Experimental evaluation of anti-rotational helmet technologies using a biofidelic human head replica embedding sensorised CSF-meninges-brain simulants

Elisa Baldoin, Giuseppe Zullo, Piergiorgio Carraretto, Andrey Koptyug, Nicola Petrone

Abstract The scientific community is deeply concerned about the social impacts stemming from the consequences of Traumatic Brain Injuries (TBIs). Therefore, Anti-Rotational Technologies (ART) were designed to mitigate TBI severity. Advanced helmet testing, involving standard rigid headforms and numerical models of the human head, faces challenges regarding biofidelity and validation against rare cadaver data. The present study uses an innovative Instrumented Human Head Replica (IHHR), including cerebrospinal fluid (CSF), meninges, and brain simulants, to tackle biofidelity concerns. The IHHR assesses severity of impacts using embedded brain and skull pressure sensors, accelerometers, and gyros. Protected drop tests were conducted from three heights, incorporating ART and balaclava, onto an inclined anvil with a motorcycle helmet. A significant height-dependent reduction in Brain Injury Criterion (BrIC) with ART was shown (p-value≤0.001), while balaclava effects were not significant. The observed relative skull-brain motion was affected by ART (p-value≤0.001) and drop height (p-value=0.003). CSF pressures were significantly affected by ART and balaclava (p-values≤0.01), showing an increase in the coup duration and a decrease in pressure peaks with ART. These findings highlight the potential of the IHHR as a valuable tool for estimating the effect of ART on the severity of TBIs, allowing the calculation of injury criteria.

Keywords Helmet, Human Head Replica, protection effectiveness, rotational impacts, CSF pressure.

I. INTRODUCTION

Traumatic brain injuries (TBIs) are a primary cause of death and disability worldwide, emphasising the importance of testing protective equipment to mitigate the risk of brain damage caused by sudden trauma [1]. Brain damage can be caused both by linear and rotational impacts [2], and for this reason an increasing number of helmets are equipped with Anti-Rotational Technologies (ART), like MIPS®, Koroid™, and Wavecell™, aimed at preventing and reducing the head rotational acceleration.

Balaclava and ART act as additional layers to reduce friction between the skin and the helmet and their influence on rotational kinematics was evaluated. Friction between a headform and a helmet is believed to influence rotational impact response [3]. In this regard, Stark *et al.* [4] evaluated static friction coefficient for headforms equipped with an outer layer mimicking skin (NOCSAE and Hybrid III headform), which is also present in the IHHR. The aim was to investigate the effect of adding a skull cap to decrease friction between skin and a helmet to produce similar friction coefficient with respect to those measured for humans. In particular, the use of a skull cap resulted in a considerable decrease in the static coefficient of friction for both the NOCSAE and Hybrid III headforms against a KASK Protone Icon helmet liner material. Moreover, Bonin *et al.* [5] investigated the effect of friction on head kinematics during oblique protected impact tests. It was observed that incorporating hair or stockings onto a bare headform decreases angular kinematics, while the inclusion of MIPS onto a headform equipped with hair or stockings additionally lowers peak angular kinematics.

The current testing methodology used for the assessment of helmet effectiveness and the numerical analysis by means of the Finite Element Method (FEM) are affected by advantages and disadvantages. Experimental testing is used in industry for the purpose of product certification because of its accuracy, repeatability, and the time-effectiveness of the testing protocols. On the other hand, there is a lack of biofidelity in the standard rigid

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headforms commonly used, like the EN960 or the Hybrid III anthropomorphic testing device (ATD) [6-7], which don't include a brain simulant. It should be noted, however, that some authors tried to develop novel headforms with more biofidelic characteristics [8-10]. There is a general confidence in the numerical models of the human head, such as the FE models developed by R. Willinger [11], D. Gilchrist [12], S. Kleiven [13], M. Ghajari [14] and their groups, or the more recent Head FE model developed by P. Pavan *et al*. [15], due to their high degree of detail. Their biofidelity means they are a good tool for the development of solutions and products, provided they are validated. Indeed, the validation of these numerical models is critical given that cadaveric data can be rare and outdated due to ethical restrictions [16-18]. Moreover, FEM models require complex mechanical characterisation of the materials and, in the case of protected impacts, the knowledge of the geometry and material properties (constitutive law) of the protective device.

The most advanced testing methodology involves a hybrid approach, carrying out several types of impact test on helmets fitted on a rigid headform. Then, kinematic data obtained by inertial sensors in the rigid headform are passed to a numerical model of the human head for the assessment of tissue-level brain damage. This approach could be used either to develop criteria or assessments that are valid in the numerical domain or to provide relationships between numerical brain damage and global kinematics, as in the case of BrIC [19]. The availability of biofidelic headforms equipped with accelerometers, gyroscopes, pressure, stress, and strain sensors inside the brain and skull surrogates could provide a cross-validation tool for FE models and improve the accuracy of experimental predictions, having more sensor outputs to be correlated with brain damage. The achievement of a satisfactory biofidelity is the main difficulty in the realisation of these types of replicas, due to the high complexity of the anatomical structures and to the peculiar physical and mechanical properties shown by biological tissues. Indeed, brain tissue is characterised both by time-independent hyperelasticity and time-dependent viscoelasticity. Therefore, quasi-static and cyclic tensile, compressive and shear tests are needed to characterise the mechanical behaviour of the brain simulant materials [20], and the choice of a synthetic material capable of accurately replicating all of these properties is complex.

To overcome the drawbacks related to the lack of biofidelity of standard rigid headforms and the challenging validation of numerical models of the human head, the biofidelic Instrumented Human Head Replica (IHHR) was developed at the University of Padua (Department of Industrial Engineering) in collaboration with Mid Sweden University [21]. This replica was adopted to perform rotational impacts, with the aid of a motorcycle helmet, to study the effect of ART, balaclava, and drop height on the evaluation of TBI severity.

II. METHODS

This experimental work belongs to an ongoing project with the objective of creating a biofidelic replica integrated with accelerometers, gyroscopes, and pressure sensors to collect data from impact tests and assess brain damage. The evolution of the project resulted in five sensorised prototypes [21], progressively improving both the biofidelity of the structures and materials and the quality of the sensors. The last version of these prototypes (IHHR) was used for this research.

The Instrumented Human Head Replica: materials and sensors

The IHHR (Fig. 1) contains a silicone rubber (PlatSil Gel-OO and Smith's Deadender, Polytek, USA) brain surrogate enveloped in open-cell polymer foam, mimicking the cushioning effect of arachnoid trabeculae in compression. Demineralised water simulates cerebrospinal fluid, while 3D-printed thermoplastic polyurethane falx and tentorium structures are glued to a polyamide skull simulant. Silicone rubber (PlatSil Gel-10, Polytek, USA) is used for skin casting.

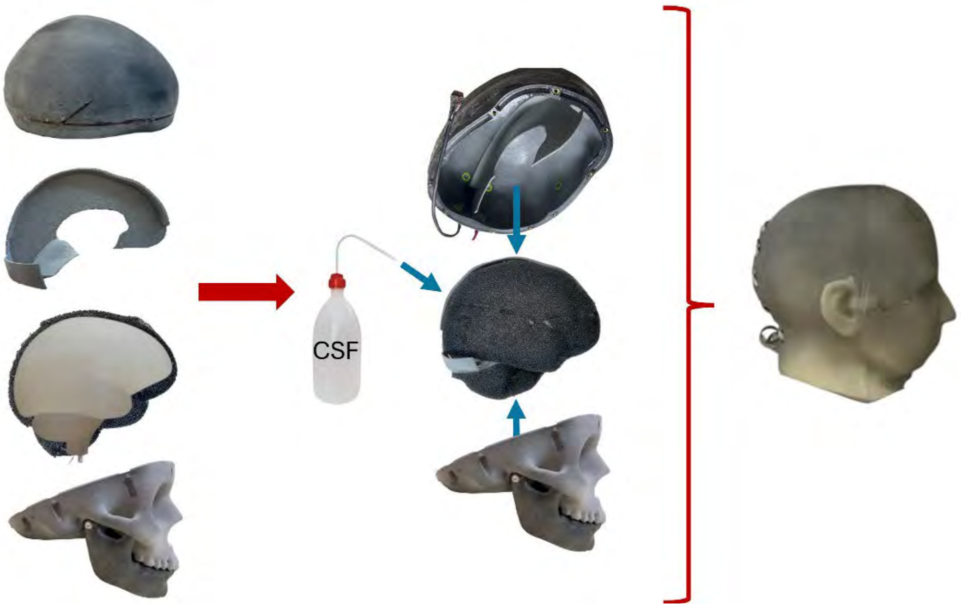

Fig. 1. Top skull (upper part of the skull), falx-tentorium, arachnoid trabeculae, brain, bottom skull (lower part of the skull) and jaw surrogates of the Instrumented Human Head Replica (IHHR) developed by the Department of Industrial Engineering (DII) of the University of Padua to be assembled, including CSF simulant, inside the skin prototype.

The replica includes a DTS 6DX PRO-A triaxial accelerometer (±500 g) and gyrometer (±8000 °/s) under the skull palate and triaxial accelerometer (Analog Devices ADXL 377, ±200 g) and gyrometer (ST Microelectronics LPR+LPY, ±2000 °/s) in the centre of mass (CoM) of the brain surrogate. These sensors allow for the assessment of relative skull-brain kinematics as an additional parameter to investigate brain damage with respect to standard rigid headforms, including sensors attached to the skull. Moreover, the prototype features 12 piezoresistive pressure sensors (TE Connectivity MS5407, 700 kPa), which are glued to the inner surface of the top skull to detect the pressure of CSF. Finally, a Multi-Axial Pressure Sensor (MAPS) detects the stress state 35 mm in front and 3 mm above the brain CoM [22]. Figure 2 shows the positions of the aforementioned sensors. CSF pressure sensors were named as Frontal, Sphenoid, Temporal, Occipital, Parietal, and Top, as shown in Fig. 3, which shows the left sagittal section of the skull surrogate with its embedded sensors. A transparency view of the brain with its sensors is also shown.

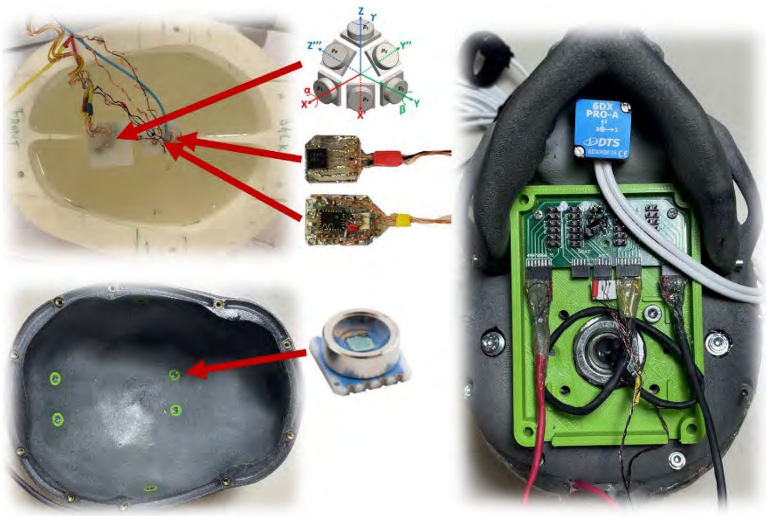

Fig. 2. Arrangement of sensors inside the Instrumented Human Head Replica (IHHR): the brain surrogate includes six pressure sensors and triaxial accelerometer and gyrometer, while the skull surrogate is equipped with 12 pressure sensors and a DTS 6DX PRO-A triaxial accelerometer and gyrometer.

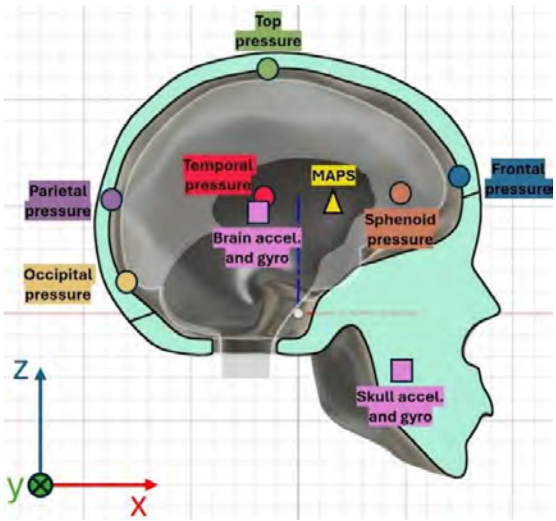

Fig. 3. Arrangement of sensors on the skull and brain surrogates: CSF pressure sensors (Frontal, Sphenoid, Temporal, Occipital, Parietal, and Top), skull DTS 6DX PRO-A triaxial accelerometer and gyrometer, brain triaxial accelerometer (ADXL 377), gyrometer (LPR+LPY), and MAPS.

This Head Replica was compared to the standard rigid Hybrid III headform by means of drop tests performed without the balaclava using the same protective helmet (Alpinestars, SM8), ART (MIPS®) and drop height (0.75 m) as in a previous work [21]. It was observed that the use of a biofidelic replica (IHHR), allowing a relative motion of the brain with respect to the skull, and the damping effect of CSF can influence the evaluation of the Brain Injury Criterion (BrIC) by a factor of 2 [21]. Therefore, experimental trials were carried out with the IHHR to evaluate the effect of several factors (drop height, ART, balaclava), using the protocol described below.

Experimental Testing: oblique impact drop tests

A motorcycle helmet (Alpinestars, SM8) was fitted on the replica to conduct protected drop tests using a drop tower constructed similar to ECE 22.06 standard. The adopted experimental rig is shown in Fig. 4. The drop tower utilises a 2.40 m linear guide paired with a ball shuttle (Harken Safety), which was connected to a cylindrical basket that supports the head while falling, thus avoiding changes in its initial alignment. A quick-release mechanism controls the beginning of the fall. The impact could be either onto a flat or onto a 45° inclined cylindrical anvil. The inclined anvil is covered with 120-grit abrasive paper. Finally, the anvils are mounted on a 44 kN load cell (LCP455, Futek, Switzerland) supported by a steel cylinder that is fixed to a seismic mass. A retaining rope prevents excessive rotation of the head after impacting, which could potentially damage the signal cables.

Protected drop tests were performed onto the 45° inclined anvil from 1 m (4.4 m/s), 1.5 m (5.4 m/s), and 2 m (6.3 m/s) heights, with and without balaclava and MIPS® ART, for a total of four configurations per height level (without balaclava and without ART, with balaclava and without ART, without balaclava and with ART, with balaclava and with ART). The highest speed of 8.0 m/s (ECE 22.06) was not explored due to the limited available tower drop height. Figure 5 shows the experimental setup for the 1 m drop test.

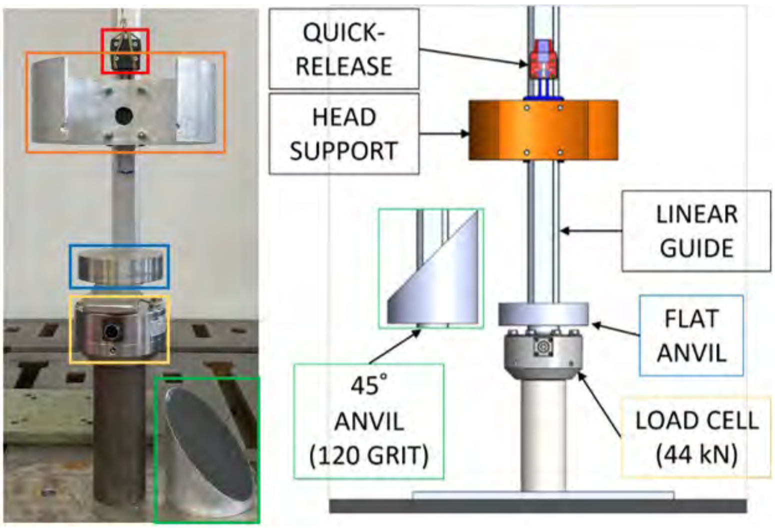

Fig. 4. Experimental rig for impact testing: the image illustrates the principal components of the rig, including the quick-release mechanism, a linear guide, the head support basket, a flat or 45° inclined anvil covered with abrasive paper, and Futek load cell. In particular, the 45° inclined anvil was adopted for the current tests.

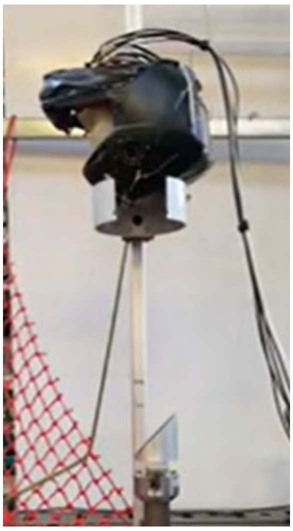

Fig. 5. Experimental rig for protected impact tests performed onto the 45° inclined anvil using the IHHR from 1 m.

A total of 32 rotational impacts were carried out, repeating the drop tests three times per configuration for the 1 m and 1.5 m tests and twice for the 2 m tests. Data were recorded at 10 kHz by a SoMat eDAQlite (HBM, Germany) and a SLICEnano (DTS, USA) synchronised with a hardware trigger.

Data analysis

Data were analysed using MATLAB R2022b (MathWorks, USA). First, data were low-pass filtered using a 4th order Butterworth filter with zero-phase and cut-off frequency of 1000 Hz. CSF pressures were also high-pass filtered using a 4th order Butterworth filter with zero-phase and cut-off frequency of 1 Hz. Then, data were cut by choosing a time window from 5 ms pre-impact to 25 ms post-impact. The load cell signal was used to determine the instant of impact.

For each trial, injury criteria related to the global kinematics were computed. This approach is supported by the existence of several injury criteria which have been developed to evaluate the risk of harm from various types of head impact, particularly in sports and road accidents. Linear and angular kinematics can be evaluated by means of Head Injury Criterion (HIC), Brain Injury Criterion (BrIC), Peak Angular Velocity (PAV), and Peak Rotational Acceleration (PRA).

Versace *et al.* [23] developed the HIC to assess head injury risks resulting from an impact, considering linear acceleration but neglecting rotational movements. In fact, HIC value is calculated with regard to the integral time of the translational acceleration, as follows:

$$HIC = max\left\{(t_2 - t_1)\left[\frac{1}{(t_2-t_1)}\int_{t_1}^{t_2} a(t)dt\right]^{2.5}\right\} \tag{1}$$

where HIC is the Head Injury Criterion (-), $a(t)$ is the resultant of the linear acceleration acquired in the head's CoM (g), and t_1 and t_2 are the limits of the time interval ($t_2 - t_1 = 15\ ms$).

However, the majority of severe head injuries occur due to rotational impacts. This rotational motion applies forces that cause the brain to twist or shear within the skull. Therefore, Takhounts *et al.* [19], along with the National Highway Traffic Safety Administration (NHTSA), developed the BrIC:

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

where $BrIC$ is the Brain Injury Criterion (-) calculated on skull data, ω_x, ω_y and ω_z are the maximum skull angular velocities (rad/s) about x, y, and z axes, and ω_{xC}, ω_{yC} and ω_{zC} are critical values (rad/s) corresponding to 66.25 rad/s, 56.45 rad/s and 42.87 rad/s, respectively.

Bonin *et al.* [5] also investigated the PAV, as follows:

$$PAV = \max(\omega(t)) \tag{3}$$

where PAV is the Peak Angular Velocity of the skull (rad/s) and ω is the angular resultant velocity acquired in the skull CoM (rad/s).

Several studies have observed a relationship between higher levels of rotational acceleration and greater severity of injuries [24-25]. Hence, PRA has been identified as significantly correlated with injury severity and it can be assessed as follows:

$$PRA = \max(\alpha(t)) \tag{4}$$

where PRA is the Peak Rotational Acceleration of the skull (rad/s^2) and α is the angular resultant acceleration acquired in the skull CoM (rad/s^2).

To this purpose, the evaluation of skull kinematics by means of palate 6DX PRO-A accelerometer and gyro allowed the assessment of HIC, PAV, PRA, and BrIC at the skull CoM. On the other hand, unlike standard rigid headforms, the presence of triaxial accelerometer and gyro inside the IHHR brain surrogate enabled the evaluation of brain kinematics. Consequently, skull-brain relative motion could be assessed in terms of relative angular velocity between the two surrogates. Given the constraining effect of the falx and tentorium on brain motion in the coronal and transverse planes, it is reasonable to assume that the brain rotates around an axis parallel to the rotation axis of the skull in response to an inclined impact. In particular, the two surrogates rotate mainly around their y-axes, which are parallel and opposite to the global y-axis shown in Fig. 3. This allows the calculation of the skull-brain relative angular velocity by directly considering the difference between the resultant angular velocities of both surrogates, rather than distinguishing their components. This simplified approach is justified by the parallel resultant angular velocities assumption, thus leading to the following assessment for the relative angular velocity:

$$\omega_{rel} = \omega_{skull} - \omega_{brain} \quad (5)$$

where ω_{rel} is the skull-brain relative angular velocity (rad/s), ω_{skull} is the resultant of the angular velocity of the skull (rad/s) and ω_{brain} is the resultant of the angular velocity recorded at the brain CoM (rad/s).

Statistical analysis measured the impact of the involved factors (height, ART, and balaclava) and their interactions on brain damage assessment, which has been quantified by means of HIC, $BrIC$, PAV, PRA, and ω_{rel}. In particular, a 3-way analysis of variance (ANOVA) was performed.

The pressure sensors embedded inside the IHHR top skull surface were needed to measure CSF pressure. In particular, data were acquired from the right set of pressure sensors, with the exception of the Temporal one, due to technical issues.

III. RESULTS

HIC, BrIC, PAV, PRA, and ω_{rel} were evaluated according to (1), (2), (3), (4), and (5) and results are reported in Figs 6–11.

Figure 6 provides HIC variations across drop height levels for the tested configurations. Statistical analysis suggested that drop height appears to affect the criterion (p-value≤0.001, η^2=0.85). Indeed, higher HIC values can be observed by increasing the fall height when performing trials both with and without ART and balaclava. On the other hand, neither ART nor balaclava seem to have a significant impact on HIC (p-values of 0.12 and 0.24, respectively).

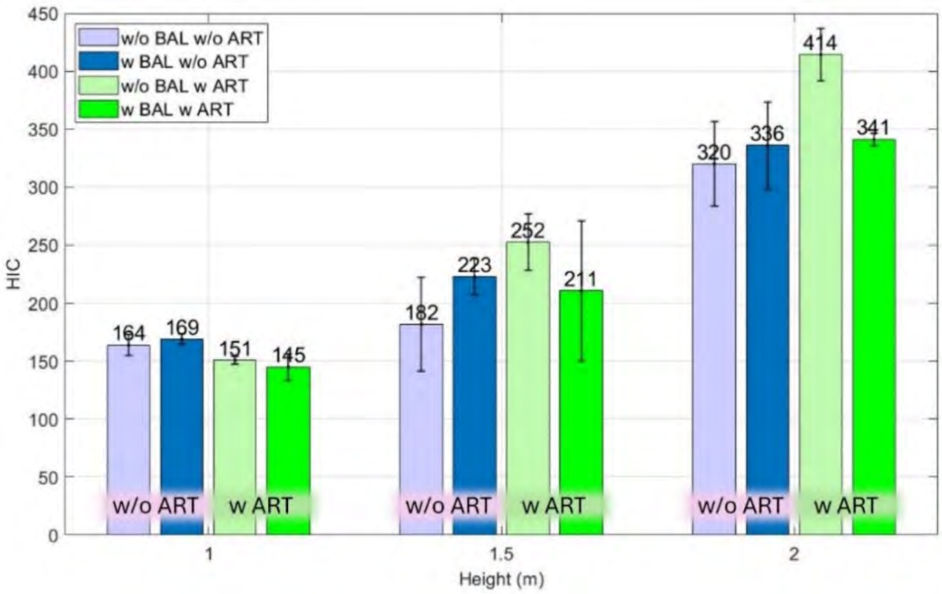

Fig. 6. Head Injury Criterion (HIC) variations across height levels (1 m, 1.5 m, and 2 m) based on balaclava (BAL) and MIPS® (ART): means and standard deviations evaluated for the four configurations (w/o BAL w/o ART, w BAL w/o ART, w/o BAL w ART, w BAL w ART) for each drop height level.

As shown in Fig. 7, the statistical analysis revealed that both ART and drop height seem to significantly affect the BrIC (p-values≤0.001, η^2=0.40), as well as their interaction (p-value≤0.001, η^2=0.13). In particular, the average percentage reduction in BrIC while using ART was about 7.6%, 21.8% and 26.2% at the three increasing drop heights, showing a more evident effect of ART as drop height raises. Conversely, the impact given by the balaclava appears negligible (p-value=0.87). As mentioned before, in a prior study regarding protected drop tests performed from 0.75 m without the balaclava, a percentage reduction in BrIC of 29.6% for the Hybrid III headform and 15.4% for the IHHR was observed while using ART [21]. Regarding the IHHR, this percentage reduction is comparable to the approximately 12.5% percentage decrease assessed in the current analysis for the 1 m drop tests performed without balaclava (light blue and green columns for the 1 m drop tests in Fig. 7). This consistency suggests a robustness of the results.

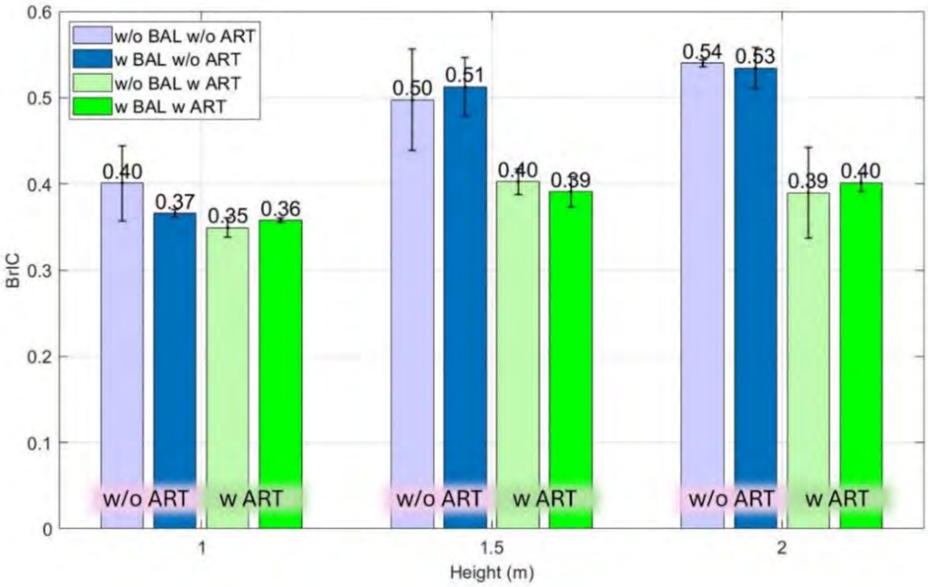

Fig. 7. Brain Injury Criterion (BrIC) variations across height levels (1 m, 1.5 m, and 2 m) based on balaclava (BAL) and MIPS® (ART): means and standard deviations evaluated for the four configurations (w/o BAL w/o ART, w BAL w/o ART, w/o BAL w ART, w BAL w ART) for each drop height level.

Figs 8 and 9 show PAV and PRA variations across drop height levels for the examined configurations. Statistical analysis evinced that PAV is significantly affected by ART (p-value≤0.0001, η^2=0.36) and drop height (p-value≤0.01, η^2=0.26). Similarly, PRA is significantly influenced by ART and drop height (p-values≤0.01, η^2=0.29). On the other hand, balaclava seems to have a negligible impact on PAV (p-value=0.38) and PRA (p-value=0.87). In contrast to what was seen for BrIC, ART and drop height interaction seem not to have a significant impact on PAV (p-value=0.65) and PRA (p-value=0.79). PAV and PRA values were comparable with experimental results from Bonin *et al.* [5].

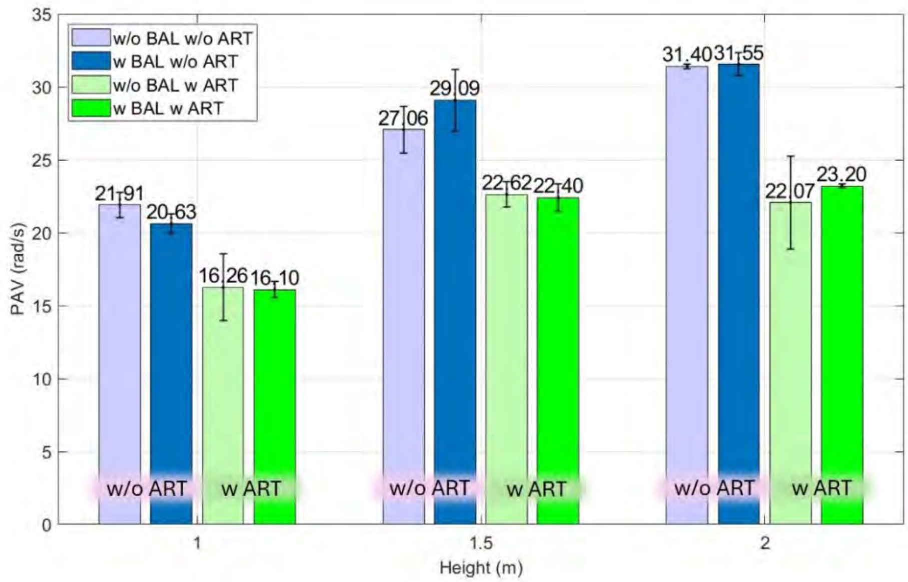

Fig. 8. Peak Angular Velocity (PAV) variations across height levels (1 m, 1.5 m, and 2 m) based on balaclava (BAL) and MIPS® (ART): means and standard deviations evaluated for the four configurations (w/o BAL w/o ART, w BAL w/o ART, w/o BAL w ART, w BAL w ART) for each drop height level.

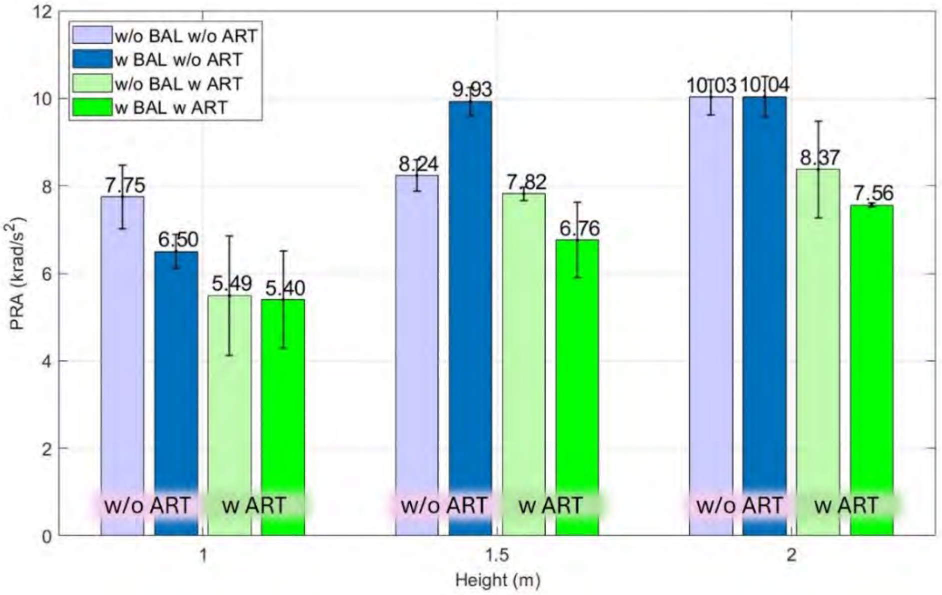

Fig. 9. Peak Rotational Acceleration (PRA) variations across height levels (1 m, 1.5 m, and 2 m) based on balaclava (BAL) and MIPS® (ART): means and standard deviations evaluated for the four configurations (w/o BAL w/o ART, w BAL w/o ART, w/o BAL w ART, w BAL w ART) for each drop height level.

Considering the movement of the skull in relation to the brain, the statistical analysis highlighted that the factors significantly influencing the maximum relative angular velocity are ART (p-value≤0.001, η^2=0.29) and drop height (p-value=0.003, η^2=0.27), while their interaction appears negligible (p-value=0.68). As for HIC, BrIC, and PRA, balaclava seems to have a negligible effect on the reduction of the peak values of relative skull-brain angular velocity for all the tested configurations (p-value=0.55). Therefore, Fig. 10 and Fig. 11 show only the results of the drop tests performed with the balaclava, without ART (Fig. 10) and with ART (Fig. 11), from the three drop heights. As expected, an increase in the peak values while increasing the drop height can be noticed both without and with ART. Moreover, these peaks decrease while using the ART both when the skull angular velocity is higher with respect to the brain one and when the brain angular velocity is higher. The percentage reduction in peaks while using ART at the three increasing drop heights was approximately 19.5%, 20.0%, and 23.0% when skull angular velocity exceeded the brain one, leading to an average percentage decrease of 20.8%. This confirms experimentally the positive effect of the analysed ART protection system. The ART causes a delay in the relative brain-skull motion, likely due to rigid body motion of the whole head induced by ART.

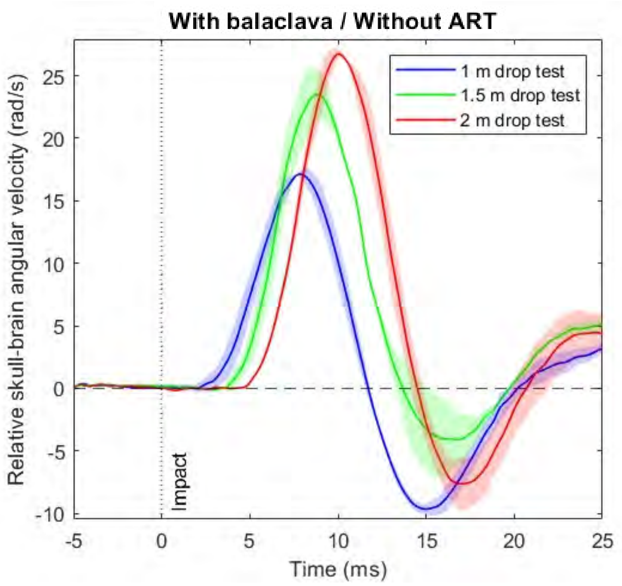

Fig. 10. Relative skull-brain angular velocity evaluated for the drop tests performed from 1 m, 1.5 m, and 2 m with balaclava/without ART.

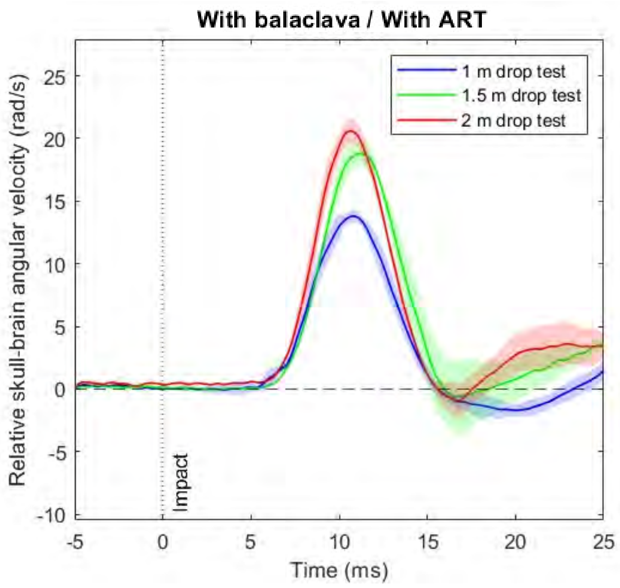

Fig. 11. Relative skull-brain angular velocity evaluated for the drop tests performed from 1 m, 1.5 m, and 2 m with balaclava/with ART.

Regarding the CSF pressure, data were analysed just for the drop tests performed from 1 m. Statistical analysis suggested that the factors significantly affecting the peak values of CSF pressure measured by Frontal, Sphenoid, and Top sensors are ART (p-value≤0.01, η^2=0.40) and balaclava (p-value≤0.01, η^2=0.30). The percentage reduction in pressure in the Frontal, Sphenoid, and Top sensors peaks while using ART was 51.5%, 48.2%, and 31.6% without balaclava, respectively (Fig. 12 and Fig. 13), and 21.7%, 18.8%, and 8.8% with balaclava, respectively (Fig. 14 and Fig. 15). Moreover, positive peak values can be noticed for all of the sensors, apart from the Occipital one, indicative of compressive pressure waves detected in the upper impact region. On the other hand, the Occipital sensor, which is placed on the opposite side, detected a negative pressure wave, corresponding with findings from Nahum [16] and Petrone [10]. Regarding Frontal, Sphenoid, and Occipital sensors, a coup-countercoup sequence can be observed. In particular, while using ART a percentage increase of 23.5% (without balaclava) and 36.4% (with balaclava) in the coup duration can be measured together with lower CSF peak values. Figs 12–15 also show the x and z components of the linear acceleration acquired in the skull and brain surrogates.

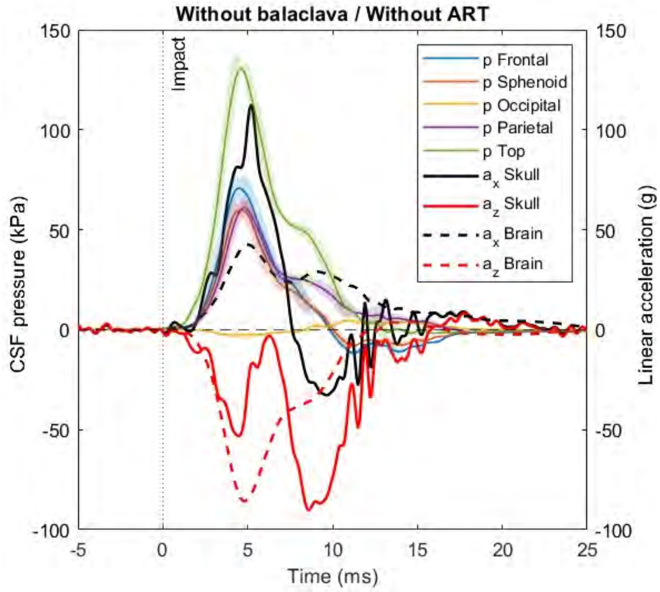

Fig. 12. CSF pressure and skull and brain linear acceleration evaluated for the drop tests performed from 1 m without balaclava/without ART.

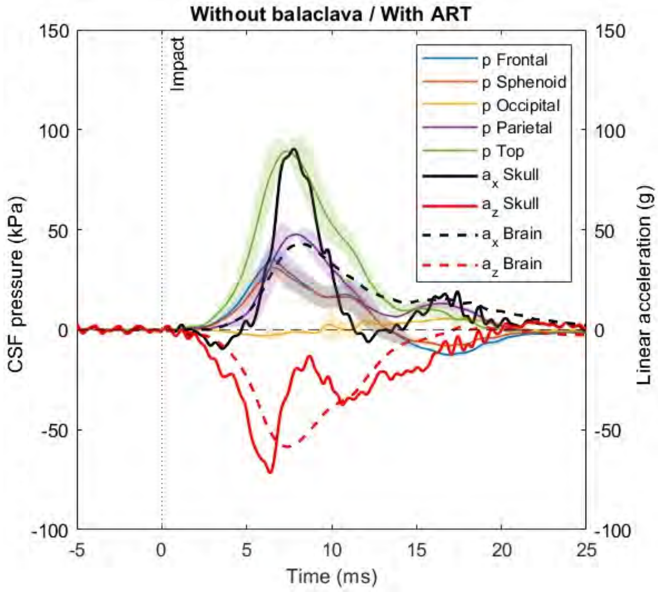

Fig. 13. CSF pressure and skull and brain linear acceleration evaluated for the drop tests performed from 1 m without balaclava/with ART.

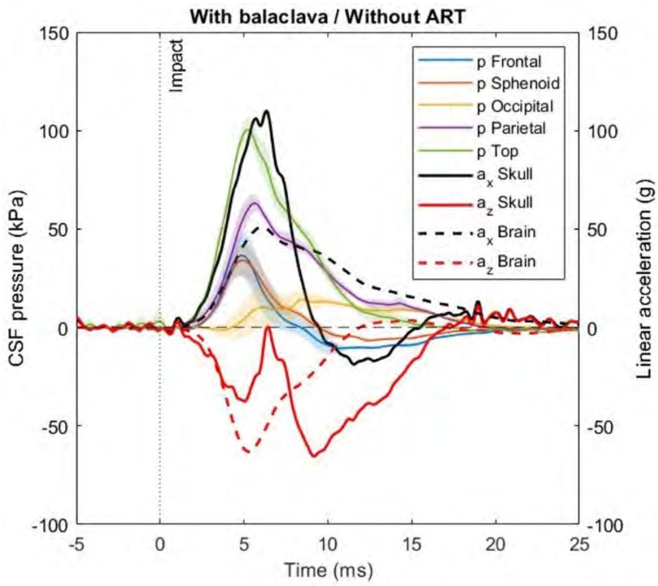

Fig. 14. CSF pressure and skull and brain linear acceleration evaluated for the drop tests performed from 1 m with balaclava/without ART.

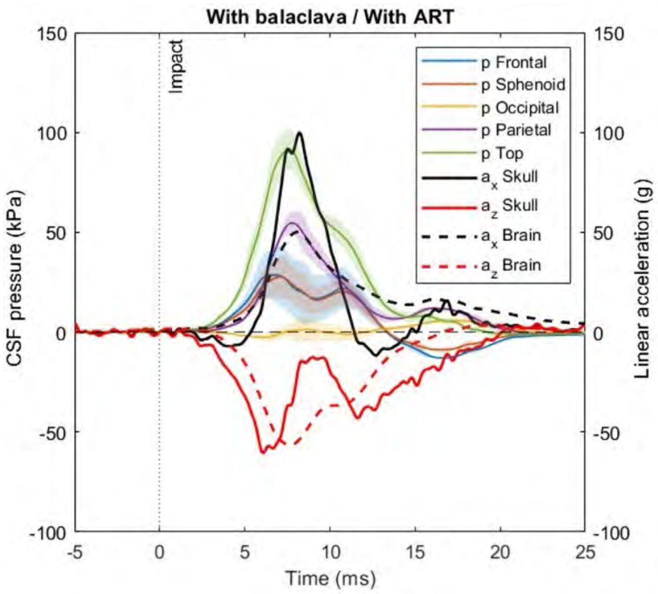

Fig. 15. CSF pressure and skull and brain linear acceleration evaluated for the drop tests performed from 1 m with balaclava/with ART.

The CSF peak pressure values were comparable with PMHS head unprotected impacts from Nahum [16], as well as with experimental results from Petrone *et al.* [10], who reproduced the Nahum experiment using the previous version of unprotected IHHR. When normalised to similar resultant skull accelerations, the present Frontal sensors detected values that are respectively 13.2% and 34.9% lower than Nahum's [16] and Petrone's [10] results: this difference has been considered acceptable since type and location of the literature impacts were different.

In conclusion, the presence of sensorised CSF-meninges-brain simulants inside the IHHR allowed us to assess not only injury criteria based on the kinematics of the head as a whole (HIC, BrIC, and PRA) but also anti-rotational effects on the relative kinematics between brain and skull and on CSF pressure during an impact.

IV. DISCUSSION

The results of this study provide valuable insights into the effectiveness of ART in mitigating head injuries during impacts. The use of a biofidelic replica (IHHR), which allows a relative motion of the brain with respect to the skull, enabled us to study the impact of various factors, including drop height, ART and balaclava on relative skull-brain angular velocity ω_{rel} in addition to injury criteria such as HIC, BrIC, PAV, and PRA. On the other hand, standard rigid headforms commonly adopted by standards for helmet testing are not capable of predicting the effectiveness of the adopted ART regarding skull-brain relative motion because of their lack of an instrumented brain surrogate. In addition, unlike heads featuring a hollow skull, the presence of a CSF simulant inside the IHHR allows brain damping, with CSF pressures waves that can be detected by the sensors installed on the inner surface of the top skull.

Experimental findings suggest, as expected, that drop height significantly influences HIC, BrIC, PAV, PRA, and ω_{rel}, with higher values observed at increasing fall heights. Moreover, all the aforementioned variables, apart from HIC, were found to be significantly affected by ART, with a major reduction in BrIC values as drop height increases. This highlights the more evident effect of ART on BrIC values for higher drop heights. Despite what was expected based on the results of Stark *et al.* [4] and Bonin *et al.* [5], friction reduction induced by a balaclava seems to have a negligible effect on the head kinematics. Nevertheless, further investigations are needed to quantify the friction coefficient between the IHHR and the adopted helmet along with its reduction when a balaclava is included. Additionally, BrIC reduction observed with the IHHR without balaclava for the drop tests performed from 0.75 m (previous analysis [21]) and 1 m (current analysis) while using ART are similar, thus confirming the consistency of the results. In the drop tests conducted in [21] from 0.75 m without the balaclava, using both the Hybrid III headform and the IHHR, a noticeable reduction in the BrIC value was observed with the implementation of the MIPS® system, indicating its efficacy in mitigating head injury risks. Specifically, the protective effectiveness assessed with the IHHR, incorporating a biofidelic brain simulant, was approximately half the protection effectiveness evaluated using the standard rigid headform in terms of BrIC. These differences, rather than to skin friction (being both heads covered of silicone rubber), may be due to mass differences or to the additional brain freedom of movement inside the skull for the IHHR. The disparity in effectiveness between the two headforms emphasises the impact of utilising a biofidelic headform on the evaluation of protective devices, suggesting that the presence of a brain surrogate and its relative motion with respect to the skull can influence the observed outcomes. This emphasises the importance of advancing the biofidelity of Head Replicas and of materials used in surrogate brain tissue to better replicate injury mechanisms and improve the accuracy of protective equipment evaluations [21]. In addition, in a hybrid approach, the use of a biofidelic headform would lead to skull kinematics that are more suitable for driving detailed numerical simulations than rigid headforms.

Considering the relative motion of the skull and brain surrogates, the observed reduction in peak angular velocity with the use of the ART indicates a positive effect in mitigating rotational forces, suggesting a potential decrease in shear deformation of the brain tissue. This finding underlines the potential of ART, such as the MIPS® protection system, in reducing the risk of TBI by minimising rotational forces that could lead to shear deformation and associated damage to brain tissue.

Regarding the analysis of CSF pressure, only the drop tests conducted from 1 m were analysed. All sensors, except for the Occipital one, were subjected to initial compressive pressure waves. Statistical analysis highlighted that ART and balaclava significantly influence the peak values of CSF pressure recorded by Frontal, Sphenoid, and

Top sensors. ART is prone to cause an increase in the coup duration, with lower CSF peak values. These peaks resulted in values similar to those detected by literature impacts, with a discrepancy that was considered satisfactory due to the varying type and location of the analysed impacts.

Despite the strengths of this study in evaluating the relative motion between the skull and brain surrogates and CSF pressures, some limitations should be noted. Moreover, recent metrics like DAMAGE [26] or the NFL football helmet evaluation [27] should be evaluated to enhance the assessment of the predictive helmet protection effectiveness. The limited number of tests and the use of a single helmet, which underwent multiple impacts, should be observed. Further research with a larger sample size would help to validate these findings. In addition, the helmet should be replaced with a new one of the same type after each test. Moreover, technical issues related to pressure sensors should be solved to evaluate CSF pressures detected by the Temporal sensor and by the left set of sensors, thus allowing for analysis of the signals acquired by the 1.5 m and 2 m drop tests.

V. CONCLUSION

In conclusion, this study demonstrates the effectiveness of using the biofidelic IHHR to assess TBI severity. According to the above findings, the IHHR is a suitable tool to experimentally evaluate the degree of protection of the adopted ART. Indeed, the use of sensors embedded in this biofidelic replica allowed for the evaluation of drop height, ART, and balaclava effects on injury criteria, such as Head Injury Criterion (HIC), Brain Injury Criterion (BrIC), Peak Angular Velocity (PAV), and Peak Rotational Acceleration (PRA), and on relative skull-brain angular velocity (ω_{rel}). Experimental findings highlight the importance of considering the relative motion between skull and brain in assessing head injury risk along with the effectiveness of protective measures like ART. The IHHR also allows us to assess CSF pressures in several zones, which is valuable for performing a validation against experimental tests, such as that carried out by Nahum.

Overall, this research contributes to advancing the reproducibility of head injury mechanisms and the study of the role of protective technologies in mitigating traumatic brain injury. Further experiments will also explore brain stress to exploit the full capabilities of the IHHR.

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