Enabling Protection for Older Children

SEVENTH FRAMEWORK PROGRAMME
THEME 7
Transport (including AERONAUTICS)

EPOCH 218744

FINAL PROJECT REPORT

Work Package 1
Task 3
The development of injury risk functions

by J A Carroll and M Pitcher
EPOCh 218744

FINAL PROJECT REPORT

The development of injury risk functions

Injury risk functions and limits for older children

by J A Carroll and M Pitcher

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Executive summary

The implementation of Directive 2003/20/EC means that children aged 3 or more years old and up to 150 cm in height must use a child restraint appropriate to their size when travelling in cars or goods vehicles fitted with seat-belts. The affect of this legislation has led to children remaining in child restraints until they are older (up to 12 years old, depending on their height).

The concept of the EPOCh project is to drive the improvement of safety for older children travelling in vehicles. To enable this, the EPOCh project will produce a 10/12 year old prototype dummy.

Once a new dummy is available, it is then necessary to specify injury criteria and accepted thresholds for these criteria which are appropriate to that age and size of occupant. Unlike the adult situation, there is very little biomechanical data from which specific injury risk functions for children can be derived. 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. However, such an approach should use the final production version of a dummy and is not feasible during the dummy development phase. As an alternative, the investigation reported here has scaled adult injury risk functions in an attempt to make them relevant for the older child dummy.

This task within the EPOCh project therefore sought to bring together the most recent knowledge and techniques on the development of injury risk functions through the scaling of adult data. This task also included a review of the material properties information available for children. Using preferred scaling techniques and the most up to date material properties, methods of calculating injury risk functions for the older child are recommended. Scaling was then applied to generate risk curves or injury threshold values for the following parameters:

- Linear head acceleration
- Neck tension, anterior-posterior bending, and shear force
- Chest compression

The input data and scaled products were compared against the limited information on child injury tolerance in the literature that could be used to check the proposed tolerance values.

Recommendations have also been given as to ways of providing better scaling results should this task be revisited in the future. These recommendations typically involve further consideration of the particular issue, and always suggest the value of having additional relevant biomechanical data.
1 Introduction

The implementation of Directive 2003/20/EC, dated 8 April 2003 (which amends Directive 91/671/EEC), means that children aged 3 or more years old and up to 150 cm in height must use a child restraint appropriate to their size when travelling in cars or goods vehicles fitted with seat-belts. The affect of this legislation has led to children remaining in child restraints until they are older (up to 12 years old, depending on their height).

The concept of the EPOCh project is to drive the improvement of safety for older children travelling in vehicles. This is to be enabled by providing a measurement tool (dummy) that represents an older child. This could then be used in child safety protocols and test procedures to account for older children. To date, European regulatory child restraint system testing has relied on the P series child dummies as restraint loading devices. The EC projects CREST and CHILD have worked on the development of the “Q” series of dummies, which have been recommended for use in the NPACS assessment programme. However, NPACS is unable to provide an assessment for children over the age of 6 years as there is currently no Q series older child (10/12 year) dummy, and there is no older child dummy appropriate for use in side impact assessments. This means that consumers will not be able to make informed decisions on the purchase of CRSs and hence provide the best protection for their older children, which is out of step with the recent changes in legislation. The Q dummies could, in the future, also replace the P-series dummies in Regulatory testing if the family included a 10/12 year old dummy. On this basis, the EPOCh project will produce a 10/12 year old prototype dummy.

Once a new dummy is available, it is then necessary to specify injury criteria and accepted thresholds for these criteria which are appropriate to that age and size of occupant. The injury thresholds (and the risk functions from which the thresholds are selected) provide the linkage between what the dummy measures and the meaning of that measurement in terms of risk of injury for a human in an equivalent impact event.

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 (PMHSs). 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. However, unlike the adult situation, there is very little biomechanical data from which specific injury risk functions for children could be derived. As alternatives, two approaches have been used recently (EEVC, 2008):

1. Perform accident reconstructions using the child dummy under development. This approach has a few key limitations, preventing its use in this project:
   a. It can sometimes be difficult to obtain sufficient case information from retrospective accident investigations to recreate the case accurately. With expert interpretation of accident case notes and a sufficient number of reconstructions, the significance of this inherent inaccuracy can be minimised. However;
   b. Accident reconstructions are expensive as they will almost certainly require a full-scale test with at least one vehicle.
   c. The prototype dummy would need to be available for the test work, limiting when the risk functions could be made available and extending the time before the dummy could be used with accompanying injury risk functions.
   d. Any modifications to the performance of the dummy would require the accident reconstruction tests to be repeated.
2. Scale adult injury risk functions and/or criteria to be relevant to the child size (dummy) being investigated.
   a. This is the approach described within this report.
   b. It is also subject to a major limitation – that the risk functions are not tailored to the dummy being developed. Any deviation in dummy performance that is not represented in the scaling process will cause an inaccuracy in the injury assessments.

This Task within the EPOCh project sought to bring together the most recent knowledge and techniques on the development of injury risk functions through the scaling of adult data. Traditionally techniques used to scale human response data from adult to child sizes have been limited by a paucity of biomechanical information regarding children. Therefore, this task also included a review of the material properties information available for children. Using preferred scaling techniques and the most up to date material properties, methods of calculating injury risk functions for the older child are recommended. Where possible and necessary, a set of injury criteria are suggested that could be used in association with the measurement tool developed in Work Package 2 of the EPOCh project.

However, it must be noted that the derivation of injury criteria for use with the dummy assumes that the dummy biofidelity will be comparable (with respect to the biofidelity targets) with the existing dummies being used as the basis for the scaling. Where human property values are used, there is a similar assumption that the dummy will have perfect biofidelity. The extent to which the dummy performance matches these assumptions can only be determined later. Deviation from the stated assumptions will reduce the accuracy and appropriateness of the injury criteria proposed within this report.

1.1 Objectives

Based on the approach described above, the specific objectives of this EPOCh project task (Task 1.3) were to:

1. Review the published literature to obtain the latest biomechanical information pertaining to injury risk assessment for children
2. Develop scaling factors
3. Develop risk functions (and injury criteria), where there is sufficient data (or an existing risk curve) to support a scaled function
2 Method

2.1 Task 1.1 conclusions

Task 1.1 involved a literature review, accident analysis and injury mechanisms identification. The latest information relating to the injuries received by older children in car accidents was reviewed. The focus was on the key injury mechanisms and the measurement capabilities needed by a dummy that represents children of this size.

In particular, the report looks at how and where these older children are being injured whilst travelling in vehicles. It establishes the main priorities for the body areas that need to be protected by restraint systems and will therefore feed into the identification of requirements for a measurement tool (i.e. dummy). The study took the following approach:

I. Review the latest literature and current research relating to older children;
II. Review accident data, relating to older children, from the previous work of CREST, CHILD, NPACS, EEVC etc. relating specifically to older children (where available);
III. Identification of injury priorities and loading conditions;
IV. Summarise injury mechanisms and prioritise for front and side impacts.

The conclusions from this report were:

- The implementation of EU Directive 2003/30/EC means the new dummy needs to be capable of sufficiently assessing a child restraint designed for a child up to 150cm or 12 years old.
- Although the majority of research of injuries sustained by children using booster restraints is limited to children under 6 years old there are comparisons with injuries sustained by children only using the adult seat belt.
- For front and side impact the main injury body region is the head. In both types of accident, injury is caused due to contact with an external rigid object. It is therefore important that the exposure risk of the head is minimised. This would mean a short excursion in front impact and good head containment in side impact.
- The abdomen and chest are the next most significant body regions to protect as this is where the majority of a child’s vital organs are located. In a front impact it is important that the child does not submarine under the lap belt. In a side impact it is important that the child restraint provides side protection from the door panel or an intruding object.
- The pelvis has also been identified as an area to protect in side impact, as again it is important the child restraint provides protection from the door panel or an intruding object.
- Limb injuries occur frequently in both front and side impacts, however they have previously been classed as low priority as they are deemed to be low in severity and difficult for the dummy to measure.

It is therefore important that the dummy has the appropriate sensors in these highlighted areas. However there are additional body regions that it would be beneficial if they were instrumented. A full list of the required instrumentation is outlined in the following section.
2.2 List of channels required

To limit the scope of the scaling that may be required, it is important to understand which parameters may need to be considered in determining injury risk. This will be determined by the sensors and potential measurements that can be recorded by the dummy during an impact. In turn the requirements for the instrumentation will be based on a review of the injuries sustained by older children and the feasibility of incorporating appropriate sensors in the dummy. Within EPOCh Task 1.2, Deliverable D1.2, a list of instrumentation required for use with the older child dummy is provided (Waagmeester et al., 2009). A summary of these channels is provided below, starting with the measurements that are considered to be key requirements:

- Head accelerations (Ax, Ay, Az)
- Upper Neck forces (Fx, Fy, Fz)
- Lower Neck forces (Fx, Fy, Fz)
- Upper Neck moments (Mx, My, Mz)
- Lower Neck moments (Mx, My, Mz)
- Thorax Spine accelerations (Ax, Ay, Az)
- Thorax 'ribcage' deflections Dx (front impact) or Dy (side impact)
- Pelvis accelerations (Ax, Ay, Az)

Other channels that are also available are:

- Head angular velocities/accelerations (ωx, ωy, ωz, ẍ, ÿ, ẑ)
- Thoracic ‘ribcage’ accelerations (Ax, Ay)
- Lower Lumbar Spine forces (Fx, Fy, Fz)
- Lower Lumbar Spine moments (Mx, My, Mz)
- Pelvis accelerations (Ax, Ay, Az)
- Pelvis angular velocities/accelerations (ωx, ωy, ωz, ẍ, ÿ, ẑ)

Other channels that may also need to be considered are:

- Abdominal penetration (internal pressure and/or external force distribution)
- Shoulder loads (side impact)
- Pubic forces
- Femur loads

It should be noted that this list of potential dummy measurements are not all applicable to every testing scenario. For example some of these measurements, would not be applicable to sled testing but would become important in accident reconstructions, where interaction with the vehicle interior may occur.

Outputs from each of the measurements listed above may require some scaling in order to convey meaningful injury risk data. The required scaling formulae, anthropometric parameters, material property values and underlying injury risk data that may be needed in the scaling process are listed in Section 1 of the report.

Based on the formulae presented below, it should be possible to provide scaling strategies for most of the channels listed here. However, certain parameters are not covered by the formulae (e.g. the lumbar spine and pelvis).

The 'other channels for consideration' (Section 2.1) were not investigated any further as they were not considered to be priorities for use with the EPOCh older child dummy.
3 Results

3.1 Scaling equations and formulae

Previously, many authors have published techniques for scaling biomechanical measurements to different sizes of subject. While the general principle behind the scaling remains consistent, each of the publications seems to adopt different specific detail. For instance, slight differences in the formulae used by each author can be observed alongside differences in the material properties considered to be the most appropriate. Also different authors may be considering slightly different injury priorities, when developing their scaling strategy, for example the inclusion or exclusion of skull fracture may affect the scaling relationship.

The following section, and the equivalent section for each body region component of the report, compares the equations used by either:

- van Ratingen et al. (1997), in the development of biofidelity requirements for the Q3 dummy
- NHTSA (1996), in the development of injury assessment values for child dummies
- Irwin and Mertz (1997), in their biomechanical basis for the CRABI and Hybrid III child dummy families
- Mertz et al. (2003), in the development of injury assessment reference values for frontal and side impacts
- EEVC (2008), output from the joint Working Group (WG) 12 and 18 activity reviewing the Q-dummy family of dummies, based largely on the work done in the CREST and CHILD project (Palisson et al., 2007)

These reference sources are grouped according to the research group represented by the authors:

1. van Ratingen et al. (1997) wrote to support the development of the Q-dummies. This formed part of the international child dummy working group activities at the time.
2. The work of Irwin and Mertz, and Mertz et al. assisting the work of the SAE (Society of Automotive Engineers) CRABI and Hybrid III dummy family task groups and the introduction of the CRABI and Hybrid III child dummy families in US regulation.
3. The EEVC report documents the European (European Enhanced Vehicle-safety Committee) position on the Q-dummies. However, through interaction with members of other international working groups, the authors collaborated with those involved in the NHTSA child dummy work.

Where a scaling ratio is specified, these should be considered as the value for the biomechanical size to which the scaling is being applied, divided by the value from the size where the tolerance information is known. For scaling from an adult to a child this means the child value divided by the adult.
3.2 Head

Head acceleration

van Ratingen et al.

\[ R_a = \frac{a}{a'} = \left( \frac{R_k}{R_{\lambda_k}} \right)^{1/3} \]

where \( R_a \) is the scaling ratio for the head acceleration; 
\( R_k \) is the ratio of head linear stiffness; and 
\( R_{\lambda_k} \) is the ratio of head mass.

Irwin and Mertz

\[ R_a = \frac{\lambda_E}{\lambda_x} \]

Where \( \lambda_E \) is the ratio of the elastic moduli of bone; and \( \lambda_x \) is the ratio of head length.

EEVC WG12&18

\[ R_a = \frac{\lambda_{\sigma_t}}{\lambda_x} \]

Where \( \lambda_{\sigma_t} \) is the ratio of calcaneal tendon failure stress; and \( \lambda_x \) is the ratio of head length.

The scaling ratio used for head acceleration differs between that published by van Ratingen et al. (1997) and that used by Irwin and Mertz (1997) and the EEVC (2008). Van Ratingen et al. assumed that the head impact test condition could be represented as a single spring-mass system, whereas the EEVC equated the force applied to the head in terms of Newton’s second law (\( F=ma \)) and the failure stress (\( F=\sigma S \), where \( \sigma \) is the failure stress and \( S \) is the cross-sectional area). Both of these approaches seem reasonable in principle; noting that one would need to consider the failure stress behaviour under dynamic conditions, rather than quasi-static, to match with the motion implied by the head acceleration. However, the approaches will give different scaling values.

Also, there is the issue as to where the head compression stiffness data will come from. Irwin and Mertz (1997) and van Ratingen et al. (1997) used cranial bone bending stiffness data. However, Irwin and Mertz (1997) demonstrated that head stiffness was dominated by the bulk modulus of the brain. This agrees with the finding from the modelling work conducted by Coats et al. (2007), which identified that the compressibility of the brain significantly affected bulk modulus and varied the principal stress in the skull. Ideally, the ratio of head compression stiffness would, therefore, reflect the bulk modulus of the brain. The bulk modulus for adult brain tissue has been reported in the literature; however, an appropriate value for children has not been determined empirically, as far as could be determined in the present study. In the absence of a bulk modulus scaling ratio, Mertz et al. (2003) and the EEVC (2008) both used calcaneal tendon failure stress as a proxy for the brain failure property. The appropriateness of using either the elastic modulus of cranial bone or tendon failure stress is discussed further later in the report.

HIC

Mertz et al.

\[ \Delta_{HIC} = \frac{\lambda_{\sigma_t}}{\lambda_x} \]

where \( \lambda_{HIC} \) is the ratio of HIC values; 
\( \lambda_{\sigma_t} \) is the ratio of calcaneal tendon failure stress; and 
\( \lambda_x \) is the ratio of sums of head circumference.

EEVC WG12&18

\[ \Delta_{HIC} = \left( \frac{\lambda_{\sigma_t}}{\lambda_x} \right)^{\frac{3}{2}} \]

where \( \lambda_{\sigma_t} \) is the ratio of calcaneal tendon failure stress; and 
\( \lambda_x \) is the ratio of head length.
Both Mertz et al. and the EEVC proposed a formula with which to scale HIC values. These formulae are similar in that they both include a term for the failure stress and another for the ratio of head dimensions. They differ in the power to which these terms are raised and it is not clear which solution offers the best approach. The differences between approaches are discussed further in the results and discussion sections of this report.

**Head rotation**

Only NHTSA (1996) have published formulae considering the scaling of rotational motion parameters. These were developed to relate primate test results to human equivalents, and were originally developed by other authors (in the 1970s and '80s). The 'm' and 'p' subscripts in the following table refer to values for man or the primate, respectively.

\[
\dot{\theta}_m = \dot{\theta}_p \left( \frac{1}{\lambda_m} \right)^\beta_m
\]

Where \( \dot{\theta} \) is the angular acceleration of the adult or child; \( \dot{\theta} \) is the angular velocity of the adult or child; \( \lambda_m \) is the head mass scaling ratio (human value divided by primate, or child divided by adult); and \( \lambda_L \) is the head length scaling ratio.

### 3.2.1 Geometric (anthropometric) measurements

The head anthropometry measurements related to the heads of young children, Q10 (10.5 year-old), Q12 (11.6 year-old) and adults are shown in Table 3.1. These values have been taken from the EPOCh Task 1.2, Deliverable 1.2 (Waagmeester et al., 2009) and are based on measurements in the CANDAT anthropometry database.

<table>
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<tr>
<th>Measurement</th>
<th>6 year old</th>
<th>Q10 (10.5 year old)</th>
<th>Q12 (11.6 year old)</th>
<th>Adult</th>
</tr>
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<tr>
<td>Head mass (kg)</td>
<td>3.38</td>
<td>3.59</td>
<td>3.68</td>
<td>4.55</td>
</tr>
<tr>
<td>Head length (mm)</td>
<td>183.0</td>
<td>187.4</td>
<td>188.4</td>
<td>195.6</td>
</tr>
<tr>
<td>Head width (mm)</td>
<td>138.0</td>
<td>143.9</td>
<td>145.1</td>
<td>155.0</td>
</tr>
<tr>
<td>Head depth (mm)</td>
<td>189.0</td>
<td>202.2</td>
<td>204.7</td>
<td>221.0</td>
</tr>
</tbody>
</table>

### 3.2.2 Material property values

This section reviews the material property data that could be used in the scaling equations shown in Section 3.2.

#### 3.2.2.1 Skull

In the study reported by Hubbard (1971), eight beam samples were cut from the layered parietal bone of four embalmed calvaria (the upper part of the skull – 'skullcap').
Symmetrical three-point bending tests were then performed on the samples. It is assumed that the calvaria were taken from adult subjects. However, this is not stated explicitly by Hubbard. It ought to be noted that the use of embalmed specimens is likely to have an effect on the measured bending response of the bone. Also it is assumed that the tests were conducted under quasi-static conditions. Material properties are likely to be different when tested at higher strain rates, more representative of the conditions experienced in an impact event.

Hubbard presents flexure characteristics of layered cranial bone from these tests. The characteristics include: Bending stiffness, elastic modulus, shear stiffness and shear modulus.

Note: Hubbard presents values for the elastic moduli from each of the tests (mean 9.7 GPa), however, Irwin and Mertz (1997) take the mean theoretical value which Hubbard calculated using layered beam theory (9.9 GPa). It is proposed that adopting the experimental value of 9.7 GPa would be more consistent with the data from the Hubbard test work.

The study by McPherson and Kriewall (1980) examined the elastic modulus of foetal cranial bone as determined by three-point bending tests. Bending tests were performed on 86 specimens obtained from six foetal calvaria. In addition, 12 full-section specimens from the parietal bone of a six-year-old child were tested for comparative purposes. The foetal cranial bone was obtained from six subjects ranging in estimated gestational age from 25 to 40 weeks. After pre-conditioning the specimens were loaded to a midspan deflection of 1.5 mm at a speed of 0.5 mm.min⁻¹ and then unloaded at the same speed. Therefore, these conditions are essentially quasi-static and comparable with the conditions assumed to have been used by Hubbard. However, again it should be noted that the biomechanical response under higher strain rates may be different.

McPherson and Kriewall fitted a first order least-squares curve to the experimental data (elastic modulus plotted against estimated gestational age). They found a moderate coefficient of determination (r-squared value) existed for both parallel fibre orientation (r = 0.74) and perpendicular fibres (r = 0.76). This seems to show an increase in the elastic modulus with increasing gestational age for either fibre orientation. The implication of this is that measurements taken in frontal, lateral, or head impacts to the top of the head potentially need different scaling.

As described above, the elastic bending moduli of bone for children and adults are published by McPherson and Kriewall (1980) and Hubbard (1971). Irwin and Mertz (1997) used a cubic spline fit (see Figure 3.1) between these data to estimate the elastic moduli for the bending of bone for the 6 month, 12 month, 18 month, 3 year old and 6 year old children and the adult. Irwin and Mertz assumed that a person’s bone elastic modulus would reach maturity (it's adult value) at between 18 and 22 years of age. After this point the modulus would remain constant until older age. They found that adjusting the age at which adulthood was reached, within the 18 to 22 range, didn't change the elastic modulus for younger children. Generally in this report it has been assumed that the material properties of human tissues reach their adult values after 25 year. However, so as not to distort the curve proposed by Irwin and Mertz, the bone elastic modulus is considered to be constant for the ages from 20 to 25. On this basis, estimates of the elastic moduli for the bending of the parietal bones of children and the adult are given in Table 3.2.

Following the development by Irwin and Mertz, the relationship showing the development of bone elastic modulus (Figure 3.1) has been used widely in other publications considering scaling relationships. This relationship seems a reasonable way of describing the development of human material properties with age. However, it is subject to certain key limitations:
- The age of the adult subjects is not known
- The adult specimens were embalmed, as opposed to fresh-frozen
- The foetal age was approximated to 0 years by Irwin and Mertz
- The testing was conducted under quasi-static conditions, not at a strain rate representative of that developed during accidental impact events

Jans et al. (1998) conducted three-point bending tests on pieces of cranial bone taken from two children aged 7 and 11 months. Material property values from the two subjects were combined and for the elastic modulus were between 1.7 and 3.3 GPa, while yield stress was between 115 and 235 MPa. A point representing these elastic modulus data has been added to Figure 3.1.

Further three-point bending tests were conducted by Margulies and Thibault (2000) on human infant cranial bone specimens. The subjects ranged from 25 weeks gestation to six months of age. These data have also been plotted on Figure 3.1, and seem to be in reasonable agreement with the initial point used by Irwin and Mertz (1997). As an academic point, the pre-term data suggest a sharp initial rise in elastic modulus which should be included in future regression analyses. However, apart from determining a realistic age 0 intercept, this is unlikely to affect the older child elastic modulus significantly.

![Figure 3.1: Elastic modulus for parietal bone bending (Irwin and Mertz, 1997)](image-url)
Table 3.2: Estimated elastic bending moduli of bone for children and adults (Irwin and Mertz, 1997)

<table>
<thead>
<tr>
<th>Age Group</th>
<th>Elastic modulus (GPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>6-month-old</td>
<td>2.8</td>
</tr>
<tr>
<td>12-month-old</td>
<td>3.2</td>
</tr>
<tr>
<td>18-month-old</td>
<td>3.6</td>
</tr>
<tr>
<td>3-year-old</td>
<td>4.7</td>
</tr>
<tr>
<td>6-year-old</td>
<td>6.6</td>
</tr>
<tr>
<td>Adult</td>
<td>9.9</td>
</tr>
</tbody>
</table>

In the study reported by Thibault *et al.* (1999), experiments were conducted on human skull and suture tissue to characterise their age-dependent material properties. Subjects ranged from three months to seven years of age. Three-point bending specimens and tension specimens were obtained from normal parietal bones, occipital bones, coronal sutures, sagittal sutures, and metopic (frontal) sutures. Once obtained specimens were tested in either three-point bending or tension. Thibault *et al.* reported that human cranial bone is extremely compliant during the first few months of life. Comparing with other sources, they commented that the elastic modulus of cranial bone increases dramatically during pre-pubescent growth and development. No detailed results are presented in the paper by Thibault *et al.*, although there is a figure showing how the data compare with other published results (e.g. McPherson and Kriewall, 1980).

The relative influence of brain size and mechanical properties on paediatric rotational inertial brain injury was investigated by Prange *et al.* (1999). They used a shear testing device to determine the shear modulus of paediatric brain tissues samples taken from five day old piglets (equivalent to less than 1 month old human infant). The mechanical properties of brain tissue for infants and adults were found to be significantly different from each other. In extending previous work testing brain tissue at small strains (< 2.5 %), the large strain data of Prange *et al.* (up to 50 %) led to the suggestion that brain tissue is non-linear. Adult tissue is stiffer than paediatric tissue at very small strains, but not at large strains. These data demonstrated that the assumption used in scaling adult inertial head injury to paediatric head injury - that the material properties are identical - does not hold true. Prange *et al.* do not provide any comment as to the level of strain at which they would expect failure of brain tissue to occur.

These properties were then used in a finite element model of a paediatric brain and an adult brain. The numerical simulations indicated that the brain mechanical properties make an important contribution toward strain distribution and magnitude during rotational, inertial loading. Prange *et al.* suggest that the brain mass and material properties of an infant have protective effects during non-impact head injury, with both factors reducing the deformation of the infant brain compared with that of an adult experiencing the same rotational inertial load. Therefore, it is important to incorporate both of these parameters when using scaled adult injury data to predict paediatric head injuries. It is also expected that head size will be an important parameter and could be protective for children, when compared with an adult.

The notion that the tolerance of the paediatric head to brain injury from non-impact events may be different from that expected when based on geometrical scaling of the adult head alone seems to agree with work reported by Margulies *et al.* (1999). Margulies *et al.* conducted a few experimental tests using pig and piglet specimens. They observed less severe neural injury in the piglets compared with the pigs, when subjected to mechanically equivalent rotational loads. Margulies *et al.* suggested that this finding may be the result of developmental variations in geometry or constitutive properties, or tissue strain injury thresholds.
Mertz et al. (2001) used the curve, developed by Irwin and Mertz (1997), for the variation of the elastic modulus of cranial bone as a function of age to derive the value for a ten-year-old child (see Table 3.3). It is of note that the units in Mertz et al. (2001) decreased by a factor of $10^6$. It is assumed that this is a mistake, a typographical error, which should be read as GPa instead.

Table 3.3: Bone elastic moduli (Mertz et al., 2001)

<table>
<thead>
<tr>
<th>Bone elastic modulus (GPa)</th>
<th>6-months</th>
<th>12-months</th>
<th>18-months</th>
<th>3-years</th>
<th>6-years</th>
<th>10-years</th>
<th>Adult</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>2.8</td>
<td>3.2</td>
<td>3.6</td>
<td>4.7</td>
<td>6.6</td>
<td>8.45</td>
<td>9.7*</td>
</tr>
</tbody>
</table>

*9.7 Hubbard (1971)

To determine the bone elastic bending modulus for a Q10 (10.5 year-old) and Q12 (11.6 year-old), the existing data (Irwin and Mertz, 1997) were interpolated, as shown in Figure 3.2. It should be noted that 9.7 (Hubbard (1971)) has been used for the bone elastic bending modulus of the adult. This resulted in a bone elastic modulus of 8.49 GPa for the Q10 and 8.78 GPa for the Q12.

Figure 3.2: Elastic bending moduli of bone for children and adults

As a proxy for the failure stress of the brain, Mertz et al. (2003) used the failure stress of calcaneal tendon, as reported by Yamada (1970, see Figure 3.3). This figure, expanded to show only the tendon tensile strength up to the age of 25, is shown again in Figure 3.4. Also shown in Figure 3.4 are the interpolated points inferred by Mertz et al. (2003).
Palisson et al. (2007) also show ratios for the calcaneal tendon stiffness, reportedly taken from Yamada (1970). However, Yamada does not report calcaneal tendon stiffness, only failure stress and elongation. This is presented alongside a curve of the relationship between stress and elongation which is clearly not linear. Neither would it be a particularly safe assumption that this curve does not vary in shape, with age of subject.
Coats and Margulies (2003) tested porcine paediatric cranial tissue at higher rates of loading than the historical quasi-static loading. From the two speeds investigated by Coats and Margulies no statistically significant effect of rate for any of the material properties of paediatric cranial bone was shown. However, when these results were compared with previous studies at more diverse strain rates, Coats and Margulies found a dramatic decrease in elastic modulus over a broad range of test rates (over two orders of magnitude). This calls into question the relevance of determining a relationship between adult and child cranial elastic modulus in quasi-static loading conditions, when testing will occur at higher loading rates. The data generated by Coats and Margulies were also used to reinforce previous findings showing that cranial bone is stiffer than suture and that the ultimate strain of paediatric cranial bone is many fold higher than that of the adult.

Through testing human paediatric cranial bone samples in three-point bending using drop test apparatus, Coats and Margulies (2006) attempted to extend their findings from porcine specimen testing. The subjects ranged in age from preterm to one year of age. The data derived compliment previous research at low rates of loading (e.g. McPherson and Kriewall, 1980 and Margulies and Thibault, 2000). However, over the relatively small range of speeds investigated (1.2 to 2.8 m.s\(^{-1}\)), strain rate was not found to have a significant effect on the modulus and ultimate stress of cranial bone from donors less than or equal to 13 months old.

In addition, Coats and Margulies investigated differences in the material properties of cranial bone with location (either parietal or occipital). They found that parietal bone was significantly stiffer and had a higher ultimate stress than occipital cranial bone. This finding would have a direct implication only if impacts to the rear of the dummy head were being considered. However, it does support assertions that tolerance to head injury is specific to the impact point and direction of loading.

Coats and Margulies also reported material properties for human paediatric cranial suture. They found parietal bone to be more than 35 times stiffer than suture. The suture also experienced strains over 100 percent before failing, 30 times more than cranial bone. However, this is perhaps less relevant for the skulls of older children, as the sutures become increasing ossified throughout childhood.

Roth et al. (2007) compared skull geometry from scaled parameters with those measured directly. They found that the thickness both for the parietal and frontal areas of the skull were quite different between the anatomical studies and curves obtained with the scaling method. The bone thickness appears to change more than the scaling method reports. Roth et al. concluded that the scaling method cannot be applied for the modelling of children’s heads under the age of six mainly because of the discrepancies in skull height and thickness. A large variation in these parameters was observed until the age of ten, with a linear evolution of this parameter after this age.

The implication of this observation is that the scaling process may require a correction to account for the difference between scaled and empirical child head shape, dimension, and structure. Also, if the difference between scaled and observed skull thickness varies in a different way for different bones of the head, any correction would need to be impact-direction specific.

### 3.2.2.2 Whole head

Post-mortem human paediatric specimens were used by Prange et al. (2004) to determine the static and dynamic properties of the whole infant head. The biomechanical tests were conducted using three unembalmed fresh-frozen human infant specimens of ages one, three, and eleven days after birth.

A series of compression tests were conducted, using a parallel plate fixture, to determine static and rate-dependent properties of the infant head in two directions (anterior-posterior and right-left). A test was conducted at 0.05 mm.s\(^{-1}\), to determine the quasi-
static stiffness of the head; then three additional tests were conducted to the same 
displacement at constant velocities of approximately 1.0, 10, and 50 mm.s⁻¹. As the 
force-deflection responses were non-linear, for each test Prange et al. used regression 
between 50 percent and 100 percent maximum displacement in order to determine a 
stiffness value. Statistical analysis revealed that the linear stiffness was dependent on 
deflection velocity but had no directional dependence.

After the compression experiments were complete, the heads were dropped onto a flat 
smooth anvil. From this, force-time histories were recorded. Each head was impacted at 
approximately 15 cm and 30 cm drop heights, once on each of five locations: vertex, 
occiput, forehead, and right and left parietal bones. Prange et al. state that, “All tests 
were conducted non-destructively so that the head specimens could be used in 
subsequent analysis.” However, there is no quantitative verification that the head impact 
response was unaffected by repeat testing of the specimens.

The whole-head drop test data from Prange et al. should provide a useful reference for 
assessing surrogate head biofidelity. In this application the average head response from 
the five impact points has been used to compare and validate the expected responses 
based on the different scaling techniques (see Section 3.2).

3.2.2.3 Summary of available values

The following table (Table 3.4) summarises the limited potential material property values 
that could be used to scale head response parameters. It should be noted that the 
values reported by Irwin and Mertz (1997) were based on the adult data from Hubbard 
(1971) and the child data from McPherson and Kriewall (1980). Information from 
Jans et al. (1998), and Margulies and Thibault (2000) have not been added to the table 
as they would not be expected to alter the curve reported by Irwin and Mertz 
substantially. All of these data points are assumed to have been derived under quasi-
static loading conditions, not dynamic.

**Table 3.4: Summary of available material property values for use in scaling 
head response data**

<table>
<thead>
<tr>
<th>Measurement</th>
<th>Source</th>
<th>Younger child (10.5 year old)</th>
<th>Q10 (11.6 year old)</th>
<th>Adult (25 year old)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skull bone modulus</td>
<td>Irwin and Mertz (1997)</td>
<td>6.6 GPa (six-year-old)</td>
<td>8.49 GPa</td>
<td>9.7 GPa (25 year old)</td>
</tr>
<tr>
<td>Calcaneal tendon failure stress</td>
<td>Yamada (1970)</td>
<td>51.94 MPa (five-year-old)</td>
<td></td>
<td>54.88 MPa (from 15 to 45 years)</td>
</tr>
<tr>
<td>Whole head stiffness</td>
<td>Prange et al. (2004)</td>
<td>22 N.mm⁻¹ (newborn)</td>
<td></td>
<td>Forehead = 1,594 N.mm⁻¹</td>
</tr>
<tr>
<td></td>
<td>Hodgson et al. (1967)</td>
<td></td>
<td></td>
<td>Side of the head = 1,067 N.mm⁻¹</td>
</tr>
</tbody>
</table>


**3.2.3 Recommendations for injury criteria and thresholds**

3.2.3.1 Head acceleration

*Linear*

UNECE Regulation 94 and the Euro NCAP tests procedures for frontal and side impacts use an adult head injury performance criterion, based on the linear acceleration (exceeded for 3 ms). In regulation the limit is 80 g (UNECE, 1995), whilst in Euro NCAP the lower performance limit is set at 88 g, whilst the higher performance limit is 72 g (Euro NCAP, 2008). The Injury Assessment Reference Value proposed for use in the U.S. with the Hybrid III mid-size male is a peak resultant acceleration of 180 g (Mertz *et al.*, 2003). This relates to a five percent risk of skull fracture (EEVC, 2008). Both of these values can be scaled to generate criteria for an older child. The 180 g value comes from the work of Mertz *et al.* (1996). It should be noted that Mertz *et al.* did not use a statistical technique to derive the 180 g threshold.

It should be remembered that as a result of the scaling the Injury Assessment Reference Value may go up instead of down. A summary of the data obtained in 15 experimental animals, by Douglass *et al.* (1968), indicated that monkeys with large heads were easier to concuss than those with small heads. Prange *et al.* (1999) also reported protective effects of brain mass and material properties in infants.

As was noted in Section 3.2.2.2, Prange *et al.* (2004) conducted drop tests using newborn PMHS head specimens. Extrapolating the results obtained at 150 and 300 mm, using a power relationship, it suggests peak accelerations of 45 and 62 g for a newborn head when dropped from 200 mm or 376 mm, respectively (see Figure 3.5). These values were compared with the expected biofidelity test levels (83 to 141 g, for an adult head dropped laterally from 200 mm (Roberts *et al.*, 1991), and 117 to 201 g, for an adult head dropped from 376 mm onto the forehead) when scaled using each of the documented formulae. In each case, the material properties used by the original authors were used again. A graph showing the comparison is shown in Figure 3.5. It should be noted that whilst Prange *et al.* observed that peak head acceleration and HIC varied with drop height, they did not change between impact location (occiput, forehead, left and right parietal regions). Therefore the acceleration data are useful for comparisons with both the scaled frontal and lateral adult test levels.

It can be seen from Figure 3.5 that the extrapolation of the experimental results of Prange *et al.*, from newborn specimens, agree most closely with the biofidelity targets scaled using the Irwin and Mertz relationship. The results from Prange *et al.* pass through the lateral head drop requirements, based on the Irwin and Mertz scaling. This is particularly interesting as the EEVC, 2008 (and Mertz *et al.*, 2003) relationship uses the same formula as Irwin and Mertz, but with calcaneal failure stress substituted for the cranium bending modulus property data. Clearly this has not improved the relationship between the adult and child head acceleration response.
Prange et al. (2004) also reported stiffness values for the newborn PMHS heads they tested, as was recorded in Section 3.2.2.2. By comparing the stiffness value of the newborn child with the adult stiffness reported by Hodgson et al. (1967), a new stiffness ratio was obtained. However, the newborn stiffness seems very small compared with the adult. This may be explained by consideration of differences between the two experimental set-ups. To confirm the experimental findings, it is hoped that more closely comparable test data, to either that of Prange et al. or Hodgson et al., could be provided.

Despite these concerns over the similitude of the test conditions, when using this new stiffness ratio in the van Ratingen et al. scaling formula, a better agreement with the newborn drop test data is produced. This is shown in Figure 3.6. Predicting the correct head impact response for the newborn in no way guarantees that the head impact response for other ages of children can be predicted accurately also. However, the most biomechanical data is available for the newborn and thus validation at this age provides a good starting point.
Based on the comparison of scaling techniques shown in Figure 3.6, it is recommended that head injury risk data should be scaled using the updated van Ratingen et al. relationship. The following section describes the application of the updated van Ratingen et al. relationship to the head injury data in order to generate head injury risk functions for older children (10 or 12 year olds). For comparison, the original van Ratingen et al. and Mertz et al. relationships have also been applied in this manner.

The static bone stiffness of the head is required for use in the scaling calculation. Two different approaches were compared to calculate the bone stiffness for the Q10 (10.5 year old) and Q12 (11.6 year old).

The first method was to interpolate between the bone stiffness data points provided by Prange et al. (2004); for an adult (1594 N.mm⁻¹) and a new born (22 N.mm⁻¹). The relationship (curve) of the bone elastic bending modulus (Mertz et al., 2001) was applied to calculate the equivalent bone stiffness (See Table 3.5).

**Table 3.5: Bone elastic bending and stiffness moduli**

<table>
<thead>
<tr>
<th>Age (years)</th>
<th>0</th>
<th>0.5</th>
<th>1</th>
<th>1.5</th>
<th>3</th>
<th>6</th>
<th>10</th>
<th>10.5</th>
<th>11.6</th>
<th>Adult</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bone Modulus (GPa)</td>
<td>2.41</td>
<td>2.80</td>
<td>3.20</td>
<td>3.60</td>
<td>4.70</td>
<td>6.60</td>
<td>8.45</td>
<td>8.49</td>
<td>8.78</td>
<td>9.7</td>
</tr>
<tr>
<td>Stiffness (N.mm⁻¹)</td>
<td>22</td>
<td>106</td>
<td>192</td>
<td>279</td>
<td>516</td>
<td>926</td>
<td>1324</td>
<td>1332</td>
<td>1396</td>
<td>1594</td>
</tr>
</tbody>
</table>

The second method for calculating the bone stiffness of the head was to factor the bone modulus by the area it is applied over (as was the approach used by van Ratingen et al., 1997) e.g.:  

![Figure 3.6: Comparison of measured peak head acceleration values with updated scaled biofidelity targets](image-url)
The bone stiffness calculated by both methods was then used in the van Ratingen et al. calculation (2.2.1) for the head risk curve factor. This calculation showed that use of the interpolated bone stiffness from an adult and newborn (Prange et al., 2004), produced a scaling ratio that was very close to 1, (0.991 for the Q10 and 1.004 for the Q12). The implication of this is that the tolerance of the older child’s head is very close to that for the adult. Based on this information the adult head injury risk function can be used. The following figure (Figure 3.7) shows the probability of skull fracture based on average acceleration drawn using a Probit logistic regression analysis. The data supporting this function are those as used by Prasad and Mertz (1985) in the development of the HIC. The data suggest that a 20 and 50 percent risk of skull fracture injury would occur with head accelerations of 75 and 244 g, respectively.

It should be noted that this is not the same function as was proposed by Prasad and Mertz. Therefore, adopting this injury risk function (as shown in Figure 3.7) would be a deviation from what has been adopted for use with other dummies. Despite the statistical validity of this function, for consistency with previous dummy limits it would be more appropriate to adopt the existing adult head acceleration limit(s).

![Figure 3.7: Head injury risk curve for an adult based on peak acceleration](image)

The van Ratingen calculation using the bone stiffness interpolated from data points provided by Prange et al. (2004) produced scaling factors of 1.029 for the Q10 and 1.041 for the Q12.

The Mertz calculation (Section 3.2) was also used to calculate a head risk curve factor for the Q10 and Q12 using the bone elastic moduli as calculated previously. This calculation produced head risk curve factors of 0.913 for the Q10 and 0.940 for the Q12.
It should be noted that in the recent EEVC (2008) scaling of injury assessment values, the authors included an additional term to describe the ratio of the loading expected to bring about tissue failure in the adult or child. This was based on the calcaneal tendon failure stress values reported by Yamada (1970). The application of such a term seems appropriate, in that the tolerance of a child’s brain to external forces may not be as high as the tolerance of an adult’s brain. For this reason it is sensible to be conservative when specifying injury risk functions for children. However, the relevance of calcaneal tendon failure properties to those of the brain is not clear. In the absence of more closely related material property data, it was decided to adopt a similar term in this analysis. On closer inspection it became apparent that the ratio of calcaneal tendon failure strength for a ten year old compared with an adult was 0.98, based on the Yamada data. It would be even closer to one for a 12 year old. For this reason, the effect of material property failure strength was assumed to be approximately equal to one.

The head injury scaling factors were then applied to data from Mertz et al. (1996) to generate head injury risk curves for the Q10 and Q12 (Figure 3.8 & Figure 3.9, respectively). The differences between the scaled curves seem negligible for low risks of injury, when compared with the width of the confidence interval.

Figure 3.8: Head injury risk curves for the Q10 based on peak acceleration
So far all methods used to calculate injury thresholds for the Q10 and Q12 have scaled down the injury risk from adult data. For comparison, an alternative method was used, which involved scaling up from the 3 year old (Q3) 3 ms head acceleration threshold values reported by the EEVC (2008). The van Ratingen scaling method using the stiffness values from Prange *et al.* (2004) and Hodgson *et al.* (1967) (interpolated with a curve of the shape reported by Mertz *et al.* {2001}) was used to calculate the head acceleration ratio for the Q10 and Q12 (Table 3.6). These ratios were then used to scale the EEVC 3 year old head injury thresholds to obtain values for the Q10 and Q12 (Table 3.7).

**Table 3.6: Q10 and Q12 head data**

<table>
<thead>
<tr>
<th></th>
<th>3 year old</th>
<th>Q10</th>
<th>Q12</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stiffness (N.mm⁻¹)</td>
<td>516</td>
<td>1332</td>
<td>1396</td>
</tr>
<tr>
<td>Head Mass (kg)</td>
<td>2.84</td>
<td>3.59</td>
<td>3.68</td>
</tr>
<tr>
<td>Ratio</td>
<td>1</td>
<td>1.43</td>
<td>1.44</td>
</tr>
</tbody>
</table>
The resulting head injury (AIS 3+) values from scaling up the EEVC Q3 thresholds are not directly comparable with the values derived from scaling down the adult head injury risk curve. The adult head injury risk is for an unspecified severity of injury and relates to the peak acceleration value rather than the value exceeded for 3 ms. The EEVC 20 percent risk of AIS 3+ head injury coincides with a 116 or 117 g, 3 ms exceedance; whereas the adult risk curve suggests a peak value of 75 g for a 20 percent risk of head injury. This suggests that the 75 g peak value is overly conservative. This should be taken into account when selecting thresholds for use with the older child dummy.

The 3 ms head acceleration values scaled up from the three year old (as shown in Table 3.7) can be compared with the adult value of 80 g, used in Regulation (R94). However, the risk of injury associated with the adult 80 g limit is not known.

Rotational

According to Prange et al. (1999) their research provides a foundation for the study of the unique aetiology and pathophysiology of paediatric head injury. One may infer from this that they recommend direct determination of rotational inertial head injury risk, rather than scaling from adult information.

For reference, the adult tolerance to peak rotational acceleration lies somewhere in the range from 1,500 rad.s\(^{-2}\), for concussion, to 16,000 rad.s\(^{-2}\) for diffuse axonal injury (Darvish and Crandall, 2002). Based on the work of Prange et al. (1999) and Margulies et al., (1999) it was suggested that the head size, brain mass and material properties of children may have a protective effect against head injury. Therefore, the adult thresholds could be adopted as conservative estimates until age-specific tolerance values can be obtained for rotational acceleration of the head.

### 3.2.3.2 Head Injury Criterion (HIC)

The HIC was derived from the short duration part of the Wayne State Tolerance Curve, which was based on six data points representing the relationship between head acceleration level and effective impulse duration found to produce linear skull fractures in cadaveric skulls. The curve was extended to longer durations with comparative animal, PMHS, and human volunteer data (Prasad and Mertz, 1985). No brain damage or skull fracture data existed where the HIC duration was greater than 15 ms. If HIC was to be used as an injury criterion, it should be limited to a duration representative of the conditions expected to cause the head injury of interest.

An analysis of field accident data by ISO Working Group 6 indicated that there were no cases of brain injuries to the three-point belt restrained occupants whose head did not impact forward interior components (Prasad and Mertz, 1985). Therefore, HIC was not developed or suggested for use in non-head impact cases. All of the biomechanical data used in the development of the HIC were head contact related. As such the suitability of HIC for non-impact injury assessment may be, and frequently is, questioned. Therefore it may only be applicable to use HIC in side impact sled testing and real world vehicle testing, where head contact is likely to occur.
Whilst HIC was principally developed to investigate the risk of skull fracture, Prasad and Mertz (1985) reported on the use of HIC in predicting brain injury. A limitation of the injury data is noted in that since only the arterial system was pressurised for the original head impact tests, damage to the venous system was not measured. Therefore venous ruptures, which result in subdural haematomas (AIS 4) and are a major cause of death in patients with a severe head injury, would not have been recorded. Neither would it have been possible using these test methods to detect diffuse axonal injuries due to brain cell damage that may result in concussion (AIS 2 to 5), or other brain injuries like cerebral oedema and swelling. On the basis that the absence of an arterial rupture does not signify the absence of brain injury, the resulting HIC risk curve may substantially underestimate the risk of brain injury for a given HIC. Therefore the appropriateness of using the HIC for determining brain injury risk should be considered, critically, before it is used for that purpose. Also, a specific brain injury HIC tolerance criterion would be needed.

Despite these limitations, HIC values related to the risk of skull fracture are presented below for the Q10 and Q12 based on the expanded head impact dataset reported by Mertz et al. (1996). It should be noted that the risk curves documented here are based on Probit logistic regression analyses. This differs from the calculation technique used by Prasad and Mertz (1985) and then Mertz et al. (1996) in that it is a statistical technique. As such, the relationship between the HIC values and skull fracture predicted by the logistic regression is limited by the supporting data.

Two methods of scaling the HIC values were available, that from Mertz et al. (2003) and the updated formula from the EEVC WGs 12 and 18 (EEVC, 2008). The Mertz et al. (2003) formula was used to produce HIC injury risk curves relating to skull fracture for the Q10 and Q12 (Figure 3.10 & Figure 3.11). It should be noted that this calculation produces scaling ratios that are greater than the adult and therefore would imply that the HIC injury tolerance would peak at around 12 and then decrease for an adult. The 20% risk of injury values for both the Q10 and Q12 are negative, and even taking into account the confidence limits the HIC risk functions have a substantial intercept (risk of injury) for a HIC of 0. This questions the reliability of these curves. The HIC 50% risk of injury is 1725 and 1729 for the Q10 and Q12.

The alternative EEVC method produces scaling ratios for the HIC that are very close to one and therefore it can be assumed the adult risk values can be used. The adult has a 50% risk of injury at a HIC value of 1642.
Figure 3.10: HIC injury risk curves for the Q10

Figure 3.11: HIC injury risk curves for the Q12
A comparison was also conducted to scale up from the Q3 HIC injury values (EEVC, 2008) (See Table 3.8). Both the Mertz et al. (2003) and EEVC (2008) scaling ratios produced higher HIC values, for a given risk of injury, for the Q10 or Q12 compared with the Q3. They suggest that a 20 percent risk of skull fracture would be associated with a HIC of around 1000 and that a 50 percent risk of injury would be associated with a HIC value somewhere between 1250 and 1350. These 50 percent risk of skull fracture values are substantially smaller than those suggested based on the scaling of the adult values and are outside of the proposed 95th percentile confidence limits. Assuming that the adult HIC values are correct, the lower values from the Q3 scaling suggest that after 10 to 12 years, older children still have a large change to undergo with respect to skull fracture tolerance (based on the HIC). However, it is not a safe assumption that the adult HIC values are correct. The Probit risk curves shown above do not appear to be wholly reliable. Equally, the Q3 risk curve may have some uncertainties associated with it.

Table 3.8: HIC injury data scaled from 3 year old

<table>
<thead>
<tr>
<th></th>
<th>Q3</th>
<th>Q10</th>
<th>Q12</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mertz et al.</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>20% (g)</td>
<td>790</td>
<td>986.88</td>
<td>989.18</td>
</tr>
<tr>
<td>50% (g)</td>
<td>1000</td>
<td>1249.21</td>
<td>1252.12</td>
</tr>
<tr>
<td>EEVC WG12&amp;18</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>20% (g)</td>
<td>790</td>
<td>1057.98</td>
<td>1063.64</td>
</tr>
<tr>
<td>50% (g)</td>
<td>1000</td>
<td>1339.22</td>
<td>1346.39</td>
</tr>
</tbody>
</table>

3.3 Neck

3.3.1 Scaling equations and formulae

As noted in Section 3.1, the scaling formulae used by van Ratingen et al. (1997), Irwin and Mertz (1997), Mertz et al. (2003), and the EEVC WGs 12 and 18 are being considered as previous examples of scaling relationships within this report. The formulae published by these authors with regard to neck loading measurements are shown below. Again, unless stated otherwise, it should be assumed that scaling ratios are the child value divided by the adult.

van Ratingen et al.

\[ R_T = \frac{b}{\lambda_x} = \frac{R_E}{\lambda_x}, \lambda_x = \frac{\lambda_x}{\lambda_x} \]

Irwin and Mertz

\[ R_M = \lambda_{\text{fr}}^2 \]

\[ \lambda_{\text{fr}} = \lambda_{\text{fr}} \]

EEVC WG12&18

\[ \lambda_{\text{fr}} = \lambda_{\text{fr}} \]

Where \( \lambda \) is the ratio of peak neck flexion moment; \( \lambda_{\text{fr}} \) is the ratio of calcaneal tendon failure stress; and
Where \( t_x \) and \( T_x \), or \( t_y \) and \( T_y \) are the resistive moment to neck bending for the adult (upper case) or child (lower case) about the x- or y-axis; 
\( R_E \) is the scaling factor for the elastic modulus; 
\( \lambda_x \) or \( \lambda_y \) are the length scale factors in the x- or y-axis.

\[ \lambda_F = \lambda_x A_E^2 \]

Where \( \lambda_F \) is the ratio of neck force, either compression or shear; 
\( \lambda_o \) is the ratio of calcaneal tendon failure stress; and 
\( \lambda_C \) is the ratio of neck circumferences.

For the neck, the authors of previous scaling studies agree on the terms to be used in the formulae for neck tension and bending. The tensile strength of the neck is governed by a ratio of soft tissue tensile strength (e.g. tendon failure strength) and the ratio of cross-sectional area. For the bending moment, an additional term is added to reflect the ratio of the moment arms.

### 3.3.2 Geometric (anthropometric) measurements

The neck anthropometry measurements related to the necks of young children, Q10, Q12, and adults are shown in Table 3.9. As with the head anthropometry, these values have been taken from the EPOCH Deliverable 1.2 (Waagmeester et al., 2009). The data for the child sizes are taken from the CANDAT anthropometry study (for the six and ten year olds), or are extrapolated from the CANDAT data (for the twelve year old). The adult measurements are taken from the Hybrid III 50th percentile adult dummy. These anthropometric measurements are plotted against age in Figure 3.12. It is interesting to see that whilst the neck depth for the Q12 is very close to that of the adult, neck width is still only 75 percent at twelve years. Indeed, from the curve shown by the neck width data, it appears as though there is a discontinuity between the child and adult measurements; and that the two will never meet. This is likely to be an issue brought about by using different sources of anthropometry for the children and adult. For instance, the child data is based on male and female subjects whereas the adult was chosen to be representative of the 50th percentile male driver, at the time of the Hybrid III design. Based on differences in the data such as this, the use of these data in the scaling process is unlikely to map the human development very closely. However, these are the measurements to which the dummies are being designed and therefore ought to be used to develop dummy-specific injury risk functions and thresholds. This assumes that the neck behaviour of the dummies depends on and varies with the neck geometry in a similar manner. If this is not the case then using these anthropometry data will lead to erroneous scaling of the risk functions and injury thresholds.

**Table 3.9: Geometric (anthropometric) measurements relating to the neck**

<table>
<thead>
<tr>
<th>Measurement</th>
<th>6 year old</th>
<th>Q10</th>
<th>Q12</th>
<th>Adult</th>
</tr>
</thead>
<tbody>
<tr>
<td>Neck circumference (mm)</td>
<td>260.0</td>
<td>288.4</td>
<td>295.1</td>
<td>383.0</td>
</tr>
<tr>
<td>Neck length (mm)</td>
<td>95.0</td>
<td>110.0</td>
<td>113.7</td>
<td>142.0</td>
</tr>
<tr>
<td>Neck width (mm)</td>
<td>78.0</td>
<td>85.6</td>
<td>87.5</td>
<td>118.0</td>
</tr>
<tr>
<td>Neck depth (mm)</td>
<td>75.0</td>
<td>81.0</td>
<td>82.8</td>
<td>84.8</td>
</tr>
</tbody>
</table>
3.3.3  Material property values

The child neck Injury Assessment Reference Values (IARVs) and the combined neck tension extension moment injury criterion used in the US are based principally on tests with anaesthetised pig subjects. These tests were replicated with a three-year-old dummy to develop injury risk functions and threshold values. One limitation noted with the porcine tests was the lack of muscle tensing. Test data with adult volunteers can provide some information on the relative contribution of muscle tensing to the biomechanical behaviour of the neck. Through the combination of the existing data sets on neck injury, Mertz and Prasad (2000) described improvements to the existing neck injury risk curves (tension, extension, and moment measurements) for all existing sizes of crash test dummies. In that paper Mertz and Prasad state that using those curves, the risk of AIS ≥ 3 neck injury can be estimated for any size and age of occupant provided the neck circumference, ligamentous failure stress and percent of maximum muscle tone are prescribed.

The exact bases for the adult neck injury criteria in use within UNECE Regulation 94 and Euro NCAP are not well understood at this time. It was considered beyond the scope of this study to determine the underpinnings of these criteria. As such, the exact contribution of muscle tone to the threshold values is uncertain. With this in mind, it was decided not to modify any of the neck scaling relationships, in the way that Mertz and Prasad had, to account for muscle tensing. In the future, with a detailed investigation of the data underlying the Regulation 94 and Euro NCAP limits, it would be possible to amend the scaling factors in this way, if necessary and appropriate.

Pintar et al. (2000) tested functional units from the cervical spine of several young caprine subjects. The average maximum tensile failure force from each of these goats is shown in Figure 3.13. This figure clearly shows an increase in failure force with increasing age. It also seems to start from some value above zero, at the newborn age and the increase in failure force appears to start levelling off between 12 and 20 years.
old. The bending stiffness for the cervical spine units were also reported by Pintar et al. and are shown in Figure 3.14, for flexion, extension, and right and left lateral bending.

Figure 3.13: Maximum tensile failure force for cervical spine units from caprine subjects, related to an equivalent human age (Pintar et al., 2000)

Figure 3.14: Bending stiffness for cervical spine units from caprine subjects, related to an equivalent human age (Pintar et al., 2000)
The study by Ching et al. (2001) examined the effect of spinal development on the tensile mechanics of isolated cervical functional spinal units (OC to C2, C3 to C4, C5 to C6, and C7 to T1) (Figure 3.15). Cadaveric baboon spines were selected for testing. The specimens spanned the age range from young paediatric to adult, resulting in an overall sample age range of two to 26 human equivalent years.

**Figure 3.15: Mean functional spinal unit tensile failure force and stiffness (Ching et al., 2001)**

Comparative figures of tensile failure force and stiffness for each of the functional spine units tested by Ching et al. and Pintar et al. are shown below (see Figure 3.16 to Figure 3.19). The scaling ratios relating child failure force and stiffness values to those for the adult are shown in Table 3.10.

**Figure 3.16: Tensile failure force for functional caprine spine units (Pintar et al., 2000)**

**Figure 3.17: Tensile failure force for functional baboon spine units (Ching et al., 2001)**
In 2002, Hilker et al. reported on additional adult caprine specimen tests. Based on the results conducted with small, medium, and large adult goats, Hilker et al. revised the scaling ratios developed for children. These revisions are shown in Table 3.10.

**Table 3.10: Scaling ratios for developing spinal tissues (Ching et al., 2001)**

<table>
<thead>
<tr>
<th>Age group (human equivalent years)</th>
<th>Pintar et al., 2000</th>
<th>Ching et al., 2001</th>
<th>Hilker et al., 2002</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Tensile failure force</td>
<td>Tensile stiffness</td>
<td>Tensile failure force</td>
</tr>
<tr>
<td>3</td>
<td>0.20</td>
<td>0.23</td>
<td>0.33</td>
</tr>
<tr>
<td></td>
<td>0.38</td>
<td>0.54</td>
<td>0.55</td>
</tr>
<tr>
<td>12</td>
<td>0.78</td>
<td>0.85</td>
<td>0.66</td>
</tr>
<tr>
<td><strong>Adult</strong></td>
<td><strong>1</strong></td>
<td><strong>1</strong></td>
<td><strong>1</strong></td>
</tr>
</tbody>
</table>

Based on testing of baboon specimens, Nuckley and Ching (2006) developed equations relating the tensile or compressive stiffness of functional cervical spine units to the age of the subject.

Following the work using caprine and baboon specimens, in 2008 Luck et al. (2008) conducted testing using paediatric PMHS. This testing matched that of the previous studies by investigating functional units of the cervical spine. The majority of the PMHS were perinatal or neonatal; however, seven specimens were paediatric (from 5 months to 168 months). The two oldest specimens were 108 and 168 months old. The results from the study were used to support the conclusion that previously published cervical spine stiffness data from juvenile animal models compare well to paediatric PMHS stiffness in the upper cervical spine. However, only limited data are reported for the two older subjects.

Ouyang et al. (2005) reported on bending and tensile testing of whole paediatric human cervical spine complexes. The ten donor subjects used by Ouyang et al. were aged between 2 and 12 years. All specimens were tested after removal of the neck musculature, which would cause reduction of the tensile failure load compared with the true *in vivo* situation.

Only average bending stiffness values are presented in the paper, which seem to agree with the young caprine data from Pintar et al. (2000). There is more information provided by Ouyang et al. on the ultimate tensile force and tensile stiffness results.
Ouyang et al. suggest a function relating ultimate tensile failure force to age of the form shown in Equation 3.1. In Equation 3.1, 'x' is age in years, and 'a' and 'b' are constants. The values proposed by Ouyang et al. were 372.7 and 72.9, respectively.

\[ F_{\text{failure}} = a \ln(x+1) + b \]  
Equation 3.1

The human tensile failure force data and trendline from Ouyang et al. (2005) are plotted in Figure 3.20 together with the caprine failure force data from Ching et al. (2001) and baboon data trendline from Nuckley and Ching (2006). From Figure 3.20 it is clear that the development of the human cervical spine does not seem to follow exactly the development of the animal surrogates. The human growth in strength appears to be substantially greater in the early years of childhood. It is important to remember that some differences may be expected between the human and animal datasets, as the animal specimens were tested as small functional units. The results from which were reported for each segment and as a mean from all tests. Ouyang et al. tested whole cervical spine specimens (from the head to the T2 vertebra). Also, the conversion of animal age to equivalent human age could introduce inaccuracies in the material property and age relationships. For example the linear animal age to human age relationship could be incorrect. More importantly for this scaling work, is that the relationship between animal and human development could be non-linear. That is, the animals may develop very rapidly shortly after birth, in comparison with human children.

![Figure 3.20: Comparison of caprine (Ching et al., 2001), baboon (Nuckley and Ching, 2006), and human (Ouyang et al., 2005) ultimate tensile failure force estimates](image)

In addition to the ultimate tensile force data, Ouyang et al. also reported that the average linear stiffness in tensile loading was 34.7 N.mm\(^{-1}\), with no statistically significant differences caused by age. On this basis, Ouyang et al. presented a mean stiffness response with plus and minus one standard deviation confidence boundaries. This could be used as a biofidelity response requirement for paediatric cervical spine behaviour.
However, the stiffness value of 34.7 N.mm\(^{-1}\) is substantially lower than has been found for adult PMHS tested under similar conditions, from which it could be inferred that some development from the young cervical spine stiffness does occur. This would call into question the appropriateness of using the Ouyang et al. stiffness response corridor for older children. Also, the finding of no statistical difference in tensile stiffness with variation in age contradicts the findings of previous authors based on animal testing. It is likely that the limited dataset analysed by Ouyang et al. could not provide the necessary resolving power to confirm such a difference. More human subject testing would be needed to confirm this suggestion.

3.3.3.1 Summary of available material property values

The following table (Table 3.11) shows the available and pertinent material property values for use in scaling the neck response parameters.

<table>
<thead>
<tr>
<th>Measurement</th>
<th>Source</th>
<th>Younger child</th>
<th>Q10</th>
<th>Q12</th>
<th>Adult</th>
</tr>
</thead>
<tbody>
<tr>
<td>Calcaneal tendon failure stress</td>
<td>Yamada (1970)</td>
<td></td>
<td>51.94 MPa</td>
<td>(five year old)</td>
<td>54.88 MPa (from 15 to 45 years)</td>
</tr>
<tr>
<td>Neck tensile strength</td>
<td>Ouyang et al. (2005)</td>
<td>798 N (six year old)</td>
<td>983 N</td>
<td>1017 N</td>
<td>1287 N (25 year old)</td>
</tr>
<tr>
<td>Neck compressive stiffness</td>
<td>Nuckley and Ching (2006)</td>
<td>5.16 MPa (six year old)</td>
<td>6.29 MPa</td>
<td>6.55 MPa</td>
<td>9.05 MPa (25 year old)</td>
</tr>
</tbody>
</table>

3.3.4 Recommendations for injury criteria

Adult upper neck injury limits for tension and shear force, and neck moment (flexion and extension) were taken from EEVC values used as the pass fail criteria for R94 and also the lower boundary for Euro NCAP. Mertz et al. (2003) proposed scaling values for similar time-dependent neck tension and extension functions. Therefore, such an approach has been adopted again. Mertz et al. (2003) scale the duration values in the time-dependent criteria. No reason for this is reported. The time scaling is proportional to the ratio of neck circumference. Why this should be the case is unclear. Scaling of the neck threshold time component may make the requirement more conservative. Therefore, until this issue is resolved it would be safer to consider the limits with time scaling. Exactly which criteria should be adopted for use with the older child dummy is recommended for future discussion. However, to enable ease of application both the time-scaled and not time-scaled thresholds will be produced in this report.

The work of Ouyang et al. (2005) provides an important source of information concerning the scaling of cervical spine tolerance to tensile loading. This ought to be incorporated in the tensile force criterion, and is used in the following analyses of scaling techniques. However, the limited dataset of Ouyang et al. did not have the statistical power to derive tensile stiffness changes with age (only failure force). Therefore, other sources of cervical spine stiffness ought to be used in setting biofidelity response requirements.

For compressive properties, the stiffness data of Nuckley and Ching (2006) could be incorporated.
The different approaches for calculating neck injury thresholds were evaluated. The Irwin and Mertz (1997) formula uses the ratio of neck circumference to calculate the neck force scaling ratio. The bending moment scaling ratio is calculated using the cross-sectional area of the neck and the neck depth perpendicular to the axis about which the bending is defined. The EEVC WGs 12 & 18 refined the Irwin and Mertz method to include the calcaneal tendon failure strength in both the force and moment scaling calculations (EEVC, 2008).

For comparative purposes, the equations for the force and moment modulus were also calculated using the stiffness data reported by Nuckley and Ching (2006) derived directly from experimental testing. Table 3.12 shows a comparison of the equation results. The formulae used to derive the scaling factors are shown in each case. The values used to derive the geometrical and material property scaling ratios are taken from Tables 3.9 and 3.11. For the Nuckley and Ching equation it should be noted that the equation only provides the stiffness value for a particular age. The scaling factor will be that result divided by the stiffness value for the age from which the response is being scaled. In preparing these values, an adult age of 25 has been used. Clearly, when the Nuckley and Ching, and Ouyang et al. formulae contain age-dependent terms, then the scaling ratios will also be dependent on the age at which adulthood is assumed to begin. It would be misleading, and give a false relationship, to assume that adulthood began at 55, for example, if the neck strength really reached an adult plateau level at 25. Therefore care must be taken when using these results to account for this age sensitivity and the potential inaccuracy that it introduces with assumed, rather than known, biomechanical development.

Table 3.12: Neck force and moment scaling factors (or ratios)

<table>
<thead>
<tr>
<th>Method</th>
<th>Formula</th>
<th>Q10 Ratio</th>
<th>Q12 Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Force</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Irwin &amp; Mertz</td>
<td>$\frac{\lambda_2}{\lambda_1} = \frac{d_2^2}{d_1^2}$</td>
<td>0.556</td>
<td>0.588</td>
</tr>
<tr>
<td>WG 12 &amp; 18</td>
<td>$\frac{\lambda_2}{\lambda_1} = \frac{A_2}{A_1} \frac{d_2}{d_1}$</td>
<td>0.679</td>
<td>0.717</td>
</tr>
</tbody>
</table>
| Nuckley and Ching| \begin{align*}
                  \lambda_2 & = \lambda_1 \frac{d_2}{d_1} \\
                  \lambda_2 & = \lambda_1 \frac{A_2}{A_1} \frac{d_2}{d_1}
\end{align*} & 0.695     | 0.723     |
| Ouyang et al.    | $F_{\text{failure}} = 372.7 \ln (\text{Age}+1) + 72.9$ | 0.763     | 0.790     |
| **Moment**       |                                      |           |           |
| Irwin & Mertz    | $\frac{R_2}{R_1} = \frac{d_2^2}{d_1^2}$ | 0.662     | 0.707     |
| WG 12 & 18       | $\frac{R_2}{R_1} = \frac{A_2}{A_1} \frac{d_2^2}{d_1^2}$ | 0.649     | 0.700     |
| Nuckley and Ching| \begin{align*}
                  R_2 & = \frac{\lambda_2}{\lambda_1} \frac{d_2^2}{d_1^2} \\
                  R_2 & = \frac{\lambda_2}{\lambda_1} \frac{A_2}{A_1} \frac{d_2^2}{d_1^2}
\end{align*} | 0.422     | 0.460     |

These calculated ratios were then used to scale the EEVC adult injury limits for neck tension and shearing force and extension and flexion moment. The calculated values are shown in Table 3.13.
Table 3.13: Neck injury values

<table>
<thead>
<tr>
<th>Method</th>
<th>Q10</th>
<th>Q12</th>
<th>Adult</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Tension (N)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Irwin &amp; Mertz</td>
<td>1834</td>
<td>1940</td>
<td>3300</td>
</tr>
<tr>
<td>WG 12 &amp; 18</td>
<td>2241</td>
<td>2365</td>
<td>3300</td>
</tr>
<tr>
<td>Nuckley and Ching</td>
<td>2294</td>
<td>2387</td>
<td>3300</td>
</tr>
<tr>
<td>Ouyang</td>
<td>2521</td>
<td>2608</td>
<td>3300</td>
</tr>
<tr>
<td><strong>Shear (N)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Irwin &amp; Mertz</td>
<td>1723</td>
<td>1822</td>
<td>3100</td>
</tr>
<tr>
<td>WG 12 &amp; 18</td>
<td>2105</td>
<td>2222</td>
<td>3100</td>
</tr>
<tr>
<td>Nuckley and Ching</td>
<td>2155</td>
<td>2242</td>
<td>3100</td>
</tr>
<tr>
<td><strong>Extension (Nm)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Irwin &amp; Mertz</td>
<td>38</td>
<td>40</td>
<td>57</td>
</tr>
<tr>
<td>WG 12 &amp; 18</td>
<td>37</td>
<td>40</td>
<td>57</td>
</tr>
<tr>
<td>Nuckley and Ching</td>
<td>24</td>
<td>26</td>
<td>57</td>
</tr>
<tr>
<td><strong>Flexion (Nm)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Irwin &amp; Mertz</td>
<td>126</td>
<td>134</td>
<td>190</td>
</tr>
<tr>
<td>WG 12 &amp; 18</td>
<td>123</td>
<td>133</td>
<td>190</td>
</tr>
<tr>
<td>Nuckley and Ching</td>
<td>80</td>
<td>87</td>
<td>190</td>
</tr>
</tbody>
</table>

The WG12 & 18 and Nuckley and Ching methods both produced very similar neck force injury values for both tension and shear for the Q10 and Q12. The Irwin and Mertz force calculation produces injury values slightly smaller in magnitude, this is because this method uses the neck circumference ratio and the adult neck circumference is significantly larger than the Q10 and Q12.

The neck injury values for moments generated by the three methods were not as similar as the force values. Experimental data from Carter et al. (2002) suggests that the adult value for flexion is in fact over estimated and the actual value should instead be closer to the neck injury extension value (57 Nm). Other than this, there is little information on which to base selection of one set of values over another. The Nuckley and Ching method produces the lowest and therefore hopefully the most conservative estimates of neck moment tolerance. However, these values would have to be considered against the feasibility of any test procedure that enforces such small values.

The EEVC adult limits use time-dependent criteria to determine a neck injury force limit corridor for both tension and shear (Table 3.14 and Table 3.15). The time points have also been scaled using the Mertz et al. (2003) technique, which factors the adult time points by the neck length ratio of child to the adult.
Table 3.14: Neck force injury tension limits

<table>
<thead>
<tr>
<th>Time (ms)</th>
<th>0</th>
<th>27</th>
<th>28</th>
<th>35</th>
<th>46</th>
<th>48</th>
<th>60</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adult (N)</td>
<td>3300</td>
<td>2900</td>
<td>1100</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Q10(N)</td>
<td>1834</td>
<td>1611</td>
<td>611</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Q12(N)</td>
<td>1940</td>
<td>1704</td>
<td>647</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 3.15: Neck force injury shear limits

<table>
<thead>
<tr>
<th>Time (ms)</th>
<th>0</th>
<th>19</th>
<th>20</th>
<th>25</th>
<th>27</th>
<th>28</th>
<th>35</th>
<th>46</th>
<th>48</th>
<th>60</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adult (N)</td>
<td>3100</td>
<td>1500</td>
<td>1500</td>
<td>1500</td>
<td>1500</td>
<td>1100</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Q10(N)</td>
<td>1723</td>
<td>834</td>
<td>834</td>
<td>834</td>
<td>834</td>
<td>611</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Q12(N)</td>
<td>1822</td>
<td>882</td>
<td>882</td>
<td>882</td>
<td>882</td>
<td>647</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

The time-dependent criteria for neck tension or shear, showing the time scaling, are plotted in Figures 3.21 and 3.22. The effect of using the neck anthropometry from Waagmeester et al. (2009) can be observed from these figures. The twelve year old limit is very close to that for the ten year old. It therefore appears that adult strength will not be reached until much later in life than has been suggested from other biomechanical information. Equivalent limit curves, but without the time scaling, are shown in Figures 3.23 and 3.24.
Figure 3.22: Neck shear injury corridor

Figure 3.23: Neck tension injury corridor no time shift
The injury thresholds calculated for the Q10 and Q12 have been scaled down from adult data. To validate this approach an alternative method was used, involving scaling up from the 3 year old. The EEVC WG 12 & 18 method was used to calculate the ratio of the Q10 and Q12 to the 3 year old (Table 3.16). These ratios were then used to scale the EEVC (2008) accident reconstruction 3 year old injury thresholds data, to calculate neck injury values for the Q10 and Q12 (Table 3.17).

Table 3.16: Q10 and Q12 neck data

<table>
<thead>
<tr>
<th></th>
<th>3 year old</th>
<th>Q10</th>
<th>Q12</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\lambda_x$</td>
<td>67.7</td>
<td>81.0</td>
<td>82.8</td>
</tr>
<tr>
<td>$\lambda_y$</td>
<td>72.6</td>
<td>85.6</td>
<td>87.5</td>
</tr>
<tr>
<td>$\lambda_{tot}$</td>
<td>0.83</td>
<td>0.98</td>
<td>0.99</td>
</tr>
<tr>
<td>$\lambda_F$</td>
<td>1</td>
<td>1.67</td>
<td>1.76</td>
</tr>
<tr>
<td>$\lambda_M$</td>
<td>1</td>
<td>1.99</td>
<td>2.15</td>
</tr>
</tbody>
</table>

Table 3.17: Neck injury data scaled from 3 year old

<table>
<thead>
<tr>
<th>Risk percentage and neck load parameter</th>
<th>3 year old</th>
<th>Q10</th>
<th>Q12</th>
</tr>
</thead>
<tbody>
<tr>
<td>20% Neck Tension Force (N)</td>
<td>1555</td>
<td>2590</td>
<td>2734</td>
</tr>
<tr>
<td>50% Neck Tension Force (N)</td>
<td>1705</td>
<td>2840</td>
<td>2998</td>
</tr>
<tr>
<td>20% Neck Flexion Moment (Nm)</td>
<td>79</td>
<td>157</td>
<td>170</td>
</tr>
</tbody>
</table>
Comparing the values between Tables 3.13 and 3.17 indicates that even the 20 percent risk of injury values scaled up from the three year old are greater than the values scaled down from the adult, using any method. However, it is again difficult to compare the values directly because the level of injury risk associated with the adult threshold values is unknown.

3.4 Thorax

3.4.1 Scaling equations and formulae

Irwin and Mertz

\[ R_d = \lambda_x \]

Where \( R_d \) is the ratio of chest deflections; and \( \lambda_x \) is the ratio of chest depth.

\[ \lambda = \frac{\lambda_s \lambda_{\text{st}}}{\lambda_{\text{Eb}}} \]

Where \( \lambda_s \) is the ratio of peak sternal deflection; \( \lambda_r \) is the ratio of rib length; \( \lambda_{\text{of}} \) is the ratio of calcaneal tendon failure stress; and \( \lambda_{\text{Eb}} \) is the ratio of bone modulus.

Assuming the thoracic organ modulus ratio, \( \lambda_{E} \), is equal to 1, then;

\[ R_d = \lambda_x \lambda_{\text{st}} \]

Further scaling formulae have been reported for impactor type testing of the shoulder, thorax, abdomen, and pelvis. Based on the expected use of the older child dummy developed within EPOCh, it is not likely that impactor type events will be used in evaluating injury risk. Therefore, based on relevance, only the chest deflection scaling formulae are reproduced

The EEVC approach seems most appropriate to this study as it includes terms accounting for geometric, stiffness, and failure ratios.

3.4.2 Geometric (anthropometric) measurements

The thoracic anthropometry measurements related to young children, Q10, Q12, and adults are shown in Table 3.9. As with the head and neck anthropometry, these values have been taken from the EPOCh Deliverable 1.2 (Waagmeester et al., 2009).

| Table 3.18: Geometric (anthropometric) measurements relating to the thorax |
|--------------------------|----------------|----------------|----------------|
|                          | 6 year old     | Q10            | Q12            | Adult          |
| Shoulder width (mm)      | 282.0          | 337.8          | 351.9          | 429.0          |
| Thorax width (mm)        | 180.0          | 217.4          | 227.0          | 305.5          |
| Thorax depth (mm)        | 134.8          | 153.9          | 158.6          | 230.0          |
| Thorax Circumference (mm)| 570.0          | 687.3          | 716.5          | 950.0          |
3.4.3 **Material property values**

Mertz *et al.* (2003) used both the parietal bone elastic modulus and calcaneal tendon failure stress in scaling the Injury Assessment Reference Values for the shoulder, thorax, abdomen, and pelvis. The reason for using material property scaling values from other body regions is likely to be due to the absence of scaling properties from the correct region. However, there was some, limited, information available with which to try and validate the scaling proposed by Mertz *et al.* and this has been added to in recent years. This has provided some interesting and novel areas of paediatric research.

3.4.3.1 **Thorax**

Cardiopulmonary resuscitation (CPR) involves the deflection of the sternum toward the spine in order to maintain oxygen and blood flow following cardiac arrest. This action is similar to the PMHS (post-mortem human subject) testing conducted previously with adult specimens (Maltese *et al.*, 2008). Various electro-mechanical devices have been developed over the past three decades to improve the quality of CPR and study the effect of the mechanics of thoracic compression on clinical outcomes. Recently, a load cell and accelerometer Force-Deflection Sensor (FDS) has been integrated into a patient monitor-defibrillator to provide visual and audio feedback on the quality of CPR chest compressions. The FDS has become the standard of care for patients of eight years or older in the Emergency Department and Pediatric Intensive Care Unit at the Children’s Hospital of Philadelphia, providing the opportunity to measure the force-deflection properties of children undergoing CPR. One type of error noted with this application of CPR measurements is that the compressions were performed on a deformable bed or stretcher. Therefore, Maltese *et al.* developed a compensation procedure for this before publishing force-deflection data from the FDS.

Eighteen subjects (11 females) ages eight to 22 years were enrolled in the study. The mean subject age was 14 ± 4 years, body mass was 47 ± 15 kg, chest depth was 185 ± 36 mm, and chest circumference was 835 ± 95 mm. According to Maltese *et al.*, maximum chest deflection did not change as a function of age (r = 0.02), nor did maximum deflection rate (r = -0.05). The thorax model elastic force at 15 percent compression (F_{e-15%}) demonstrated the highest correlation of the parameters that describe thoracic elasticity (r = 0.78). Whilst there is a figure in the paper by Maltese *et al.*, showing the distribution of F_{e-15%}; unfortunately, no values for this parameter are given in the paper.Crudely, the elastic force at 15 percent compression are described in the following way: "Subjects under age 10 (n = 2) generated on average ~100 N of elastic force, subjects aged 12 to 17 generated ~150 N (n = 9, excluding poor fit cases) and the 22 year old subjects (n = 2) generated ~340 N.

In a paper by Ouyang *et al.* (2006), a series of thoracic impact testing of paediatric PMHS were reported. In this case injury information was provided, although the testing was at a severity where six of the nine PMHS suffered a pneumothorax and another PMHS had a bleeding thymus gland. Again chest deformation responses are provided for younger (two to four years old) or older (five to twelve years old) children. Ouyang *et al.* also comment on the stiffness of the thorax for the younger and older children. The young cohort demonstrated an average initial stiffness of approximately 60 N.mm⁻¹, whereas the corresponding average stiffness recorded for the old cohort was approximately 75 N.mm⁻¹. They compare the thorax force–deformation responses from the older cohort (tested with a 3.5 kg, 75 mm diameter impactor) with the biofidelity corridor proposed for testing with a 3.8 kg, 82 mm diameter impactor (although it is not obvious where exactly this corridor came from). The responses and corridor are reasonably similar with the PMHS having a slightly higher peak force, at peak deflection, than suggested by the corridor boundaries.
3.4.3.2 Summary of available material property values

The following table (Table 3.19) shows the lack of data with which to scale the thoracic response data to the Q10 or Q12 size.

In the van Ratingen et al. (1997) relationship for scaling thoracic response to impactor tests, femoral bone elastic moduli are used. These data are taken from the summary paper of Stürz (1980), although the original source is Currey and Butler (1975). The Currey and Butler data are shown in Table 3.19 for comparison. Using the femoral modulus data instead of the skull bone modulus data would have the effect of increasing the scaling rates. It is not known whether the femoral or skull bone moduli are more closely related to the moduli of rib bone. However, adopting the EEVC approach of not scaling the modulus will give more conservative tolerance values for children. Therefore this approach is used, subsequently in the report.

<table>
<thead>
<tr>
<th>Measurement</th>
<th>Source</th>
<th>Young child</th>
<th>Q10</th>
<th>Q12</th>
<th>Adult</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skull bone modulus</td>
<td>Irwin and Mertz (1997)</td>
<td>6.6 GPa (six-year-old)</td>
<td>8.45 GPa</td>
<td>8.68 GPa</td>
<td>9.7 GPa (25 years assumed)</td>
</tr>
<tr>
<td>Femoral bone modulus</td>
<td>Currey and Butler (1975)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Calcaneal tendon failure stress</td>
<td>Yamada (1970)</td>
<td>51.94 MPa (five-year-old)</td>
<td></td>
<td></td>
<td>54.88 MPa (from 15 years to 45)</td>
</tr>
<tr>
<td>Thoracic organ modulus</td>
<td>Assumed (EEVC, 2008)</td>
<td>Equal to adult</td>
<td>Equal to adult</td>
<td>Equal to adult</td>
<td></td>
</tr>
</tbody>
</table>

3.4.4 Recommendations for injury criteria

A relationship between the risk of significant thoracic injury (AIS ≥ 3) and Hybrid III dummy sternal deflection for shoulder belt loading was developed by Mertz et al. (1991). This relationship forms the basis for the thoracic compression criteria in US and European regulation, as well as other regions of the world. Based on the data points underlying this adult risk curve, it has been, and is, possible to scale the relationship for other sizes of occupant.

The cardiopulmonary resuscitation data of Maltese et al. (2008) provide some interesting information regarding changes in thoracic stiffness with age. They found that normalising the force applied to the chest at maximum deflection by the maximum deflection of the chest was not an adequate descriptor of general thoracic elasticity. On this basis, it seems that scaling forces applied to the thorax must account for the thoracic stiffness as well as simply the geometry. Also, as suggested for the EEVC (2008) scaling work, injury risk functions must also take account of differences in the failure properties of human tissue. The scaling formulae proposed for the EEVC are the only formulae to include all the necessary terms for this process and should therefore be used in preference to earlier Mertz et al. derivations or the van Ratingen et al. biofidelity scaling relationships. Using slightly different strategies for the biofidelity and injury risk function scaling will produce some deviation between the dummy and the injury risk function. However, it may be appropriate to introduce that deviation to account correctly for the varying material properties between young children, older children and adults.
Any scaled injury risk curve should be compared with the recent work of Ouyang et al. (2006), accepting that the older group of children in that study (the five children from five years to twelve years old) were all injured (four at a severity of AIS ≥ 3). The study by Ouyang et al. could also be used to inform scaling parameters with respect to thoracic stiffness. Unfortunately, equivalent thoracic stiffness data for lateral impacts is not available at this time.

The EEVC equation for calculating the chest scale factor incorporates the ultimate tensile strength of the calcaneal tendon for the Q10 and Q12. This was calculated from data provided by Yamada (1970), from which a linear gradient was applied, based on information known for a 5 year old and 15 year old (Figure 3.25). This provided values of 53.4 MPa for the Q10 and 54.0 MPa for the Q12.

![Figure 3.25: Interpolation of calcaneal tendon tensile failure strengths to derive values for Q10 and Q12](image)

Having the tendon failure strength values allowed the EEVC equation (Section 3.4.1) to be applied, with the thoracic organ modulus ratio \( \lambda_{SS} \) assumed to be 1. This equation results in a chest scaling factor of 0.653 for the Q10 and 0.677 for the Q12.

These chest injury scaling factors were then applied to data from Mertz et al. (1991) to generate injury risk curves for chest AIS 3+ injuries for the Q10 and Q12 (Figure 3.26). The data from Mertz (1991) measured the average sternal deflection from sled tests (accident reconstructions) using the Hybrid III 50\(^{th}\) percentile dummy.

The graph (drawn using a Probit logistic regression analysis) shows that, based on average sternal deflection data used by Mertz, there is a 20 percent risk that thoracic AIS 3+ injury would occur with a sternal deflection of 24.0 mm for the Q10 and 24.9 mm for a Q12. There is a 50 percent risk of thoracic AIS 3+ injury with a sternal deflection of 33.3 mm for the Q10 and 34.5 mm for the Q12. This shows that a 50% risk of injury is caused by a 22% chest compression for both the Q10 and Q12.

The Mertz data were also used to calculate injury risk curves for chest AIS 4+ injuries for the Q10 and Q12. These data suggest that there is a 20 percent risk that thoracic AIS 4+ injury would occur with a sternal deflection of 26.0 mm for a Q10 and 27.0 mm for the Q12. There is a 50 percent risk of thoracic AIS 4+ injury with sternal deflection of 34.3 mm for the Q10 and 35.6 mm for the Q12.
The study by Ouyang et al. (2006) showed that there was an injury onset at 45 mm peak chest deformation (40.9% chest compression) for the younger set of children (2 to 4 years old). However there was no obvious correlation for the older set of children (5 to 12 years old), except that all suffered injuries with a peak compression of at least 24.2%. This series of injuries is consistent with the chest compression percentage (22%) calculated from the 50 percent risk of chest injury using the EEVC formula. That is, a greater than 50 percent incidence of injury was observed at chest compressions above 22 percent.

These injury thresholds calculated for the Q10 and Q12 have been scaled down from adult data. To validate this approach an alternative method was used, this involved scaling up from the 3 year old. The EEVC WG 12 & 18 method was used to calculate the ratio of the Q10 and Q12 to the 3 year old (Table 3.20). These ratios were then used to scale the EEVC 3 year old injury threshold data, to calculate chest compression injury values for the Q10 and Q12 (Table 3.21).
Table 3.20: Q10 and Q12 chest data

<table>
<thead>
<tr>
<th>3 year old</th>
<th>Q10</th>
<th>Q12</th>
</tr>
</thead>
<tbody>
<tr>
<td>Chest Depth (mm)</td>
<td>122.0</td>
<td>153.9</td>
</tr>
<tr>
<td>Tendon (MPa)</td>
<td>46.6</td>
<td>53.6</td>
</tr>
<tr>
<td>Elastic Modulus</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Ratio</td>
<td>1</td>
<td>1.46</td>
</tr>
</tbody>
</table>

Table 3.21: Chest compression injury data scaled from 3 year old

<table>
<thead>
<tr>
<th>3 year old</th>
<th>Q10</th>
<th>Q12</th>
</tr>
</thead>
<tbody>
<tr>
<td>20% Head Injury Risk (mm)</td>
<td>36</td>
<td>52</td>
</tr>
<tr>
<td>50% Head Injury Risk (mm)</td>
<td>53</td>
<td>77</td>
</tr>
</tbody>
</table>

It is suggested that the 50 percent risk of injury values scaled up from the three year old are not useful as they are likely to be in excess of the measurement range within the dummy. This also brings into question the validity of the 20 percent risk of injury values calculated in this way. It seems unlikely that only one in five 10 year olds would receive an AIS $\geq 3$ thoracic injury with 52 mm of chest compression. In this regard the values produced when scaling down from the adult seem more reasonable.

Considering compression as a proportion of chest depth further illustrates the problem with the results scaled up from the EEVC risk function for the Q3 based largely on accident reconstruction data (EEVC, 2008). Using the compression measurements in Table 3.21 and the chest depth from Table 3.20, the percentages of chest compression associated with either a 20 percent or 50 percent risk of thoracic injury are 29.5 and 43.4 % respectively (for the three year old). For the Q10 the percentages are 33.8 and 50.0 %. These chest compression percentages seem unrealistically high for those risks of injury and reinforce the concerns over the reliability of the results scaled from the EEVC (2008) values. It would seem more appropriate to use the new results scaled down from the adult. These maintain the chest compression at 22 percent, for any size of dummy, for an associated 50 percent risk of thoracic AIS $\geq 3$ injury.

### 3.4.4.1 Viscous Criterion (V*C)

A viscous tolerance criterion was proposed by Viano and Lau (1990). The velocity sensitive tolerance is represented by the maximum product of velocity of deformation and compression, which is derived from the chest compression response. In this paper, an injury risk curve for the probability of an AIS $\geq 4$ thorax injury for an adult human subject is proposed. However, at that time Viano and Lau commented that the tolerance level for the Hybrid III dummy had not been established. Subsequently that research was completed and the Viscous Criterion is used around the world with the Hybrid III adult dummies.

Mertz et al. (1997) commented on the Viscous Criterion in the introduction to their paper on injury risks for children and adults. They suggested that the Viscous Criterion curve can be applied to adults and children because equal viscous criterion levels experienced
by both children and adults will produce equal thoracic organ stresses. In principle this seems appropriate as the compression term in the criterion is given as a ratio of the chest size for the subject. However, care should be taken as the available chest compression in the human may differ from that in a dummy. For instance, in a dummy the whole chest cavity may not be deformable. There are likely to be rigid elements in the dummy such as a spine box onto which the compression sensor is mounted. In which case, the force-deflection response of the dummy may deviate from similarity to the human when close to its mechanical limit of compression. At which point the Mertz et al. comment on equivalent thoracic organ stresses for a given $V^*C$ would no longer be appropriate. This is likely to be an issue for all dummies; however, in the small volume of a child’s thorax it may be more difficult to accommodate the instrumentation and instrumentation mounting points. Therefore, it is suggested that $V^*C$ can probably be transferred to any dummy size based on two stipulations: sufficient biofidelity (viscous compression response) of the dummy thorax and that it has sufficient space within the thoracic cavity to provide reliable results up to the injury threshold selected. The latter point should be confirmed in the worst case conditions of the dummy’s expected application.

3.4.4.2 Acceleration

Real world accidents involving both child car occupants (seated in Römer-Peggy I or II restraints) and pedestrians were recreated by Stürtz (1980) using VIP 3c and 6c dummies. On the basis of nine cases (including six reversible or non-injury cases) Stürtz proposed limits of 105 g for the peak resultant thoracic acceleration and 85 g when exceeded for 3 ms. Apparently these values represented protection criteria covering 75 percent of the reversible injuries and 25 percent of the irreversible injuries. Alternative limits of 55 g for both acceleration measures were also suggested for representing 50 percent of reversible and irreversible cases. The current UNECE R44 chest acceleration criteria is also 55 g (3ms exceedence)

It is not known how the biofidelity of the VIP child dummies would compare with the Q-series. It is unlikely that a dummy-specific injury risk function developed for the VIP series could be reliably transferred to the Q-series. However, the use of only nine cases to develop the proposed thresholds for the VIP dummies means that the tolerance values can only really be considered as indicative. On the basis of providing a general guide to a child’s tolerance, rather than a specific risk function, the values suggested by Stürtz provide at least a little information.

These values, reported by Stürtz, are around the level proposed for use in US regulation with adult subjects or surrogates of 60 g; although the Stürtz child values are slightly higher than the adult, when looking at the 75:25 risk values. There is no clear basis as to why the child tolerance would be expected to be higher than for adults. Also, if the adult value was maintained it should provide a conservative estimate of injury risk for a child. Therefore it is proposed to maintain the European 55 g limit, if one is needed.

3.5 Abdomen

It was the ambition for this section of the report to document potential scaling techniques and biomechanical data that could be used to validate abdominal and pelvic response data. Unfortunately, the limited data do not support the derivation of scaled injury risk functions. However, the information gathered concerning these two body regions is reported below for completeness.

Within the study of Chamouard et al. (1996), investigating submarining risk for dummies on a booster, 12 quasi-static compression tests were carried out with six child volunteers. Belt displacements of up to 60 mm were recorded. Chamouard et al. observed that the initial stiffness of the abdomen of the children aged three to ten years
showed relatively little scatter (in the range of about 1 to 2.5 N.mm⁻¹). They therefore drew a corridor in which three-to-ten year impact test dummies must be situated in order to be considered as being representative.

Johannsen (2006) recommended the use of average surface pressure for the assessment of abdominal injury risk. However, Johanssen also acknowledges that whilst the measurement principle is good, better sensors need to be developed.

In the same year, the biomechanical response of the paediatric abdomen under dynamic belt loading was evaluated by Kent et al. (2006) using a porcine experimental model. After investigation of the young porcine development, a size of pig was selected as being appropriate for comparison with the six year old human child. A series of 18 quasistatic and 47 dynamic tests were then carried out on 47 different swine to quantify the penetration response of the abdomen when loaded by a transversely orientated ‘lap’ belt. Each subject was ‘euthanized’ immediately prior to dynamic testing. The test was performed immediately after cardiac and respiratory activity was confirmed to have ceased.

In quasistatic tests, the loading phase occurred over several seconds (about six). These tests revealed response changes both when the belt position was shifted, and when muscle stimulation was applied. The upper abdomen belt placement generated an effectively stiffer response compared with the lower abdomen. At 30 mm of belt penetration, the mean reaction force with lower abdominal loading was approximately 90 N while it was approximately 170 N at the upper abdomen. The effective stiffness also increased when the muscles were stimulated. The degree of increase varied across subjects, but the mean reaction force at the lower abdomen, with 30 mm of penetration, increased from 78 N with no stimulation to 103 N when the muscles were actively tensed.

Corridors representing the upper and lower bounds of the force versus belt penetration data published by Chamouard et al. (1996) were plotted again the data from the quasistatic porcine tests as a limited assessment of the validity of the porcine model. The quasistatic porcine response represented the quasistatic human data well.

A linear regression model was used by Kent et al. to indicate that the maximum force did vary as a function of the maximum abdominal deflection and weakly as a function of the maximum velocity of deformation. Kent et al. also show the exact nature of such changes by grouping the tests into ‘bins’ of similar loading rates and comparing the reaction force across the bins. This provided some indication of the weak trend for the reaction force (at 25 % penetration) to increase with increasing loading rate.

Kent et al. do not provide any information regarding injuries sustained by the porcine specimens.

3.6 Pelvis

In 2003 Ouyang et al. reported on a series of lateral impacts to the pelves of child PMHS. This work is invaluable for providing descriptive information on the force deflection behaviour of paediatric pelves. It yields biofidelity corridors for pelvis impacts to younger (two to four years old) or older (five to twelve years old) children. However, whilst this information should be useful in defining biofidelity targets for child dummies, the testing was conducted at a sub-injurious level. Therefore, pelvic injury risk functions cannot be derived from this research.
4 Discussion

As noted by Cory and Jones (2006);

“The scaling technique in general makes the assumption that the child is a scale model of the adult and their mass and material differences vary by relatively simple mathematical relationships. If these assumptions hold true, the approximations can be considered to be good. However, the more these assumptions are not valid, the more the translated physical measurements may be distorted from their true levels.”

The following sections of the report discuss whether the assumptions used in the scaling are appropriate approximations, based on the limited information available for defining scaling ratios and validation of the scaled products.

4.1 Head injury

Comparing the results from the head response scaling reveals that the Mertz method produces an injury risk curve for the Q10 or Q12 that is different to that produced with the van Ratingen method. This was expected as van Ratingen et al. (1997) used the cranial bending stiffness data in their scaling; whereas, Mertz et al. (2003) and the EEVC (2008) both used calcaneal tendon failure stress as a proxy for the brain failure property. As previously mentioned the van Ratingen method shows a better correlation to actual test data and therefore this method will be used. However, this does not account for the fact that Irwin and Mertz (1997) and Coats et al. (2007) have reported that head stiffness and principal stress in the skull are dominated by the bulk modulus of the brain. For this reason it seemed appropriate to try and incorporate the stiffness values from the empirical testing of Prange et al. (2004). This led to much better agreement with the measured head acceleration values from that experimental research; although further validation against another test series is strongly recommended.

The porcine testing of Coats and Margulies (2003) indicated a decrease in elastic modulus of cranial tissue with increasing loading rate. If the relationship of adult to child stiffness is not equivalent at higher loading rates, then it may be misleading to determine such a relationship in quasi-static loading conditions, when testing will occur at higher loading rates. This fact gives added weight to the adoption of head stiffness values derived from realistic impact tests rather than using existing quasi-static test data. It also might go some way to explaining the improved result when using the stiffness values reported by Prange et al. from drop tests, rather than quasi-static bending or tensile test data.

The risk curves produced by the van Ratingen method are similar with a 20 percent risk of skull fracture produced from 77 g and 78 g, respectively for the Q10 and Q12. These are very similar to the adult 20 percent risk of injury of 75 g. The equivalent values for the 50 percent risk of injury are 251 g, 254 g, and 244 g, respectively for the Q10, Q12, and adult. The original analysis of the adult head injury data by Mertz et al. (1996) indicated that a head acceleration of 80 g was associated with a risk of skull fracture that was less than 0.1 percent. Those authors suggested 180 g as being equivalent to a 5 percent risk of skull fracture. Clearly the risk curve derived using a statistical technique is markedly different. Due to the technique used by Mertz et al. to derive their risk curve, the peak head acceleration will underestimate the risk of skull fracture, as supported by the reported data. The interpretation of head acceleration values should take into account the risk of skull fracture which the data justify. However, the threshold value used in test work may or may not need to be adjusted depending on the balance between the allowable risk of injury and the feasibility of providing that level of protection.
It must also be remembered that the original head impact data were being used to investigate the risk of skull fracture from head contacts. The appropriateness of using this dataset to define risk functions for the EPOCh older child dummy must be reviewed.

The numerical simulations conducted by Prange et al. (2004) indicated that the brain mechanical properties make an important contribution toward strain distribution and magnitude during rotational, inertial loading. At present there does not appear to be a proposed technique that incorporates these parameters when scaling adult injury data to predict paediatric head injuries. Also there is no agreed adult injury risk function for rotation-induced brain injury. Therefore no scaling of rotational measurements has been conducted. However, this may have further implications as the brain mechanical properties must also have a bearing on the tolerance of a child’s head to linear inertial loading. The lack of terms reflecting these properties is clearly a limitation of the head injury scaling processes presented to date.

Current regulatory use of child dummies considers only frontal and rear impact events (not side) through sled-based testing. Under these conditions, it is unlikely that a severe head contact will occur and gross head excursion is limited based on the dummy kinematics. In consumer testing a need has arisen to consider head contact both through side impact contacts with the wings of a child restraint system or an intruding structure and in frontal tests using a full-scale vehicle (where contact with a part of the vehicle interior is likely). Linear acceleration values (peak resultant, 3 ms exceedance) are used at the moment for assessing these head impacts with child dummies. Such measures are thought to provide a useful assessment of the injurious nature, or otherwise, resulting from the contact and are also intuitive to design towards. With adult dummies the Head Injury Criterion (HIC) is routinely used to evaluate the severity of head impacts and one might want to use the HIC with child dummies. For this reason, HIC-based injury risk functions and values have been scaled from the Q3 and adult for use with the Q10 or Q12. Deviation of the adult HIC skull fracture risk curve from a 0, 0 intercept and large discrepancies between the results from the adult or Q3 scaling have called into question the reliability of the results. Also, as noted in Section 3.2, there are some concerns over the appropriateness of HIC for use in padded, not very short duration, head impacts. With these issues in mind, it seems difficult to specify precise HIC values for a given level of injury risk. Instead it seems more likely that these risk functions and values could serve as supporting evidence for more pragmatic assessment thresholds. Such thresholds probably need to be set with due consideration of the test procedure for in which they are to be used, current child restraint system performance, and the feasibility of implementing a HIC threshold. However, it is strongly suggested that the appropriateness of the HIC be reviewed with respect to the intended use before a threshold value is introduced as a CRS assessment criterion.

4.2 Neck injury

Although there is currently only limited neck test data to enable an injury risk curve to be created, upper neck injury thresholds have been calculated for the Q10 and Q12 based on the EEVC adult upper neck injury limits. The Irwin and Mertz method for neck force uses the neck circumference ratio squared. However it was noted that the neck circumference of the adult is very large in comparison to the Q10 and Q12.

The neck moment ratio was altered slightly from the original Irwin and Mertz formula as cubing the neck depth ratio did not produce a sensible result. There is concern over the neck depth measurement and hence the neck depth ratio between the Q10 or Q12 and the adult. The scaling ratio was therefore altered to the cross-sectional area (depth x width) multiplied by the distance it was applied over, as used by the EEVC (2008).

Lower neck injury thresholds have not been considered independently of the upper neck within this scaling work. No scaling techniques for the lower neck have been previously published. However, it would be possible to derive lower neck injury thresholds if these are needed for use with the older child dummy. This need is yet to be established, as the
existing Q-series are often used with only an upper neck load cell. Mertz et al. (2003) suggested that the limits for lower neck flexion, extension and lateral bending moments should be twice those for the upper neck. The exact reason for this ratio of upper to lower neck bending strength is not provided. It does seem reasonable that the lower neck would be able to tolerate a greater bending moment than the upper neck based on muscle and neck size. The functional unit testing of neck specimens that has been reported within Section 3.3.3 should be able to provide information pertinent to the untensed neck at superior and inferior locations. Based on those data, a more accurate ratio of upper to lower neck strength could be derived and used to produce lower neck limits. This derivation is considered to be beyond the scope of this work, particularly as no immediate need for the resulting threshold values has been raised to date.

4.3 Chest compression

The WG12 & 18 method for calculating chest compression was used to calculate injury risk curves for AIS ≥ 3 and AIS ≥ 4 thoracic injury. These curves were very similar, as was reported with the original adult data (Mertz et al., 1991). As it is slightly more conservative, it is suggested that the AIS ≥ 3 risk curve is used to define injury threshold values for thoracic injury, unless AIS ≥ 4 injuries are of particular interest.

Within the Q Dummies report (EEVC, 2008) chest deflections for 20% and 50% AIS ≥ 3 injury risk were calculated for the Q3 dummy. The values of 36 mm (20%) and 53 mm (50%) are a little higher than those calculated for the Q10 and Q12 using the same equation. This suggests that more chest compression would be allowed for the Q3 than for the Q10 or Q12. It is not understood how this can be the case if both sets of authors genuinely used the same formula and values. The EEVC report also calculated values for the other Q-series dummies by scaling the Q3 chest deflection value. However these results seem to suggest that smaller children would be injured at higher chest deflections than older ones, which is counterintuitive as they would have smaller chest cavities. This suggestion comes about as a result of the assumption that a child’s rib cage can sustain a greater deformation than an adult’s before rib fracture. As noted by the EEVC authors, any additional flexibility of a child’s rib cage needs to be considered alongside the potential for injury to the underlying thoracic organs. Therefore we have used the scaled Hybrid III adult risk function to provide recommendations on injury criteria for the Q10 or Q12.

In a similar manner it would be possible to scale the deflection risk functions from side impact dummies to derive lateral thoracic injury criteria. However, recently within the EC Aprosys project concerns were raised over the appropriateness of using 5th percentile female risk functions, where the scaling was from and to a WorldSID. The issues mostly related to assumptions over the similitude of thoracic design from one size of dummy to the other. These assumptions would be stretched even further when scaling to a Q-series dummy. Also, the lateral thorax response of child dummies in child restraint systems can be dominated by the extent to which the arm provides a bridging structure between the shoulder and the pelvis. This provides a protective affect on the thorax deflection, which is unlikely to be mimicked by a human. Care has been taken with adult side impact dummies to avoid this behaviour through the use of half arms (too short to provide an effective bridging structure in normal circumstances) and specifying a flexed position of the shoulder for consumer and regulatory testing. Pure scaling of the adult response alone is unlikely to account for such kinematic differences.

The exercise of scaling the lateral injury risk function could be undertaken; however, based on these two concerns it is recommended that such an approach would need some form of validation before application to the new Q10 or Q12. This validation would be beyond the remit of this investigation and therefore the lateral injury risk scaling has not been carried out at this time.
5 Conclusions

The available published methods for scaling biomechanical responses from the adult to older children were collated. Necessary information required for use in the scaling functions was gathered. Scaling was then applied to generate risk curves or injury threshold values for the following parameters.

In each case, the input data and scaled products were compared against information in the literature that provided, or could be used to advise as to the, expected tolerance values.

To come to a consistent set of tolerance values for the complete Q dummy family in line with the Q10/Q12 values recommended in this report it is recommended to apply the method of this report the smaller Q-dummy family members. It is also recommended to assess the feasibility of the application of the tolerance values in line with the validation given in EEVC (2008) chapter 5.

**Head acceleration (linear)**

![Figure 5.1: Head injury risk curve for an older child dummy (Q10 or Q12) based on peak linear acceleration](image)

The adult 3 ms head acceleration exceedence limit of 80 g was also considered to be a reasonable and conservative threshold for use with the older child dummy.

Limitations as to the interpretation of skull fracture risk, based on measured head acceleration were identified. Risk curves based on statistical techniques have now been derived. However, the appropriateness of the underlying data, with respect to the expected use of the EPOCh older child dummy remains uncertain.

The risk curves produced by the van Ratingen method showed there was a 20 percent risk of skull fracture at 77 g and 78 g, respectively for the Q10 and Q12. There is a 50 percent risk of skull fracture at 251 g and 254 g, respectively for the Q10 and Q12.
Functions for the risk of skull fracture, based on Q10 or Q12 HIC (Head Injury Criterion) values were developed through scaling the adult risk function or risk values for use with the Q3. Concerns were raised over the validity of either approach and therefore it was suggested that HIC injury risk threshold values should be adopted with caution and a degree of pragmatism.

It is proposed that tolerance values for rotational acceleration can be transferred from the adult to the child to give conservative thresholds. The adult tolerance to peak rotational acceleration lies somewhere in the range from 1,500 rad.s$^{-2}$ for concussion, to 16,000 rad.s$^{-2}$ for diffuse axonal injury.

**Neck tension, anterior-posterior bending, and shear**

The neck tension and shear corridors are reproduced below.

![Figure 5.2: Neck tension injury corridor](image)
Figure 5.3: Neck shear injury corridor

Figure 5.4: Chest injury risk curve using average sternal deflection
The risk curves produced by the Mertz method showed there was a 20 percent risk of chest AIS 3+ injury at 24.0 mm and 24.9 mm, respectively for the Q10 and Q12. There is a 50 percent risk of skull fracture at 33.3 mm and 34.5 mm, respectively for the Q10 and Q12.

The Viscous Criterion may also be transferred to the child although some validation of the appropriateness of using the adult threshold is recommended.
6 Recommendations

The scaling techniques that have been published previously and used again here rely heavily on a few data points regarding the biomechanical properties of children. These data are interpolated to define changing biomechanical properties with age. The exact interpolation used with these limited data may cause differences in results of the scaling process. This use of the original data must be treated with caution, and the potential for inaccuracy considered with respect to the intended use of the scaled results. More fundamentally, it must be questioned as to the appropriateness of using calcaneal tendon failure strength and cranial bone bending moduli to scale parameters relating to all body regions. In certain circumstances it is expected that such an approach will lead to substantial inaccuracies in the products of the scaling. There is a clear need for further biomechanical data to validate the outputs of studies such as this.

If possible, the head testing of Prange et al. (2004) should be extended to include subjects of older ages. This would provide better validation of scaling techniques used in relation to head impact responses of children. If testing with adult specimens could also be included, then it would mitigate concerns as to the similitude of test conditions when comparing the results from the work of Prange et al. with those of early test series (e.g. Hodgson et al., 1967).

A limitation of head scaling techniques is that they do not account properly for the brain mass and material properties. New scaling relationships are needed to address this limitation, along with sufficient biomechanical data to support the use of those relationships.

Another limitation of the scaling of head injury risk for children is that the adult risk functions for HIC and head acceleration have been based on head contact impacts, of a very short duration. These impacts may not be very relevant to the conditions under which the EPOCh dummy will be used. In side impacts head contact is expected, but this will be with a padded structure (unless there is gross excursion of the dummy or intrusion of a vehicle structure). In frontal impacts in a car body, head contact may again be expected with a part of the vehicle interior; however, frontal sled-based tests would not be expected to bring about head contact with anything other than perhaps another part of the dummy. Therefore it is strongly recommended that alternative data sources are found which are more relevant to the conditions expected for children in CRSs during accidents. This should also help to resolve the issue of the difference between HIC values scaled from either the adult of Q3. Generating such a dataset is beyond the remit of this study.

On the assumption that brain mass and material properties for children have a protective effect when compared with adults, it is suggested that adult tolerance values for rotational inertial loading could be used with an older child dummy to give a conservative threshold. However, this assumption would benefit from further validation. Also, the appropriate tolerance value would need to be selected for the injury or injuries against which protection is being encouraged (e.g. diffuse axonal injury, traumatic brain injury, or mild traumatic brain injury). This selection may be far from trivial, particularly where different authors have proposed different limits or assessment methods for the same injury group.

From their testing of paediatric neck specimens, Ouyang et al. (2005) found no statistical difference in tensile stiffness with variation in age. This was in contradiction to the findings of previous authors based on animal testing. To confirm that the Ouyang et al. result was due to the limited size of the dataset, and hence the available resolving power, more human subject testing is recommended, where ethics allow.

It is recommended that the appropriateness of scaling adult lateral thoracic injury criteria for use with child dummies seated in child restraint systems is investigated. This investigation would need to consider whether the scaling duly accounts for differences in dummy design between the adult and child as well as differences in kinematic behaviour.
Such an investigation was considered beyond the scope of the investigation undertaken for this report.
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