Q-DUMMIES FRONTAL INJURY CRITERIA

1. Introduction

No child biomechanical data directly usable for Q-dummies is currently available in the literature. Only very few cadaver tests were performed using children (Kallieris 1976). Moreover those tests only provide a comparison between the child and the dummy response and do not provide any information regarding the injury mechanisms or thresholds.

Studies performed using animal testing and a GM 3 year old dummy (Mertz et al 1982, Prasad and Daniel 1984) proposed animal injuries paired with dummy measurements. Tolerance data were obtained for the head, the neck, the thorax and the abdomen. However, the main limitation of these data is that the dummy used was the GM 3 year old dummy.

Accident reconstructions were performed (Planath et al 1992, Newman and Dalmotas 1993). Injury criteria were derived from these tests for the dummies used in the reconstructions.

The CREST project, co-funded by the European Comission, included an extensive program where 56 real world accident were reconstructed using P and Q dummies. The project was completed in 2000. However, the number of tests using Q-dummies was not large enough to construct reliable injury risk curves.

A second project of accident reconstructions called CHILD was lunched in September 2002 to continue the development of the Q-dummies and to define the injury risk curves. For that purpose, the injuries observed in the real world accidents were paired with the Q-dummy measurements. Injury risk curves were drawn for the head and the thorax.

This document is a proposition of injury criteria specific to Q-dummies for frontal impacts corollating results of the CHILD project with scaling injury criteria available in literature.

2. Method

2.1. Scaling method

The scaling technique is used in biomechanics to derive the response and the injury thresholds of a specimen from the response and the injury thresholds of another subject, the size of which is different. For that purpose, the variations of stiffness, geometry and failure stress are either observed from tests or assumed, as a function of age or size of the specimen. The mass density is assumed to be equal for children and adults (Melvin 1995).

In our study, this technique is used:

- To derive the information regarding the Q dummies from the information available for the 50th centile male adult. The injury criteria, applicable for the Q-dummies are derived from the injury criteria available for the Hybrid III midsize adult male dummy.

- To derive the information regarding the Q3 dummy from the information available for the Hybrid III 3 years old dummy.
- To derive Q3 dummy values from Q dummies of different ages

2.1.1. Reference data

2.1.1.1. Adult and child dummy references regarding the anthoropometry

The Hybrid III midsize adult male and Hybrid III 3 years old dimensions are based on Irwin and Mertz (1997).
The Q dummy dimensions are those used in the specs of the last version of the Q-dummies. They were provided by TNO and were based on the CANDAT database (van Ratingen et al 1997)

2.1.1.2. Adult and child references regarding the material properties

- **Testing data**

Yamada (1970) reported an extensive study of the mechanical properties of the human soft and hard tissues. The calcaneal tendon stiffness E_t and failure stress σ_t are reported for fetuses (5; 6; 7 and 8 gestational month old), for children (newborn; 4.5; 14.5 years old) and for adult.
Based on this data, the calcaneal tendon stiffness and stress are interpolated for the 6 month, 12 month, 18 month, 3 year old and 6 year old children (Table 1).

Table 1 : Calcaneal tendon stiffness ratios and failure stress ratios

	0 year	1 year	1,5 year	3 years	6 years
λ_{Et}	0,48	0,58	0,61	0,77	0,88
$\lambda_{\sigma t}$	0,63	0,70	0,75	0,85	0,96

Mc Pherson and Kriewall (1980) performed a study where the mechanical properties of fetal cranial bone are reported. The stiffness of the skull, E_b, was measured for fetuses and for a six-year old child.
Based on this data, the skull stiffness is interpolated for the 6 month, 12 month, 18 month, 3 year old and 6 year old children (Table 2).

Table 2 : Skull bone stiffness ratios

	0 year	1 year	1,5 year	3 years	6 years
λ_{Eb}	0,24	0,32	0,36	0,47	0,67

- **Assumed data**

Melvin (1995) reports that the development processes of the collagenous and ligamentous tissues are observed to be equivalent. Therefore, it is assumed that the variations of the

mechanical properties of the brain (Mertz 1998) of the neck ligamentous tissue (Melvin 1995), as a function of age, are the same as those of the calcaneal tendon.
The cranial bone data are also used to scale biomechanical data for bone structures (Melvin 1995).
Since no age dependent failure stress data were available in literature Mertz et al (1997) assumed the heart failure stress is independent with age.
All the assumed ratios are summarized in Table 1

Table 3 : Assumed ratios

	Brain	Neck ligamentous tissu	Rib	Heart
λ_E	/	$=\lambda_{Et}$ (Calcaneal tendon)	$=\lambda_{Eb}$ (Skull bone)	=1
λ_σ	$=\lambda_{\sigma t}$ (Calcaneal tendon)	$=\lambda_{\sigma t}$ (Calcaneal tendon)	$=\lambda_{\sigma t}$ (Calcaneal tendon)	/

2.1.2. Determination of the scaling ratios

2.1.2.1. Head

The definition of head scale factors is based on (Mertz 2003):

$F=m\gamma$ (1)

$F=\sigma S$ (2)

Where:

F is the force applied on the head
m is the mass of the head
γ is the acceleration of the center of gravity of the head
σ is the head failure stress
S is the head cross sectional area

Equations 1 and 2 can be combined to give the acceleration:

$\gamma=\sigma S/m$ (3)

So acceleration ratio is:

$\lambda_\gamma= \lambda_\sigma \lambda_S/ \lambda_m$ (4)

Since child and adults head were assumed to be of equal density $\lambda_m= \lambda_x\lambda_y\lambda_z$, where x is the head length, y is the head breadth and z is the chin to vertex distance
Surface ratio is $\lambda_S=\lambda_y\lambda_z$
Irwin and Mertz (1997) have shown that the brain modulus is of first order on cranial modulus therefore the head stiffness depends on brain stiffness. Lastly as it was assumed that the variations with age of the brain tissue and of the calcaneal tendon are the same $\lambda_{\sigma}= \lambda_{\sigma t}$

So the acceleration ratio is:

$\boxed{\lambda_\gamma= \lambda_{\sigma t}/\lambda_x}$ (5)

HIC ratio is:

$\lambda_{HIC}= (\lambda\gamma)^{2.5}/\lambda_T$ (6)

Where $\lambda_T= \lambda_x/(\lambda_\sigma)^{1/2}$

So the combination equations 5 and 6 gives:

$\boxed{\lambda_{HIC}= (\lambda_{\sigma t})^3/(\lambda_x)^{1.5}}$ (7)

The head scaling factors from the Hybrid III midsize adult male dummy to the Q dummies are summarized in Table 4

Table 4 : Head scaling factors from the Hybrid III midsize adult male dummy to the Q dummies

	Q0	Q1	Q1.5	Q3	Q6
λ_γ	0,99	0,84	0,87	0,94	1,03
λ_{HIC}	0,49	0,45	0,53	0,71	0,98

The head scaling factors from the Q dummies of different ages to the Q3 dummy are summarized inTable Table 5

Table 5 : Head scaling factors from the Q dummies to the Q3 dummy

	Q0	Q1	P1.5	Q3	Q6
λ_γ	0,95	1,12	1,07	1	0,91
λ_{HIC}	1,49	1,59	1,35	1	0,72

2.1.2.2. Neck

The scaling method is based on muscular moment arm and cross-sectional area of the neck muscles (Mertz 1989). The axial force F can be expressed as:

$F=\sigma S$ (1)

Where

σ is the neck failure stress

S is the neck area

The axial force ratio is:

$\lambda_F= \lambda_\sigma\lambda_S$ (2)

As it was assumed that the variations with age of the neck tissue and of the calcaneal tendon are the same $\lambda_\sigma= \lambda_{\sigma t}$

Surface ratio is $\lambda_S=\lambda_x\lambda_y$ (3)

Where x is neck depth and y is neck width

Combining equations 2 and 3 gives:

$\boxed{\lambda_F= \lambda_{\sigma t}\, \lambda_x\lambda_y}$ (4)

The bending moment can be expressed as:

$M=Fx$ (5)

The moment ratio is:

$\lambda_M= \lambda_F\lambda_x$ (6)

Combining equations 4 and 6 gives:

The bending moment ratio is:

$\boxed{\lambda_M= \lambda_{\sigma t}\, \lambda_x^2\lambda_y}$ (7)

The neck scaling factors from the Hybrid III midsize adult male dummy to the Q dummies are summarized in Table 6

Table 6 : Neck scaling factors from the Hybrid III midsize adult male dummy to the Q dummies

	Q0	Q1	Q1.5	Q3	Q6
λ_F	0,13	0,29	0,33	0,41	0,56
λ_M	0,07	0,22	0,25	0,33	0,50

The neck scaling factors from the Hybrid III 3 years old dummy to the Q3 dummy are summarized in Table 7

Table 7 : Neck scaling factors from the Hybrid III 3 years old dummy to the Q3 dummy

	Q1
λ_F	1,2
λ_M	1,5

The neck scaling factors from the Q dummies of different ages to the Q3 dummy are summarized in the Hybrid III midsize adult male dummy Table 8

Table 8 : Neck scaling factors from the Q dummies of different ages to the Q3 dummy

	Q0	Q1	Q1.5	Q3	Q6
λ_F	3,12	1,42	1,25	1	0,74
λ_M	4,76	1,49	1,30	1	0,67

2.1.2.3. Thorax

<u>Peak sternal deflection due to shoulder belt loading:</u>

The rib is represented as a bending beam:

The moment applied to the rib is:

$M=Fy/4$ (1)

Where F is the force and y is the rib length

The rib failure stress is:

$\sigma_b=Mc/I$ (2)

Where c is the distance to neutral fibre and I is the inertial moment

The rib deflection is

$\delta=Fy^3/(48E_bI)$ (3)

Where E_b is bone modulus

The combined equations 1, 2 and 3 give:

$\delta=\sigma_b\, y^2/(12cE_b)$ (4)

As it was assumed that $\lambda_{\sigma b} = \lambda_{\sigma t}$ the rib deflection ration is:

$\boxed{\lambda_{\delta}= \lambda_y\, \lambda_{\sigma t}/\lambda_{Eb}}$

<u>Peak sternal deflection due to bag loading:</u>

Thoracic organ stress is:

$\sigma= \delta/xE$ (1)

Where

δ is the deflection

x is the toracic depth
E is the thoracic organ modulus, it is assumed to be independent of age: $\lambda_E=1$

The deflection ratio is:

$$\boxed{\lambda_\delta = \lambda_x \lambda_{\sigma t}}$$

Peak acceleration:

The definition of acceleration factor is based on:

$$F=m\gamma \qquad (1)$$

$$F=\sigma S \qquad (2)$$

Where:

F is the force applied on the thorax
m is the thoracic mass
γ is the acceleration of the center of gravity of the head
σ is the rib stress
S is the thoracic cross sectional area

Equations 1 and 2 can be combined to give the acceleration:

$$\gamma=\sigma S/m \qquad (3)$$

So acceleration ratio is:

$$\lambda_\gamma = \lambda_\sigma \lambda_S / \lambda_m \qquad (4)$$

Since child and adults thorax were assumed to be of equal density $\lambda_m = \lambda_x \lambda_y \lambda_z$, where x is the torso depth, y is the torso width and z is the torso height

Surface ratio is $\lambda_S=\lambda_y\lambda_z$

As it was assumed that the variations with age of the bone and of the calcaneal tendon are the same $\lambda_\sigma = \lambda_{\sigma t}$

So the acceleration ratio is:

$$\boxed{\lambda_\gamma = \lambda_{\sigma t}/\lambda_x} \qquad (5)$$

The thorax scaling factors from the Hybrid III midsize adult male dummy to the Q dummies are summarized in Table 9

Table 9 : Thorax scaling factors from the Hybrid III midsize adult male dummy to the Q dummies

	Q0	Q1	Q1.5	Q3	Q6
$\lambda_{\delta\ belt}$	0,84	1,03	0,98	0,93	0,84
$\lambda_{\delta\ bag}$	0,20	0,33	0,36	0,44	0,56
λ_γ	1,8	1,50	1,51	1,58	1,63

The thorax scaling factors from the Q6 dummy to the Q3 dummy are summarized in the Hybrid III midsize adult male dummy Table 10.

Table 10 : Thorax scaling factors from the Q6to the Q3 dummy

	Q3	Q6
$\lambda_{\delta\ belt}$	1	1,1

2.2. CHILD project method

2.2.1. Data

The data used to develop the injury criteria were drawn from CHILD and CREST cases that had been validated by the both projects. The validation process of the reconstructions was an in-depth comparison of the reconstructions and the real world accidents including vehicle internal and external deformations, child restraint systems deformation and evidence of occupant kinematics. Around 70 cases were validated in this way. Initially reconstructions were performed with P dummies. These P dummies measures were not taken into account in the analysis except for the P1 ½ which is much closer to a Q dummy of that size. This process resulted in some 40 cases being available for the analysis for Q0, Q1, Q3, Q6 and P1 ½ dummies in frontal impacts with head, neck, thorax, abdomen, pelvis and lumbar spine measures.

2.2.2. Data analysis

The methodology used to develop the injury criteria was to compare the injuries observed in the real world accidents with the validated crash reconstruction dummy measurements. As the reconstructions were performed on dummies ranged from 0 to 6 years old, all data were scaled to a given age. The scaling methodology was the one proposed by Mertz (2003) and already described in the part 2.1 of this document, but instead of scaling adult data to child data, all age child data were scaled to a child given age. If the sample was considered big enough then injury risk curves were constructed by Certainty Method and Logistic Regression

3. INURY CRITERIA

3.1. HEAD

The existing EEVC adult head injury criteria are the Head Injury Criteria HIC 36ms=1000 and the acceleration 3ms=80g. These values scaled to the Q3 correspond to HIC 36ms=710 and acceleration 3ms = 75 g.

3.1.1. Head injury criteria issued from scaling adult data

The head reference data are the mid-size adult injury criteria reported in the Injury Assessment Reference Values (IARVs). They are defined for use with the Hybrid III midsize adult male dummy (Mertz 2003).
Two criteria are used to assess the severity of head injuries: the Head Injury Criterion (HIC) and the resultant peak acceleration of the center of gravity of the head. HIC value is referred to as 15ms HIC. The 15ms HIC Injury Assessment Criteria limit is 700 for the midsize adult male. It corresponds to 5% risk of skull fracture and to 5% risk of AIS≥4 brain injury (Prasad and Mertz 1985, Mertz et al.1996). The peak resultant acceleration Injury Assessment Criteria is 180 G for the midsize adult male, which corresponds to 5% risk of skull fracture (Mertz et al.1996).

3.1.2. Head injury criteria issued from the CHILD project

The head data were drawn from around 40 dummies. The real world accident injuries were directly paired with head linear accelerations and HIC 15ms values. Data were scaled in order to correspond to the 3 years old equivalent value. Table 5 gives the head scaling factors for the 3 years old child from other age Q-dummies. In the Child database there are very few cases AIS≥4 and very few cases with skull fracture. Head injury risk curves for 3ms acceleration and HIC 15ms were constructed with certainty method and logistic regression.

3.1.3. Comparison between results issued from both methods

3.1.3.1. HIC 15ms

Figure 1 Child data points and AIS4+ injury risk curves for Q3 Head Injury Criteria 15ms

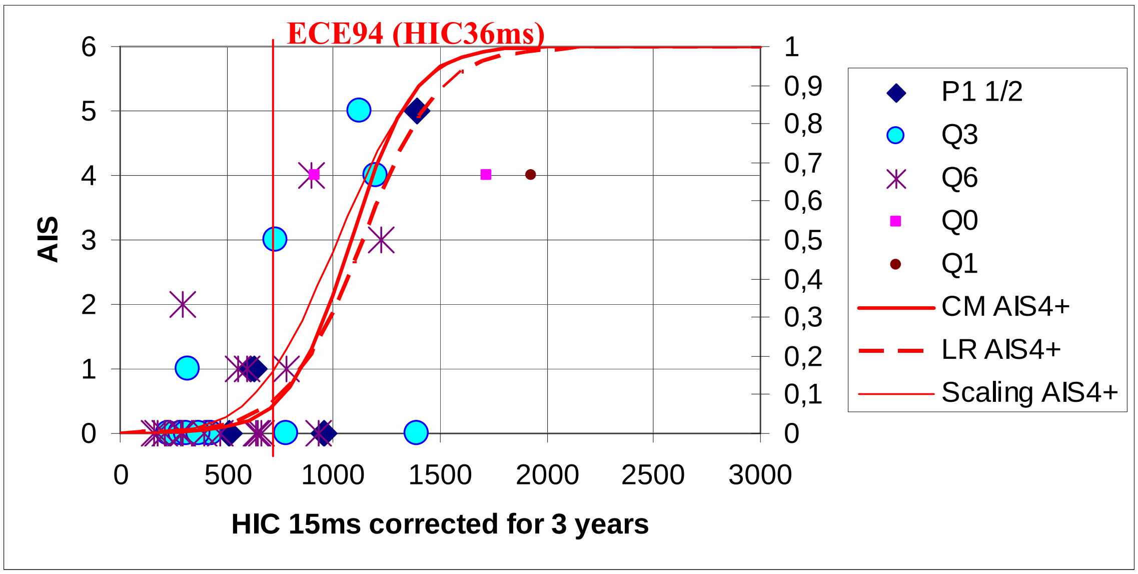


Figure 2 Child data points and AIS3+ injury risk curves for Q3 Head Injury Criteria 15ms

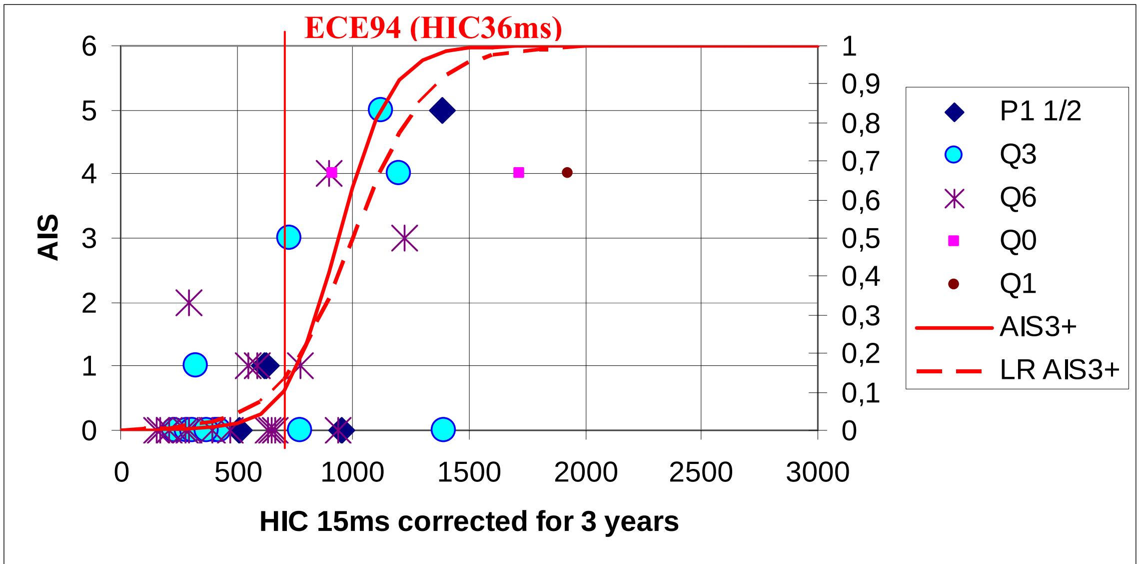
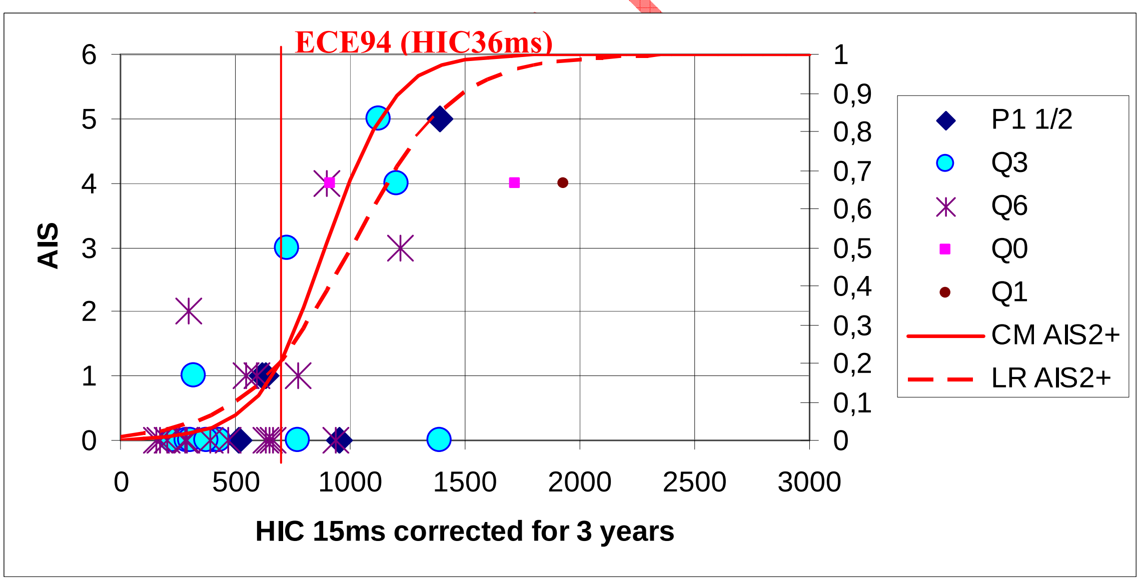


Figure 3 Child data points and AIS3+ injury risk curves for Q3 Head Injury Criteria 15ms



The HIC 15ms difference observed between the AIS4+ injury risk curve issued from scaling adult data and the curves issued from the Child project is about 100 between 0 and 50% of risk. As the AIS4+ injury risk curves issued from the Child project were constructed with few data no conclusion is possible comparing both methods. For AIS4+, AIS3+, AIS2+ the logistic regression and certainty method give similar injury risk curves between 0 and 50% of risk. AIS3+ which corresponds to a sever injury is the best injury threshold. Therefore the HIC 15ms values proposed for injury risk are AIS3+ 20% and 50% of risk (Table 11)

Table 11 : Q3 Head AIS3+ injury risk

HIC 15ms	20%	50%
Calculated with Certainty method	790	940
Calculated with Logistic regression	780	1000

ECE94 scaled
HIC36ms=710

3.1.3.2. Head 3ms acceleration

Figure 4 Child data points and AIS4+ injury risk curves for Q3 Head Acceleration 3ms

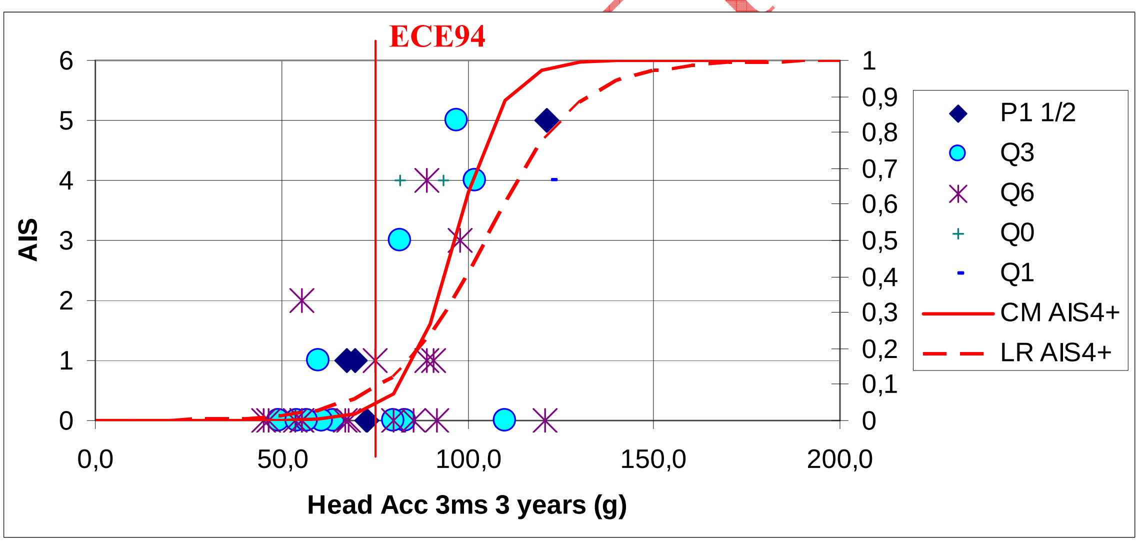


Figure 5 Child data points and AIS3+ injury risk curves for Q3 Head Acceleration 3ms

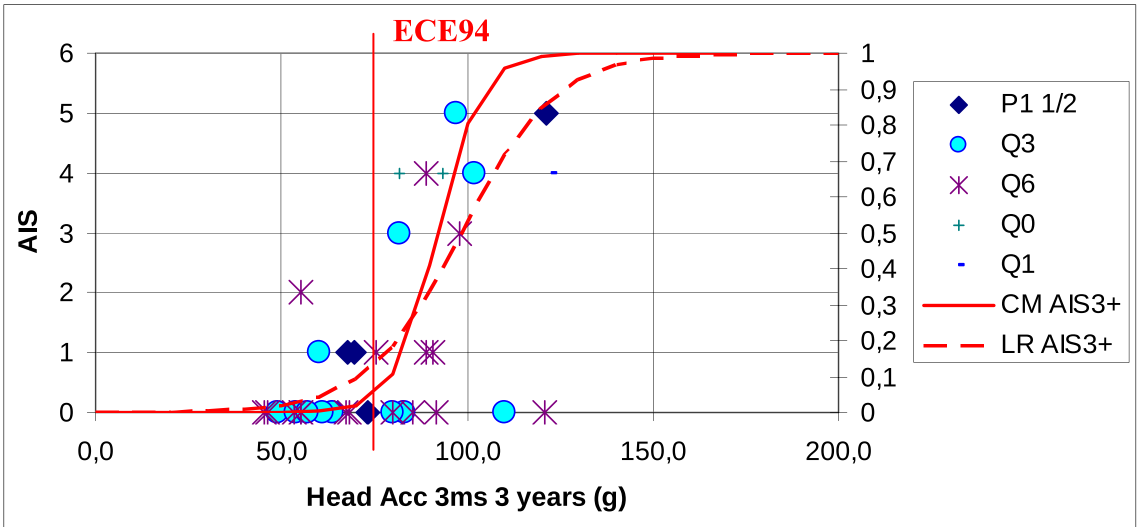
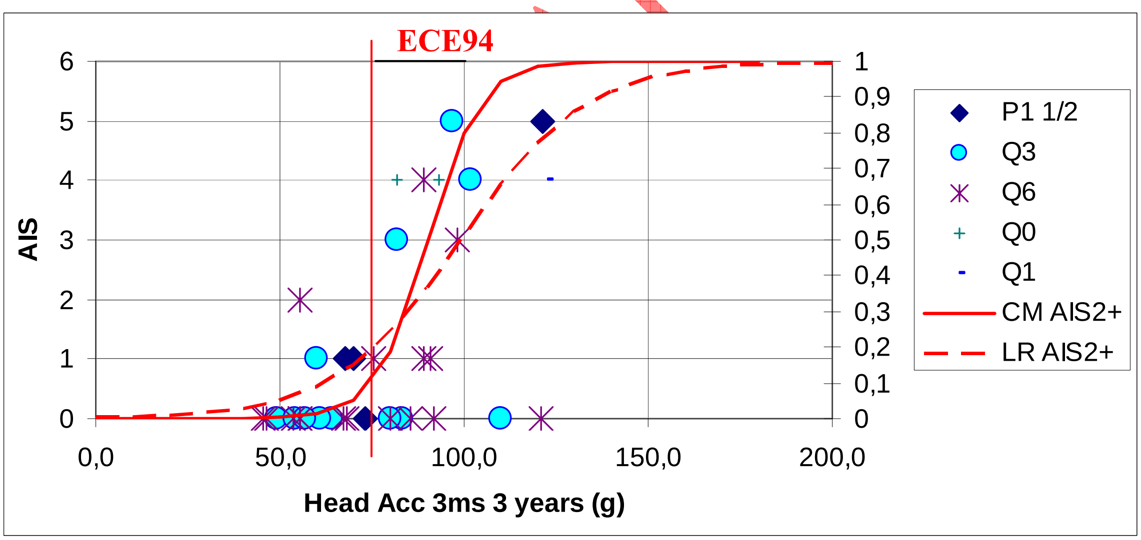


Figure 6 Child data points and AIS2+ injury risk curves for Q3 Head Acceleration 3ms



No injury head acceleration 3ms data is available in literature, therefore we just can compare the injury risk curves issued from the Child project and performed with the Certainty Method and the Logistic Regression. The injury risk curves are quite similar between 0 and 50% of risk. AIS3+ which corresponds to a sever injury is the best injury threshold. The acceleration 3ms values proposed for the head injury risk are AIS3+ 20% and 50% of risk (Table 12).

Table 12 : Q3 Head AIS3+ injury risk

Acceleration 3ms	20%	50%
Calculated with Certainty method	84g	92g
Calculated with Logistic regression	81g	99g

ECE94 scaled
75g

3.2. Neck

The existing EEVC adult neck injury criteria are:

- the tension force Fz<3,3kN at 0ms, Fz<2,9kN at 35ms, Fz<1,1kN at 60ms,
- the shearing force Fx<3,1kN at 0ms, Fx<1,5kN between 25 and 35ms, Fx<1,1kN at 60ms
- the flexion moment My<190Nm

These values scaled to the Q3 correspond to :

- the tension force Fz<1,35kN at 0ms, Fz<1,2kN at 35ms, Fz<0,45kN at 60ms,
- the shearing force Fx<1,27kN at 0ms, Fx<0,6kN between 25 and 35ms, Fx<0,45kN at 60ms
- the flexion moment My<63Nm

3.2.1. Neck injury criteria issued from scaling adult and child data

For in-position testing the neck injury criteria are the peak values of the axial forces (tension and compression), the bending moments (extension, flexion and lateral flexion) (Mertz 2003). Peak tension and peak extension moment are based on animal testing paired with a 3 year old child dummy and correspond to 3% for the tension and to 5% for extension moment of AIS3+ injury risk. Peak flexion moment and peak compression are based on volunteer testing (Mertz and Patrick 1967, 1971), and non-injurious accident reconstructions (Mertz et al 1978, Nyquist et al 1980). These values correspond to a AIS3+ injury risk inferior to 5%.

3.2.2. Neck injury criteria issued from the CHILD project

The neck data are drawn from around 40 dummies in frontal crash. The method is a detailed analysis of the real world accident neck injuries and mechanisms in order to associate good physical parameters to each kind of injury. The physical parameters are the shearing and traction forces, and flexion moment. Data are scaled in order to correspond to the 3 years old equivalent value. Table 8 gives the neck scaling factors, force and moment factors, for the three years old child. There are very few cases with injuries for each parameter, and not enough to enable the construction of injury risk curves.

3.2.3. Comparison between results issued from both methods

Figure 7 Child data points and AIS3+ injury risk curve for Q3 neck tension

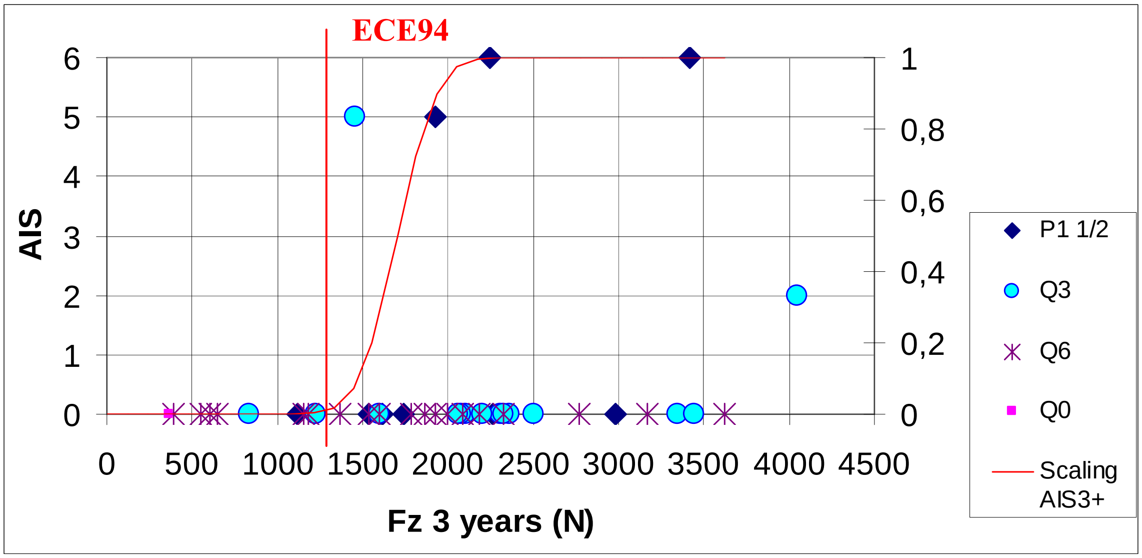
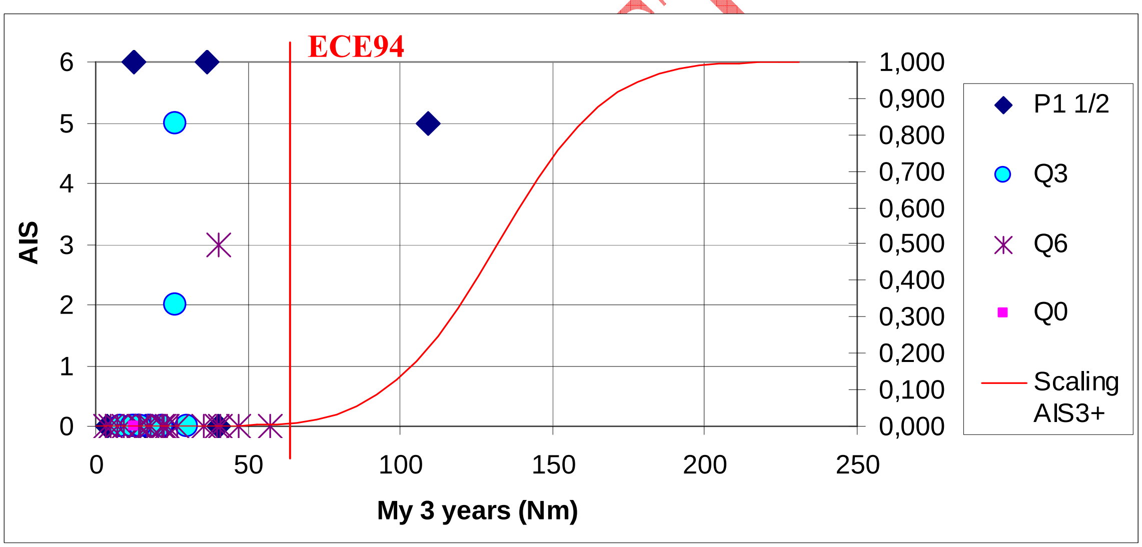


Figure 8 Child data points and AIS3+ injury risk curve for Q3 neck flexion moment



The comparison between both methods is possible only for tension and flexion moment but there are not enough injury cases to do an accurate comparison. For neck tension the scaled injury risk curve seems coherent with the Child data. No neck injury is observed below 1450N of traction force in the Child database and the scaled AIS3+ injury risk curve indicates a 3% risk for a 1220N tension (Table 13). As far as the flexion moment is concerned there is no coherence between the Child injury data and the scaled injury risk curve. Therefore the proposed neck injury risk values are tension values (Table 13):

Table 13 : Q3 Neck AIS3+ injury risk

	3%	20%	50%	Child project First injury	ECE 94 scaled
Fz issued from scaling	1220N	1555N	1705N	1457N	1350N
My issued from scaling		106Nm	129Nm	13Nm	63Nm

3.3. Thorax

The existing EEVC adult thorax injury criteria are the chest deflection d=50mm and the deflection rate VC=1m/s. These values scaled to the Q3 correspond to d=46,5mm and VC=1m/s

3.3.1. Thorax injury criteria issued from scaling adult and child data

In frontal impact the thorax injury criteria are the peak sternal deflection, the peak sternal deflection rate and the peak thoracic spine acceleration.

The predominant thorax injury in the AIS≥3 data-base is the rib fracture. However because of the low elastic modulus of their ribs, children can undergo large sternal deflections without rib fractures but with organ injury. The risk of AIS≥4 thoracic organ injury, particularly heart injury, must be taken into account.

- Peak sternal deflection due to shoulder belt loading:

The sternal deflection risk curve of AIS≥3 was defined (Mertz et al 1991) for the 3-point-belt restrained midsize male Hybrid III dummy. The IARV of 50 mm sternal deflection due to belt loading corresponds to 50% risk thorax injury AIS≥3.

- Peak sternal deflection due to airbag loading:

Mertz et al (1997) have published an injury risk curve for AIS≥4 thoracic injury. These curves, based on cadaver impact data (Kroell et al 1972 et 1974), are defined for sternal deflection due to a distributed loading. The IARV of 64,3 mm sternal deflection due to distributed loading corresponds to 5% risk thorax injury AIS≥4.

- Peak sternal deflection rate:

The injury risk curve for AIS≥4 thoracic injury based on sternal deflection was developed using the animal and the GM 3-year old dummy data from Mertz et al (1997). Because of behavior differences between the GM dummy and the 3 years old Q-dummy, the injury risk curve defined on the GM dummy should not be used directly for the Q-dummy family. No peak sternal deflection rate based on adult testing exists for adults.

- Peak thoracic spine acceleration:

The spine acceleration provides an assessment of how well the restraint loads are balanced between the neck, lumbar spine, clavicles, ribs and internal thoracic organs (Mertz 2003). Therefore, to limit the distortion between these segments the limit thoracic acceleration for the Hybrid III midsize male dummy was defined as 60g (Mertz 1984).

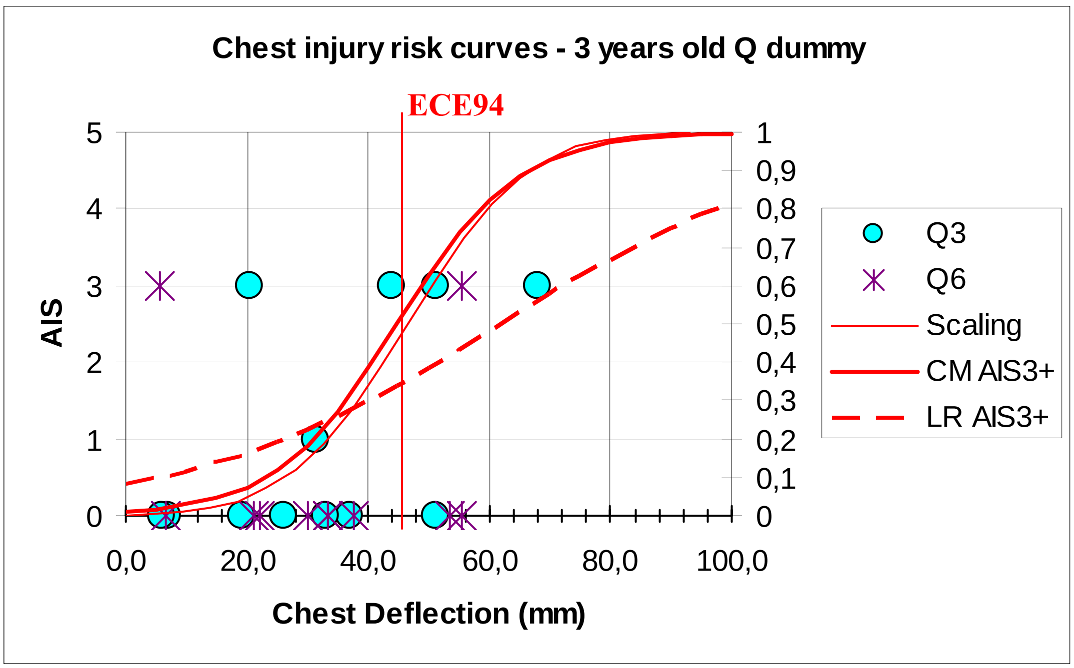
This value was based on the results of volunteer tests (Stapp 1970, Mertz and Gadd 1971).

3.3.2. Thorax injury criteria issued from the CHILD project

The chest data were drawn from 24 dummies. The real world accident injuries were directly paired with the deflection dynamic measurements acquired with the Q3 and the Q6 dummies. Data were scaled in order to correspond to the 3 years old equivalent value. Table 10 gives the chest scaling factors for the Q3 dummy from the Q6 dummy. Chest injury risk curves were constructed with certainty method and logistic regression.

3.3.3. Comparison between results issued from both methods

Figure 9 Child data points and AIS3+ injury risk curve for Q3 chest deflection



The comparison is possible only for peak sternal deflection due to shoulder belt. A good match is observed in the curve issued from scaling and the curve issued from the Child database calculated with Certainty Method. Therefore the chest deflection values proposed for the thorax injury risk are 20% and 50% of AIS3+ injury risk (Table 14):

Table 14 : Q3 Chest AIS3+ injury risk

Chest deflection	20%	50%
Issued from scaling	33mm	46mm
Issued from the Child database	31mm	45mm

ECE94 scaled
46,5mm

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