Assessment of an Anti-Rotation Technology effect and Pedelec Bicycle Helmets based on Certimoov methodology

Nicolas Bourdet, Caroline Deck, Frank Meyer, Remy Willinger

Abstract This study evaluates the protective performance of commercial bicycle and Pedelec helmets and investigates the effectiveness of the rotational mitigation technologies MIPS. Helmets were tested using the Certimoov helmet assessment methodology, which combines experimental impact testing with numerical brain injury modelling. A sets of 46 adult helmets, including models with and without MIPS and helmets certified under EN 1078 or additionally under the NTA 8776 Pedelec standard, was subjected to six impact configurations: three linear impacts on a flat anvil at 5.5 m/s, and three oblique impacts on a 45° anvil at 6.0 m/s. Each configuration was repeated three times using an instrumented Hybrid III headform. The measured six-degree-of-freedom head kinematics were used as input to the SUFEHM finite element head model to estimate intracerebral Von Mises stress and the associated risk of moderate neurological injury.

Results show that helmets equipped with MIPS significantly reduce rotational kinematics and brain injury risk in oblique impacts, while their benefit remains limited in purely linear impacts. In contrast, helmets certified under the NTA 8776 Pedelec standard did not demonstrate superior protection compared with conventional EN 1078 helmets. Overall, the combined experimental-numerical approach highlights substantial variability in helmet performance and underlines the importance of considering rotational impacts and brain-based injury criteria in helmet evaluation standards.

Keywords Pedelec helmet, Rotational-mitigation, Brain injury risk, Oblique impacts, Helmet performance.

I. INTRODUCTION

The rapid growth of micromobility modes of transport, such as bicycles, electric bicycles, and electric scooters, has noticeably transformed urban transportation systems. While these modes offer environmental and mobility benefits, they also raise important safety concerns for vulnerable road users (VRUs), who travel without structural protection. In urban environments where pedestrians, cyclists, and users of personal mobility devices increasingly share the same infrastructure, collisions and fall events can result in severe injuries. Epidemiological studies consistently show that head injuries are among the leading causes of severe trauma and fatalities for cyclists and other VRUs involved in traffic accidents [1].

Helmet use is widely recognised as one of the most effective protective measures for reducing the risk of traumatic brain injury (TBI). Current bicycle helmet certification standards, such as EN-1078 [2], assess helmet performance using vertical drop tests onto rigid anvils and primarily guarantee a minimal protection through peak linear acceleration thresholds. However, real-world accidents frequently involve oblique impacts, where the head experiences both linear and rotational motion [3-4]. Numerous biomechanical studies have demonstrated that the rotational kinematics of the head play a key role in the mechanisms of brain injury, particularly in the development of diffuse axonal injuries, which has been recognised for several decades. Early biomechanical investigations by Holbourn [5] demonstrated that rotational loading of the head could generate significant shear stresses within brain tissue, potentially leading to cerebral concussion. This hypothesis was later supported by the experimental work of Ommaya *et al*. [6], who investigated rotational head dynamics in primate models and proposed scaling approaches to estimate concussion thresholds in humans.

Subsequent research has consistently confirmed the importance of rotational head kinematics in brain injury mechanisms. Studies conducted by Gennarelli *et al.* [7], Deck *et al*. [8], Kleiven [9], Zhang *et al*. [10], and Takhounts *et al*. [11] highlighted the relationship between rotational acceleration, intracranial loading, and the relative

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motion between the brain and skull, which are key factors in the occurrence of neurological injuries such as diffuse axonal injury and subdural haematoma.

Furthermore, experimental and accident reconstruction studies have shown that rotational head motion is primarily generated by tangential forces during oblique impacts. Mills *et al*. [12] and Bourdet *et al*. [13] demonstrated that such tangential loading can induce significant rotational accelerations of the head. These findings are consistent with accident analyses reported by Otte *et al*. [14] and Harrison *et al*. [15], which revealed that a large proportion of head impacts in motorcycle crashes occur under oblique impact configurations.

In response to these limitations, new helmet technologies have been introduced to mitigate rotational loading during oblique impacts. Among them, rotational management systems, such as the Multi-directional Impact Protection System (MIPS), aim to reduce rotational forces transmitted to the head by allowing relative motion between the helmet and the head during impact. At the same time, the development of high-speed electric bicycles has led to the emergence of new helmet categories, such as Pedelec helmets certified under the NTA-8776 helmet standard, which have introduced stricter impact conditions than conventional bicycle helmet standards.

Overview of the Multi-directional Impact Protection System (MIPS)

The MIPS system is a rotational management technology designed to reduce rotational head kinematics during oblique impacts. The system incorporates a low-friction slip layer within the helmet that allows limited relative motion between the head and the helmet shell, thereby reducing the rotational forces transmitted to the head. By mitigating rotational acceleration, MIPS aims to reduce brain tissue shear strains associated with traumatic brain injuries such as concussion and diffuse axonal injury [7][9][16]. MIPS-based systems are now widely implemented in commercially available bicycle helmets [17].

Overview of the EN 1078 Standard for Bicycle Helmets

The EN 1078 standard defines the safety requirements and testing procedures for helmets intended for cyclists, skateboard users, and roller skaters using conventional bicycles and low-speed micromobility devices. Helmet impact performance is evaluated through linear drop tests performed at impact velocities of 5.42 m/s on flat anvils and 4.57 m/s on kerbstone anvils. Tests are conducted at predefined helmet impact locations using EN 960 instrumented headforms to measure linear head accelerations. Compliance with the standard is assessed based on peak resultant linear acceleration, with a maximum allowable threshold of 250 g. It should be noted that the current EN 1078 standard does not assess rotational head kinematics during impacts.

Overview of the NTA 8776 standard for S-Pedelec helmets

The NTA 8776 helmet standard specifies requirements and test procedures for helmets intended for riders of high-speed electrically power-assisted bicycles (S-EPAC), commonly referred to as Pedelec, which can reach assisted speeds of up to 45 km/h. This Dutch Technical Agreement was developed in response to the emergence of this new vehicle category and the absence of a dedicated European helmet standard for S-EPAC users. Consequently, the Dutch bicycle industry and standardisation bodies established NTA 8776 to provide an intermediate certification framework adapted to the higher speeds of these vehicles.

The standard builds upon the conventional bicycle helmet standard EN 1078, but introduces several modifications intended to increase the level of protection. These changes mainly concern the shock-absorbing requirements and the protected area of the helmet. In particular, the test area is extended to improve coverage of the temporal and occipital regions, which are considered vulnerable zones in bicycle crashes.

Regarding impact testing, helmets are subjected to higher impact velocities than those specified in EN 1078. The test procedure includes impacts on a flat anvil at 6.5 m/s and on a kerbstone anvil at 5.42 m/s, while the maximum peak linear acceleration must remain below 250 g. Tests are conducted using EN 960 headforms instrumented with tri-axial accelerometers.

Although NTA 8776 increases the severity of linear impact tests and enlarges the protective area, the evaluation remains mainly based on peak linear acceleration criteria (PLA). Consequently, the standard does not explicitly address rotational head kinematics, which are increasingly recognised as a key factor in TBI mechanisms during oblique impacts.

Several studies investigating helmet performance under oblique impacts [12][17-19] have focused mainly on peak rotational head acceleration, often neglecting the time evolution and direction of rotation. To better account for the complex mechanical response of the brain, researchers have increasingly relied on tissue-level injury criteria derived from finite element (FE) head models. Deck *et al*. [20] showed that helmet optimisation strongly depends on the head surrogate and injury criteria used, while more recent studies have coupled helmet and human head FE models to improve biomechanical evaluation [9-10][21-22].

Experimental methods have also been proposed to investigate tangential and oblique impacts, including rotating disc tests [18] and sliding anvil configurations [17]. Similar approaches have been applied to bicycle, motorcycle, and hockey helmets, combining measurements of linear and rotational head kinematics with FE-based injury metrics to better assess helmet protective performance [23-30].

Objectives of the study

The objective of this study was to evaluate the protective performance of commercially available bicycle helmets using an experimental-numerical methodology designed to better represent real-world accident conditions. More specifically, the study aims to compare helmets incorporating the MIPS system, with helmets without such systems in order to quantify their effectiveness in reducing head kinematics and brain injury risk during impact.

In addition, the study investigates the protective capabilities of helmets designed for Pedelec helmets, which are certified under the NTA 8776 standard. Their performance is compared with that of conventional bicycle helmets certified only under EN 1078, in order to determine whether the stricter certification requirements translate into improved biomechanical protection.

To achieve these objectives, a series of realistic linear and oblique helmet impact tests was conducted using the Certimoov method [31] combined with the Strasbourg University Finite Element Head Model (SUFEHM) to estimate brain injury risk. This approach provides a comprehensive assessment of helmet protective performance and enables a comparative evaluation of current helmet technologies available on the market.

II. METHODS

This section presents the methodology developed to comparatively evaluate the protective performance of bicycle helmets, based on the approach previously described by Deck *et al.* [24]. Particular attention was given to comparing helmets incorporating the MIPS rotational management system with helmets without such technology, as well as helmets specifically designed for electric bicycle use (Pedelec helmets) with conventional bicycle helmets.

A total of 46 adult bicycle helmet models were analysed in this study, listed in Table II. These models span a wide price range and are representative of the current consumer market. The evaluation relied on realistic impact tests based on the Certimoov testing protocol [24], developed at Strasbourg University. The protocol includes impacts on both horizontal and oblique anvils and is combined with numerical simulations using the SUFEHM. This combined experimental-numerical approach enables the estimation of brain injury risk associated with helmet impacts.

The finite element head model SUFEHM, illustrated in Fig. 1, incorporates realistic constitutive laws for both brain and skull tissues, and enables the computation of tissue-level brain responses under impact loading. Validation of this head model was previously reported in earlier studies [8-9], against experimental local brain motion data [10-11], as well as against intracranial pressure measurements [12-13]. Following this validation, 125 real-world head trauma cases were reconstructed using SUFEHM in order to derive brain injury criterion. An extensive logistic regression analysis [14-15] was performed on several computed intracerebral parameters and associated percentiles. The Nagelkerke coefficient, used to assess the quality of the regression, identified the maximum brain von Mises stress (VMS) as the most relevant predictor of moderate concussion or short coma corresponding to AIS2+ (Abbreviate Injury Scale) injuries. A brain injury tolerance threshold corresponding to a 50% risk of this reversible injury was established at 36.5 kPa [16-17]. The corresponding injury risk function, obtained using binary logistic regression applied to the VMS values computed with SUFEHM, is provided in Eq. (1) and illustrated in Fig. 1. Since the curve was derived using the best mathematical fit rather than being constrained to pass through the origin, it does not predict a zero injury risk at zero stress.

$$Risk = \frac{1}{1 + e^{3.178 - 0.087.VMS}} \tag{1}$$

where VMS is the maximum value of brain Von Mises Stress in kPa.

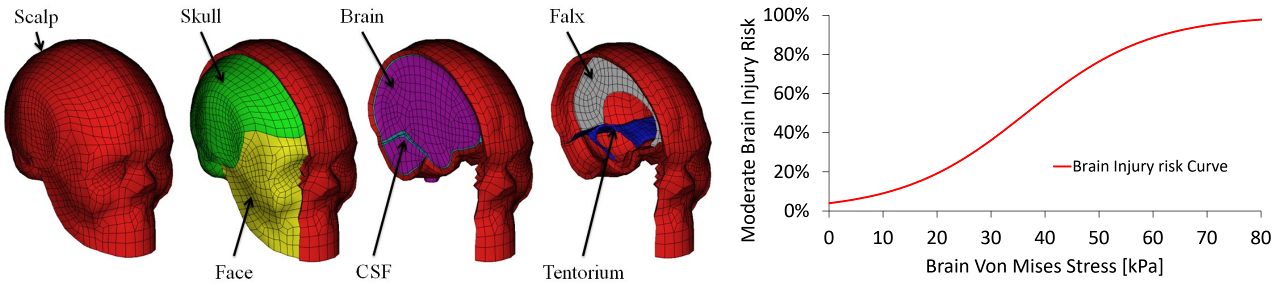

Fig. 1. Illustration of SUFEHM and the related brain tolerance curve.

The methodology described in Deck *et al* [24], consists of six helmeted headform impact configurations: three linear impacts on a flat anvil, and three oblique impacts anvil. To ensure experimental repeatability, each impact configuration was repeated three times according to the test matrix reported in in Table I. For this purpose, six sample helmets were used, each being impacted three times at locations sufficiently separated to avoid influencing adjacent impact areas. This was verified through the test matrix, which ensured that each impact configuration was performed as a first impact condition.

TABLE I

TEST MATRIX FOR THE THREE LINEAR AND THREE OBLIQUE IMPACT TESTS INVOLVING SIX DIFFERENT HELMETS

Impact configurations	Helmet sample ID	First Impact	Second Impact	Third Impact	Impact Velocity
Linear impacts	H1	FRONTAL	OCCIPITAL	LATERAL	5.5 m/s
	H2	LATERAL	FRONTAL	OCCIPITAL	5.5 m/s
	H3	OCCIPITAL	LATERAL	FRONTAL	5.5 m/s
Oblique impacts	H4	YROT	XROT	ZROT	6.0 m/s
	H5	ZROT	YROT	XROT	6.0 m/s
	H6	XROT	ZROT	YROT	6.0 m/s

Therefore, for each helmet model, the methodology involved 18 experimental impact tests performed on six helmet samples, followed by numerical simulations to compute the brain response and estimate the corresponding injury risk for each impact configuration. More specifically, the experimental procedure, illustrated in Fig. 2, consists of three linear impacts against a horizontal flat anvil at a velocity of 5.5 m/s, in accordance with the standard impact velocity conditions (FRONTAL, OCCIPITAL, and LATERAL) and three oblique impacts against a 45° inclined anvil covered with P40 grit abrasive paper. The oblique impacts were conducted at a velocity of 6.0 m/s. The experimental procedure followed the methodology described by Deck *et al.* [24], where the rationale for the selection of impact configurations, impact velocities is provided.

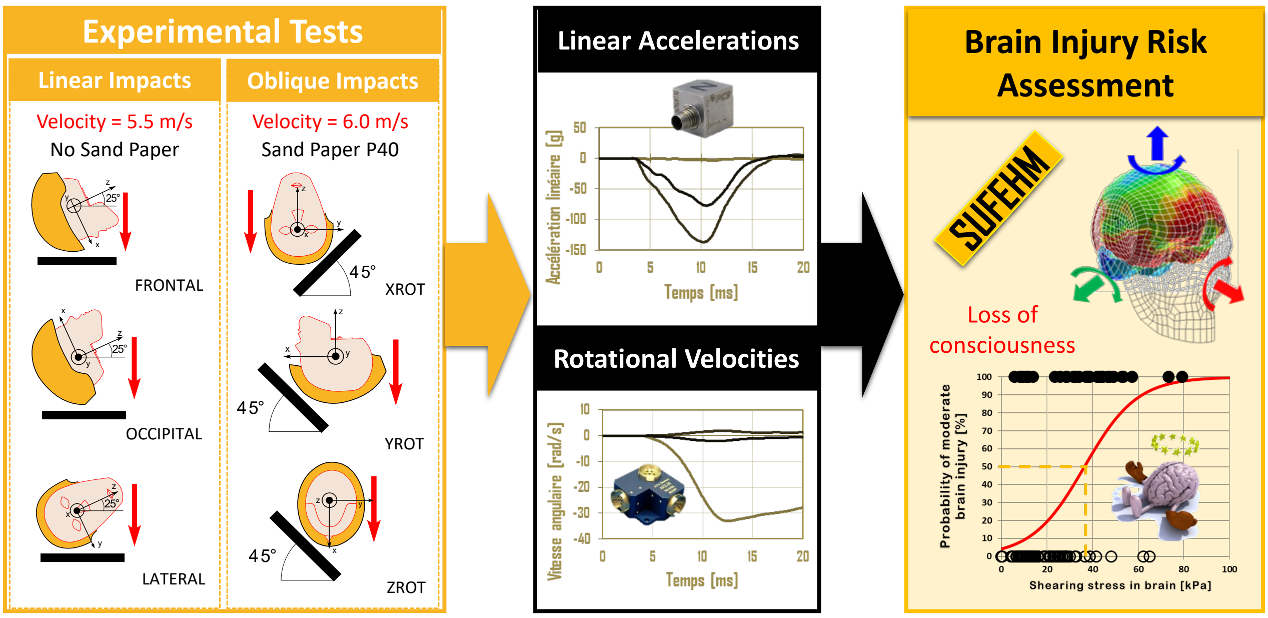

Fig. 2. Illustration of the coupled experimental and numerical test method. The experimental acceleration of the headform is considered as an input condition for the numerical simulation of the head impact followed by the brain injury risk assessment.

For each test, six-degree-of-freedom (6DoF) kinematic data were recorded, including three linear accelerations and three rotational velocities, and used as input for the SUFEHM to evaluate brain injury criteria at the tissue level.

Based on the computed SUFEHM injury risk, a comparative rating score was additionally assigned to each helmet in order to facilitate global comparison between helmet models and impact configurations. The rating system was not intended to replace the underlying biomechanical metrics, but rather to provide a simplified synthesis of the injury risk results derived from the tissue-level analysis. Consequently, the scientific interpretation of helmet performance in the present study primarily relies on the measured kinematic parameters and the SUFEHM-based injury criteria. The evaluated helmets were rated on a scale ranging from 0.5 to 5 stars, with increments of half a star, according to the calculated average risk, as illustrated in Fig. 1 and Equation 1. A rating of 5 stars corresponds to an average injury risk below 6%, indicating a high level of protection, while a rating of 0.5 star is assigned when the calculated injury risk exceeds 86%, corresponding to the lowest level of protection.

$$Rating\ Score\ (risk) = \begin{cases} \dfrac{\left\lfloor \mathbf{10} - \dfrac{\lfloor \mathbf{100} \cdot \boldsymbol{risk} \rfloor}{\mathbf{10}} + \mathbf{0.5} \right\rfloor}{\mathbf{2}} & if\ \neq 0 \\ \mathbf{0.5}\ if\ \dfrac{\left\lfloor 10 - \dfrac{\lfloor 100 \cdot risk \rfloor}{10} + 0.5 \right\rfloor}{2} = 0 \end{cases} \tag{1}$$

For all experimental impact tests, a more realistic head surrogate was used to measure both linear and rotational kinematics over time, with a more biofidelic moment of inertia than the EN 960 headform used in current standards Deck *et al* [24]. The EN 960 headform was therefore replaced by an instrumented Hybrid III dummy head equipped with PCB tri-axial linear accelerometers and ATA-type angular velocity sensors. Linear accelerations were measured using PCB Piezotronics 356B21 accelerometers (±500 g), with sensitivities of 10 mV/g. Rotational velocities were recorded using RS-06 triaxial MHD angular rate sensor arrays with a sensitivity of 50 mV/rad/s. Data were acquired at 25.6 kHz and filtered using CFC 1000 for linear accelerations and CFC 180 for rotational velocities according to Deck *et al* [24].

Helmet performance was evaluated using seven injury metrics: PLA, peak rotational velocity (PRV), peak rotational acceleration (PRA), the Head Injury Criterion (HIC) [41], the Brain Injury Criterion (BrIC) [11], and the tissue-level brain injury metric derived from SUFEHM simulations. PRA was calculated from the rotational velocity time histories by numerical differentiation using a central finite-difference method, while HIC and BrIC were calculated according to Equations (2) and (3), respectively:

$$HIC = max_{(t_1,t_2)} \left\{ (t_2 - t_1) . \left(\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t)dt \right)^{2.5} \right\} \tag{1}$$

$$BrIC = \sqrt{\left(\frac{\omega_x}{\omega_{xC}}\right)^2 + \left(\frac{\omega_y}{\omega_{yC}}\right)^2 + \left(\frac{\omega_z}{\omega_{zC}}\right)^2} \tag{2}$$

where ω_x, ω_y and ω_z are maximum angular rates on X-, Y-, and Z-axis, respectively, and ω_{xC}, ω_{yC} and ω_{zC} are the critical angular velocities in their respective directions: ω_{xC} = 66.25 rad/s ; ω_{yC} = 56.45 rad/s ; ω_{zC} = 42.87 rad/s.

A total of 46 adult bicycle helmets, listed in Table II, were tested in 2024 using the linear and oblique impact methodology: 14 EN 1078-certified helmets equipped with MIPS technology, 17 EN 1078-certified helmets without rotational management systems, and 15 helmets certified under both NTA 8776 and EN 1078 standards, some of which also incorporated MIPS technology.

From this overall dataset, two specific helmet subsets were considered in the present study. The first subset consisted of 12 MIPS-paired helmet models, corresponding to a total of 24 helmets (12 equipped with MIPS and 12 without MIPS), and was used to compare the effect of the MIPS rotational management system. The second dataset, including all 46 helmet models, was used for the broader comparison between Pedelec helmets certified under NTA 8776 and conventional helmets certified under EN 1078.

The 46 helmet models involved six helmet samples subjected to three impacts, for a total of 828 impacts.

Differences between helmet groups were assessed using the Mann and Whitney U test [42], a non-parametric test comparing the distributions of two independent samples without assuming normality. The test evaluates the null hypothesis that both groups originate from the same population, with a *p-value* below the significance threshold (typically 0.05) indicating a statistically significant difference. Effect sizes were additionally calculated using the rank-biserial correlation [43] to quantify the magnitude of group differences independently of scale. In this study, only a larger effect size was considered. This combined approach strengthens the robustness of the comparisons despite unequal sample sizes.

TABLE II
LIST OF TESTED HELMET MODELS WITH SIZE AND REGULATION TESTED IN 2024.

N°	BRAND	MODELS	SIZE	STANDARD	N°	BRAND	MODELS	SIZE	STANDARD
1	GIRO	AGILIS-MIPS	M	EN 1078	10	LAZER	STRADA	L	EN 1078
2	GIRO	SYNTAX-MIPS	M	EN 1078	11	ABUS	GAMECHANGER	L	EN 1078
3	ENDURA	FS260-PRO-MIPS-HELMET-II	M/L	EN 1078	12	MET	Allroad	L	EN 1078
4	FOX	SPEEDFRAME-PRO-MIPS	L	EN 1078	13	Abus	Skurb	L	EN 1078
5	MET	Allroad-MIPS	L	EN 1078	14	Abus	Urban-I-3.0	L	EN 1078
6	Abus	Skurb-MIPS	L	EN 1078	15	Abus	Moventor-2.0	L	EN 1078
7	Abus	Moventor-2.0-MIPS	L	EN 1078	16	MET	MOBILITE	M/L	EN 1078
8	MET	MOBILITE-MIPS	M/L	EN 1078	17	MET	MILES	L	EN 1078
9	MET	MILES-MIPS	L	EN 1078	1	ENDURA	SPEED-PEDELEC	L/XL	NTA 8776
10	Abus	Urban-I-3.0-MIPS	L	EN 1078	2	Abus	Pedelec-2.0	L	NTA 8776
11	UNIT1	FARO-MIPS	M	EN 1078	3	ABUS	Pedelec-2.0-MIPS	L	NTA 8776
12	ABUS	GAMECHANGER-MIPS	L	EN 1078	4	BBB	Indra-Faceshield	L	NTA 8776
13	FOX	SPEEDFRAME-RS-MIPS	L	EN 1078	5	BONTRAGER	Charge-Wavecel	L	NTA 8776
14	BERN	MACON-2.0-MIPS	L	EN 1078	6	CASCO	Speedairo-2-RS	L	NTA 8776
1	URGEBIKE	CAB	L	EN 1078	7	CRATONI	Commuter	L	NTA 8776
2	Egide	INO	M/L	EN 1078	8	LAZER	Anvez-NTA-MIPS	L	NTA 8776
3	GIRO	AGILIS	M	EN 1078	9	POLISPORT	E-City	L	NTA 8776
4	GIRO	SYNTAX	M	EN 1078	10	LUMOS	Ultra-E-BIKE-MIPS	M/L	NTA 8776
5	ENDURA	FS260-PRO-HELMET-II	M/L	EN 1078	11	FOX	DROPFRAME-PRO-MIPS	L	NTA 8776
6	thousand	Chapter	M	EN 1078	12	LIMAR	LUMIERE-E-BIKE-MIPS	M	NTA 8776
7	thousand	Heritage-2.0	M	EN 1078	13	Bern	Hudson-MIPS	L	NTA 8776
8	LUMOS	Ultra	M/L	EN 1078	14	MET	Urbex-MIPS	L	NTA 8776
9	LIMAR	DELTA	L	EN 1078	15	UNIT1	AURA-MIPS	M	NTA 8776

III. RESULTS

Results from the experimental impact tests and the associated brain injury risk computed using SUFEHM are presented below. In a first part, the analysis compares helmets equipped with the MIPS rotational management system with helmets without such a system. In a second step, Pedelec helmets certified under NTA 8776 are compared with conventional helmets certified only under EN 1078 in order to assess the influence of certification requirements on helmet protective performance. All peak kinematic values presented in the results are resultant values.

Comparison of helmets with and without MIPS

This section compares the experimental results obtained for helmets equipped with the MIPS rotational management system and the same helmet models without such technology, based on a subset of 12 different helmet models. In total, 432 tests and simulations were performed across six impact configurations, enabling a detailed evaluation of helmet protective performance. Figure 3 shows the distribution of PLA, PRA, and PRV measured during the helmet impact tests for models with and without the MIPS system. Overall, helmets equipped with MIPS tended to exhibit lower kinematic values than helmets without such technology. On average, helmets without MIPS produced a peak linear acceleration of 138 ± 22 g, compared with 134 ± 28 g for helmets equipped with MIPS (*p-value* < 0.05). A more pronounced difference was observed for rotational kinematics, with peak rotational acceleration decreasing from 6.1 ± 2.2 krad/s² for helmets without MIPS to 5.0 ± 2.0 krad/s² for helmets with MIPS (*p-value* < 0.05). Similarly, peak rotational velocity decreased from 23 ± 9 rad/s to 20 ± 7 rad/s (*p-value* < 0.05).

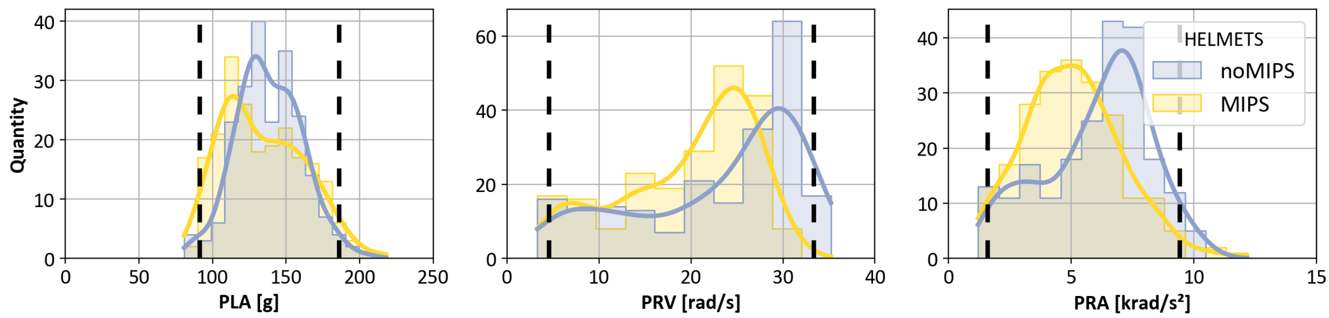

Fig. 3. Distribution of the data in terms of PLA, PRV as well as PRA obtained from all the helmet tests, all configurations combined according to standard group.

Consistent trends were also observed for injury criteria, with mean HIC values decreasing from 712 ± 184 (without MIPS) to 665 ± 232 (with MIPS), and mean BrIC values decreasing from 0.42 ± 0.18 to 0.36 ± 0.14. The corresponding mean values for all tests are summarised in TABLE III and in Fig. 4. These differences were confirmed by the calculation of the Mann–Whitney U test with *p-values < 0.05*.

TABLE III
MEAN VALUES OF PEAK RESULTANT LINEAR ACCELERATION, PEAK ROTATIONAL ACCELERATION, PEAK ROTATIONAL VELOCITY, HIC, AND BRIC OBTAINED FROM HELMET TESTS ACCORDING TO THE PRESENCE OR ABSENCE OF MIPS TECHNOLOGY (P-VALUES < 0.05)

TYPE	Impact Counts	PLA [g]	PRA [krad/s²]	PRV [rad/s]	HIC	BrIC
no MIPS helmets	216	138 ± 22	6.1 ± 2.2	23 ± 9	712 ± 184	0.42 ± 0.18
MIPS helmets	216	134 ± 28	5.0 ± 2.0	20 ± 7	665 ± 232	0.36 ± 0.14
All	432	136 ± 25	5.5 ± 2.1	21 ± 8	689 ± 210	0.39 ± 0.16

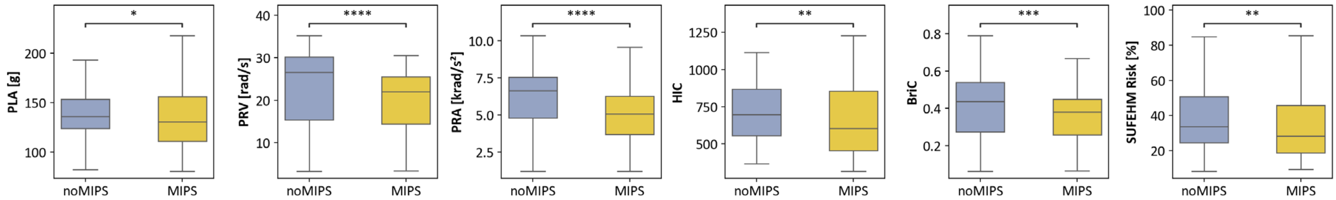

Fig. 4. Comparison of the distributions of global kinematic parameters and mean neurological injury risk for all helmet models (without MIPS; with MIPS). *Increasing numbers of asterisks indicate increasing levels of statistical significance (p decreasing), while ns indicates a non-significant difference (p > 0.05).*

Fig. 5 (below) presents the distribution of the different injury metrics for each impact configuration for helmets equipped with the MIPS system and those without it. For linear impacts on a horizontal flat anvil (FRONTAL, LATERAL, and OCCIPITAL), the effect of the MIPS system appears limited, with no statistically significant differences observed for most parameters (ns). In some cases, slightly higher values were observed for MIPS-equipped helmets, particularly for lateral and occipital configurations. In contrast, for oblique impacts (XROT, YROT, and ZROT), the presence of the MIPS system led to a systematic reduction in rotational kinematics and associated injury criteria. Significant decreases were observed for PRV, PRA, BrIC, and SUFEHM risk, with several configurations showing strong statistical significance (*p-value < 0.001*).

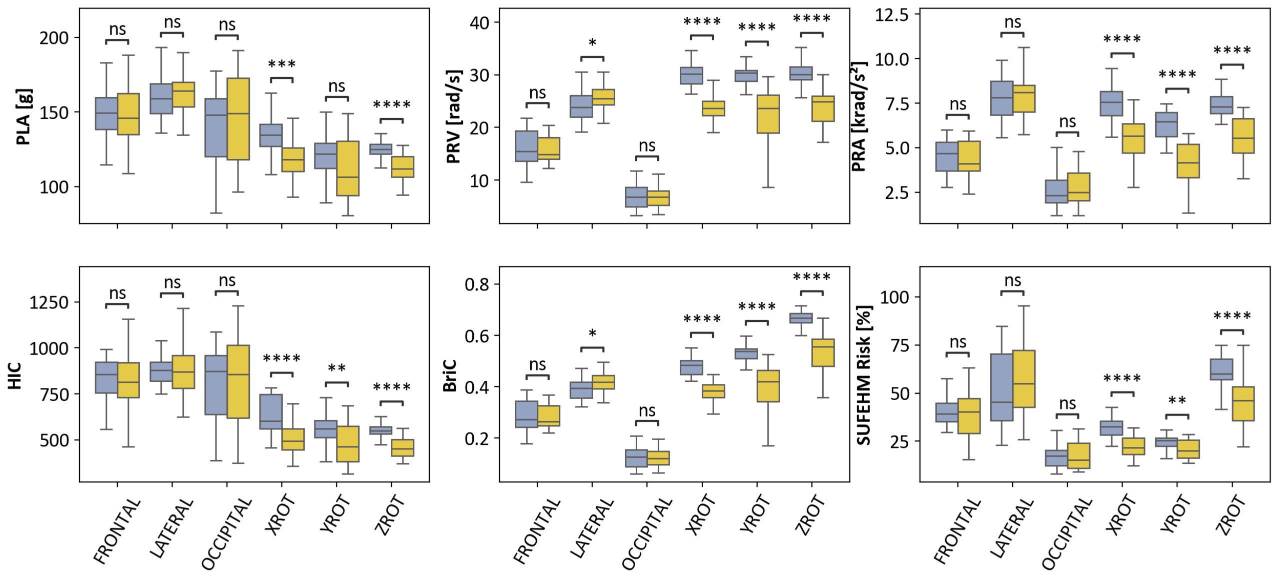

Fig. 5. Comparison of the distributions of global kinematic parameters and mean neurological injury risk for each impact configuration across all helmet models (without MIPS; with MIPS). *Increasing numbers of asterisks indicate increasing levels of statistical significance (p decreasing), while ns indicates a non-significant difference (p > 0.05).*

Table IV reports the mean values and standard deviations of the different kinematic and injury metrics obtained from the 12 paired-MIPS helmet models. The table includes PLA, PRV, PRA, HIC, BrIC, SUFEHM VMS, and SUFEHM risk. Gray cells indicate statistically significant differences between the two helmet groups at the 95% confidence level (*p-value* < 0.05), as determined using the Mann–Whitney U test. The corresponding effect size is reported using the rank-biserial correlation. Configurations marked with an asterisk (*) denote a large effect size, whereas the absence of an asterisk indicates a small to moderate effect. Although the statistical analysis revealed significant differences between group means, the associated standard deviations were often large and, in several cases, exceeded the mean differences between the two groups. This trend was observed across most parameters and impact configurations, highlighting the substantial variability inherent in the experimental results.

TABLE IV

MEAN VALUES OF THE DIFFERENT PARAMETERS. GRAY CELLS CORRESPOND TO A SIGNIFICANT DIFFERENCE BETWEEN MEANS WITH 95% CONFIDENCE (*P-VALUE*<0.05), AS DETERMINED BY THE MANN-WHITNEY U TEST. EFFECT SIZE IS REPORTED USING THE RANK-BISERIAL CORRELATION; A STAR (*) DENOTES A LARGE EFFECT SIZE, WHILE THE ABSENCE OF A STAR INDICATES A SMALL TO MODERATE EFFECT.

		FRONTAL	OCCIPITAL	LATERAL	XROT	YROT	ZROT
PLA [g]	No MIPS	149.3 ± 16.7	140.2 ± 26.1	161.6 ± 16.8	*134.0 ± 14.8	120.0 ± 15.1	*124.9 ± 6.1
	MIPS	147.9 ± 21.6	145.0 ± 28.5	164.9 ± 18.8	119.0 ± 14.9	112.5 ± 19.4	115.3 ± 15.0
PRV [rad/s]	No MIPS	16.0 ± 3.5	7.1 ± 2.4	24.2 ± 2.6	*30.1 ± 2.1	*30.0 ± 2.0	*30.5 ± 2.3
	MIPS	15.8 ± 2.4	6.9 ± 2.1	25.5 ± 2.5	23.2 ± 3.0	22.4 ± 5.7	23.9 ± 2.9
PRA [rad/s²]	No MIPS	4508 ± 984	2604 ± 958	7805 ± 1195	*7672 ± 1273	*6309 ± 787	*7477 ± 697
	MIPS	4319 ± 933	2806 ± 1027	7984 ± 1376	5492 ± 1238	4107 ± 1316	5573 ± 1062
HIC	No MIPS	830 ± 120	804 ± 208	894 ± 108	*626 ± 106	559 ± 98	*559 ± 59
	MIPS	806 ± 173	830 ± 234	885 ± 151	507 ± 89	489 ± 116	474 ± 86
BrIC	No MIPS	0.29 ± 0.06	0.13 ± 0.04	0.39 ± 0.04	*0.48 ± 0.03	*0.53 ± 0.04	*0.67 ± 0.05
	MIPS	0.28 ± 0.04	0.13 ± 0.04	0.42 ± 0.04	0.38 ± 0.05	0.40 ± 0.10	0.53 ± 0.07
SUFEHM VMS [kPa]	No MIPS	31.3 ± 5.5	18.0 ± 5.0	37.5 ± 10.0	*28.0 ± 2.8	23.3 ± 2.7	*41.7 ± 4.1
	MIPS	30.7 ± 7.2	18.0 ± 5.9	41.8 ± 11.6	21.8 ± 3.8	21.0 ± 3.5	34.3 ± 6.1
SUFEHM Risk [%]	No MIPS	40 % ± 10 %	17 % ± 6 %	51 % ± 19 %	*32 % ± 5 %	24 % ± 4 %	*61 % ± 8 %
	MIPS	39 % ± 14 %	18 % ± 7 %	59 % ± 19 %	22 % ± 5 %	21 % ± 5 %	45 % ± 13 %

Regarding the comparative helmet evaluation (rating), Figure 6 presents the individual helmet scores and the distribution of ratings for each impact configuration. For linear impacts, the results show that the MIPS system does not significantly improve helmet performance. No statistically significant differences were observed between helmets with and without MIPS, and in some configurations, particularly LATERAL, a slight decrease in the rating was observed for MIPS-equipped helmets. In contrast, for oblique impacts (XROT and ZROT), helmets equipped with MIPS demonstrated a significant improvement in rating, corresponding to an increase of approximately 0.5 to 1 star.

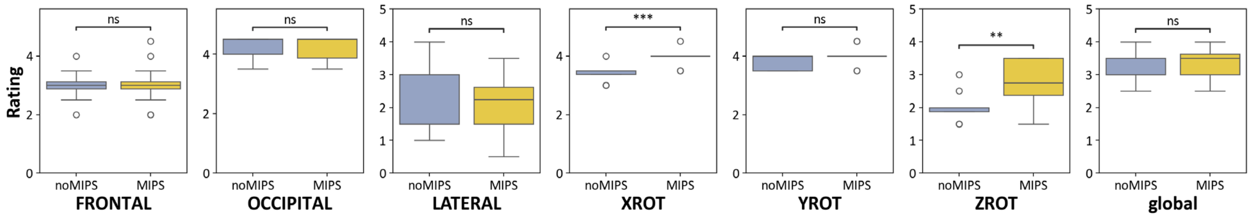

Fig. 6. Average scores of the comparative helmet evaluations for the 12 tested models without MIPS and with MIPS. *Increasing numbers of asterisks indicate increasing levels of statistical significance (p decreasing), while ns indicates a non-significant difference (p > 0.05).*

Comparison Between Pedelec and Standard Bicycle Helmets

This section compares the 15 Pedelec helmets certified under the NTA 8776 standard with the 31 helmet models certified exclusively under the EN 1078 standard. As shown in Fig. 7 and Table V, helmets compliant with the NTA 8776 standard tended to exhibit slightly lower head kinematics compared with helmets certified only under EN 1078. On average, PLA decreased from 139 ± 28 g for EN 1078 helmets to 134 ± 28 g for NTA 8776 helmets (*p-value* < 0.05). Similarly, PRA decreased from 5.7 ± 2.3 krad/s² to 5.2 ± 2.1 krad/s² (*p-value* < 0.05), while PRV decreased from 21 ± 9 rad/s to 20 ± 8 rad/s (*p-value* < 0.05).

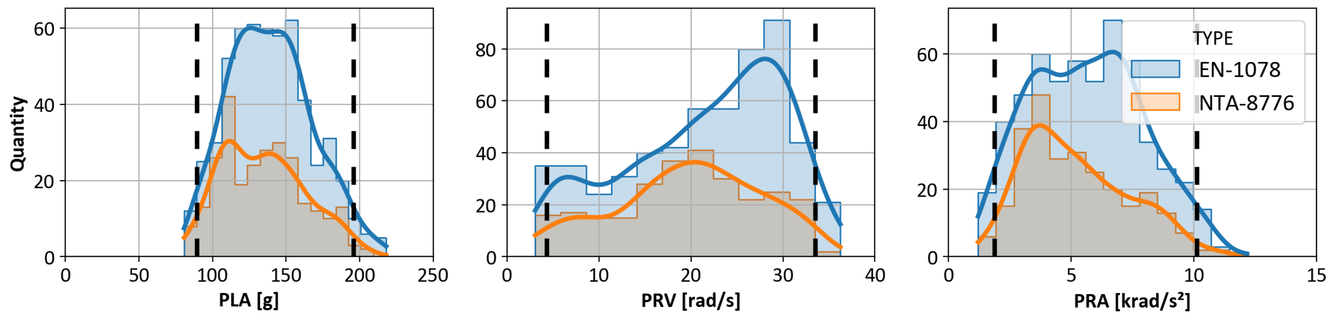

Fig. 7. Distribution of PLA, PRV, and PRA for helmets certified under EN 1078 only and helmets certified under NTA 8776 (Pedelec).

Consistent trends were also observed for the injury criteria, with mean HIC values decreasing from 707 ± 239 to 690 ± 243 and BrIC values decreasing from 0.39 ± 0.17 to 0.36 ± 0.16. However, although the other parameters showed significant differences, HIC values did not differ significantly between groups (*p-value > 0.05*), as indicated by the "ns" label in the boxplot in Fig. 7. These results suggest that helmets certified under the NTA 8776 standard provide slightly improved impact mitigation compared with helmets certified only under EN 1078. However, the magnitude of these differences remains relatively moderate, indicating that the higher test velocities specified in the NTA 8776 certification do not necessarily translate into substantially improved protective performance under the impact conditions considered in this study.

TABLE V
MEAN VALUES OF PEAK RESULTANT LINEAR ACCELERATION, PEAK ROTATIONAL VELOCITY, AND PEAK ROTATIONAL ACCELERATION FOR HELMETS CERTIFIED UNDER EN 1078 ONLY AND HELMETS CERTIFIED UNDER NTA 8776 (PEDELEC)

TYPE	Impact Counts	PLA [g]	PRA [krad/s²]	PRV [rad/s]	HIC	BrIC
EN 1078 Helmets	558	139 ± 28	5.7 ± 2.3	21 ± 9	707 ± 239	0.39 ± 0.17
NTA 8776 Helmets	270	134 ± 28	5.2 ± 2.1	20 ± 8	690 ± 243	0.36 ± 0.16
All	828	138 ± 28	5.5 ± 2.2	21 ± 8	701 ± 240	0.38 ± 0.16

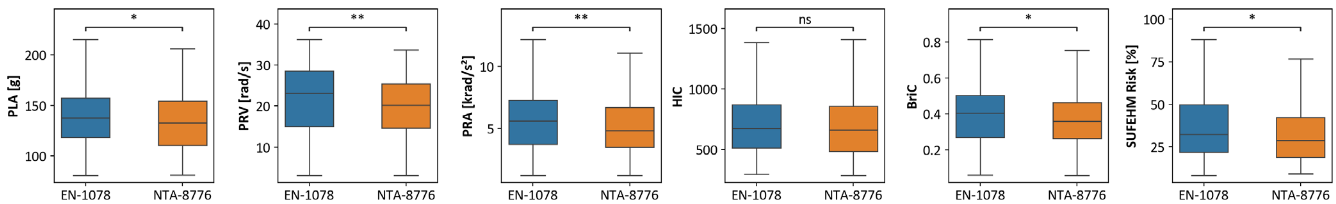

Fig. 8. Comparison of the distributions of global kinematic parameters and mean neurological injury risk for all helmet models (EN 1078; NTA 8776). *Increasing numbers of asterisks indicate increasing levels of statistical significance (p decreasing), while ns indicates a non-significant difference (p > 0.05).*

The results presented in Fig. 9 compare the PLA, PRV, and PRA measured during helmet impact tests for helmets certified under EN 1078 and those additionally certified under NTA 8776, according to the different impact configurations, as well as HIC, BrIC and SUFEHM risk computed using the data measured.

For impacts on horizontal anvil (FRONTAL, OCCIPITAL, and LATERAL), no statistically significant differences were observed between the two helmet groups for most metrics (*ns*). The distributions of PLA were very similar between EN 1078 and NTA 8776 helmets, indicating that the additional NTA 8776 certification does not significantly modify helmet performance under linear direct impact conditions.

For oblique impacts, the results showed a more pronounced effect. In the XROT and ZROT configurations, a significant slight reduction in PLA was observed for NTA 8776 helmets (from 129 ± 18 g to 120 ± 17 g in XROT configuration and from 120 ± 14 g to 108 ± 18 g in ZROT configuration). However, rotational velocity and rotational acceleration differences remained statistically non-significant. In the YROT configuration, peak rotational velocity and rotational acceleration were significantly lower for NTA 8776 helmets (reduction of 13% on the PRV and of 18% on the PRA). Although these differences were not always statistically significant, consistent reductions in mean kinematic values were observed across all oblique impact tests; however, the interpretation of this trend is limited by the large variability and broad distribution of the results.

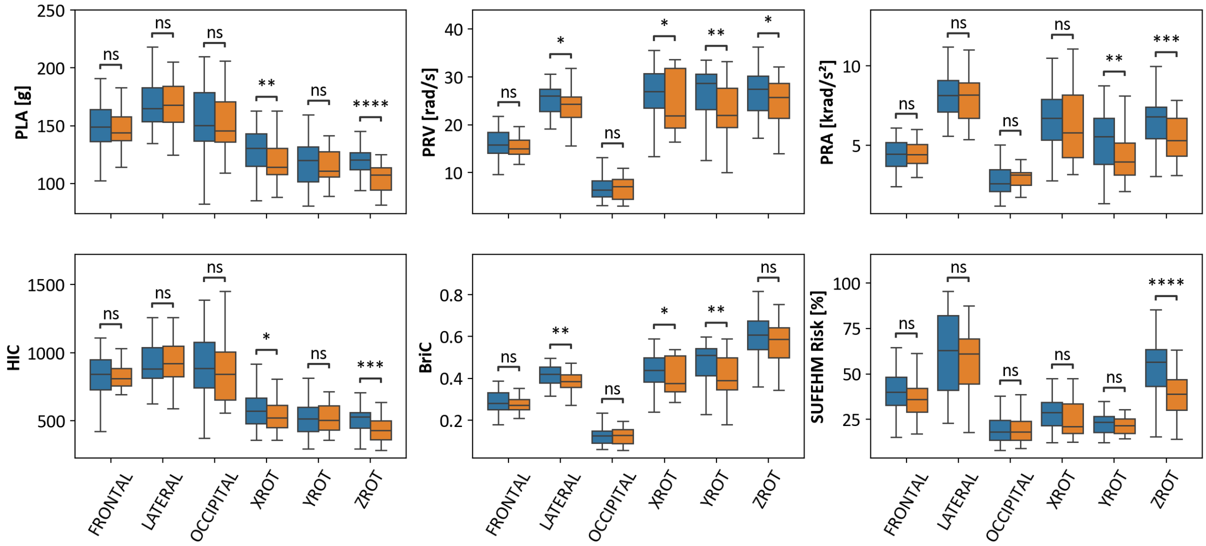

Fig. 9. Comparison of the distributions of global kinematic parameters and mean neurological injury risk for each impact configuration across all helmet models (EN 1078; NTA 8776). *Increasing numbers of asterisks indicate increasing levels of statistical significance (p decreasing), while ns indicates a non-significant difference (p > 0.05).*

We then compared the injury criteria (HIC, BrIC, and SUFEHM risk) obtained for helmets certified under EN 1078 and helmets additionally certified under NTA 8776 across the different impact configurations. Overall, the results showed only moderate differences between the two helmet groups. No statistically significant differences in HIC values were observed for the linear impact configurations (FRONTAL, OCCIPITAL, and LATERAL) or for the YROT configuration. Although statistically significant reductions were identified in the XROT and ZROT configurations, the magnitude of these differences remained limited. When all impact configurations were considered together, the global HIC values did not differ significantly between the two helmet groups. Concerning BrIC, slight differences were observed in the LATERAL, XROT, and YROT configurations. While the mean BrIC values showed statistically significant reductions for the NTA 8776 helmets, the numerical differences remained small, decreasing for example from 0.43 to 0.40 in the YROT configuration. With respect to the SUFEHM-based brain injury risk, which accounts for tissue-level brain response, only limited differences were observed for most configurations. No statistically significant differences were found for the linear impact conditions or for the XROT and YROT configurations. However, a significant reduction in injury risk was observed for the ZROT configuration (*p-value < 0.0001*), with the predicted neurological injury risk decreasing from 56% for EN 1078 helmets to 38% for NTA 8776 helmets.

TABLE VI

MEAN VALUES OF THE DIFFERENT PARAMETERS. GRAY CELLS CORRESPOND TO A SIGNIFICANT DIFFERENCE BETWEEN MEANS WITH 95% CONFIDENCE (*P-VALUE*<0.05), AS DETERMINED BY THE MANN-WHITNEY U TEST. EFFECT SIZE IS REPORTED USING THE RANK-BISERIAL CORRELATION; A STAR (*) DENOTES A LARGE EFFECT SIZE, WHILE THE ABSENCE OF A STAR INDICATES A SMALL TO MODERATE EFFECT.

		FRONTAL	OCCIPITAL	LATERAL	XROT	YROT	ZROT
PLA [g]	EN 1078	148.8 ± 21.5	152.4 ± 30.4	168.5 ± 20.7	129.4 ± 18.2	117.7 ± 20.0	*119.7 ± 14.4
	NT 8776	147.1 ± 16.2	151.3 ± 23.9	166.4 ± 19.6	119.5 ± 16.9	114.9 ± 14.3	107.5 ± 17.8
PRV [rad/s]	EN 1078	16.1 ± 2.8	7.0 ± 2.5	25.4 ± 2.9	26.8 ± 5.0	26.2 ± 6.2	27.0 ± 4.9
	NT 8776	15.5 ± 2.0	6.8 ± 2.3	23.9 ± 3.7	24.4 ± 6.1	22.8 ± 5.9	24.7 ± 4.7
PRA [rad/s²]	EN 1078	4440 ± 917	2803 ± 905	8164 ± 1409	6720 ± 2035	5266 ± 1790	6547 ± 1607
	NT 8776	4469 ± 824	2888 ± 698	7898 ± 1325	6206 ± 2233	4317 ± 1581	5479 ± 1331
HIC	EN 1078	811 ± 167	897 ± 260	919 ± 160	584 ± 132	518 ± 124	513 ± 99
	NT 8776	819 ± 124	878 ± 234	934 ± 160	541 ± 124	520 ± 106	446 ± 104
BrIC	EN 1078	0.29 ± 0.05	0.13 ± 0.04	0.41 ± 0.05	0.44 ± 0.07	0.47 ± 0.11	0.60 ± 0.11
	NT 8776	0.28 ± 0.03	0.13 ± 0.04	0.39 ± 0.07	0.41 ± 0.09	0.41 ± 0.10	0.56 ± 0.11
SUFEHM VMS [kPa]	EN 1078	30.8 ± 6.7	19.5 ± 5.5	43.5 ± 13.1	25.6 ± 5.2	22.2 ± 3.8	38.0 ± 7.8
	NT 8776	30.0 ± 5.4	19.2 ± 5.4	40.3 ± 8.7	23.6 ± 6.0	21.3 ± 3.2	30.9 ± 8.1
SUFEHM Risk [%]	EN 1078	39 % ± 13 %	20 % ± 7 %	61 % ± 22 %	29 % ± 9 %	23 % ± 6 %	53 % ± 15 %
	NT 8776	37 % ± 10 %	19 % ± 7 %	57 % ± 17 %	26 % ± 10 %	21 % ± 5 %	39 % ± 15 %

Table VI summarizes the mean values obtained from the 828 helmet impact tests for both helmet standards. Gray cells indicate statistically significant differences between groups. Effect sizes, assessed using the rank-biserial correlation, were only large for ZROT and PLA, as indicated by the asterisks (*). Although several parameters differed significantly, the large standard deviations, often greater than the differences between group means, highlight the substantial variability observed across all impact configurations.

Figure 10 compares the average ratings obtained by the 46 helmet models for each impact configuration and for the global evaluation. Overall, helmets certified under NTA 8776 achieved ratings similar to those certified only under EN 1078, with no statistically significant differences observed for frontal, occipital, lateral, XROT, YROT, or global scores. A statistically significant improvement was identified only for the ZROT configuration, where NTA 8776 helmets showed slightly higher ratings. Despite this isolated difference, the broad overlap of the score distributions and the large variability between helmet models indicate that the stricter NTA 8776 certification does not systematically translate into higher overall protective performance. Instead, Pedelec helmets generally ranked within the same performance range as conventional bicycle helmets.

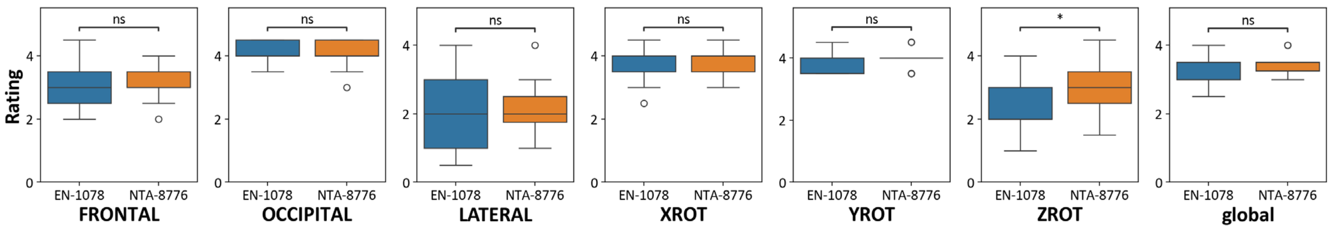

Fig. 10. Average scores of the comparative helmet evaluations for all 46 models (EN 1078; NTA 8776). *Increasing numbers of asterisks indicate increasing levels of statistical significance (p decreasing), while ns indicates a non-significant difference (p > 0.05).*

Overall synthesis

Global brain injury risks computed with the brain von Mises stress extracted from SUFEHM in accordance with the presented methodology, showed slight improvement, as reported in Table VII. Indeed, regarding the effect of rotational mitigation system in the helmets, a substantial reduction on brain injury risk can be observed in XROT and ZROT configurations. Nevertheless, even if these effects are statistically significant, the coefficients of variation (COV) are high and therefore moderate the improvement. In the same way, statistically, the effect can only be observed on ZROT configuration, although this configuration is not considered in the NTA 8776 standard.

TABLE VII

GAIN IN BRAIN INJURY RISK OBTAINED WITH THE MIPS ROTATIONAL SYSTEM AND THE NTA 8776 (PEDELEC) CERTIFICATION, TOGETHER WITH THE COEFFICIENT OF VARIATION (COV)

		FRONTAL	OCCIPITAL	LATERAL	XROT	YROT	ZROT	GLOBAL
MIPS Effect	GAIN	2%	-2%	-14%	31%****	14%**	25%****	10%**
	COV	31%	35%	38%	27%	21%	25%	52%
Pedelec Effect	GAIN	5%	2%	6%	10%	6%	26%****	11%*
	COV	31%	34%	38%	34%	24%	34%	54%

Note: **Statistical differences between groups were assessed using the Mann–Whitney U test. Increasing numbers of asterisks indicate increasing levels of statistical significance (p decreasing), while ns indicates a non-significant difference (*p-value* > *0.05*).**

IV. DISCUSSION

This study focused on the comparative evaluation of modern bicycle helmets. By systematically assessing and comparing the performance of different helmet categories, including models incorporating the rotational mitigation technologies MIPS and those certified for pedelec use, the analysis provides insight into the effectiveness of current helmet safety solutions. Such comparative approaches allow researchers and industry stakeholders to identify potential areas for improvement and innovation, contributing to the ongoing development of helmets designed to better reduce the risk of head injuries and enhance rider safety. It should be noted that this study focused exclusively on shock-absorption performance under ambient temperature conditions and did not address penetration resistance or environmental conditioning effects. It should also be noted that the objective of the present study was not to perform a direct within-model comparison between helmets with and without MIPS, but rather to evaluate commercially available helmet populations representative of current consumer products. Consequently, observed differences between MIPS and non-MIPS helmets may also reflect variations in overall helmet design, geometry, liner properties, or shell construction, and cannot be attributed exclusively to the presence of the MIPS system alone. In particular, some differences observed under linear impact configurations may be related to intrinsic model design characteristics rather than to the rotational management system itself.

A first limitation of this study concerns the number of helmet models included in the analysis. Although the sample size was relatively limited, it was sufficient to provide an exploratory comparison between the investigated helmet categories [44]. Future studies including a larger number of helmet models would allow a more comprehensive assessment of market variability. Another limitation is related to the experimental set-up. The impact tests were conducted under free-fall conditions using a headform without a neck, thereby neglecting potential neck contributions to head kinematics during impact. Bland *et al.* [45] showed that neck boundary conditions influence measured head kinematics during oblique impacts. However, it should be noted that the Hybrid III neck is relatively stiff and does not fully reproduce biofidelic neck behaviour under rotational loading conditions, as previously discussed by Rémy Willinger *et al.* [46]. Nevertheless, recent FE investigations suggest that the influence of the neck on brain response is relatively small. Feist and Klug [47], using the THUMS human body model, and Fahlstedt *et al*. [48], using the KTH head-neck model, reported that the neck effect typically modifies brain response by approximately 10-15% depending on the impact conditions.

The Hybrid III headform was used in this study. Its friction coefficient at the head–helmet interface is higher than that of the magnesium EN 960 headform (0.75 ± 0.06 vs 0.16 ± 0.026). Trotta *et al*. [49] reported a scalp–liner friction coefficient of 0.29 ± 0.07, suggesting that current headforms may not fully reproduce realistic head–helmet interaction. Several recent studies have highlighted the strong influence of headform friction properties on measured kinematics during oblique helmet impacts. Stark *et al.* [50] demonstrated that both headform friction and inertial properties significantly affect rotational kinematics and helmet ranking outcomes during oblique testing. Similarly, Bonin *et al*. [51]and Juste-Lorente *et al.* [52] reported that headform–helmet friction conditions can substantially influence measured rotational accelerations and the apparent effectiveness of rotational management systems such as MIPS. A new headform (EN 17950 [53]) is currently under development within CEN/TC158-WG11 for future multi-sport helmet standards. Moreover, only a Hybrid III 50th percentile male headform was used for all impact tests. Consequently, the present results are primarily representative of male head anthropometry and inertial properties. Since female head geometry and mass distribution may differ and influence rotational kinematics, caution should be exercised when generalizing these findings to other populations.

Concerning the metric used, a number of studies that evaluated helmet performance used to consider global parameters as PLA, PRV, PRA as well as HIC and BrIC [40-47]. A key limitation of these metrics is that they represent the complex temporal evolution of brain loading using a single scalar quantity. Finite element head models have been used for more than two decades to assess brain injury risk at the tissue level. Unlike global kinematic metrics, such as HIC or BrIC, these models account for both linear and rotational head loading and allow the computation of intracranial stress and strain. Several injury metrics have been proposed using FE head models, including Maximum Principal Strain with the GHBMC [62] and KTH models [9]. In this study, the SUFEHM model was used. Based on simulations of 125 documented head trauma cases originating from bicycle, motorcycle, and pedestrian accidents and other accident scenarios, Deck *et al*. [40] established a tolerance threshold for moderate brain injury (AIS2+), corresponding to a 50% injury risk at a brain von Mises stress of 36.5 kPa. Nevertheless, it should be noted that the present study relies on a single finite element head model (SUFEHM), and the use of additional validated brain models would strengthen the robustness and generalizability of the findings.

By comparing the paired-MIPS helmet groups, the results appear consistent with the intended function of rotational management systems, which are designed to reduce rotational head kinematics generated during oblique impacts. Such rotational motions are strongly associated with brain shear strains and diffuse axonal injury (DAI) mechanisms. By allowing limited relative motion between the helmet shell and the head, the MIPS layer contributes to reducing rotational loading and the associated brain injury risk.

The observed reductions mainly concern rotational kinematic parameters and SUFEHM-based brain injury metrics during oblique impacts, whereas no meaningful improvement was observed under predominantly linear loading conditions. These findings therefore suggest that the MIPS system primarily mitigates rotational loading rather than enhancing protection against linear impacts.

These observations are in agreement with the findings reported by Bonin *et al.* [51], who also reported reductions in rotational head kinematics for MIPS-equipped helmets during oblique bicycle helmet impacts. Their study further highlighted that the effectiveness of MIPS depends on impact orientation, impact speed, and headform surface conditions. Similar to the present results, the reductions were mainly observed for rotational kinematic parameters, while limited effects were identified for linear acceleration metrics.

Although the trends observed for global kinematic metrics (PLA, PRA, PRV, and BrIC) were generally consistent with those obtained using the SUFEHM-based injury criteria, the tissue-level analysis provided additional biomechanical interpretation beyond peak headform kinematics alone. Unlike conventional certification metrics based primarily on resultant peak values, the finite element approach accounts for the temporal evolution, directionality, and combined translational-rotational loading applied to the brain. Consequently, two impacts producing similar peak rotational accelerations may generate different intracerebral stress distributions and associated injury risks. This was particularly relevant for oblique impacts, where reductions in rotational kinematics did not always translate proportionally into reductions in predicted brain tissue stress. The SUFEHM-based analysis therefore enabled a more physiologically relevant interpretation of helmet protective performance and highlighted the limitations of certification approaches relying primarily on peak linear acceleration thresholds without considering tissue-level brain response.

Despite being designed for higher-speed cycling conditions, Pedelec helmets certified under NTA 8776 did not demonstrate substantially improved protection against head injuries compared with conventional bicycle helmets certified only under EN 1078. Although some differences reached statistical significance on kinematics values, these differences remained small and were associated with substantial variability. In addition, in Fig. 10, which presents the distributions of the comparative helmet ratings for all tested models, no clear overall advantage was observed for helmets certified under the NTA 8776 standard. Only the ZROT configuration showed a statistically significant improvement for NTA 8776 helmets, whereas all other impact configurations and the global rating exhibited non-significant differences.

Overall, the tested Pedelec helmets tended to rank within the mid-range of conventional helmet models, with few statistically significant differences when considering all impact configurations together. This observation was particularly noticeable for linear impact configurations, despite the fact that the NTA 8776 certification specifies higher test velocities than EN 1078. Although consistent reductions in some kinematic metrics were observed for NTA 8776 helmets, the interpretation of these trends remains limited by the broad distribution and variability of the results.

These findings suggest that the stricter requirements of the NTA 8776 standard do not necessarily translate into improved protective performance under the impact conditions considered in this study. One possible explanation is that the NTA 8776 certification primarily increases test severity in terms of linear impact energy, whereas the mechanisms associated with traumatic brain injuries, particularly diffuse axonal injuries, are more strongly linked to rotational head kinematics. Consequently, helmet designs optimized to satisfy higher linear acceleration requirements may not necessarily provide improved mitigation of rotational loading, which remains a critical factor in brain injury risk.

The present results also highlight the need for further evolution of bicycle helmet certification procedures. While EN 1078 and NTA 8776 primarily rely on peak linear acceleration thresholds, several recent helmet assessment approaches increasingly incorporate oblique impacts and rotational injury metrics. For example, modern motorcycle helmet regulations such as ECE 22.06 include oblique impact testing procedures intended to evaluate rotational head kinematics. Similarly, independent consumer-oriented rating systems, such as the Virginia Tech STAR methodology [63], combine multiple impact configurations and rotational injury predictors to provide a more comprehensive assessment of helmet protective performance. Compared with these approaches, the current bicycle helmet standards remain relatively limited in their ability to evaluate rotational brain loading mechanisms.

Future improvements to bicycle helmet standards could therefore include systematic oblique impact configurations, the use of more biofidelic instrumented headforms, and the integration of rotational or tissue-level injury criteria derived from finite element head models. Such developments would provide a more representative assessment of real-world accident conditions and could facilitate the optimisation of future helmet designs specifically intended to mitigate rotational brain loading. In addition, combining experimental kinematics with tissue-level brain response metrics may improve the interpretation of helmet protective performance beyond simple peak acceleration thresholds.

V. CONCLUSION

This study presented a comparative evaluation of commercially available bicycle helmets using an experimental-numerical methodology combining realistic impact tests and FE head injury modelling. The results highlight the importance of considering both linear and rotational head kinematics, as well as tissue-level brain injury criteria, when assessing helmet protective performance.

The analysis shows that on 12 helmet models incorporating rotational mitigation technologies, such as MIPS, it tended to reduce rotational head kinematics and associated brain injury risk during oblique impacts, which are commonly observed in real-world accident scenarios. However, their influence remains limited in purely linear impact configurations, where no significant improvement was observed, and in some cases slightly higher kinematic values were recorded.

The comparison between conventional EN 1078 helmets and Pedelec helmets certified under the NTA 8776 standard indicates that the stricter certification requirements associated with higher impact velocities do not necessarily lead to improved protection against brain injury. The tested Pedelec helmets generally performed within the mid-range of conventional helmet models and did not demonstrate a clear advantage in terms of predicted injury risk.

Overall, these findings underline the limitations of current helmet certification approaches that primarily rely on peak linear acceleration criteria. The results support the need for helmet evaluation methodologies incorporating oblique impacts and brain injury metrics derived from biomechanical models, which provide a more realistic representation of head loading conditions during accidents. Such approaches may contribute to improved helmet design, better consumer information, and the development of more representative helmet safety standards.

VI. ACKNOWLEDGEMENTS

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