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EUROPEAN ENHANCED VEHICLE-SAFETY COMMITTEE

The use of thoracic deflection criteria balanced with abdomen pressure criteria for the Q-Series in frontal impacts

Working Group 12 Report,
Document No. D661,
April 7, 2016





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EUROPEAN ENHANCED VEHICLE-SAFETY COMMITTEE

The use of thoracic deflection criteria balanced with abdomen pressure criteria for the Q-Series in frontal impacts

EXECUTIVE SUMMARY

The Q dummies were developed to replace the P-dummies in the ECE Regulation 44 and in consumer testing. EEVC Working Groups 12 (Biomechanics) and 18 (Child Safety) collaborated on joint research to evaluate the Q-Series and recommended that they replace the P-Series in UN Regulation 44 (Wismans et al., 2008, [1]). This 2008 report also proposed injury criteria and injury assessment reference values for use with the dummies in the Regulations. In 2008 the Q-Series comprised Q0, Q1, Q1.5, Q3 and Q6 dummies only. The Q10 was more recently added to represent the older children and a review of its performance in frontal impact was performed within EEVC Working Group 12 (Wismans et al., 2015 [2]).

The aim of the current report is to complement the previous efforts by looking at thoracic and abdominal injury criteria and dummy kinematics. As shown in Chapter 2 abdominal and thoracic injuries are relevant priorities for prevention for children of various ages in frontal impact events. Estimating the loads transferred by the restraint system to the thorax and abdomen is important therefore for children restrained in integral as well as non-integral Child Restraint Systems (CRS).

Multiple measurements principles have been considered in the past to attempt to quantify directly abdominal loads, including the Abdominal Pressure Twin sensors (APTS) for the Q dummies developed by Ifsttar. The development and evaluation with different versions of the APTS design are summarized in Chapter 3. The production version of the APTS design and the corresponding abdomen section of the dummy are now close to finalization. With the potential to generate abdominal loading in sled tests and the ability of the APTS to detect that loading, the Q-Series dummies are capable to assess abdominal injury risk in non-integral as well as in integral child restraint systems.

Chapter 3 also contains abdominal tolerance risk curves, which were obtained for Q3 and Q6 based on accident reconstruction in belted configuration. Pressure levels slightly over 1 bar seem to correspond to a limit between behaviours which are visually acceptable or

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unacceptable. Based on the risk estimate, such a limit is likely to relate to a risk of AIS 3+ injury of between 20 and 50%. For the Q10, the same separation may occur at slightly higher pressure levels. For the Q1.5, there is currently no such equivalence as the dummy has not been tested in three-point belted configurations.

The section 4 discusses Thoracic Injury Criteria, and thoracic deflection in particular and their scaling for the Q dummies of different sizes. Reliable thoracic injury risk functions however are still lacking. A brief review of candidate risk curves is proposed and discussed.

The section 5 provides background about the interaction issues between the Q dummies and the three point belt, and discuss possible countermeasures. This concerns in case of non-integral CRS movement of the diagonal part of the seat belt towards the neck and penetration of the lap part of the seat belt into the gap between the legs and the pelvis.

Chapter 6 concludes this reports with a Discussion and Conclusion section including a number of recommendations concerning, among others, the implementation of the APTS in the Q dummies for regulatory testing, both for integral and non-integral CRS, the use of abdominal pressure and chest deflection as injury criteria, the use of hip inserts to describe submarining behaviour in a more realistic way, the future research concerning shoulder belt thorax interaction and the seating procedures for non-integral CRS.

1 INTRODUCTION

1.1 Background

UN Regulation 44 specifies performance requirements that are assessed during front and rear impact tests with the P-Series “family” of child dummies. These dummies were developed during the 1970s and were the most sophisticated child dummies in Europe when the Regulation was introduced in 1981. The P-Series have been instrumental in improving the quality of child restraints and have proven to be extremely durable for regulatory testing. Nevertheless, the dummies are often referred to as simple loading devices with minimal instrumentation. The anatomy and behaviour of the internal structures of the body are not represented, which is one of the fundamental shortcomings of these dummies. In addition, the method used to detect abdomen loading (a clay insert between the lumbar spine and abdomen), is somewhat subjective and does not allow for a complete assessment of injury risk.

The Q-Series were developed as possible successors to the P-Series in regulatory (and consumer) testing. Work on the dummies was initiated in 1993 by the International Child Dummy Working Group and continued during the European Framework Programme projects CREST (Child REstraint System sTandard, 1996-2000), CHILD (CHild Injury Led Design, 2002-2006), CASPER (Child Advanced Safety Project for European Roads, 2009-2012) and EPOCh (Enabling Protection of Older Children, also 2009-2012). EEVC Working Groups 12 (Biomechanics) and 18 (Child Safety) collaborated on joint research to evaluate the Q-Series and recommended that they replace the P-Series in UN Regulation 44 (Wismans et al., 2008, [1]). They also proposed injury criteria and injury assessment reference values for use with the dummies in the Regulation.

In 2007, the UN Informal Group on Child Restraint Systems was established by the Working Party on Passive Safety (GRSP) to develop a new UN Regulation on the approval of “Enhanced Child Restraint Systems”. The new Regulation No. 129 introduces the “i-Size” concept (a stature-based system of classification for child restraints), the new family of child dummies (the Q-Series) and a side impact test procedure. The first phase of the new Regulation was completed in 2011 and sets out performance requirements and test methods for integral ISOFIX child restraints. It was adopted by the World Forum for Harmonisation of Vehicle Regulations (WP.29) during its 158th Session held in Geneva in November 2012. The second phase is underway and incorporates non-integral child restraints.

The UN Informal Group has based many of its decisions regarding the use of the Q-Series in the new UN Regulation on the work carried out by EEVC Working Groups 12 and 18. A production version of a new Q-Series dummy, the Q10, could provide an “upper limit” dummy for the new Regulation. Its possible use in frontal impact was the focus of a recent EEVC report (Doc 642, Wismans et al., 2015 [2]). In parallel, a broader advice was also asked on the interaction between the Q-Series dummies and the three-point seat belt (in non-integral child restraints) and the use of performance thresholds for chest compression and abdomen pressure. This will be the focus of the current report.

1.2 Motivation

A pragmatic limit for chest acceleration of 55g is specified in the new UN Regulation, based on comparisons between the P-Series and Q Series undertaken by EEVC WGs 12 and 18 (Wismans et al., 2008 [1]). Applying a chest deflection performance limit would also ensure that child restraint systems do not transfer compressive loading to the chest to minimise other parameters that are regulated. However, the CASPER project was unable to derive an injury risk function for the chest (deflection or acceleration) for the Q-Series using accident reconstructions. Also, whilst EEVC WGs 12 and 18 proposed scaled deflection limits for the Q Series, they have not been adopted in the new UN Regulation or by the wider child safety community.

At present, the Q-Series has no means of detecting abdomen loading. It can't be fitted with a clay insert and requires another solution. Prototype abdomen pressure sensors (fitted within a modified abdomen block) were developed and evaluated in the CASPER project and appear to display consistent results (Beillas et al., 2012, [3]); however, these sensors are not currently part of the original dummy equipment. The CASPER project proposed abdomen injury risk curves and performance thresholds (based on these sensors) for the Q3 and Q6 dummies. The work on these sensors continued after the CASPER project and new versions have been developed that make them usable with Q1.5 dummies, which could be relevant for the evaluation of integral CRS. Various tests have also been conducted with the Q10 dummy (as reported in Wismans et al., 2015, [2]). The use of chest compression and abdomen pressure measurement and performance thresholds could offer benefits for the assessment of child restraints in the new Regulation (particularly in front impact) assuming that the kinematic behaviour is adequate. However issues have been raised in the past about belt interactions with the dummies.

1.3 Objectives and format of the report

The objective of this report is to review the status of the Q dummies concerning the capabilities for thorax and abdominal injury detection and develop recommendations for performance thresholds for thorax and abdominal injuries. The work described in this report provides information on the use of chest compression and abdomen pressure measurements with the Q-Series. Consideration is also given to whether the instrumentation is sensitive to different loading conditions and if is capable of providing a reliable measurement in a regulatory test. With regards to the abdomen, the assessment is based initially on the prototype sensors developed during the CASPER project, but covers also all subsequent developments, including their integration in modified dummy abdomens.

The report starts with a brief overview of child injury causation for children with a focus on chest and abdomen (Section 2).

Section 3 focuses on the abdomen pressure measurement instrumentation for the Q series, including density and mass definition, biofidelity, sensitivity of response, repeatability, durability, performance in tests and injury assessment potential.

Section 4 then provides an overview of chest deflection measurement with the Q series, as well as the information about the chest deflection and abdomen pressure sensors. The report documents evaluations of the interaction between the Q-Series and three-point seat belts. It

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considers whether the interaction phenomena described above are realistic and the biofidelity of the dummy at the whole-body level (rather than just at the component level).

Section 5 provides background on the belt interactions with the Q dummies, the possible issues associated with these interactions and discusses possible solutions. Some comments have been made regarding the kinematics of the dummy including unloading and behaviour in the rebound phase, however these issues are beyond the scope of the current report.

Chapter 0 concludes this report with a discussion and conclusion section including a number of recommendations. The work has focussed on the Q-Series dummies, but to an extent also provides consideration of potential changes to front impact test procedures, which might in the short term, improve the assessment of non-integral child restraint systems. Ultimately the recommendations provide an opinion on whether longer term actions are needed, such as: improvements in shoulder and thorax biofidelity to improve the interaction with the diagonal part of the seat belt.

Background and detailed information is provided in Annexes.

2 ACCIDENT DATA, ABDOMINAL AND THORACIC INJURY CAUSATION

2.1 Introduction

It is essential that the measurement capabilities of a dummy reflect the injuries that are observed in the field among occupants of the same size. This enables the dummy to be used reliably in test procedures that target specific priorities for injury reduction. The current report is focused specifically on the abdominal and thoracic injuries, which are believed to be especially significant for children in certain age ranges.

As part of the EEVC report “Advanced child dummies and injury criteria for frontal impact”, EEVC WG18 reviewed the European accident statistics with respect to child car occupants. A broad range of sources were examined, including: CREST (Child Restraint STandard, a European collaborative research project), CCIS (the Cooperative Crash Injury Study in the UK), GIDAS (German In-Depth Accident Study), GDV (German Insurance), IRTAD (International Road Traffic Accident Database) and LAB (Laboratory of Accidentology and Biomechanics, France). Although these sources used very different definitions and data collection methods, which made it difficult to merge data for analysis, sufficient information was available to classify injury causation according to the different groups of child restraint systems that were used. The EEVC findings have subsequently been updated and refined with more recent sources, such as CASPER (Child Advanced Safety Project for European Roads, a European collaborative research project) (see Kirk, 2012, [4]). Grouping the findings by child restraint type was necessary to highlight where performance improvements were needed. However, in many of these sources (including the most recent, from CASPER), restraint types that are now intended for older children only (i.e. booster systems), were also used by younger children in significant numbers (due to differences in the legislation from the time of the collisions). It is difficult, therefore, to separate the experiences of children of a specific age restrained by a specific restraint system.

However, a new, comprehensive analysis of the European accident data was not within the scope (and resources) of EEVC WG12 in preparing this report. This section summarises some findings from the literature whilst also taking into account the original work of EEVC WG18, the work of CASPER and research carried out by TRL for the European Commission (Visvikis at al., 2014, [5]) for older children. Data from the TRL study and some complimentary data from the Rhône Registry Injury Database in France are provided in ANNEX A.

2.2 Results and discussion

There is limited in-depth and representative real-world data available for children of all ages [5]. Such data is needed to draw meaningful comparisons between the type and frequency of injuries that children experience in collisions. Although analyses of the sources described above cannot quantify statistically which body regions are more frequently injured, they can be used to highlight general trends.

For children in front impact collisions the most important findings are:

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- The head appears to be one of the most frequently injured body regions (at AIS \geq 2 levels). Head injuries in front impacts typically occur due to head contact with the interior of the vehicle (rather than through non-contact, deceleration-only mechanisms). Results from the Rhône registry (Appendix A), suggest that the rate is higher for younger children and decreases with age.
- Neck injuries appear to be relatively rare compared to other body regions, especially in older children.
- Injuries to the chest are observed and many researchers note their importance due to the presence of vital organs in the chest cavity. In frontal impact, it is believed that loading by the restraint system (diagonal part of the seatbelt, harness or shield) could be involved.
- Abdomen injuries are also found in both younger and older children. However, there were no abdominal injuries in children below 18 months in the Rhône Registry and no abdominal injuries in rear-facing restraint systems in the GIDAS and CCIS samples. For young children restrained by an integral CRS, some injuries have been documented in the literature and will be discussed in more details below. For older children correctly restrained by the belt, these occur through loading by the lap part of the seat belt, and are also observed in non-integral child restraint systems (as well as belt-only cases). Arbogast et al. (2007) [6] studied cases of abdominal injuries in children restrained by the belt. They found that direct compression by the lap or diagonal belt was the most common injury mechanism. They also identified three kinematic patterns leading to injuries: a “pre-submarining” condition where the lap belt is improperly positioned over the abdomen before the crash, a “classic submarining” condition where the lap belt is over the pelvis before the crash but the pelvis moves under the belt while the torso reclines, and a “submarining and jack-knifing” condition where the pelvis slides under the belt while the torso flexes forward.

Regarding integral CRS, two main architectures seem currently present: 5 point harnesses and shield systems. Attempting to compare them, Langwieder et al (1999) [7] found a trend of lower injury severity with shields. However, the group 1 frontal sample included no serious injuries in the sense of the AIS scale (19 shield cases: 10 AIS1; 30 harness cases: 24 AIS1 and 4 AIS2). Also, 4 point harnesses, which are not in use anymore, were included in the harness sample. Based on the CHILD and CASPER Project accident databases, Johannsen et al. (2012) [8] reported that thoracic and abdominal injuries (including spine) represented 42% of the MAIS2+ with shields vs. 17% with harnesses. However, the results were only based on 31 cases for shields. These cases were sometimes old, included some group 2 shield CRS (which would not be relevant anymore), in both frontal and side impact. In the USA, Edgerton et al. (2004) [9] found higher risks of abdomen/pelvis and chest MAIS2+ with shield booster seat than 5 point harness (12 injuries for 16 cases vs 1 injury for 30 cases, respectively). The risk differences were statistically significant, with large odd ratios but with wide confidence intervals due to the low number of samples. For the abdomen, there was no injury for the harness sample and 25% of AIS2+ (4 out of 16) for the shield sample. However, besides the small sample numbers, the study grouped both frontal and side impacts together. Overall, due to relatively low case numbers, methodological differences and other limitations, these published results seem too limited to support definitive conclusions on real world abdominal injury risk associated with the use of shield CRS in frontal impact.

Beyond their limitations, these studies may have included different CRS evolutions and designs, distributing differently the restraining loads on the pelvis, abdomen and thorax.

2.3 Conclusions

Abdominal and thoracic injuries are relevant for children of various ages. Injuries to older children (belted only or on booster CRS) seem due to the loading from the adult seat belt. Different kinematic scenarios (submarining, improper belt positioning, etc) have been proposed for the abdominal injuries. Injuries to younger children in integral CRS (e.g. shield or harness), could occur through loading from the restraint system, and relative risks associated to different architectures are still being debated. Overall, estimating the loading transferred by the restraint system to the thorax and abdomen seems therefore relevant for children of various age ranges.

When applying these conclusions, it should be noted that there is very limited representative data with enough depth to identify (with any statistical confidence) needs and priorities for improving the performance of child restraint systems for specific age groups, or compare risks associated with CRS types. Although meaningful statistical analysis could not be carried out, trends in the data were highlighted, which can serve as a reference point until representative data are available.

3 ABDOMEN PRESSURE TWIN SENSORS (APTS) FOR Q1.5 TO Q6

3.1 Background and dimensions

Numerous abdominal injury criteria have been proposed in the past (as reviewed by Kent et al., 2008 [10]) including peak belt force, abdominal penetration and more recently pressure related variables (e.g. Kremer et al., 2011 [11]). Most criteria cannot be used without specific dummy instrumentation.

In the P dummies, the risk of abdominal injury has been assessed by looking at the deformation of a clay block behind the abdominal foam. This assessment can be subjective and it does not provide measurements that could be correlated to the possible injury severity but only information about possible abdominal penetration. The belt force can be easily measured, but specific instrumentation such as an Antero Superior Iliac Spine (ASIS) load cell is required to separate the loads transmitted to the pelvis (which are not believed to be injurious) from the loads transmitted to the soft abdomen. For paediatric dummies, Ono et al. (2005) [12] used such a load cell with an HIII 3 YO restrained using different CRS. However, estimating the abdominal loading can be challenging as it may require comparing lap belt and iliac loads (which may have different direction) and does not account for possible loading of the diagonal belt below the ribs. Multiple measurements principles have been implemented to attempt quantify directly abdominal loads. These include the measure of the abdominal compression in the rate sensitive abdomen for Hybrid III dummies [13] & [14], and the Force Matrix Sensor and the Abdominal Pressure Twin sensors (APTS) for Q dummies.

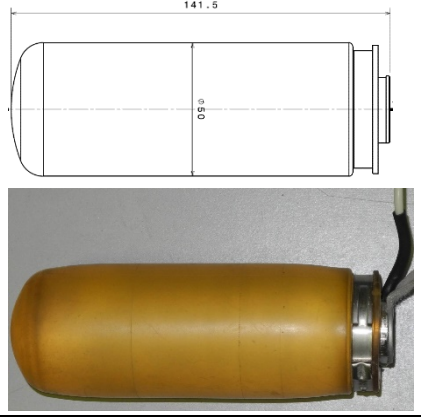
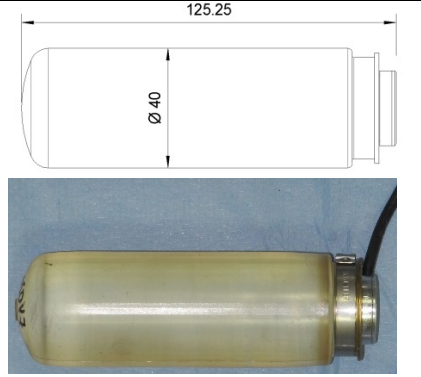
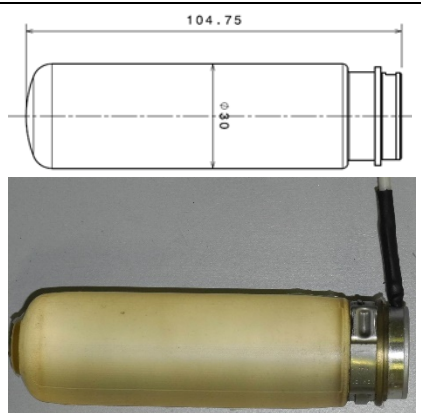
The EEVC report Doc 642 [2] included a detailed description of the current status for APTS in Q10 dummies. This chapter focuses on APTS for other Q dummies, but will provide a brief update on the integration of the APTS in Q10 dummies.

APTS were developed by Ifsttar (formerly INRETS), for Q dummies during the Child European project and improved during the European CASPER and French Proetech projects. The APTS are two soft cylindrical polyurethane bladders closed by aluminium caps and filled with liquid. The pressure in the fluid is measured by a pressure cell implanted in the cap. Three versions (V1, V2 and V3) were developed successively. APTS V1 (50mm nominal diameter) were used with Q3 and Q6 until 2011. They were replaced by APTS V2 and then V3 (50mm nominal diameter) that were used in Q3, Q6, Q10 and Thor dummies. Evolution of the APTS to V3 was mainly due to changes to non-deformable components (cap, wire protection). Smaller APTS V3 (30 mm nominal) are also available for Q1.5 and could be used with Q1 dummies as these two dummies share the same abdomen. The nominal diameter corresponds to the diameter in the mould. Once filled, the maximum external diameter increases slightly. Illustrations of the APTS V3 with their main specifications are provided in table 1. First designed and manufactured for academic research use, the APTS V3 are now produced and sold by Transpolis SAS (France). Transpolis has recently introduced APTS V3 in 40mm diameter form in order to facilitate the integration of the APTS in the abdomen of the Q3 and Q6.

The APTS are mounted inside the Q dummy abdomen in two blind holes parallel to the lumbar spine. The APTS are inserted with the aluminium cap down and the sensor cables

come out of the abdomen through small holes at the base of the abdominal block. The APTS should be held in place by the base, but should otherwise be free to slide within the hole.

Table 1: APTS V3 specifications. Tolerances will be refined with production. All use the same pressure sensors (10 bar nominal measuring range). Data provided by Ifsttar and Transpolis SA.

<p>APTS V3 50mm</p>		<p>Nominal length: 141.5mm Length of rigid region: 16 mm Max. external dia. (filled): 51.5±1 mm Mass: 272±10 g</p> <p>Used in Q10 drilled and final abdomen Used in Q3/Q6 drilled abdomen</p>
<p>APTS V3 40mm</p>		<p>Nominal length: 125.25mm Length of rigid region: 16mm Max. external dia. (filled): 41.5±1 (mm) Mass: 160 ± 10 g</p> <p>Used in Q3/Q6 final abdomen</p>
<p>APTS V3 30mm</p>		<p>Nominal length: 104.75mm Length of rigid region: 16.5mm Max. external dia. (filled): 31±1 (mm) Mass: 81.5±10 g</p> <p>Used in Q1.5 drilled and final abdomen</p>

In the research phase, the holes were drilled using a boring saw while the abdomen was held in a plaster cast. To prevent unwanted slippage during testing, the APTS were held in the holes using fabric hook and loop fasteners (e.g. Velcro) attached to the cap and the lower abdomen. Talc (baby) powder was used to limit friction between bladder and foam.

More recently, there have been efforts to integrate the APTS into abdomens moulded with the required hole. Abdomens were manufactured for the Q10 (for APTS V3 50mm), the Q6 and Q3 (APTS V3 40mm) and recently for the Q1.5 (APTS V3 30mm). A thin Lycra fabric socket closed by a small rope in the upper end is used around the APTS to reduce the friction with the skin of the hole. The socket should be attached to the bottom of the hole to keep the

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sensor in place. Foam plugs are also provided with the moulded abdomen to facilitate current certification testing on the block without the presence of the APTS.

While, the APTS 40mm and moulded abdomens are now available, most research data was collected using previous components. Illustrations of all sensors and abdomens types used in this report are shown with their specifications in table 1, table 2 and table 3.

Table 2: Current specifications for the integration in research abdomens. All tolerances are estimated based the first abdomens.







Drilled abdomens (used in research)				
	Q1.5	Q3	Q6	Q10
Dummy				
Orient.	Parallel to spine (within 5 degrees), symmetric location (left/right, within 2mm)			
Dia (mm)	32.5±2	53±2		
Spacing (mm)	51±2	65±2	61±2	100±2
Location (mm)	22±4 from plane	66±4 from axis of cylinder in back of abdomen	70±4 from plane in back of abdomen	84±3 from plane in back of abdomen
Drilling depth	Stop just before reaching the skin. The skin should be preserved (except for holes needed to pass cables and Velcros. Holes should be finished manually (e.g. with a blade)			

Table 3: Illustrations and specifications of final abdomens.

	Final abdomens				
	Q1.5	Q3	Q6	Q10	
					
	Q1	Q1.5	Q3	Q6	Q10
	036-0000	048-0000	020-0000	033-0000	010-0000
Abdomen					
Drawing nr (standard)	036-5000		020-5000	033-5000	010-4300
Drawing nr (with holes)	036-5010		020-5010	033-5010	010-4310
Width (total)	152 mm		183 mm	173 mm	243 mm
Depth (total)	132 mm		153 mm	128 mm	153 mm
Depth (spine CL to front)	95.4 mm		132.7 mm	118.4 mm	137.3 mm
Height (total)	123 mm		178 mm	155 mm	202 mm
Lumbar recess Radius	Square recess not		22.5	22.5 mm	26.5 mm
Length	engaged with rubber column		52	74 mm	68 mm
Mass			790 ±30 gram	830 ±30 gram	1390 ±45 gram
Volume	470 ±25 gram		2094 cm ³	1587 cm ³	3218 cm ³
Density (average)	1097 gr/cm ³		0.377 gr/cm ³	0.523 gr/cm ³	0.432 gr/cm ³
	0.429 gr/cm ³				
Lumbar and Pelvis					
Diameter	44 mm		44 mm	44 mm	48 mm
Rubber column length	52.5 mm		52.5 mm	82.5 mm	82.5 mm
ASIS Width	134 mm		157 mm	162 mm	192 mm
	(ASIS: 45° tangent point)				
APTS					
Diameter (filled)	31 ±1 mm		41.5 ±1 mm		51.5 ±1 mm
Length (total)	104.75 mm		125.25 mm		141.5 mm
Sensor disk length (btm)	16 mm		16 mm		16.5 mm
Mass (single sensor)	81.5 ±10 gram		160 ±10 gram		272 ±10 gram
Position in Abdomen					
Hole diameter	35 mm		45 mm	45 mm	55 mm
Pitch (CL-CL)	52 mm		62 mm	62 mm	92 mm
APTS CL fwd of spine CL	52 mm		67 mm	67 mm	72 mm
Mass	340 ±45 gram		573 ±45 gram	600 ±45 gram	1030 ±45 gram
Volume	898 cm ³		1661 cm ³	1207 cm ³	2547 cm ³
Density (average)	0.379 gr/cm ³		0.345 gr/cm ³	0.497 gr/cm ³	0.404 gr/cm ³
					
ASIS position (X)	31.7 mm rear CL		15.4 mm rear CL	1.4 mm fwd CL	1.0 mm fwd CL
ASIS position (Z)	19.0 mm above btm		20.0 mm above btm	16 mm above btm	19.3 mm above btm
					

Note: Tolerances will be refined with production.

3.2 Density and mass of the instrumented abdomen

The anatomical region represented by the dummy abdomen is mostly composed of soft tissues and fluids with densities similar to the one of water (1g/cm^3). The APTS are composed of different materials: the aluminium cap is the most dense while the paraffin oil is the least dense (around 0.8g/cm^3). Dividing the APTS masses from table 1 by the volume occupied by the APTS (computed as shown in figure 1) results in average densities of 1.02g/cm^3 for the APTS 50mm and 1.14g/cm^3 for the APTS 30mm. The density of the APTS 40 mm is expected to be in between previous two, based on the proportions of metallic and other components. While slightly above the density of water, such density could be considered as soft tissue like.

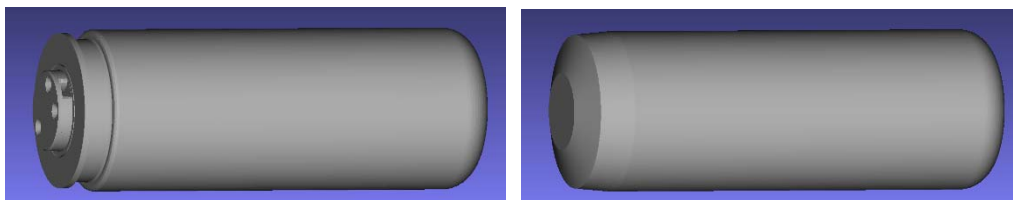


Figure 1: APTS nominal volume. Left: APTS 50mm. Right: A convex hull is used to represent the volume taken by the bladder. The volume of the convex hull is 266 cm^3 for the APTS 50mm and 71.2 cm^3 for the APTS 30mm.

For the standard abdomen (without APTS), the foam has a much lower density. The standard Q3 abdomen weighs around 790 grams (nominal) for a volume of 2094 cm^3 (nominal), resulting in a density of approximately 0.38 g/cm^3 . Similarly, the standard Q6 abdomen weighs around 830g (nominal) for a volume of 1587g (nominal) and a density of 0.52 g/cm^3 .

Partially replacing the foam by APTS therefore results in a mass and density increase. In the case of the APTS V3 with Q3 and Q6 research versions, this is amplified by the fact that the APTS can protrude outside of the abdomen (as shown in table 2). In that case, the mass increase could be up to 352g for the Q3. This mass increase will be mitigated by the use of the 40 mm APTS which have a lower volume and do not protrude outside the abdomen. Also, moulded abdomens with reduced mass were manufactured by Humanetics along with more standard abdomens, leading to further potential reductions. For example, the first two abdominal blocks moulded for APTS 40 mm had masses of 769 grams for the normal abdomen, and 574 grams for a lightweight version. Adding the APTS 40 mm masses (2×160 grams) and using 790 grams as a reference for the standard abdomen, this would lead to increases of 299 grams and 104 grams for the normal and lightweight abdomen, respectively. A complete table including both research and final abdomen is provided for all dummies in table 4.

Table 4: Mass increase due to the abdomen instrumentation. The number of abdomens (n) used for the mass estimates is indicated between parentheses. The source of the data is provided in footnote when relevant. The weight of the sleeves was neglected in the case of final abdomens.

Dummy	Standard abdomen mass (grams)	APTS V3		Abdomen for APTS		Abdomen with APTS (grams)	Mass increase (grams)
		Model (mm)	Av. mass of pair (grams)	Type	Mass (grams)		
Q1.5	470±25	30	163	Moulded	332 (n=1)	495	25
				Research	409 (n=1)	572	102
Q3	790±30	40	320	Moulded	Normal: 769	1089	299
		50	544	Research	Light: 574	894	104
Q6	830±30	40	320	Moulded	Heavier (n=1) 718	1038	208
				Research	Lighter (n=1) 598	918	88
Q10	1390±45	50	544	Moulded	651 (n=1)	1195	365
				Research	Normal: 1222	1766	376
					Light: 1030	1574	184
					1192	1736	346

Based on a preliminary analysis, reducing the mass further could be problematic. For the APTS, lighter fluids exist but seem either flammable or toxic.

Three options could be considered for future use:

- 1) Attempt to reduce the mass of the APTS or abdomen further by using more radical design changes. Such a change would result in an overall abdomen density that would be further away from the human tissues being represented.
- 2) Maintain the whole dummy mass by reducing the mass of other components. Since the mass of the complete dummy matched the original design target, it can be assumed that the mass of the human abdomen was integrated into other components. This would require modifying other components (e.g. pelvis mass).
- 3) Accept an increase of the whole dummy mass to accommodate the APTS. This approach could be justified if the mass increase had limited effect on the dummy response, and/or if an increased mass target for the dummy was considered acceptable (e.g. due to possible changes in anthropometric targets with time). Also, it must be noted that all past tests with APTS were performed with added mass. This includes many accident reconstructions performed within CHILD and CASPER. As a result, risk curves proposed by these projects may include tests with added mass, not only for the abdomen but also for other regions.

The third approach could be the easiest to implement and the mass increase could be mitigated by the use of lightweight abdomens in their final version. Accepting the mass increase was the option tentatively selected by the GRSP IG and present in the working document for the regulation R129 updates.

3.3 Biofidelity and version comparison

3.3.1 Biofidelity response of research version

Belt compression tests and associated response corridors derived by scaling from porcine tests (Rouhana et al, 1989, [15]), define the biofidelity target of the abdomen for the Q-Series dummies. Results for the Q10 were summarized in Wismans et al. 2015 [2] and shown in figure 2. Instrumenting the abdomen with APTS V2 was found to increase the stiffness for the Q10 but the response curve remained on the edge of the corridors. The test had been performed with an early version of the abdominal block and comparisons with other blocks will be provided in the next section. Tests were not repeated for the APTS V3 which share the same pressure cell, dimensions and deformable components as the APTS V2.

Results for the Q6 dummies instrumented with APTS 50mm V2 are presented in figure 3. They suggest that the effect of the sensor on the biofidelity response are minimal, the response being on the upper side of the corridor in all cases.

For the Q1.5, the test setup was modified to have a tighter coupling between the dummy and the mount and to minimize vibrations (figure 4a). Two pairs of sensor were tested in combination with two abdominal blocks. One of the blocks included an internal defect (bubble) that was not visible prior to drilling (figure 4b). An illustration of the test setup and results are provided in figure 4c. The abdomen with bubble had an overall stiffer responses but was still within the response corridor. All other combinations led to very similar responses which were difficult to distinguish from one another. The repeatability seemed good in all cases.

For the Q3, results from older tests with the normal mount, APTS V1 and APTS V2 (Beillas et al., 2012, [3]) are shown in figure 5a. Results from new tests performed with another abdomen, APTS V3 and an inverted mount (as for the Q1.5) are shown in figure 5b. The results are very similar and both suggest that the effect of the APTS on the abdominal response is limited for the Q3.

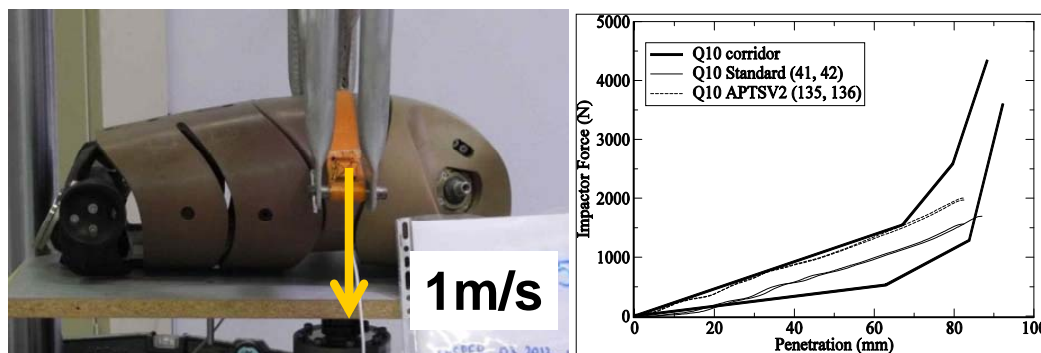


Figure 2: Q10 biofidelity testing: responses for the normal and instrumented abdomen. From Beillas et al, 2012 [16]. Two tests are plotted for each condition.

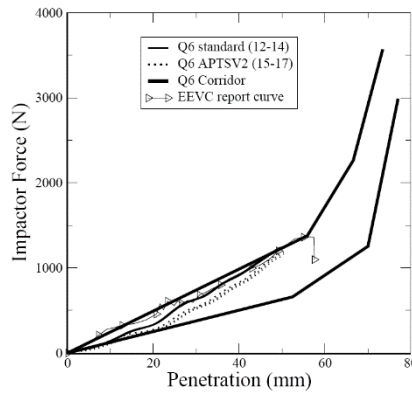


Figure 3: Q6 biofidelity testing: responses for the normal and instrumented abdomen (APTS V2, 50mm) in the biofidelity conditions (Curves from [17]). EEVC report curve refers to the example curve in [1]). Test setup is the same as in figure 2. Three tests are shown for each condition.

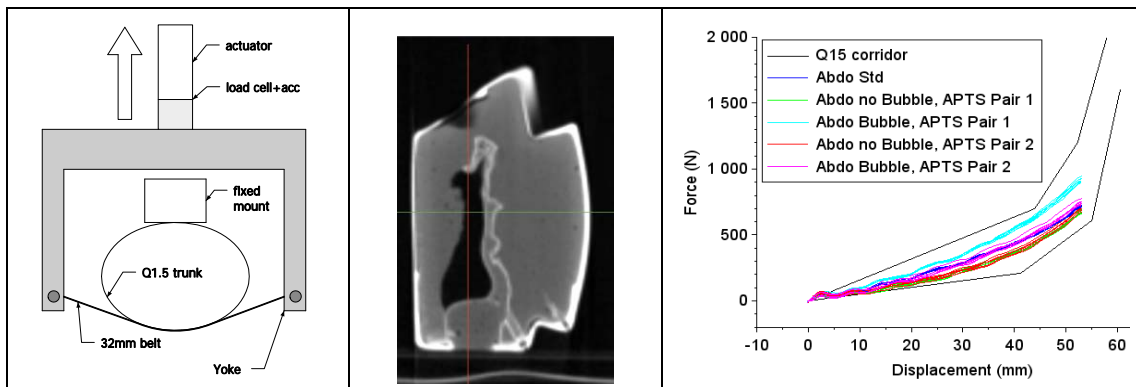


Figure 4: Q1.5 biofidelity testing with APTS 30mm (V3). Left: Test Setup. Middle: CT scan of an abdominal block with a large bubble inside the foam (sagittal section). The abdomen had passed the current certification procedure. Right: Response curves for two abdominal blocks and two pairs of sensors, with 6 repetitions per conditions. The outlier set of curves corresponds to the abdomen with bubbles. All other sets (which include an intact abdomen) are very close to one another.

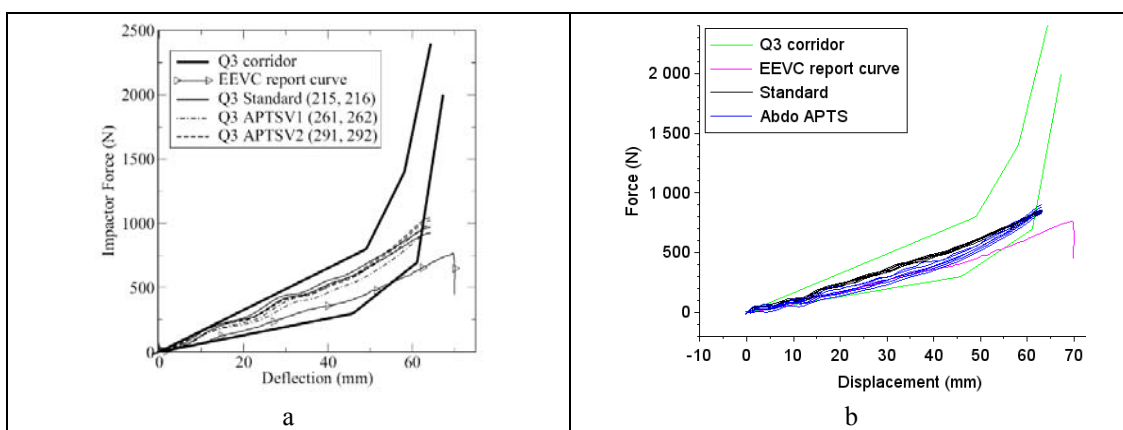


Figure 5: Q3 biofidelity testing a) Test with the mount as in figure 2, standard abdomen, or abdomens instrumented with APTS v1 or APTS v2. Two tests per condition. b) Tests with the inverted mount as in figure 4, standard and abdomen instrumented with APTS V3. Six tests per condition. ‘EEVC report curve’ refers to the example curve in [1].

3.3.2 Pressure response in the biofidelity conditions of the research version

Corresponding pressure response curves in the biofidelity condition are shown in figure 6 for the four dummies. The pressure responses had a small toe zone for all dummies (more marked for Q1.5 and Q3 than Q6 and Q10) followed by a linear portion. The left vs. right response was mostly symmetric, with some discrepancies observed for the Q6 (about 15% difference between the two sides). Also, for the Q1.5 and all combinations tested, one pair of sensors combined with the abdomen with no bubble behaved like an outlier compared to the three other conditions (which were almost superimposed). No explanation was found for these variations.

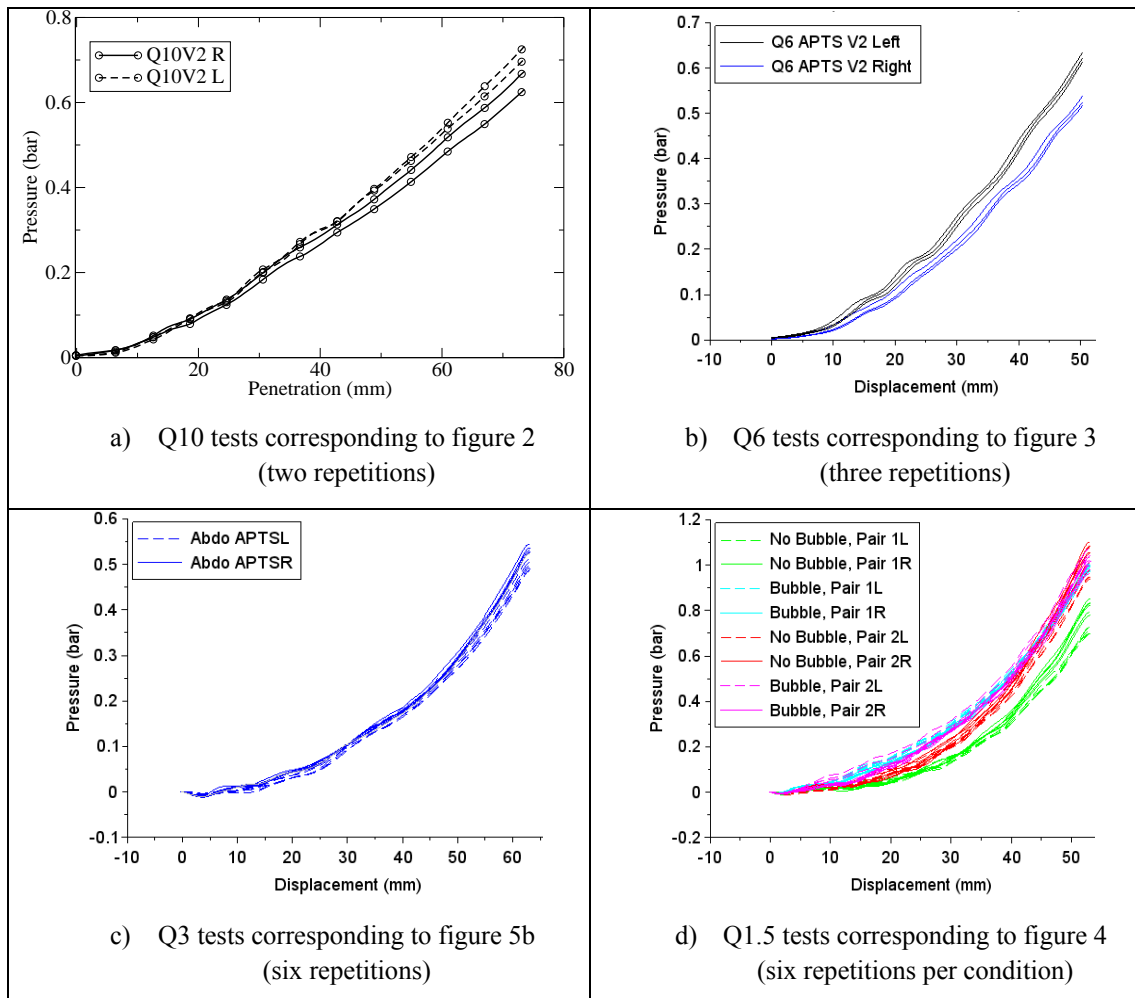


Figure 6: Pressure responses for the biofidelity tests

3.3.3 Comparisons of the responses between versions

A few tests were conducted to compare the responses of the research versions in a drilled abdomen and the final versions with a moulded abdomen. In order to facilitate testing and remove the need for the availability of a full dummy, a simplified test method was developed when comparing Q10 abdominal blocks [2]. The isolated abdominal block, instrumented or not, is positioned on a support similar to the one used in the abdomen certification test. It is then compressed at 1m/s using a 25mm rigid bar. It was observed using this approach that the stiffness response was similar to belt loading up to 60mm, with some stiffening with the bar

after that. The bar also led to higher pressures for the same displacement (which could be expected). Overall, the method seemed adequate to compare abdominal blocks.

Responses for various abdominal blocks and sensors are provided in figure 8 for the Q10, figure 9 for the Q3 figure 12 for the Q6, and figure 11 for the Q1.5.

For the Q10, of the five abdomens tested (which all passed the certification test), two groups of responses can be seen: a stiffer group with the early prototype and a moulded abdomen, and a softer one with two drilled abdomen and one standard block (not instrumented). As such, it appears that the variability of the foam response may be as important as the instrumentation with APTS 50mm. Despite the variations and based on the figure 7, all curves could be within the corridor but are close to the limits in some cases.

For the Q3, five abdomens were tested with the 25mm bar. All abdomens responded very similarly up to 50mm compression (figure 9b). Then the response became stiffer for the instrumented abdomen than the standard abdomen. The stiffening occurred in the compression range where the standard abdomen is too soft because its response is linear. It is therefore not expected to be an issue in terms of biofidelity. The combinations of instrumented abdomens all had similar responses (within less than 10%, figure 9c). Pressure responses were symmetrical within approximately 10% for one pair of APTS40, and much more symmetric for the other cases (overlap of left and right response curves). Once averaged (left and right), the pressure responses were approximately within 10% for all combinations. The APTS50 led to the lowest pressure (and the highest force) but only one abdomen was tested. The lightweight and normal abdomen had similar responses despite their mass differences.

For the Q6, three abdomens (one drilled for APTS50 and two moulded for APTS40) and two pairs of APTS40 were tested, leading to five test combinations. The setup was otherwise identical to the tests for the Q3. To the contrary of the response for the Q3, the abdomens with different densities had different responses: the heavier abdomen (Abd2, 718g) was stiffer than the lighter one (Abd1, 518g) and it has a response similar to the drilled block with APTS50 (figure 10a). The difference between the blocks was lower than the width of the corridor but their exact position with respect to the corridor is not known as the test setup is different. Regarding the pressure response, one pair (APTS401) was less symmetric than the other two (figure 10b) but the differences were less than 10%. For their amplitude, the pressures were similar (figure 10c) to the stiffness results (figure 10a). The response with the heavier abdominal block was within 10% of the research version. Coefficients of variation for all tests were computed and are shown in the repeatability section (3.5.1).

Finally, for the Q1.5, the first production abdomen was tested in the same simplified test setup and compared to the response of the abdomen with the foam plugs and a drilled abdomen (with the bubble defect). The APTS were found to increase the non-linearity of the response, leading to higher forces than the abdomen with plugs above 30mm displacement. The response was also stiffer than with the drilled abdomen (figure 11a). The differences were relatively limited considering the width of the corridor but when considering the results for the belt loading figure 4, it seems possible that the response of the moulded abdomen is towards the edge of the corridor. The symmetry of the response was good (figure 11b) with the production abdomen.

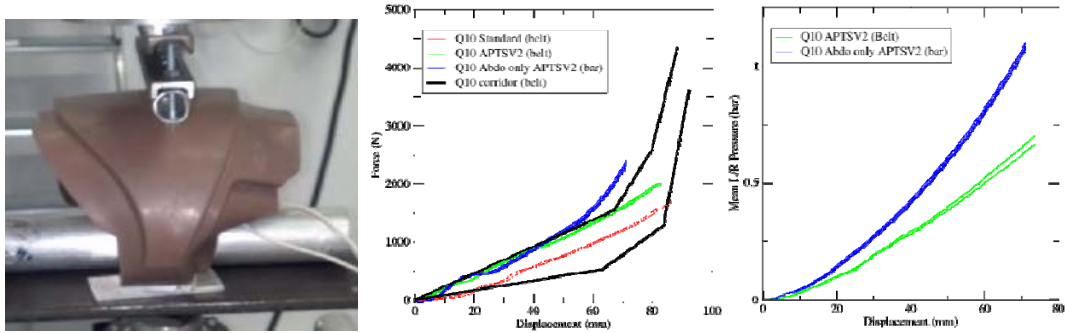


Figure 7: Simplified test setup used for the comparisons. Illustration with the Q10. Left: test setup. Centre: force responses in the 32mm belt testing (biofidelity condition, figure 2) and the 25mm bar compression. The responses are very similar up to 60mm compression. Right: corresponding pressure responses. The rigid bar led to higher pressures (e.g. 0.818 bar vs. 0.51 bar at 60mm displacement).

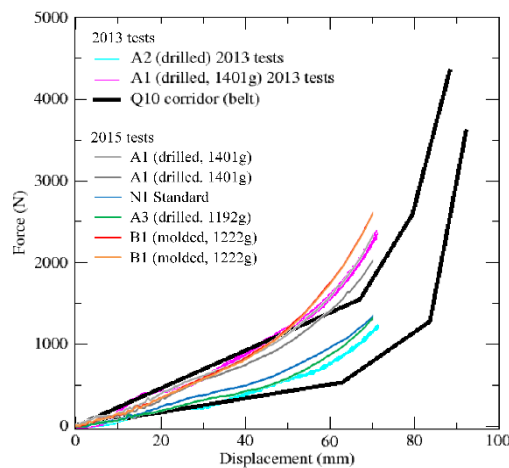


Figure 8: Q10 isolated abdomen response in 25 mm bar tests. Tests were performed in two series in 2013 and 2015, with one block tested twice. The blocks include a standard block (not instrumented, N1), three research blocks (drilled, A1 to A3) and one moulded block (B1). The superimposition between the 2013 and 2015 tests was generated by image overlay.

The use of thoracic deflection criteria balanced with abdomen pressure criteria for the Q-Series in frontal impacts

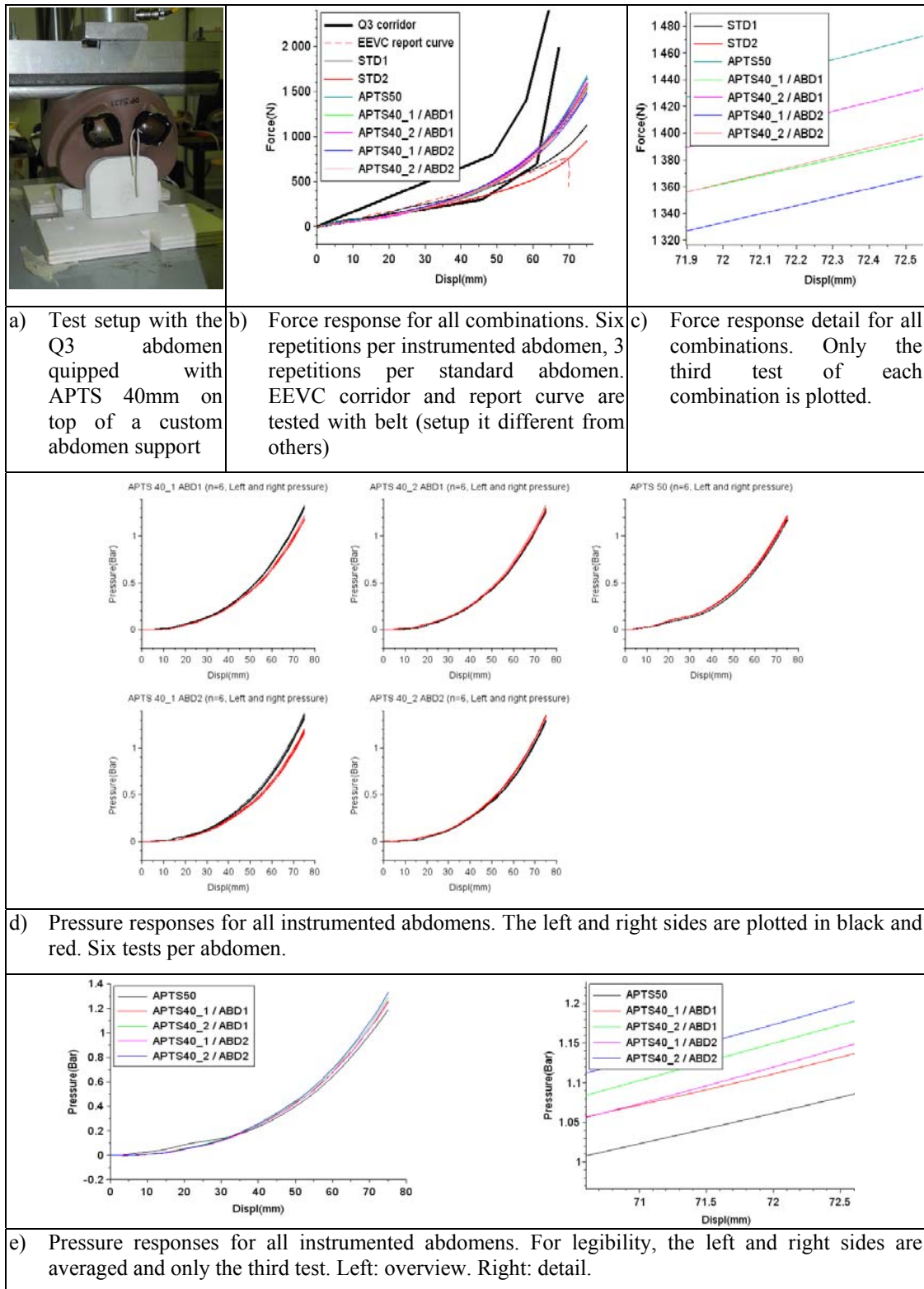


Figure 9: Q3 isolated abdomen response in 25 mm bar tests. The tests included two standard abdomens (no holes, STD1: 779g, STD2: 815g), one abdomen drilled (595g) with APTS 50, two abdomens (ABD1: normal, 769g, ABD2: lightweight: 574g) moulded for APTS40 (two pairs tested APTS40_1, APTS40_2). The Q3 corridor and EEVC report example curve are just provided as context as these were obtained in different test conditions (belted)

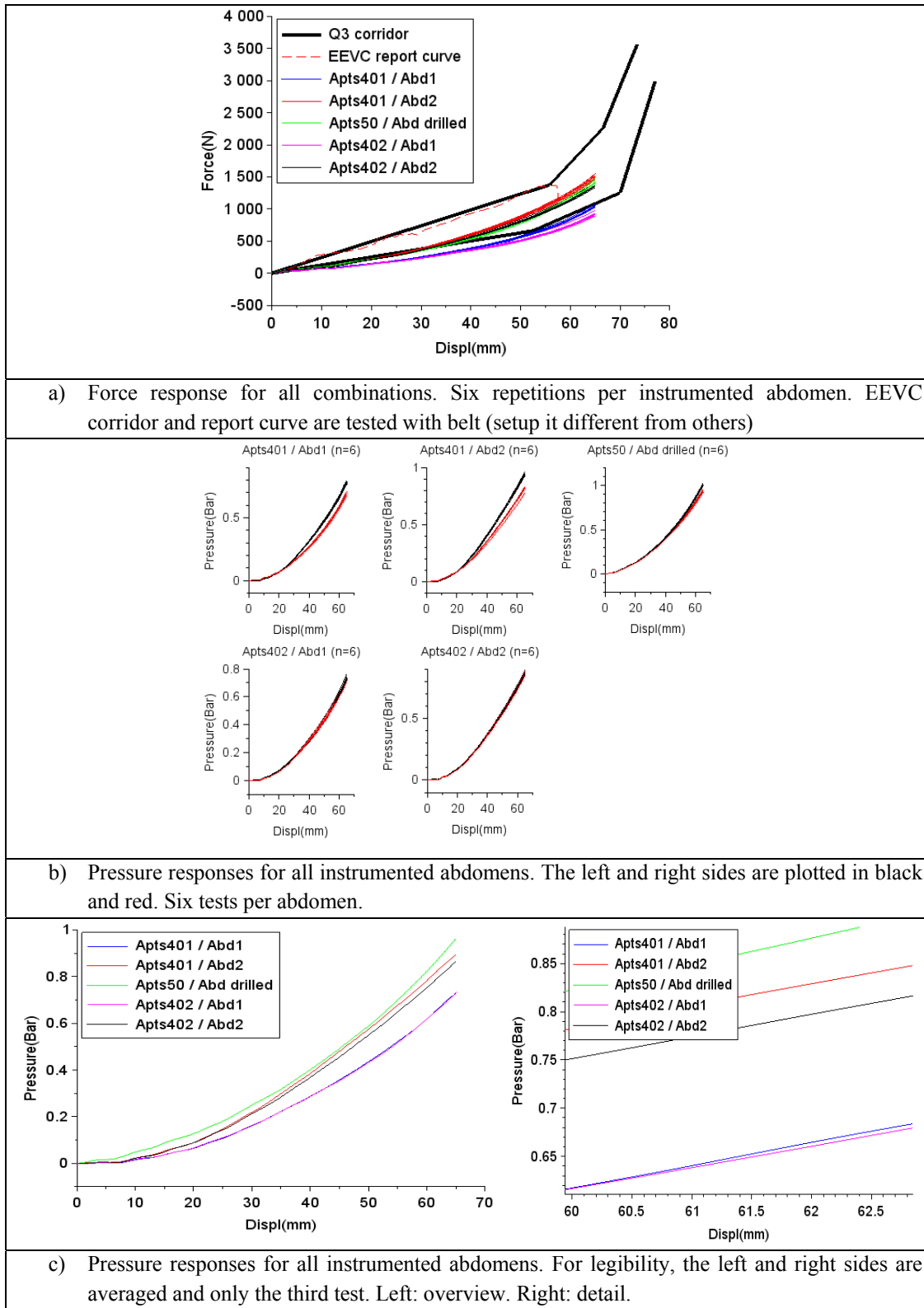


Figure 10: Q6 isolated abdomen response in 25 mm bar tests. The tests included one drilled abdomen with APTS 50, two abdomens (Abd1: 598g, Abd2: lightweight: 718g) moulded for APTS40 (two pairs tested APTS401, APTS402). The Q6 corridor and EEVC report example curve are just provided as context as these were obtained in different test conditions (belted).

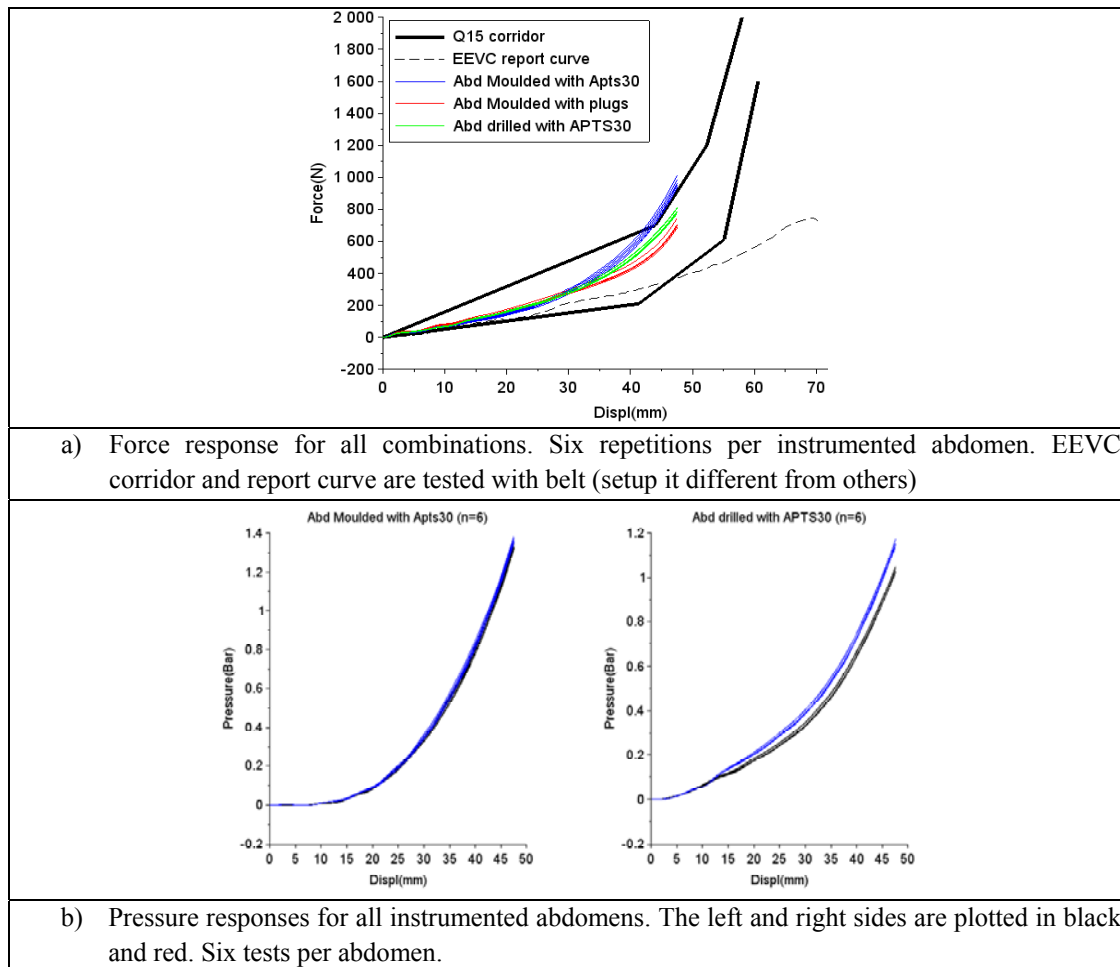


Figure 11: Q1.5 isolated abdomen response in 25 mm bar tests. The tests included the first production abdomen (332g) equipped with APTS30 production sensors provided by Transpolis or with the foam plugs used for the certification testing. For comparison, the abdomen with the bubble defect shown in the previous section but equipped with the same sensors was also tested. The Q1.5 corridor and EEVC report example curve are just provided as context as these were obtained in different test conditions (belted).

3.3.4 Discussions and conclusions

Overall, the effect of the instrumentation of the abdomen on the biofidelity was limited. The largest effect (stiffening) was observed for the Q10. In that case, the response in the biofidelity tests was at the upper limit of the corridor. However for some other Q10 abdomens tested in a simplified setup, the response is expected to be closer to the middle of the corridor in the biofidelity test conditions, suggesting that the effect of the instrumentation could be within the abdomen variability. The response of the moulded abdomen was similar to one of the research (drilled) abdomen.

For Q1.5, Q3 and Q6, the effect of adding the APTS was much more limited. The response was within the corridor in all cases, and near the upper limit for the Q6 with APTS 50mm (as for the standard block tested) and the Q1.5. The pressure response was mostly symmetric, with variations that could be up to approximately 10% in some cases. The APTS 40mm had a response similar to the APTS 50 mm (within less than 10%) when tested in the Q3. Similarly, the results with the APTS 40 mm and one of the Q6 moulded block were similar to the APTS 50 mm (the other moulded block tested being softer).

The results mix tests performed on early prototypes (e.g. one early Q10 abdomen, the first moulded abdomens, the first APTS 40mm) and production components (e.g. standard block). Despite this, overall the results were consistent and sample to sample variations could be expected to decrease during production stage.

Particular attention will be needed for the definition of the abdominal blocks in terms of mass and response. It is recommended to check the abdomen parts together with liquid filled APTS bladders in the abdomen certification test. This to better represent the final configuration. Furthermore it is recommended to explore the possibility to narrow down the current abdomen certification corridor. In terms of recommendation, it should be ensured that the performance of the production components remains in line with the biofidelity target.

3.4 Sensitivity of the research version

3.4.1 Effect of the belt position on the pressure response

The main purpose of the sensor is to distinguish between load transferred to the abdomen and loads transmitted to other body regions. In particular, loads applied to the pelvis through the lap belt – which are not believed to be injurious – should not result in large pressure measurements.

To test this sensitivity, the trunk of the dummy was subjected to compressive loading at 1m/s by a belt as in the biofidelity testing. The belt position was moved along the trunk of the dummy, from the pelvis (below the antero superior iliac spine) to the lower thorax. Results for the Q10 were presented in [2] and will not be repeated here. They suggested that the instrumentation worked as intended by having a low sensitivity over the pelvis and the thorax, and a higher sensitivity over the abdomen. The same protocol was applied to the Q3 (Beillas et al., 2012 [3]), Q6 (Beillas [17]) and Q1.5 dummies. Results, summarized in figure 12, are similar to the ones obtained for the Q10, with (1) a much lower sensitivity over the pelvis and the thorax than the abdomen and (2) similar sensitivities for small position changes over the soft abdomen.

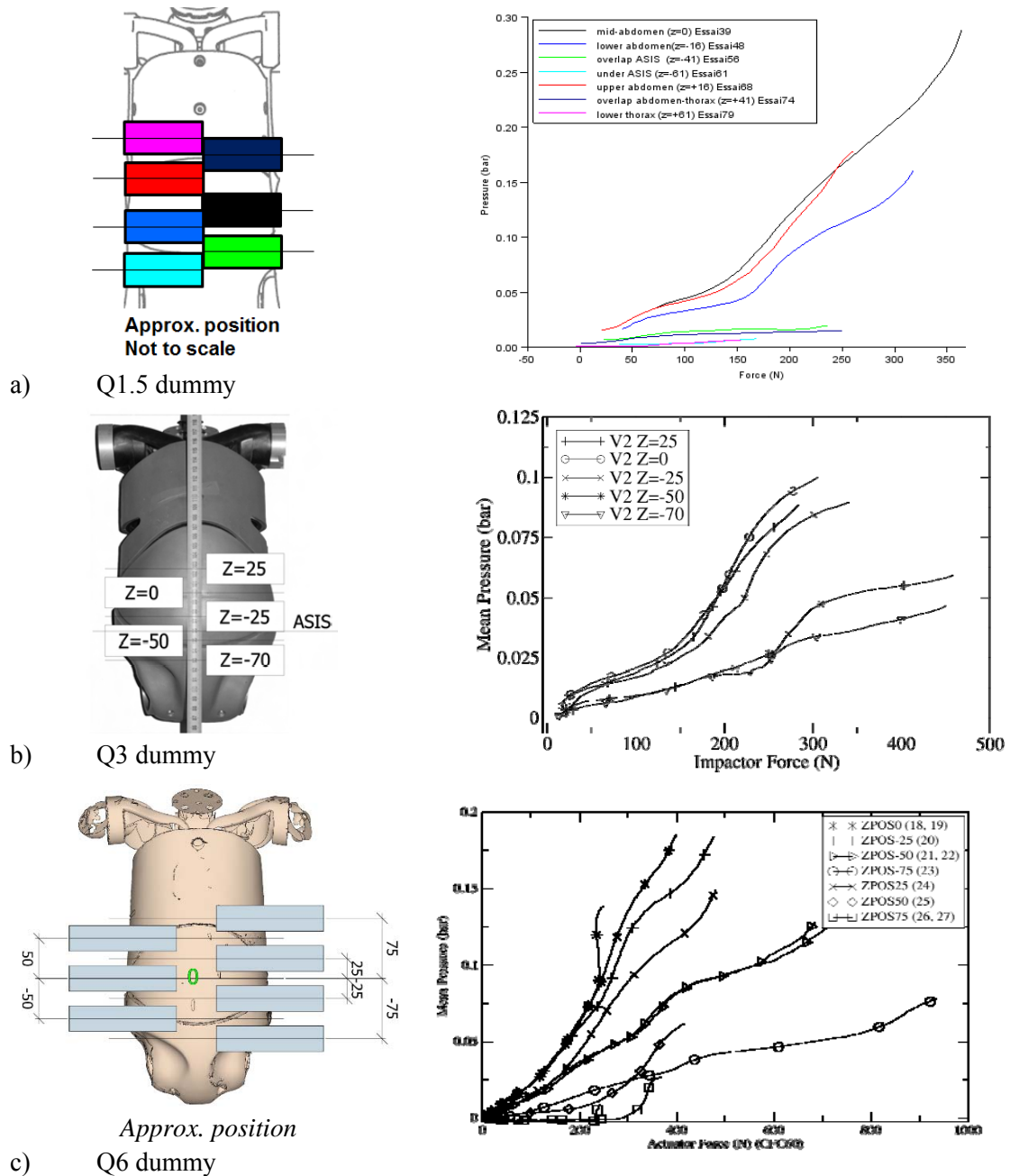


Figure 12: Sensitivity of the abdominal pressure to the position of the belt on the abdomen during belt compression at 1m/s. All tests with APTS V2 except Q1.5 with APTS V3.

3.4.2 Effect of the loading surface on the pressure response

The effect of the loading surface was also studied within the context of shield CRS. The results are described in detail in Beillas and Soni 2014 [18] and will be summarized here.

A Q3 dummy was in inverted position and tested for abdomen/thorax compression at 1m/s as described in figure 13. The compression was applied using a narrow belt (32mm, positioned on the mid abdomen), a standard 50mm belt (positioned to on the mid abdomen or at the limit between abdomen or pelvis), a baseline shield (SH1, with three positions to engage more or less the thorax, abdomen or pelvis), and two more shields (SH2 and SH3, tested only in one position).

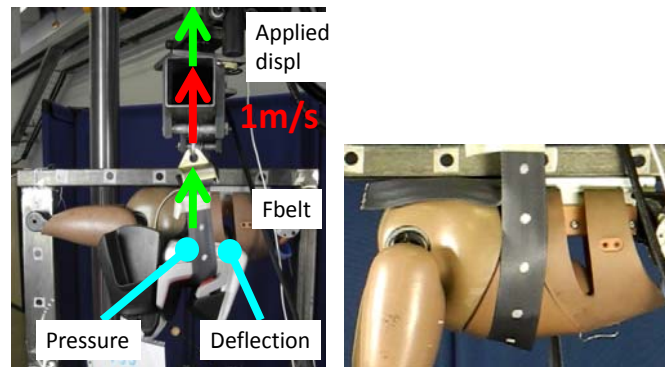


Figure 13: Test setup. The Q3 is mounted face down inside a frame and held at the thoracic spine. The shield (left) or the belt (right) is positioned on the abdomen or torso. The belt is pulled at 1m/s from above the dummy. Measurements include belt force (both sides), applied displacement, abdominal pressure (APTS 50) and thoracic deflection.

The results are summarized in the figure 14. For the belts (figure 14a), the stiffest response (in terms of force vs. displacement) was obtained for the wide belt in the lowest position (partially engaging the pelvis), followed by the wide belt on the mid abdomen and the narrow belt on the mid abdomen. Force vs. abdominal pressure relationships were in the reverse order: for a given force, the highest pressure was obtained for the narrow belt, and the lowest pressure was obtained for the wide belt when the upper pelvis was engaged. Pressures vs. belt displacement relationships were similar for all cases. For the baseline shield SH1 (figure 14b), the stiffness responses were between lower and mid wide belt configurations, and the effect of the shield location on the response was somewhat limited. For thoracic deflection and pressure vs. force responses, the variations were larger and seemed to be related to the geometrical involvement of the pelvis or thorax. For example, when the shield was low, the thoracic deflection was the smallest (as a function of force or displacement) and similar to the wide belt low configuration. Conversely, chest deflections were largest for the mid and upper shield locations. Pressures for belt mid abdomen and high shield (both cases without pelvis involvement) were also similar up to 1000 N loading while the belt low and other shield configurations were similar up to 2000 N (cases with some pelvis involvement). The displacements vs. pressure characteristics were similar for all SH1 configurations and belts. The response with SH3 was similar to SH1 but SH2 had a lower stiffness response and a higher pressure for a given force than SH1 or the wide belt in the mid abdomen (figure 14c). Pressures vs. applied displacement characteristics were again similar with these shields.

The use of thoracic deflection criteria balanced with abdomen pressure criteria for the Q-Series in frontal impacts

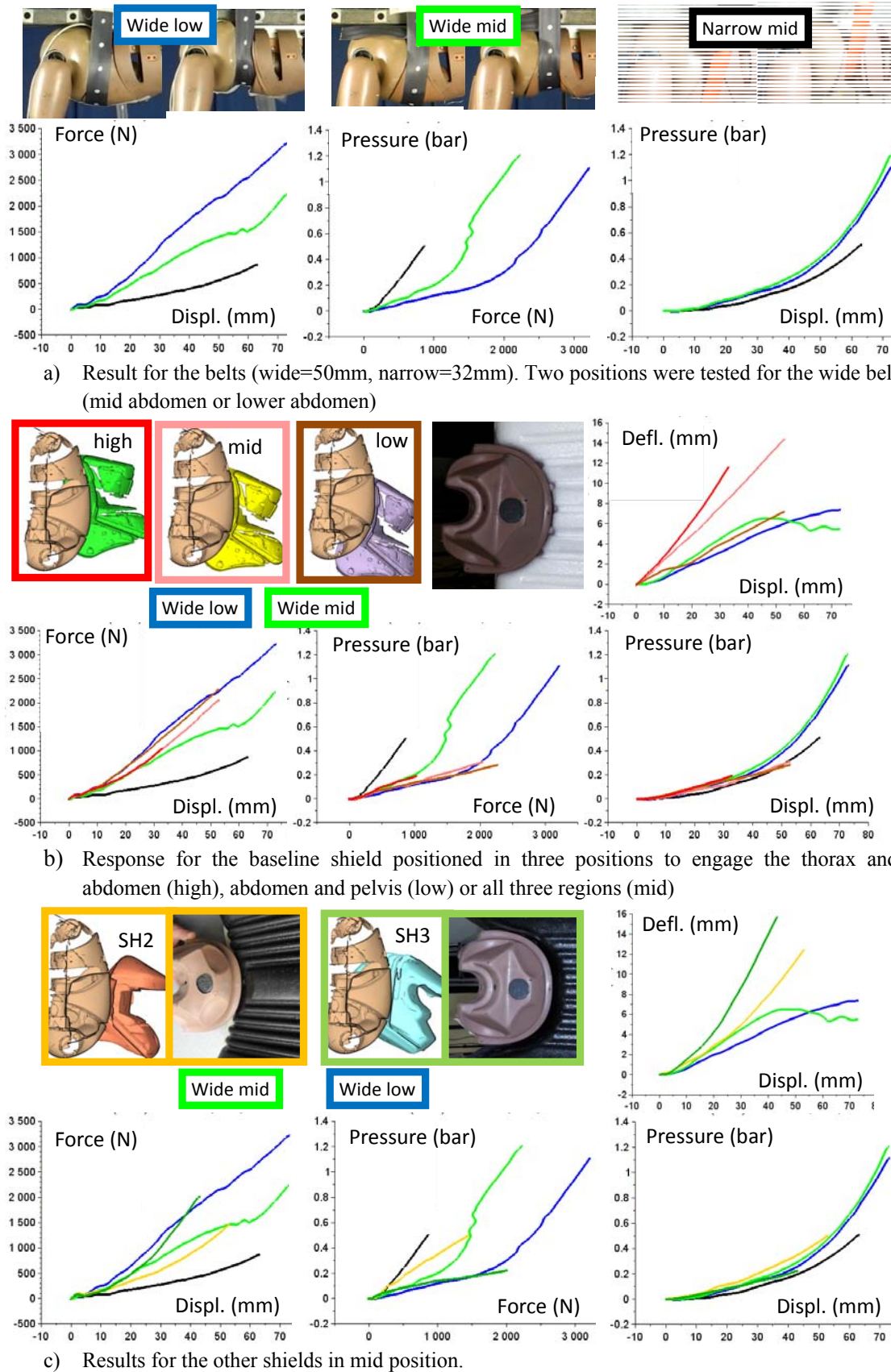


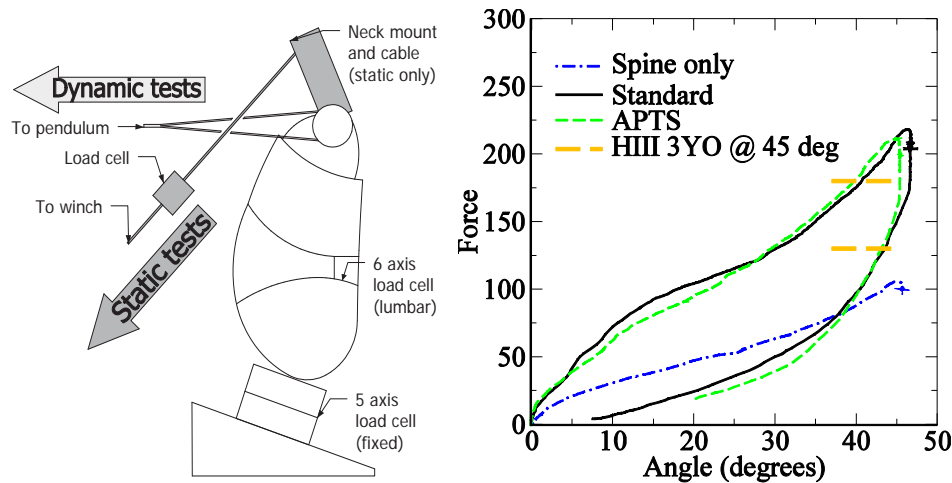
Figure 14: Effect of loading surface (belt, shield) on the response of a Q3 in controlled abdominal and thoracic loading. In all cases, the curve legends (color coding) are the frames of the pictures.

Overall, these tests suggested that, in some cases, shields can effectively distribute loads compared to belts (i.e. higher force for a given abdominal pressure), moreover, this distribution is very sensitive to the shield position. Abdominal pressure and thoracic deflection seemed to reflect the shield interactions with the pelvis, thorax and abdomen. When the pelvis was not engaged at all (SH1 high), the responses were similar to direct abdominal loading in the mid abdomen by a standard lap belt. One reason may be that a standard width belt covers most of the abdomen of the Q3 dummy (similarly to a shield just above the pelvis) and that the shape of the SH1 shield was found to conform to the Q3 abdomen (similar to a belt, figure 14a). When the pelvis was engaged (upper pelvis region), pressures and deflections were lower. The responses of shields SH1 and SH3 in mid position (engaging the pelvis) were similar to the standard belt engaging the pelvis. In contrast, the abdominal pressure was higher for a given force for the shield SH2 (until 1500N approx.). To the contrary of SH1 and SH3, the SH2 shape does not match the Q3 abdomen at the beginning of the loading (figure 14c), which may explain this initial additional pressure. In all cases, it seemed that the abdominal pressure represented the displacement applied to the belt more than the force. This could be explained by the fact that, for the configurations tested, the belt displacement always resulted in abdominal compression while the forces were sometimes transferred to other structures (thorax and pelvis). The main limitation of these tests is that the displacement metric used to represent the compression is the applied displacement (piston). This does not account for variations of belt extensions or shield deformations in various tests. While it is believed that this effect may have been minimized by the use of a similar start position and that the deformations are relatively small considering the applied forces, this was not demonstrated and the lack of video data in most tests prevented a more detailed analysis. Another limitation is that the forces applied and resulting pressures were much lower than those observed in sled testing. This may have prevented the observation of nonlinear phenomena later in the response.

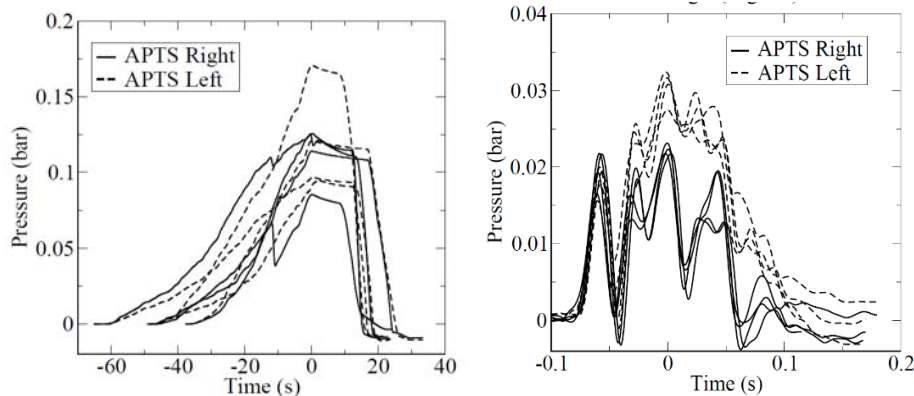
3.4.3 Flexion response

Currently, the sensitivity to lumbar flexion was only evaluated for the Q3 dummy equipped with APTS V1 and tested in a fixed pelvis configuration (Beillas et al., Stapp 2012) [3]. The test setup and essential results are summarized in figure 15. Overall, the effect of flexion on pressure was found to be limited in both static and dynamic testing as shown by the low pressure levels reached despite the large flexion angle. This was attributed to the fact that compression of the abdomen due to the thorax occurs anteriorly to the sensors, and that the sensors are left free to slide in their holes.

The same design principles were used in all the Q dummies and, in the final version, a socket is used to ensure that sliding can occur. However, the tests have not been repeated with final components or dummies.



- a) Test setup (left) and static response in a condition similar to the Part 572 Hybrid III tests (right). The presence of the abdominal block was found to contribute for approximately 50% of the stiffness in torso flexion (right).



- b) Pressure response in static (left, 4 tests) and dynamic (right, 4 tests) conditions.

Figure 15: torso flexion, pressure response and dummy stiffness

3.4.4 Conclusions on the sensitivity

Different test setups were used to test the sensitivity of the pressure measurements to parameters such as the loading location and direction and the loading surface area. Overall, the pressure measurements were found to be sensitive to the loading path: the involvement of the pelvis or thorax led to a decrease in measured pressure. This was true for all dummies in belt testing, and for the Q3 in shield testing. This was to be expected considering the design of the sensors. From a biomechanical point of view, this is a desirable behaviour for the pelvis as pelvis loading is not expected to generate abdominal injuries. For the thorax, it is less clear as the solid abdominal organs (e.g. liver, spleen, kidneys and pancreas), which are frequently injured, are located in the lower thorax. The lack of sensitivity could be acceptable as loading to the lower thorax would be expected to be detected using the deflection sensor (as confirmed in the shield testing).

In terms of sensitivity to the loading surfaces area, loading with shields or belts in the Q3 led to similar responses when they loaded similar anatomical regions and surface areas. A narrow belt led to higher pressure for the same force (as it could be expected) but it is not clear if the increase would be directly correlated to an increased risk of injury.

Regarding lumbar flexion, the lack of pressure sensitivity was targeted in the APTS design. From a biomechanical standpoint, the diaphragm and lower ribs could be expected to apply some loading to the abdomen during flexion. However, there is to our knowledge no quantification of this effect, and furthermore, it is not known whether this could affect the injury risk.

3.5 Repeatability and reproducibility

3.5.1 Repeatability

Repeatability has been assessed mainly using the test results collected to study the biofidelity and sensitivity shown in previous sections. The responses curves from tests corresponding to test repetitions were always close from one another (qualitatively). Examples of quantifications are presented in table 5 for the APTS30, table 6 for the APTS50 and APTS40 in Q3, and in table 7 for the APTS50 and APTS40 in Q6. The coefficient of variation was always lower than 4%. It may have been higher for the tests with the belt than with the bar, which could be explained by differences of complexity and number of components involved.

The only sled tests found for which the number of repetitions was sufficient to compute coefficient of variations are reported in Beillas et al (2012) [3]. A series of 6 tests with Q3 equipped with APTS V2 is reported. The dummy is seated on a booster seat on a R44 bench and subjected to a R44 pulse. The belt is positioned directly on the abdomen to generate significant abdominal loading. The average maximum pressure reached in the tests varied between 2.20 and 2.42 bar, for a coefficient of variation of 4.4%.

Table 5: Repeatability for the APTS30 in the Q1.5 the biofidelity setup (figure 4) (computed at 60mm displacement, six tests per configuration)

	Mean			CV		
	Force (N)	Press L (bar)	Press R (bar)	Force	Press L	Press R
Abdo Std	631	-	-	0.74%	-	-
Abdo no Bubble, APTS Pair 1	581	0.70	0.61	2.23%	3.00%	1.13%
Abdo Bubble, APTS Pair 1	814	0.87	0.87	1.81%	1.45%	1.45%
Abdo no Bubble, APTS Pair 2	602	0.93	0.84	2.99%	1.92%	3.66%
Abdo Bubble, APTS Pair 2	665	0.88	0.92	3.31%	1.48%	2.74%

Table 6: Repeatability for the APTS40 and APTS50 isolated bar setup with the Q3 (figure 9) (computed at 70mm displacement, six tests per configuration)

	Mean			CV		
	Force (N)	Press L (bar)	Press R (bar)	Force	Press L	Press R
apts50	1302	0.97	1.00	1.11%	0.26%	1.34%
apts401abd1	1245	1.08	0.98	1.18%	0.79%	1.55%
apts402abd1	1266	1.05	1.06	1.71%	1.14%	1.62%
apts401abd2	1219	1.10	0.97	2.78%	1.42%	1.71%
apts402abd2	1228	1.06	1.10	1.60%	0.99%	0.50%

Table 7: Repeatability for the APTS40 and APTS50 isolated bar setup with the Q6 (figure 10) (computed at 65mm displacement, six tests per configuration)

	Mean			CV		
	Force (N)	Press 1 (bar)	Press 2 (bar)	Force	Press 1	Press 2
Apts401/ Abd1	1048	0.78	0.69	1.73%	1.49%	1.52%
Apts401/ Abd2	1514	0.94	0.82	2.09%	1.63%	2.58%
Apts50/ Abd drilled	1431	1.00	0.93	2.03%	1.16%	1.67%
Apts402/ Abd1	924	0.74	0.72	3.15%	1.83%	1.53%
Apts402/ Abd2	1369	0.87	0.86	1.84%	1.27%	1.45%

3.5.2 Reproducibility

Since variations of the response in abdomen testing characterize the variation of all components and test setup, a simplified certification test was designed to evaluate the isolated APTS response and quantify variations between APTS. An illustration of the setup and a description of the procedure are provided in figure 16, with results from tests performed by Transpolis SA on 18 APTS50, 4 APTS40 and 6 APTS30 (all v3) summarized in table 8. The CV on the deflection (which would be linked to the compliance) and the measured pressure (sensitivity) was 7% or less.

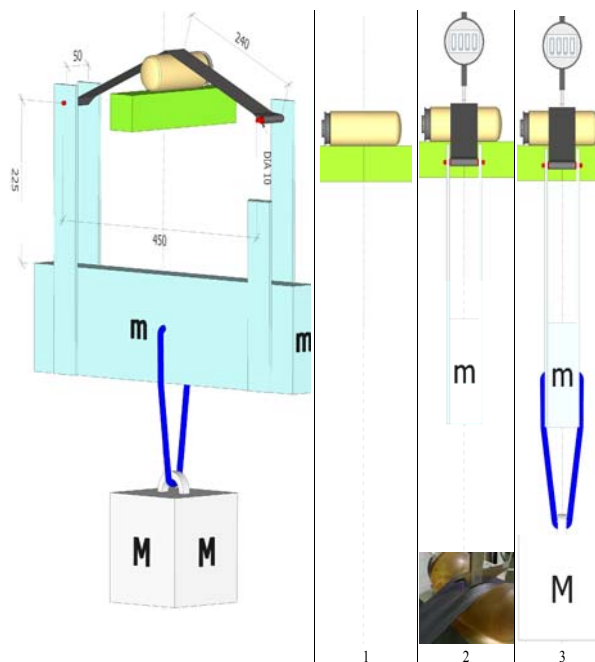


Figure 16: Static verification procedure (certification). Left: setup overview. A standard belt (46mm wide, 480mm long) is mounted on a U-shaped frame (mass 2kg). An additional mass (25.516kg) can be added to the frame. Right: verification steps (the bladder deformation is not shown on the diagrams): (1) the bladder is positioned on a horizontal beam (2) the belt is positioned over the centre of the bladder with the frame hanging below the beam. Initial height (calliper, with example of photo) and pressure are measured (3) The mass M is added and the measurements are repeated. The displacement and pressure variations are computed.

Table 8: Results from the static verification procedure on a series of APTSv3. Data provided by Transpolis SA.

Model (all APTSv3)	N	Mean		Standard deviation		CV	
		Press. increase (bar)	Deflection (mm)	Press. increase (bar)	Deflection (mm)	Press. Increase	Deflection
APTS 50	18	0.60	15.98	0.035	0.90	5.84%	5.63%
APTS 40	4	0.63	16.28	0.036	1.12	5.71%	6.88%
APTS 30	6	1.01	10.63	0.070	0.58	6.93%	5.44%

3.5.3 Discussion and conclusions

The results on repeatability and reproducibility are currently based on a limited number on parts, making it difficult to assess with certainty the variability of the combination of the abdominal blocks and sensors.

For repeatability, results from tests performed on isolated sensors, and combinations of production candidate abdomen and sensors tested in the simplified biofidelity setup (table 5 to table 7) all suggest coefficients of variations below 4%, and the few sled tests performed to study repeatability with the belt directly on the abdomen also provided variations of the same amplitude. This could be adequate for the intended use.

Reproducibility assessed on 28 isolated sensors in a simplified setup led to coefficients of variation on deflection and pressure that are below 7% (table 8). However, when combined with the abdomen, larger variations have been observed.

First, the asymmetry of the response (assessed in the simplified biofidelity setup) reached 7% in average (based on the 14 combinations presented in table 5 to table 7) with a maximum of 14%. The effect of the setup on the asymmetry should be further investigated. In particular, the fact that the displacement is imposed (unlike in a sled test) could amplify the phenomenon. Furthermore, the abdomen is supported by the certification fixture but the efforts applied in the test are much higher than in the certification test and the support may not always be adequate.

Second, significant variations of the stiffness (force for a given displacement) were observed for the Q1.5 (perhaps due to an abdomen manufacturing issue), Q6 (for which abdomens with two densities were manufactured) and Q10 (figure 8). They were not observed for the Q3 despite abdomens manufactured with different densities. In some cases (example of the Q10 in particular), groups of abdomen responses can be made. While the differences between blocks are typically lower than the width of the corridor, the fact that the stiffness affects the pressure measurement in the current test conditions suggests that a reduction of the variability of the abdomen could be helpful. The variations may be amplified by the test configuration (applied displacement in fixed back condition). Also, most blocks tested in this report were not manufactured according to the final specifications listed in table 3.

In conclusion, while the first results are overall encouraging, it is recommended to:

- Continue following the variability of both sensors and abdomens as the number of manufactured parts increase. This is especially relevant for the abdomen for which only few molded blocks have been manufactured according to the final specifications listed in table 3.

- Investigate the effect of the test setup to assess the variability of the abdomen, and the combination of abdomen and sensors. An updated test setup (different from the certification test and simplified biofidelity test) may be useful in the future.

3.6 Durability

No failure of the bladder, or opening of the sensors was reported for the APTS V2 or APTS V3 30mm or 50mm in Q1.5, Q3, Q6 and Q10 tests, both for the prototype version and first sensors in the field. The maximum pressure reached (7 bar) occurred in a Q6 equipped with 50mm sensors during a sled misfire with the belt on the abdomen. The test resulted in the damage of the lumbar spine, the sled and a belt load cell but the APTS remained functional. Issues reported with APTS V2 include small fluid leaks making the sensor greasy to touch, and cable output damage requiring several repairs. APTS V3 were designed in an attempt to solve these issues with a better cable attachment and sealing (production candidate).

The experience with 40mm sensors is currently limited as these were introduced more recently. No damage was reported however up to now. No issues have been reported for the abdominal block either.

Damage of an IR-TRACC in Q3 sled tests was reported by user however (who suggested that the use of smaller APTS could help mitigating the issue).

3.7 Performance in sled testing and full car tests

3.7.1 Introduction

The APTS in Q dummies have been used in many sled testing configurations over the years. However, the results were mostly obtained using research versions with Q3 and Q6. Results are more limited for the Q1.5, Q10, and in general for all final versions (moulded abdomens, APTS 40).

Results with Q10 dummies are described in detail in [2] and will not be repeated here. A few additional tests (accident reconstructions) are now available for the Q10 with APTS 50 in Tylko and Bussi eres 2014 [19]. The results are in line with the report on the Q10 [2] suggesting that the sensors are able to detect abdominal loading. The authors also stated that: "Of interest is the intra-abdominal pressure recorded in the Q10. The elevated values for the left side (2.54 bar) and the right side of the Q10 (2.16 bar) appear consistent with the child injuries. In contrast, pressure measured in the most serious case of belt migration during the barrier tests was 1.16 bar on the left, inboard side and 0.76 bar on the right, outboard side. This pressure sensor adds an interesting dimension to injury risk assessment."

3.7.2 Belted configuration with research version

Numerous tests have been performed as part of research projects or at facilities that have purchased APTS. However, the results are not always public. A few publicly available studies for including APTS for Q1.5, Q3 and Q6 in their research version include (by chronological order):

- Series of sled tests were performed by LAB with a Q6 dummy to study the effect of misuses (CASPER project, reported in Beillas et al. 2012 [3]). The dummy was seated on the back of a Peugeot 807 car body mounted on a reverse sled and subjected to a R44 pulse. Six configurations were tested: one normal restraint condition, with the child installed on a booster with a proper belt fit, and five degraded conditions (no CRS,

relaxed posture, belt under the arms or above the armrest, ...). The restraint performance was first assessed visually using the video. Configurations in which the restraint conditions were deemed to be poor were associated with visible abdominal loading and/or high pressures (above 1 bar). The highest pressures were obtained in the cases with the diagonal belt under the armpit in which both lap and diagonal belt. In the normal restrain condition, there was no visible abdominal loading and the maximum pressure was 0.31 bar.

- Nine sled tests aiming to study the repeatability of the APTS and to compare the response of the APTS V1 and APTS V2 were also reported in [3]. The lap belt was positioned directly on the abdomen of a Q3 dummy seated on a R44 bench and subjected to a R44 pulse. The six tests with APTSV2 had a coefficient of variation of 4.4% (mean peak pressure: 2.32 bar) which was about 10% higher than the average with the APTS V1 (three tests). The study suggests that differences are therefore limited between the versions.
- Still in the same study [3], series of accident reconstructions performed in the CASPER and CHILD projects were reanalysed. The study focused on frontal cases with Q3 or Q6 restrained by the three point belt (n=27). Cases in which the suspected injury mechanisms could not be reproduced were removed (e.g. no visible abdominal loading despite severe abdominal injuries to the child occupant). Two cases with suspected APTS malfunction (belt in the space between upper abdomen and lower thorax, possible upwards sensor migration prior to the loading) were also removed, leading to 19 remaining cases (12 Q6, 7 Q3, equipped with 50mm APTS V1 or V2). The abdominal pressures seemed in overall agreement with the visible abdominal loading and the results were used to develop a risk curve (presented in section 3.8).
- As part of the CASPER project, a silicone prototype was built by Ifsttar to attempt filling the gap at the thigh hip junction in the Q3 dummy [20] & [21]. A series of 34 sled tests with a R44 bench and pulse were conducted on a Q3 restrained by the three point belt on a booster cushion to evaluate the performance of the prototype, along with a prototype developed by Humanetics (patches). Parameters included the position of the lap belt anchorage (moved up from the normal position), the dummy position (e.g. normal or relaxed), a misuse (no use of the belt guides), and a reduced friction between seat pan and dummy. In most cases, there was none or limited visible abdominal loading and the pressures were low (always lower than 1 bar). In the case with high anchorage and relaxed dummy posture, the belt slipped onto the abdomen and resulted in pressures superior to 1.2 bar even when no insert was used. In all cases, the insert prevented penetration into the gap at the thigh pelvis junction but there was one case where the belt was suspected to have moved between the pelvis and abdomen, i.e. below the APTS and in which the pressure was lower than 1 bar while being over the pelvis. Illustrations of the test setup and results are shown in figure 17.
- Series of sled tests by TRL with a Q3 dummy performed in view of the regulatory evolutions (Visvikis et al., 2014, [5]). Sixteen front impact experiments were carried out on a deceleration sled at TRL and in conditions meeting the requirements of UN Regulation 129. Thirteen experiments were with two non-integral child restraint systems under various different conditions. A Q3 (6 tests) and a Q10 (7 tests) were used in these experiments. The three further experiments were carried out with integral child restraint systems and the Q3 to provide supplementary information about the sensitivity of the dummy and abdominal sensors to child restraint design. It was concluded that the sensors did not influence the kinematics of the Q10 greatly or its interaction with the three-point

seat belt. The lap part of the belt followed the same path around the pelvis of the Q10 and was not affected by the presence of the abdominal sensors. There was some evidence for reduced lumbar spine flexion and head excursion (with the sensors), but it was fairly weak. The tests with the Q3 demonstrated sensitivity of the dummy and sensors to restraint system type. There was a marked increase in abdominal pressure when the Q3 was seated in impact shield child restraint systems (figure 18). However, testing in non-integral CRS either with or without lap belt guides showed no discernible difference in the path of the lap belt. The belt stayed on the pelvis in both child restraints, with some intrusion into the gap with the legs (restricted by the pelvis inserts). Visvikis et al. commented that belt guides are not needed to keep the belt on the pelvis of the Q3 dummy in Regulation 129 test conditions.



Figure 17: Illustration of test series to check the performance of inserts to fill the gap at the thigh and pelvis junction. Top: silicon insert using tested and one of the test configurations (here: relaxed and higher anchorage position). Bottom: resulting abdominal loading (associated with pressures above 1.2 bar)

Similarly, the lap part of the seat belt stayed on the pelvis in both non-integral child restraints with the Q10; with and without belt guides. Despite minor differences in the height of the lap belt on the pelvis, the abdomen pressure results were reasonably similar across the two child restraints (figure 19). It was concluded that: “the test procedure, in case of non-integral child restraints, was not able to distinguish between different means for abdominal injury protection since the child restraints performed equally, regardless of the presence of guides to position the lap belt and keep it in place. Furthermore the test procedure was capable of distinguishing between integral child restraints with impact shields and non-integral child restraints. The level of abdominal loading was markedly higher with impact shields so the test procedure may be useful in encouraging improvements in this type of child restraint.”

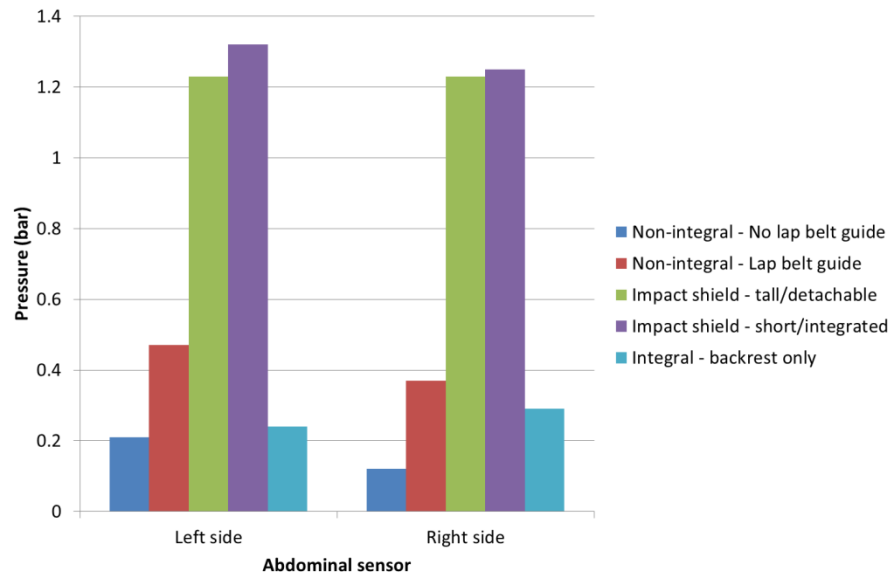


Figure 18: Q3 abdominal pressure measurements in different child restraint configurations, as reported by Visvikis et al. (2014 [5])

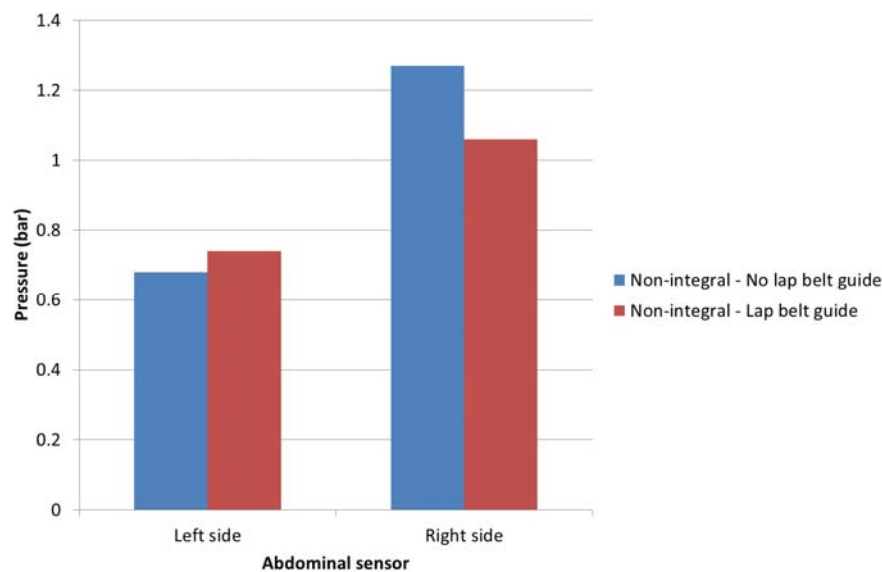


Figure 19: Q10 abdominal pressure measurements in non-integral child restraints with and without lap belt guides, as reported by Visvikis et al. (2014 [5])

- Prototypes of inserts to fill the gap at the pelvis thigh junction were also developed by Dorel and tested in sled tests with APTS in the configuration considered for R129. The results were reported to the GRSP IG on CRS in two occasions: for Q6, Renaudin and Brunetière (2015a) [22] and for Q10, Renaudin and Brunetière (2015b) [23]. For the Q10 without insert, there was no visible abdominal loading (even without CRS or with an inflatable CRS) and the abdominal pressures were low (0.7 bar or less). However, with the insert, abdominal loading and submarining was visible without CRS and with the inflatable CRS. Pressures were above 2.5 bar. Submarining did not occur however with a “normal” group 2/3 CRS and pressures were lower than 0.7 bar too. For the Q6, similar

observations could be made but with less contrast on the pressure results. Without insert, no abdominal loading was visible (even without CRS and with prototype hip shields from Humanetics) and the maximum pressure was 0.83 bar. With the insert, submarining and abdominal loading were visible without CRS and the pressures were 1.21 bar. The pressure raised to 1.64 bar with the addition of the hip shield on top of the inserts. With CRS and inserts no abdominal loading was visible.

3.7.3 Integral CRS with research versions

For forward seating CRS, there are at least two architectures – harness and shield – that appear to generate different loading patterns on the abdomen. Several studies were conducted, partly motivated by the fact that shield systems are currently rated well in consumer tests which did not assess thoracic and abdominal loading. A summary of some publicly available studies is provided hereafter.

- A series of 8 sled tests aiming to study the restraint conditions of a Q3 with APTS with harness or shield CRS are reported in Johannsen et al. (2013) [24]. The four tests with harness led to abdominal pressures of 0.4 bar or less while the tests with shields led to pressures between 1.8 and 2.7 bar, suggesting very significant abdominal loading. This is in line with results available in [16] where a harness and shield CRS had been tested on a NPACS bench with the same accident reconstruction pulse. In these tests, maximum pressures with harness were 0.3 bar while maximum pressures with shields were 1.5 bar or above.
- Series of sled tests performed with Q1.5 and Q3 at TRL to study the conditions of restraints by shield CRS and harness (Visvikis et al., 2014, [25]). Eighteen front impact experiments were carried out on a deceleration sled, complying with the requirements of UN Regulation 129. A sample of five typical harness systems and five typical shield systems were selected for the programme. Test conditions were intended to reflect those used by ADAC in their child safety sled tests, based on an offset frontal impact (64 km/h and a deceleration pulse that peaked at around 40 g). A Q1.5 and Q3, fitted with abdomen pressure sensors were used in the experiments. Visvikis et al. found that shield systems tended to perform better than harness systems for the parameters measured by ADAC and those specified for use in Regulation 129 (albeit with a different pulse) – head excursion, head acceleration, HIC, upper neck resultant force, chest acceleration (vertical and resultant). However, harness systems performed much better than shield systems for upper neck flexion moment, chest deflection and abdomen pressure. With the Q1.5 dummy, the pressure measurements with shield systems were six to twelve times greater than the mean of the harness systems. They were three to eight times greater with the Q3. Although, abdomen pressure was considerably higher with shield systems than with harness systems, one shield system achieved lower abdomen pressure than the others (for both dummy sizes). This indicates that the abdomen pressure measurements are able to distinguish between, not only different styles of integral child restraint system (shield or harness) but also, different systems which may direct various levels of loading through the abdomen.
- As pressures seemed to be consistently higher with shields than harnesses, a study was performed to try to understand some of the mechanisms involved (Beillas and Soni, 2014, [18]). The study combined a study on the sensitivity of the response to the loading surface (shown in section 3.4.2) with four sled tests on a Q3, a geometrical analysis and simulations with a 3 Y.O. human model. As a summary of the findings, the geometrical

analysis suggested that the Q3 and human model had similar geometrical characteristics when seated, and that the edge of the shield was above the bony pelvis, which could lead to abdominal loading. Abdominal loading was observed in the human simulation, with the pelvis passing below the shield. The sled tests were conducted on a R129 bench (normal or with a reduced seat angle) with a R44 pulse. In a sled test without backrest, the motion of posterior rods used to track the pelvis kinematics and the excursion of the knees suggest that the pelvis passed below the shield leading to abdominal loading. This is similar to the loading shown in section 3.4.2 with the shield above the pelvis. This behaviour is also consistent with the human simulations. The corresponding maximum pressure was 1.3 bar. This pressure further increased to 2.1 bar when the bench angle was reduced by using a modified cushion. The pressure was lowered to 1.8 bar when the CRS was strapped to prevent inertial loading. While these results must be considered carefully as there were no repetition, they may suggest that the pelvis is not restrained by the shield but that the pelvis and CRS are partially restrained by the seat angle. Illustration of this study are provided in figure 20.

- Full car EuroNCAP ODB tests were performed with Q1.5 and Q3 to study possible ejection and dummy loading when using shield CRS. The results are summarized in a document presented by the expert of France at the 55th GRSP meeting in Geneva [26]. Two tests included Q3 dummies with APTS. In the first test, it is reported that the abdominal pressure in the Q3 was twice the limit (1.13 bar) proposed in Johannsen et al (2012) [27]. In the second test with a different vehicle and shield, the pressure was approximately three times higher as the provisional threshold.

3.7.4 Results with moulded abdomens and 40mm APTS in Q3 and Q6

Results from a series of sled tests performed by TRL for the European Commission using the APTS40 sensors and moulded abdomens (for the Q3, 'DP5221' 576 g and for the Q6 'DP5273' 598 g) became available recently. However, contrary to testing with the research versions of the abdomens, no fabric hook and loop fasteners were used with the moulded abdomens to limit movement of the sensors in the holes. Whilst no obvious movement of the sensors was detected after testing, the use of hook and loop fastening is recommended for future testing. These tests also made use of pelvis inserts from DOREL for the Q3 and a production ready hip liner from Humanetics for the Q6.

The dummies were tested with either 'No CRS', a 'Poor' CRS (modified to encourage submarining by blocking the lap belt guides) or a 'Good' CRS (an unmodified CRS deemed to have lower belt guides to position the seat belt correctly on the pelvis of the dummy). To encourage abdominal loading further a higher severity pulse was used in some tests and a more slouched initial position (based on the UMTRI positioning procedure and as used with the ten-year-old dummy in FMVSS No. 213) was also used.

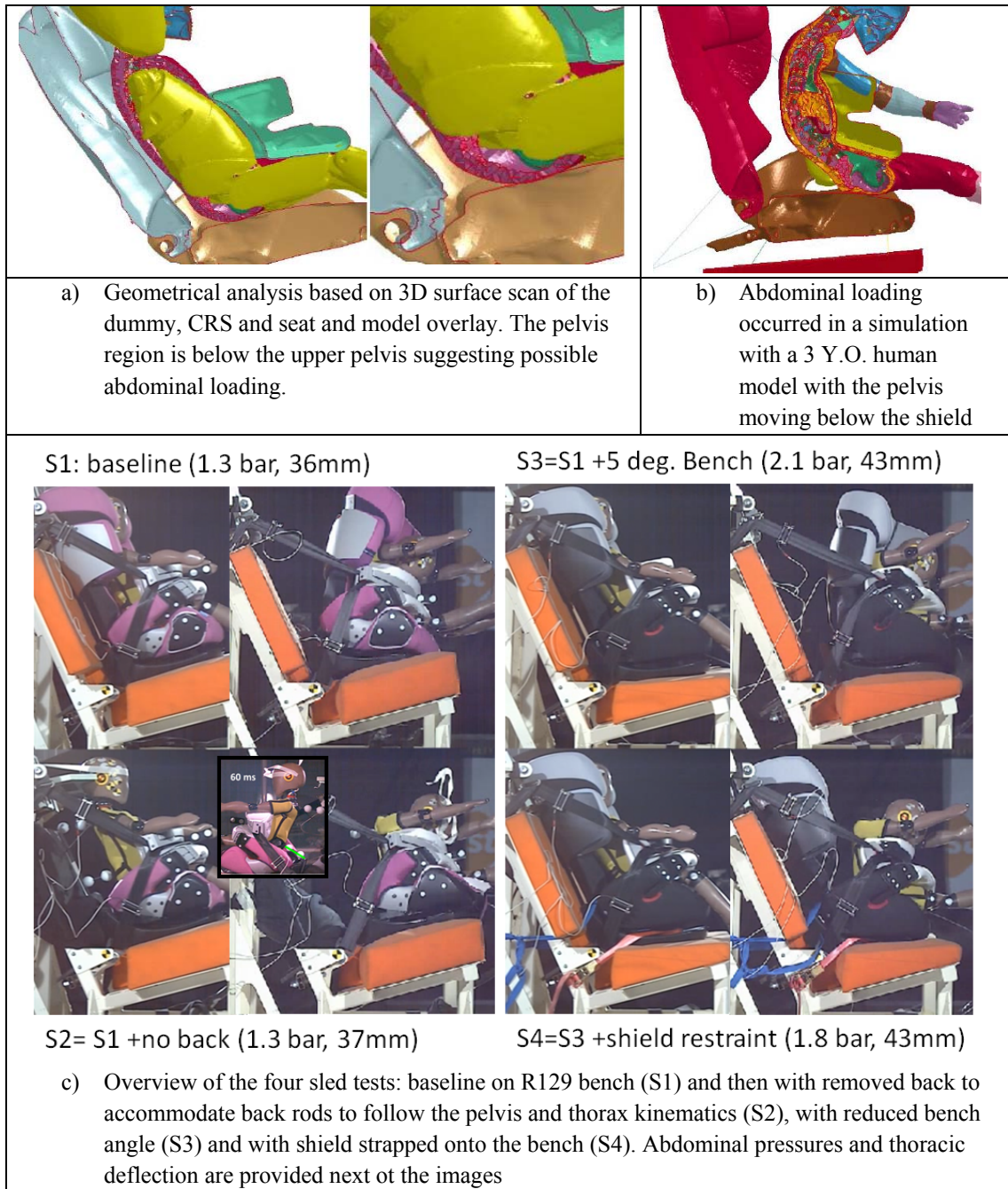


Figure 20: Illustration of an analysis of the shield restraint conditions and abdominal loading (Beillas and Soni, 2014, [18]).

The abdomen results from the sled tests with the Q6 dummy are shown in table 9. In this context submarining is used to mean: a slouching motion with a backward pelvis tilt and slipping forward of the pelvis, resulting in poor routing and interaction of the seat-belt with the body, potentially causing injuries to the abdomen. A condition which leads to the restraint of the child through the lap belt no longer involving the pelvis. table 9 also includes the column, ‘Belt location’. This is to show when there was some loading to the abdomen, pelvis region or upper thighs observable in the video footage of the tests. Abdomen loading could have been severe or it could have been superficial and unlikely to be injurious. It does not necessarily imply that the belt had slipped up and off the anterior superior iliac crest. In some cases, while the belt seems to load the lower anterior aspect of the abdomen, it seems that the

belt remains below the ASIS and that the pelvis skin is mostly involved in the deformation. This scenario is different from more optimal cases where the belt remains lower on the thighs and from the cases where it slips over the ASIS onto the instrumented abdomen. These cases are illustrated in figure 21 to figure 23 for the Q6 and discussed below. The results are also further illustrated in Appendix B.

The Q6, fitted with the Humanetics hip liner, did not submarine when placed directly on the R129 test bench, using any of the different test conditions. Analysis of the videos did show the lap belt appeared to load the lower abdomen of the dummy to some extent and the pelvis skin region (figure 22). The angle of the belt seems to suggest that the loading could have involved only the lowest part of the abdomen and that the pelvis could still be involved too. The peak pressure was highest in the test using the UMTRI installation procedure (left 1.09 / right 1.22 bar). When in the CRS with poor lap belt guidance the Q6 submarined using R129 test conditions (figure 23). The recorded pressures (0.93/0.93 bar) were higher than those in the test without a CRS (0.83/0.89 bar). Using the UMTRI installation method the Q6 again submarined in the CRS with poor lap belt guidance and the measured pressures increased. Equally, using the R129+ pulse meant the Q6 submarined. The combined increase in test pulse severity and UMTRI installation procedure created very high recorded pressures on both sides of the abdomen (1.81/1.39 bar) as the dummy submarined under the lap belt. Using a CRS where the lap belt restrains the pelvis correctly, with the same test condition (i.e. most severe UMTRI and R129+ pulse conditions), the abdomen pressures recorded by the Q6 were 0.32 bar on both the left and the right sides.

Table 9: Q6 abdomen loading results from TRL tests with APTS40 sensors, moulded abdomen and Humanetics hip liner

Test	CRS	Submarining observed	Belt location	Pressure (bar)	
				Left	Right
R129	No	N	Pelvis skin / lower abdomen (figure 22)	0.83	0.89
	Poor	Y	Abdomen. (figure 23)	0.93	0.93
UMTRI installation	No	?	Abdomen / pelvis skin	1.09	1.22
	Poor	Y	Abdomen	1.21	1.18
R129+ Pulse	No	N	Lower abdomen / pelvis skin	1.12	1.17
	Poor	Y	Abdomen	1.16	1.13
UMTRI and R129+ pulse	Poor	Y	Abdomen	1.81	1.39
	Good	N	Pelvis. (figure 21)	0.32	0.32



Figure 21: Good CRS R129+ & UMTRI



Figure 22: No CRS R129 conditions



Figure 23: Poor CRS R129 conditions

The abdomen results from sled tests with the Q3 are shown in table 10 and illustrated in figure 24 to figure 26. The results are also further illustrated in Appendix B. The Q3, fitted with the DOREL pelvis insert and using the CRS with poor lap belt guidance did not submarine under R129 test conditions (figure 25). Analysis of the videos showed that the lap belt could interact with the lower abdomen of the dummy to some extent, although it seems that it is mostly the pelvis skin that is involved and that the belt could have remained below the instrumented part of the abdomen. The peak pressures measured by the APTS were 0.39 bar on the left and 0.47 bar on the right. Similar kinematics of the belt loading the abdomen were also seen when using the UMTRI installation method or the R129+ pulse. However the combined increase in test pulse severity and UMTRI installation procedure did result in the Q3 dummy submarining (figure 26). The recorded pressures in this case were 1.06 bar on the left and 0.95 bar on the right. The pressures measured by the Q3 in a CRS where the lap belt restrains the pelvis correctly, using the most severe UMTRI and R129+ pulse conditions were 0.22 bar on the left and 0.41 bar on the right.

Table 10: Q3 abdomen loading results from TRL tests with APTS40 sensors, moulded abdomen and DOREL pelvis insert

Test	CRS	Submarining observed	Belt location	Pressure (bar)	
				Left	Right
R129	Poor	N	Pelvis skin (Figure 25)	0.39	0.47
UMTRI installation	Poor	N	Pelvis skin / lower abdomen	0.51	0.56
R129+ pulse	Poor	N	Pelvis skin / lower abdomen	0.56	0.58
UMTRI and R129+ pulse	Poor	Y	Abdomen Figure 26)	1.06	0.95
	Good	N	Pelvis (Figure 24)	0.22	0.41



Figure 24: UMTRI and R129+ pulse, Good CRS



Figure 25: Poor CRS R129 conditions



Figure 26: Poor CRS UMTRI installation & R129+ pulse

3.7.5 Conclusions

Numerous tests using different test CRS, test setups, dummies and APTS versions were reported in the literature or made available for this report. Most tests have focused on the restraint conditions with booster seat or cushion and with Q3 and Q6 dummies. Within each study, the abdominal pressures that were measured seem to be in line with the video analysis and the visible loading of the abdomen. Configurations that would be deemed as good restraint conditions (no visible abdominal loading, certified CRS, ...) typically led to the smallest pressures, typically much lower than 1 bar in all studies. Visually severe abdominal loading associated with large pelvis motion (“submarining”) or severe misuse (e.g. diagonal

belt under the arm) led to the largest pressures, most of them (well) above 1 bar. One exception was the TRL study with final sensors, where the restraint with a degraded CRS led to visually severe abdominal loading but a peak pressure slightly below 1 bar namely: 0.93 bar. Studies where the lower abdomen was loaded with possible pelvis involvement were associated with lower pressures, mostly around 1 bar and lower in the TRL study (Q6 and Q3). The same trends had been observed with the Q10 results presented in a previous report [2], perhaps with higher pressure levels corresponding to the transition between appropriate and inappropriate restraint (between 1 and 1.5 bar). It must also be noted that, to the contrary of the Q10, the diagonal belt did not seem to contribute to the abdominal loading as it moved toward the neck in most cases.

Overall, these results seem to confirm the ability of the system to detect abdominal loading in sled configurations and is in line with results on sensitivity from the section 3.4.1.

While they are generally consistent from study to study, more detailed comparisons between studies are made difficult by the fact that most tests were not repeated, that test setups were different (e.g. bench, pulse, posture), that sensors and abdomens sometimes differed (version or variability) and that inserts were sometimes used to prevent lap belt penetration into the gap at the junction between thigh and pelvis in order to allow submarining. In the case of submarining, it must also be noted that the loading seen by the abdomen may be dependent on the timing of the belt slippage and the duration/magnitude of the pulse, which can be affected by test conditions (inserts, pulse, etc.). This will be further discussed in the section on dummy kinematic behaviour.

For integral CRS, the situation is somewhat more complex given the different restraint system approaches available. All tests with a harness system, when used correctly, yielded low values of abdominal pressures (typically less than 0.5 bar). This is in line with the almost complete absence of abdominal injuries observed with that restraint condition, and the fact that the loads seem to be transferred predominantly to the pelvis and shoulder. Tests with shields consistently led to much higher pressures whether in Q1.5 or Q3. The results seemed sensitive to the shield model however with significant variations between models and configuration (from 1.3 bar to over 3 bar). It appears that these pressures may be attributed to direct abdominal loading alone as the pelvis does not seem to be restrained by the shield due to the restraint system geometry. The issue had however to be observed indirectly (backless test, isolated test, human simulation) as the abdomen cannot be seen directly in the test. The same phenomenon had been identified by Tanaka et al. (2009) [28]. The pressure levels in sled tests seem to correspond to severe cases of abdominal loading by the belt. This observation matches the component tests summarised in section 3.4.2 for which pressure levels with a shield were similar to mid abdomen belt loading when the pelvis was not engaged.

3.8 Injury assessment: risk curve and scaling

3.8.1 Introduction

The abdominal sensors are able to discriminate between abdominal loading and loading to the thorax and abdomen, both in direct belt loading (sensitivity tests) and in sled (detection of visible loading). The pressure is increasing with the force or the compression applied to the abdomen, which are both injury predictors (e.g. Kent et al., 2008, [10]). The pressure measured in vessels has also been correlated to the risk of injury. Kremer et al (2011) [11] proposed 50% AIS3+ risk at 0.64 bar based on PMHS tests. However establishing

relationships between parameters collected on elderly PMHS or porcine specimen and paediatric subjects is challenging due to geometrical and possible material changes due to age and/or species. Linking those human subject parameters to dummy readings would be another challenging step due to structural differences between the dummy and the human.

An alternative approach consists of performing accident reconstructions with instrumented dummies and attempting to relate the dummy readings with the injuries observed in the field.

3.8.2 Risk curve based on accident reconstruction

Numerous accident reconstructions involving child occupants have been performed as part of the CREST, CHILD, and then CASPER projects. Of these, 19 reconstructed cases involving children restrained by a three point belt with or without booster seat have been selected as valid and were used to build risk curves. The cases include 12 cases reconstructed with Q6, and 7 reconstructed with Q3, all equipped with APTS 50 mm. Details about the selection, methodology used to build the risk curves and injury metrics can be found in Beillas et al. (2012) [3]. Risk curves using the maximum pressure filtered according to SAE CFC 180 are shown in figure 27. Correcting the Q3 data by scaling using the calcaneal tendon stiffness or failure stress as proposed in Palisson et al (2007) [29] did not help in reducing the confidence interval on the combined Q3 and Q6 results.

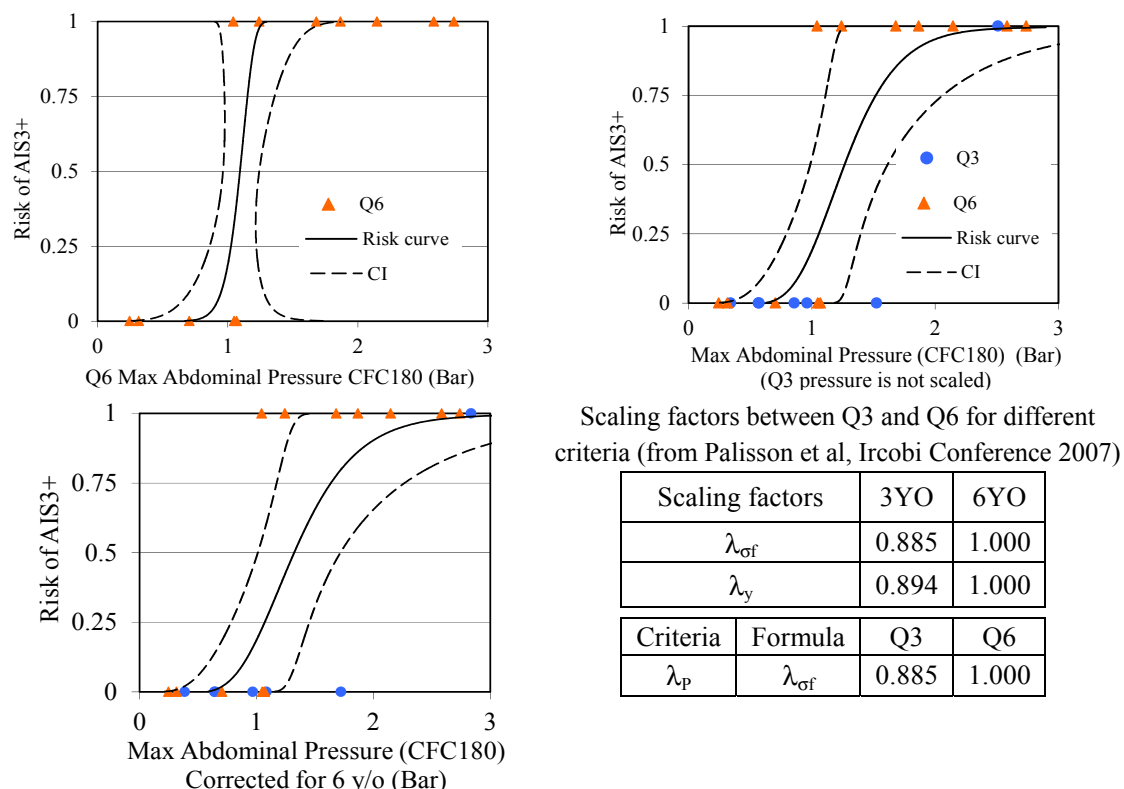


Figure 27: Risk curves for maximum pressure filtered CFC180 based on 19 accident reconstructions (12 Q6, 7 Q3). From [3]. Top left: risk curve for the 12 Q6 (1.09 bar at 50% risk). Top right: risk curve for all cases combined (no scaling, 1.26 bar at 50% risk). Bottom: risk curve for the Q6 and scaled Q3 values (1.32 bars at 50% risk). The scaling factors are provided next to the plot.

3.8.3 Discussion on scaling and conclusions on risk curves

Results obtained by accident reconstructions with Q3 and Q6 restrained by belts appeared as the most relevant for the estimation of injury risk curves. The curves should be considered cautiously considering the limited number of cases and the inherent limitations of accident reconstruction based approaches. Also, none of the reconstructions were performed with a Q10, Q1.5, or with final versions of the APTS and abdomen and only cases with belt restraints were considered (no cases with harness or shield).

Scaling the tolerance between dummies could be attempted but the validation of the scaling assumptions would be problematic. More specifically, for pressure scaling, the method based on the use of calcaneal tendon properties did not help improving the confidence on the risk curves shown in the figure 27. Alternative scaling approaches that would be more directly based on pressure could be attempted. However, these would require (1) assuming that the pressure measured in the APTS is equivalent to the pressure measured in hollow organs or solid organs for all dummies/ages (2) obtaining corresponding pressure values for the possible age ranges. While some age dependent properties are available in Yamada (1970) [30] for the intestines and the kidney, the age dependency is not detailed within the age class 0-9 years old but the standard deviations within the age classes suggest that they could be relatively small. The first assumption on pressure equivalence would still need to be verified however (e.g. using adult dummies of similar proportions and corresponding PMHS tests).

Scaling using an intermediate variable (e.g. force, compression) that has also been shown to be an injury predictor could represent an alternative. Abdominal compression would be the most obvious candidate for such scaling as force could depend on the loading path as shown in previous sections on sensitivity to the loading surface for example (3.4.2). Kent et al. (2008) [10] have suggested 38% abdominal compression for a 50% risk of AIS3+ based on belt tests performed on a porcine paediatric model. While the penetration speed in case of submarining may be smaller than some of the speeds used in these test (3 to 8 m/s), the number could still be used to obtain equivalences between dummies.

However, even for this variable, scaling could be challenging as the abdominal depth covered by the abdominal block is very dummy dependent (as shown on figure 28) and may differ from both porcine specimens and human subjects of different ages. For example, still based on dimensions shown figure 28, 38% compression on the Q1.5 would lead to $0.38 \cdot 135 = 51$ mm penetration for a 71mm soft abdomen (leaving only 20mm in front of the lumbar spine to include the abdomen skin and bladder envelope) while 38% on the Q3 would lead to $0.38 \cdot 162 = 62$ mm penetration for a 111 mm soft abdomen. It is unlikely that these two cases could be equivalent in a human in terms of pressure response.

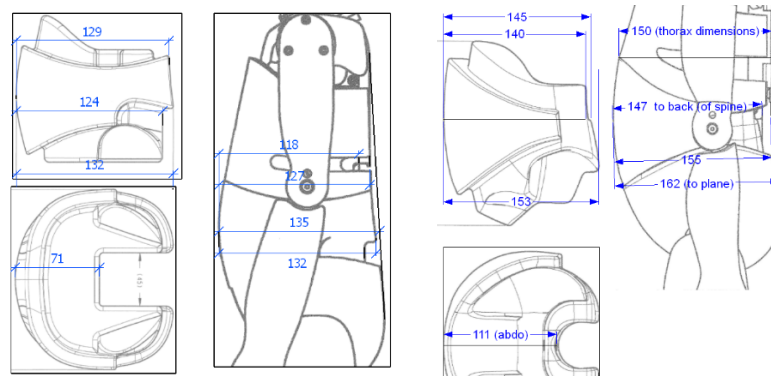


Figure 28: Dimensional difference between dummies: examples of the Q1.5 and Q3. The soft abdomen of the Q1.5 (left) represents only $71/135=53\%$ of the abdominal depth while the soft abdomen of the Q3 (right) represents $111/162=67\%$ of the abdominal depth. All dimensions (mm) were measured approximately on the drawing provided in the R129 phase 1.

Overall, the numerous limitations of the scaling approaches and the difficulty in verifying the underlying assumptions question the direct application of a scaling approach for other dummies. Looking at real world performance may be more appropriate to establish a pragmatic correspondence between dummies. For example, in the case of the Q10, it was found in Wismans et al. (2015) [2] that the transition between visually acceptable and unacceptable behaviours appeared between 1 and 1.5 bar, which is in the same range as the Q3 and Q6 tolerance levels. There is currently no such equivalence available for the Q1.5.

3.9 Discussion and conclusions

The current status of the APTS for Q1.5 to Q6 dummies is summarized in the current section. It should be considered along with the report on the Q10 [2] to cover the whole Q dummy family seated in forward facing configuration.

The APTS design and abdomen is now close to finalization with the development of moulded abdominal blocks for all dummies and the introduction of 40mm APTS that better fit in the Q3 and Q6. With lighter abdomens, they could limit the mass increase of the dummy to 200 grams in the Q10, around 100 grams or less in the Q3 and Q6, and 25 grams in the Q1.5. Considering the fact that numerous tests were conducted with higher mass increase due to larger sensors, heavier blocks, and inserts at the hip, nothing in the analysed data suggests that this mass increase could be unacceptable.

The effect of APTS on the biofidelity response, analysed in biofidelity testing or simplified testing with a rigid bar, seems acceptable considering the width of the reference corridor and the variability of the foam. The abdomen properties need to be finalized keeping in mind the biofidelity requirements and efforts to reduce the foam variability should be considered.

The sensors were found to detect abdominal loading both in tabletop tests and in sled testing (sensitivity). Repeatability and reproducibility testing was more limited. Reducing the variability between abdomens and between sensors should be considered with the final production versions. The variability requirement have to be considered within the context of sensitivity and of the response and the fact that acceptable and unacceptable behaviours are often widely separated in sled tests. To better characterize the variability, the certification

The use of thoracic deflection criteria balanced with abdomen pressure criteria for the Q-Series in frontal impacts

requirements may need to be updated as abdomen and APTS are currently tested separately using simplified quasi-static loading under very small loads for the abdomen.

Concerning the selection of a possible tolerance, risk curves were obtained for Q3 and Q6 based on accident reconstruction in belted configuration. Pressure levels slightly over 1 bar seem to correspond to a limit between behaviours which are visually acceptable or unacceptable. Based on the risk estimate, such a limit is likely to relate to a risk of AIS 3+ injury of between 20 and 50%. For the Q10, the same separation may occur at slightly higher pressure levels (See [2]).

For the Q1.5, there is currently no such equivalence as the dummy has not been tested in belted configurations. Regarding the use with integral CRS, it was found in Q3 tests that loading the abdomen by a shield covering most of the abdomen gives similar results to loading by a wide belt also covering most of the abdomen. While the result is limited to one study, it may suggest that the risk could be equivalent for both systems.

4 THORACIC INJURY CRITERIA

4.1 Background

As identified in Section 0, chest injuries are less common than those of the head, abdomen and extremities for children in non-integral child restraint systems (Kirk, 2012 [4]). Nevertheless, the chest is an important body region to protect because it contains organs that are essential for life, such as the heart and lungs. There are, in general, two main mechanisms for chest injury in vehicle occupants: localised forces applied directly to the chest by the interior of the vehicle (including the restraint system) or sudden chest deceleration and its effect on the internal organs (Cavanaugh, 2002 [31]).

4.1.1 Chest acceleration

Chest acceleration (sustained for 3 ms or longer) is a performance criterion in UN Regulation 129. The measurement is made on the thoracic spine of the Q-Series dummies at a location that corresponds to the fourth thoracic vertebra (T4) in humans. A pragmatic limit of 55 g is specified, which is the same as that used with the P-Series dummies in UN Regulation 44. This limit was chosen based on comparisons between the P-Series and the Q-Series undertaken by Working Groups 12 and 18 of the European Enhanced Vehicle Safety Committee (Wismans et al., 2008 [1]).

Chest acceleration provides a general indication of how well a restraint system allows the occupant to “ride down”, or absorb the collision forces in a manner that avoids injury. The basis for the measurement as a predictor of serious chest injury stems from the results of human volunteer rocket sled testing by Stapp (1970 [32]) as well as instrumented stunt diver tests by Mertz and Gadd (1971 [33]). However, Kroell (1971 [34], 1974 [35]) demonstrated the importance of chest deformation, using blunt thoracic impacts of unembalmed (adult) cadavers. Chest acceleration may not, therefore, distinguish fully the potential contribution of compression loading (i.e. if it is used as the sole means of assessing the risk of chest injury). It may be useful only as a crude predictor of risk from a well-distributed load (assuming that the duration is short and that the load does not result in significant chest deformation). However, it should be noted that the chest of the Q-Series dummies is designed to be deformable (in contrast with the P-Series dummies). Deformation of the chest during an impact event may be associated with a reduction in chest acceleration compared with a rigid thorax. Therefore, care should be taken to avoid the situation that chest deflection is used as a means to reduce chest acceleration measurements.

4.1.2 Chest deflection

The Q-Series dummies are equipped with a compliant ‘rib cage’ and are capable, therefore, of measuring chest deflection. The measurement is made with an IR-TRACC (Infra-Red - Telescoping Rod for Assessment of Chest Compression) at a point that corresponds to the mid-sternum in humans. EEVC Working Groups 12 and 18 proposed scaled deflection limits for the Q-Series, but they were not adopted in the new Regulation. One of the main problems with these scaled limits is that they are based on the risk of rib fracture only and do not account for injuries to the organs and soft tissues (see also 4.2). In adults, particularly those from middle age and above, serious chest injuries tend to occur with rib fractures (Ziegler and Nikhileshwer, 1994 [36]; Holcomb et al., 2003 [37]); a fracture-based limit might be appropriate, therefore. However, for children (and young adults) serious chest injury can

occur without rib fracture (Arbogast et al., 2013 [38]). Therefore, whilst a chest injury threshold that takes significant account of rib fracture may be appropriate for adults, the link between deflection and rib fracture occurrence and serious chest injury is likely to be less robust for children.

A group of interested experts has formed a Q-Series chest and abdomen injury criteria task force (Johannsen et al., 2013 [39]). One of their key aims is to derive a chest injury risk curve, based on deflection, by repeating a number of accident reconstructions from the CASPER project. The CASPER project was unable to derive an injury risk function for chest deflection (or acceleration) of the Q-Series using accident reconstruction. In many cases, the original reconstruction was valid in terms of reproducing the original collision, but the chest deflection measurement was undermined in some way. It is hoped that repetition of the tests may allow some improvements in the chest deflection data available from reconstructions and hence the hope for an injury risk function. The efforts of this group continue now, but so far have not led to the generation of sufficient data to provide an improved injury risk function for chest deflection measurements from the Q-Series of dummies.

The use of chest deflection measurement with an appropriate performance threshold could offer benefits for the assessment of child restraints in the new Regulation (particularly in front impact). The following section considers potential thoracic injury criteria assuming that the loading to the chest is representative of the loading mechanism that would be responsible for the injury or injuries being predicted.

4.2 Proposals for chest deflection thresholds

As occupants get larger, it seems reasonable to assume that the chest deflection threshold for dummies should increase for a given chest injury risk level (to account for the greater chest depth). However, the chest deflection thresholds proposed by Wismans et al. (2008) [1] on behalf of EEVC Working Groups 12 and 18 displayed the opposite trend. They predict that very high chest deflection can be tolerated by younger children for injuries at the AIS \geq 3 level and that their tolerance reduces as they get larger. This might be plausible for the risk of rib fracture, but it does not seem plausible for organ injuries. These thresholds are shown in table 11.

Table 11: Chest deflection thresholds proposed for the Q-Series by EEVC Working Groups 12 and 18 (Q1 to Q6) [1] and EPOCh (Q10) [40]

	Q1	Q1.5	Q3	Q6	Q10
20% risk of AIS \geq 3 injury	40	38	36	33	-
50% risk of AIS \geq 3 injury	59	56	53	49	56

Table 12 shows a series of thresholds derived using an alternative scaling formula proposed by Wismans et al. (2008) [1]. This formula differed from that used to derive the thresholds in table 11 in that it excluded the ratio of bone modulus. Material properties were considered by the inclusion of the calcaneal tendon failure stress ratio.

Table 12: Chest deflection limits in mm based on Wismans et al. (2008) [1] (so-called bag loading scaling)

Base level	Q0*	Q1	Q1.5	Q3	Q6	Q10	HHIII 50th %
50 % risk of AIS \geq 3	11.5	16.6	18.4	22.5	28.2	32.7	50.0

* Note that in the Q0 deflection cannot be measured

These thresholds are a potential means of assessing the capacity of child restraints to restrict compressive loading to the chest. The thresholds included in table 12 were proposed for use in UN Regulation 129 (Visvikis, 2014 [41]).

Wismans et al. (2008) [1] compared the performance of contemporary child restraint systems with the limits in table 11. They found that only a few forward facing non-integral products would fail the 50 % risk of injury level thresholds (in case of tests with the Q6 dummy). For some of these dummies the thresholds are above the measuring range. A substantial proportion (at least one third) would fail the 20 % risk thresholds both with the Q3 and Q6. Based on this information and subsequent experience, it is expected that fewer current or near-to-market child restraint systems would be able to meet chest deflection requirements based on the values in table 12 as these are lower than the ones in table 11.

Although sparse, the accident data available for children, assumed to be correctly restrained in modern child restraint systems in frontal impact collisions, suggest that the risk of sustaining an AIS \geq 3 thoracic injury in R129 conditions should not be very high. Assuming that many products exceed the values in table 12, then there would be a potential conflict between the implications associated with these values in laboratory testing and the real world experience. This is likely to be supported by cardiopulmonary resuscitation (CPR) data: compressions between 35 and 47mm were measured in actual CPR performed on children between 8 and 10 years old (Maltese et al., 2011) 1, which suggests that minimal severe injury risk should occur at such levels which are higher than those in table 12.

Table 13 shows a further series of thresholds derived using another alternative scaling formula, as proposed by Mertz et al. (2003) [43]. This formula used geometric scaling only (based on the ratio of chest depth, so that the threshold increases with dummy size); material properties were not taken into account, which explains why the limits in table 13 also slightly differ from those in table 13. Mertz et al. noted that, at the AIS \geq 3 level, the thresholds were aligned with the risk of rib fractures, whereas at the AIS \geq 4 level, they were aligned with organ injuries (particularly to the heart).

When compared with the CANDAT anthropometry data for chest depth, as used in the design requirements for the Q-Series, then the chest deflection values from table 13 can be seen to correspond to a consistent percentage compression across the Q-Series dummy sizes. The 50 % risk of an AIS \geq 3 chest injury level of deflection corresponds with about 21.7 % compression and the AIS \geq 4 level is about 28.0 % compression.

Table 13: Chest deflection limits in mm based on Mertz et al. (2003) [43] belt AIS \geq 3 and bag AIS \geq 4 scaling and chest depth from CANDAT anthropometry data

Base level	Q0*	Q1	Q1.5	Q3	Q6	Q10	HIII 50th %
50% risk of AIS \geq 3	18.3	23.7	24.6	26.5	29.3	33.5	50.0
AIS \geq 4	23.5	30.5	31.6	34.1	37.7	43.1	64.3
Chest depth (CANDAT)		109	113	122	135	155	{221} [†]
Percentage compression AIS \geq 3 (%)		21.7	21.8	21.7	21.7	21.6	22.6
Percentage compression AIS \geq 4 (%)		28.0	28.0	28.0	27.9	27.8	29.1

* Note that in the Q0 deflection cannot be measured, † Hybrid III value from technical specifications.

From a pragmatic viewpoint, the two sets of thresholds in table 12 and table 13 seem to be more useful than those in table 11 because:

- i. The threshold gets larger as the dummy size increases, which seems to be more plausible for a given risk of chest injury (i.e. all injuries, not just rib fracture);
- ii. All of the dummies are physically capable of measuring deflection up to the threshold.

To provide more achievable limits for use in Regulation 129, a variety of other, pragmatic suggestions have been made around the activities of the UN Informal Group on Enhanced Child Restraint Systems (for example, Lawall and Klimitsch, 2014 [44]). However, these proposals seem to lack a robust scientific basis and seem to be more concerned with feasibility than with objective developments.

The need to provide feasible thresholds for a given application is acknowledged. As such, it still remains for somebody to provide a robust suggestion for chest deflection thresholds to be used in Regulation 129 (or any equivalent application) which fits the necessary compromise between matching real world injury expectations, being achievable in the specified test conditions and reflecting the available measurement range of the Q-Series family of dummies. The pragmatic concept of having a consistent level of chest compression (as in table 13) was endorsed by Carroll [45] who proposed a 35% compression level at a meeting of the Informal Group. Therefore, until such time as further evidence becomes available, it is endorsed by this group to use a consistent compression threshold between dummies that fits the necessary compromise previously mentioned. The AIS \geq 4 injury values from table 13 seem a good candidate for this threshold and is suggested to be used as deflection criteria with the Q-Series family of dummies.

It should be noted that none of the threshold values proposed to date and as presented in tables 12, 13 and 14 account for the dummy stiffness. There is an implicit assumption that because the dummies come from the same family of design, that they will all match biofidelity requirements in a similar and reasonable way. However, no accounting has been made for the detailed dummy-to-dummy variations in this respect. Also, given that the

dummy responses are not completely within the biofidelity corridors, care should be taken transferring human tolerance information to the dummies. Understanding this limitation the threshold values from table 13 are still recommended. However, dummy specific injury risk curves and injury prediction thresholds should be sought in the future. Therefore, activities to resolve issues with the kinematic behaviour of the dummies and to generate new injury risk curves are welcomed (for instance, through the efforts of the chest and abdomen injury criteria task force).

As reported in the earlier report [2], the Q10 has a multi-point chest deflection measurement system. As such there is scope for combining the data from the two sensors to give more information about potential injury risk. For instance, it may be that loading one part of the chest in isolation is more injurious than distributed loading over the whole chest. In this case, there may be benefit to using an advanced criterion which accounts for force distribution somehow. As stated in that report, the appropriateness of such a proposal would need further research and validation than has been possible to date. In the meantime it remains the suggestion that any deflection limit is applied to the peak value from either sensor.

4.3 Other concerns for chest deflection thresholds

In recent test work for the European Commission evaluating the potential application of chest deflection thresholds for use in Phase 2 of Regulation 129, an issue was noted with the Q3 measurements. Regulation 129 tests were carried out using either a booster cushion or a booster seat or booster seat with added mass. In six tests all with a different type of seat, contact was noted between the clavicle of the dummy and its lower neck bracket. The typical location of contact on the clavicle is shown in figure 29. This contact artificially limits the displacement of the clavicle and hence will influence the motion of the rib cage at the level of the IR-Tracc. As such it seems unlikely that the dummy will be able to achieve ~ 40 mm of compression during such dynamic tests with non-integral child restraint systems where the adult belt is used to restrain the child. This means that any threshold set at around 40 mm is likely to be ineffective.



Figure 29: Image showing the typical area of the Q3 clavicle that makes contact with the lower neck load cell bracket during dynamic front impact tests of UN Regulation 129

Furthermore, in testing with the Q3 and Q6, it was observed that the maximum value recorded by the chest compression sensor(s) is due to the chin contacting the chest, rather than the belt loading the thorax. It is likely that this contact between the chin and the chest is not biofidelic, with an expected contact for a child's chin occurring further down the body than the chest, if a contact does occur. Instead, this behaviour could be a product of the rigid thoracic spine in the dummy, constraining upper torso movement to the head and neck rather than including the spine as well. The implications for the dummy measurements are that the peak loading coming directly from a non-integral restraint system can be masked by this contact. If the assumption is that belt loading is most likely to be responsible for children's chest injuries in the real world (and not head to chest contacts), then this creates the issue that the injurious loading is not identifiable in certain test conditions (e.g. the proposed Regulation 129 – Phase II).

4.4 Discussion and conclusions

Chest acceleration was used as an injury metric in R44 to limit the general severity of loading to the child via the restraint system. While the chest deflection would be a better criteria to assess the chest injury risk, the determination of a reliable injury risk function is still an issue. Efforts are continuing to define a dummy-specific chest deflection injury risk function for use with the Q-Series dummies. However, no definitive guidance on specific thresholds is available yet. In lieu of an up-to-date dummy-specific injury risk function, other efforts have scaled limits to make them applicable for the various Q-Series dummies. Section 4.2 describes a variety of proposals. The scientific validity of each may be questioned, and the value of each weighed against the next, but they serve to highlight the numerous options available to choose a pragmatic threshold. Based on the options presented, it is recommended that the AIS ≥ 4 injury severity thresholds from table 13 (28%) are used in conjunction with the chest deflection measurements from the Q-Series family of dummies.

Fundamentally, care should be taken choosing any threshold without first validating that it can be measured reliably and that the kinematics of the dummy behaviour make it a meaningful assessment of safety. This last point will be largely discussed in the next section. In the end, due to belt interaction issues, it could be desirable to keep using chest acceleration for a limited time. However, with forward-facing child restraint systems, deformation of the compliant chest of the Q-Series will influence the acceleration. Therefore, and for the time-being only, to prevent unwanted strategies where high deflection is used to reduce chest acceleration or minimize other parameters, it is strongly recommended by EEVC WG12 that a deflection limit is used in regulation to avoid the potential design strategy of using compression of the thorax to minimise other parameters.

5 INTERACTION WITH SHOULDER AND LAP BELT

5.1 Introduction

In the CREST project (1996-2000) concerns were raised regarding the belt interaction with the dummy both at the pelvis and the shoulder. For the pelvis, the entrapment of the lap belt at the junction between thigh and pelvis was pointed as possibly preventing submarining, and for the shoulder, the slippage of the diagonal belt towards the neck more or less independently of the restraint condition was suspected to prevent a reliable assessment of the thorax loading using chest deflection.

Both issues are however irrelevant for the interaction with integral CRS (e.g. harness or shield), for which no major problem was raised up to now. Interactions with integral CRS could be affected however by the rigid design of the thoracic spine which has been known to affect the overall kinematics (e.g. Sherwood et al., 2003). This, among other possible issues are reviewed in a CASPER public report [20].

The interactions with the shoulder belt and the lap belt for non-integral CRS, as well as possible countermeasures, will be detailed hereafter.

5.2 Interaction with the shoulder belt

As mentioned in Section 0, injuries to head and abdomen are common for children in non-integral child restraint systems. However, chest injuries are also reported for belt-restrained children involved in frontal impact collisions. There are, in general, two main mechanisms for chest injury in children: localised forces applied directly to the chest by the interior of the vehicle (including the restraint system); or sudden chest deceleration and its effect on the internal organs (Cavanaugh, 2002 [31]). It may be inferred that many chest injuries result from a compression mechanism and for belt-restrained children in the back of a car, this is likely to be attributable to the diagonal portion of the seat belt.

With a compliant 'rib cage' and chest deflection instrumentation, it was intended that the Q-Series dummies would be able to assess the severity of loading from a seat belt relevant to the compressive injury mechanism, when the belt was passing over the chest in a frontal impact event. Some issues concerning the selection of an appropriate performance threshold were introduced in Section 4. Also, to relate the dummy measurements to a risk of injury for a child in an equivalent restraint system, there is a necessity that the dummy kinematics and behaviour is broadly consistent with that expected for the child. This is needed to ensure that the dummy measurements align in some way with the mode of deformation expected to cause the injuries in a child.

Experiments have shown that the diagonal portion of a three-point seat belt can slide over the chest of the Q-Series dummies during a dynamic frontal impact event (e.g. Visvikis et al, 2014 [5]). With a standard set-up and position of the shoulder belt upper anchorage position the belt slides from the middle of the clavicle towards the neck of the dummy, in the Q3, Q6 as well as the Q10 (Bohman and Sunnevång, 2012 [46] and Croatto and Masuda, 2013 [47]).

Arbogast et al. (2013) [30] compared the performance of the Q10 with child volunteers in low acceleration sled tests (4.3 g sled acceleration). The subjects were seated directly on the test

bench and restrained with a three-point seat belt, no child restraint system. Although, the position of the upper anchorage was adjusted for each subject to give a similar path for the diagonal portion of the belt over the shoulder and a 55° angle for the diagonal belt with respect to the horizontal. In these tests, Arbogast et al. determined that the belt slipped towards the neck for all subjects. However, the magnitude of the movement with the Q10 was greater than with the volunteers. On average, the relative lateral motion of the shoulder belt was 44 % ± 1 % of the total Q10 single shoulder width, compared with 24 % ± 9 % of the single shoulder width for the human volunteers.

In a comparative study between a Q6 and a human mathematical model, Beillas et al. (2014) [48], observed that such slippage did not occur in the human model. A similar slippage could be observed after making the spine and ribs of the human model rigid and reducing the friction with the belt. Maintenance of the belt in position was attributed to multiple factors including the clavicle forward motion and the spine flexibility. Also, in the same study, table top tests performed on the Q6 ribcage have shown that the IR-TRACC sensor underestimated the applied deflection by about 7% each 10mm of belt shift towards the neck and that the lower thorax was stiffer than the upper thorax. This movement towards the neck occurs relatively early in an impact and takes the belt routing away from the chest deflection sensor(s).

Several solutions, that will be discussed below, have been proposed to address this limitation in behaviour. They include the inclusion of additional instrumentation, modification of test parameters and modification of the dummy.

5.2.1 Additional instrumentation

To make sure that the diagonal part of the belt does not slip to an un-instrumented part of the chest, Visvikis et al. (2014 [5]) proposed the use of a second chest deflection measurement point. After seeing initial prototypes, the chest and abdomen injury criteria taskforce adopted the concept. Modifications to the Q3 and Q6 have now been designed which facilitate the use of a string potentiometer mounted on the thoracic spine (between the scapula mouldings) and attached to the clavicle retainer (figure 30).

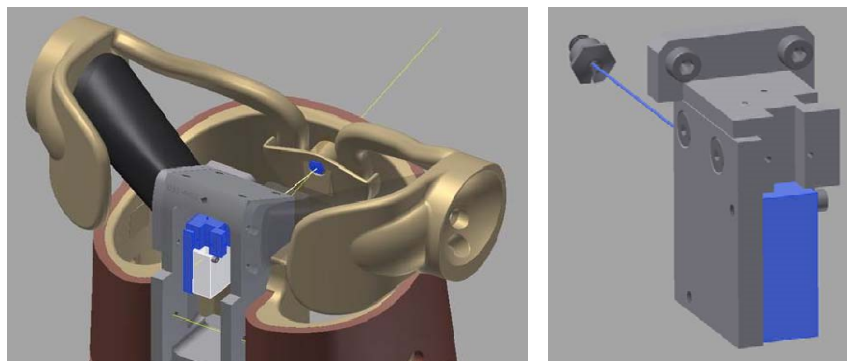


Figure 30: String potentiometer

This modification is useful in understanding the loading to the chest of the dummy. However, it does not resolve the underlying issues preventing a robust chest deflection measurement from being obtained; namely, the belt slippage towards the neck and the hard contacts between the clavicles and the neck bracket of the dummies.

5.2.2 New upper anchorage position

For Q3 and Q6 the slippage of the diagonal belt may be observed in the R129 test environment are there also reports of this slippage in the car environment. For Q10 it is not observed in the R129 bench tests at all however mentioned in many reports about dummy performance in the car environment. The objective of Visvikis (2014) for Q6 was to look for countermeasures to prevent slippage and the objective of Eggers et al. (2015) for Q10 to see how the slippage can be promoted in the R129 environment.

Visvikis et al. (2014) [5] tried moving the upper anchorage position when testing with the Q3 and Q6 dummies on the R129 test bench in a booster seat. The anchorage was moved outboard by 100 mm and downwards by 75 mm. Those authors found that moving the upper anchorage did not have any significant effects on the interaction between the dummy and the belt. Eggers et al., 2015 [41] also tried moving the upper anchorage position in tests with the Q10 dummy. In these tests the dummy was positioned on a booster cushion and the anchorage position was moved either 70 mm outboard, 75 mm upwards or 100 mm rearwards. In the test with the far position there was less slippage of the diagonal portion of the seat belt towards the neck of the dummy and away from the chest deflection measurement positions. Therefore, the deflections measured in that tests were the highest whilst the acceleration was the lowest.

Recent tests at TRL used an even more extreme upper anchorage position. For the “extreme” position, the upper anchorage was moved 60 mm forwards, 100 mm outwards and 50 mm downwards, compared to the existing R129 position. As identified by Eggers et al., using the extreme position with the Q10 led to higher chest deflection measurements, this was confirmed even when seated in a booster cushion. With the Q6, the use of the extreme upper anchorage position prevented the belt from slipping towards the neck. This led to the clavicle making a hard contact against the spine box of the dummy. However, even this extreme position was unable to stop the diagonal belt sliding towards the neck with the Q3 dummy. The exception being one test where the belt started very wide on the shoulder of the dummy. Such a wide position would not be viable for securely retaining a child throughout an impact event.

5.2.3 Modification of the dummy

Possible countermeasures that could affect the interaction with the diagonal belt have been discussed in the past. In a study where the responses of a Q6 and a human model were compared [48], modifying a single parameter in the human simulation (e.g. friction, rib stiffness, spine stiffness) was not sufficient to reproduce the belt slippage observed in the Q6. To reproduce a similar behaviour, stiffening the ribs, spine and reducing friction were all required as illustrated in figure 31. Also, the shoulder flexibility seemed to play a role in keeping the belt in position. While this does not indicate that the exact reverse changes would be required on the dummy, it may help to formulate hypotheses for dummy changes (e.g. thoracic spine and shoulder flexibility) that could be tested on a dummy model first.

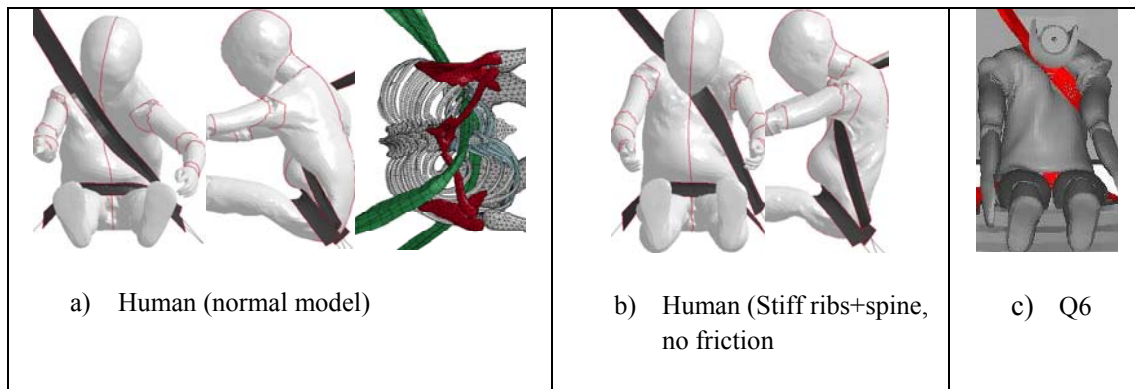


Figure 31: Comparison of a human model and a Q6 model response (Adapted from [48])

5.3 Interaction with the lap belt

Submarining is believed to be one of the mechanisms leading to abdominal loading and sometimes injury in children [49]. The use of booster seats and cushions is known to reduce this risk [50]. As described in Section 6, Abdominal Pressure Twin Sensors (APTS) have been developed for the Q1.5, Q3, Q6 and Q10 dummies. These sensors are able to detect abdominal loads produced by child restraint systems. With the standard Q-Series dummies in a regulatory testing environment and using the adult belt it has not been possible to generate significant load on the sensors even when placing the dummy directly on the bench, without a booster seat or cushion. During such a test, the lap portion of the belt becomes trapped in the gap between the pelvis and thighs (this has been described extensively in the literature as early as the late 1990's for the Q3 prototype or more recently for the Q10 by Hynd et al., 2011 [33]).

Simulations performed by IFSTTAR using a 6 year old human body model in a Regulation 129 environment indicated that a child pelvis is not able to catch the lap belt, producing submarining when a child just sits on the bench. The introduction of a booster in the simulation changed the belt routing and load distribution on the pelvis. In this configuration, the presence of a booster cushion prevents submarining behaviour. With the Q6 dummy, simulation and tests with or without a booster do not produce any abdominal loading when tested under the same conditions (Beillas et al., 2013 [51]).

This absence of submarining and abdominal loading in a regulation environment is therefore believed to be an issue for the assessment of the protection provided by booster seats and booster cushion. This has triggered research efforts to develop dummy accessories to improve the interaction with the belt and to adjust the regulatory environment to make it more challenging.

5.3.1 Dummy accessory prototypes

Various dummy accessories have been proposed by respectively: Humanetics for the Q3 to Q10, Ifsttar for the Q3 and Dorel for the Q6 to address the potential for the lap part of the seat belt to penetrate the gap between the thighs and pelvis.

For the Q10, Humanetics had proposed the use of hip shields. These are sculpted plastic parts that sit on the outside of the suit to prevent the belt entering the gaps in the flesh. These followed preliminary versions (hip patches) that were tested within the CASPER project for Q3 (e.g. [20]) and also for Q10. Illustrations are shown in figure 32a and figure 32c.

Ifsttar produced and tested a soft silicone insert prototype for the Q3 dummy [20] & [21]. The insert was moulded directly on the dummy inside the suit (figure 32b). The insert was also used by other test laboratories. As the insert was only for Q3, Dorel designed a modified hip shield worn below the jacket (figure 32d). Due to the lack of unsatisfactory performance, Dorel recently produced a silicon pelvis insert for the Q6 and Q10 using principles similar to the Q3 Ifsttar insert (figure 32e). Finally, Humanetics worked with Dorel to create a production-ready 'hip-liner' version of the insert for the Q3, Q6 and Q10 (figure 32e).

5.3.2 Pelvis inserts and 'hip liner' evaluations

The suit patches did not prevent entirely the penetration into the gap especially for the Q10 (e.g. [16]). The hip shields were evaluated by Visvikis et al. (2014 [5]). They found that the belt loaded the pelvis higher when the hip shields were used. They also noted small changes in dummy measurements in an equivalent test conducted with or without the hip shields, without however any fundamental change in kinematics. Hip shields may however not be a preferred solution though, as it is difficult to keep them in place during the test (as illustrated for the Q10 in [2] p46 and p348) and are prone to be misused.

In tests with the pelvis insert prototype for the Q3 (already described in the section 3.7.2), the insert prevented large belt penetration in the gap but the effect was limited otherwise: belt slippage into the abdomen did not occur in the nominal test conditions but it occurred when the lap belt anchorage points were raised to their highest position whether the insert was there or not. In the intermediate position, slippage seemed to occur with the inserts but not without it. However, it was not associated with a pelvis rotation as it could be expected in case of submarining. Visvikis et al. also evaluated this pelvis insert prototype. This clearly reduced belt intrusion into the gap between the thighs and pelvis, but did not prevent it entirely.

The Dorel insert prototype for the Q6 were tested by putting the dummy directly on the R129 bench (without CRS) or on a CRS (presented in Section 3.7.2). The insert prevented the penetration into the gap and generated the abdomen loading results. In that case, the kinematic response looked more like submarining than with the Q3. The combination of hip shields and insert led to higher abdominal loading. The use of the Dorel pelvis insert for the Q10 further reinforced the tendency towards submarining with the Q-Series of dummies (Renaudin et al., 2015 [52]). There was no penetration into the gap and a kinematics that looked like submarining in poor restraint conditions. Further reclining the dummy back also contributed to this tendency, especially for smaller dummies. There is still the underlying size contributor, where the heavier Q10 is more likely to submarine than the Q6 which has a greater ability to submarine than the Q3. Also, there is little pelvis rotation in the Q3 compared to the other dummies.

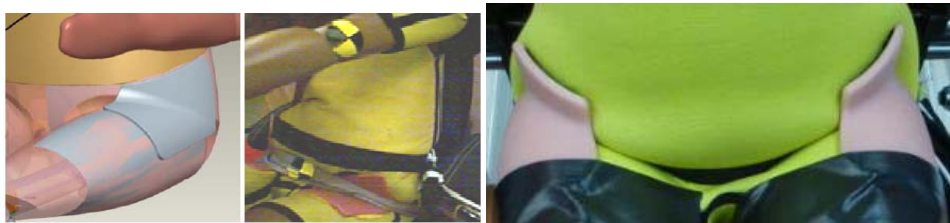
The accessory concepts allow submarining/abdominal loading to be observed with all three dummies under certain conditions. This submarining with the Q-Series leads to a clear difference in the kinematic behaviour regarding interaction between the seat belt and the dummy. Also, the APTS are able to distinguish between tests where the belt has moved up from the pelvis to load the abdomen and those where it has remained on the pelvis. The combination of inserts and APTS with selected test conditions (e.g. dummy in a relaxed seating position, perhaps associated with a reduced bench angle and a higher pulse) enables the detection of poor restraint conditions such as the absence of or a poorly designed booster.



a) Q suit patches from Humanetics for Q3 [20] and Q10 [16]



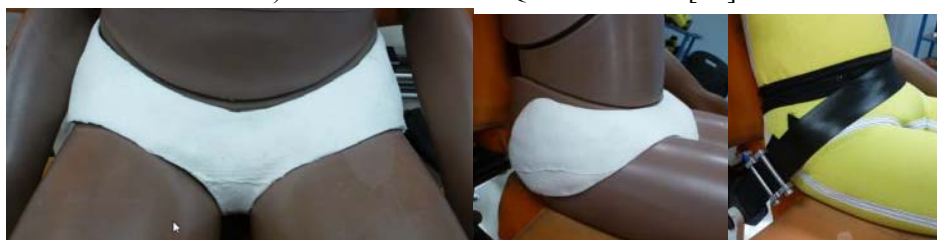
b) Q3 silicone insert from Ifsttar [20]: the insert is worn inside the jacket



c) Hip shields from Humanetics for Q10 (left, Lemmen et al., 2013 [53]) and Q6 [22]



d) Plastic insert for Q6 from Dorel [22].



e) Q6 (left, [22]) and Q10 (right [23]) silicone inserts from Dorel



f) Hip liner from Humanetics. Q3, Q6 and Q10 (left to right). Production candidates.

Figure 32: illustration of the various prototypes and production-ready designs to improve the interactions between the dummy and the lap belt.

As yet, there have been no studies attempting to define whether the new behaviour, which results mostly from preventing an inappropriate belt interaction, is biofidelic. However, we are not aware of any reference data to assess this biofidelity in the paediatric population.

5.4 Discussion and conclusions

No major issues have been raised with the Q-Series dummies regarding their interaction with integral child restraint systems, whether that be harness or shield systems. In either case, it seems that the Q series dummies are able to assess the severity of loading placed on the chest by the restraint system. Equally, with the addition of the APTS, it seems feasible to assess the severity of loading placed on the abdomen.

However, the same is not true for non-integral systems or restraint types where the adult seat belt provides the primary restraining component. In these cases concerns have been raised over the migration of the belt from a correct position on the shoulder towards the neck and the implications this has with regard to the sensitivity of the chest deflection instrumentation. Also, there has been a concern with the ability of the Q-Series dummies to represent the tendency for children to slide under the lap portion of the seat belt (submarining) during a frontal impact event.

Movement of the diagonal portion of the seat belt has been observed with all Q-Series dummies used in these restraint types in particular the Q3, Q6 and Q10. It is most pronounced in the Q3 and least in the Q10. With each of the Q-Series dummies, when the belt starts close to the neck, or slides there during the dynamic event, the measurements of chest deflection are smaller than when a path over the centre of the clavicle is maintained. This may offer an incentive to keep the diagonal belt from the outer edge of the shoulder, but conflicts with concerns about the diagonal belt rising too high and loading the neck of the child.

The seat belt slipping into the neck may also have an effect on the other dummy measures like the head excursion. With the belt moving to the neck, the torso may be restrained more aggressively than if the belt had remained on the shoulder. This may reduce head excursion. Therefore, it is conceivable that Q-Series dummy kinematics underestimate the potential head excursion that a child would undergo in an equivalent event.

The addition of an upper 'clavicle' chest deflection sensor in the Q3 and Q6 helps to understand the loading to the thorax, but is still limited by the belt interaction and the available measurement space in the thoracic cavity of the dummy. While possible countermeasures have been discussed in the literature, these may require significant changes to the dummy design and no solution is likely to be introduced in the short term future.

Modifying the anchorage position used in testing may have the potential to keep the diagonal portion of the seat belt lower on the thorax of the Q-Series dummies, than with existing setups. However, an extreme upper anchorage position may not be a good representation of the anchorage points found in most modern vehicles. It would also not assess the performance of belt guides to keep the diagonal belt from the neck of the child. Ideally, belt guides would keep the diagonal portion of the seat belt over the centre of the clavicle and over the rib cage. They should be discouraged from forcing the belt to either extreme position; too far towards the shoulder or too close to loading the neck. As such it seems difficult to assess the performance of belt guides in dynamic tests.

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Abdominal injuries are a priority for prevention for children when using the seat belt in frontal impact collisions. Though the efficiency of booster seats in preventing these injuries has been indicated, dynamic testing with the Q-Series has traditionally not been able to show that efficiency. A problem was identified with a lack of belt slippage from the pelvis to the abdomen, and hence, an inability to generate abdominal loading. Through a variety of concepts and prototypes it now seems that dummy accessories designed to fill the gap between pelvis and thigh flesh in the dummies have reached the stage of being effective. They allow the belt to slide up onto the abdomen and create a path which generates loading detectable with the APTS.

Improving the dummy interaction using dummy accessories results in dummy kinematic changes leading to submarining like behaviour in cases where no CRS is used with Q6 or Q10 (discussed in 5.3.2). As such, these results are in line with epidemiological studies highlighting the importance of CRS to prevent abdominal loading and are considered to be a step forward for the evaluation of CRS. Issues which still remain to be determined:

- whether the dummy kinematics are now a reasonable representation of a child's behaviour, which will be difficult to assess in the absence of clearly defined biofidelity targets.
- the precise use conditions and requirements that generate 'unacceptable' loading when a booster seat is not used or is misused.
- the behaviour of the production-ready hip liner in comparison with the prototypes.

However, it seems as though the abdominal loading issues could be at least partially addressed with the adoption of hip liner dummy parts and a more realistic seating position, such as that proposed by UMTRI. This could represent a significant improvement over the current situation where no CRS is needed to pass the candidate regulatory test.

6 SUMMARY, CONCLUSIONS AND RECOMMENDATIONS

The head, abdominal and thoracic injuries are relevant priorities for prevention for children of various ages in frontal impact events. Estimating the loads transferred by the restraint system to the thorax and abdomen is important therefore for children restrained in integral or non-integral CRS.

In the European regulation UNECE Regulation 44 CRS's are homologated through testing with P-dummies that were developed in the 1970s by TNO. The dummies were mainly designed to act as a loading device with appropriate dimensions and mass distribution, but with limited measurement capability.

In 1993 the development of the Q-series of child dummies started as successor to the P-series. In 2007, the UN Informal Group on child restraint systems was established (GRSP Passive Safety group of WP29) to develop a new UN Regulation on the approval of "Enhanced Child Restraint Systems". The first phase of the new Regulation was completed in 2011 and sets out performance requirements and test methods for integral ISOFIX child restraints and includes the Q-Series dummies.

The objective of this report is to review the status of the Q dummies concerning the capabilities for thorax and abdominal injury detection and develop recommendations for performance thresholds for thorax and abdominal injuries. An important part of the report deals with a documentation of the development and evaluation of the Abdomen Pressure Twin Sensors (APTS).

6.1 APTS design and performance

Multiple measurements principles have been considered in the past to attempt to quantify directly abdominal loads, including the Abdomen Pressure Twin sensors (APTS) for the Q dummies developed by Ifsttar. The APTS are two soft cylindrical polyurethane bladders closed by aluminium caps and filled with liquid to be located within the abdominal section of the dummy. The pressure in the fluid is measured by a pressure cell in the cap. Many tests with different versions of the APTS design have been carried out as summarized in Chapter 3 and in [2] for the Q10.

The production version of the APTS design and the corresponding abdomen section of the dummy are now close to finalization. With the potential to generate abdominal loading in sled tests and the ability of the APTS to detect that loading, the Q-Series dummies are capable to assess abdominal injury risk in non-integral as well as integral child restraint systems.

Changes to the dummy brought about through the addition of the APTS have been characterised and their performance and sensitivity to applied loading types is satisfying. The Q-Series dummies are sensitive to initial posture and quality of restraint system. Also, the APTS instrumentation is sensitive to abdominal loading from the seat belt in subsystem and sled test conditions. The effect of APTS on the biofidelity response, analysed in biofidelity testing and simplified testing with a rigid bar were small.

6.2 Abdominal and thoracic injury risk curves

Concerning the selection of a possible abdominal tolerance, risk curves were obtained for Q3 and Q6 based on accident reconstruction in belted configuration. Pressure levels slightly over

1 bar seem to correspond to a limit between behaviours which are visually acceptable or unacceptable. Based on the risk estimate, such a limit is likely to relate to a risk of AIS 3+ injury of between 20 and 50%. For the Q10, the same separation may occur at slightly higher pressure levels. For the Q1.5, there is currently no such equivalence as the dummy has not been tested in belted configurations.

For chest deflection injury assessment information based on scaling is available including the values proposed in table 13 of Chapter 4. These are proposed for use with the Q-Series family of dummies until such time as further information becomes available to update the injury risk association with chest deflection measurements from the Q-Series dummy family. Also, before these values are adopted, care should be taken to ensure that they can be used to give meaningful results in a specific application.

6.3 Evaluation of integral CRS versus non-integral systems

The biomechanical requirements placed on a test device for assessing integral child restraint systems are different to those for non-integral seats. In the latter the dummy needs to represent the complex interaction of a child with the adult seat belt. The use of the Q-Series dummies in integral child restraint systems was brought about with the introduction of Phase I of Regulation 129. It followed recommendations on the suitability of the dummies for such an application given by Wismans et al. (2008 [1]). Very few issues have been raised regarding this application of the dummies and both chest deflection and abdominal pressure seem relevant for this configuration.

Phase II of the Regulation is concerned with non-integral restraint systems. Since initial evaluations of the Q-Series dummies, concerns were expressed over the interactions with the seat belt in non-integral seats. Primarily the issues relate to:

1. Movement of the diagonal part of the seat belt towards the neck (and its implications for the assessment of head displacement and chest injury protection); Due to this the dummy is not well able to assess the effect of different diagonal belt guide designs. Additionally, it seems important that non-integral child restraints which position the three-point belt correctly across the torso of the dummy, are not penalised by failing harsh chest compression criteria.
2. Penetration of the lap part of the seat belt into the gap between the legs and the pelvis (and its implications for the assessment of abdomen injury protection). The pelvis insert or hip liner accessories can help preventing this belt entrapment, resulting in an improved kinematic response and submarining in poor restraint conditions.

6.4 Recommendations

- Finalize the APTS design and the abdomen keeping in mind the biofidelity requirements and, if needed, update the certification procedure to better control the variability.
- Implement the APTS in the Q dummies for regulatory testing both for integral and non-integral CRS.
- Concerning abdominal injury thresholds implement a threshold around 1 bar pressure to distinguish between acceptable and non-acceptable abdominal penetration. For the Q1.5

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the level should be further evaluated as the biomechanical evidence and number of test results is limited in comparison to Q3 and Q6 and to some extent to the Q10.

- The Q6 hip liner (insert) provided by Humanetics is effective to avoid unrealistic belt entrapment that prevents submarining behaviour with this dummy and is recommended to use in regulatory tests. Equivalent production-ready versions of the Q3 and Q10 inserts are available and as soon as these inserts are positively evaluated, it is recommended to use these parts in regulatory tests as well.
- An adequate installation procedure should be used to evaluate protection from submarining in frontal impact tests with non-integral child restraint systems. The UMTRI installation procedure is recommended as it was shown to be effective for Q3, Q6 and Q10 dummies.
- For chest deflection, initial recommendations for injury thresholds are available taking the AIS ≥ 4 threshold values from table 13. Considering the assumptions made to select these values, they should be evaluated for the specific applications in which the dummies are to be used. The case of integral CRS – for which no significant issues with the dummy kinematic behaviour was put forward – needs to be separated from the use of belted configurations for which recommendations are provided hereafter.
- Due to issues with diagonal belt interactions, the value and feasibility of adopting such chest deflection values as the only assessment of the chest loading in Regulation 129 Phase 2 remains to be proven. It is recommended to continue research to improve the diagonal belt interactions for the Q3 and Q6. In particular, belt slippage towards the neck should be reduced.
- In the meantime, until such issues are fixed, assuming that chest acceleration is kept as an injury criteria in that configuration, use chest deflection in addition to prevent strategies where the deformable ribcage of the dummy would be excessively compressed to lower the acceleration or other injury metric.

REFERENCES

- [1] Wismans, J at al. (2008) Advancement of Child Dummies and Injury Criteria. EEVC Report., Document Number 514 EEVC. Feb 12, 2008. EEVC.
- [2] Wismans J, et al. (2015) Q10 Dummy report: Advanced Child Dummies and Injury Criteria for Frontal Impact. EEVC Working Group 12 Report Document Number 642. Jan 19, 2015.
- [3] Beillas, P., Alonzo, F., Chevalier, M-C., Lesire, P., Leopold, F., Trosseille, X., Johannsen, H. (2012) Abdominal Twin Pressure Sensors for the Assessment of Abdominal Injuries in Q Dummies: In-Dummy Evaluation and Performance in Accident Reconstructions. Stapp Car Crash Journal 56: 387-410.
- [4] Kirk, A. (2012). Summary of CASPER accident database. Paper presented at Cover Child Safety Final Workshop: CASPER and EPOCH, 13-15 June 2012, Berlin. Retrieved October 17, 2012 from: <http://www.biomechanics-coordination.eu/site/en/documenten.php>.
- [5] Visvikis C., Pitcher M., Carroll J., Cuerden R., Barrow A. (2014) New UN regulation on child restraint systems - assessment of amendments to the new regulation, front and side impact procedures and Q-Series dummy family injury criteria. TRL Final report. CLIENT PROJECT REPORT CPR1801. 221 pages.
- [6] Arbogast, K. B., Kent, R. W., Menon, R. A., Ghata, Y., Durbin, D. R., & Rouhana, S. W. (2007). Mechanisms of abdominal organ injury in seat belt-restrained children. *The Journal of trauma*, 62(6), 1473-1480.
- [7] Langwieder K., Hummel Th., and Finkbeiner F. (1999) Injury risks of children in cars depending on the type of restraint Special Session on Child Occupant Protection, Sitges, September.
- [8] Johannsen H., Beillas P., and Lesire P. (2012) Safety Performance of Shield Systems in Comparison to 5-Point-Belt Systems. 10th International Conference Protection of Children in Cars Munich, December 6-7.
- [9] Edgerton, E.A., Orzechowski, K.M. , and Eichelberger, M.R. (2004) Not all child safety seats are created equal: the potential dangers of shield booster seats. *Pediatrics*. Vol 113; N 3; March.
- [10] Kent. R., Stacey. S., Kindig. M., Woods. W., Evans. J., Rouhana. S. W., et al. (2008). Biomechanical response of the pediatric abdomen. Part 2: injuries and their correlation with engineering parameters. *Stapp Car Crash Journal*, 52, 135-166.
- [11] Kremer, M.A., Gustafson, HM., Bolte, J.H., Stammen J., Donnelly B. (2011) Pressure-based abdominal injury criteria using isolated liver and full-body post-mortem human subject impact test. *Stapp Car Crash Journal* 55 p317-350.
- [12] Ono Y., Komiyama Y., Takatori O. (2005) Method of Evaluating Abdominal Injury in Japan's Child-Restraint-System Assessment Program. 19th ESV Conference 2005, Paper 05-292-W.
- [13] Rouhana, S.W., Elhagediab, A.M., Walbridge, A. Hardy, W.N. et al. (2001) Development of a reusable, rate-sensitive abdomen for the Hybrid III family of dummies. *Stapp Car Crash Journal* 45:1-27.
- [14] Elhagediab, A.M., Hardy, W.N., Rouhana, S.W., Kent, R.W., Arbogast, K.B., Higuchi, K. (2007): Development of an Instrumented Rate-Sensitive Abdomen for the Six Year Old Hybrid III Dummy. JSAE Annual Congress, Paper no. 20075245.

The use of thoracic deflection criteria balanced with abdomen pressure criteria for the Q-Series in frontal impacts

- [15] Rouhana. S.W., Viano D.C., Jedrzejczak E.A. and J.D. McCleary (1989) Assessing Submarining and Abdominal Injury Risk in the Hybrid III Family of Dummies. In Proceedings of the 33rd Stapp Car Crash Conference. pp 257–279. SAE Paper no. 892440. Warrendale. Pa. Society of Automotive Engineers.
- [16] Beillas, P., Alonzo, F., Chevalier, M.C., Johannsen, H., Renaudin, F., and Lesire, P. (2012). Abdominal Pressure Twin Sensors for the Q dummies: from Q3 to Q10. Proceedings of the ICrash Conference, Turin, July 18-20, 2012.
- [17] Beillas, P., (2013) Abdominal Twin Pressure Sensors for the Assessment of Abdominal Injuries in Q Dummies: Status update. WP29 GRSP IG 36th Meeting – Jan 2013. Available online: <https://www2.unece.org/wiki/display/trans/CRS+36th+meeting>.
- [18] Beillas P. and Soni, A. (2014) Assessment of abdominal loading by shield CRS in frontal impact: comparison between direct loading in isolated tests and sleds. 12th International Conference Protection of Children in Cars. Dec 4-5, Munich. Germany.
- [19] Tylko S and Bussi eres A. (2014) Paired Comparison of the Hybrid III 10 year old and Q10 Seated in the Rear Seats of Vehicles Undergoing Frontal Crashes. 12th International Conference Protection of Children in Cars. Dec 4-5, Munich. Germany.
- [20] Beillas P., Alonzo F., Chevalier M.C., (2010) Report associated with the Deliverable D.1.2: Auxiliary equipment for Q3 and Q6 to improve belt interaction response. Sustainable Surface Transport, Collaborative Project on Child Advanced Safety Project for European roads, European funded project SCP7-GA-2009–218564 CASPER. Public report available online at <http://casper-project.eu/>.
- [21] Beillas P. (2012) Activities related to Q3 and Q6 submarining / belt interactions. Task 1.2 of the Casper Project. Presented at the COVER Workshop , March 13-15 2012, Berlin, Germany. Publicly available. Also shown at the 31st GRSP IG meeting April 11, 2012.
- [22] Renaudin F. and Bruneti ere Y. (2015a) Q6 Abdominal penetration. Tests results with different insert proposals. GRSP IG on Child CRS 50th Meeting. Clepa Presentation. Available online at:
<https://www2.unece.org/wiki/download/attachments/25265983/CRS-50-10e%20CLEPA%20Presentation%20Q6%20abdominal%20penetration.pdf?api=v2>
Accessed Aug 20, 2015.
- [23] Renaudin F. and Bruneti ere Y. (2015b) Q10 Abdominal penetration. Tests results with insert proposal. GRSP IG on Child CRS 52th Meeting. Clepa Presentation. Available online at <https://www2.unece.org/wiki/download/attachments/25887433/CRS-52-07e%20Dorel%20Presentation%20Q10%20abdominal%20penetration.pdf?api=v2>
Accessed Aug 20, 2015.
- [24] Johannsen H., Beillas P., Lesire P., (2013) Analysis of the Performance of Different Architectures of Forward Facing CRS with Integral Restraint System: Comparison of forward-facing seat with an integral harness and with an impact shield. Proceedings of the 23rd International Technical Conference on the Enhanced Safety of Vehicles, Seoul, South Korea, May 27-30, paper N 13-0226. Published by U.S. Department of Transportation NHTSA.
- [25] Visvikis C., Pitcher M. and Carroll J. (2014) Evaluation of shield and harness systems in frontal impact sled experiments. 12th International Conference Protection of Children in Cars. Dec 4-5, Munich. Germany.
- [26] Informal document GRSP-55-39 (2014) A study on shield systems. Presented by the expert of France. 55th Session of the UN ECE Working Party on Passive Safety GRSP 19-23 May 2014. Agenda Items 15 & 18. Available online on the UNECE website.

- [27] Johannsen H., Trosseille X., Lesire P., Beillas P., (2012) Estimating Q-Dummy Injury Criteria Using the CASPER Project Results and Scaling Adult Reference Values, IRCOBI Conference 2012, Dublin, Ireland, Sept 11-13.
- [28] Tanaka Y., Yonezawa H., Hosokawa N., Matsui Y., Mizuno K., Yamaguchi M., Yoshida R. (2009) Responses of Hybrid III 3YO and Q3 Dummies in Various CRSs Tested Using ECE R44 Impact Conditions. ESV Conference, Paper 09-0242.
- [29] Palisson A., Cassan F., Trosseille X, Lesire P., Alonzo F. (2007) Estimating Q3 Dummy Injury Criteria for Frontal Impacts using the CHILD Project Results and Scaling Reference Values. Ircobi Conference. Maastricht. The Netherlands. September. Pp 263-276.
- [30] Yamada H. (1970) Strength of biological material. Ed. G. Evans. The Williams & Wilkins Company, Baltimore. 298pp.
- [31] Cavanaugh, J. (2002). Biomechanics of thoracic trauma. In: Nahum, A. M. and Melvin, J. W., eds. Accidental Injury Biomechanics and Prevention. New York, NY: Springer.
- [32] Stapp, J. P. (1970). Voluntary human tolerance levels. In: Gurdjian, E. S., Lange, W. A., Patrick, L. M., and Thomas, L. M., eds. Impact Injury and Crash Protection. Springfield, IL: Charles C Thomas.
- [33] Mertz, H. J. and Gadd, C. W. (1971). Thoracic tolerance to whole-body deceleration (Paper No. 710852). In: Proceedings of the 15th Stapp Car Crash Conference. Warrendale, PA: SAE International.
- [34] Kroell, C. K., Schneider, D.C., and Nahum, A. M. (1971). Impact tolerance and response to the human thorax (Paper No. 710851). In: Proceedings of the 15th Stapp Car Crash Conference. Warrendale, PA: SAE International.
- [35] Kroell, C. K., Schneider, D.C., and Nahum, A. M. (1974). Impact tolerance and response to the human thorax II (Paper No. 741187). In: Proceedings of the 18th Stapp Car Crash Conference. Warrendale, PA: SAE International.
- [36] Ziegler, D. W., and Nikhileshwer, N. (1994). The morbidity and mortality of rib fractures. *Journal of Trauma-Injury Infection and Control* 37(6). 975-979.
- [37] Holcomb, J. B., McMullin, N. R., Kozar, R. A., Lygas, M. H., and Moore, F. A. (2003). Morbidity from rib fractures increases after age 45. *Journal of the American College of Surgeons* 196(4). 549-555.
- [38] Arbogast, K. B., Locey, C. M., Bohman, K. and Seacrist, T. (2013). Relative kinematics of the shoulder belt and the torso: comparison of the Q10 ATD and pediatric human volunteers (Paper No. IRC-13-27). In: IRCOBI Conference Proceedings, 11-13 September 2013, Gothenburg, Sweden. Zurich, Switzerland: International Research Council on Biomechanics of Injury (IRCOBI).
- [39] Johannsen, H., Beillas, P., Bendjellal, F., Lesire, P., Müller, G, Renaudin, F., Visvikis, C., and Waagmeester, K. (2013). Chest and abdomen injury criteria task force. In: Proceedings of the 11th International Conference Protection of Children in Cars, 5-6 December 2013, Munich, Germany. Munich, Germany: TÜV SÜD.
- [40] Hynd M., McGrath M., Waagmeester K., Salters E., Longton L. & Cirovic S. (2011). EPOCH Project dissemination. In: Proceedings of the 9th International Conference Protection of Children in Cars, December 2011, Munich, Germany. Munich, Germany: TÜV SÜD.
- [41] Visvikis, D. (2014) Validation of proposed chest deflection thresholds. Presentation to the 47th meeting of the UN Informal Working Group on Enhanced Child Restraint Systems. Document CRS-47-10e; available from the UNECE website: <https://www2.unece.org/wiki/display/trans/CRS+47th+meeting>.

- [42] Maltese M.R., Arbogast K.B., Wang Z., Craig M. (2011) Scaling Methods Applied to Thoracic Force Displacement Characteristics Derived from Cardiopulmonary Resuscitation. Paper Number 11-0313. ESV Conference.
- [43] Mertz, H. J., Irwin, A. L. and Prasad, P. (2003). Biomechanical and scaling basis for frontal and side impact injury assessment reference values (Paper No. 2003-22-0009). In: Stapp Car Crash Journal Volume 47: Papers Presented at the 47th Stapp Car Crash Conference. Warrendale, PA: SAE International.
- [44] Lawall, C. and Klimitsch, C. (2014) Chest deflection. Presentation to the 47th meeting of the UN Informal Working Group on Enhanced Child Restraint Systems. Document CRS-47-08e; available from the UNECE website: <https://www2.unece.org/wiki/display/trans/CRS+47th+meeting>.
- [45] Carroll, J. (2014) Further development of chest deflection proposals. Presentation to the 48th meeting of the UN Informal Working Group on Enhanced Child Restraint Systems. Document CRS-47-08e; available from the UNECE website: <https://www2.unece.org/wiki/display/trans/CRS+48th+meeting>.
- [46] Bohman K. and Sunnevång C. (2012). Q10 child dummy performance in side and frontal sled tests. In: Proceedings of the 10th International Conference Protection of Children in Cars, 6-7 December 2012, Munich, Germany. Munich, Germany: TÜV SÜD.
- [47] Croatto S. and Masuda M. (2013). Q10 and HIII-10 YO in frontal impact: sensitivity to restraint systems. Proceedings of the 23rd International technical conference on the Enhanced Safety of Vehicles (ESV); Seoul, Republic of Korea.
- [48] Beillas P., Soni A., Chevalier M-C., Johannsen H., Müller G., Holtz J., Q6 (2014) Dummy Thoracic Response and Diagonal Belt Interactions: Observations based on Dummy Testing and Human and Dummy Simulations. IRCOBI conference. Berlin, Germany.
- [49] Arbogast, K. B., Kent, R. W., Menon, R. A., Ghata, Y., Durbin, D. R., & Rouhana, S. W. (2007). Mechanisms of abdominal organ injury in seat belt-restrained children. *The Journal of Trauma*, 62(6), 1473–1480.
- [50] Durbin, D. R., Elliott, M. R., & Winston, F. K. (2003). Belt-positioning booster seats and reduction in risk of injury among children in vehicle crashes. *JAMA*, 289(21), 2835–2840.
- [51] Beillas P., Soni A., Renaudin F. and Johannsen J. (2013). Comparison of the submarining behaviours of a 6 years old human model and a Q6 dummy in sled testing. In: Proceedings of the 11th International Conference Protection of Children in Cars, December 2013, Munich, Germany. Munich, Germany: TÜV SÜD.
- [52] Renaudin F., Brunetiere Y., Pitcher M., Carroll J., Burleigh M. and Waagmeester K. (2015). Improvement in Q-Series dummy submarining behaviour in non-integral CRS. In: Proceedings of the 13th International Conference Protection of Children in Cars, 3 December 2015, Munich, Germany. Munich, Germany: TÜV SÜD.
- [53] Lemmen P., Waagmeester K., Burleigh M., Lakshminarayana A., Korschdon K., Visvikis C., Carroll J., Hynd D., Pitcher M. (2011) Development of the Q10 10 Year-Old Child Crash Test Dummy. ESV paper 13-0437, 2013.
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ANNEX A: INJURIES TO CHILDREN IN EUROPE

There is lack of representative European data providing sufficient detail about the restraint and accident configuration as well as injury patterns as a function of age. A query of the Rhône Registry Injury Database in France covering the period 1996 – 2012 provided 77,608 car occupants sustaining all levels of injuries. Of these, 78 children aged 0 to 12 years received an injury at the AIS \geq 3 level. The distributions of injuries sustained by children of different age classes are shown in figure A1. The head was the most commonly injured anatomical region but its percentage decreased with age. Abdominal injuries were most common in the 4-8 age range (24%) for which they were more common than thoracic injuries. In other age classes, the thorax was the second most commonly injured anatomical region. There were no abdominal injuries prior to 1.5 YO.

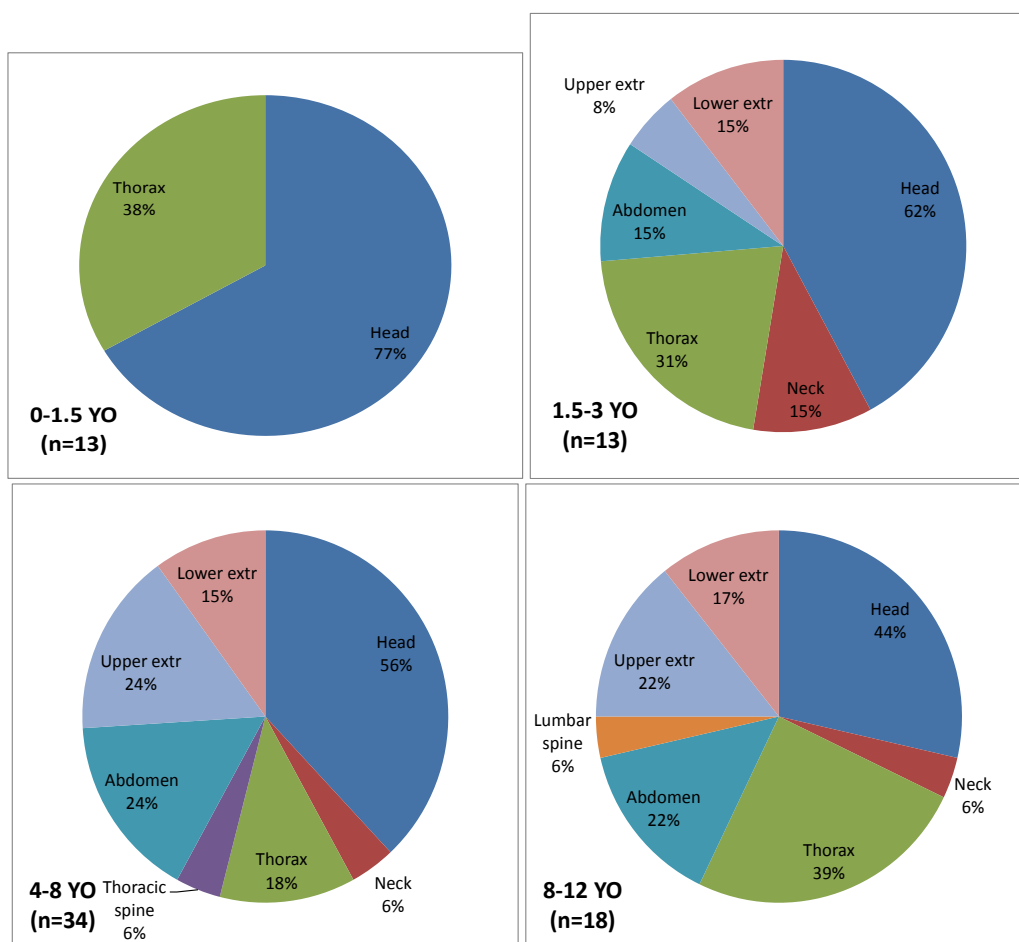


Figure A1: Distribution of AIS \geq 3 injuries among children (front impact, all restraint conditions) Data derived from the Rhône Registry, described by Laumon et al., 1997.

An analysis of representative, in-depth collision databases from Great Britain and Germany was undertaken by Visvikis et al., 2014 to investigate which body regions need to be protected in front and side impacts and hence where the regulatory performance evaluation should be targeted.

Visvikis et al. also found that there were very few injuries to children at AIS \geq 2 in the representative samples from Great Britain and Germany. This is illustrated in figure A2 which

The use of thoracic deflection criteria balanced with abdomen pressure criteria for the Q-Series in frontal impacts

shows the distribution of AIS \geq 2 injuries by body region for each restraint type in front impact collisions. It was impossible to quantify which body regions need to be protected in a statistically robust way. However, there were clear trends and for targeting the regulatory performance evaluation, it was found for all restraint types in frontal impacts that there is good evidence that the focus should be on the protection of the head, chest, arms and abdomen.

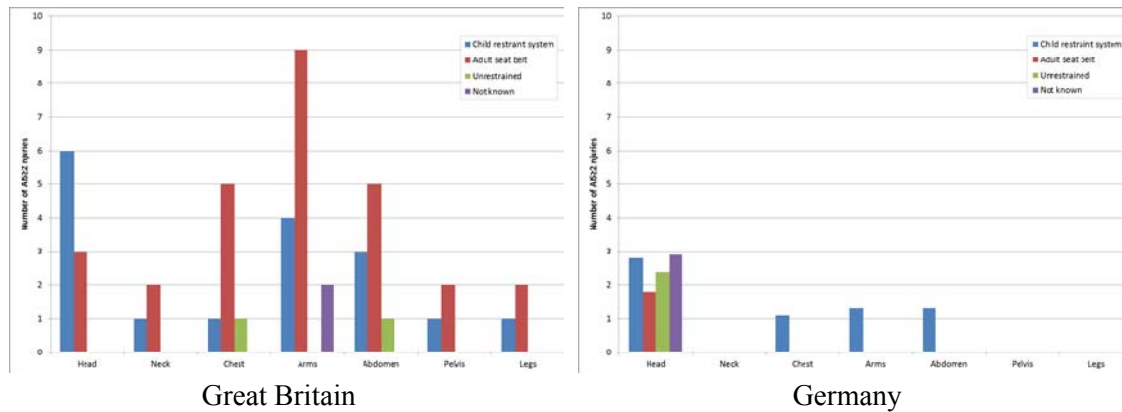


Figure A2: Body region injured by restraint type (AIS \geq 2) – Front impact






REFERENCES

Laumon, B., Martin, J.L., Collet, P., Chiron, M., Verney, M.P., Ndiaye, A. and Vergnes, I. A French road accident trauma registry: First results in 41st AAAM conference. 1997. Orlando, Florida: AAAM.

Visvikis C., Pitcher M., Carroll J., Cuerden R., Barrow A. (2014) New UN regulation on child restraint systems - assessment of amendments to the new regulation, front and side impact procedures and Q-Series dummy family injury criteria. TRL Final report. CLIENT PROJECT REPORT CPR1801. 221 pages.

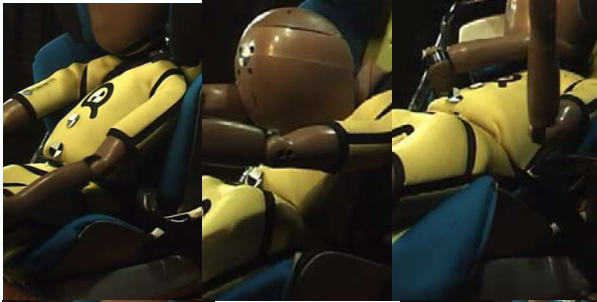


ANNEX B: ILLUSTRATION OF SLEDS Q3 AND Q6 WITH 40MM APTS AND MOULDED ABDOMENS (TRL STUDY)

Table B-1: illustration of the kinematics and belt location for the Q6 tests performed at TRL

Configuration	Photos	Sub mari ning	Belt location	Pressure (bar)	
				L	R
R129 No CRS (I34)		N	Pelvis skin / lower abdome n	0.83	0.89
R129 Poor CRS (I39)		Y	Abdom en	0.93	0.93
UMTRI installation No CRS (I38)		?	Abdom en / pelvis skin	1.09	1.22
UMTRI installation Poor CRS (I40)		Y	Abdom en	1.21	1.18
R129+ Pulse No CRS (I47)		N	Lower abdome n / pelvis skin	1.12	1.17






The use of thoracic deflection criteria balanced with abdomen pressure criteria for the Q-Series in frontal impacts

Table B-1 (continued): illustration of the kinematics and belt location for the Q6 tests performed at TRL

Configuration	Photos	Sub mari ning	Belt location	Pressure (bar)	
				L	R
R129+ Pulse Poor CRS (I41)		Y	Abdom en	1.16	1.13
UMTRI and R129+ pulse Poor CRS (I45)		Y	Abdom en	1.81	1.39
UMTRI and R129+ pulse Good CRS (I46)		N	Pelvis	0.32	0.32

The use of thoracic deflection criteria balanced with abdomen pressure criteria for the Q-Series in frontal impacts

Table B-2: illustration of the kinematics and belt location for the Q3 tests performed at TRL

Test	Photos	Subma rining	Belt location	Pressure (bar)	
				L	R
R129 Poor CRS (I58)		N	Pelvis skin	0.39	0.47
UMTRI installatio n Poor CRS (I59)		N	Pelvis skin / lower abdomen	0.51	0.56
R129+ pulse Poor CRS (I60)		N	Pelvis skin / lower abdomen	0.56	0.58
UMTRI and R129+ pulse Poor CRS (I61)		Y	Abdomen	1.06	0.95
UMTRI and R129+ pulse Good CRS (I62)		N	Pelvis	0.22	0.41

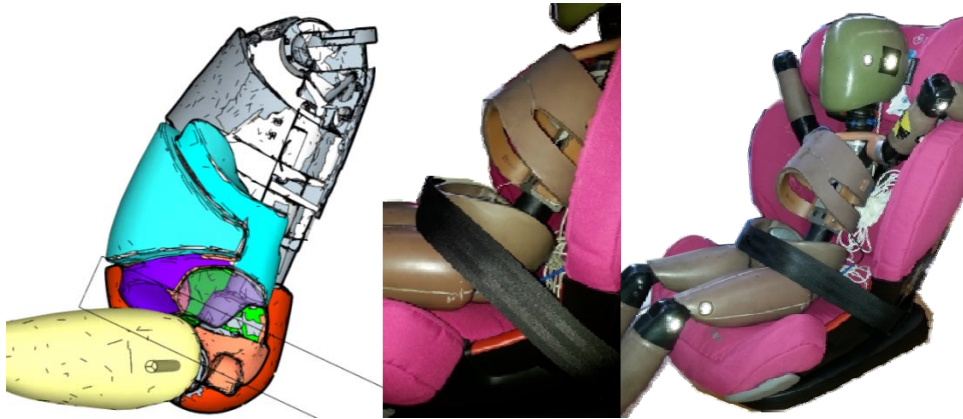


Figure B-1: Illustration of the geometry of the Q3 in the region of the interaction with the lap belt. Left: schematic illustration showing the pelvis skin (in orange) with respect to the iliac wing (in green) and thigh (in yellow). The abdomen (in cyan) has a small region (in purple below the ASIS and inside the pelvis skin (in purple)). The lines represent a standard belt dimensions just for proportion. The abdominal sensors are mostly sensitive in the blue region per their design (see section on the sensitivity to the belt location). Center and right: illustration of possible interactions in that region. The jacket and abdomen were removed for clarity.