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Head impact kinematics and injury risks during E-scooter collisions against a curb

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ABSTRACT

E-scooters as a mode of transportation is rapidly growing in popularity. This study evaluates head impact conditions and injury risk associated with E-scooter crashes. A multibody model of E-scooter falls induced by wheel-curb collision was built and compared with an experimental E-scooter crash test. A total of 162 crash scenarios were simulated to assess the effect of fall conditions (E-scooter initial speed and inclination, obstacle orientation, and user size) on the head impact kinematics. The forehead hit the ground first in 44% of simulations. The average tangential and normal impact speeds were 3.5 m/s and 4.8 m/s respectively. Nearly 100% of simulations identified a risk of concussion (linear acceleration peak >82 g and rotational acceleration peak >6383 rad/s^2) and 90% of simulations suggested a risk of severe head injuries (HIC>700). This work provides preliminary data useful for the assessment and design of protective gears.

1. Introduction

Over the last 5 years, the increasing popularity of E-scooters in urban areas has been associated with a rapid increase in the incidence of injuries. The French Road Safety department reported a 40% increase in the number of E-scooter injuries in 2020 [1] versus the year before. Injuries caused by E-scooter accidents pose a significant financial burden for healthcare systems, with an average of 1000 euros incurred for each injury [2]. Nearly half of the victims in this context sustained head injuries [3,4]. The [3] reported that 48% of injured riders had lacerations or abrasions to the head, 15% experienced traumatic brain injuries such as concussions, subarachnoid haemorrhages or subdural hematoma, and 3% presented with a skull fracture [4]. showed that head injuries occur at a rate of 17.9% in the head and 33.3% in the face. E-scooter accidents are most often the result of individual falls on the road. This injury mechanism is cited by over 80% of victims in epidemiological studies [4–7]. Poor road conditions and loss of control associated with excessive speed are recognized as the main causes of falls [2,3].

Despite the high risk of injury for E-scooter users (112 injured/million kilometres) compared to motor vehicle users (0.65 injured/million kilometres)b [8], the accident biomechanics of this new means of transportation is understudied. Experimental E-scooter

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crashes are limited to a few videos of accidents with crash test dummies or volunteers. Ptak et al. [9] studied E-scooter crashes against a vehicle using numerical simulations and provided preliminary data on the head impact kinematics. They reported linear head accelerations between 30 g and 80 g during impact with the car. Posirisuk et al. [10] numerically reproduced E-scooter falls caused by potholes and identified head impact conditions. The average impact speed was 6.3 m/s and all head-ground impacts were oblique. However, the risk of injury and risk factors were not specifically studied. Further studies also need to investigate other types of individual E-scooter falls, such as falls induced by collisions with a curb.

The objective of this study is to assess the effect of crash conditions on head impact kinematics and the risk of head injury by numerically reproducing E-scooter crashes induced by a collision with a curb. The crash conditions studied are: the initial speed, the E-scooter inclination, the curb orientation, and the size and position of the rider.

2. Materials and methods

2.1. Development and evaluation of the E-scooter model

A multibody model of E-scooter falls caused by a collision with a curb was built on MADYMO software v2020.2 (Siemens Digital Industries Software, USA). The E-scooter model was developed based on the geometry, mass, and centre of gravity of a commercially available E-scooter (ESA 1919 EKFV, KSR Group GmbH) having the following characteristics: 1.08 m length, 1.14 m height, 24-cm diameter wheels, and a 13 kg weight. The E-scooter model was composed of one rigid body with six ellipsoids, and its mass was located at the centre of gravity calculated from its CAD model (Catia V5). All ellipsoids were considered rigid, except for the wheels, for which a damping coefficient of 200 kg/s and a contact force function (1200 N/mm) were defined to absorb some energy at the contact with the curb. The 50th percentile male pedestrian model (1.74 m, 75.7 kg) from MADYMO was chosen to represent a typical E-scooter rider. The model was originally developed and validated to assess the safety of vulnerable road users [11,12]. It has been widely used in the simulation of pedestrian, bicycle, and E-scooter accidents [9,10,12–14]. The human model was positioned on the E-scooter with the hands hooked on the handlebar, the torso bent slightly forward, and one foot on the front and the other at the back of the platform, which is a common posture on E-scooters. A coefficient of friction of 0.3 was selected to represent the contact between the rider's feet and the E-scooter's chassis, similarly to Shang et al. [15]. The hands of the model were attached to the handlebar with a mechanical joint that was deactivated when the force in the joint exceeded 0.01 N, to simulate the hand releasing the handlebar upon impact. At the beginning of a simulation, an initial velocity was applied to the pedestrian and E-scooter models. A curb represented by a rigid body with a rectangular ellipsoidal surface was placed 5 cm in front of the E-scooter to initiate the fall. The coefficient of friction was 0.5, between the E-scooter wheels and the ground to simulate a dry road, and 0.6, between the E-scooter rider and road surface, according to previous multibody studies involving vulnerable road users [10,13,16,17]. Simulations were run with a timestep of 2×10^{-5}s.

The definition of the contact force between the E-scooter wheels and the curb, as well as the validation of the model's kinematics during the fall, were based on an experimental E-scooter crash. The numerical model was compared to an experimental test reproducing an E-scooter forward fall with an anthropometric test device (ATD) conducted at DEKRA Automobil GmbH facilities. In that experiment, a sled test was used to launch a Hybrid III 50th Male ATD standing on an E-scooter (ESA 1919 EKFV, KSR Group GmbH) at a speed of 6 m/s. The hands of the ATD were initially closed around the handlebar. A metallic curb was placed perpendicular to the trajectory of the E-scooter, 50 cm after the launch of the E-scooter. The collision with the curb produced a forward fall of the ATD with a head impact to the ground. The head's experimental acceleration was measured from 3-axis accelerometers (sampling rate 20 kHz) placed at the centre of gravity of the ATD head. The fall was recorded with a high-speed camera at 1000fps (Redlake Motion Os7). The displacement and velocity of the head during falls were measured by tracking markers on the head in high-speed videos using the Kinovea v0.9.5 motion analysis software (http://www.kinovea.org). In order to evaluate the multibody model, the collision parameters of this experimental test (50th percentile male pedestrian, speed of 6 m/s, perpendicular curb and neutral inclination) were reproduced numerically.

In a first step, the frictionless contact between the E-scooter wheels and the curb, defined by a force/displacement curve, was adjusted to minimise the discrepancies between the numerical and experimental distance between the head impact location and the curb and duration of the fall. The fall duration studied was measured between the initial E-scooter/curb collision and the head impact against the ground. A deviation of 3% (±10 cm in distance and ±0.02 s in duration) was considered satisfactory. The obtained stiffness modulus of the E-scooter wheels was 1200 N/mm.

The global kinetics, the horizontal and vertical head displacements from the start of the simulation and the peak head acceleration were compared between simulations and experiments for validation purposes. In addition, the root mean square error (RMSE) (**Equation (1)**) of the displacements during the airborne phase of the simulation was calculated from the following equation:

$$\mathrm{RMSE} = \sqrt{\frac{\sum_{i=1}^{n}(dexp - dnum)^2}{n}} \quad (1)$$

where d_{exp} and d_{num} are the experimental and the numerical fall displacement (m) and n is the number of points along the curves (n = 15).

2.2. Parametric study of E-scooter falls

A parametric study was performed using a full factorial design of experiments to investigate the influence of fall conditions on the head impact kinematics and on the resulting risk of injury. Fig. 1 reports the parameters and corresponding levels of the study. The E-scooter speed (3 m/s, 6.5 m/s, and 10 m/s) and inclination (neutral, away from the obstacle or to the left, and towards the obstacle or to the right), curb orientation (22°, 45° and 90°), the rider size (5th, 50th, and 95th percentiles), and its positioning on the E-scooter (upright and bent) were varied from the initial E-scooter falls multibody model. The speed range was set to study the average speeds of use, i.e., 3.7 m/s according to Pestour [18], as well as the maximum speeds allowed in current regulations, i.e., from 5.5 m/s to 10 m/s [19–22]. The curb orientations reflect three typical angles of impact between a curb and an E-scooter. The morphological differences of adult riders [23] were considered by studying three sizes of human body models. Two postures were simulated: chest upright with the feet side by side or chest bent 10° with the feet one behind the other. These postures were identified in E-scooter crash videos found on the Internet (Bailly et al., 2022) and the two positions of the feet had been described by Garman et al. [24]. The three inclinations of the E-scooter represent either a destabilisation or a turn initiation prior to the impact [2]. The result was a full factorial design of 162 fall scenarios, simulated using MADYMO software v2020.2 (Siemens Digital Industries Software, USA).

The assessment of the head impact kinematics included the head impact location, the normal and tangential impact velocity, and the linear and rotational accelerations at impact. The head impact location was determined using a Finite Element mask dividing the head into 6 zones (forehead, face, back, sides, top, underside), which only served to determine the contact location, and did not have any contact stiffness. The head impact velocity and acceleration were directly extracted from the numerical simulation results and were filtered using a Butterworth low-pass filter with a 1000 Hz cut-off frequency, as recommended by the SAE J211 standard [25].

The risk of head injury was evaluated based on linear and rotational accelerations, and 3 injury criteria were calculated: Head Injury Criteria (HIC) **(Equation (2))** based on linear accelerations, Brain Injury Criteria (BrIC) **(Equation (3))** based on angular velocities, and the combined probability of concussion (CP) **(Equation (4))** calculated from both linear and rotational accelerations.

$$\mathrm{HIC} = \max\left[\frac{1}{t2 - t1}\int_{t1}^{t2} a(t)dt\right]^{2.5}(t2 - t1) \tag{2}$$

where a is the linear acceleration (g), and t2 and t1 are initial and final times (s) maximising the HIC [26].

$$\mathrm{BrIC} = \sqrt{\left(\frac{\omega_x}{66.25}\right)^2 + \left(\frac{\omega_y}{56.45}\right)^2 + \left(\frac{\omega_z}{42.87}\right)^2} \tag{3}$$

where ω_x, ω_y, ω_z are the maximum angular velocities along the x-, y-, z-axis, respectively (rad/s) [27]

$$CP = \frac{1}{1 + e^{-(-10.2 + 0.0433*a + 0.000873*\alpha - 0.000000920*a\alpha)}} \tag{4}$$

where a is the peak linear acceleration (g) and α the peak rotational acceleration (rad/s^2). [28].

The HIC_{15} was compared to the injury assessment reference value ($HIC_{15} = 700$) to assess a 5% risk of severe traumatic brain injury

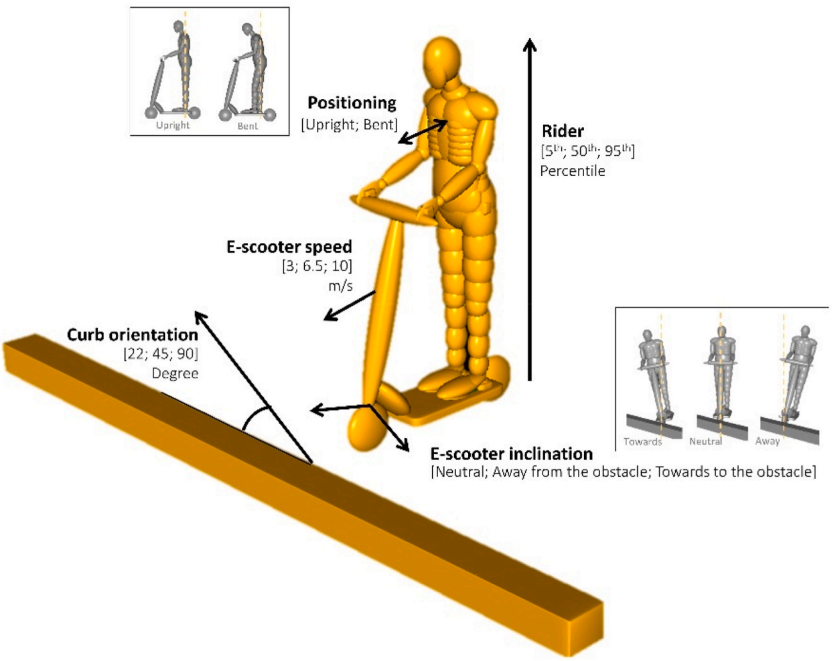

Fig. 1. E-scooter fall parameters.

(TBI) (AIS4+) [26]. The BrIC was compared to a threshold of 0.5 and 1, corresponding respectively to a 50% risk of mild and severe TBI [27]. The resulting linear and rotational head accelerations were compared to acceleration thresholds of 82G and 6383 rad/s^2, respectively, corresponding to a 50% risk of mild TBI (mTBI) according to previous impact reconstructions [29] and on-field acceleration measurements [30]. Finally, the head impact speed and the linear head acceleration were compared to the impact speed and acceleration threshold used in bicycle helmet standards as no specific helmet standard exists for E-scooters.

A factorial analysis of variance (ANOVA) was performed on Minitab v18 software (Minitab LLC, USA) to determine the effect of the studied parameters, with a significance level set at $p < 0.05$, on the outcome responses, i.e., impact velocities and accelerations, HIC and BrIC.

3. Results

3.1. Evaluation of the E-scooter model

The comparison of numerical and experimental E-scooter falls is presented in Fig. 2. Both fall kinematics show that after the E-scooter front wheel hits the curb, the rider is thrown forward. Both experimentally and numerically, the head/ground collision occurs at 0.56 s after the E-scooter hits the curb. The lower limbs are the first to hit the ground, followed by the thorax, the upper limbs, and finally, the head **(Fig. 2a).** In the experimental test, the hands remain on the handlebar longer than in the numerical model, in which the handlebar is released from the hand at the early stage of the collision, leading to a difference between the experimental and numerical harm kinematics. The fall distance between the head impact location and the curb is 3.30 m experimentally, and 3.24 m numerically, i.e., less than a 2% difference. The horizontal and vertical head displacements of the numerical model are consistent with those recorded on the ATD. The RMSE for the vertical head displacement is 0.20 m, and 0.22 m, for the horizontal displacement. The most significant differences are found at the beginning of the vertical displacement. The numerical rider fall starts earlier (100 ms after curb contact) than does the ATD in the experimental fall (200 ms) (Fig. 2b). The head impact of the numerical model also occurs 14 ms before the ATD, but their peak head accelerations are equivalent. The peak acceleration measured from the accelerometers positioned at the centre of the head was 576 g and 580 g for the experimental and numerical head rider model, respectively, i.e. less than a 1% difference (Fig. 2c).

3.2. Parametric study of E-scooter falls

Among the 162 simulations of the parametric study, different fall kinematics were observed at low (3 m/s) and high speeds (6.5 m/s and 10 m/s) (Fig. 3). At low speed, the rider falls close to the obstacle, 2.1 [1.4–3.0] m from the obstacle, and maintains contact with the E-scooter during the fall. At high speed, the higher the velocity, the farther the rider is launched from the obstacle, 3.8 [2.3–4.9] m

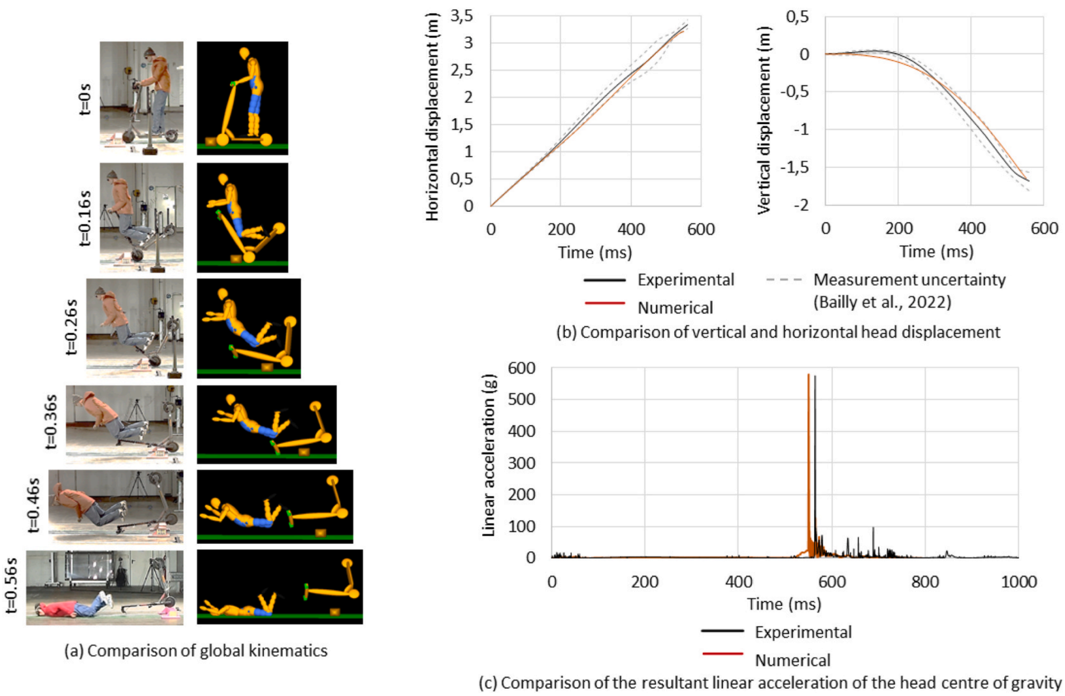

Fig. 2. Comparison of numerical and experimental falls: (a) global kinematics, (b) horizontal and vertical head displacements, (c) linear acceleration.

for 6.5 m/s and 5.7 [3.4–8.5] m for 10 m/s. The head hit the ground in all simulations, but it was the first hit zone in only 6% of the cases, while the lower and upper limbs and torso hit the ground first in 57%, 34% and 1% of the fall simulations, respectively.

The head impact locations are shown on Fig. 4a. The main head impact locations were the forehead (44%) and the face (18%). The other head areas individually accounted for 15% or less of all impacts. All the head impacts were oblique. For all simulations, the average impact speed was 6.5 m/s, decomposed as average tangential and normal impact velocities of 3.5 m/s and 4.8 m/s, respectively (Fig. 4b). The average normal head impact velocity was nearly the same as that in bicycle helmet standards evaluations (ASTM F1447-18, NF EN 1078) [31,32]. However, in 40% of all fall scenarios, the normal head impact velocity was greater than 5.42 m/s, i.e., the impact velocity used in helmet standard tests.

The risk of head injury was assessed for all simulated fall scenarios. Head accelerations and injury criteria with respect to injury thresholds are reported in Fig. 5. The average peak linear acceleration was 571 g and the average peak rotational acceleration was 25 580 rad/s^2. The thresholds corresponding to a 50% risk of mTBI for linear acceleration (82 g) and rotational acceleration (6383 rad/s^2) were exceeded in 99% and 96% of the simulations, respectively. According to the combined probability of concussion (CP) of Rowson and Duma [28] **(Equation (4))**, the risk of concussion was greater than 95% in more than 95% of cases. The average HIC was 5999. More specifically, the average HIC was 3545, 7553 and 6900 for an initial E-scooter speed of 3 m/s, 6.5 m/s and 10 m/s, respectively. Thus, increasing the riding speed before the crash from 3 m/s to 6.5 m/s doubled the average HIC. In 88% of cases, the HIC was greater than 700, which corresponds to 5% of all severe injuries (AIS>4) [26]. The average BrIC value was 0.5, corresponding to a 50% risk of moderate injuries (>AIS 2) [27]. At a riding speed of 3 m/s, less than 3% of BrIC values were above 0.5, while it was 41% at higher speeds. In less than 1% of all simulated crashes, the BrIC was above 1, corresponding to 50% of severe injuries (>AIS4) [27]. The acceleration thresholds in bicycle helmets standards of 250 g [31] and 300 g [32], respectively, were exceeded in 88% and 80% of the simulations (Fig. 5).

The ANOVA (Table 1) quantifies the influence of the crash conditions on injury parameters. The initial E-scooter velocity was the main factor explaining the impact speed and injury metrics: increasing the riding speed significantly increased the head impact speed (especially tangential velocity) as well as the values of all injury predictors. The inclination of the E-scooter away from the obstacle also significantly increased the normal impact speed and all the injury risk predictors (Fig. 5). Finally, the rider's size significantly increased the normal head impact velocity and reduced the BrIC, and the E-scooter inclination affected the normal impact speed, the HIC, the Rotational acceleration and the BrIC.

4. Discussion

In this study, a multibody model of E-scooter falls caused by a collision with a curb was developed and validated against an experimental E-scooter fall. The model was used to assess head impact conditions, the risk of head injury and the effect of crash conditions on injury risks. A high risk of head injury was found in most of the simulated scenarios and the riding speed was the main factor explaining the head impact speed and the risk of injury.

The head hit the ground in all simulations, but the lower and upper limbs were the first body parts to hit the ground in most of the cases. This is consistent with epidemiological data showing that 48% of victims suffer multiple injuries in E-scooter accidents [4] and that injuries to the extremities account for 1/3 of all injuries [33]. Any part of the body impacting the ground first may have a protective effect on head acceleration and on the risk of head injury. For instance, for a collision at 10 m/s, the head acceleration was 30% higher when the head was the first to impact the ground, as compared to when the upper hands impacted the ground first. Because the current model did not include muscle activity, protective reflexes of the upper arm were not modelled. These reflexes are expected to affect the upper hand kinematic as well as the head acceleration and risk of injury to the head. In the simulation, the head most often hit the ground in the forehead (44%) and face (18%) areas, which is consistent with the findings of Posirisuk et al. [10], who observed that 56% of impacts occurred to the forehead and 44% to the face, where there was no helmet protection.

The average resulting head impact velocity was 6.5 m/s, again in agreement with Posirisuk et al. [10], who found a head/ground impact velocity of 6.3 ± 1.4 m/s. The normal impact speeds were between 3.5 m/s and 6.2 m/s (1st and 3rd quartile), and the average was 4.8 m/s, which is also consistent with Posirisuk et al. [10]. This suggests a similar head injury mechanism between E-scooter falls induced by a curb and by a pothole. The impact velocity used in the helmet standard (5.42 m/s) is within the range of normal head impact velocities identified in the simulations of E-scooter falls. This suggests that the current bicycle helmet standards (ASTM F1447-18, 2018; NF EN 1078) are well adapted to evaluate helmets for such crashes [31,32]. However, current standards only test helmets with a linear impact normal to the head. In both our simulations and those of Posirisuk et al. [10], the head impact with the ground was oblique, with a significant tangential impact speed (3.5 m/s on average). This is problematic as a tangential impact speed is strongly

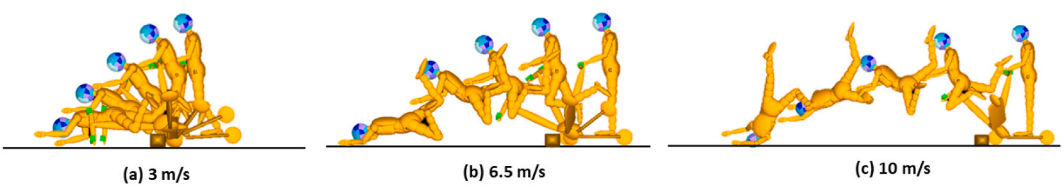

Fig. 3. Typical fall kinematics for the three initial E-scooter speeds tested: 3 m/s (a), 6.5 m/s (b) and 10 m/s (c). The blue color represents the Finite Element mask which is used to determine contact location and has no contact stiffness.

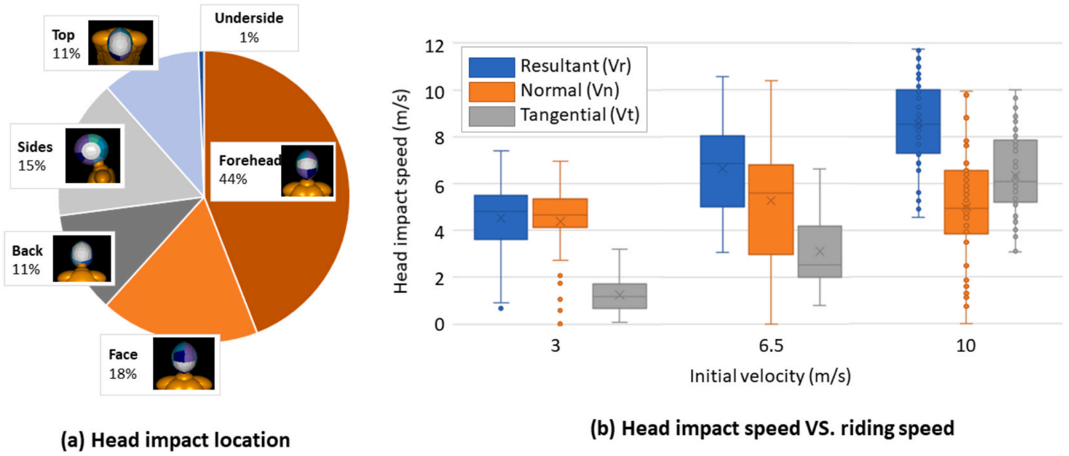

Fig. 4. Distribution of head impact: (a) location and (b) velocity, among the 162 E-scooter crash scenarios simulated.

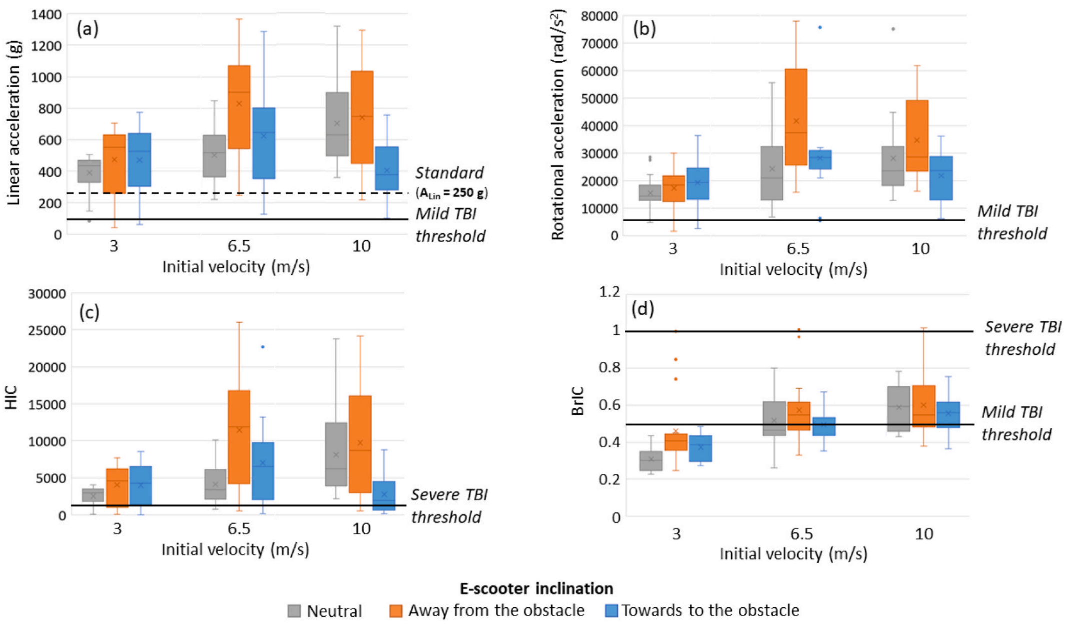

Fig. 5. Average peak linear (a) and rotational (b) accelerations, HIC (c) and BrIC (d) for different initial speeds and inclinations of the scooter.

linked with rotational head accelerations, and in our simulations, the levels of rotational acceleration were well above the thresholds for mild head injuries [30]. This suggests a need to supplement the current helmet evaluation with oblique impact conditions and the evaluation of head rotations [34,35].

A risk of mTBI was identified in almost all the E-scooter fall simulations using both linear and angular accelerations. This is consistent with epidemiological studies showing that head injuries are common in E-scooter accidents [3,7]. A risk of severe head injury was also predicted using the HIC in 88% of the simulated falls. This was not expected as severe head injury is a rare event in E-scooter falls, and usually, less than 3% of injured riders suffer an intracranial haemorrhage after an accident [5,7]. The high risk of severe head injury found in our simulations might be attributable to the fact that falls induced by curbs are among the most severe types of crashes. It might also be due to a lack of protective reflexes in our models as suggested by Ref. [36] for bicycle accidents. Nevertheless, with linear head accelerations almost always above the current thresholds in helmet evaluation standards (ASTM F1447-18, 2018; NF EN 1078) [31,32], this study strongly suggests the importance of wearing a helmet to reduce the risk of head injuries. The current lack of helmet use with E-scooters [6] must thus be addressed, for instance, through prevention campaigns.

This parametric study showed that the most important examined factors affecting the risks of head injury are the initial speed of the E-scooter, followed by the E-scooter inclination. Increasing the E-scooter riding speed from 6.5 m/s to 10 m/s led on average to an increase of 29% in the resulting head impact speed on the ground (Fig. 4). This is consistent with the study of Posirisuk et al. [10], which showed that increasing the initial E-scooter speed from 5.5 m/s to 8.3 m/s led to a 14% increase in the average head impact

Table 1
Contribution of factors and their interactions on responses (%).

	Normal impact velocity	Tangential impact velocity	Linear acceleration	HIC (Head Injury Criteria)	Rotational acceleration	BrIC (Brain Injury Criteria)
Linear	**12.87**	**73.11**	**20.16**	**22.02**	**25.44**	**41.84**
E-scooter speed	3.79*	70.02*	10.23*	9.58*	16.48*	27.99*
E-scooter inclination	4.25*	0.71	7.75*	9.29*	7.22*	4.3*
Rider size	2.9*	0.64	0.62	1.74	0.49	7.92*
Curb orientation	0.59	1.37*	0.48	0.71	1.05	0.09
Rider positioning	1.35	0.36	1.09	0.7	0.19	1.54*
2-factor interaction	**30.96**	**7.11**	**31.75**	**32.4**	**26.7**	**19.91**
Rider*inclination	9.55*	0.32	7.6*	9.19*	6.29*	3.26*
Speed*inclination	7.64*	0.73	9.83*	9.93*	5.85*	2.38
Rider*positioning	3.28*	0.45	4.41*	2.92*	3.01*	4.35*
Orientation*inclination	2.72	0.89	2.89	2.91	4.13*	3.66*
Orientation*positioning	1.36	0.03	2.63*	2.29	0.99	0.6
Positioning*inclination	0.64	0.08	0.18	0.09	0.01	2.12*
Orientation*rider	1.73	1.8*	0.82	0.64	1.37	1.19
Speed*orientation	2.91	1.01	2.59	3.45	1.29	1.49
Speed*rider	0.47	1.57	0.55	0.79	3.57	0.5
Speed*positioning	0.64	0.16	0.23	0.19	0.19	0.37
Interaction >2 factors	**56.17**	**19.78**	**48.09**	**45.58**	**47.86**	**38.25**

Statistically significant factor for $\alpha = 0.05$; color code: red = large contribution >10%; yellow = medium contribution >5%; green = small contribution <5%, No color = effect not statistically significant.

speed. It is also consistent with the general literature, which shows that speed is an aggravating factor in road accidents [10,36]. Interestingly, in the case of the crash scenarios where the E-scooter was inclined toward the obstacle, head acceleration and HIC were lower for an initial speed of 10 m/s compared to 6.5 m/s. This was because the highest speed was associated with longer falls and higher tangential head impact velocities, but also with lower normal head impact velocities compared to 6.5 m/s (Fig. 4). E-scooter inclination away from the obstacle also significantly increased linear and rotational accelerations versus a neutral orientation, particularly at high speed. Rider size, in interactions with the initial speed and the E-scooter inclination, significantly affect the risk of head injury. Posirisuk et al. [10] also found that the size of the rider may increase the head speed and force on impact, but did not consider the rotational head injury criteria in their study. Both studies used an adult model with large anthropometric variability (5th, 50th, and 95th percentiles), but no children were simulated. Current regulations allow E-scooters to be used from the age of 12 in some countries [19–22]. E-scooter falls with children or adolescent driver models should be simulated in further studies to generalize the current findings.

This study presents some limitations. First, only one type of common E-scooter fall was simulated, namely, a fall caused by a collision with a curb. This fall is common [4,5,7], but other types of falls and collisions against obstacles or vehicles must be studied to have an overall view of head impact conditions in E-scooter crashes. In particular, future work should focus on collisions with vehicles [9], which are rare, but generally more serious and often fatal for E-scooter riders [37]. Second, not all crash parameters were studied, and we might have overlooked some effects. For instance, the size of the curb, the friction between the curb and the wheel as well as the E-scooter dimensions and weight might affect the fall kinematics; Xu et al. [14] recommended using a heavier scooter, while Posirisuk et al. [10] suggested to use larger wheels to improve safety. Further work is thus needed to investigate the influence of the E-scooter geometry and weight on injury risk. Also, we chose to use the fast and robust multibody simulation method to access the overall fall kinematics and impact conditions in various fall scenarios, and to use published thresholds of head accelerations to predict head injury. However, the modelling provided no information on local brain loadings or on the effect of helmet use on injury occurrence. Also, as there is no clear consensus on the injury threshold for linear head acceleration, the predicted injury risks need to be considered with caution. In future work, the use of human finite element models would enable to access this information for a given crash [38–40]. Another limitation of this study is that the numerical model did not include muscle activity. In real-world falls, riders may have protective reflexes and may actively use their arms to protect against impacts. This behaviour may help reduce the head impact severity and the risk of injury. Thus, our simulation may represent the worst-case scenarios, where, for some reason (e.g., alcohol and drug consumption), the protective reflex did not occur. Further work may look at active models like those presented by Meijer et al. [41] to reproduce the protective reflex during the fall and to improve the prediction of injury risks. Finally, while some head-impacted zones seem to be included in the coverage zone of the helmet, further studies should analyse precisely which head zones are protected and which type of helmet is most effective.

To conclude, our objective was to assess the head impact conditions and injury risk in a typical E-scooter accident. This paper presents the development and validation of a multibody model of E-scooter falls and the simulation of 162 fall scenarios involving a collision with a curb. We confirmed that E-scooter users are vulnerable, with high levels of head acceleration at impact, leading to a significant risk of mild and severe head injury. Our study provides preliminary information that can be used to improve the prevention and protection of head injury of E-scooter users. Reducing the riding speed can be an interesting prevention strategy as speed is a key factor of the risk of injury during the fall. The evaluation and design of protection gears must consider realistic impact conditions, especially the effect of both normal and tangential impact velocity.

Author contribution statement

Marion Fournier: Conceived and designed the experiments, Performed the experiments, Analyzed and interpreted the data, Wrote the paper. Nicolas Bailly, Yvan Petit: Conceived and designed the experiments, Analyzed and interpreted the data, Wrote the paper. Andreas Schäuble: Conceived and designed the experiments, Contributed reagents, materials, analysis tools or data, Wrote the paper

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Data availability statement

Data will be made available on request.

Declaration of competing interest

The authors declare the following financial interests/personal relationships which may be considered as potential competing interests:Yvan Petit reports financial support was provided by Natural Sciences and Engineering Research Council of Canada.

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