 30 days after training
‘pre’	Mean multiple of pre-intervention measures
T1 – T5	Training sessions 1 to 5

5

PERTURBATION-BASED BALANCE TRAINING TO REDUCE FREEZING OF GAIT IN AN INDIVIDUAL WITH PARKINSON’S DISEASE: A CASE STUDY

PERTURBATION-BASED BALANCE TRAINING TO REDUCE FREEZING OF GAIT IN AN INDIVIDUAL WITH PARKINSON'S DISEASE: A CASE STUDY

Travis Gawler, DPT, NCS

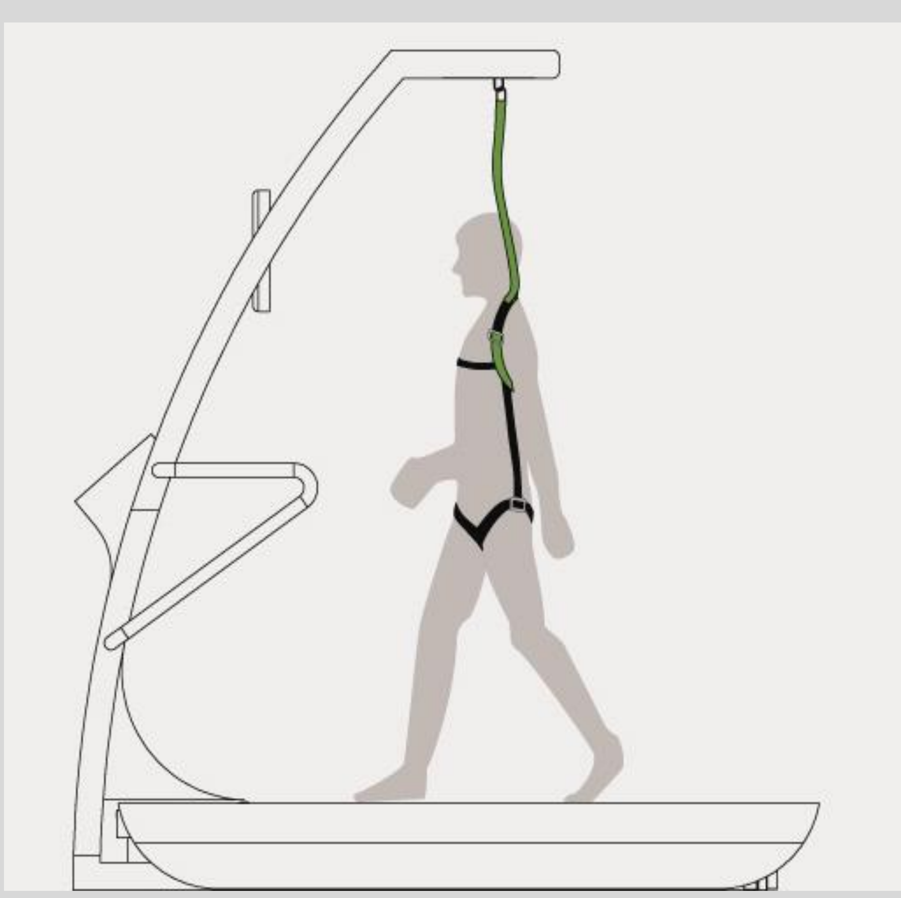
PALMETTO HEALTH

Introduction

Freezing of gait is a disorder frequently observed in Parkinson's disease(PD) in which individuals describe a feeling of their feet being briefly glued to the floor while walking.[1] It is often very difficult to treat and poorly understood.[1] Given the limited benefit of medication on alleviating postural instabilities and subsequent falls for individuals with PD, physical therapy approaches are necessary. Current literature has shown that perturbation training in individuals with PD leads to improved gait performance however this has only been shown in individuals without freezing of gait.[2]

Purpose

The purpose of this study is to describe the use of a novel intervention of perturbation training to alleviate freezing and falls in an individual with Parkinson's disease.



Case Presentation

A 62 y.o. male with a chief complaint of increasing falls over the past 1-2 months when getting out of a car and when carrying objects with a medical history of Parkinson's disease and bilateral deep brain stimulation surgery in 2004.

Initial evaluation included:

- Completion of Mini-BESTest,
- Functional Gait Assessment (FGA)
- Five time sit to stand test
- Timed Up and Go (TUG) test for basic, cognitive and manual subtypes.

Findings revealed:

- Impaired corrective and protective balance strategies
- Difficulty with dual tasking while walking
- Postural dysfunction
- Gait impairment

The treatment intervention included progressive multi-directional perturbation training on the MediTouch Balance Tutor perturbation system. Interventions were progressed from a static base of support(BOS) to a dynamic BOS with patient walking on the Balance Tutor™ system at a speed of 1.1 - 1.4 mph. Over the course of 5 treatments knowledge of the direction of perturbation was varied, amount of perturbation was increased and then manual and cognitive loads were added by having the patient carry weighted bolsters, complete a unilateral manual puzzle and/or perform retro-counting or naming tasks. The patient received treatment twice a week for four weeks, totaling seven visits however two of these were for evaluation/re-evaluation purposes. At discharge, the patient was transitioned to a tailored home exercise program including multi-directional stepping with and without manual and cognitive components.

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Results

Medication dosage and DBS settings were unchanged over the course of treatment. Most functionally significant change was falling frequency of 2-3 per week to no falls over the 4 weeks leading to discharge. The Mini-BESTest improved by 7 points which exceeds minimal detectable change of 5.52 points.[3] TUG-manual improved by 12 seconds however statistical significance cannot be determined as MDC values are not established. Difference between TUG-manual and basic has improved to 3.9 seconds which is now below 4.5 second cut-off for increased fall risk.[4] There was not a significant change in FGA, TUG-basic, TUG-cognitive and Five time sit to stand test scores.

Weekly Fall Frequency	
First Session	2-3
Last Session	0

Fig 1. Weekly Fall Frequency

Mini-BESTest	
First Session	21/28
Last Session	28/28
Improvement	7

Fig 2. Mini-BESTest scores

	Timed Up and Go Test			
	TUB-basic	TUG-cognitive	TUG-manual	Diff.(manual-basic)
First Session	10.3	12	26	15.7
Last Session	10.3	11	14	3.7
Improvement	0	1	12	12

Fig 3. Timed Up and Go scores(in seconds)

Discussion

The patient reported:

- Resolution of falls, decreased frequency of freezing with walking, and increased confidence with his mobility.
- The patient's balance reactions through use of ankle, hip and stepping strategies are significantly more efficient following perturbation training using the MediTouch Balance Tutor System.
- While a cause-and-effect relationship cannot be inferred from a case report, it is plausible that perturbation training using the Balance Tutor system may decrease severity and frequency of freezing of gait with walking.
- Physical Therapy needs to continue to assess and refine treatment protocols for this condition.

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6

Clinical implementation of a reactive balance control assessment in a sub-acute stroke patient population using a ‘lean-and-release’ methodology

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Clinical implementation of a reactive balance control assessment in a sub-acute stroke patient population using a 'lean-and-release' methodology

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ABSTRACT

Reactive balance control, specifically performance of rapid stepping responses, is associated with falls, but not routinely assessed in clinical practice. Challenges to clinical assessment may include a lack of available methods that are safe, standardized and able to quantify the balance responses. We implemented a reactive balance control assessment, using lean-and-release methodology, in an inpatient stroke rehabilitation program. Through retrospective chart review of all admissions ($n = 183$) over a 1-year period, we evaluated the clinical uptake and patient-specific factors associated with its use. Seventy-seven of 183 (42%) patients were administered the assessment, on average, 16.2 (SD 13.1) days post-admission. Patients who received the assessment were younger, at an earlier time post-stroke, with a shorter rehabilitation length of stay, with less lower-limb impairment, higher levels of functional balance, less motor and cognitive impairment, greater recovery of functional mobility, and were more likely to have the capacity to walk (all measures $p < 0.0001$), compared to those who did not receive the assessment. This study demonstrates the potential for clinical uptake of the lean-and-release assessment among patients with stroke, who are progressing in their functional and mobility status over the course of their inpatient rehabilitation. However, the results suggest limitations in application to patients with greater disability or who demonstrate slower recovery of functional mobility. Ongoing research is required to develop clinical approaches to reactive balance control assessment that are effective, efficient and relevant to clinical populations and feasible for clinical practice.

1. Introduction

Your article is registered as a regular item and is being processed for inclusion in a regular issue of the journal. If this is NOT correct and your article belongs to a Special Issue/Collection please contact m.radhakrishnan@elsevier.com immediately prior to returning your corrections.''.Falls risk post-stroke is high with significant physical and psychosocial consequences contributing to decreased independence, activity and participation [1]. A critical factor underlying falls is the lack of ability to recover from a loss of balance, specifically using a rapid stepping response [2]. Reactive stepping performance is predictive of falls among community-living elderly [3,4], associated with falls among those with stroke in inpatient rehabilitation [5] and predictive of falls after discharge to the community [6].

However, reactive balance control is not routinely assessed in physiotherapy practice [7]. Commonly used clinical measures focus on volitional limb control and self-governed speed of movement [7]. Limited availability of clinical tools is among the top three cited barriers to improving assessment practices of reactive balance control [8]. Assessments that probe reactive balance control in response to a push or pull do exist [9,10]. Additional challenges to clinical implementation may include safety concerns [11] and difficulties in using a standardized protocol with a controlled method of perturbation and quantification of response.

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Observation-based measures may also mask mechanisms underlying performance [12,13]. For example, delays in step initiation phases, that occur prior to observable limb movement, can influence the overall success or failure of a response [5,6,14].

Many methodologies, for example translating platforms [15], are prohibitive for clinical use due to the cost, space and technological support required. Use of a weight-drop cable-pull has been proposed as a clinical tool [16], and a bidirectional cable system was piloted in stroke [17], but considered too complex for routine application. Alternatively, experimental paradigms using a lean-and-release methodology [18] may be feasible. Specifically, the patient is attached to a safety harness and leans forward on a horizontal cable that is suddenly released, simulating a forward fall. The cable load associated with the lean angle can be easily measured, providing a means to control perturbation magnitude. This approach (detailed below) was, therefore, developed as a standard technique in a stroke rehabilitation setting, to address the current gap in assessment practice of reactive balance control.

This study evaluated the clinical uptake of the lean-and-release methodology as a reactive balance control assessment for patients within inpatient stroke rehabilitation, specifically by determining: (i) to what extent was the assessment used and implemented clinically, and (ii) what were the patient-specific determinants that influenced use of this assessment methodology?

2. Methods

This study was a retrospective chart review approved by the institution's research ethics board.

2.1. Setting and participants

Assessments were administered within a novel on-site clinic that partners researchers and clinicians, with an aim to accelerate new technology and findings into practice. An assessment, that integrates technological (e.g. forceplates, pressure-sensitive mats) and clinical measures to assess balance and gait, was collaboratively developed [19]. The lean-and-release assessment is one component of the larger assessment. The 'front-line' physiotherapists administer the assessment within the clinic as part of routine practice. Trained health-care students assist with equipment operation and post-processing of forceplate data for the clinical report.

A retrospective review was conducted for consecutive admissions to the Stroke inpatient rehabilitation program over a 1-year period, specifically October 2011 to September 2012. From the 189 patients identified, 1 patient was excluded due to an extremely short length of stay (1 day) and 5 patients were excluded who were discharged to acute care, due to medical status change, and did not complete inpatient rehabilitation. A final sample of 183 patients was included in subsequent analyses.

2.2. Lean and release methodology

The lean-and-release balance perturbation assessment simulates a forward fall (see Fig. 1). A commercially available, over-bed transfer gantry (Prism Medical, Concord, ON, Canada) was modified to provide a low-cost harness system that allowed for unrestricted movements but ensured safety if the patient was unable to recover balance. The individual is held in a static forward lean position by means of a horizontal cable, attached at the level of his/her chest, which is released unexpectedly.

2.2.1. Magnitude of perturbation

A standardized template for pre-perturbation standing position was adopted [20]. A load cell [A-Tech Instruments Ltd., Scarborough, ON, Canada] was mounted in series with the tether cable to measure pre-perturbation cable load (during the forward lean) and quantify the perturbation magnitude when the cable was released. Cable load was monitored by the physiotherapist across all trials. Patients were encouraged to lean forward from the ankles such that 8–10% of their body weight was consistently supported. For reference, 10–12% body weight corresponds to a whole-body lean angle of approximately 9° from vertical and is of sufficient magnitude to consistently elicit a stepping response in healthy young adults [21]. Lesser lean angles were allowed at the therapist's discretion, according to functional ability of the patient.

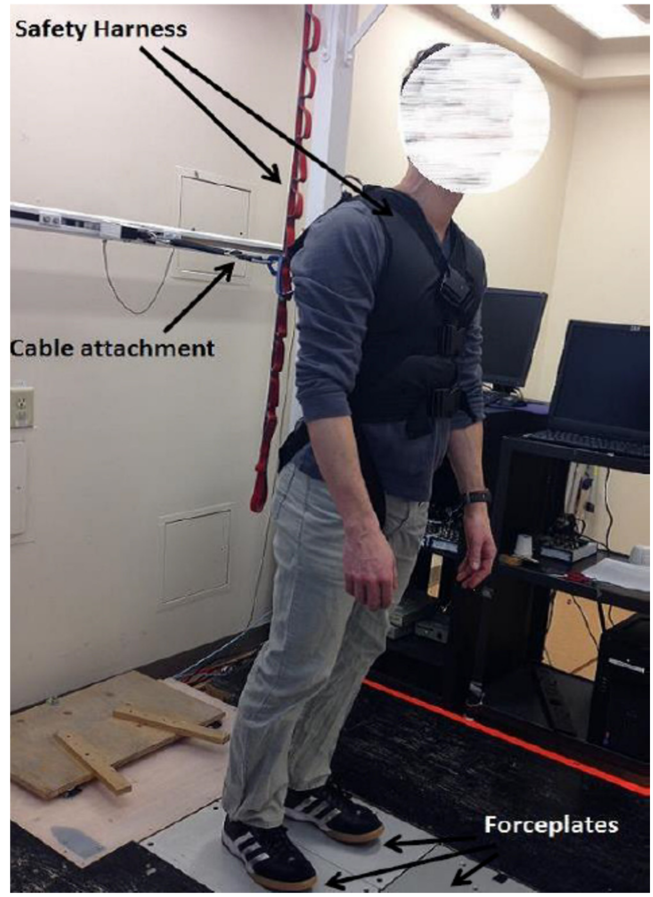

Fig. 1. The 'lean-and-release' balance perturbation method. A 4-post overbed transfer gantry was modified: legs were elevated to accommodate standing beneath the frame and a cross-bar was added for cable attachment. The patient wears a safety harness that is attached to the overhead support structure. The patient leans forward on a cable connected horizontally to the frame. The cable load associated with the lean angle is measured to determine the magnitude of the perturbation. The cable is released unexpectedly inducing a forward fall. The assessment is video-recorded to document patient performance, including need for assistance, number of steps required and stepping limb preference. Forceplates were added to provide measures of pre-perturbation symmetry (body weight beneath both feet) and timing of response. The forceplate positioned in front captures foot contact time of the stepping limb.

2.2.2. Measurement

The assessment was video-recorded to enable review of patient performance (e.g. need for assistance, stepping-limb preference, number of steps required) [22]. We additionally used forceplates (Advanced Medical Technology Inc., Watertown, MA) to measure patients' stance symmetry (percent body weight under each lower limb) and timing of stepping responses (e.g. foot off, swing and foot contact time) referenced to the onset of perturbation [5,14].

2.2.3. Tasks and instructions

The assessment protocol included two task conditions: 5 trials each of preferred-response (PREF) and encouraged-used (ENC). In PREF, patients were instructed to 'respond however you would naturally to recover your balance'. In ENC, the preferred stepping limb (most frequently used in PREF) was blocked by the physiotherapist, by placing his/her hand or foot approximately 5 cm in front of the patient's shin, to encourage stepping with the opposite limb. The patient was instructed to 'respond however you would naturally to recover your balance, knowing that I have blocked this limb'.

This set-up was deemed to provide a safe, quantifiable and standardized method for measuring reactive balance control.

2.3. Measures

Patients were classified as to whether they received (REACT) or did not receive (NoREACT), the reactive balance control assessment. Patient characteristics extracted from health records included gender, age, height, weight, time post-stroke, affected side and length of stay (LOS). Patient clinical profile was determined from the following measures: Chedoke-McMaster Stroke Assessment Impairment Inventory (CMSA) [23] (lower-limb impairment); Berg Balance Scale (BBS) [24], St. Thomas Risk Assessment Tool in Falling (STRATIFY) [25], and history of falls during inpatient stay (functional balance and fall risk); Functional Independence Measure [26], total (FIM-T), motor (FIM-M) and cognitive (FIM-C) subscores (functional disability), and Clinical Outcome Variables Scale (COVS) [27] (functional mobility and walking status).

2.4. Data analysis

Statistical analyses were performed with SAS 9.3 (SAS Institute, Inc., Cary, NC, USA). Descriptive characteristics were used to characterize the patient sample. Frequency values were used to describe the prevalence of clinical use of the assessment and trials and conditions successfully completed. Unpaired *t*-tests and Fisher's Exact tests were used to detect mean and proportional differences in patient characteristics across REACT and NoREACT groups. A two-way analysis of variance was conducted with factors of group (REACT and NoREACT) and time (admission and discharge), to determine differences in clinical profile across patient groups. For all statistical analyses, $\alpha = 0.05$.

3. Results

3.1. Clinical use and implementation

Seventy-seven of 183 (42%) stroke inpatient admissions were administered a lean-and-release reactive balance control assessment during their rehabilitation stay. Of those assessed, 27/77 (35%) completed both initial and discharge assessments whereas 50/77 (65%) received only one assessment. On average, the first assessment occurred 16.2 (SD 13.1) days post-admission; this equated to the midpoint (50.6% SD 26%) of patients' overall LOS. Post hoc analyses revealed that patients who received one assessment received this significantly later ($p < 0.0001$) in their rehabilitation (20.0 SD 14.5 days; 61% SD 25% LOS) than those who received both initial and discharge assessments (9.1 SD 5.9 days; 30% SD 14% LOS). There were no other differences (all comparisons $p > 0.25$) in patient characteristics (gender, age, height, weight, time post-stroke, affected side, inpatient length of stay) or clinical profile during rehabilitation (CMSA leg/foot, BBS, fall risk score, history of falls, FIM-total/motor/cognitive, COVS, walking status) between those who received one versus two assessments.

Across patients, 98.5% (1065/1081 trials) of perturbations elicited a stepping response, with a mean cable load of 7.9% (SD 3.0%) body weight. Nine of the 16 trials that did not elicit a step required support from the therapist and/or harness system for balance recovery ($n = 4$ trials), elicited a grasp response where the patient reached to the therapist for support ($n = 1$ trial) or both ($n = 4$ trials). At initial assessment, 75/77 (97.4%) and 66/77 (85.7%) of REACT patients completed ≥ 3 of 5 trials of the PREF and ENC conditions, respectively. Eight of 77 (10.4%) patients did not complete any ENC trials: five patients exhibited failed responses in the preceding PREF trials (including trials with total assistance and no observable stepping reaction); two patients had difficulty understanding instructions, due to English as a second language; and one therapist administered additional PREF trials (where observably the patient used both their paretic and nonparetic limb).

3.2. Patient-specific determinants of reactive balance control assessment

Patient characteristics and clinical profile for REACT and NoREACT patients are found in Table 1. The distribution of individual patient scores on functional balance, disability and mobility measures, for NoREACT and REACT patients, is displayed in Fig. 2. At time of admission, REACT patients were younger ($p = 0.002$), earlier post-stroke ($p = 0.005$) and had a shorter LOS ($p = 0.005$) than NoREACT patients. There was a significant group by time interaction for COVS ($F_{(1,152)} = 5.32$; $p = 0.023$); REACT patients made greater change in functional mobility than NoREACT during rehabilitation. Post hoc analyses revealed significant between group-differences in COVS scores at time of admission ($p = 0.003$) and discharge ($p < 0.0001$). There were no other significant interaction effects. There was a main effect of time for all other clinical measures (all $p < 0.0001$). There was a main effect of group: REACT patients had significantly less lower-limb impairment (CMSA leg/foot), higher functional balance scores (BBS), less functional disability (FIM-T, FIM-M, FIM-C), and were more likely to be able to walk without assistance (all $p < 0.0001$) compared to NoREACT patients. There were no differences in identified fall risk (STRATIFY) but REACT patients were less likely to experience a fall during rehabilitation than NoREACT patients ($p = 0.005$). Post hoc review revealed that 7 of 8 REACT and 24 of 29 NoREACT patients who fell, did so during wheelchair transfers while the remaining patients fell during upright stance or walking.

4. Discussion

This study aimed to determine the clinical uptake of a standardized lean-and-release assessment method adapted for use within an inpatient stroke rehabilitation program. The results suggest that the reactive balance control assessment was adopted for clinical use with a substantial proportion, although not majority, of patients. Clinical profiles suggest that most patients assessed had considerable post-stroke balance and mobility impairments. However, on average, patients more likely to be assessed versus not, were younger, at an earlier time post-stroke, with shorter length of stay, less lower-limb impairment, higher levels of functional balance, less motor and cognitive disability, greater recovery of functional mobility, and more likely to have the capacity to walk without physical assistance. The assessment consistently elicited a reactive stepping response. The majority of patients were able to complete both preferred and encouraged-used trials.

The results of this study demonstrate the potential for clinical uptake of reactive balance control assessment methodologies that may better reveal and quantify underlying dyscontrol. The clinical profiles of REACT patients suggests that, on average, physiotherapists are likely to administer this assessment as the patient progresses during the course of their inpatient rehabilitation. Plausibly, an evaluation of a patient's capacity to use a reactive step for balance recovery increases in importance as the patient gains independence in standing and walking. The majority of patients after stroke regain the capacity to walk [28]; however, walking is the most common activity preceding falls after discharge to the community [1]. It would be appropriate, then, for physiotherapists to prioritize assessment of a patient's reactive balance control in the course of their rehabilitation and as they prepare for discharge from the hospital.

However, the results also suggest that there are limitations to the use of this assessment within early stages of post-stroke recovery that require further investigation. Differences in clinical profiles between REACT and NoREACT patients suggest there are perceived limitations to administering this more challenging balance assessment to those in early stages of rehabilitation who present with greater disability. Whereas the inability to stand would be an obvious and real barrier, further research is required to determine the factors that would limit application of this methodology to the lower-functioning individual who may benefit

Table 1
Clinical profile of patients who received a lean-and-release reactive balance control assessment (REACT) versus those who did not (NoREACT) during inpatient stroke rehabilitation.

	REACT $n = 77$		NoREACT $n = 106$		Group comparison
	Admission	Discharge	Admission	Discharge	p-Value
Patient characteristics					
Sex (male:female)	45:32	–	53:53	–	0.29
Age (yrs)	63.9 SD 14.7	–	70.4 SD 12.8	–	0.002
Height (cm)	167.4 SD 8.3	–	165.4 SD14.0	–	0.22
Weight (kg)	73.0 SD 17.1	–	71.6 SD 14.6	–	0.59
Time post-stroke (days)	14.1 SD 12.0	–	19.2 SD 15.0 ($n = 103$[a])	–	0.005
Affected side (L:R:both:no)	36:35:5:1	–	53:39:3:11	–	0.43
Inpatient rehab LOS (days)	30.6 SD 15.8	–	39.7 SD 27.4	–	0.005
Lower limb impairment					
CMSA – leg (out of 7)	4.7 SD 1.7 ($n = 73$)	5.1 SD 1.0 ($n = 57$)	3.9 SD 1.3 ($n = 83$)	4.2 SD 1.2 ($n = 63$)	$F_{(1,158)} = 22.1$; $p < 0.0001$
CMSA – foot (out of 7)	4.3 SD 1.4 ($n = 73$)	4.6 SD 1.3 ($n = 57$)	3.3 SD 1.3 ($n = 82$)	3.7 SD 1.5 ($n = 62$)	$F_{(1,156)} = 21.4$; $p < 0.0001$
Functional balance & fall risk					
BBS (out of 56)	35.4 SD 16.6	50.4 SD 5.1 ($n = 71$)	22.8 SD 17.8 ($n = 104$)	35.6 SD 14.6 ($n = 89$)	$F_{(1,180)} = 43.8$; $p < 0.0001$
STRATIFY 'high' fall risk	27/75 (36%)	–	51/104 (49%)	–	0.09
In-patient fall history (≥1 fall)	–	8/77 (10%)	–	29/106 (27%)	0.005
Functional disability					
FIM-T (out of 126)	86.7 SD 19.6 ($n = 72$)	115.5 SD 7.7 ($n = 72$)	73.0 SD 23.7 ($n = 99$)	98.7 SD 19.9 ($n = 98$)	$F_{(1,169)} = 31.6$; $p < 0.0001$
FIM-M (out of 91)	60.5 SD 18.0 ($n = 72$)	84.2 SD 6.1 ($n = 72$)	50.3 SD 20.4 ($n = 99$)	70.9 SD 16.5 ($n = 98$)	$F_{(1,169)} = 25.6$; $p < 0.0001$
FIM-C (out of 35)	26.2 SD 5.5 ($n = 72$)	31.4 SD 3.8 ($n = 72$)	22.7 SD 5.9 ($n = 99$)	27.8 SD 5.2 ($n = 98$)	$F_{(1,169)} = 21.9$; $p < 0.0001$
Functional mobility					[b]
COVS (out of 91)	63.8 SD 16.7	80.5 SD 6.6 ($n = 70$)	55.8 SD 17.0 ($n = 102$)	67.9 SD 13.3 ($n = 84$)	*Ax*: $F_{(1,177)} = 9.3$; $p = 0.003$ *DC*: $F_{(1,152)} = 52.1$; $p < 0.0001$
Walk without assistance: yes (%)	46/77 (60%)	70/77 (91%)	40/106 (38%)	67/106 (63%)	$F_{(1,181)} = 26.9$; $p < 0.0001$

REACT and NoREACT group are $n = 77$ and $n = 106$, respectively unless otherwise stated.
[a] $n = 3$ outliers were removed from analysis where time post-stroke at admission ranged 159–243 days.
L = left. R = right. LOS = length of stay. CMSA = Chedoke-McMaster Stroke Assessment Impairment Inventory. BBS = Berg Balance Scale. STRATIFY = St. Thomas Risk Assessment Tool in Falling. FIM-T, FIM-M, FIM-C = Functional Independence Measures, total, motor and cognitive subscores, respectively. COVS = Clinical Outcome Variables Scale. Walk without assistance: extracted from COVS Item 5 'Performance of Ambulation' and included all patients who scored ≥ 4 i.e. could walk with supervision or independently with/without walking aid.
[b] COVS analyses revealed group by time interaction therefore group comparisons were performed separately at admission (Ax) and discharge (DC).
For reference, high scores on the CMSA, BBS, FIM and COVS mean less lower-limb impairment, greater functional balance abilities, less functional disability and greater functional mobility, respectively.

from the assessment. Some patients who were assessed also did not complete all trials and conditions; further investigation and modifications to the protocol may be warranted, to determine the optimal number of trials and conditions required to yield important clinical information.

The majority of patients (65%) were assessed approximately ten days before discharge which may limit the therapist in using the assessment to guide treatment at this late stage of rehabilitation. Effective methods to train reactive balance control [14,29] are emerging and there is encouraging evidence to suggest that rapid adaptations of balance control can occur post-stroke with only a few sessions of training [14]. However, the necessary training dose that will translate to effective and learned responses when patients are exposed to real-life challenges is not yet known. A subgroup of REACT patients (35%) were administered the assessment earlier in their rehabilitation, allowing for therapeutic use of information gained and evaluation of interventions through follow-up assessment. There were no obvious clinical differences to those who received one versus two assessments. Further research is therefore required to determine factors that may pose barriers to timely administration of the assessment.

Other patient-specific characteristics or comorbidities, not included in this study (e.g. unstable medical status, recent surgery, low exercise tolerance, musculoskeletal disorder or pain), may have influenced the choice to use the assessment or not, or timing of its implementation. Patient preferences or anxiety may have contributed to therapists' clinical decision-making [19]. Alternatively, it is possible that the present results underestimate the potential clinical use: that many NoREACT patients were clinically appropriate for assessment, but other factors specific to the therapist or practice environment posed as barriers. Therapist time constraints have previously been cited as barriers to completing balance and gait assessments [8,19]. It may be noteworthy that, at time of discharge, the mean clinical scores of functional balance, disability and mobility for NoREACT patients were equal to, or surpassed, the REACT patients' admission scores. Plausibly, some NoREACT patients were clinically appropriate for assessment but were nearing their discharge from the hospital. It is possible, then, that considerations of both patient readiness for assessment juxtaposed with residual length of stay, influenced therapist decision-making. This is worthy of attention given current recommendations to accelerate the transfer of acute patients to inpatient rehabilitation, enabling earlier, intense therapy within a reduced length of stay [30]. Reactive stepping performance at time of discharge from stroke rehabilitation is predictive of falls upon return to the community [6]. Further, fall rates are highest in the early stages after discharge from the hospital [1]. The present results suggest that patients with slower rates of functional mobility recovery after stroke may not be adequately assessed (or trained). Given the importance of reactive balance control, we should continue to prioritize the development of effective yet also efficient clinical methodologies and processes that would support assessment and training of patients at this critical transition from rehabilitation to community living.

Interestingly, reactive balance control was not more likely to be assessed with patients who fell. It is possible that REACT patients received different treatment leading to fewer falls. However, the majority of falls occurred during wheelchair transfers, not during

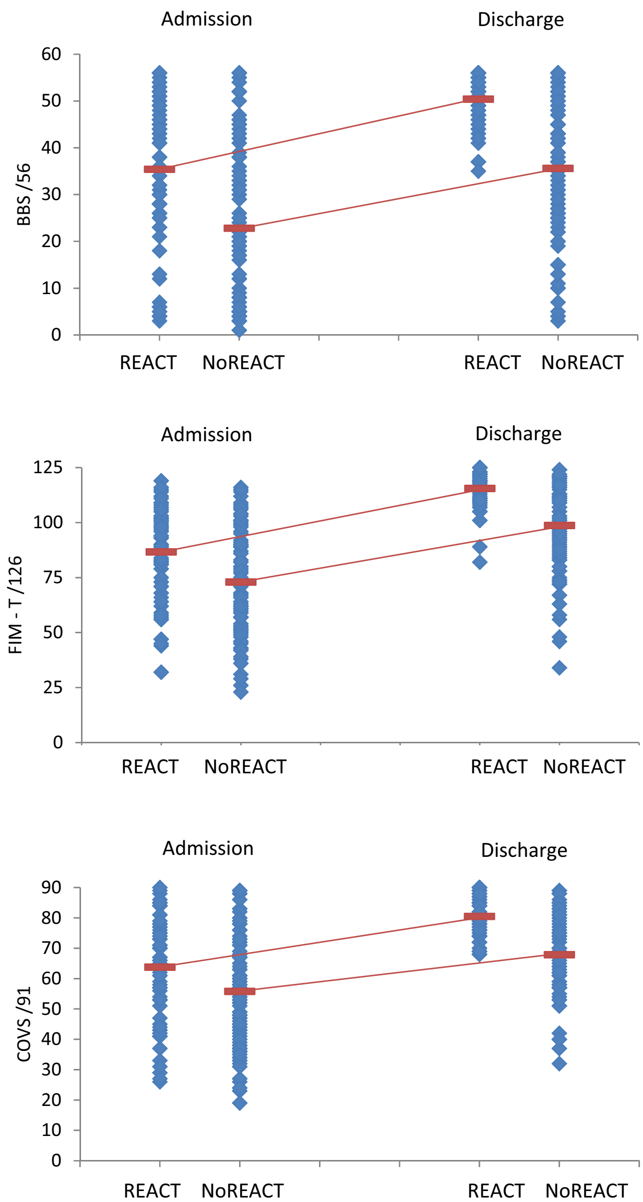

Fig. 2. Distribution of scores for those who received the reactive balance control assessment (REACT $n = 77$) versus those who did not (NoREACT $n = 106$) on measures of functional balance (Berg Balance Scale – BBS), disability (Functional Independence Measure (FIM-T) and mobility (Clinical Outcomes Variables Scale – COVS). Data points represent individual patient scores; rectangles represent mean scores of measures at admission and discharge.

standing or walking. Such falls may be associated with circumstances where patients act against instructions or recommended supervision/aid [1]. The absence of an assessment for inpatient fallers may be more consistent with the NoREACT clinical profile of lower overall functional, motor and cognitive ability. However, given that a history of inpatient falls predicts future falls post-discharge [31], the need to develop assessment approaches that are appropriate for those with slower rates of recovery is further underscored.

The collaboration between both clinicians and researchers, in developing and adapting the lean-and-release assessment for clinical care [19], most likely influenced its clinical uptake. Strong clinical:research partnerships can promote research utilization and foster best practices [32]. We recognize, however, that access to, and support for the administration of, the technology associated with this assessment would not be widely available in other clinical settings. However, it is important to differentiate between the equipment required to implement the assessment methodology versus additional measurement technology used. A standardized methodology for assessing reactive balance control was implemented with relatively simple modifications to a transfer gantry, commercially available at a cost equivalent to that of other commonly used therapeutic equipment (e.g. stationary bike). We added forceplate measures, to gain information related to timing of responses, but meaningful clinical information can still be attained without this technology. Patient responses including need for assistance, decreased foot clearance and attempts to step with the blocked limb have been linked to falls post-stroke [5,6]. Further, physiotherapists have advocated for the use of a harness system to safely incorporate reactive balance control assessment and training into practice [19], suggesting support for its clinical use. It is also noteworthy that inexpensive technological measures (e.g. game-based force plates, accelerometers) are being advanced [33], that can provide meaningful information in a more immediate and user-friendly format. Collectively, these factors bode well for the development of such methodologies and measures into clinically meaningful, accessible and feasible tools.

5. Conclusions

A standardized approach to reactive balance control assessment, using the lean-and-release methodology with measurement technology, was developed for clinical use within a stroke inpatient rehabilitation setting. This assessment provides a safe and controlled method of balance perturbation. Added measurement technology allows for quantification of the patient's response that may reveal underlying balance control issues masked by observation-based methods. This study demonstrated the potential for clinical uptake of such assessment methods among patients with stroke, who were progressing in their functional and mobility status over the course of inpatient rehabilitation. However, patients with lower levels of cognitive and motor status and slower to progress with functional mobility, were not routinely assessed prior to discharge. Ongoing research is required to develop clinical approaches to reactive balance control assessment that are effective, efficient, relevant to clinical populations and feasible for clinical practice.

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Conflict of interest

The authors declare they have no conflict of interest.

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