POLITECNICO DI TORINO

Master's Degree in Biomedical Engineering

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Center Airbag Out-of-Position: recommended test procedures and biomechanical analysis for evaluating occupant injury risk.

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Summary

Airbags are restraining safety devices aimed at reducing both mortality and morbidity from crashes, but their activation may sometimes produce an increase in the number of injuries when the vehicle occupant is very close to the airbag or in direct contact when it deploys. In fact, when this occurs, the force resulting from the energy released by the airbag can be stronger than claimed by the manufacturer. This kind of force is very important when the vehicle occupants are in different positions from those considered normal, called out-of-positions. For the Side Airbag and Curtain Airbag, there is already a protocol in place recommending procedures for assessing out-of-position injury risks through injury criteria. For the Center Airbag there is not such a protocol. Therefore, in this paper, the implementation of the protocol is described based on 'Recommended procedures for evaluating occupant injury risk from deploying side airbags' prepared by Technical Working Group.

This protocol analyses the test devices (dummies), instrumentation, test procedures, and performance guidelines that should be used for evaluating the interaction injury risk. The risk curves, derived from instrumentation present on the dummies, correspond to a statistical model of biomechanical data in order to predict injury. The subsequent biomechanical analysis of these curve allows to determine the suitability of the airbag in question and, if it is within the limits imposed by National Highway Traffic Safety Administration, proceed with further testing for approval.

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Chapter 1

Automotive Safety

The National Highway Traffic Safety Administration - is an agency of the U.S. federal government related to transportation safety - reported data on the estimated number of motor vehicle fatalities in the first half of 2021. An estimated 20.160 people died in road accidents in the first half of 2021, an increase of +18.4% compared to 2020: 8.935 people were estimated to have died in road accidents in the first quarter of 2021 and 11.225 died in the second quarter. The fatality rate in this period increased to 1.34 deaths per 100 million miles driven, compared to the expected rate of 1.28 deaths per 100 million miles driven in the first half of 2020.[**1**]

Automotive safety is constantly evolving. With the aim of limiting the high number of road accidents and offering maximum protection to the victims involved, there has been an ever-increasing spread of safety systems. These are the main precautionary measures and can be divided into passive and active restraint systems. Many of these systems have, over time, become mandatory from a regulatory point of view. For this reason, it is increasingly common for new vehicles to undergo crash tests carried out by independent bodies to realistically assess the risk posed by today's vehicles.

It is useful to establish the difference between active and passive safety systems, which is mainly based on the timing of intervention: active safety has the role of preventing an accident, passive safety has the role of limiting the consequences when an accident has already occurred.

1.1 Active safety

Active safety refers to the set of devices or systems that should prevent an accident from occurring, with a preventive function. This category includes many devices in vehicles, such as brakes, lights, steering, tyres, and shock absorbers. Even the windscreen wiper itself, which allows a good view under certain conditions. Other active safety devices or systems are:

1. ABS, an anti-lock braking system
2. ESP, an electronic dynamic stability control system which also enhances the functionality of ABS
3. TCS, a system that reduces wheel spin and improves stability

The use of these devices is closely linked to the role they are intended to play, and it is important to know their limits and conditions of effectiveness in order to avoid improper use. Furthermore, the most complex or technologically advanced devices do not protect against possible sudden failures, may require special maintenance and periodic overhaul, and must be handled by expert personnel.[**2**]
At the heart of vehicle safety there is the driver, who must always drive carefully and respect road rules.

1.2 Passive safety

Passive safety devices and systems are intended to reduce the consequences of an accident once it has occurred. Their main objective is to absorb the energy released in the event of an impact so as not to cause damage. The management of the kinetic energy possessed by the vehicle and its occupants at the moment of impact is the typical field of application of passive safety devices and the main subject of research into it. According to the definition, this class includes devices such as:

1. Safety belts
2. Child seats
3. Airbags
4. Headrests
5. Chassis

This list may also include roadside guard-rails, which have the function of containing and absorbing the kinetic energy of the vehicle thanks to their deformability.

Steering, pedals, doors, and other parts of the vehicle must comply with the Community's regulations on passive safety concerning the type-approval of various parts or devices.[**3**]

1.3 Airbag

Airbag is a vehicle automatic occupant-restraint system using a bag designed to inflate very quickly to reduce injuries and fatalities in the event of an accident. It is inflated in 30-50 thousandths of a second at a speed of approximately 320 km/h. It can only be used once and should be replaced after use. The appropriate power level is based on the sensor input readings, which can typically detect occupant size, seat position, occupant's seat belt use and accident severity. Major factor controlled is the pressure inflation because the force of the airbag inflation causes most of the injuries or deaths. Airbags drastically reduce both morbidity and mortality from crashes, but with the increased use of airbags there has been a corresponding increase in the number of injuries attributable to these devices.[**4**]

The variable that quantifies occupant protection is the stiffness of the airbag. It is not a constant variable but is related to both force and displacement. One can consider the airbag as a spring of constant k which can be described as

$$k = \frac{d}{ds}[F(s)] = \frac{d^2}{ds^2}[W(s)] = \frac{d^2}{dV^2}[P(s)] \tag{1.1}$$

The stiffness values must be suitable for its operation. The stiffness should not be too low or too high, as it will not function properly under either condition. A compromise is sought for this value in order to comply with the standards specifying different possible types of accident.

1.3.1 The history

The airbag was invented by John W. Hetrick, an American industrialist who came up with the idea after an accident involving his family and took out a patent in 1953, calling it a 'safety shock absorber assembly for automotive vehicles'. Around the same years, the German Walter Linderer proposed a type of airbag also based on a system of compressed air, released by the contact of the bumper or the driver, but it proved ineffective because it was not able to inflate the airbags quickly enough. In fact, over the years the restraint system was improved and modified until it was used in cars in the early 1960s with the refinement of other components to make it work properly. In 1964, Japanese automotive engineer

Yasuzaburou Kobori developed an airbag that used an explosive device to trigger its inflation, for which he obtained patents in 14 countries. In 1967, Allen K. Breed invented a ball-in-tube mechanism for shock detection: an electromechanical sensor with a steel ball attached to a tube by a magnet would inflate an airbag in less than 30 milliseconds. For the first time, a small explosion of sodium azide was used instead of compressed air during inflation. The mechanism was also adopted in 1973 by General Motors, a US car manufacturer, and installed in a fleet of cars for government use, and the same year in the first passenger-side car for public use, an Oldsmobile Toronado. The following year it was also fitted to other vehicles from the car manufacturer, Buick, and Cadillac. The reception in the market was not positive, initially not everyone approved and shared the actual validity of the airbags, and this led to further delays in the spread of this device that proved, in retrospect, effective. In 1967, Mercedes began researching a type of airbag that would be installed in the most luxurious cars from 1980 onwards.

In the first half of the 1980s, the airbag started to become widespread, but with one recommendation: one should not travel in a vehicle equipped with an airbag without a seat belt. It was believed that the airbag, on its own and without seat belt support, would increase the possibility of injury when deployed. In 1994, TRW, a global supplier of automotive products, began production of the first gas-powered airbag. It was not introduced in Europe until the first half of the 1990s and was not finally accepted as a standard until after 2000.[**5**]

1.3.2 Structure

The airbag, as previously mentioned, is nothing more than a balloon (inner tube), which, in the event of an accident, is inflated by a small explosion, a chemical reaction or the instantaneous release of compressed gas. The airbag is made of a polyamide fabric, usually nylon, or polyethylene terephthalate (PTE) which is very strong and resistant to ageing. In addition, it must have a low friction coefficient in order to have a light deployment and non-abrasive skin contact. The inside of the cushion is fitted with retaining straps which keep the inflatable cushion in the desired shape when it bursts. On the back, there are outflow openings through which the gas escapes. There are two ways of deploying the inflatable cushion: the standard deployment and the star deployment. Star deployment has a reduced expansion towards the driver and is advantageous when the position of the passenger seats is not correct (Out-of-Position).[**6**]
An airbag is made of:[**7**]

1. **Sensors**, that analyse the impact, measure the vehicle's abrupt deceleration and thus the intensity of the impact.

2. An **electronic control unit** (ACU - airbag control unit) which receives the signal from the sensors, processes it and sends the command to a detonator.

3. A **detonator** (or gas generator) which triggers the substance, usually sodium azide NaN_3 and potassium nitrate KNO_3, contained in the explosive capsule by means of an electric current or the impact of a tip. When it explodes, it develops enough gas to inflate the container.

4. A possible **second capsule**, present in hybrid airbags, which contains pre-compressed inert gas that inflates the bag.

5. The airbag itself, the **bag**, which is usually made of synthetic material and has holes in the back.

The airbags in the vehicle are controlled by a central airbag control unit (ACU) which monitors several related sensors within the vehicle (accelerometers, impact sensors, side pressure sensors, wheel speed sensors, gyroscopes, brake pressure sensors and seat occupancy sensors). ACUs often record this and other sensor data into a circular buffer to provide a snapshot of the crash event. An ACU typically includes capacitors within its circuitry so that the module remains powered and able to deploy the airbags if the vehicle's battery connection to the ACU is interrupted during an accident.
Once the required threshold is reached or exceeded, the ACU activates a gas-generating propellant to rapidly inflate the bag. When the vehicle occupant impacts and crushes the bag, the gas escapes in a controlled manner through small ventilation holes. The volume of the airbag and the size of the openings in the bag are adapted to each type of vehicle, to distribute the force experienced by the occupant over time and over the occupant's body.

Signals from the various sensors are sent to the airbag control unit, which determines from them the angle of impact, severity or force of the crash, along with other variables. Each restraint is typically activated with one or more pyrotechnic devices, commonly called initiators or electric matches. The electric match consists of an electrical filament wrapped in a combustible material. When the filament overheats, it starts the gas generator, which in turn initiates the chemical reaction of the solid fuel: the extremely rapid reaction between the two elements produces an enormous amount of nitrogen in the gaseous state, which instantly inflates the module at speeds of over 300 km/h.

1.3.3 Mechanism of action

Airbag deployment is divided into three stages:[**8**]

1. Detection - crash sensors detect sudden deceleration due to a rapid impact. The sensors activate a unit containing sodium azide that starts to burn and releases nitrogen gas that inflates immediately the bag.

2. Inflation - occurs in the first 100 ms at a very high average speed (around 150 mph). Numerous other by-products such as carbon dioxide, metal oxides, sodium hydroxide and other gases are also released with nitrogen gas, creating a highly corrosive alkaline aerosol. Sodium azide can lead to the production of toxic and explosive products due to the reaction with water. These products can cause abrasive damage and skin injuries.

3. Deflation - takes place within 2 seconds, involves venting gases through exhaust ports or porous panels while the cushion cools. Deflation is often accompanied by the release of dust-like particles and gases into the interior of the vehicle called effluent. Most of this dust consists of corn-starch, french chalk or talc, which are used to lubricate the airbag during deployment.

Before the impact there is a constant pressure, and consequently a constant speed of motion that can be considered as a laminar motion, while during impact the fluid assumes a variable speed that generates a totally turbulent motion. Before reaching the phase of complete deployment (regime) of the device, there is a transitory phase in which the motion is totally random. Once the regime has been reached, a laminar type of motion returns.

1.3.4 Types of airbag

In terms of type, airbags differ according to their components and can therefore be pyrotechnic, hybrid or dual stage:[**9**]

1. Pyrotechnic airbags are characterized by a metal mesh between the first capsule and the bag that cools the gases that inflate it, preventing solid particles from entering the bag.

2. Hybrid airbags are not characterized by the net because the second capsule containing the inert gas does not contain such particles since the gas is colder than the smoke generated by the explosion of the first capsule.

3. Dual stage airbags are characterized by a pair of pyrotechnic or hybrid airbags but with only one bag. On the basis of the information received, the control unit chooses whether to activate a single pyrotechnic charge by partially inflating it or to trigger both but at a temporary distance.

The difference between a hybrid and a pyrotechnic airbag lies in the fact that the former is faster than the latter and is therefore mainly used for side and curtain types since space is limited and they need to deploy as quickly as possible. Another advantage of hybrids is that they require a smaller amount of charge than pyrotechnics and, as a result, production is easier and cheaper.

They can be further divided according to where they are placed and what protective function they are to perform. The types currently on the market are:

1. **Frontal**: All cars are equipped with two frontal airbags - one for the driver (mounted on the steering wheel) and one for the passenger next to the driver (mounted on top of the front dashboard). Airbags that deploy in two or more steps, depending on the force of the impact, are called adaptive.

2. **Side (SAB)**: These have become very popular in recent years and are used to protect the occupants in the car in the event of a side impact, also offering protection in the event of a rollover. Airbags in this category consist of two chambers that are more compact and softer than frontal airbags. There are three types: head, thorax and combined.

3. **For the knees**: installed under the steering wheel to protect the driver, and under the glove box for the front passenger.

4. **Curtain (CAB)**: Protects the head of front and rear passengers in the event of an accident or impact with transverse forces. The airbag deploys between the passenger's head and the window. They can be placed at the front or rear of the car roof and also between the pillars.

5. **Center (CeAB)**: is installed between the front seats and deploys between them.

6. **For pedestrians**: designed to minimize pedestrian injuries in the event of a collision with the car and is located at the bottom of the bonnet, close to the windscreen.

Figure 1.1: Frontal airbag

Figure 1.2: Side airbag

Figure 1.3: For knees airbag

Figure 1.4: Curtain airbag

Figure 1.5: Center airbag

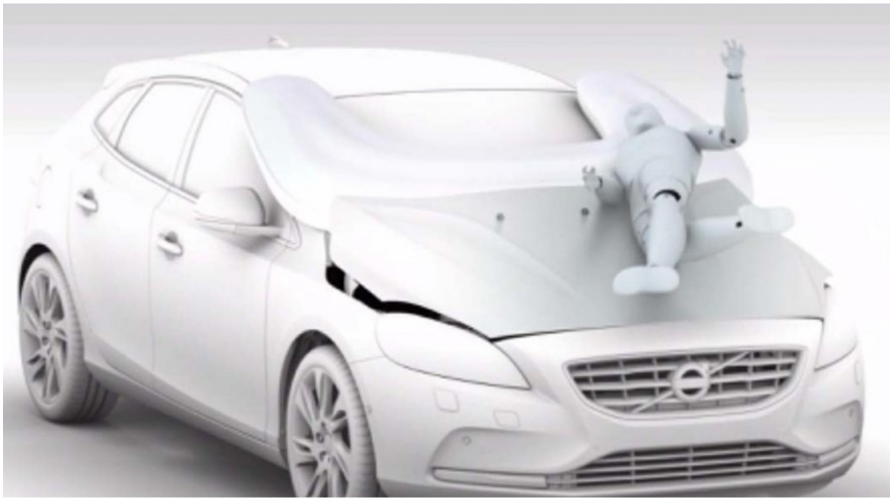


Figure 1.6: For pedestrians airbag

1.4 Center Airbag

The Center Airbag (CeAB) is a vehicle occupant-restrain system designed to help reduce the risk of crash injury to the vehicle occupants. It is an airbag integrated into the front seat (driver and/or passenger) that it deploys between the front seating positions. The bag is intended to deploy upward and forward and to wrap around the occupant to protect the torso and the head.[**10**]

The CeAB has several aims:

1. To prevent front passengers from colliding with each other.

2. To reduce occupant's lateral motion across the vehicle due to intrusion components and the striking vehicle.

3. To help protect the driver from the impact zone on the passenger side if there is not a front passenger.

The CeAB is only activated in near side impacts, far side impacts and rollovers. The airbag does not deploy in frontal or rear impacts, but it does if the vehicle is involved in a multiple crash event in which a subsequent side impact or rollover occurs.

1.5 Out-of-Position

Out-of-position (OoP) is an American protocol related to airbag construction and it denotes a situation in which the vehicle occupant is out of a normal seating position (upright and forward-facing position) but is close to the airbag module at the time of inflation. In fact, these tests are performed with positions chosen to block the airbag deployment path and to align the dummy's measurement systems to measure the uncommon dummy-airbag interaction effects. The interaction between occupant and airbag module is referred to as the 'worst-case' scenario, potentially more dangerous than common positions.

The requirements are divided for frontal airbags - more stringent than side airbags - and side airbags. As they are considered separately, the standards governing these devices are different. For frontal airbags, there is the FMVSS - 208 regulation (Federal Motor Vehicle Safety Standard regulates automotive occupant crash protection and its purpose is to reduce the number of fatalities and the number and severity of injuries to occupants involved in frontal crashes), while for side airbags there is the TWG (Technical Working Group, which will be discussed in chapter 3). In both there is a sub-category dealing with out-of-position.

These tests are of interest because small changes in a vehicle occupant position can have profound effects on the results of OoP testing. In the interim-final rule of FMVSS that NHTSA issued in May 2000, OoP static test becomes a mandatory requirement of new regulation and has been implemented since 2003.

Out-of- position tests, focusing on few parameters, are static for reasons of higher repeatability of test conditions and were mainly developed to evaluate the inflation-injury risk of airbags.

OoP situations in real world accidents can occur when:[**11**]

1. delayed ignition of the airbag during collision.

2. close by seating positions.

3. forward movement of the occupant due to pre-impact braking.

During contact phase under OoP conditions, different loading mechanisms (analysed by Horsch et al.) occurs:[**12**]

1. Punchout: in an early phase of the inflation, a small portion of the bag has escaped from the module.

2. Membrane loading: the fabric wraps around a body region when the bag inflation is termined, exerting a relatively distributed force on the occupant.

3. Bag-slap: related with inflation-induced injuries, the occupant is struck by a small but rapidly moving portion of the bag. To reduce biomechanical loads under OoP conditions, is necessary centered on design and conceptual measures for the module cover and the inflator.

Chapter 2

Anthropomorphic test devices

Anthropomorphic test devices (ATDs), also called dummies, are mechanical surrogates of the human used to measure occupant injury risk of various types of restraint systems in vehicle crashes.[**13**] Dummies are used as test devices to determine compliance with both static and dynamic crash test requirements of FMVSS 208 for frontal impact protection and the FMVSS 214 for side impact protection. Also, these dummies must meet the requirements of the NHTSA.

The main characteristics of ATDs are:[**14**]

1. **Biofidelity**: ability to replicate as faithfully as possible the reactions of the human body and it is assumed to be suitable when the replica has the same anthropometry and the impact conditions.
2. **Durability**: ability to maintain performance over time.
3. **Reproducibility**: smallness of variability to be obtained between dummies tested under the same conditions.
4. **Repeatability**: similarity of results to be obtained in repeated testing of a dummy under the same conditions.

Each dummy is designed to mimic human physical characteristics such as body size, shape, mass, stiffness. They are instrumented with transducers and sensors that measure accelerations, deformations, and loading exerted on various body parts in a crash event. In fact, through comparison between mechanical responses and kinematic responses obtained by this instrumentation, it is possible to assess biofidelity and the efficacy of restraint system designs.
Specifically, the main sensors used are:

1. Accelerometers: a resistance that changes due to the acceleration caused the impact. The change in resistance results in a change in voltage and each voltage value corresponds to an acceleration value. It is essential to know not only the absolute value of acceleration but also the time in which it is applied.

2. Load cells: applied to the bones. These cells are made up of one or more resistances that generate a voltage when compressed.

3. Chest compression sensors: used to assess the compression of the chest and its speed, particularly due to seat belts.

Dummies are classified according to size, age, sex, and impact direction: there are child dummies of different ages, adult male and female dummies of different sizes, dummies for frontal, for side and for rear impact collisions.

2.1 Frontal impact dummies

Frontal impact dummies are instrumented to protect occupants in front impact crashes. The THOR and Hybrid III models are the most widely used for frontal and offset frontal automotive crash testing today, which are discussed in more detail in the following paragraphs. The most vulnerable parts of the body are the head, neck, chest, and knees.[**15**]

2.2 Side impact dummies

Side impact dummies (SIDs) are instrumented to protect occupants in side impact crashes. In these collisions, unlike frontal impacts, there is less vehicle structure to protect occupants, with the most considerable forces traveling through the pelvis, shoulder and head from one side to the other.

Four SIDs have been developed: SID, SID-HIII, EuroSID-1 and BIOSID. The SID is a Hybrid II dummy modified for side impact testing. This dummy has no arm or shoulder structure, and its chest cannot simulate the human chest response for its material. The SID-HIII is a SID in which head and neck have been replaced with those of Hybrid III to improving biofidelity. The EuroSID-1, European Side Impact Dummy version 1, has been developed in the mid-1980s to improve biofidelity, durability and instrumentation. The SID and EuroSID were evaluated by ISO (International Standards Organization) negatively due to their insufficient biofidelity. So, the BIOSID, Biofidelic Side Impact Dummy, has been required for more biofidelity (head, neck, shoulder, thorax, abdomen, and pelvis) and additional measurement capability.

In 1997, the ISO developed a more biofidelic side impact dummy, WorldSID dummy that was based on the medium size of men. The reproducibility, the durability and the sensitivity have been significantly improved compared to previous dummies.[**16**]

2.3 Rear impact dummies

Rear impact dummies (RIDs) are instrumented to protect occupants in low-speed rear impacts. Chalmers University, Volvo Car Corporation and Saab Automobile AB developed the biofidelic BioRID dummy that was designed to represent a 50^{th} percentile male in Europe with a vertebral column consisted of 24 separate vertebrae to provide realistic movements. This dummy, compared with Hybrid III, has human-like characteristics of neck and vertebrae.[**17**, **18**]

2.4 CRABI dummies

Figure 2.1: CRABI

CRABI, Child Restraint Airbag Interaction, dummies have been developed because of necessity to evaluate the injury potential associated with the interactions of deploying passenger airbags and rearward-facing child restraints when they are placed in the vehicle front seat. The dummies heights and weights were based on anthropometry studies.
The dummy's regulations are described in 49 CFR Part 572-Subpart V and are

used in FMVSS 208 that regulates automotive occupant crash protection and in FMVSS 213 that regulates child restraint system.[**19**]

2.5 Hybrid III

In 1971, the Hybrid I was developed and was used to measured head, chest triaxial acceleration and femur load. In 1972, First Technology Safety Systems (FTSS), with the support of the U.S. automotive giants, developed the Hybrid II making several modifications to the previous dummy to achieve better results. In 1973, ATD 502 dummy was developed to achieve a more human-like seating posture and a better repeatability. In 1976, General Motors (GM) developed the Hybrid III through significant improvement in the body regions of Hybrid II and ATD 502. This evolution from the Hybrid I to the Hybrid III dummies was mainly influenced by three reasons:

1. Lack of biomechanical response from dummy's component.

2. The previous dummies were lacking in measurement of biomechanical parameters and in design purpose.

3. The documentation of the previous dummies was not enough for replication.

The first objective of the new dummy was to improve the existing ones and to make component's responses more consistent with available biomechanics data. The second was the ability to measure additional parameters and to document the realization in detail to guarantee the reproducibility. Therefore, the dummy's features must be serviceability, durability, and setup stability.[**20**]
Eventually, in fact, Hybrid III features a biomechanically based head design, human-like automotive seating posture, constant torque primary joints and detailed documentation for fabrication.
The Hybrid III Dummies Family involves a 3-year-old and a 6-year-old child dummy, a small adult female and a midsize adult male dummy.
Nowadays, the Hybrid III 50^{th} percentile male dummy has been commonly used in the field of car crash tests.

Figure 2.2: Hybrid III dummies family

2.5.1 Hybrid III 3-Year-Old Child Dummy

This dummy replicates the average 3-year-old child and was developed to evaluate airbag aggressiveness when a child is close to its deployment path in vehicular front and side impact. The dummy's final design was based on a combination of two dummies: scaled-down version of Hybrid III 50^{th} percentile male and scaled-up version of CRABI dummy. The dummy should be durable and repeatable during the test conditions and should be instrumented to measure head/neck loading, chest compression and the Viscous Criterion. In fact, the dummy is equipped with several linear accelerometers, different load cells and one displacement transducer. The dummy's regulations are described in 49 CFR Part 572-Subpart P and are used in FMVSS 208 that regulates automotive occupant crash protection and in FMVSS 213 that regulates child restraint system.[**21**]

Figure 2.3: Hybrid III 3-Year-Old

2.5.2 Hybrid III 6-Year-Old Child Dummy

This dummy replicates the average 6-year-old child and was developed to evaluate airbag aggressiveness when a child is close to its deployment path in vehicular front and side impact. The dummy design was modified to include:

1. patches of skin under the chin and at the occipital condyles to decrease the probability of bag punctures.

2. neck shield at the back of head and neck due to a non-humanlike junction to decrease the probability of bag trapping.

Also, the neck and the lumbar were equipped with nylon inserts to prevent signal noise. The dummy should be durable and repeatable during the test conditions and should be instrumented to measure head/neck loading, chest compression and the Viscous Criterion. In fact, the dummy is equipped with several linear accelerometers, different load cells but fewer than Hybrid III 3-Year-Old and one displacement transducer.
The dummy's regulations are described in 49 CFR Part 572-Subpart N and are used in FMVSS 208 that regulates automotive occupant crash protection and in FMVSS 213 that regulates child restraint system.[**22**]

Figure 2.4: Hybrid III 6-Year-Old

2.5.3 Hybrid III 5^{th} Percentile Adult Female Dummy

Figure 2.5: Hybrid III 5^{th} Percentile Adult Female

This dummy represents a 5^{th} percentile adult female and a 12–13-year-old adolescents. It was designed for the evaluation of front impact countermeasures and was

a scaled-down version of the Hybrid III 50^{th} percentile dummy. This dummy is a suitable substitute for the SID-IIs, especially in the present of head and neck injuries due to identical values. The thorax, abdomen and pelvis injuries should be tested with the SID-IIs.[**23**]
The dummy's regulations are described in 49 CFR Part 572-Subpart O.

2.5.4 Hybrid III 50^{th} Percentile Adult Male Dummy

This dummy represents the average adult male, and it is the most widely developed crash test dummy for testing restraint systems in vehicle crash situations. It offers consistent results and many measurement capabilities, in fact the dummy is equipped with several linear accelerometers, different load cells and two displacement transducers. This dummy is also used in non-automotive applications such as medical and sports equipment.
The dummy's regulations are described in 49 CFR Part 572-Subpart E.[**24**]

Figure 2.6: Hybrid III 50^{th} Percentile Adult Male

2.5.5 Hybrid III 95^{th} Percentile Large Male Dummy

This dummy represents the largest size in adult population and was a scaled-up version of Hybrid III 50^{th} percentile adult dummy with similar kinematics. It was developed by the Sierra Engineering Company in the late 1940's and, in fact, it is

known as 'Sierra Sam'. It is utilized global for the evaluation of automotive and military safety restraints. The dummy is equipped with several linear accelerometers, load cells and two displacement transducers.[**25**]
The instrumentation of the Hybrid III family of dummies is quite extensive and is shown in the following table:

Dummy instrumentation	HIII-5 female	HIII-50 male	HIII-95 male	HIII-3-yr child	HIII-6-yr child
Head					
Accelerations (A_x, A_y, A_z)	Yes	Yes	Yes	Yes	Yes
Neck					
Head/C1 (F_x, F_y, F_z, M_x, M_y, M_z)	Yes	Yes	Yes	Yes	Yes
C7/T1 (F_x, F_y, F_z, M_x, M_y, M_z)	Yes	Yes	Yes	Yes	Yes
Shoulder					
Loads (F_x, F_z)	Yes	Yes	Yes	No	Yes
Thorax					
Spine acceleration (A_x, A_y, A_z)	Yes	Yes	Yes	Yes	Yes
Sternal deflection (δ_x)	Yes	Yes	Yes	Yes	Yes
Sternal acceleration (A_x)	Yes	Yes	No	Yes	Yes
Abdomen					
Lumbar (F_x, F_y, F_z, M_x, M_y)	Yes	Yes	Yes	Yes	Yes
Pelvis					
Acceleration (A_x, A_y, A_z)	Yes	Yes	Yes	No	No
Ilium (F_y)	Yes	Yes	No	No	Yes
Lower extremities					
Femur (F_x, F_y, F_z, M_x, M_y, M_z)	Yes	Yes	Yes	No	Yes
Tibia/femur displacement (δ_x)	Yes	Yes	Yes	No	No
Knee clevis (F_z)	Yes	Yes	Yes	No	No
Tibia loads and moments (F_x, F_y, F_z, M_x, M_y, M_z)	Yes	Yes	Yes	No	No

Figure 2.7: Hybrid III 95^{th} Percentile Large Male

2.6 SID-IIs

Figure 2.8: SID-IIs

The instrumentation for side impact dummies is shown in the following table:

Dummy instrumentation	SID	SID-HIII	EUROSID-1	BIOSID	SID-IIs
Head					
Accelerations (A_x, A_y, A_z)	No	Yes	Yes	Yes	Yes
Neck					
Head/C1 ($F_x, F_y, F_z, M_X, M_y, M_z$)	No	Yes	No	Yes	Yes
C7/T1 ($F_x, F_y, F_z, M_x, M_y, M_z$)	No	No	Yes	Yes	Yes
Shoulder					
Loads (F_x, F_y, F_z)	No	No	Yes	Yes	Yes
Deflection (δ_y)	No	No	No	Yes	Yes
Arm (A_x, A_Y, A_z, M_Y, M_z)	No	No	No	No	Yes
Thorax					
Spine acceleration (A_x, A_Y, A_z)	Yes	Yes	Yes	Yes	Yes
Rib deflection (δ_y)	No	No	Yes	Yes	Yes
Rib acceleration (A_y)	Yes	Yes	Yes	Yes	Yes
Abdomen					
Force (F_y)	No	No	Yes	No	No
Deflection (δ_y)	No	No	No	Yes	Yes
Lumbar (F_x, F_y, F_z, M_X, M_y)	No	No	F_y, F_z, M_x	Yes	Yes
Pelvis					
Acceleration (A_x, A_y, A_z)	Yes	Yes	Yes	Yes	Yes
Ilium (F_y)	No	No	No	Yes	Yes
Acetabulum (F_y)	No	No	No	No	Yes
Pubic (F_y)	No	No	Yes	Yes	Yes
Lower extremities					
Femur ($F_x, F_y, F_z, M_X, M_y, M_z$)	No	No	F_z	Yes	Yes
Knee clevis (F_z)	No	No	No	Yes	Yes
Tibia loads and moments ($F_x, F_y, F_z, M_x, M_y, M_z$)	No	No	No	Yes	Yes

The dummy is a small, second-generation side impact dummy. It is based on the anthropometry of Hybrid III 5^{th} female but also based on the height and weight of 12-13-year-old adolescents. The SID-IIs dummy is instrumented in the head, thorax, abdomen and pelvis. This enables to evaluate side impact countermeasures for small occupants. The dummy weighs only 45 kg, designed for the development of side bags, in particular to assess the risks in case of OoP. The dummy is equipped with several linear accelerometers, different load cells and displacement transducers. The dummy's regulations are described in 49 CFR Part 572-Subpart R and are used in FMVSS 214 that regulates side impact protection.[**26**]

2.7 THOR

At present, the Hybrid IV, also known as THOR (Test device for Human Occupant Restraint), is being developed as a successor to the Hybrid III. This features significantly expanded instrumentation, more sensitive sensors, and more biomechanical improvements. In fact, compared with the Hybrid III dummy, this dummy has better damage prediction ability and has more human-like characteristics. EuroNCAP is considering the use of this ATD for future frontal impact tests as part of their ongoing commitment to improving road safety.[**27**]

Figure 2.9: THOR

2.8 Dummy construction

The **head** consists of an aluminum casting covered by vinyl skin. The thickness of vinyl skin was chosen to improve biomechanical fidelity and repeatability of head response during impacts. The design is based on ATD 502 because it represents the state-of-the-art knowledge of human geometry, weight, inertia and biomechanical response. There are three accelerometers mounted orthogonally at the center of gravity. Head response is the resultant of these acceleration measurements.
The **neck** consists of three rigid aluminum vertebral elements that are shaped in a durometer butyl elastomer chosen for its high damping characteristic. Molded aluminum end plates allow the attachment to the head and neck bracket (that allows the head levelling) and a steel cable crosses the neck center to provide a high level of axial strength. This design is based on concept developed by the General Motors Research (GMR). There are transducers used to measure loads and moments about the occipital condyle.
The **thorax** consists of a welded steel spine, six steel ribs backed by a poly-viscose damping material to provide the correct chest dynamic response in case of impact and a urethane bib to help distribute loads. There are a triaxial accelerometer located at the assembly center of gravity and a rotary potentiometer (deflection transducer) connected to the sternum through a rod and slider mechanism. This potentiometer measures the longitudinal displacement of the sternum relative to the thoracic spine.
The **lumbar spine** is a curved polyacrylate elastomer member constating of end plates, that allows the attachment to the pelvis and thoracic spine, and of two cables to provide lateral seating stability. This construction provides a human-like seated posture.[**20**] As previously stated, ATDs with all their instrumentation are used in crash tests to evaluate the vehicles safety and there is the need of established relationships between measurements of variables made on the dummy and the probability of a human supporting a specific type and severity of injuries. The process to develop these relationships is called injury criteria.

2.9 Injury criteria

Inflation-induced injuries are defined as those injuries that occur due to the presence of the occupant in the airbag deployment zone at the initiation of its deployment. Less severe injuries (i.e. abrasions, burns and fractures) are not considered as inflation-induced.[**28**] These injuries occur in the second phase of airbag deployment when the inflating airbag has pushed body parts away from the sternum and have been associated with spontaneous force during deployment where the airbag action has caused greater injury than the impact otherwise would have. A higher risk of

injury is associated with particular occupant positions. In fact, injury risk analysis is carried out mainly for OoP tests. These injury criteria have been developed to analyze the mechanical responses of crash test dummies in terms of injury risk to a living human being and to quantify limit values of different body regions. In fact, these criteria link the probability of trauma to mechanical parameters and without them the severity of traumas in accident reconstructions cannot be evaluated. They are based on the engineering principle whereby the internal responses of a mechanical structure are exclusively governed by the structure's geometric and material properties and the forces and motions applied to its surface, regardless of the size.[**29**] The development of human injury tolerance levels is difficult due to physical differences between humans. The criteria have been derived from experimental methods performed on cadaver and animal testing and, only more recently, on crash test dummies. These injury criteria are developed for one size dummies, usually the Hybrid III 50^{th} percentile adult male dummy, and only after they are also applied to the other size dummies through the scaling process described by Mertz et al. (1997), which considers size, mass and tissue properties variations with age.[**30**] An important indicator of injury risk is Abbreviated Injury Scale (AIS): this is an anatomically based injury severity scoring system that classifies, with a static approach, each injury by body region on a 6-point scale.
The AIS score refers to the single injury in a body region, whereas an individual in a car accident has several injuries.
Then, based on AIS score, several injury criteria were developed that led the definition of two classes of injury values, in case of OoP tests:

1. **Injury Reference Values**

2. **Injury Research Values**

The first values have a strong scientific support and are listed in Figure 2.10. The second, instead, have less scientific support or insufficient test experience to underline their validity and are listed in Figure 2.11.
This classification is dictated by the level of scientific understanding which is not the same for every injury risk of potential out-of-position. This is due to the fact that considering injury values with less scientific bases could lead to reject some airbag systems that are, as a matter of fact, more promising at reducing injury risks during cashes.
The next airbag systems are recommended to design according to the Injury Reference Values but, at the same time, also are recommended to consider the Injury Research Values where feasible. In fact, future airbag designs that follow the Injury Reference Values should not be rejected simply to follow Injury Research Values.[**31**]
The most valid method to assess injury risks to humans is the injury risk assessment

based on anthropomorphic dummy responses. However, being an imperfect science, it is still possible to get invalid results.

AIS Score	AIS Evaluation	Injury	Body region
1	Minor	Distortions, minor contusions, abrasions	Head, neck, pelvis, abdomen, thorax, arts
2	Moderate	Sprains, lacerations, contusions, slight fractures	Head, neck, pelvis, abdomen, thorax, arts
3	Serious	Fractures, breaks, severe contusions, concussions	Head, neck, pelvis, abdomen, thorax, arts
4	Severe	Severe lacerations, severe contusions, fractures	Head, neck, pelvis, abdomen, thorax, arts
5	Critical	Concussion, fractures, bleeding, severe lacerations	Head, neck, thorax, abdomen
6	Fatal	Decapitation, laceration, fracture, resection	Head, neck, thorax, abdomen

Table 2.1: Injury severity score and abbreviated injury scale.

2.9.1 Dummy Injury Reference Values

Depending on the airbag design, the body regions subject to injuries are different but the main ones are head, neck and thorax. These injuries should be minimized because cannot be made zero with any inflatable restraint systems.

Reducing injuries can be accomplished, on a practical level, by choosing injury values that would indicate approximately a 5 percent risk of AIS 4 or greater injury for the head and thorax or AIS 3 or greater for the neck. This risk percentage does not mean that the 5 percent risk of injury applies to all occupants, but that if the airbag deploys, the occupant is as severely out of position and if the dummy responses are below the specified injury values, then the risk of serious or severe injury from the airbag is very low. The lower score for the neck injuries is due to the fact that these injuries have been the most common fatal injury in out-of-position interactions.[**31**]

Head Injuries

The only criterion allowed by NHTSA to evaluate head injury risks, according to FMVSS 208, is Head Injury Criterion (HIC). This criterion was developed from Wayne State Tolerance Curve (WSTC) that describes relationship between head acceleration and its duration: the linear acceleration is inversely related to impact duration. The HIC equation derives from power regression analysis of the WSTC.[**30**]
HIC is based on the time history of the linear acceleration of the head center of gravity and, in fact, it is calculated by measurements of an accelerometer mounted at the center of gravity of a crash test dummy's head. HIC is defined as

$$HIC = \max_{t_1,t_2}\{(t_2 - t_1)[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)\,dt]^{2.5}\} \tag{2.1}$$

where t_1 and t_2 are the initial and final times chosen to maximize HIC, a is head acceleration expressed as multiple of g. This formula confirms the relationship described above. The time duration, t_2 - t_1, is limited to a maximum value of 36 ms due to a better assessment of short-duration impacts hazard and a proper measurement of longer events. To simplify the calculation, it is recommended to use 15 ms as time interval of the search for the maximum HIC value. This reduction was followed by a lowering of the corresponding threshold HIC value. For the Hybrid III 50^{th} percentile adult male dummy, 700 is estimated to represent a 5 percent risk of a severe injury. To obtain the Injury Reference Values for the other sizes of dummies, it is necessary to scale with a method that uses geometric and material failure scaling.

Neck Injuries

The neck injuries are the most common fatal injuries in the out-of-position interactions because of rupture of the connective tissues between the head and neck (occipital condyles). Also, these injuries are a result from direct impact to the head or occur via inertial loading from the head.
To evaluate neck injury risk there are two approaches:

1. To impose a limit on the peak force and moment values measured by the upper neck load transducer.
2. To impose a limit on an index, N_{ij}, that is a linear combination of axial loads and bending moments.

The criterion allowed by NHTSA, according to FMVSS 208, is the use of the index N_{ij} due to its strong foundation in biomechanics. The performance limit of this index is 1.0 in any of the four loading mechanisms: compression-extension, compression-flexion, tension-extension and tension-flexion.[**31**]
The N_{ij} can be expressed as sum of the normalized loads and moments

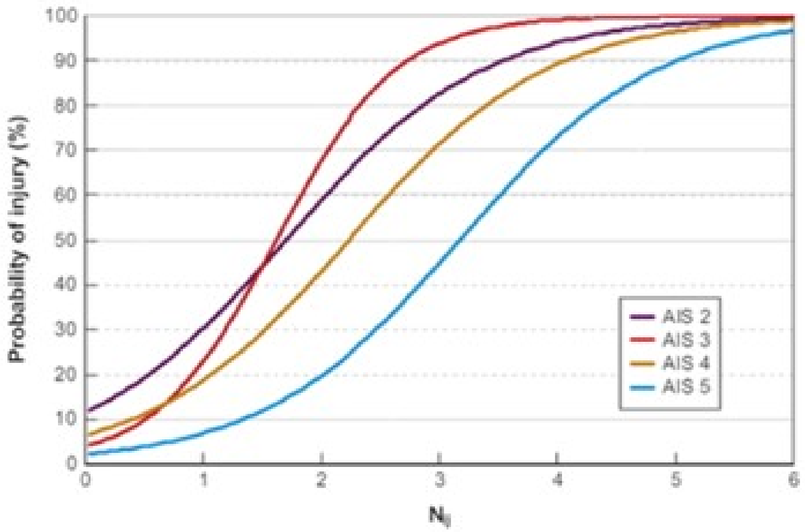
$$N_{ij} = \frac{F_Z}{F_{int}} + \frac{M_Y}{M_{int}} \tag{2.2}$$

Where F_Z is the axial load (tension or compression), M_Y is the flexion-extension bending moment, F_{int} and M_{int} are respectively the critical intercept value of load and of moment used for normalization. The following table shows the N_{ij} thresholds for the different loading modes adopted in FMVSS 208 for evaluation with the Hybrid III dummy family.

Dummy	**Axial load (N)**	**Flexion (Nm)**	**Extension (Nm)**
50^{th} %ile male	6806 tension 6160 compression	310	135
5^{th} %ile male	4287 tension 3880 compression	155	67
3 Year-Old Child	2120	68	27
6 Year-Old Child	2800	93	37

Table 2.2: N_{ij} thresholds values currently in use in FMVSS 208.

The critical intercept moments were scaled up and down to all other dummy sizes, while the critical intercept loads values were only scaled from the three-year-old. The choice of the limit value is dependent on the test condition. For OoP testing, the limit values for N_{TE} are lower than their in-position testing limit to reduce the occurrence of OoP injuries in airbag deployment accidents by placing a more severe restriction on OoP testing of airbag designs. According to this American regulation, unit value of this index represents a 22 percent risk of AIS 3+ injury for any type of occupant. The following figure shows AIS curves as a function of N_{ij}:



The Alliance developed injury risk curves for tension-extension moment and imposed the limit for this moment at 2 percent risk of AIS $\geq$ 3 neck injury, based on animal tests. In addition to this, the Alliance proposed to limit the peak tension and the peak compression: the first, was set at 3 percent risk of AIS $\geq$ 3 neck injury while the second, was set as Injury Assessment Reference Value.[**31**]

Thoracic Injuries

Thoracic injuries are associated with a significant mortality rate because of the potential damage to vital anatomic organs contained within the thoracic cavity.
The measurement of chest deflection is critical for the assessment of thoracic injury since deflection and the viscous response are two of the most commonly accepted criteria for chest injury, in particular the chest deflection is used in FMVSS 208, and the viscous criteria is used in European regulations. The chest compression and the compression rate are treated as thoracic Injury Reference Values for out-of-position testing of airbags. AAMA recommended the chest compression threshold in out-of-position conditions to be 64 mm for the Hybrid III 50^{th} percentile male which corresponds to a 5 percent probability of an AIS $\geq$ 4 thoracic injury. For the 3-year-old and 6-year-old dummies, the values chosen are based on AAMA recommendation, in 1998, for frontal airbag out-of-position testing and these values are scaled: their IARVs for 5 percent risk of AIS $\geq$ 3 thoracic injury surpass their IARVs for 5 percent risk of AIS $\geq$ 4 injury due to the AIS = 3 classification is based on rib fractures, and the ribs of children can be subject to large deflections without fracturing. For the SID-IIs, the chest compression rates are similar for frontal and side impacts, and this has been observed through tests based on animal.

So, the Injury Reference Value for the chest compression rate is the same of the Hybrid III 5^{th} percentile female associated with approximately a 5 percent risk of AIS ≥ 4 thoracic injury in frontal impacts.[**31**]

Chest Deflection (Compression) Rate

There are two methods to determine chest deflection rate:

1. Integration of the difference between rib/sternum and spine accelerations.

2. Differences between the deflection data from the sternum or ribs.

These should give the same result due to noise or delayed measures and are discussed in detail in the Appendix B of TWG, that states that both methods can be used to assess the chest compression rate, but recommends that if the differentiation method is used, the result is verified by performing the integration method as well.

In the following figure Injury Reference Values for OoP for each size dummies obtained by scaling process are listed:

Body Region/Injury Measure	Dummy			
	Hybrid III 3-Year-Old Child	Hybrid III 6-Year-Old Child	Hybrid III Small Female	SID-IIs
Head				
15 ms HIC	570	723	779	779
Upper Neck				
N_{ij}	1	1	1	1
Intercepts				
F_T (N)	2120	2800	3880	3880
F_C (N)	2120	2800	3880	3880
M_F (Nm)	68	93	155	155
M_E(Nm)	27	37	61	61
Tension (N)	1130	2070	2070	2070
Compression (N)	1380	1820	2520	2520
Thorax				
Deflection (mm)	36	40	-	34
Deflection rate (m/s)	8.0	8.5	-	8.2

Figure 2.10: Dummy Injury Reference Values for Out-of-Position Testing

2.9.2 Dummy Injury Research Values

The Injury Research Values are critical indicators of potential injury but there is not enough biomechanical and scientific support to determine the correctness of this values predicting injury risks. For this, additional research and work are needed to provide acceptable test experience to obtain their accuracy. It is necessary that a wide range of scientific data confirms the correct value and not only verifies the value published as Injury Research Value. The same Injury Research Value can become an Injury Reference Value, if it is sufficiently well understood. So, the TWG should periodically review the scientific status of these Injury Values. The manufacturers and suppliers are recommended to consider these values, where feasible.[**31**]

Upper Neck Load Cell

There were not many experiences measuring the lateral bending and twisting of the neck on the dummy and the Injury Research Values recently proposed were based on the judgment of the biomechanics experts. The twist moment values and the extension values were at the same level while the lateral bending moment values were set midway between the extension and flexion values because the amount of muscle and connective tissue that resists lateral bending is greater than the amount that resists extension bending, but not greater than the amount that resists flexion. For the neck, the weakest bending mode was the extension and the strongest bending mode was the flexion.[**31**]

Lower Neck Load Cell

After conducting some research, it was observed that the upper neck load cell measurements were not the best indicators for neck injury risks. So, injury indicators measured at the lower neck load cell could be considered as Injury Research Values.

Assuming that the ratios of upper to lower bending moments and the ratios of upper to lower extension moments share the same values, the Injury Research Values for lower neck flexion and lateral bending moment limits were calculated from the corresponding values for the upper neck.
The Injury Research Values for extension bending moment were obtained for the other dummies scaling, through procedures given by Mertz et al. (1997), the IARV based on research reported by Prasad, Kim, and Weerappuli (1997) in which this value corresponds to the lowest force observed in cadaver spines. This recommended value was 154 Nm for the Hybrid III 50^{th} percentile male dummy, obtained by multiplying the corrected upper neck moments values with a ratio of 3.28. This should provide a very low risk of AIS 3+ injury.
For lower neck tension forces, compression forces and twist moment the Injury Research Values are the same as for the upper neck because there is not biomechanical or anatomical reason to believe that these forces could cause different risks to

upper or lower neck.[**31**]

Thoracic Injuries

The spine acceleration was considered as an Injury Research Value in OoP testing, although not necessarily indicating a thoracic injury in the absence of excessive compression or compression rate. The values showed are scaled from the value specified by FMVSS 208 for the Hybrid III 50^{th} percentile male dummy. NHTSA has established a tolerance level for the resulting accelerations on the thorax: this parameter is measured with a triaxial accelerometer placed on the dummy spine and its value is 60 g in the 3 ms interval.[**31**]

Abdominal and Pelvic Injuries

The only dummy that allows to measure abdominal and pelvic injuries is the SID-IIs.
The measures to be performed on abdominal are compression and compression rate: the first was obtained scaling the value of the Hybrid III 50^{th} percentile male dummy (39 mm) proposed based on cadaver and animal testing while the second was the same value used as Injury Reference Value for thoracic compression rate.

There are two injury criteria that have been used in OoP testing for the pelvis: the pubic symphysis force and the iliac crest force that representing the force levels at which fractures can occur. These values were obtained scaling from IARVs suggested for the BioSID based on cadaver testing.[**31**]

Arm Injuries

There are two injury criteria required for the upper extremities: the bending moments for the humerus (upper arm) and for the ulna (forearm). Furthermore, in this case, the only dummy that allows to measure moments of ulna and humerus is the SID-IIs. The value of bending moment of the ulna was obtained scaling down IARV for 50^{th} percentile male dummy, based on the two different average moments: 89 Nm derived from the work of Begeman et al. (1999) measured from a sample of adult cadavers and 94 Nm derived from the work of Pintar et al. (1998). The value of bending moment of the humerus was recommended by Kirkish et al (1996) for the SID-IIs with instrumented arm.[**31**]

In the following figure Injury Research Values for OoP for each size dummies obtained by scaling process are listed:

Body Region/Injury Measure	Dummy			
	Hybrid III 3-Year-Old Child	Hybrid III 6-Year-Old Child	Hybrid III Small Female	SID-IIs
Upper Neck				
Lateral moment (Nm)	30	42	67	67
Twist moment (Nm)	17	24	39	39
Lower Neck				
Flexion moment (Nm)	83	119	190	190
Extension moment (Nm)	34	48	77	77
Lateral moment (Nm)	60	84	134	134
Twist moment (Nm)	17	24	39	39
Tension (N)	1330	1490	2070	2070
Compression (N)	1380	1820	2520	2520
Thorax				
Spine acceleration (max g, 3 ms)	55	60	-	73
Abdomen				
Deflection (mm)	-	-	-	32
Deflection rate (m/s)	-	-	-	8.2
Pelvis				
Pubic symphysis load (N)	-	-	-	4000
Iliac load (N)	-	-	-	4000
Arm				
Resultant bending moment, ulna (Nm)	-	-	-	54
Resultant bending moment, humerus (Nm)	-	-	-	130

Figure 2.11: Dummy Injury Research Values for Out-of-Position Testing

2.9.3 Scaling Process

As previously declared, the injury criteria are developed for one size dummy, i.e. Hybrid III 50^{th} percentile adult male dummy and are translated to other size dummies through a process called scaling influenced of geometric and material differences. This process was born out of the need not to limit biomechanical approach to vehicle safety to 50^{th} percentile adult male but extending it to the entire population exposed to risk. The scaled values are good approximations of the expected values demonstrated by engineering experience.[**32**]

The most common scaling process is dimensional analysis. This kind of process allows the unknown responses of a given system to be estimated from the known responses of a similar system. For structural analysis three fundamental scaling factors have been established that based on ratios between properties characterizing these two systems: length, mass density and modulus of elasticity (or stiffness).[**29**]

These dimensionless ratios are defined as

1. *Length scale ratio* $\lambda_L = L_1/L_2$
2. *Mass density ratio* $\lambda_\rho = \rho_1/\rho_2$
3. *Modulus of elasticity ratio* $\lambda_E = E_1/E_2$

Where the subscript 1 refers to the model to be scaled to, and the subscript 2 refers to the standard to be scaled from, in this case Hybrid III 50^{th} percentile adult male dummy.

The modulus of elasticity and mass density are assumed equal when scaling biomechanical data between adult subjects. This assumption is due to the geometric similitude and all data can be scaled as function of length scale ratio. When scaling data from adults to children, the assumption of the same value of two modulus of elasticity is no longer valid because of the developing mechanical properties of the body tissues in children. The other quantities associated with impact responses can be formed by combining these ratios, assuming equal density (body mass specific weight not change with age):

1. *Mass* $\lambda_\rho = \lambda_3^L$
2. *Time* $\lambda_T = \lambda_L/\lambda_E^{1/2}$
3. *Velocity* $\lambda_V = \lambda_E^{1/2}$
4. *Acceleration* $\lambda_A = \lambda_E/\lambda_L$
5. *Force* $\lambda_F = \lambda_L^2\lambda_E$
6. *HIC* $\lambda_{HIC} = \lambda_E^2/\lambda_L^{1.5}$

These definitions of quantities allow to define the response of one subject size based on measurements of another subject size and are characterized by stress equal. In reality, the stress levels and failure strain of biological tissue may be age dependent, hence it is more correct to scale failure threshold levels by the failure stress λ_σ or strength ratio. In fact, failure stress ratio was used in the scaling process between various dummy sizes and, for this reason, there is a change in some scaling relationships listed above.[**29**]

1. *Failure Strength* $\lambda_\sigma = \sigma_1/\sigma_2$
2. *Acceleration* $\lambda_A = \lambda_\sigma/\lambda_L$
3. *Force* $\lambda_F = \lambda_L^2\lambda_\sigma$
4. *Moment* $\lambda_M = \lambda_L^3\lambda_\sigma$
5. *HIC* $\lambda_{HIC} = \lambda_\sigma^{2.5}/\lambda_L^{1.5}$

To scale the biomechanical parameters used as Injury Assessment Reference Values, it is necessary to specify the relative lengths of the body regions of interest and the relative modulus ratios.

Chapter 3

Protocol OoP CeAB

The Alliance of Automobile Manufactures (Alliance) together with the Association of International Automobile Manufacturers (AIAM), Automotive Occupant Restraints Council (AORC) and Insurance Institute for Highway Safety (IIHS) have sponsored the Technical Working Group (TWG) for the developing of a protocol that analyzes the risks associated with Side Airbag (SAB) and Curtain Airbag (CAB) deployments and describes how to minimize those risks. The need for this protocol arises because of the aggressiveness of side and curtain airbags. During their deployment, these airbags release an amount of energy that results in greater forces between the airbag itself and the nearby occupant than intended. Therefore, this protocol describes the instrumentation, test devices and procedures, performance guidelines and injuries criteria that should be used to assess the out-of-position risks of interaction between SAB-CAB and a vehicle occupant. The test procedures described in the protocol offer as complete an evaluation as possible for current state-of-the-art airbag designs.

For Center Airbags, there is not a protocol in place recommending procedures for assessing out-of-position injury risk. For this reason, with ZF as supplier, together with Italdesign, I proposed to use a testing workshop to compile a protocol dedicated to CeAB, based on ‘Recommended procedures for evaluating occupant injury risk from deploying side airbags’ prepared by TWG.

The recommendations proposed by the TWG address three fundamental areas:[**31**]

1. The test devices (dummies) best suited for assessing OoP injury risk from the close-range deployment of CeAB.

2. Performance criteria against which to assess the injury risk indicated by the forces measured on the dummies.

3. A standard set of test procedures (occupant positions) for assessing CeAB inflation-injury risk associated with different airbag designs.

In order to draft the protocol, it was necessary to visit ZF headquarters in Alfdorf, Germany. The specifics for the dummy seating positions were laid out in the company's laboratory using a draft document provided by the client as a starting point. First, the document defines the test objective – which is to establish the highest interaction between a body region of the dummy and the airbag module - and the position of the dummy. Once a local reference system (x, y, z) of vehicle is established, the next step is to take measures for each position so as to ensure repeatability of the test with another vehicle. Next, a static test is executed to simulate the impact and, after, airbag deployment allowing to calculate the injury risk curves. This test also allows to examine whether the inflatable restraint system would ensure the vehicle occupants even in a worst- case scenario.
The risk of CeAB inflation-injury have been assigned using dummies representing the small female and adolescent (SID-IIs), the 6-year-old child, and the 3-year-old child.

3.1 Dummy Preparation

3.1.1 General

The dummy should be in great condition and able to meet its performance requirements. The dummy wears tight fitting cotton knit shirt and pants. It is possible, to prevent the CeAB from getting caught in the seam, by using electrical tape or by applying 4 mm tape on the skullcap seam. To achieve acceptable frictional characteristics, baby powder is used and whereas alcohol is used to clean the dummy's head skin.[**31**]

3.1.2 Test Temperature

The temperature should be within a temperature range of 20.6-22.2 °C and a relative humidity of 10-70 percent after a soak period of at least 4 hours prior to its application in a test.[**31**]

3.1.3 Instrumentation

Through accelerometers, load cells and transducers it is possible to obtain injury risk measures and all measurements should be recorded by high-speed cameras with a minimum speed of 1000 frames per second (3000 fps is recommended) and, later, should be filtered. These cameras should be positioned so that to capture the entire field of vision. The recommended instrumentation and relative measurements, for each dummy, are shown in the following figure:

Dummy	Body Region	Instrumentation Measure
Hybrid III 3-Year-Old Child Dummy	Head	3 accelerations (x, y, z)
	Neck	
	Upper	3 forces and 3 moments (x, y, z)
	Lower	3 forces and 3 moments (x, y, z)
	Thorax	
	Upper spine (~T1)	3 accelerations (x, y, z)
	Sternum	
	Upper	1 acceleration (x)
	Center	1 deflection (x)
	Lower	1 acceleration (x)
	Spine (~T4)	3 accelerations (x, y, z)
	Spine at level of Rib 3	1 acceleration (x)
	Lower spine (~T12)	3 accelerations (x, y, z)
Hybrid III 6-Year-Old Child Dummy	Head	3 accelerations (x, y, z)
	Neck	
	Upper	3 forces and 3 moments (x, y, z)
	Lower	3 forces and 3 moments (x, y, z)
	Thorax	
	Upper spine (~T1)	1 acceleration (x)
	Sternum	
	Upper	1 acceleration (x)
	Center	1 deflection (x)
	Lower	1 acceleration (x)
	Spine at level of Rib 1	1 acceleration (x)
	Spine (~T4)	3 accelerations (x, y, z)
	Lower spine at level of Rib 6	1 acceleration (x)
Hybrid III 5th Percentile Adult Female	Head	3 accelerations (x, y, z)
	Neck	
	Upper	3 forces and 3 moments (x, y, z)
	Lower	3 forces and 3 moments (x, y, z)
SID-IIs	Head	3 accelerations (x, y, z)
	Neck	
	Upper	3 forces and 3 moments (x, y, z)
	Lower	3 forces and 3 moments (x, y, z)
	Thorax	
	Upper spine (~T1)	3 accelerations (x, y, z)
	Ribs	3 accelerations (y) and deflections (y)
	Spine box, opposite each rib	3 accelerations (y)
	Abdomen	
	Ribs	2 accelerations (y) and deflections (y)
	Spine box, opposite each rib	2 accelerations (y)
	Lower spine (~T12)	3 accelerations (x, y, z)
	Pelvis	3 accelerations (x, y, z)
	Acetabulum	1 force (y)
	Pubic symphysis	1 force (y)
	Iliac	1 force(y)
5th percentile Arm	Humerus	2 moments (x, y)
	Ulna	2 moments (x, y)

Figure 3.1: Test Devices (Dummies) and Recommended Instrumentation for Assessing OoP Injury Risk for CeAB

3.1.4 Electrical Grounding

All of this instrumentation, vehicle and test devices must be grounded. The cables attached to the dummy's head, thorax, abdomen and pelvis shall always be connected to earth ground. There is a high likelihood for electrostatic discharges because of the inflating airbag and to prevent this, between tests, an anti-static spray is used.[**31**]

3.2 Test Procedures

The tests specified for the SID-IIs are relevant to driver and passenger seating positions, while those for the 3-year-old and 6- year-old child dummies are relevant only to passenger positions. The test dummy positions were chosen to block the deployment path of the deploying airbag and also to align the dummy's measurement systems to determine the effects of the resulting dummy-airbag interaction. Evaluations should be conducted with representative seats and center console located in the vehicle design position. In some positions, there should be a different load case due to the difference of vehicle design, such as absence of or lower center console.
These test positions represent nominal 'worst case' scenarios. Each manufacturer should verify if this requirement is satisfied and adjust the test positions accordingly in case of negative response.[**31**]

3.2.1 General seat preparation procedure

This is followed by a set of instructions that apply to the vehicle seat and, also in this case, test engineers can modify the instructions to make them more consistent with their particular system and meet the objectives of the individual test:[**31**]

1. To aid dummy positioning, identify and mark the centerline of the seat back and seat cushion. For the CeAB, draw a horizontal line on the seat corresponding to the top edge of the airbag module.

2. Tests are to be conducted with the seat in the rearmost and lowest adjustment. The seat back should be adjusted to the manufacturer's design angle or to achieve a torso angle of 25 degrees as measured on the SAE J826 H-Point machine (in our case, a torso angle of 21 degrees). If any of these adjustments is found to interfere with the inflation of the airbag or with the stated test objective, then the seat track position and or seat back angle may be adjusted the minimum amount necessary to avoid obstruction and fulfill the required test objective with the seat still in a nominally normal position for travel.

3. The head restraint is adjusted to its full-down position.
4. The center console is adjusted to its lowest position to avoid interaction with the CeAB.
5. All windows on the tested (inflation) side of the vehicle should be in the closed position, unless otherwise specified.

In the table are shown for each of the three test devices, the test positions to assess Out-of-Position injury risk for the CeAB airbag:

Dummy	Section	Test Position	Body Region
Hybrid III 3-Year-Old Child	3.3.3.2	Rearward facing with knees on Center Console	Head, neck, thorax
	3.3.3.2	Rearward facing	Head, neck, thorax
	3.3.4.1	Inboard facing with arm on Center Console	Head, neck, thorax
Hybrid III 6-Year-Old Child	3.3.3.3	Lying on seat with head on Center Console	Head, neck
SID-IIs	3.3.3.6	Outboard facing	Head, neck, thorax, abdomen, pelvis
	3.3.3.7	Arm on Center Console with Instrumented Arm	Arm,forearm

Table 3.1: Recommended Test Procedures.

Each test position was performed with normal seat, but additional tests were carried out with sport seat, also to confirm the validity and accuracy of results, for the first three positions.

3.3 Test Positions

As previously stated, the six OoP positions of the dummy are described in detail in the drafting of the protocol of CeAB and of which the biomechanical analysis of the curves obtained from the static test will be carried out in the following chapter.

3.3.3.2. Rearward facing with knees on Center Console (Passenger Positions with CeAB)

Figure 3.2: Rearward facing Hybrid III 3-Year-Old Child Dummy with knees on center console

Test Objective: To maximize the chest interaction by aligning the sternum with the top of the center airbag module and the dummy head with the centerline of the vehicle.

The dummy is placed on the center console with its head centered between the two headrests, kneeling and facing rearward. Its wrists are in contact with the headrest so as to have the chest as closer as possible to CeAB module. The dummy must be stable: it is possible with the position of knees and aligning the legs parallel to center console. Then it is possible to deploy CeAB and record the dummy channels.

3.3.3.2. Rearward facing (Passenger Positions with CeAB)

Figure 3.3: Rearward facing Hybrid III 3-Year-Old Child Dummy

Test Objective: To maximize the head interaction with CeAB by aligning the head center of gravity with the top of the center airbag module.

The dummy is placed on the passenger seat in a kneeling position facing the driver seat and keeping the head in its neutral orientation. The dummy should be leaning inboard until its head contacts the top edge of CeAB module and the chest with the center console. The dummy arms and hands should be hanging against the sides of the torso (left arm in contact with seat back and right arm in contact with center console). Then it is possible to deploy the CeAB and record the dummy channels.

3.3.4.1.Inboard facing with arm on Center Console (Passenger Positions with CeAB)

Figure 3.4: Inboard facing Hybrid III 3-Year-Old Child Dummy with arm on center console

Test Objective: To maximize the head/neck interaction with CeAB by aligning the head center of gravity with the deployment path of the airbag.

The dummy is placed on the passenger seat in a kneeling position facing the driver seat by aligning the head center of gravity with side seam of the seat in the direction of the airbag module. The dummy arms are placed on the center console in such a way as to give the dummy stability (left elbow in contact with seat backrest) while the legs are misaligned. The dummy pelvis should be closer as much as possible to the center console so that the chest is in contact with the latter. After, it is possible to deploy the CeAB and record the dummy channels.

3.3.3.3. Lying on seat with head on Center Console (Passenger Positions with CeAB)

Figure 3.5: Lying on seat Hybrid III 6-Year-Old Child Dummy with head on center console

Test Objective: To maximize the head/neck interaction with CeAB by aligning the head center of gravity with the deployment path of the airbag.

In this position, it is possible to use a booster foam block to achieve the positioning requirement and prop up the dummy pelvis. The dummy is placed on the passenger seat lying on its left arm. Its back is in contact with the seatback and the right shoulder with the center console. The head is on the center console so as to achieve the test objective. In addition, adjust the torso so that it creates 90 degrees with the thighs and the forearms to an orientation 45 degrees to the upper arm. The passenger seat height adjusts itself until the dummy head and pelvis are aligned. Afterwards, it is possible to deploy the CeAB and record the dummy channels.

3.3.3.6. Outboard Facing (Driver or Passenger Positions with CeAB)

In this position, due to the difference of vehicle design, there are two different load case:
1. Driver seat - Presence of a center console higher than the seat cushions;
2. Passenger seat - Loadcase 1 is not verified or absence of center console.

Figure 3.6: Outboard facing SID-IIs (Loadcase 1)

Figure 3.7: Outboard facing SID-IIs (Loadcase 2)

Test Objective: To maximize chest interactions by aligning the center of the upper thoracic rib with the top edge of the airbag module.

Disassemble the front door and, based on a loadcase, adjust the seat forward to allow for a dummy body horizontal configuration. Position the dummy sitting facing toward the outboard of the vehicle with its arm against the seatback and adjust the relative seat rearward until the dummy leg touches the vehicle. Keeping the horizontal orientation for the arm, slide the dummy's pelvis inboard until the dummy back is in contact with the center console, in presence of the latter or the dummy torso is aligned with deployment path of CeAB, in presence of the loadcase 2. Verify that the dummy position meets the test objective. Once all these instructions have been carried out, it is possible to deploy the CeAB and record the dummy channels.

3.3.3.7. Arm on Center Console with Instrumented Arm (Driver Positions with CeAB)

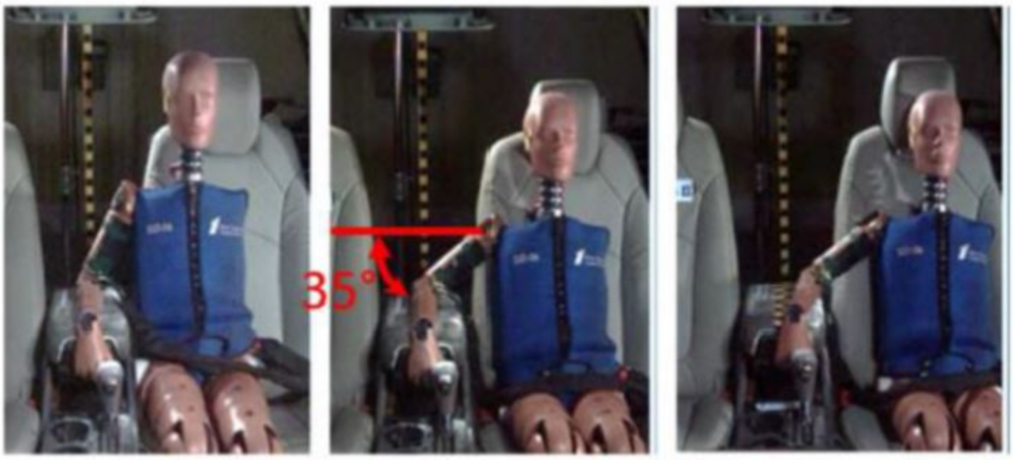


Figure 3.8: Arm on Center Console with Instrumented Arm

Test Objective: To maximize upper arm interaction with the CeAB module by adjusting the seat to the mid-seat track position.

Adjust the driver seat to the mid-seat track position so that the dummy is placed close the inboard edge of the seat with elbow on the center console. The upper arm is leaning to have a 35-degree orientation relative to the dummy shoulder and the hand should be leaned to have a 45-degree orientation relative to the center console surface. The forearm should be at rest on the center console. Afterwards, the CeAB may be deployed, and the channels of the dummy recorded. For this position, the test is not carried out as it would only result in injury research values and not reference values, therefore the test is considered of less importance for biomechanical analysis.

3.4 Positions Results

As previously mentioned, for each position, significant measurements were taken from the local reference system (x, y, z) and are shown in the tables below.
The local reference system is based on Y plane with the following coordinates: (X = 1000 mm, Z = 1000 mm). The common measures for all positions are related to the seat: height, length adjustment, lean angle and seat-cushion inclination.

Normal seat

3.3.3.2 Rearward facing with knees on Center Console

Length adjustment	**Lean angle**	**Cushion inclination**	**Seat height**
-160 mm (Rear)	72.6°	lowest	lowest

From	**To**	**Measures [mm]**
COG right	Z1000	-210
Head middle	Y0	0
Head top	Z1000	-110
Occipital scr.lower right	X1000	685
Shoulder right	X1000	546
Hip screw right	X1000	322
Hip screw right	Z1000	-330
Knee(distance)	Knee	70
Knee pivot point right	X1000	417
Wristlets	Headrest bar	8
Wristlets	Headrest bar	8
Buttocks back	X1000	270

3.3.3.2 Rearward facing

Length adjustment	Lean angle	Cushion inclination	Seat height
-160 mm	72.6°	lowest	lowest

From	To	Measures [mm]
COG left	Y0	25
COG right	X1000	466
Shoulder right	X1000	334
Shoulder right	Z1000	-400
Knee left	Seat back	22
Pelvis screw right	X1000	224
Head top	Z1000	-278
Pelvis screw right	Z1000	-574
COG right	Z1000	-388

3.3.4.1 Inboard facing with arm on Center Console

Length adjustment	Lean angle	Cushion inclination	Seat height
-160 mm (Rear)	72.6°	lowest	lowest

From	To	Measures [mm]
Head top	Z1000	-292
COG right	X1000	263
COG right	Z1000	-394
Shoulder right	X1000	210
Shoulder right	Z1000	-379
Shoulder screw left	Y0	18
Pelvis screw right	X1000	233
Pelvis screw right	Z1000	-532
Elbow screw right	X1000	197
COG left	Y0	133

3.3.3.3. Lying on seat with head on Center Console

Length adjustment	**Lean angle**	**Cushion inclination**	**Seat height**
-160 mm (Rear)	72.6°	lowest	highest

From	**To**	**Measures [mm]**
Booster block		Dimensions according to TWG
COG right	X1000	-271
COG right	Z1000	-404
COG right	Y0	44
Head top	Seat cheek	108
Shoulder screw right	X1000	18
Shoulder screw right	Z1000	296
Shoulder	Centre arm console	Contact
H-point	Z1000	-370
H-point	Z1000	268

3.3.3.6 Outboard facing

Length adjustment	**Lean angle**	**Cushion inclination**	**Seat height**
-70 mm	72.6°	lowest	lowest

From	**To**	**Measures [mm]**
COG right	X1000	249
COG right	Z1000	-103
Shoulder right	X1000	136
Shoulder right	Z1000	-318
Pelvic screw right (center)	X1000	60
Pelvic screw right (top)	Z1000	-624
Center Console	Back	41
COG left	Y0	287

Sport seat

3.3.3.2 Rearward facing with knees on Center Console

Length adjustment	Lean angle	Cushion inclination	Seat height
-160 mm (Rear)	72.6°	lowest	lowest

From	To	Measures [mm]
COG right	Z1000	-148
Head top	Z1000	-40
Occipital scr. lower right	X1000	629
Shoulder right	X1000	496
Hip screw right	X1000	310
Hip screw right	Z1000	-332
Knee (distance)	Knee	100
Knee pivot point right	X1000	417
Wristlets left and right	Headrest bar	15

3.3.3.2 Rearward facing

Length adjustment	Lean angle	Cushion inclination	Seat height
-160 mm	75.85°	lowest	lowest

From	To	Measures [mm]
COG left	Y0	24
COG right	X1000	455
Shoulder right	X1000	323
Shoulder right	Z1000	-410
Knee left	Seat back	47
Pelvis screw right	X1000	224
Head top	Z1000	-291
Pelvis screw right	Z1000	573
COG right	Z1000	-400

3.3.4.1 Inboard facing with arm on Center Console

Length adjustment	Lean angle	Cushion inclination	Seat height
-160 mm	75.85°	lowest	lowest

From	To	Measures [mm]
Head top	Z1000	-294
COG right	X1000	261
COG right	Z1000-394	396
Shoulder right	X1000	201
Shoulder right	Z1000	380
Shoulder screw left	Y0	18
Pelvis screw right	X1000	222
Pelvis screw right	Z1000	-535
Elbow screw right	X1000	190
COG left	Y0	133
Thorax	Center Console	Contact

Chapter 4

Biomechanical Analysis

Biomechanics is the discipline that applies the principles of mechanics to living organisms allowing to describe the movement of various body segments and to evaluate the forces acting on these segments. Impact biomechanics integrates knowledge of body forces and movements with a systematic understanding of functional anatomy, human physiology and human biological tissue mechanics to explore the possible relationships between external events and associated injuries. It generally addresses the prediction and prevention of injuries to the human body enabling the development of applications in the field of vehicle safety. To achieve this, researchers must have a clear understanding of the mechanism of injury, be able to describe the mechanical response of the tissues involved and have basic information on human impact tolerance, making use of study models to assess injury dynamics.

This branch of biomechanics is the starting point of passive safety, i.e. the search for methods to reduce injuries during a transport accident.
It has been shown in extensive studies, supported by numerous tests on cadavers or volunteers, that there is a direct link between the acceleration produced by the impact on the body segment under examination and the physiological damage caused to it.
In fact, a risk curve corresponds to a statistical modelling of biomechanical data in order to predict the risk of injury. Risk curves relate quantifiable parameters to AIS injury severity scales to assess occupant protection in impact tests.

Chapter 2 defined the limits of the examined injury criteria and, according to the values obtained from the static OoP tests, the percentages. These percentage values determine the color, which in turn enables the severity of the injury to be classified and to determine whether the restraint system is fulfilling its function. The colors are linked to injury levels as follows:

1. Green color: includes values from 0 to 80% and it represents acceptable values and the correctness of the restraint system;

2. Yellow color: includes values from 80 to 100% and it represents a situation in the balance that requires the dummy repositioning and a possible modification of the system;

3. Red color: includes all values greater than 100% and it represents a serious situation where a modification of the restraint system is necessary.

The steps to conduct the biomechanical analysis are the following:

1. **Video analysis**, to verify the accuracy of the position and if test objective has been achieved;

2. **Preliminary analysis of injury criteria**, to visually estimate the body region interacting with the airbag and therefore exposed to the risk of injury;

3. **Airbag vs Dummy**, to determine the contact time (first contact and maximum interaction contact) and to make, based on the interaction, an initial estimate of the forces and moments acting on the dummy;

4. **Curve analysis**, to effectively verify that the position meets the test objective and whether the previous steps are correct;

5. **Test objective curves** and, if necessary, additional curves (highest values).

The observation and data recording intervals are of 300 ms, but the injury criteria specified in FMVSS 208 must be met if calculated from data recorded within 100 ms after initial airbag deployment. Typically, the maximum recorded data occurs in the first half (tens of ms) of this latter interval, as there must be proximity between the airbag module and the body region of interest to fulfil the test objective and maximize the desired interaction.

3.3.3.2 Rearward facing with knees on Center Console

Normal seat

In this case, the test objective involves chest interaction with CeAB module. From the first video analysis, it emerges that the reliability of position is optimal, but the desired interaction is not maximal. In order to maximize it, the dummy should have been rotated towards the module. However, this would have caused the center line of the positioning to be lost. Hence, the dummy repositioning would have proved difficult.

From the preliminary analysis of the injury criteria, it is clear that the interaction with the head and the pelvis is zero. The most affected by injury body regions are the upper neck and the thorax. The first contact is at approximaly 10 ms and the maximum unfolding around 17 ms.

From the video observation, it is possible to see that the airbag deployment transmits to the neck a tractive force ($F_z > 0$) and its main movement is extension ($M_y < 0$). From the curve analysis, precisely N_{TE} curve, it is possible to verify that the preliminary analysis is correct: the dummy is subject to a tension-extension along Y and its maximum is around 14 ms. The obtained maximum value from N_{TE} is 25%.

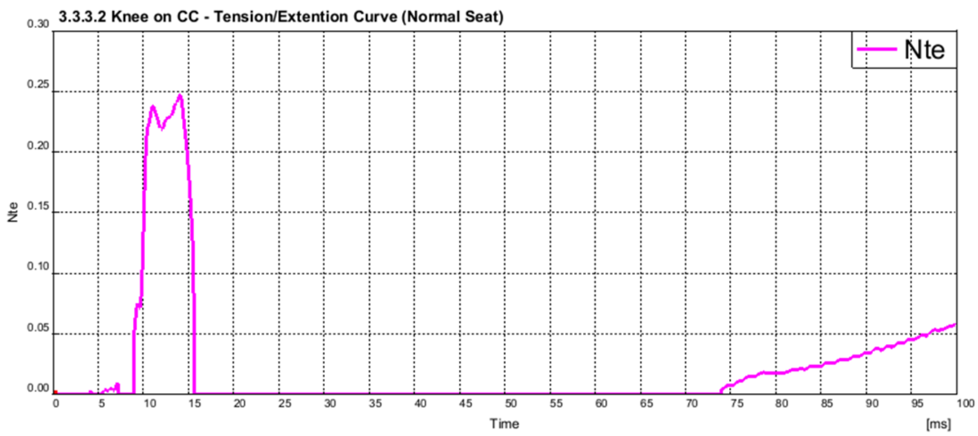


Figure 4.1: Tension-Extention Curve N_{TE}

To better understand this last curve, it is possible to superimpose two curves: M_y and F_z. The force's contribution is bigger than that of the moment from 8 to 16 ms but the amplitude is intensified by the presence of the latter. In fact, it is possible to verify that the initial peaks of the force and N_{TE} curves coincide.

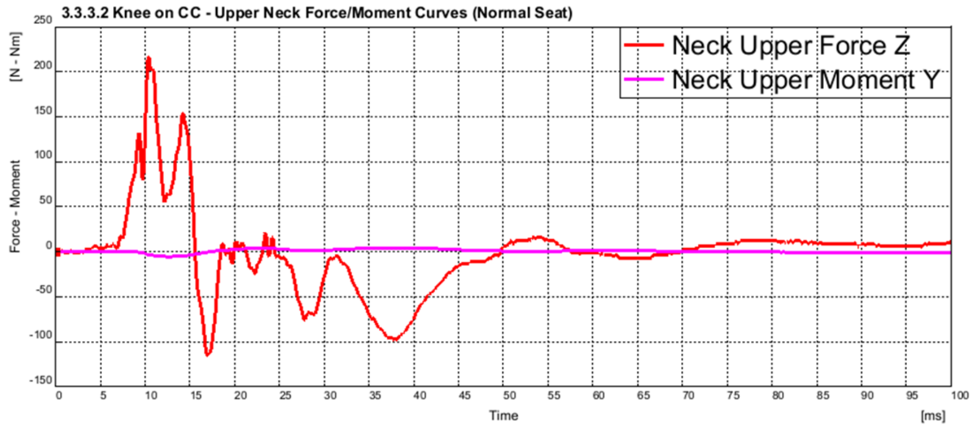


Figure 4.2: Upper Neck Force and Moment Curves

Regarding the body region of interest of test objective, the highest value is relative to the chest deflection rate around 20 ms in which the maximum deployment of the airbag occurs. In the following graphic, there are two curves: the green curve is obtained by integration of the acceleration and the orange curve by derivation of the displacement. The two curves have a similar trend, but the green one is 'clearer' because of the integration operation allows to keep more signal and remove noise. In this case, to calculate the maximum deflection rate, the orange curve is chosen because it represents the worst-case.

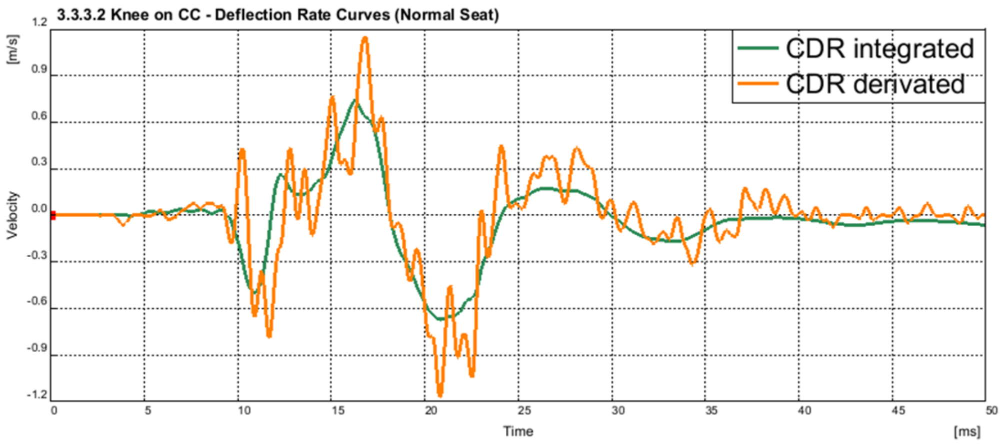


Figure 4.3: Deflection Rate Curves

Sport seat

As in the previous case, the test objective involves the chest interaction but, in this case, the desired interaction is maximal. The center line of positioning is maintained: the seat design allows the dummy to maintain a higher position and to have a shorter distance between dummy chest and airbag module. In this way, the airbag has less space for its deployment and will exert a greater force on the chest.

Also in this case, from the preliminary analysis of the injury criteria, it is clear that there is no interaction with the head and the pelvis, and the greatest interaction occurs for the neck upper and for the thorax. The first contact is at approximaly 11 ms and the maximum unfolding around 19 ms.

From the curve analysis, precisely chest curves, it is possible to verify that the preliminary analysis is correct: the airbag exerts a force on the chest which results in maximum deflection along X at 19.30 ms and the respective deflection rate at 20.85 ms with a value of 27.8%.

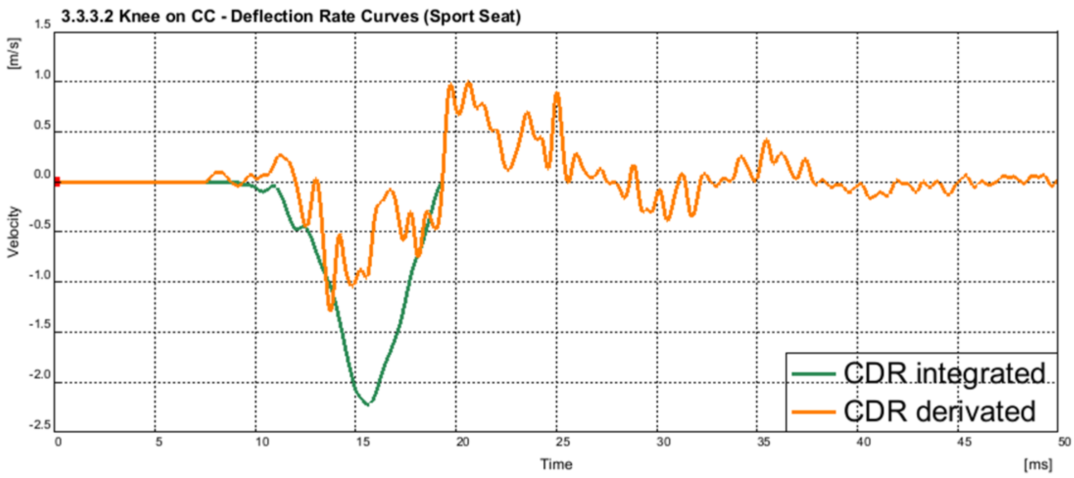


Figure 4.4: Deflection Rate Curves

Another high value obtained from the curves is that relating to the upper neck, index N_{TE}: the dummy is subject to a tension-extension along Y and its maximum is at 15.25 ms. The neck initially undergoes a negative rotation with its maximum at 15.25 ms and subsequently a tractive force with its maximum at 18.50 ms.

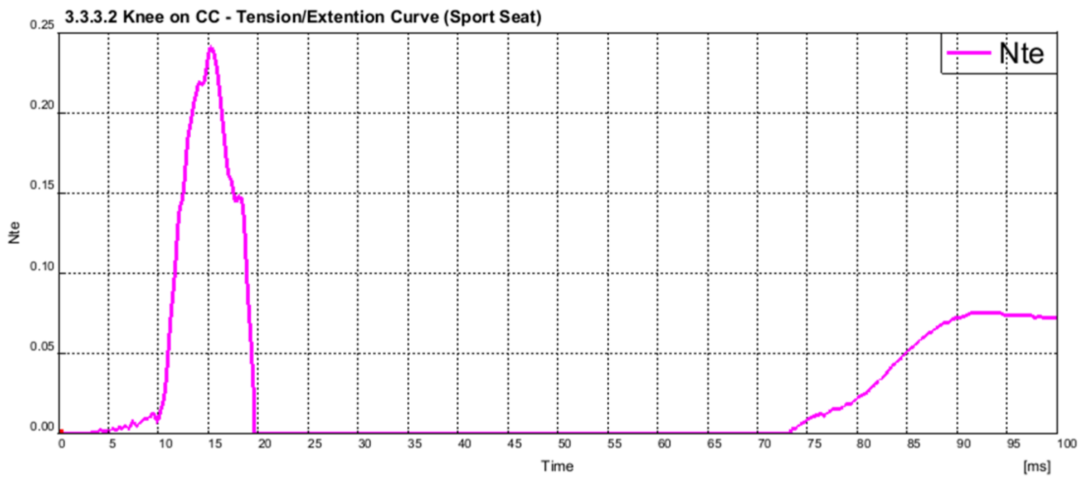
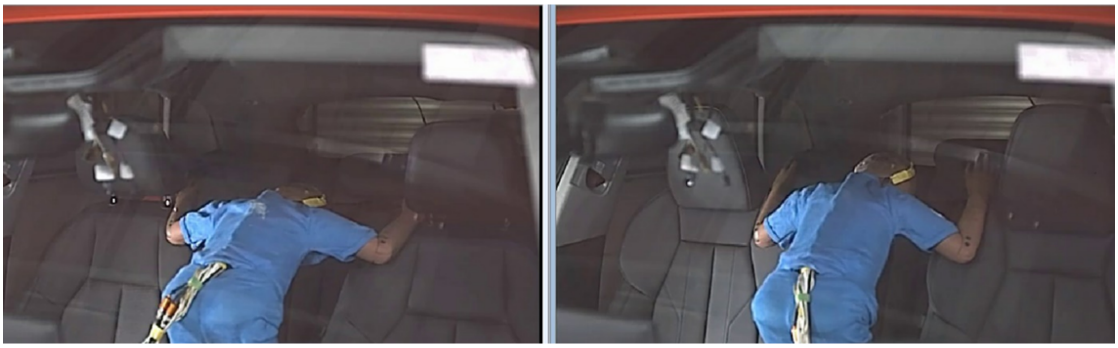


Figure 4.5: Tension-Extention Curve N_{TE}

COMPARISON: Normal Seat vs Sport Seat



In this figure, it is possible to see that the dummy positioning is different: with normal seat the arm arch is greater than with sport seat and for this reason, the distance between the dummy chest and the airbag module is shorter in case of sport seat. The seat design, for the first case, allows a faster airbag deployment. This is possible to see in this comparison graphic: the blue curves represent the case of normal seat while the red curves represent the case of sport seat. The curves morphology is similar, but in case of normal seat is anticipated given the faster deployment and so an anticipated contact of bag with the dummy chest.

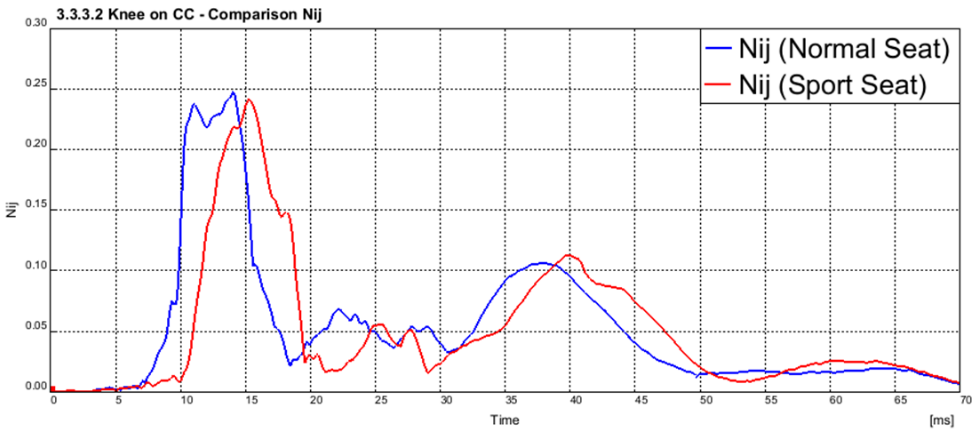


Figure 4.6: Comparison N_{ij}

Regarding the curves of the thorax (deflection rate), the sport seat presents a higher percentage: respectively of 27.8% in comparison of 8.4% of the normal seat. This can be explained as the space available for the airbag deployment is smaller, because of the dummy position, and the airbag becomes partially trapped exerting a greater force on the dummy chest.

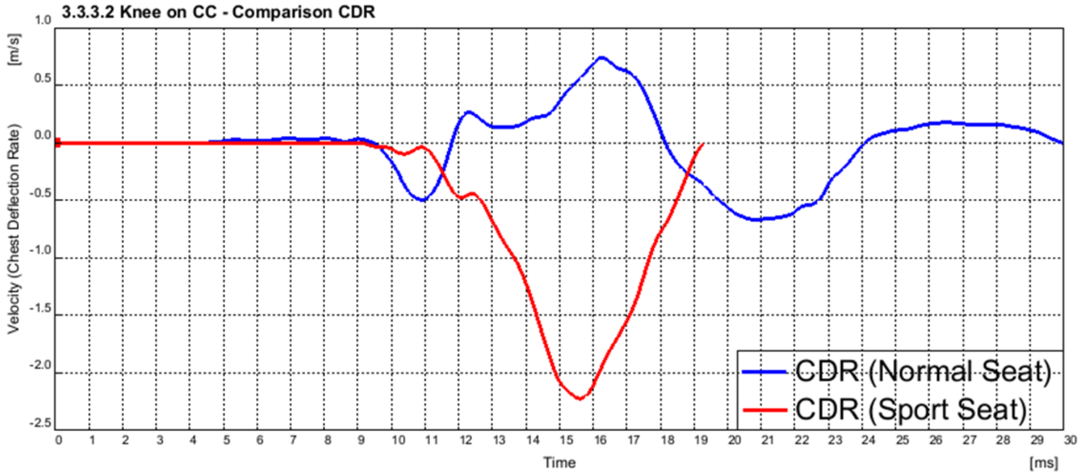


Figure 4.7: Comparison Deflection Rate Curves

In conclusion, it is possible to state that the impact with a normal seat is more severe.

3.3.3.2 Rearward facing

Normal seat

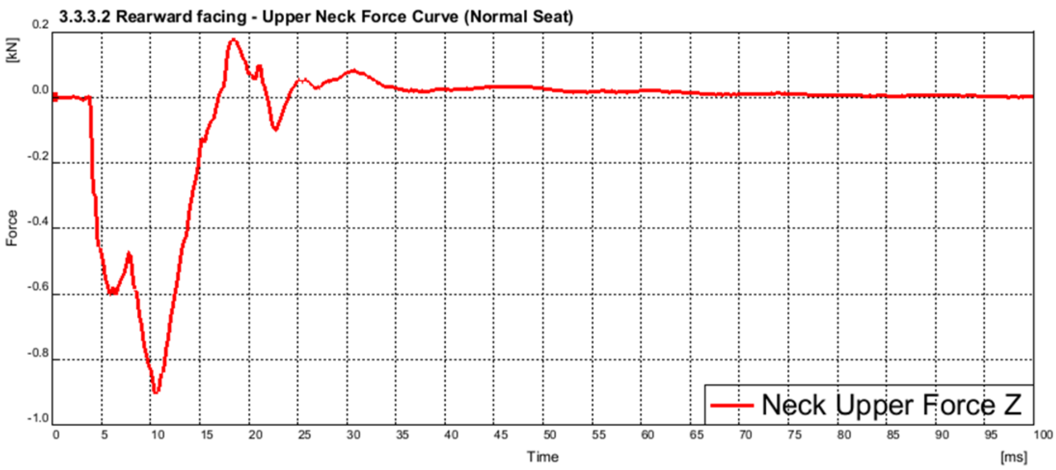


Figure 4.8: Upper Neck Force Curve

The test objective requires maximizing the head interaction with CeAB by aligning the head center of gravity with the top of the center airbag module. From the first video analysis, it can be seen that the maximum interaction occurs with the head and neck while the chest interaction is almost zero as there is no contact between them.

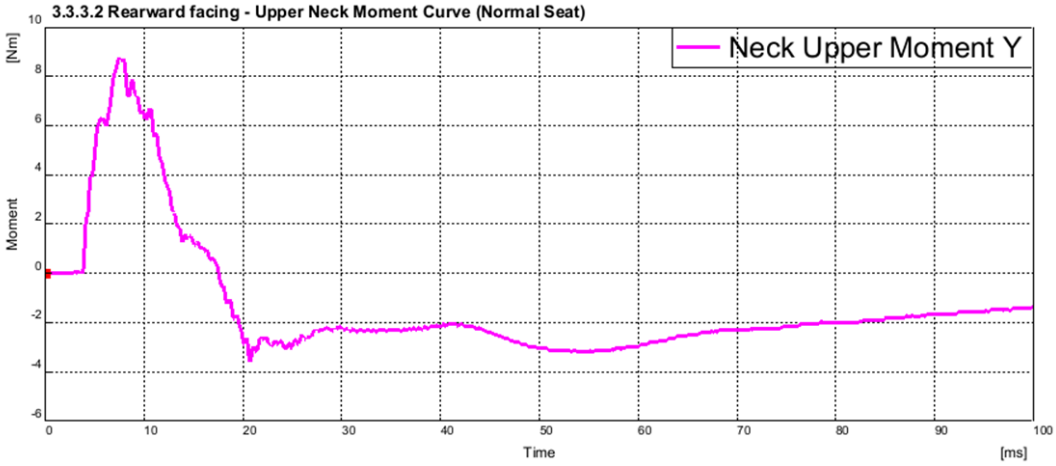


Figure 4.9: Upper Neck Moment Curve

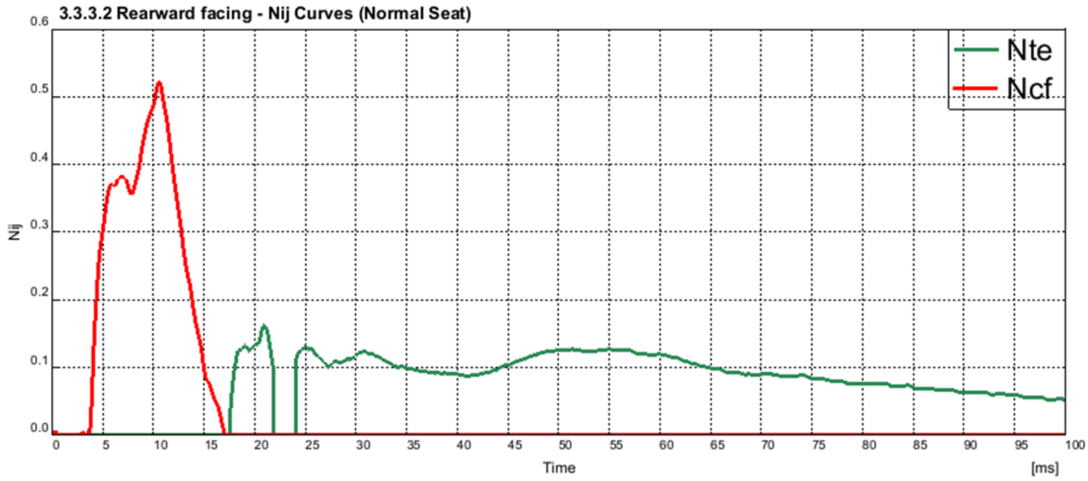


Figure 4.10: N_{ij} Curve

From the preliminary analysis of the injury criteria: regarding the head, it can be determined that it is subject to compression in the first 20 ms due to the initial deployment of the airbag and then to tension when the lower section, in contact with the dummy, inflates more as it gives to dummy head a negative rotation. For the same reason, the movements related to the neck are the same and it is possible to see in the figure 4.8 and 4.9. The percentage value of force to which the upper neck is subject is 65.5%. The rotation translates, initially, into flexion and, after maximum deployment, into extension. The first contact is at approximaly 4 ms and the maximum unfolding around 12 ms. From the curve analysis, precisely N_{ij} curves, it is possible to verify that the preliminary analysis is correct: the dummy is subject to a compression-flexion along Y and its maximum is around 11 ms. The obtained maximum value from N_{CF} is 52.3%. The red curve, in the figure 4.10, represents this behavior. At around 18 ms, as previously indicated after maximum deployment, the neck undergoes a tension-extension described by the green line.

Sport seat

As in the case of the normal seat, the test objective involves the head interaction with CeAB. From the first video analysis, it is possible to declare that the aim has been met and the position can be considered accurate.

Unlike the previous case, despite the different seat design, the time of first contact is the same. From the preliminary analysis of the injury criteria, it is possible to state that, due to the deployment of the airbag, the head and consequently the neck are initially compressed and after about 20 ms undergo a tension with negative rotation. The thorax curve should be flat because of no interaction. The first contact is at approximaly 4 ms and the maximum unfolding around 10 ms.

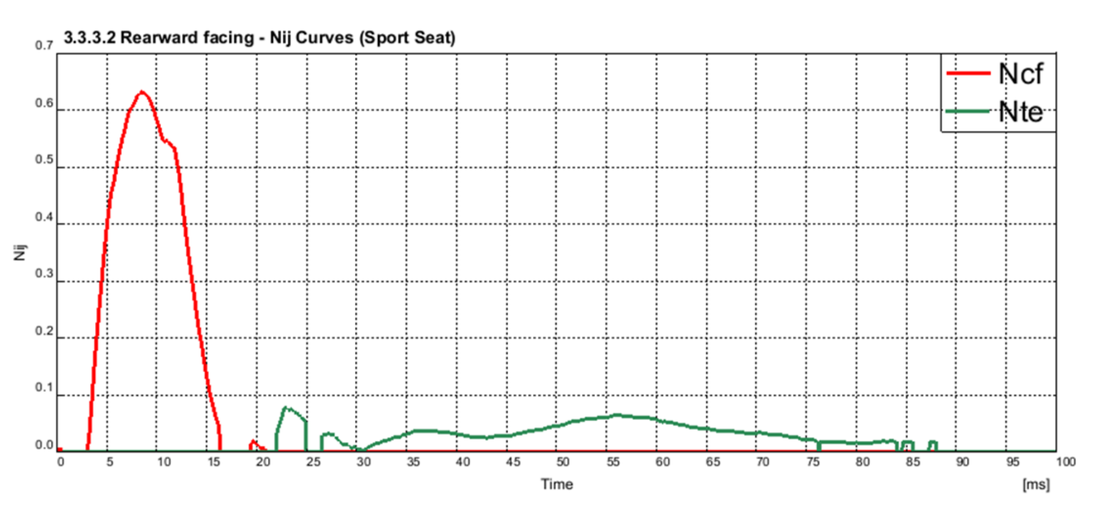
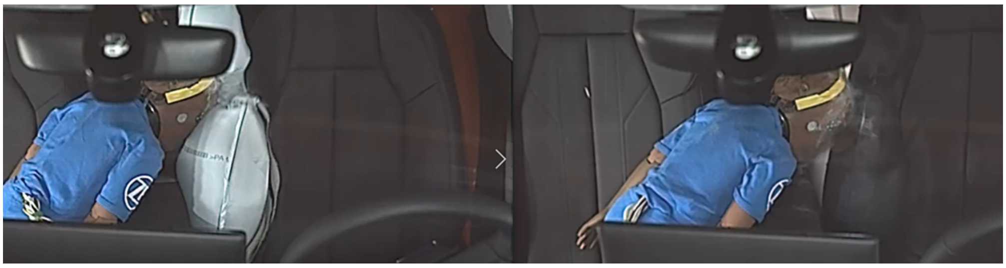


Figure 4.11: N_{ij} Curve

From the curve analysis, precisely N_{ij} curves, it is possible to verify that the preliminary analysis is correct: the dummy is subject to a compression-flexion along Y and its maximum is around 9 ms. The obtained maximum value from N_{CF} is 63.1%. After this initial section, it is possible to observe part of the curve in green representing the tension-extension governed mainly by the moment.

COMPARISON: Normal Seat vs Sport Seat



In this case, as in the previous one, the morphology of the N_{ij} curve, for both types of seats, is similar with some minor differences in the first ms due to the deployment of the airbag, in turn due to the design of the seat:

- in the case of the normal seat, there is a faster deployment and the maximum value is near the maximum airbag unfolding;
- in the case of the sport seat, there is a slightly slower deployment time and the maximum value occurs close to the first contact with the dummy.

This difference can be seen in the graph below:

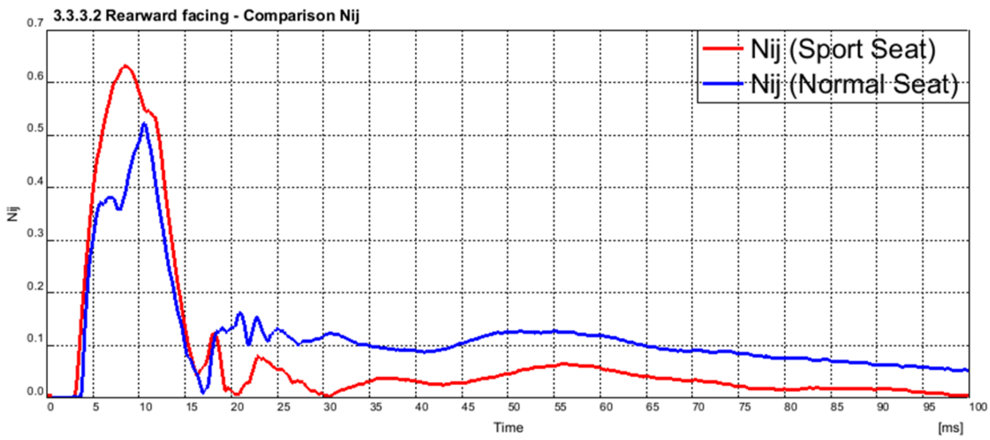
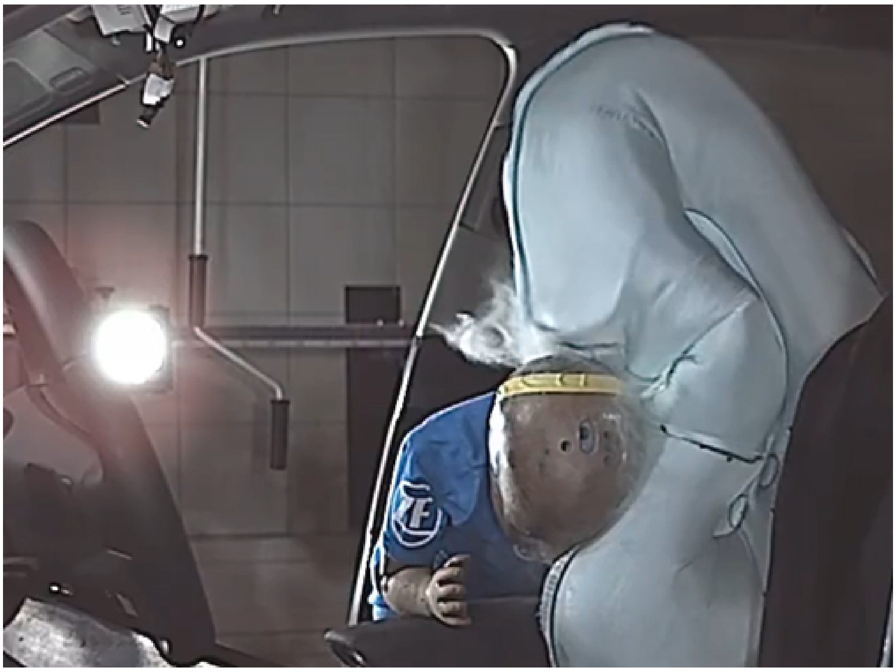


Figure 4.12: Comparison N_{ij} Curve

3.3.4.1 Inboard facing with arm on Center Console

Normal seat

The test objective involves the head/neck interaction with CeAB by aligning the head center of gravity with the deployment path of the airbag.



From the first video analysis, it is evident that the body regions stressed by the airbag deployment are the head and the neck as stated in the objective while the chest is not involved. In fact, due to its design, the airbag completely envelops the upper part of the dummy during its deployment. The lower section, initially, gives the neck compression and therefore a negative force, but once more inflated the section of the airbag to the left of the head COG causes tension-extension with backward movement of the head. The first contact is at approximaly 9 ms and the maximum unfolding around 22 ms. All this can be seen from the curves which underline the correctness of the preliminary analysis.

In this case the index N_{ij} curve is analyzed: the dummy is mainly subject to a tension-extension along Y, after maximum airbag deployment, and its maximum is around 25 ms. The obtained maximum value from N_{TE} is 33%: this is due to a positive force and a negative momentum.

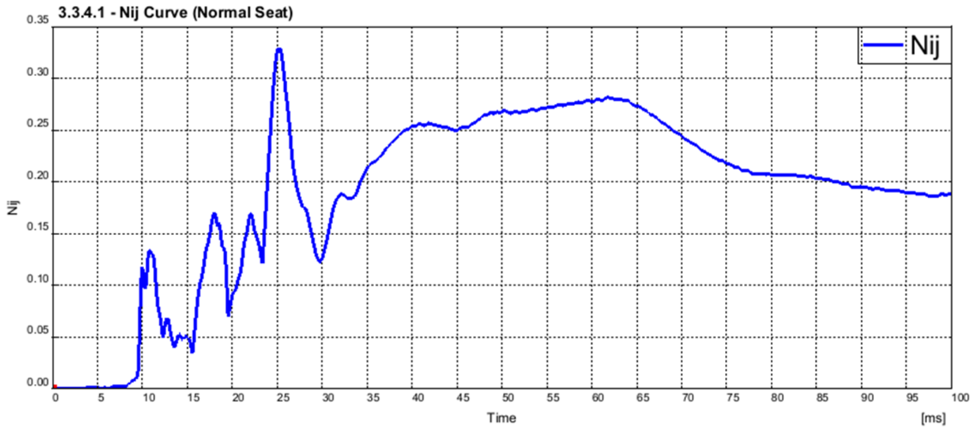


Figure 4.13: N_{ij} Curve

As previously mentioned, the force acting on the upper neck is mainly positive but, as shown in the graph below, around 268 ms there is contact of the dummy with the vehicle dashboard which causes compression on the neck.

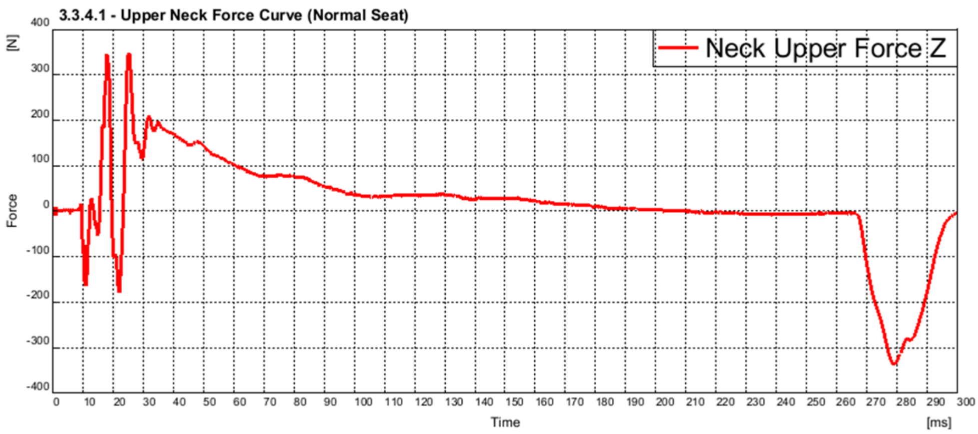


Figure 4.14: Upper Neck Force Curve

Regarding the upper neck, it is mainly dominated by negative values characterizing the extension but in the first 10 ms there is a slight flexion due to the very first contact with the bag before it envelops the whole neck of the dummy.

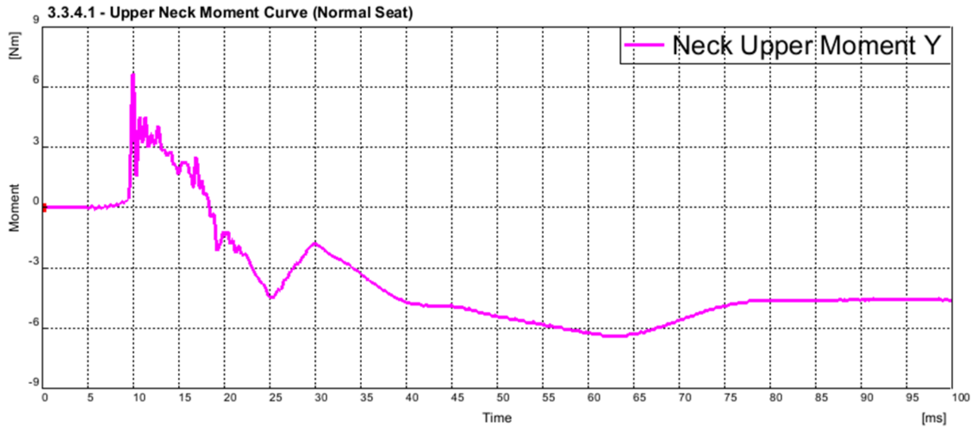


Figure 4.15: Upper Neck Moment Curve

Sport seat

As in the case of the normal seat, the test objective involves the head/neck interaction with CeAB and from the first video analysis, the stressed body regions are respected. So, the test objective has been met.

From the preliminary analysis, the upper neck is subject to the same forces and moments as in the case of the normal seat with a dominance of the tension-extension movement. Also in this case, despite the difference in design, the timing is the same: the first contact is at approximaly 9 ms and the maximum unfolding around 22 ms.

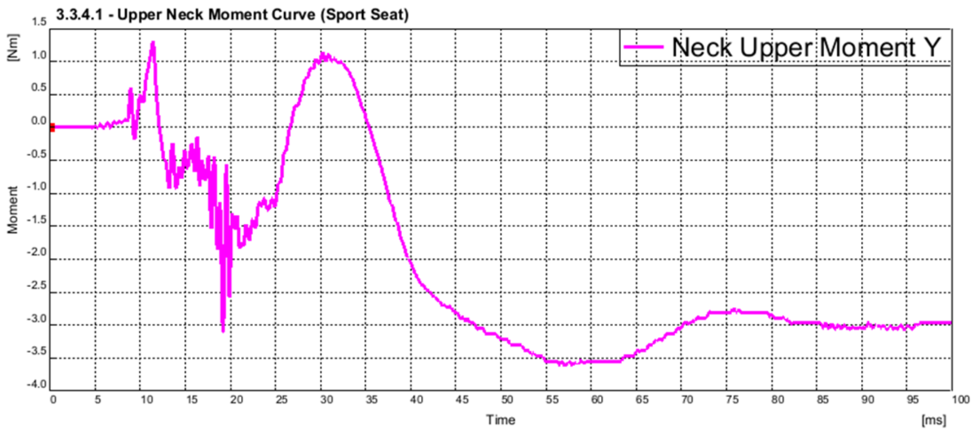


Figure 4.16: Upper Neck Moment Curve

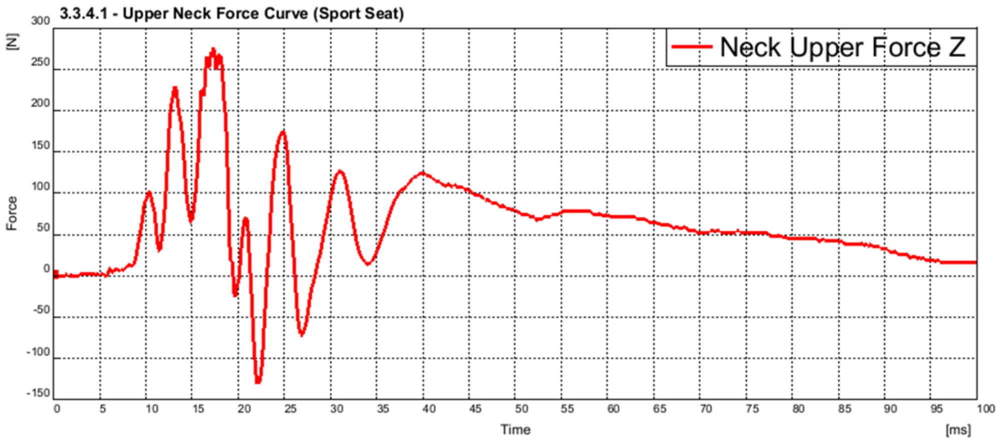


Figure 4.17: Upper Neck Force Curve

In this case, looking at the injury curves, precisely N_{ij} curves, it is possible to verify that the preliminary analysis is correct: the dummy is subject to a tension-extension along Y and its maximum is around 57.10 ms. The obtained maximum value from N_{TE} is 17%.
Due to the airbag and seat design, the force exerted on the upper neck of the dummy is purely tension. Even at the beginning, as is clearly visible in the video, the lower section of the bag does not act from below by compressing but laterally by causing a positive force.

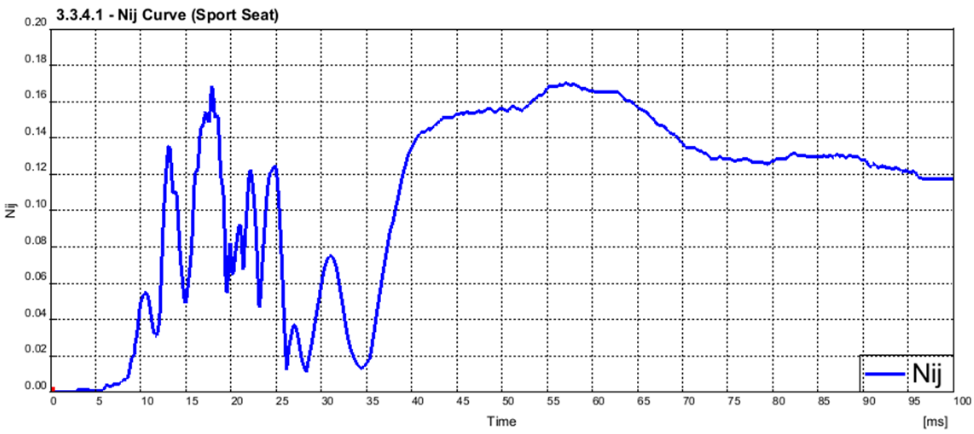
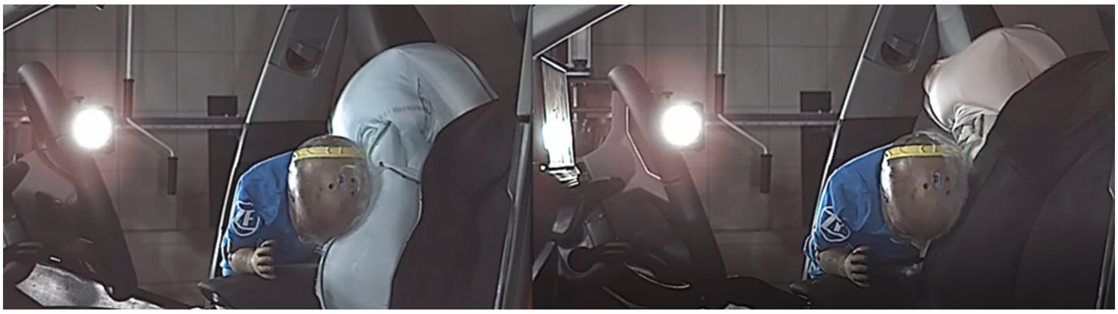


Figure 4.18: N_{ij} Curve

COMPARISON: Normal Seat vs Sport Seat



Apart from the different deployment times, the main difference between normal and sport seat is related to the deployment way and it occurs in the first milliseconds. This has a great impact on the force along Z relative to the upper neck.
It is possible to observe that, due to the design of the seat, the deployment of the airbag is different:

- in the normal seat, the first contact with dummy occurs with the lower section of the airbag and this is the cause of the compression (negative) force in the first 15 ms;
- in the sport seat, this contact occurs later and the first is with the upper section of the bag. For this reason, the contact exerts on dummy neck a tension (positive) force.

This difference can be seen in the graph below:

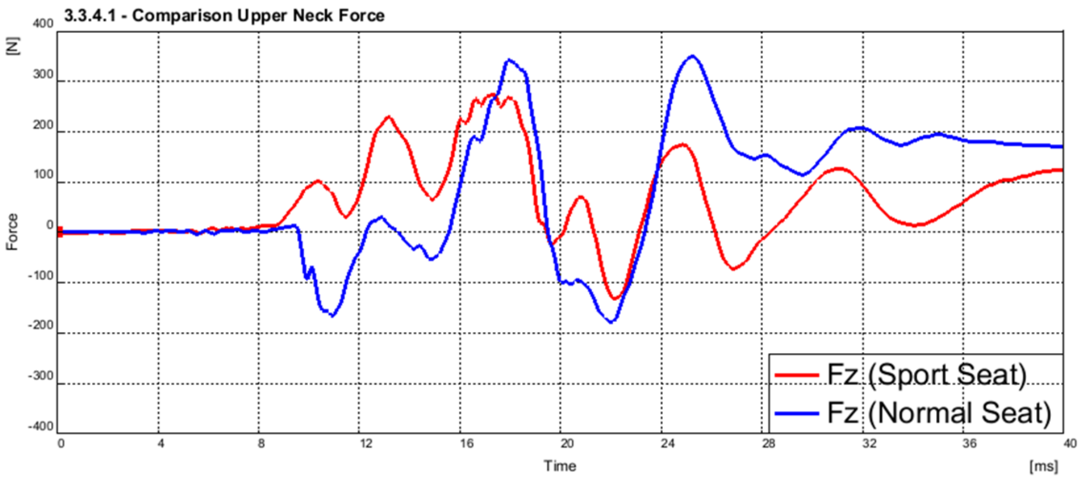


Figure 4.19: Comparison Upper Neck Force Curve

Since the force along the Z-axis has this trend, it is easy to understand the behavior of the N_{ij} index being influenced by it. The most evident difference falls in the first ms while in the remaining ms only the amplitude changes, greater for the normal seat.

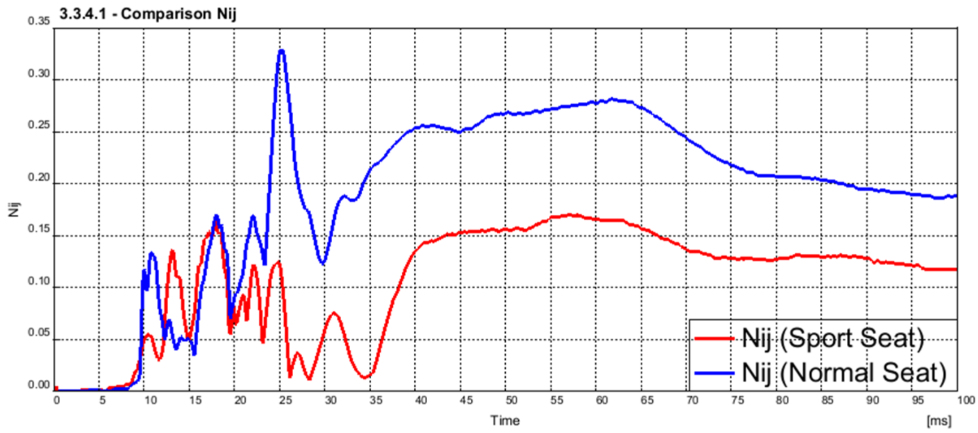


Figure 4.20: Comparison N_{ij} Curve

Increasing the observation time of the index N_{ij} to 300 ms, it is possible to observe:

- in the case of the normal seat the final contact of the dummy with the dashboard causing a compression-flexion;
- in the case of the sports seat as the deployment time of the airbag is slower, it is not well visible, but the contact takes place and can be underlined by the increasing red curve at 292 ms.

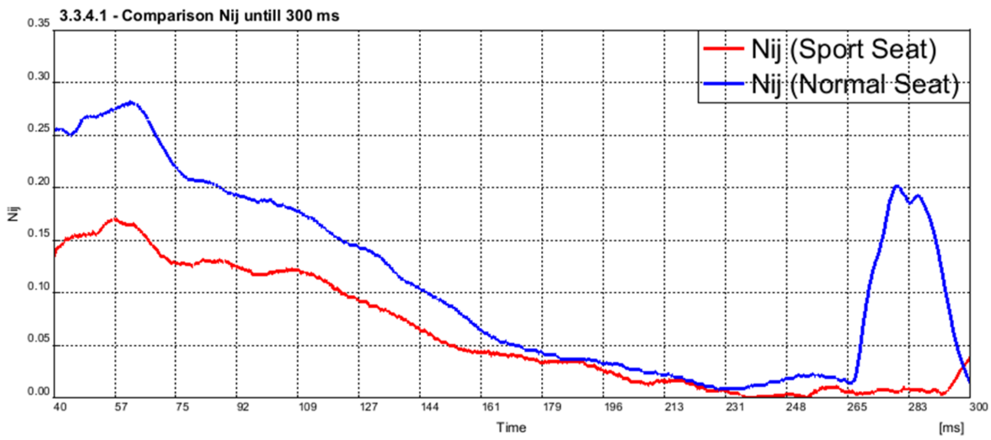


Figure 4.21: Comparison N_{ij} Curve until 300 ms

3.3.3.3 Lying on seat with head on Center Console

The test objective involves head/neck interaction with CeAB by aligning the head center of gravity with the deployment path of the airbag. From the first video analysis, it is possible to state that the body regions involved in the deployment are the head and neck. So, the aim is achieved and the positioning is correct.

From the preliminary analysis of the injury criteria, it is clear that the lower section of airbag, due to its inflation, gives the head and consequently the neck a negative force and a positive moment. Regarding the thorax, there is not interaction and so, the curve was not recorded. The first contact is at approximaly 9 ms and the maximum unfolding around 20 ms.

From the curve analysis, precisely N_{ij} curve, it is possible to observe that the preliminary analysis is correct: the neck in subject to a compression-flexion along Y and its maximum is around 22 ms. The obtained maximum value from N_{CF} is 18.9%. This graphic is an alternation of compression-flexion and of tension-flexion. This can be seen in the graph below showing the behavior of force and moment in the first 70 ms.

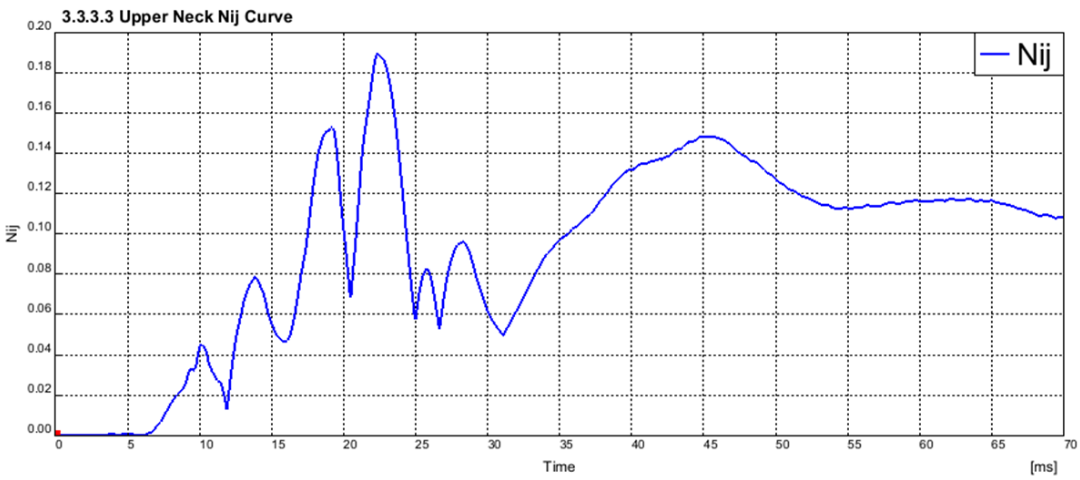


Figure 4.22: N_{ij} Curve

There is always a positive moment that results in flexion, but the force alternates between positive and negative values due to the interaction head-airbag. Positive force results in tension and negative force in compression. For this reason, as the index is a combination of force and moments, at negative force we will have the contribution of N_{CF} and at positive force the contribution of N_{TF}. The presence of the moment allows to amplify the amplitude of N_{ij}.

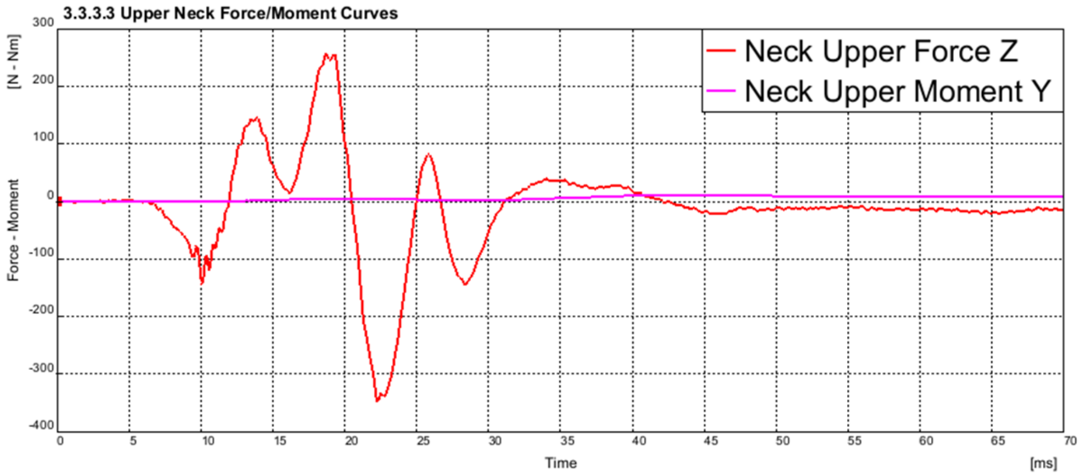
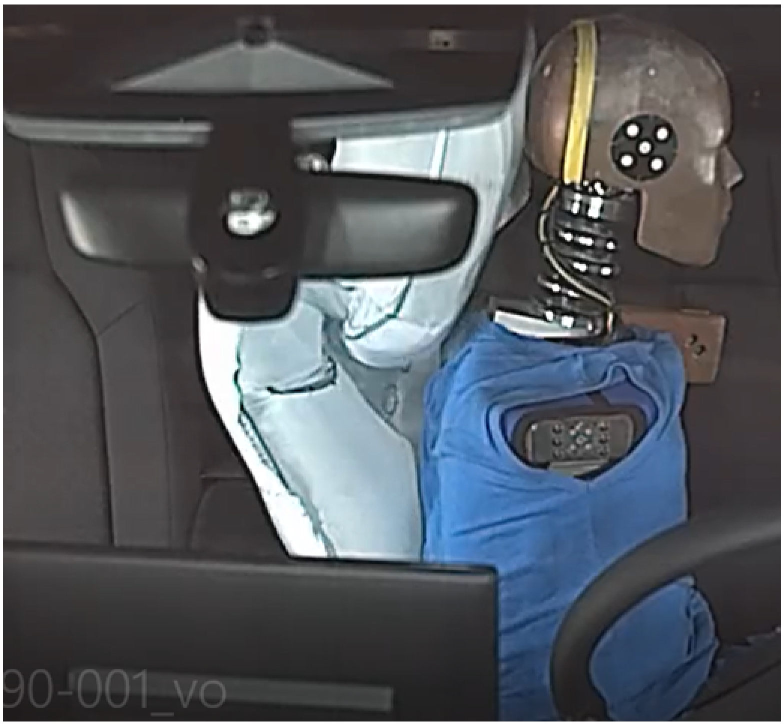


Figure 4.23: Upper Neck Force and Moment Curves

3.3.3.6 Outboard facing

This test can be conducted for driver and passenger positions, but to analyze curves to assess the OoP injury risk the driver position was chosen.



The test objective involves chest interaction by aligning the center of the upper thoracic rib with the top edge of the airbag module. From the first video analysis, it emerges that the body region that interacts most with the bag is the chest and so, the reliability of position is optimal, and the test objective has been achieved.

From the preliminary analysis of the injury criteria, it is clear that the interaction with the head is zero and so, the curve will be almost flat. The force will be exerted exclusively on the thorax and subsequently, it produces a tension-extension movement at the neck. The first contact is at approximaly 8 ms and the maximum unfolding around 20 ms.

From the curve analysis, precisely thorax curves, it is possible to verify that the preliminary analysis is correct. Acceleration, deflection and deflection rate are analyzed resulting in sinusoids which are then filtered. It is possible to obtain these curves through two methods: integration and derivation. For the first case, starting from acceleration, the first integration gives the velocity and the second the deflection. For the second one, starting from deflection, the first derivation gives the velocity and the second the acceleration. In fact, where the acceleration and the deflection are maximum, the velocity is around zero while where the acceleration and the deflection are zero, the velocity has its maximum. Between two methods, the integration is preferred because keeps the signal intact and removes more noise components.
As previously indicated, depending on the airbag deployment, it is mainly the lower section that acts on the thorax and this can be observed in the following graphic of deflection.

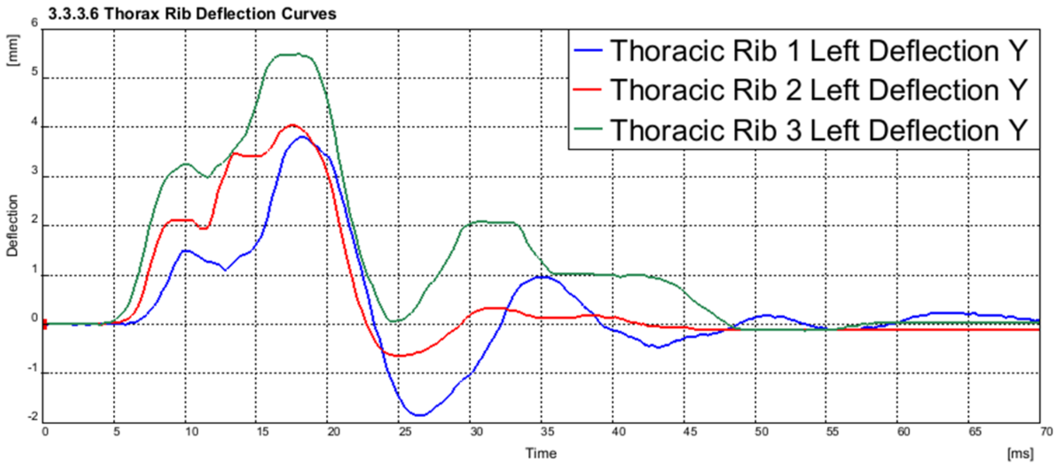


Figure 4.24: Thoracic Rib Deflection Curves

In fact, there is a greater deflection along Y from bottom to top: initially a maximum for thoracic rib 3, which is the lowest, at 17.95 ms and subsequently, a maximum for thoracic rib 2 at 17.55 ms and a maximum for thoracic rib 1 at 18.25 ms.

Regarding deflection rate curves, the observation interval is up to 18 ms as there is the maximum thorax-bag interaction and then, due to the impact, the dummy moving towards the door does not interact with CeAB anymore.

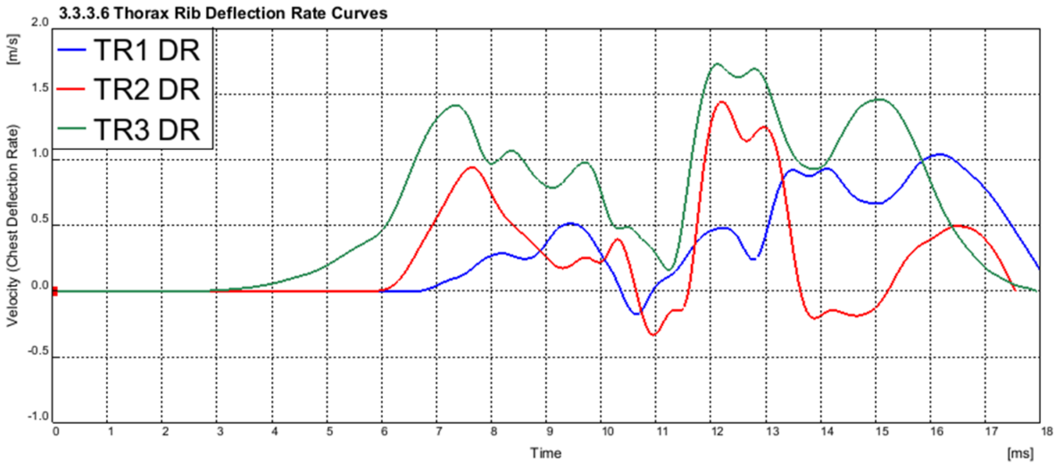


Figure 4.25: Thoracic Rib Deflection Rate Curves

It is possible to see that the morphology of the curves is rather similar because the thoracic ribs from bottom to top, are subject to the same acceleration from which the deflection rate is derived. More precisely, it is observed that thoracic ribs 2 and 3 present the same morphology due to the design of the lower section of the CeAB, which involves slightly less rib 1 and is demonstrated by the lower acceleration to which it is subjected. The morphology is time-shifted as the deployment of the airbag first involves rib 3 and after a few ms rib 2.

Conclusion

As previously announced in Chapter 4, in order to verify the adequacy of the airbag involved in the out-of-position tests, it is necessary to carry out the biomechanical analysis of the curves. Using the limit values imposed by the NHTSA and the values obtained from the instrumentation present on the dummy during the static OoP tests carried out in the laboratory at the ZF headquarters, it is possible to determine the percentages for each criterion of injury characterising each position. The percentages make it possible to classify the severity of the injury, since a high result implies that the value obtained by the instrumentation is close to the limit value.
For the eight positions tested, 5 for the normal seat and 3 for the sport seat, only values lower than the limit values are obtained, implying percentage values between 0 and 80%: actually, the highest value obtained is that of the N_{CF} index, for the '3.3.3.2 Rearward facing' position, equal to 63.1%. In this way, the association of the green colour is obtained, which represents the acceptability and correctness of the restraint system in question. Moreover, it is fundamental to state that each position respects the test objective - first step of the biomechanical analysis - and in the case under examination, this is verified except for the position '3.3.3.2 Rearward facing with knees on Center Console' as to have the maximum interaction with the chest one should turn the dummy towards the module losing the alignment of the head with the origin of the local reference system of the vehicle and making the repositioning more difficult.

On the basis of the data obtained and analysed, it is possible to conclude that the Center Airbag under examination with this specific shape and characterised by a given deployment mode is optimal and can continue with the performance of further tests for approval.
The main objective of this study was the drafting of the ISO Manual OoP Testing Protocol for the new airbag, since the NHTSA has made static testing a mandatory requirement of the new regulation, implemented since 2003. It can be said that this goal has been successfully achieved as the document has already been published as a first revision. It is called first revision because the protocol can be updated in the

years to come. This is because by involving several manufacturers, they can modify it according to their needs and vehicle design. An example of the different vehicle design is the case of position '3.3.3.6 Outboard facing' which sees the presence of two cases due to the presence or absence of the centre console. Obviously, the changes made by the manufacturer with respect to the recommended positions must still be within a reasonable range representing the typical 'worst case' conditions.

In light of the tests conducted in the laboratory at ZF in Alfdorf, I had the opportunity to personally observe how the dummies were positioned and the static OoP tests executed. For this reason, in this paper I could combine theory with practical experience which allowed me to have a better understanding of the subject in question and, consequently, being better equipped to write the protocol, especially the details of the numerous procedures. Being able to witness the tests first-hand, take part in the discussions during the experiments and observe all the work behind each one made me more committed to pursue a career as biomedical engineer in the automotive world.
Being able to write my thesis at Italdesign, a leading global provider of development services in the automotive industry, gave me the opportunity to observe the job market and give my contribution to it. It was a very formative experience in a stimulating and dynamic environment. It allowed me to learn by putting my knowledge and theoretical skills to test and to be tutored by experts, to enrich and form myself while growing professionally. Right from the beginning, I noticed and appreciated the inclusive working atmosphere within the company, and this made it possible for me to take part in several meetings on the subject of my work.
In conclusion, this job has given me the opportunity to learn and fully address issues related to biomechanical engineering that are not covered in depth in the academic track I choose but are, a nevertheless, fundamental to the career path I would like to take.

Bibliography

[1] D'Ambrosi D. *Incidenti stradali USA: con gli spostamenti +18% di vittime nel 2021.* Sicuroauto.it

[2] ACI: Automobile Club D'Italia - Cosa si intende per sicurezza attiva

[3] ACI: Automobile Club D'Italia - Cosa si intende per sicurezza passiva

[4] Wallis, L.A. and Greaves, I. 2002. Injuries associated with airbag deployment. Emerg Med J; 19:490–493.

[5] Lifehackk - *Storia e invenzione di Airbag per il settore automobilistico*

[6] HELLA TECH WORLD - Componenti elettrici ed elettronica automotive. *SISTEMA AIRBAG SRS*

[7] Wikipedia - *Airbag: I Componenti*

[8] Corazza M. and Trincone S. and Virgili A. 2004. *Effects of Airbag Deployment: Lesions, Epidemiology, and Management.* Am J Clin Dermatol 5 (5), 295-300.

[9] Come funziona l'airbag un valido ed efficace dispositivo di sicurezza. MotoAutoGratis.it

[10] Thomas, S.and Wiik, R. and Brown, J. 2013. *The Front Center Airbag.* SAE Int. J. Trans. Safety 1(2), 352-363.

[11] Spiess O. and Marotzke T. and Zahn M. 1997. *Development Methodology of an Airbag Integrated Steering Wheel in Order to Optimize Occupant Protection Balanced Against Out-of-Position Risks.* SAE Transactions, Vol. 106, Section 6: JOURNAL OF PASSENGER CARS: Part 1, 1314-1326.

[12] Horsch, J. et al; *Assessment of Airbag Deployment Loads*, SAE Paper 902324, Society of Automotive Engineers, Warrendale, PA

[13] Mertz H.2002. *Anthropomorphic Test Devices.* In: Nahum A.M., Melvin J.W. (eds) Accidental Injury. Springer, New York, NY.

[14] Kallieris D. and Rizzetti A. and Mattern R.1995.*The biofidelity of Hybrid III dummies* Institute of Legal Medicine, University of Heidelberg.

[15] Humanetics, Humanetics Group, Anthropomorphic test devices,Frontal Impact ATDs

[16] Roger P. D. and Irwin A. and Athey J. and Balser J. and Eichbrecht P. and Hultman W.R. and Kirkish S. and Kneisly A. and Mertz H. and Nusholtz G. and Rouhana S. and Scherer R.and Salloum M. and Smrcka J. 1995. *Technical*

Specifications of the SID-IIs Dummy SAE Transactions Vol. 104, SECTION 6: JOURNAL OF PASSENGER CARS, Part 2, pp. 3212-3241

[17] Xu T. and Sheng X. and Zhang T. and Liu H. and Liang X. and Ding A.2018. *Development and Validation of Dummies and Human Models Used in Crash Test* Hindawi, Applied Bionics and Biomechanics.

[18] Humanetics, Humanetics Group, Anthropomorphic test devices, Rear impact ATDs

[19] Humanetics, Humanetics Group, Anthropomorphic test devices, Child ATDs, CRABI Series

[20] Foster J.K. and Kortge J.O. and Wolanin M.J.2018. *Hybrid III — A Biomechanically-based Crash Test Dummy* Safety Research and Development Lab, Environmental Activities Staff, General Motors Corp.

[21] Humanetics, Humanetics Group, Anthropomorphic test devices, Child, Hybrid III children series, 3-years-old.

[22] Humanetics, Humanetics Group, Anthropomorphic test devices, Child, Hybrid III children series, 6-years-old.

[23] Humanetics, Humanetics Group, Anthropomorphic test devices, Frontal Impact ATDs, Hybrid III 5th Female

[24] Humanetics, Humanetics Group, Anthropomorphic test devices, Frontal Impact ATDs, Hybrid III 50th Male

[25] Humanetics, Humanetics Group, Anthropomorphic test devices, Frontal Impact ATDs, Hybrid III 95th Male

[26] Humanetics, Humanetics Group, Anthropomorphic test devices, Side impacts, SID-IIs.

[27] Humanetics, Humanetics Group, Anthropomorphic test devices, Frontal Impact ATDs,THOR-5F

[28] Melvin J.M. and Mertz H.J. 2002. *Airbag Inflation-Induced Injury Biomechanics* In: Nahum A.M., Melvin J.W. (eds) Accidental Injury. Springer, New York, NY.

[29] Eppinger R. and Sun E. and Bandak F. and Haffner M. and Khaewpong N. and Maltese M. and Kuppa S. and Nguyen T. and Takhounts E. and Tannous R. and Zhang A. and Saul R. 1999. *Development of Improved Injury Criteria for the Assessment of Advanced Automotive Restraint Systems - II.*

[30] Simms C. and Wood D. *Injury Mechanisms and Injury Criteria.* In: Pedestrian and Cyclist Impact. Solid Mechanics and Its Applications, vol 166. Springer, Dordrecht.

[31] Lund, A. K. 2003. *Recommended procedures for evaluating occupant injury risk from deploying side airbags.* The Side Airbag Out-of-Position Injury Technical Working Group.

[32] Irwin A. and Mertz H.J.1997. *Biomechanical Basis for the CRABI and Hybrid III Child Dummies.* General Motors Corp.