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Preliminary effectiveness assessment of an airbag-based device for riders' leg protection in side impacts

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Abstract

PTWs usually are cheaper than cars, they have lower energy consumption and fewer polluting emissions than larger vehicles. So, over the last decade there has been a significant increase of PTW circulating park. When riders are involved in an accident, legs are the most frequently injured part of the body. The aim of this paper is to assess the performance of a preliminary concept of inflatable leg protector mounted on a motorcycle. Five impact configurations were simulated in a finite element virtual environment, with the car impacting laterally on the motorcycle. Both stationary and moving motorcycle crashes were performed, while the car speed was 50*km/h* in every configuration. The rider was represented by a Hybrid III dummy model. Since the model was not validated for side impacts, a comparative analysis was performed. A set of safety parameters was used to assess the performance of the device. Their reduction was obtained in specific impact conditions, although mixed with load increases. Despite a widespread parameter reduction was not achieved, the results demonstrated the potential of the proposed device for lower limb protection, and they allowed the identification of the most severe conditions to be used in future development

Keywords: Airbag; PTW; Leg protector; Lower limb injuries; Side impact.

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1. Introduction

Road safety is still a relevant topic in the scientific community, because millions of people die on the roads every year worldwide (about 1.35 million dead and 50 million injured in 2017 (WHO (2018))). In addition, "road traffic crashes are predicted to become the third most common cause of disability worldwide by 2020" (de Rome et al. (2012)). In 2016, Powered Two-Wheeler (PTW) riders represented 17% of all road deaths in EU area (EC (2018)). In 2010 EU proposed an ambitious target of halving the overall number of road casualties by 2020 (EC (2011)). In the last decades EU also promoted research activities to increase PTW and rider safety. PTW riders are considerably more exposed and vulnerable than drivers; indeed, according to European Commission (EC) "the improvement of the safety of vulnerable road users, in particular motorcyclists for whom accidents statistics are particularly worrying" (EC (2010)), is a priority. In fact "motorcycling is the mode of transport for which the number of fatalities decreased least between 2006 and 2015" (EC (2017)). Motorcycles and roads are safer than before, but the number of casualties is still too high; safer PTWs and improved riders' protections are still necessary. These technical targets represent a difficult challenge because riders, unlike car drivers, do not have any structure or frame to protect them. Passive systems mounted on the motorcycle are partially effective because of the separation between rider and motorcycle due to a crash or a fall (although recent research activities investigated also restraint systems integrated on the PTW (Grassi et al. (2018a))). An effective way of providing the required protection is certainly the use of protective clothing. Personal protective equipment is resistant to abrasion, cuts and tears, and the integrated reinforcement absorbs and distributes the forces of direct impacts. It is well known that wearing a helmet can significantly reduce the probability of fatal head injuries. According to National Highway Traffic Safety Administration (NHTSA) "a non-helmeted rider is 40% more likely to incur a fatal head injury […] than a helmeted motorcyclist" (Crompton et al. (2010)).

Upper body injuries are often fatal, due to the presence of the internal organs, but they occur less frequently than lower limbs injuries (Sporner et al. (1990); Meredith et al. (2013); Aarts et al. (2016); Piantini et al. (2019); Serre et al. (2012)). It is estimated that "70% of motorcyclists sustain some kind of leg injury during a crash" (Rizzi (2015)). Other studies (Serre et al. (2012); Lateef (2002)) have a slightly lower estimation for the same injury type (i.e. approximately 60%). In the European-focused Motorcycle Accidents In-Depth Study (MAIDS) report (ACEM (2009)), 31.8% of the injuries, reported by riders, were lower extremity injuries, instead 18.4% were head injuries. Lower limbs are the second most frequently injured body region according to the ISO 13232 standard (Grassi et al. (2018b)). Anyway, leg injuries can cause long standing or permanent disabilities (Nordentoft et al. (1984)). Typical reported injuries are abrasions and skin excoriations, fractures and soft tissue injuries. Abrasions and skin excoriations are very frequent, but they can be mitigated using protective clothing (Nordentoft et al. (1984)). Fractures mainly affect long bones such as tibia and femur. Tibia and fibula fractures are very common, as reported in (Piantini et al. (2019)); they can break because of bending moment (e.g. impact between lower leg and car bumper). When the leg hits a larger surface (e.g. the radiator grille) the rider can report also severe soft tissue injuries. According to Piantini et al. (2016), femur is the most injured bone, with a predominance of diaphysis fractures.

For these reasons different types of leg protector have been studied since the 1980s. In (Tadokoro et al. (1985)) a crushable leg protector was tested: the device was comprised of two deformable aluminium honeycomb structures on each side of the motorcycle. It was tested with a car impacting on the motorcycle at the collision angle of 45° and 90°. Tests showed a reduction of the lower leg fractures and an increase in thigh region injuries. In Chinn et al. (1984) both a rigid structure (like a "shield") and a deformable structure (made of polyurethane foam) were tested and compared. These devices were mounted on the motorcycle, which impacted against a barrier inclined 30° with respect to the longitudinal direction of the vehicle. According to the authors the deformable structure was more effective in terms of protection compared to the rigid one. Nonetheless both prototypes were never brought to the market. Rogers and Zellner (1998) described tests, performed by International Motorcycle Manufacturers Association (IMMA), in which a special hull, designed by Transport Research Laboratory (TRL), was tested. They reproduced seven impact configurations and stated that the proposed structure could reduce the severity of lower leg injuries. In particular, it was able to prevent the fractures of the tibia and the direct contact with the car. However, the structure increased both the compressive load and the bending moment applied to the femur, causing its fracture. Moreover, in some configurations the locking of the knee and the chest rotation resulted a twisting action on the thigh that caused the fracture of the femur by torsion. This phenomenon represents the main limit of the leg protectors tested so far. Crash-bars are the only leg protectors integrated on the PTWs (especially on large touring motorcycles) and present on the market. They are rigid structures made of tubular bars, which can prevent leg from being trapped between the motorcycle and the road. Their effectiveness is still not uniquely assessed: according to some authors, crash-bars effectiveness "is limited to a restricted range of accidents and circumstance" (Nairn and Partners Pty Ltd (1992)); for other researchers crash-bars can decrease the severity of leg injuries in motorcycle accidents (Mohaymany and Eghbalian (2007)). The main drawback of leg protectors is the alteration of the rider's kinematics, which usually turns into upper body injuries (Tadokoro et al. (1985)). Despite a few research activities, no manufacturer has currently introduced leg protectors on the market (except crash-bars), because the cited negative effects could not be eliminated.

In this paper an innovative solution, based on inflatable components, is proposed and studied. In the device concept, four airbags (two per side) are mounted on a motorcycle to protect legs in side impacts, since lower limbs are very vulnerable in these configurations due to the riding position. In the past, motorcycle airbags were already investigated (Nairn and Partners Pty Ltd (1992); Mohaymany and Eghbalian (2007); Elliott (2003); Iijima et al. (1998); Kuroe et al. (2005); Yamazaki et al. (2001); Barbani et al. (2012)), but they were used only to reduce upper body injuries and prevent rider's ejection in frontal impacts. This research aims to perform a preliminary verification of the protective effectiveness of the device, in a small set of critical impact conditions. As suggested by past research activities, the identification of changes in rider kinematics and generally of any adverse effect introduced by the device is included within the objectives of this paper.

Nomenclature

EC	European Commission
FE	Finite Element
HIC	Head Injury Criterion
IMMA	International Motorcycle Manufacturers Association
MAIDS	Motorcycle Accidents In-Depth Study
MATD	Motorcyclist Anthropometric Test Device
NCAC	National Crash Analysis Center
NHTSA	National Highway Traffic Safety Administration
PTW	Powered Two-Wheeler
TRL	Transport Research Laboratory

2. Methods

2.1. Virtual environment

In order to assess the effectiveness of the proposed device, five impact configurations were defined, and crash tests were simulated using Finite Element (FE) models. In every scenario a passenger car impacted a dual touring motorcycle on the right side. Simulations with both stationary and moving (5*m/s*) motorcycle were performed; car speed was always 13.9*m/s* (50*km/h*). According to MAIDS report (ACEM (2009)), in the most frequent impact scenario (25% of cases) the motorcycle and the car have the same direction of travel, but in the second one (17% of cases) the vehicles are perpendicular. This configuration (*C 90*) was included in this study as the main configuration (Fig.1) since it is the most frequent lateral impact. The other configurations were defined so that the sample could be fully representative of the side impacts with a limited number of scenarios. Every impact configuration can be identified by the angle formed by axes of the two vehicles before the impact (Fig.1). The impact points were chosen so that the car hit one of the two airbags instead of the leg.

A rigid FE motorcycle model was created from a CAD model, provided by the manufacturer. A rigid model was implemented since 1) this is a preliminary study, 2) in the real impact, motorcycle parts (as motorcycle frame or engine block) are much stiffer than the rider's leg, pushed against them. Therefore, using a rigid motorcycle was acceptable because it has introduced a limited approximation. Specifically, the motorcycle was modeled as a shell using the external surfaces of all components, to provide reference and support to the virtual rider. The model was complemented with the inertial properties of the real motorcycle (i.e. mass and the inertia moments), assigned to the center of gravity.

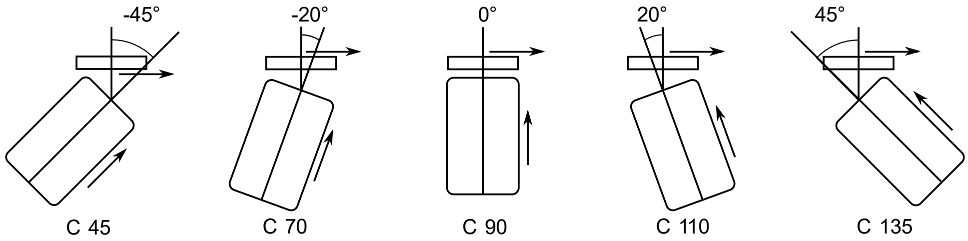

Fig. 1. Impact configurations. The biggest boxes represent the car, while the smallest represent the motorcycle; arrows represent the vehicles directions of travel and indicate the direction to which the front axle is pointing.

Airbags were modelled with 2D elements with a stretched and non-folded shape without vent holes (Fig.2). Their material was modelled as a fabric with a thickness of 0.4*mm* and each airbag had a volume of 7.5*l* when completely opened. A defined pressure model was used for the airbags: the pressure curve had an initial linear ramp, followed by a constant value.

The virtual environment was developed exploiting the expertise of previous studies (Barbani et al. (2014a); (2014b). The car type can influence the results because differences in characteristics and geometries, e.g. the size of the bonnet, can alter the rider's kinematics. A sedan was chosen in agreement with the ISO 13232 standard (ISO (2005)). Even if the indications of this ISO standard were not fully followed, this study was developed with as little deviation as possible from the only reference currently available. The FE model of the car was developed and validated by National Crash Analysis Center (NCAC).

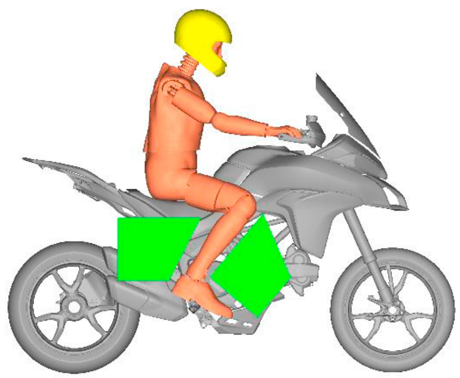

Fig. 2. Representation of the device: the green trapezoids represent the airbags.

The rider was represented by a FE model of a crash test dummy, to facilitate future comparison with experimental tests. ISO 13232 (ISO (2005)) suggests the use of the Motorcyclist Anthropometric Test Device (MATD), but no FE model was available. MATD was developed from Hybrid III, so the latter was considered a valid alternative together with EuroSID. Dummies are equipped with different set of sensors, especially in lower limbs. EuroSID has no load cell in lower leg, unlike Hybrid III; so, evaluating lower leg injuries was impossible using EuroSID model. Human body models (e.g. THUMS) could have been a viable alternative, but they were discarded because using this kind of models would have greatly complicated the comparison with experimental data. Thus, the rider was modelled using a Hybrid III 50th percentile dummy and fitted with a full-face helmet (Pratellesi et al. (2011)). The dummy model was not validated for side impacts. Thus, a comparative analysis was performed, i.e. results collected with and without the device were compared. It is important to note that in every configuration the impact point did not change passing from stationary condition (of the motorcycle) to moving condition or from configurations without to configurations with the device.

2.2. Performance assessment

Because of the limitations due to the chosen dummy, mentioned above, biomechanical indices, as Tibia Index, were not considered in this paper and the protective performance of the device was evaluated considering loads applied directly to the limbs. Biomechanical limits are evaluated with tests on human bones or limbs; so, they can not be directly compared with the loads estimated with the dummy. The only injury criterion considered in this paper is HIC_{36} because it is calculated just considering the linear head acceleration. Even if the neck of the Hybrid III dummy is not validated for side impacts, it does not alter the acceleration peaks due to head impacts. So, some safety parameter (Table 1) were defined in order to assess the performance of the device. A maximum value was identified for each parameter, considering all simulations without airbags (both stationary and moving motorcycle). These numeric values were considered as limit and all parameters were compared with them in each simulation. According to this method, in tests with the device a given parameter may worsen compared to the same configuration without it, as long as it does not exceed its limit. So, the dummy provided information about the loads acting on the rider without an injury estimation, as it is not validated for this kind of impacts. Regarding loads acting on the upper body, only chest acceleration and HIC_{36} were considered. The first one provided information about the inertial effects, the latter about the head injuries due to the impact of the head on the windscreen (or bonnet) of the car.

In this paper only right leg was considered because it was directly involved in the impact. Hybrid III dummy was equipped with a load cell in the middle of each femur; so, in this paper axial force and both bending and twisting moments, acting on this bone, are considered. Only bending moment and axial force on the lower leg were considered. Since each tibia was equipped with two load cells, i.e. the first one was in the upper tibia area and the latter was near the ankle, in each simulation the loads estimated by both were considered and then the maximum values were reported.

The performance of the device will be positively rated if the limit values, defined in the configurations without airbags are not exceeded in any of the configurations with the installed device. In fact, the comparative approach will allow to state that the device reduces the maximum loads on the dummy and thus it introduces a beneficial effect for the rider, while no adverse effects are observed. An increment of a load parameter, but still under the limit value, is accepted because a higher value was anyway observed in one of the configurations without the device.

3. Results

The results of the simulations were analyzed with the assessment scheme presented in the previous section; the limit values determined for each parameter are reported in Table 1. Results of the assessment are reported in Tables 2 and 3 for configurations with stationary and moving motorcycle respectively. In these tables, results are expressed as a percentage of the limit values.

Table 1. Limit values for each parameter estimated in simulations without airbags.

Parameter	Limit value	Configuration
Femur Bending Moment [Nm]	652	C 135 (Moving)
Femur Twisting Moment [Nm]	280	C 45 (Moving)
Femur Axial Force [kN]	7.3	C 45 (Moving)
Tibia Bending Moment [Nm]	1238	C 110 (Moving)
Tibia Axial Force [kN]	7.4	C 110 (Stationary)
HIC_{36}	1920	C 45 (Stationary)
Chest Acceleration [g]	134	C 110 (Stationary)

In configurations with stationary motorcycle (without airbag), the bending moment on the tibia tended to be greater than 50% of the limit, highlighting the relevance of the load applied to this area. It is important to note that the airbags significantly reduced this parameter in almost all configurations, but in *C 45* where the bending moment on the tibia increased to 96% of the limit. The axial force on the tibia tended to reach high values in configurations without the device (the limit value was defined in C 110), but it was reduced by the introduction of the device (Table 2). In the simulations with the airbags, the bending moment applied to the right femur was greater than 50% of the limit in most cases and increased in two configurations, especially in *C 90*. Twisting moment was greater than 50% of the limit in all cases (both with and without airbags) and the device led to significant increases: the values exceeded the limit in *C 70* and *C 45* (in the latter, this parameter was doubled); differently the maximum reduction was 4% in *C 90*. However, the airbag device was effective to reduce the axial force on the femur in each scenario.

Table 2. Results of simulations with stationary motorcycle. Red cells represent loads increased by the presence of the device; white text values represent a load greater than limit.

Parameter	C 45		C 70		C 90		C 110		C 135	
	W/o	W	W/o	W	W/o	W	W/o	W	W/o	W
Femur Bending Moment	95%	91%	76%	57%	47%	72%	79%	42%	52%	59%
Femur Twisting Moment	63%	131%	91%	111%	76%	72%	70%	68%	66%	78%
Femur Axial Force	65%	42%	48%	27%	47%	38%	70%	35%	82%	71%
Tibia Bending Moment	68%	96%	48%	43%	76%	48%	73%	42%	62%	30%
Tibia Axial Force	47%	56%	19%	39%	40%	27%	100%	23%	46%	24%
HIC_{36}	100%	81%	61%	68%	24%	79%	14%	13%	5%	2%
Chest Acceleration	72%	56%	25%	24%	36%	39%	100%	146%	54%	83%

In *C 45*, HIC_{36} reached the limit value in the configuration without airbags, but the device reduced the index value by 20%. In other configurations it reached lower values. Head injuries were substantially unchanged except in *C 90*, where HIC_{36} more than tripled in the simulation with the device (from 24% to 79% of the limit), but still under the limit. In *C 110* the chest acceleration reached the limit in the simulation without the device, although it assumed a high value also in *C 45*. In the latter configuration the largest reduction of this parameter was obtained. In *C 110* and *C 135* the chest acceleration increased: especially in *C 110* the limit value was exceeded by 46%.

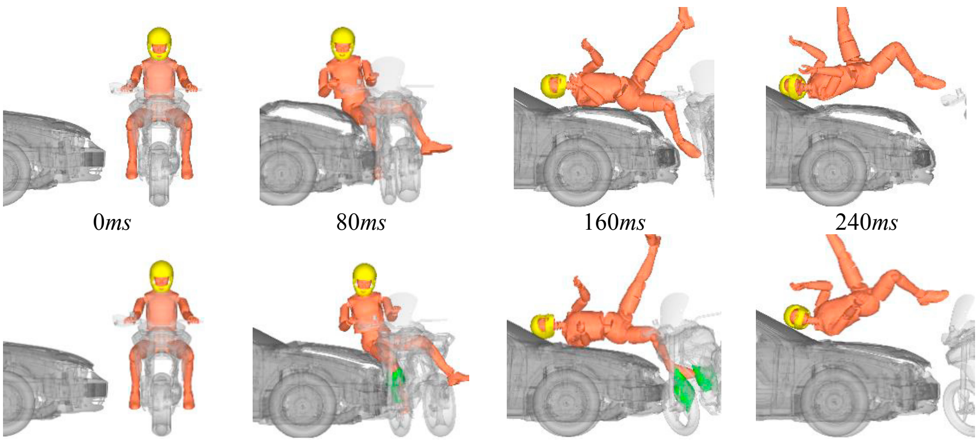

Fig. 3. Dummy kinematics in *C 90* (stationary motorcycle) without (upper) and with (lower) airbags.

Among all the studied configurations, *C 90* is the only one reported in the ISO 13232 standard (ISO (2005)), and thus it was reported in Fig. 3. In this scenario without airbags, the car hit the right ankle and trapped it between the two vehicles. Subsequently all the right lower leg was crushed by the car against the motorcycle, the pelvis slid sideways and rested on the bonnet. In the simulation with the device, the car impacted simultaneously against the airbag and the lower leg. Afterwards the rider impact kinematics was similar in the two simulations, but it was influenced by a large deformation of the bonnet in the configuration without device. In this case the bonnet deformation, changed the impact point of the helmet with the windscreen, and thus the injury outcome. In the simulation with the device, after the initial compression, both airbags expanded causing a yaw motion, absent in the simulation without protector. This movement reduced the pressure on the right calf.

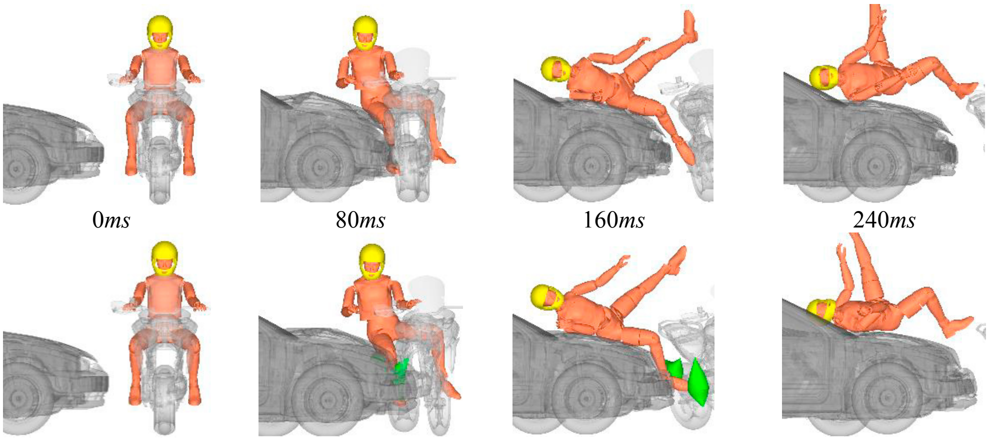

Fig. 4. Dummy kinematics in *C 110* (stationary motorcycle) without (upper) and with (lower) airbags.

In *C 110* two parameters (axial force on the tibia and chest acceleration) reached their limit in the case without airbags. Moreover, all other parameters (except HIC_{36}) were greater than 70% of their limits in this scenario, but they were significantly reduced by the device; only chest acceleration increased, as previously mentioned. For this reason, *C 110* kinematics is reported in Fig.4. In the simulation with the airbags, the car initially hit the front airbag which was completely crushed. The presence of the device did not cause significant variations to the dummy kinematics. The most significant differences were in the final part of the simulation, when a more pronounced roll movement of the dummy was noticeable. As a matter of fact, in the simulation with the airbags, the maximum chest acceleration was reached in final part, around 240*ms*. The other configurations showed no significant variations of the parameters, except for the twisting moment on the upper leg, that exceeded the limit value in the *C 45* and *C 70*.

Table 3. Results of simulations with moving motorcycle. Red cells represent loads increased by the presence of the device; white text values represent a load greater than limit.

Parameter	**C 45**		**C 70**		**C 90**		**C 110**		**C 135**	
	W/o	**W**	**W/o**	**W**	**W/o**	**W**	**W/o**	**W**	**W/o**	**W**
Femur Bending Moment	64%	50%	84%	43%	60%	54%	72%	88%	100%	116%
Femur Twisting Moment	100%	54%	79%	54%	58%	59%	53%	75%	59%	55%
Femur Axial Force	100%	21%	68%	37%	66%	63%	89%	76%	98%	63%
Tibia Bending Moment	54%	27%	85%	33%	83%	67%	100%	63%	78%	32%
Tibia Axial Force	42%	43%	14%	30%	41%	38%	42%	48%	89%	109%
HIC_{36}	55%	46%	26%	23%	12%	19%	5%	4%	2%	7%
Chest Acceleration	37%	32%	40%	69%	52%	79%	54%	60%	81%	89%

In simulations with moving motorcycle (without airbags) the bending moment on the tibia was always greater than 50% of the limit, reaching it in *C 110* (Table 3). In other scenarios (except *C 45*) this parameter was extremely close to the limit value. The data highlighted the importance of this area also in moving-moving configurations. However, the device greatly reduced the tibia bending moment in all configurations.

The axial force on the tibia was lower than 50% of the limit in all scenarios (both with and without the device), except for *C 135*. In this configuration, the parameter was close to the limit in the case without airbags, but it exceeded the limit after their introduction. This was a critical configuration without the device because other parameters were very near the limit value. Bending moment on the femur was significantly reduced only in *C 70*, but it was increased in *C 110* and exceeded the limit in *C 135*. On the other hand, twisting moment was greater than 50% of the limit in every configuration without airbags and reached the limit in *C 45*. In *C 45* and *C 70* this parameter was significantly reduced, while in C 110 it was increased up to 75% of the limit. The axial force had its limit set in *C 45*, but the device significantly reduced this load. Regarding upper body injuries, HIC_{36} changed slightly, while the chest acceleration increased in almost all scenarios, reaching high values.

In *C 70* (Fig. 5) only the chest acceleration and the axial force on the tibia were increased by the airbags; all moments on the leg were near the limit in the simulation without the airbags, but they were subsequently reduced. In the simulation without airbags the car hit the right ankle and trapped it between the two vehicles. Then the car crushed the lower leg against the motorcycle that performed a yaw motion. At this point, the rider's pelvis slid on the seat till the right thig and the pelvis itself leaned against the bonnet. The upper body performed a roll motion and the head was pushed against the windshield. In the simulation with the airbags the rider's kinematics was essentially the same, but the airbags reduced the contact with the dummy.

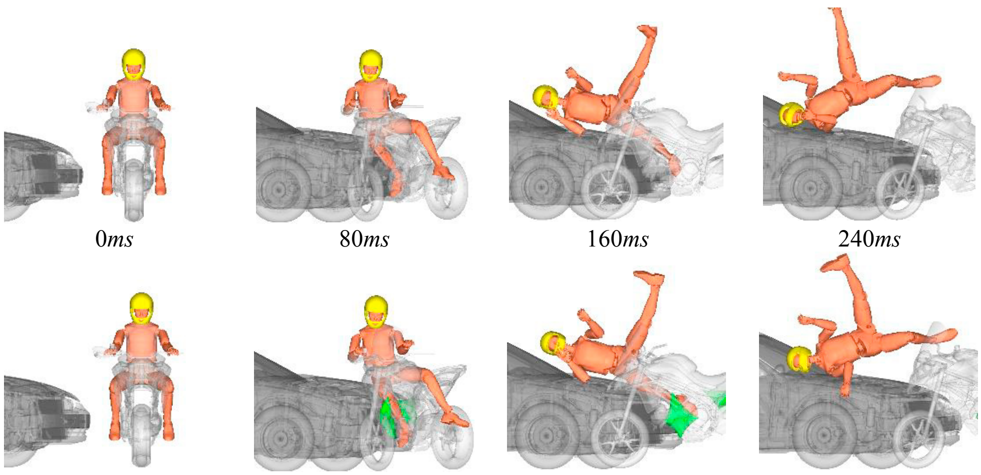

Fig. 5. Dummy kinematics in *C 70* (moving motorcycle) without (upper) and with (lower) airbags.

4. Discussion

4.1. Analysis of the results

In *C 90* (with stationary motorcycle) upper body injuries were increased, as reported in literature, and in the same scenario (but with moving motorcycle) head injuries were significantly increased. These results highlighted the inertial effects due to the initial velocity of the rider. In fact, it is important to note that in every configuration with the moving motorcycle, the impact point of the head was shifted forward (in the direction of travel of the motorcycle). In *C 110* (with stationary motorcycle) head injuries did not increase and leg injuries were reduced. On the contrary, in this configuration the chest acceleration reached the maximum value compared to all the other simulations. Overall, in this scenario the airbags provide the best protection to the rider with stationary motorcycle. Instead, the results (with moving motorcycle) were rather different; four parameters were increased. In *C 70* with stationary motorcycle three parameters increased and one of them exceeded its limit. On the contrary, the same scenario with moving motorcycle presented significant reductions of the safety parameters that tended to be lower than the 50% of their limits. So, it highlighted the inertial effect due to the rider's initial speed. However, this configuration showed the largest increase of the chest acceleration.

In *C 45*, with stationary motorcycle, the airbags reduced some of the safety parameters, but they tended to be too high in both simulations with the device and without it. Indeed, in this configuration the twisting moment on the femur was increased from 63% of the limit to the 131%. On the contrary, in the same configuration, with moving motorcycle, almost all parameters were reduced by the airbags and tended to be lower than the 50% of their limits.

Results showed that lower leg was the most frequently loaded zone in the studied impact configurations, in fact it was the only body part directly hit during the impact, i.e. when the car had the maximum kinetic energy, and it was crushed between the two vehicles in every configuration. As predictable consequence load applied to this area tended to be greater than the 50% of their limits in every configuration both with stationary and moving motorcycle without airbags; they were reduced in almost all configuration with the proposed device. Although severe impact conditions were selected for the simulations, the results showed that the airbags were able to reduce significantly the loads on the lower leg in the majority of configurations. In simulations with stationary motorcycle, femur twisting moment had high values in every configuration and the studied device did not reduce it enough, so torsional fractures were very probable. In the simulations with moving motorcycle, without airbags, twisting moment was lower than in the corresponding configurations with the stationary motorcycle (except for *C 45* where it was equal to the limit). Moreover, in *C 45* and *C 70* (with stationary motorcycle) this parameter was increased by the device and exceeded the limit.

The study showed selective reduction of loads in specific configurations and impact conditions. Specifically, the device offered a protective performance: 1) to the lower leg in all configurations but *C 45* (stationary-moving) and *C 135* (moving-moving); 2) globally in *C45* and *C 70* moving-moving configurations, i.e. in those configuration where the motorcycle and the car had a concurrent component of their velocities. However, the device failed to produce a widespread reduction of the loads and the avoidance of critical values for the parameters in all configurations. A re-design of the airbags to improve their protective performance and generate a widespread decrease of the injuries is necessary. The results suggested that the performance of the front airbag, active in *C 110* and *C 135*, is more critical than the rear one. Increased volume and/or maximum inflation pressure will be tested in the prosecution of the research, together with the introduction of vent holes.

4.2. Limitations

A constant pressure model was used for the airbags, as it doesn't require any assumption on vent holes. Its simplicity conflicts with a detailed representation of the real airbag behavior. In the continuation of the study the current airbag model will be replaced with one capable to represent the evolution of the internal pressure and comprehensive of vent holes.

The impact conditions are extremely severe, and they might hamper the evaluation of the device effectiveness, since most of the parameters are close to their limit. The definition of refined impact speeds, based on analysis of crash databases, will provide better guidance for the design of the airbags. Therefore, an accidentological study will be carried out to determine the speeds of the most frequent impact conditions.

The use of the Hybrid III model represents a further limitation. Its use was decided to facilitate future comparison with experimental data. Nonetheless the usage of numerical human body models, might provide more representative results. In addition, the employed Hybrid III model did not allow to assess the right ankle injury, although the simulations demonstrated that it is often the first impact point of the car. In addition, usually the rider's head was pushed against the windshield by the rest of the body in the final part of the simulations: this movement caused large deflections on the neck. However, neck injuries could not be assessed, because the neck of the Hybrid III was designed and validated for frontal impacts and its behavior is not reliable in side impacts. Eventually, the entire dummy is not validated for side impact so, we could only compare data collected in simulations with and without airbags, but the absolute values of biomechanical indices could not be quantitively compared with their limits. However, a qualitative comparison shows that the limits, reported in Table 1, could correspond to critical injuries (e.g. HIC_{36} limit in this work is almost double of the biomechanical limit). Also the limit values of both the bending and twisting moment on the femur are much greater than the biomechanical limits reported in literature, 373*Nm* (Martens et al. (1986)) and 175 *Nm* (Fildes et al. (1994)) respectively. Similar considerations apply to the limits on the chest acceleration and the bending moment on the tibia. However, the biomechanical limits were not used in this work, since the dummy was not validated for the specific impact conditions. Nonetheless the reduction of the values for the parameters used in this study is reputed a good indication for the improvement of the protective device. Reducing all parameters below at least 50% of their respective limits will be one of the main objectives in the next studies.

5. Conclusion

This study investigated the well-known problem of leg protection, using tools not yet easily accessible in the 1980s as FE modelling. A new concept of leg protection device, based on airbags, was proposed and its effectiveness was assessed in side impacts. This device does not interfere with the rider while he is riding, because it is mounted on the motorcycle and not on the rider, and it is not inflated in normal conditions. Both increases and decreases in loads were achieved. The proposed device has shown a potential for lower limb protection, but it needs further development to obtain a widespread reduction of the safety parameters considered in the work.

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