Head impact conditions in case of equestrian accident

Nicolas Bourdet, Rémy Willinger

Abstract In France, the number of people practicing horse-riding represents 3.6% of the population. The mortality rate varies between 0.6 and 1.7 per 1 million people and between 7.8 and 10 per 100,000 riders. The most common injury mechanism is falls from the horse, and the head is one of the most commonly injured parts. Currently, there is very little information available concerning the head impact condition for this kind of accident. The objective of the present work, therefore, is to identify the initial condition of head impact in the case of a horse rider experiencing a fall accident. A parametric study using multibody modelling to simulate a number of virtual accidents based on detailed real-world situations allowed us to propose realistic rider's head impact conditions in terms of normal and tangential velocity. Five parameters, such as the human posture, the initial horse velocity, horse kinematic, the orientation of the falls and the human size, have been varied. The results showed three main impact areas: frontal; parieto-occipital; and temporo-facial. The head impact velocities typically range from 6.6m/s to 7.5m/s, with an inclination versus normal to surface between 20deg and 30deg. These results will contribute to an improvement in standard tests for equestrian helmets, especially introducing motivated tangential impacts conditions.

Keywords horse rider, fall accident, multibody simulations, head impact conditions.

I. INTRODUCTION

In 2013, the French equestrian federation (FFE) counted 700,000 licenses, i.e. 1% of the French population. Among these licensees, 160,000 equestrians participated in contests [1]. Currently, the number of people in France practicing horse-riding is about 2 million, i.e. 3.6% of the population. The mortality rate varies between 0.6 and 1.7 per 1 million people and between 7.8 and 10 per 100,000 horse riders. In the general population, the helmet is worn by only 9–20% of victims [2-4]. According to Pounder (1984) [5], nine types of mechanism can cause trauma for the horse rider: fall from horse; crushed by horse; take a shoe shot; be struck by an object (branch, tree); get stuck in the stirrup; take a lanyard shot; take a horse whim; be stepped on by the horse; be bitten by the horse. The most common mechanism is fall from horse, which represents 60–87% of accidents [2][4][6-10]. The second mechanism is crushing by the horse after a fall (11–16%).

The most common injured parts are located on the upper body (head and upper extremity) and represent 30%, as reported by Ball et al. in 2007 [2] and Cripps in 2000 [11], as well as by Moss et al. in 2002 [7]. Regarding injuries, musculoskeletal injuries are the most common. But head injuries cause the most serious injuries in 55–100% of cases, including neurological damage and intracranial injuries (17–25%). These lesions are responsible for the death of a rider or, in most cases, for serious permanent injuries [6][12].

In the specific case of horse-riding contests, accident mechanisms are falls or ejection from horse and taking a shoe shot. From studies conducted by Balendra et al. in 2007 and 2008 [13-14], it was found that falls from horse are more common for non-professional riders than for professionals, which means they have more risk of serious injury. According to Waller in 2000 [15], injuries occur in 44% of cases after ejection from the horse, in 10% of cases after being crushed by the horse and in 7% of cases after fall alone from the horse. Also in his study, Waller adds that 14% of injuries occur in the turns and 16% on the straight. In the professional area, the most injured body parts during an accident are the head, shoulders and chest. According to Waller [15], insurance data collected in the USA shows that the majority of lesions are localised on the head and neck. Moreover, fractures are the most common injuries followed by brain and cervical injuries, as also reported by Balendra (2008) [14].

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For head protection, the equestrian helmets are certified according to EN 1384 standard [16] in Europe. The shock absorption of the standard test consists of a free fall of a helmeted headform impacting a flat, fixed steel anvil, with a drop velocity of 5.42 m/s. The pass/fail criterion is the headform acceleration, which must not exceed 250 g and the duration over 150 g must not exceed 5 m/s. This certification test is similar to the standard for bicycle helmets, EN 1078 [17]. The main criticism of these standards is that the tangential component of the impact velocity vector is not taken into account. Indeed, these aspects lead to an angular acceleration of the head and potentially create intracerebral lesions, as explained in the studies by Gennarelli et al. in 1972 and 1987 [18-19]. Whether during a bicyclist's accident, a pedestrian accident or a horse-riding accident, the head impact conditions in real-world situations are unknown.

Several studies investigated the kinematics of the victims by simulating accidents from detailed accident cases, as in the studies of Singh et al. (2007) [20], who studied accidents between cyclist and bus, while Short et al. (2006) [21] investigated pedestrian and cyclist collisions, and Maki et al. (2001; 2003) [22-23] and Serre et al. (2007) [24] investigated cyclist and car accidents. More recently, in 2011, Bourdet et al. [25] simulated real bicyclists accidents in order to extract the head impact condition and better understand the behaviour of the bicycle helmet during impact. Moreover, the authors have simulated a large number of virtual bicyclist accidents in order to estimate the head impact conditions in case of a single fall, as well as the most frequently impacted impact area [26]. The results demonstrated that head impact points are very often located around and under the helmet rim, that the normal head initial speed is close to 5.5 m/s and that the head velocity presents a significant tangential component.

Regarding the horse falls, very few authors have focused on the head impact conditions. Avanessian et al. (1994) [27] reconstructed typical cases of horse-riding accidents experienced by polo players, when the horse suffered a heart attack and fell to the ground. The objective of this work was to determine force levels on the rider's head during and after body impact with the ground.

The aim of the present study is to further investigate the horse-rider kinematics during falls, especially concerning the unhelmeted head impact conditions just before contact with the ground as well as the impact point on the head compared with the current horse helmet standard tests. A first step was to analyse a large number of real accidents and to classify the different falls based on the video observation. A second step was to establish and to validate a horse-rider multibody model and to conduct a parametric study on fall situations in order to evaluate the head impact velocity vector and head impact location at the time of impact with the ground.

II. METHODS

A total of 23 real-world accident cases were analysed, based on videos provided by France Galop. Two main types of fall were identified:

- the typical fall of forward ejection (Fig. 1);
- the imbalance causing a more lateral fall (0).

From these videos, the majority of falls are caused by a sudden tilting of the horse, causing forward ejection of the jockey, so that the jockey's head impacts onto the ground. Among these accidents, 80% of the head trauma caused an injury less than AIS2.

Fig. 1. Illustrations extracted from the 23 real-world accident cases in the situation of forward ejection fall.

Fig. 2. Illustration extracted from the 23 real-world accident cases in the situation of imbalance lateral fall.

As the videos were recorded for TV show, the resolutions of pictures are too much poor and it was not possible to track points to extract any velocities or displacements.

Based on the observation of these falls, the second step was to calculate the jockey kinematics for a very large number of virtual falls by varying parameters such as the horizontal speed of the horse, the orientation of the horse, the kinematics of horse during falling (angle and rotation speed), the initial posture of the jockey as well as his size. For each case, the initial conditions of the head impact in terms of normal speed and tangential speed, as well as the location of the impact point, will be extracted. In order to evaluate accurately the initial head velocity and position just before the impact, it is necessary to simulate the human body kinematic properly~~,~~ via multi-body modelling. Madymo® is a software, from TNO Automotive, that uses multibody computation dedicated to human kinematic analysis. The principle of solving multibody system is to define a set of rigid bodies represented by ellipsoids and connected by joints. Unlike finite elements (FE) methods, contact between two bodies is not a deformable surface but a penetration force defined by a function. The computational time for this multibody approach is strongly reduced in comparison with FE simulation.

For the parametric study, the human model representing the horse-rider is the TNO human model. This model was developed by TNO in 2001. It is described in the studies Hoof et al. (2003) [28] and De Lange et al. (2005) [29]. The model consists of 64 ellipsoids attached to 52 rigid bodies. These rigid bodies are interconnected by kinematic connections, such as spherical, translational and revolution joints. The average model corresponds to the 50th percentile male. The size is 1.74 m, with a mass of 75.7 kg. Inertial values of the different body segments come from the study of Schneider et al. (1983) [30]. The connection stiffness was obtained through tests on volunteers and cadavers.

The horse model is developed using ellipsoids in such a way that the geometry is respected, as depicted in 0. The head-neck system of the horse is tied to the trunk by a rotation joint just to position the head in an initial position. Thus, the head can be positioned at different angles, depending on the posture of the horse at the time of accident. Three rotation joints are implemented in each leg, giving the possibility to move each part. All the joints used in this horse model were blocked during the simulations. The horse model is used only as a catapult for the jockey.

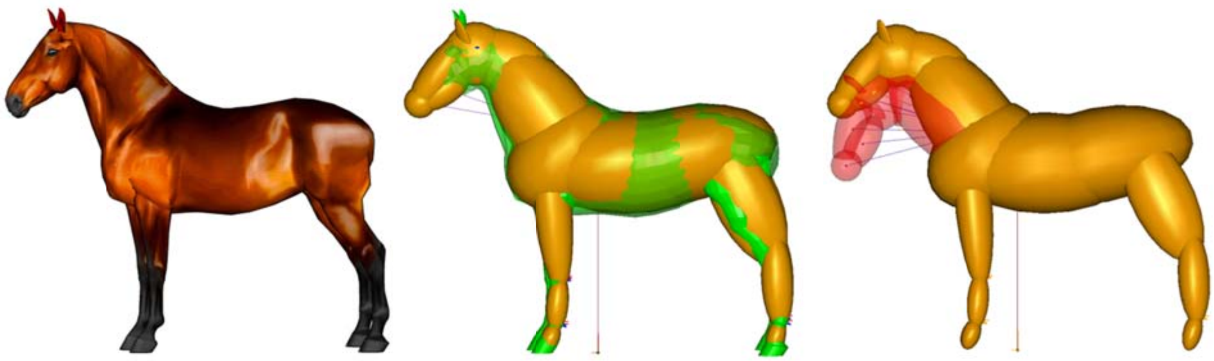


Fig. 3. Representation of the horse geometry available (*left*), the multibody modelling superimposed on the geometry (*middle*) and the neck articulation (*right*).

An important hypothesis during the simulation is that the horse is not deformed during the accident. Therefore, the validation of the horse-rider model can be established through simplified experimental falls, as follows. To validate the kinematic behaviour of the coupled horse-rider model, experimental tests were carried out with a volunteer on a simulator device of a jockey's ejection. The device uses a mechanical horse and simulates a forward tipping configuration. The volunteer is ejected forward and falls on a landing mat, as illustrated in 0 and 0. During the fall, the horse's and the volunteer's kinematics are recorded using a high-speed camera with 500 FPS and at a resolution of 1696x1710 px. The horse kinematics was extracted in terms of displacement and rotation as a function of time and implemented in the multibody model of the horse, as illustrated in 0, in order to procure the initial conditions of the experimental horse motion.

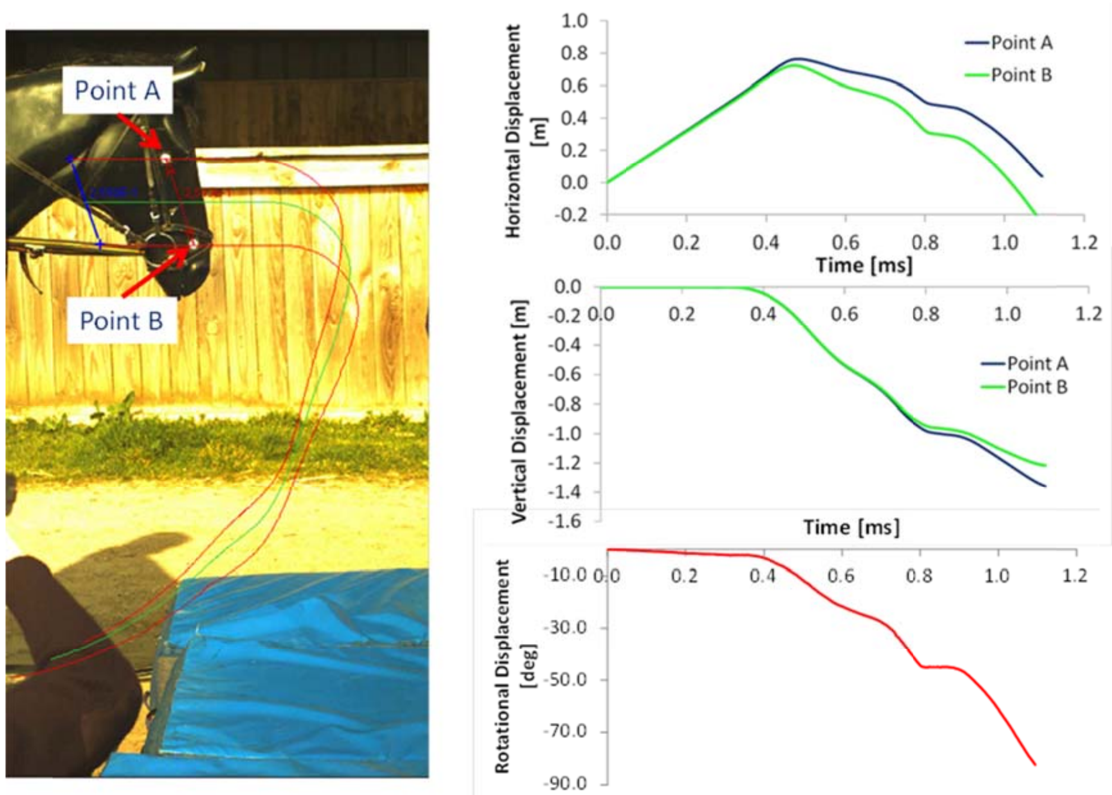


Fig. 4. Representation of the experimental horse kinematics in terms of linear and rotational displacement extracted from the high-speed video. This constitutes the initial conditions of the model validation.

0 shows the kinematics of the volunteer and the human body model in the same initial conditions. The kinematic of the model is comparable with that of the volunteer kinematics during the experiment. The superimposition of the experimental volunteer head displacement curves with the human model's head is illustrated in 0. The head velocity upon contact is 5.2 m/s for the volunteer, and the head impact speed obtained with the simulation is 5.6 m/s. It can be concluded that the simulation results are in accordance with the experiments, thus validating the horse-rider multibody model.

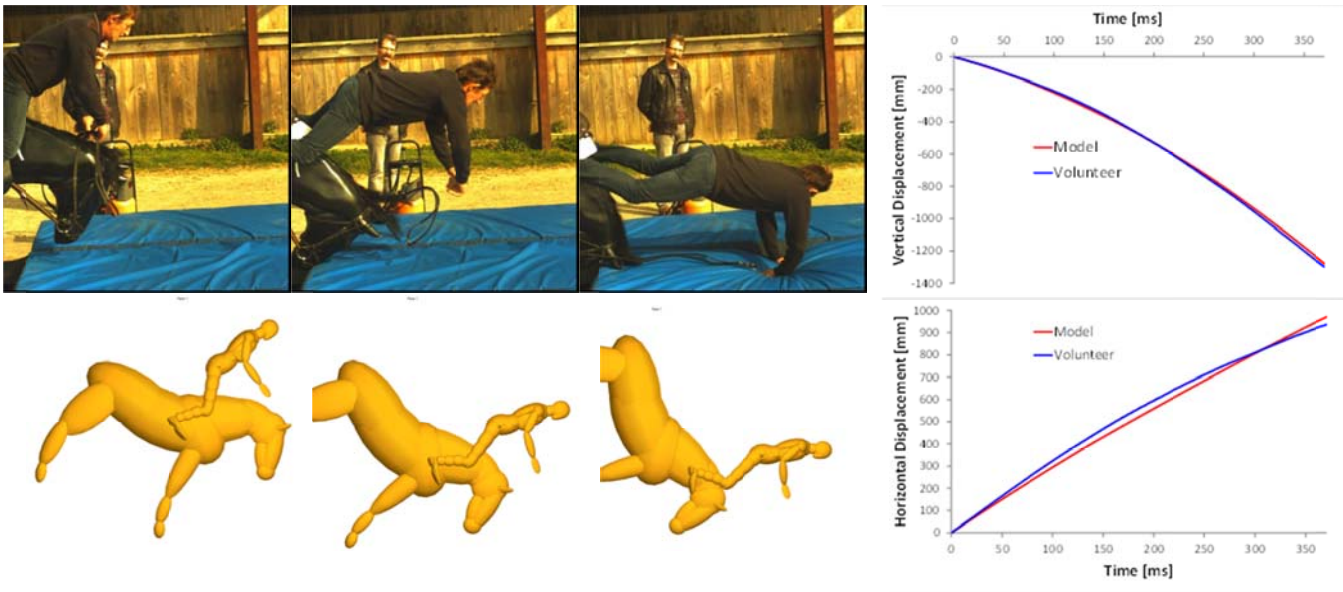


Fig. 5. Representation of the horse kinematics in terms of linear and rotational displacement extracted from the high-speed video.

The parametric study with the above developed horse-rider model consists of simulating a large number of horse-riding accident situations to better assess the impact conditions of the head against the ground. The coupled human-horse multibody model is used to compute the rider kinematics for a large set of accident situations. A total of five parameters have been selected: three concerning the horse and two characterising the rider position, as shown in 0. As a whole, no less than 1,920 accident simulations have been performed.

The first parameter is the horse orientation. It consists of orientating the horse at four angles (0 deg, 30 deg, 60 deg and 90 deg) versus the direction of the horse, the displacement remaining the same. The second parameter is the horse initial velocity. Six speeds were selected according to the three main gaits of the horse, i.e. walk, trot and gallop, as expressed in Table I.

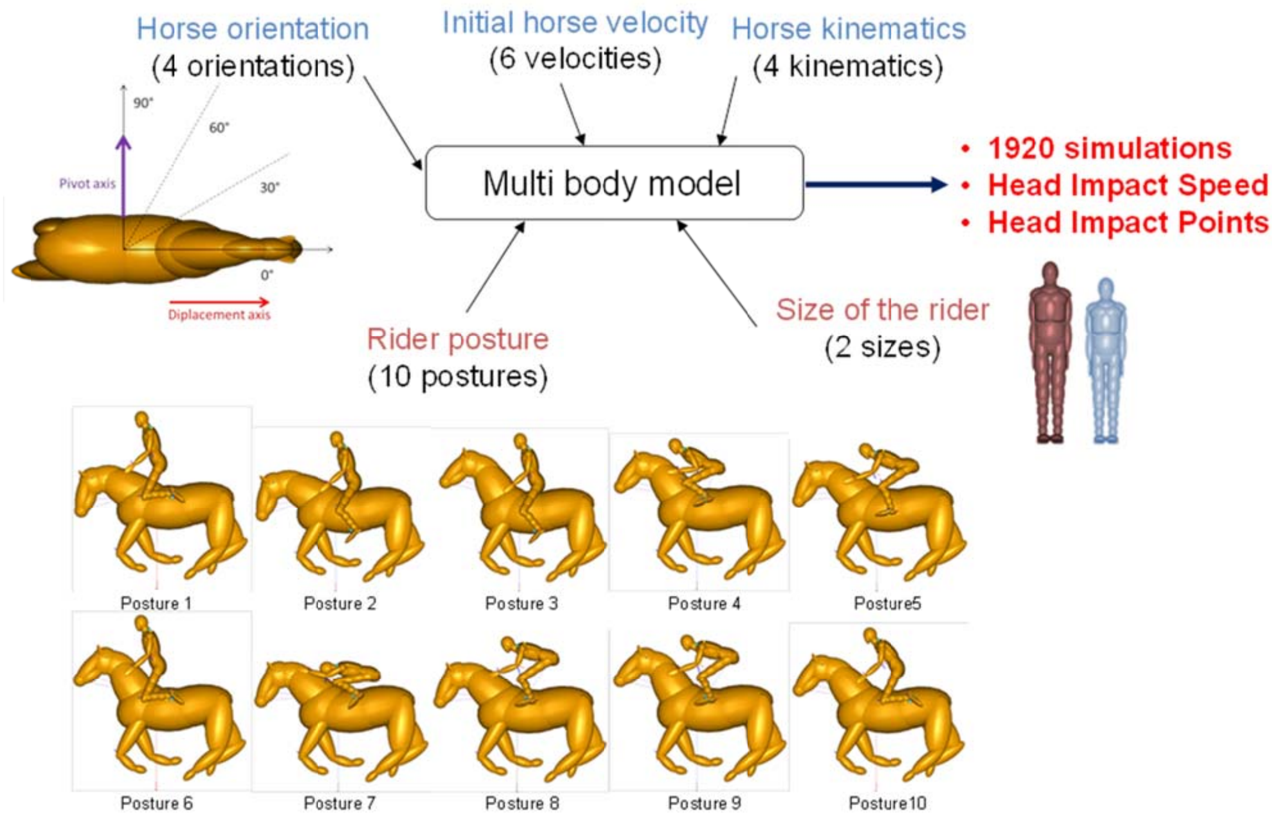


Fig. 6. Illustration of the parametric study on rider accidents, leading to 1,920 fall configurations.

TABLE I
INITIAL VELOCITIES USED FOR HORSE-RIDING ACCIDENT SIMULATIONS

Gaits	Walk	Trot		Gallop		
Velocities [m/s]	2	4	6	9	12	15
Velocities [km/h]	7.2	14.4	21.6	32.4	43.2	54

The third parameter involved in this study relates to the horse and concerns its fall-over kinematics. The tilting movement is based on the experimental simulator shown in Fig. 4 and divided into four kinematics by modifying the time of fall. The horizontal displacement is determined by an initial velocity and a stopping distance based on the equation 1.

$$\gamma = \frac{V_0^2}{2 \cdot d_b} \text{ and } t_b = \frac{V_0}{\gamma} \qquad \text{Equation 1}$$

where V_0 is the initial velocity of the horse, γ the constant deceleration of the horse, d_b the stopping distance and t_b the stopping time.

Finally, the two parameters characterising the rider are the size (175 cm with a mass of 75 kg, and 160 cm with a mass of 60 kg), as well as the posture on the horse, as illustrated in 0. The selected postures list is a non-exhaustive list captured from videos.

Concerning the output parameters, the head impact velocities as well as the head impact points at the time of impact were computed. Thus, for each simulation the determination of the initial contact time of the head was extracted using the computed contact force curve where the force is zero, as illustrated in 0. The location of the impact point is then determined at this time in terms of a square zone of 20mm large. The head impact velocity vector is also extracted and projected to the normal and tangential axis at the impact point. The impact velocity angle is computed according to **Error! Reference source not found.**.

$$\theta_{Impact\ Velocity} = \tan^{-1}\frac{V_{tangential}}{V_{normal}}$$ Equation 2

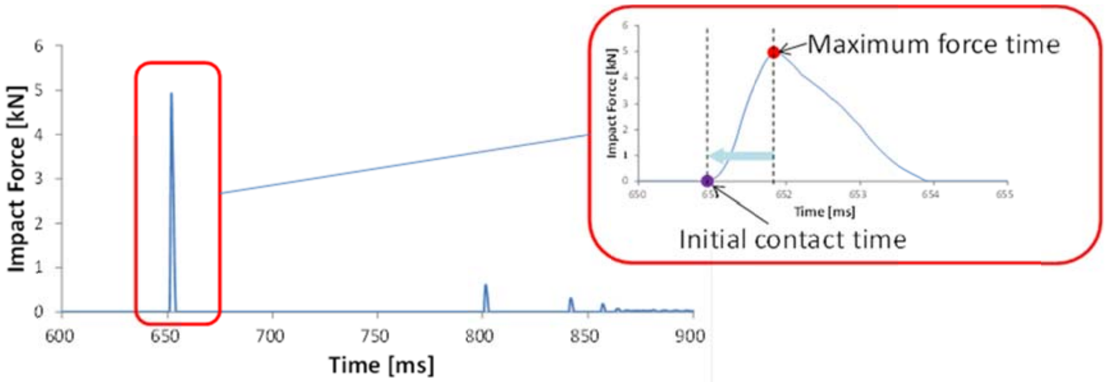


Fig. 7. Illustration of the initial contact time extraction.

III. RESULTS

A total of 1,920 virtual jockey accidents have been simulated in order to assess the rider's kinematics. For all of these simulations, the head velocity vector just before contact with the ground has been extracted, as well as the location of the contact point on the head. Among the 1,920 simulations, a head impact appeared for 1,918 cases. 0 illustrates three excerpts among the 1,920 simulations. 0a shows the distribution of the impact point density on the head for all of the simulations. The most commonly affected area is the frontal and facial area (red). The blue area is 100 times less affected and is located in the lateral area at the ear level and superior occipital region, as well as the chin area.

TABLE 2
MEAN VALUES AND EXTREMA OF HEAD IMPACT SPEEDS AND INCLINATION VERSUS NORMAL FOR ALL THE SIMULATIONS

	Mean	Minimum	Maximum
Resultant Velocity [m/s]	7.0 ± 1.3	0.9	10.6
Normal Velocity [m/s]	6.3 ± 1.4	0.2	9.0
Tangential Velocity [m/s]	2.8 ± 1.2	0.0	6.7
Impact Velocity Angle [deg]	25 ± 12	0	87

From the results, it can be observed that the maximum head impact velocity is 10.6 m/s compared with the initial riding velocities for gallop of 9, 12 and 15 m/s. This can be explained by the fact that the kinetic energy is a combination of a linear component and rotational component. Initially the rotation component is null, but when the head impacts the ground the angular velocity is not null. Then a part of the linear kinetic energy is transferred to an angular kinetic energy. Moreover, the potential energy can increase the kinetic energy either for linear or rotation. Concerning the impact velocity of the head, at first the human body model used is a multibody model and not a rigid one. The total kinetic energy of the full body is divided into several kinetic energy for each sub body. Then the velocity of one part is not the same as for another part as in the kinematic of particles. Moreover, when the head impacts the ground, it is not always the first contact of the body on the ground. Hence the head impact velocity is lower.

Moreover, the head impact speeds range from 0.9 m/s to 10.6 m/s, as reported in 0, with a mean value of 7.0 ± 1.3 m/s. The normal velocity is 6.3 ± 1.4 m/s and the tangential component is 2.8 ± 1.2 m/s, i.e. an inclination of the velocity vector versus normal of about 25 deg. These extreme values are observed for the impacts on the side area of the head. In contrast, the lower values appear at the parieto-occipital area, as illustrated in 0b. Globally, on the upper region of the head, the impact velocity is between 4 m/s and 6 m/s and in the frontal area about 7 m/s. Concerning the tangential effect, 0c shows that further away from the fronto-parieto-vertex area the impact point is, the more tangential is the velocity vector. Indeed, the areas with a tangential impact are located on the chin and the occipital part, with an angle greater than 45 deg. However, in these areas the speed does not exceed 6 m/s for the resultant and 2.5 m/s for the normal component.

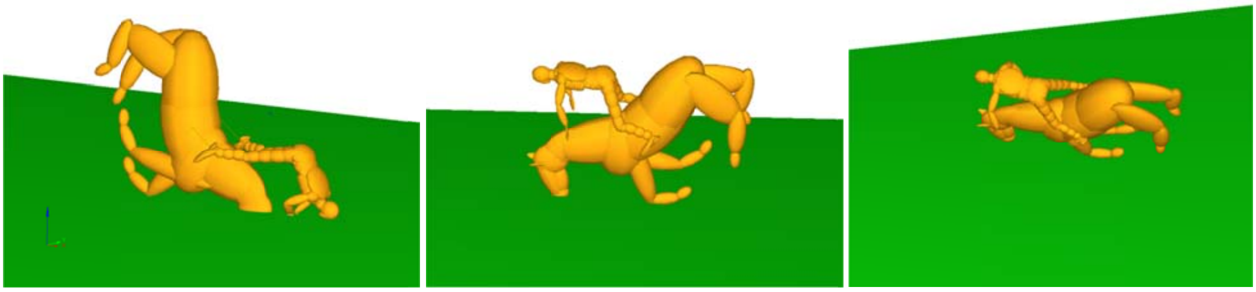
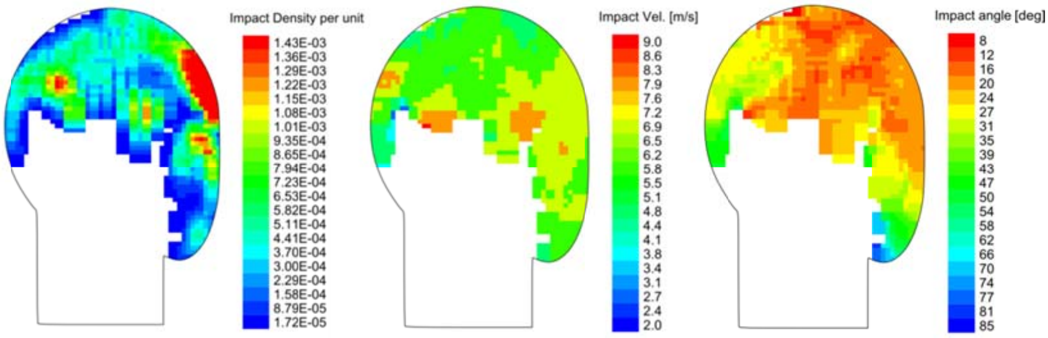


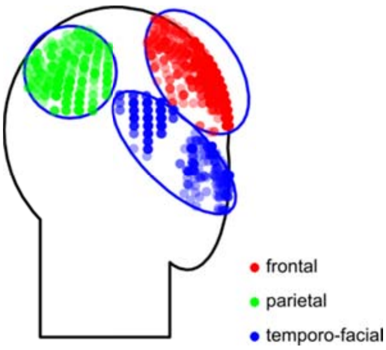
Fig. 8. Examples of three fall configurations among the 1,920 simulations.

Three main areas where the impact point densities on the head are the highest can be selected. An average of the head impact velocities were then computed for each zone, according to the impact speed. 0 shows the three main areas, including proportions and speeds. It is observed that the frontal area concerns 40% of the impacts, with an average speed of 7.1 ± 1.1 m/s. The parietal and temporo-facial areas gather about the same proportion of impact points (20% each), but the average impact velocities are slightly different. Indeed, on the parietal region the impact velocity is 6.6 ± 1.5 m/s, with an inclination of 29 ± 13 deg versus normal. While in the temporo-facial area, the average speed is 7.5 ± 1.1 m/s, with an inclination of 26 ± 8 deg.



(a) (b) (c)

Fig. 9. Representation of the impact point density on the head (a) and the repartition of the resultant head impact velocities (b) plus the impact velocity angle (c).



Zones	Frontale	Pariéto-occipitale	Temporo-faciale
Proportion	40 %	20%	19%
$V_{resultant}$ [m/s]	7.1 ± 1.1	6.6 ± 1.5	7.5 ± 1.1
$V_{normale}$ [m/s]	6.5 ± 1.0	5.8 ± 1.6	6.7 ± 1.1
$V_{tangential}$ [m/s]	2.6 ± 1.1	3.0 ± 1.1	3.2 ± 1.0
Impact velocity angle [deg]	21 ± 9	29 ± 13	26 ± 8

Fig. 10. Representation of the three main impact points on the head along with related proportions and velocities.

IV. DISCUSSION

Few authors studied the kinematics of the horse-rider during a horse fall accident, especially concerning the head conditions just before contact with the ground – an important aspect when it comes to helmet testing. Several authors focussed on modelling of the equestrian helmet during impact in order to study the performance of the protection: Forero Rueda et al. [31-33], Cui et al. [34] and Mills et al. [35]. Although they highlighted that the rotational effect, caused by tangential impact, on the brain injury risk is very relevant and that the consideration of the linear acceleration is not sufficient to optimise the helmet according to biomechanical criteria, all these authors have simulated helmeted head impacts using a normal velocity, only without any tangential component introducing rotation. Moreover, in the European Standard EN 1384:2012 [16], the absorbing standard test to homologate equestrian helmets consists of dropping the helmeted EN 960 headform onto a flat anvil with an impact velocity of 5.42 m/s. No tangential component of the impact velocity is included in this standard test. The maximum head acceleration must not exceed 250 g and the time duration at a head acceleration of 150 g must not exceed 5 ms. The protection area is defined by the grey area illustrated in 0a. The present work permitted an evaluation of the head impact conditions for a large number of virtual equestrian fall accidents in order to extract the location of the impact on the head, as well as the impact velocity vector. In the present study several parameters have been varied, for example, horse speed or rider posture. The results provide interesting inputs to the discussion regarding the helmet test methods with regard to impact velocity vector.

The main limitation of this study is that the list of postures is non-exhaustive and could gather more possible configurations. Nevertheless, the ten computed postures represent a wide range of realistic configurations, which allow the involvement of a large number of kinematics. On the other hand, only horse race fall accidents were considered, and an important next step would be to consider horse show-jumping accidents. Finally, limitations also exist at human body modelling level where a simple state-of-the-art model was applied as for pedestrian and bicycle accident simulation. It is possible that, in real life, the human would turn its head to avoid facial impact. This limitation affects the analysis of the impact location, but probably not the results in terms of impact velocity vector.

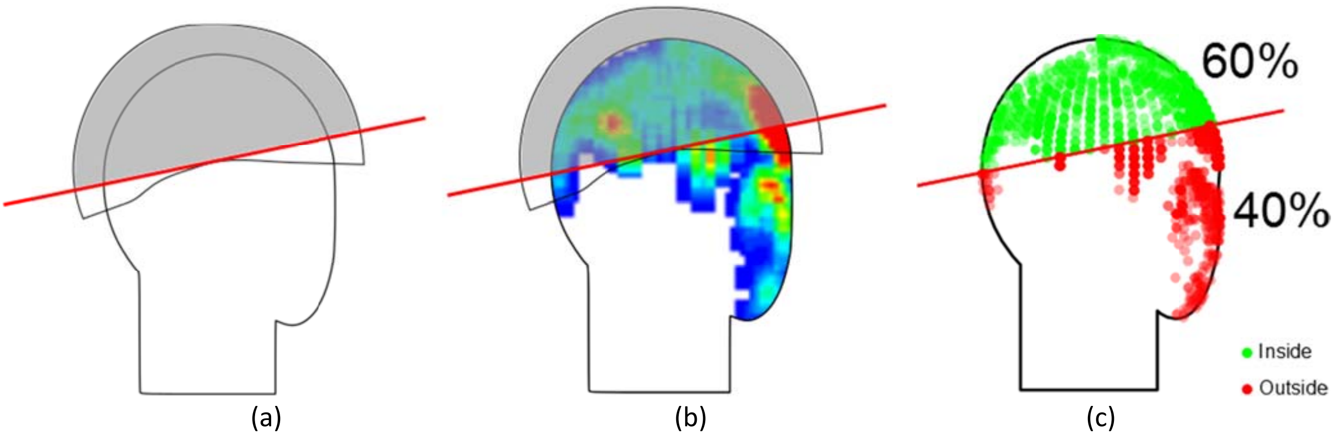


Fig. 11. (a) Representation of the test surface depicted in the standard, (b) superimposition of the computed impact points density with the test surface, (c) representation of the proportion of impact points inside the test surface (green) and outside the test surface (red).

The head impact point density is superimposed to the test surface, as shown in 0(b) and the proportion of impact points is computed (0(c)). At first, it can be observed that 40% of the impact points are below the test line, which is represented in red in 0. This means that only 60% of potential head impacts were considered in the standard, which constitutes a significant limitation to the evaluation of the real performance of the protective system. In addition, the highest head impact velocities are located close to the test line limit, as shown in 0(b). Moreover, inside the test surface the mean resultant impact velocity is 6.8 ± 1.3 m/s, with a normal component of 6.1 ± 1.4 m/s and a tangential one of 2.6 ± 1.2 m/s. The European standard impact velocity is lower, at 5.42 m/s, and contains no tangential component. However, it is well known that this component generates a rotational acceleration of the head and thus increases the injury risk, as reported by Halldin et al. (2001) [36] and by Deck et al. (2007) [37]. Therefore, the standard test for equestrian helmets should in future include this tangential component. Previous studies by Bourdet et al. (2012; 2013) [26][38] on bicyclist accidents have simulated real accident cases as well as bicyclist falls. They concluded that most of the impact locations on the head are outside the helmet "test areas". Moreover, this study demonstrated that the bicyclist head impacted with a resultant velocity of 6.8 m/s, a normal component of 5.5 m/s and a tangential one of 3.4 m/s. Thus the results led to the proposal to improve bicycle helmet standards, as described in study of Deck et al. (2012) [39], by including tangential test and advanced model-based head injury criteria.

A first proposal for the improvement of the equestrian helmet standard could be to change the test line position. Indeed, 0(a) and (b) represents the proportion of impact point inside surface and outside a proposal test. The proportion of impact points inside the test surface could rise up to 82%. A second proposal to test specific points on the helmet that represent the most common impacted areas is illustrated in 0(c).

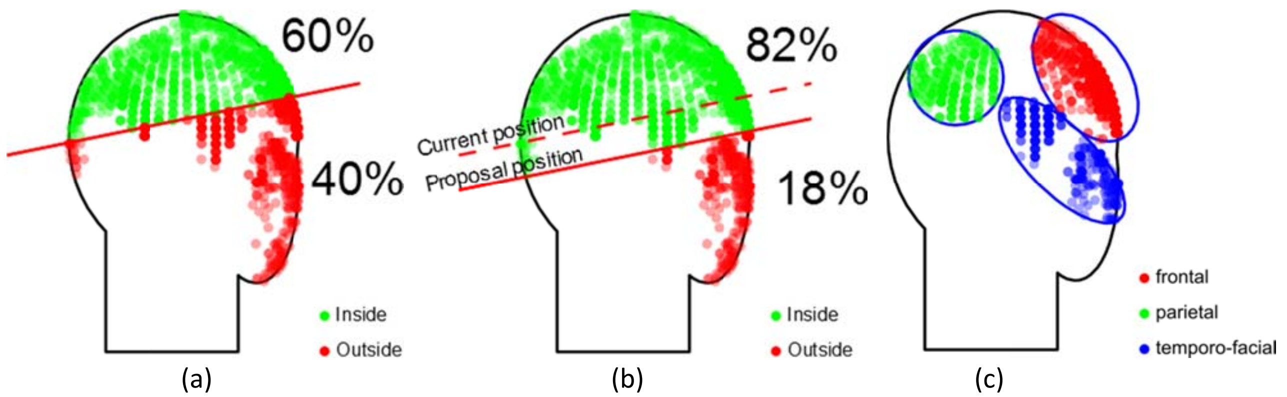


Fig. 12. Representation of the proportion of impact points inside (green) and outside (red) the test surface for (a) the current test line and (b) the proposal test; and (c) the three most common impacted areas.

V. CONCLUSIONS

This study focuses on horse-rider head impact conditions in the case of fall accident, an aspect poorly reported in the literature. A multibody model of the jockey-horse complex has been developed and validated against volunteer fall data and applied to an extensive parametric study involving no less than 1,920 virtual accident simulations. Ten postures of the riders have been considered at six initial speeds of the horse (2 m/s to 15 m/s). Results demonstrate that head impact points are very often located around and under the helmet rim, that the normal head initial speed is close to 6 m/s and that the head velocity presents a significant tangential component. Three main impact locations can be extracted, gathering 80% of the total impact points: frontal, parieto-occipital and temporo-facial, with specific impact velocities. For frontal impact, which represents 40% of the impacts, the vertical drop velocity is 6.5 m/s, with an inclined surface of 21 deg. The second area is the parieto-occipital region, which represents 20% of the total impact points. The drop velocity is 5.8 m/s, with an impact angle of 29 deg. The last zone is temporo-facial, in which the velocity is about 6.7 m/s and an inclined surface of 26 deg. It is considered that the results of this study contribute to the discussion related to new helmet standard and should be considered in helmet design.

VI. ACKNOWLEDGEMENTS

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