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DOI
10.1016/j.forsciint.2024.112068

Publication date
2024

Document Version
Final published version

Published in
Forensic Science International

Forensic Science International 361 (2024) 112068

Contents lists available at ScienceDirect

Forensic Science International

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

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ARTICLE INFO

Keywords:
Forensic biomechanics
Forensic reconstruction
Falls from height
Numerical modelling
Human body model
Simcenter Madymo

ABSTRACT

Forensic reconstruction and scenario evaluation are crucial in investigations of suspicious deaths related to falls from a height. In such cases, distinguishing between accidental falls, being pushed or jumping is an important but difficult task, since objective methods to do so are currently lacking. This paper explores the possibility of repurposing a passive rigid body model of a human from commercially available crash simulation software for forensic reconstruction and scenario evaluation of humans dropping from heights. To use this approach, a prerequisite is that the human body model can produce realistic movements compared to those of a real human, given similar environmental conditions. Therefore, this study assessed the validity of the commercially available Simcenter Madymo Pedestrian Model (MPM) for simulating human fall movements. Experimental kinematic and kinetic data was collected from nine participants, who dropped from a height in three different ways: passively tilting over, getting pushed, and jumping. Next, the performance of the MPM in reproducing the kinematics of the experimental falls was assessed by comparing the orientation of the body 0.3 s after platform release. The results show that the MPM currently does not consistently reproduce the experimentally recorded falling movements across multiple falling conditions and outcome measures. The MPM must therefore be adapted if to be used for forensic reconstruction and scenario evaluation, for example by implementing active movement.

1. Introduction

Finding a body at the base of a staircase or a building often leads to a challenging investigation for law enforcement. The individual could have tripped or fallen accidentally, but the fall could also have been a result of being pushed or a case of self-inflicted harm through jumping. Alternatively, the body could also have been moved post-mortem to stage it as an accident or suicide following a homicide. Distinguishing between an accident, a suicide, and a crime is a complex task that may require an extensive forensic investigation. The ultimate goal of such a forensic investigation is to reconstruct the events that could have led to the observed situation at the scene and to make a probability statement about possible scenarios. This is done based on, for instance, the position and orientation of the body at a scene, observed injuries, and tactical information such as witness testimonies.

During the process of forensic reconstruction, there is a need for efficient and validated methods to objectively evaluate possible scenarios. Unfortunately, such methods for falling incidents are currently lacking and as a result, forensic investigations of this kind often lead to dead-ends. One method that has been applied in literature is using a crash test dummy for the physical reconstruction of standing fall scenarios [1–3]. An interesting alternative is to use numerical simulations with human body models for forensic reconstruction and scenario evaluation. For this approach, a computer model of the human body is needed with an appropriate set of initial conditions and parameters — such as similar fall height as observed at the scene or similar body proportions as the victim. Then, based on physics, the kinematics and dynamics of this human body model are computed forward in time, which is generally done using simulation software. For example, commercial simulation software has been developed for (car) crash simulations, including human body models. Dedicated software for simulating humans dropping from heights is currently not available.

While not specifically designed for simulating human falling behaviour, crash simulation software has previously been used for reconstructing human falling incidents for application in forensics. For instance, the crash-simulation software PC-Crash has been used to execute a large number of fall simulations with varying input conditions such as jumping, tripping, being pushed or being thrown [4].

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https://doi.org/10.1016/j.forsciint.2024.112068
Received 7 March 2024; Received in revised form 14 May 2024; Accepted 23 May 2024
Available online 25 May 2024
0379-0738/

Subsequently, these simulation results were applied to the investigation of a number of forensic fall cases [4]. PC-Crash has also been used in several forensic case reconstructions of falling incidents, where some of the simulations led to a plausible scenario [5,6]. Another commercially available crash simulation software package used in literature is Simcenter Madymo (Siemens Industry Software and Services BV). Madymo has already been used in at least two forensic case reconstructions of falling incidents [7]. Additionally, Madymo was used to perform simulations of fall incidents using initial conditions based on experimental recordings of volunteers falling [8,9]. Some of these studies found a set of initial conditions that could explain the traces at the scene of the incident being reconstructed [6,8]. However, until now none of the commercially available human body models have been validated for application to fall simulations, making the reliability of such simulations in a forensic setting unknown.

To enable applying numerical simulation of humans dropping from heights in forensic reconstruction, it is crucial to have a good understanding of how well the given software and the human body model perform compared to real-life falling. Consequently, with an accurate human body model, it is desirable to make probability statements on the likelihood of occurrence of the observed outcome for any set of considered input conditions. This can for example be done by using a Monte Carlo approach. In a Monte Carlo simulation, a large number of simulations are performed, for a range of varying input conditions such as position and posture. Therefore, for effective and (cost and time) efficient use, a human body model should be as simple as possible while maintaining its accuracy for falling movements.

The current study focuses on the applicability of the Madymo Pedestrian Model (MPM) for forensic fall reconstruction for several reasons. First of all, the MPMs have already been extensively validated for application in pedestrian-vehicle impact simulations, proving that they accurately reproduce the kinematics of the human body for that application [10,11]. The MPMs are passive, no active elements are modelled (i.e., no muscle activity is simulated), and therefore it is a relatively simple model. Of these pedestrian models, the Madymo 50th percentile ellipsoid pedestrian model is the most commonly used model for pedestrian impact reconstruction [12]. This model is robust, easy to scale to different body dimensions and computationally efficient [10]. Another practical reason to use Madymo for fall reconstruction is that Madymo has built-in tools for predicting injury outcomes, such as skull fracture, which is promising for application to forensics.

The goal of the current study was to determine to what extent the relatively simple, passive Madymo 50th percentile ellipsoid pedestrian model can reproduce the kinematics of experimentally recorded human falls. Consequently, the aim was to validate the use of the Madymo 50th percentile ellipsoid pedestrian model in reconstructing these experimental falls. For this, experimental data was collected from healthy volunteers performing several types of falling movements and the performance of the Madymo 50th percentile ellipsoid pedestrian model to reproduce the experimentally recorded falls was evaluated.

2. Methods

2.1. Experiment

2.1.1. Participants

Nine healthy participants were recruited for the study: three females and six males. The mean (SD) age, height, and weight of the participants were 26 (4) years, 181 (9) cm, and 82 (15) kg, respectively. The study was approved by the TU Delft Human Research Ethics Committee (approval nr. 1667). Participants provided written informed consent before partaking in the experiment.

2.1.2. Setup and materials

The experiment was conducted in an exercise hall equipped with a foam pit (a large container filled with soft foam blocks) that participants could safely fall into (*Gymworld Zoetermeer, FreeRun Academy*). A platform was positioned at the edge of the pit, about 1.5 m above the edge of the pit (Fig. 1). A custom-made force plate (*ForceLink BV, Culemborg, The Netherlands*) on top of the platform recorded the ground reaction force vector and its point of application with a sampling frequency of 1000 Hz.

The kinematics of the participants were tracked using an IMU-embedded Xsens motion capture suit (*MVN MT9, Xsens Technologies B. V., Enschede, The Netherlands*) and the corresponding software package MVN studio (*Version 2.6.5, Xsens Technologies B.V., Enschede, The Netherlands*) with a sampling frequency of 120 Hz.

During a subset of the trials, a custom-made device ("pusher") was used to push the participant into the foam pit. To record the magnitude of the push force, the pusher was equipped with a uni-axial force transducer (*Keli Transducers Co. Model: PST, capacity 150 kg*). Made from lightweight materials, the pusher was designed such that friction and inertia effects were negligible compared to the load applied to the participant. The pusher was mounted on a height-adjustable column tripod (*Manfrotto, 161MK2B*).

2.1.3. Experimental protocol

Falls were initiated with the participants in an upright position with their arms hanging by their sides, and their feet placed about 25 cm apart at the marks on the force plate (Fig. 2). Falls were performed under six sets of initial conditions, divided into three categories:

2.1.3.1. Passive. The participants were instructed to slowly lean into the direction of the pit to initiate the fall while trying to minimise active corrections during the fall. Participants performed three passive backwards falls *(PaBa)* and three passive sideways falls *(PaSi)* (Fig. 3a and Fig. 3b).

2.1.3.2. Pushed. Using the pusher, a short, firm push force was applied to the participant such that they fell. The peak push force was measured to vary between 69 and 152 N. The participant was warned before they were pushed to enhance the comfort of the participant, considering the uncomfortable nature of unanticipated pushes. Participants performed three backwards pushed falls *(PuBa)*, and three sideways pushed falls *(PuSi)* (Fig. 3c and Fig. 3d). In the backward trials, the push force was applied at the sternum. In the sideways trials push force was applied to

Fig. 1. A snapshot from one of the experimental trials. Participant falling from the force plate (a) that is located on the platform (b) into the foam pit. The participant wears an Xsens motion-capture suit (c). The device for pushing the participant into the pit can be seen on the left (d).

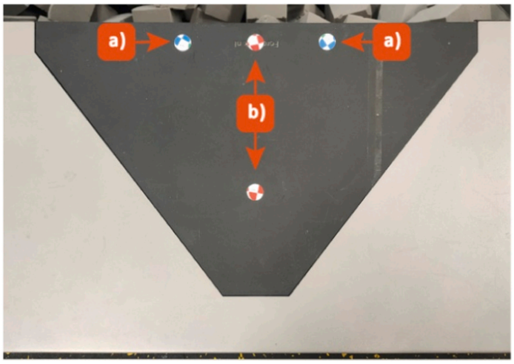

Fig. 2. Top view of the force plate used to record participant's ground reaction forces. The circular blue indicators to the left and the right (a) mark the foot positions during forward and backward falls, red indicators in the centre (b) indicate the locations for the participant's feet during sideways falls while participants were facing right.

the upper arm, approximately 10 cm below the shoulder.

2.1.3.3. Active. Participants were instructed to jump forward *(JuFo)* towards the centre of the pit, which was about 1.5 horizontal meters from the edge of the platform (Fig. 1). In addition, participants were instructed to step off the edge of the platform in a forward direction *(StFo)*, by placing a foot forward off the edge of the platform. Participants performed three falls jumping forward and three falls stepping forward (Fig. 3e and Fig. 3f).

Fig. 3 shows snapshots of typical trials for each of these falling conditions. Eleven trials were excluded from the analysis due to data-acquisition issues resulting in missing data. No passive and pushed falls in the forward direction were conducted due to the risk of causing back or neck injuries. The order of the fall types for each participant was assigned using a balanced Latin square for participants 1–7 and was randomised for participants 8 and 9. The active falls were recorded for future use and were not included in the subsequent analysis and simulations.

2.1.4. Synchronization

To ensure accurate, offline synchronisation of the force plate, kinematics and (for some trials) push force data with respect to each other the following pre-trial synchronisation process was followed: The participant briefly stood still on the force plate, with their feet at the marked locations (Fig. 2), and stamped down their left foot twice. This action generated a distinctive synchronisation peak in the force plate data and Xsens systems.

For trials where the participant was pushed, the force plate and pusher were synchronised by placing a wooden beam on the force plate while leaning it against the pusher's head. A distinct pair of synchronisation peaks in the force signals of both the force plate and pusher was made by tapping the beam twice.

2.2. Modelling

2.2.1. Madymo

Forward-dynamic simulations were executed for every trial from the passive and pushed experiments using crash reconstruction software Simcenter Madymo™ (*Version 2020.2, Siemens Industry Software and Services BV*). Simulations for the active experiments were not executed, as the Madymo Pedestrian Model (MPM) is inherently passive. A Madymo scene was built consisting of a platform for the MPM to drop from, and a floor 2 m below the platform (Fig. 4). The 0.5 m height difference between the Madymo platform and the experimental platform drop was made to ensure enough space for the MPM to complete the fall. This height difference did not impact the results, as the focus was on assessing the model's position and orientation after platform release (see *Outcome Measures*) and not the final state at landing. The used human body model was the scalable Madymo Pedestrian Model (*version 5.2, 'h_ped50el_inc.xml'*). A participant-specific MPM was made by scaling the standard model to the participants' sex, weight and height using tooling supplied with Madymo. The participant-specific MPM is henceforth referred to as the Madymo dummy. The Madymo dummy was initialised on top of the digital platform.

2.2.2. Initial conditions and modelling choices

To determine to what extent the MPM can reproduce the kinematics of the experimentally recorded falls, the initial conditions for the simulations needed to be carefully set. Firstly, the initial posture of the Madymo dummy was aligned with the posture of the participant recorded with Xsens. However, there were dissimilarities in joint coordinate system definitions and joint degrees of freedom between the Madymo and Xsens models. To be able to use the recorded Xsens data as input for the Madymo dummy, certain joints in the MPM had to be adjusted. These adjustments ensured that the coordinate systems and degrees of freedom of the MPM matched with those of the Xsens model (Appendix A) and that their initial posture aligned. Moreover, the Madymo software constructs the Madymo dummy around a virtual joint located in the middle of the pelvis between the hips, named the h-point. The h-point links the Madymo dummy to the environment and permits input of its global orientation and velocity. Accurately determining the h-point's height while considering the lower body's posture prevented the model from floating above the platform or penetrating the platform at the start of the simulation.

Secondly, to initiate the movement for the passive simulations, an initial velocity had to be applied to the Madymo dummy. To simulate different phases of falling, the participant's velocity was extracted at four specific time points during the fall (*T1, T2, T3* and *T4*) and used as the initial velocity for the simulation. *T1-T4* were linearly spaced in time between the last time instance at which the participants stood still before the fall and the time instance at which contact with the platform was lost (zero ground reaction force), henceforth referred to as the 'platform release' (see Fig. 5). The mean (SD) time difference between T1 and T4 was 1.48 s (0.23). The linear and angular velocities of the participant's pelvis segment relative to the world (resp. v_{pelvis} and ω_{pelvis}) at these time instances were applied at the model's h-point. Simulations were executed with the corresponding recorded postures at time instances *T1, T2, T3* and *T4*. To investigate the impact of individual joint movements on the simulations, additional simulations were performed applying not only the measured v_{pelvis} and ω_{pelvis} to the model, but applying also all other joint angular velocities (ω_{joints}) measured at *T1-T4*.

For the pushed falls, participants stood still before they were pushed during these trials, resulting in an initial velocity of zero. Hence, the movement of the Madymo dummy was initiated by the push force. Therefore, the push force magnitude over time was extracted from the pusher output and applied at the sternum or shoulder of the model, for backward and sideways falls respectively.

Lastly, to prevent the passive model from collapsing instantly, the knee and hip joints were locked in their starting posture for all falls during the full simulations. With this locking of the lower body and the fact that the Madymo dummy is completely passive, the Madymo dummy can be seen as an inverted pendulum. Therefore, the Madymo dummy is inherently unstable and must be balanced to stay in its initial position. This balancing was done by applying a stabilising torque at the pivot point, which is the ankle joint. The magnitude of this stabilising ankle torque (T) depends on the angle of the centre of mass of the Madymo dummy relative to the vertical (α), the length of the vector from the ankle to the centre of mass (l) and the mass of the Madymo dummy (m), see Fig. 6. This relationship can be expressed as $T = mgl\sin\alpha$ with g being the gravitational acceleration. The centre of mass was

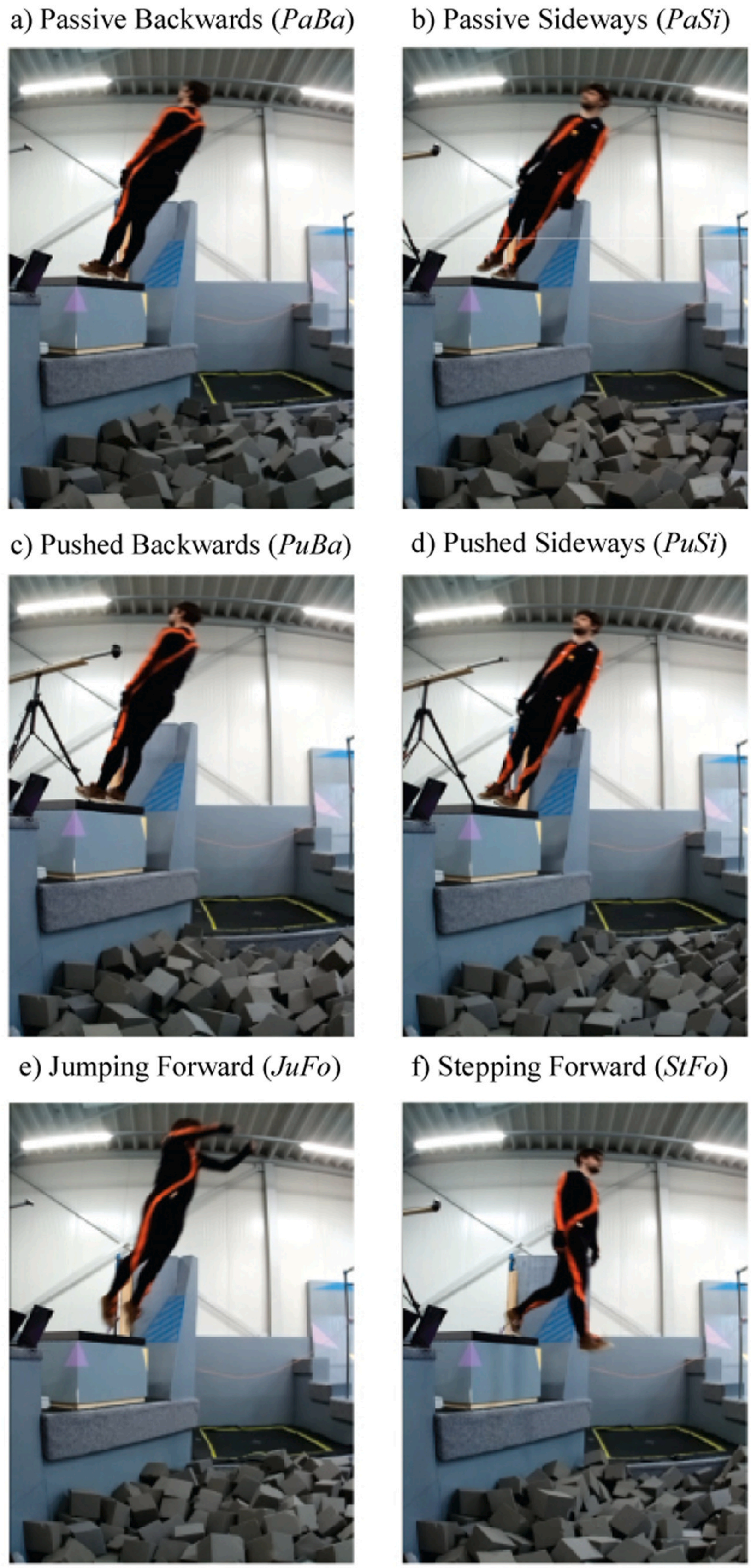

Fig. 3. Snapshot of typical trials for each of the falling conditions, a) passive backwards, b) passive sideways, c) pushed backwards, d) pushed sideways, e) jumping forward, f) stepping forward.

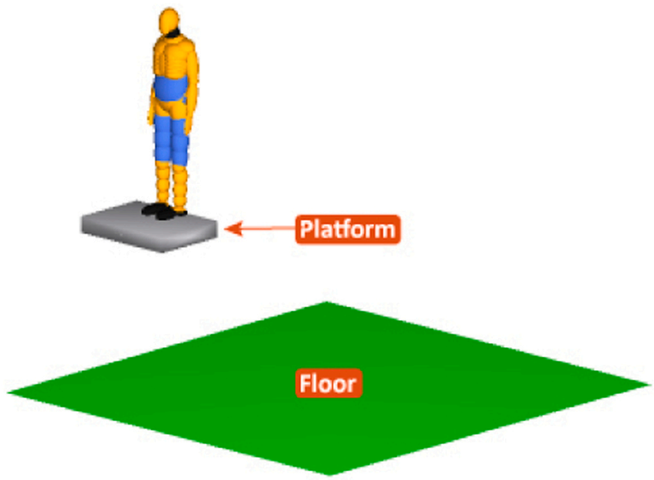

Fig. 4. The scene used for the Madymo simulations. The Madymo dummy is standing on top of the digital platform from which the falls are initiated.

assumed to be located at the h-point. To assess the effect of this stabilising torque, simulations were performed with and without adding this stabilising torque to the original ankle rotation stiffness implemented in the MPM [10]. An overview of all the simulation setups that were executed, and the terms used to refer to them, can be found in Appendix B.

2.3. Analysis

2.3.1. Outcome measures

To assess the accuracy of the simulation results compared to the experiment, three outcome measures were defined (Fig. 7): 1) the vertical angle, 2) the ledge angle and 3) the longitudinal angle. To compare the experiment and simulation during a comparable phase of the fall, the outcome measures were determined at 0.3 seconds after platform release for both the experiment and model. As this time instance was found to be just before participants hit the foam pit in the experiment. Platform release was defined as the time instance that the participant/ Madymo dummy dropped off the force plate/platform and the vertical component of the ground reaction force fell below 5 N. The outcome measures were determined for each trial, for both the experimental and the simulated falls.

In each of the outcome measures, the vector pointing from the pelvis to the head was defined as the body's vertical axis, represented by the green solid arrow in Fig. 7. The vertical angle (blue dashed arrow, Fig. 7a) is defined as the angle in degrees between the participant's body's vertical axis and the global vertical axis pointing upwards in the direction of gravity (red dotted arrow, Fig. 7a). The ledge angle (blue dashed arrow, Fig. 7b) is defined as the angle in degrees between the body's vertical axis and the horizontal edge of the platform from which they are falling (red dotted arrow, Fig. 7b). For both the vertical and ledge angle the axis of rotation is pointing into the screen, such that rotations towards the right and away from the platform in Fig. 7a and rotations to the right in Fig. 7b are positive. The longitudinal angle (blue dashed arrow, Fig. 7c) is defined as the rotation in degrees of the participant or Madymo dummy around their local axis relative to their initial position. All angles are mapped to the interval of -180° to 180°, with 0° the upright middle position.

2.3.2. Statistics

The experimental data originally consisted of 18 data subsets per participant – three outcome measures determined across three fall types (i.e., passive, pushed or active) and two fall directions (i.e., backwards/ sideways). The distribution of these datasets, including the data of the active trials, is shown using box plots providing the median, lower and upper quartiles. Hereafter, the data of the active trials is omitted, resulting in 12 data subsets. A Kolmogorov-Smirnov test showed that only one of these 12 subsets significantly differed from the normal distribution. As an Analysis of Variance (ANOVA) tends to be robust for non-normality, the effect of fall type and fall direction on the outcome measures of the experimental data was assessed using a two-way ANOVA with Bonferroni correction. For these 12 subsets, the mean and standard deviation of the outcome measures are provided.

The simulation data consisted of 120 data subsets – three outcome measures determined across the different modelling choices. The modelling choices are: fall type (i.e., passive or pushed), fall direction (i.e., backwards or sideways) and stabilising torque (i.e., with or without stabilising ankle torque) and in case of modelling passive falling initial velocity type (i.e., $v_{pelvis} + \omega_{pelvis}$ or $v_{pelvis} + \omega_{pelvis} + \omega_{joints}$) and time (i.e., *T1*-*T4*) are added. For well-founded model development, the simulation data would preferably be also analysed using N-way ANOVAs to assess the effects of the different modelling choices on the outcome measures. However, within the 120 simulation datasets, a Kolmogorov-Smirnov test showed that 43 of these datasets significantly differed from the normal distribution. Using non-parametric tests for these non-normally distributed datasets would involve an unfeasible amount of individual tests without providing insight into the interaction effects. This would barely provide any additional value compared to choosing

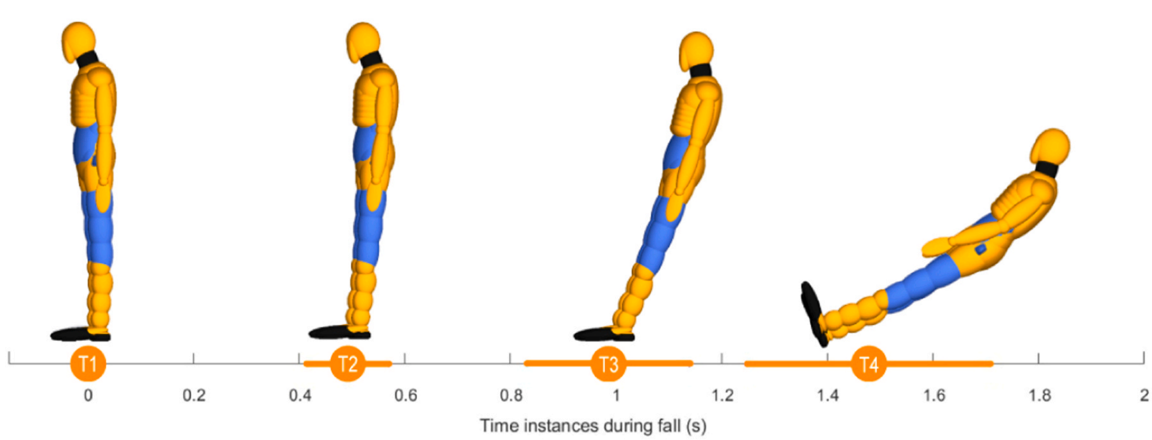

Fig. 5. Timeline and visual representation of time points T1 through T4 in the fall simulation study. T1 represents the last time instance participants stood still before initiating a fall, while T4 corresponds to the time instance at platform release. Above the timeline, a Madymo model illustrates a typical example of the body orientations at each time point. The circles and whiskers on the timeline indicate the mean and standard deviation of time differences relative to T1. T2 through T4 are ranges as participants independently initiated their fall, resulting in varied fall durations across participants.

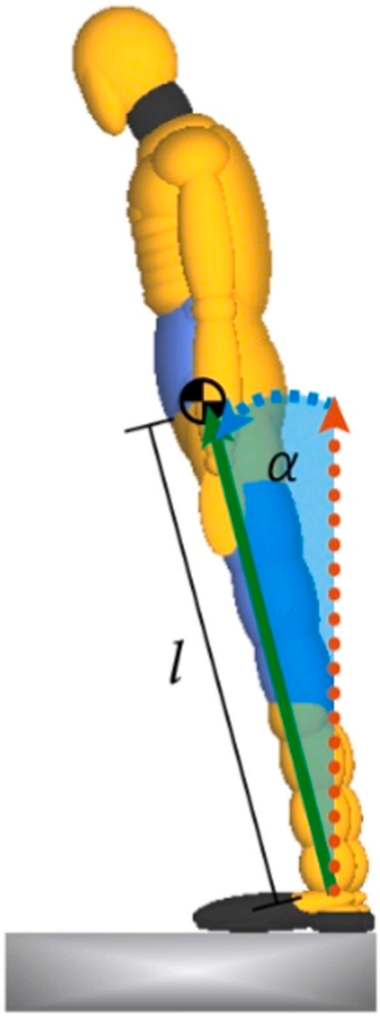

Fig. 6. Relevant parameters for determining the stabilising ankle torque, illustrated using a Madymo dummy.

the best-performing model settings for each individual fall type and fall direction. Therefore, for every modelling choice, the root mean squared error (RMSE) of the simulation results was calculated with respect to the experimental data. The completion rate (CR) is presented as the ratio of completed simulations to valid trials, accounting for instances where simulations were not completed due to numerical instabilities or convergence problems. Based on the RMSE and CR, the best-performing model for each fall type and direction was selected. Subsequently, a Wilcoxon signed-rank test was conducted to compare the data from the chosen model with the experimental data. The median and interquartile range (IQR) of the outcome measures for both experiments and simulations are provided.

3. Results

3.1. Experimental trials

During the experimental trials, notable variations were observed in the way in which participants performed falls. For example, while some participants tried to maintain an upright position by actively adjusting their posture, other participants did not and exhibited bigger vertical angles. This is also shown in Fig. 8, where the vertical angles in the PaBa, PuBa, and PuSi are sometimes negative, indicating that the participants' body vertical axis rotated slightly to the left and towards the platform in Fig. 7a. Table 1 provides the means and standard deviations for all outcome measures for all passive and pushed falls, while Fig. 8 provides the median and lower and upper quartiles for all fall types, including the active trials (see *Experimental Protocol*). The entire dataset, including analysis files and raw data, can be found at the 4TU repository (data.4tu.nl) under DOI 10.4121/20054996 [13].

A significant effect of fall type (i.e., passive or pushed) was found on the longitudinal angle ($p = 0.022$). A significant effect of fall direction (i.e., backwards/sideways) was found on the ledge angle ($p = 0.008$). Additionally, a significant interaction effect of fall type and fall direction for the longitudinal angle ($p = 0.02$) was found.

3.2. Modelled trials

3.2.1. Passive trials model analysis

The model with the lowest RMSE was taken for examination as this was considered the best-performing model (see *Statistics*). For simulations of the backward falls (PaBa), this was the model with initial conditions extracted at T3, with stabilising ankle torque and only applying v_{pelvis} and ω_{pelvis} (Table 2). For simulations of the sideways falls (PaSi), the model with the lowest RMSE was the model with initial conditions extracted on T2, with stabilising ankle torque and only applying v_{pelvis}

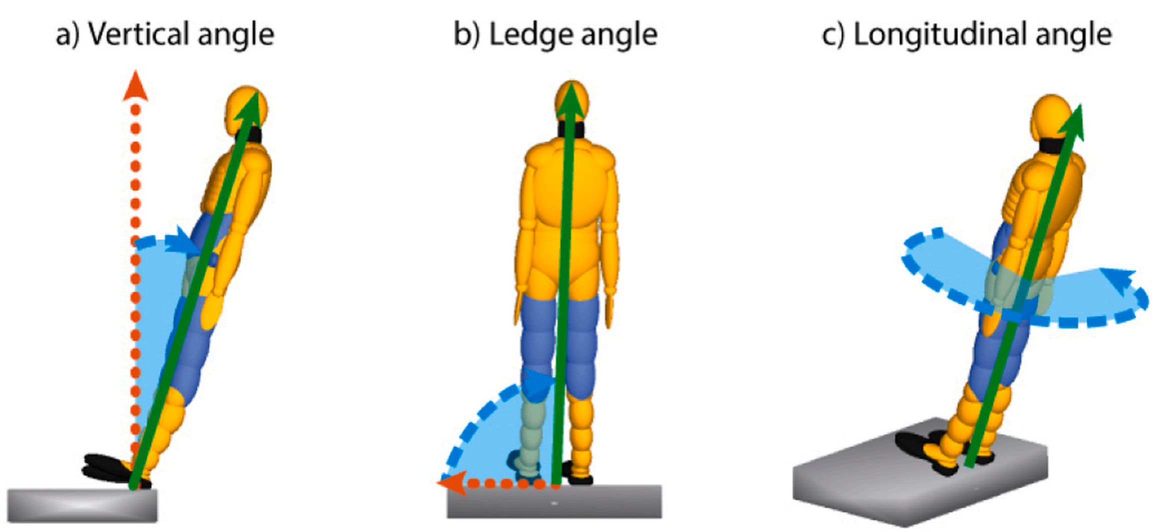

Fig. 7. The three outcome measures based on which the experiment and simulations were compared, illustrated using a Madymo dummy: a) vertical angle, b) ledge angle and c) longitudinal angle.

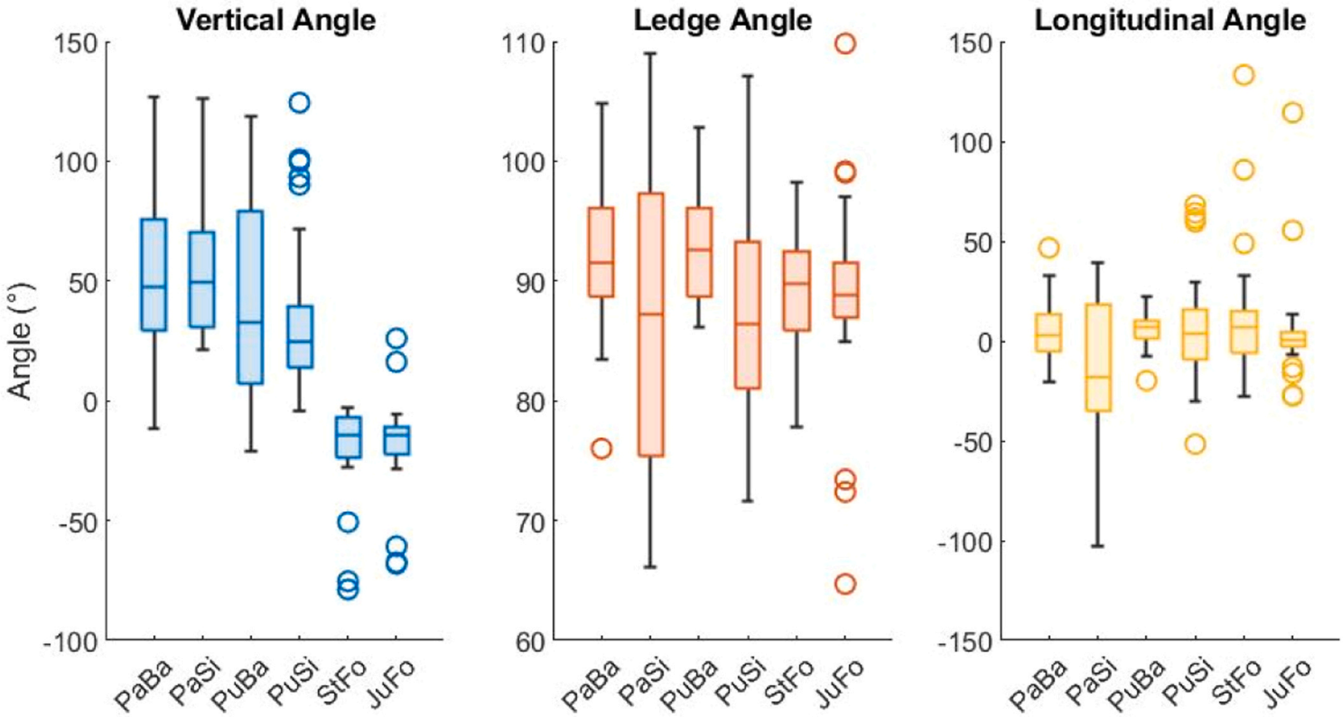

Fig. 8. Box plots for the outcome measures across all conditions. The line in centre of box represents the median. The lower and upper end of the box represent the lower and upper quartiles respectively. The whiskers extending from the box indicate variability outside the upper and lower quartiles. Outliers are displayed with circles. A positive vertical angle represents rotation to the right and away from the platform in Fig. 7a, a ledge angle larger than 90° represents rotation towards the right edge of the platform in Fig. 7b, a positive longitudinal angle represents anti-clockwise rotation around the body's vertical axis in Fig. 7c.

Table 1
The means (SD) of the outcome measures for the passive and pushed fall types and both fall directions.

		Vertical angle (°)	Ledge angle (°)	Longitudinal angle (°)
Passive	PaBa	51.9 (37.5)	92.5 (6.5)	92.5 (6.5)
	PaSi	56.5 (29.2)	87.0 (13.3)	-18.0 (38.6)
Pushed	PuBa	43.4 (43.8)	92.7 (4.6)	5.3 (8.9)
	PuSi	37.7 (35.2)	88.0 (9.4)	7.4 (29.9)

and ω_{pelvis} (Table 2).

Wilcoxon signed-rank tests showed that there were significant differences between the model and the experimental data for the vertical angle for PaBa and PaSi (both $p < 0.001$). No significant differences were observed for the ledge and longitudinal angle. The median and interquartile range (IQR) of the outcome measures for both experiments and simulations are shown in Table 3. Fig. 9 shows Bland-Altman plots of the level of agreement between the experiment and model setup per outcome measure.

3.2.2. Pushed trials model analysis

For simulations of the backward falls (PuBa), the model with the lowest RMSE was the model with stabilising ankle torque and with all joints locked (Table 2). For simulations of the sideways falls (PuSi), the model with the lowest RMSEs was the model with a stabilising ankle torque and with the lower body joints locked (Table 2). Wilcoxson sign-rank tests showed that there were significant differences between the model and the experimental data for PuBa and PuSi for the vertical angle (both $p < 0.001$), the ledge angle (resp., $p = 0.04$ and $p = 0.07$), and the longitudinal angle (resp., $p = 0.039$ and $p < 0.001$). The median and interquartile range (IQR) of the outcome measures for both experiments and simulations are shown in Table 3. Fig. 9 contains the Bland-Altman plots showing the level of agreement between the experiment and the model setup per outcome measure.

4. Discussion

In the current study, we explored the applicability of the Madymo Pedestrian Model for forensic reconstruction of falls using numerical modelling. The experiments with volunteers provided much insight into the fall kinematics of people dropping off heights. Despite exploring an extensive set of modelling parameters, the results suggest that even the best-performing model settings did not consistently replicate the experimentally recorded falling movements across multiple falling conditions and outcome measures (see Table 2 and Fig. 9).

The experiments showed large variations in falling kinematics and behaviour between participants. Initially, it was expected that the outcome measures of the different fall types (i.e., passive or pushed) in the experiment would differ significantly. However, a significant effect of fall type was found only in the longitudinal angle, and a significant effect of fall direction (i.e., backwards or sideways) was only found in the ledge angle. During the experimental trials, notable variations in falling behaviour were observed in the way participants actively corrected their falls. For example, some participants were startled during the fall and tried to maintain an upright position by actively adjusting their posture, likely in an attempt to prevent them from landing on their heads. The lack of the effects of fall type and fall direction on the outcome measures may be partially due to participants being aware that the fall was safe and knowing that and when the fall was going to happen. Furthermore, for safety reasons, the fall height was rather limited and no forward passive or pushed falls were conducted. As a result, the falling movements exhibited by participants during the experiment may differ from those in actual dangerous or lethal falls. A real situation might provoke an even stronger corrective response than was observed in the current experiments, especially in the pushed falls condition. These observations suggest that active behaviour may play a considerable role during dropping off heights.

The Madymo Pedestrian Model is designed for high-impact studies for which a passive human model generally suffices [10]. Although not

Table 2
The root mean square error (RMSE) and completion rate (CR) for all model choices. The shaded cells show the model choice with the lowest RMSE for the fall type. Velocity placement *Pelvis* indicates application of $v_{pelvis} + \omega_{pelvis}$ and velocity placement *All* indicates application of $v_{pelvis} + \omega_{pelvis} + \omega_{joints}$.

	Stabilised	Velocity placement	Backwards RMSE (°)	Backwards CR (%)	Sideways RMSE (°)	Sideways CR (%)
Passive T1	No	Pelvis	83.15	96%	77.45	100%
		All	86.01	96%	92.27	100%
	Yes	Pelvis	66.29	96%	79.51	96%
		All	64.17	92%	93.11	96%
Passive T2	No	Pelvis	44.97	96%	51.6	100%
		All	48.57	96%	53.31	100%
	Yes	Pelvis	41.75	96%	46.96	100%
		All	51.97	96%	48.81	100%
Passive T3	No	Pelvis	46.04	100%	50.64	100%
		All	50.31	96%	50.63	96%
	Yes	Pelvis	37.91	100%	51.31	100%
		All	47.56	100%	50.16	92%
Passive T4	No	Pelvis	39.75	84%	61.47	76%
		All	52.44	88%	83.15	56%
	Yes	Pelvis	42.02	64%	76.04	56%
		All	49.27	64%	83.28	48%
	Stabilised	Locking type	RMSE	CR	RMSE	CR
Pushed	No	LB lock	81.44	100%	70.43	96%
		All lock	77.77	100%	69.49	100%
	Yes	LB lock	51.16	100%	62.85	96%
		All lock	45.52	100%	64.84	100%

designed for forensic reconstruction of falls, it would have been of great practical benefit if such passive models could directly be applied to such problems. However, the current results suggest that this is not the case even though a number of different modelling choices were tried. Due to the passive nature of the MPM, the dummy was inherently unstable. To balance the dummy, one of the modelling choices was to configure the dummies with a stabilising ankle torque. Implementation of this stabilising ankle torque only seemed to considerably decrease the RMSE with respect to the original ankle stiffness in the pushed trial simulations and in the passive trial simulation for T1. This could be due to participants leaning slightly forward at T1, positioned with their centre of gravity towards their toes, causing Madymo dummies to also lean forward. This posture creates a mechanical advantage for maintaining the vertical angle due to the distribution of the mass of the foot and the model and the direction of the applied torque. In contrast, for *T2-T4* the Madymo dummies were already tipping towards the fall direction, leading to a mechanical disadvantage in maintaining the vertical angle. It is important to note that the stabilizing torque's magnitude was derived from a simplified model treating the dummy's mass as a point at the pelvis centre. This approach may not accurately reflect the true centre of mass, especially when upper body movement is involved, possibly leading to inaccurate estimation of the required stabilizing torque.

For the passive falls the points of application of the initial velocity varied and the time when the initial posture and velocity were extracted (*T1-T4*) were also varied. The points of application of the initial velocity did not seem to have a considerable effect on the RMSE. Since the lower body was locked, the angular joint velocity only influenced the movements of the upper body. These results suggest that either the joint angular velocities of the participants in the experiment were low or that the upper body joint movements for passive falls did not have an impact on the Madymo dummy in terms of the defined outcome measures. The time on which the initial posture and velocity were extracted (*T1-T4*) did seem to have an effect on the RMSE. It was expected that the RMSE would decrease as a later time instance was used. Although the RMSE for T1 is considerably higher than the RMSE for T2-T4, overall, there is not much difference in RMSE between T2-T4. Surprisingly, the RMSE for T4 for the sideways seems to be increased. However, the completion rate for T4 in general is low, likely due to numerical instabilities or problems with the initial posture. For the pushed falls the number of locked joints was varied to check whether there was any absorption of the push force in the upper body. This seemed to only have a minor contribution as the

Table 3
The median (IQR) of the outcome measures for the passive and pushed falls for both experiments and simulations.

			Vertical angle (°)	Ledge angle (°)	Longitudinal angle (°)
Passive	Backwards (PaBa)	Exp	47.28 (27.33, 77.97)	91.54 (88.68, 96.22)	3.09 (-5.04, 13.78)
		Sim	98.60 (90.32, 109.44)	94.28 (89.74, 97.41)	5.66 (-7.01, 17.72)
	Sideways (PaSi)	Exp	49.60 (29.57, 71.05)	87.20 (75.03, 97.41)	-18.46 (-35.17, 19.18)
		Sim	104.67 (94.43, 110.93)	84.34 (65.44, 100.82)	-45.58 (-59.05, -15.23)
Pushed	Backwards (PaBa)	Exp	37.76 (5.73, 80.66)	92.53 (88.60, 96.07)	6.56 (1.07, 10.34)
		Sim	105.02 (97.27, 109.07)	96.72 (92.62, 101.01)	10.38 (0.22, 24.71)
	Sideways (PaSi)	Exp	24.87 (13.93, 47.79)	86.45 (80.84, 93.77)	3.72 (-9.61, 17.99)
		Sim	113.66 (97.17, 119.95)	61.51 (52.70, 78.17)	-58.55 (-78.66, -25.72)

type of locking did not considerably affect the RMSE.

The best-performing model settings were selected based on the RMSE. The adjusted passive MPM with these settings does not — and if this was indeed the best achievable result, may not even be able to — accurately reproduce the experimentally recorded falling movements across multiple falling conditions and outcome measures. The differences between the Madymo dummies and the experimental data were most prominent in the vertical angle. The average vertical angles of the Madymo dummies were between 45° to 55° larger than those found in the experiment. Since the model is completely passive, the observed active corrective behaviour of the participants in the experiment is not reproduced in the model. Therefore, before the Madymo Pedestrian Model can be used in forensic fall reconstruction, it should be adapted such that it can balance itself well and predict corrective movements by, for example, incorporating a form of active control. In addition, it is desirable to have one model that can accurately predict passive and pushed (and active) falls.

In conclusion, while several of the used simulation setups produced outcome values that were quite close to those in the experiment, this held only for specific conditions, participants or outcomes. As such, the model in its current form is not yet suitable for reconstructing falls for forensic purposes and needs to be adapted. Hence, in court cases, any such simulation results should be considered with caution and with a thorough understanding of the modelling approach and its limitations. Particular attention should be paid to avoiding interpreting a coincidental match between model and reality as proof of its validity.

Funding

This research did not receive any specific grant from funding agencies in the public, commercial, or not-for-profit sectors.

CRediT authorship contribution statement

Arjo J. Loeve: Writing – review & editing, Supervision, Methodology, Formal analysis, Conceptualization. **Winfred Mugge:** Writing – review & editing, Methodology, Formal analysis, Conceptualization. **Jan Peter van Zandwijk:** Writing – review & editing, Supervision, Methodology, Investigation, Formal analysis, Conceptualization. **Kim Hutchinson:** Writing – review & editing, Visualization, Validation, Software, Methodology, Investigation, Formal analysis, Conceptualization. **Vera de Vette:** Writing – original draft, Visualization, Validation, Software, Methodology, Investigation, Formal analysis, Conceptualization.

Declaration of Competing Interest

The authors declare no conflicts of interest.

Appendix A

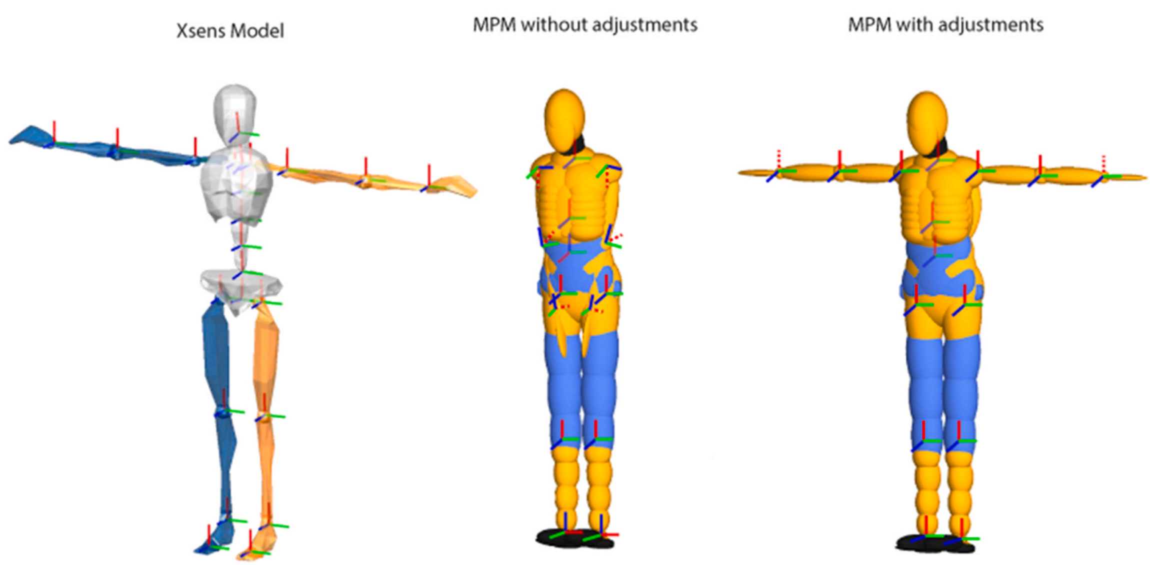

Figure A1. Xsens model, Madymo Pedestrian Model (MPM) without and with adjustments in neutral position. Joint coordinate systems are displayed per joint, blue indicating the x-axis, green indicating the y-axis and red indicating the z-axis. Unpermitted joint angle inputs are displayed using the dotted lines.

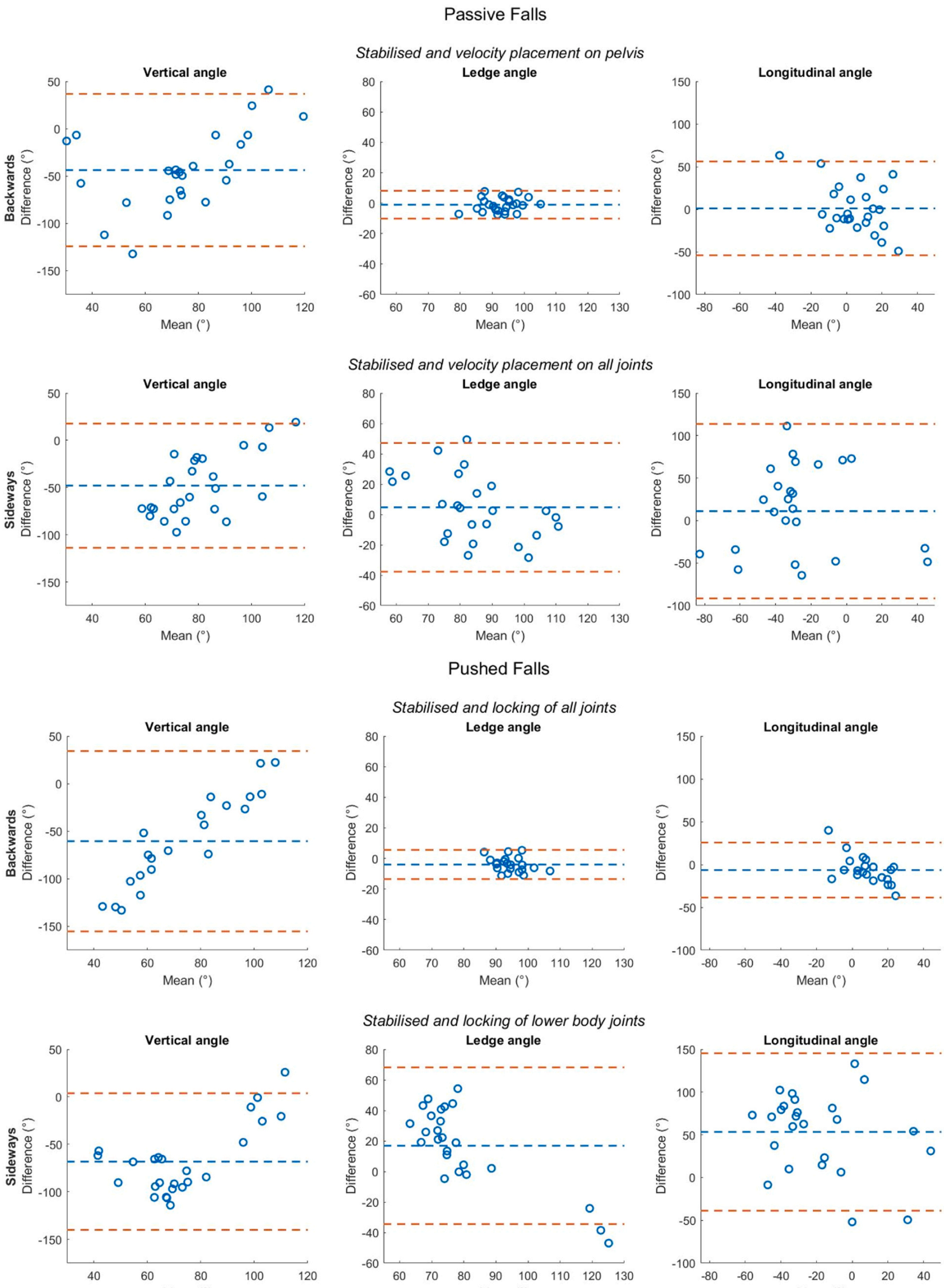

Fig. 9. Bland-Altman plots showing the level of agreement per outcome measure between the experimental data and the best simulation setup. The blue dashed line indicates the mean value of the difference, and the orange dashed lines represent the 95% limits of agreement. Positive values indicate that the simulation model underestimates the angle compared to experimental data, negative values indicate that the simulation model overestimates the angle compared to the experimental data.

Table A1
Adjustments of the Madymo Pedestrian Model explained per joint.

Joint	Previous configuration	Updated configuration	Transformation for Xsens Compatibility
Shoulder	Universal joint (2DOF)	Spherical joint (3DOF)	Rotation by 90° over the x-axis and subsequently over the y-axis. Removal of 98.35° offset around the x-axis and 9° offset around the y-axis such that the arms are parallel to the ground in the neutral position and align with Xsens neutral position.
Elbow	Universal joint (2DOF)	Spherical joint (3DOF)	Rotation by 90° over the y-axis and subsequently by -90° over the x-axis.
Lumbar low	N/A	N/A	Rotation by -90° over the y-axis.
Ankle	N/A	N/A	Rotation by 90° over both the x-axis and the y-axis.
Shoe	N/A	N/A	Adjust the sole of the front side of the shoe with 5.84° so that it is parallel to the heel of the shoe.

Appendix B

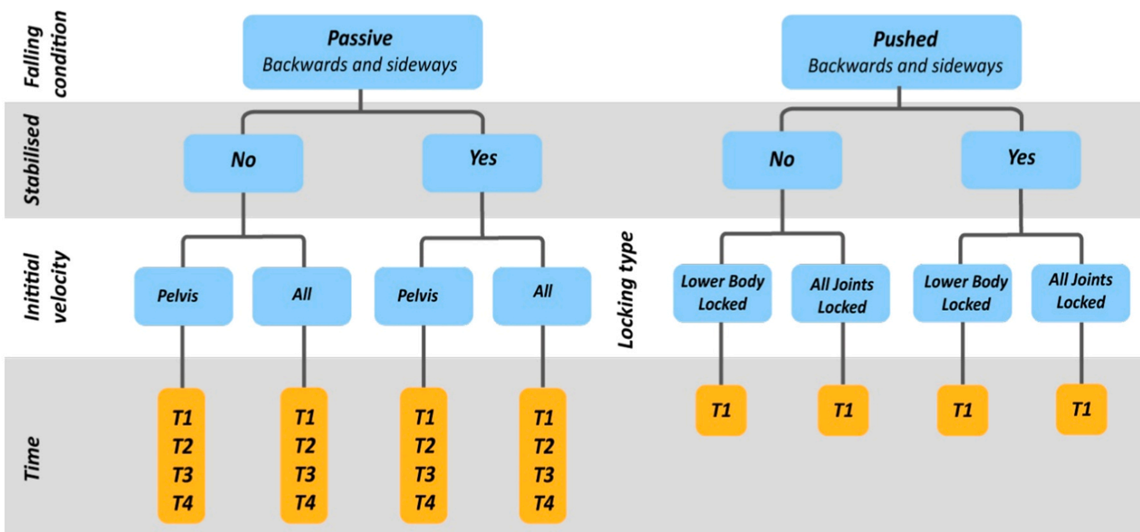

Figure A2. Overview of all the simulation setups that were executed, and the terms used to refer to them.

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