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UNIVERSIDAD PONTIFICIA
ICAI

MASTER OF ENGINEERING IN MOBILITY AND SAFETY

MASTER'S THESIS

Experimental testing and data analysis of cervical airbags used in bicycle helmets

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Supervisor: Francisco José López Valdés

Madrid

17 of July 2023

Declaro, bajo mi responsabilidad, que el Proyecto presentado con el título

Experimental testing and data analysis of cervical airbags used in bicycle helmets

en la ETS de Ingeniería - ICAI de la Universidad Pontificia Comillas en el

curso académico **2022/23** es de mi autoría, original e inédito y

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El Proyecto no es plagio de otro, ni total ni parcialmente y la información que ha sido

tomada de otros documentos está debidamente referenciada.

Fdo.: Jaime Álvarez Fernández Fecha: 24/ 07/ 23

Jaime Álvarez

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EL DIRECTOR DEL PROYECTO

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Chapter 1. INTRODUCTION

According to the Global Road Safety report performed by the World Health Organization (WHO), over 1.35 million people die each year in road traffic crashes on public roads. Road traffic collisions are therefore the eighth leading cause of death worldwide, and the first cause in groups of age comprehended between 5-29 years. In terms of percentage this means that about 2.5% of annual deaths worldwide are road-related. The WHO reports that over 26% of the total annual deaths due to road traffic crashes have involved cyclists or pedestrians. As a consequence, the WHO places pedestrians, cyclists, and motorcyclists as one of the most vulnerable groups, since they represent over 50% of the annual deaths on public roads. In addition, the WHO reported that over 86% of the bicycle trips were carried out in public roads that had not been conditioned for such use, leading to a considerable increase in the risk for the users [1].

The European Commission produced a report in 2021 entitled "European Road Safety Observatory", which dealt with Facts and Figures for bicycles. The report suggested that, although cycling is becoming more and more popular, the bicycle is the only vehicle that has not shown any significant improvements in safety measures to reduce fatalities in the last thirteen years. This research stated that most of the fatalities in Europe were of people over the age of 65 and happened in urban roads. Northern countries showed the highest number of fatalities due to the wide use of bicycles as a form of transportation [1].

The National Highway Traffic Safety Administration (NHTSA) stated that in 2020 there were 938 fatalities among bicycle users, which represented 2.4% of the total fatalities reported in the United States in road traffic crashes. In addition to these fatalities, around 38,900 people were registered in hospitals as traffic-related victims. The NHTSA observed that the number of males implicated in road traffic crashes was significantly higher than the number of females [3].

The figures presented show a clear vulnerability of cyclists and expose the need to improve road safety for these road users. The data gathered presents the urge to upgrade the infrastructure to guarantee safe conditions for cycling, as well as the mandatory and regulated use of helmets on public roads.

Nowadays, the helmet is only used in urban roads, where 33% of road traffic crashes occur [4]. This issue needs to be dealt with efficiently and quickly; cyclists may suffer severe consequences, sustaining serious injuries to the regions of the neck and head that might lead to death in the worst-case scenario. Specifically, three out of four deaths on public roads are due to injuries to the neck or head regions [5].

In 2013, the Spanish insurance company MAPFRE reported in the study "*Cyclist: Helmets and head injuries*" [5] that injuries to the head could be reduced between 63% and 88% just by using helmets. Nevertheless, the study also suggested that the use of helmets was only effective in the upper and middle regions of the head, leaving the lower head and the neck vulnerable to severe traumas. The study also stated that helmet use could provide similar protection to the rider's head regardless of whether only the bike itself or motor vehicles were involved in the crash.

The lack of protection in the regions of the neck and lower part of the head is consistent with what was observed by Naess in 2020 [6], which stated that two thirds of road users involved in traffic crashes, who were riding bicycles and were registered in a hospital, had injuries in said regions. In addition, one third of the cyclists sustained severe head and neck injuries. Therefore, to decrease these types of injuries, it is necessary to search for passive safety elements that guarantee reductions in the severity of injuries in the neck and lower region of the head.

Airbags are being incorporated into helmets to address this issue. They have been claimed to limit both linear and angular accelerations of the head, thus improving head kinematics. This involves significant reductions in head and neck injuries and achieves significant improvements in the values obtained for the Head Injury Criterion (HIC) [7].

Chapter 2. STATE OF ART

This chapter explains basic concepts regarding injury biomechanics and presents a more detail review of the literature and the causes of the hyperextension of the neck during cycling. The first section of the chapter gives a brief description of what is injury biomechanics and what injuries criteria are available to predict the possible injuries suffer by a cyclist in the head and neck regions. The second section explains the possible injuries that cyclist may face during road traffic crashes and presents the available data regarding the effectiveness of airbag in helmets for cyclists.

Chapter 2.1. INJURY BIOMECHANICS

Injury biomechanics is a specific discipline focussed on the research of mechanical forces and their consequences on the human body during traumatic events such as sport injuries or road traffic crashes. The aim of injury biomechanics is to understand the behaviour of the human body structure and the possible different injuries suffer due to the different forces applied. By researching the physiological behaviour of human body, this discipline is able to prevent different injuries, determine the severity of the injuries, design safety measures in cars, or create new protection gear.

This section is going to study several aspects of injury biomechanics needed for a better understanding of the master thesis. The discussion will cover several injury criteria to provide a better understanding of the risk and severity of injuries suffer by cyclist. The different injuries criteria will help assess the severity, likelihood and the forces implicated during a bicycle crash. The section will focus on the concept of injury criterion and the Abbreviated Injury Scale, the Neck Injury Criterion, the Head Injury Criterion and the Brain Injury Criterion. The understanding of these concepts and criteria will provide the premises for evaluating the effectiveness of airbag helmets in preventing the hyperextension of the neck which is the objective of the study.

INJURY CRITERION

Injury criterion is quantified measure that helps assess the likelihood and severity if the different possible injuries sustained by a subject that suffers a traumatic event. It provides a normalized measure of the different injuries that a subjects may hold based on the forces, displacements, deformations and bending moments withstand. “Injury criteria have been developed in terms that address the mechanical responses of crash test dummies in terms of risk to life or injury to a living human. They are based on an engineering principle that states that the internal responses of a mechanical structure, no matter how big or small, or from what material it is composed, are uniquely governed by the structure’s geometric and material properties and the forces and motions applied to its surface. The criteria have been derived from experimental efforts using human surrogates where both measurable engineering parameters and injury consequences are observed and the most meaningful relationships between forces/motions and resulting injuries are determined using statistical techniques" [8].

As previously mentioned, the more relevant injury criteria for this study are:

ABBREVIATED INJURY SCALE

The Abbreviated Injury Scale (AIS) is a scoring system widely used around the world to determine the severity of the injuries that a patient may sustain. The AIS was developed by the American Association for Automobile Medicine (AAAM) in the 1960s to assess the severity of the injuries that occur in a road traffic crash and to classify them [9].

The AIS system gives a score comprehended between 1 (minor injury) and 6 (critical severity). Initially, the aim of the AIS was to determine the severity of each of the injuries a subject may incur during a car crash individually, rather than the overall outcome of the multiple injuries that might be obtained. Nowadays, the AIS has become a widely used scoring system that provides standardized terminology to describe the types and severities of the injuries, available for any type of events.

The AIS has undergone several revisions over the years allowing a more complete version that classifies the injuries by the body region and severity. The last version is from 2015. In order to be able to assess the codification correspondent to each of the possible injuries incurred the users must fulfil a course to learn the procedures and techniques applied.

Some of the applications that the AIS provides is the evaluation of the different injuries that a patient might face and their severity, and the determination of the correct treatment and possible outcome. The use of this system is applied both at the clinical level in hospitals and at the academic level in research studies.

NECK INJURY CRITERION

The Neck Injury Criterion (NIC) is used to determine the likelihood of incurring on a neck injury during an impact where any type of force is applied to that region. The NIC was developed in the 1970s by General Motors and the Insurance Institute for Highway Safety (IIHS).

The method for evaluating the probability of suffering a neck injury is based on the study of the accelerations and displacements of the neck. The NIC covers, on one hand, the displacement and maximum rotation that could help assess damage in the soft tissue and spinal cord injuries and, on the other hand, the axial, shear and bending moments that could allow to predict the type of injury and the overall dynamic motion of the area [10].

The NIC is calculated as:

$$NIC\ (t) = a_{rel}\ (t) \times 0.2 + [v_{rel}]^2\ , \quad (1)$$

$$a_{rel}\ (t) = a_x^{T1} - a_x^{Head}\ , \quad (2)$$

$$v_{rel}(t) = \int a_{rel}\ (t)\ , \quad (3)$$

Where a_x^{T1} is the acceleration in the X direction on the first thorax spine and a_x^{Head} is the acceleration at the Centre of Gravity (CG) of the head [11]. The data usually is filtered with SAE J211 CFC 1000 for accelerations and velocities according to FMVSS regulation from NHTSA from 2008 [12].

The associated risk curve of the NIC is:

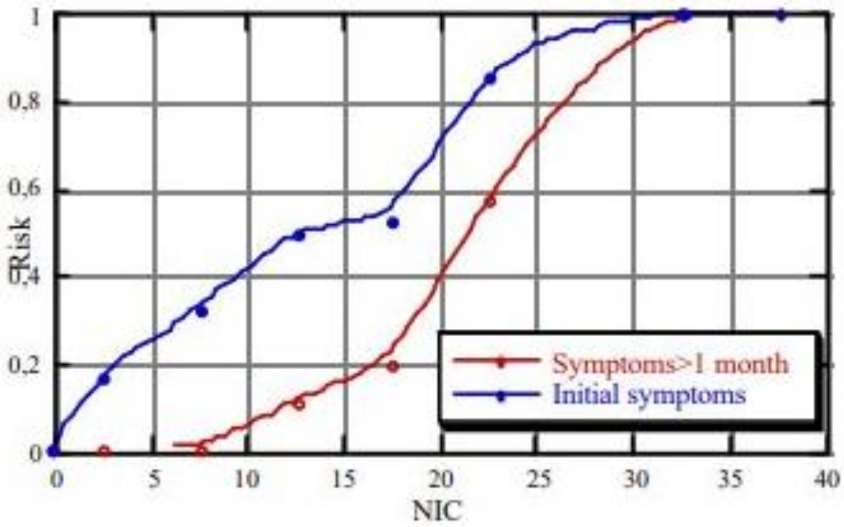

Figure 1. Associated risk function for NIC [13]

While it is possible to apply the NIC to Post-Morten Human Surrogates (PMHS) testing the use of this criterion is most commonly applied to Anthropometric Test Devices (ATDs) testing. The use of the NIC is widely spread among medical and forensic settings, in automotive safety research and in regulatory testing institutions such as Euro NCAP.

HEAD INJURY CRITERION

The Head Injury Criterion (HIC) is designed to help assess the risk of inducing a head injury on an occupant that have been involved in a crash. The HIC was developed in the 1970s when researchers realised that further investigations were required for a better understanding of the intrinsic risk of the head during a crash [14]. The criterion was developed based on a previous Criterion name Gadd Severity Index (GSI) that was formulated with data from skulls fractures.

The HIC is based on the study of the accelerations of the head to evaluate the likelihood of inducing a head injury. This criterion has undergone several revisions over the years given in each of the new versions a more complete and clarifying understanding of head injuries and the probability of suffering one.

The HIC is calculated as:

$$HIC = \max\left\{(t_2 - t_1)\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)\, dt\right]^{2.5}\right\}, \quad (4)$$

The parameters t1 and t2 represent the interval of time thar is going to be studied. The critical HIC value should use intervals of time under 15ms to ensure precision. The parameter a (t) represents the result acceleration of the center of mass of the head [11]. The data is filter with SAE J211 CFC 1000. The associated risk function is as follows:

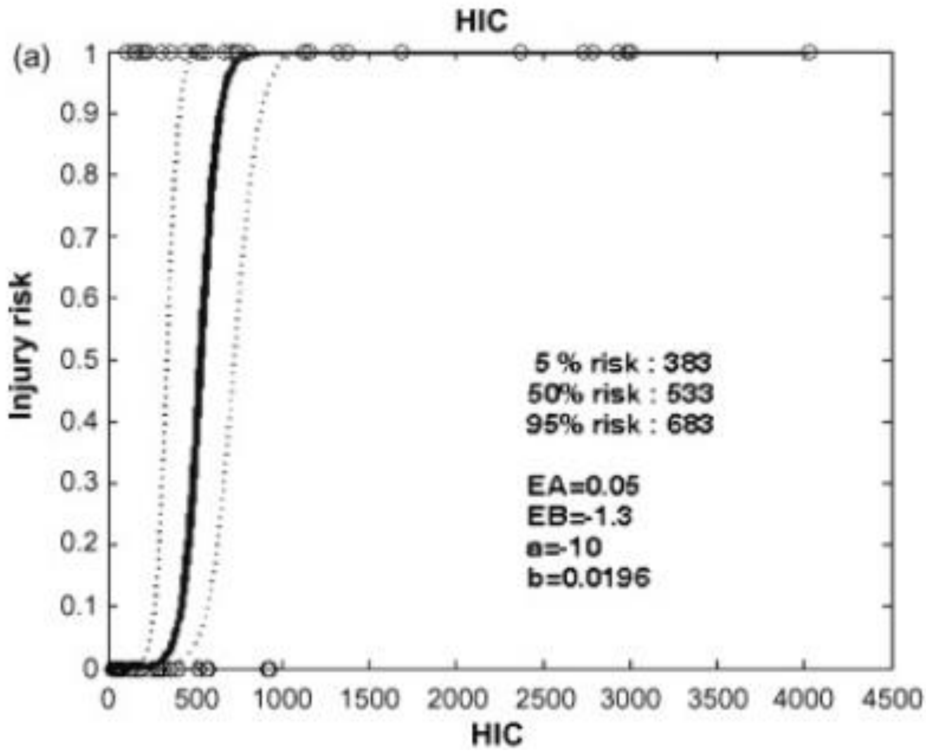

Figure 2. Associated risk function for HIC [15]

The HIC is used to evaluate the effectiveness of prototypes, airbags restraint systems, etc. This allows engineers to redesign new measures to improve safety in the region of the head during a road traffic crash. Several regulations and standards incorporate the HIC to evaluate and regulate the head injuries, such as the Federal Motor Vehicle Safety Standard (FMVSS) by the NHTSA, or the European Union ECE R21 established by United Nations Economic

Commission for Europe (UNECE). The HIC is also well established among the research safety environment and in regulatory testing in institutions such as Euro-NCAP.

BRAIN INJURY CRITERION

The Brain Injury Criterion (BrIC) was designed to help assess the severity of the brain injuries due to a road traffic crash. The BrIC was designed in the 1940s with the contributions of several researchers and organizations from the field of biomechanics related to road safety. The mayor contributions and most significant development were conducted by the National Highway Traffic Safety Administration (NHTSA) and the Society of Automotive Engineers (SAE); therefore, the development of this criterion cannot be attributed to a single person or collective [16].

The BrIC study the influence of the head rotation in the occurrence of a brain injury. The initial studies were developed with animal testing to confirm or reject the first hypothesis.

The BrIC is calculated as:

$$BrIC = \sqrt{\left(\frac{w_x}{w_{xc}}\right)^2 + \left(\frac{w_y}{w_{yc}}\right)^2 + \left(\frac{w_z}{w_{zc}}\right)^2} \, , \qquad (5)$$

Where W_x, W_y and W_z are the maximum angular velocities and W_{xc}, W_{yc} and W_{zc} represent the critical angular velocities for each direction. Bet The associated risk function is as follows:

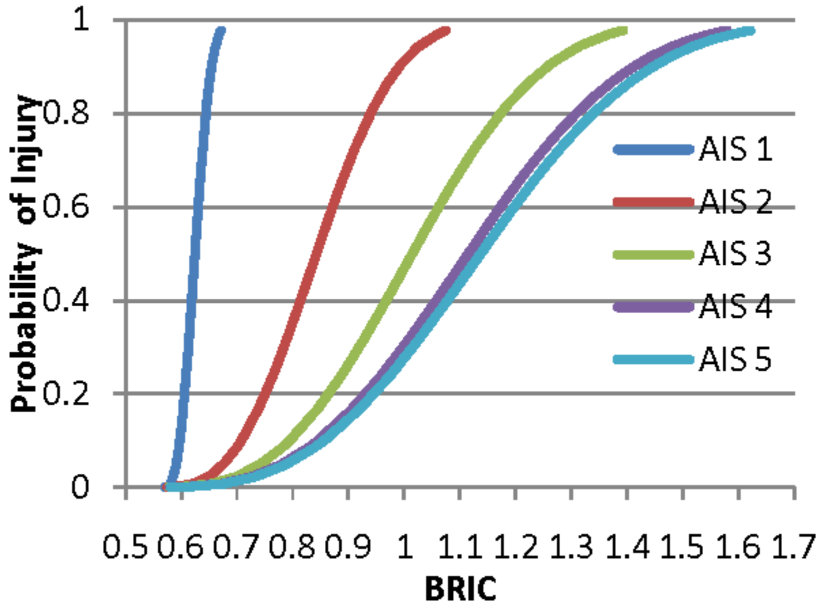

Figure 3. Associated risk function for Brain Injury Criterion [16]

The main application of the BrIC is related to automotive safety helping design and testing new helmet prototypes or implement new safety regulations for a better protection against brain injuries, although it´s use is not very stablished among safety regulations protocols.

Chapter 2.2. LITERATURE REVIEW

The improvement in road safety for cyclists is an unresolved issue that causes many deaths and serious injuries every day. Several studies have been conducted with the objective of analysing the most common injuries that can be found in road traffic crashes that involved cyclists, and determining the type and severity of the injuries that cause death in this vulnerable segment of population. Many studies reported that the most common injuries depended to a large extent on the level of protection that the user had incorporated, and the vehicles associated to the crash. Emmanuelle Amoros stated in the report "*The injury epidemiology of cyclists based on a road trauma registry*" of 2011 that the type and severity of the injuries presented by cyclists were strictly related to the characteristics of the crash. Amoros linked the severity of the injuries and the increase of injuries to the internal organs to the involvement of motor vehicles in the crash [17].

INJURIES IN CYCLISTS

The most common injuries were found in two regions, in the head and in the cervical spine. With respect to the totality of injuries sustained during road traffic crashes involving cyclists, injuries in the head area accounted for 19.1% in cyclists that used a helmet and 47.6% for cyclists that did not use it [18].

Head injuries could be found for cases in which cyclists did not wear any type of protection, which were usually of a serious nature and would lead, in many cases, to permanent injuries or, in the worst cases, to the death of the cyclist. In order to determine the severity of the injuries produced, most studies used an anatomically-based injury severity scoring system. This system is known as the Abbreviated injury scale (AIS) system. The study "*Associations between helmet use and brain injuries amongst injured pedal- and motor-cyclist: A case series analysis of trauma centre presentation*" from 2013 reported that the use of a bicycle helmet could reduce the occurrence of AIS 2+ injuries from 60% to 34% [19]. This reduction in the risk of incurring AIS 2 + injury reflects the need to legislate the mandatory use of bicycle helmets as a vital measure in the fight to reduce severe injuries and fatalities in bicycle traffic road crashes.

Limited studies have been conducted to analyse cervical spine injuries related to cyclists [20]. Traditionally, neck sprains, hyperextensions of the neck and cervical fractures have been associated with American football, but sports such as cycling have seen their numbers increased more and more in the latest years. Thus, cycling has become the second cause of neck sprains and the first one for cervical spine fractures for men, and the second cause of cervical fractures for women. Several studies reflect a slight increase in the incidence of cervical spine fractures from 6.5 per million in 2000 to 8.8 per million in 2015 for all type of sports. This has also been the case in cycling, where the increase has gone from 0.67 per million in 2000 up to 2.7 per million in 2015, corresponding to a 400% increase. Furthermore, reports from Ireland, France, and Australia present evidence of a 200% increase in the number of traumatic spine injuries from 2000 to 2013, of which up to 70% of them corresponded to cervical spine injuries [21].

Injuries related to the upper cervical spine area appear when the head withstands forces during the crash that are transmitted through the neck to the rest of the upper body. The most typical cervical spine injuries are the C6/C7 fracture, the occipital condyle fracture and C5/C6 fracture. The occipital condyle fracture is usually related to the rotation and compression in the C0/C1 joint, which usually happens during the first stages of the collision [22], Which may lead to the hyperextension of the cervical spine.

Two studies performed in 2002 and 2007 by Roger W. Nightingale [23][24] analysed the average bending moment that a cervical spine could withstand during the hyperextension phase of the neck of a human surrogate. The study suggested that females had a higher tolerance than males in the hyperextension of the upper cervical spine (O-C2) due to a pure bending moment in the sagittal plane. The average angles that males could resist were 42.4º with a standard deviation up to 8º while females could withstand angles up to 50.2º with a standard deviation of 11.4º. Nevertheless, the methodology used during these studies for calculating the hyperextension angles differ from the use in our experiments, therefore these experiments can't be contrasted against our study.

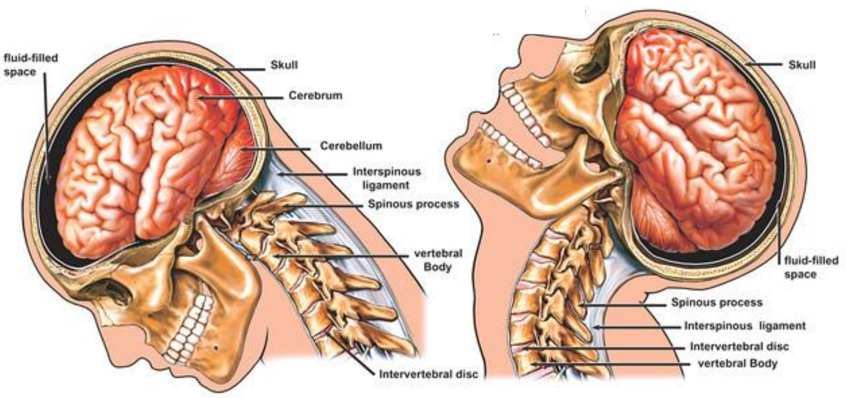

Figure 4. Hyperextension of the neck [25]

A study performed by Frederick P Rivara in 1997 entitled "*Epidemiology of bicycle injuries and risk factors for serious injury*" suggested that crashes where motor vehicles were implicated increased the risk of cervical injuries and the severity of the injury by 4 times.

Besides, the study also confirmed that the expectation of survival for cyclists with neck-related injuries where 15 times lower. No relation between the use of helmet and the appearance of neck injuries was found [26].

AIRBAGS FOR HELMETS

The protection against cervical spine injuries is limited, in fact several studies have focused on the influence of the helmet in the appearance of those types of injuries. Different conclusions were reached in these studies. Some studies like "*Adult Bicycle Collisions: Impact of Helmet Use on Head and Cervical Spine Injury*" from Vyas from 2020 [27] stated that the helmets increased the risk of incurring on a cervical spine injury; however, others maintained that the results were statistically non-significant and that the effectiveness of helmets in reducing the risk of suffering cervical spine injuries could not be proven [28][29][30]. Nevertheless, all the studies reached the same conclusion that the incorporation of airbag systems to protect the area of the cervical spine was needed for better protection.

Limited studies have been undergone using systems that incorporated airbags for the protection of the neck. The study "*The Influence of Vehicle Low Impact Velocity over the Helmet Airbag Deployment and Cyclist Injuries*", performed by Ovidiu Andrei Condrea in 2020, tested two different setups, one including the airbag prototype and the other one without it, at different velocities. The authors stated that the use of the airbag helped reduce the Neck Injury Criterion (NIC); although the effectiveness of the airbag system seemed difficult to compare due to differences between the tests such as velocity [7].

Further investigations have been conducted being able to observe that air pressure could be adapted to adjust the mechanical behaviour of the neck to the most favourable. The study "*Modeling and Optimization of Airbag Helmets for Preventing Head Injuries in Bicycling*" from 2017 by Kurt [31] suggested that an airbag helmet would be able to absorb more impact energy before engaging the impact phase and that the Hövding airbag prototype could reduce the peak linear acceleration and the HIC values compared to a standard helmet. In the paper "*Consumer Testing of Bicycle Helmets, presented at the International Research Council on* *the Biomechanics of Injury*" from 2017 Stigson [32] confirmed the results of the research performed by Kurt and reckon a decrease in the rotational acceleration of the head also.

Figure 5. Hövding Airbags prototype for cyclists [33]

It seems obvious that there is a potential reduction on cervical spine injuries but, further research is needed to fully understand the kinematics of the neck area during the hyperextension phase, determine the possible injuries that might appear relative to the cervical spine area and design new devices to reduces the impact.

Chapter 3. METHODS

The aim of this research is to evaluate the neck hyperextension and thus be able to compare the different airbag prototypes proposed by the company Evix. A structure is designed to develop this sequence of experiments and guaranteed a repetitive process easy to compare. The experiment used an Anthropometric Test Device (ATD) also known as crash test dummy (or dummy, for simplicity) for the set of tests proposed. The tests used the Hybrid dummy III 50th percentile male and although it is an ATD intended for frontal impact car collisions tests, since we are interested in a very specific region of the body, it is valid as a first approximation. Although the purpose of the dummy is frontal impacts and some rear impacts crashes where the neck of the dummy produces a motion in the sagittal plane, due to the similar characteristics in motion expected the Hybrid III is chosen for the execution of the tests. However, it must be taken into account that the similarities of the neck assembly of the dummy presents limitations when comparing with a human surrogate. Therefore, fit of the airbags was adjusted a close as possible to a human surrogate in order to obtain biofidelic results. It must be also considered that the neck assembly of the dummy presents a higher rigid structure than the humans.

Chapter 3.1. TEST MATRIX

The test matrix of the experiments carried out in this work is shown below::

Table 1. Tests matrix

Airbag use		Test number	Pressure in airbag (bar)
	*Figure 6. No airbag*	26	-
		28	-
		29	-
	*Figure 7. Airbag 1*	14	0.1
		15	0.1
		16	0.15
		17	0.15
		18	0.2
		19	0.2
	*Figure 8. Airbag 2*	8	0.1
		13	0.1
		9	0.15
		12	0.15
		10	0.2
		11	0.2
	*Figure 9. Airbag 3*	20	0.1
		21	0.1
		22	0.15
		23	0.15
		24	0.2
		25	0.2

Chapter 3.2. TEST SETUP AND TEST FIXTURE

The experiment takes as its axes the following coordinate system:

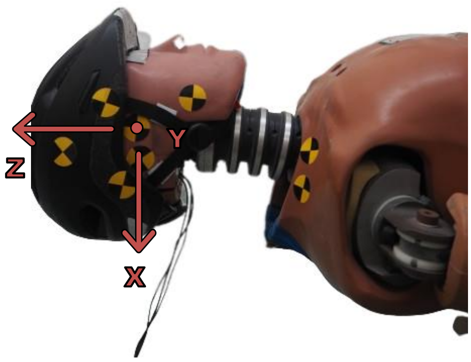

Figure 10. Coordinate system

The initial structure designed consisted of a set of aluminium bars that formed the frame along which wooden board slid guided, allowing only a translation in the x axis *Figure 11. Original structure 3x1x2m*. Due to the intensity of the test and the slenderness of the structure, the downward motion produced vibrations in the steel bars, for which reason the structure had to be stiffened by means of bracing *Figure 13*. To avoid excessive buckling of the table, longitudinal steel reinforcements and transverse wooden reinforcements were placed. These reinforcements serve two purposes, on the one hand they helped to stiffen the board and on the other hand they helped to get the board to go down simultaneously and at the same speed on all four rails, thus avoiding the possibility of getting stuck.

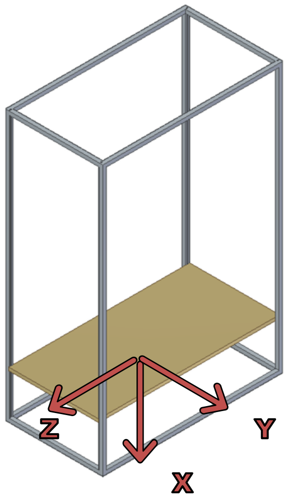

Figure 11. Original structure 3x1x2m

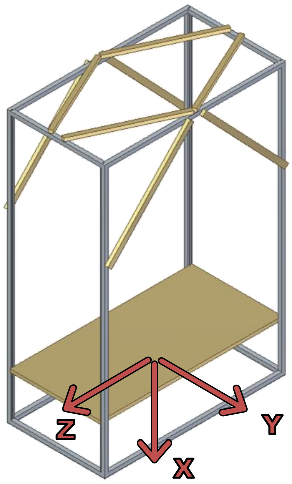

Figure 12. Final structure 3x1x2m

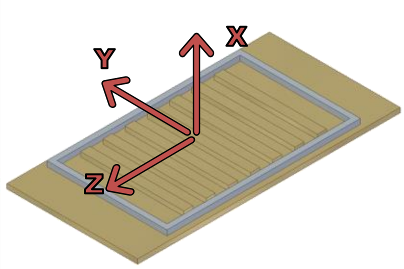

Figure 13. Reinforcements under the wood board 1x2m

The tests were carried out at a height of 50cm above the impact surface, from where it was released by means of an electromagnet and left in free fall. At the time of the impact the lower part of the wooden board hit a surface of concrete blocks that were placed high enough to allow the hyperextension of the neck. The dummy initial position was kept constant during the different test performed with the help of a lashing strap. The dummy was resting on the wooden table and centered in the Y direction defined for the tests.

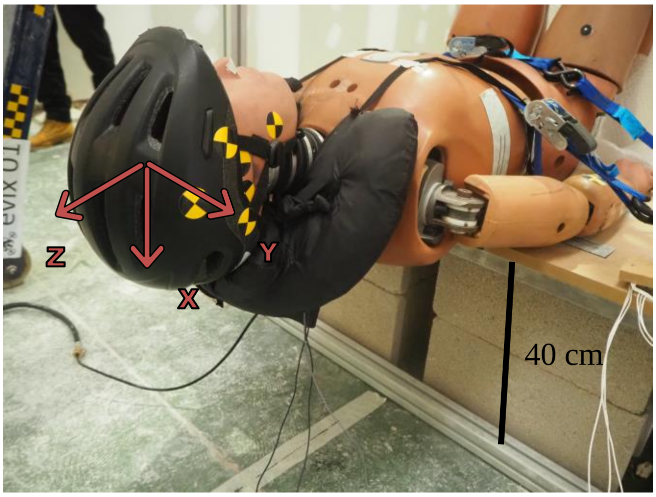

Figure 14. Test set up

The set of tests performed consisted of 21 tests under the same environment, varying only the model of airbag used and the pressure configuration. The airbags were pressurized under three different values: .10 bar, 0.15 bar and 0.20 bar. For each of the different airbags and pressures two test were performed. The airbags were inflated before the start of the test and kept pressurized to the target value during the execution of the test. The helmet used during the experiment corresponds to a standard helmet available in the market.

Chapter 3.3. TEST INSTRUMENTATION

The tests were recorded with a high-speed camera placed to visualise de complete motion of the hyperextension of the neck. The tests were recorded at a frequency of 1000 Hz. Photo targets were placed in the dummy's head and helmet. These photo targets allowed tracking the motion followed by the region of interest and to assess the effect of the airbags incorporated. The photo targets were used during the tracking analysis of the tests that were performed without airbag and with the airbags 1 and 2. The tracking of the motion of the dummy for the airbag 3 was performed calculating the relative angles of the nose between the frame prior to the impact with the rigid surface and at the frame with the highest hyperextension of the neck due to lack of visibility of the photo targets. The hyperextension angle of the neck was calculated as the difference between the average initial angle of the tests without airbag and the angle of the moment of maximum rotation of the head. The maximum, minimum and average values were calculated for each of the tests performed.

The kinematics of the head in the sagittal plane were recorded by linear acetometers at a frequency of 10,000 Hz. The accelerometers measure the acceleration in the X and Z directions. An Angular Rate Sensor (ARS) measured the rotation of the head in the Y direction. The sensor was located in the centre of gravity of the head. All the signals obtained were filtered with a CFC 300 class filter.

Chapter 3.4. Data analysis

Head kinematics and neck hyperextension were analysed to determine the effect of the different types of airbags at the three possible target pressures imposed. The Head Injury Criterion and Brain Injury Criterion were also calculated and analysed to evaluate the behaviour of the dummy with or without airbag.

Chapter 4. RESULTS

LINEAR ACCELERATION OF THE HEAD

The linear acceleration of the head in the X direction is presented in *Figures 15-17* classified by the different pressure levels that were targeted. The results for airbags 1 and 2 were slightly higher than airbag 3 and the baseline (Without airbag) in their maximums regardless of the pressure configuration selected. Nevertheless, it was observed that for the pressure 0.15 bar the maximum peaks values of airbags 1 and 2 were minimized. In the case of the airbag 1, no influence was observed for the minimum's peak values regardless of the different pressure configurations proposed. Instead, in the case of airbag 2, the minimum peak was defined for a pressure of 0.15 bars at an acceleration of 6.3 g. In the case of airbag 3, a clear trend was observed in which, as the pressure increased, the acceleration obtained decreased.

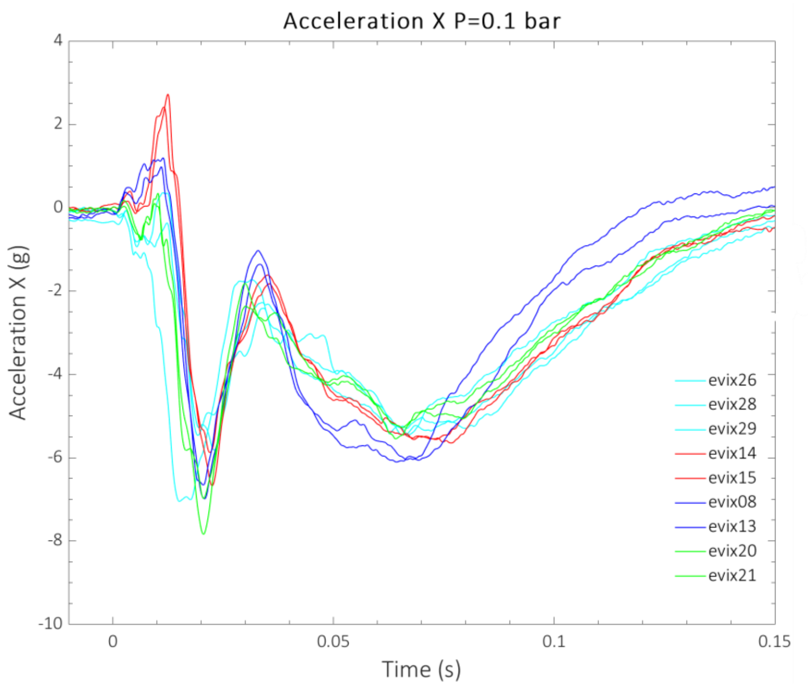

Figure 15. Head Linear Acceleration in X direction for P=0.10 bars (light blue-Without airbag, red-airbag 1, blue-airbag 2, green-airbag3)

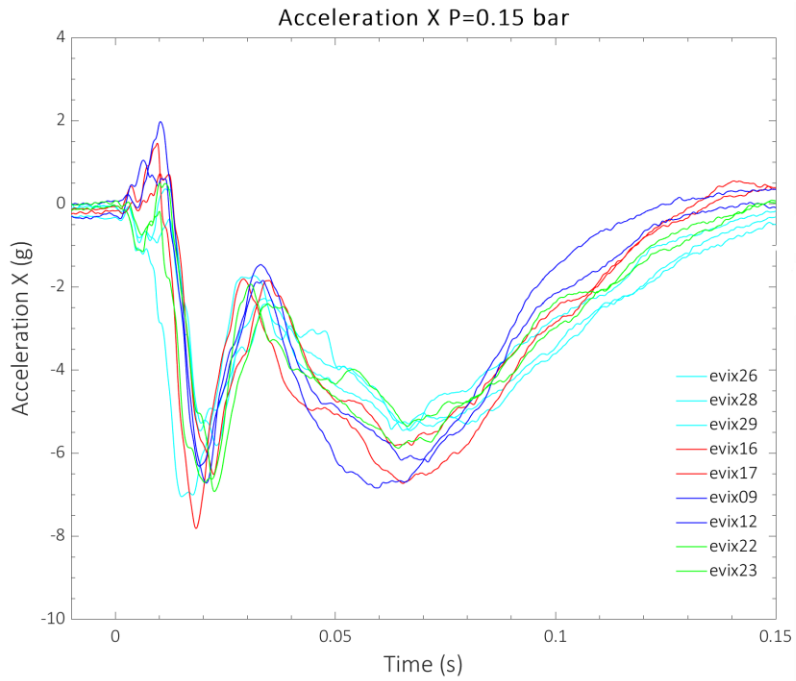

Figure 16. Head Linear Acceleration in X direction for P=0.15 bars (light blue-Without airbag, red-airbag 1, blue-airbag 2, green-airbag3)

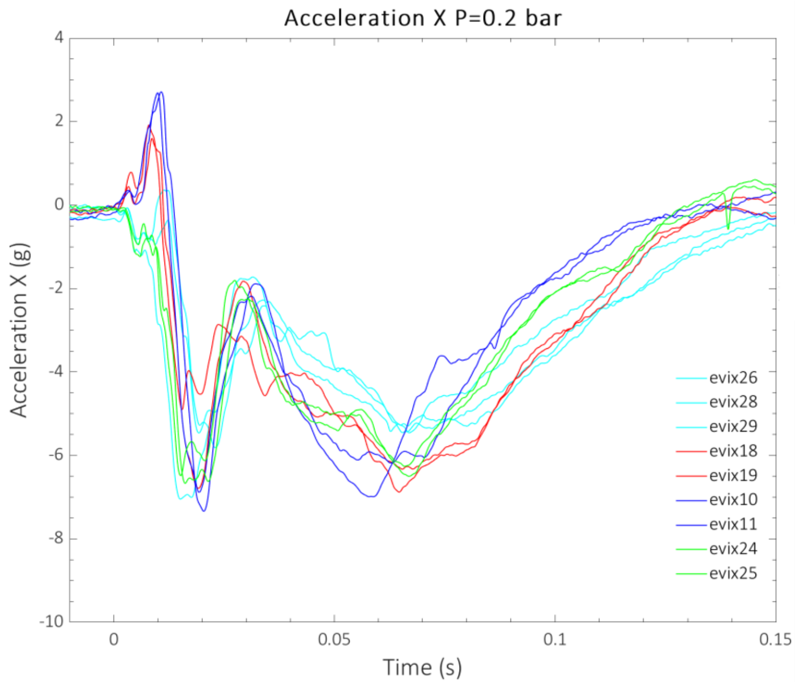

Figure 17. Head Linear Acceleration in X direction for P=0.20 bars (light blue-Without airbag, red-airbag 1, blue-airbag 2, green-airbag3)

The evaluation of the head linear acceleration for Z direction leaded to significant differences as showed in *Figures 18-20*.

The results presented for the airbag 1 and 2 showed improvements in the maximum peaks, especially in the case of the airbag 2 where a reduction over 7g can be observed.

In the case of airbag 1 and 2 the minimum peaks did not reveal any improvement. However, it is observed that the effect of the airbag 3 is not noticeable for the z-axis, since the values for the maximum and minimum peaks obtained were higher than the baseline.

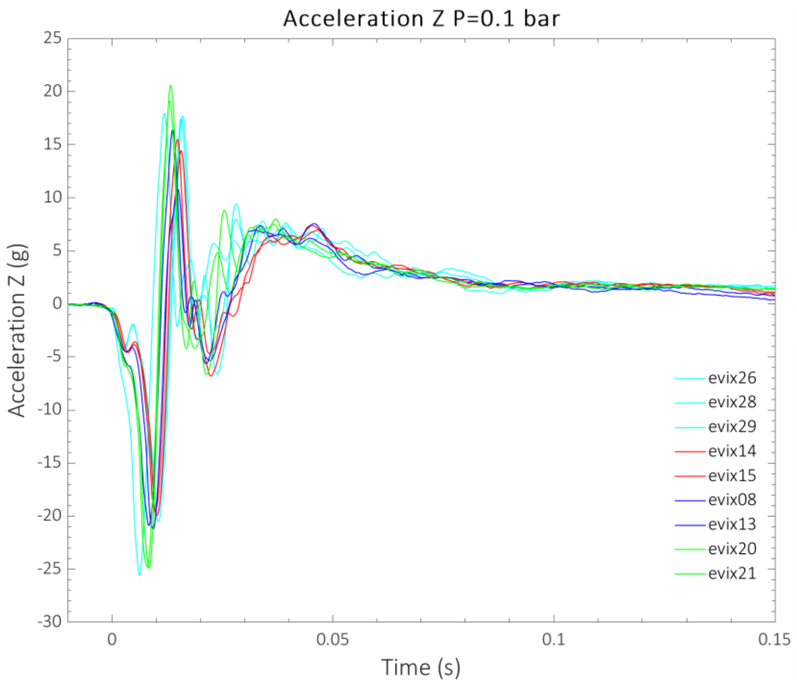

Figure 18. Head Linear Acceleration in Z direction for P=0.10 bars (light blue-Without airbag, red-airbag 1, blue-airbag 2, green-airbag3)

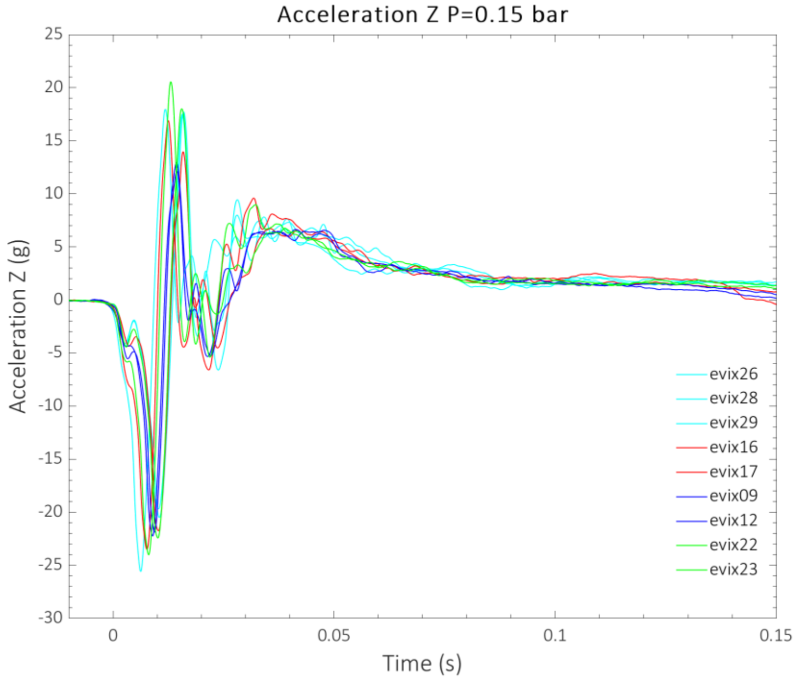

Figure 19. Head Linear Acceleration in z direction for P=0.15 bars (light blue-Without airbag, red-airbag 1, blue-airbag 2, green-airbag3)

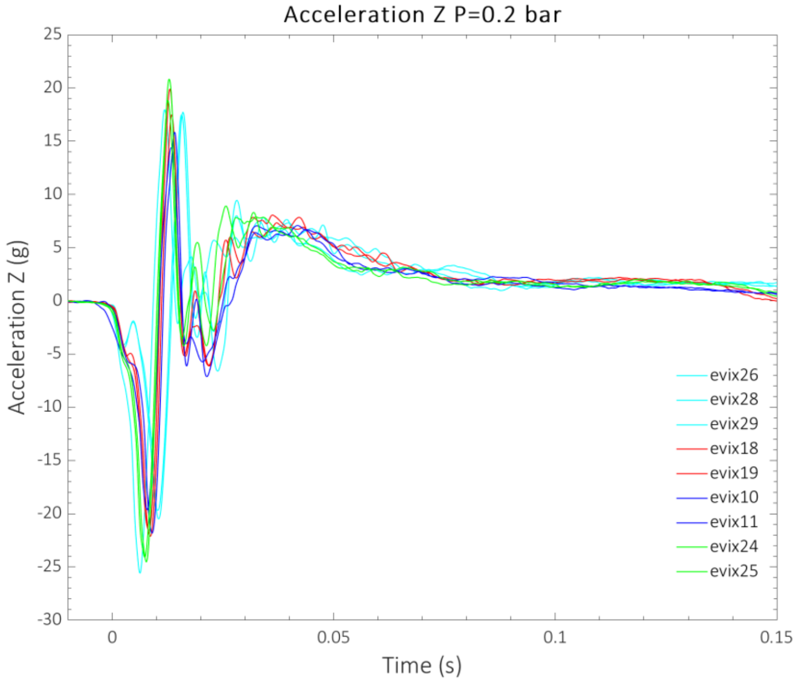

Figure 20. Head Linear Acceleration in z direction for P=0.20 bars (light blue-Without airbag, red-airbag 1, blue-airbag 2, green-airbag3)

A summary of the minimum and maximum accelerations values is shown in *Table 3*:

Table 2. Head linear acceleration peak values summary

Airbag used	Test number	Airbag pressure (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	-20.5
	29	-	0.36	-5.82	17.47	-19.69
Airbag 1	14	0.1	2.73	-6.67	14.44	-19.94
	15	0.1	2.41	-5.87	15.5	-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	0.2	1.6	-6.79	17.5	-22.13
	19	0.2	1.92	-5	19.9	-21.56
Airbag 2	8	0.1	0.98	-6.99	16.44	-20.89
	13	0.1	1.2	-6.66	10.83	-21.21
	9	0.15	1.05	-6.32	12.94	-21.9
	12	0.15	1.98	-6.72	12.14	-22.24
	10	0.2	2.69	-6.89	14.41	-19.64
	11	0.2	2.71	-7.34	15.85	-21.84
Airbag 3	20	0.1	0.35	-7.84	19.13	-24.81
	21	0.1	0.1	-6.98	20.61	-24.95
	22	0.15	0.5	-6.93	18.01	-22.43
	23	0.15	-0.02	-6.71	20.56	-24.02
	24	0.2	-0.08	-6.67	20.83	-24.55
	25	0.2	-0.01	-6.49	18.82	-24.09

ANGULAR VELOCITY OF THE HEAD

The angular velocity of the head in the Y direction was classified by the different types of airbag prototypes available for the project as shown in *Figures 21-23*. To determine the effect of the different pressure configurations the tests without airbag were kept out the figures, however the figures that include these tests in the comparison are located in the *Annex I Graphs*.

It should be noted that there was a relationship between the volume size of the airbag, the optimum pressure held by the airbag and the angular velocity obtained for the head. Airbag 1 behaviour reflected that the lowest maximum angular velocity appeared with the lowest imposed pressure configuration. As the volume of the airbag grew, it was observed that the maximum peaks gave better results with higher pressure settings. This trend is observed in both airbag 2 and airbag 3, where the lowest maximum peak values are given for pressures of 0.15 bar and 0.20 bar, respectively.

Therefore, for pressures of 0.10 bars airbag 1 presented the best behaviour while for pressures of 0.15 bars the top prototype was airbag 2. Nonetheless, for the pressure of 0.20 bars no significant improvement was found compared to the test without airbag.

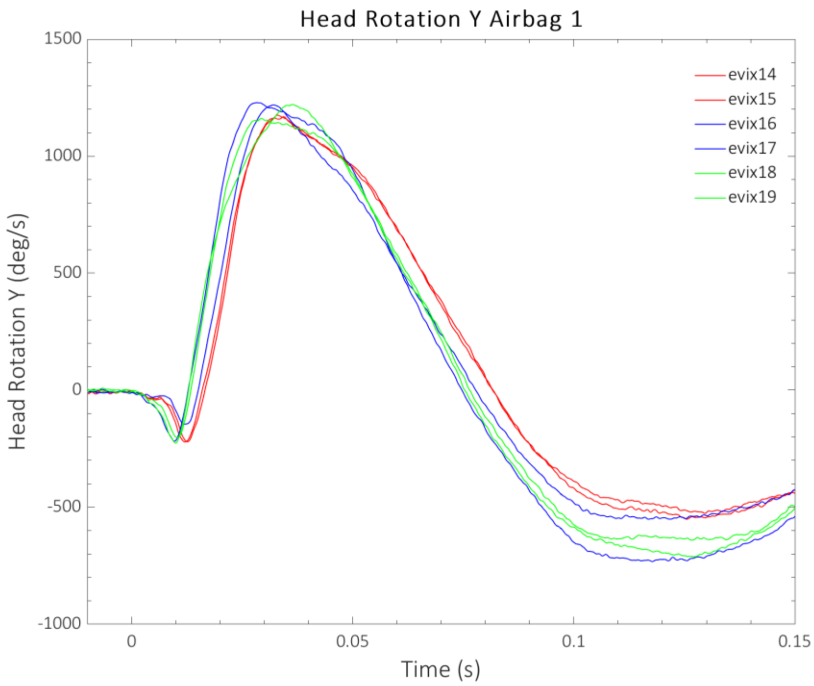

Figure 21. Angular rotation in y direction for airbag 1 (red-P=0.10bar, blue-P=0.15bar, green-P=0.20bar)

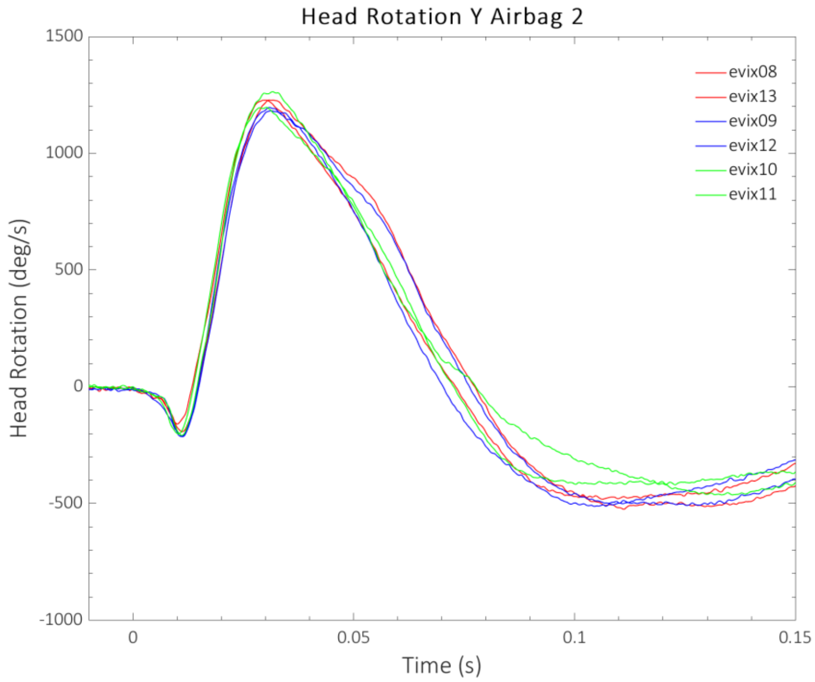

Figure 22. Angular rotation in y direction for airbag 2 (red-P=0.10bar, blue-P=0.15bar, green-P=0.20bar)

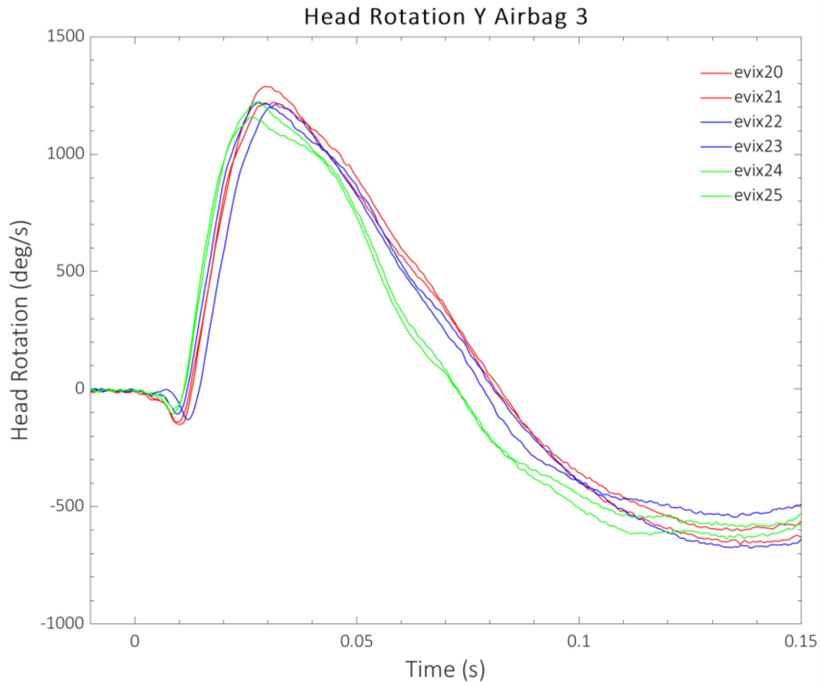

Figure 23. Angular rotation in y direction for airbag 3 (red-P=0.10bar, blue-P=0.15bar, green-P=0.20bar)

Table 4 summarizes the peak rotational speed values observed in the tests:

Table 3. Head angular rotation peak values summary

Airbag used	Test number	Airbag pressure (bar)	Maximum w_y (deg/s)
No Airbag	26	-	1310.02
	28	-	1188.93
	29	-	1180.29
Airbag 1	14	0.1	1175.42
	15	0.1	1168.63
	16	0.15	1219.38
	17	0.15	1228.78
	18	0.2	1161.47
	19	0.2	1221.19
Airbag 2	8	0.1	1227.93
	13	0.1	1228.12
	9	0.15	1196.65
	12	0.15	1182.29
	10	0.2	1196.77
	11	0.2	1263.86
Airbag 3	20	0.1	1289.16
	21	0.1	1222.23
	22	0.15	1215.97
	23	0.15	1221.01
	24	0.2	1222.66
	25	0.2	1159.42

HYPEREXTENSION ANGLES OF THE HEAD

Regarding the hyperextension angles, notable differences can be found if the tests without airbag are compared to the tests that incorporate any airbag prototype. The airbag 2 gave the best behaviour with a reduction of the hyperextension angle of 13 deg compared to the tests without airbag. The airbag 2 could reduce the hyperextension angles in a 34% . If the airbags were ordered by performance the airbag 2 would be followed by airbag 1 and airbag 3 with a reduction of 16% and 7% respectively. The results obtained presented significant improvements in terms of angle reduction, further research is needed to fully understand the behaviour of hyperextension of the neck and the implication that different air volumes may have.

Table 4 summarizes the hyperextension angles of the neck observed in the tests:

Table 4. Head hyperextension angles values summary

Airbag used	Test number	Airbag pressure (bar)	Hyperextension angle (deg)	Mean ± SD (deg)
No Airbag	26	-	48.12	50.06 ±1.73
	28	-	49.75	
	29	-	52.32	
Airbag 1	14	0.1	42.12	41.99 ±1.29
	15	0.1	42.65	
	16	0.15	39.35	
	17	0.15	43.52	
	18	0.2	41.93	
	19	0.2	42.37	
Airbag 2	8	0.1	39.76	37.20 ±2.05
	13	0.1	38.13	
	9	0.15	35.96	
	12	0.15	39.26	
	10	0.2	36.19	
	11	0.2	33.9	
Airbag 3	20	0.1	47.16	46.53 ±2.21
	21	0.1	50.11	
	22	0.15	44.11	
	23	0.15	48.31	
	24	0.2	44.16	
	25	0.2	45.31	

INJURY CRITERIA

The HIC and BrIC were calculated using equations (6) and (7) and are presented in the *Table 6. Injury Criterion values summary*. The evaluation of the HIC results reported that only the airbag 1 reduced slightly the HIC of the baseline. Despite this reduction, there was not a difference large enough to be able to determine that the results obtained were significant. The study of the BrIC stated that all the airbags reduced slightly the values of the baseline, although as the HIC no statistic significancy was found to help determine the best configuration.

The calculated HIC and BrIC values are shown in *Table 6*:

Table 5. Injury Criterion values summary

Airbag used	Test number	Airbag pressure (bar)	HIC	Mean ± SD (deg)	BrIC	Mean ± SD (deg)
No Airbag	26	-	8.31	5.71±0.01	0.405	0.379±0.018
	28	-	5.71		0.368	
	29	-	5.70		0.365	
Airbag 1	14	0.1	5.00	5.92±0.97	0.363	0.370+0.009
	15	0.1	5.06		0.361	
	16	0.15	5.17		0.377	
	17	0.15	7.17		0.38	
	18	0.2	6.30		0.359	
	19	0.2	6.83		0.378	
Airbag 2	8	0.1	5.85	5.14±0.56	0.38	0.376±0.008
	13	0.1	4.44		0.38	
	9	0.15	4.97		0.37	
	12	0.15	4.81		0.366	
	10	0.2	4.98		0.37	
	11	0.2	5.76		0.391	
Airbag 3	20	0.1	8.55	8.37±0.70	0.399	0.378±0.012
	21	0.1	8.88		0.378	
	22	0.15	7.02		0.376	
	23	0.15	8.66		0.378	
	24	0.2	8.85		0.378	
	25	0.2	8.24		0.358	

Chapter 5. DISCUSSION

Cyclists are presented as one of the most vulnerable groups on public roads with the highest risk of injuries and fatalities. Although the effectiveness of helmets has been proven in the past few years, the only significant improvements have been seen in the protection of cyclists against head and brain injuries. It has been observed how, despite the improvement in terms of reduction of deaths and severity of injuries, cyclists have continued to persevere as a high-risk group [19]. Several studies have shown that helmets alone are insufficient for the protection of the cervical spine [20], therefore the incorporation of cervical airbags might be needed to alleviate this trend. This project has studied the cervical hyperextension of the Hybrid III 50th percentile male dummy for an impact velocity of 11.3 km/h in the XZ plane (Sagittal plane). Some of the most frequent injuries sustained by cyclists in a road traffic crash are to the cervical spine [22]. As a result, the aim of this project was to evaluate the effectiveness of cervical airbags in reducing the severity of these injuries.

Regarding the head kinematics, it was noted that in general for the X direction there was no significant improvement regardless of the prototype or selected air pressure configuration. In the case of airbags 1 and 2, there was a small improvement for the pressure value of 0.15 bars, which led to lower maximum peak values for the linear acceleration than in the baseline. Nevertheless, for airbag 1 the improvement didn't represent a significant influence despite of the different air pressure configurations used. In case of airbag 2 the lowest maximum peak value was defined for 0.15 bars of air pressure, but the improvement was not relevant. Airbag 3 presented a clear tendency where, as the pressure of the airbag was increased, the a_x decreased. Nonetheless, the results were very similar to those in the baseline. All these maximum peaks happened within 0.01-0.02s of the impact, which could explain the difficulty of reducing these peak values. Due to the short period of time, the airbag prototypes were not able to absorb the amount of energy generated, therefore no important improvements were observed in terms of linear acceleration in the x direction when airbags were used.

For the Z direction the linear acceleration was improved with airbags 1 and 2. Airbag 1 presented robust results in terms of reduction of the maximum peak values of the linear acceleration for the air pressure configurations of 0.1 bars and 0.15 bars. In the case of airbag 2, the maximum peak values were improved regardless of the air pressure. However, the improvement for both prototypes were not enough to be considered as statistically significant; thus, the decrease could be explained by the better fitting of airbags 1 and 2. Airbag 3 did not present any improvements.

Past studies have also presented similar results corroborating the reduction of the head linear acceleration when using airbag systems [31-32], although the results obtained in evix's project presented higher values than expected. However, there were important differences in terms of airbag design between these studies and the current one. In the aforementioned tests airbags that covered the head, neck, and helmet were used, while in this study the airbag prototypes only covered the neck region. Further research is needed to fully understand the risks and advantages that these new cervical airbags designs may bring.

With respect to the rotation of the head, the three prototypes reduced significantly the hyperextension angles of the neck in comparison with the tests performed without airbag protection. The best performance was achieved by airbag 2, followed by airbag 1, and lastly airbag 3. Airbag 2 at 0.2 bars presented the best results with a 34.6% improvement with respect to the average values of the tests without airbag. For airbag 1 the improvement was of 19.2% when considering the average values for the tests with airbag 1 and the tests without any airbag, and the best results were obtained for 0.15 bars of pressure. In the case of airbag 3 the improvement was of 7.7%. A past study confirmed that the airbag helmet HÖVDING 2.0 could improve the angular acceleration of the head, nevertheless this study did not analyse the hyperextension of the neck[34]. With regards to the head angular speed, it was observed that the greater the volume capacity of the airbag, the higher the air pressure required to improve the performance of the airbag. Therefore, the best results were obtained for airbag 1, 2 and 3 at airbag pressures of 0.10 bars, 0.15 bars and 0.20 bars, respectively.

Regarding the injury criteria calculated, the average BrIC and HIC values were not statistically significantly different. The study performed presented small improvements in terms of reduction of the BrIC values for all airbags prototypes and for the HIC values only the airbag 2. Past studies corroborate that the incorporation of an airbag to protect the neck and head region could prevent these kinds of injuries and reduce the HIC values obtained [31]. This could also be true for cervical airbags, but further research is needed to understand the effectiveness of the different airbag designs.

Chapter 6. CONCLUSIONS

The study performed to evaluate the hyperextension of the neck of the Hybrid III 50th percentile male dummy was constituted by a set of 21 tests. During the set of tests three airbag's prototypes were analysed. For each of the airbag's prototypes 6 tests were performed at three different airbag pressures and 3 tests were executed without airbag to be used as a baseline. The study covered the analysis of the body kinematics and assessed the likelihood of incurring on a brain and head injury.

The most relevant differences could be found in the hyperextension angles obtained during the different tests. The parameters evaluated presented a significant reduction in the angles of neck hyperextension, which could lead to considerable reductions in cervical spine injuries. Therefore, further studies must be performed to confirm the effectiveness of these new airbag's prototypes.

Airbag 2 presented the best results in terms of hyperextension angles for the neck, despite the pressure imposed to the airbag. In addition, a clear correlation between volume capacity and pressure imposed to the airbag was noticed. This relation led to significant reductions in the values presented in the kinematics parameters.

Further investigations must be conducted to rate the effectiveness of the cervical airbag prototypes in reducing the cervical spine injuries.

Chapter 7. FUTURE WORKS & LIMITATIONS

The study carried out is presented as a good basis for the field of research on airbags for bicycle helmets, since it is a subject that has not been studied in depth so far. Although it is true that there are some limitations in the experiment, the results obtained mark the path to follow from now on. Some of the limitations that the experiment has faced is that the ATD used satisfies partially the aim of the project , this is due to the fact that the Hybrid III dummy was designed for vehicle frontal impact tests, so although it is a good approximation, it limits the data reliability. In addition, another limitation that was found was the impossibility of obtaining data in the area of the neck of the dummy due to de lack of sensors.

The results presented is currently being used to develop a simulation model that evaluates the performance of the airbags through a Finite Element Model, which is the reason for selecting the control environment that was designed. Consequently, the performance of the airbag is limited to the control environment and might be different under more realistic road traffic crash scenarios with cyclist implicated.

Therefore, some future works that might arise from this initial project are:

- ✓ Testing with Post-Mortem Human Surrogates (PMHS).
- ✓ New airbags prototypes designs.
- ✓ New testing configurations and setups to ensure the good behaviour of the airbag's helmets.
- ✓ Can be used to validate Finite Element Model simulations.

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ANNEX I GRAPHS

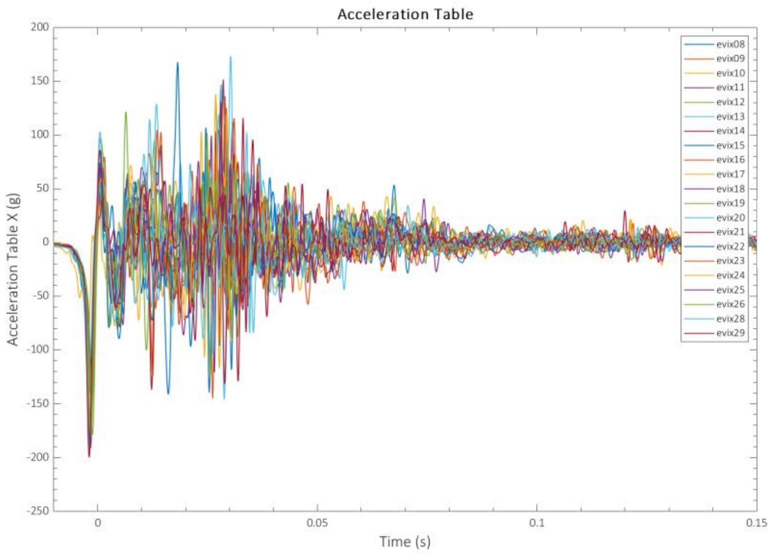

Figure 24. Table acceleration for each test

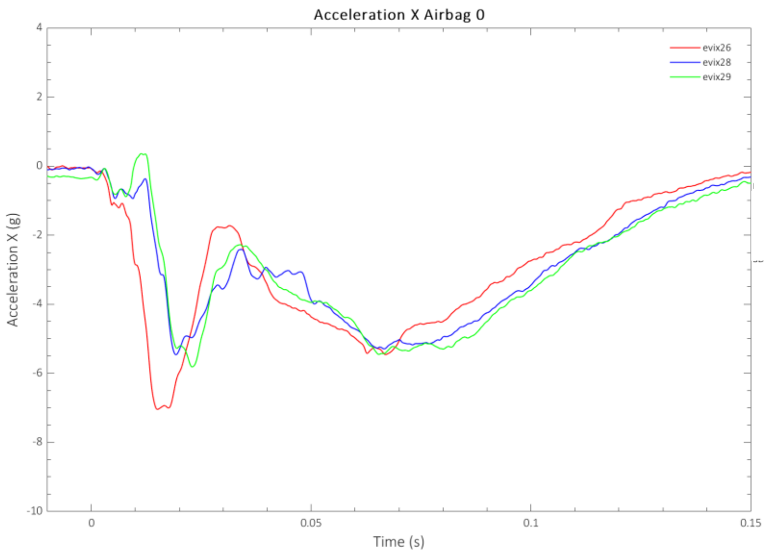

Figure 25. Linear acceleration in the X direction for test without airbag

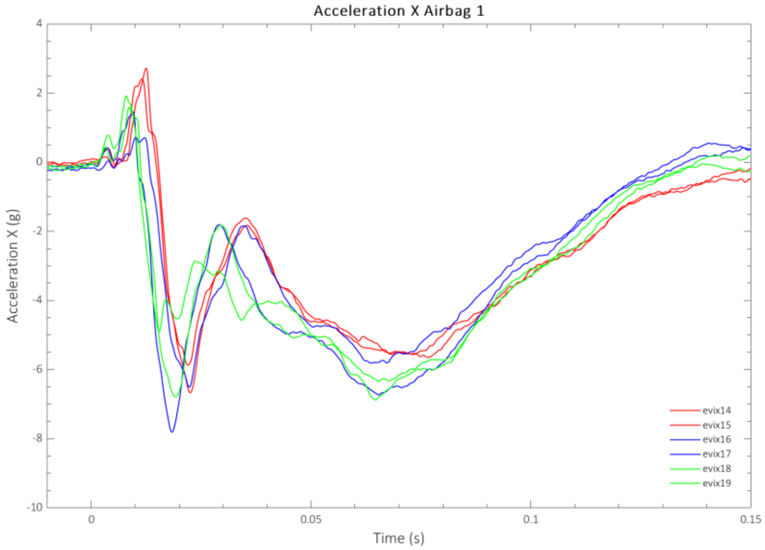

Figure 26. Linear acceleration in the X direction for airbag 1

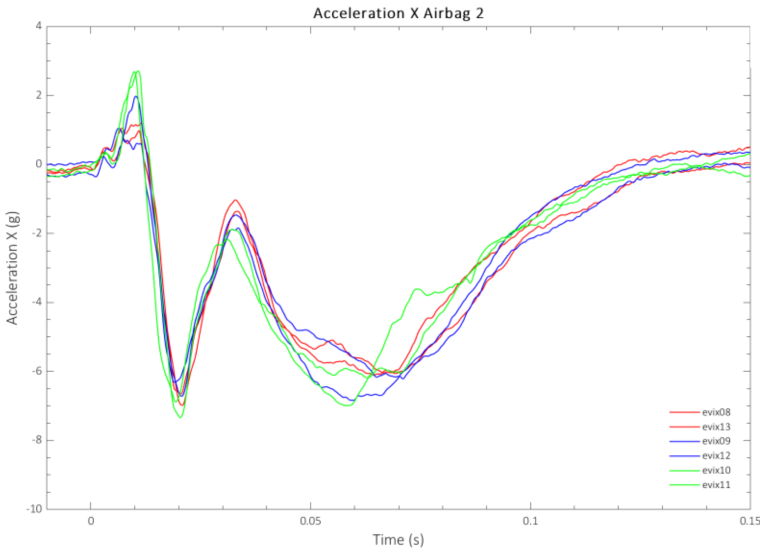

Figure 27. Linear acceleration in the X direction for airbag 2

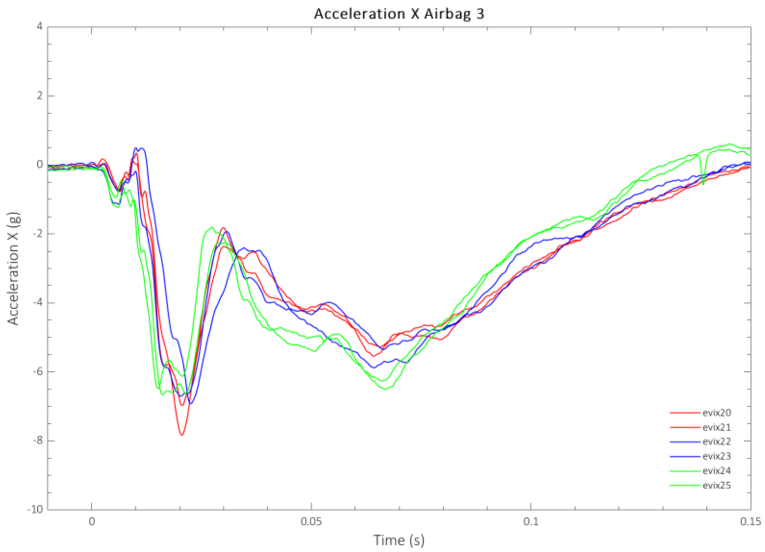

Figure 28. Linear acceleration in the X direction for airbag 3

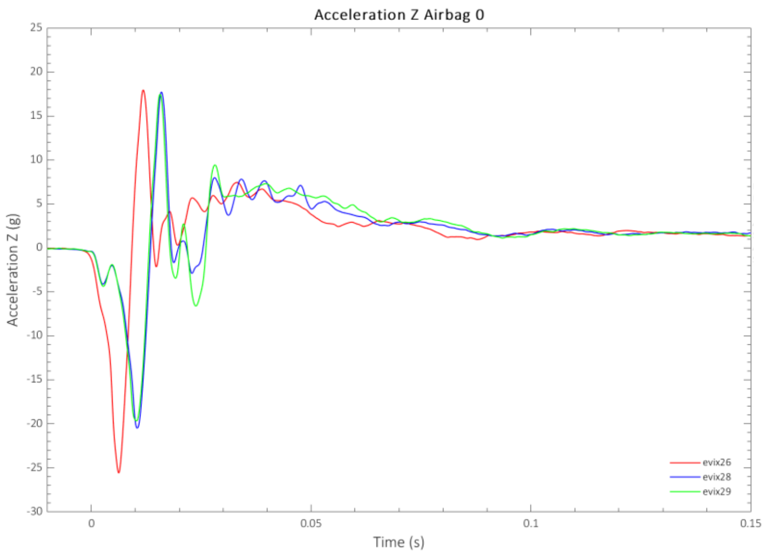

Figure 29. Linear acceleration in the Z direction for test without airbag

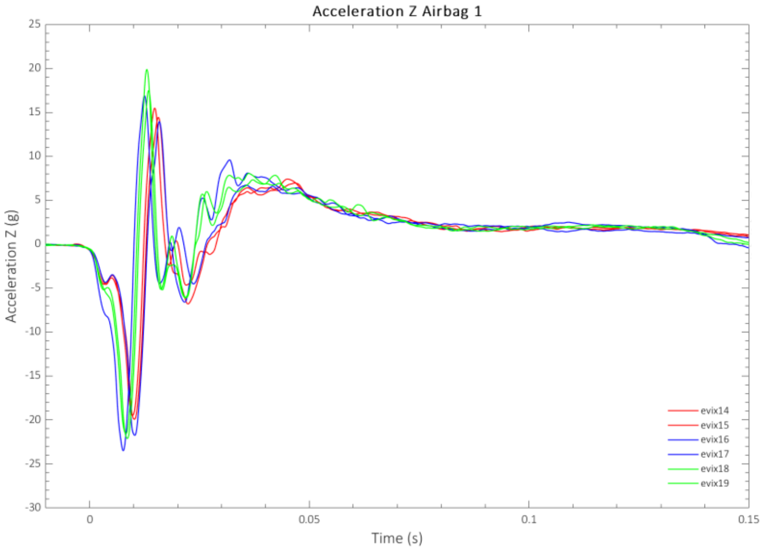

Figure 30. Linear acceleration in the Z direction for airbag 1

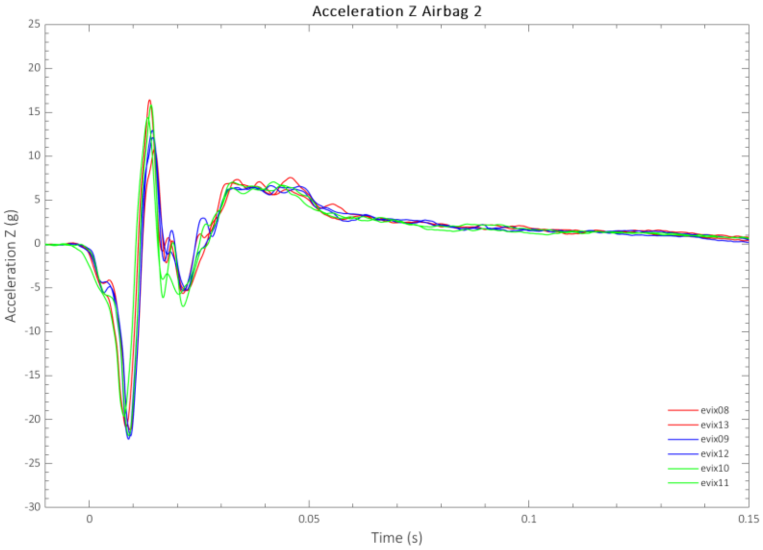

Figure 31. Linear acceleration in the Z direction for airbag 2

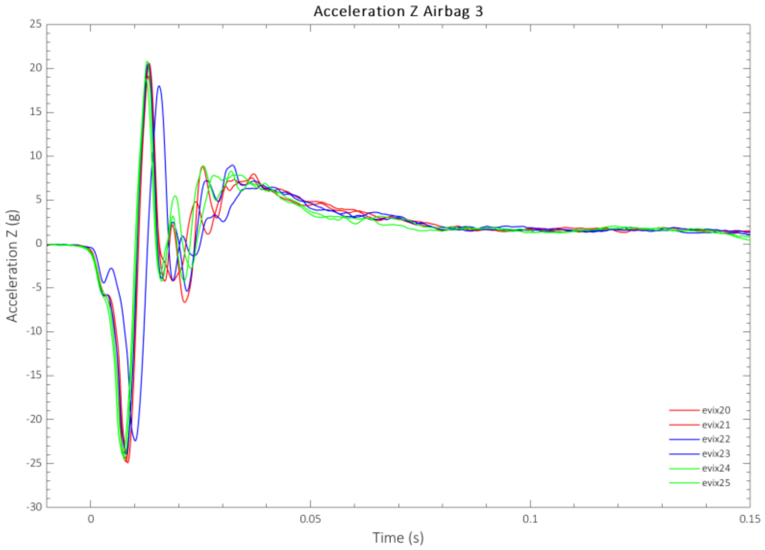

Figure 32. Linear acceleration in the Z direction for airbag 3

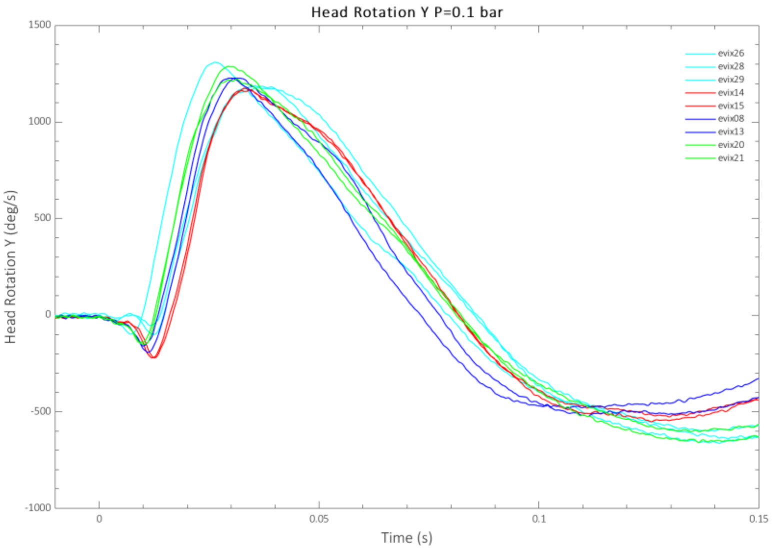

Figure 33. Head rotation for Y direction P=0.10 bar

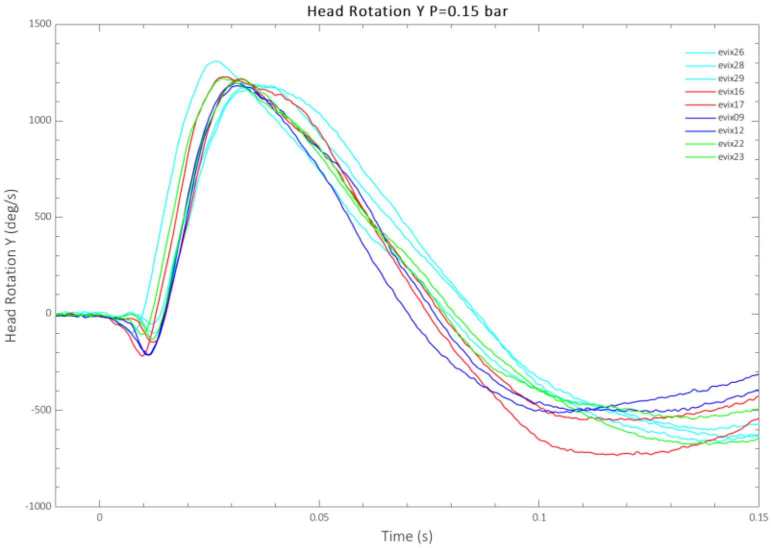

Figure 34. Head rotation for Y direction P=0.15 bar

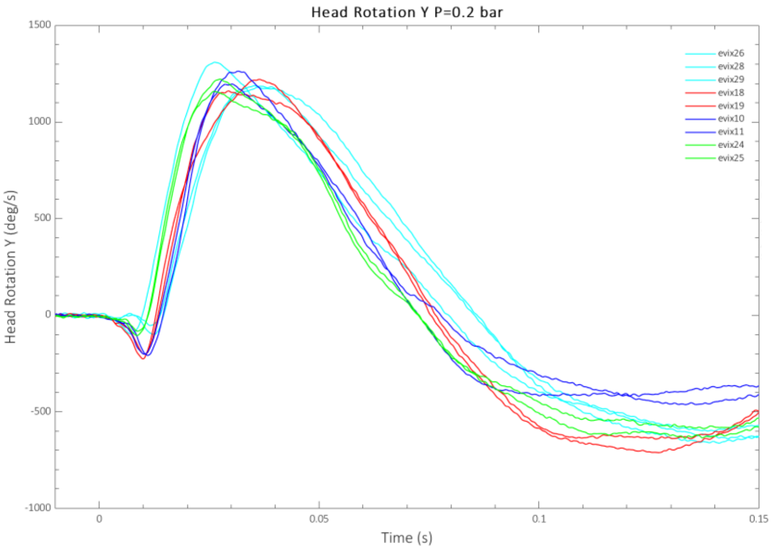

Figure 35. Head rotation for Y direction P=0.20 bar

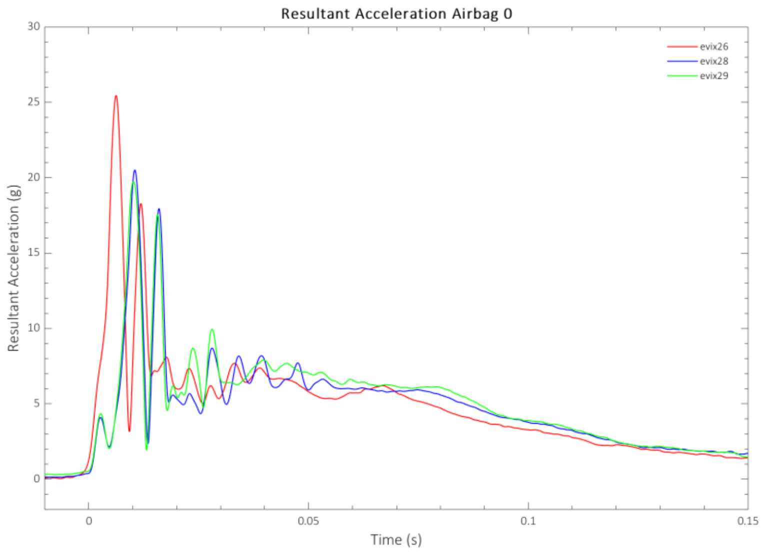

Figure 36. Resultant linear acceleration for tests without airbag

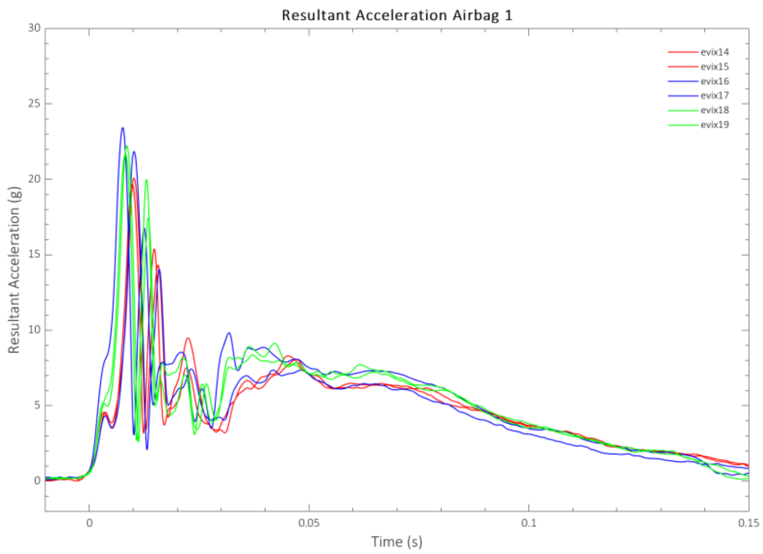

Figure 37. Resultant linear acceleration for airbag 1

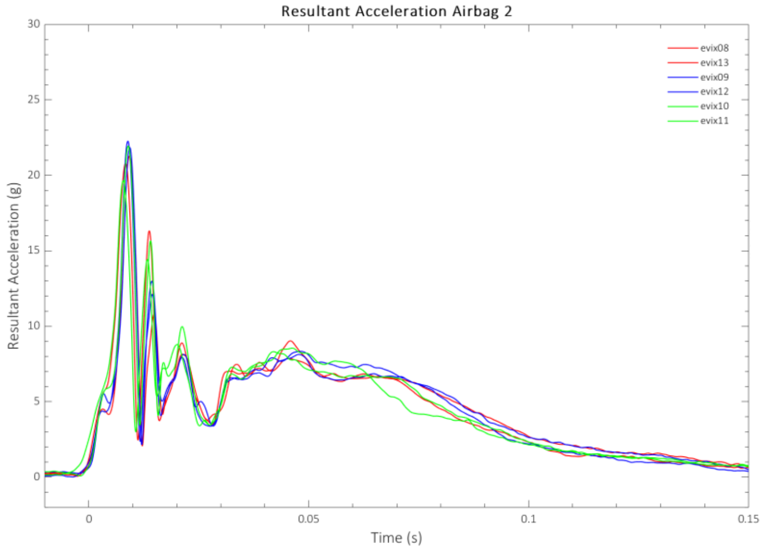

Figure 38. Resultant linear acceleration for airbag 2

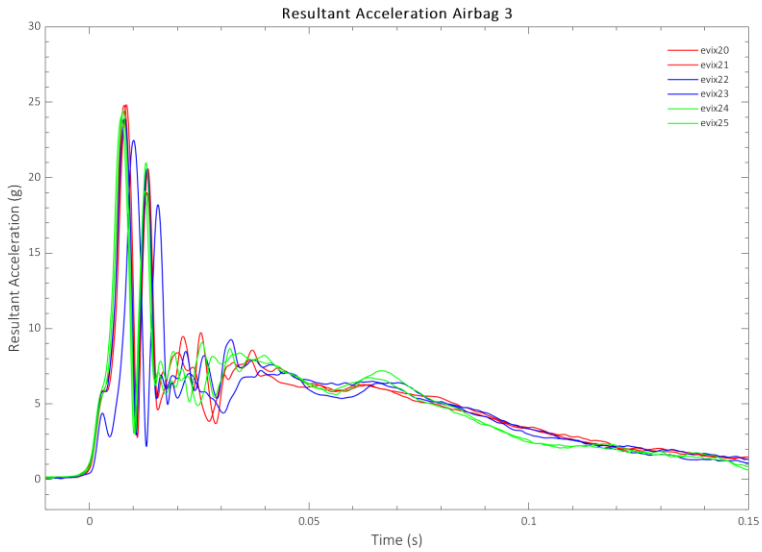

Figure 39. Resultant linear acceleration for airbag 3

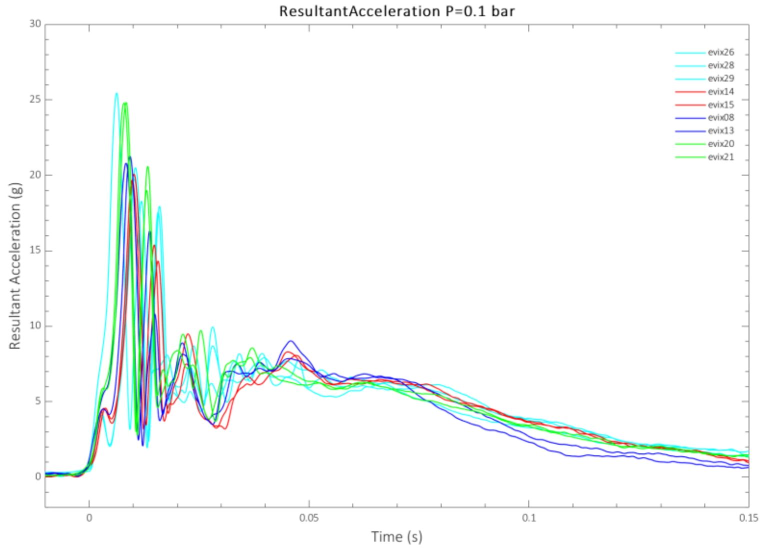

Figure 40. Resultant linear acceleration for P=0.10 bar

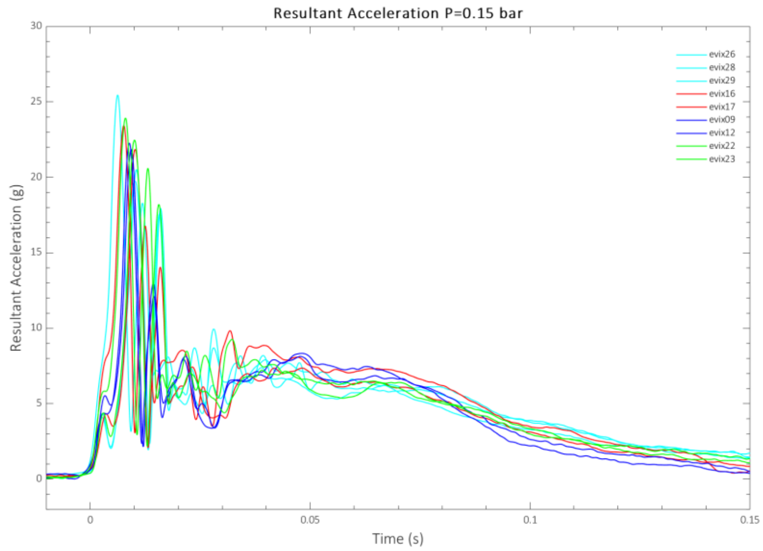

Figure 41. Resultant linear acceleration for P=0.15 bar

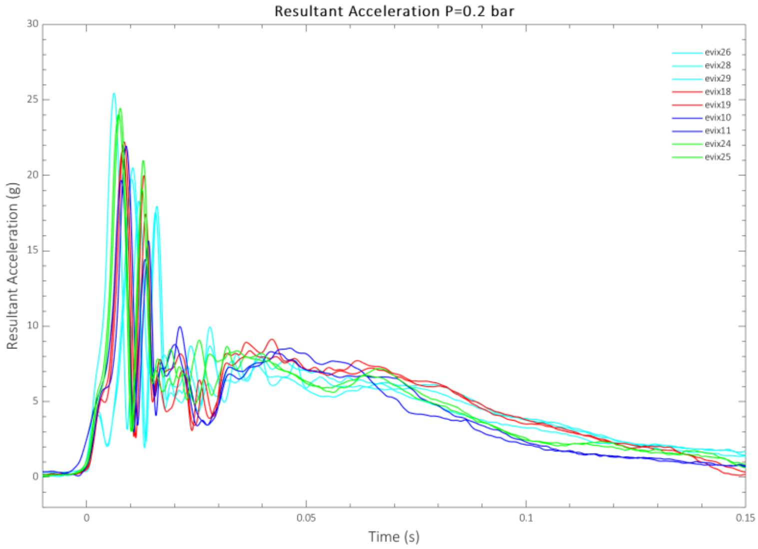

Figure 42. Resultant linear acceleration for P=0.20 bar

ANNEX II PAPER

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

Carmen M. Vives-Torres, Manuel Valdano, Jaime Alvarez-Fernandez, Juan M. Asensio-Gil, Carlos Rodriguez-Morcillo, Marc Millet-Solanelles, Nil Oleaga-Ortega, Lucas Llobet-Cusí, Francisco J. Lopez-Valdes

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).

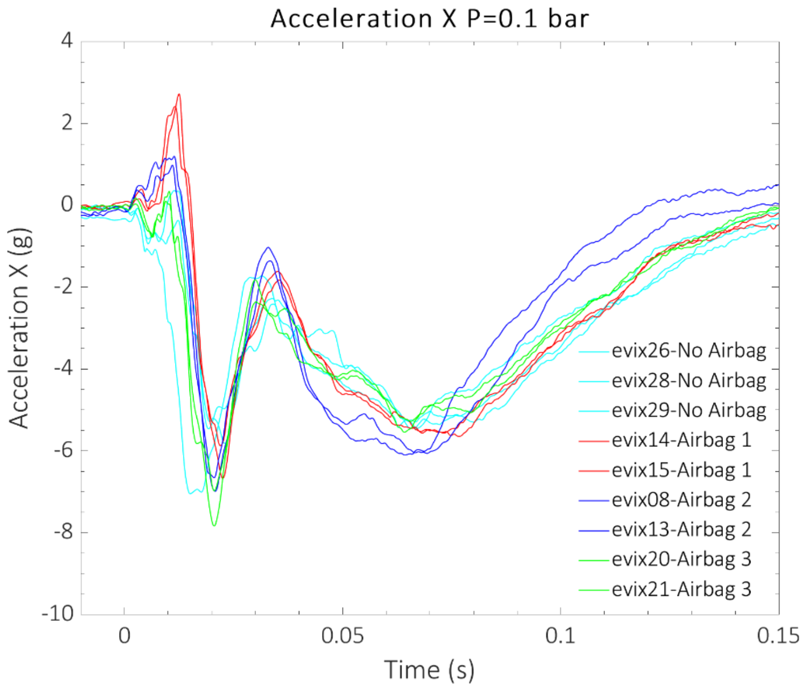

Fig. 3. Head linear acceleration in the x-axis for the tests without airbag (light blue) and with airbags at 0.10 bar (red, dark blue, and green).

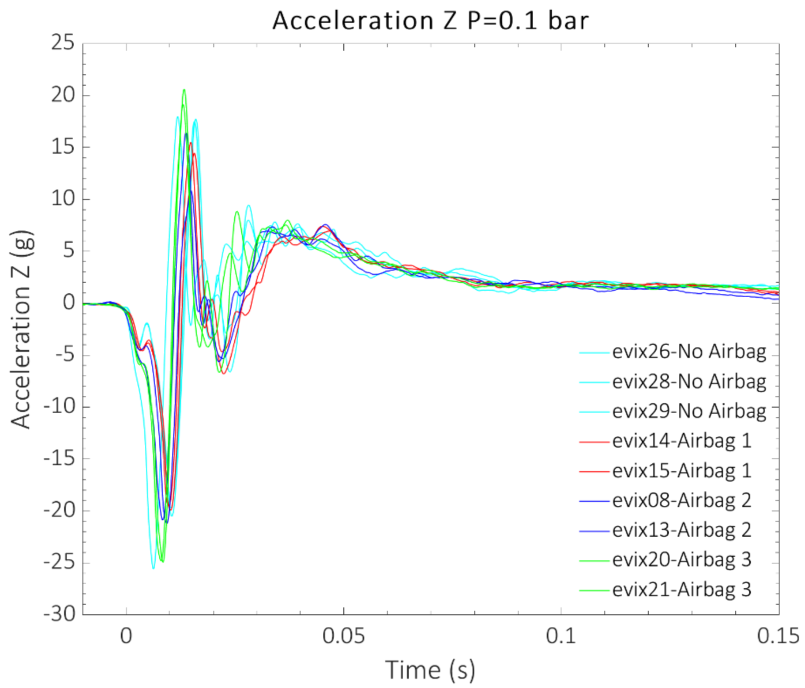

Fig. 4. Head linear acceleration in the z-axis for the tests without airbag (light blue) and with airbags at 0.10 bar (red, dark blue, and green).

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.

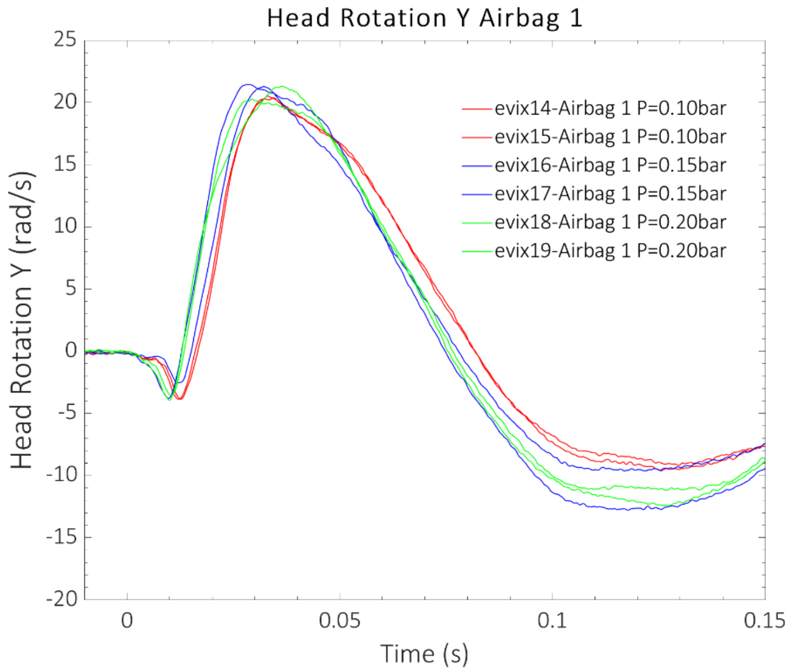

Fig. 5. Head angular velocity for tests with Airbag 1 at 0.10 bar (red), 0.15 bar (blue), and 0.20 bar (green).

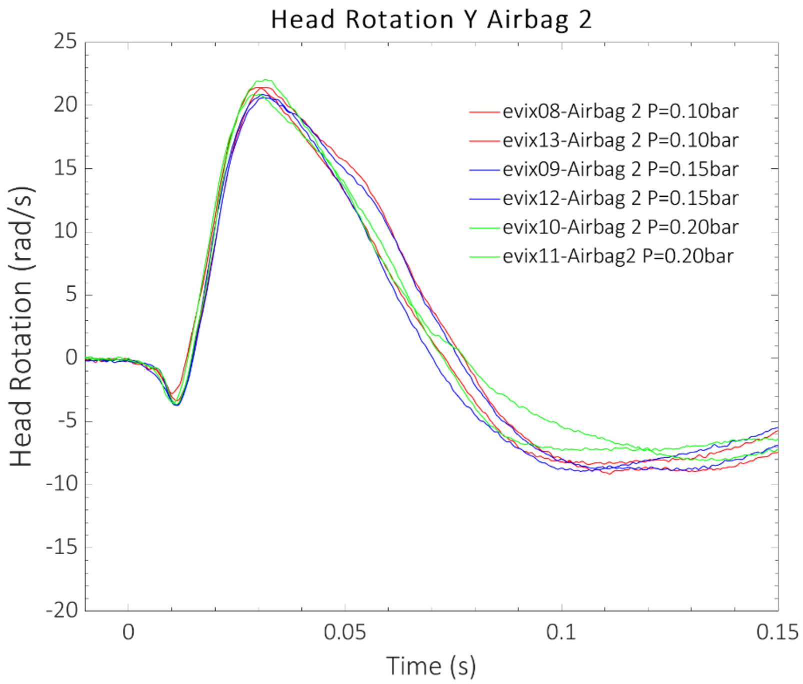

Fig. 6. Head angular velocity for tests with Airbag 2 at 0.10 bar (red), 0.15 bar (blue), and 0.20 bar (green).

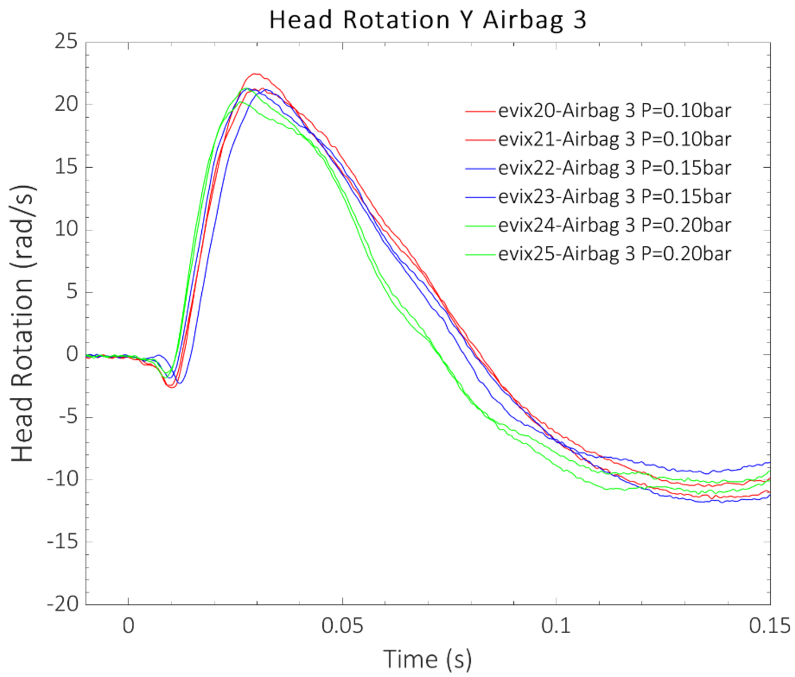

Fig. 7. Head angular velocity for tests with Airbag 3 at 0.10 bar (red), 0.15 bar (blue), and 0.20 bar (green).

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

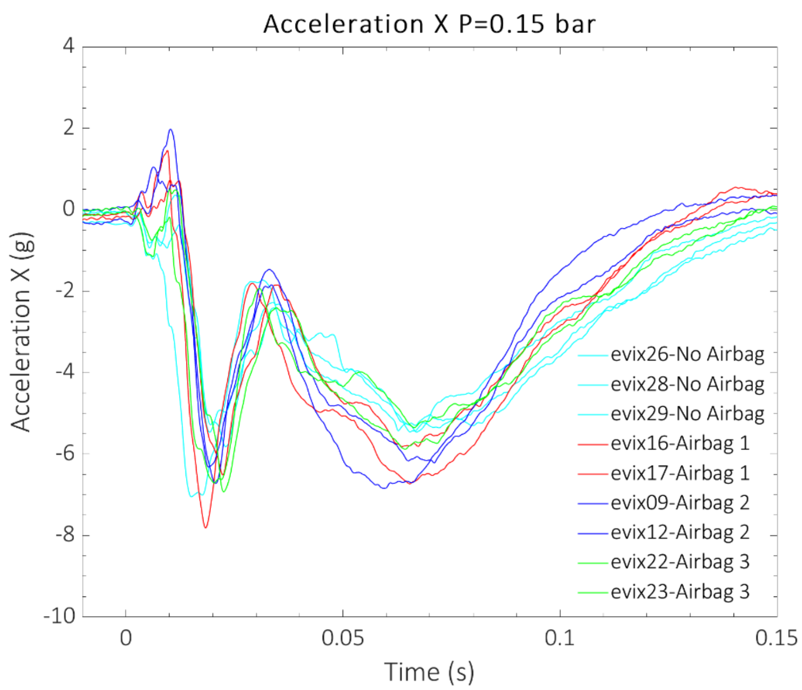


Fig. A1. Head linear acceleration in the x-axis for the tests without airbag (light blue) and with airbags at 0.15 bar (red, dark blue, and green).

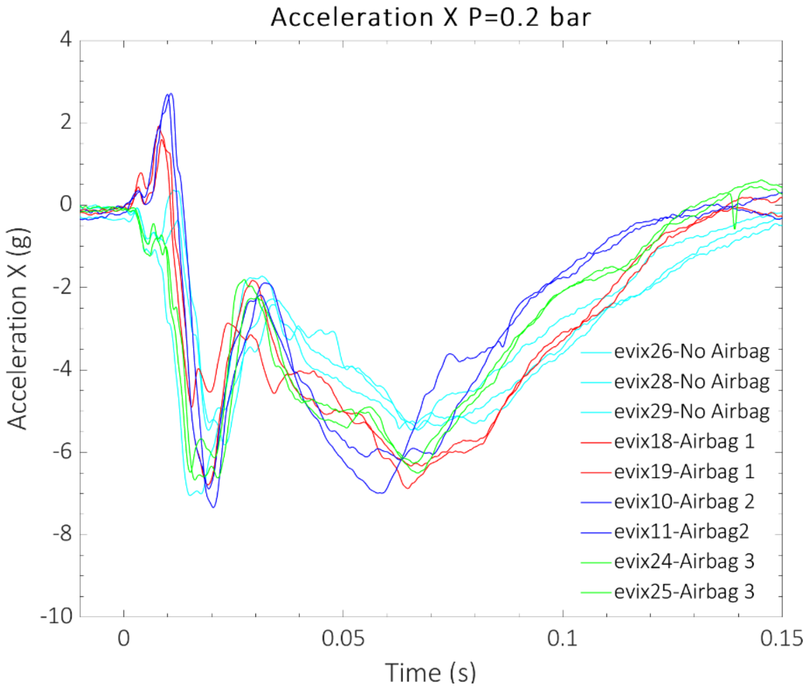


Fig. A2. Head linear acceleration in the x-axis for the tests without airbag (light blue) and with airbags at 0.20 bar (red, dark blue, and green).

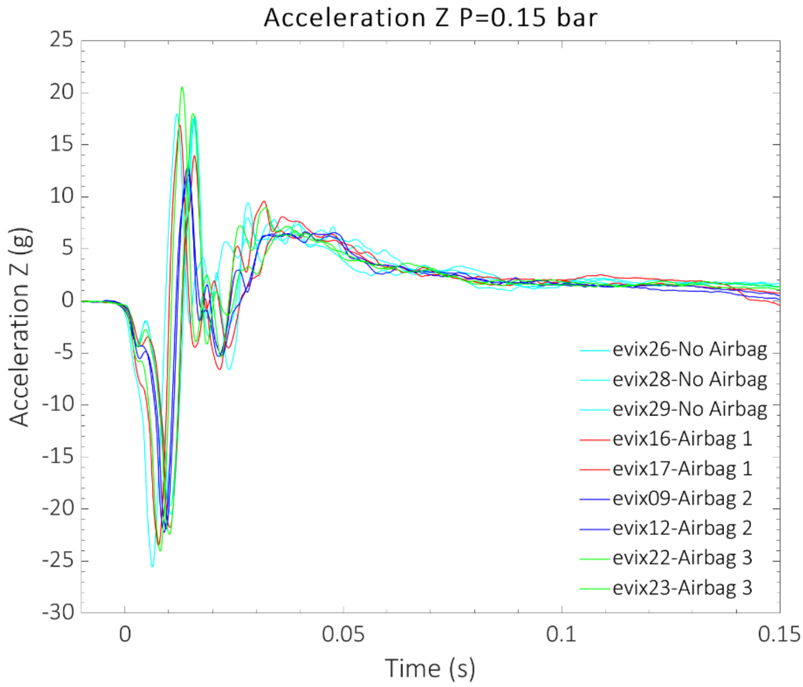


Fig. A3. Head linear acceleration in the z-axis for the tests without airbag (light blue) and with airbags at 0.15 bar (red, dark blue, and green).

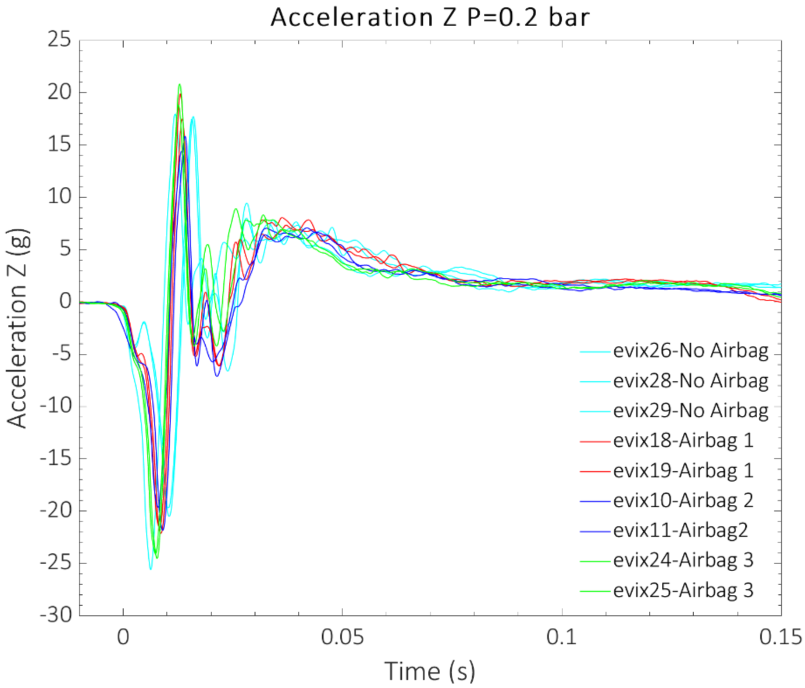
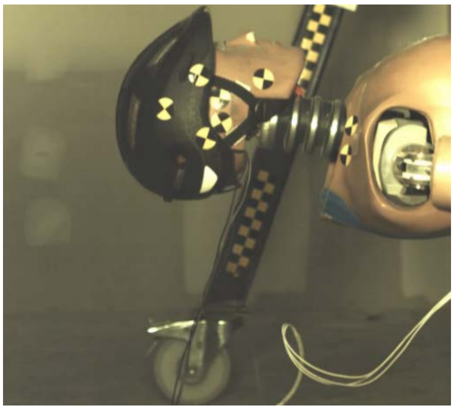


Fig. A4. Head linear acceleration in the z-axis for the tests without airbag (light blue) and with airbags at 0.20 bar (red, dark blue, and green).



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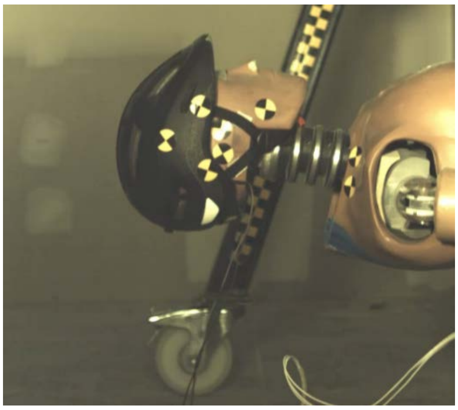
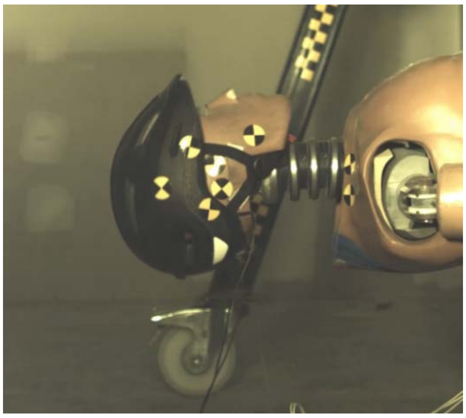


T = 0 ms (trigger)



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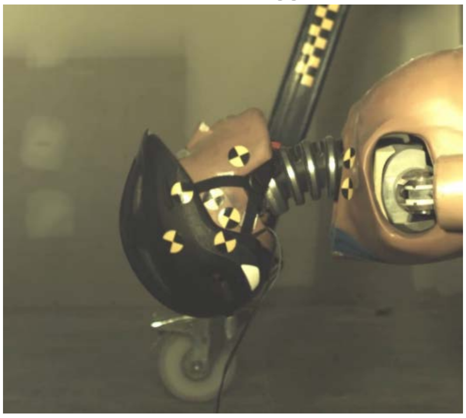
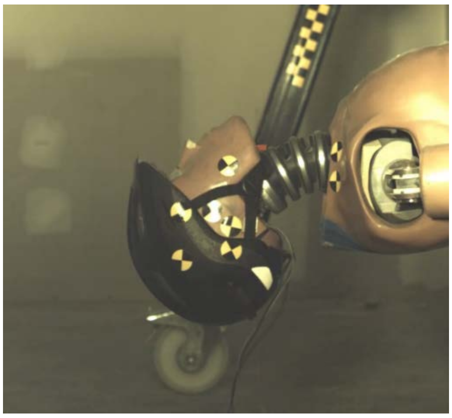
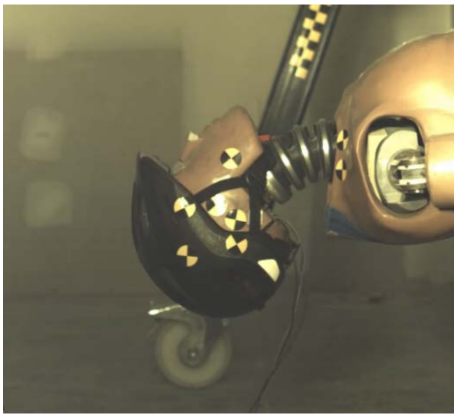


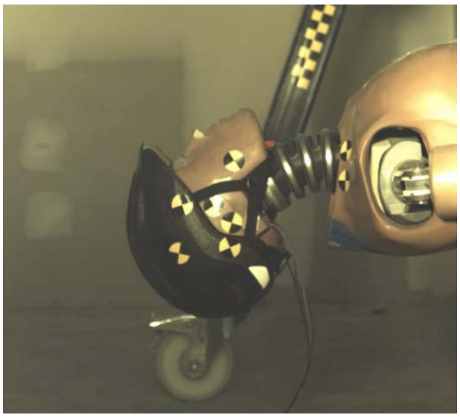
T = 40 ms



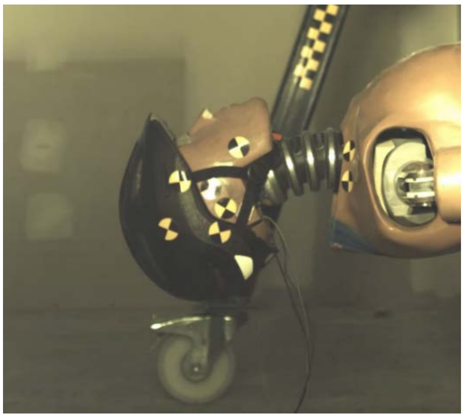
T = 60 ms



T = 80 ms

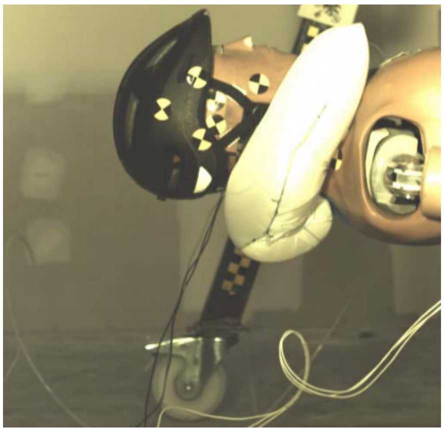


T = 100 ms

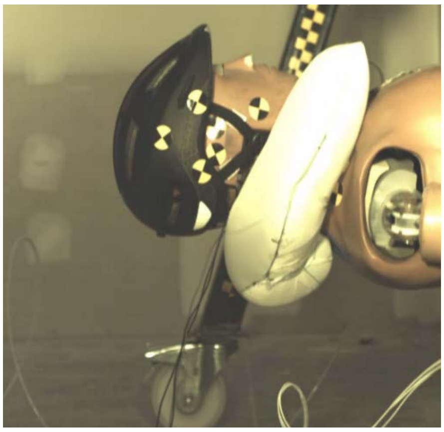


T = 120 ms

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



T = -20 ms



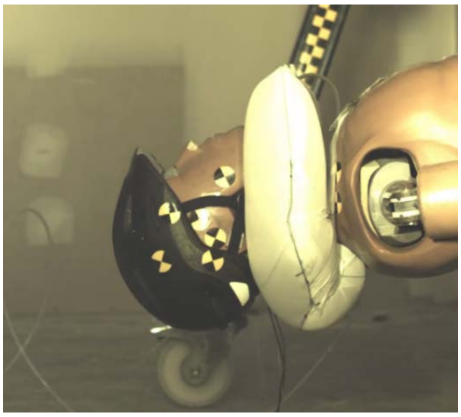
T = 0 ms (trigger)

T = 20 ms

T = 40 ms

T = 60 ms

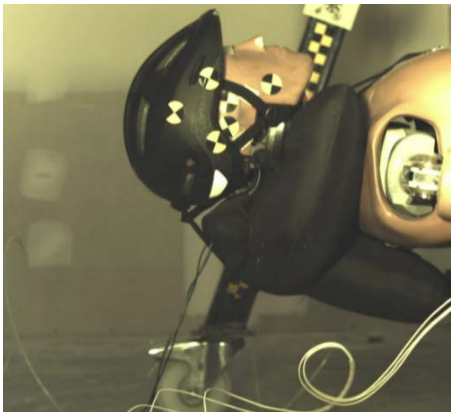
T = 80 ms



T = 100 ms

T = 120 ms

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

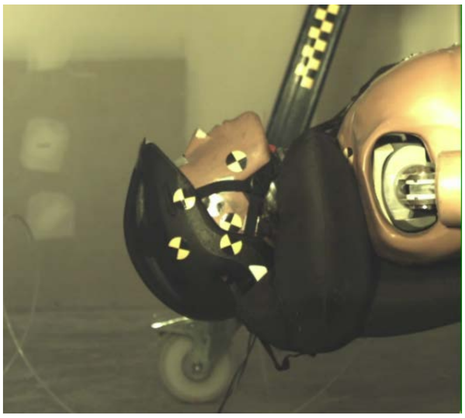


T = -20 ms

T = 0 ms (trigger)

T = 20 ms

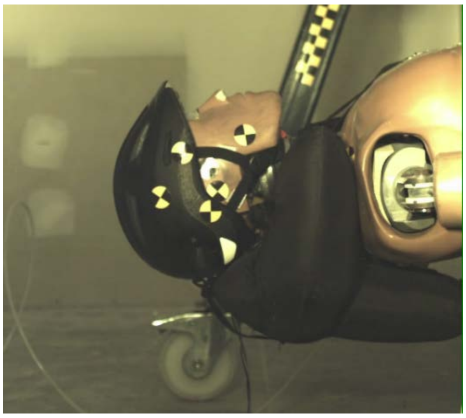
T = 40 ms



T = 60 ms

T = 80 ms

T = 100 ms



T = 120 ms

Fig. 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.

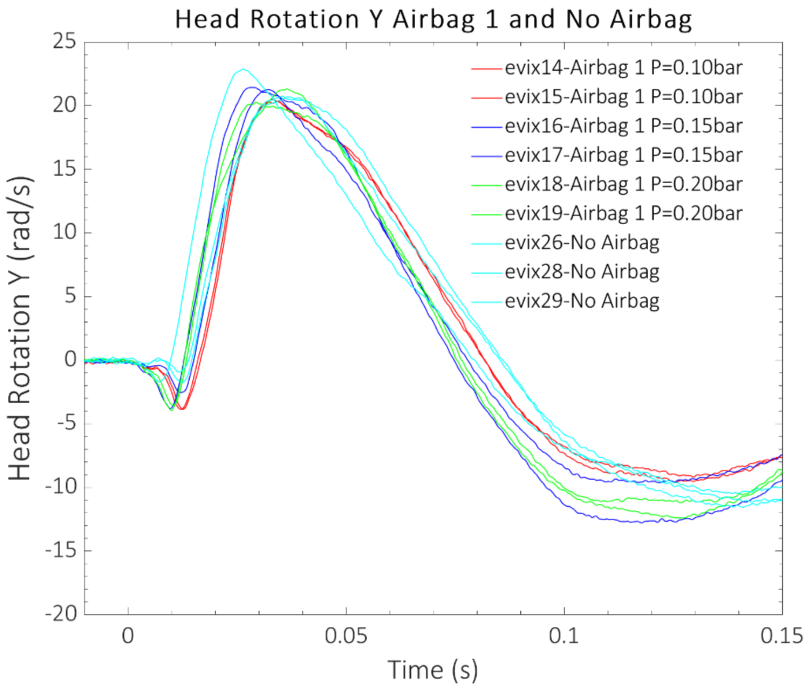


Fig. A9. Head angular velocity for tests with Airbag 1 (red, dark blue, and green) and without airbag (light blue).

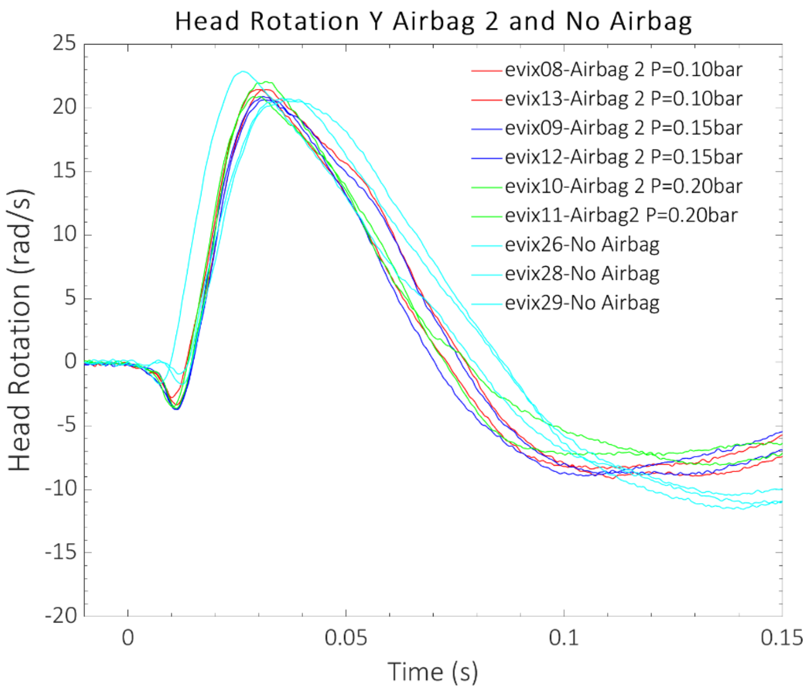


Fig. A10. Head angular velocity for tests with Airbag 2 (red, dark blue, and green) and without airbag (light blue).

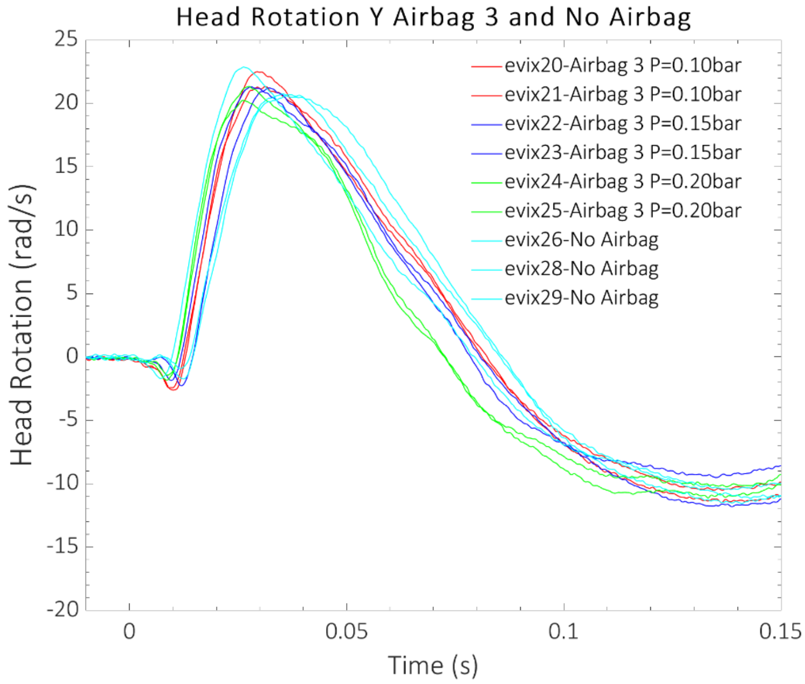


Fig. A11. Head angular velocity for tests with Airbag 3 (red, dark blue, and green) and without airbag (light blue).