Corey A. Pew[1]
Mechanical and Industrial Engineering,
Montana State University,
Bozeman, MT 59715
e-mail: Corey.Pew@montana.edu

Soroosh Sadeh
Department of Kinesiology and Health Education,
The University of Texas at Austin,
Austin, TX 78712
e-mail: Soroosh.Sadeh@utexas.edu

Hao-Yuan Hsiao
Department of Kinesiology and Health Education,
The University of Texas at Austin,
Austin, TX 78712
e-mail: HHsiao@austin.utexas.edu

Richard R. Neptune
Walker Department of Mechanical Engineering,
The University of Texas at Austin,
Austin, TX 78712
e-mail: rneptune@mail.utexas.edu

Development of a Novel Perturbation Platform System for Balance Response Testing and Rehabilitation Interventions

Balance perturbations are often used to gain insight into reactive control strategies used to prevent falls. We developed a perturbation platform system (PPS) that can induce perturbations in both vertical and angled directions. The PPS was evaluated using human subject testing to verify its function and performance. The final system consisted of two box platforms that can individually perform vertical and angled surface perturbations. Following a perturbation, the system can automatically reset for the next iteration under the weight of the standing participant. The PPS achieves a peak downward acceleration of 4.4 m/s^2 during drop events that simulate sudden surface changes. The experimental testing revealed that the perturbation induced a peak limb loading of 280 ± 38% of body weight (BW) during vertical drops and that participants' center of mass displacements were consistent with previous balance studies evaluating vertical perturbations. The system can be used in a laboratory or clinical setting to better understand balance response and control mechanisms and assist in rehabilitation training to improve balance control and help mitigate the incidence of falls. [DOI: 10.1115/1.4056831]

Keywords: balance, biomechanics, falling, perturbation, rehabilitation, uneven ground

1 Introduction

Falling is one of the leading causes of both fatal and nonfatal injuries in the United States. In 2017, falling accounted for over 36,000 deaths and 8 million nonfatal visits to the emergency room [1,2]. Each year 30–60% of older adults experience a fall, with falling being the leading cause of death due to accidental injury, which has seen a 31% increase in incidents from 2007 to 2016 [3,4]. Individuals poststroke experience balance control deficits that lead to 73% of stroke survivors falling within the first six months after discharge [5,6]. Similarly, individuals with lower-limb amputations experience major health risks due to falling with more than half reporting falls over the previous year and 75% of those report multiple falls [7,8]. In addition to the potential for significant injuries, falling can have negative consequences related to psychological and social aspects of daily living for those who experience a fall, often limiting their physical activity and social interactions due to the fear of falling [9].

With falling being a critical clinical problem, researchers and clinicians often seek to develop assessment and intervention methods to identify those at risk of falling and help reduce the risk of future falls. An individual's postural response to an unexpected perturbation can take the form of a corrective response (nonstepping) or a protective response (stepping) to regain their balance [10]. To elicit these reactions in a laboratory or clinical setting, motor-driven cable systems [11,12] have been used to apply forces to the waist to study reactive stepping in older adults [13] and individuals poststroke [14]. Another common method used to induce balance perturbations is via movable support surfaces, which can induce a slip-like perturbation in the middle of a walkway [15] and unexpected forward and backward platform translations to study nonstepping postural reflexes in individuals poststroke [16]. Treadmill acceleration/deceleration and sideways translations have also been used to study balance control during walking in young and older adults [17]. Common limitations of these existing systems include the need for multiple individuals to operate the device, the size and complexity of the system that limits their use to laboratory settings, and single function capability.

Previous standing and walking support surface perturbation systems have focused primarily on horizontal perturbations. More recent work has utilized vertical perturbations to assess startle reactions, balance control [18], and limb loading [19]. Hsiao et al. utilized electromagnets to control a vertical standing surface drop perturbation [19]. However, a limitation of the system was that following each trial the participant was required to step off the platform while it was manually reset, and therefore reduced the feasibility of being used as a rehabilitation tool for repetitive exercise. Other studies have used a pneumatically actuated platform to facilitate weight transfer during step initiation in individuals with Parkinson's disease [20,21]. In addition to platform translations, studies have used tilted surfaces to simulate uneven terrain. A static device that could vary the surface angle up to 15 deg in inversion or eversion was designed to assess balance recovery mechanisms on uneven terrain in individuals with a lower-limb amputation [22]. This device was designed to sit on a standard in-ground force plate while the individual walked over the uneven surface on an elevated walkway. The angle of the surface was modulated manually, with two individuals being needed to manipulate the platform into the orientations of interest. These perturbation systems were designed to induce rapid changes in the body center of mass position that simulate stumbling or falling and trigger postural responses [13,18,23].

Unfortunately, current standing platform perturbation systems are often limited by: (1) the need for multiple individuals to operate the device, (2) downtime between trials to reset the platform, (3) single direction functionality, and/or (4) the ability to only test unilateral or bilateral perturbations. The purpose of this work was to design and test a novel perturbation platform system (PPS) that can perform unilateral and/or bilateral vertical or angled platform displacement perturbations with a powered reset function capable of lifting a subject's entire body weight.

2 Methods

2.1 System Design Specifications. The design of the PPS was guided by the desire to create a system capable of simulating

[1]Corresponding author.
Manuscript received December 22, 2021; final manuscript received January 26, 2023; published online April 17, 2023. Assoc. Editor: Elizabeth Hsiao-Wecksler.

various types of drop perturbations while being operated remotely by a single individual. In addition, the system was designed to reset to the start position on command while supporting the participant's body weight. To accommodate the widest range of testing scenarios, the system was designed with the following specifications:

(1) The system should provide a vertical displacement of the standing support surface up to 76 mm and provide either unilateral or bilateral perturbations. This distance was selected to trigger a rapid postural reaction for balance recovery and to allow the resulting impact force (∼ one bodyweight) to be safe for repeated exposures [18].
(2) The platform surface should be capable of tilting up to 15 deg in both directions to provide ankle inversion/eversion or plantarflexion/dorsiflexion perturbations, depending on the orientation. These angle magnitudes were chosen to provide a maximum perturbation while minimizing the risk of injury [22].
(3) The platform reset function should be able to lift a maximum body weight of 135 kg, which would accommodate 95% of individuals in the United States [24].
(4) The system size and weight should facilitate portability such that it can be easily moved for use in a laboratory or clinical setting. In addition, the system should fit over in-ground force plates to allow for kinetic measurements.

2.2 System Performance Testing. Verification of performance under weighted conditions consisted of experimental testing where five participants (two male, age = 25 ± 3 years, mass = 69.2 ± 9.5 kgs) provided informed consent to a protocol approved by The University of Texas at Austin Internal Review Board. To characterize the motion of the system and participants, a 12-camera Vicon motion capture system (Vicon, Centennial, CO) was used to track the motion of the lower limbs using a Plug-In-Gait marker set and four markers attached directly to each platform. Vertical ground reaction forces were measured using two force platforms (Bertec, Columbus, OH) located beneath each of the drop platforms and sampled at 1000 Hz. During plantarflexion/dorsiflexion and vertical drop trials the individual stood in a modified tandem stance (one foot on each drop platform) (Fig. 1 Left) with the distance between the heel of the leading limb and toe of the trailing limb was one-third of their foot length [25]. During Eversion/Inversion trials participants used a parallel stance over the two platforms (Fig. 1 Right). Five drops of each condition (vertical, plantarflexion/dorsiflexion and eversion/inversion) both unilaterally (dominant limb only) and bilaterally were performed (50 trials per participant) [26]. Vertical drops were performed at the 76 mm setting while angled drops were performed at the 10 deg setting to minimize risk of ankle injury to the participants. Perturbations were induced to the leading limb during vertical and plantarflexion/dorsiflexion, unilateral trials. Testing with participants included both the drop perturbations and resetting the platforms with individuals standing on top of the platform.

Quantification of platform performance utilized the peak downward acceleration of the platform surface during perturbations with participants, calculated as the second time derivative of the platform marker position during the drops. In addition to downward acceleration, the distance traveled by the platform, drop duration (time from start to end of movement), minimum vertical force during drop (% body weight experienced by the participant), and initiation offset time (the time difference between the start of motion for the two platforms given the drop command at the same time) were analyzed.

2.3 Postural Response Testing. To verify the PPS's ability to elicit responses consistent with falling, the participants' center of mass (COM) translation during the various perturbation conditions was analyzed [27]. In addition, the maximum vertical limb loading force was calculated and normalized to body weight [19,20].

2.4 Data Processing. The derived downward platform acceleration signals were smoothed using a 6 Hz, lowpass, Butterworth filter for each drop averaged across drop trials. Vertical ground reaction forces were normalized to each participant's body weight (BW). The body COM displacement was calculated in Visual3D software (C-Motion, Germantown, MD). Maximal body COM displacements from onset to end of movement were calculated in the vertical, mediolateral, and anteroposterior directions. Comparisons of performance between the two individual platform units utilized a one-way ANOVA to determine if performance varied by platform conditions. In addition, interaction effects between participant mass and drop time were investigated to determine if platform performance varied with changes in user mass. All statistical testing was performed using the statistical toolbox in MATLAB (MathWorks, Natick, MA).

3 Results

3.1 System Design. The resulting PPS design consists of three main components: (1) two movable platform surfaces, (2) a high-pressure air source, and (3) a remote-control interface (Fig. 2). The operation of each platform utilizes four, double acting pneumatic pistons (Space Saver Low Profile, SS-150, Mead

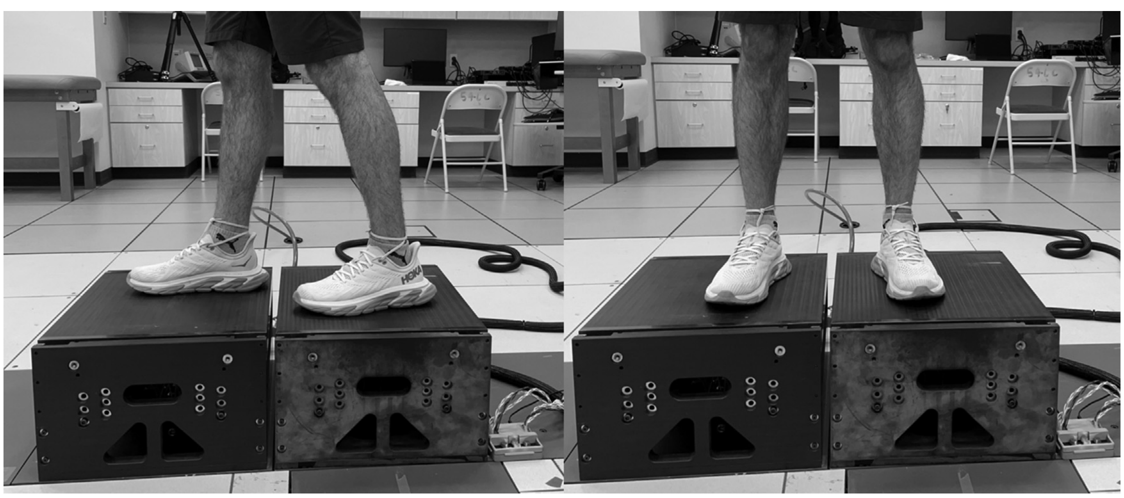

Fig. 1 Modified tandem stance (left) and even stance (right) used in postural testing

Fig. 2 Main components of the perturbation platform system: two platform units and a tethered, hand-held, remote-control unit (front). High pressure air source located in adjacent room with air line under floor.

Fluid Dynamics, Chicago, IL) (Fig. 3*A*). The system is controlled via a 5/3 directional air valve (IMI Norgren, K81EA00KV0KV02W1, Littleton, CO) (Fig. 3*B*). The valve is controlled with two 12-volt solenoids that can be activated or deactivated to raise and lower the pistons. The use of the 5/3 valve allows for directional control of the pistons along with a third mode where airflow is closed, which is activated when no electrical signal is applied. This provides a safety feature in the event of power loss that will lock the pistons in place without dropping the subject unexpectedly. High-pressure air is provided from an external air compressor (Fig. 2) and is conveyed to each platform via a single inlet line (Fig. 3*C*). Currently, the compressor source is in a separate room with the air line routed through the floor to provide quiet and discreet operation of the platforms. Air is supplied and exhausted from the pistons evenly via distribution blocks (Fig. 3*D*). An exhaust muffler is also used to quiet exiting air to prevent startling the participants or alerting them to the timing of upcoming perturbations (Fig. 3*E*). Lastly, each piston is fitted with a one-way variable flow control valve at the inlet/outlet to allow for speed control of the piston drop and return.

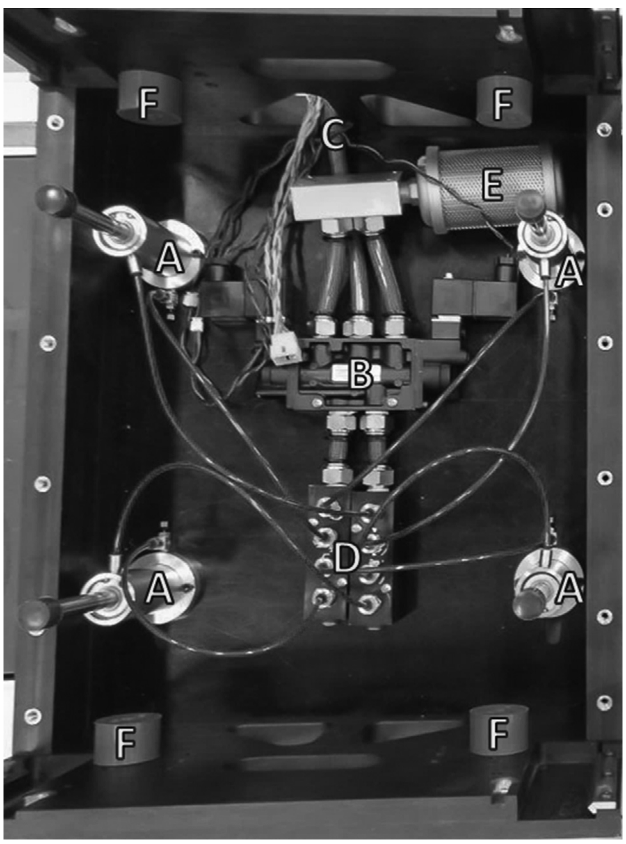

Fig. 3 Internal view of a single platform. (A) Four pneumatic pistons, (B) 5/3 control valve, (C) high-pressure inlet line, (D) distribution blocks, (E) exhaust muffler, and (F) bump stops to arrest the platform at specific drop heights and angles.

The linear movement of the platform is guided up and down on four V-Groove track roller and rail elements (Fig. 4*A*, *B*, respectively). Movement of the platform (linear or angled) is controlled via four, 12-volt, sealed linear solenoids (Magnet Shultz of America, S-07791, Westmont, IL) (Fig. 4*C*). The linear solenoids are threaded into brass inserts (Fig. 4*D*) that are pressed into the moving platform and contain a hardened steel pin. When a 12-volt signal is applied, the pin is retracted into the solenoid. When the signal is removed, an internal spring pushes the pin outward to mate with a corresponding brass insert (Fig. 4*E*) in the end plates of the platform box. The solenoids and pins allow for the selection of different drop types (vertical versus angle drop). When all pins are retracted, the plate is guided by the track rollers and drops down linearly. When a single, coaxial pair of pins is engaged, the platform rotates around that axis to enable angled drops in both directions (i.e., plantarflexion/dorsiflexion or inversion/eversion). During angled drops, the track rollers lose contact, and the motion of the platform is guided solely by the rotation about the pins. To enable different levels of vertical and angle drops, adjustable stops are used to provide an end position for the platform after a drop (Fig. 3*F*).

To address the specific design specifications, the following points are considered:

(1) To facilitate different vertical drops, the vertical displacement is determined by six different positions (three for vertical drops and three for angle drops) via predrilled holes in

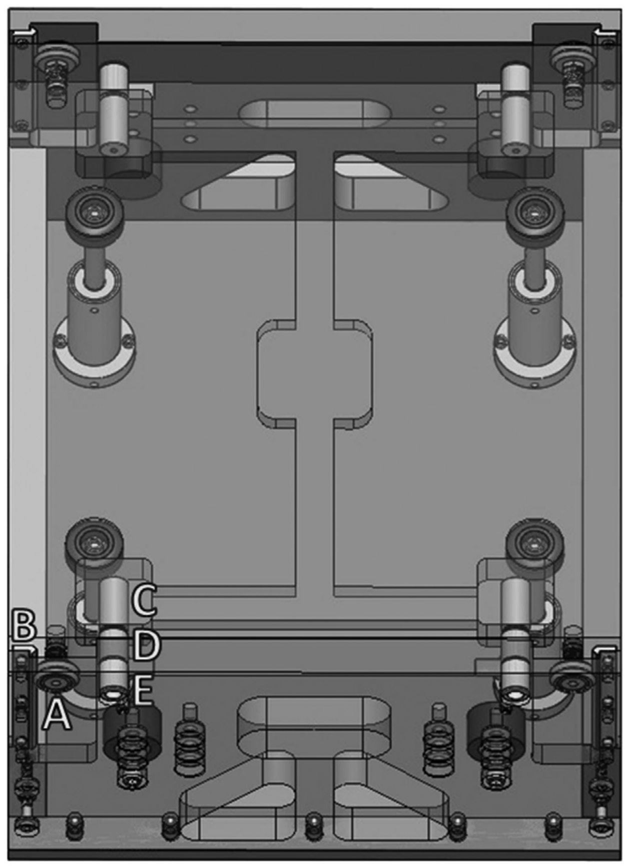

Fig. 4 Perturbation Platform System design drawing with platform motion control components revealed. Individual components (same for all four corners): (A) V-groove track roller, (B) V-groove rail, (C) sealed linear solenoid, (D) platform brass insert for solenoid mount and pin support, and (E) end plate brass insert for pin support.

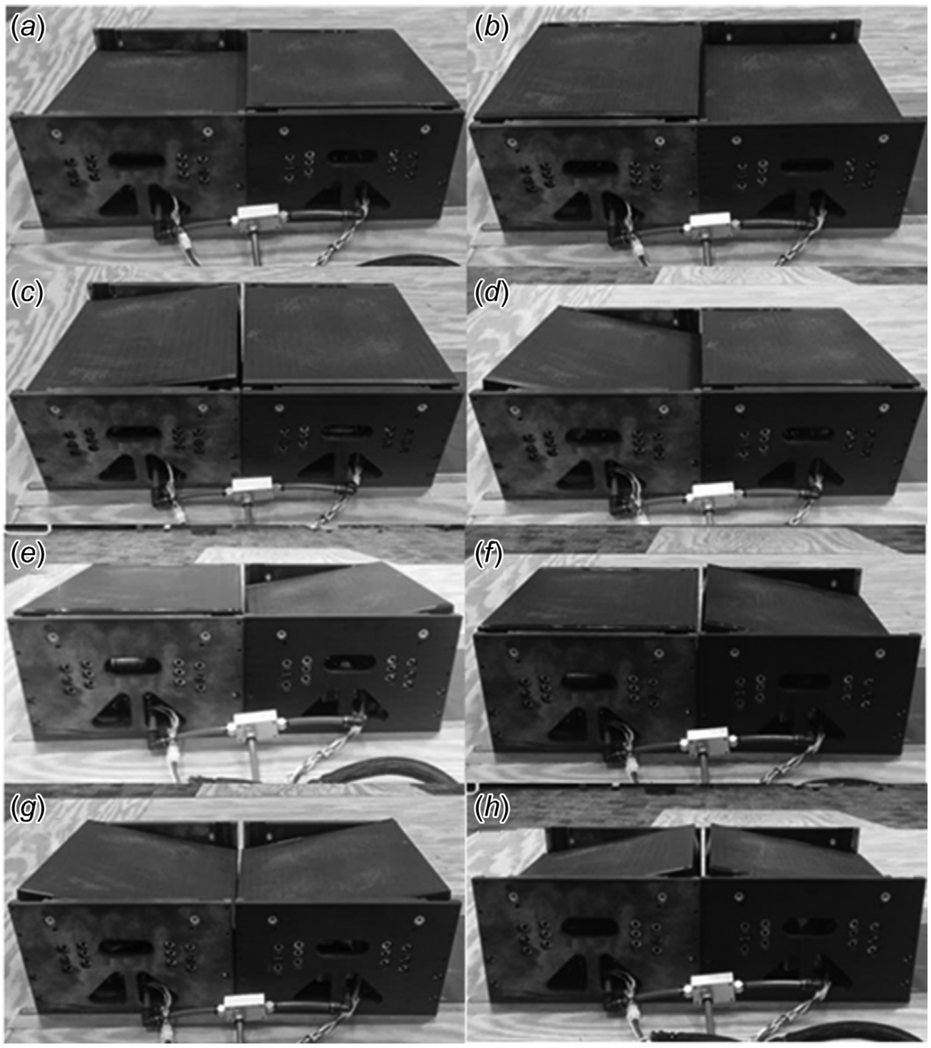

Fig. 5 Different configurations available. (*a*) Vertical drop left, (*b*) vertical drop right, (*c*) inversion left, (*d*) eversion left, (*e*) eversion right, (*f*) inversion right, (*g*) eversion bilateral, and (*h*) inversion bilateral. Not pictured but available are bilateral up, and vertical drop bilateral.

the endplates of the platform. The outermost holes facilitate 25.4 mm, 50.8 mm, and 76.2 mm vertical drop heights.

(2) To facilitate angle drops, the inner set of holes allows for 5 deg, 10 deg, and 15 deg tilt angles.
(3) To reset the system, each piston has the capability to lift 80 kg at a working pressure of 6.9 Bar. With four pistons per platform, each platform can lift a 320 kg person or 640 kg when the two systems are used together (i.e., bilateral perturbations). During all conditions the platforms successfully reset to their top position with the participants on top of the platform.
(4) To facilitate portability and reduce system weight, the main components of the system are constructed from ABS plastic with each platform unit weighing 25 kg, measuring 406 mm wide × 508 mm long × 236 mm tall, and include handles for ease of transport (Fig. 2). The system is portable, with each component being able to be carried by a single individual or all components together on a dolly and could be transported in the trunk of a vehicle or stored in a closet when not in use. Final implementation in the laboratory setting includes in-ground force plates directly beneath each platform. An elevated surface surrounding the system may be used to facilitate over ground walking trials as well as helping to alleviate potential anxiety due to the sensation of height when standing on the platforms.

To enable functionality by a single operator, a control unit utilizing a Teensy 3.2 microcontroller (PRJC Sherwood, OR) (Fig. 2) was developed. The controller is hard wired to the two platforms with connectors at each end to allow for ease of assembly and disassembly. The 4-meter lead allows for remote operation of the system by a single operator from a central location in the laboratory. The controller consists of four inputs. The two on either side allow the drop mode of each platform to be set individually and includes LED indicators to specify the current mode of operation (vertical, inversion/plantarflexion, eversion/dorsiflexion, or no drop). The two vertical buttons in the center initiate the drop or return for both platforms simultaneously for simple activation of each drop condition. Varying modes of operation can be accomplished with the two platforms combined (Fig. 5), which allows a wide range of unilateral and bilateral testing scenarios.

3.2 System Function Testing. Testing of the various configurations showed in all drop conditions, the platforms were able to achieve acceptable final orientations and then return to the initial starting position under the participant's body weight. Both platforms were tested and compared to each other. Platform A represents the unit in the forward position for vertical and dorsiflexion/plantarflexion drops and on the right for inversion/eversion drops. Drop distance and angle met or exceeded expectation at

Table 1 Performance of the two platforms. Platform A represents the front/right system shown in Fig. 1.

	Vertical drop data			Stats			
	Platform A		Platform B	Platform A Univs Bi		Platform A versus B	
	Unilateral	Bilateral	Bilateral	Difference	*p* value	Difference	pvalue
Drop distance (mm)	82.2 ± 2.6	79.9 ± 2.3	79.3 ± 0.3	0.2 ± 0.0	**0.002**	0.6 ± 0.2	0.440
Peak acceleration (m/s^2)	−4.0 ± 1.7	−4.4 ± 1.2	−4.3 ± 1.7	0.3 ± 0.4	0.467	0.1 ± 0.8	0.885
Drop time (s)	0.36 ± 0.08	0.32 ± 0.07	0.34 ± 0.08	0.03 ± 0.02	0.143	0.02 ± 0.02	0.421
Minimum force (%BW)	1.6 ± 3.2	5.5 ± 5.9	6.2 ± 5.9	3.6 ± 1.0	**< 0.001**	0.7 ± 9.0	0.696
Maximum force (%BW)	211 ± 53	261 ± 62	280 ± 38	52.1 ± 14.8	**0.001**	19.2 ± 50.1	0.221
Initiation offset time (s)	0.02 ± 0.01						
	Angle drop data			Stats			
	Platform A		Platform B	Platform A Univs Bi		Platform A versus B	
	Unilateral	Bilateral	Bilateral	Difference	*p* value	Difference	*p* value
Drop distance (deg)	10.4 ± 0.6	10.4 ± 0.3	9.9 ± 0.2	0.01 ± 0.06	0.916	0.5 ± 0.4	**< 0.001**
Peak acceleration (deg/s^2)	−9.5 ± 12.7	−9.1 ± 6.4	−7.1 ± 4.6	0.36 ± 1.4	0.802	2.0 ± 8.7	**0.012**
Drop time (s)	0.34 ± 0.04	0.32 ± 0.02	0.37 ± 0.03	0.01 ± 0.0	**0.024**	0.05 ± 0.04	**< 0.001**
Minimum force (%BW)	1.7 ± 3.1	5.3 ± 5.6	5.6 ± 5.9	3.4 ± 0.5	**< 0.001**	0.3 ± 7.5	0.678
Maximum force (%BW)	152 ± 29	157 ± 25	167 ± 26	5.1 ± 3.8	0.177	10.6 ± 25.5	**0.004**
Initiation offset time (s)	0.04 ± 0.03						

Note: Bold values indicate significant differences, $p < 0.05$.

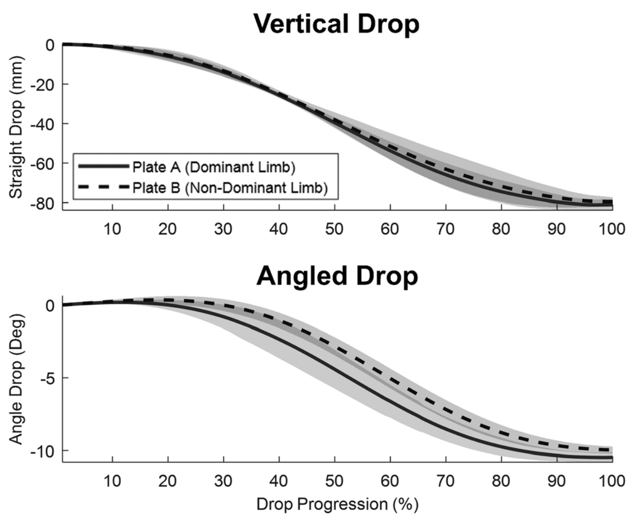

Fig. 6 Drop distance (mm, top) and angle (degrees, bottom) for plate A (dominant limb and unilateral and bilateral drops) and plate B (non-dominant limb and bilateral drops only)

79.9 ± 2.3 mm and 79.3 ± 0.3 mm for bilateral vertical drops for platforms A and B, respectively and angled drops at 10.4 ± 0.3 and 9.9 ± 0.2 deg for platforms A and B, respectively (Table 1 and Fig. 6). Peak acceleration reached −4.4 ± 1.2 m/s^2 (downward). Peak values for other measures included 0.36 ± 0.08 s for drop time, 6.2 ± 5.9% BW for minimum force, and 0.04 ± 0.01 s for initiation offset. No significant interaction was found between the drop time of each platform and the mass of the participants ($p = 0.595$).

3.3 Postural Response Testing. Vertical COM displacements were observed in all conditions (Table 2 and Fig. 7, Top). The greatest vertical COM displacement (−85.5 ± 5.4 mm) occurred during bilateral Vertical drop perturbations. Participants showed anterior COM displacements in response to unilateral and bilateral Dorsiflexion/Plantarflexion perturbations and during unilateral Vertical drop perturbation (Table 2 and Fig. 7, middle). Lateral COM displacements toward the perturbed limb side were observed during unilateral Inversion (29.9 ± 7.4 mm) and Eversion (9.6 ± 8.7 mm) (Table 2 and Fig. 7, Bottom). Peak limb loading was 280 ± 38% body weight during Vertical drops and 167 ± 26% body weight during angle drops (Table 1 and Fig. 8).

4 Discussion

We designed and tested a novel Perturbation Platform System to perform multiple perturbation types (vertical or angled, unilateral or bilateral) that can automatically reset for the next perturbation while supporting the participant's body weight. Results from testing the Vertical drop of the system indicate that the platform acceleration was similar to previous perturbation devices [18,20]. In addition, the PPS induced body COM displacements that were comparable to previous studies [20,23,27,28].

4.1 System Design. The physical design of the system was able to meet or exceed the specified objectives in constructing a functional and portable perturbation system. In reference to the design specifications the PPS has various position settings to accommodate numerous combinations of vertical/angled and unilateral/bilateral perturbations (specifications 1 and 2), during system reset each platform can lift 320 kg or 640 kg when the two systems are used together (specification 3), and the system is able to be moved, setup, and broken down by a single individual in a laboratory or clinical setting (specification 4).

4.2 System Performance Testing. While the PPS was unable to reach full free-fall acceleration (-9.81 m/s^2), the acceleration was consistent with other devices. Sanders et al. (2019) achieved an acceleration of −4.9 m/s^2, which is similar to our peak value of −4.4 m/s^2 [18]. Rogers et al. (2011) displaced 15 mm in 0.1 s, while we achieved approximately 80 mm of displacement in 0.33 s [20]. Greater than the expected 76 mm Vertical Drop was due to rubber coated end stops that compressed when loaded. In addition, we show that the minimum force experienced on the participants' feet was between 5 and 6% of their body weight on average, with 55% of trials achieving 0% body weight during the drop, indicating that near free-fall was experienced briefly during the drop. For vertical drops we found no significant differences between the performance of the two platforms, and while statistically significant performance was noted during angle drops, the magnitude of those differences (0.5 deg angle, 0.05 s time) are not

Table 2 Center of mass displacement values are mean ± SD (mm). A/P is anterior/posterior motion, positive values forward; M/L is medial/lateral motion, positive toward dropped foot, Vert is vertical, and negative is down.

	Unilateral			Bilateral		
	A/P	M/L	Vert	A/P	M/L	Vert
Vertical	47.1 ± 16.6	1.0 ± 3.2	−52.8 ± 9.0	−0.2 ± 4.4	0.8 ± 2.9	−85.5 ± 5.4
Dorsiflexion	24.6 ± 7.7	−0.1 ± 3.3	−41.1 ± 5.3	14.6 ± 5.1	0.4 ± 3.6	−68.1 ± 3.9
Plantarflexion	34.6 ± 14.1	0.6 ± 3.5	−28.0 ± 5.6	20.8 ± 4.7	−0.4 ± 1.8	−59.0 ± 5.0
Inversion	6.5 ± 8.3	29.9 ± 7.4	−32.5 ± 4.0	0.8 ± 4.8	−1.5 ± 3.0	−56.3 ± 7.8
Eversion	2.3 ± 8.6	9.6 ± 8.7	−40.7 ± 6.3	0.4 ± 5.0	0.6 ± 2.9	−64.6 ± 4.1

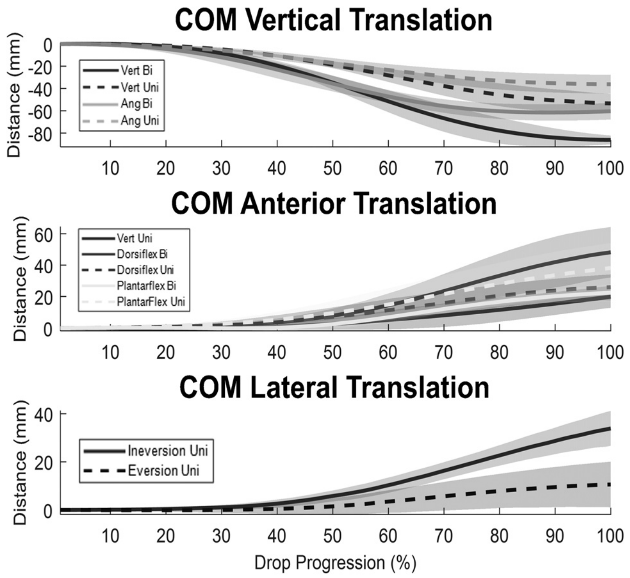

Fig. 7 Center of mass (COM) translation. Top: vertical (downward) COM translation during vertical (vert) and angled (ang) during both bilateral (bi) and unilateral (uni) trials. Middle: anterior (forward) translation during vert, dorsiflexion, and plantarflexion trials. Bottom: lateral (toward dropped side) during inversion and eversion trials.

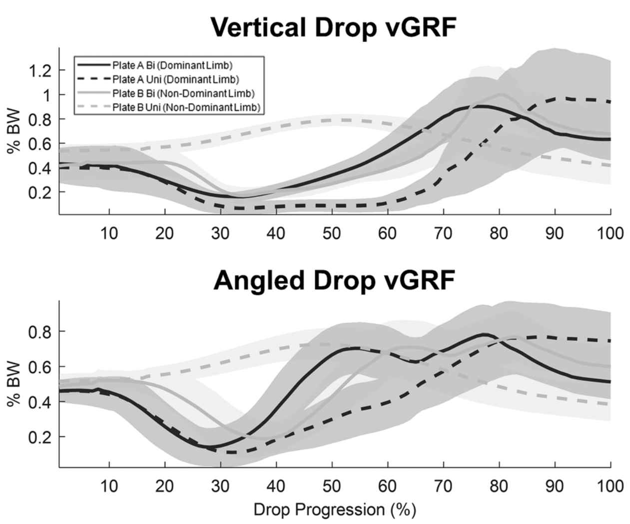

Fig. 8 Vertical ground reaction force (vGRF) during vertical (top) and angled (bottom) drops. Plate A (dominant limb) and plate B (non-dominant limb) during unilateral (uni) and bilateral (Bi) trials. For the unilateral trials, only plate A was dropped.

expected to be functionally noticeable by the participants. Angled drops achieved expected drop angles of 10 deg of platform motion. This indicates that our device will be able to produce comparable or improved drop perturbations in comparison to previously described systems. In addition, the platforms functioned similarly between participants.

4.3 Postural Response Testing. Following the perturbation onset, participants showed similar postural reactions compared to previous balance perturbation studies. Vertical COM displacements during all conditions as well as anterior/posterior displacements during vertical and dorsiflexion/plantarflexion unilateral drops, and Medial/Lateral displacements during unilateral inversion drops were similar to Schmid et al. (2011) who applied mediolateral support surface oscillations during standing [27]. In addition, participants demonstrated peak vertical forces greater than one bodyweight. This amplitude is similar to those previously reported by Rogers et al. (2011) who were seeking to improve weight transfer in persons with Parkinson's Disease [20] and Hsiao et al. who applied vertical standing support surface perturbations to characterize biomechanical limb loading abnormalities in individuals poststroke [19]. These results confirm that the PPS system can provide perturbations that challenge balance control and weight-bearing without inducing excessive forces that could potentially lead to injuries.

4.4 Clinical Implementation. The PPS is designed for both laboratory and clinical settings. Similar perturbation testing has been used to understand the physical and neurological implications of advanced age, stroke, lower limb amputation, and Parkinson's disease on the response to sudden, unexpected perturbations [18,20,22]. The PPS can be used to induce repetitive exposures to perturbations that trigger neuromuscular reactions that are important for balance control, and therefore has the potential to be used as a rehabilitation intervention [4,20,29] and potentially reduce fall risk in the elderly [30,31] and other clinical populations. We plan future studies using the device to identify neuromuscular and biomechanical abnormalities during balance perturbations in older adults and will assess adaptive changes in balance control following bouts of perturbation exercises.

One limitation of the current system is the use of a hard-wired controller. However, the Teensy microcontroller is capable of Bluetooth wireless communication. In future implementations of the device, versatility and usability can be improved by making the control system wireless. In addition, control can be integrated into an existing gait lab computer system to allow activation of the platforms in unison with other lab components such as motion capture and/or force plates. This would facilitate more advanced testing conditions that synchronize the system automatically with specific kinematic or kinetic events.

5 Conclusions

This work outlined the design and testing of a novel Perturbation Platform System that allows for unilateral or bilateral surface perturbations during standing or walking in a laboratory or clinical setting. The system allows for both vertical and angle drop functions and can automatically reset to the original position under the load of the participant's body weight. The system can be used in a laboratory or clinical setting to better understand balance response and control mechanisms when falling and assist in rehabilitation training to help mitigate the incidence of falls.

Funding Data

- This work was funded by a University of Texas at Austin College of Education Small Grant Award (Funder ID: 10.13039/100008562).

Data Availability Statement

The datasets generated and supporting the findings of this article are obtainable from the corresponding author upon reasonable request.

References

[1] Centers for Disease Control and Prevention, 2017, "10 Leading Causes of Injury Deaths by Age Group Highlighting Unintentional Injury Deaths, United States – 2017," Centers for Disease Control and Prevention, Atlanta, GA, accessed Feb. 7, 2023, https://www.cdc.gov/injury/wisqars/pdf/leading_ causes_of_injury_deaths_highlighting_unintentional_2017-508.pdf

[2] Centers for Disease Control and Prevention, 2017, "National Estimates of the 10 Leading Causes of Nonfatal Injuries Treated in Hospital Emergency Departments, United States – 2017," Centers for Disease Control and Prevention, Atlanta, GA, accessed Feb. 7, 2023, https://www.cdc.gov/injury/wisqars/pdf/leading_causes_of_nonfatal_injury_2017-508.pdf

[3] Burns, E., and Kakara, R., 2018, "Deaths From Falls Among Persons Aged ≥65 Years — United States, 2007–2016," Morbidity and Mortality Weekly Report, **67**(18), pp. 509–514.

[4] Rubenstein, L. Z., 2006, "Falls in Older People: Epidemiology, Risk Factors and Strategies for Prevention," Age Ageing, **35**(Suppl 2), pp. ii37–ii41.

[5] Forster, A., and Young, J., 1995, "Incidence and Consequences of Falls Due to Stroke: A Systematic Inquiry," BMJ, **311**(6997), pp. 83–86.

[6] Ashburn, A., Hyndman, D., Pickering, R., Yardley, L., and Harris, S., 2008, "Predicting People With Stroke at Risk of Falls," Age Ageing, **37**(3), pp. 270–276.

[7] Miller, W. C., Deathe, A. B., Speechley, M., and Koval, J., 2001, "The Influence of Falling, Fear of Falling, and Balance Confidence on Prosthetic Mobility and Social Activity Among Individuals With a Lower Extremity Amputation," Arch. Phys. Med. Rehabil., **82**(9), pp. 1238–1244.

[8] Miller, W. C., Speechley, M., and Deathe, B., 2001, "The Prevalence and Risk Factors of Falling and Fear of Falling Among Lower Extremity Amputees," Arch. Phys. Med. Rehabil., **82**(8), pp. 1031–1037.

[9] Kempen, G. I., van Haastregt, J. C., McKee, K. J., Delbaere, K., and Zijlstra, G. R., 2009, "Socio-Demographic, Health-Related and Psychosocial Correlates of Fear of Falling and Avoidance of Activity in Community-Living Older Persons Who Avoid Activity Due to Fear of Falling," BMC Public Health, **9**(1), pp. 1–7.

[10] Rogers, M. W., and Mille, M.-L., 2018, "Balance Perturbations," *Handbook of Clinical Neurology*, Elsevier, New York, Chap. 5.

[11] Singer, J. C., Prentice, S. D., and McIlroy, W. E., 2016, "Age-Related Challenges in Reactive Control of Mediolateral Stability During Compensatory Stepping: A Focus on the Dynamics of Restabilisation," J. Biomech., **49**(5), pp. 749–755.

[12] Pidcoe, P. E., and Rogers, M. W., 1998, "A Closed-Loop Stepper Motor Waist-Pull System for Inducing Protective Stepping in Humans," J. Biomech., **31**(4), pp. 377–381.

[13] Andrew Sawers, X., Pai, Y.-C., Bhatt, T., and Ting, L. H., 2017, "Neuromuscular Responses Differ Between Slip-Induced Falls and Recoveries in Older Adults," J. Neurophysiol., **117**(2), pp. 509–522.

[14] Gray, V. L., Yang, C. L., Fujimoto, M., McCombe Waller, S., and Rogers, M. W., 2019, "Stepping Characteristics During Externally Induced Lateral Reactive and Voluntary Steps in Chronic Stroke," Gait Posture, **71**, pp. 198–204.

[15] Borrelli, J., Creath, R. A., Pizac, D., Hsiao, H., Sanders, O. P., and Rogers, M. W., 2019, "Perturbation-Evoked Lateral Steps in Older Adults: Why Take Two Steps When One Will Do?," Clin. Biomech., **63**, pp. 41–47.

[16] Marigold, D. S., Eng, J. J., and Timothy Inglis, J., 2004, "Modulation of Ankle Muscle Postural Reflexes in Stroke: Influence of Weight-Bearing Load," Clin. Neurophysiol., **115**(12), pp. 2789–2797.

[17] Roeles, S., Rowe, P. J., Bruijn, S. M., Childs, C. R., Tarfali, G. D., Steenbrink, F., and Pijnappels, M., 2018, "Gait Stability in Response to Platform, Belt, and Sensory Perturbations in Young and Older Adults," Med. Biol. Eng. Comput., **56**(12), pp. 2325–2335.

[18] Sanders, O., Hsiao, H. Y., Savin, D. N., Creath, R. A., and Rogers, M. W., 2019, "Aging Changes in Protective Balance and Startle Responses to Sudden Drop Perturbations," J. Neurophysiol., **122**(1), pp. 39–50.

[19] Hsiao, H. Y., Gray, V. L., Borrelli, J., and Rogers, M. W., 2020, "Biomechanical Control of Paretic Lower Limb During Imposed Weight Transfer in Individuals Post-Stroke," J. Neuroeng. Rehabil., **17**(1), p. 140.

[20] Rogers, M. W., Hilliard, M. J., Martinez, K. M., Zhang, Y., Simuni, T., and Mille, M. L., 2011, "Perturbations of Ground Support Alter Posture and Locomotion Coupling During Step Initiation in Parkinson's Disease," Exp. Brain Res., **208**(4), pp. 557–567.

[21] Creath, R. A., Prettyman, M., Shulman, L., Hilliard, M., Martinez, K., MacKinnon, C. D., Mille, M. L., Simuni, T., Zhang, J., and Rogers, M. W., 2013, "Self-Triggered Assistive Stimulus Training Improves Step Initiation in Persons With Parkinson's Disease," J. Neuroeng. Rehabil., **10**(1), pp. 1–10.

[22] Yeates, K. H., Segal, A. D., Neptune, R. R., and Klute, G. K., 2016, "Balance and Recovery on Coronally-Uneven and Unpredictable Terrain," J. Biomech., **49**(13), pp. 2734–2740.

[23] Jonsson, E., Henriksson, M., and Hirschfeld, H., 2007, "Age-Related Differences in Postural Adjustments in Connection With Different Tasks Involving Weight Transfer While Standing," Gait Posture, **26**(4), pp. 508–515.

[24] Fryar, C. D., Kruszon-Moran, D., Gu, Q., and Ogden, C. L., 1999, "National Health Statistics Reports," NHSR No. 122, Dec. 20, 2018.

[25] Gillette, J. C., and Abbas, J. J., 2003, "Foot Placement Alters the Mechanisms of Postural Control While Standing and Reaching," IEEE Trans. Neural Syst. Rehabil. Eng., **11**(4), pp. 377–385.

[26] Holcomb, A. E., Hunt, N. L., Ivy, A. K., Cormier, A. G., Brown, T. N., and Fitzpatrick, C. K., 2022, "Musculoskeletal Adaptation of Young and Older Adults in Response to Challenging Surface Conditions," J. Biomech., **144**, p. 111270.

[27] Schmid, M., Bottaro, A., Sozzi, S., and Schieppati, M., 2011, "Adaptation to Continuous Perturbation of Balance: Progressive Reduction of Postural Muscle Activity With Invariant or Increasing Oscillations of the Center of Mass Depending on Perturbation Frequency and Vision Conditions," Hum. Mov. Sci., **30**(2), pp. 262–278.

[28] Nakazawa, K., Kawashima, N., Akai, M., and Yano, H., 2004, "On the Reflex Coactivation of Ankle Flexor and Extensor Muscles Induced by a Sudden Drop of Support Surface During Walking in Humans," J. Appl. Physiol., **96**(2), pp. 604–611.

[29] Vistamehr, A., Kautz, S. A., Bowden, M. G., and Neptune, R. R., 2019, "The Influence of Locomotor Training on Dynamic Balance During Steady-State Walking Post-Stroke," J. Biomech., **89**, pp. 21–27.

[30] Pai, Y.-C., and Bhatt, T. S., 2007, "Repeated-Slip Training: An Emerging Paradigm for Prevention of Slip-Related Falls Among Older Adults," Phys. Ther., **87**(11), pp. 1478–1491.

[31] Shimada, H., Obuchi, S., Furuna, T., and Suzuki, T., 2004, "New Intervention Program for Preventing Falls Among Frail Elderly People: The Effects of Perturbed Walking Exercise Using a Bilateral Separated Treadmill," Phys. Med. Rehabil., **83**(7), pp. 493–499.