DEVELOPMENT OF THE Q10 10 YEAR-OLD CHILD CRASH TEST DUMMY

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Paper Number 13-0438

ABSTRACT

In Europe, the law requires children up to 12 years or up to 150 (or 135) centimetres to be restrained in a child restraint system (CRS) when travelling in cars. The EC FP7 project EPOCh developed test procedures and tools for impact tests for CRS designed to protect older and larger children in vehicle collisions. The EPOCh project involved TRL, Humanetics, IDIADA, DOREL and University of Surrey. One of the main EPOCH deliverables was the development of a prototype 10-year-old Q dummy.
This paper reports on the development and assessment of the Q10 dummy within the EPOCh project, as well as the subsequent refinement and evaluation of the dummy based on third party testing.

INTRODUCTION

According to data from the European Road Safety Observatory [1], the latest data available for child fatal injuries is the year 2011. For that year 815 children in the age below 15 years old were killed on European roads. Since 1995 the figure has dropped around 56%. Whilst this is a good achievement, there is still a need to reduce the level of child fatalities and all types of child injury in the EU in the various transport modes. Child fatality numbers for some EU Member States in 2011 are provided in Figure 1.
Motivated by these numbers the EU FP7 projects CASPER [2] and EPOCh [3] addressed the safety of children transported in passenger cars. While CASPER worked on improvements of existing Q-series dummies, their application in test procedures and development of numerical models, including Human body models, the EPOCh project realised a Q family member representing adolescents and investigated its application in UN Regulation No. 44 and NPACS type test procedures.

The paper will outline the development of the Q10, including: a summary of anthropometry and

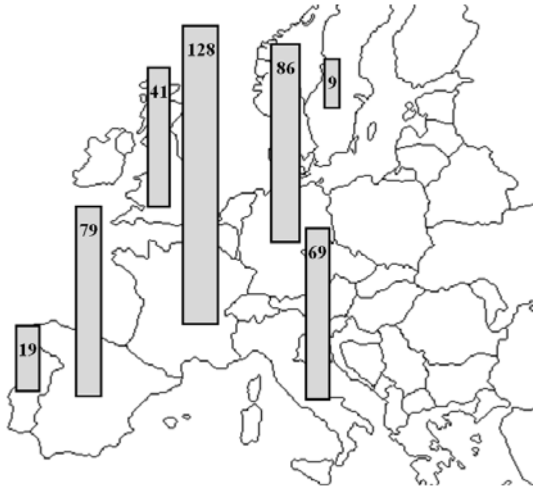

Figure 1. Numbers of child fatal injuries for the year 2011 in some Member States (source: www.erso.eu).

biofidelity specifications, prototype design realisation, prototype performance and proposals for injury thresholds obtained by scaling values from adult and smaller child dummies. A brief list of findings from testing by OEMs, suppliers and test houses world-wide will be provided followed by design updates implemented in a production version of the Q10 dummy. First results on the performance of the production version dummy are included as well. Finally, recent work to evaluate abdomen loading sensors related to the detection of belt penetration in the pelvis-thigh region will be presented. The sensors were developed in CASPER and transferred to the Q10 dummy.

SIZE SELECTION

Under the current regulation (2003-20-EC) countries have an option to select the maximum size of a child that has to use a CRS when traveling in a car. Two different statures are mentioned in the regulation: 135 or 150 cm. This corresponds roughly with 50th percentile children of 8.9 and 11.6 Years Old (further indicated as YO) respectively. The current regulation specifies a P10 dummy (Stature 1.38 m, mass 32 kg, approximately 50th percentile 9.4 YO). The EPOCh team presented their recommendation on the size selection in a stakeholder's forum meeting in June 2009. The conclusion from extensive discussions was that the age of 10.5 YO was felt to be most appropriate to represent the group of oldest CRS users [4].

REQUIREMENTS AND PROTOTYPE DESIGN

Anthropometry

The dummy was sized according to the 50th percentile anthropometry of the selected age: stature 1443 mm, seating height 748 mm, shoulder height 473 mm and total body mass 35.5 kg. In line with the rest of the Q-dummy family the Q10 anthropometry was based on CANDAT (Child ANthropometry DATabase) [5]. Figure 2 provides data for mass versus seating height. The yellow boxes give envelops for 5th to 95th percentile extremes according to CANDAT around the Q-dummy ages 1, 1.5, 3, 6 and 10.5 and 11.6 YO. Note that the latter relates to the age with average stature of 150 cm in CANDAT. Dimensions and masses as abstracted from the database are included in Table 1 and Table 2. References to the measures included in Table 1 are depicted in Figure 3. Additional information related to the skeletal geometry of the pelvis was obtained from [6]. This reference specifies coordinates for important landmarks in the pelvis, based on CT scans from 81 children of 5 to 11 years old. A full description of the dimensional requirements is included in the Q10 design brief [4, 7].

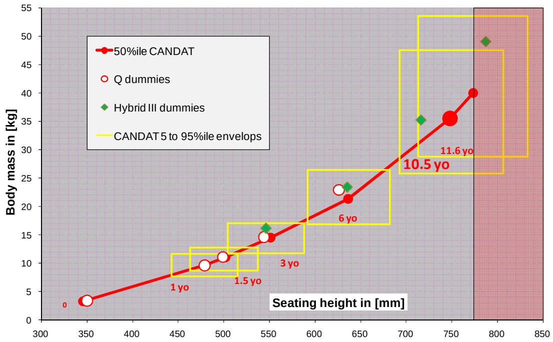

Figure 2. Mass versus seated height for the Q and Hybrid III series with the 5th to 95th percentile envelopes for 1, 1.5, 3, 6, 10.5 and 11.6 YO.

Biofidelity

In defining the Q10 dummy, priority was put on the frontal impact performance because the current UN Regulation No. 44 addresses frontal impacts only.

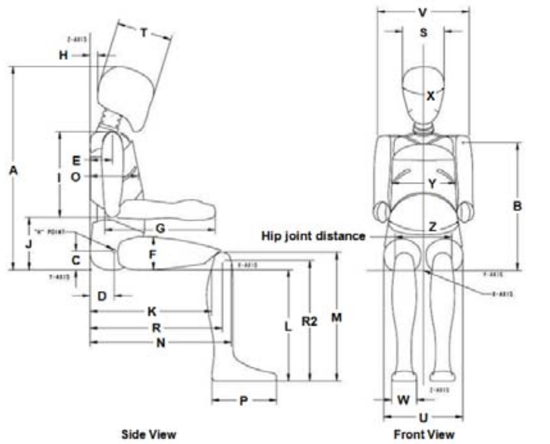

Figure 3. Q10 Overall dimensions.

Table 1. Q10 anthropometry requirements and realised dimensions in the prototypes from EPOCh

Description	Requir. [mm]	Actual [mm]
A1 – Seated Height (head tilt)	747.6	733.7
A2 – Seated Height (via T1)	747.6	748.4
B - Shoulder Height (top of arm)	473	472.5
C - Hip Pivot Height	65.9	65.9
D - Hip Pivot from Back Plane	90.4[1]	90.4
- Hip Joint Distance	130.0[1]	132.0
F - Thigh Height	114.0	114.0
G - Lower Arm & Hand Length	374.7	374.2
I - Shoulder to Elbow Length	292.9	291.6
J - Elbow Rest Height	189.6	181.0
K - Buttock Popliteal Length	417.5	414.9
L - Popliteal Height	405.7	405.7
M - Floor to Top of Knee	445.6	446.0
N - Buttock to Knee Length	488.4	485.4
O - Chest Depth at Nipples	171.2	171.0
P - Foot Length	220.0	220.0
- Standing Height (head tilt)	1442.5	1441.2
- Standing Height (via T1)	1442.5	1455.5
R - Buttock to Knee Joint	(none)	445.7
R2 - Floor to Knee Joint	(none)	414.0
S - Head Breadth	143.9	144.0
T - Head Depth	187.4	186.5
U - Hip Breadth	270.4	271.5
V - Shoulder Breadth	337.8	337.8
W - Foot Breadth	86.0	86.0
X - Head Circumference	534.5	534.0
Y - Chest Circum at Axilla	687.3	604.6
- Chest Circum at Nipples	684.9	633.6
Z - Waist Circumference	593.5	664.6

1) The data from Reed *et al.* [6] are transformed from standing to sitting and scaled from 10 YO stature of 137.4 cm to 144.3 cm for Q10.

Table 2. Q10 mass requirements and actual values prototypes and ballasted versions used for testing

Body part	**Required** [kg]	**Original prototype** [kg]	**Ballasted prototype** [kg]
Head	3.59	3.59	3.59
Neck	0.60	0.63	0.63
Upper torso	5.15	5.14	5.14
Lower torso	9.70	8.04	9.02
Upper arm[1]	1.09	1.05	1.10
Lower arm[1,2]	0.90	0.83	0.90
Upper leg[1]	3.71	3.70	3.70
Lower leg[1,2]	2.53	2.44	2.44
Total mass	**35.5**	**33.4**	**34.7**

1) Each arm / leg
2) Including hands / feet respectively

Nonetheless provisions were included for side impact applications wherever possible. This includes for instance a pelvis design based on the WorldSID 5th dummy and a neck design based on the Q3s.

With regards to frontal biofidelity targets the approach defined for other members of the Q family was adopted. It consists of scaling of adult targets based on anthropometry and bone properties. By applying this approach Q-dummy research results from the previous EU FP projects CREST [8] and CHILD [9] become applicable for the Q10.

Biomechanical requirements for the various body parts were described in [10]. As an example, scaled performance corridors for neck My moment versus flexion and thorax force versus deflection for 4.27 m/s frontal impact are provided in Figure 4 and Figure 5 respectively. For the thorax impact tests an impactor with a mass of 8.76 kg is defined. Corridors for Q6 and Hybrid III 50th percentile male are given for reference. It can be observed that the Q10 targets are in between those of the smaller child and the adult dummy.

Instrumentation

In the development of the prototype Q10 the following provisions for instrumentation were included:

- 2D Displacements –Upper and lower rib cage.
- 3D Linear accelerations – Head, spine T4 and T12, sternum and Pelvis.
- 3D Angular rate sensors – Head, thorax and Pelvis.
- 6 axis loadcells – Upper and lower neck, Lower lumbar spine, Sacro-Iliac and Femur.
- Single axis loadcell – Pubic force.

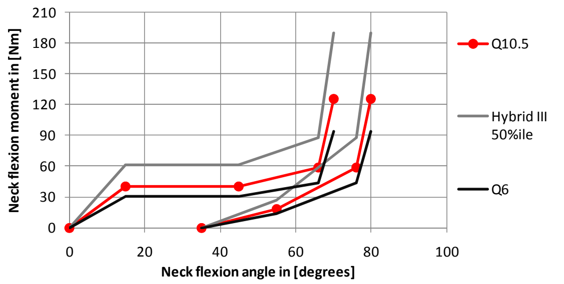

Figure 4. Q10 Neck flexion corridors.

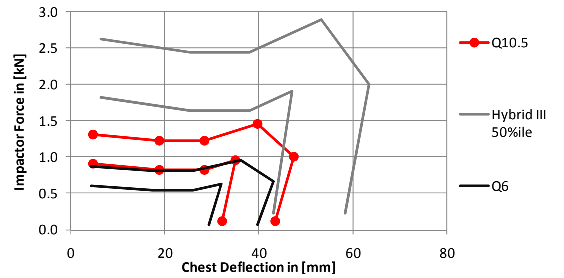

Figure 5. Thorax impact performance corridors.

For side impact application provision for alternative channels was implemented, such as 2-D chest displacement, T1 Ay acceleration and pubic symphysis load.

In addition to the above provisions for tilt sensors were included in various locations to facilitate positioning of the dummy.

Prototype Design Realisation

Using the requirements defined above, three prototype dummies were realised in the EPOCh project. The design and prototype evaluation are described in [10] and [11]. Figure 6 through Figure 9 give some impressions of the prototype design. The following summarises the performance of the prototypes realised as well as some key features of the Q10.

A comparison of the realised dimensions of the prototypes with the targets is given in Table 1. In Table 2 the actual mass distribution is compared with the requirements. As can be seen actual dimensions and masses correlate well with specifications set.

An apparent deviation in sitting and standing height is explained by the fact that these dimensions are measured in a fully erect posture while the dummy is assembled with the head tilted 27° forward. To enable comparison with an erect posture the dimensions measured via T1 are included in Table 1 as well. For this dimension good correlation is obtained.

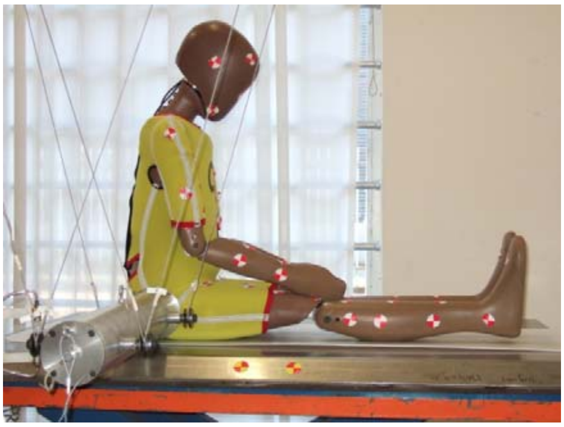

Figure 6. Q10 Prototype in pelvis certification test.

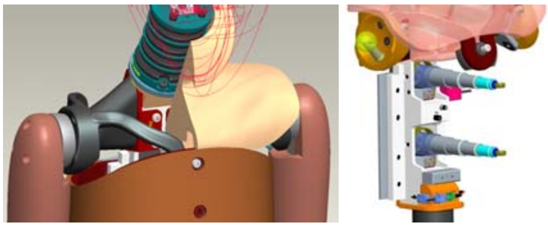

Figure 7. Shoulder design (initial concept) and double IR-Traccs in thorax.

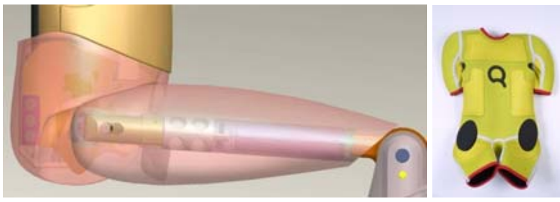

Figure 8. Detail of sit-standing pelvis design and suit with patches intended to prevent belt intrusion.

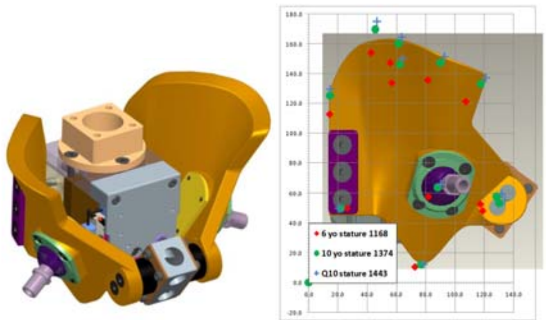

Figure 9. Pelvis design based on WorldSID 5th and comparison against bone anthropometry data from [6].

In addition to the sitting and standing height the chest circumferences show deviations. Actual dimensions are smaller than specified values because the soft muscle tissue at nipple and axilla level is not represented in the dummy. When considering the thickness of the suit (6 mm) which, to a degree, represents the flesh dummy values are much closer to the target.

The mass of some prototype parts revealed to be low within the mass tolerance for the upper and lower arms. With an addition of some ballast items to the upper arms, ~50 gram each, lower arms, ~70 gram each, and pelvis, ~980 grams, the mass was increased close to nominal weight. In design updates for the production version this was corrected adding mass to the related body parts.

For the head biofidelity three criteria for drop tests on a rigid plate were defined and evaluated [10]:

- Frontal 130 mm drop height: Biofidelity corridor limits based on EEVC scaling 113.1 – 194.2 *g*.
- Lateral 130 mm drop height: Biofidelity corridor limits based on EEVC scaling 116.4 – 200.0 *g*.
- Lateral 200 mm drop height: Biofidelity corridor limits based on ISO TR9790 107 – 161 *g*.

Figure 10 gives resulting accelerations for these conditions together with the corridors. The head of the prototype meets the frontal (130 mm) and lateral (130 mm) low in the EEVC corridors and meets the lateral (200 mm) high in the ISO/TR9790 corridor. During the prototype evaluation the approach to aim for simultaneously compliance with the EEVC and ISO/TR9790 corridor was found to be not feasible, therefore the EEVC corridors were selected to be the priority. As the skin stiffness generally will increase over time it was recommended to increase the stiffness of the head such that its performance is at the lower side close to the middle corridor [11].

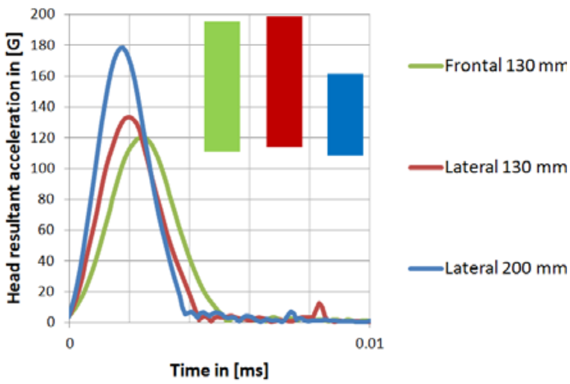

Figure 10. Head drop biofidelity results.

For the neck biofidelity requirements in flexion, extension and lateral flexion were defined. Figure 11 shows neck flexion bending performance in a Part

572 pendulum test with respect to the corridors defined [11]. The flexion response is in the lower range of the corridor and the rise in stiffness that should occur around 35° of head rotation is slightly late, starting around 45°. The slope in stiffness rise is correct. An improved performance could be obtained by increasing the rubber stiffness, but that would affect the fracture toughness and therefore the durability of the part. Another possibility is to change the neck mould, but this may affect the response in other directions which was not preferred.

Figure 12 gives neck extension bending performance in a Part 572 pendulum test in comparison with the biofidelity corridor [11]. It can be concluded that the extension performance fits the corridor very well. No further adjustment was deemed necessary.

Figure 13 shows the neck lateral flexion bending performance in a Part 572 pendulum test in comparison with the biofidelity corridor [11]. Up to 45° of head lateral flexion the performance is in the right order of magnitude but the rise in stiffness for large rotations is not followed.

Figure 14 shows shoulder pendulum force versus timein comparison with the biofidelity corridor. The test is done using an 8.76 kg pendulum at an impact speed of 4.5 m/s. It can be observed that the initial response of the shoulder is too stiff. In relation to this

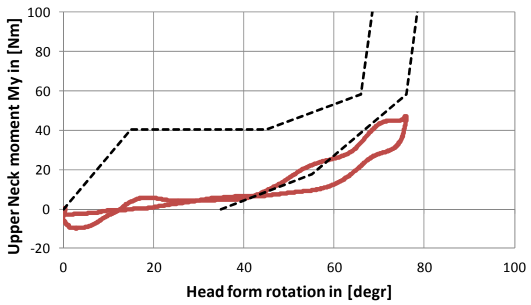

Figure 11. Neck flexion moment versus head rotation at 6.6 m/s.

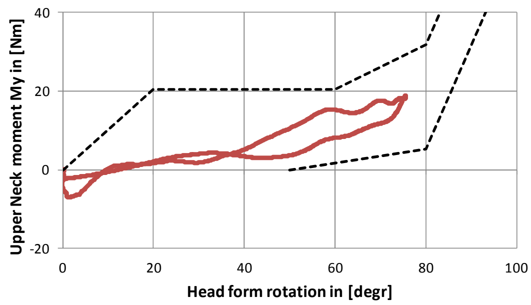

Figure 12. Neck extension moment versus head rotation at 4.3m/s.

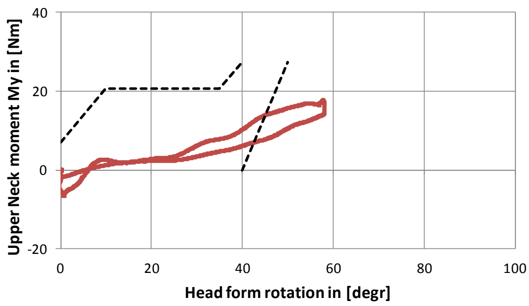

Figure 13. Neck lateral flexion moment versus head rotation at 3.9 m/s.

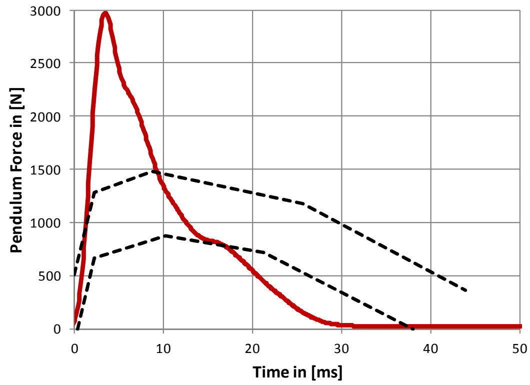

Figure 14. Lateral shoulder impact force versus time.

result it should be noted that the Q10 is an omni-directional dummy and performance tuning in either direction will affect the performance in the other direction. In the EPOCh project an optimal balance was sought in both directions with the focus on frontal impact. As shown below identical trends with regards to lateral impact performance are observed for thorax and pelvis region. Hence the stiffness distribution in side is balanced between these body regions, avoiding dominance of a single body segment in absorbing loads.

For the frontal biofidelity of the thorax two pendulum test impact speeds were specified: 4.31 and 6.71 m/s. Figure 15 provides pendulum force versus rib displacement in impact direction. Rib displacements are obtained by averaging the measured values in upper and lower chest IR-Traccs. It can be observed that the rib cage response meets the corridors reasonably well, especially for the 6.71 m/s impact. For the lower impact speed at 4.31 m/s the response is somewhat above the corridor. However, compared to other Q family members the frontal thorax performance is much better [12].

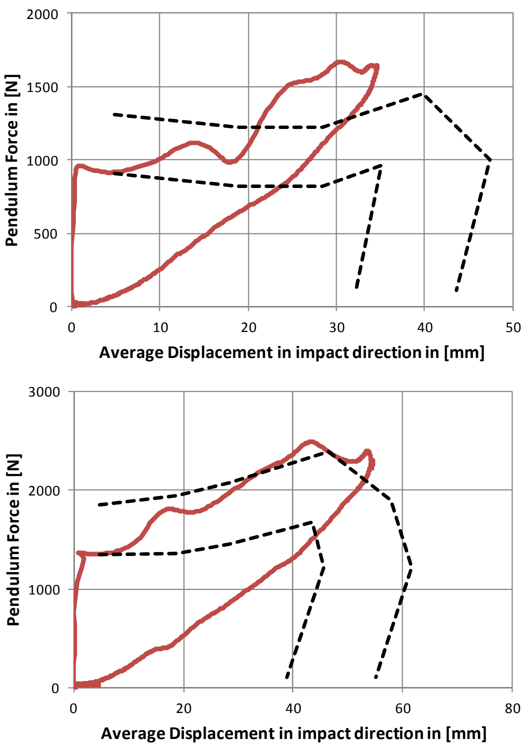

Figure 15. Thorax frontal pendulum impact 4.31 m/s (top) and 6,71 m/s (bottom).

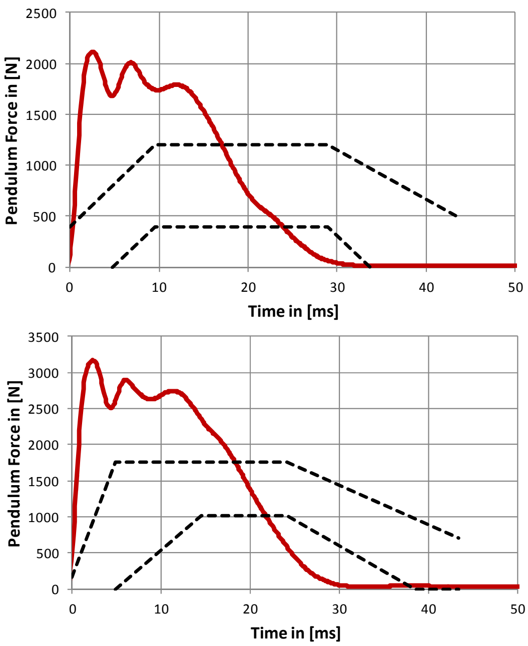

Figure 16. Thorax lateral pendulum impact 4.31 m/s (top) and 6,71 m/s (bottom).

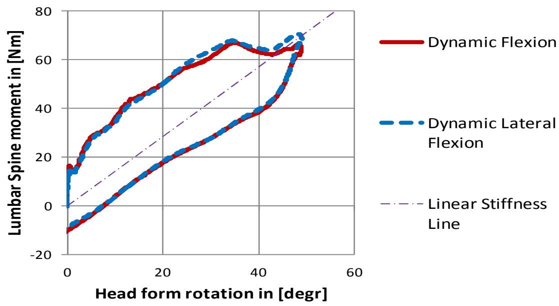

Figure 17. Lumbar Spine dynamic and static stiffness's (pendulum impact speed 4.53 m/s).

For the lateral biofidelity of the thorax two pendulum test impact speeds were specified: 4.31 and 6.71 m/s. Figure 16 gives results in comparison to the biofidelity corridors [11]. As for the shoulder the initial response overestimates the stiffness. Although performance tuning might be applied this would affect the frontal performance and introduce an imbalance with the shoulder and pelvis performance under lateral loadings.

The lumbar spine is made of a cylindrical rubber column; hence performances in frontal and lateral flexion are identical. Figure 17 gives test results obtained from pendulum impact tests. The dynamic stiffness is about 80 Nm/56° = 81.9 Nm/rad. This is slightly higher than the targets set for flexion (68.6 Nm/rad) and for lateral flexion 71.4 Nm/rad [11]. This result was considered to be acceptable.

Figure 18 gives the lateral pelvis impact performance in terms of pendulum force versus time. Again results are shown in comparison with the biofidelity corridor [11]. The pelvis response is in line with the lateral shoulder and thorax responses, showing too high stiffness. During the lateral pelvis impact tests a bottoming-out contact between the iliac wing and the sacrum block was observed for impact speeds of 4.0 m/s and above. This should not occur until the impact speed exceeds 5.2 m/s.

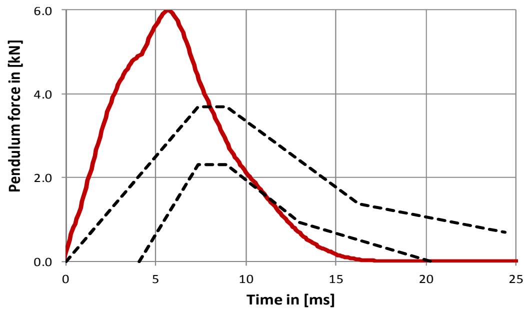

Figure 18. Pelvis lateral pendulum impact at 5.2 m/s.

When considering the side impact performance in the different regions it can be concluded that shoulder, thorax and pelvis pendulum impact responses largely exceed corridors in the initial phase. The stiffness in all regions is too high and further improvements for this configuration is needed (and being worked on as described below).

INJURY RISK FUNCTIONS

To apply the dummy it is necessary to specify injury criteria, risk functions and thresholds which are appropriate to this age and size of occupant. With adult humans the conventional approach taken to derive injury risk functions has been to conduct representative tests around the injury threshold with Post-Mortem Human Subjects (PMHS). These tests are then repeated with the dummy and the relevant dummy output compared against the observed risk of injury for the PMHS. By following this process, dummy-specific injury risk functions are defined directly relating a dummy measurement with the risk of injury for a human. Unlike the adult situation, however, there is very little biomechanical data from which specific injury risk functions for children can be derived. As alternatives, two approaches have been used [12]:

- Perform accident reconstructions using the child dummy under development.
- Scale adult injury risk functions and/or criteria to be relevant to the child size (dummy) being investigated.

The European Enhanced Vehicle-safety Committee (EEVC) Working Groups 12 and 18 used the accident reconstruction data developed within the European Commission (EC) CREST and CHILD projects to help develop risk functions for the Q3 dummy [12]. These functions were updated within the CASPER project [13].
Accident reconstructions with the newly developed Q10 were beyond the scope of the EPOCh project. As an alternative the EPOCh project took the second approach and scaled adult injury risk functions in an attempt to make them relevant for the older child dummy. To provide a comparison, the risk functions developed for the Q3 by EEVC WGs 12 and 18 were scaled up to the Q10 using the same formulae.

Previously, many authors have published techniques for scaling biomechanical measurements to different sizes of subject [14], [15], [16], [17]. While the general principle behind the scaling remains consistent, each of the publications seems to adopt different specific details. EPOCh therefore reviewed available scaling methods for each body region and dummy measurement. The review considered whether there are any new material property data available to aid the scaling process and if the output was reasonable.
The scaled injury risk functions or criteria were then compared with initial test results with the Q10 dummy under Regulation 44 conditions. Associated with this comparison has been an assessment of the feasibility for CRS manufacturers to meet prospective criteria. This has also been balanced with pragmatic expectations of how well the criteria may relate to current CRS performance and real world accidental injury incidence. The limits for the Q10 dummy resulting from this approach and to be used in UN Regulation No. 44 frontal impact conditions are summarised in Table 3. A more detailed description of the full EPOCh injury risk review is provided in [11]. Further work on the thresholds is needed to arrive at final values for frontal and derive values for side impact.

Table 3. Proposed injury criteria for use with the Q10 dummy in UN Regulation No. 44 frontal impact conditions [11]

Measurement	Threshold
Head 3 ms exceedence	80 *g*
Head horizontal excursion	465 mm
Head vertical excursion	885 mm
Neck tension	†
Neck flexion	125 Nm
Neck extension	37 Nm
Chest deflection (either IR-Tracc)	56 mm
Chest 3 ms exceedence	45 *g*

† To be set after further testing with the Q10

THIRD PARTY EVALUATION

Following the EPOCh evaluations two instrumented Q10 prototype dummies were made available to third parties for further evaluation testing. A wide variety of tests were performed by research labs, restraint manufacturers, OEMs and consumer organisations world-wide to check on the dummy performance in a vehicle environment. The tests included sled tests on a body in white as well as full-scale crash tests. Different restraint configurations were tested considering belts with and without pretensioner and belts with and without load limiter in combination with different child restraint types. Variations in test conditions included tests with sled buck rotation left and right, with and without seat in front and partial overlap to the left and right. In some tests the dummies were equipped with add-on features, such as abdomen sensors and lap shields to prevent belt intrusion between the pelvis and the thigh.

Figure 20 shows some typical test results. Values for head 3 ms exceedence and upper neck bending My are depicted in cross plots against chest 3 ms exceedence. Results are normalised with respect to thresholds from Table 3. Results from the EPOCh project are included for reference (grey coloured markers). Chest acceleration 3 ms is generally high in NPACS sled tests (45 to 70 *g*) performed in EPOCh and low in BIW sled tests (35 to 30 *g*). The UN Regulation No. 44 tests performed in EPOCh show values in the middle range (25 to 50 *g*). For the upper neck My (extension) high values are observed in some tests. In most cases this can be attributed to impact loads upon rebound. In tests with pretensioner and load limiter a high extension moment may occur at the end of the loading phase (see e.g. Figure 19). Studies using a detailed finite element model of the Q10 dummy are on-going to investigate the root cause of this phenomenon. The high head acceleration 3 ms exceedence results found in some tests was also attributed to rebound impacts.

Feedback was provided on several dummy performance issues, the most important of which included:

1) Belt interaction in the pelvis region: lap belt penetrating between pelvis and thigh;
2) Belt slip towards the neck: observed after introducing Cordura top layer in the suit for durability;
3) Side impact performance: to be improved and, if possible, instrumentation to be added in shoulder region (e.g. shoulder load cell);
4) Durability of various dummy parts.

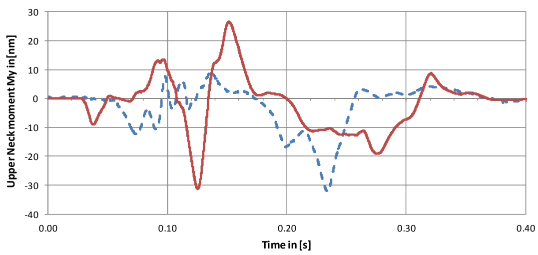

Figure 19. Upper Neck moment My for test with pretensioner and load limiter (red, solid line) and without (blue, dashed line).

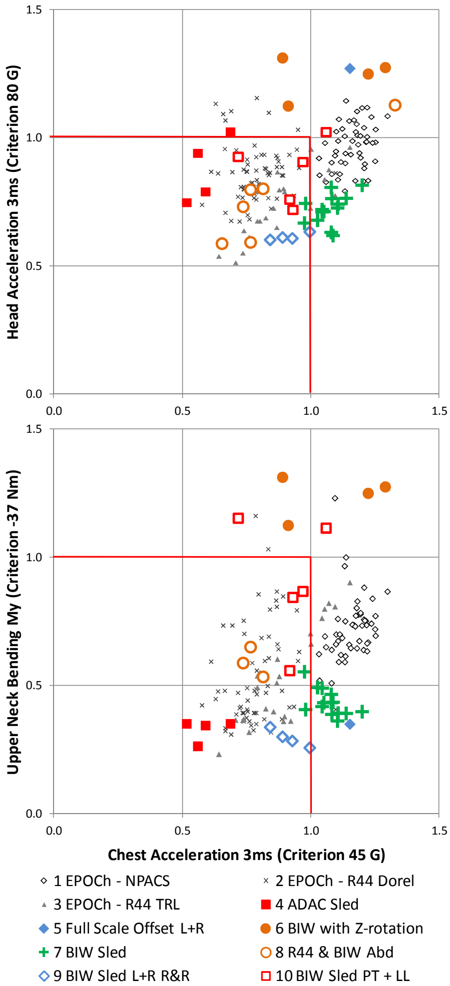

Figure 20. Maximum parameter values in EPOCh and 3rd party tests (normalised w.r.t. criteria set): Head acceleration 3 ms exceedence versus chest acceleration 3 ms (top); and upper neck My versus chest acceleration 3 ms exceedence (bottom).

DUMMY UPDATES FOR Q10 PRODUCTION VERSION

Following the testing with prototypes efforts were made to develop a production version Q10 dummy. Remarks related to durability, performance and handling of the dummy forwarded by the EPOCh and third party test programs were addressed. In the following a brief overview of the main changes is provided. A full overview is provided in [18].

Head and Neck - Additional dimple markers were added to the head for accurate dummy positioning.

Also the biofidelity was aligned with the EEVC requirements by making the response stiffer.
For neck no changes were introduced other than adding a lifting strap for handling, and reorientation of some of the washers.

Thorax / shoulder complex - The rib cage of the prototype dummies was produced using Ureol material which was recently banned under the REACH regulation. The new material formulation, also applied in other Q dummies was adopted. Figure 21 and Figure 22 compare the performance of production version rib cages with new material to the prototypes in frontal and lateral impactor biofidelity tests. Essentially identical responses are obtained.
Shoulder durability issues were reported, with the rubber tearing from the end plate. This failure was caused by high belt loading at top of arm in tests with the belt slipping over the shoulder onto the arm. As a countermeasure a stronger internal wire was applied in combination with a stress relief at the attachment of the rubber to the end plate (see Figure 23). This updated design was already used in the third party testing program. No further failures were reported, indicating that durability was improved.

Abdomen and Lumbar Spine - During tests with the prototypes severe ballooning of the abdomen was observed (see Figure 24). This part consists of a foam block with PVC skin. It is probable that air could not escape from the single vent hole fast enough, so six more holes were added evenly to the top rear and bottom of the skin. As the abdomen is soft it is not expected that this modification influences the dynamic response. However, this is to be evaluated in future testing. The lumbar spine only had very minor changes (e.g. change to screws with socket heads).

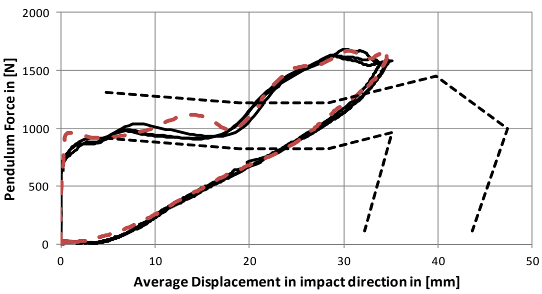

Figure 21. Performance of production version dummy in frontal thorax biofidelity test (solid black lines production dummy tests (3 off), dashed red line prototype).

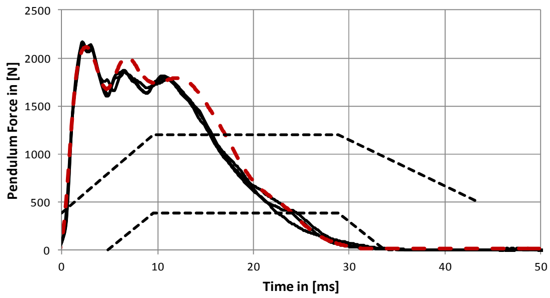

Figure 22. Performance of production version dummy in lateral thorax biofidelity test (solid black lines production dummy tests (3 off), dashed red line prototype).

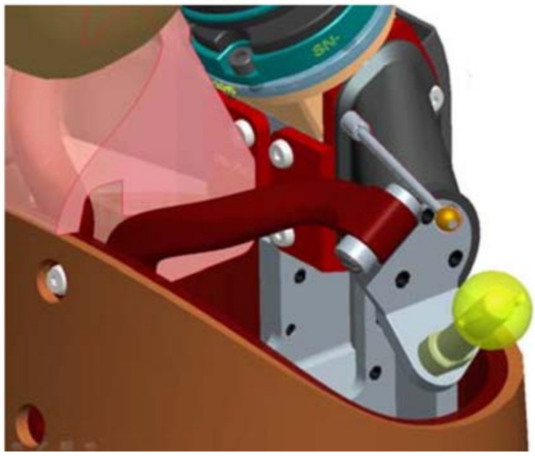

Figure 23. To reinforce the shoulder under loads from outboard belts the internal cable was reinforced and oriented more horizontally.

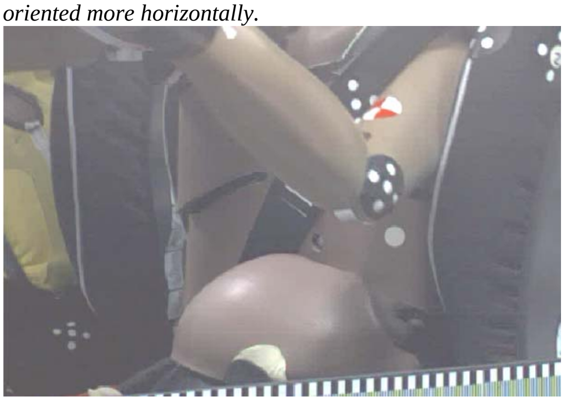

Figure 24 –Test without suit showing ballooning of the abdomen.

Pelvis - In the lower torso region the prototypes were 1660 grams too light and a temporary ballast weight (980 grams) was added to the pelvis to compensate. For the production dummy mass was added to the abdomen, pelvis flesh and pelvis bones, and some steel parts in the pelvis were replaced with tungsten.

DAS ballast was also added to allow for DAS integration into the sacrum. The changes resulted in a total mass increase of ~1550 grams bringing the total mass of the lower torso up to 9590 grams. This is close to target (9700 grams) specified in Table 2. The remaining 110 grams is consumed by the wire mass.
As with the thorax the pelvis flesh and bone material was banned due to the REACH regulation. Replacement materials as applied in the other Q dummies were introduced and tuned to provide identical performance and improved durability.
In lateral pelvis impactor tests contact between the iliac wing and the sacrum block was observed from impact speeds of 4.0 m/s onwards. To raise this contact to a higher impact speed the clearance between iliac wings and sacrum block was increased. Also the pubic buffers were stiffened. Future lateral pelvis testing is planned to evaluate the performance of these updates.

Arms - In the third party testing fracturing of the lower arm flesh at the wrist section occurred upon impact against the front seat. To resolve this failure tougher flesh material has been introduced in combination with rounded edges to the lower arm bone end. Like other items the performance of this solution is to be investigated in future testing. In case fracturing in the wrist section remains it might be decided to change the geometry of the hands from stretched to fist configuration.
Other modifications to the arms included the introduction of locking threads for the shoulder joint screws to maintain 1 *g* friction and adding friction screws on both sides of the elbow to balance loading and prevent damage in this joint. For handling purposes marker dimples were added to the wrist.
Mass was raised to the weight specified in Table 2, avoiding the ballast weights applied in the prototypes.

Legs - As for the arms tougher flesh material was introduced to the lower leg improve durability. Also the lower leg bone was extended for this purpose. In various tests excessive sliding of the upper leg flesh along the femur bone towards the knee was observed, introducing damage to the flesh in the pelvis area. To prevent sliding of the flesh along the metal bones observed in the production version, the flesh is locked to the bone by adding a retainer in an access hole for the femur load cell (see Figure 25).
Various minor modifications were introduced in the knee such as anti-fretting plastic washers and friction stops to the knee stop pins.
Mass was raised to the weight specified in Table 2.

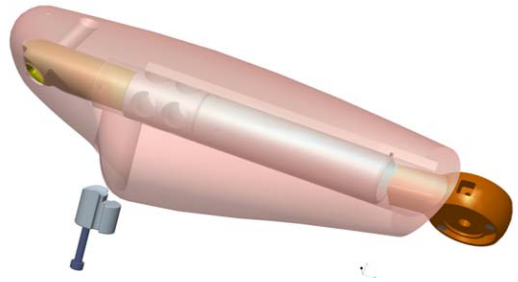

Figure 25. Upper leg flesh with insert.

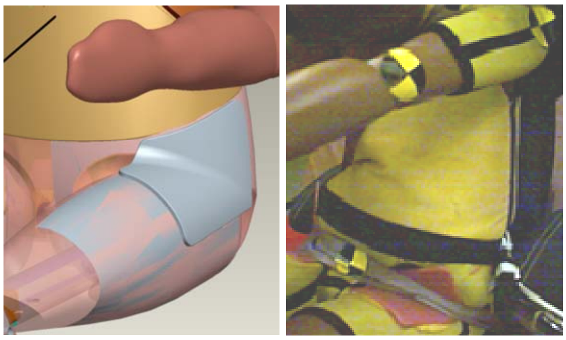

Figure 26. Lap belt shield design (left) and shield performance in dynamic condition (right).

Suit - In the prototype, hip patches where introduced to try and prevent lap belt intrusion into the hip pelvis gap. This did not work and profiled, stiffer, hip shields have been incorporated for further evaluation testing (see Figure 26).
A zip was added in the abdomen area to help with fitting and umbilical wire access. The front chest panel was covered with Cordura for better durability.

SIDE IMPACT PERFORMANCE – FE STUDY

In an attempt to define future improvements for side impact performance a finite element study into possible concepts has been conducted (see Figure 27). The initial focus was on the shoulder and thorax performance for interaction with vehicle restraint systems. To avoid any negative effects on the frontal impact performance it is proposed to introduce a side impact kit with a minimal number of components that are to be exchanged for side impact applications. The configuration evaluated through simulation included:

1) Omitting the lower arms;
2) Plastic upper arm bone (instead of aluminium);
3) Softened shoulder rubber and arm flesh.

Results for shoulder pendulum impact forces are depicted in Figure 28. Omitting the lower arm only does not have a significant influence of the shoulder impact force as this part is remote from the impact location and due to interaction of the stiff arm (aluminium bone) with the thorax. When introducing a plastic bone in the upper arm (as used for the WorldSID 5th female dummy), shoulder forces were reduced. Softening of the shoulder rubber and arm flesh results in a further reduction, bringing forces close to the corridor. This design concept for improved side impact performance will be explored in more detail and if feasible realised in hardware for further evaluations.

An additional item for the side impact performance is the realisation of a shoulder load cell. Initial design efforts have started (see Figure 29). However, due to the shoulder concept with ball joint on an oblique pin, issues arise related to cross talk, non-linearity and hysteresis. Further efforts will be needed to realise an adequate solution.

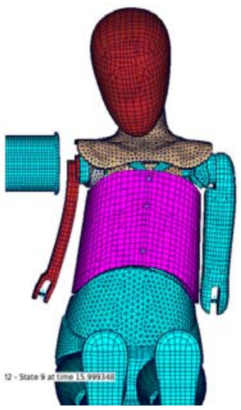

Figure 27. Deformed configuration for simulation with lower arms removed and upper arms realised with plastic bone (arm flesh in left arm removed for visualisation).

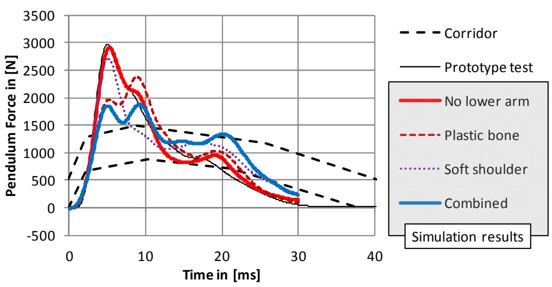

Figure 28. Shoulder impact pendulum force versus time Prototype test result (for reference) and simulation results: No lower arm, No lower arm + Plastic upper arm, No lower arm + Soft shoulder rubber and all three measures combined.

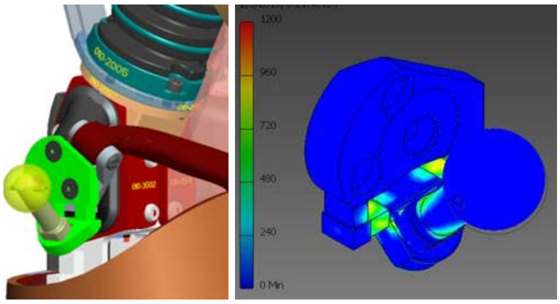

Figure 29. Shoulder load cell (green item in left plot) concept packaging and feasibility study.

MEASURING ABDOMEN LOADING WITH THE Q10 – FIRST STEPS

UN Regulation No. 44 specifies performance requirements for head excursion, chest acceleration and abdomen loading (by means of a clay insert between the lumbar spine and the foam abdomen block), using the P-Series dummies. Collision studies undertaken by EEVC WG 12 and 18, the CASPER and EPOCh projects indicate that it would be desirable to maintain the assessment of child restraint system performance at these body regions when the Q-Series is introduced into legislation [12, 19, and 20]. At present, the Q dummies have no method for detecting abdomen loading. It cannot be fitted with a clay insert and will require another solution. A potential solution is provided by the Abdominal Pressure Twin Sensors (APTS) developed at IFSTTAR during the CHILD and CASPER projects [21]. The APTS sensor consists of two cylindrical soft polyurethane bladders filled with a gel solution (see Figure 30). They are closed by aluminium caps in which pressure sensors are located. The bladders are implanted in holes drilled in the abdominal block of the dummy. In the CASPER project, risk curves were derived for sensors installed in the Q3 and Q6 [e.g. 13]. Risk curves for the Q10 are yet to be developed.

Various studies have shown that the lap part of the seat belt can become trapped in the gap between the legs and the pelvis in the Q-Series dummies [22]. The pelvis design of the Q10 differs from that of the other Q-Series dummies and minimises this gap. In addition, patches were stitched into the suit of the prototype Q10 to further limit belt intrusion. Nevertheless, the phenomenon was observed by participants in the third-party testing described earlier and a new solution has been developed consisting profiled hip shields (see Figure 26).

To investigate the use of the APTS sensors with the Q10 dummy and the capacity of the hip shields to prevent belt intrusion into the gap between the legs and the pelvis a programme of three impact sled experiments was carried out. The Q10 was seated in a non-integral ISOFIX child restraint system (a booster seat) on the test bench defined in the draft new UN Regulation on “Enhanced Child Restraint Systems”. It was exposed to a pulse defined in the draft Regulation (which is identical to that it UN Regulation 44). Three tests were done:

- Test #1: Reference test, no APTS or hip shields
- Test #2: Using APTS sensor but no hip shields
- Test #3: Using both APTS sensor and hip shields

The booster seat did not feature guides for the lap part of the seat belt, although severe abdomen loading was not expected in these experiments because the seat was approved to UN Regulation 44.

Figure 32 shows the interaction between the dummy and the seat-belt during each experiment. The abdomen sensors did not seem to influence the kinematics of the dummy or the way it interacted with the seat-belt (tests 1 and 2). The lap parts of the belt intruded partially into the gaps between the legs and the pelvis, but not to the same extent as that reported for other Q-Series dummies [see e.g. 18]. The abdomen ‘ballooned’ over the lap belt, but this seemed to reduce when the hip shields were used (test 3). The hip shields also helped to keep the lap part of the belt higher on the pelvis and away from the gap between the legs and the pelvis. Table 4 compares some of the main dummy measurements from the sled experiments. The Q10 had been used extensively prior to these experiments as part of the third-party testing described earlier. Although the dummy was inspected for damage, its schedule between the laboratories did not allow for regular certification and hence these data should be viewed in that context. The table shows that the abdomen sensors and the hip shields in this small programme, did not tend to influence the broader dummy measurements, but there were some exceptions and it would be worthwhile to continue to investigate this further.

Figure 30. APTS Abdominal pressure sensor (left) and abdomen with APTS inserted (right).

Evaluation tests as done by Takata in the third party testing programme also showed small changes in dummy readings when using the hip shields. See Figure 31. This was explained by small changes in the dummy kinematics when using the shields.

Relatively low levels of pressure were recorded by the APTS sensors, which was consistent with the low levels of abdomen loading observed in the videos and highlighted in Figure 32. The pressure was higher on the right side of the dummy, on the buckle-side of the seat belt. Beillas *et al.* [21] also reported low levels of abdomen pressure in booster seats that were used correctly, under UN Regulation 44 conditions; test bench and pulse.

These experiments were part of a wider programme supported by the European Commission (DG Enterprise and Industry) with the aim of investigating the state-of-the-art of the Q-Series dummies and sensors for measuring and assessing abdomen loading in the draft new UN Regulation. Further experiments and analyses by TRL and other members of the UN Informal Working Group on Child Restraint Systems are on-going with a view to proposing and validating a solution that can be implemented in the new Regulation.

Table 4. Q10 dummy measurements during the experiments

	Test #1 No sensor, no shield	**Test #2 Sensor, no shield**	**Test #3 Sensor and shield**
Head exc. (mm)	350	326	346
Head acc. 3ms (g)	62.9	66.3	78.6
Upper neck Fx (kN)	-0.98	-1.00	-1.17
Upper neck Fz (kN)	3.63	3.79	4.80
Upper neck My (Nm)	-16.6	-13.5	-18.5
Chest acc. 3ms (g)	34.0	32.8	35.4
Chest def. upp. (mm)	49.1	47.9	47.3
Chest def. low. (mm)	47.2	42.4	39.7
APTS left (bar)	No sensor	0.72	0.68
APTS right (bar)	No sensor	1.22	1.27

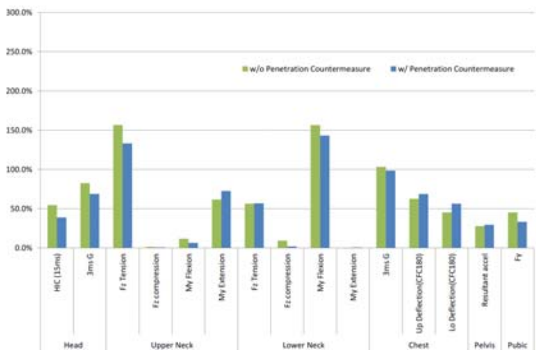

Figure 31. Influence of hip shields on dummy reading (data provided by Takata)

Figure 32. Q10 interaction with the seat belt: Test #1 No abdomen sensor and no hip shields (left); Test #2 Abdomen sensors without hip shields (middle); Test # 3 Abdomen sensors and hip shields (right).

DISCUSSION

Real-world car crash records shows that good achievements have been made over the past decades in reducing the number of fatally and severely injured children on European roads. Nonetheless further efforts are needed to improve the safety for this group of road users. The EU FP7 project EPOCh contributed to this by developing a Q10 dummy that represents adolescents.

In the EPOCh project prototype Q10 dummies were realised and extensively evaluated in UN Regulation No. 44 and NPACS test conditions [11]. Subsequently two dummies were forwarded to OEM's, suppliers and test houses world-wide to evaluate the dummy performance in a vehicle environment. Feedback and recommendations on design updates were collected for implementation in a production version of the Q10 dummy. The main remarks on the dummy performance included:

1) Belt interaction in the pelvis region: lap belt penetrating between pelvis and thigh;
2) Belt slip towards the neck: observed after introducing Cordura top layer in the suit for durability;
3) Side impact performance: to be improved and, ir possible, instrumentation to be added in shoulder region (e.g. shoulder load cell);
4) Durability of various dummy parts.

Various studies have shown that the lap part of the seat-belt can become trapped in the gap between the legs and the pelvis in the Q-Series dummies [22]. Based on this experience the pelvis of the Q10 was designed to have a minimal gap while maintaining the sit-standing concept of the Q dummies. This, even in combination with patches stitched into the suit, did not prevent the belt entrapment. A solution consisting of a profiled hip shields (see Figure 26) was developed and first evaluations do indicate that belt entrapment is reduced (see Figure 32). Sled tests also showed that the shields may affect dummy readings to some extent, requiring further investigations.

The shoulder belt slippage towards the neck was not that profound in EPOCh testing when a soft neoprene suit was used. To prevent reported damage to the suit under belt loading a Cordura top layer was added for use in the third-party testing programme. This resulted in an improved durability (no damage was reported). However, the reduced friction between belt and suit might promote the belt slippage towards the neck. Further studies are necessary on this item also considering location of the instrumentation in the dummy chest as well as the influence of external items like belt geometry. Recent studies have shown the influence of the belt position on the belt slippage in other Q dummies [23].

A key item for future improvement concerns performance in side impact. Although it is generally recognised and acknowledged that the dummy design was focused on frontal performance, updates in side impact were recommended to support restraint design for this configuration. To avoid any negative effects on the frontal impact performance it is proposed to introduce a side impact kit with modified arm and shoulder rubber which are to be exchanged for side impact applications. Concepts are being explored using a finite element model of the Q10. Simulations show that a significant improvement of the side performance can be obtained, with shoulder impact loads close to the corridor, but this is to be further evaluated by realising a design. In view of the size of the Q10 dummy, being close to an adult, it is also proposed to evaluate its performance in a full-body side impact configuration. Previous PMHS tests from Wayne State University on small females may serve as basis for this. Results from these tests in which the shoulder, thorax and pelvis were impacted simultaneously in well-defined conditions served as basis for the requirement definition of the small female WorldSID 5th dummy. For application to the Q10 dummy response data and requirements are to be scaled assuming changes in material properties and geometry and test should be repeated using the Q10 dummy to evaluate its performance in the distributed loading conditions. In support of these activities a shoulder load cell is currently being developed.

During the EPOCh and third party testing various issues related to durability and handling of the Q10 dummy were listed. Issues on the shoulder as raised by EPOCh were implemented before testing by third parties and found to be effective (no further damage was reported). Various other items like local failures in the flesh are being addressed in the production version of the dummy which is currently being realised.

First proposals for frontal injury criteria were made by the EPOCh project considering scaling of existing adult data. Results were cross checked by scaling available child data. As for studies in risk curves and thresholds in general caveats had to be taken into consideration when defining these first proposals for the limits. Each of the scaling approaches available from literature makes numerous approximations to keep the formula relatively simple to calculate. As an example geometric similitude is often cited as an assumption, so that the smaller body is the same as the larger in all aspects but size. Such assumptions and approximations will affect the scaling ratios and results. However, it is hoped that those effects are relatively small to the other aspects being taken into consideration.
In view of the above it can be concluded that further work on proposed criteria and thresholds is needed. In this work pragmatic decisions based on an expected real world performance of CRS could be considered.

CONCLUSION

The Q-Series dummy family is being completed with the development of a Q10 dummy representing larger children. Prototype dummies were developed and evaluated in the EU FP7 EPOCh project in UN Regulation No. 44 and NPACS conditions. Further testing was performed by OEM's, suppliers and test houses world-wide to explore the performance of the dummy in a vehicle environment. The testing resulted in various items for improvement which are currently being realised in a production version of the dummy. Further work is needed though to improve the performance of the dummy in side impact. For frontal impact phenomena related to belt slip on the thorax, belt penetration in the pelvis region, neck loadings observed when using advanced restraints and the implementation of abdominal sensors need further investigations. In parallel further work is needed on injury criteria and thresholds, including establishing means for abdominal loadings.

ACKNOWLEDGEMENT

The authors would like to thank the European Commission for commissioning and funding the EPOCh project in which the Q10 prototype dummies were developed. The authors would also like to thank the Commission for funding the experiments to measure abdomen loading with the Q10. Any views expressed in this paper are not necessarily those of the European Commission. Special thanks also to all those that contributed to the evaluation testing of the Q10 dummy providing valuable feedback on the dummy design.

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