The Effectiveness of Cervical Airbags in the Control of Head and Neck Kinematics

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Abstract Cyclists represent a significant percentage of seriously or fatally injured road users. Head and brain injuries in cyclists have been extensively studied, but less focus has been given to cervical injuries. Airbags are being designed to mitigate or prevent injuries in cyclists. The objective of this study was to assess the effectiveness of three airbag prototypes designed primarily to prevent hyperextension cervical injuries in cyclists. A test series was conducted with a Hybrid III 50th percentile. The performance of the airbags was assessed by comparing head kinematics and selected injury criteria. The most noticeable differences were obtained for hyperextension angles. The average angle without airbag was 50.06 ±1.73 degrees, compared to 41.99 ±1.29, 37.20 ±2.05, and 46.53 ±2.21 degrees, respectively, for the tests with the three different airbags. No substantial differences in peak linear acceleration and head angular velocity were obtained in the tests; however, a relation between volume capacity and airbag pressure was observed. There were no relevant reductions in the brain injury criterion. The lowest values were obtained using Airbag 1, with an improvement of 2.4 % in the average brain injury criterion. Further research is required to evaluate the effectiveness of airbags in the occurrence of cervical trauma.

Keywords Cervical airbag, cyclist injuries, head kinematics, hyperextension, neck injuries.

I. INTRODUCTION

Compared to other means of transportation, cycling remains the only transportation mode in which the number of fatalities has not decreased since 2010 [1]. Worldwide, pedestrians and cyclists represent 26% of all traffic-related deaths [2]. In Europe, the proportion of cyclists injured with respect to the total number of road users injured rose from 7% to 9% from 2010 to 2019. The same percentages were observed for fatally injured cyclists, corresponding to 2,035 cyclist deaths in Europe in 2019 [1]. In the United States, 38,886 cyclists were injured and 938 died in 2020, the latter corresponding to 2.4 % of all traffic-related fatalities from that year [3].

While head and brain injuries in cyclists have been extensively studied in the past, spinal injuries have received less attention. Specifically, upper cervical spine injuries (uCSIs) are frequent and occur when the head sustains forces during trauma [4]. This anatomical region has complex supporting structures that allow weight to be transferred between the head and upper body, enabling the motion of the neck [4]. Cycling-related spinal injuries have increased in recent years [5-6]. Neck injuries are more likely to occur in collisions between cyclists and motor vehicles, and cyclists sustaining these injuries are 15 times more likely to die than those without such injuries [7]. In a recent study, cycling was identified as the second most frequent cause of cervical sprains and the most common for cervical fractures in men, and the second most frequent cause of cervical fractures in women [5].

In a study evaluating CSIs in the south-east region of Norway between 2015 and 2019, 12% of the documented CSIs were related to cycling. The most frequent injury occurring concomitantly with CSI was traumatic brain injury, present in 48.2% of cyclists with cervical injuries. The most common CSIs in cyclists were C6/C7 fractures, occipital condyle fractures, and C5/C6 fractures. Occipital condyle fractures are frequently caused by the rotation and compression at the C0/C1 joint, which may occur when the cyclist falls over the handlebars and hits the ground headfirst [8]. This could induce hyperextension of the cervical spine.

The most effective passive safety equipment currently used by cyclists is helmets. There is wide agreement on the effectiveness of helmets in reducing head and brain injuries in cyclists [8-10]. Helmet use in cyclists has been

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associated to reductions of up to 51% in head injuries, 33% in face injuries, and 65% in fatal head injuries [11]. Until now, helmets have been primarily designed to mitigate or prevent injuries to the head [12]. However, there is less agreement on the relation between helmet use and neck injury [7][9–14]. Some current helmet designs are integrating airbags that could provide further protection to the head and even to the cervical spine [14-15]. Several designs have been evaluated so far: in some, the airbag is worn around the neck, and in others it even surrounds the helmet when deployed [15–17]. The inclusion of an airbag to the helmet would allow the head to absorb more impact energy before a maximum force level is reached during the blow [17]. If filled with air, the pressure could be adapted to meet the desired mechanical behaviour of this protective device [17].

The Swedish Hövding 2.0 is an airbag helmet designed to improve cyclist protection. When deployed, the airbag surrounds the helmet and the neck. Past studies have evaluated the effectiveness of this equipment, measuring reductions in the peak linear acceleration of the head, in the rotational acceleration of the head, and in head injury criterion (HIC) values when compared to other helmets without airbags. In the shock absorption test, the Hövding 2.0 resulted in a peak linear acceleration of 48 g, which was three times lower than the average 175 g of the other conventional helmets. In the oblique test, lower strains were measured with the airbag helmet than without the airbag [16-17].

In another study, an anthropomorphic test device (ATD) wearing a helmet airbag was used during two crash tests at 6.86 m/s and 11.1 m/s impact velocity, respectively. In the first test there was no airbag deployment while in the second test the airbag deployed. The effectiveness of the helmet airbag is difficult to compare in this study as only two crash tests were performed, and they were carried out at different impact velocities. Nevertheless, neck injury criterion (NIC) values were lower in the test where the airbag deployed even though this test was done at higher impact velocity [15].

In a recent study, finite element model simulations were used to compare the efficiency of an airbag helmet in mitigating traumatic brain injuries versus a conventional helmet. The airbag helmet lowered the impact energy, therefore reducing peak forces applied to the head. There was also a decrease in the peak linear acceleration values and a delay in the time at which this peak occurred, resulting in lower HIC36 values. Maximum principal strain was reduced with the airbag helmet [18].

There is still not enough information on the performance of helmet airbags in real-world scaled tests and on its effectiveness in mitigating or preventing cervical injuries in cyclists. The objective of the current study was to assess the effectiveness of three airbag prototypes designed primarily to prevent hyperextension cervical injuries in cyclists. The performance of the airbags was assessed by comparing the resulting linear and rotational head kinematics, the brain injury criterion, and the amount of cervical hyperextension. Despite not being representative of specific real-world falls in cyclists, this test setup was chosen to have a controlled loading environment for the hyperextension of the neck, which could later be used to validate finite element model simulations with the airbag prototypes.

II. METHODS

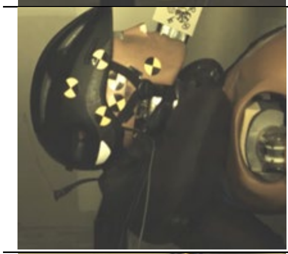
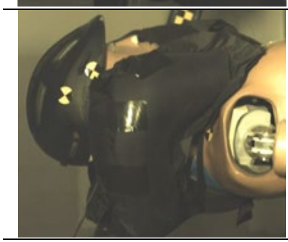
A customised test rig was designed and built to produce the hyperextension of the cervical spine of the Hybrid III 50th percentile dummy. The dummy was positioned flat and supine onto a rigid horizontal plate at the resting position leaving the head free to rotate without any contact throughout the experiment, which can be seen in Fig. 1. The dummy's initial position was kept constant throughout the tests. The plate and dummy were then elevated 50 cm from the resting position. An electromagnet that was supporting the dummy and the plate was deactivated to let the structure fall guided along vertical rails. The plate was then abruptly stopped using a rigid surface, resulting in the hyperextension of the dummy's cervical spine. The coordinate system used in this study is presented in Fig. 2.

A total of 21 tests were carried out under the same conditions, varying only the airbag prototype (if used) and the inflation pressure of the airbags. Three different prototypes were used at three pressure levels: 0.10, 0.15, and 0.20 bar. Two repeats were conducted for each airbag at each pressure level. They were inflated to the desired pressure prior to testing using an air compressor and placed around the neck. The pressure was kept constant throughout the test. A conventional helmet available in the market was used in all the tests. The test matrix is shown in Table I.

Fig. 1. Test setup and initial position of the ATD.

Fig. 2. Coordinate system used.

TABLE I
TEST MATRIX

Airbag used		Test number	Pressure airbag (bar)
	No airbag	26	-
		28	-
		29	-
	Airbag 1	14	0.10
		15	0.10
		16	0.15
		17	0.15
		18	0.20
		19	0.20
	Airbag 2	8	0.10
		13	0.10
		9	0.15
		12	0.15
		10	0.20
		11	0.20
	Airbag 3	20	0.10
		21	0.10
		22	0.15
		23	0.15
		24	0.20
		25	0.20

The motion was recorded at 1,000 Hz using a high-speed video camera. Photo targets were placed on the head of the dummy to enable the subsequent tracking to calculate the hyperextension angles. This technique was used for all the tests without airbag and the ones with Airbag 1 and Airbag 2. The design of Airbag 3 obstructed these photo targets, so the nose angle was measured instead. The hyperextension angle was defined as the difference in the angle between the moment of maximum rotation and the initial angle. The latter was defined as the average initial angle in the tests without airbag. The average hyperextension angles were calculated for each test.

Dummy head kinematics in the sagittal plane were measured at 10,000 Hz. This included the linear acceleration in the x and z directions and the angular velocity in the y axis at the centre of gravity (CG) of the head. These data were processed and filtered using CFC 300 filters. Head kinematics and hyperextension angles were analysed to compare the effectiveness of the different airbags with respect to the baseline case, in which the dummy was not equipped with any airbag.

The Brain Injury Criterion (BrIC) was calculated for each test using the following equation:

$$BrIC = \sqrt{\left(\frac{w_x}{w_{xc}}\right)^2 + \left(\frac{w_y}{w_{yc}}\right)^2 + \left(\frac{w_z}{w_{zc}}\right)^2}, \quad (1)$$

where w_x, w_y, and w_z are the maximum angular velocity components, and w_{xc}, w_{yc}, and w_{zc} are the critical values for each orthogonal direction [19].

III. RESULTS

The data from the tests are presented in the following sections. In addition, pictures from the high-speed cameras are shown in Figures A5 - A8 in the Appendix.

Linear Acceleration of the Head

The linear accelerations of the head at the CG are shown in Figures 3 and 4 for airbags inflated at 0.10 bar, and the rest in Figures A1 - A4 in the Appendix. The peaks are presented in Table II. With respect to a_x, the acceleration curves are presented at each pressure level, always including the tests without airbags for better comparison. Airbags 1 and 2 resulted in higher initial peaks than in the tests with Airbag 3 and without airbag. These maximum values were minimised at 0.15 bar for both Airbags 1 and 2. In addition, the following peaks occurring at approximately 0.02 seconds varied considerably with airbag pressure. In the case of Airbag 2, the minimum peak occurred at 0.15 bar (-6.32 g). When considering Airbag 3, this peak was reduced as airbag pressure increased. No influence of the inflation pressure was observed for Airbag 1.

Regarding a_z, more noticeable differences were observed. Overall, the third airbag did not improve the linear acceleration in the z-axis compared to when no airbag was used. However, more relevant differences were seen with Airbags 1 and 2. Even though the differences in the minimum values between these two airbags and the tests without airbags were not substantial, there were improvements in terms of maximum accelerations. Specifically, Airbag 2 consistently achieved lower maximum acceleration magnitudes regardless of the airbag pressure, with reductions of up to 7 g (40 % with respect to the worst value from the baseline case).

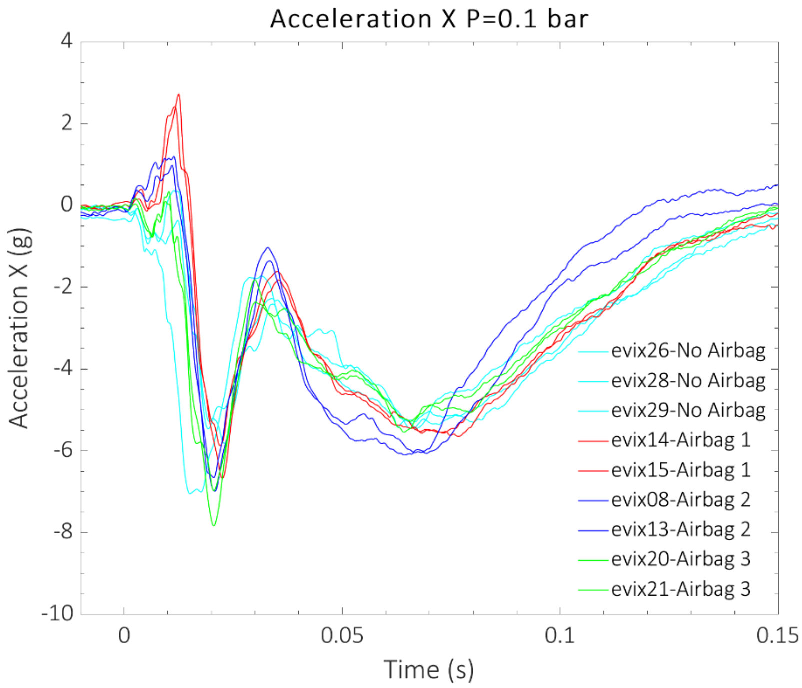


Fig. 3. Head linear acceleration in the x-axis for the tests without airbag (light blue) and with airbags at 0.10 bar (red, dark blue, and green).

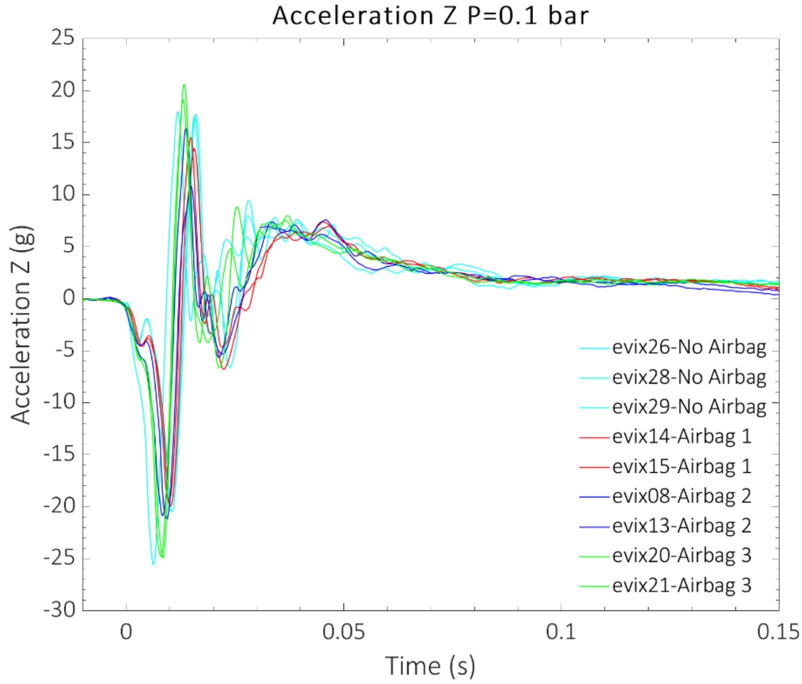


Fig. 4. Head linear acceleration in the z-axis for the tests without airbag (light blue) and with airbags at 0.10 bar (red, dark blue, and green).

TABLE II

PEAK A_X AND A_Z VALUES OF THE CENTRE OF GRAVITY OF THE HEAD FOR EACH TEST

Airbag used	Test number	Pressure airbag (bar)	Maximum a_x (g)	Minimum a_x (g)	Maximum a_z (g)	Minimum a_z (g)
No airbag	26	-	0.01	-7.05	17.97	-25.59
	28	-	-0.02	-5.46	17.72	-20.50
	29	-	0.36	-5.82	17.47	-19.69
Airbag 1	14	0.10	2.73	-6.67	14.44	-19.94
	15	0.10	2.41	-5.87	15.50	-19.49
	16	0.15	0.73	-6.51	13.96	-21.76
	17	0.15	1.46	-7.81	16.91	-23.51
	18	0.20	1.60	-6.79	17.50	-22.13
	19	0.20	1.92	-5.00	19.90	-21.56
Airbag 2	8	0.10	0.98	-6.99	16.44	-20.89
	13	0.10	1.20	-6.66	10.83	-21.21
	9	0.15	1.05	-6.32	12.94	-21.90
	12	0.15	1.98	-6.72	12.14	-22.24
	10	0.20	2.69	-6.89	14.41	-19.64
	11	0.20	2.71	-7.34	15.85	-21.84
Airbag 3	20	0.10	0.35	-7.84	19.13	-24.81
	21	0.10	0.10	-6.98	20.61	-24.95
	22	0.15	0.50	-6.93	18.01	-22.43
	23	0.15	-0.02	-6.71	20.56	-24.02
	24	0.20	-0.08	-6.67	20.83	-24.55
	25	0.20	-0.01	-6.49	18.82	-24.09

Hyperextension Angles

As aforementioned, hyperextension angles were calculated for each test and the results are shown in Table III. The hyperextension angles were highest when no airbag was used for that test (baseline case), with an average angle of 50.06 ±1.73 deg. These angles were improved when any of the three airbag prototypes were used. The lowest reduction was observed when Airbag 3 was employed, with an average hyperextension angle of 46.53 ±2.21 deg. Airbag 1 resulted in an average value of 41.99 ±1.99 deg, which corresponded to a 16% improvement with respect to the baseline case. Airbag 2 was the most effective in decreasing this angle, with reductions of more than 25%, corresponding to an average value of 37.20 ±2.05 deg. No clear trends were observed when the airbag pressure was increased.

TABLE III
HYPEREXTENSION ANGLE FOR EACH TEST

Airbag used	Test number	Pressure airbag (bar)	Hyperextension angle (deg)	Mean ±SD (deg)
No airbag	26	-	48.12	
	28	-	49.75	
	29	-	52.32	50.06 ±1.73
Airbag 1	14	0.10	42.12	
	15	0.10	42.65	
	16	0.15	39.35	
	17	0.15	43.52	
	18	0.20	41.93	
	19	0.20	42.37	41.99 ±1.29
Airbag 2	8	0.10	39.76	
	13	0.10	38.13	
	9	0.15	35.96	
	12	0.15	39.26	
	10	0.20	36.19	
	11	0.20	33.90	37.20 ±2.05
Airbag 3	20	0.10	47.16	
	21	0.10	50.11	
	22	0.15	44.11	
	23	0.15	48.31	
	24	0.20	44.16	
	25	0.20	45.31	46.53 ±2.21

Angular Velocity of the Head

The angular velocity of the CG of the head in the y-axis was recorded for each test. These curves are shown for the tests with the three airbags in 5 - 7. The curves for the tests without airbag are not included in these figures to show the effect of inflation pressure more clearly. However, the figures with both set of curves have been included in the Appendix. The peak values for each test are included in Table IV.

Regarding each of the three airbag prototypes, there seemed to be a relation between airbag size and optimum pressure level. The airbag prototypes are numbered in order of volume capacity from lowest to highest. Overall, the lowest maximum values of head angular velocity for Airbag 1 occurred when it was inflated at 0.10 bar; for Airbag 2, at 0.15 bar; and for Airbag 3, at 0.20 bar.

Overall, the highest peak head angular velocity occurred in test number 26, which was carried out without an airbag. For the tests performed with airbags at 0.10 bar, Airbag 1 reduced the maximum values the most, with peaks of 20.51 and 20.40 rad/s. At 0.15 bar, Airbag 2 was the most effective of the three, with peak values of 20.89 and 20.63 rad/s; however, there were no relevant differences with respect to the tests without airbag. At 0.20 bar, no airbag consistently reduced the maximum angular velocity without airbag.

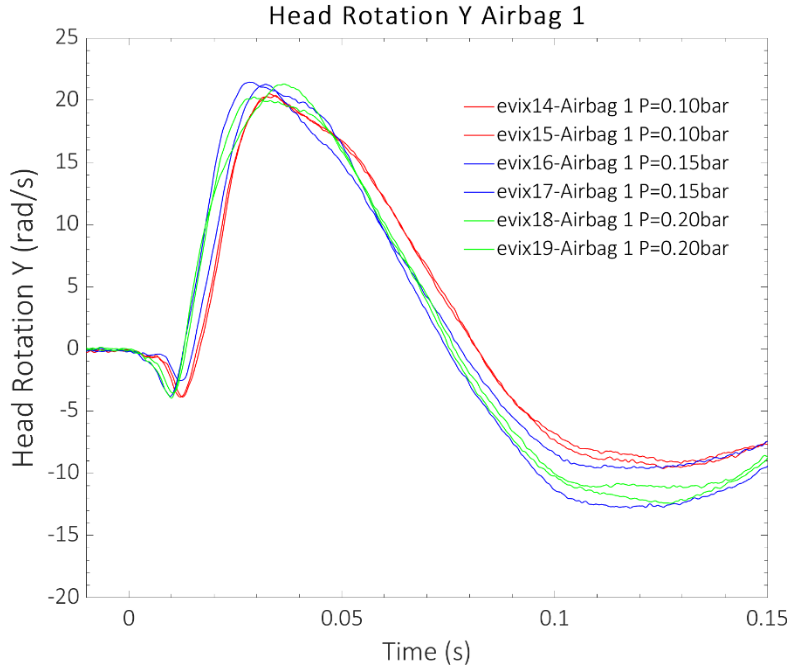


Fig. 5. Head angular velocity for tests with Airbag 1 at 0.10 bar (red), 0.15 bar (blue), and 0.20 bar (green).

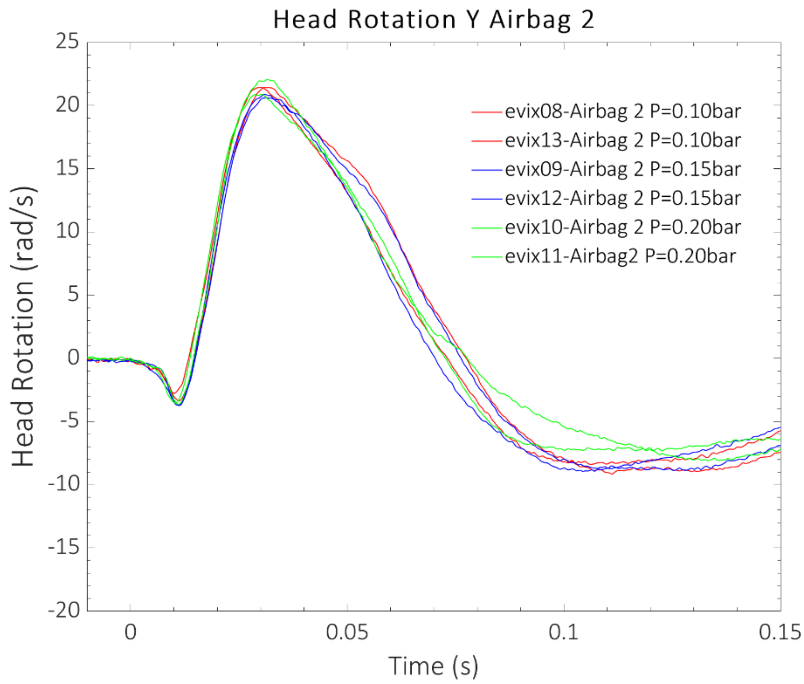


Fig. 6. Head angular velocity for tests with Airbag 2 at 0.10 bar (red), 0.15 bar (blue), and 0.20 bar (green).

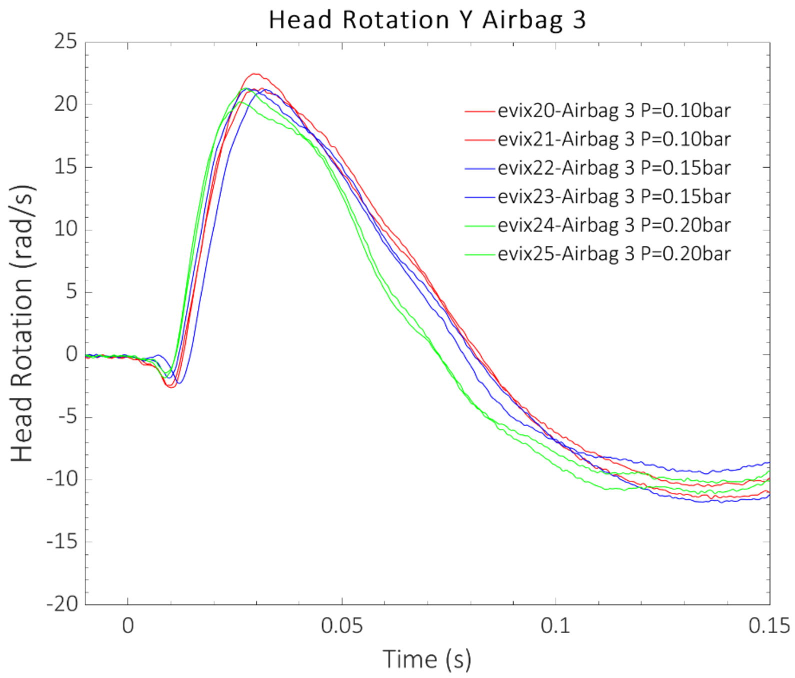


Fig. 7. Head angular velocity for tests with Airbag 3 at 0.10 bar (red), 0.15 bar (blue), and 0.20 bar (green).

TABLE IV
PEAK VALUES FOR THE ANGULAR HEAD VELOCITY FOR EACH TEST

Airbag used	Test number	Pressure airbag (bar)	Maximum w_y (rad/s)
No airbag	26	-	22.86
	28	-	20.75
	29	-	20.60
Airbag 1	14	0.10	20.51
	15	0.10	20.40
	16	0.15	21.28
	17	0.15	21.45
	18	0.20	20.27
	19	0.20	21.31
Airbag 2	8	0.10	21.43
	13	0.10	21.43
	9	0.15	20.89
	12	0.15	20.63
	10	0.20	20.89
	11	0.20	22.06
Airbag 3	20	0.10	22.50
	21	0.10	21.33
	22	0.15	21.22
	23	0.15	21.31
	24	0.20	21.34
	25	0.20	20.24

Brain Injury Criterion

BrIC values were calculated for all tests and they are presented in Table V. There was a small reduction in the average BrIC using Airbag 3 (0.378 ±0.012), which was further improved with Airbag 2 (0.376 ±0.008) and Airbag 1 (0.370 ±0.009). However, almost negligible differences were obtained between the tests with and without airbag. Consequently, the probabilities of AIS 2+ and 3+ injury were also similar in all the tests. The highest probabilities corresponded to the tests without airbag, with average p(AIS 2+) and p(AIS 3+) of 23.66 ±2.87% and 6.43 ±0.87%, respectively. The lowest average p(AIS 2+) was 22.17 ±1.28%, and the lowest average p(AIS 3+) was 5.97 ±0.38%, both corresponding to the tests performed with Airbag 1.

TABLE V
BRIC VALUES AND PROBABILITIES OF AIS 2+ AND AIS 3+ INJURY FOR EACH TEST

Airbag used	Test number	Pressure airbag (bar)	BrIC	Mean ±SD	p(AIS2+)	p(AIS3+)
No airbag	26	-	0.405			
	28	-	0.368			
	29	-	0.365	0.379 ±0.018	23.66 ±2.87%	6.43 ±0.87%
Airbag 1	14	0.10	0.363			
	15	0.10	0.361			
	16	0.15	0.377			
	17	0.15	0.380			
	18	0.20	0.359			
	19	0.20	0.378	0.370 ±0.009	22.17 ±1.28%	5.97 ±0.38%
Airbag 2	8	0.10	0.380			
	13	0.10	0.380			
	9	0.15	0.370			
	12	0.15	0.366			
	10	0.20	0.370			
	11	0.20	0.391	0.376 ±0.008	23.11 ±1.29%	6.25 ±0.39%
Airbag 3	20	0.10	0.399			
	21	0.10	0.378			
	22	0.15	0.376			
	23	0.15	0.378			
	24	0.20	0.378			
	25	0.20	0.358	0.378 ±0.012	23.40 ±1.79%	6.34 ±0.54%

IV. DISCUSSION

Cyclists still represent a significant proportion of seriously or fatally injured road users. The effectiveness of helmets in mitigating or preventing certain injuries in cyclists has been proven and widely analysed in the past [8-9][11-12]. Specifically, this piece of equipment has been shown to reduce the risk of sustaining head and traumatic brain injuries. However, there is no consensus on the relation between helmets and neck injuries. Airbags have been proposed as a possible way to reduce the risk of suffering neck injuries [14-15].

This study analysed head kinematics in an ATD wearing a helmet and a cervical airbag during the hyperextension of the neck. The impact occurred at 11.3 km/h and all the movement was assumed to be in the sagittal plane. Cyclists frequently suffer falls or collisions that lead to the hyperextension of the neck, during which neck injuries could potentially be minimised using cervical airbags. Thus, the experiments performed had the objective of evaluating the effectiveness of cervical airbags during the hyperextension of the neck.

With respect to linear kinematics of the head, peak maximum acceleration in the z-axis was improved with Airbags 1 and 2. The maximum magnitudes of a_z were reduced when using Airbag 1 at 0.10 bar and 0.15 bar, and they were consistently lower with Airbag 2 regardless of airbag pressure. Prior to this peak there was another at approximately 0.01 s in the opposite direction. However, the differences in these peak values were minimal. This could be due to the amount of time available for the airbags to absorb some of the impact energy; at first there is not enough time for the airbags to reduce acceleration, but more time has passed when the second peak occurs. Airbags 1 and 2 presented a better fit around the neck than Airbag 3, which could explain these reductions in peak acceleration compared to Airbag 3. Nevertheless, there were no relevant differences in terms of peak acceleration in the x direction. The most important difference was in the initial peak in these curves, between the start and 0.01 s approximately. The initial acceleration in the x-axis with Airbag 3 followed the same tendency as without airbag; on the other hand, there was an initial peak when Airbags 1 or 2 were used. This could also be related to the fit of the airbags, which, in turn, could have also influenced the type of loading.

Past studies have presented reductions in peak acceleration with airbag helmets [16–18]. This decrease in linear acceleration could be more significant than in the current study due to the differences in airbag design and test setup. In these past studies, airbag helmets were used, in which the inflated airbag surrounded the neck and helmet, and drop tests were performed where the head contacted an impactor. However, the airbag prototypes used in the current study only surrounded the neck once deployed and the head was free to rotate without any contact. These differences in contact area between the airbag and subject and experiment setup are important when considering energy absorption [18]. Further research is needed to fully understand the differences resulting from the two airbag designs and the best tests to evaluate their effectiveness.

Moreover, regarding the rotation of the head, more considerable differences were found in hyperextension angles. All airbags used reduced the average hyperextension angles. The highest reduction was achieved with Airbag 2, followed by Airbags 1 and 3, respectively. Airbag 2 decreased the average angle by 25.7% and presented the lowest hyperextension angle of the test series when inflated at 0.2 bar (33.90 degrees). This was a 35.2% improvement with respect to the worst-case scenario without airbag. In a past study, the airbag helmet improved rotational acceleration of the head with respect to the tests without airbag [16]. However, no focus was given to rotational angles in said study. Two other analyses evaluated extension angles in the upper cervical spine (0-C2) in tests with pure bending moments [20-21]. The average angle at which injury at the upper cervical spine occurred in extension was 50.2 ±11.4 degrees for females and 42.4 ±8.0 degrees for males [20-21]. However, these values should not be directly compared to the hyperextension angles obtained in the present analysis, as the angles were not measured following the same methodology and the direct transfer between the Hybrid III and human is not possible, but they can serve as a first approximation for possible injury tolerances.

When considering head angular velocity, there was a relation between volume capacity and airbag pressure. Airbag 1 had a volume capacity of 6.2 L, Airbag 2 of 8 L, and Airbag 3 of 11 L. The lowest peaks in rotational velocity were achieved at 0.10 bar with Airbag 1 (lowest pressure level); at 0.15 bar with Airbag 2 (middle pressure level); and at 0.20 bar with Airbag 3 (highest pressure level). Therefore, the bigger the airbag, the higher the pressure level that was needed to obtain the optimum results regarding rotational velocity for each airbag

There were no relevant differences in BrIC in the cases evaluated. In the present study, even if HIC values were also calculated for all the tests they were not included in this manuscript as the obtained values were associated to probabilities of injury close to zero. In past studies, HIC values were reduced when helmet airbags were used [17-18]. The reductions in the probability of injury in these studies could have been more significant than in the current one due to airbag design and test conditions; the airbags surrounded the neck and helmet and there was direct impact between the head form and surface. Therefore, there was more surface area for energy absorption, leading to higher peak acceleration reductions, which in turn resulted in lower HIC values. In another study, two crash tests were performed at different impact velocities using a helmet airbag [15]. In the lower impact velocity test, there was no airbag deployment; in the higher one, the airbag deployed. HIC values were higher in the second test even though the airbag deployed. Nevertheless, NIC values were calculated in the cited study, and the use of airbag led a reduction of 33%. These two results highlight the need for further research regarding injury criteria when airbags are used.

There were some limitations of this study that need to be discussed. The Hybrid III was the ATD chosen for these tests. Although this dummy was not specifically designed for this purpose, it is used in frontal and in some rear crash tests where the movement of the head is primarily contained in the sagittal plane. This was also the case in these experiments, thus the choice of ATD. Nevertheless, limitations in the representation of the human neck in the Hybrid III need to be considered before transferring these results to humans. Moreover, tightness of fit was an important factor in these tests. The position of the airbags was repeated in all the tests as best as possible, but this could have influenced the results obtained. Although the ATD was tightly secured to the horizontal plate to prevent any movement of the torso and lower body, there might have been minor uncontrolled displacements of the dummy during the tests. A procedure was designed to place the ATD in the same position in all the tests, but some variation could have occurred. In addition, comparison with past studies is limited as the airbag design was different to the studies mentioned. The improvement in hyperextension angles and peak accelerations best indicate the potential of these airbag designs of reducing the risk of suffering cervical injuries during the hyperextension of the neck. Future work using these data is in process and it aims to evaluate the performance of the airbags through finite element model simulations. The results from the current study will be used to validate the simulations, which was the rationale behind the chosen controlled loading environment. It is expected that the performance of the airbags would be different under more realistic impact scenarios, like the ones expected for cyclist collisions or falls.

V. CONCLUSIONS

The hyperextension of the neck of the Hybrid III 50th percentile was analysed during 21 tests. Three airbag prototypes were used in 18 of these tests to evaluate different kinematic parameters and injury criteria. The most important differences were obtained for hyperextension angles. This parameter could indicate possible reductions in NIC which should be further studied. Airbag 2 achieved the best results in terms of hyperextension angles. Volume capacity and airbag pressure were related; differences in kinematic parameters were observed for each airbag depending on pressure, specifically for head angular velocity and peak linear acceleration. Further research is warranted to evaluate the effectiveness of cervical airbags in reducing cervical injuries in cyclists.

VI. ACKNOWLEDGEMENT

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VII. REFERENCES

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VIII. APPENDIX

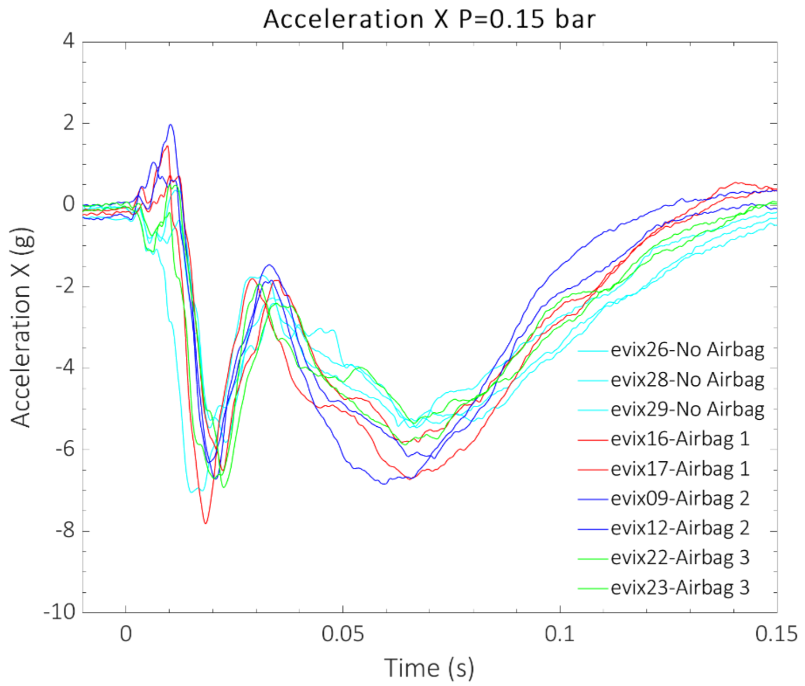


Fig. A1. Head linear acceleration in the x-axis for the tests without airbag (light blue) and with airbags at 0.15 bar (red, dark blue, and green).

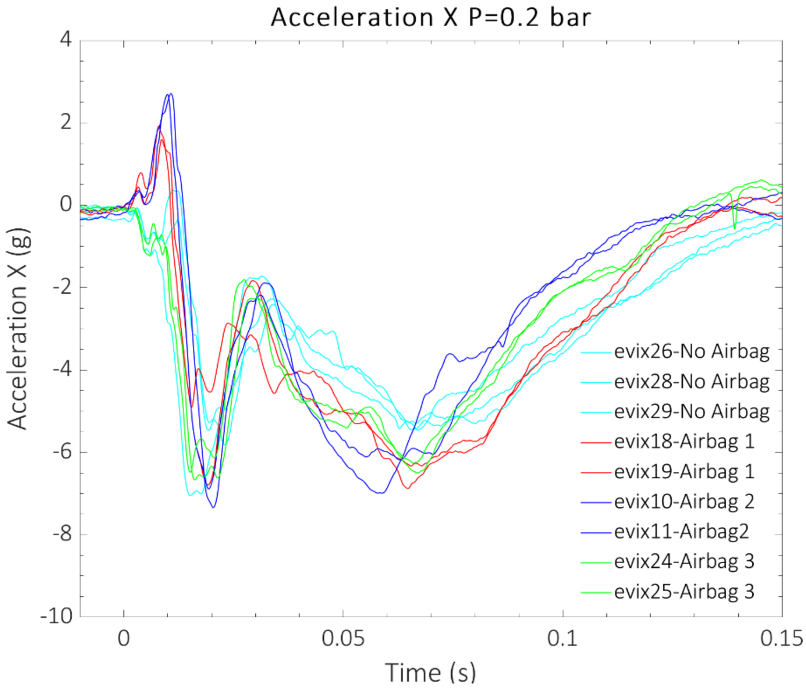


Fig. A2. Head linear acceleration in the x-axis for the tests without airbag (light blue) and with airbags at 0.20 bar (red, dark blue, and green).

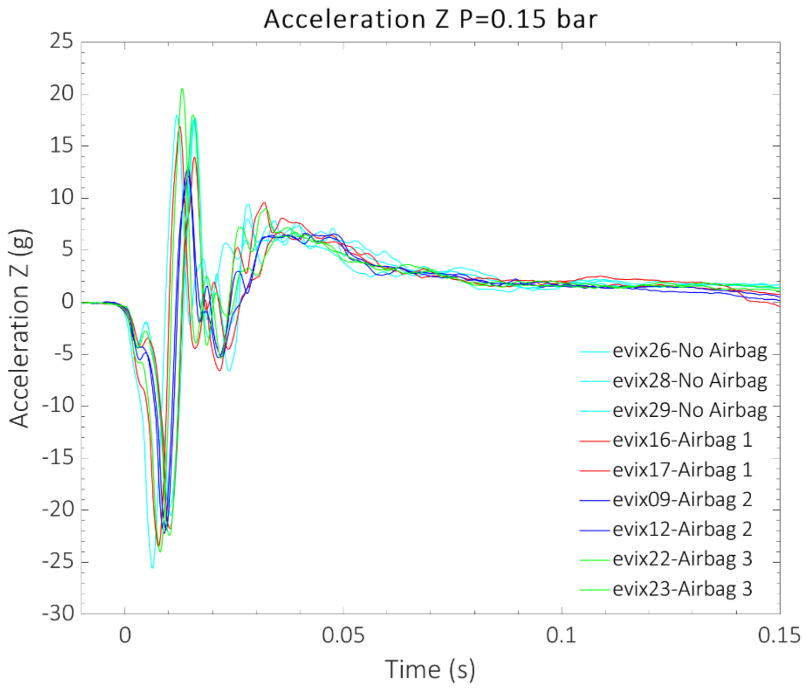


Fig. A3. Head linear acceleration in the z-axis for the tests without airbag (light blue) and with airbags at 0.15 bar (red, dark blue, and green).

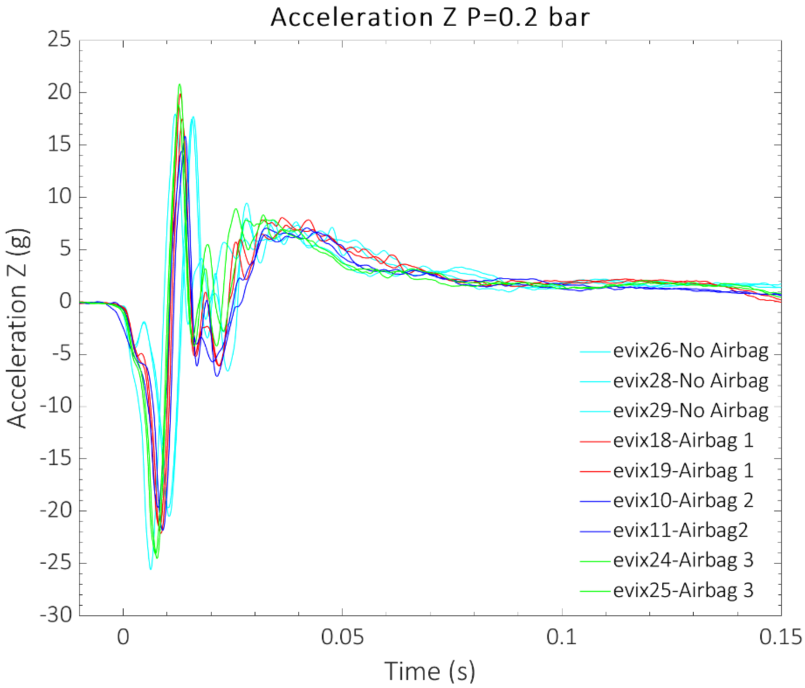
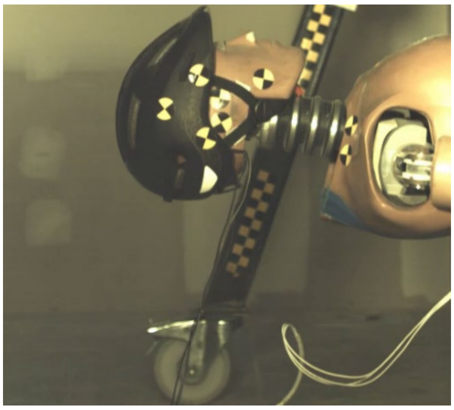


Fig. A4. Head linear acceleration in the z-axis for the tests without airbag (light blue) and with airbags at 0.20 bar (red, dark blue, and green).



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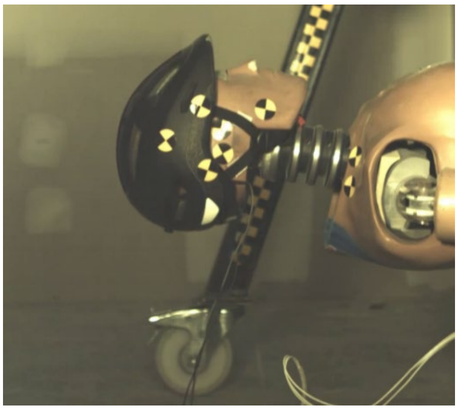
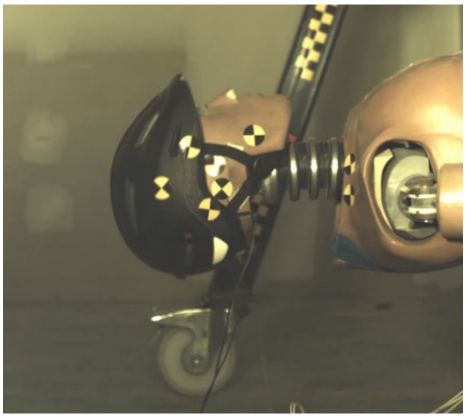


T = 0 ms (trigger)



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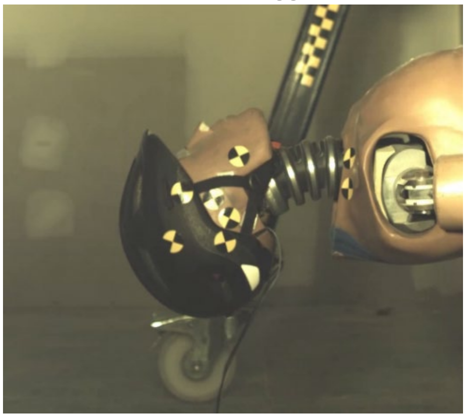

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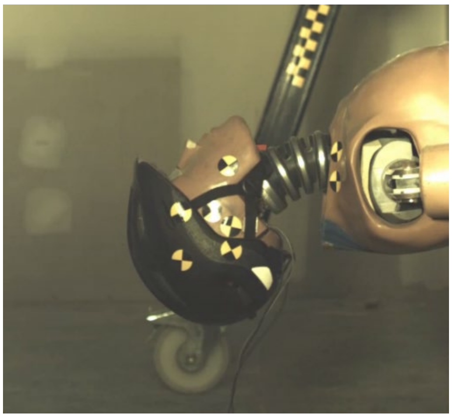

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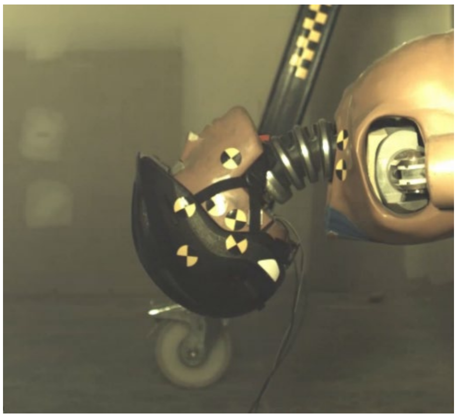

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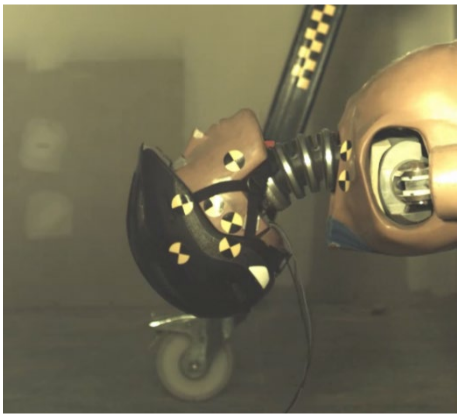

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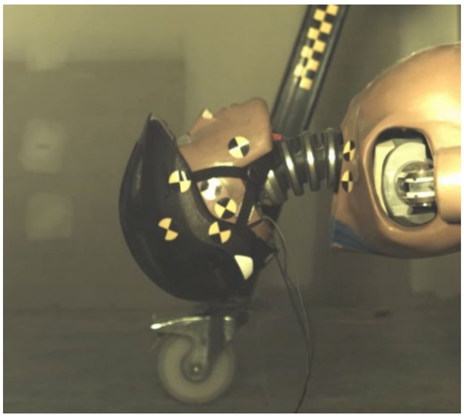
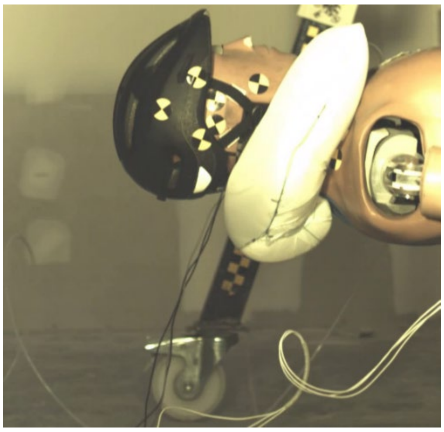


T = 120 ms

Fig. A5. Head kinematics during the test EVIX 26 without any airbag. Video stills every 20 ms.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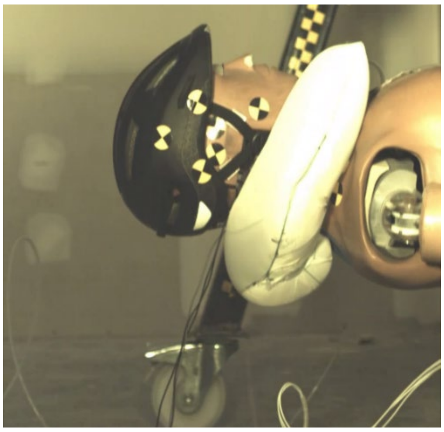
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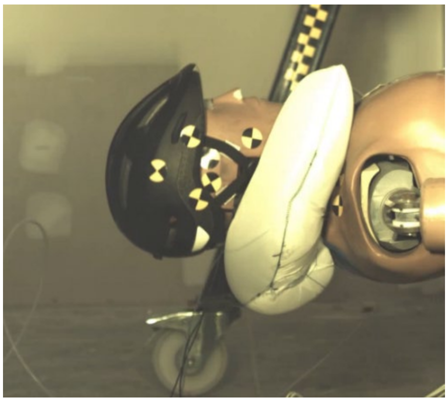
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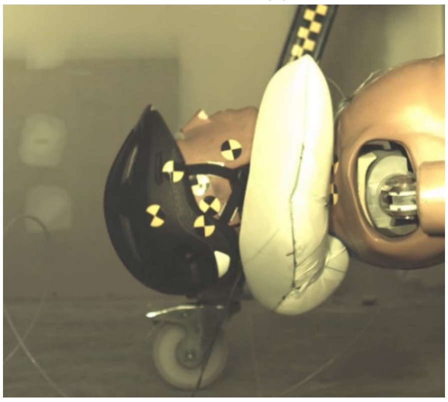
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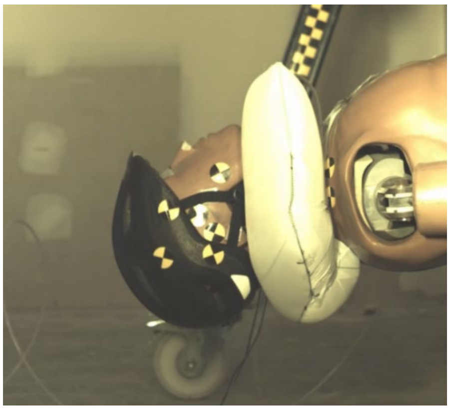
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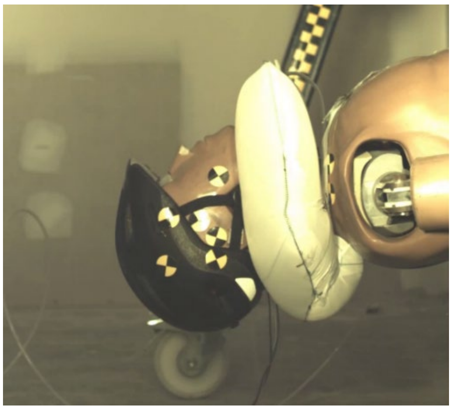
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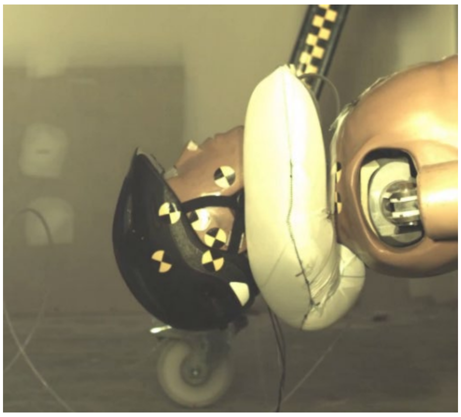
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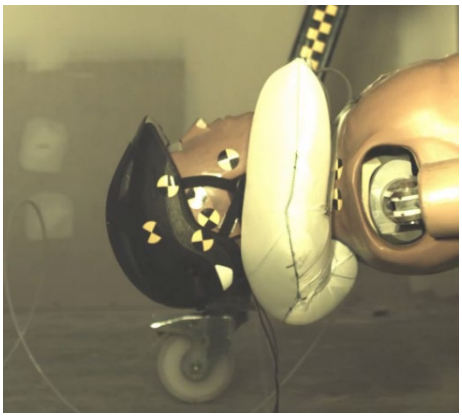
T = 120 ms

Fig. A6. Head kinematics during the test EVIX 14 with Airbag 1. Video stills every 20 ms.

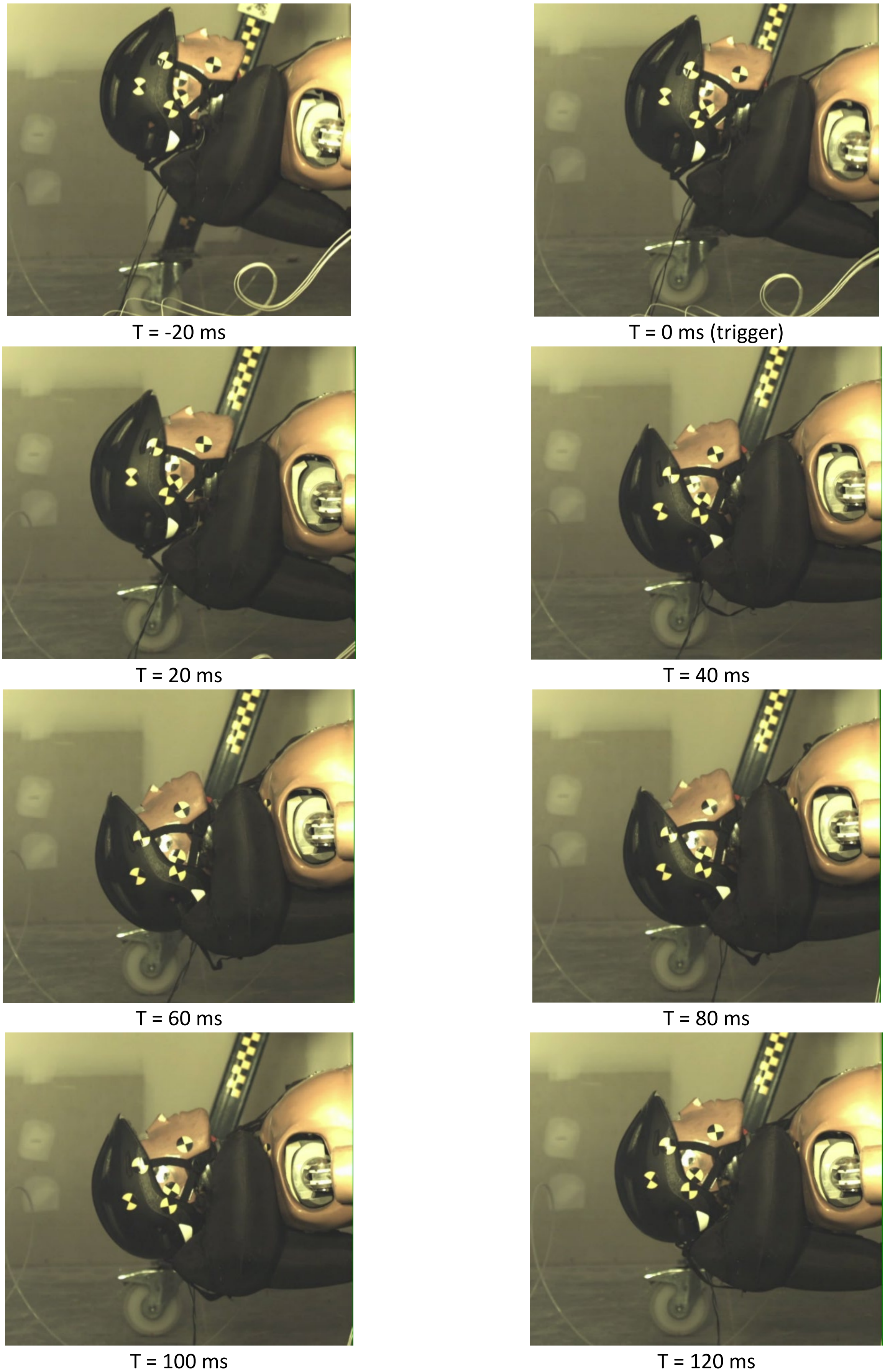

Fig. A7. Head kinematics during the test EVIX 08 with Airbag 2. Video stills every 20 ms.


T = -20 ms


T = 0 ms (trigger)


T = 20 ms


T = 40 ms


T = 60 ms


T = 80 ms


T = 100 ms


T = 120 ms

Fig. A8. Head kinematics during the test EVIX 20 with Airbag 3. Video stills every 20 ms.

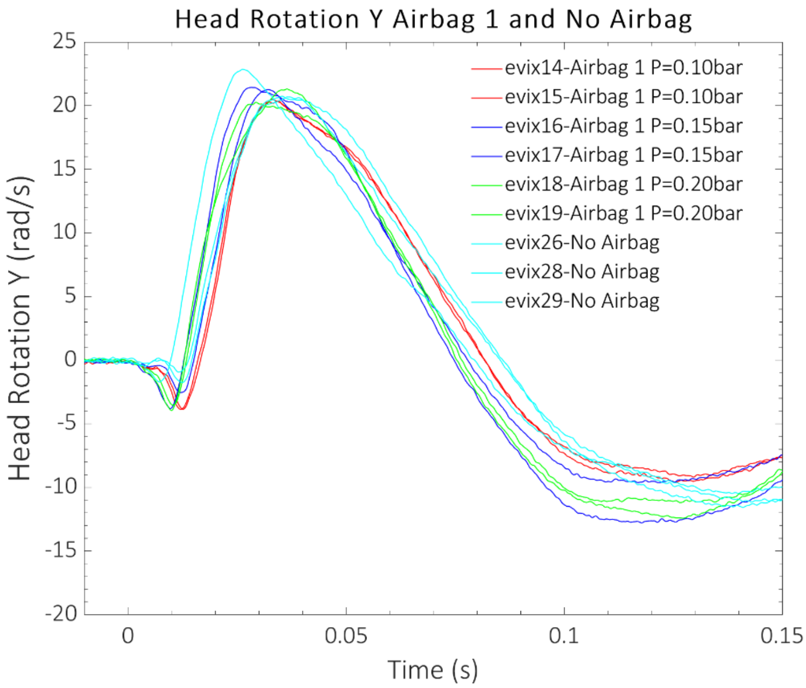

Fig. A9. Head angular velocity for tests with Airbag 1 (red, dark blue, and green) and without airbag (light blue).

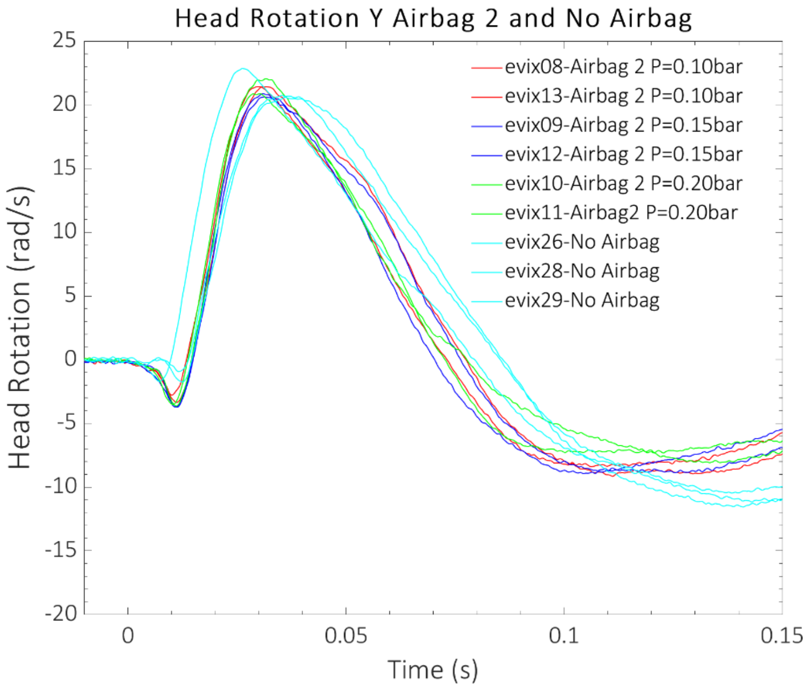

Fig. A10. Head angular velocity for tests with Airbag 2 (red, dark blue, and green) and without airbag (light blue).

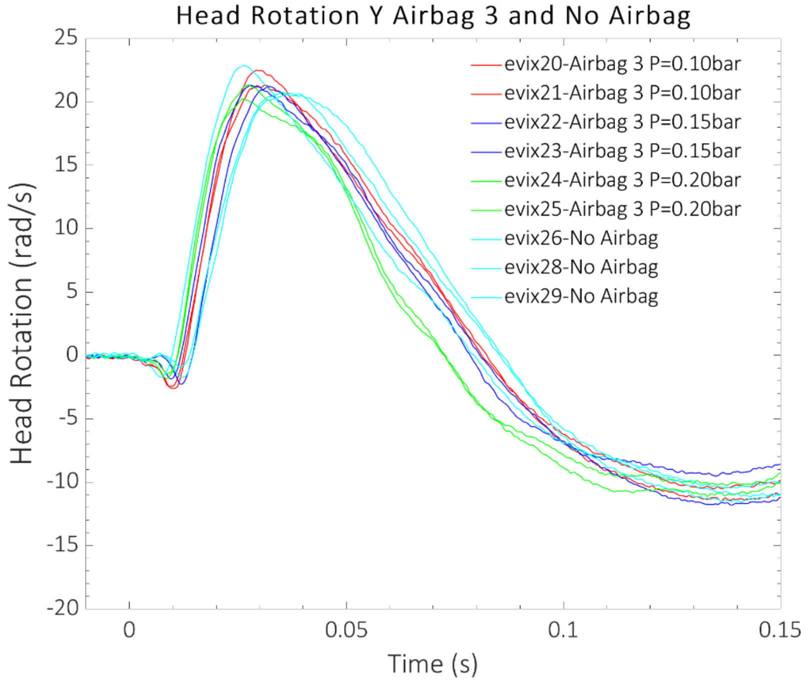

Fig. A11. Head angular velocity for tests with Airbag 3 (red, dark blue, and green) and without airbag (light blue).