Coupling a Wearable Airbag with a Back Protector: What are the Effects on Motorcyclists' Safety?

Oscar Cherta-Ballester, Maxime Llari, Tanguy Bros, Valentin Honoré, Pierre-Jean Arnoux

Abstract This work aims to assess the protection provided by an airbag, a back protector and their combination under blunt back impacts which could cause thoracic and spinal trauma. Finite element modelling was used to simulate normal impacts to the HUMOS2 model at L1 level, one of the most vulnerable regions of the spine. Multiple impact conditions, based on the results of previous multibody accident simulations, were simulated on the human model by testing the protectors and reference simulations without any protection. The behaviour of airbag and back protector models was previously validated from drop impact experimental data. Impact force, intervertebral joint rotation from T1 to L5, organs internal energy and back compression were measured to estimate impact severity. Peak contact force mitigation is the main benefit of back protectors, especially for penetrant impacts. In addition to impact force and intervertebral rotation, the airbag decreases back compression and organ deformation due to its higher energy absorption capacity. Coupling the airbag with the back protector increases the effectiveness of the inflatable device by distributing the impact on a larger area of the bag providing the best protection for both flat and penetrant impact conditions.

Keywords Back protector, injury, motorcycle, safety device, wearable airbag.

I. INTRODUCTION

Motorcyclists are considered as vulnerable road users due to their relative lack of protection and their exposure to impacts when involved in accidents. Personal Protective Equipment (PPE) such as helmets, clothing, back protectors or wearable airbags are available to attenuate injuries.

Back protectors are arrangements of energy absorbing and/or impact spreading materials covering at least a portion of the back [1]. They were initially designed to avoid abrasion caused by the sliding on the road surface and produced using synthetic hard shell with little padding material to offer comfort [2]. An additional foam inner liner was added to the traditional hard shell to provide shock absorption and theoretically prevent injuries by reducing forces transmitted in direct blows to the back and spine [3].

Wearable airbags consist of inflatable bags embedded in the garments worn by motorcyclists which are activated in response to an accident [4]. The areas of the human body covered by these devices vary between manufacturers, but they are generally designed to protect the chest and the back of the rider. Some airbag devices are coupled with a back protector outside to protect the user from hard elements of the system, such as the gas generator, in case of no inflation and/or to distribute the impact on a wider area of the airbag.

Several works have already quantified the effectiveness of airbags in reducing the severity of thoracic injuries [5-7] which are, after head trauma, the main cause of death for motorcyclists [8-9]. However, no studies have been performed to evaluate PPEs against blunt impacts on the back. This type of impact can cause injuries on the thorax and spine [10-11] leading to haemorrhagic, respiratory and neurological issues for the victim. For instance, 40% of AIS2+ thoraco-lumbar spinal injuries in motorcycle accidents are caused by direct impacts to the back [12].

The objective of this work is to evaluate the effectiveness and the mechanisms of protection of an airbag, a back protector and their combination under back impacts in order to support the assessment and development of motorcyclists PPEs.

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II. METHODS

Finite element modelling was used to simulate blunt impacts to the back of a human body model, with and without protection, in order to assess the effectiveness of an airbag, a back protector and their combination. The software used in this study is the solver Radioss V2021.

Human Body, Airbag and Back Protector Modelling

The human body model used in this work is the HUman MOdel for Safety (HUMOS II) representing the 50th percentile adult male. It includes the description of compact and trabecular bones, internal organs, ligaments, muscles, tendons and skin [13]. The occupant model was chosen for its posture closer to that of motorcyclists and to facilitate the simulation of rear impacts due to the curvature of the back.

The airbag model was developed and validated in a previous work [5] based on a prototype designed by the manufacturer In&motion. The mechanical properties of the fabrics were obtained from tensile tests and the dynamic response of the model was correlated based on experimental drop impact tests.

The numerical models of two types of back protectors (called "Back protector 1" and "Back protector 2") currently included into In&motion airbag systems were developed from the CAD provided by the manufacturer. Both back protectors were made in foam, without hard shell, and have a thickness of 18mm. Surface variations and ventilation holes were not considered for modelling and geometries were filled with constant thickness to have a homogeneous behaviour on the entire surface. The models were meshed using 8-node hexahedron elements of 4 mm. The material properties were modelled with Radioss law 70, which is an experimental based tabulated strain rate dependent law for viscoelastic materials, recommended to reproduce the behaviour of foams.

Drop impact tests, an impactor falling onto the protector spread over an anvil, were carried out to evaluate the response of back protectors and obtain data for material law definition. The target strain rate was 0.25 s-1, corresponding to an impact velocity of 4.5 m/s to be comparable to the 1621-2 European standard for back protectors (4.43 m/s) [1]. Impacts with 1621-2 impactor shape, 7.6 and 14.9 kg masses, and velocities between 1.8 and 5.1 m/s were performed to obtain the deformation velocities necessary for the creation of material laws. A flat anvil of 28cm*28cm was chosen to simplify the calculation of foam deformation. A force cell installed under the anvil was used to measure transmitted forces, while a laser sensor monitoring the vertical displacement of the impactor allowed the calculation of foam deformation. Simulations of drop tests at 4.5-4.7 m/s with 7.6 kg 1621-2 impactor and the flat anvil were carried out and compared to the experimental results to correlate the behaviour of the models.

Impact Simulations and Evaluation of Safety Devices

The human model was coupled with the airbag and "Back protector 1" to obtain four main modelling configurations (Figure 1):

1. Without protection (NO PPE)
2. With back protector (BP)
3. With airbag (AB)
4. With airbag and back protector (AB+BP)

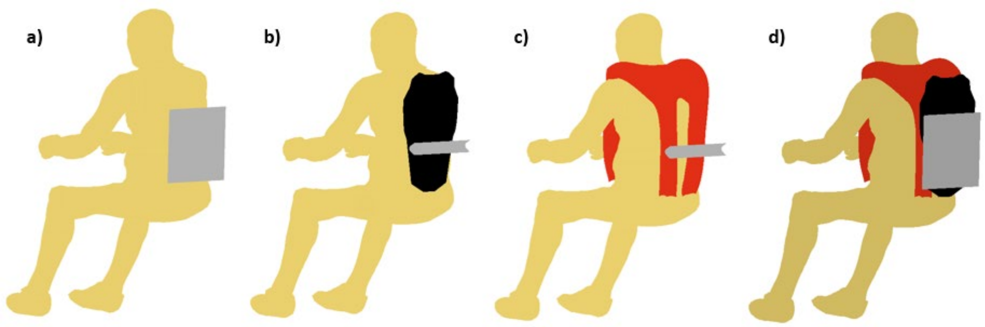

Fig. 1. Back impact simulations: (a) NO PPE; (b) BP; (c) AB; (d) AB+BP.

Two impactor shapes (kerb stone and plate), already used in a previous work [5], were defined to consider penetrant, e.g., a kerb or a pole, and flat obstacles with larger contact surfaces, e.g., the door of a vehicle or the ground. The kerb stone is rectangular on top with a length of 30 cm and a width of 5 cm. The impact face is a cylindrical face with a radius of 1.25 cm (Figure 1b). The plate is a square with a side of 30 cm having sharp edges (Figure 1a).

Impactors were meshed with 2D shell elements and modelled as rigid bodies with only one degree of freedom corresponding to the direction of the impact. Impactor centre of gravity was aligned with that of the vertebra L1, identified as one of the most vulnerable regions of the spine [14-15]. The plate impactor covered the back from T8 to L5, while the kerb stone only impacted the area over L1. Linear normal impacts were performed at 3, 5 and 7 m/s. Impact velocities were based on impact conditions obtained from motorcycle accident simulations, where the maximum normal impact velocity of the back against the ground was 7m/s and more than 75% of these impacts happened below 3 m/s [16]. The mass of the impactors was 23.4 kg to be comparable to previous research [5][10]. The movements of the human model were not constrained, that means it was not seated on any support and gravity acceleration was not considered.

Four magnitudes were measured to estimate impact severity and the level of protection of the devices:

1. Maximum impact force on the back.
2. Maximum of the intervertebral rotations measured on each segment from T1 to L5.
3. Maximum of thoraco-abdominal organ internal energies.
4. Back compression at L1 level.

Human model vertebrae were modelled with 2D shell elements and defined as rigid bodies. The anterior-posterior force component, measured as the sum of impactor-back and protector(s)-back contact forces, were analysed. Intervertebral rotations were measured on the joints between vertebrae, modelled with springs, in the sagittal plane of the human model. For internal organ energy, the heart, lungs, aorta, liver, kidneys and spleen were considered. Back compression was computed from the deflection, measured between a node on the thoracoabdominal skin and a node on the back skin at L1 level, and the corresponding initial torso depth (288 mm).

For each impactor shape and impact velocity, the results measured on the human model alone (NO PPE) were defined as reference values. Percentages of the corresponding reference value were calculated from simulations with protector (BP, AB and AB+BP) in order to quantify the benefits provided by each protective device.

III. RESULTS

Back Protector Models Validation

Drop impact tests at 4.5-4.7 m/s carried out with the mass of 7.6 kg were taken as reference for the validation of the models. Force versus time curves obtained from experimental tests were compared with those coming from simulations (Figure 2). The peak, duration and shape of the numerical curves are consistent with experimental data.

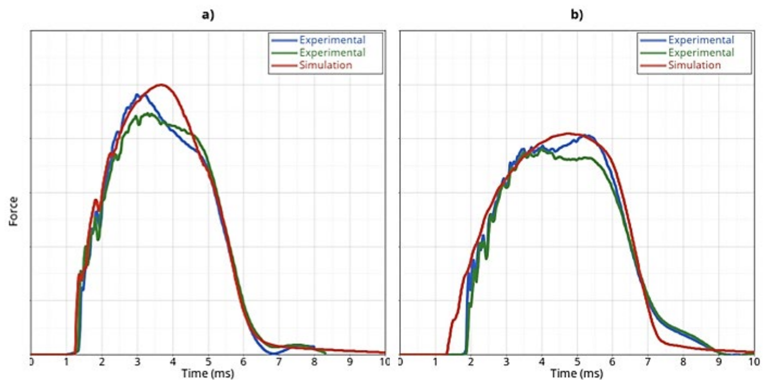

Fig. 2. Force versus time curves from drop impact tests: (a) Back protector 1; (b) Back protector 2.

Evaluation of Safety Devices

The parametric study of 24 back impact simulations showed that penetrant impacts are more severe for the motorcyclist considering intervertebral rotations, organ internal energies and back compression (TABLE I). Only contact forces were lower due to the wider impact surface of the plate leading to a more important deceleration of the impactor. Impact severity tends to increase with impact velocity for both impact surfaces. For all the simulations, the highest intervertebral rotations and internal organ energies were observed at T12-L1 joint and on the liver, respectively.

TABLE I
MAXIMUM VALUES OBTAINED FROM SIMULATIONS WITHOUT PPE

Impactor	Velocity (m/s)	Impact force (N)	Intervertebral rotation (deg)	Internal energy (mJ)	Back compression (%)
Plate	3	4402	7,3	782	31
	5	8201	9	2018	45
	7	15396	11,8	4284	62
Kerb stone	3	2775	16,3	943	42
	5	5076	23,5	2949	62
	7	11376	29	5876	82

Focusing on impact forces (Figure 3), the airbag offers higher levels of reduction (12-24%), compared to the back protector (0-2%), for the kerb stone impactor at 3 and 5 m/s. Contrary to the airbag, the contribution of the back protector increases with impact velocity. At 7 m/s the back protector reduces the maximum force by 39%, while the airbag by 9%. The highest mitigation was observed wearing the airbag and the back protector together (22-47%). For the plate impactor, low benefits (4-9%) of the back protector were observed below 7 m/s. In contrast to the back protector, the airbag performs better for wider impact surfaces reducing forces by 57% at 3 m/s and by 24% at 5 m/s. The contribution of both devices is closer at 7 m/s with a decrease of 29% for the back protector and 22% for the airbag. Coupling the two devices provides the best level of protection at 5 m/s (-33%) and 7 m/s (-37%).

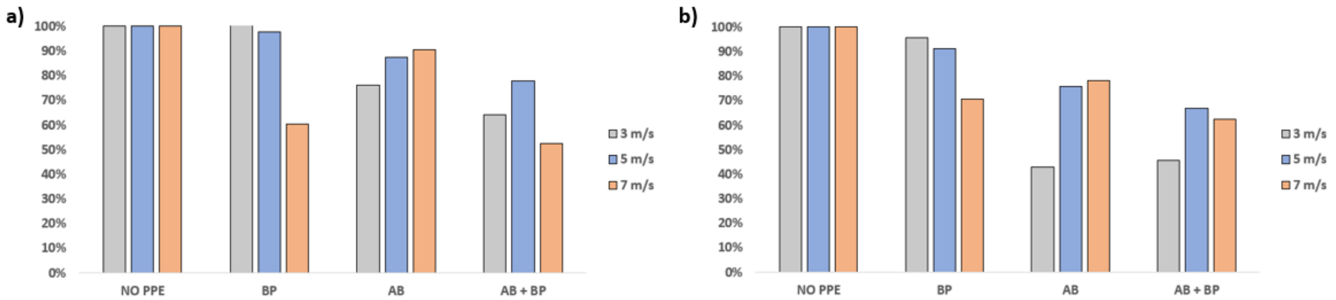

Fig. 3. Impact force: (a) Kerb stone impactor; (b) Plate impactor.

Regarding the intervertebral rotation (Figure 4), the back protector offers higher reductions for impacts against the kerb stone (16-19%) compared to plate impacts (0-3%). The effect of the airbag is more significant for flat impacts, decreasing rotation between 15% and 66% versus 7-30% if the back is impacted with the kerb stone. Considering both impactor shapes, the combination airbag-back protector enables the highest gain (21-67%).

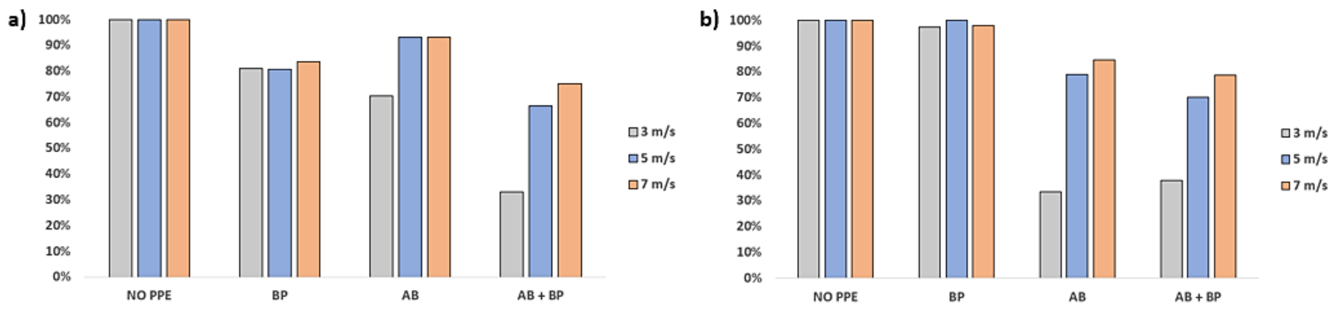

Fig. 4. Intervertebral rotation: (a) Kerb stone impactor; (b) Plate impactor.

Concerning the internal energy measured on the organs (Figure 5), the highest reduction wearing the back protector is 5%. The benefit of the airbag is more important for the flat impact surface (9-66%) compared to penetrating impacts (5-32%). Coupling the airbag and the back protector offered the greatest reduction of organ deformation energy, i.e., between 22% and 81% for the kerb stone and from 20% to 65% for the plate.

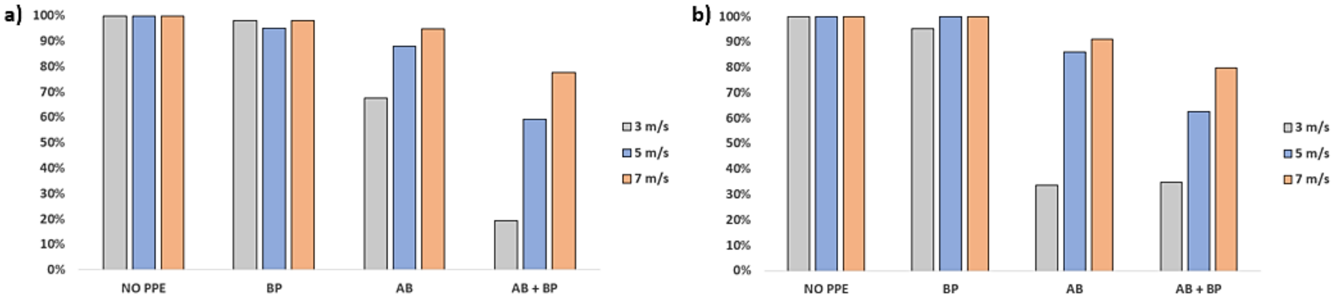

Fig. 5. Thoraco-abdominal organs' internal energy: (a) Kerb stone impactor; (b) Plate impactor.

Regarding the compression of the back (Figure 6), the effect of wearing the back protector is very low for kerb stone impacts (5-9%) and almost nil if the back is impacted by the plate (0-3%). The gain provided by the airbag is more significant, especially if the impactor is flat (decrease between 63% and 79%). In case of penetrant impacts, the benefit was lower (37-63%). The gain of coupling the airbag and the back protector is quite constant over different velocities and impactors with compression reductions between 63% and 73%.

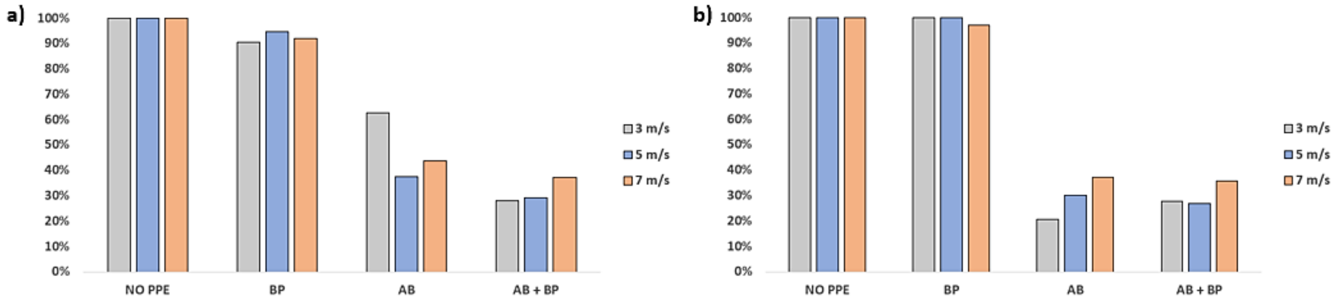

Fig. 6. Back compression: (a) Kerb stone impactor; (b) Plate impactor.

IV. DISCUSSION

The assessment of two PPEs for motorcyclists, an airbag and a back protector, and their combination under blunt impacts on the back was performed in this work. A parametric numerical study was carried out and impact forces, intervertebral rotations, internal energy of thoraco-abdominal organs as well as back compression were analysed as indicators of impact severity.

The main benefit of the back protector was the reduction of peak forces, which is coherent with the tests performed to certify these devices [1], at the highest impact velocity. This could be explained by their capacity to avoid direct contact between the human body and the impactor and distribute the impact on a wider surface leading to decreased intervertebral rotation in case of penetrant impact as well. Correlation of foams' behaviour at normal impact velocities higher than 4.7m/s and closer to 7 m/s, which could not be performed due to the height limit of the dropping apparatus, would be useful to give more confidence on these results. The energy absorption capacity of back protectors is low due to the stiffness of the material and its limited thickness. Indeed, the deformation of the torso and therefore of the ribcage and the internal organs were not mitigated.

In contrast with the back protector, the airbag reduced the compression of the back and internal organs deformation due to its higher energy absorption capacity. It also attenuated impact forces and rotations, especially for impacts with the plate. The performance of the airbag increases with impactor surface because a bigger amount of gas is displaced inside the bag increasing its internal pressure and therefore its compression resistance. The absorption capacity of an airbag is linked to its thickness, internal pressure and volume [5-7], between others, and a more exhaustive study would be necessary to define the optimal design to protect the back of the user.

The combination of the airbag with the back protector improves the effectiveness of airbag protectors providing the best protection for any type of impact. In case of flat impacts the benefits were slightly lower to those of the airbag alone at 3 m/s but better at 5 and 7 m/s. At 3 m/s the airbag was not fully compressed and the combination with the back protector reduced the contact surface between the impactor and the bag decreasing the displaced amount of gas. For penetrant impacts the back protector spread the shock over a larger surface of the bag increasing its performance. The back protector also helps to attenuate impacts if the airbag is completely compressed or not inflated.

To the best of our knowledge, this is the first research evaluating motorcyclist PPE against blunt impacts on the back comparing the effectiveness and the mechanisms of protection of air and foam safety devices. The lack of spinal and thoracic injury criteria for the simulated impacts is the main limitation of this work. The development of specific injury risk indicators seems necessary to quantify more accurately the severity of the injuries and the level of protection offered by each device. In the meantime, recommended biomechanical thresholds of impact force (4,73 kN) and torso deflection (67.1 mm) at T1 level [10] as well as spinal T12-L1 range of motion (6.7°) [17] show the severity of the simulated impacts, in particular at 5 and 7 m/s. In the simulations where these limits were exceeded, only the airbag (plate 3, 5 and 7 m/s and kerb stone 5 m/s) and/or the combination of airbag and back protector (plate 3, 5 and 7 m/s and kerb stone 3 and 5 m/s) decreased some of them below the thresholds. Intervertebral rotations and back compression were overestimated in this work because spinal movements are not limited by the contact between vertebrae (modelled as rigid bodies). Complementary research using other whole human body models would be interesting to check the robustness of the results of this work. Models with more refined mesh and more detailed spine modelling (3D mesh, intervertebral discs, ligaments, etc.), would be useful to go further in understanding the mechanisms of injury to the spine, internal organs and ribcage. Measuring strains could be interesting to estimate the risks of bone fracture and organ laceration [18]. Another possibility could be a multimodel approach by using the outputs of the whole human model as input loadings for an isolated spinal model such as SM2S [19]. The correlation of human models' behaviour, as performed by [20-21] at T1, T6 and T8, should also be considered to evaluate injury risks more accurately. For the present work, no biomechanical data for impacts at L1 has been found in literature. Analysing the influence of the method to measure back compression as done by [22] for the thorax could be another perspective to improve the biofidelity of human body models.

In view of improving the effectiveness of the evaluated safety devices, future studies should test other airbag and back protector designs in terms of shape, thickness, material properties or internal pressure. Further investigations could also consider other impact angles in order to analyse the effect of tangential impact velocities, which are higher than normal ones in motorcycle accidents [16], on injury risk. This would need a validation of the foams' response under impact conditions with shear effects, as done by [23] for helmet foams, and not only under perpendicular impacts. It would also be interesting to impact other areas of the back as dimensions, shape, mechanical properties and range of motion differ between vertebrae and spinal regions. The variation of impactor mass should also be studied in order to consider the effective weight of the rider depending on real impact conditions. Coupling the airbag with a chest protector to quantify the effect of combining an airbag and a foam under front chest impacts, based on long-established thoracic injury criteria already applied by [5-7], could be another perspective. Previous investigations have already showed the limits of foam protectors in attenuating chest injuries [24], but they could be a good way to improve airbag performance.

Understanding injury mechanisms is essential to develop effective protective devices and prevent injuries. In the present work, direct impacts at L1 level have been considered which do not cover all the possible causes of injury on this zone of the spine. For instance, normal impacts with the plate impactor higher up the spine could create shear injuries at L1. Previous research has pointed out that 50% of thoraco-lumbar spinal trauma in motorcycle accidents comes from indirect impacts [12] and most of these injuries are compression fractures caused by axial/bending loadings [2][14]. Since airbag and back protectors could also limit flexion-extension movements of the torso and the neck, quantifying the effect of these devices on spinal mobility seems necessary to evaluate their potential benefits in reducing bending. Performing dedicated epidemiological and biomechanical studies would be helpful to identify the type, occurrence and causes of spine injuries in order to develop protectors and evaluation standards more adapted to real accidentology and user needs.

V. CONCLUSIONS

This research has investigated the effect of wearing a back protector, an airbag and their combination on the attenuation of loadings on the back and the spine of motorcyclists under posterior-anterior impacts. Based on impact simulations, the mechanisms of protection of each type of PPE (foam, airbag and combination) were identified and their main benefits for the user were better understood. The back protector reduced peak forces and intervertebral rotations in cases of penetrant impacts. In addition to force and spinal rotation attenuation, especially for flat impacts, the airbag decreased the energy transmitted to the human body. Coupling the airbag with the back protector provided the best protection for any type of the impacts to the back by distributing the shock over a wider area of the airbag and increasing its absorption capacity.

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ERRATUM

Coupling a Wearable Airbag with a Back Protector: What are the Effects on Motorcyclists' Safety?

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After final paper submission, a mistake was found in the unit of strain rate. The targeted strain rate for the creation of back protectors' material laws was 250s^{-1} (=0.25ms^{-1}) and not 0.25 s^{-1}.

It was also discovered that for some simulations the compression of the back was incorrectly calculated. An updated Figure 6 and the corresponding paragraph describing it are presented below. Considering the new results, only the airbag (plate 3 and kerb stone 5 m/s) and/or the combination of airbag and back protector (plate 3 and 5 m/s and kerb stone 3 and 5 m/s) decreased some of the measured magnitudes below the biomechanical thresholds indicated in the article. These modifications do not change the overall conclusions of the paper.

Regarding the compression of the back (Figure 6), the effect of wearing the back protector is very low for kerb stone impacts (8-9%) and almost nil if the back is impacted by the plate (0-3%). The gain provided by the airbag is more significant, especially if the impactor is flat (decrease between 18% and 79%). In case of penetrant impacts, the benefit was lower (8-37%). The gain of coupling the airbag and the back protector is quite comparable for both impactors with compression reductions between 20% and 73%.

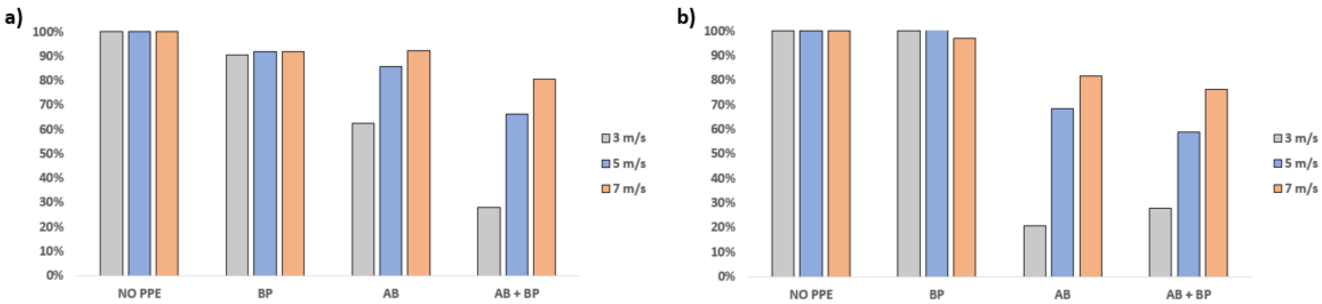

Fig. 6. Back compression: (a) Kerb stone impactor; (b) Plate impactor.