MethodsX 9 (2022) 101702

Contents lists available at ScienceDirect

MethodsX

journal homepage: www.elsevier.com/locate/mex

Method Article

Test-retest reliability of the FALL FIT system for assessing and training protective arm reactions in response to a forward fall

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ABSTRACT

The use of the hands and arms is an important protective mechanism in avoiding fall-related injury. The aim of this study was to evaluate the test-retest reliability of fall dynamics and evokd protective arm response kinematics and kinetics in forward falls simulated using the FALL simulator For Injury prevention Training and assessment system (FALL FIT). Fall FIT allows experimental control of the fall height and acceleration of the body during a forward fall. Two falls were simulated starting from 4 initial lean angles in Experiment 1 and with 4 different fall accelerations in Experiment 2. Fourteen younger adults (25.1±3.5 years) and 13 older adults (71.3±3.7 years) participated in Experiment 1 and 13 younger adults (31.8±5.7 years) participated in Experiment 2. Intraclass correlation coefficients (ICC) were used to the evaluate absolute agreement of single measures at each condition and averages across conditions. Average measures of fall dynamics and evoked kinematics and kinetics exhibited excellent reliability (ICC(A,4)>0.86). The reliability of single measures (ICC(A,1) > 0.59) was good to excellent, although 18% of single measures had a reliability (ICC(A,1)) between 0.00 and 0.57. The FALL FIT was shown to have good to excellent reliability for most measures. FALL FIT can produce a wide range of fall dynamics through modulation of initial lean angle and body acceleration. Additionally, the range of fall velocities and evoked kinematics and kinetics are consistent with previous fall research.

- The FALL FIT can be used to gain further insight into the control of protective arm reactions and may provide a therapeutic tool to assess and train protective arm reactions.

DOI of original article: 10.1016/j.humov.2021.102914

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https://doi.org/10.1016/j.mex.2022.101702

A R T I C L E I N F O
Method name: FALL simulator For Injury prevention Training and assessment
Keywords: Upper extremity, Falls, Injury
Article history: Received 3 January 2022; Accepted 9 April 2022; Available online 16 April 2022

Specifications table

Subject Area:	Medicine and Dentistry
More specific subject area:	Rehabilitation
Method name:	FALL simulator For Injury prevention Training and assessment
Name and reference of original method:	*n/a*
Resource availability:	*n/a*

Method details

Introduction

Falls have been a leading cause of morbidity and mortality in the past three decades [1]. Falls continue to be a common cause of traumatic brain injuries [2,3] and nearly all hip fractures [4–6]. Although falls infrequently cause injury [7], they result in more frequent injuries and death in adults over 65 years old [8,9]. In response to an unavoidable fall, protective arm reactions are thought to reduce the likelihood of injury [10–13]. Moreover, there is a decline in the effectiveness of protective arm reactions with older age. For example, although older adults in long-term care have been shown to orient their hands in advance of ground impact [14–16], these efforts often prove ineffective in preventing head and/or hip injuries [14,16].

Between 19 and 48% of falls among older adults are initially in the forward direction [15,17–19]. As many as one-third of falls initially directed forwards result in injury [19] with head impact occurring more frequently [15]. Some of the earliest investigations of protective arm reactions involved self-induced forward falls where participants were instructed to fall, at an instant of their choosing, with the body held straight and the arms extended [20,21]. More recent studies of forward falls have used an externally released cable restraint that initially supported participants in a forward leaning position, allowing for an unpredictable onset of perturbation [22–30].

In the previously mentioned studies of forward falls, fall characteristics were constrained through instructions given to the participants or through some experimental control (for example, see [29]) to initially fall forward and have a forward landing configuration. However, in actual falls or unconstrained forward falls in a laboratory, roughly one half of falls resulted in a forward directed landing configuration and one half resulted in lateral or backward landing configurations [15]. Additionally, impact velocity has been shown to vary despite similar initial fall conditions (i.e. height, perturbation direction) in unconstrained lateral [12] and posterior [11] falls. Prior to impact, individuals frequently attempt to recover balance by trying to grasp something, stepping, or using an upper limb reaction [17,19], which may modulate the fall acceleration and associated impact velocity [31,32].

Experiments investigating protective arm reactions generally fall into two categories, where the fall is constrained (e.g. [20–27,30]) and where there are no (or few) constraints (e.g. [11,12,14–17,19,33]). Although constraints on the experiments investigating forward falls are a limitation, the variability of fall characteristics (fall direction, fall velocity, etc.) and evoked protective reaction vary more within and between participants in experiments without constraints compared to experiments with constraints relatively speaking. The variability in fall characteristics and evoked reactions may make it difficult to isolate the effect of protective arm reactions from the whole-body reaction. This may limit the utility of this experimental paradigm in investigating and/or training specific components of protective biomechanics resulting from differences in fall characteristics. On the other hand, studies with constraints often use verbal instructions to limit the contribution of the lower extremities during the fall and subsequent impact. However, when participants are given instructions constraining the fall characteristics, e.g. participants instructed to fall with their body straight [20,21], it is not clear how much ankle or hip strategies alter impact velocity. Additionally, studies with constraints control the impact velocity through fall height, with falls from greater heights resulting in greater fall velocities at impact. This may allow for preplanning and precludes investigation of varying levels of impact velocity modulation, which may occur due balance recovery efforts prior to impact, at constant fall height.

Developing a means of controlling fall acceleration and associated velocity at impact in constrained forward fall experimental paradigms will add a degree of ecological validity by making the fall velocity unpredictable, allow examination of the effect of modulation of impact velocity on protective arm reactions, isolate arm reactions, and potentially allow assessment and training of protective arm reactions. In response to the foregoing limitations, we have developed the FALL simulator For Injury prevention Training and assessment system (FALL FIT) to isolate protective arm reactions and allow the fall velocity to be controlled and unpredictable. Accordingly, the aim of this study was to determine the reliability of the FALL FIT dynamics in eliciting falls and of the biomechanical characteristics of evoked protective arm reactions.

Methods

Device

The FALL FIT (Figure 1) was designed using Solidworks (Concord, MA). Parts and raw materials were sourced from McMaster-Carr Supply Company (McMaster-Carr, Elmhurst, IL) and machining of raw materials was carried out by the Micro-Fabrication, Machining, and Electronics Technical Service Center (Baltimore, MD). The FALL FIT simulates a forward fall with a controlled acceleration (Figure 1A and 1B). Participants are secured to the support platform via a belt around their torso. Fall velocity at impact is modulated by varying the initial lean angle (period of time when the body is accelerated before impact) and/or the counterweight load (controls the acceleration of the body and support platform) with smaller initial lean angles and counterweight loads resulting in smaller velocity at impact.

The initial lean angle is controlled by varying the distance between an electromagnet (MagneTool Inc., Troy, MI) and the anchor point (Figure 1C). A counterweight system, similar to what has been used previously in studies of vertical drop landings [34–37], is used to control acceleration of the fall. The angular acceleration can be reduced by increasing the load of the counterweight (Figure 1D).

Protocol

Two separate experiments were performed. The first experiment investigated the reliability of the tests with varying initial lean angle and the second examined the reliability of tests with varying counterweight loads. In each experiment, the orientation of the participant's body and instructions given to the participants were the same. Participants began each trial secured to the support platform by a belt at the trunk (Figure 1A). The ankles were maintained in a relaxed position and participants were instructed to grasp the top lateral edge of the support platform such that rapid orientation of the hands/arms was required prior to impact in order to absorb the energy of the fall. The shoulders were approximately 12 inches distal to the end of the pendulum resulting in each shoulder being approximately centered over each gymnastic mat after falling. The support platform was manually released by the investigator at an unpredictable time. Participants were given minimal instruction regarding performance of the protective arm reaction. Participants were instructed to “land on your hands as softly as possible.” Each hand landed on a separate gymnastic mat (North Carolina Gym Supply, Hillsborough, NC) with each mat placed over a force platform (AMTI, Watertown, MA). Each mat was 30 cm thick with the bottom 20 cm being comprised of foam with density of 23 kg/m^3 and the top 10 cm being comprised of foam with density of 29 kg/m^3. The indentation load deflection rating was 45 and 34 for the bottom and top portions of foam respectively.

Participants experienced 4 conditions in Experiment 1 and in Experiment 2 with a repeated trial for each condition. In Experiment 1, the counterweight load was set to zero and the initial lean angle

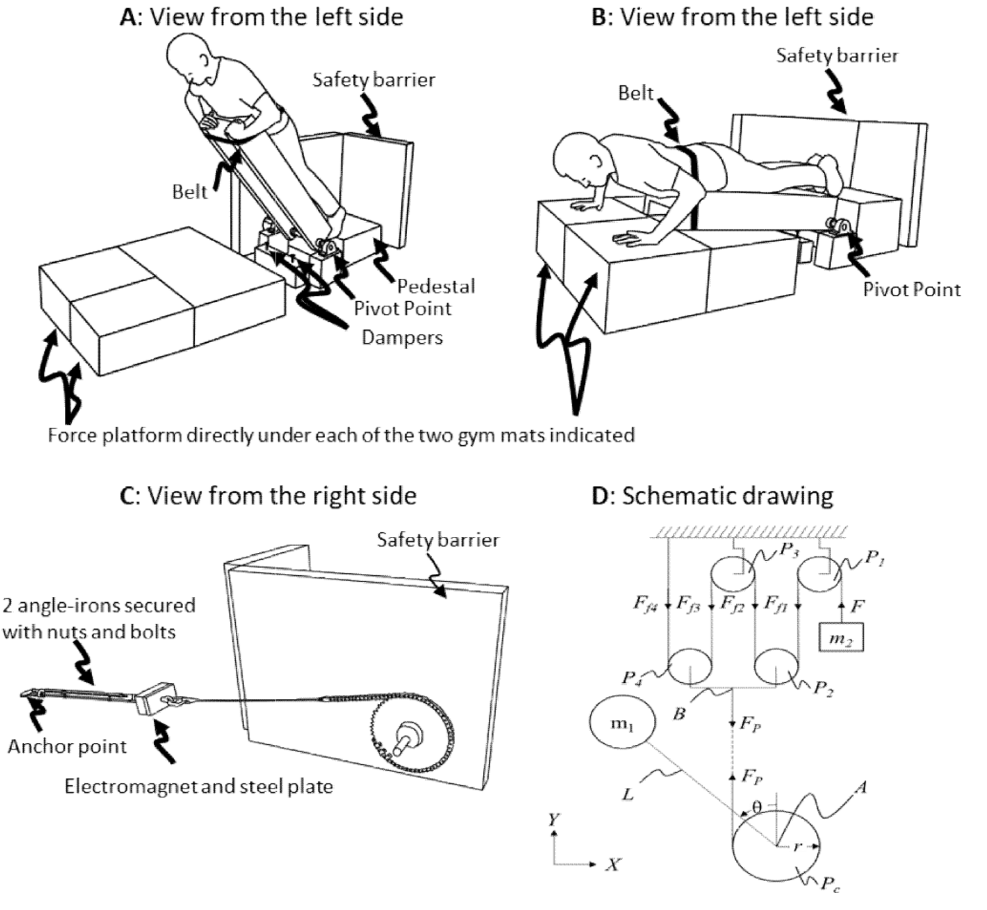

Fig. 1. Illustration of the FALL simulator For Injury prevention Training and assessment system (FALL FIT). Participants lay on top of the support platform. At an unpredictable time, the support platform and the participant are released from a leaning position (panel **A**). Following perturbation onset, the participant rapidly orients their hands and arms to prepare to absorb the impact energy following impact with the landing surface (panel **B**). The device is held in position using an electromagnet. The initial lean angle is controlled by varying the distance between the anchor point and the electromagnet (panel **C**). A counterweight system is used to reduce the angular acceleration of the support platform. A schematic diagram of the FALL FIT with a counterweight system is shown in panel **D**. The participant is modeled as an inverted pendulum with mass m_1 and height L. When the system is released, the angle θ increases, pulling point B downward. The counterweight m_2 moves upward at a rate that is 4 times greater than point B moves downward. Four pulleys, P_1, P_2, P_3, and P_4, were used to effectively multiply the counterweight m_2. The pulley system was an effort to compensate for the relatively small moment arm the counterweight load acts through compared to the inverted pendulum, r versus L respectively. Physical constraints limited the maximum allowable radius r of the counterweight pulley (P_c). The cable system needed to be sufficiently far from the area where the arms could move during the trials. Additionally, the pivot point A had to be higher above the ground than the radius of the counterweight pulley. Adapted from Borrelli, Creath, and Rogers (2020) with permission from Elsevier.

was varied. Participants were suspended at 4 initial lean angles (Low: 65±3, Medium: 52±3, Medium-Hight: 45±3, and High: 39±3 degrees from vertical). Lean angles were chosen to give a range of drop durations comparable in timing to reported perturbation-evoked reach-to-grasp and protective arm reactions [11,38–40].

In Experiment 2, the initial lean angle was held constant (37±1 degrees from the vertical) and the counterweight was varied (Small: 5.9±0.3, Medium: 10.7±0.6, Medium-Large: 15.5±0.9, Large: 25.1±1.4 percent body weight when including the 14.4 kg weight of the support platform) using four counterweights (2.6, 4.7, 6.8, and 11 kg). Trials were performed within 2 minutes of one another. The two trials for each condition are subsequently referred to as test and retest. Presentation of test conditions, initial lean angle in Experiment 1 and counterweight load in Experiment 2, were randomly selected for each participant and the test and retest of each condition were presented sequentially. Prior to starting, participants performed a single familiarization trial using the same initial lean angle or counterweight load that would be used in the first 2 trials. A familiarization trial reduces exaggerated responses typically observed during responses evoked from a novel perturbation [35].

Participants

Test-retest reliability of the FALL FIT was estimated using data from previous studies investigating the effect of initial lean angle [38,41] and counterweight load on evoked protective arm reactions. Experiment 1 included a convenience sample of 14 younger adults (5 males/9 females; mean age: 25.1±3.5 years (22-32 years); mean height: 173.9±10.0 cm (159-191 cm); mean body mass: 78.9±4.6 kg (54.4-103.0 kg)) and 13 older adults (9 males/4 females; mean age: 71.3±3.7 years (65-77 years); mean height: 175.2±9.0 cm (157-189 cm); mean body mass: 77.3±15.0 kg (56.3-104.8 kg)). Experiment 2 included a convenience sample of 13 younger adults (12 males/1 female; mean age: 31.8±5.7 years (25-41 years); mean height: 185.7±6.1 cm (174-193 cm); mean body mass: 82.4±5.6 kg (77.4-92.1 kg)). Only one participant participated in both Experiment 1 and Experiment 2 with more than one year between testing sessions. The FALL FIT evolved from an inverted pendulum to an inverted pendulum with a counterweight system. The experiments proceeded as follows: experiment 1 with younger adults, experiment 1 with older adults, and then experiment 2. Experiment 2 occurred almost 2 years after the first experiment. Experiment 2 occurred mostly between semesters when younger adult (graduate) student participants were more generally available. Older adults were not included in experiment 2 due to timing and delays/constraints associated with COVID.

Participants were excluded if they presented with self-reported injury or surgery within the past 6 months, osteoporosis/osteopenia, neurological disorder, musculoskeletal restrictions at the shoulder, elbow, wrist, back, neck, or medical condition that precludes participation in regular exercise, or dizziness or unsteadiness. The study was approved by the Institutional Review Board at the University of Maryland School of Medicine and all participants provided written informed consent prior to participation.

Data recording and processing

Kinematics of the arm and the fall simulator were recorded using a 10 camera Vicon motion capture system with Nexus Software (Vicon, Denver, CO). Retroreflective markers were placed bilaterally on the ulna- and radius- styloid process, the medial and lateral epicondyle of the humerus, and the scapula-acromioclavicular joint. Additional markers were placed on the support platform and axle that the platform pivots about to track the angle of the support platform. Kinetic data were recorded with a pair of force platforms (ATMI, Watertown, MA). All data is reported for the preferred arm/hand. Force platform and motion capture data were recorded at 1500 Hz and 150 Hz respectively. Force platform and motion capture data were low-pass filtered using a dual-pass second-order Butterworth filter with a cut-off frequency of 30 Hz and 10 Hz, respectively.

Fall or drop duration and angular velocity prior to contact with the landing surface were included to characterize the perturbation of the FALL FIT. The maximum vertical ground reaction force (vGRF) has been shown to be affected by several factors. For example, the impact velocity measured at the hand and the neck, elbow angle (EA, 180 degrees is full extension) and angular velocity (EAV) at impact, [21,24,42–45].

Drop duration was defined as the time between perturbation onset (when the angular acceleration of the table exceeded 10 degrees/s^2) and hand contact (force platform load exceeding 10 N). The angular velocity of the support platform and participant was measured at 70 degrees with respect to the vertical. Seventy degrees was empirically determined to be the largest angle prior to any participant contacting the gym mats. Elbow angle was calculated as the angle between the vector connecting midpoint of the ulnar and radial wrist markers and the midpoint of the lateral and medial elbow markers and the acromion and the midpoint of the lateral and medial elbow markers. The maximum vGRF, elbow angle at impact, and elbow angular velocity at impact were recorded for the preferred hand or the hand the participant used to write.

Statistical analyses

Intraclass correlation coefficients (ICCs) were calculated for each dependent variable to determine test-retest reliability [46–48] of single measures and average measures. A two-way model testing for absolute agreement (ICC(A,1)) was used to estimate reliability of single measures in each experimental condition, i.e., initial lean angle or counterweight load. A two-way model of absolute agreement (ICC(A,4)) was used to estimate reliability of the measures averaged across the test condition and retest condition. ICC reliability values of less than 0.40 were considered to be poor, between 0.40 and 0.59 fair, between 0.60 and 0.74 good, and between 0.75 and 1.0 excellent [49]. The minimal differences for the error to be real (MD) were also reported [50]. The MD is equal to the product of 1.96 and the standard deviation of the test-retest differences giving 95% confidence that the error is real.

Results

Experiment 1: variable initial lean angle

Table 1 presents the test-retest reliability of the FALL FIT and the evoked kinetics and kinematics for younger adults in Experiment 1. The test-retest reliability was fair to excellent for 82% of single measures (ICC(A,1)>0.51) and excellent for averaged measures (ICC(A,4)>0.89).

Similarly, for older adults (Table 2), the test-retest reliability was fair to excellent for 89% of single measures (ICC(A,1)>0.53) and excellent for averaged measures (ICC(A,4)>0.87) in Experiment 1.

Experiment 2: variable counterweight load

Table 3 details the test-retest reliability for Experiment 2. The ICC for single measures was good to excellent for 75% of the variables ((ICC(A,1)>0.60). Test-retest reliability was good to excellent for average measures (ICC(A,4)>0.87).

Discussion

We developed the FALL FIT to allow control over the dynamics of the body in a simulated forward fall involving protective arm movements. Control of the initial lean angle and counterweight load allow simulation of a variety of fall conditions and evoked arm kinetics and kinematics comparable to what has been observed in naturally occurring falls and in falls evoked in the laboratory. The fall characteristics of the FALL FIT were shown to have excellent reliability for average measures and good to excellent reliability for single measures with some exceptions. In addition, the evoked arm kinetics and kinematics also exhibited good to excellent reliability with some exceptions.

The reliability for single measures (ICC(A,1)<0.99) and good to excellent for average measures (ICC(A,4)>0.86). While the reliability of the majority of single measures exhibited good to excellent reliability, there were several measures that exhibited poor to fair reliability. This may be partially explained by the relatively narrow range of dispersion of these results [51]. More importantly, the mean difference between trials was less than the minimum error for the difference to be real. This suggests that the differences are within an acceptable range of measurement error [50,52,53]. However, the level of control afforded by the varied fall heights and counterweight system may allow protective arm responses to a wide range of fall velocities and accelerations to be investigated, assessed, or trained.

Hand contact has been reported to occur between about 680 ms [11] and 702 ms [54] after loss of balance in falls from standing height. The time between perturbation onset and hand contact in the present study was about 553-575 ms at the high initial lean angle (39°). Although the increased drop duration reported in previous studies may be the result of greater fall heights employed, the FALL FIT

Table 1
Variable Initial Lean Angle-Younger Adults.

	Initial Lean Angle	Mean (SD)	ICC	Mean Trial Difference (MD)
Angular Velocity (°/s)	L	82 (18)	0.91 (0.75-0.97)	1 (16)
	M	131 (23)	0.98 (0.94-0.99)	1 (9)
	MH	142 (15)	0.84 (0.53-0.95)	4 (15)
	H	160 (13)	0.84 (0.57-0.95)	1 (15)
	Avg.	129 (12)	0.99 (0.97-0.99)	1 (15)
Drop Duration (ms)	L	366 (28)	**0.53 (0.02-0.82)**	5 (55)
	M	459 (34)	**0.56 (0.09-0.83)**	15 (59)
	MH	521 (28)	**0.53 (0.06-0.82)**	10 (52)
	H	575 (45)	0.84 (0.57-0.95)	1 (53)
	Avg	480 (27)	0.91 (0.71-0.97)	5 (56)
Maximum vGRF (%BW)	L	55 (11)	0.82 (0.46-0.94)	4 (12)
	M	63 (13)	0.79 (0.47-0.93)	2 (17)
	MH	69 (13)	0.90 (0.72-0.97)	1 (11)
	H	72 (15)	0.80 (0.48-0.93)	1 (19)
	Avg	65 (11)	0.97 (0.91-0.99)	1 (15)
V_{neck} (m/s)	L	2.4 (0.3)	0.71 (0.26-0.90)	0.1 (0.4)
	M	2.9 (0.3)	0.82 (0.53-0.94)	0.1 (0.3)
	MH	3.1 (0.2)	0.70 (0.31-0.89)	0.1 (0.3)
	H	3.2 (0.2)	0.84 (0.58-0.94)	0.1 (0.3)
	Avg	2.9 (0.2)	0.93 (0.79-0.98)	0.1 (0.3)
V_{hand} (m/s)	L	3.3 (0.8)	0.66 (0.20-0.88)	0.1 (1.4)
	M	3.5 (0.5)	**0.58 (0.12-0.84)**	0.2 (0.8)
	MH	3.5 (0.4)	**0.52 (0.05-0.81)**	0.2 (0.7)
	H	3.6 (0.3)	0.81 (0.47-0.94)	0.1 (0.4)
	Avg	3.5 (0.4)	0.89 (0.63-0.97)	0.1 (0.9)
EA (°)	L	100 (17)	0.69 (0.29-0.89)	4 (26)
	M	121 (15)	0.68 (0.25-0.98)	1 (24)
	MH	127 (14)	0.94 (0.83-0.98)	1 (10)
	H	131 (12)	0.92 (0.71-0.98)	3 (8)
	Avg	120 (12)	0.98 (0.94-0.99)	1 (19)
EAV (°/s)	L	261 (242)	0.77 (0.44-0.92)	47 (323)
	M	163 (169)	0.60 (0.15-0.85)	64 (286)
	MH	104 (127)	0.67 (0.26-0.88)	47 (191)
	H	102 (148)	0.85 (0.55-0.95)	42 (148)
	Avg	158 (145)	0.94 (0.81-0.98)	27 (255)

Mean (standard deviation) of the dependent variables at the low, medium, medium-high, and high initial lean angles and the average across initial lean angles. The intraclass correlation coefficient (95% confidence interval) was estimated for single measures (ICC(A,1)) and average measures (ICC(A,4)). Intraclass correlation coefficients with poor-fair reliability are in bold (ICC<0.6). The mean trial difference and the minimum difference for the error to be real (MD) are also listed. L = low initial lean angle, M = medium initial lean angle, MH = medium-high initial lean angle, H = high initial lean angle, Avg = average, vGRF = vertical ground reaction force, V_{neck} = vertical neck velocity at impact, V_{hand} = vertical hand velocity at impact, EA = elbow angle at impact, EAV = elbow angular velocity at impact.

can be configured to yield a wide range of drop durations by manipulating the counterweight. For example, falls from the high initial lean angle and a counterweight load greater than 10.7%BW results in a drop duration greater than 648 ms.

The maximum vGRF was reported to be between 118%BW and 129%BW in simulated forward falls from an initial lea angle in younger adults [22,23]. In the present study, the maximum vGRF was 63%BW and 72%BW, for older and younger adults respectively, which was observed at the high initial lean angle with no counterweight. The difference in maximum vGRF between the studies may be attributed to the increased thickness of the foam used in the present study [25,26,55]. This suggests that the cushioning material used in the present study may be on an overly conservative safety measure. Therefore, thinner cushioning material may be used in the landing surface in future studies allowing simulation of falls on more realistic non-compliant landing surfaces, e.g., concrete, hardwood, linoleum, etc.

Table 2
Variable Initial Lean Angle-Older Adults.

	Initial Lean Angle	Mean (SD)	ICC	Mean Trial Difference (MD)
Angular Velocity (°/s)	L	58 (28)	0.77 (0.40-0.92)	10 (35)
	M	123 (17)	0.84 (0.43-0.96)	7 (23)
	MH	129 (22)	0.75 (0.37-0.91)	1 (27)
	H	133 (18)	0.88 (0.67-0.96)	1 (20)
	Avg	111 (16)	0.95 (0.82-0.98)	4 (29)
Drop Duration (ms)	L	351 (47)	0.81 (0.46-0.94)	13 (50)
	M	442 (56)	0.80 (0.49-0.93)	18 (83)
	MH	507 (54)	**0.53 (0.02-0.82)**	10 (133)
	H	553 (64)	0.75 (0.39-0.91)	14 (97)
	Avg	463 (48)	0.90 (0.70-0.97)	14 (91)
Maximum vGRF (%BW)	L	41 (12)	0.85 (0.60-0.95)	2 (12)
	M	53 (16)	0.63 (0.21-0.86)	5 (29)
	MH	59 (16)	0.84 (0.56-0.95)	1 (18)
	H	63 (17)	0.77 (0.43-0.92)	1 (20)
	Avg	54 (13)	0.98 (0.93-0.99)	1 (20)
V_{neck} (m/s)	L	2.3 (0.4)	0.70 (0.30-0.89)	0.1 (0.5)
	M	3.9 (0.3)	0.66 (0.20-0.88)	0.1 (0.6)
	MH	3.2 (0.2)	0.74 (0.36-0.91)	0.1 (0.5)
	H	3.2 (0.2)	0.69 (0.27-0.89)	0.1 (0.5)
	Avg	2.9 (0.2)	0.87 (0.63-0.96)	0.1 (0.4)
V_{hand} (m/s)	L	2.8 (0.9)	0.68 (0.26-0.88)	0.1 (1.4)
	M	3.2 (0.9)	0.80 (0.49-0.93)	0.3 (1.2)
	MH	3.3 (0.8)	0.79 (0.45-0.93)	0.1 (1.0)
	H	3.6 (0.7)	0.81 (0.50-0.94)	0.1 (0.8)
	Avg	3.2 (0.7)	0.95 (0.84-0.98)	0.1 (1.1)
EA (°)	L	85 (21)	0.81 (0.50-0.94)	0.1 (0.5)
	M	113 (12)	0.85 (0.62-0.95)	3 (19)
	MH	123 (9)	0.78 (0.43-0.92)	2 (16)
	H	126 (9)	**0.41 (0.00-0.77)**	1 (18)
	Avg	112 (9)	0.94 (0.81-0.98)	1 (19)
EAV (°/s)	L	159 (230)	**0.55 (0.04-0.83)**	27 (439)
	M	65 (281)	0.73 (0.35-0.90)	84 (255)
	MH	14 9260	0.93 (0.81-0.98)	3 (224)
	H	-26 (246)	0.92 (0.77-0.97)	25 (187)
	Avg	53 (219)	0.96 (0.88-0.99)	9 (321)

Mean (standard deviation) of the dependent variables at the low, medium, medium-high, and high initial lean angles and the average across initial lean angles. The intraclass correlation coefficient (95% confidence interval) was estimated for single measures (ICC(A,1)) and average measures (ICC(A,4)). Intraclass correlation coefficients with poor-fair reliability are in bold (ICC<0.6). The mean trial difference and the minimum difference for the error to be real (MD) are also listed. L = low initial lean angle, M = medium initial lean angle, MH = medium-high initial lean angle, H = high initial lean angle, Avg = average, vGRF = vertical ground reaction force, V_{neck} = vertical neck velocity at impact, V_{hand} = vertical hand velocity at impact, EA = elbow angle at impact, EAV = elbow angular velocity at impact.

In falls from an initial leaning position where the lower extremities are unconstrained, contributions from the lower extremities and trunk may reduce the impact velocity. The neck velocity at impact was reported to be 2.3-2.7 m/s for simulated forward falls where the shoulder was about 1 m above the landing surface [22,23]. In the present study, at the high initial lean angle with no counterweight, the neck velocity at impact was 3.2 m/s with the shoulder being about 0.9 m above the landing surface implying that a lack of contributions from the lower extremities and trunk increased impact velocity. Considering that the average height of the neck and head are about 1.44 m and 1.6 m above the ground for an average male [56], further support for contributions of the lower extremities and trunk in reducing impact velocity is provided by the fact that falls from standing height have shown vertical head velocity at ground impact to be 2.9 m/s [22] which is also smaller than the fall velocity reported here despite a greater fall height. Therefore, efforts to reduce the fall acceleration may be approximated by increasing the counterweight with the FALL FIT resulting in a reduced impact velocity.

Table 3
Variable Counterweight-Younger Adults.

	CounterweightLoad	Mean (SD)	ICC	Mean Trial Difference (MD)
Angular Velocity (°/s)	S	133 (13)	0.79 (0.44-0.93)	1 (18)
	M	126 (13)	0.64 (0.14-0.86)	1 (22)
	ML	116 (11)	0.75 (0.31-0.92)	4 (13)
	L	96 (8)	**0.00 (0.00-0.30)**	7 (23)
	Avg	118 (8)	0.87 (0.55-0.96)	3 (20)
Drop Duration (ms)	S	601 (40)	**0.32 (0.00-0.95)**	2 (95)
	M	648 (77)	0.84 (0.55-0.95)	3 (91)
	ML	701 (63)	**0.57 (0.07-0.84)**	16 (116)
	L	816 (95)	0.69 (0.25-0.89)	15 (1551)
	Avg	692 (44)	0.91 (0.70-0.97)	1 (114)
Maximum vGRF (%BW)	S	64 (12)	0.85 (0.59-0.95)	1 (13)
	M	56 (14)	0.91 (0.72-0.97)	1 (12)
	ML	53 (11)	0.94 (0.81-0.98)	1 (8)
	L	46 (11)	0.79 (0.46-0.93)	3 (13)
	Avg	55 (11)	0.98 (0.94-0.99)	1 (12)
V_{neck} (m/s)	S	3.1 (0.2)	**0.57 (0.03-0.85)**	0.1 (0.4)
	M	2.8 (0.2)	0.74 (0.36-0.91)	0.1 (0.3)
	ML	2.5 (0.3)	**0.04 (0.00-0.57)**	0.1 (0.9)
	L	2.2 (0.3)	0.61 (0.09-0.86)	0.1 (0.6)
	Avg	2.7 (0.2)	0.92 (0.74-0.98)	0.1 (0.6)
V_{hand} (m/s)	S	3.3 (0.4)	**0.35 (0.00-0.74)**	0.1 (0.8)
	M	2.9 (0.4)	0.79 (0.47-0.93)	0.1 (0.5)
	ML	2.8 (0.4)	0.74 (0.37-0.91)	0.1 (0.5)
	L	2.4 (0.4)	**0.48 (0.00-0.81)**	0.1 (0.9)
	Avg	2.9 (0.3)	0.89 (0.67-0.97)	0.1 (0.7)
EA (°)	S	132 (10)	0.88 (0.66-0.96)	1 (10)
	M	130 (11)	0.93 (0.79-0.98)	1 (9)
	ML	128 (11)	0.87 (0.63-0.96)	1 (12)
	L	126 (14)	0.86 (0.60-0.95)	1 (16)
	Avg	129 (11)	0.98 (0.93-0.99)	1 (12)
EAV (°/s)	S	75 (132)	0.80 (0.46-0.94)	42 (150)
	M	55 (139)	0.86 (0.57-0.96)	37 (136)
	ML	89 (156)	0.76 (0.41-0.92)	38 (209)
	L	110 (125)	0.65 (0.20-0.88)	27 (207)
	Avg	82 (125)	0.97 (0.89-0.99)	23 (181)

Mean (standard deviation) of the dependent variables at the low, medium, medium-high, and high initial lean angles and the average across initial lean angles during trials with a counterweight. The intraclass correlation coefficient (95% confidence interval) was estimated for single measures (ICC(A,1)) and average measures (ICC(A,4)). Intraclass correlation coefficients with poor-fair reliability are in bold (ICC<0.6). The mean trial difference and the minimum difference for the error to be real (MD) are also listed. S = small counterweight load, M = medium counterweight load, ML = medium-large counterweight load, L = large counterweight load, Avg = average, vGRF = vertical ground reaction force, V_{neck} = vertical neck velocity at impact, V_{hand} = vertical hand velocity at impact, EA = elbow angle at impact, EAV = elbow angular velocity at impact.

The hand velocity at impact was about 3.6 m/s for the high initial lean angle with no counterweight, about 0.6 m/s greater than that reported in simulated forward falls from an initial leaning position [22,23]. Studies of falls from standing height also evoke smaller hand velocity at impact, between 2.5 and 2.9 m/s [11,22,23,54]. Use of a medium, medium-large, or large counterweight in conjunction with the high initial lean angle results in a vertical hand velocity of 2.4-2.9 m/s consistent with the results reported by others.

The elbow angle at impact in simulated forward falls from an initial lean angle have been reported to be between 155 and 168 degrees of extension [22,23]. In contrast, the evoked elbow angle at impact in trials beginning with the high initial lean angle, the present study found the elbows to be more than 24 degrees less extended. The elbow angle at impact does not appear to be modulated as a result of varying counterweight load and therefore it may not be possible to evoke an elbow angle comparable with previous studies. However, the these studies [22,23] did not require the hands and arms to be oriented during the fall because they were in an orientation conducive to absorbing impact at the beginning of the fall. In a study requiring rapid orientation of the hands and arms, the elbow angle at impact was about 116 degrees of extension [30]. Although the elbows were extended to a lesser extent than what is reported here, the body was in a more upright orientation at impact and there was likely reduced need for absorbing the impact through elbow flexion and/or reduced maximum vGRF. Finally, the elbow extension angular velocity at the high initial lean angle was 102 and -26 degrees/s compared to 80 and 69 degrees/s for younger and older adults [30]. These differences may be attributed to differences in the fall dynamics. For example, the drop duration was nearly 100 ms longer and the max vGRF was almost twice as large in the present study. The evoked arm kinematics have been shown to be modulated with fall height/dynamics [38].

Although the FALL FIT can provide unpredictable fall acceleration and onset of a fall, participants are likely able to pre-plan a reaction strategy due to the knowledge that a fall is unavoidable. During an actual fall, it is more likely that the fall parameters, such as potential for recovering balance, fall direction, impact velocity, availability of handholds that can be reached and grasped, etc., are unpredictable. However, there may be a point in time when it is clear that a fall is unavoidable. Studies have documented varying levels of success in preventing head/body impact as a result of deploying protective arm reactions in experiments of falls from standing height [11,12] and in naturally occurring falls [15,17,33]. Following a loss of balance, the reaction strategy for using the hands and arms may first aim to recover balance by redirecting body momentum and the associated movements of the hands and arms may be directed in a direction that is not compatible with ultimately protecting the body [57]. A protective strategy must be deployed at some time sufficiently prior to impact to allow the hands and arms the time necessary to be oriented in a manner conducive to preventing injury. Therefore, it is conceivable that the successful orientation of the hands and arms results from knowledge that the fall is unavoidable sufficiently in advance of impact.

From a clinical perspective, the present approach may also allow for novel assessment of protective arm reactions and serve as a rehabilitation tool for sensorimotor training to enhance fall recovery arm reactions and reduce the risk of injuries. With regard to assessment, protective arm reactions in younger adults have been characterized [38] and age-related changes have been observed which show older adults exhibit reduced movement time, require additional elbow angular flexion to absorb a comparable fall as younger adults which may put them at greater risk of head/body impact [41]. Task specific, perturbation-based balance training has been shown to result in postural adaptations that reduced the likelihood that a fall would occur [58]. Moreover, reactive, volitional, and exercise-based training have been shown to have benefits in reducing fall risk [59–62], particularly when modalities are combined, e.g. power training and task specific perturbation training [63]. Although, information on prevention training for fall injury avoidance focused on the upper extremities is quite limited [64], it seems reasonable that augmenting reactive, volitional, and strength/power-based training with task specific perturbation training using the FALL FIT could yield enhanced outcomes. By progressively manipulating fall intensity via changes in fall height and velocity, the approach described here may be useful for systematically enhancing postural challenge allowing progressive protective arm reaction training through trial-and-error practice to promote sensorimotor adaptations and fall injury safety.

Among the limitations of the study, the samples were relatively homogeneous healthy younger and older adults. The results presented here can only be extended to other populations that have similar variability to this study's population [53]. Future work will be required to determine the reliability and suitability of the FALL FIT to improve protective arm responses in older adults and other populations at risk for falls and fall related injuries. These results are further limited by the instructions given to the study participants. Instructions have been shown to affect the vGRF and affect other biomechanical factors associated with protective arm reactions [22,23,42,65]. Participants were instructed to land as softly as possible which may represent the "best-case" [32] and offer a safer condition than landing with stiff or straight elbows [45]. Finally, the FALL FIT approximates a forward fall where the body acts as an inverted pendulum. While this arrangement allows control over the falling acceleration of the body, this may not best represent the body kinematics during an actual fall. As we have mentioned above, falls where the body is not constrained to the same level as used here result in reduced acceleration as indicated by the fall velocity at impact. The FALL FIT may be representative of a more severe type of fall where the lower extremities are unable to contribute to reducing the fall velocity and/or recovering balance (e.g., a trip where protective steps are unsuccessful or not possible). However, the counterweight system may allow simulation of a range of successful and less successful balance recovery efforts that modulate the impact velocity.

Conclusion

We have developed the FALL FIT to allow control of the fall acceleration and associated fall velocity. The test-retest reliability of the fall dynamics and the evoked reaction was generally good to excellent with some exceptions. The FALL FIT is suitable for investigating a wide variety of fall conditions. Additionally, the FALL FIT may prove a viable method for assessing and training dysfunctional protective arm reactions and/or individuals with limitations.

Declaration of Competing Interests

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Acknowledgements

JB was supported by the University of Maryland Advanced Neuromotor Rehabilitation Research Training (UMANRRT) Program (NIDRR 90AR50280, NIDILRR 90AR5004 formerly H133P100014). This project was supported by the University of Maryland Claude D Pepper-OAIC NIH/NIA grant P30AG028747. The assistance of Ruth Akinlosotu at the University of Maryland Baltimore and Kathleen Simpson at the Baltimore VA Medical Center, Geriatric Research, Education and Clinical Center for support with data collection and study participant recruitment is gratefully acknowledged.

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