Biofidelity requirements for the THORAX project

EUROPEAN COMMISSION
DG RTD

SEVENTH FRAMEWORK PROGRAMME
THEME 7
TRANSPORT - SST
SST.2007.4.1.2: Human physical and behavioral components
GA No. 218516

THORAX
Thoracic injury assessment for improved vehicle safety

Deliverable No.	THORAX D2.1	
Deliverable Title	Biofidelity requirements and for the THORAX project	
Dissemination level	Public	
Written By	David Hynd and Jolyon Carroll (TRL) Johan Davidsson (Chalmers) Philippe Vezin (IFSTTAR)	2013-07-03
Checked by	Johan Davidsson (Chalmers) and Cor van der Zweep (Uniresearch)	2013-07-05
Approved by	Paul Lemmen (Humanetics)	2013-07-05
Issue date	2015-07-16	

Public Executive Summary

Biofidelity requirements are to be used to ensure that a crash test dummy loads the vehicle and restraint system in an accident in a similar way to the human, and to ensure that the response of the dummy to this loading is relevant to the prediction of injury risk in simulated crashes that are representative of real-world accidents. The main aim of this deliverable is to provide a set of biofidelity requirements for the thorax and shoulder for evaluation of an advanced frontal impact crash test dummy.

This report has reviewed existing thorax and shoulder biofidelity requirements for frontal crash test dummy evaluations. The load cases used in these requirements were compared to the loads in actual collisions. It was identified that inclusion of additional requirements in which the thorax is exposed to various types of distributed and belt-only loads would be beneficial.

To identify additional tests, post mortem human subject data and volunteer data were reviewed and test conditions and available data documented. Inclusion criteria used to assess e.g. the quality of documentation of a data set, or the representativeness of the subjects that were tested, were established. Using these criteria the reported test conditions and results were analysed with the target to specify biofidelity requirements and engineering guidance for the design of an enhanced dummy shoulder-thorax complex. None of the available datasets were ideal for specifying biofidelity requirements for frontal impacts in modern restraint systems as available in the market now. These systems typically include belt pretensioning and force-limiting and carefully combined belt and airbag contributions to the occupant protection. Instead a broad set of requirements has been used in an attempt to capture biofidelity under various restraint system types and load conditions. It is hoped that a dummy with a good level of biofidelity throughout this broad range of conditions will still demonstrate an appropriate level of biofidelity in modern restraint systems and common crash conditions.

The biofidelity requirements document a well-defined set of test conditions and the dummy responses that are required in those loading conditions. The engineering guidance includes biomechanical data that will be used to define relative - rather than absolute - targets for dummy performance. These relative targets are useful to guide the design of an enhanced dummy.

Various methods used to normalise the response data to that of a standard size of subject or scale data to other sizes were reviewed, benefits and limitations discussed and recommendations were made.

Finally, a set of biofidelity target corridors for the 50th percentile male are presented in the Appendix B to Appendix K.

Contents

1 Introduction

Biofidelity requirements are specified in order to ensure that the dummy loads the vehicle and restraint system in an accident in a similar way to the human, and to ensure that the response of the dummy to this loading is relevant to the prediction of injury risk in real-world accidents, regulatory and consumer information crash tests, and product development tests. Both the dummy used in the European directive today (the Hybrid III 50th percentile male dummy) and its intended successor (THOR) have been assessed extensively with respect to their ability to predict the risk of thorax injury and to discriminate between what are known as effective and poor safety systems, based on field performance data. Both dummies are considered limited in this capacity, although the THOR design has better biofidelity than the Hybrid III dummy. Nevertheless, improved thorax biofidelity and injury assessment, particularly the ability to discriminate between modern restraint systems of similar performance, is a key objective of the THORAX project.

The main objectives of Work Package 2 (WP2) of the THORAX project are to:
- Define a set of biomechanical requirements for an enhanced shoulder-thorax complex for crash-test dummies of different sizes and ages; and
- Define a set of injury risk functions for vehicle occupants of different sizes and ages.

This deliverable provides a set of biomechanical requirements for the evaluation of the thorax and shoulder of a frontal impact crash test dummy. A first draft of the requirements was developed in the first half of 2010, and discussed with Stakeholders (including representatives of regulators, industry and research organisations) at a meeting in September 2010. The present document represents an updated set of requirements based on discussion with Stakeholders and on new information which has become available to the THORAX consortium. This encompasses both biofidelity requirements and engineering guidance for the design of an enhanced dummy shoulder-thorax complex. The former document a well-defined set of test conditions and the dummy responses that are required in these loading conditions. The latter includes biomechanical data that will be used to define relative - rather than absolute - targets for dummy performance. These relative targets are useful to guide the design of an enhanced dummy.

The main objective of this report is to define a set of biomechanical requirements for the thorax and shoulder of an advanced frontal impact crash test dummy. The definition of biofidelity requirements and an overview of how they are developed is giving in Section 2. Existing thorax and shoulder biofidelity requirements are reviewed in Section 3 of this report. These come from a range of government and industry bodies, as well as previous EC Framework projects. The merits and limitations of these requirements are also discussed. Section 4 gives an overview of the types and severity of loading to the thorax and shoulder for which it is relevant to define biofidelity requirements. This is informed by a range of considerations, such as:
- The types of frontal impact accident in which people sustain injuries to these regions (identified in the accident analyses of the FP7 EC Framework projects COVER and THORAX);
- The types of restraint system used;
- The rate of loading in relevant accident scenarios.

Furthermore, this section documents additional inclusion criteria used to assess e.g. the quality of documentation of a data set, or the representativeness of the subjects that were tested. Section 5 then gives an overview of the selected biofidelity data sets, with a detailed review of all of the candidate data sets in Appendix A, and detailed test conditions documented in Appendix B to Appendix K.

In order to develop biofidelity targets from biomechanics data sets, many authors have attempted to normalise the responses in the data set to that of a standard size of subject. This processing is intended to make the data more representative of a particular occupant group, and to reduce the scatter in the observed biomechanical responses of different subjects. The various methods that have been used for this normalisation process are reviewed and discussed in Section 6. These methods have often been applied to relatively simple data, such as pendulum impactor data sets. Section 6 also examines the options for normalising the much more complex kinematics of an occupant in a sled test, and evaluates whether these options are appropriate. Normalisation methods are also sometimes used to scale biofidelity requirements to a completely different size of occupant, e.g. from 50^{th} percentile male to 5^{th} percentile female, and the benefits and limitations of this are also discussed.

Finally, Discussion and Conclusions may be found in Section 7 and 9 respectively, and Section 8 contains the risk register for this task.

The presented laboratory test procedures, impact response requirements data and injury data will serve as data in Task 2.6 (Development of injury risk functions), Task 3.1 (Demonstrator requirements) and in Task 3.3 (Demonstrator validation tests) in the THORAX project.

2 Background

As noted in Section 0, biofidelity requirements are defined in order to ensure that:

- The dummy loads the vehicle and restraint system in a crash test or accident reconstruction in a similar way to the human;
- The response of the dummy to this loading is similar to that of a human in the same loading condition;
- The measurements made with the dummy are equivalent to similar measurements made on human subjects that have been correlated with the risk of injury to those subjects.

Biofidelity requirements consist of a collection of biomechanical response requirements, plus a detailed description of the test conditions, that define the typical response of a human in a given loading condition. For instance, this may be head acceleration in an impact with a rigid surface, or mid-sternum chest compression when impacted by a pendulum of a certain size, mass and velocity. In general, the broader the range of conditions for which the biofidelity of the dummy is confirmed, the more confidence users can have that the dummy response and prediction of injury will be valid in a wide range of loading conditions.

Biofidelity requirements are typically defined for a particular occupant grouping, such as a 50th percentile (average stature) male or 5th percentile (small stature) female. The requirements are typically defined based on tests performed with volunteers (at very low loading levels and therefore low injury risk) and Post Mortem Human Subjects (PMHS - usually at higher, potentially injurious, loading levels).

Groups of volunteers and PMHS are usually selected to be as representative as possible of a particular occupant grouping, but are never identical to the target group. Many authors have therefore developed normalisation techniques in an attempt to:

- Improve the representativeness of the responses;
- Reduce the scatter in the responses, and therefore improve the confidence in the resulting biofidelity requirements.

These normalisation processes usually use subject characteristics (e.g. height, mass, chest depth and so forth) or response characteristics (e.g. effective mass of the thorax in a sled test) to scale the magnitude and timescale of the response of each subject. The options for normalisation are discussed in detail in Section 6. In addition to this, the individual subject response data is sometimes time-shifted, e.g. in order to align the peak in each subject's response.

Once the data have been grouped (and possibly normalised), several approaches have been used to define the response requirements for a dummy, such as:

- Drawing a simple corridor that approximates the maximum and minimum of the responses in the data set (see, for example, Figure 2-1). This has the advantage that the corridor can be defined in terms of a simple set of co-ordinates, but the disadvantage that the corridor width may be very wide for some data sets and therefore not provide very strong guidance for the design of a dummy.
- Defining the mean and standard deviation of the responses, and then drawing a simple corridor to approximate the ±1 standard deviation responses. This has the advantage that the corridor can be defined in terms of a simple set of co-ordinates, but the disadvantage that the corridor width may still be quite wide for some data sets and therefore not provide very strong guidance for the design of a dummy. In some cases where the original data consists of a small number of subjects with similar responses, the standard deviation may be very small and this gives an unrealistically challenging target for the design of the dummy. Corridors defined in this way may be so narrow at

some points in the response that it is essentially impossible for a dummy to entirely meet the requirement.

- Defining the mean and standard deviation of the responses, and then statistically comparing the dummy response with these. This has the advantage that the requirement is more precisely defined, but the assessment of the dummy will depend on the scatter in the original data. For instance, if the confidence limits are very small, a dummy with good biofidelity for most of the impact response, but which exceeds the confidence limits for part of the response, may be rated as less good than if it had been assessed against a data set with more scatter. This is sometimes handled by weighting different biofidelity requirements to account for the confidence in the requirement, e.g. a lower weighting for a smaller sample.

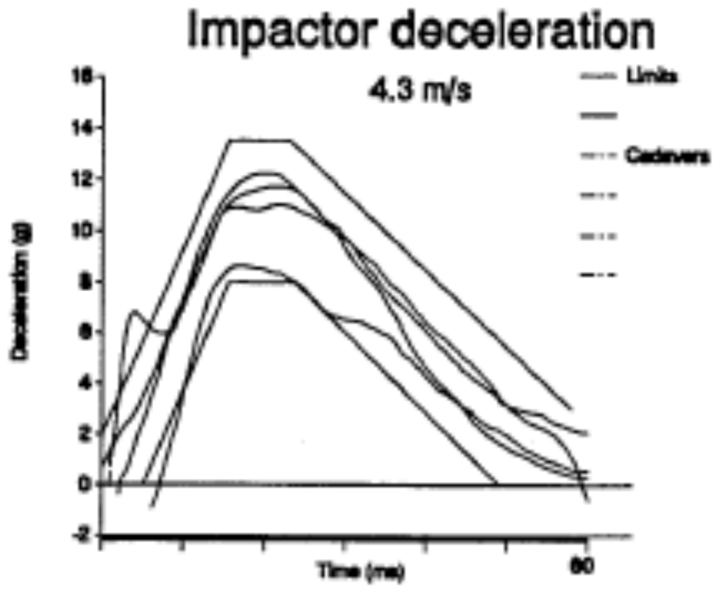


Figure 2-1: Example biofidelity corridor (with the original PMHS responses) (adopted from Roberts *et al.*, 1991b)

Furthermore, a number of terms are used to describe these response requirements, depending on how the requirement has been defined and how it will be applied. Examples include:

- Biofidelity requirements (which implies that the dummy must meet the requirement);
- Biofidelity corridors;
- Biofidelity target corridors (which implies a target for dummy design, and that it is expected that the dummy may not meet all targets for all body regions).

The biofidelity of a whole dummy is sometimes assessed by combining the results from all of the biofidelity assessments. This is the approach that has been used in the ISO side impact biofidelity requirements (ISO, 1999), but the approach has not been used by the EEVC. When this approach is used, it is usual that the biofidelity requirements are weighted according to their importance (e.g. a high weighting for a body region with a high risk of fatal injury), or the confidence with which the requirement is defined (e.g. a high weighting for requirements based on well-defined test conditions, with a large number of subjects, and with a small scatter or standard deviation of the responses of the subjects).

3 Existing frontal impact biofidelity requirements

The following sections briefly describe the main existing thorax (including shoulder and spine) biofidelity requirements that have been used for the development and evaluation of frontal impact dummies over the last decade. In most cases these are a subset of a broader set of requirements that have been defined for all body regions of a frontal impact dummy.

3.1 Trauma Assessment Device Development Program (TADD) requirements (1989)

The Trauma Assessment Device Development Program was sponsored by NHTSA in the late 1980's and early 1990's to develop an improved thorax and abdomen for the Hybrid III 50th percentile male crash test dummy, which could also be used as a component of an entirely new frontal impact dummy. This followed-on from previous work on the Advanced Anthropomorphic Test Device (AATD) Development Programme. One of the early outputs from the TADD programme was the publication of design requirements and specifications for the thorax (and abdomen) (Schneider *et al.*, 1989). This included definition of the relevant crash and restraint types, relevant injury severities, and also biofidelity requirements. The focus had shifted somewhat towards the assessment of belt and airbag restraint systems at lower loading rates, although thorax-to-steering-wheel and thorax-to-dashboard loading was still considered important, and (as a lower priority) out-of-position airbag testing.

Priority was given to low velocity impactor and static deflection rather than high speed impactor test conditions:

> *'The new subcomponent should be designed to provide biofidelity in impact response to a rigid 150-cm diameter, 23-kg impactor at the mid-lower sternum and left and right lower ribcage over the spleen and liver, in accordance with the force-deflection curves outlined and discussed in … this document. Since the priority in future impact testing is shifting toward the need for a device with humanlike response to restraint systems and from the need for biofidelity of unrestrained ATDs with vehicle components, the priority in meeting dynamic response corridors should be shifted toward achieving the 4.3-m/s corridor rather than the 6.7-m/s corridor.'*
>
> *'The new subcomponent should have improved biofidelity in response to low-velocity and static loading. This implies a significant reduction in static stiffness for the first two inches of internal deflection than now offered by the Hybrid III thorax.'*
>
> *'Given that all loading-rate response corridors cannot be equally achieved, the priorities in achieving biofidelity should be the following: (1) 4.3 m/s* [impactor tests]*; (2) static F-δ* [force-deflection response]*; and (3) 6.7 m/s* [impactor tests]*.'*

3.2 ADRIA requirements (1999)

The ADRIA project (ADRIA, 1999) listed biofidelity requirements for the neck and thorax of a frontal impact dummy. For the thorax, the Kroell impactor tests were selected and the THOR prototype dummy was assessed against the Kroell sternum force-deflection requirements. No requirements for the shoulder or spine were given. The ADRIA report also noted that:

> *'The Kroell impactor tests are not representative for loading of the thorax during a frontal impact: the impactor mass is large and the impactor shape is not related to seat belt loading in real accidents. Dynamic belt loading tests would be more representative. Cesari and Bouquet (1994), Kallieris et al. (1995) and Morgan et*

al. (1994) have carried out thorax tests with more representative loading of the thorax, however no performance requirements have been developed up to now based on these test results.'

3.3 NHTSA THOR requirements (2001)

NHTSA biofidelity response requirements for the THOR-Alpha frontal impact dummy were published by GESAC Inc. as part of the documentation package for the dummy (GESAC, 2001). Firm requirements for the thorax were defined, as well as proposals for additional requirements for the thorax and spine.

NHTSA thorax requirements

- Neathery (1974). Pure frontal mid-sternum impactor PMHS tests using a 23.4 kg impactor, at a velocity of 4.3 m/s and 6.7 m/s. Mid-sternal force-compression response corridors were defined at each speed, with a 'correction' of 667 N applied to the force to account for the stiffening due to muscle tensing that would be expected to occur in a living human car occupant.

NHTSA spine proposals

- Six data sets were evaluated for their potential to be used to define the biofidelity of the spine, but none were considered to be suitable to be used.

NHTSA thorax proposals

- Schneider *et al.* (1992). Coupling between different regions of the thorax under quasi-static loading conditions, defined by relative deflections at eight locations on the thorax.
- Cesari and Bouquet (1990). Regional loading under belt impact, defined by deflections at various locations on the thorax.

The Neathery (1974) requirements and Schneider *et al.* (1992) proposed requirements were the main test conditions documented in the TADD biofidelity requirements.

3.4 FID THOR requirements (2003)

The EC 5th Framework project FID 1999 reviewed the available biomechanical data, as well as new biomechanical data developed within the project, and proposed a comprehensive of biofidelity requirements for an advanced front impact dummy (van Don *et al.*, 2003). In the FID project, potential biofidelity data sets were excluded if the test apparatus was not available to the project and would be difficult to reproduce within the project timescale. The requirements included a description of the test conditions and the target response corridors. The following limited set of thorax and shoulder biofidelity requirements were defined:

FID thorax requirements

To evaluate the performance of the thorax three tests were proposed, based on the data of Kroell *et al.*, 1971, Yoganandan *et al.*, 1997 and Vezin *et al.*, 2002b.

- Kroell *et al.*, 1971. Pure frontal mid-sternum impactor PMHS tests using a 23.4 kg impactor, at a velocity of 4.3 m/s. Mid-sternal force-compression response requirements were defined.
- Yoganandan *et al.*, 1997. Oblique lower ribcage pendulum impactor PMHS tests using a 23.5 kg impactor with a 40 mm thick Ensolite padding on the impact face, and a velocity of 4.3 m/s. Force-time and displacement-time responses were defined. Chest band deflection histories were normalised with respect to the initial chest depth. All force-deflection responses in the original paper were also scaled to a standard body

weight of 75 kg using the method proposed by Eppinger, 1976, although this appeared to increase the scatter in the force-deflection responses.

- Vezin *et al.*, 2002b. PMHS sled tests with seat-belt, 4 kN load limiter, and airbag at 50 km/hr, and seat-belt and 4 kN load limiter (no airbag) at 30 km/hr. Corridors for upper and lower sternum resultant acceleration versus time were defined. It was noted in van Don *et al.* that additional data for defining biofidelity requirements with more confidence was required (only three PMHS were tested in each restraint configuration) and would become available in future projects. Van Don *et al.* also defined T1, T8, T12 and sacrum resultant acceleration versus time corridors based on the Vezin *et al.* tests. The corridor definitions were based on mean ± standard deviation from three tests, which resulted in corridors that are in places very narrow. It should be noted that no additional tests were performed in later projects, so this data set remains comprised of only three subject per test condition.

FID shoulder requirements

One data set was proposed for shoulder requirements in the FID project.

- Vezin *et al.*, 2002b. PMHS sled tests with and without airbag (as above). Left and right acromion resultant acceleration, and left and right upper humerus resultant acceleration versus time response requirements were defined. Van Don *et al.* noted that the corridors were based on at most three tests, and that further testing should therefore be performed to refine the corridors provided.

3.5 EEVC THOR requirements (2003)

The EEVC adopted the frontal impact requirements proposed by van Don *et al.* (EEVC WG12, 2003). The EEVC subsequently held a workshop to compare the THOR-NT (developed by GESAC under contract to NHTSA) and THOR-FT (developed by the FID project), and make recommendations on which of the two designs was preferred and what - if any - design revisions were necessary. In the preamble to the report (EEVC WG12, 2006), the EEVC noted the following background information:

> *‘Anticipating the need for a next-generation frontal dummy, NHTSA took the lead by initiating the development of an advanced frontal impact dummy, working with GESAC during 1994 to 2005; this dummy is known as THOR. During this period, NHTSA was in contact with Europe regarding the requirements for the dummy. In Europe, it was recognised that there would be the need in the future for a next generation dummy that would give improved injury risk indications for, for example, the more complex interactions between the chest and seatbelts on their own or in combination with airbags and also the steering wheel and also a better measure for the risk of injury to the feet and legs.*
>
> *‘Europe wished to ensure that an advanced frontal impact dummy could interact and respond correctly to the European restraint systems, which are primarily seat belts, normally in association with airbags for front seat occupants. These European airbags act as supplementary restraint systems, rather than the primary restraint system, and often differ significantly to the US airbag design. Thus it was deemed necessary to evaluate the THOR-Alpha, the NHTSA first version, for the European condition. This was undertaken through two EC projects; ADRIA, which evaluated the THOR dummy between 1997 and 1998 and FID (2000 – 2003), which aimed at establishing design improvements. As a result of comments from EEVC and others, NHTSA introduced a revised version, manufactured by GESAC, called THOR–NT. Also FTSS produced a version to the EEVC recommendations, called THOR-FT. This has resulted in two different designs for THOR.’*

This report made a direct comparison of the EEVC and NHTSA biofidelity requirements, of which the shoulder, spine and thorax requirements are shown in Table 3-1.

Table 3-1 EEVC comparison of shoulder and thorax biofidelity requirements (from EEVC, 2006)

Test	EEVC Requirements (van Don *et al.*, 2003)	NHTSA Requirements (GESAC, 2001)	Comments
Shoulder	Vezin (2002) - NB: more tests required	None defined	Additional WG12 requirement
Spine	Vezin (2002)	None defined	Additional WG12 requirement
		Numerous tentative proposals	
Thorax	Kroell (1971)	Neathery (1974)	Identical requirements, referenced differently
	Yoganandan (1997)	Yoganandan (1997)	Identical requirements
	Vezin (2002)		Additional WG12 requirement
		Proposed - Q-S thorax regional coupling - Schneider et al. (1992)	
		Proposed - Belt loading - Cesari and Bouquet (1990)	

3.6 NHTSA THOR requirements (2005)

The NHTSA 2001 requirements were updated with new data for the specification of the THOR-NT dummy (GESAC, 2005). Again, the document includes firm requirements and proposals for possible new requirements.

NHTSA thorax requirements

- Neathery, 1974. Pure frontal mid-sternum pendulum impactor PMHS tests using a 23.4 kg impactor, with a velocity of 4.3 m/s and 6.7 m/s. Based on new data with tensed and untensed porcine subjects, the ‘correction’ of 667 N applied to the force to account for the stiffening due to muscle tensing that would be expected to occur in a living human car occupant (see NHTSA 2001 requirements above) was removed for the 4.3 m/s force-deflection response corridor. The correction was retained for the higher-speed requirement. Furthermore, NHTSA specified that the primary requirement should be the 4.3 m/s requirement, followed by the 6.7 m/s requirement.
- Yoganandan *et al.*, 1997. Oblique lower ribcage pendulum impactor PMHS tests using a 23.4 kg impactor with a 19 mm thick Rubatex padding, with a velocity of 4.3 m/s. (NB: the original paper quotes an impactor mass of 23.5 kg, and a 40 mm thick Ensolite padding; no reason is given for the change of specification in the NHTSA requirements.) Force-time and displacement-time responses were defined.

Identical proposals for additional spine and thorax requirements were made as in the 2001 NHTSA requirements.

3.7 ACEA/ISO frontal impact biofidelity requirements (draft 2010)

ISO TR9790, published in 1989, defined biofidelity corridors and test procedures in lateral impacts for the assessment of side impact dummies. ACEA is currently supporting work within the ISO/TC22/SC12/WG5 Frontal Biofidelity Specification International Task Force to develop biofidelity requirements for frontal impact dummies. To date, the Task Force has focussed on pendulum impactor requirements from Kroell-type tests (such as those used in the EEVC and NHTSA requirements above), and the most recent seat-belt only sled tests from UVA (five tests sponsored by NHTSA and three by JARI/JAMA). A draft report is available on the pendulum impactor requirements which covers:

- Selection of the included data sets
- Normalisation of the data
- Alignment/shifting of the data
- Construction of the biofidelity corridors
- Application of a correction for muscle tensing

Good communication between the THORAX WP2 partners and the ACEA/ISO Task Force has been maintained in order to avoid duplication of effort, share the knowledge and experience in both groups, and harmonise on the approach to determining the biofidelity corridors where appropriate.

It should be noted that ACEA/ISO have put no limit on the extent or severity of injury to the PMHS for inclusion in their draft frontal impact biofidelity requirements. The draft ACEA/ISO frontal impact biofidelity requirements include subjects with over 20 rib fractures. In contrast with this, the ISO side impact biofidelity requirements documented in ISO TR 9790 excluded PMHS with more than five rib fractures, except for two subjects with seven and nine rib fractures respectively that were included in one of the six thorax biofidelity test conditions. It is understood that severely injured subjects were also excluded from the EEVC side impact requirements (Roberts *et al.*, 1991b) in order to ensure that the biofidelity requirements were representative of the human response before significant injury occurs (assuming that the stiffness of the thorax decreases as the number of rib fractures increases).

The reason for the difference is that in the frontal impact requirements, ACEA/ISO have chosen to exclude subjects if their impact response was markedly different from the typical response in the data set. A difference in response may be due to, for example, different anthropometry, intra-individual differences, or to the level of structural damage to the rib cage. It is also possible that intra-individual differences could mask differences due to structural damage of the rib cage. For example, a subject who was stiffer than average could appear to have average stiffness if the rib cage stiffness was compromised by multiple rib fractures. Including this subject would not increase the available information on the response of an uninjured or slightly injured subject, but would appear to improve the confidence in the result.

3.8 Discussion

The recent frontal impact thorax, shoulder and spine biofidelity requirements from EEVC, NHTSA and ISO are shown in Table 3-2. All groups have included pendulum impactor data, with EEVC and NHTSA including both fully frontal and oblique requirements. The most recent NHTSA requirements have removed the higher-speed pendulum impactor tests.

Table 3-2 Comparison of recent biofidelity requirements

Body region	EEVC Requirements (ESV 2003)	NHTSA Requirements (2001)	NHTSA Requirements (2005)	ACEA/ISO (Draft June 2010)	Comments
Shoulder	Vezin (2002) - NB: more tests required • Sled: seat-belt only at 30 kph; belt-and-airbag at 50 kph (4 kN load limit for seat-belts in both)	None defined	None defined	None defined	Shoulder requirements were only defined by EEVC
Spine	Vezin (2002) - NB: more tests required • Sled: see shoulder	None defined *Numerous tentative proposals*	None defined *Numerous tentative proposals*	None defined	Spine requirements only defined by EEVC, although numerous proposals made by NHTSA
Thorax	Kroell (1971) • Frontal rigid impactor: 23.4 kg; 4.3 and 6.7 m/s Yoganandan (1997) • Oblique padded impactor: 23.4 kg; 4.3 m/s	Neathery (1974) • Frontal rigid impactor: 23.4 kg; 4.3 and 6.7 m/s Yoganandan (1997) • Oblique padded impactor: 23.4 kg; 4.3 m/s	Neathery (1974) • Frontal rigid impactor: 4.3 m/s Yoganandan (1997) • Oblique padded impactor: 23.4 kg; 4.3 m/s	Pendulum impactor tests based on Kroell (1971), INRETS, and CEESAR data • Frontal rigid impactor: 23.4 kg; 4.3 and 6.7 m/s	EEVC 2003 and NHTSA 2001 are identical requirements, referenced differently. NHTSA 2005 requirements drop the higher-speed pendulum tests. ISO requirements based on similar data, but lack oblique requirements
	Vezin (2002) - NB: more tests required • Sled: see shoulder			Proposed Shaw (2009) • Sled: lap and diagonal seat-belt, at 40 kph	Sled test requirements (with restraints) included in EEVC and (draft) ISO. No sled tests requirements in NHTSA 2001 or 2005.
		Proposed - Q-S thorax regional coupling - Schneider *et al.* (1992)	Proposed - Q-S thorax regional coupling - Schneider *et al.* (1992)		Proposals for possible additional requirements from NHTSA
		Proposed - Belt loading - Cesari and Bouquet (1990)	Proposed - Belt loading - Cesari and Bouquet (1990)		

In addition to the pendulum impactor tests, the EEVC requirements include sled tests that provide biofidelity requirements for the shoulder, spine and thorax (in seat-belt only and belt-and-airbag loading conditions), although the EEVC document notes that the dataset is very small and a larger dataset is recommended. ISO is currently working on including the most recent seat-belt only sled tests from the University of Virginia. NHTSA have also proposed several other test conditions that would broaden the range of loading severity and distribution over which the dummy is validated, but full requirements based on these proposals have not been published.

In general, it can be observed that the available biofidelity requirements are primarily based on pendulum impactor tests. Whilst potentially useful, particularly when oblique impacts are also included, this loading condition is of relatively low relevance to the speed and distribution of loading from modern restraint systems. The Vezin sled tests use loading conditions that are more relevant to the in-vehicle situation. When these tests were specified by the EEVC it was expected that additional tests would be added to the dataset to increase the confidence in the biofidelity corridors, because only three PMHS had been tested with each restraint configuration. However, no further tests have been undertaken and this dataset therefore remains smaller than is desirable for setting biofidelity requirements.

All of the requirements reviewed cover a limited range of impact types, load distributions and loading rates. The NHTSA requirements have started to address this by proposing additional data sets that could be used to define possible future biofidelity requirements. This is in line with earlier recommendations from the EEVC (1996), which was keen to ensure that the dummy had good biofidelity across a range of impact severities and load distributions relevant to real-world accidents, and particularly to ensure that any future frontal impact dummy should be suitable to assess the risk of injury in more frequent, lower severity impacts.

4 Criteria for the selection for biofidelity requirements

4.1 Prioritisation of biofidelity test conditions

The selection of biofidelity requirements for a crash dummy must take into account a range of factors, such as:

- The intended application for the dummy (which defines the likely loading severity, load distribution etc.);
- The level of injury severity that is targeted for reduction; and
- The availability of suitable biomechanical data (including the response of volunteers or PMHS in relevant loading conditions, and detailed descriptions of the test conditions that enable the tests to be reproduced accurately).

Within the THORAX project, it was identified that the velocity of loading (the loading rate) to the thorax from modern restraints was typically 1 $m.s^{-1}$, based on crash test results with the Hybrid III dummy and on PMHS tests (Been *et al.*, 2010). It was also identified that higher loading rates could occur in some circumstances (such as certain combinations of occupant, seating position, and restraint tuning), and it was therefore recommended that the THORAX demonstrator should be designed for a loading rate of 1-4 $m.s^{-1}$.

It was also considered important that a range of loading conditions are represented in the biofidelity test conditions, to maximise the likelihood that the THORAX demonstrator will respond appropriately whatever combination of restraint loads are applied to it. It was therefore proposed to target the design, and therefore biofidelity requirements, to the following loading conditions:

- Focus on the type of loading typically applied by modern restraint systems
    - No Out-Of-Position (OOP) capability required if this is likely to compromise the in-position assessment of modern restraint systems
    - Restraint loading with a loading rate of 1-4 $m.s^{-1}$
    - Capable of discriminating between different restraint conditions (such as different combinations of airbag and seat-belt loading)
- Distributed and localised loading
    - Including human-like rib stiffness distribution over the chest height
- Low-inertia loading
- Must be sensitive for 5% risk of AIS 3 (lower end of Euro NCAP green rating)

This information was presented at a Stakeholder workshop held in London on the 11th of May, 2010. The Stakeholders included representatives from regulators, industry, academia and research laboratories from around the world. The Stakeholders recommended that the THORAX demonstrator should definitely include:

- Planned updates to THOR
- Improved shoulder based on the SD-2
- Improved rib-to-rib stiffness distribution
- Instrumentation that facilitates investigation of advanced (e.g. rib strain or curvature related) injury assessment potential and future R&D

The Stakeholders' recommendations imply that confirmation of chest curvature (e.g. by comparison with chest-band[1] measurements) under various loading conditions, as well as localised rib cage stiffness requirements, should be included in the biofidelity requirements.

[1] A transducer for reconstructing the external shape of the rib cage based on multiple (typically 40) strain gauge sensors distributed around the circumference of the thorax.

Furthermore, the industry partners within the THORAX project were asked to comment on the expected load range and loading velocity expected for future restraint systems. The Partnership for Dummies and Biomechanics (PDB) and Continental provided the following comments (see Been *et al.*, 2010):

- Modern restraint systems already use the available space to decelerate the occupant, sometimes with individually optimised load levels (multi-stage airbags and load limiters), so it is not expected that the loads that act on the occupant in a standard crash will be reduced dramatically within the next few years
- A reduction of the loads might be possible by active pre-crash systems
- However, there will always be scenarios, where a pre-crash detection is not possible, and the normal crash pulse will act on the occupant

This suggests that a typical loading velocity of 1 $m.s^{-1}$, and an upper loading velocity of approximately 4 $m.s^{-1}$, would be an appropriate range of loading severity for the biofidelity requirements.

Furthermore, PDB and Continental provided the following additional recommendations for the THORAX demonstrator:

- *'Biofidelity*
    - *Dummy should be able to reliably detect relevant injury patterns*
    - *Development of injury risk curves according to the ISO-methods*
    - *Biofidelity assessment according to ISO proposal*
    - *A balance between biofidelity and usability has to be considered (more human-like → more complexity → more risk for variances)*

- *Sensitivity:*
    - *Capability to distinguish different restraint systems (good – acceptable – poor)*
    - *High sensitivity is required for modern restraint systems with standard or adaptive airbags and belt systems with pretensioners and load limiters in the range of (2-5 kN)*
    - *Good sensitivity at lower loading conditions (e.g. < 20 mm chest deflection equivalent to Hybrid III)'*

The considerations in this section set limits on the type and severity of test that should be considered for setting biofidelity target corridors. For instance, high-speed rigid impactor data, such as the high-speed Kroell-type impactor test specified in most previous biofidelity requirements (with the exception of the most recent NHTSA requirements), is not considered relevant, because it has neither the load distribution or loading rate that the dummy will experience in crash tests. Lower speed impactor data may still be considered, although the loading type is not of the highest priority because it is not representative of loading from modern restraint systems.

4.2 Inclusion criteria

In addition to the general requirements on the type of test, severity of loading and so forth, criteria were defined to the quality of each relevant data set. These criteria included, for example, whether the test conditions are well documented, and whether the subjects were representative. The following inclusion criteria were agreed by the THORAX project consortium:

- Only include tests where the setup can be adequately reproduced:
    - Well documented test conditions
    - Preferably with matching dummy test data available (repeat tests with the same dummy type can be used to confirm that the test conditions have been reproduced correctly)
    - Restraints can be purchased/reused: (e.g. airbag, seat-belt, pre-tensioner, load limiter, seat, and knee bolster)
    - For chest deformation: either film targets or chest band
        - For three-point belt, a chest band with >40 strain gauges
        - For airbag-only restraint, fewer gauges is acceptable
- Body Mass Index (BMI) of subjects in the range 18 – 27 (i.e. not slight or obese)
- If PMHS subjects used, fresh or frozen subjects only (no embalmed PMHS)
- No ‘excessive’ degenerative changes to the ribcage and spine
- Proper pre-test seating posture (not out-of-position tests)
- No more than 12 rib fractures

Priority was given to data sets with larger numbers of suitable subjects and without repeated tests with the same subject. Where any exceptions were made to the above criteria these are made clear in the relevant Appendix. In particular, some requirements include subjects with more than 12 rib fractures or with a wider range of BMI where it was not possible to identify a difference in response due to the injury or BMI level.

4.3 Comparison with the ACEA/ISO exclusion criteria

The ACEA/ISO draft frontal impact biofidelity requirements (Lebarbé, 2010) used the following exclusion criteria

- Insufficient description of the test set-up *(equivalent to the THORAX inclusion criterion)*
- Force-deflection measurements not available
- Multiple impacts to the same PMHS were assessed on a case-by-case basis, particularly with respect to the impact speed and injury outcome
- Poor PMHS condition *(equivalent to the THORAX inclusion criterion)*
- Embalmed PMHS *(equivalent to the THORAX inclusion criterion)*

Furthermore, some data sets were considered ‘useable’ for biofidelity corridor definition, but were not included in the work reported to date because of their small sample size. No limit was set on the number of fractures sustained by a subject. Instead, the shape of the force-deflection response was assessed independent of the injury level. If the response was similar to the other tests in the same test condition, it was included, otherwise it was excluded.

In general, the THORAX inclusion criteria and the ISO exclusion criteria result in a similar definition of suitable test conditions. The most important difference relates to the exclusion of excessively injured subjects (≥12 rib fractures) using the THORAX criteria. These subjects may be included in the ISO requirements, provided that the form of the force-deflection response was similar to that of less severely injured subjects.

5 Review of candidate biofidelity data sets

A detailed literature review was undertaken to identify candidate biomechanics data sets that could be considered for use in setting biofidelity requirements. Candidate data sets were identified from a number of sources:

- The European Biomechanical Experiments Database, which is currently hosted on the Humos2 EC Framework Project web site
- The NHTSA Biomechanics Test Database, which is hosted on the NHTSA web site
- Previous documentation of frontal impact biofidelity requirements (e.g. Section 2)
- Publications in the scientific literature

In total, more than 45 candidate data sets were reviewed. All of the identified candidate data sets were assessed against the data prioritisation guidelines and inclusion criteria identified in Section 4.1 and 4.2 respectively. The reviews of the candidate datasets may be found in Appendix A, and the outcome of the review is summarised below.

The main reasons that data sets were excluded were:

- The use of embalmed PMHS subjects;
- Insufficient information on the test set-up to allow the tests to be reproduced accurately with a dummy, and this information probably cannot be recovered; or
- No force-compression, or similar detailed chest response data available.

A sub-set of the evaluated data sets were considered as suitable for defining biofidelity requirements, *and* suitable for replication within the THORAX project (or at any well-equipped crash test or biomechanics laboratory). These requirements are summarised in Table 5-1 and Table 5-2. Table 5-1 shows the selected data sets that give absolute biofidelity requirements, i.e. those that result in, for example, specific targets for the deflection of the mid-sternum in a particular loading condition. Table 5-2 shows the selected data sets that give relative biofidelity requirements, i.e. those that result in, for example, relative compression targets for different regions of the thorax, or under different load distributions. **Error! Reference source not found.** to Appendix K show the detailed biofidelity requirements for the selected data sets.

Another sub-set of the evaluated data were identified that were assessed as being potentially suitable for defining biofidelity requirements with several constraints:

- In some cases additional information on the test set-up would be required in order to reproduce the test conditions accurately, but it was considered that it may be possible for this information to be recovered because the original authors are still active in the field. In most cases, contacts have been made with the original authors, but - as yet - no further information has become available. It is expected that these data sets cannot be converted to biofidelity requirements within the timescale of the THORAX projects, but they may become available in the future.
- The complexity of the test set-up was judged to be high, such that it is probably only practicable for the original authors to reproduce the tests with a dummy.
- Some data sets were judged to be of good quality, but very small (e.g. only three subjects per test condition). Where the test set-up would be expensive to reproduce it would not be practicable to reproduce the tests within the project. However, it should be noted that it is possible that these tests could be reproduced by the original authors, which would add to the evaluation of the Demonstrator.

These potential biofidelity requirements are summarised in Table 5-3 and Table 5-4 for absolute and relative requirements respectively. Table 5-1 to Table 5-4 also define whether the data set is suitable for defining thorax biofidelity requirements or shoulder biofidelity requirements, and whether it should be considered for thorax injury assessment tests in WP2.6.

Some of the reviewed data sets were considered not to be suitable for defining biofidelity requirements, but possibly suitable for defining injury risk functions. These data sets are highlighted in Table 5-5 as a contribution to THORAX Task 2.6.

Table 5-1 Recommended data sets – Biofidelity requirements / absolute requirements

Author / Appendix	Description	Restraints / Loading	Chest compression measurement	Shoulder displacement measurements	Injuries	Thorax biofidelity requirements	Shoulder biofidelity requirements	Injury risk functions
Lebarbé (2010) including data from Nahum *et al.* (1970) Kroell *et al.* (1971), Kroell *et al.* (1974), Bouquet *et al.* (1994)	Pendulum tests, Response corridors using 11 PMHS. 22.3 or 23.4 kg at an impactor velocity of 4.7 or 4.3 m/s	Hub	Deflection and load on sternum	None	Multiple rib fractures	✓	✗	✓
Yoganandan *et al.* (1997)	Pendulum tests. Oblique. 7 PMHS, of which 5 meet the inclusion criterion, exposed to a 23.5 kg pendulum impact at 4.3 m/s	Hub	Deflection and load on chest	None	Commonly 2-4 fractures per subject	✓	✗	✓
Forman *et al.* (2006a) (not including Shaw *et al.* (2000))	Sled tests. Passenger position. 9 PMHS of which 8 PMHS meets the inclusion criterion at 48 and 29 km/h in Ford Taurus 1997 buck	3-point belt; force limited 3-point belt; and 3-point belt with airbag	Chest bands, high resolution, in addition to T1, T8, and T12 accelerometer arrays	Photo targets on clothing/wrapping only, difficult to observe the T1 photo target	Dependent on restraint: low or none for all but one of the airbag tests	✓	✗	✓
Bolton *et al.* (2006)	Sled tests. Passenger position. 3 PMHS at 48 and 29 km/h in Ford Taurus 1997 buck	Airbag with knee bolster and lap belt	Chest bands, high resolution, in addition to T1, T8, and t12 accelerometer arrays	Photo targets on clothing/wrapping only	No rib fractures	✓	✗	✓
Törnvall *et al.* (2008)	Sled tests. 9 tests with 3 PMHS in 0° full frontal 45° far-side and 30° near-side collisions at 27 km/h in rigid seat	3-point belt	Three-dimensional film targets on sternum and T1 only	Three-dimensional film targets on the shoulder and T1, upper arm mounted accelerometer	No fractures (detection method palpation)	✗	✓	✗

Table 5-2 Recommended data sets – Engineering guidelines / relative requirements

Author / Appendix	Description	Restraints / Loading	Chest compression measurement	Shoulder displacement measurements	Injuries	Thorax biofidelity requirements	Shoulder biofidelity requirements	Injury risk functions
Cavanaugh *et al.* (1988)	Table top tests. Static. 2 PMHSs	Chest loading plate; 4.5 cm x 10 cm	Chest deflection at eight location	None	Sub-injury and injury level	✓	✗	✗
Césari and Bouquet (1990)/ Riordain *et al.* (1991)/ Césari and Bouquet (1994)	Table top tests. Dynamic. 7 PMHS, of which 3 meet the inclusion criterion, at 3 m/s loading rate using a 22.4 kg impactor	Diagonal shoulder belt	Chest deflection at eleven locations	Mid clavicle bone displacement	No fractures	✓	✓	✓
	Table top tests. Dynamic. 7 PMHS, of which 4 meet the inclusion criterion, at 7.3 m/s loading rate using a 22.4 kg impactor	Diagonal shoulder belt	Chest deflection at eleven locations	Mid clavicle bone displacement	0-17 rib fractures (0-7 for those that meet the inclusion criterion	✓	✓	✓
	Table top tests. Dynamic. 7 PMHS, of which 4 meet the inclusion criterion, at 2.4 m/s loading rate using a 76.1 kg impactor	Diagonal shoulder belt	Chest deflection at eleven locations and with two chest bands	Mid clavicle bone displacement	2-18 rib fractures	✓	✓	✓
Kent *et al.* (2004)	Table top tests. Quasi-static. 15 PMHS, of which 12 meet the inclusion criterion, loading rate 1 m/s	Hub, 2 point belt, 4 point belt, or belt for distributed load	Chest deflection and reaction force	None	Sub-injury and injury level	✓	✗	✓
Shaw *et al.* (2007)	Table top tests. Quasi-static or dynamic. 5 PMHS using load rate 1 m/s	Chest loading plate; 6,2 cm x 6,2 or 11,3 cm	3-D chest deflection at approximately 10 locations and reaction force	None	Ranged from 4 to to 17 rib fractures per subject	✓	✗	✓

Table 5-3 Potential data sets – Biofidelity requirements / absolute requirements

Author	Description	Restraints / Loading	Thorax measurements	Shoulder measurements	Injuries	Thorax biofidelity reqs	Shoulder biofidelity reqs	Injury risk functions
Yoganandan *et al.* (1993)	Sled tests. Driver position. 14 PMHS at 32 or 47 km/h in Ford Tempo 1986 buck	Different restraints. Important data set comprise 5 PMHS tests using an airbag with lap belt	Chest bands (24 to 40 gauges)	Not available	For lap belt and airbag in combination only 2 rib fractures in average	✓	✗	✓
Rouhana *et al.* (2003)	Sled tests. 7 PMHS of which 7 meet the inclusion criterion, at 40 km/h in specially made seat	Mainly force limited 4 point belt system incl. dual lap belt pretensioner	Trans-thoracic rod technique in combination with film analysis or string potentiometer	Not available	5 rib fractures in average for the 4 point belt tests, 24 rib and clavicle fractures in 3 point belt, spine injuries and sternum fractures for both restraints	✓	✗	✓
Shaw *et al.* (2009b)	Sled tests. 8 PMHS, of which 6 meets the inclusion criterion, at 40 km/h in rigid/cable seat	3-point belt (separate lap and shoulder belt)	Multiple 3D-film targets on rib cage, sternum and along the spine and accelerometer arrays	Three-dimensional film targets on shoulder (left and right acromion)	5 rib fractures in average, clavicle fractures in 2 and 1-2 sternum fractures in all but one subject	✓	✓	✓

Table 5-4 Potential data sets – Engineering guidelines / relative requirements

Author	Description	Restraints / Loading	Thorax measurements	Shoulder measurements	Injuries	Thorax biofidelity reqs	Shoulder biofidelity reqs	Injury risk functions
Fayon *et al.* (1975)	Table top tests. Static. 7 PMHS and volunteers	Diagonal shoulder belt	Sternum	None	8 rib fractures in average	✓	✗	✓
Salzar *et al.* (2008) and Lessely *et al.* (2008)	Table top tests. Quasi-static. 3 PMHS using ramp-hold or sinusoidal load at rates 05 to 1.2 m/s.	Loaded by a diagonal shoulder belt	Chest deflection and reaction force	Arms were on representative posture	-	✓	✗	✗

Table 5-5 Data sets recommended for injury risk functions only

Author	Description	Restraints / Loading	Thorax measurements	Shoulder measurements	Injuries	Thorax biofidelity reqs	Shoulder biofidelity reqs	Injury risk functions
Stalnaker *et al.* (1973)	Pendulum tests, PMHS exposed to 10 kg probe impacts at 5.35-6.71m/s	Hub	Sternum deflection and contact load between pendulum and subject	None		✘	✘	✓
L'Abbe *et al.* (1982)	Table top tests, Dynamic. 10 volunteers. 3.6 kN over 60 ms	Diagonal shoulder belt	Chest deflection at eleven locations	Mid clavicle bone displacement	None	✘	✘	✓
Yoganandan *et al.* (1991)	Sled tests. 15 PMHS of which 9 PMHS meet BMI and NRF inclusion criterion at 24 or 50 km/h in Ford Tempo 1986 buck	3-point belt, no steering assembly	Chest bands (either 24 or 34 gauges)	Not available	Multiple rib fractures and sternum and clavicle fractures	✘	✘	✓
Petitjean *et al.* (2002)	Sled tests. 4 PMHS at 64 km/h using production seat and restraints	Force limited 3-point belt and airbag combinations	Not available	Not available	15 rib fractures on average	✘	✘	✓
Vezin *et al.* (2002)	Sled tests. 6 PMHS, of which 5 PMHS meets the inclusion criterion, at 30 km/h and 50 km/h using a and rigid seat and commercial restraint systems	Force limited 3-point belt (separate lap and shoulder belt) and force limited 3-point belt with airbag	Spine accelerometer data only	Not available	6 rib fractures in average for the airbag test and 2 rib fractures in average for the belt only tests	✘	✘	✓
Lebarbé *et al.* (2005)	Out-of-position tests, 9 PMHS stationary distributed on 5 different loading conditions	Airbag deployment, membrane and punch-out loads	None	None	Ranged from 0 to 23 rib fractures	✘	✘	✓
Trosseille *et al.* (2008)	Out-of-position tests, 8 PMHS stationary (2 of these were frontal impacts)	Pendulum and airbag, membrane loads	None	None	Range from 3 to	✘	✘	✓

6 Normalisation and scaling

In order to specify biofidelity requirements for a mid-size average male dummy, it is important that the requirements are consistent with expectations for an occupant of 50th percentile male size. This is important because a 50th percentile dummy will have to meet the appropriate anthropometry measurements (such as the length and mass of each body segment). However, the subjects that were tested are unlikely to have been exactly the same size as the dummy specification. Every subject will have unique anthropometry, including their individual geometry and inertial properties. Any deviations between the 50th percentile and the individual subjects need to be taken into account and typically this is done by adjusting the response according to the anthropometry. Making adjustments to the responses of test subjects, to bring them in line with the average size, is called normalisation.

Sometimes it is necessary to specify requirements for a size other than the average male, a size outside of the general range of subjects in the test sample. Care must be taken when scaling beyond the sample anthropometry range because there may not be any immediate validation of the process being used. However, in principle this can be done using exactly the same algorithms as would be used in normalisation within the sample size range. As a general concept we can, therefore, treat scaling as normalisation to a body size other than that of the 50th percentile.

In addition to considering only subject anthropometry, depending on what is known about the test subjects and their responses, it is possible in certain situations to account for dynamic properties. For instance, the effective mass and effective stiffness of the thorax can be taken into account; although stiffness considerations are often simplified and based on knowledge of subject anthropometry.

6.1 Review of available normalisation and scaling methods

The ACEA Task Force on frontal biofidelity performed a substantial review of scaling techniques which could be used in the normalisation of thorax biofidelity responses. Many of these discussions and the background work is relevant to this task of the THORAX project. Therefore, throughout this section due regard will be given to the efforts of the ACEA group.

In working for the ACEA Task Force, Lebarbé, 2010 identified three distinct normalisation methods from the literature. Each method was based on one of the following concepts.

- Use of a regression equation taking into account subject details (statistical method)
- Mass-based scaling of responses
- A spring-mass simplification of the physical system to derive the important parameters

The following sections review the different normalisation and scaling methods that have been used grouped either by type (e.g. statistical methods) or by the type of test for which they have been used (e.g. for mass-spring models, for which the model may be applicable only to a specific loading condition).

6.1.1 Statistical methods

As an example of the regression equation approach, Neathery, 1974 used this method to predict peak force (during the plateau period in the impactor test data) based on subject anthropometry and impact conditions. This allowed Neathery to normalise peak force and deflection values knowing the subject anthropometry and impact conditions for each test. In fact, two approaches to the regression were used: one where parameters were selected for inclusion based on their expected relevance to the impact response; and one where many more parameters were included, and less relevant parameters excluded through the statistical modelling. In this case, both approaches resulted in similar normalisation factors. The same approach was then used to develop corridors for the 5th and 95th percentile occupant.

Krause, 1984 also used regression techniques to normalise side impact PMHS dynamic response data. In this case the measured force during drop tests or Heidelberg sled tests were equated on the basis of subject chest depth, weight, and the test drop height or impact speed.

The statistical approaches to normalisation and scaling have the benefit that they allow many parameters to be evaluated for their possible influence on the subjects' responses. Combinations of parameters can also be evaluated; however, the number of parameters and complexity of combinations that can be assessed is limited by the sample size in most biomechanics data sets.

6.1.2 Methods using simple global scaling factors

Eppinger *et al.*, 1984 set out a regime for response scaling on the basis of subject mass. A premise behind the formulae derived by Eppinger *et al.* was that mass density and modulus of elasticity were constant between test subjects. This led to fairly simple scaling equations for acceleration (*A*), length (*L*), time (*T*), and force (*F*); as shown below.

$$A_s = \lambda_m^{\frac{-1}{3}} A_i$$

$$L_s = \lambda_m^{\frac{1}{3}} L_i$$

$$T_s = \lambda_m^{\frac{1}{3}} T_i$$

$$F_s = \lambda_m^{\frac{2}{3}} F_i$$

For this application,

$$\lambda_m = \frac{M_s}{M_i}$$

Where: M_s is the standard mass, and M_i is the mass of the test subject.

For scaling the time, the interval between data points should be scaled by the appropriate factor. Then the data will need to be re-sampled to provide a common rate between all tests.

Using this method, forces are scaled based only on the mass of the subject. Other methods have included a characteristic length (such as thorax depth) as an approximation to account for individual variations in thorax stiffness (based on the geometrical assumption that a smaller diameter rib cage is likely to be stiffer). As noted by Eppinger *et al.*, their scaling algorithm does not change the velocity of the test. This makes such an approach unsuitable for normalising e.g. some of the frontal impact pendulum test series where a variety of impact speeds were used. On this basis the ACEA Task Force ruled out the use of this method. Instead they opted for a mass-spring method.

Along with the regression equation and mass-based scaling techniques, a mass-spring method was also reported by Mertz, 1984. As in the case of the regression equation, this method was developed as a means of normalising PMHS lateral thoracic impact response data (whole body drop tests). As with all scaling and normalisation procedures, care should be taken when these are applied to a different data set to that for which they were developed (see Section 6.1.5).

The Mertz normalising factors for time (*t*), force (*f*), acceleration (*a*), velocity (*v*), and displacement (*x*) are shown below.

$$R_t = (R_m)^{\frac{1}{2}}(R_k)^{\frac{-1}{2}}$$

$$R_f = (R_m)^{\frac{1}{2}}(R_k)^{\frac{1}{2}}$$

$$R_a = (R_k)^{\frac{1}{2}}(R_m)^{\frac{-1}{2}}$$

$$R_v = 1$$

$$R_x = (R_m)^{\frac{1}{2}}(R_k)^{\frac{-1}{2}}$$

Where the mass and spring stiffness ratios, R_m and R_k, are the mass of the standard subject divided by the mass of the particular subject tested, and similarly the stiffness for the standard divided by the particular subject.

To derive the stiffness ratio between subjects, Mertz approximates the structures of thoraces to be geometrically similar. By then assuming that the elastic moduli are equivalent between all subjects, Mertz deduced that the stiffness is proportional to some characteristic length. This also sets the proportionality constant between the characteristic length and a generic length in that dimension to be equivalent for all subjects.

However, the assumption that the elastic moduli are equivalent for all subjects is quite crude. For example, Burstein *et al.*, 1976 showed that the femur tensile elastic modulus for a sample of subjects aged 40-50 years was 17.7±4.45 GPa, which gives ratios of 0.75-1.25 compared with the mean. This is comparable to a chest depth range of 180 to 300 mm, for a mean chest depth of 240 mm, which is similar to the range for the Kroell-type subjects listed by Lebarbé, 2010. That is, subject factors based on bone moduli are likely to have a similar range to those based on geometry, but have been assumed to be unity.

Deviation of the subjects from these assumptions and approximations will reduce the validity of the normalisation. The effective stiffness can be determined if force-deflection data is available from a tests, but the effective stiffness of the target group (e.g. 50th percentile male) is unknown and is typically estimated for a given data set using parameters such as subject mass.

6.1.3 Normalisations used for specific types of data

6.1.3.1 Normalisation of impactor test data

In 1989 Viano adapted the Mertz mass-spring system to include a second mass (Viano, 1989). This was used in the normalisation of responses in blunt lateral impacts to the thorax, abdomen, and pelvis. The normalisation factors were adapted to include terms which describe the masses of the pendulum, PMHS, and standard subject.

The formulae developed by Viano were used again by Shaw *et al.*, 2006 when considering the oblique and lateral response of the PMHS thorax, to allow for inter-subject comparisons. This is also the method selected by the ACEA Task Force (Lebarbé, 2010), who began their development of frontal impact requirements by investigating the pendulum testing of Kroell *et al.*, 1971.

To make use of the Viano equations, a method for calculating the effective mass of each subject's thorax is needed. Conventionally, the effective mass is based on conservation of momentum and balancing the impulse for the pendulum and the subject. The velocity of the subject and pendulum together is obtained by integrating the thoracic acceleration measured at the spine, where available. Unfortunately for the ACEA Task Force, only nine of the pendulum impact tests had spine accelerations available. This meant that another method of deducing the combined pendulum and subject velocity was required, and two options which used the peak pendulum deflection were proposed. It was subsequently decided that the following equation gave the best reduction in the scatter of the subjects' responses.

$$m_2 = \frac{2E_d m_1}{m_1 v_0^2 - 2E_d}$$

Where,	m_2	=	effective mass of the subject
	m_1	=	mass of the pendulum
	v_0	=	impact velocity
	E_d	=	energy of deformation

Here $E_d = \frac{1}{2}\left(\frac{m_1 m_2}{m_1+m_2}\right) v_0^2 = \int_{x=0}^{x\,max} F.dx$

To find the most appropriate stiffness ratio, the ACEA Task Force used the assumption that normalisation necessarily shrinks the envelope of a set of curves. Thus, for a given set of tests, the best stiffness ratio is the one that provides the smallest envelope of response curves. In order to compare, in an objective manner, the efficiency of the normalisation induced by each stiffness ratio, the cumulative CV (coefficient of variation), which indicates the degree to which the envelope of responses is 'collapsed', was calculated for each method and compared from one to the next. It was found that using the ratio of chest depth provided a useful stiffness ratio for this particular data set; conforming to the assumptions that the PMHS are geometrically similar and the Young's Modulus of tissues is consistent from one PMHS to another.

Despite using this approach, Lebarbé lists particular shortcomings of the mass-spring model:

- Linearity of the spring and the absence of damping in modelling the thorax. No viscous effect is taken into account in this model. As shown below, the assumed response (Figure 6-1) is markedly different to the typical thorax force-compression response (e.g. Figure 6-2).
- The use of an effective mass for the PMHS which is theoretically significant only at the time for which it has been calculated – the time when the deflection is maximum – in the sense that it represents the mass necessary to fulfil the equation of energy.
- In addition, the scaling ratios generally use macroscopic PMHS characteristics – the anthropometry data – to calculate the stiffness ratio. This requires gross assumptions as described previously.

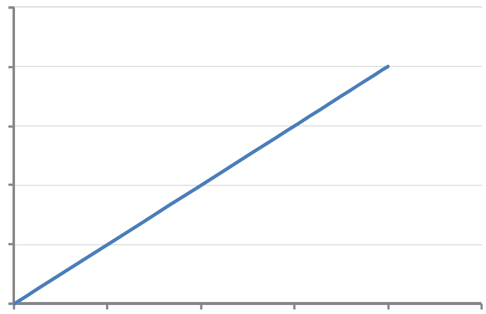


Figure 6-1 Assumed linear spring thorax response

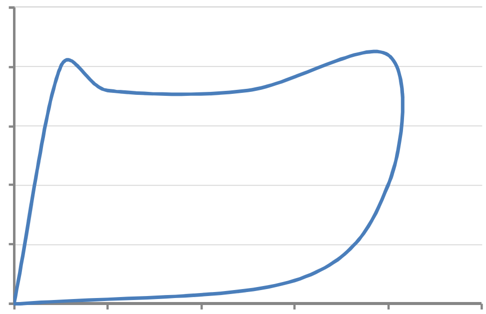


Figure 6-2 Typical actual thorax response

Lebarbé suggested that one way to evaluate the relevance of the normalisation process would be to:

- Select a subject from the data set
- Normalise all of the other responses using e.g. the chest depth of the selected subject as the standard value
- Compare the normalised responses with the response selected subject

If the normalisation is reliable, the normalised responses should be close to the response of the selected subject. However, it should be noted that this assumes that each subject has a response that is typical of subjects of similar size and this will not always be the case. In theory this check could be performed for every subject in the dataset to increase the confidence in the normalisation process, and this may be particularly important if the same process is to be used to scale to an outlying subject stature such as a 5th or 95th percentile.

6.1.3.2 Normalisation of sled test data

Test subject behaviour during a sled test represents a complicated interaction between the subject and the seat and restraint system. Many factors will influence the response of the subject. This makes normalisation and scaling a difficult task, because in an ideal world all the complex interactions should be modelled to provide a robust prediction of a typical response. Unfortunately, it is not realistic to try and consider every aspect of a test in the normalisation process. Instead a simple way of accounting for subject-to-subject differences is required, for which reasonably simple mathematical functions can be used to describe such differences.

Lebarbé, together with the ACEA Task Force, attempted to satisfy this need by applying a modified mass-spring model to the belt-restrained frontal impact test environment. This approach was compared with the Eppinger mass-based scaling and it seems as though there is little benefit of using one technique over the other for the kinematic measurements.

From a subjective point of view, it seems advantageous to use the same scaling practice for all types of testing. However, in practical terms, this may be less than ideal. For instance, at the very least, the characteristic length when scaling different body regions will change. Also, different types of test (impactor, sled, table-top) may need refinements or changes in the scaling process to remain relevant. Whilst Lebarbé is considering only the thorax segment in his definition of biofidelity requirements based on the Shaw *et al.*, 2009b tests, the THORAX partners have identified that it is also important to define the global kinematics of the subjects in order to ensure that the loading to the thorax segment in dummy tests is representative of that to the PMHS subjects (i.e. to ensure that the input to the thorax is representative of the PMHS tests). This means that equivalent corridors for other global responses are required.

The THORAX Project is seeking to define requirements at several different levels throughout the body. This puts considerable pressure on the universality of a particular scaling technique.

If the selection of different characteristic lengths for different body regions is considered then, according to the assumptions stipulated by Mertz, this is unnecessary because the relationship between one characteristic length to another (in the same dimension) was deliberately set to be equivalent for all subjects. In practice this will not be the case. It is very unlikely that a nominally 5th percentile person would be 5th percentile in every aspect. Any normalisation needs to pick the appropriate characteristic length to describe the response. If this is not the case, then the consequence of the deviation from the assumption is that the normalisation for a given subject may stretch them in an inappropriate way. To demonstrate this problem it may help to consider a short person with long arms or vice versa. For such a person scaling arm measurements according to overall stature would be inappropriate.

As a further example of the difficulties in using a single standard scaling approach for all body regions, consider the shoulder or head in a belt-restrained frontal impact. Here the difference between thorax response scaling and head scaling in a mass-spring model is that the spring for the head must also include the behaviour of the neck and all joints between the thorax and occipital condyle. More fundamentally, the influence of other kinematic aspects may affect the response at the head differently to the response at the thorax. For instance, the amount of global head excursion is likely to be related to the ability of the subject to rotate (yaw) out of the shoulder belt during the impact. This will be related to the initial position of the shoulder belt on the clavicle and the amount the belt slips towards the shoulder during the impact. It may be that conventional scaling based on subject anthropometry does not describe, very effectively, differences such as these.

On a similar note, Shaw *et al.*, 2009b note that the thorax displacements in their tests were related to the distance between the shoulder belt prior to the test and the measurement position being considered. They show a correlation between x-axis deflection and y-axis distance of the belt from the measurement site. Factors such as this may be more influential to the response than simple chest depth and occupant mass. In which case, traditional scaling will never account for the main part of the sample variance. It remains to be determined if normalisation of a sample such as the Shaw *et al.* PMHS tests is a reliable, robust, and useful technique when a single scaling technique is applied to several body regions. It is possible that opting for no normalisation, or simple normalisation according to mass or overall stature, could provide most of the benefit that a complex scaling would offer without incurring such a long processing time. If this is the case then a pragmatic decision to choose a simple or no scaling could be considered. This is evaluated further in Section 6.2.

6.1.3.3 Normalisation of table-top test data

For table-top tests, the mass-spring considerations may be reduced to just a spring. Normalisation would consider only the stiffness of the subject. In this case scaling according to chest depth may be the most appropriate approach, although this omits intra-individual differences in bone modulus, as discussed in Section 6.1.2. However, the THORAX project has defined table-top test data as relative biofidelity requirements; because normalisation scaling factors are multiplicative and relate to the thorax as a whole they will not change relative requirements, so normalisation is not required.

6.1.4 Previous EEVC and ISO normalisation methods

This section gives a brief overview of the scaling methods previously used by EEVC and ISO for developing biofidelity requirements.

6.1.4.1 EEVC side impact biofidelity requirements

The EEVC biofidelity requirements that were used to assess the EuroSID-1 and ES-2 side impact dummies (Roberts *et al.*, 1991b) used a normalisation process proposed by Mertz, 1984 and Lowne (reported in Roberts *et al.*, 1991b). Impactor data was normalised using the Lowne mass-spring-mass model; sled and drop test data were normalised using the Mertz single mass-spring model (see Section 6.1.2) because the effective mass of the striking object was infinite. A standard effective mass of the shoulder (in impactor tests) of 20.5 kg, of the thorax (in impactor tests) of 29.6 kg, and of the upper wall (which was impacted by the thorax and arm in sled tests) of 37.0 kg was defined. The effective masses were derived from the original PMHS data, e.g. by deriving the effective mass when the impactor velocity was equal to the velocity of the spine for thorax impactor tests. It was noted that the exact effective mass used is not too critical for normalisation, provided that the same approach is used both for the dummy and the PMHS.

Prior to normalisation, all of the responses were time-shifted ‘by eye’, because no clear time-zero definition existed for most of the response data. No adjustment was made to account for muscle tone. Furthermore, the EEVC applied the same normalisation process to the dummy data when evaluating the biofidelity of the EuroSID-1 dummy (Roberts *et al.*, 1991a) prior to that dummy being introduced in to regulations. The dummy normalisation factors listed in Roberts *et al.* ranged from 0.73 to 1.26, depending on the body region and test type.

6.1.4.2 ISO TR 9790 side impact biofidelity requirements

ISO used essentially the same normalisation method as EEVC, except that different standard effective masses were used, and a different sub-set of the available PMHS data was selected for inclusion. Not all of the ISO requirements were normalised, but where the PMHS data had been normalised the same normalisation process was required for the dummy data (see e.g. Appendix N of ISO, 1999).

6.1.5 Discussion

The goals of normalisation are twofold:

1. Shift the response data to represent better a particular occupant, e.g. 50th percentile male;
2. Reduce the scatter in the observed responses, which is considered to improve the confidence in the resulting biofidelity target.

It is easy to appreciate why these goals are considered to be desirable. However, the normalisation has usually been attempted using very simple models to represent the complexity of the structures and response of the human body to loading. Furthermore, the most appropriate model is sometimes determined by trying several and selecting the one that reduces the scatter the best. This has several limitations. Firstly, there is no explicit check that the shift in the data (Goal 1 above) is appropriate - i.e. that it really shifts the data *towards*, not away from, the 50th male response. Secondly, the selected method may be different for different data sets even if the type of test is very similar, because of the natural variability in the small samples that are typical of biomechanical studies.

This may be adequate for localised responses in localised tests, such as pendulum impacts to the thorax, where the influence of other body regions on the response may be relatively small. However, in the complex environment of a sled test - where the subject may interact with a seat, knee bolster, lap belt, diagonal belt, and airbag - it is not clear how the different approaches could be applied to give consistent normalisation for all responses.

As a minimum, it is recommend that the checks on the normalisation process proposed by Lebarbé, 2010 are used. However, it should be noted that this approach assumes that each subject is representative of other subjects of similar stature, but this is unlikely to be the case for all subjects.

Both the EEVC and ISO defined the requirement to normalise the dummy test data in exactly the same way as the human subject data. This ensures the comparability of the dummy data and the biofidelity corridors. This is particularly important when the normalisation factors include parameters derived from the test data (such as the effective mass, or effective stiffness at peak compression).

In summary, the normalisation of isolated response characteristics - such as chest force-compression in an impactor test - seems to be a reasonable approach. However, the utility of the various approaches is much less clear e.g. for normalising all of the kinematic data for various body regions in a sled test. In particular, the effect of using different normalisation approaches for different parameters within the same data set appears to have significant risks. Section 6.2 evaluates a number of these issues further for an example data set.

6.2 Normalisation of Shaw *et al.*, 2009 sled test kinematic data

Within the THORAX project, the appropriateness of different scaling parameters was assessed by considering the affect on the responses, in relation to a few kinematic measurements taken from the recent PMHS sled tests by the University of Virginia (Shaw *et al.*, 2009b), also known as the 'gold standard' sled tests. The parameters considered included factors related to the total body mass of the subject, stature, initial pelvis to head distance, and combinations of these.

For previous sled tests at the UVA (University of Virginia) Forman *et al.* (2006) normalised all occupant trajectories using the subject stature (height) as a characteristic length. When applied to the Shaw *et al.* data, consideration of stature alone was not sufficient to reduce the scatter in the responses. For instance, subject 443, used in test 1378 was relatively tall, but did not experience a large head excursion. The x-axis displacement of the head for the subjects in the gold standard (Shaw *et al.*, 2009b) tests is shown in Figure 6-3 without any normalisation and in Figure 6-4 after normalisation based on subject stature.

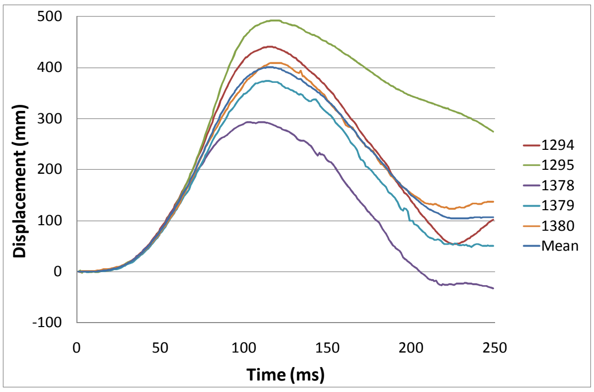


Figure 6-3: Head x-axis displacement from the gold standard tests (adopted from Shaw *et al.*, 2009b) without normalisation

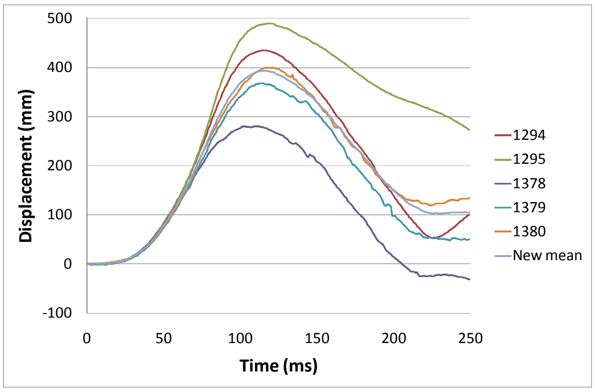


Figure 6-4: Head x-axis displacement from the gold standard tests (adopted from Shaw *et al.*, 2009b), normalisation based on subject stature

Using a factor related to the initial distance between the head and pelvis provided a better compression of the head displacement measurements than the stature alone. However, this effect was not continued at the shoulder level. Instead the characteristic length needed to be changed to the pelvis to shoulder distance. Unfortunately, whilst this improved the x-axis displacement variation, it was not as effective with regard to the z-axis. The response for the subject's right shoulder displacement, without normalisation are shown in Figure 6-5 for the x-axis displacement and Figure 6-7 for the z-axis. Equivalent figures showing the effect of normalisation based on the initial pelvis to shoulder length are shown in the corresponding Figures 6-6 and 6-8.

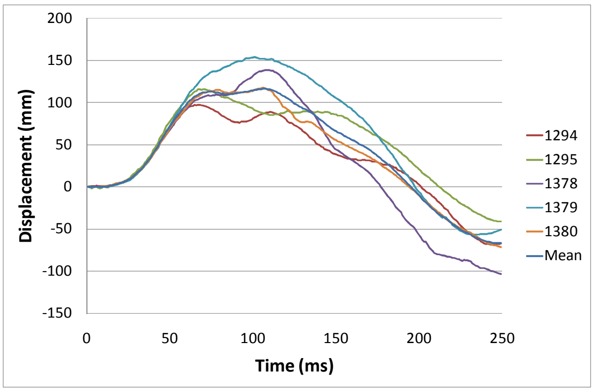


Figure 6-5: Right shoulder x-axis displacement from the gold standard tests (adopted from Shaw *et al.*, 2009b without normalisation

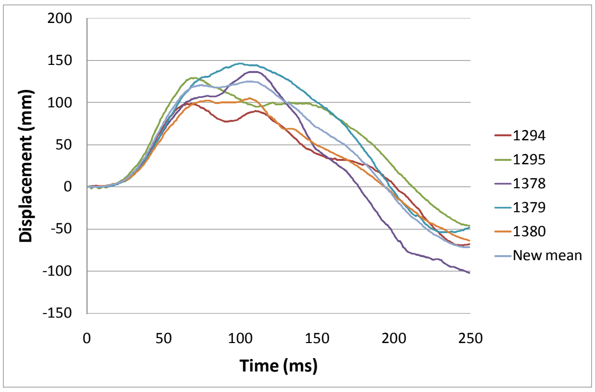


Figure 6-6: Right shoulder x-axis displacement from the gold standard tests (adopted from Shaw *et al.*, 2009b), normalisation based on initial pelvis to shoulder length

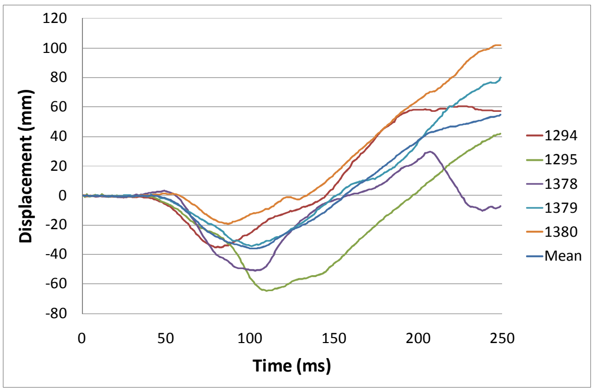


Figure 6-7: Right shoulder z-axis displacement from the gold standard tests (adopted from Shaw *et al.*, 2009b) without normalisation

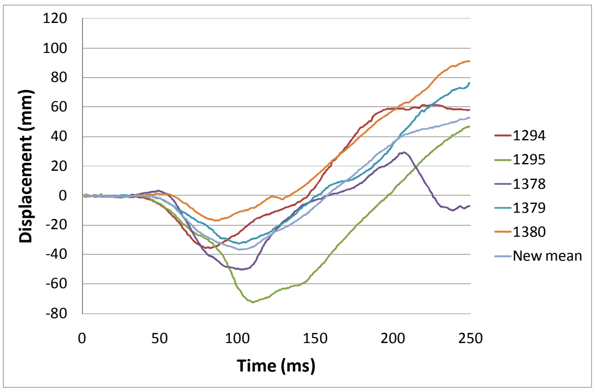


Figure 6-8: Right shoulder z-axis displacement from the gold standard tests (adopted from Shaw *et al.*, 2009b), normalisation based on initial pelvis to shoulder length

Combining this characteristic length (initial pelvis to X distance) with a factor for total body mass, using the mass-spring equation shown above did not reduce the scatter. This suggests that the mass-spring model would have to be updated before it could be applied to this type of sled testing and the various different measurement points and body regions. It is considered to be beyond the scope of this analysis to develop new mass-spring (or other) models and scaling equations for all body regions in sled tests. Also, it is quite likely that in developing new models, there may be a requirement for different characteristic lengths to be used. These lengths may not have been reported by Shaw *et al.*, 2009b. As such this work item would require continued interaction with Shaw *et al.* to identify exactly what measurements can be provided retrospectively.

Eppinger's mass-based scaling was not effective in reducing response deviation when applied to the head displacements. Evidently factors other than subject mass are more influential for the head excursion.

In summary, several factors and approaches have been used to scale kinematic responses from the Shaw *et al.* tests. It seems that none of these relatively simple approaches will provide a consistent improvement and reduction in the response scatter. As such, it is recommended that the responses are considered as broadly representative of a 50th percentile male occupant, without normalisation being applied. Scaling to other occupant sizes will need further, careful, consideration of the influential parameters. This should be undertaken in collaboration with Shaw *et al.* who may have access to additional measurements and a clearer insight into the relevant parameters.

6.3 Scaling to other body sizes

Primarily, the previous few sections have discussed normalisation of assorted responses from human subject testing under a variety of loading conditions. However, as mentioned in the introduction to normalisation, there can also be a need to scale to sizes of person other than a standard mid-sized male. Accident analysis within Work Package 1 of the THORAX project has suggested that there may be a safety benefit (in terms of mitigation of torso injuries) in testing with either a large male or a small female dummy as well as a mid-sized male.

One obvious way to define biofidelity requirements for occupants of different sizes would be to perform tests with subjects that have a similar stature to the target stature. This has been done for side impact biofidelity requirements, where Yoganandan *et al.*, 2004 repeated a series of sled tests using smaller female subjects in order define small female specific biomechanical corridors.

Without size-specific biofidelity data, it becomes necessary to scale the existing information to the particular size of interest. This is the situation for frontal impact biofidelity, where no small female or large male data are available as complete datasets. Instead there may be one or two occupants matching those broad descriptions within the groups of subjects conventionally used to define requirements for mid-sized occupants.

Where data exist for two sizes, it presents the opportunity to scale from one occupant size to another in order to check the validity of such scaling. This would be possible with the small female side impact sled test data of Yoganandan *et al.*, 2004, because matching results under equivalent conditions had been reported already for mid-sized occupants by Maltese *et al.* (2002). However, there is no publicly reported statement as to whether scaling results between these two sets of data is reliable. Furthermore, any such scaling would be different from the scaling approaches described for the frontal impact condition because the impact conditions are not equivalent.

The biomechanical response requirements for the THOR-5F (the prototype 5th female THOR dummy) were published by Shams *et al.*, 2003, based on scaled THOR 50th male requirements GESAC, 2001. Shams *et al.* used the method described by Mertz *et al.*, 1989. Shams *et al.* reported that, due to the lack of 5th female response data in frontal impact, the Task Force formed by the Mechanical Human Simulation Subcommittee of the Human Biomechanics and Simulation Standards Committee of the Society of Automotive Engineers agreed to scale the Hybrid III responses by using the mass and geometric scale factors generated from the ratio of the corresponding elements of the 5th female and the 50th male sizes.

6.3.1 Scaling of the pendulum requirements

To provide some biofidelity requirement guidance for the 5th percentile female, the mid-size response to pendulum testing was scaled. This replicates the work of Shams *et al.*, 2003 who scaled the complete set of biofidelity requirements to be relevant for the development of the THOR 5th percentile female dummy. The requirements proposed by Shams *et al.* cannot be used directly because, for example, the pendulum requirements would not include the work reported by the ACEA/ISO Task Force on frontal biofidelity.

In the low speed pendulum test sample used by Lebarbé, 2010 there were some subjects who were substantially smaller than the 50th percentile size. These tests offer some scope to check the scaling procedure to see that it is in agreement with the empirical effects of size differences. Test number MS589 involved a male PMHS who weighed 60 kg, was 1.69 m in height and had a chest depth of 180 mm (approximately 15th percentile mass and 19th percentile height, based on the UMTRI AMVO anthropometry, Schneider *et al.*, 1983). To make use of the potential to check the scaling procedure, all tests were normalised, using exactly the same procedure as before, to the size of the MS589 PMHS. The following three figures show the results of this normalisation:

- Figure 6-9 shows the responses included in the low-speed test sample, without normalisation
- Figure 6-10 shows the low-speed responses normalised to the size of the 50th percentile (effective mass = 24 kg, chest depth = 230 mm)
    - This figure also includes normalisation to the impact conditions of the test involving MS589 (impactor mass = 23.7 kg, speed = 4.4 m.s^{-1})
- Figure 6-11 shows the low-speed responses normalised to the size of the PMHS used in test MS589 (effective mass = 9.7 kg, chest depth = 180 mm) and again to the impact conditions

Comparing Figure 6-9 with Figure 6-10 demonstrates how effective the initial normalisation is in reducing the scatter of the data. This is largely due to normalisation of the test conditions (i.e. impactor mass and velocity), not subject characteristics.

Test MS589 has one of the lowest peak force values of the tests shown in Figure 6-9. It is interesting to note how much this force level increases in Figure 6-10. This change reflects the expected effect if the PMHS for test MS589 was 50th percentile instead. The response now fits well within the general spread of the other tests, indicating that there is nothing peculiar about this particular test.

When the normalisation was changed for the size for the MS589 PMHS, then a similar spread of responses can be seen, as with the 50th percentile normalisation. However, the peak force has been reduced for all test responses. The peak force range is now between 1 kN and 1.7 kN, whereas for the 50th percentile it was between about 1.5 kN and 2.5 kN. With the scaling to the MS589 size, all responses follow a general spread around a mean which would not be too dissimilar from the original MS589 response. This is encouraging because it suggests that the normalisation approach used by Lebarbé is broadly appropriate for scaling this data set within the bounds of the original data. Extrapolation to the 5th percentile female size is the next step and requires confidence in the scaling approach. It is hoped that this check of the procedure within the existing data range gives some confidence that the scaled results would not be too far from that expected with a subject of that size.

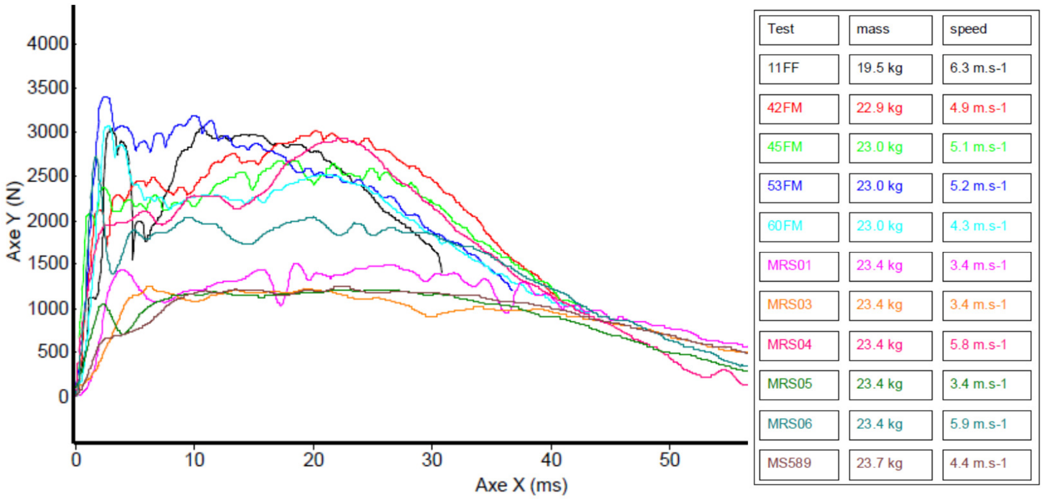


Figure 6-9: Original force time responses for the low-speed pendulum tests (Figure Lebarbé, 2010) without normalisation

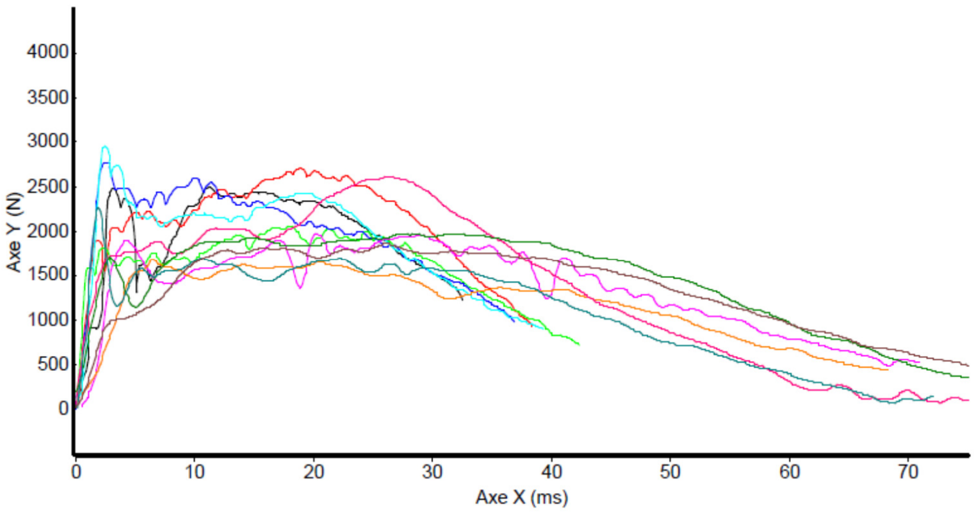


Figure 6-10: Original force time responses for the low-speed pendulum tests (Lebarbé, 2010) normalised to the 50th percentile size

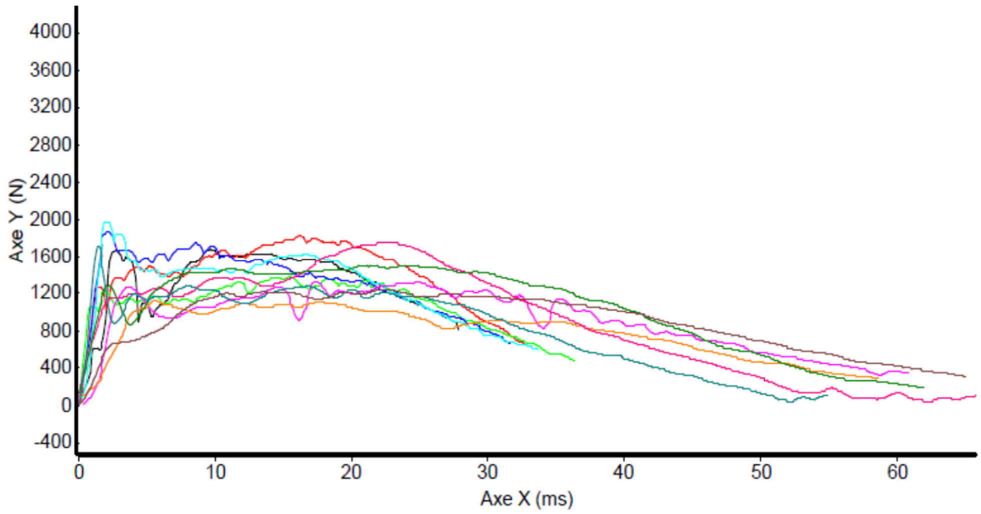


Figure 6-11: Original force time responses for the low-speed pendulum tests (Lebarbé, 2010) normalised to the size of PMHS from test MS589

7 Discussion

One of the stated objectives of the project is to develop a demonstrator thorax that is better able to discriminate between restraint systems of similar, reasonably good performance, and to discriminate better at low risks of serious thorax injury. Ideally this would be achieved using a demonstrator thorax that is able to replicate human response across a range of loading types and severities. However, in practice this is likely to be beyond the scope of the THORAX project, so the demonstrator thorax will have to be well tuned to replicate the human response at the range of loading types and severities that are of most interest. This is primarily modern combined airbag and seat-belt restraints that apply loading to the thorax at a modest rate of about 1 m/s, although the thorax should be reliable for higher loading rates up to about 4 m/s (see Section 4).

Despite the large number of frontal impact volunteer and PMHS tests that have been performed over the last four decades, only a relatively small number have been identified as suitable for defining biofidelity requirements (or engineering guidelines) for the thorax and/or shoulder of a frontal impact crash test dummy. For example, for sled tests, it was often found that it would not be possible to reconstruct the complex test conditions, either due to insufficient description of the test set-up in the original publications, or because the restraint systems (including seat-belts, airbags and the seat itself) are no longer available. In other cases, thorax deflections had not been recorded, or had been recorded with chest bands that had fewer gauges than currently recommended for reconstructing accurate force-compression data.

Nevertheless, it has been possible to recommend a set of requirements that are more relevant to current restraint loading conditions than previous frontal impact biofidelity requirements that have been published. For some of these data sets, appropriate normalisation methods have been defined and suitable normalised biofidelity corridors have been produced, for example the Kroell-type mid-sternum impactor data analysed by Lebarbé (2010).

Some of the data sets that were reviewed were excluded because of the low number of subjects. However, a number of these included several different restraint conditions, with different injury outcomes for each condition. It is recommended that WP4 of the THORAX project considers reconstructing these data sets as part of its work to evaluate the sensitivity of the THORAX demonstrator to different restraint conditions.

The best defined set of sled tests recommended in this report (Shaw 2009) is relatively severe, in terms of the sternum compression rate, maximum sternum compression, and the level of injury of the subjects. The sternum compression rate is approximately 1.9-3.8 m/s, with a typical value of approximately 3 m/s. This is at the high end of the relevant velocity range defined in Section 4 and higher than the preferred loading rate of approximately 1 m/s.

A number of previous biofidelity requirements, and the current ISO Kroell pendulum impactor requirements, have defined force-compression responses. The NHTSA has defined force-time and compression-time requirements for the Yoganandan *et al.*, 1997 data set, but indicated that these are a secondary priority compared with meeting the defined force-compression requirement. Meeting the force-compression requirement may be sufficient for injury metrics based on compression. However, this may not be enough to demonstrate that the biofidelity is sufficient to give confidence that injury metrics based on rate of compression (e.g. VC) are reliable. For this reason, the force-time and compression-time requirements for the Yoganandan data have been prioritised, and similar requirements have been defined for the Kroell tests.

That said, the normalisation methods used for the Kroell data (by Lebarbé) and the Yoganandan data are quite different. It is possible that this could lead to contradictory requirements that cannot be met by a single dummy design. It is therefore recommended that the Kroell (Lebarbé) requirements are the highest priority.

If it is desired to develop a 5th percentile female frontal impact dummy in the future, it is recommended that tests are undertaken with small female subjects in at least two test conditions to provide unscaled requirements for this group. If at least one of these test conditions was chosen to replicate one of the biofidelity requirements proposed in this report, it would also be possible to use this data to check the validity of the scaling techniques that have been proposed for deriving 5th female biofidelity requirements from test with subjects that better represent the 50th percentile male.

7.1 Comparison with Previous Requirements

Table 7-1 shows a comparison between the EEVC 2003, NHTSA 2005, draft ACEA/ISO 2010 and draft THORAX 2010 biofidelity requirements. The shoulder, thoracic spine and thorax requirements are discussed below.

7.1.1 Shoulder

The THORAX shoulder requirements are more comprehensive than the EEVC requirements, with high severity PMHS sled tests; moderate severity, multidirectional PMHS sled tests, and quasi-static volunteer tests. Although the number of suitable subjects is low in the Törnvall data set, and each subject was tested several times, it is considered that this data set provides a useful confirmation of performance at an intermediate severity level. Neither NHTSA nor ISO have thus far defined shoulder biofidelity requirements.

7.1.2 Thoracic spine

There are very few data sets suitable for defining dynamic biofidelity requirements for the thoracic spine in a restraint loading condition. EEVC recommended the spine accelerations published by Vezin *et al.*, 2002a; 2002b, although it was noted that the data set was small (three subjects) and additional data was recommended. NHTSA has not set biofidelity requirements for the thoracic spine, but has proposed a number of data sets that could be considered for biofidelity requirements. These include dynamic sled tests with embalmed PMHS (which would not meet the inclusion criteria set by THORAX) and volunteers, as well as quasi-static tests of the bending stiffness of the spinal column. The NHTSA biofidelity requirements document (GESAC, 2005) notes that this information indicates that the stiffness of the lumbar and thoracic spines is comparable. To date, ACEA/ISO have not defined thoracic spine biofidelity requirements. THORAX has replaced the Vezin tests with the Shaw 2009 tests, which gives a larger sample size and detailed trajectory data. This data set is also used for shoulder and thorax requirements, and the requirements on the thoracic spine accelerations and trajectories are a useful check that the loads input to the shoulder and thorax are a good match for the original PMHS tests, and therefore give added confidence in the shoulder and thorax requirements.

7.1.3 Thorax

Three types of biofidelity requirement have been defined for the thorax: impactor, sled, and table-top tests. The impactor and sled tests generally give absolute requirements on the thorax compression in the defined loading condition. The table top tests, however, have been defined as giving relative compression at different regions of the thorax in a single test condition, or relative compression at a single thorax region in different loading conditions. This recognises that the inertial loading of the organs on the ribs in a dynamic front impact with a restraint system will lead to a different thorax stiffness than that observed in these tests, so the absolute compression should not be set as a requirement.

EEVC has not defined biofidelity requirements based on table-top tests, and NHTSA have proposed two data sets that potentially could be used to define relative biofidelity requirements. The ACEA/ISO group has defined one biofidelity requirement based on the Kent *et al.*, 2004 table-top tests. The THORAX project has defined several biofidelity requirements based on table top tests.

For the impactor test conditions, all four sets of requirements are very similar except that ACEA/ISO has not defined an oblique impactor requirement. For pure frontal impactor requirements, all four sets are based primarily on the same data, but they have been processed in different ways that lead to somewhat different requirements. THORAX has adopted the ACEA/ISO lower-speed impactor requirements, but not the higher-speed requirements because they were considered to be too severe for a dummy intended for in-position vehicle crash tests featuring modern restraint systems.

The THORAX sled test conditions are more extensive than other biofidelity requirements; for instance, the ACEA requirements currently include one sled test requirement, although three other sled tests conditions were identified as suitable provided volunteers could be found to work on the data. The sled test conditions have the advantage that the loading condition more accurately reflects the crash test condition in terms of load distribution and inertial loading of the rib cage by the internal organs, although the thorax loading velocity is markedly higher than observed a typical modern vehicle legislative or consumer information tests.

7.2 Gaps in the draft requirements

The biofidelity requirements defined by the THORAX project consortium cover a wide range of loading conditions; however, there is little data that exactly matches the intended application of the dummy – combined seat-belt and airbag loading with a thorax loading velocity comparable to that observed in typical legislative or consumer information test procedures. The closest match is the Forman (2006) data, which includes passenger three-point belt and airbag tests.

The THORAX partners also discussed the use of several data sets that do not contain chest deflection measurements, for example the sled tests of Petitjean *et al.*, 2002, and the OOP airbag deployment tests of Lebarbé *et al.*, 2005. It is recommended that these would be more appropriate as sensitivity tests in Work Package 4.

Table 7-1 Comparison of recent biofidelity requirements

NB: cells shaded in light grey indicate potential biofidelity data sets

Body region	Type	Absolute / Relative	EEVC (ESV 2003)	NHTSA (GESAC 2005)	ACEA/ISO (Draft June 2010)	THORAX (Draft October 2010)
Shoulder	Sled	Absolute	Vezin (2002) • Two restraint conditions; two speeds	None defined	None defined	Törnvall (2008) • 3-pt belt, three impact directions, 26.5 kph
						Shaw (2009) • 3-pt seat-belt, 40 kph
	Quasi-static	Absolute	None defined	None defined	None defined	Davidsson (2010) • THORAX tests
	Table-top	Relative	None defined	Schneider (1992) • Quasi-static thorax regional coupling	None defined	Cesari and Bouquet (1990) (plus L'Abbe (1982), Riordain (1991), and Cesari and Bouquet (1994) Belt loading – relative regional compression; PMHS and volunteer
				Cesari and Bouquet (1990) Belt loading – relative regional compression		
Thoracic spine	Sled	Absolute	Vezin (2002) - NB: more tests required • Sled: see shoulder	Cesari and Bouquet (1990) • Belt loading – relative regional compression	None defined	Shaw (2009) • 3-pt seat-belt, 40 kph
	Quasi-static component	-	None defined	*Several tentative proposals (mostly embalmed PMHS)*	None defined	None defined
Thorax	Impactor	Absolute	Kroell (1971) • Frontal rigid impactor: 23.4 kg; 4.3 and 6.7 m/s	Neathery (1974) • Frontal rigid impactor: 4.3 m/s	Lebarbé (2010) • Pendulum impactor tests based on Kroell (1971), INRETS, and CEESAR data • Frontal rigid impactor: 23.4 kg; 4.3 and 6.7 m/s	Lebarbé (2010) • Based on Kroell (1971), INRETS, and CEESAR • Frontal rigid impactor: 23.4 kg; 4.3 m/s
			Yoganandan (1997) • Oblique padded impactor: 23.4 kg; 4.3 m/s	Yoganandan (1997) • Oblique padded impactor: 23.4 kg; 4.3 m/s		Yoganandan (1997) • Oblique padded impactor: 23.4 kg; 4.3 m/s

Body region	Type	Absolute / Relative	EEVC (ESV 2003)	NHTSA (GESAC 2005)	ACEA/ISO (Draft June 2010)	THORAX (Draft October 2010)
	Sled	Absolute	Vezin (2002) - NB: more tests required • Sled: see shoulder	None defined	Proposed Shaw (2009) • Sled: lap and diagonal seat-belt, at 40 kph	Bolton (2006) • Lap belt & airbag, two speeds Forman (2006) (not inc. Shaw 2000) • Chest bands with various restraints Yoganandan (1993) • Lap belt and airbag; two speeds Rouhana (2003) • Mostly four-point belt restraint Shaw (2009) • 3-pt seat-belt, 40 kph
	Table-top	Relative	None defined	Schneider (1992) • Quasi-static thorax regional coupling Cesari and Bouquet (1990) • Belt loading – relative regional compression	None defined	Cavanaugh (1988) (Same as Schneider (1992)) • Quasi-static thorax regional coupling Kent (2004) • Relative chest compression in different loading conditions Shaw (2007) • Relative regional compression Cesari and Bouquet (1990) (plus L'Abbe (1982), Riordain (1991), and Cesari and Bouquet (1994) • Belt loading – relative regional compression; PMHS and volunteer Fayon (1975) • Relative regional compression, belt and hub loading Salzar (2008) and Lessely (2008) • Relative regional compression, belt loading

8 Risk register

Risk No.	What is the risk	Level of risk[2]	Solutions to overcome the risk
1	Electronic results for all of the recommended data sets will not become available to the project partners.	2	Maintain communication with data owners and other research groups to maximise the options for access to the original data.
2	THORAX may not have sufficient budget to perform a full ISO-style data selection and normalisation process if all of the electronic data become available.	2	Prioritise data sets based on the relevance of the loading condition and the size of the sample.
3	The wider biomechanics community may not agree with all of the recommended biofidelity requirements.	1	Maintain communication with other research groups, such as the ACEA/ISO working group, and align requirements where appropriate.
4	The THORAX partners may not be able to recreate some of the more complex test conditions	2	Consider sub-contracting some tests to the original authors.

[2] Risk level: 1 = high risk, 2 = medium risk, 3 = Low risk

9 Conclusions

This report has reviewed existing frontal impact dummy thorax biofidelity requirements with respect to the loading environment (e.g. restraint type, distribution of load, velocity of loading) observed in modern cars during a typical frontal impact. It was found that existing biofidelity requirements were biased towards higher loading velocities than is now recommended and unrepresentative loading conditions. Therefore, a review of the available data sets that could be considered for defining thorax biofidelity requirements for an advanced frontal impact dummy was undertaken. As part of this review, this report documents a set of objective criteria by which the relevance of candidate data sets can be assessed.

From this review, a set of sled, impactor and table-top tests have been recommended for use within this project. Several data sets have been recommended for defining biofidelity requirements, while several others have been recommended for use in defining ‘engineering guidelines’. Typically the latter have been defined either where the original data was sampled from only a small number of subjects, and where relative (not absolute) requirements are defined – for example the relative stiffness of the upper, middle and lower part of the thorax, not the absolute stiffness of each of these regions.

Some of the recommended data sets have been documented in detail as appendices to this report, and basic biofidelity requirements have been defined.

A review has been undertaken of the normalisation techniques that have been used in the biomechanics literature in an attempt to make volunteer and PMHS test data more representative of a particular car occupant group. The limitations of various techniques for different types of test and different types of data have been explored. From this, recommendations have been made for appropriate normalisation techniques for some types of data, such as local force-compression responses in impactor tests. However, it was also found that the normalisation methods reviewed did not provide a consistent result when used with data from more complex loading environments, such as whole-body kinematic sled test data. In this case, it was recommended that no normalisation is performed. With this recommendation it is particularly important that the test subjects were representative of the occupant group – in this case 50th percentile male – for which biofidelity requirements are being developed. Further work will also be required to understand how this data should be scaled to represent significantly different occupant sizes, such as a 5th percentile female.

Where an appropriate normalisation method has been identified, normalised biofidelity requirements have been defined, either by adopting requirements from the literature, from current parallel work by an ACEA/ISO expert group, or developed within the project.

Acknowledgments

The authors would like to acknowledge the following people for their contributions to this report:

Mathieu Lebarbé, CEESAR, France
Greg Shaw, Jeff Crandall and Richard Kent, University of Virginia, USA

Glossary

Term	Definition
3pt	3-point seat-belt system
4pt	4-point seat-belt system
AB	Airbag
FL	Force-limited seat-belt
KB	Knee bolster - a support for the knee (typically padded) that limits forward motion of the knee-femur-pelvis region
Lap	Lap-belt only
OOP	Out-of-Position
PT	Pre-tensioned seat-belt
SB	Standard seat-belt
THOR	Test device for Human Occupant Restraint frontal impact crash test dummy

References

ADRIA (1999). *Final report for publication.* ADRIA 4th Framework EC Project, PL96-1074. 19 April, 1999.

Ali T, Kent R, Murakami D and Kobayashi S (2005). *Tracking rib deformation under anterior loads using computed tomography imaging.* Journal of Passenger Cars - Mechanical Systems 114(6). 111-125.

Arbogast K, Maltese M, Nadkarni V, Steen P and Nysaether J (2006). Anterior-posterior thoracic force-deflection characteristics measured during cardiopulmonary resuscitation: comparison to post-mortem human subject data. 50th Stapp Car Crash Conference, Dearborn, MI, USA, 6-8 November, 2006. Paper number 2006-22-0006. Society of Automotive Engineers, Warrendale, PA, USA, pp.131-145.

Ash J H, Shaw C G, Lessley D J, Crandall J R (2012a) PMHS restraint and support surface forces in simulated frontal crashes, Japanese Society of Automotive Engineering Annual Congress, Yokohama, Japan, No. 20125221.

Ash J H, Lessley D J, et al. (2012b) Whole-Body Kinematics: Response Corridors for Restrained PMHS in Frontal Impacts, Proceedings of IRCOBI Conference, Dublin, Ireland, IRC-12-21, pp. 142-154.

Backaitis S and St-Laurent A (1986). *Chest deflection characteristics of volunteers and Hybrid III dummies.* 30th Stapp Car Crash Conference, San Diego, California, USA, 27-29 October, 1986. Paper number 861884. Society of Automotive Engineers, Warrendale, PA, USA, pp.157-166.

Been B, Bortenschlager K and Steeger B (2010). *Design specification for demonstrator development.* Deliverable 3.1. THORAX EC FP7 Project: Thoracic Injury Assessment for Improved Vehicle Safety. SST.2007.4.1.2: Human physical and behavioral components GA No. 218516.

Bolton J, Kent R and Crandall J (2006). Passenger Air Bag Impact and Injury Response using a Hybrid III Test Dummy and Human Surrogates. NHTSA Biomechanics Test Database, test 8377-8379.

Bouquet R, Ramet M, Bermond F and Cesari D (1994). *Thoracic and pelvis human response to impact.* 14th International Technical Conference on the Enhanced Safety of Vehicles, Munich, Germany, 23-26 May, 1994. US Department of Transportation, National Highway Traffic Safety Administration, pp.100-109.

Burstein A, Reilly D and Martens M (1976). *Aging of bone tissue: mechanical properties.* Journal of Bone and Joint Surgery 58A. 82-86.

Cesari D and Bouquet R (1990). *Behaviour of human surrogates thorax under belt loading.* 34th Stapp Car Crash Conference, Orlando, Florida, USA, 4-7 November, 1990. Paper number 902310. Society of Automotive Engineers, Warrendale, PA, USA, pp.73-81.

Cesari D and Bouquet R (1994). *Comparison of Hybrid III and human cadaver thorax deformations.* 38th Stapp Car Crash Conference, San Antonio, Texas, USA, 31 October - 2 November, 1994. Paper number 942209. Society of Automotive Engineers, Warrendale, PA, USA, pp.65-76.

Cheng R, Yang K, Levine R and King A (1984). *Dynamic impact loading of the femur under passive restrained condition.* 28th Stapp Car Crash Conference, Chicago, Illinois, USA, 6-7 November, 1984. Paper number 841661. Society of Automotive Engineers, Warrendale, PA, USA, pp.101-118.

Cavanaugh J, Jepsen K, King A. (1988) Quasi-static loading of the Thorax of Human Cadavers and the Hybrid III Dummy. 16th Annual International Workshop on Human Subjects for Biomechanical Research. Atlanta, Georgia, October 16, 1988, pp. 3-20.

Crandall J (2008a). ATD Thoracic Response Test Development, 40 km/h Frontal Sled Tests, Gold Standard Buck: THOR NT Advanced Frontal Crash Test Dummy, UVA1286 Standard NT Shoulder, UVA1289 Chalmers SD-1 Shoulder, CADAVERS 411, 403, UVA1294, 1295. A Report Prepared for NHTSA: Cooperative Agreement No. DTNH22-93-Y-07028. Internet: http://www-nrd.nhtsa.dot.gov/database/aspx/searchmedia2.aspx?database=b&tstno=9547&mediatype=r&r_tstno=9547

Crandall J (2008b) ATD Thorac Response Test Development, 40 km/h Frontal Sled Tests, Gold Standard Buck, THOR NT Advanced Frontal Crash Test Dummy; UVA1286, 1287, 1289, 1290, 1291, Internet: http://www-nrd.nhtsa.dot.gov/database/aspx/searchmedia2.aspx?database=b&tstno=9548&mediatype=r&r_tstno=9548

Duma S, Stitzel J, Kemper A, McNally C, Kennedy E and Matsuoka F (2005). *Non-censored rib fracture data from dynamic belt loading tests on the human cadaver thorax.* 19th International Technical Conference on the Enhanced Safety of Vehicles, Washington, DC, USA, 6-9 June, 2005. Paper number 05-0360. US Department of Transportation, National Highway Traffic Safety Administration.

EEVC WG12 (2003). *Biofidelity Impact Response Requirements for an Advanced Mid-Sized Male Crash Test Dummy.* EEVC WG12 Public Document. European Enhanced Vehicle-safety Committee (www.eevc.org). May 1996.

EEVC WG12 (2006). EEVC Recommendations on the Future of the Harmonised THOR Dummy: Report on the outcome of the Special Workshop on 'Harmonisation of THOR - The Advanced Frontal Dummy', held 4-5 May 2006 at TRL, UK. EEVC WG12 Public Document. European Enhanced Vehicle-safety Committee (www.eevc.org). 15th August, 2006.

Eppinger R (1976). *Prediction of thoracic injury using measurable experimental parameters.* 6th International Conference on Experimental Safety Vehicles, Washington, DC, USA, 12-15 October, 1976. US Department of Transportation, National Highway Traffic Safety Administration, pp.770-780.

Eppinger R, Marcus J and Morgan R (1984). *Development of dummy and injury index for NHTSAs thoracic side impact protection research program.* SAE Government/Industry Meeting and Exposition, Washington, DC, USA, 21-24 May, 1984. Society of Automotive Engineers, Warrendale, PA, USA.

Fayon A, Tarriere C, Walfisch G, Got C and Patel A (1975). *Thorax of three-point-belt wearers during a crash (experiments with cadavers).* 19th Stapp Car Crash Conference, San Diego, California, USA, 17-20 November, 1975. Paper number 751148. Society of Automotive Engineers, Warrendale, PA, USA, pp.195-223.

FID (1999). Improved Frontal Impact Protection through a World Frontal Impact Dummy.

Forman J, Lessley D, Kent R, Bostrom O and Pipkorn B (2006a). *Whole-body kinematic and dynamic response of restrained PMHS in frontal sled tests.* 50th Stapp Car Crash

Conference, Dearborn, MI, USA, 6-8 November, 2006. Paper number 2006-22-0013. Society of Automotive Engineers, Warrendale, PA, USA, pp.299-336.

Forman J, Lessley D, Shaw C, Evans J, Kent R, Rouhana S and Prasad P (2006b). *Thoracic response of belted PMHS, the Hybrid III, and the THOR-NT mid-sized male surrogates in low-speed, frontal crashes.* 50th Stapp Car Crash Conference, Dearborn, MI, USA, 6-8 November, 2006. Paper number 2006-22-0009. Society of Automotive Engineers, Warrendale, PA, USA, pp.191-215.

Forman J, Lessley D and Kent R (2008). Standard Belt Rear Seat Sled Tests Report Covering Tests 1332, 1333, 1334, 1335. NHTSA Biomechanics Test Database, test 10031-10034.

Forman J, Lopez-Valdes F, Lessley D, Kindig M, Kent R, Ridella S and Boström O (2009). Rear seat occupant safety: an investigation of a progressive force-limiting, pretensioning 3-point belt system using adult PMHS in frontal sled tests. 53rd Stapp Car Crash Conference, Savannah, Georgia, USA, 2-4 November, 2009. Paper number 2009-22-0002. Society of Automotive Engineers, Warrendale, PA, USA, pp.49-74.

GESAC (2001). Biomechanical response requirements of the THOR NHTSA Advanced Frontal Dummy (Revision 2001.02). GESAC-01-04. Boonsboro, MD, USA: GESAC Inc. November, 2001.

GESAC (2005). Biomechanical response requirements of the THOR NHTSA Advanced Frontal Dummy (Revision 2005.1). GESAC-05-03. Boonsboro, MD, USA: GESAC Inc. March, 2005.

ISO (1999). ISO TR 9790: Road vehicles - anthropomorphic side impact dummy - lateral response requirements to assess the biofidelity of the dummy. ISO. First edition 1 December 1999.

Kallieris D, Mellander H, Schmidt G, Barz J and Mattern R (1982a). *Comparison between frontal impact tests with cadavers and dummies in a simulated true car restrained environment.* 26th Stapp Car Crash Conference, Ann Arbor, Michigan, USA, 20-21 October, 1982. Paper number 821170. Society of Automotive Engineers, Warrendale, PA, USA, pp.353-367.

Kallieris D, Schmidt G and Schüler F (1982b). Thoraxbelastung und Verletzungsmuster bei der Schrägkollision [Thorax load and pattern of injury in oblique collisions]. Beiträge zur gerichtlichen Medizin 40. 275-281.

Kallieris D, Stein K, Mattern R, Morgan R and Eppinger R (1994). *The performance of active and passive driver restraint systems in simulated frontal collisions.* 38th Stapp Car Crash Conference, San Antonio, Texas, USA, 31 October - 2 November, 1994. Paper number 942216. Society of Automotive Engineers, Warrendale, PA, USA, pp.165-175.

Kallieris D, Rizzetti A, Mattern R, Morgan R, Eppinger R and Keenan L (1995). *On the synergism of the driver air bag and the 3-Point belt in frontal collisions.* 39th Stapp Car Crash Conference, San Diego, California, USA, Paper number 952700. Society of Automotive Engineers, Warrendale, PA, USA, pp.389-401.

Kent R, Crandall J, Bolton J, Prasad P, Nusholtz G and Mertz H (2001). *The influence of superficial soft tissues and restraint condition on thoracic skeletal injury prediction.* 45th Stapp Car Crash Conference, San Antonio, Texas, USA, 15-17 November, 2001. Paper number 2001-22-0008. Society of Automotive Engineers, Warrendale, PA, USA, pp.183-203.

Kent R, Lessley D and Sherwood C (2004). *Thoracic response to dynamic, non-impact loading from a hub, distributed belt, diagonal belt, and double diagonal belts.* 48th Stapp Car Crash Conference, Nashville, Tennessee, USA, 1-3 November, 2004. Paper number 2004-22-0022 Society of Automotive Engineers, Warrendale, PA, USA, pp.495-519.

Krause P (1984). *Normalisation of side impact cadaver dynamic response data utilising regression techniques.* SAE Government/Industry Meeting and Exposition, Washington, DC, USA, 21st to 24th May, 1984. Society of Automotive Engineers, Warrendale, PA, USA, pp.1-12.

Kroell C, Schneider D and Nahum A (1971). *Impact tolerance and response of the human thorax.* 15th Stapp Car Crash Conference, San Diego, California, USA, 17-19 November, 1971. Paper number 710851 Society of Automotive Engineers, Inc., Warrendale, PA, USA, pp.84-134.

Kroell C, Schneider D and Nahum A (1974). *Impact tolerance and response of the human thorax II.* 18th Stapp Car Crash Conference, Ann Arbor, Michigan, USA, 4-5 December, 1974. Paper number 741187. Society of Automotive Engineers, Warrendale, PA, USA, pp.383-457.

L'Abbé R, Dainty D and Newman J (1982). *An experimental analysis of thoracic deflection response to belt loading.* International IRCOBI Conference, Cologne, Germany, 8-10 September, 1982. IRCOBI, pp.184-194.

Lebarbé M, Potier P, Baudrit P, Petit P, Trosseille X and Vallancien G (2005). *Thoracic injury investigation using PMHS in frontal airbag out-of-position situations.* 48th Stapp Car Crash Conference, San Antonio, TX, USA, 9-11 November, 2008. Society of Automotive Engineers, Warrendale, PA, USA, pp.323-42.

Lebarbé M (2010). Contribution to the Definition of Biofidelity Corridors in Frontal Impact for an Adult Male 50th Percentile - Thorax Impactor Report – April 2010. Frontal Biofidelity Specification International Task Force, ISO/TC22/SC12 – Working Group 6. Unpublished. Nanterre, France: CEESAR.

Lebarbé M. (2011) Road Vehicles – Anthropometric Frontal Impact Dummy – Frontal Impact Response Requirements to Assess the Biofidelity of the dummy: Part 1: Thorax and Part 2: Abdomen. ISO/TC22 N 3088.

Lessley D, Shaw G, Ash J, Crandall J (2012) Displacement response of the spine in restrained PMHS during frontal impacts, Proceedings of the Japanese Society of Automotive Engineering Annual Congress, Yokohama, Japan, No. 20125182.

Lessley D, Salzar R, Crandall J, Kent R, Bolton J, Bass C and JL F (2008). *Kinematics of the thorax under dynamic belt loading conditions.* International Crash Worthiness Conference, Kyoto, Japan. Taylor & Francis.

Mertz H (1984). *A procedure for normalising impact response data.* SAE Government/Industry Meeting and Exposition, Washington DC, USA, 21-05-1984 to 24-05-1984. Society of Automotive Engineers, Warrendale, PA, USA.

Mertz H, Irwin A, Melvin J, Stalnaker R and Beebe M (1989). *Size, weight and biomechanical impact response requirements for adult size small female and large male dummies.* SAE Congress, 1 March, 1989. Paper number 890756. Society of Automotive Engineers, Inc., Warrendale, PA, USA.

Michaelson J, Forman J, Kent R and Kuppa S (2008). *Rear seat occupant safety: kinematics and injury of PMHS restrained by a standard 3-point belt in frontal crashes.* 52nd Stapp Car Crash Conference, San Antonio, TX, USA, 3-5 November, 2008. Paper number 2008-22-0012 Society of Automotive Engineers, Warrendale, PA, USA, pp.295-326.

Moorhouse K (2008). *An Improved Normalization Methodology for Developing Mean Human Response Curves.* Injury Biomechanics Research: Proceedings of the Thirty-Sixth International Workshop.

Nahum A, Gadd C, Schneider D and Kroell C (1970). *Deflection of the human thorax under sternal impact.* 700400. Society of Automotive Engineers, Inc., Warrendale, PA, USA.

Neathery R (1974). *Analysis of chest impact response data and scaled performance recommendations.* 18th Stapp Car Crash Conference, Ann Arbor, Michigan, USA, 04th to 05th December, 1974. Society of Automotive Engineers, Warrendale, PA, USA, pp.459-493.

Nusholtz G, Kaiker P, Huelke D and Suggitt B (1985). *Thoraco-abdominal response to steering wheel impacts.* 29th Stapp Car Crash Conference, Washington, DC, USA, 9-11 October, 1985. Paper number 851737. Society of Automotive Engineers, Warrendale, PA, USA.

Nusholtz G, Hsu T and Byers L (2007). *A proposed side impact ATD bio-fidelity evaluation scheme using cross-correlation approach.* 20th International Technical Conference on the Enhanced Safety of Vehicles, Lyon, France, 18-21 June, 2007. US Department of Transportation, National Highway Traffic Safety Administration.

Patrick L (1981). *Impact force-deflection of the human thorax.* 25th Stapp Car Crash Conference, San Francisco, California, USA, 28-30 September, 1981. Paper number 811014. Society of Automotive Engineers, Warrendale, PA, USA, pp.471-496.

Petitjean A, Lebarbé M, Potier P, Trosseille X and Lassau J-P (2002). *Laboratory reconstructions of real world frontal crash configurations using the Hybrid III and THOR dummies and PMHS.* 46th Stapp Car Crash Conference, Ponte Vedra Beach, Florida, USA, 11-13 November, 2002. Paper number 2002-22-0002. Society of Automotive Engineers, Warrendale, PA, USA.

Riordain S, Ramet M, Cesari D and Bouquet R (1991). *A comparison of Hybrid III and cadaver thorax response under diagonal belt loading.* 13th International Conference on Experimental Safety Vehicles, Paris, France, 4-7 November, 1991. US Department of Transportation, National Highway Traffic Safety Administration, pp.860-862.

Roberts A, Beusenberg M, Cesari D and Glaeser K-P (1991a). *The biofidelity of the production version of the European Side Impact Dummy 'EuroSID-1'.* 13th International Conference on Experimental Safety Vehicles, Paris, France, 04th to 07th November, 1991. US Department of Transportation, National Highway Traffic Safety Administration, pp.850-859.

Roberts A, Lowne R, Beusenberg M and Cesari D (1991b). *Test procedures for defining biofidelity targets for lateral impact test dummies.* 13th International Conference on Experimental Safety Vehicles, Paris, France, 04th to 07th November, 1991. US Department of Transportation, National Highway Traffic Safety Administration, pp.956-967.

Rouhana S, Bedewi P, Kankanala S, Prasad P, Zwolinski J, Meduvsky A, Rupp J, Jeffreys T and Schneider L (2003). *Biomechanics of 4-point seat belt systems in frontal impacts.* 47th Stapp Car Crash Conference, San Diego, California, USA, 27-29 October, 2003.

Paper number 2003-22-0017. Society of Automotive Engineers, Warrendale, PA, USA, pp.367-399.

Salzar R, Bass C, Lessley D, Crandall J, Kent R and Bolton J (2008). *Viscoelastic response of the thorax under dynamic belt loading.* International IRCOBI Conference, Bern, Switzerland, 17-19 September, 2008. IRCOBI, pp.73-85.

Salzar R, Bass C, Lessley D, Crandall J, Kent R and Bolton J (2009). *Viscoelastic response of the thorax under dynamic belt loading.* Traffic Inj Prev 10(3). 290-296.

Schneider L, Robbins D, Pflüg M and Snyder R (1983). *Anthropometry of Motor Vehicle Occupants.* Final Report UMTRI-83-53-1. University of Michigan Transportation Research Institute. December 1983.

Schneider L, King A and Beebe M (1989). Design Requirements and specifications: Thorax and Abdomen Development Task. Interim Report. Trauma Assessment Device Development Program. UMTRI-89-20 / DOT HS 807 511. Michigan: University of Michigan Transportation Research Institute. November, 1989.

Schneider L, Ricci L, Salloum M, Beebe M, King A, Rouhana S and Neathery R (1992). *Design and Development of an Advanced ATD Thorax System for Frontal Crash Environments, Volume 1.* UMTRI-92-22-1 / DOT HS 808 138. Michigan: University of Michigan Transportation Research Institute. June 1992.

Shams T, Huang T-J, Rangarajan N and Haffner M (2003). *Design requirements for a fifth percentile female version of the THOR ATD.* 18th International Technical Conference on the Enhanced Safety of Vehicles, Nagoya, Japan, 19-22 May, 2003. US Department of Transportation, National Highway Traffic Safety Administration.

Shaw G, Crandall J and Butcher J (2000). *Biofidelity evaluation of the THOR advanced frontal crash test dummy.* International IRCOBI Conference, Montpellier, France, 20-22 September, 2000. IRCOBI, pp.11-29.

Shaw G, Lessley D, Evans J, Crandall J, Shin J, Portier P and Paoloni G (2007). *Quasi-static and dynamic thoracic loading tests, cadaveric torsos.* International IRCOBI Conference, Maastricht, The Netherlands, 18-21 September, 2007. IRCOBI.

Shaw C G, Parent D, et al. (2009a) Frontal Impact PMHS Sled Tests for FE Torso Model Development, Proceedings of International IRCOBI Conference, York, UK, pp. 341-356.

Shaw G, Parent D, Purtsezov S, Lessley D, Crandall J, Kent R, Guillemot H, Ridella S, Takhounts E and Martin P (2009b). *Impact response of restrained PMHS in frontal sled tests: skeletal deformation patterns under seat belt loading.* 53rd Stapp Car Crash Conference, Savannah, Georgia, USA, 2-4 November, 2009. Paper number 2009-22-0001. Society of Automotive Engineers, Warrendale, PA, USA.

Shaw J, Herriott R, McFadden J, Donnelly B and Bolte J (2006). *Oblique and lateral impact response of the PMHS thorax.* 50th Stapp Car Crash Conference, Dearborn, MI, USA, 6-8 November, 2006. Paper number 2006-22-0007. Society of Automotive Engineers, Warrendale, PA, USA, pp.147-167.

Stalnaker R, McElhaney J, Roberts V and Trollope M (1973). *Human torso response to blunt trauma.* Human Impact Response - Measurement and Simulation. Proceedings of the Symposium on Human Impact Response, Warren, MI, USA. New York: Plenum Press, pp.181–198.

Törnvall F, Holmqvist K, Martinsson J and Davidsson J (2004). Comparison of shoulder range-of-motion and stiffness between volunteers, Hybrid III and THOR Alpha in static frontal impact loading. International Crashworthiness Conference, ICrash, San Francisco, California, USA.

Törnvall F, Svensson M, Davidsson J, Flogaard A, Kallieris D and Haaland Y (2005). *Frontal impact dummy kinematics in oblique frontal collisions: Evaluation against post mortem human subject test data.* Traffic Injury Prevention 6(4). 340-350.

Törnvall F, Holmqvist K, Davidsson J, Svensson M, Gugler J, Steffan H and Håland Y (2008). *Evaluation of dummy shoulder kinematics in oblique frontal collisions.* International IRCOBI Conference, Bern, Switzerland, 17-19 September, 2008. IRCOBI, pp.195-210.

Trosseille X, Baudrit P, Leport T and Vallancien G (2008). *Rib cage strain pattern as a function of chest loading configuration.* 52nd Stapp Car Crash Conference, San Antonio, TX, USA, 3-5 November, 2008. Paper number 2008-22-0009. Society of Automotive Engineers, Warrendale, PA, USA.

UVA (2010). *ATD thoracic response test development, 40 km/h frontal sled tests, gold standard buck.* Report prepared for NHTSA (U.S. Department of Transportation, National Highway Traffic Safety Administration).

van Don B, van Ratingen M, Bermond F, Masson C, Vezin P, Hynd D, Owen C, Martinez L, Knack S and Schaefer R (2003). *Biofidelity impact response requirements for an advanced mid-sized male crash test dummy.* 18th International Technical Conference on the Enhanced Safety of Vehicles, Nagoya, Japan. US Department of Transportation, National Highway Traffic Safety Administration.

Vezin P (2002a). Human Thorax/Shoulder Behaviour during Frontal Impact with Airbag. FID Project, Work Package 2, D5 Part 1.

Vezin P (2002b). Human Thorax/Shoulder Behaviour during Frontal Impact without Airbag. FID Project, Work Package 2, D5 Part 2.

Vezin P, Bruyere-Garnier K and Bermond F (2002a). *Human response to a frontal sled deceleration.* International IRCOBI Conference, Munich, Germany, 18th to 20th September, 2002. IRCOBI, pp.84-98.

Vezin P, Bruyere-Garnier K, Bermond F and Verriest J-P (2002b). *Comparison of Hybrid III, THOR-Alpha and PMHS response in frontal sled tests.* 46th Stapp Car Crash Conference, Ponte Vedra Beach, Florida, USA, 11-13 November, 2002. Paper number 2002-22-0001. Society of Automotive Engineers, Warrendale, PA, USA, pp.1-26.

Viano D (1989). *Biomechanical responses and injuries in blunt lateral impact.* 33rd Stapp Car Crash Conference, Washington DC, USA, 04th to 07th October, 1989. Society of Automotive Engineers, Warrendale, PA, USA, pp.113-142.

Yoganandan N, Skrade D, Pintar F, Reinartz J and Sances AJ (1991). *Thoracic deformation contours in frontal impact.* 35th Stapp Car Crash Conference, San Diego, California, USA, 18-20 November, 2001. Paper number 912891. Society of Automotive Engineers, Warrendale, PA, USA, pp.47-63.

Yoganandan N, Pintar F, Skrade D, Chmiel W, Reinartz J and Sances AJ (1993). *Thoracic biomechanics with air bag restraint.* 37th Stapp Car Crash Conference, San Antonio, Texas, USA, 8-10 November, 1993. Paper number 933121. Society of Automotive Engineers, Warrendale, PA, USA, pp.133-144.

Yoganandan N, Pintar F, Kumaresan S, Haffner M and Kuppa S (1997). *Impact biomechanics of the human thorax-abdomen complex.* IJCrash 2(2). 219-228.

Yoganandan N, Pintar F and Gennarelli T (2004). *Small female-specific biomechanical corridors in side impact.* International IRCOBI Conference, Graz, Austria, 22-24 September, 2004. IRCOBI, pp.137-152.

Appendix A Summaries of available biofidelity data

A literature review was conducted to present sources of data from which the biofidelity of a dummy thorax and shoulder complex could be assessed. A summary of the most relevant studies, i.e. either data from test with volunteers or un-embalmed PMHS, is given below. Studies that included test with injuries as well as those with no injuries are included. The data are described in chronological order per type of test family, e.g. pendulum and sled tests are presented separately; however, where several works are based on the same experimental test set-up, they are described together. In the first section (A.1), data that mainly address thorax response while in the second section (0), data with the particular aim to assess the shoulder response are addressed.

Data sets that included only embalmed subjects are not included in this appendix. Tests highlighted in grey are those for which the raw data is not available in electronic form. Tests that are struck through are those for which the subject was outside the BMI range defined in Section 4.2.

Table A-1 shows the abbreviations used in this Appendix.

Table A-1 List of abbreviations used in Appendix A

Term	Definition
3pt	Three-point seat-belt system
4pt	Four-point seat-belt system
AB	Airbag
FL	Force-limited seat-belt
Hub	The loading surface was made to resemble the hub of a steering wheel
KB	Knee bolster - a support for the knee (typically padded) that limits forward motion of the knee-femur-pelvis region
Lap	Lap-belt only
PMHS	Post-Mortem Human Subject
PT	Pre-tensioned seat-belt
SB	Standard seat-belt

A.1 Thorax

A.1.1 Sled tests

Laboratory-based sled tests have been designed to simulate inertial effects of real vehicle frontal crashes. For many years, PMHS sled-based testing was used for investigations on the dynamic biomechanical response of the human thorax during a horizontal deceleration. The studies of PMHS in a sled test environment are useful for understanding the occupant kinematics and the injury mechanisms according to the restraint systems employed. It is more difficult to quantify the thoracic force-deformation response in this test condition, due to the fact that current instrumentation does not allow the accurate measurement of the force applied on the chest by the restraint systems.

Fayon 1975

Fayon *et al.*, 1975 exposed 31 fresh, unembalmed PMHSs to 44 km/h sled impacts using a standard three-point seat-belt system and different standard seats to study the resulting injury to the thorax. The subjects' lungs were reinflated and the vascular system repressurised prior to testing. Seat-belt forces and some chest deformation data was recorded and presented. Seat-belt characteristics were varied throughout the test series; the number of retractors varied and preload actuators were introduced. Impact speed was approximately 50 km/h and 63 km/h and the stopping distance was varied from 400 mm to 1000 mm (excluding one test). Both passenger and driver occupant positions were adopted. In some experiments chest compression was recorded using either film analysis or a potentiometer. For both methods, a rod was inserted through the thorax, one end being attached to the sternum while the other end was made to be visible some 300 mm behind the back of the PMHS. In most of the experiments the number of rib fractures was above 10 per subject. Force-deflection histories were given for two subjects tested with standard belt and two subjects tested with preload devices. Graphs of peak force vs. number of rib fractures were given for all tests, but no tabulated data was presented.

Conclusion: These tests are not recommended for setting biofidelity requirements because force-compression data is available for too few subjects, and the restraint configurations may be difficult to reproduce accurately (no matching dummy tests were available to check the quality of reconstruction).

Cheng 1984

Cheng *et al.*, 1984 ran sled tests with a limited number of unembalmed PMHSs in either a rigid seat or a VW car seat, at 48 km/h. The subjects' lungs were ventilated repeatedly prior to the test, but not pressurised at the time of the test, and the vascular system was repressurised. The restraints were either a pre-inflated and non-venting airbag, or a shoulder belt only in parallel with a knee bar. There is insufficient information about the restraint systems to enable them to be reconstructed accurately. The injuries sustained varied, from minor to severe including rib, sternum, clavicle bone and vertebrae fractures. Peak seat-belt loads and steering column loads were presented, but no chest deformation data were presented.

Conclusion: These tests are not recommended for setting biofidelity requirements because force-compression data was not recorded, and it is not possible to reconstruct the restraint systems accurately.

Kallieris 1982/1994/1995

Kallieris carried out a number of series of PMHS experiments at the University of Heidelberg using different body-in-white, seats, and restraint systems (Kallieris *et al.*, 1982a; 1982b; 1994; 1995). In Kallieris *et al.*, 1982a, a Volvo 240 interior was reproduced and dummy responses were compared to that of PMHSs. Kallieris *et al.*, 1982b reported on a series of oblique crashes, impact angles were 15°, 30° and 45°, using PMHSs in VW seats. These experiments have been reconstructed by Törnvall *et al.*, 2005. The tests were performed in both near side belt-geometry, i.e. the case where the occupant moves into the shoulder belt anchor, and far side belt-geometry i.e. the case where the occupant moves away from the shoulder belt anchor. Törnvall *et al.* were able to reconstruct eight of the original PMHS tests (of which four PMHSs were exposed to identical loading conditions).

In the study from 1994, Kallieris reported results from experiments with 16 unembalmed PMHS positioned in the driver's seat of a BMW 3-series and exposed to impact velocities from 48 to 56 km/h with a mean deceleration pulse of 17 *g*. The restraints used were a three-point seat-belt and airbag in combination with a knee bolster, or a three-point seat-belt with airbag (Table A-2). The PMHS thoraces were instrumented with a 12-accelerometer array. Biomechanical responses and the thoracic deformation contours and deflection time histories were also reported from chest bands (normally 24 gauges were used of the 40 gauges available). Photographic targets were applied to the clothing/wrapping only, with no direct connection to the underlying skeletal structure. Thoracic injuries occurred in 11 out of the 16 tests.

In the airbag and knee bolster tests, the subjects were additionally restrained at about 60 ms by the head contacting the windscreen. The trunk of the PMHS was not always restrained symmetrically by the driver airbag and a rotation of both the head and torso was sometimes observed. The head-to-windscreen contact occurred well before maximum chest compression (which occurred at ~85-95 ms depending on the thorax level assessed), and could well have affected the maximum chest compression. This constraint would be particularly difficult to reproduce accurately, and there were no dummy tests undertaken in this restraint condition that would improve the confidence in the quality of reconstruction.

In the seat-belt (plus knee bolster) tests, the PMHS kinematics included a tangential impact of the head against the steering wheel. The exact nature of the head impact was reported to be dependent on the PMHSs seated height. Again, this may have affected the peak thorax loads, and therefore the compression profiles, in a way that would be difficult to reproduce with the dummy. No confounding contacts were reported for the seat-belt and airbag restraint system tests, and dummy tests were conducted in this restraint configuration that could be used (if the data is available) to check the accuracy of the reconstruction. However, combined belt and airbag loading is likely to be the most difficult type of loading to reproduce accurately, particularly if the original restraint systems are no longer available.

The seat-belts used are defined in terms of their percentage elongation (6% or 16%), but the test conditions for this are not defined. The load limiters and airbags are not defined in any detail, and the vehicle buck is described simply as 'the front part of a mid-sized passenger vehicle', with no information on the seat-type used. There is no mention in the papers of any perfusion of the vascular system or inflation of the lungs during the testing.

Table A-2 Test matrix for a selection of the tests carried out by Kallieris *et al.*, 1982a; 1994; 1995

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
B2731	9014	AB + KB	M	31	70	1,70	24	0	48
B2888	9207		M	25	74	1,84	22	0	49
B2889	9212		M	38	79	1,74	26	0	47
B2730	9013	3pt SB + KB	M	34	71	1,80	22	NA	48
B2894	9216		M	20	86	1,77	27	0	56
B3018	9309		M	29	68	1,84	20	2	49
B3019	9310		F	52	68	1,68	24	1	48
B3020	9311	3pt SB + AB + KB	F	47	76	1,69	27	Yes	48
B3021	9312		M	32	85	1,85	25	NA	48
~~B3254~~	~~9501~~	3pt FL + AB + KB	~~F~~	~~63~~	~~90~~	~~1,67~~	~~32~~	~~1~~	~~48~~
B3255	9502		M	58	80	1,76	26	2	48
B3256	9503		M	50	75	1,77	24	0	48

Conclusion: These tests were carried out with restraints that were available in the early 1990s and as such these test conditions will be difficult to reproduce. In addition there were too many undefined constraints (such as head-to-windscreen contacts in the airbag + knee bolster test *condition) that may affect the thorax loading. It is therefore recommended that these tests are not used for defining biofidelity requirements.*

Yoganandan 1991

Yoganandan *et al.*, 1991 and co-workers performed sled tests with PMHS at an impact velocity of 50 km/h with a deceleration pulse of 16 *g* of which six of these tests were published in Yoganandan *et al.*, 1991 (Table A-3). A three-point belt was used to restrain the PMHSs and the restraining forces were measured in each test. The seat-belt characteristics and anchorage locations were not defined in the paper. Chest deformation contours were measured using a chest band (EPIDM, 24 gauges used in a majority of the tests and 34 gauges in the four most recent tests) and other outputs included chest acceleration, belt forces and deformation at the upper and lower chest level. The PMHSs were unembalmed and pressurised to approximate the *in-vivo* pulmonary and vascular characteristics. All six PMHS demonstrated multiple rib fractures. In addition, clavicle and sternum fractures were found.

Table A-3 Test matrix for tests in which standard belts only were used and conducted at Medical College of Wisconsin and made available through the NHTSA data base

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
RC101M		3pt SB	M	58	82	1,75	27	10	49
RC102D			M	57	73	1,78	23	12	49
RC103V			M	66	77	1,73	26	8	50
RC104F			M	58	64	1,74	21	13	50
RC105A			M	67	73	1,70	25	19	50
~~RJ106J~~			~~M~~	~~44~~	~~86~~	~~1,75~~	~~28~~	~~9~~	~~50~~
RC107R			F	63	77	1,70	27	22	49
RC108E			M	57	73	1,72	25	12	48
~~RC109N~~			~~M~~	~~59~~	~~91~~	~~1,83~~	~~27~~	~~12~~	~~49~~
RC110V			F	63	61	1,60	24	25	50
~~RC111T~~			~~F~~	~~65~~	~~75~~	~~1,58~~	~~30~~	~~14~~	~~34~~
RC120P			M	51	66	1,73	22	8	24
RC121C			M	67	66	1,83	20	0	24
RC122S			F	81	60	1,57	24	4	24
RC123G			F	67	68	1,65	25	1	24

Conclusion: Seats similar to those used in the study will be difficult to obtain. The number of strain gauges used may provide unsecure data when the torso was restrained by a shoulder belt only. These tests are therefore not recommended to be used to define biofidelity requirements but may be suitable for injury assessment in case a vehicle buck, seat, and restraint systems can be sourced.

Yoganandan 1993

In a further paper, Yoganandan *et al.*, 1993 used fourteen un-embalmed re-pressurised PMHSs in deceleration sled tests at velocities of 32 or 47 km/h in a Ford Tempo 1986 buck with airbag from the same make and model. In these tests various restraint system configurations were used: air bag with knee bolster, air bag with lap belt and air bag with three point belt (Table A-4). The subjects were pressurised to approximation the *in-vivo* pulmonary and vascular characteristics. Two chest bands recorded thoracic deformations (the number of gauges varied from 24 to 40) and the belts were instrumented with load cells. In each case the results were normalised with respect to the initial chest depth. The results indicated that under any restraint combination, regional differences exist in the deformation response between the upper and lower thoracic levels. It was seen that the response of the human thorax was very different between the air bag with three point belt loading compared to the air bag with knee bolster and airbag with lap belt restraint combinations.

Table A-4 Test matrix for a selection of the tests in which airbag in combination with other restraints have been used and conducted at Medical College of Wisconsin and available through the NHTSA data base

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
RC112F		Lap belt + AB	F	67	50	1,64	19	3	49
RC113C			M	64	70	1,66	25	3	49
RC114B			M	58	73	1,84	22	0	49
RC129Y			M	59	78	1,75	25	8	32
RC130P			M	56	63	1,68	22	4	32.7
RC118U		AB + KB	F	29	41	1,70	14	0	46.5
~~RC119~~			~~M~~	~~71~~	~~81~~	~~1,69~~	~~28~~	~~11~~	~~45.4~~
~~RC124~~			~~M~~	~~76~~	~~81~~	~~1,68~~	~~29~~	~~0~~	~~31.6~~
RC125Z			F	75	85	1,80	26	9	43.8
RC126W			F	64	54	1,68	19	6	34.7
RC127E			M	81	62	1,82	19	3	34.4
RC115H		AB + 3pt SB	F	67	57	1,50	25	13	48.0
RC116C			M	68	59	1,74	19	10	48.3
RC128L		AB + KB + 3pt SB	F	67	46	1,54	19	3	29.9

Conclusion: Distributed loading to the thorax through air bag only is considered an important test condition. The chest deformation is uniform and does thereby require less number of chest band strain gauges compared to standard belt configurations to estimate the thorax deformation. The test conditions using knee bolster only have been reported to be difficult to reproduce and should be excluded for that reason. For the two tests combing airbag and 3-pt standard belts the number of gauges on the chest band does not meet our inclusion criteria. The loading condition in which airbag, knee bolster and 3-pt standard belt was combined should be excluded due to the low number of subjects.

Seats and airbags similar to those used in the study will be difficult to obtain. It is recommended that these tests could be used to define biofidelity requirements if a suitable vehicle buck, seat, and restraint systems can be sourced. It should be noted that this is unlikely to be possible within the THORAX project. The data could be used for injury assessment with the above caveat.

Shaw 2000 / Kent 2001 / Forman 2006

A number of series of frontal impact PMHS tests using the same body in white on a sled system were carried out at UVa from 1999 to 2006 (Shaw *et al.*, 2000, Kent *et al.*, 2001 and Forman *et al.*, 2006b. Data from these series were analysed, normalised and reported by Forman *et al.*, 2006a for the purpose of HBM validations. The authors excluded some tests from the previous studies, e.g. due to failure of the pretensioner to deploy, or the 'extreme number of rib fractures (23)' occurring in one test. The data from Shaw *et al.*, 2000 are not recommended for setting biofidelity requirements because the PMHS were embalmed; the other tests used frozen PMHS. In the Kent *et al.*, 2001 and Forman *et al.*, 2006b test series', a vehicle buck from a mid-sized 1997 vehicle was used.

The data sets include results from both female and male PMHS (Table A-5). The ages of the PMHS were within a rather small range, but the BMI was below or above recommended limits for three of the nine PMHS that were used in the three latter tests. The female subject with a rather high BMI (test UVA667, Table A-3) suffered from four-times more rib fractures compared to the other subjects. In the report deformation contours of the upper and lower chest from high-resolution (40 gauges in Kent *et al.*, 2001; 59 gauges in Forman *et al.*, 2006b) chest bands are provided in parallel with accelerometer and restraint force data. Matching tests with the Hybrid III and THOR-NT dummies were presented in Forman *et al.*, 2006b.

Forman *et al.*, 2006a provide a detailed description of the test configuration, subject anthropometry, and subject positioning, and note that detailed measurements for the anchorage positions are available from the authors on request. Additional information, e.g. restraint make and model, are available through reports available through the NHTSA data base.

Table A-5 Test matrix for the tests presented by Forman *et al.*, 2006a (not including the embalmed subjects from Shaw *et al.*, 2000)

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
UVA577	111	Passenger 3pt FL belt + AB	M	57	70	1,74	23	0	48
UVA578	107		F	69	52	1,55	22	4	48
UVA580	105		F	57	57	1,77	18	0	48
UVA665	112	Passenger 3pt SB + AB	M	55	85	1,76	27	3	48
UVA666	115		M	69	84	1,76	27	3	48
~~UVA667~~	~~120~~		~~F~~	~~59~~	~~79~~	~~1,61~~	~~30~~	~~13~~	~~48~~
UVA1094	322	Passenger 3pt SB	M	49	58	1,78	18	0	29
UVA1095	323		M	44	77	1,72	26	0	29
UVA1096	327		M	39	79	1,84	23	0	29

Conclusion: The tests from Shaw et al., 2000 in the first series should not be included since the subjects were embalmed prior to testing. In addition the seat and restraint models are from an older car model and likely difficult to obtain which makes accurate reconstruction of these tests difficult. The tests using standard belt only produced no rib fractures and may for this reason serve as a complimentary data set to the Shaw et al., 2009b series (although in the latter, due the fact that there was no retractor - just two independent pieces of webbing that retrain the pelvis and the chest, and the knees restrained with a plate - there was an earlier pelvis restraint/occupant coupling to the sled). For reconstruction of that data set, it may be useful to develop a generic instrumentation board and seat. The belt and airbag configurations may then be of interest since the test environment is partially already developed.

Petitjean 2002

Petitjean *et al*., 2002 carried out a series of PMHS sled tests in which the belt force limiter and airbag combination was varied (Table A-6). The restraint, airbag and belt system were from a production vehicle, and it is understood that these components are still readily available to the THORAX project partners. The seat was a modified production seat, with load cells in the subframe, and a knee bolster was included. The data set includes only two subjects per restraint group and may, due to subject variation, not be representative of the population at risk. In one group a female was included while in the other the two subjects were males. This could influence the results, but the effect is likely to be small since the female's stature and mass was similar to that of the male subjects. Three of the subjects received a large number of rib fractures which could partly be attributed to the high delta-V used. The test series is well documented, but unfortunately there is not chest deformation data available. The subjects' vascular systems and lungs were repressurised piror to testing. Matching tests with the Hybrid III and THOR-Alpha dummies are available.

Table A-6 Test matrix for the test reported by Petitjean *et al*., 2002

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
536		3pt 4 kN FL belt + AB + KB	F	78	70	1,69	25	6	64
539			M	81	60	1,72	20	18	64
542		3pt 6 kN FL belt + KB	M	76	67	1,74	22	21	64
543			M	75	70	1,69	25	16	64

Conclusion: These tests are not suitable for defining biofidelity requirements, because thorax deformation was not recorded; however, they may be reconstructed for the purpose of injury assessment. As such, these tests would add data representative of high delta-V crash conditions.

Vezin 2002

Vezin *et al.*, 2002a and Vezin, 2002a; 2002b reported on six PMHS sled tests carried out at INRETS as part of the FID project (Table A-7). The purpose of the tests was to establish injury and biofidelity data for the neck, thorax and shoulder in two frontal impacts configurations. These two configurations were only force limited three-point seat-belt at 30 km/h and force limited three-point seat-belt in combination with an airbag at 50 km/h. One of the subjects in the 30 km/h group was very thin (BMI=14) whereas the other subjects had rather similar proportions (BMI=19-25). Only resultant acceleration of the sacrum, T1, T8, and T12 were provided.

Table A-7 Test matrix for the FID tests (Vezin *et al.*, 2002a)

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
FID11		4 kN FL 3pt belt + AB	M	46	63	1,83	19	12	50
FID12			M	83	69	1,68	24	6	50
FID13			M	74	67	1,68	24	0	50
FID14		4 kN FL 3pt belt	M	78	82	1,8	25	2	30
FID15			M	81	58	1,67	21	4	30
~~FID16~~			~~M~~	~~90~~	~~45~~	~~1,77~~	~~14~~	~~0~~	~~30~~

Conclusion: These tests may be reconstructed for the purpose of injury assessment. As such, these tests would add data representative of low delta-V crash conditions resulting in a few rib fractures. These data are not recommended for biofidelity assessments because of to the small size of the sample, and lack of chest deformation measurements and displacement data.

Rouhana 2003

Rouhana *et al.*, 2003 carried out sled tests to study the effect of load distribution, by introducing four-point seat-belts, on the number of rib fractures (Table A-6). In these tests eight PMHS were included of which six were restrained by a four-point seat-belt system. The seat used was rigid but covered with foam and trim from a production bucket seat, and the locations of the anchorage points were specified in the paper. Some type of force limitation for the upper anchorage point in combination with restricted pelvis was adopted that allowed large torso forward rotations. Further information on the force-limiting system and seat configuration would be required in order for these tests to be reproduced accurately, although matching Hybrid III and THOR dummy test data is available which could be used to check the quality of reconstruction. Single-point sternum compression measurements using a trans-thoracic rod technique data are available. The male PMHS in the four-point configuration were all slightly above average weight (although within the range specified for the inclusion criteria), while the two females varied (BMI 18 and 26). In six of the tests (tests 209 to 222 in Table A-8), the lungs were inflated three times prior to testing and partially inflated at the time of the test. No vascular repressurisation was performed.

Table A-8 Test matrix for the tests conducted jointly by UMTRI and Ford (Rouhana *et al.*, 2003)

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
208	206	3pt SB	M	75	72	1,75	24	32	40
209	474		M	72	82	1,78	26	16	40
210	853	4pt FL + PT belt	M	75	81	1,80	25	12	40
217	247		M	41	82	1,75	27	0	40
218	639		M	60	91	1,83	27	3	40
221	683		F	69	42	1,52	18	12	40
222	657		F	79	59	1,52	26	3	40

Conclusion: These tests are of interest since a four-point belt system was used which produced distributed loads and symmetric cadaver response, and three-point seat-belt loading is available for comparison. The displacement measurement data limits the usefulness of reproducing the test series. In case the tests with females and three point belts are excluded, only three tested subjects remain. Further information is required on the test configuration before these tests could be reproduced, and it is expected that this information will not be forthcoming during the timescale of the THORAX project.

Bolton 2006

Bolton *et al.*, 2006 reported on deceleration sled tests in which unembalmed PMHS restrained by a combination of a full-powered airbag, knee bar and lap belt was undertaken to study occupant to passenger bag interaction and establish a body of baseline data to be used for dummy evaluation (Table A-9). The vehicle interior mimicked that of the right front passenger side of a 1997 Ford Taurus (DN101). Hybrid III test data is available in order to assess the performance of any generic car interior. Detailed positioning data for the dummy and PMHSs are provided in the report. The PMHS, which had been frozen prior to testing, were fitted two chest bands on the 4th and 8th rib (38 and 39 gauges), tri-axial accelerometer arrays on the T9 and L1 vertebra, and the posterior pelvis, and a uni-axial accelerometer on the upper part of the sternum. In addition, accelerometer packages were attached to mounts on the back of the head and T1 vertebra. Belt forces and instrument panel-to-PMHS contact time were also recorded. Prior to test the lungs were ventilated and the vascular system of the PMHSs were repressurised. The data was not scaled nor was response corridors estimated.

The tests produced a distributed load with no/only minor injuries.

Table A-9 Test matrix for the UVa airbag, knee bolster, and lap belt sled test series (Bolton *et al.*, 2006)

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
UVA650	124	Lap belt + AB + KB	M	40	47	1,50	21	4	49
UVA651	121		M	70	57	1,76	18	0	49
UVA652	118		M	46	74	1,75	24	0	49

Conclusion: These tests are of interest since the restraints used offer a distributed load on the thorax and in addition the pelvis excursion was well controlled. The latter provided a controlled interaction between bag and occupant torso-head complex. The vehicle interior, seat system and restraint system may be reconstructed based on available information and were identical to the tests presented by Forman et al., 2006a.

Michaelson 2008

Michaelson *et al.*, 2008, Forman *et al.*, 2008 and Forman *et al.*, 2009 studied the response of eight unembalmed PMHS in the rear seat to 48 km/h delta-V full frontal impacts (Table A-10). The same seat buck model, which originated from a 2004 mid-sized sedan, was used in the three studies while the restraints varied. In the study by Michaelson *et al.*, 2008 a standard belt system was used and in the two studies by Forman *et al.*, 2008; 2009 a force limited and pretension belt system was used. Forman *et al.*, 2008 also tested a Thor NT, a Hybrid III 50% male and a Hybrid III 5% female in the different configurations.

The PMHS were fitted an upper and a lower multi-point position sensing chest bands, accelerometer arrays on the upper, mid and lower spine and pelvis and finally a single axes accelerometer on the sternum. In addition the belt loads were recorded. The combination of collision speed, seat buck characteristics and standard belt produced an average of 18 NRF while the more advanced restraints produced an average of 10 NFR. The pulmonary and cardiovascular systems were repressurised prior to testing. Of the eight subjects used in the 10 tests, only three meet the inclusion criterion on Body Mass Index. Of these three, one submarined beneath the lap belt and two did not submarine, leaving a maximum sample size of two subjects per restraint condition which is not sufficient for defining biofidelity requirements.

Forman *et al.*, 2009 normalised their results using the same method described for the UVA sled tests above Forman *et al.*, 2006a.

Table A-10 Test matrix for UVa rear seat tests (Michaelson *et al.*, 2008; Forman *et al.*, 2008, Forman *et al.*, 2009)

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
~~1262~~	~~362~~	3pt SB	~~M~~	~~51~~	~~55~~	~~1,75~~	~~17.9~~	~~13~~	~~48~~
~~1263~~	~~394~~		~~F~~	~~57~~	~~109~~	~~1,65~~	~~40~~	~~29~~	~~48~~
1264	367		M	57	59	1,79	18	13	48
~~1332~~	~~404~~	3pt FL + PT belt	~~M~~	~~54~~	~~124~~	~~1,89~~	~~35~~	~~-~~	~~30~~
~~1333~~	~~404~~		~~M~~	~~54~~	~~124~~	~~1,89~~	~~35~~	~~7~~	~~49~~
~~1334~~	~~400~~		~~M~~	~~53~~	~~151~~	~~1,82~~	~~46~~	~~-~~	~~29~~
~~1335~~	~~400~~		~~M~~	~~53~~	~~151~~	~~1,82~~	~~46~~	~~2~~	~~48~~
1386	-	3pt FL + PT belt	M	67	69	1,75	23	12	48
1387	-		M	69	67	1,71	23	2	48
1389	-		M	72	73	1,83	22	17	48

Of the subjects that meet the inclusion criteria defined in Chapter 4, one submarined beneath the lap belt and two did not. This gives a maximum sample size of two per restraint condition, which is not sufficient to define biofidelity requirements. These data are therefore not recommended.

Shaw 2009

Shaw *et al.*, 2009b carried out frontal sled tests with restrained unembalmed PMHS at 40 km/h. In total eight PMHS were tested and the data may provide corridors for shoulder response, global response and chest deformation for upper and lower as well as right and left chest. Most of the PMHS included in the study are close to 50th percentile male compared with other studies utilising PMHS (Table A-11). CT-scans were used to exclude any subjects with ribcage pathology etc. The PMHS were seated on a rigid planar seat and restrained by a custom three-point shoulder and lap belt. Detailed information on the test set up, including seat dimensions, belt anchorage points etc, can be found in Crandall, 2008a. Instead of chest bands, which have traditionally been used to assess chest compression, video analysis provided three-dimensional trajectories of multiple skeletal sites on the torso relative the spine. Such quantifications were enabled by attaching rigid film targets to various ribs and vertebrae and the design of the seat that allowed video recordings of the spine throughout the main part of the tests. The subjects' lungs were inflated with 2.5 litres of air immediately prior to the test, and then left open to the atmosphere. In addition to rib deformations, kinematics of the head, spine and the belted and the unbelted shoulder have been provided. These can be used to check that the general kinematics of the dummy are representative of the tests, and therefore that the input to the thorax from the restraint system is likely to be representative. However, it should be noted that this would require an extensive 3D camera system and 6D analysis similar to the Vicon system used by the original authors, and that such a system is not available at most laboratories.

It should also be noted that these experiments produced a large number of rib fractures and as such these tests are at the high end of the severity range, and higher than ideal for defining the biofidelity requirements.

Conclusions: Due to the stiffness of the used restraints and seat cushion these tests were rather severe and produced a large number of rib, sternum and clavicle fractures. This occurred although the sled velocity change was not high at 40 km/h. As such, the test setup is fairly representative of severe crashes in which the occupants are restrained by a diagonal and lap belt only without pretension or force limiting belt systems or airbag. As such the use of this data set, for evaluation of HBM and dummy performance, should be complemented with test in which airbag, belt pre-tension and load limiting system are introduced.

Table A-11 Test matrix for experiments conducted by Shaw *et al.*, 2009

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
1294	411	3pt SB* + KB	M	76	70	1,78	22	7	40
1295	403			47	68	1,77*	22	27	40
1358	425			54	79	1,77	25	15	40
1359	426			49	76	1,84	22	9	40
1360	428			57	64	1,75*	21	5	40
1378	443			72	81	1,84	24	9	40
1379	433			40	88	1,79*	27	10	40
1380	441			37	78	1,80	24	2	40

* Belt system comprised two separate belt webbings for the lap and the shoulder.

A.1.2 Table top tests

The loading conditions and the interaction between the restraints and the PMHS are usually better controlled in a table top test compared to sled tests. The reasons being that in most of the table top tests, the PMHS is placed on its back on a flat surface, loading is usually applied in the direction of interest (only load that compresses the ribcage) and the loading rate is relatively low. By these arrangements the rotations of the torso, the effect of the bending of the spine and the vertical or lateral displacements of the torso, which commonly occur in sled tests, are kept to a minimum. However, there is a lack of inertia effects of the internal organs and in case dummies are evaluated in which the mass distribution is not humanlike, e.g. the spine box is heavier than the spine of a human, table top test data will mainly provide relative stiffness measures and deformation patterns.

In principle data from three table top test configurations are available:

- Dynamic and static load from a single diagonal belt. This configuration is mainly useful to study ribcage deformation pattern, but also to study relative stiffness of the ribcage to different load rates.
- Static and dynamic tests with indenters of different size and at different positions on the rib cage. These table top tests may be very useful for assessment of the relative stiffness of different thorax regions (and the THORAX project has a specific recommendation from the stakeholders to improve this aspect of the dummy) and the coupling between these regions.

- Dynamic and quasi-static tests in which the chest response as a function of different loading configurations (single or double diagonal belts, hub and distributed loading configurations) are studied. Such data can be used to assess the capability of the dummy to predict the relative sensitivity to different loading configurations.

Furthermore, the contact between the back of the PMHS and the table-top can influence the results, and may be different for the dummy. For instance, for tests with the PMHS lying on the table top, the contact between the PMHS and the table-top is via the spine *and* the ribs, whereas in the dummy this may be only via the (relatively rigid) spine. These differences are not apparent in car crashes in which the occupant is without back support/constraints. Therefore some tests have been performed where the PMHS is supported at the spine, and the ribs are not loaded by the table-top. This condition is more readily reproducible with a dummy.

Fayon 1975

Fayon *et al.*, 1975 reported a study in which seven fresh unembalmed male PMHSs and an unspecified number of volunteers (possibly 14 subjects) were positioned on their backs on a flat table and statically loaded by a diagonal seat belt or hub/disk (Table A-12). During the tests sternum and rib deflections were recorded using either a rod through the thorax technique or string potentiometer/film analysis technique.

The loads applied in the volunteer tests were sub-injury level and as such resulted in peak sternum deflections of about 25 mm (approximately 60 kg). Force vs. sternum deflection data was presented for both belt and disk loads from the volunteer tests. In addition deformation data for different regions of the ribcage (2nd rib, sternum and 9th rib) are provided as a function of belt load (it is assumed from the plotted data that the load is from a diagonal belt and not the hub).

Table A-12 Test matrix for static table top tests reported by Fayon *et al.*, 1975

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
1	42	1 Diagonal belt or Hub	M	43	50	1,59	20	NA
2	43		M	53	65	1,72	22	7
3	48		M	42	75	1,67	27	6
5	50		M	70	77	1,84	23	10
6	51		M	63	61	1,75	20	13
7	52		M	43	63	1,74	21	2.5

Conclusion: These tests are of some interest since deformation of the ribcage was measured at three different locations, the combination of hub and single diagonal belt restraints, and the inclusion of both volunteers and PMHS. The loads applied to the volunteers are, however, very low. In addition, the description of the test set-up and the test procedures are unfortunately limited. These experiments are therefore only recommended if additional information on the test set-up can be obtained.

L'Abbé 1982

L'Abbé *et al.*, 1982 reported a series of tests that included dynamic and static seatbelt loading tests to examine the thoracic deflection characteristics of human volunteers. A large number of tests were carried out with ten volunteers lying supine on a rigid table and loaded by a diagonal seatbelt passing from the left clavicle down to the lower right ribs. The belt was centred on the sternum and was at an angle of 36° to the mid-sagittal plane. The deflection of the thorax was measured at the eleven locations on the thorax including mid-clavicle. Dynamic chest loading was applied through an impact mechanism that had a pre-load and a pendulum striker. The mean age of the volunteers was 21 years, with a mean mass of 77 kg and a mean sitting height of 90.3 cm. Many other mean anthropometric data for the thorax were given, but no data for individual subjects. Furthermore, matching tests with the Hybrid III dummy were reported.

The belt tests were performed both statically and dynamically.

- For the dynamic belt loading tests the peak belt load was restricted to 3.6 kN over 60 ms to avoid injury to the volunteers. Muscle tension was assessed through repeated tests with the volunteers both relaxed and tensed, the lungs being inflated to approximately 50% of their maximum volume.
- Static loads ranging from 15 to 20 kg were applied through the axis of each deflectometer (at each of the 11 measurement points) by a 30 mm diameter steel pad.
- Static belt loads ranged from 0 to 75 kg.

The resulting data was further analysed and presented by Backaitis and St-Laurent, 1986. They found that the dynamic testing of the relaxed volunteers showed that the largest displacement occurred at the right 7th rib (37 mm), and at the upper and mid-sternal locations (32 and 40 mm, respectively). For the tensed volunteers the largest displacement was again seen at the 7th rib (35 mm) and the upper and mid-sternal locations.

Conclusion: These dynamic and static table top tests were performed on volunteers and are therefore of special interest. The static loads applied to the volunteers were, however, rather low and deemed not especially useful. The dynamic tests produce important deformation patterns whereas the stiffness data provided in the study are of limited value. The latter is due to the load level applied, being about 20% of that the loads experienced in a vehicle collision, and the expected differences in mass distribution between a dummy and the volunteers. These tests may be of interest for defining injury risk functions.

Cesari 1990

Continuing this work, Cesari and Bouquet, 1990 reported the results from 16 tests with 13 PMHS using a very similar test configuration to that used by L'Abbé *et al.*, 1982. The aim of these tests was to extend the severity of the tests on volunteers by L'Abbe up to and including injury level using PMHS. In these tests the impactor which loaded the belt had a mass of either 22.4 or 76.1 kg and the impact velocities ranged from 3-9 m/s. The PMHS were unembalmed and the pulmonary system was pressurised before the tests and let open to the atmospheric pressure during the tests. Instrumentation attached to the subject was able to monitor the chest deformation and the tension force in the belt.

Results included the number of rib fractures (NRF) vs. impact energy, NRF vs. belt force, the NRF vs. Viscous Criterion and peak deformation of the 10 locations on the thorax. Data caution must be taken because the PMHSs sustained high number of rib fractures and some even flail chest. It was shown that six rib fractures correspond to a belt load of 10 kN or impact energy of 830 J. Numerous matching Hybrid III dummy tests were also performed. The pulmonary system was inflated and then left open to the atmosphere for the tests.

Conclusion: Part of a larger test series - see conclusion below.

Riordan 1991

Riordain *et al.*, 1991 presented the results of 33 belt loading tests with 20 unembalmed PMHS. The same set-up was used as Cesari and Bouquet, 1990. The paper includes the results on 13 PMHS from the 1990 paper and extends the data for a further seven PMHS using a high mass (76.1 kg) impactor. The lungs of the PMHS were inflated to restore the correct thoracic shape and let open to the atmospheric pressure during the tests. The tolerance of rib fracture was shown to be significantly less than the previous reported value. Six rib fractures corresponded to a belt load of 5.6 kN and impact energy of 420 J. One rationale for this may have been that the variability between the results for different cadavers was large.

Conclusion: Part of a larger test series - see conclusion below.

Cesari 1994

Cesari and Bouquet, 1994 presented the results from a further nine unembalmed PMHS tests in the same belt loading test configuration. The same two impactor masses were used as before and the velocity was varied from 2.38 to 7.3 m/s. Matching tests with the Hybrid III were also performed. In addition to the chest compression measurements used in the previous tests, two chest bands with 16 gauges each were used at the upper and lower thorax. Thorax stiffness data is presented in the appendix of the paper.

In total 34 tests were undertaken with PMHS subjected to diagonal belt table top tests in the INRETS table top test programme. Table A-13 to Table A-15 present the subset of these tests (preliminary version) that are readily available from the references given above.

Table A-13 Test matrix for low-speed, low-mass experiments conducted by INRETS (Cesari and Bouquet, 1990 (first three tests) and Cesari and Bouquet, 1994 (last four tests))

Test	PMHS No.	Test number with the same PMHS	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
~~THC61~~	~~K~~	~~Assumed No 1~~	1 Diagonal belt	~~M~~	~~72~~	~~53~~	~~1,83~~	~~16~~	~~0~~
~~THC64~~	~~L*~~	~~2~~		~~M~~	~~71~~	~~41~~	~~1,70~~	~~14~~	~~0~~
~~THC68~~	~~M*~~	~~2~~		~~M~~	~~40~~	~~56~~	~~1,83~~	~~17~~	~~0~~
THC76	Q	1		F	64	49	1,64	18	0
~~THC78~~	~~R~~	~~1~~		~~M~~	~~43~~	~~54~~	~~1,86~~	~~16~~	~~0~~
THC90	S	1		M	67	67	1,80	21	0
THC92	T	1		M	63	56	1,76	18	0

* The tested PMHS had been exposed to a low-sped and low-mass impact before this test.

Table A-14 Test matrix for high-speed, low-mass experiments conducted by INRETS (Cesari and Bouquet, 1990)

Test	PMHS No.	Test number with the same PMHS	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
~~THC11~~	~~A~~	~~1~~	1 Diagonal belt	~~F~~	~~47~~	~~92,5~~	~~1,70~~	~~32~~	~~8~~
THC12	B	1		F	17	58,5	1,64	22	0
~~THC13~~	~~C~~	~~Assumed No 1~~		~~F~~	~~86~~	~~43~~	~~1,60~~	~~17~~	~~17~~
~~THC14~~	~~D~~	~~1~~		~~M~~	~~69~~	~~82~~	~~1,73~~	~~27~~	~~16~~
THC15	E	1		M	60	69	1,77	22	3
THC16	F	1		M	59	62	1,70	21	4
THC17	G	1		M	71	75	1,77	24	7

Table A-15 Test matrix for low-speed, high-mass experiments conducted by INRETS (Cesari and Bouquet, 1990 (first six tests) and Cesari and Bouquet, 1994 (last five tests))

Test	PMHS No.	Test number with the same PMHS	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
~~THC18~~	~~H~~	~~1~~	~~1 Diagonal belt~~	~~M~~	~~67~~	~~47~~	~~1,74~~	~~16~~	~~6~~
~~THC19~~	~~I~~	~~1~~		~~F~~	~~83~~	~~43~~	~~1,55~~	~~18~~	4
THC20	J	~~1~~		M	70	63	1,60	25	18
~~THC62~~	~~K*~~	~~Assumed No 2~~		~~M~~	~~72~~	~~53~~	~~1,83~~	~~16~~	4
~~THC65~~	~~L*~~	~~3~~		~~M~~	~~71~~	~~41~~	~~1,70~~	~~14~~	~~10~~
~~THC69~~	~~M*~~	4		~~M~~	~~40~~	~~56~~	~~1,83~~	~~17~~	~~1~~
~~THC75~~	~~P*~~	~~2~~		~~M~~	~~60~~	~~44,5~~	~~1,60~~	~~17~~	~~6~~
THC77	Q*	2		F	64	49	1,64	18	6
~~THC79~~	~~R*~~	~~2~~		~~M~~	~~43~~	~~54~~	~~1,86~~	~~18~~	~~3~~
THC91	S*	2		M	67	67	1,80	21	2
THC93	T*	2		M	63	56	1,76	18	10

* The tested PMHS had been exposed to a low-sped and low-mass impact before this test.

Conclusion: Of the subjects that meet the inclusion criteria defined in Section 4, there are only three tests in the low-mass and low-velocity group, and four tests in each of the two more severe loading groups. In addition these tests can mainly be used to assess the dummy deformation pattern to diagonal belt loading. In addition, the type and amount of information on each subject and on the response of each subject is quite variable between the different publications. However, additional information has been made available to the project and has been used to define biofidelity requirements.

Kallieris 1987

In quasi-static loading table top tests Kallieris (reported in Schneider *et al.*, 1989) studied the effect of different type of quasi-static loading applied to the thorax of PMHSs on ribcage deformation using PMHSs. A polyurethane foam filled airbag produced a uniform deformation of about 54 mm in average for three measurement locations on two sides while a diagonal belt produced deformation that ranged from 10 to 85 mm for these six measurement points. Data and reports have not been located.

Conclusion: The information available is not sufficient to define biofidelity requirements. These data are therefore not recommended.

Cavanaugh 1988

Cavanaugh *et al.* (1988) (reported in Schneider *et al.*, 1992) statically loaded the chest of two supine unembalmed PMHS (Table A-16) using a 4.5 cm x 10 cm rigid and unpadded loading plate while measuring chest deflection at eight locations. The upper, mid, and lower sternum were loaded by an Instron testing machine, as were the ribs at upper, mid, and lower regions. A triaxial load cell in the loading device was used to record the forces acting on the ribcages.

Sternal loading was performed under two support conditions:

1) Support at the spine only, with a rigid aluminium bar supported on unistrut,
2) Support of the spine and ribs posteriorly. This rib support extended bilaterally approximately 7 cm lateral to the midline.

Rib loading was performed only for the second support condition. Loading rates ranged from 1.7 mm/s to 102 mm/s and the stroke was usually set at 25.4 mm. The original data include force-time and deflection time histories.

Table A-16 Subject data for the PMHS included in the experiments by Cavanaugh *et al.* (1988) (reported in Schneider *et al.*, 1992)

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
AATD2; test 1-10	986	Multiple tests at different locations	M	29	70	1,73	24	23
AATD5; test 1-30	115		M	57	58	1,79	26	18

Conclusion: The test series reported in Schneider et al., 1992 includes data from two subjects. Regardless of this, the data provide a target for coupling stiffness of the ribcage, i.e. relative deflections at the various regions of the chest for loading at one region, the data recommended for use as an engineering guideline in the THORAX project.

Kent 2004

Kent *et al.*, 2004 carried out 67 tests on 15 unembalmed PMHS lying supine on a rigid bench by either a two-point diagonal belt, a hub, a pair of two-point diagonal belts (in a crossed configuration), or a distributed load in random order (Table A-17). The posterior boundary condition was a rigid flat plate on which the subject was laid. The subject was free to move on the plate and the spinal curvature was not controlled other than by the flat plate interface. Prior to testing the subjects' pulmonary systems were pressurised to typical mean full-inspiration volume immediately prior to testing. The airway remained occluded throughout loading. A high-speed material testing machine applied the load at rate of 1.0 m/s. A load cell measured the reaction force on the PMHS back support. Mid-sternal chest deflection was obtained from string potentiometers attached to the belt, band or hub. All PMHS were tested five times, the first four times up to non injurious levels with the four different loading cases. The fifth test was injurious and this loading case varied between the specimens.

Kent calculated force-deflection corridors for each load case using whole body mass and modulus scaling factors for the reaction force. The mid-sternal chest deflection and the reaction force at the PMHS back were used to calculate thorax stiffness.

Conclusion: This data set is well documented, the test setup is reproducible without the use of vehicle components that will become out of date, and presents thorax stiffness as a function of loading conditions for a reasonably large set of subjects. The subjects were subjected to multiple testing which could compromise the results. However, the loading sequences were in random order and appear to have no or very small effect on the calculated stiffness values. This data set is recommended for use in the THORAX project.

Table A-17 Subject data for the PMHS included in the experiments by Kent *et al.*, 2004

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	
	176	Multiple test using different loadings conditions	F	85	58	1,57	24	
	182		F	80	65	1,57	26	
	177		F	79	48	1,61	19	
	155a		F	71	54	1,66	20	
	173		F	67	57	1,62	22	
	~~147~~		~~F~~	~~63~~	~~45~~	~~1,61~~	~~17~~	
	186		F	58	61	1,78	19	
	157		F	55	74	1,68	26	
	189		M	79	57	1,59	23	
	190		M	79	73	1,73	24	
	170		M	75	65	1,78	21	
	178		M	73	81	1,82	24	
	~~188~~		~~M~~	~~71~~	~~85~~	~~1,73~~	~~28~~	
	145		M	54	88	1,92	24	
	~~187~~		~~M~~	~~54~~	~~113~~	~~1,78~~	~~36~~	

Ali 2005

Ali *et al.*, 2005 used computed tomography during quasi-static belt like and distributed like load on four PMHS to further study deformation patterns of the chest (Table A-18). In these experiments the spine was laid on spinal brackets that allowed unconstrained rib motion.

Table A-18 PMHS characteristics for the subjects tested by Ali *et al.*, 2005

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
~~-~~	~~1~~	Distributed	~~M~~	~~75~~	~~49~~	~~1,70~~	~~17~~	~~8~~
-	2		F	57	50	1,65	18	13
-	3	Diagonal belt	F	50	59	1,67	21	22
~~-~~	4		~~M~~	~~48~~	~~50~~	~~1,67~~	~~17.9~~	~~16~~

Conclusion: Only one subject per loading condition meets the inclusion criteria defined in Chapter 4. The data is therefore not recommended for inclusion.

Duma 2005

Duma *et al.*, 2005 carried out table top tests in which 47 strain gauges were mounted to the on ribs of the two PMHS included in the study (Table A-18). The strain gauges were primarily installed to improve the measurement of the exact rib fracture timing. A single diagonal belt loaded the PMHS chest at a rate of 1.5 m/s.

Table A-19 Data for the test carried out by Duma *et al.*, 2005

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
-	SF33	Various restraints	F	73	45	1,54	19	Multiple
~~-~~	~~SM35~~	~~Various restraints~~	~~M~~	~~73~~	~~84~~	~~1,73~~	~~28~~	~~Multiple~~

Conclusion: Only one PMHS meets the inclusion criteria defined in Chapter 4 and only one type of loading condition was used, so this data is not recommended.

Arbogast 2006

Arbogast *et al.*, 2006 presented data from a special device that allowed the recording of chest compression and load applied to the chest when providing CPR. In total data from 91 subjects were reported on. For these average loading rate was 0.26 m/s and the vertical motion was 45 mm. It has been put forth that during these CPR measurements the back support has varied.

Conclusion: This data set is includes a large number of subjects and the loading conditions are well described. However, the loading rate is lower than that of a typical loading rate in a frontal collision and the test conditions (back support) varied and difficult to reproduce. Due to these limitations in combination with lack of resources in the THORAX project these are recommended to be excluded the THORAX biofidelity requirements. It may be that these tests could be useful for controlling the response of the flesh on the back of the dummy's thorax, provided that the loading conditions can be reproduced adequately.

Shaw 2007

Shaw *et al.*, 2007 exposed the torsos of five male unembalmed PMHS to quasi-static (1 m/s) loading to the anterior by rectangular indenters (Table A-20). The loading conditions resembled those use in the study by Cavanaugh, but loading rate, dimensions of the indenter and position were different. Chest compression was recorded using advanced film analysis that enabled the capture of 3D displacements of photo targets attached to the ribs and sternum. This was an advantage over experiments in the past in which only vertical rib motion were recorded.

The paper presents thorax force-deflection data for static loading and non-injurious dynamic loading (<30 mm compression) and injurious dynamic loading (approximately 75 mm compression). In addition, the paper presents deformation pattern (coupling data) for non-injurious and injurious dynamic test conditions.

Table A-20 Test subjects included in the study by Shaw *et al.*, 2007

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF*
1	343	Multiple tests at different locations	M	72	66	1,80	20	15
2	342		M	75	73	1,83	22	10
3	320		M	48	68	1,68	24	4
4	319		M	52	77	1,79	24	17
5	203		M	67	77	1,70	27	15

* Number of fractures in the final injurious ramp-and-hold tests

Conclusions: These tests may be reconstructed for the purpose of assessment of relative stiffness of the upper, middle, and lower region of the thorax. As such the data is mainly recommended for engineering guidance in the development phase of the THORAX demonstrator.

Salzar 2008/2009 and Lessely 2008

Salzar *et al.*, 2008; 2009 and Lessley *et al.*, 2008 carried out table top test on PMHSs that were fixed in a configuration comparable to that of a person in a severe frontal crash (Table A-21). As compared to the study by Kent *et al.*, 2004, the test fixture used allowed each subject's spine to be rigidly mounted without constraining the costo-vertebral or costo-transverse joints. In these studies the chest was loaded by a diagonal belt and load rates were varied from 0.5, 0.9 to 1.2 m/s. The load were either ramp-and-hold or sinusoidal wave form and the applied displacement of the belt was 10%, 15% or 20% of the chest depth. Also injurious experiments applying a ramp-and-hold belt load at 40% compression were deployed. The deformation was varied to produce and not produce injuries. The upper arms were lifted so that the direction of the arms simulated a driving posture and shoulder and clavicle response was also recoded.

Table A-21 Test conditions used in the study by Salzar *et al.*, 2008; 2009 and Lessley *et al.*, 2008

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
-	412	Multiple loading trough a diagonal belt	M	62	68	1,75	22	None*
-	413		M	54	68	1,75	22	None
-	419		M	31	90	1,93	24	None

* Fractures were detected prior to the fourth test with this PMHS and hence no additional tests were carried out.

Conclusion: These experiments, although the number of specimen tested were limited, could be useful in the assessment of the rate dependency of the thorax stiffness. However, the study found substantial variability in the instantaneous elastic response among the three test subjects. These differences were explained to be due to differences in size and age among the tested subjects. The raw force-deflection data compare fairly well with the Kent et al., 2004 data. Despite the low number of PMHS, the data is considered a potential biofidelity requirement but additional knowledge is needed prior to inclusion in the THORAX biofidelity requirements.

A.1.3 Impactor tests

Nahum 1970 and Kroell 1971

Nahum *et al.*, 1970 and Kroell *et al.*, 1971 provided response for blunt sternal impacts to the thorax of 15 unembalmed PMHS (Specimen 05-11 and 11-28, respectively) of both sexes, age 19-81 years, mass 53-82 kg, stature 1.56-1.89 m). The PMHS were placed in a seated position on a flat horizontal surface without back support and the arms and legs were extended horizontally and parallel to the mid-sagittal plane. The horizontal impactor (diameter 6-inch) was an unpadded flat wooden form with a 12.7 mm edge radius. The subject was placed in a position such that the surface of the thorax in line with the impactor centreline was vertical with the longitudinal centreline of the impactor at the same vertical height as the mid-sternum and guided in the mid-sagittal plane of the subject. In twelve of the tests the thoracic aorta was liquid pressurised at impact.

Total chest deflections (including interface effects such as compression of superficial tissues and non-square alignment of impactor and chest at contact) exceeded the true sternal deflection. The force deflection responses were presented in terms of the measured total deflection, with a recommendation to subtract 12.7 to 19.1mm from the curves to approximate the force vs. sternal deflection relationship.

The test series were further analysed and hence conclusions are to be found further below.

Kroell 1974

Kroell *et al.*, 1974 proposed performance requirements in the form of response corridors for anthropometric dummies when they are impacted in a similar manner to the impacts in his PMHS studies. The corridors were based on the data generated in the papers published in 1970 and 1971 as mentioned above, and 23 additional tests with unembalmed PMHS (specimen 30-64). Low speed 19.9 kg, 4.92 m/s tests and high speed 23.1 kg, 7.15 m/s tests were specified. These corridors were based on an approximate average of the collected response data. In the low speed tests, skeletal deflection had been measured so no adjustment of the response was necessary. However, for the high-speed tests, 12.7mm was subtracted from the measured penetration to account for tissue thickness and obliqueness of impact etc. As such an estimated sternal deflection was used to develop the high speed corridor. An allowance was also made for muscle tone in developing these corridors.

The test series were further analysed and hence conclusions are to be found further below.

Neathery 1974

As a result of the fact that the Kroell and Stalnaker data were being used to represent the 50th percentile adult response when the average masses from the two studies were 64 kg and 54.9 kg respectively the chest impact data of Kroell and Stalnaker were re-examined by Neathery, 1974. The objective of the study was to determine the relationships that might exist between the physical characteristics of PMHS, the impact conditions and subsequent responses. The aim was to be able to predict average 50th percentile responses from non-50th percentile PMHS data and also to extend this to other population such as 5th, 50th, and 95th percentiles.

It was found that the interrelation of the physical characteristics of the PMHS in the Kroell data was similar between males and females. An empirical equation was developed by multiple linear regression analysis to describe this relationship. On the basis of a further detailed analysis of the results it was concluded that the Kroell and Stalnaker data could not be considered as a common database.

Using the Kroell male data, empirical equations were also developed for predicting the response of humans corresponding to various sizes of crash test dummies. Neathery, 1974 proposed response corridors similar to those proposed by Kroell, using force-deflection response data of 10 PMHS tests (three PMHS for the low-speed corridors and 7 PMHS for the high-speed corridor), but scaled by the equations developed in the study were proposed for the 5th, 50th, and 95th percentile crash dummies.

Conclusions: A new analysis of the same data set as used by Neathery, 1974 *have been carried by ACEA/ISO (Lebarbé, 2010) and which is presented below.*

Bouquet 1994
Bouquet *et al.*, 1994 presented results from four unembalmed PMHS tests that had been subjected to a blunt impact to the central sternum. The mass of the impactor was 23.4 kg and the striking surface was a 152 mm diameter disk. Two tests were performed on each PMHS, the first at a sub-injury level (impact speed ~ 3.5 m/s) and the second at injury level (impact speed ~5.8 m/s). In the injury level test with the first subject, extensive rib fractures occurred and as a result the impact velocity was lowered for subsequent tests.

Corridors for the force-time history during the sub-injury test were presented. The force time results for the high-speed tests were difficult to interpret, therefore no corridor was developed. Corridors for the acceleration time history could not be considered for the low speed impacts, as the measured accelerations were too low (below 5 *g* in all tests) but a corridor for the 5.8 m/s tests was proposed. It should be noted that the results from this study were normalised to the 50th percentile based on the weight and height of the specimens tested.

Conclusions: The data is considered for included in the study by ACEA/ISO (Lebarbé, 2010) which is presented below.

Lebarbé 2010
New thorax pendulum response corridors were developed and proposed by Lebarbé 2010 for the on behalf of ISO/TC22/SC12/WG5 as part of a larger project in which a Task Force were to develop a worldwide-accepted set of biofidelity specifications for a 50th percentile frontal car crash test dummy.

A few steps were identified and adopted in the process of developing the new response corridors. These were mainly determination of the test sample, normalisation of the data, corridor construction and muscle tensing correction.

Determination of the test sample
The inclusion criteria defined by Lebarbé allowed doing a first sorting of the impactor database. The result is a set of 38 data sets which meet the inclusion criteria and for which data is available:
- One tests from CEESAR/LAB
- Seven tests from INRETS pendulum test series which were reported by Bouquet *et al.*, 1994
- Twenty tests from the General Motors test program and which were reported by Nahum *et al.*, 1970, Kroell *et al.*, 1971 and Kroell *et al.*, 1974

The test configuration was the same in all tests series: a rigid 6 inch diameter flat impactor shape centred on the mid sternum impacted the PMHS thorax but the data set presents various impactor masses, initial velocities and various PMHS anthropometries.

There after a shape analysis was carried out and three subsets were identified:

- High speed-low mass
- High speed-high mass
- Low speed-high mass

Normalisation
A mass-spring model was used for normalisation of the data. The scaling factors used were calculated based on the mass and speed of the impactor and to the stiffness and effective mass of the PMHS.

Corridor construction
The VRTC's method was used for response corridor construction.

Muscle tensing correction
No force shift in the PMHS response corridor was adopted for muscle response correction.

Results
The study presents response corridors for an impactor mass of 22.3 kg at an impactor velocity of 4.7 and 7.9 m/s. These were based on 30 PMHS tests (11 tests for the lower and 14 tests for the higher impactor speed (Table A-22 and A-23)). The author of the report also presented response corridors built for 23.4 kg impactor mass and 4.3 and 6.7 m/s impact speed to enable the use of the traditional Kroell based test conditions. The corridors for the higher impactor mass and slightly lower impactor velocities were recommended by the author for future inclusion in the frontal biofidelity requirements.

Conclusions: The lower impactor velocity response corridors and associated test are suggested to be included in the THORAX biofidelity requirements. These corridors were established using data from tests with eleven subjects. Out of these, seven subjects meet the inclusion requirements as specified in Chapter 4 while stature was not specified in the original publications for four of the subjects.

Table A-22 Specimen data for the tests included in the low speed corridor suggested by Lebarbé (2010)

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
60	11FF	Pendulum impact to the mid sternum, hub type	F	60	59	1,60	23	11
171	42FM		M	61	54	-	-	0
177	45FM		M	64	64	-	-	10
189	53FM		M	75	77	-	-	3
200	60FM		M	66	79	1,80	24	9
MRS01	MRT01		M	76	82	1,73	27	?
MRS03	MRT02		M	57	76	1,74	25	?
MRS04	MRT02		M	57	76	1,74	25	1
MRS05	MRT03		M	66	69	1,72	23	-
MRS06	MRT03		M	66	69	1,72	23	11
IMP574	MS589		-	-	-	-	-	20

- Data not available.

Table A-23 Specimen data for the tests included in the high speed corridor suggested by Lebarbé (2010)

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
61	12FF	Pendulum impact to the mid sternum; hub type	F	67	63	1,63	24	22
~~65~~	~~13FM~~		~~M~~	~~81~~	~~76~~	~~1,68~~	~~27~~	~~21~~
66	14FF		F	76	58	1,56	24	7
69	15FM		M	80	53	1,65	19	13
76	18FM		M	78	66	1,77	21	14
77	19FM		M	19	71	1,96	19	0
~~79~~	~~20FM~~		~~M~~	~~29~~	~~57~~	~~1,80~~	~~17~~	~~0~~
83	22FM		M	72	75	1,83	22	17
85	23FF		F	58	61	1,63	23	23
86	24FM		M	65	82	1,83	24	24
93	31FM		M	51	75	1,83	22	14
94	32FM		M	75	54	1,71	19	20
96	34FM		M	64	59	1,78	19	13
99	36FM		M	52	75	-	-	7
104	37FM		M	48	74	-	-	9
178	46FM		M	46	95	-	-	0
190	54FF		F	49	37	-	-	7
191	55FF		F	46	81	-	-	8
204	64FM		M	72	63	1,63	24	6

Patrick 1981

Sub-injury force-deflection curves were obtained by Patrick, 1981 from eight impacts to a volunteer with a 10 kg, 153 mm diameter padded striker at impact velocities of 2.4 to 4.6 m/s. The volunteer was 50 years of age, weighted 73 kg and had a stature of 1,73 m. The resulting BMI for the test subject was 24. The data was generated under similar conditions as in the Kroell study using unrestrained back.

Conclusions: The few tests carried out on a single volunteer in the experiments by Patrick does not produce data representative of the average occupant. The data set is therefore not recommended for inclusion in the THORAX biofidelity requirements.

Nusholtz 1985

Nusholtz *et al*., 1985 reported on a series of tests using eight unembalmed re-pressurised PMHS. The focus of the research was on the trauma to the soft-tissue organs surrounded by the thoracic cage, as well as the kinematic response of the thoracic cage. The stationary vertical PMHS were struck by a steering wheel assembly affixed to a 65 kg or 25 kg impactor, at three velocities in the range of 2.7–11 m/s.

The response of the thorax of a repressurised PMHS to direct loading from a steering wheel system was measured in four ways. The force obtained from a load cell placed directly behind the setting wheel hub, accelerometers attached to the thorax skeletal structure, pressure transducers placed in the descending aorta and high speed film analysis of the kinematics.

The results showed a complex three-dimensional kinematic response of the thorax that was dependent upon the initial configuration of the steering wheel relative to the test subject. Severe injuries involving the major organs/arteries protected by the ribcage were shown to be impact position dependent, and as such it was concluded that impact tolerance levels based on deflection and velocity or a combination or both might be inappropriate for steering wheel impacts. In particular the location of the liver with respect to the impact device was felt to be an important criterion to be addressed in thoraco-abdominal impact.

Conclusions: The thorax response highly sensitive to test setup. The data set is therefore not recommended for inclusion in the requirements.

Yoganandan 1997

Oblique lower ribcage pendulum impactor tests were carried out on seven embalmed PMHS (Table A-22) by Yoganandan *et al*., 1997. A 23.5 kg heavy impactor, with a front surface diameter of 150 mm and with a 40 mm thick Ensolite padding on the impact face, impacted the lower region of the ribcage at a velocity of 4.3 m/s. In these tests, the torso was initially rotated from right to left by 15°, such that the impact occurred on the right antero-lateral ribcage. The posterior region of the torso was unsupported. The thoracic vasculature and pulmonary system were both repressurised prior to testing.

The subjects received multiple fractures; commonly 2-4 fractures on the impacted side, but in a few cases also fractures on the un-impacted side of the ribcage.

Chest band deflection histories were normalised with respect to the initial chest depth. All force-deflection responses in the original paper were also scaled to a standard body weight of 75 kg using the method proposed by Eppinger, 1976, although this appeared to increase the scatter in the force-deflection responses. Also force-time and displacement-time responses were defined in the paper.

Table A-22 Specimen data for the oblique lower ribcage PMHS carried out and reported by Yoganandan *et al.*, 1997

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
~~1~~	~~1~~	Oblique pendulum impacts, padded hub type	~~M~~	~~72~~	~~82~~	~~1,70~~	~~28~~	4
2	2		M	81	63	1,75	21	4
3	3		M	84	68	1,68	24	0
4	4		M	86	56	1,70	19	2
5	5		M	62	61	1,74	20	3
~~6~~	~~6~~		~~M~~	~~70~~	~~91~~	~~1,69~~	~~32~~	4
7	7		M	68	83	1,78	26	6

Conclusions: This test series provide unique loading conditions. As such it recommended for inclusion in the THORAX biofidelity requirements.

Stalnaker 1973

Stalnaker *et al.*, 1973 used a similar test configuration with the impact velocity varied from 5.35-6.71 m/s, but an impacting mass of only 10 kg. There were 8 male and 2 female PMHS and the average mass of all of the subjects was 54.9 kg.

In contrast to the Kroell study the impacting mass was stopped after some fixed displacement of the impactor (unlike the Kroell tests where normally the impactor was stopped completely by the PMHS). The Stalnaker data represents a relaxed individual, no adjustments to the data to take account of muscle tensing were made and the data represents impactor penetration whereas the Kroell corridors represent sternal deflection. No formal performance corridors were developed, rather bands of data.

See conclusions under Neathery 1974 above.

A.1.4 Out-of-position and non-standard seating position experiments

Lebarbé 2005

In the study by Lebarbé *et al.*, 2005 nine embalmed PMHS on a on a static test bench were submitted to a frontal airbag deployment in out-of-position configuration (Table A-23). Two phases of this event and the combination of these were studied; the punch-out loading of the thorax that occur when the bag is being unfolded, and the membrane effect that occur when the bag is totally unfolded but pressurised.

Prior to tests, the PMHS were pressurised so that the vascular pressure was close to physiological conditions. Instrumentation included spine and sternum accelerometer arrays and rib strain data. No thorax deformation measurements except strain data from a limited number of gauges were provided. The study presented thoracic injuries generated by the three tests conditions.

Table A-23 Data for the specimen included in the study by Lebarbé *et al.*, 2005

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
M13_1	554	AB; Membrane 13	M	76	77	1.70	27	12
M13_2	555		M	67	65	1.75	21	15
M78_1	559	AB; Membrane 78	M	73	67	1.74	22	11
~~M78_2~~	~~561~~		~~M~~	~~72~~	~~83~~	~~1.73~~	~~28~~	~~0~~
~~M128_1~~	~~560~~	~~AB; Membrane 128~~	~~F~~	~~74~~	~~73~~	~~1.60~~	~~29~~	~~0~~
P52_1	557	AB; Punch	M	79	70	1.66	25	0
P52_2	558		F	80	64	1.58	26	10
C52_1	562	AB; Combined	M	80	62	1.67	22	15
C52_2	565		M	72	60	1.70	21	23

Conclusions: The study includes few subjects per test group, each subject suffer from a large number of rib fractures and lack thorax deformation measurements. As such the data set is not recommended for inclusions in the THORAX biofidelity requirements, but may be of interest for defining injury risk functions.

Trosseille 2008

In the study by Trosseille *et al.*, 2008 eight unembalmed and stationary PMHS were submitted to different type of loading to study rib strain patterns (
Table A-24). The loading conditions were either a 23.4 kg impactor propelled at 4.3 m/s or deployment of an unfolded available airbag. The direction of the load was varied in the tests; in 0° tests the load was applied from the front, in 60° the load was applied more from the side than from the front, and in 90° the load was applied from the side.

Prior to tests the artery system of the PMHS was pressurised. The ribs, in some subjects also the cartilage, were heavily instrumented with strain gauges in order to study the rib deformation pattern when submitted to different loading conditions.

Table A-24 Selected data for the tests included in the study by Trosseille *et al.*, 2008

Test	PMHS No.	Restraint *	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
575	575	AB 60°R	M	72	59	1,56	24	4
577	577	AB 90°R	M	76	66	1,62	25	6
585	585		M	89	64	1,68	23	14
586	586	Pendulum, Hub 90°L	M	74	77	1,76	25	4
587	587		M	82	78	1,80	24	9
588	588	AB 90°R	M	88	69	1,67	25	0
589	589	Pendulum, Hub 0°	M	88	60	1,69	21	14
594	594	AB 0°	M	78	65	1,70	22	3

* The number refer to the direction of the loading in the horizontal plane; 90° being perfectly lateral and 0° being perfectly frontal. R = Right, L = Left

Conclusions: The study includes two frontal tests, one of each loading type, and a single oblique test with an airbag. Due to the small size of the data set, it is not recommended for *inclusions in the THORAX biofidelity requirements, but may be of interest for defining injury risk functions.*

A.2 Shoulder complex

Biomechanical research concerning the behaviour of the shoulder during frontal impact conditions is rare. A few of the studies concerning the loading of the thorax have also presented some data on the behaviour of the shoulder. These studies will be discussed in detail in this section.

A.2.1 Sled tests

Törnvall 2008

Törnvall *et al.*, 2008 carried out test with unembalmed PMHS to study the belt-to-shoulder interaction in 45° far-side and 30° near-side collisions (Table A-25 and Figure A-12). For comparison, also full frontal tests were carried out using the same PMHS that were used in the oblique crashes. The subjects were exposed to multiple trauma, but the collision severity was moderate and no injuries were detected (by palpation only). A hard seat, with 50 mm deformable foam placed between subject and seat cushion, and standard belts were used in the study (anchorage points, belt configurations, seat dimensions etc are available upon request).The seat and restraint system was mounted to a sled system that could be adjusted to facilitate different collisions angles. The subjects, three in total, were equipped with three-dimensional film targets (head, T1, shoulder and sternum), which were rigidly attached to the underlying bone by the use of screws. No chest band or other means to study chest compression was used. The subjects were ventilated once prior to test and the ventilation system left open during the test.

Table A-25 Test matrix for experiments conducted jointly by Chalmers and Graz (Törnvall *et al.*, 2008)

Test	PMHS No.	Restraint*	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
16	1	3pt SB, full front	F	84	62	1,64	23	0	27
19	2		M	59	61	1,66	22	0	27
~~22~~	~~3~~		~~M~~	~~71~~	~~94~~	~~1,79~~	~~29~~	~~0~~	~~25~~
~~23~~	~~3~~		~~M~~	~~71~~	~~94~~	~~1,79~~	~~29~~	~~0~~	~~26~~
17	1	3pt SB, 45° far-side	F	84	62	1,64	23	0	27
20	2		M	59	61	1,66	22	0	28
~~24~~	~~3~~		~~M~~	~~71~~	~~94~~	~~1,79~~	~~29~~	~~0~~	~~26~~
18	1	3pt SB, 30° near-side	F	84	62	1,64	23	0	27
21	2		M	59	61	1,66	22	0	27
~~25~~	~~3~~		~~M~~	~~71~~	~~94~~	~~1,79~~	~~29~~	~~0~~	~~26~~

* Near side means that the PMHS moved towards the upper belt anchorage point during the test. Far side means that the PMHS moves away from the upper belt anchorage point during the test.

Conclusion: These tests are low severity, test conditions well documented, seat and restraint system are available and all three tests configurations are relevant. The lack of thorax instrumentation reduces and the number of subjects tested reduces the benefit of using this data set to evaluate the thorax biofidelity. The particular data set is mainly useful in the *assessment of the shoulder-to-belt interaction in oblique collisions. As such it is rather unique and therefore recommended for inclusion in the requirements.*

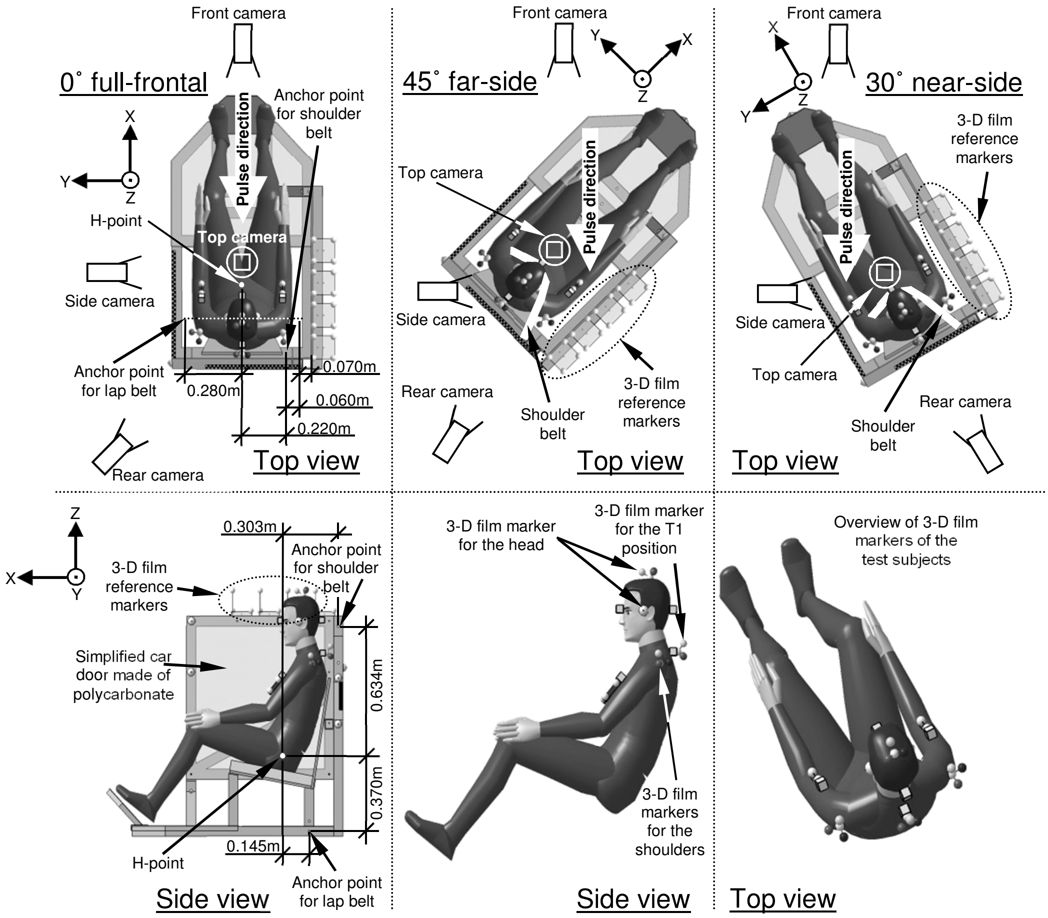


Figure A-12 Schematic of the 0° full-frontal, 45° far-side and 30° near-side test set-up used by Törnvall *et al. (adopted from Törnvall et al.*, 2008)

Shaw 2009

Shaw *et al.*, 2009a and b carried out frontal sled tests with eight restrained un-embalmed PMHS at 40 km/h. On top of the provision of detailed thorax compression and spine kinematics during the collisions, the tests provide 3 dimensional displacement data on the belted and unbelted shoulder and belt loads in full frontal loading conditions using a single standard shoulder and lap belt. For details on the subjects and test setup please refer to Appendix A.1.1

Conclusions: These tests are carried out at a rather high severity, test conditions are very well documented, seat and restraints systems are available and the number of testes subjects larger than in many other PMHS and volunteer studies. In addition the film targets used to assess the shoulder displacements were designed and firmly attached to enable calculation of the kinematics of the relevant position inside the shoulder complex. As such this data set is rather unique and therefore recommended for inclusion in the requirements.

A.2.2 Table top tests

L'Abbé et al., 1982

L'Abbé *et al.*, 1982 conducted tests with ten volunteers to study the thorax response, including the mid clavicle deflections, under belt compression. Please refer to Appendix A.1.2 for additional details.

Conclusions: The belt loads applied in these tests are about 20% of those in a representative collision. Due to this and lack of information on belt to clavicle bone position during the test this data set is not recommended for inclusion in the biofidelity requirements.

Cesari 1990

Cesari and Bouquet, 1990; 1994 carried out table top tests on 13 PMHS using a pendulum mass of 22.4 kg or 76 kg that loaded the belt system at impact velocities of 3 to 9 m/s. The seat belt was placed across the torso in a geometrical layout similar to that of a car driver wearing a shoulder belt. Instrumentation attached to the subject was not only able to monitor the chest deformation but also the mid clavicle deformation in the vertical direction (for-aft displacement of the clavicle bones). In addition belt forces were recorded. The PMHS data caution must be taken because the PMHS sustained a high number of rib fractures and some even flail chest.

Conclusions: The subjects that meet the inclusion criteria as specified in Chapter 4 are limited to four per relevant loading conditions. The data can be used to assess the clavicle repose in case additional data (belt position in relation to the clavicle bone and measurement location relative the length of the clavicle bone) are made available to the project.

A.2.3 Static shoulder range-of-motion and stiffness tests

Törnvall 2004

Törnvall *et al.*, 2004 measured the range-of-motion and the shoulder stiffness of seated volunteers in simulated collisions scenarios. The shoulder displacements relative to the chest were recorded using photometry when the arms of the volunteers were pulled either forward, forward-upward or upward while the chest was prohibited from forward and upward motions. In a complimentary analysis of the data by Törnvall *et al.*, 2008, it was observed that the spine curvature changed as a function of applied load to such a large degree that the data could not be used as published in 2004. In the reanalysis of the data, Törnvall estimated the shoulder displacements relative to T1 from the contour of the subjects' necks. This analysis enabled a better comparison with the crash test dummies.

Conclusions: Due to spine motions during the loading of the volunteers a comparison of the dummy and the volunteers shoulder responses are deemed to be inexact. In addition no load case in which the shoulder is pulled rearward by the belt is included. The data set is not recommended for inclusion in the biofidelity requirements.

Appendix B **Lebarbé (2010)**

Recommended Biofidelity Requirement:
Absolute chest deflection
Pendulum impactor tests

Biofidelity corridors for a 50th percentile adult male, based on frontal impact thorax impactor tests, were defined by Lebarbé, 2010. These were developed on behalf of ACEA (European Automobile Manufacturers Association) and the Frontal Biofidelity Specification International Task Force of the ISO (TC22/SC12) Working Group 5. The aim of this task force is to develop a set of biofidelity specifications for a 50th percentile frontal car crash test dummy which can be accepted worldwide. Therefore, the task force has a remit to define biofidelity requirements for all body regions. However, the first report from the group related to thorax impactor tests, as reported by Lebarbé, 2010.

The report by Lebarbé documents:

- The exhaustive list of the PMHS impactor tests gathered from the literature
- The criteria developed and used to select the tests to build the biofidelity specifications
- The description of all the methods necessary to process the records and construct the corridors
- The thorax impactor corridors

Impactor biofidelity corridors have been available for many years, as they were developed by Neathery, 1974. However, recent studies dealing with biofidelity corridors have treated the data in a way which attempts to reduce the amount of subjectivity in the corridor definition process. Therefore, it was deemed necessary for the Task Force to investigate these recent processes and the possible inclusion of updated impactor requirements.

In the development of new biofidelity requirements, a five-step process was defined:

1. Determination of the test sample
2. Normalisation
3. Signal alignment / deflection shift
4. Corridor construction
5. Muscle tensing correction

Additionally, there were decisions taken on the inclusion or exclusion of tests, and where necessary test responses were digitised from traces published in the original papers.

The test database used by the Task Force was put together from a review of both the literature and the NHTSA database. The Task Force opted to consider test conditions similar to those used by Nahum *et al.*, 1970 as including tests similar to these would lead to the largest number of tests in the subset identified. Other impact conditions were reviewed but not included in the limited biofidelity specifications of the Task Force.

An initial dataset of tests was picked out, which included tests from three different series:

- CEESAR/LAB (1 test)
- INRETS (7 tests)
- General Motors (36 tests)

The test configuration was the same for all of the initially selected tests in that a 6 inch diameter (152 mm), flat-face, impactor struck the PMHS on the sternum (centred at mid-sternum or level with the 4th rib). The 44 tests meeting the requirements of the Task Force are listed in Table B-1.

Apart from Test MS589, the data were not available in numerical format for analysis by Lebarbé, therefore the Task Force digitised the curves from the pdf format of the original paper.

Table B-1 Initial impactor test dataset selected by the Frontal Biofidelity Task Force (Lebarbé, 2010), NRF: Number of rib fractures ITC: Invertube contact time

Lab/author	Test No.	Impactor mass (kg)	Impactor speed ($m.s^{-1}$)	Age	NRF	Stature (m)	Mass (kg)	Chest depth (mm)	ITC
Kroell	05FM	19.28	5.14	60	3	1.85	86.18	257	
Kroell	06FM	19.28	5.14	83	11	1.83	77.11	254	
Kroell	07FF	19.28	4.02	86	11	1.68	37.65	200	
Kroell	09FM	19.28	5.14	73	0	1.85	76.2	238	
Kroell	10FF	19.28	4.92	82	12	1.60	43.1	168	
Kroell	11FF	19.5	6.3	60	11	1.60	59.0	208 (est)	30.8
Kroell	12FF	22.9	7.2	67	22	1.63	62.6	187	65
Kroell	13FM	22.9	7.4	81	21	1.68	76.2	246	65
Kroell	14FF	22.9	7.3	76	7	1.56	57.6	216	65
Kroell	15FM	23.6	6.9	80	13	1.65	53.1	200	65
Kroell	18FM	23.6	6.7	78	14	1.77	65.8	219	65
Kroell	19FM	23.6	6.7	19	0	1.96	71.2	203	65
Kroell	20FM	23.6	6.7	29	0	1.80	56.7	203	65
Kroell	21FF	23.6	6.8	45	18	1.74	68.5	213	65
Kroell	22FM	23.6	6.7	72	17	1.83	74.8	226	65
Kroell	23FF	19.5	7.8	58	23	1.63	61.2	226	65
Kroell	24FM	22.9	9.7	65	24	1.83	81.6	251	25.11
Kroell	25FM	5.5	13.9	65	18	1.68	54.4	207	65
Kroell	26FM	1.9	11.3	75	0	1.73	63.5	248	65
Kroell	28FM	1.6	14.6	54	0	1.83	68.0	238	65
Kroell	30FF	1.6	13.3	52	3	1.56	40.8	180	65
Kroell	31FM	23	10.2	51	14	1.83	74.8	238	30.9
Kroell	32FM	22.9	9.9	75	20	1.71	54.4	248	18.9
Kroell	34FM	19	8.3	64	13	1.78	59.0	241	35.4
Kroell	36FM	19	7.2	52	7	1.83	74.8	226	65
Kroell	37FM	22.9	9.8	48	9	1.79	73.9	248	23.2
Kroell	42FM	22.9	4.87	61	0	1.83	54.4	216	65
Kroell	45FM	23	5.1	64	10	1.81	64.0	254	65
Kroell	46FM	19.3	7.4	46	0	1.78	94.8	286	65
Kroell	53FM	23	5.2	75	3	1.74	77.1	241	65
Kroell	54FF	19.6	6.71	49	7	1.63	37.2	2085	24.9
Kroell	55FF	19.6	9.92	46	8	1.77	81.2	241	21.6
Kroell	60FM	23	4.3	66	9	1.80	79.4	222	65
Kroell	62FM	9.98	6.93	76	(AIS 4)	1.74	50.3	245	65
Kroell	63FM	23	6.93	52	4	1.83	88	225	
Kroell	64FM	23	6.93	72	6	1.63	63.0	216	49.2
INRETS	MRS01	23.4	3.36	76	?	1.73	82.0	250	65
INRETS	MRS03	23.4	3.43	57	?	1.74	76.0	230	65
INRETS	MRS04	23.4	5.81	57	1	1.74	76.0	230	65
INRETS	MRS05	23.4	3.39	66	?	1.72	69.0	210	65
INRETS	MRS06	23.4	5.88	66	11	1.72	69.0	210	65
INRETS	MRS07	23.4	3.40	69	?	1.64	52.0	220	65
INRETS	MRS08	23.4	5.77	69	11	1.64	52.0	220	65
CEESAR	MS589	23.67	4.4	88	20	1.69	60.0	180	65

In many of the force response curves, an initial spike in force was noted by Lebarbé. Earlier treatment of the data was to round-off this spike believing it to be an artefact; an oscillation excited by the release of the striker. According to Lebarbé, modern use of inertia compensated force measurement has shown this spike to be due to both a measurement artefact and a true inertial response of the subject. The Task Force decided to keep the early spike in the force/deflection corridor definition but to define the portion of the curve with the early spike as not mandatory for the biofidelity requirement.

Once the digitised data were available, the force-time and deflection-time measurements were processed with a CFC_600 filter (with only a few exceptions where a CFC_180 filter was used instead). Lebarbé makes no mention in the Task Force report of the filtering used by the original authors. Indeed Nahum *et al.*, 1970 and Kroell *et al.*, 1971 don't make any direct statements of any filtering that was applied in their test work. However, Kroell *et al.* comment that the light beam oscillograph used in the data recording had a flat (within 5 %) frequency response up to 1,000 Hz. Therefore use of a CFC_600 filter seems appropriate to match a high frequency cut-off around 1,000 Hz.

For six of the original 44 tests, the response including the early inertial spike was not available; therefore the sample size was reduced to 38 before any signal analysis was carried out.

Shape correlation using the Nusholtz *et al.*, 2007 method was used to determine if any of the responses were atypical. The shape analysis identified three samples to be used separately for the development of biofidelity corridors:

- High speed, low mass
- High speed, high mass
- Low speed, high mass

Lebarbé proceeds to define biofidelity requirements for the two high mass groups. However, for the purposes of the THORAX biofidelity requirements, only the low speed impact condition meets the impact speed definition for the dummy demonstrator. For this reason only the low speed requirement is considered further in this review.

Following the signal analysis and derivation of consistent samples, 11 tests comprised the low speed, high mass group. The sample group used by Lebarbé is shown in Table B-2.

Table B-2 Low speed, high mass, impactor test dataset selected by the Frontal Biofidelity Task Force (Lebarbé, 2010), NRF: Number of rib fractures

Test No.	Impactor mass (kg)	Impactor speed ($m.s^{-1}$)	Age	NRF	Stature (m)	Mass (kg)	Chest depth (mm)
11FF	19.5	6.3	60	11	1.60	59.0	208 (est)
42FM	22.9	4.87	61	0	1.83	54.4	216
45FM	23	5.1	64	10	1.81	64.0	254
53FM	23	5.2	75	3	1.74	77.1	241
60FM	23	4.3	66	9	1.80	79.4	222
MRS01	23.4	3.36	76	?	1.73	82.0	250
MRS03	23.4	3.43	57	?	1.74	76.0	230
MRS04	23.4	5.81	57	1	1.74	76.0	230
MRS05	23.4	3.39	66	?	1.72	69.0	210
MRS06	23.4	5.88	66	11	1.72	69.0	210
MS589	23.67	4.4	88	20	1.69	60.0	180

As mentioned in Section 6.1.3.1, The Task Force agreed to use a mass-spring model for the normalisation process. The normalisation accounted for impactor mass and speed, and subject stiffness and effective mass.

No signal alignment was required further to the original marking of time zero corresponding to a contact time mark on the oscillograph and on the movie. Regarding the use of total deflection or skeletal measurement, the Task Force opted to use the total external (surface) deflection. The dummy biofidelity therefore needs to be assessed on that basis. Dummy total deflection should be measured using high speed video, for instance, or extrapolated from an internal deflection measurement. In some cases it was necessary for Lebarbé to remove a deflection offset to generate the total deflection measure. This offset was 13 mm (~½ inch).

The corridors were developed using the method from the Vehicle Research and Testing Centre, Shaw *et al.*, 2006. This computes an ellipse based on the standard deviation around the mean in both the x-axis and y-axis measurement. This is shown in the subsequent figures by a shaded area. No correction for muscle tensing was used in this analysis.

The corridors presented in Figure B-1 were built for the loading phase of tests with a 23.4 kg impactor mass and a 4.3 m.s^{-1} impact speed.

- The corridor in red dotted lines is the Kroell corridor as defined in Neathery, 1974.
- The corridor in red continuous lines is the same Kroell corridor, but without muscle tensing correction (-670 N) and shifted to the right (+13 mm) to represent total deflection.
- The corridor shaded in grey is the corridor defined by Lebarbé. The black continuous line represents the average curve. This is the high mass, low speed corridor recommended by Lebarbé.

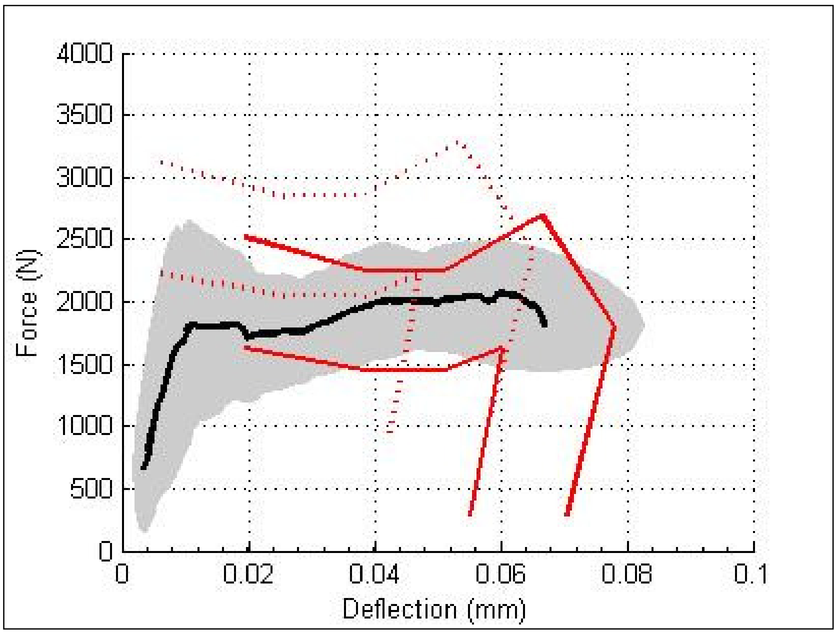


Figure B-1 Low speed corridors, 23.4 kg impactor mass, 4.3 m.s^{-1} impact speed (Lebarbé, 2010)

Figure B-2 to Figure B-4 show the biofidelity requirements for the 50th percentile male for the loading phase up to maximum mean force; from maximum mean force to maximum mean deflection; and from maximum mean deflection through the unloading phase. These are presented separately because the ellipses from which the unloading corridor are constructed substantially obscure the corridor for the loading phase if the whole corridor is plotted on one graph.

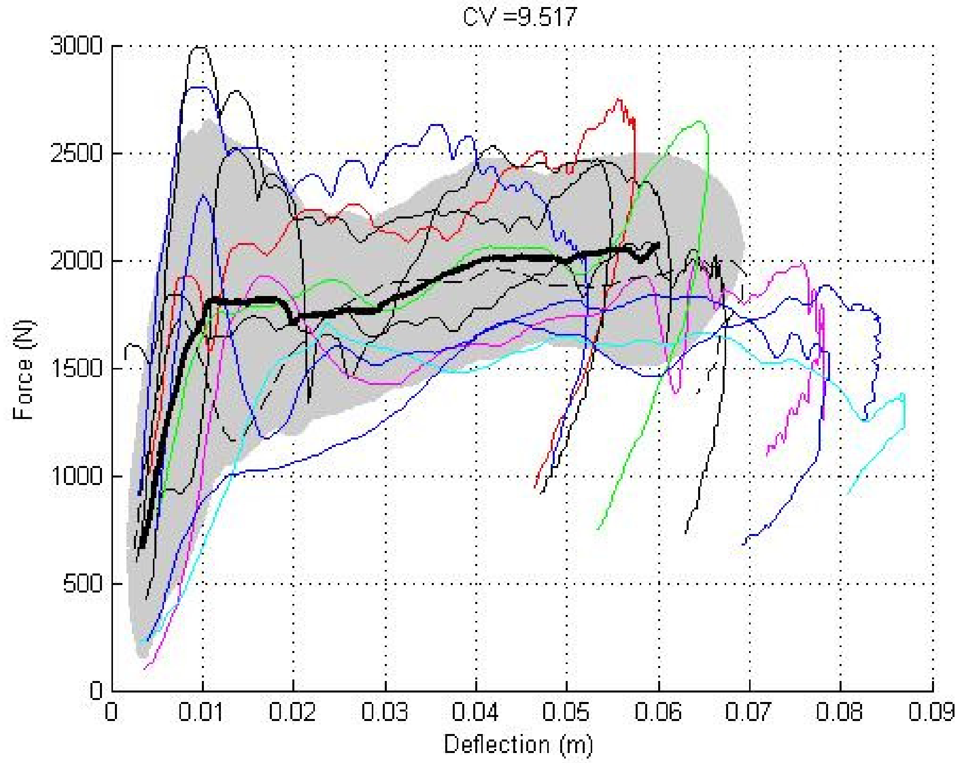


Figure B-2 Low speed corridor, 23.0 kg impactor mass, 4.6 m.s^{-1} impact speed – loading phase up to maximum mean force

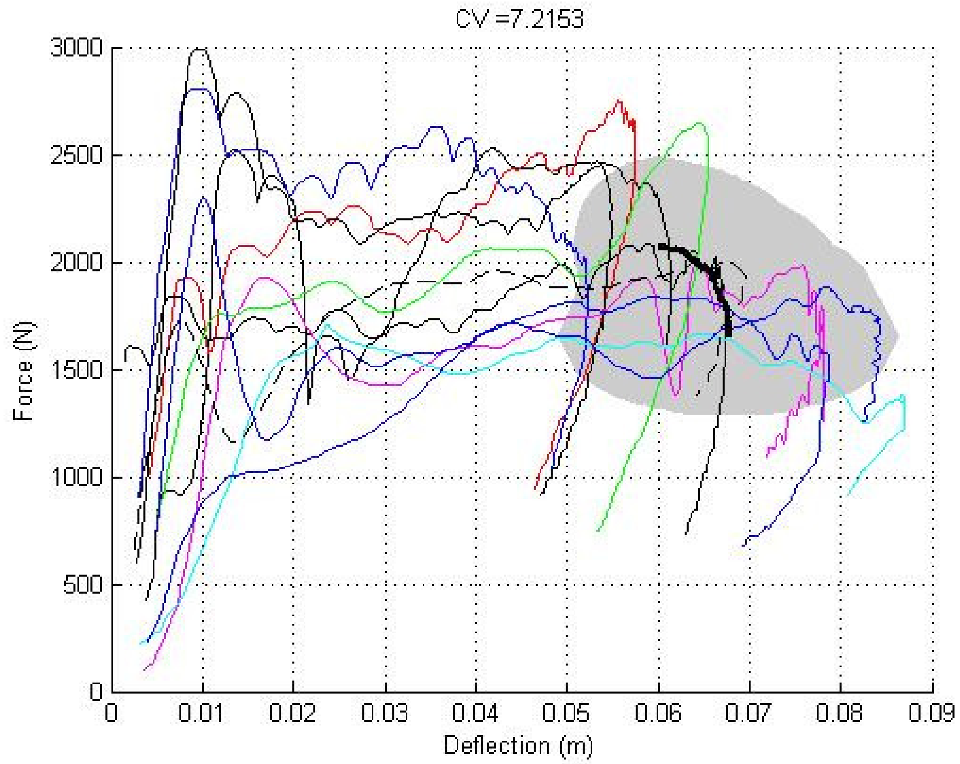


Figure B-3 Low speed corridor, 23.0 kg impactor mass, 4.6 m.s^{-1} impact speed – from maximum mean force to maximum mean deflection

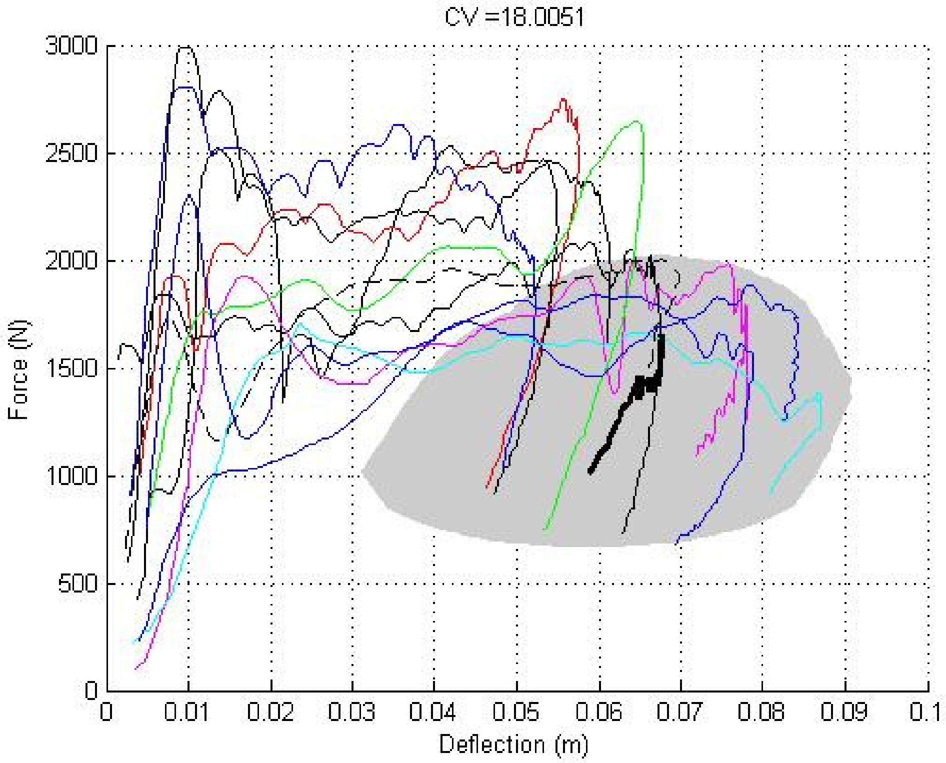


Figure B-4 Low speed corridor, 23.0 kg impactor mass, 4.6 m.s^{-1} impact speed – from maximum mean deflection through the unloading phase

For clarity, Figure B-5 and Figure B-6 also show the force-time and displacement-time requirements respectively.

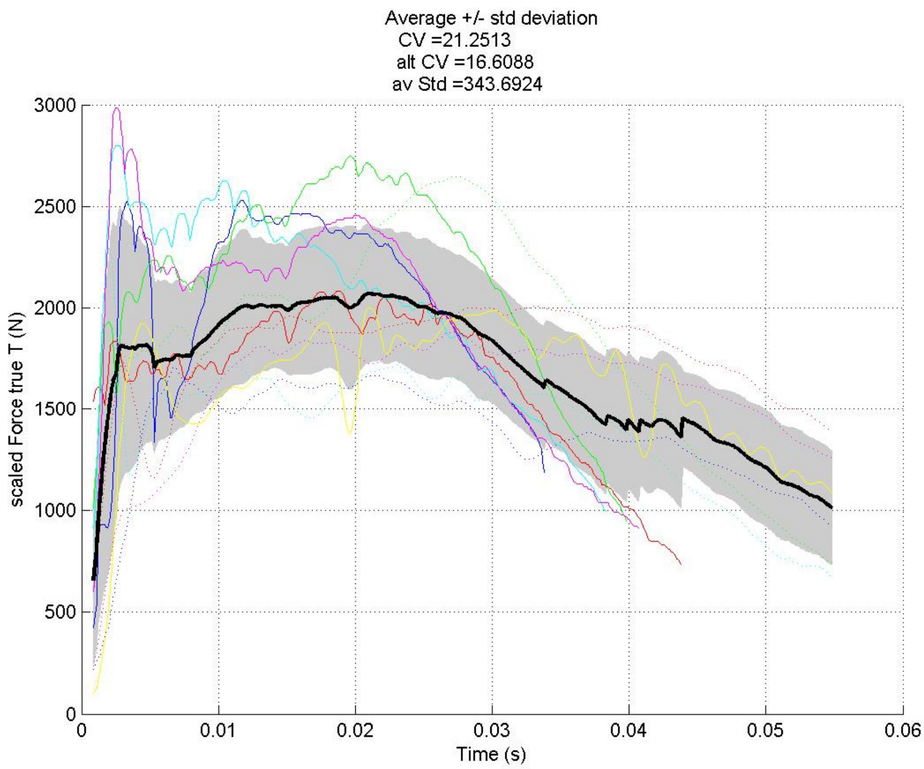


Figure B-5 Force-time corridor, 23.0 kg impactor mass, 4.6 m.s^{-1} impact speed

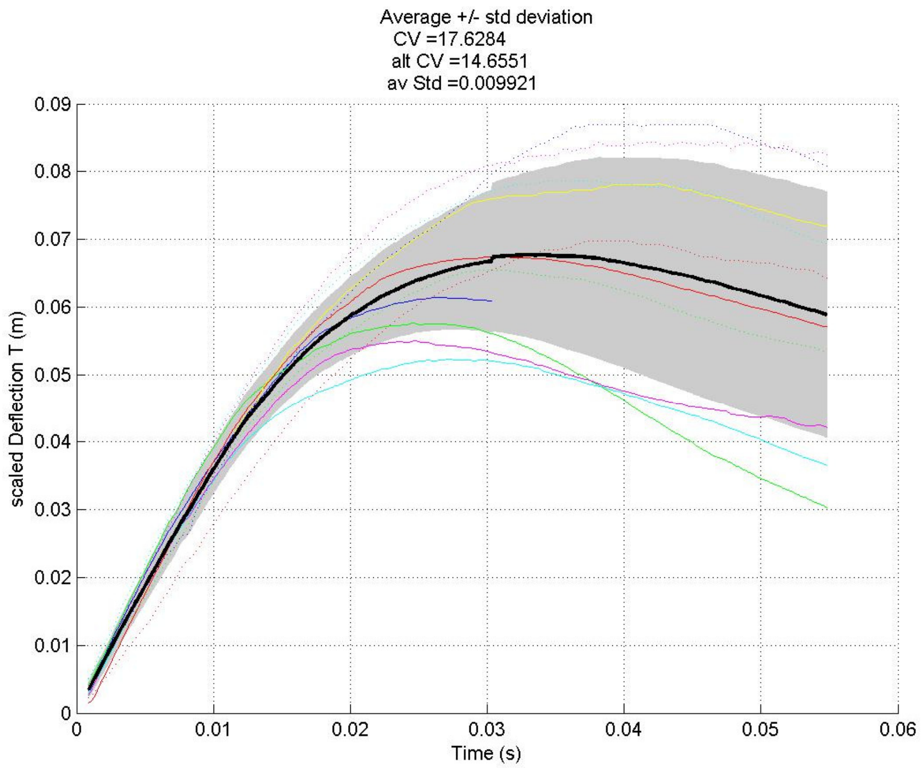


Figure B-6 Deflection-time corridor, 23.0 kg impactor mass, 4.6 m.s^{-1} impact speed

Using the same technique as the normalisation to develop the corridor shown in Figure B-1, the low speed, high mass responses were scaled to the size of the 5th percentile female. The fifth percentile size was defined with an effective mass of 15.1 kg and a chest depth of 175 mm. Impact conditions of an impactor mass of 23.0 kg and an impact speed of 4.6 m.s^{-1} were used as these gave an energy close to the mean for the low speed high mass group. This selection also kept the mass and speed close to the mean values for the group (22.9 kg and 4.5 m.s^{-1}).

The resulting corridor and scaled individual responses representing the fifth percentile female are shown in Figure B-7. This mean response and corridor (based on the standard deviations in force and deflection) could be used to assess the biofidelity of a fifth percentile female dummy. Scaling of this requirement to other sizes of dummy can be undertaken should another size of frontal impact dummy become available.

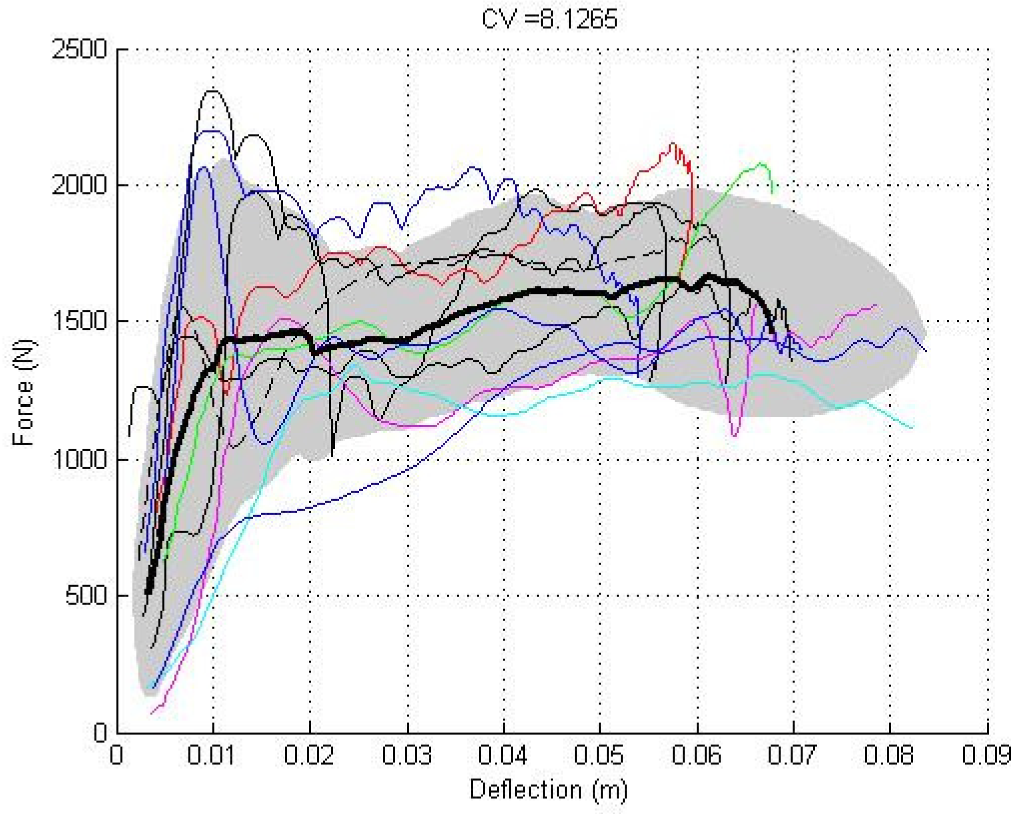


Figure B-7 Low speed corridor, 23.0 kg impactor mass, 4.6 m.s^{-1} impact speed, scaled to the 5th percentile female anthropometry (effective mass, 15.1 kg; chest depth, 175 mm)

It should be noted that the impactor size or mass used to define this biofidelity requirement has not been reduced in order to limit the energy imparted in a biofidelity test of this type. Previously, authors have reduced the impactor size and mass when specifying response requirements for a fifth percentile female. This may be to account for fears over robustness of the dummy or in side impacts to limit the number of ribs engaged by the face of the impactor. However, this dataset of PMHS tests includes small subjects which were struck with a high mass impactor. Therefore, ideally, a small dummy should be struck with a high mass impactor to replicate the conditions of the original PMHS tests. Departing from the ideal situation it may be that a small dummy is not able to accommodate the full deflection range implied by this corridor. Consideration may then be given to limiting the impact energy, somehow, to evaluate biofidelity over a reduced deflection range only. This should be reviewed alongside the specification and requirements for use of the dummy.

Appendix C **Yoganandan *et al.* (1997)**

Recommended Biofidelity Requirement:
Absolute chest deflection
Pendulum impactor tests

C.1 Short overall description

This requirement is based on 15° oblique impacts to the lower thorax performed by Yoganandan *et al.*, 1997. Seven PMHS subjects were tested once each, and normalised force-deflection requirements were defined. Subsequently, NHTSA defined a biofidelity requirement including force-deflection, force-time and deflection-time requirements (GESAC, 2005). The information concerning the test configuration is based on the information in: Yoganandan *et al.*, 1997; GESAC, 2005; and the test report on subject PC101, which is available from the NHTSA biomechanics database (report number B03085).

C.2 Test set-up

The following test configuration should be used:

- A pendulum impactor, with a load cell (mass 0.7 kg, based on current specifications for an Interface Model 1210) and 152 mm diameter impact plate (mass 1 kg) mounted on the striking face of the impactor. The total mass of the impactor, load cell, impact plate and foam padding should be 23.4 kg.
- The impact plate shall be instrumented with a uniaxial accelerometer for inertia compensation of the force measurement.
    - The impactor shall have a front face consisting of 19 mm thick, 152 mm diameter Ensolite padding (which may be substituted by 19 mm thick Rubatex padding conforming to the force-deflection characteristics given in GESAC, 2005.
- The impactor shall be aligned 15° left of full-frontal, such that the impactor strikes the right anterior-lateral thorax at the level of the 8th rib (approximately aligned with T12). For dummy biofidelity testing, impacts should also be conducted with the impactor aligned 15° right of full frontal to check that the dummy response is symmetrical. It was noted that there was no significant rotation of the torso during the loading period.
- The impact velocity shall be 4.3 m.s^{-1}.
- All data acquisition shall comply with SAE J211-1:2003. Electronic data shall be acquired at 10-20 kHz, and marker data shall be acquired at a minimum of 2250 Hz.
- The dummy should be seated upright on
    - A Teflon-coated surface (Yoganandan *et al.*, 1997), or
    - A thin Teflon sheet (GESAC, 2005).
- The dummy back of the dummy should be unsupported.
- The legs of the dummy should be fully stretched forward, and the arms extended forward and outward sufficient to allow the pendulum to contact the thorax without interfering with the arms.
- The dummy should be dressed in long underwear (GESAC, 2005).

C.3 Biofidelity requirements

The outputs required from this test are force-deflection, force-time and deflection-time curves. The impact force should be determined from the inertia-compensated load cell measurement. Alternatively, a standard pendulum impactor of mass 23.4 kg could be used and the force estimated from the impactor mass multiplied by the impactor acceleration. In the original PMHS tests, chest deflection was measured using a 40 gauge chest band aligned with the centre of the impactor, as well as by tracking markers on the impactor and T12 using a camera mounted perpendicular to the direction of travel of the pendulum. Either method may be used; if marker tracking is used it is recommended that a frame rate of at least 1000 frames per second is used (the original data was recorded at 2250 frames per second.

C.4 Data processing and normalisation

All seven subjects from the original Yoganandan *et al.*, 1997 data set were used in the definition of the biofidelity requirements. Two subjects had a BMI greater than the range specified in the inclusion criteria (PC101 had a BMI of 28.4 and PC106 had a BMI of 31.9. However, the response of both of these subjects was within the range for the sample. Several normalisation methods were attempted, including:

- Lebarbé, 2010 – using the following equation to estimate the effective mass of the subject (spine accelerations were recorded in the tests, but were not available for analysis so the conventional calculation for effective mass could not be used):

  $$m_2 = \frac{2.E_d.m_1}{m_1.v_0^{\,2} - 2.E_d}$$

  Where

  Ed = Energy (area under force-deflection) up to max deflection
  m1 = mass of pendulum
  v_0 = initial velocity of the pendulum

  and using the ratio of subject chest depth to 50th percentile male chest depth to represent the effective stiffness.
- ISO 12350, but using the effective mass calculated as above.
- Moorhouse, 2008, using the effective mass as calculated above, and using the effective stiffness calculated at maximum chest deflection. The standard (50th percentile) effective mass was estimated by averaging the percentage of effective mass to total body mass for each subject, and then multiplying that average by the known total body mass of the 50th percentile male (76 kg); this gave a standard effective mass of 22.97 kg. The standard effective stiffness was estimated using the mean effective stiffness of the seven subjects (52.34 $N.mm^{-1}$).

The modified Moorhouse method was the most effective at reducing the scatter in the results and was therefore selected. The same normalisation processing shall be applied to the dummy results as was applied to the PMHS results. The subject data for the normalisation calculations is given in Table C-1:

Table C-1 Specimen data for the oblique lower ribcage PMHS carried out and reported by Yoganandan *et al.*, 1997

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
1	1	Oblique pendulum impacts, padded hub type	M	72	82	1,70	28	4
2	2		M	81	63	1,75	21	4
3	3		M	84	68	1,68	24	0
4	4		M	86	56	1,70	19	2
5	5		M	62	61	1,74	20	3
6	6		M	70	91	1,69	32	4
7	7		M	68	83	1,78	26	6

C.5 Target corridors

The following target corridors shall be used:

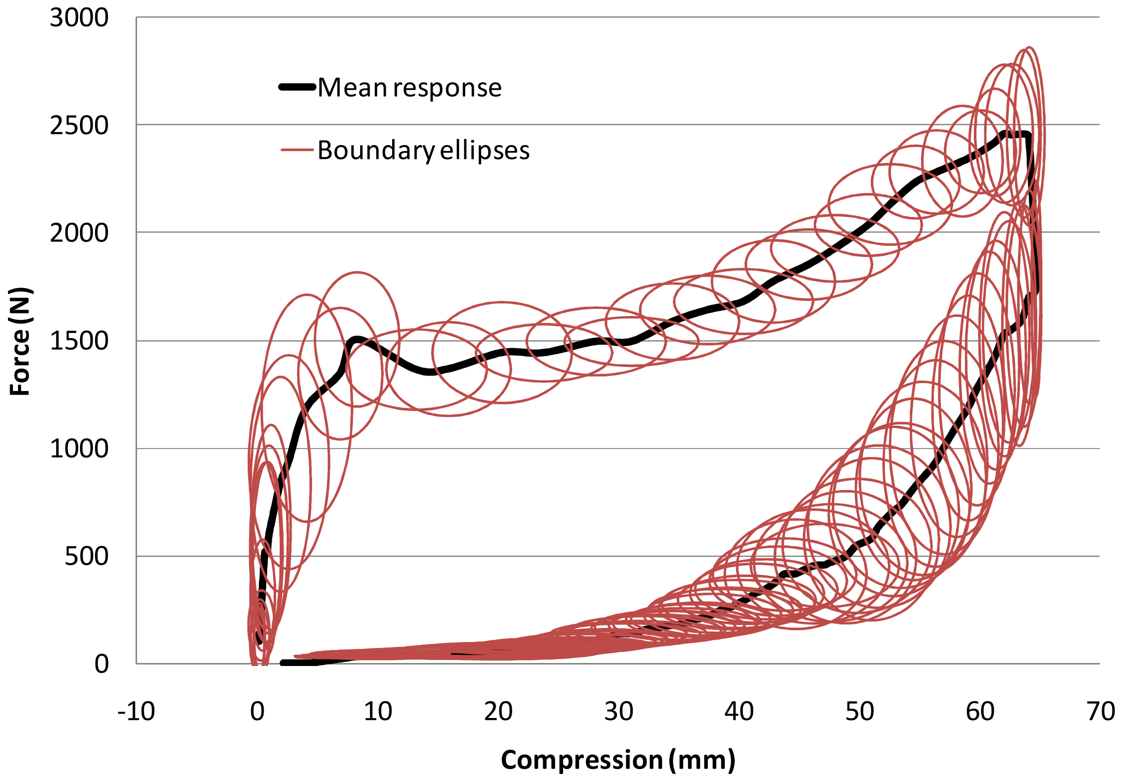


Figure C-1 Yoganandan *et al.*, 1997 normalised force-normalised deflection target corridor

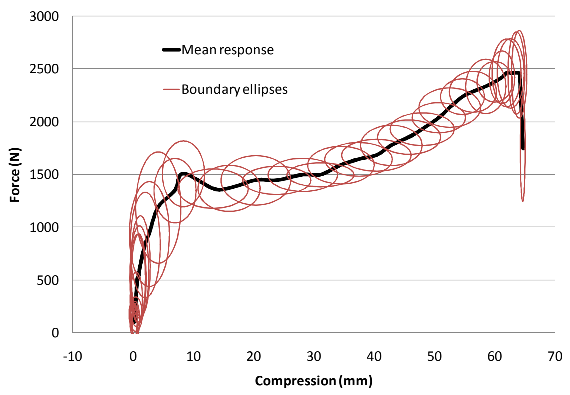


Figure C-2 Yoganandan *et al.*, 1997 normalised force-normalised deflection target corridor – loading phase

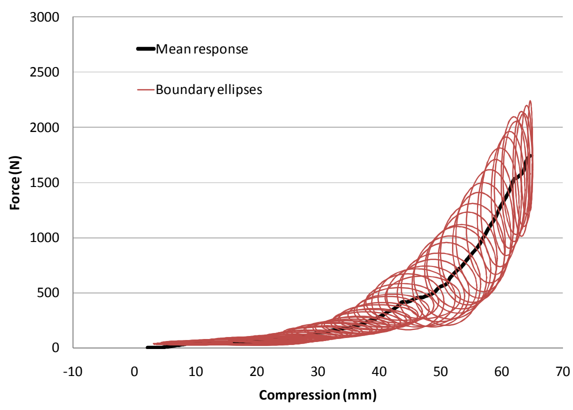


Figure C-3 Yoganandan *et al.*, 1997 normalised force-normalised deflection target corridor – unloading phase

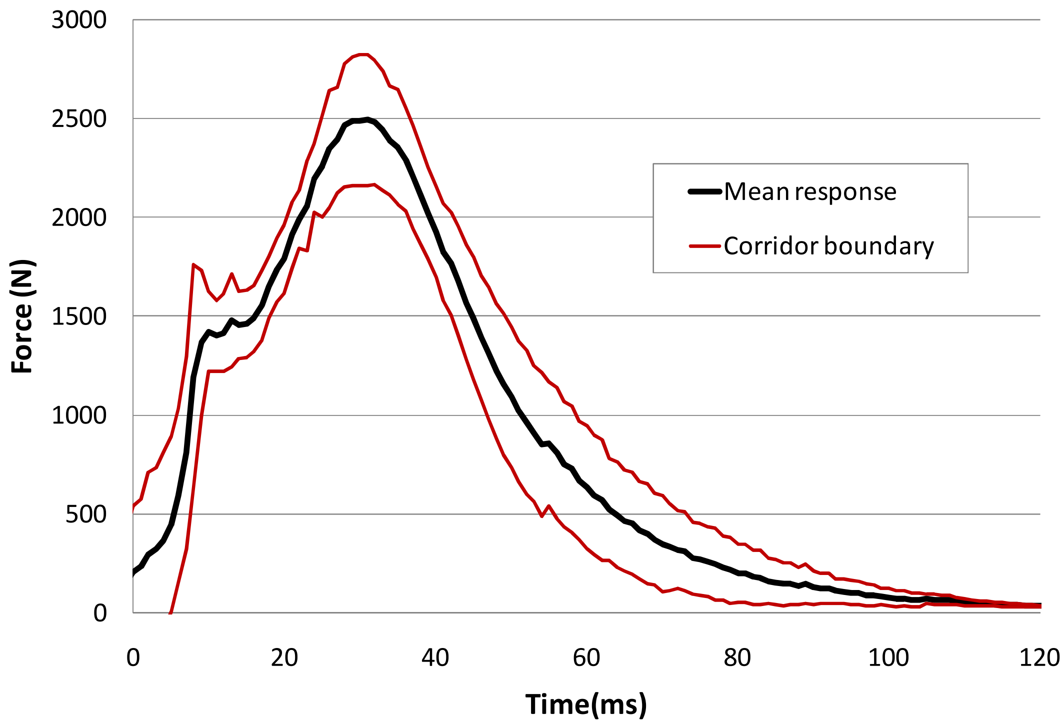


Figure C-4 Yoganandan *et al.*, 1997 normalised force-normalised time target corridor

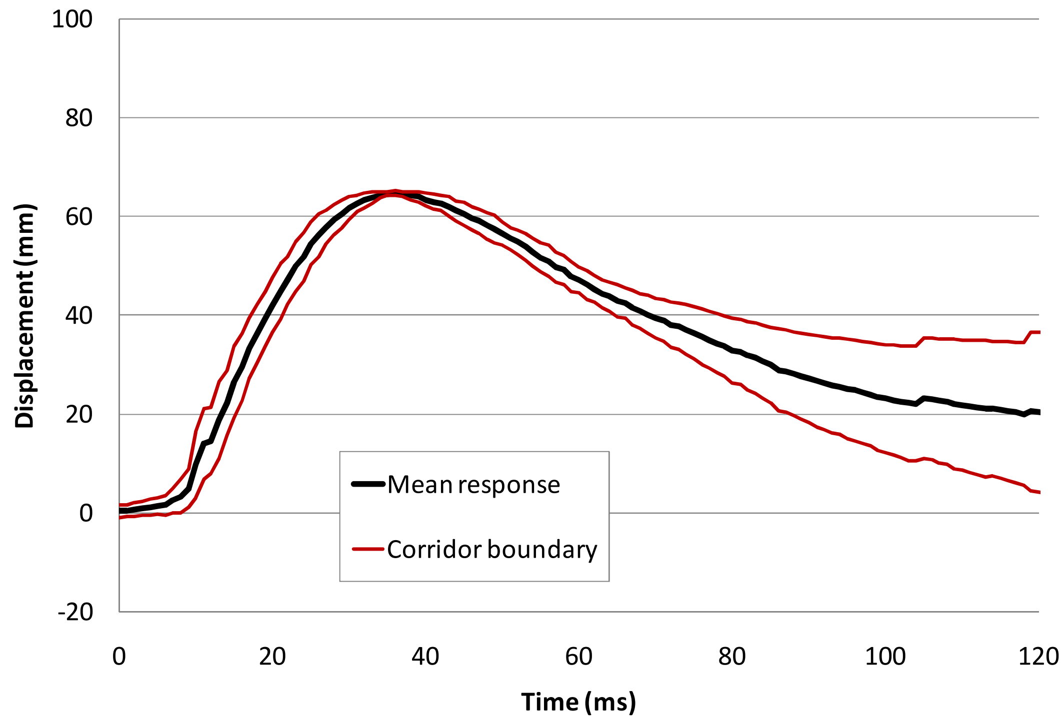


Figure C-5 Yoganandan *et al.*, 1997 normalised deflection-normalised time target corridor

The following difference with the GESAC, 2005 biofidelity requirements are noted:

- Padding given as square in GESAC, 2005, not circular
- The GESAC, 2005 requirements normalised the force and deflection using the method of Eppinger, 1976, using a standard 50^{th} percentile male body mass of 75 kg

Appendix D **Forman *et al.* (2006a), including Bolton *et al.* (2006)**

Recommended Biofidelity Requirement:
Absolute chest deflection
Sled tests

As noted in appendix A, Forman *et al.*, 2006a reported on series of sled tests carried out at UVa from 1999 to 2006. In addition Bolton et al. reported on a series of tests in 2006 that used the same buck, measurements techniques etc. as those reported by Forman. Certain sets of these test series are recommended as providing useful biofidelity requirements. These series are identified in the following table (Table D-1). In the table some tests appear crossed-through. This indicates tests which do not fulfil all of the requirements of the inclusion criteria. Moreover, all the tests as candidates to be used in defining a 3-point belt group fail to meet the inclusion criteria for the chest deformation since they were performed with chest bands using less than 40 gauges.

Table D-1 Sled tests from the University of Virginia (Forman *et al.*, 2006a) and Bolton et al. (2006)

Test	Restraint*	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)	Instrumentation	Chestband
UVA577	Passenger 3pt FL belt + AB	M	57	70	174	23.1	0	47.4	3-axis on T1 3-axis on T8 3-axis on T12 1-axis on suprasternum	40 / 2nd 40 / 5th
UVA578		F	69	53	155	21.9	4	47.6		40 / 2nd 40 / 5th
~~*UVA579**~~		*F*	*72*	*59*	*156*	*24.3*	*11*	*47.6*		*40 / 2nd* *40 / 5th*
UVA580		M	57	57	177	18.2	0	47.6		40 / 2nd 40 / 5th
UVA1094	Passenger 3pt SB	M	49	58	178	18.3	0	29.9	3-axis on T1 3-axis on T8 3-axis on T12 1-axis on suprasternum	40 / 2nd 40 / 5th
UVA1095		M	44	77	172	26.1	0	29.8		40 / 2nd 40 / 5th
UVA1096		M	39	79	184	23.5	0	29.4		40 / 2nd 40 / 5th
~~UVA665~~	Passenger 3pt SB + AB	M	55	85	176	~~27.4~~	3	48.9	3-axis on T1 3-axis on T8 3-axis on T12 1-axis on suprasternum	40 / 2nd 40 / 5th
~~UVA666~~		M	69	84	176	~~27.1~~	3	48.1		40 / 2nd 40 / 5th
UVA650	Lap belt + Passenger AB	M	40	47	150	20.8	4	48.9	3-axis on T1 3-axis on T8 3-axis on T12 1-axis on suprasternum	40 / 2nd 40 / 5th
UVA651		M	76	57	176	18.4	0	48.1		40 / 2nd 40 / 5th
UVA652		M	46	74	175	24.3	0	49.7		40 / 2nd 40 / 5th

* *UVA579: The pretensioner did not work properly but kinematics of subject do not seem to have been affected*

D.1 Normalisation and scaling

In the THORAX Project it was decided to put an inclusion criterion on the BMI, but not on the anthropometry. Consequently, important deviations between the 50th percentile and the individual subjects could exist. For instance a small female (height = 1.56 m, mass = 60 kg) has the same BMI as a tall man (2 m, 98 kg) that correspond to the BMI (24.6) of the 50th percentile (1.77 m and 77 kg) but will not have the same biomechanical response.

These deviations need to be checked and, if necessary, taken into account by adjusting the response according to the anthropometry. This can be done using a normalisation procedure as discussed in Chapter 6. No consensus has been reached regarding an appropriate normalisation method for sled tests. This is because the discrepancies between the tests may be caused by several factors in addition to the differences in size and weight. For these reasons, the normalisation procedure in several cases does not provide benefit.

Further investigation is needed to select a normalisation procedure. For that reason, the biofidelity requirements provided here are not built from normalised data. However, it should be noted that the corridors presented here can be easily updated once an accepted and well defined procedure of normalisation has been agreed within the consortium.

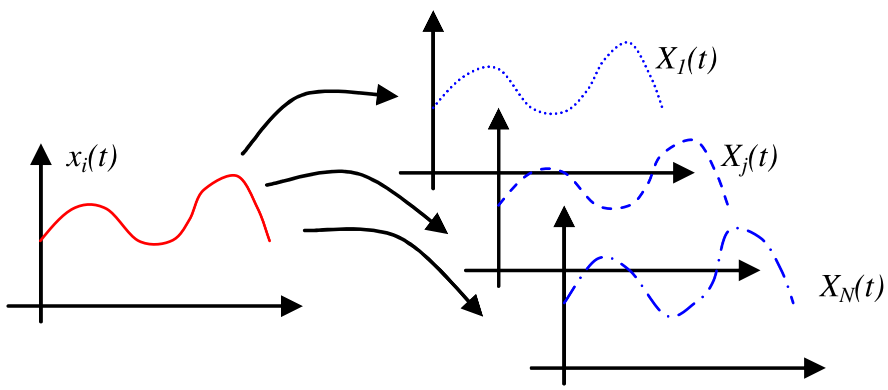
D.2 Biofidelity requirements

The biofidelity requirements are defined for different restraint system types using the tests selected in the Table D-1. Requirements are based on those defined by the EEVC WG12 after the FID EC project (EEVC WG12, 2003). EEVC proposed corridors for the sternum, the upper spine (T1), the chest (T8) and the lower spine (corresponding to T12/L1) resultant acceleration, and for the chest deformation. All these physical parameters are measurable on the dummy. However, in the selected tests only the x-axis sternum acceleration was measured so the corridor for the sternum resultant acceleration cannot be established.

Finally, corridors were drawn by using the method proposed by the ACEA/ISO Working Group to ensure a good coherence and harmonisation between the two initiatives (ACEA/ISO and THORAX Project). The corridors were defined following steps similar to those used by the ACEA/ISO working group and described hereafter:

- Step 1 – Determination of the test sample by signal analysis

The quality of the signals (T1, T8 and T12 resultant and chest deformation) from all of the tests was checked. To do it in an objective manner, the method set up by Nusholtz *et al.*, 2007, was used. This method uses the cross-correlation between the N signals from the same set of data in order to detect any outliers or bad signals. For that purpose the cross-correlations between a signal $x_i(t)$ and each signal $x_j(t)$ with $j = 1, N$ were computed.



The cross-correlation between two signals is defined by:

$$R_{x,y}(\tau) = \frac{\int_{-\infty}^{+\infty} x(t)\, y(t+\tau)\, dt}{\sqrt{\int_{-\infty}^{+\infty} x^2(t)\, dt \int_{-\infty}^{+\infty} y^2(t)\, dt}} \qquad \text{(Eq. 1)}$$

The coefficient for the shape correlation is defined by $R_{x,y}(\tau = 0)$, two signals have identical shape if:

$$R_{x,y}(\tau = 0) = 1 \qquad \text{(Eq. 2)}$$

The phase correlation is given by the value of the time delay τ where the value of $R_{x,y}(\tau)$ is at a maximum.

The magnitude correlation corresponds to ratio of the norms of the signal:

$$\frac{\|x(t)\|}{\|y(t)\|} = \frac{\int_{-\infty}^{+\infty} x^2(t)dt}{\int_{-\infty}^{+\infty} y^2(t)dt} \qquad \text{(Eq. 3)}$$

Both shape, magnitude and phase cross-correlation were analysed. The level of the shape correlation for which a test is assumed to be "similar" to the others was set as 0.6. The level for the ratio of magnitude correlation was defined also as 0.6 and the phase correlation was assessed as less than 10 ms. Using this method the inclusion/exclusion of the tests in grey in Table D-1 can be decided, objectively, based on this signal processing approach.

Consequently, the sled tests UVA534, UVA535, and UVA579 can be used in the definition of the biofidelity requirements. However, the x-axis value of the sternum acceleration does not show sufficient similarities between the different tests to be used to define a corridor with sufficient confidence. Moreover, the series UVA650-652 (not provided by University of Virginia, but download from NHTSA database) show very noisy signals despite the fact that the report explained that a CFC180 was used. These data should checked before inclusion.
The final set of tests based on this analysis is given in
Table D-2.

Table D-2 Sled tests from the University of Virginia (Forman *et al.*, 2006a) included after signal comparisons

Test	Restraint*	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)	Instrumentation	Chestband
UVA577		M	57	70	174	23.1	0	47.4		40 / 2nd 40 / 5th
UVA578	Passenger 3pt FL belt + AB	F	69	53	155	21.9	4	47.6	3-axis on T1 3-axis on T8 3-axis on T12	40 / 2nd 40 / 5th
UVA579		F	72	59	156	24.3	11	47.6		40 / 2nd 40 / 5th
UVA580		M	57	57	177	18.2	0	47.6		40 / 2nd 40 / 5th
UVA665		M	55	85	176	27.4	3	48.9		40 / 2nd 40 / 5th
UVA666	Passenger 3pt SB + AB	M	69	84	176	27.1	3	48.1	3-axis on T1 3-axis on T8 3-axis on T12	40 / 2nd 40 / 5th
UVA667		F	59	79	161	30.6	13	48.4		40 / 2nd 40 / 5th
UVA668		F	54	55	162	20.9	23	48.6		40 / 2nd 40 / 5th
UVA1094		M	49	58	178	18.3	0	29.9		40 / 2nd 40 / 5th
UVA1095	Passenger 3pt SB	M	44	77	172	26.1	0	29.8	3-axis on T1 3-axis on T8 3-axis on T12	40 / 2nd 40 / 5th
UVA1096		M	39	79	184	23.5	0	29.4		40 / 2nd 40 / 5th

Since there is no consensus regarding normalisation procedures for frontal impact sled test data, no normalisation was applied to the data used for the definition of the present corridors. Equally, no signal alignment or time-shift was applied to the data since the instant of the impact in sled tests was clearly defined and the evaluation (Step 1) of the phase correlation between signals did not detect any important shift between them.

In accordance with the method used by the ACEA/ISO task force to develop response corridors, the VRTC method (Shaw *et al.*, 2006) was used. The corridors presented in the present report have been computed using the VRTC's method. It should be noted that for acceleration vs. time or force vs. time corridors, the method defines a plot of the mean response with a corridor set at ± one standard deviation.

D.2.1 Three-point force-limited belt and passenger airbag requirements

These series of sled tests were performed at the University of Virginia with a 2000 Ford Taurus Buck with a reinforced OEM front passenger seat with anti-submarining pan. The PMHS were restrained with an OEM 3 point-belt with a nominal 4 kN force-limited retractor and a buckle-side pretensioner. A standard ford Taurus 1998 passenger de-powered airbag was deployed.

Four tests are available; a failure of the pretensioner was identified for the test UVA579. The signal analysis of the channels of this test shows that there are no significant differences between with the other signals. Consequently, this test was included in the series of tests.

The sled deceleration for this series, with a mean delta-*v* of 47.5 ± 0.1 km/h is shown in Figure D-.

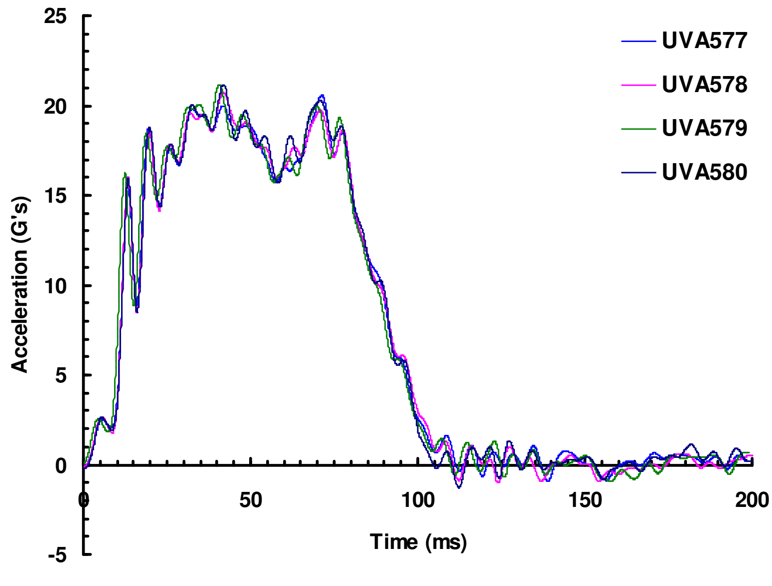


Figure D-6: Sled deceleration for three-point force-limited belt and passenger airbag sled test

As before, the following figures present the biofidelity requirements, in all cases the corridor is given with and without the curves of the tests comprising the series.

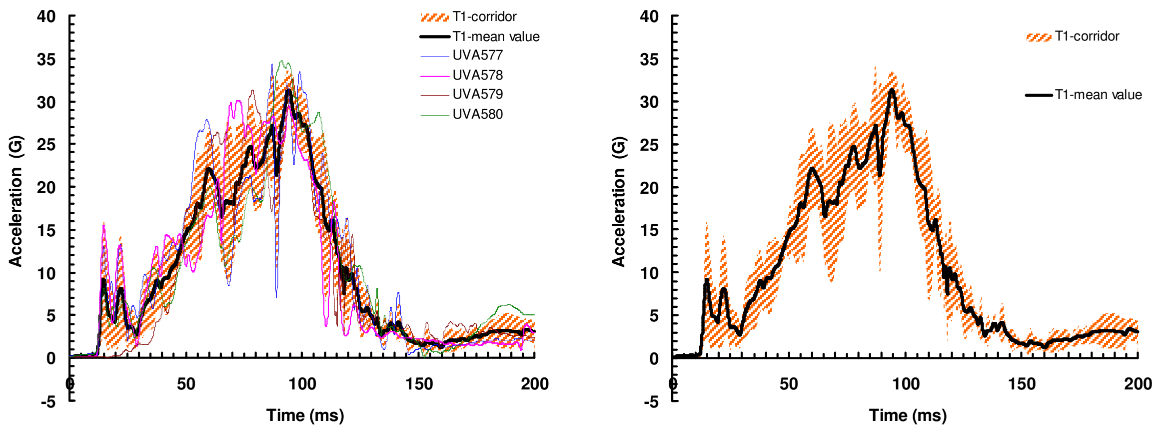


Figure D-7: T1 resultant acceleration for three-point force-limited belt and passenger airbag sled test

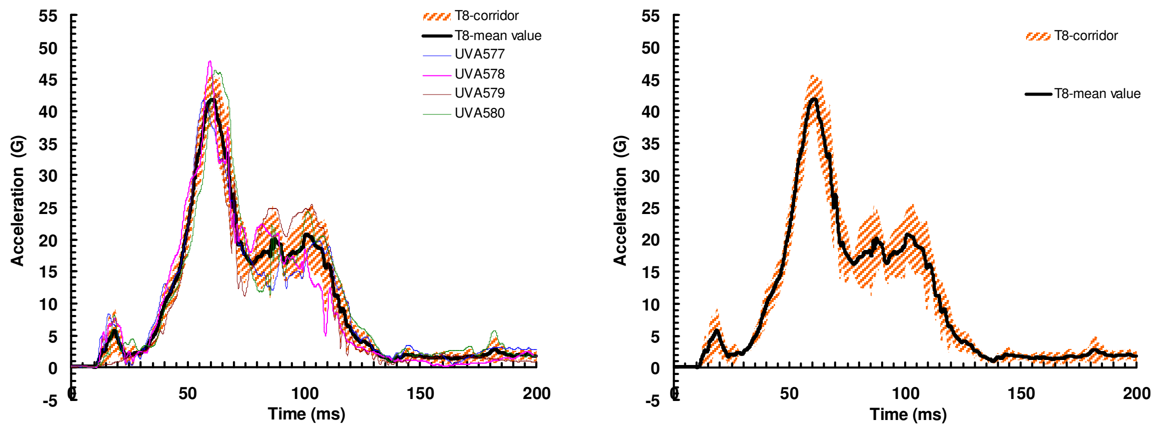


Figure D-8: T8 resultant acceleration for three-point force-limited belt and passenger airbag sled test

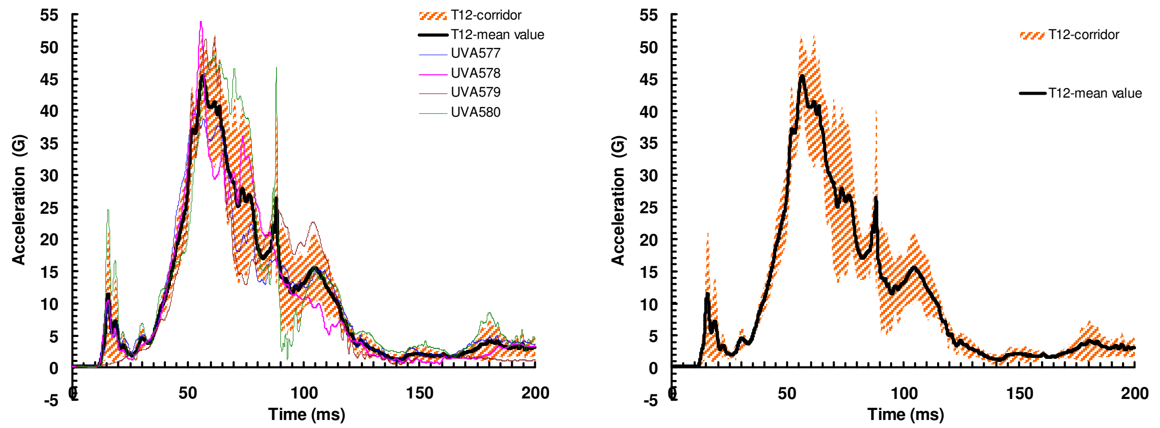


Figure D-9: T12 resultant acceleration for three-point force-limited belt and passenger airbag sled test

Kinematics data plots of the tests are shown below. X and Z displacements versus time as well as the trajectory are represented. The original curves are represented by dashed lines and the normalized data are with solid lines. The average curve (of the normalized data) is represented by a black solid line.

Particularly some drawbacks have been found in this test configuration. In the NHTSA report (obtained from the database) there are no data of the T1 kinematics (the report shows the shoulder kinematics). The shoulder kinematics is shown using the same process and the T1 kinematics are taken from the Forman et al 2006.

The hip kinematics data have also problems. It is labelled as femur instead of hip, verified by comparing the data plots with the results show in Forman et al 2006. This comparison shows that there is coincidence with the 577 and 578 tests but the results shows in NHTSA report of the test 580 are different of the results in Forman et al 2006 and the no comparison of 579 test could be performed due to this test was rejected by Forman in his publication.

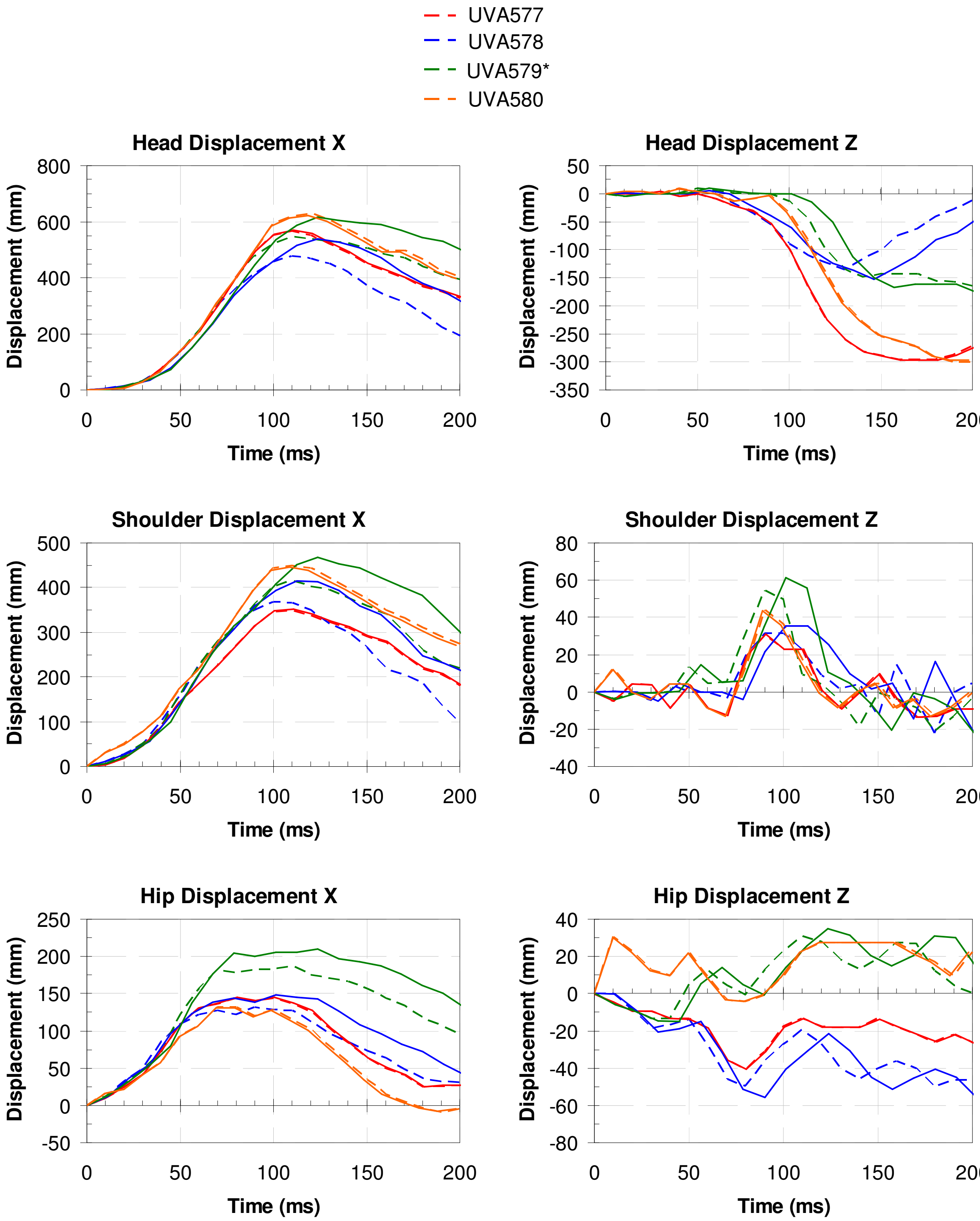


Figure D-10: Head, shoulder and hip trajectories for three-point force-limited belt and passenger airbag

D.2.2 Three-point belt and passenger airbag requirements

These series of sled tests were performed at the University of Virginia with a 1997 Ford Taurus Buck with a reinforced OEM front passenger seat with anti-submarining pan. The restraint system was composed of a 1998 Ford Taurus 3 point-belt with standard (without force limiter) retractor and a standard Ford Taurus 1998 passenger de-powered airbag.

The sled deceleration for this series, with a mean delta-*v* of 48.5 ± 0.34 km/h is shown in Figure D-11.

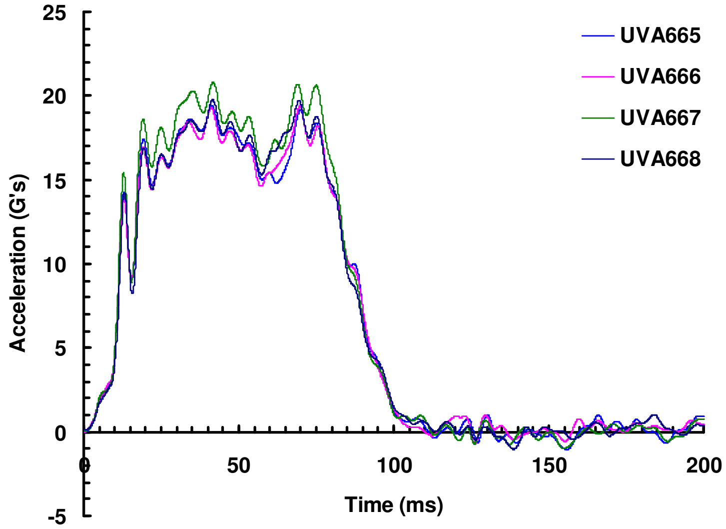


Figure D-11: Sled deceleration for three-point belt and passenger airbag sled test

As in the previous two sections, the following figures present the biofidelity requirements, in all cases the corridor is given with and without the curves of the tests comprising the series.

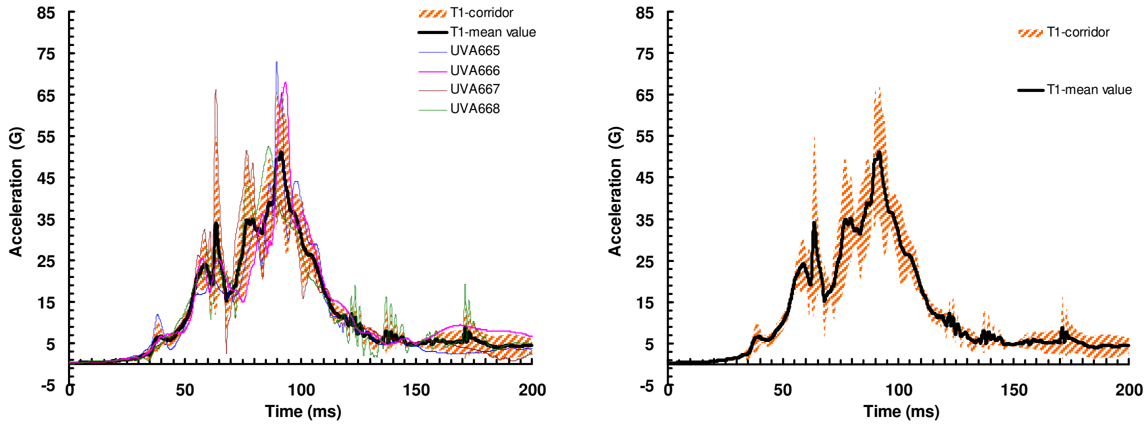


Figure D-12: T1 resultant acceleration for three-point belt and passenger airbag sled test

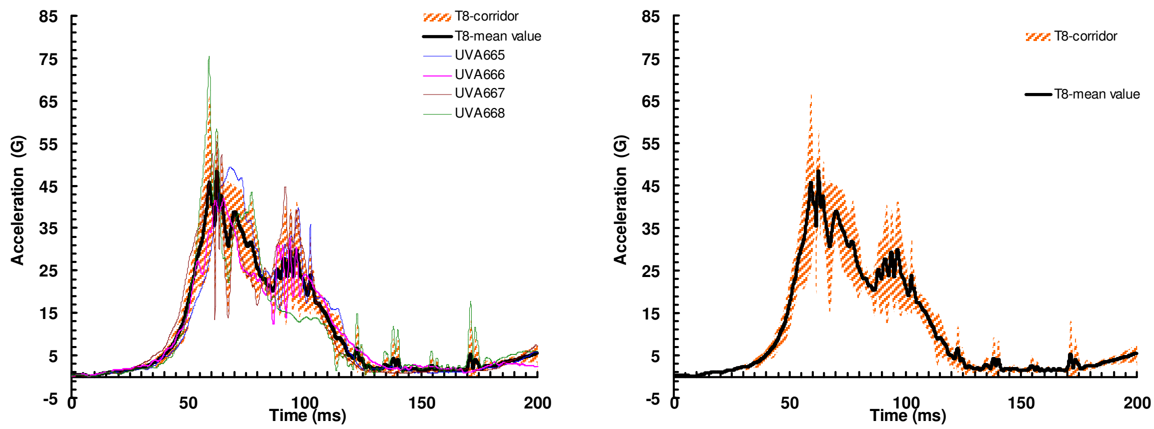


Figure D-13: T8 resultant acceleration for three-point belt and passenger airbag sled test

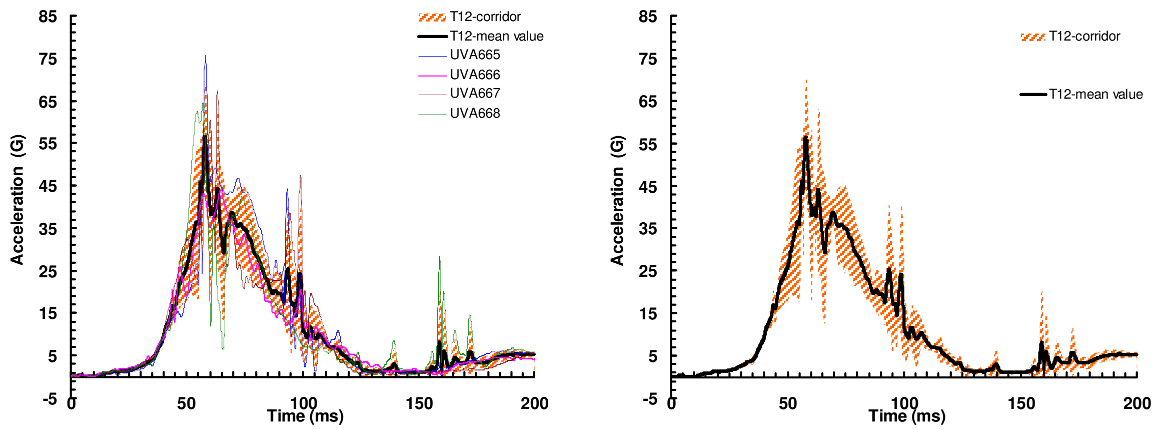


Figure D-14: T12 resultant acceleration for three-point belt and passenger airbag sled test

Kinematics data plots of the tests are shown below. The line typology (solid or dashed) is the same as defined in the previous configuration.

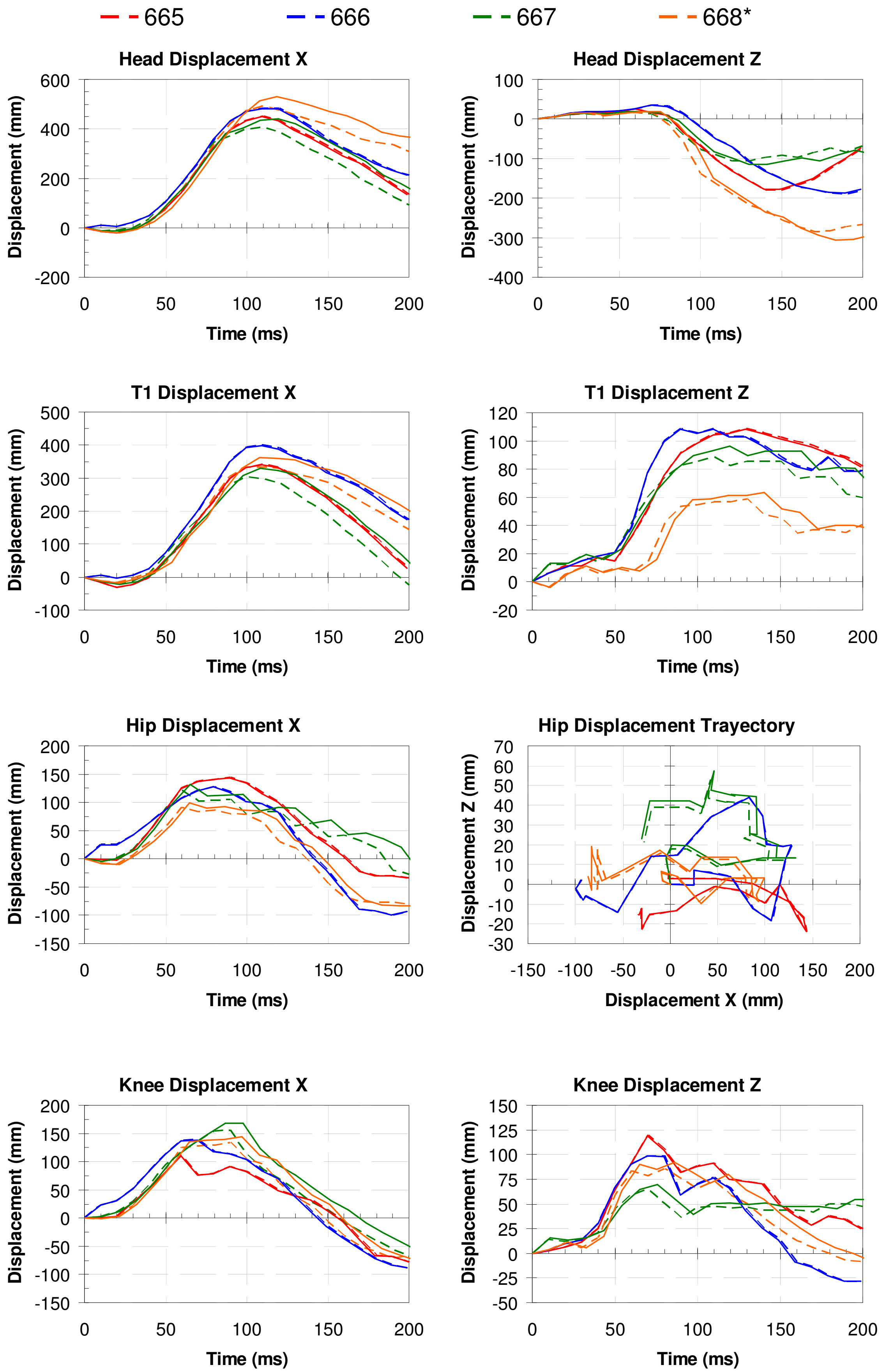


Figure D-15: Head, T1, hip and knee displacements for three-point belt and passenger airbag sled test

D.2.3 Three-point belt low-speed requirements

These series of sled tests were performed at the University of Virginia with the same set-up as that of the previous series, three-point belt, but without the airbag.

The sled deceleration for this series, with a mean delta-*v* of 29.7 ± 0.26 km/h is shown in Figure D-16.

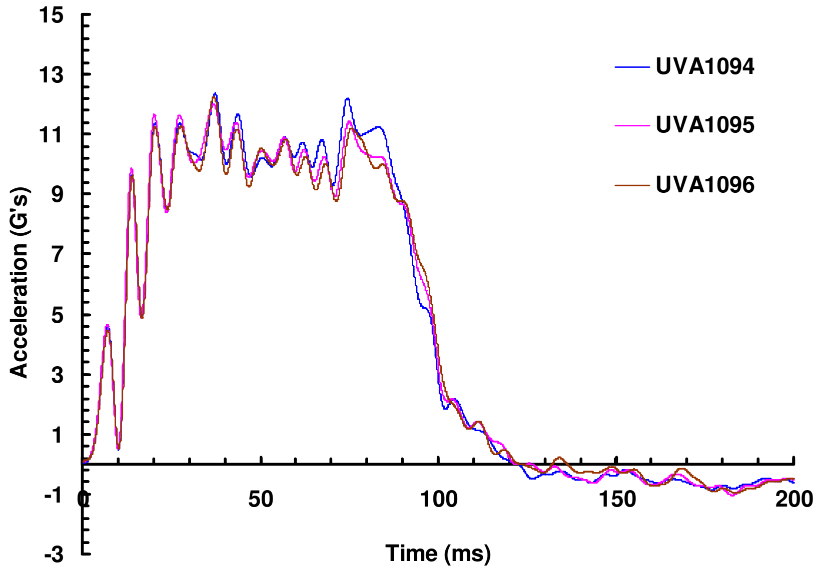


Figure D-16: Sled deceleration for three-point belt and passenger airbag sled test

As in the previous two sections, the following figures present the biofidelity requirements, in all cases the corridor is given with and without the curves of the tests comprising the series.

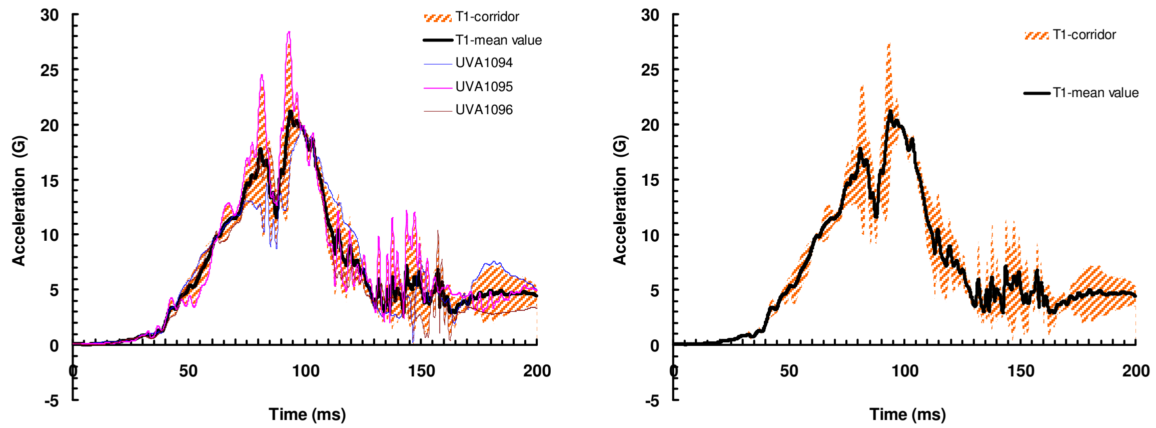


Figure D-17: T1 resultant acceleration for three-point belt low speed sled test

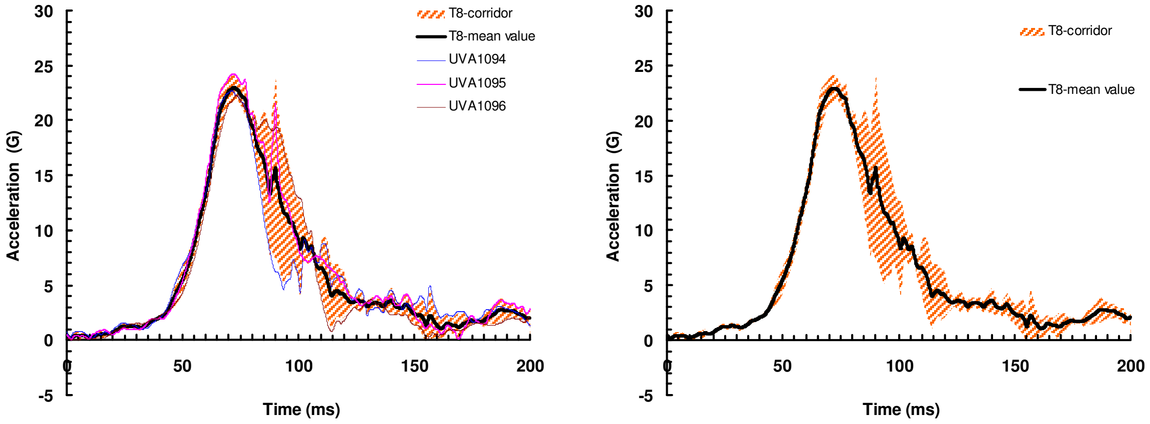


Figure D-18: T8 resultant acceleration for three-point belt low speed sled test

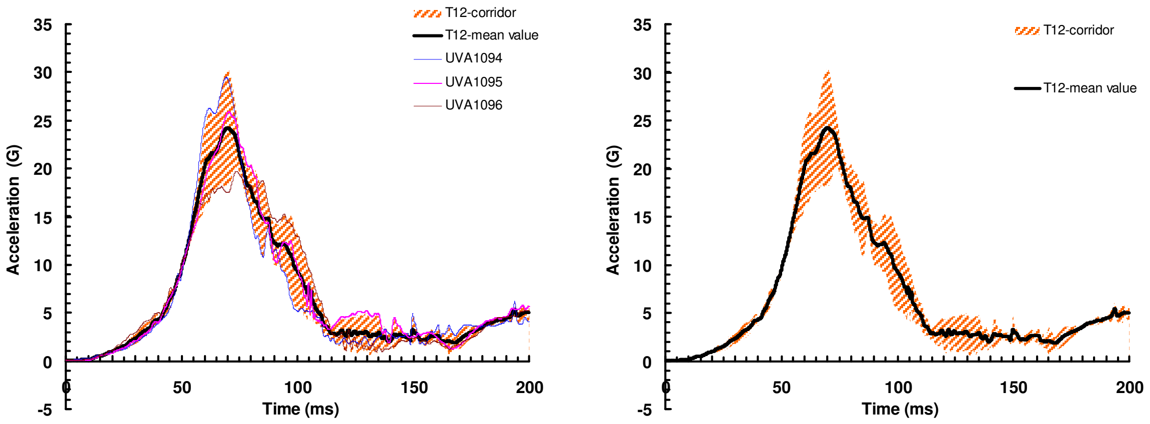


Figure D-19: T12 resultant acceleration for three-point belt low speed sled test

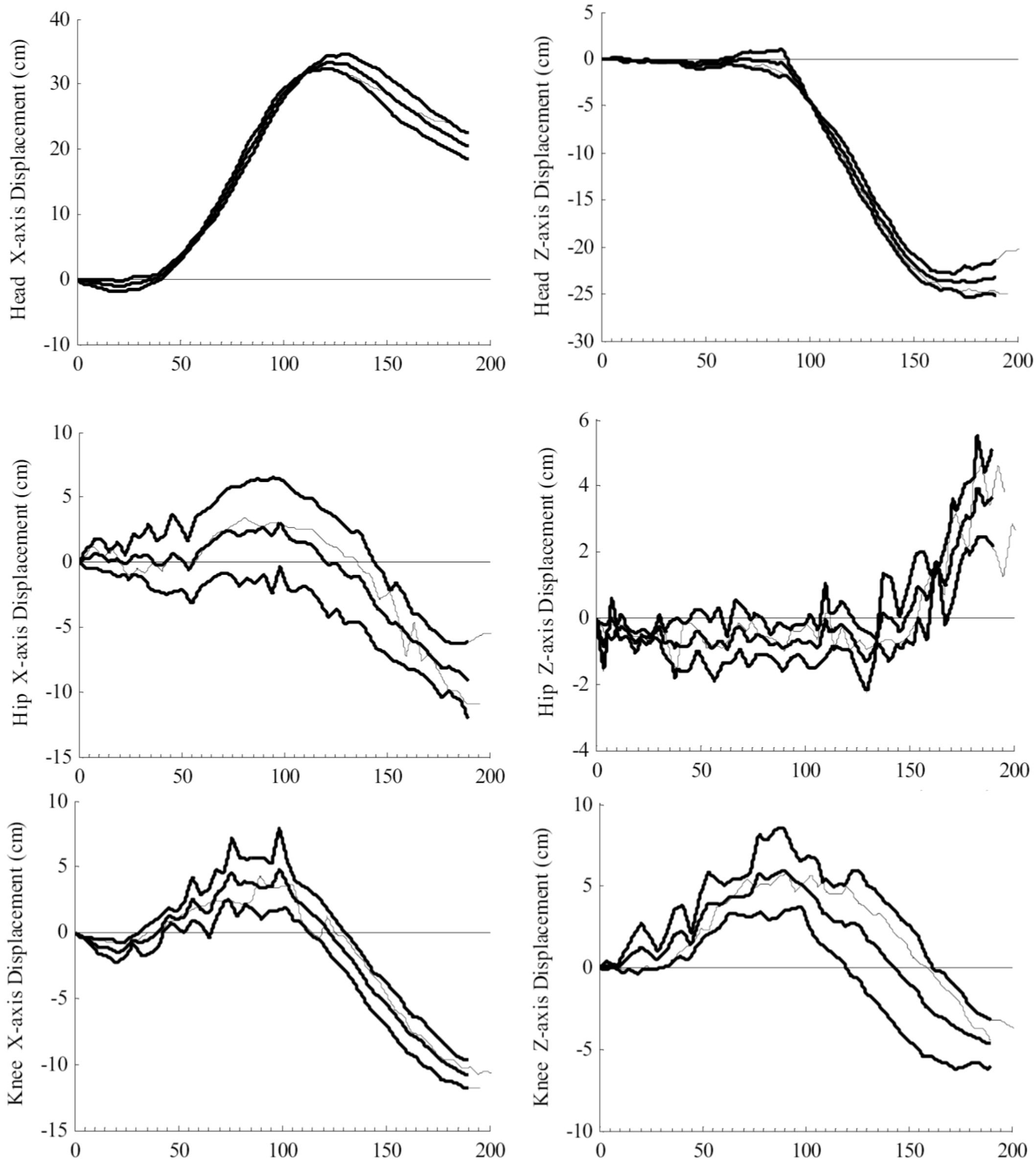


Figure D-20: Head, hip and knee trajectory for three-point belt low speed (adopted Forman et al 2006a)

D.2.4 Two-point belt and passenger airbag requirements (Bolton et al. 2006))

These series of sled tests were performed at the University of Virginia with a 1997 Ford Taurus Buck with a reinforced OEM front passenger seat with anti-submarining pan. The PMHS were restrained with an aftermarket 2 point lap belt. A standard Ford Taurus 1997 passenger full-powered airbag was deployed.

The sled deceleration for this series, with a delta-v of 48.6 – 49.7 km/h is shown in Figure D-21.

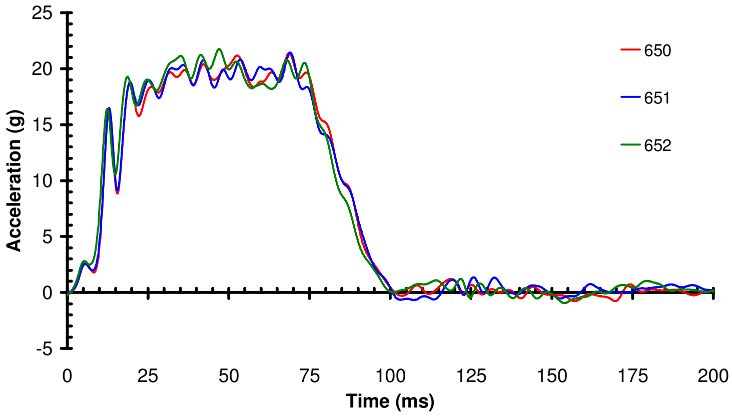


Figure D-21: Sled deceleration for two-point lap belt and passenger airbag sled test

As in the previous three sections, the following figures present the biofidelity requirements, in all cases the corridor is given with and without the curves of the tests comprising the series.

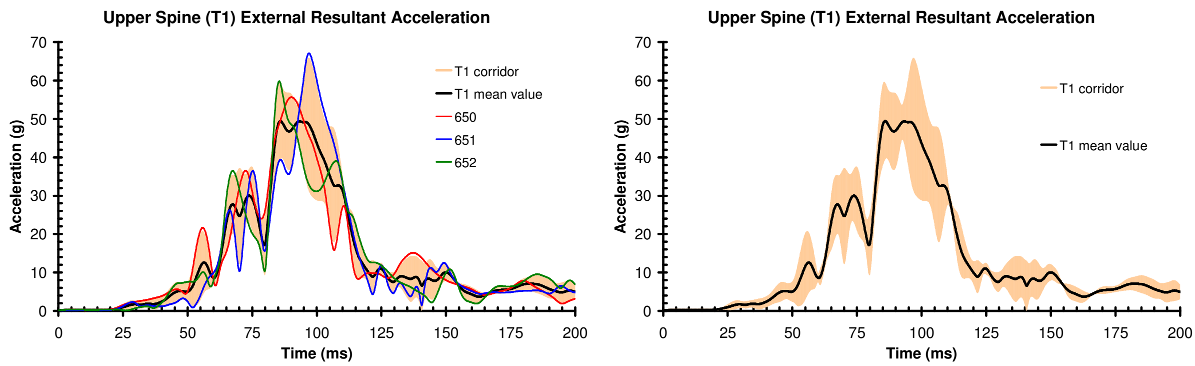


Figure D-22: T1 resultant acceleration for two-point lap belt and passenger airbag sled test

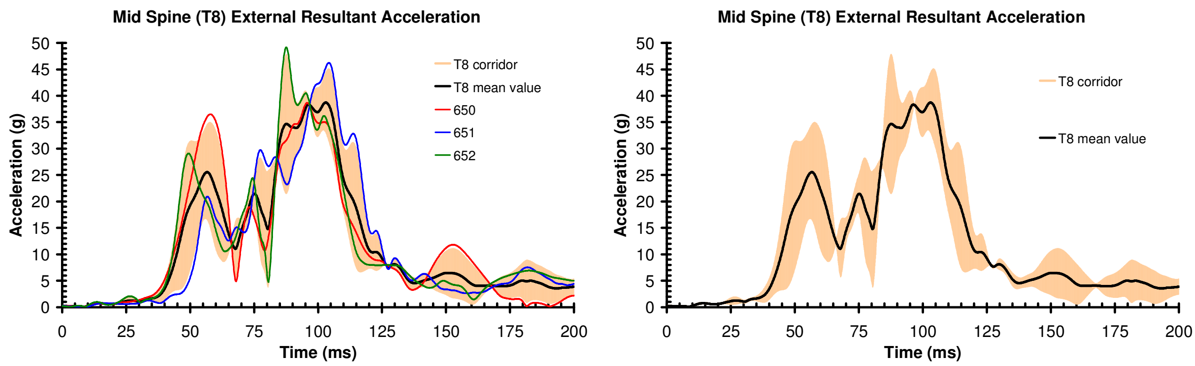


Figure D-23: T8 resultant acceleration for two-point lap belt and passenger airbag sled test

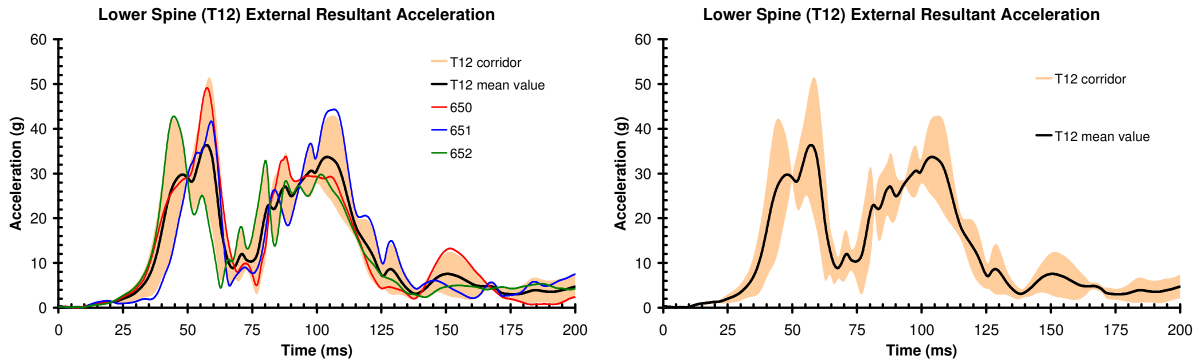


Figure D-24: T12 resultant acceleration for two-point lap belt and passenger airbag sled test

Kinematics data plots of the tests are shown below. The line typology (solid) is the same as defined in the previous configuration.

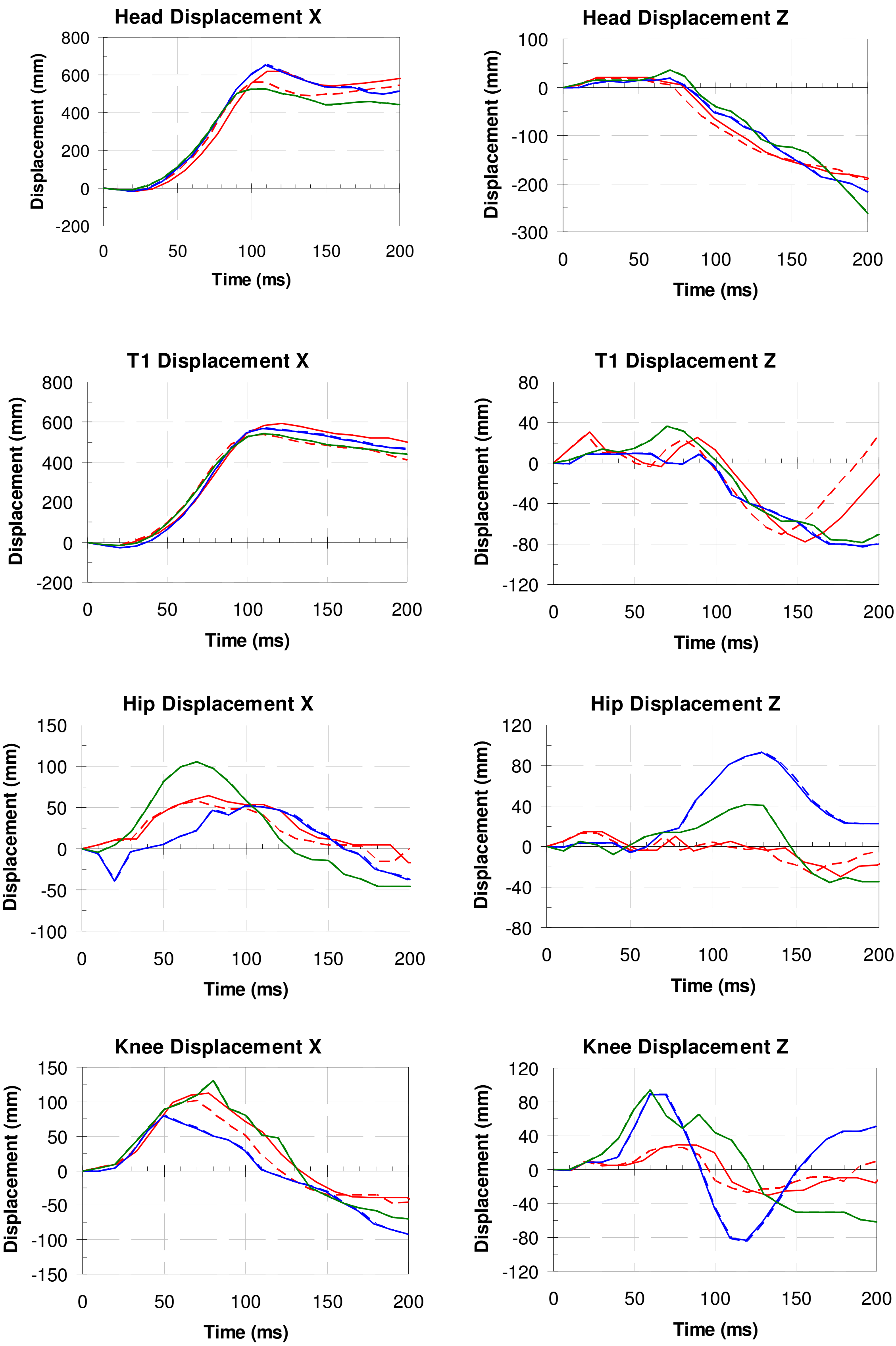


Figure D-25: Head, T1, hip and knee trajectories for two point lap belt and passenger full powered airbag

Appendix E **Törnvall *et al.* (2008)**

Recommended Biofidelity Requirement:
Absolute shoulder kinematics
Sled tests

E.1 Summary

The aims of the report is to present post mortem human subject (PMHS) kinematics, especially shoulder kinematics in 45° far-side collisions (away from the shoulder belt anchor) and in 30° near-side collisions (towards the shoulder belt anchor) and full-frontal collisions for evaluation of the dummy shoulder response.

The report includes data from ten tests with three PMHSs that were conducted for three collision angles: 45° far-side, 30° near-side and 0° full-frontal. The PMHS were fitted three-dimensional film targets and accelerometers to the bones. The subjects were seated in a well-defined seat system, without steering wheel or air bag, and were restrained with a three-point lap-shoulder belt without force limitation or pretension system, before they were accelerated. The average peak acceleration was 13.3 g and the average Δv was 26.5 km/h. The kinematics of the test subjects was captured using four video cameras.

The results of this study include accelerometer data, belt forces and film target data. The film target data is provided in a global coordinate system, in sled coordinate system and relative the first thoracic vertebra (T1).

E.2 METHODS

E.2.1 Experimental Design

Each subject was tested at three angles: first 0° full-frontal, second 45° far-side, and third 30° near-side, in which the subjects moved towards a simplified car door interior (Table E-1 and Figure E-1). However, PMHS 3 was tested twice in the 0° full-frontal configuration, before the 45° far-side and the 30° near-side test configurations. The 0° full-frontal test was repeated due to an instrumentation failure.

Table E-1 Oblique frontal collision test conditions

Type of test	PMHS	Test number	Δv [km/h, CFC 60]	Maximum peak acceleration [g, CFC 60]
0° full-frontal	PMHS 1	16	27.0	13.3
	PMHS 2	19	27.1	13.6
	PMHS 3	22	24.9	12.6
	PMHS 3	23	25.6	12.5
Averages for the 0° full-frontal tests			**26.2**	**13.0**
45° far-side	PMHS 1	17	27.0	13.7
	PMHS 2	20	27.5	14.4
	PMHS 3	24	25.5	13.2
Averages for the 45° far-side tests			**26.7**	**13.8**
30° near-side	PMHS 1	18	27.1	13.5
	PMHS 2	21	27.0	13.6
	PMHS 3	25	25.7	12.9
Averages for the 30° near-side tests			**26.6**	**13.3**

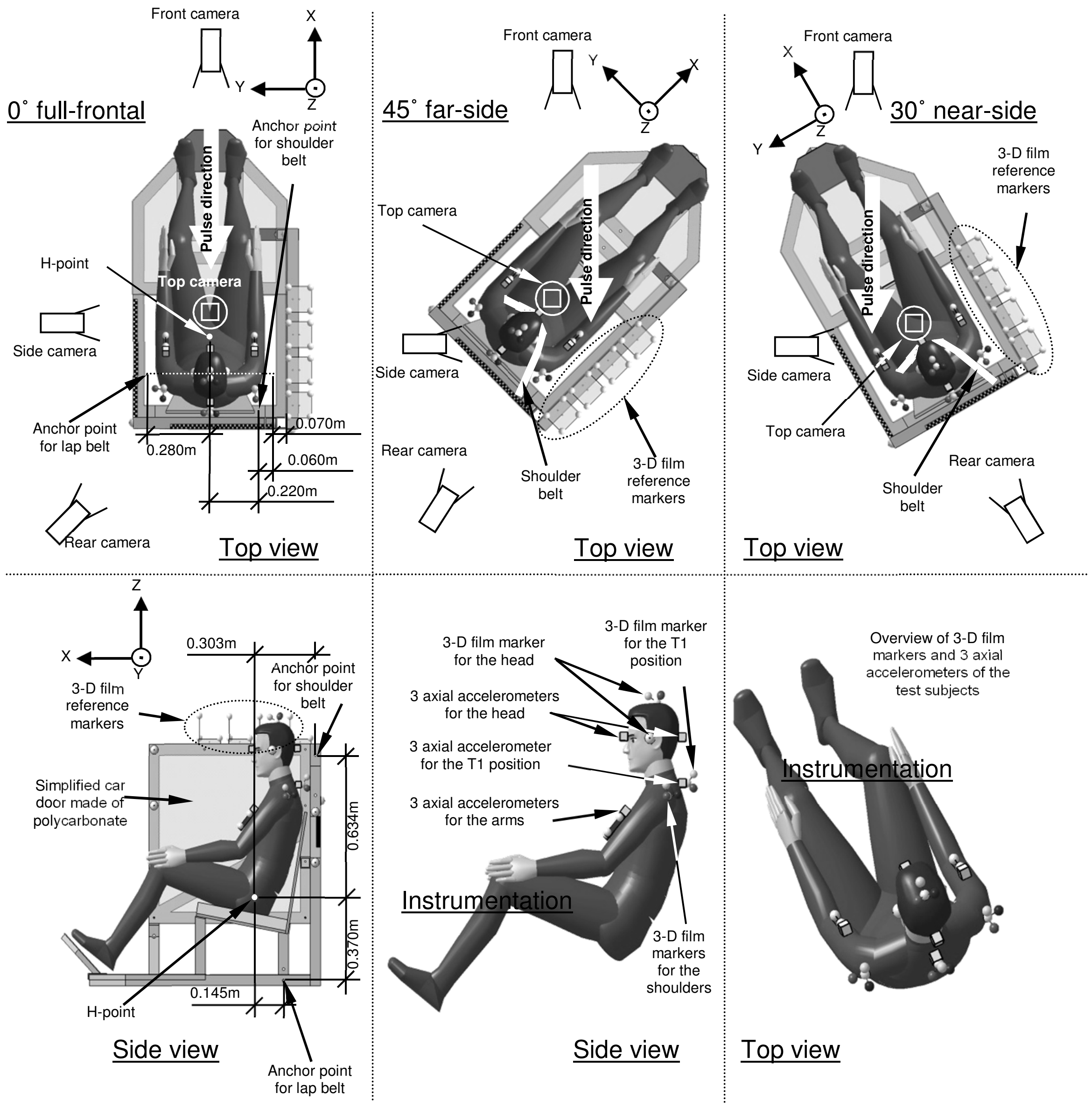


Figure E-1 Test set-up of 0° full-frontal, 45° far-side and 30° near-side tests, and instrumentation of the test subjects (adopted from Törnvall et al., 2008)

E.2.2 Test Subjects

The test subjects were dressed in cotton clothing and positioned in a laboratory seat. Thereafter the PMHS were positioned in a laboratory seat, with the H-point corresponding to the R-16 standard position (United Nations, 1958b) of the H-point, and their backs resting against the back rest. Their heads were held up by a cable, connected to an electro magnet that was released 0–3 ms after the sled system started to move. Their hands were placed in their laps. It was noticed during the analysis of PMHS 1 and 2 that the upper torso was difficult to position in a repeatable manner. Consequently, the upper torso of PMHS 3 was stabilized prior to the test by means of a cable, around the upper torso, which was connected to an electro magnet that was released 0–3 ms after the sled system started to move. The shoulders were then pushed back as much as possible. A few minutes prior to testing, their pulmonary systems were inflated (pressurized) once as a pre-conditioning procedure and left open to ambient atmospheric pressure a few seconds before test.

A new shoulder belt was positioned on the right shoulder for every test and each subject. Photos of the belt position relative the shoulder were taken and compared with the starting position of the belt in the subsequent tests, to facilitate repeatability throughout the series. Then the test subjects were accelerated. After each completed test, the clavicles of the PMHSs were examined, by palpation, for fractures or dislocations.

E.2.3 Instrumentation

All test subjects were equipped with 3-D markers which were screwed to the bone: at the top of the skull, at the posterior tip of the first thoracic vertebra (T1), at the posterior part of the acromion on the right and left shoulders, and at the middle of the right and left humeri, as shown in Figure E-1. Each of the test subjects was also equipped with triple-axial accelerometers which were screwed to the bone: at the anterior and posterior of the skull in line with the head centre of gravity (CG), at T1, and at the front in the centre of the right and left humeri (Figure E-2). 0 shows the initial position of the x, y and z axes of all of the triple-axial accelerometers used in the test set-up. Each was made up of three ±50 g, ADXL150 surface micro-machined, accelerometers which were sealed in an aluminum box.

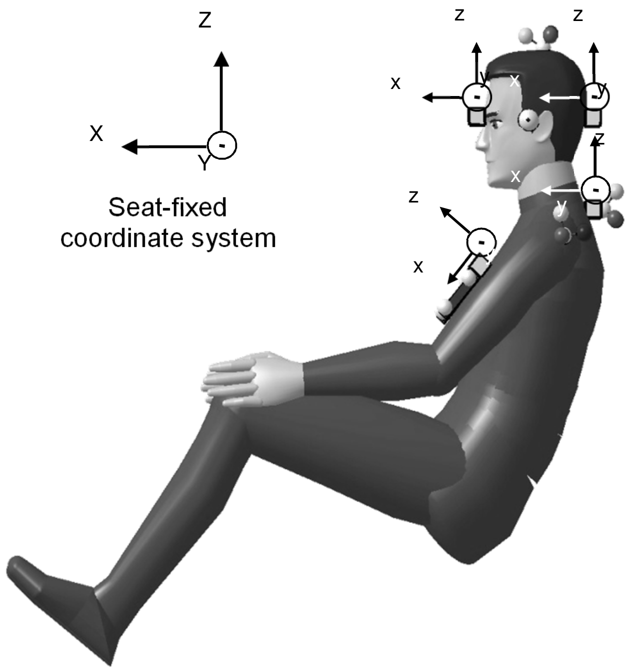


Figure E-2 Initial coordinate system for each accelerometer fastened on the test subjects (x, y, z) to correspond with the seat-fixed coordinate system (X, Y, Z), (adopted from Törnvall et al., 2008)

E.2.4 Sled and seat system

The sled was the pneumatic acceleration (high-G) type. Figure E-14, 1–3 shows the acceleration pulses, filtered according to SAE J211 (SAE, 1995), using Channel Frequency Class (CFC) 60. Table E-1shows the Δv for each test and the maximum peak acceleration. Average peak acceleration was 13.3 g and the average Δv was 26.5 km/h.

The tests were conducted using an R-16 seat (United Nations, 1958b) with 50 mm deformable foam (on top of the seat surface) which was made of polyethylene 220-E, 35 kg/m^3 and had a stiffness of 40 kPa at 10% compression, 55 kPa at 25% compression and 110 kPa at 50% compression. The R-16 seat and 45° angled foot plate were produced and positioned according to the R-16 standard. No steering wheel or air bag was used. A fixed three-point belt with a standard buckle and conveyer belt without force limitation were used. The belt was made of a high-stretch band with 16% stretch at 11.3 kN (produced by Autoliv, production number 570 4196 00H). Attachment points for the belt were within the R-14 regulation corridors (United Nations, 1958a) (Figure E-1Figure E-1).

The simplified car door used in the 30° near-side tests was made of a 10 mm thick polycarbonate board to allow high-speed video recording through the door during testing. The polycarbonate board was positioned vertically, parallel to the median plane of the seat with an offset of 0.35 m (Figure E-1).

E.2.5 Data Acquisition and Analysis

Each PMHS test was recorded by four MotionXtra HG-LE high-speed digital video cameras (1000 fps, resolution 752*1128), with a front view, side view, rear view and top view, as shown in Figure E-1. The top and side view cameras were equipped with fixed 14 mm focal length lenses, while the front and rear cameras had fixed 20 mm focal length lenses. The 3-D marker displacements (kinematics) were retrieved from the high speed videos and analyzed with TEMA Automotive™ Software. The kinematics from the head 3-D marker was recalculated to represent the kinematics of the head CG. All kinematic data was filtered using CFC 20.

The x accelerations for the front and the rear of the head were filtered and added together, as were the y and z accelerations. These were then divided by two to get the filtered acceleration for the head CG in the x, y and z directions. All accelerations given in this study are the resultants of the x, y and z accelerations for each of the accelerometers. These resultants were calculated by squaring each filtered x, y and z acceleration for each triple-axial accelerometer and then adding the x, y and z values to extract the root of the value. All acceleration signals were filtered using CFC 60. The lap belt force and the shoulder belt force were also measured for each test and filtered using CFC 60.

E.2.6 Coordinate System

All displacement data were produced according to the seat-fixed coordinate system (Figure E-1). Each displacement trajectory was taken as zero at impact start. The X axis is referred to as the anterior-posterior direction, the Y axis as the lateral-medial direction, and the Z axis as the superior-inferior direction of the test subjects' displacement in this study. All accelerations in this study were measured in a local coordinate system (x, y, z), where the axis rotated with the accelerometer attachments, and were taken as zero at impact start. The initial position of the accelerometers with corresponding initial coordinate systems is shown, see Figure E-2.

E.3 Results

The data analysis focuses on an overview of the human response, in particular the shoulder motions of the three PMHSs in the three test configurations. Detailed kinematics from the 3-D motion analysis and acceleration data can be found in Figure E-7 (X-Y trajectories), Figure E-8, Figure E-9 and Figure E-10 (displacements in relation to the sled),

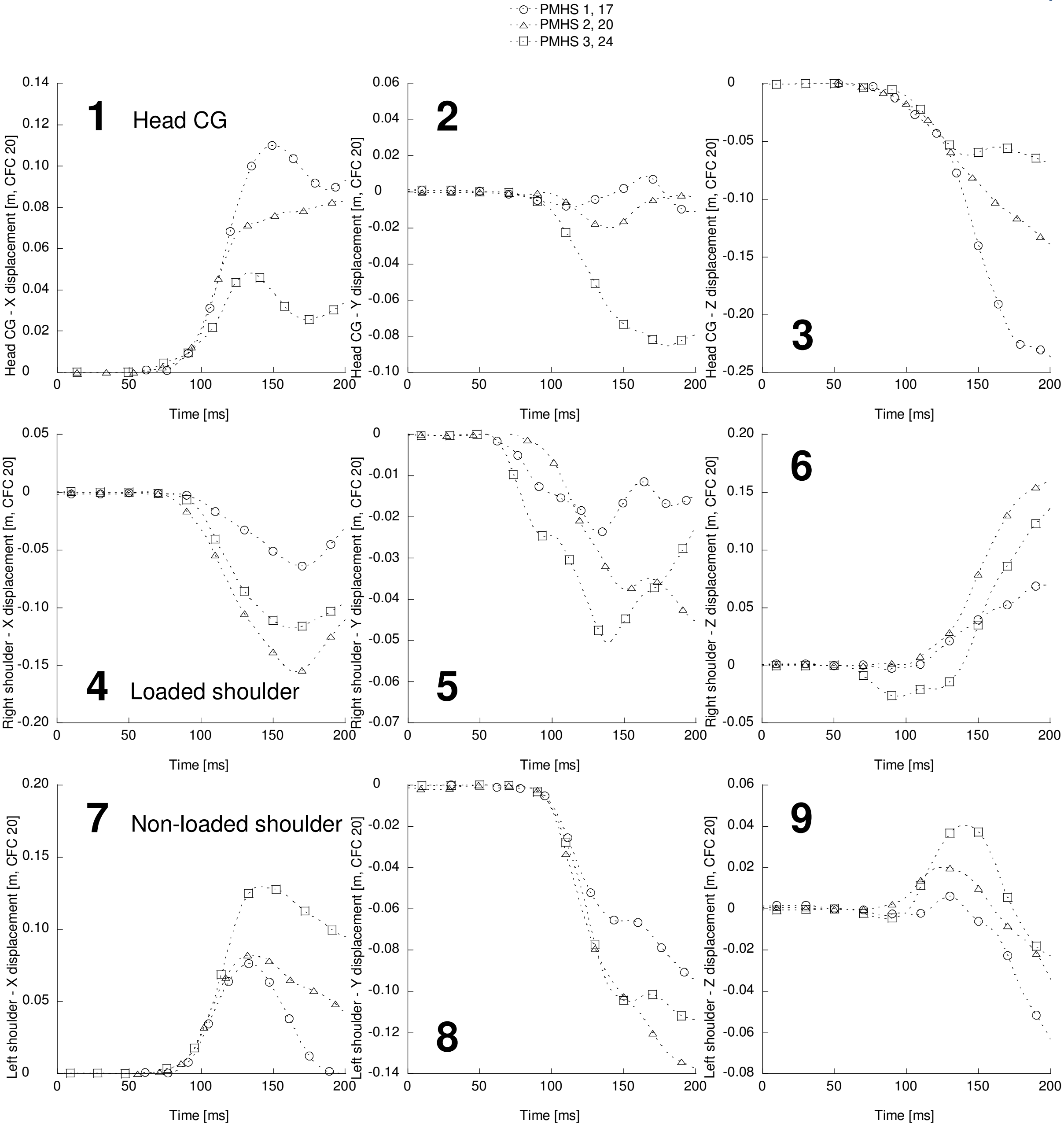


Figure E-11,

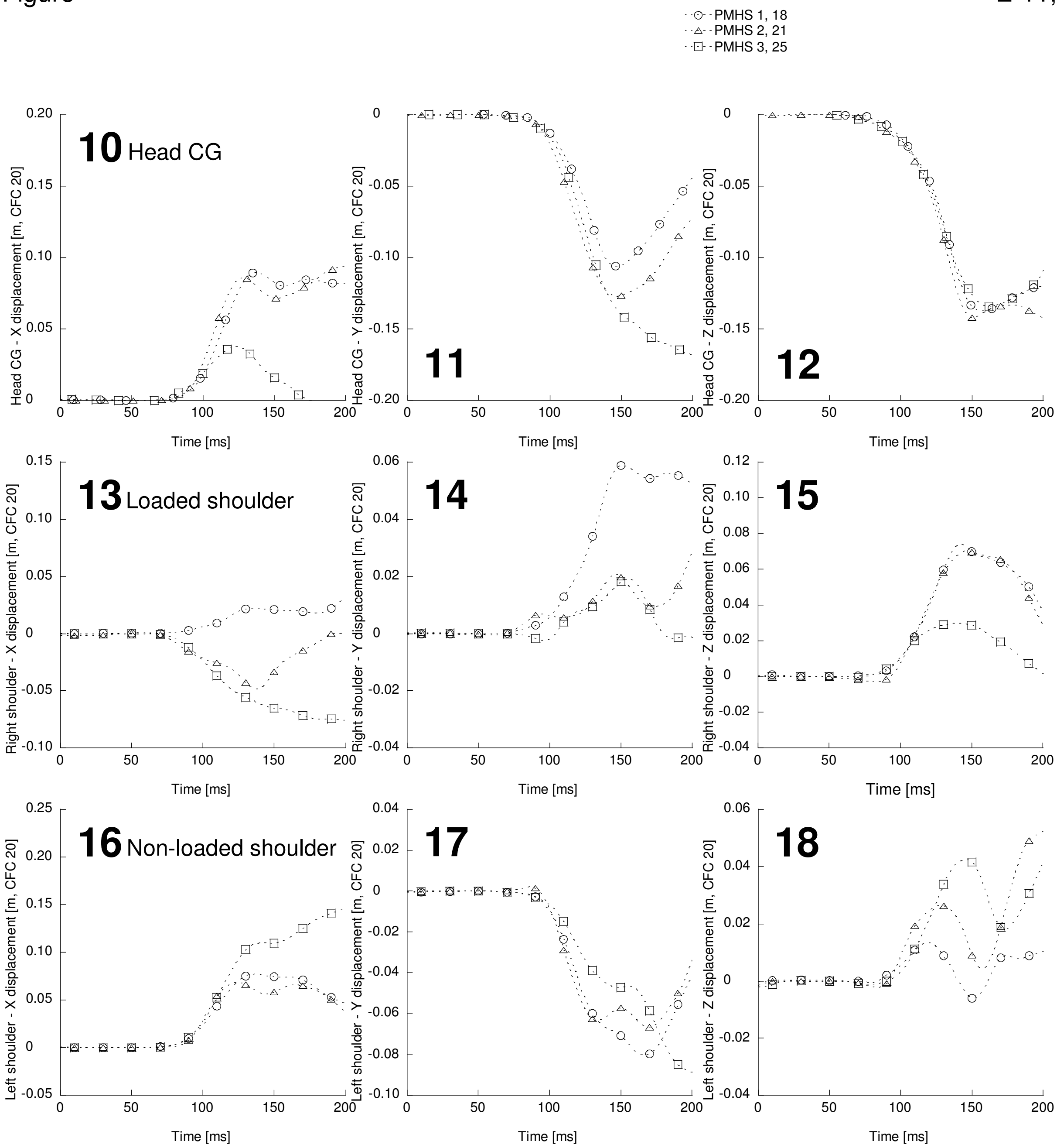


Figure E-12 and

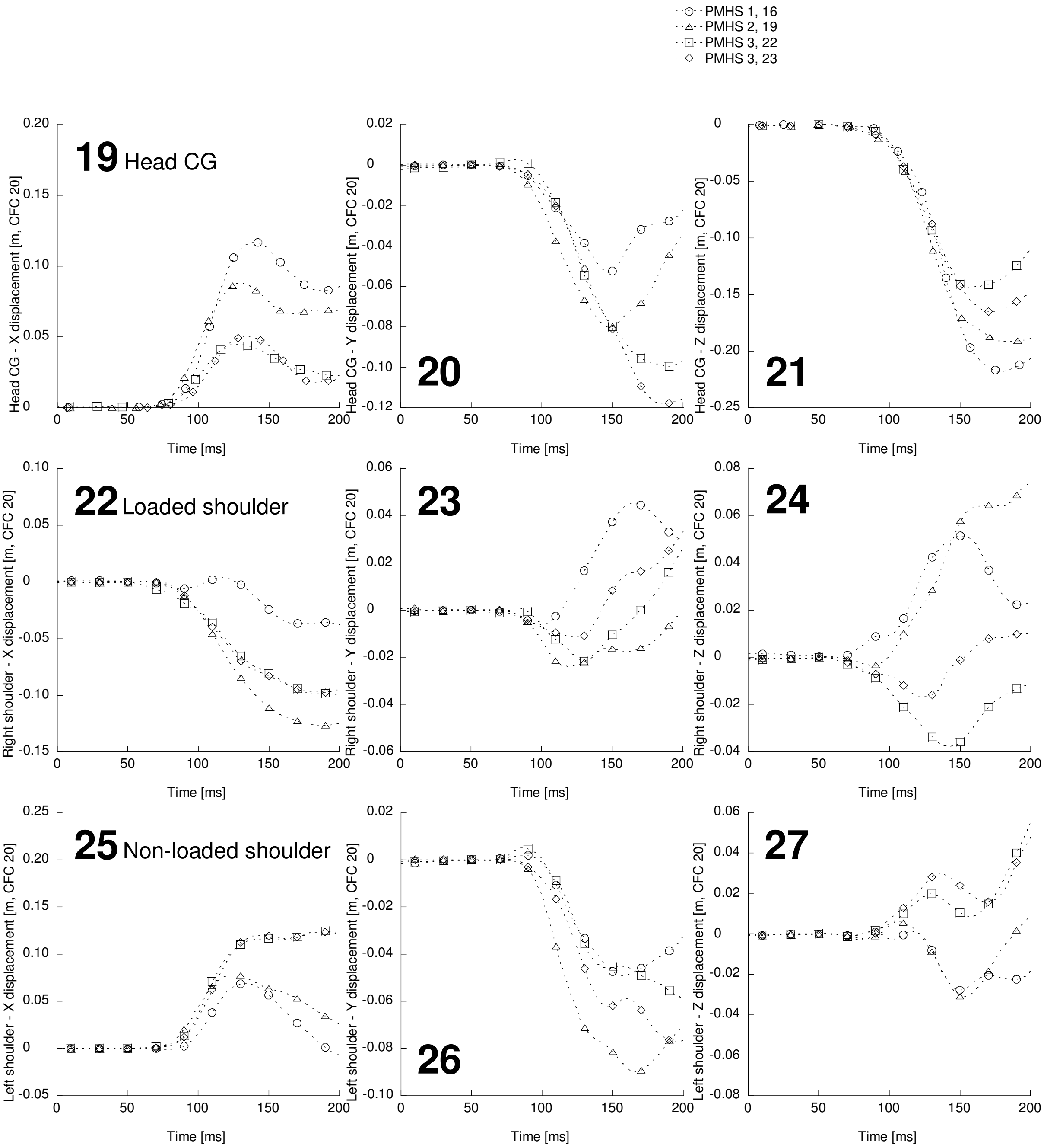


Figure E-13 (displacements in relation to T1), Figure E-14(sled accelerations and belt forces) and Figure E-15 (body accelerations) of this paper.

E.3.1 45° Far-Side Sled Tests

The high-speed video data show that the test subjects' T1 markers moved forward and laterally, in the reverse direction to that of the sled acceleration (Figure E-3 and Figure E-7). During the forward and lateral motion of T1, the shoulder belt slipped along the clavicle from the sternal to the acromial end for all test subjects (Figure E-4). The test subjects' heads rotated and translated forward in the reverse direction to that of the sled acceleration (Figure E-7, 1–2).

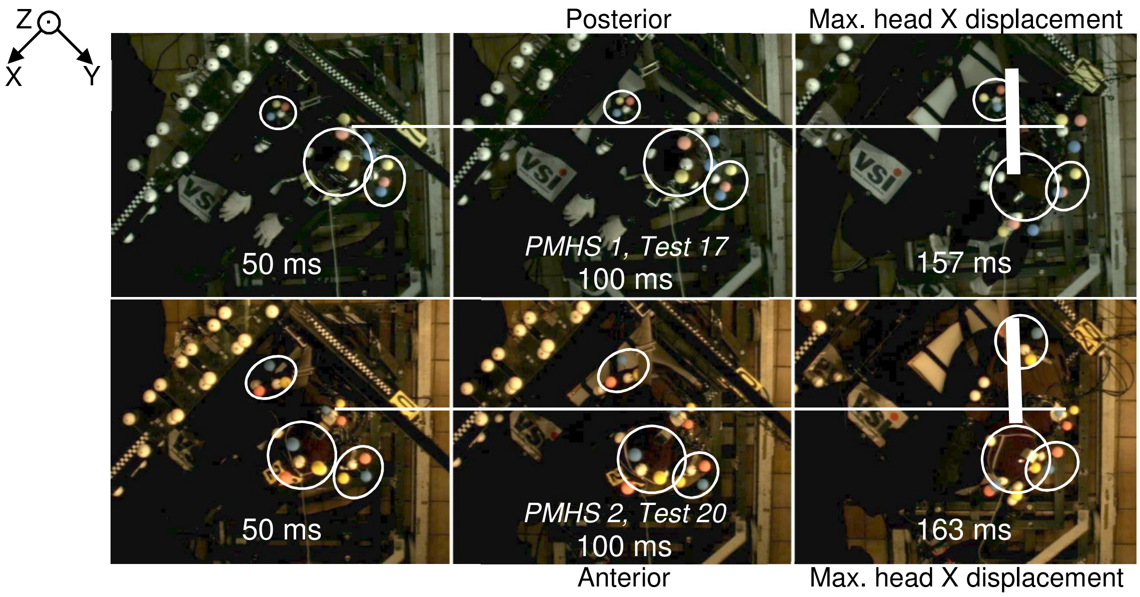


Figure E-3 Top view of PMHSs 1 and 2 in the 45° far-side test set-up at 50 ms, 100 ms and at maximum anterior head CG displacement (X direction) in relation to the sled. The horizontal line in each row of photos is set at T1 level. The middle circle in each photo is the head position. The smaller white circles and ellipses represent the shoulder markers. The wide white bars (almost vertical) at maximum anterior head CG displacement (X direction) highlight the path of the shoulder belt. (Test 24 with PMHS 3 is not represented in the figure due to top camera failure), (adopted from Törnvall et al. 2008)

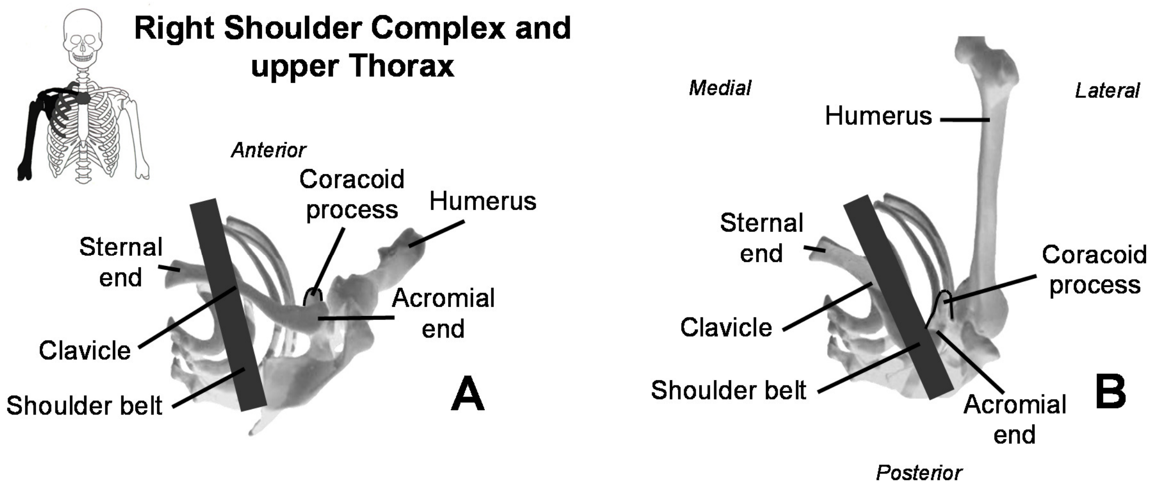


Figure E-4 A: Initial position of the clavicle when the arm is hanging down at the side of the body. B: Position of the clavicle when the arm is flexed (adopted from Törnvall et al. 2008)

The test subjects reached their maximum anterior head CG displacement at 157 ms (PMHS 1), 163 ms (PMHS 2) and 167 ms (PMHS 3) (Table B-2 and Figure E-8, 1). The loaded shoulder displacement relative to T1 (the right shoulder in contact with the

shoulder belt) for all test subjects included both a posterior and a superior motion (

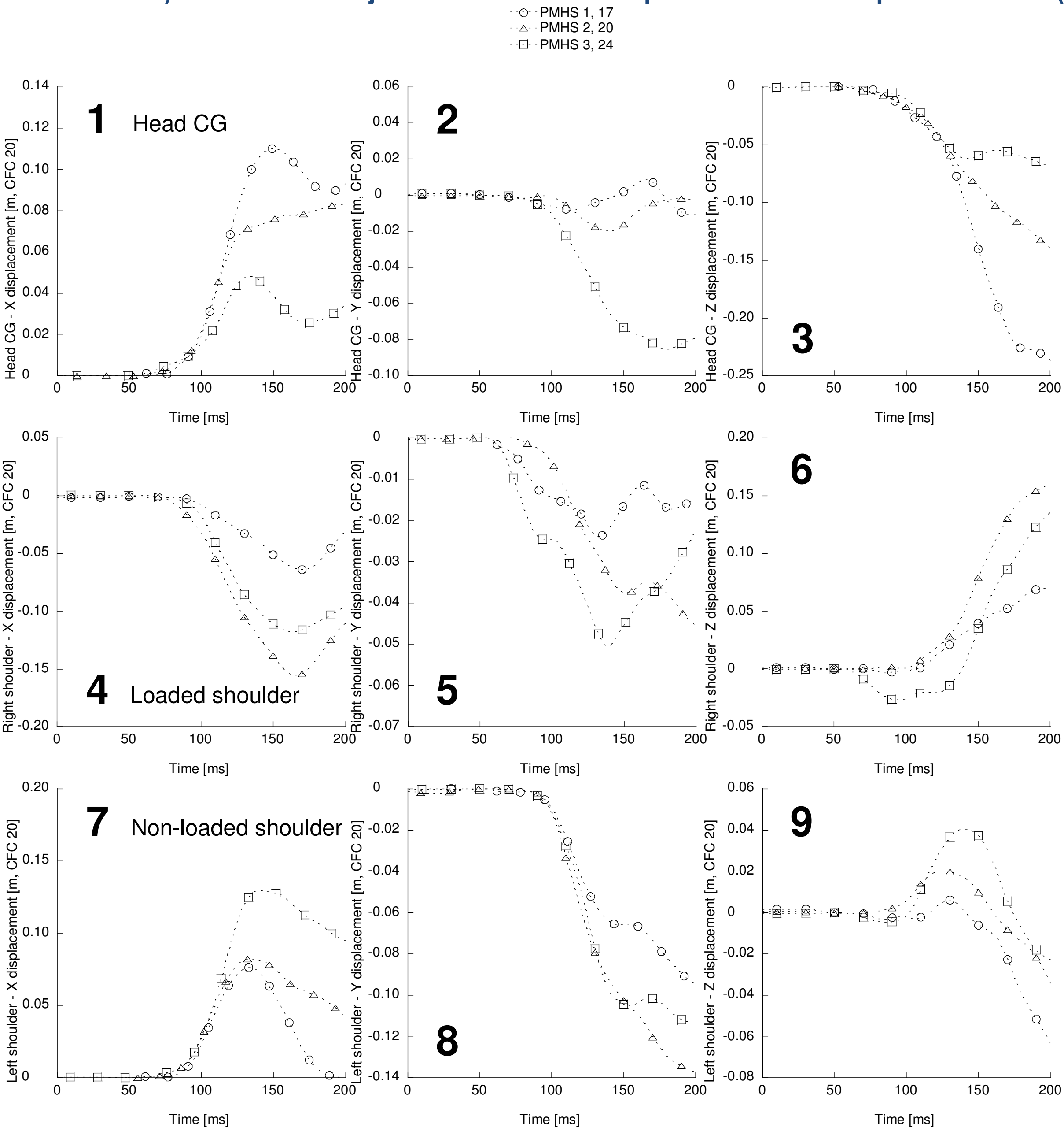


Figure E-11, 4 and 6). However, the non-loaded shoulder displacement relative to T1 (the left shoulder not in contact with the shoulder belt) for all test subjects is an anterior

motion with an initially superior motion (

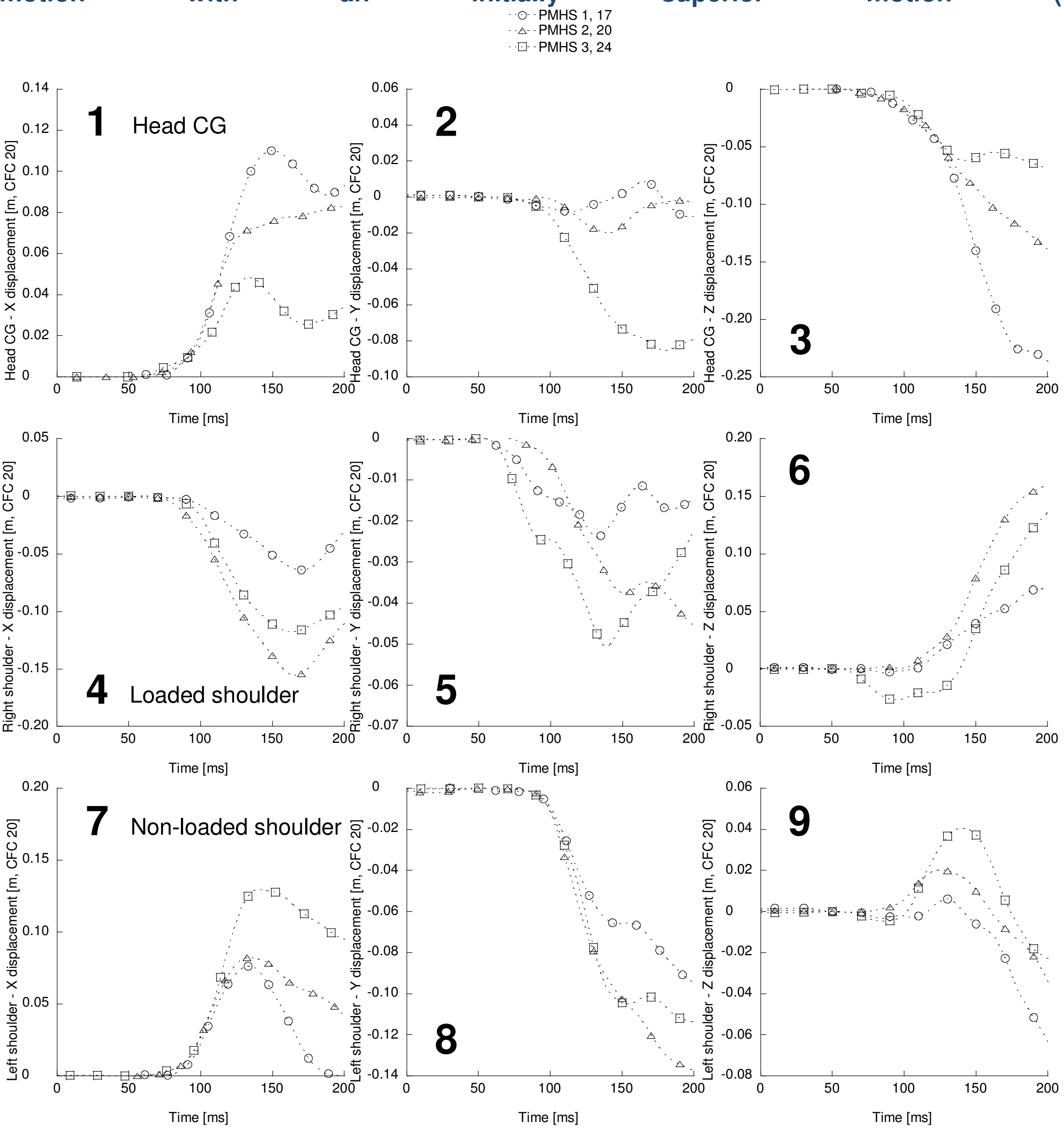


Figure E-11, 7 and 9).

Table E-2 Maximum anterior head displacement, lap belt forces and shoulder belt forces

Type of test	Dummy type or PMHS	Test number	Time for maximum anterior head displacement [ms]	Maximum anterior head displacement [cm]	Lap belt force [kN]	Shoulder belt force [kN]
45° far-side	PMHS 1	17	157	25.5	3.1	4.4
	PMHS 2	20	163	20.9	3.1	4.5
	PMHS 3	24	167	20.7	3.6	5.9
30° near-side	PMHS 1	18	137	22.7	2.2	3.5
	PMHS 2	21	132	18.9	2.1	3.6
	PMHS 3	25	129	14.6	2.3	4.7
	PMHS 1	16	152	28.0	2.2	3.7

0° full-frontal	PMHS 2	19	145	24.3	2.3	4.0
	PMHS 3	22	147	22.4	2.6	5.4
	PMHS 3	23	147	22.6	2.7	5.5

None of the test subjects slipped out of the shoulder belt before maximum anterior head CG displacement occurred (Figure E-3 and Table E-3). The point defined as the time when the belt slipped off the shoulder, was when the belt had slipped over an approximate line drawn from the coracoid process and passing through the humerus joint. This was determined by visual inspection of the high speed videos. A complement to identifying the escape time was to look for sudden changes in the shoulder belt force. The time at which the test subjects slipped out of the seat belt was divided into two categories: “on-loading slip” was defined as the event when the subject slipped out before the maximum anterior head CG displacement, in relation to the sled, had occurred; “off-loading slip” took place when the subject slipped out after the maximum anterior head CG displacement, in relation to the sled, i.e. during the rebound phase.

Table E-3 Belt slippage during the 45° far-side tests

Belt geometry	Test subject	Belt slippage	On-loading slip or Off-loading slip
45° far-side	PMHS 1, Test 17	–	–
	PMHS 2, Test 20	–	–
	PMHS 3, Test 24	Belt slippage	Off-loading slip after approximately 170 ms

E.3.2 30° Near-Side Sled Tests with Door Interaction

During the collisions, the test subjects’ T1 and head CG markers moved forward, in the reverse direction to that of the sled acceleration (Figure E-50 and Figure E-7). During this forward motion the shoulder belt slipped along the clavicle towards the neck, from the acromial to the sternal end, for all the test subjects. The shoulder belt compressed the soft tissue in the inferior region of the neck. The test subjects’ heads rotated forward, in the reverse direction to that of the sled acceleration, when the test subjects started to load the shoulder belt (Figure E-7, 1–2).

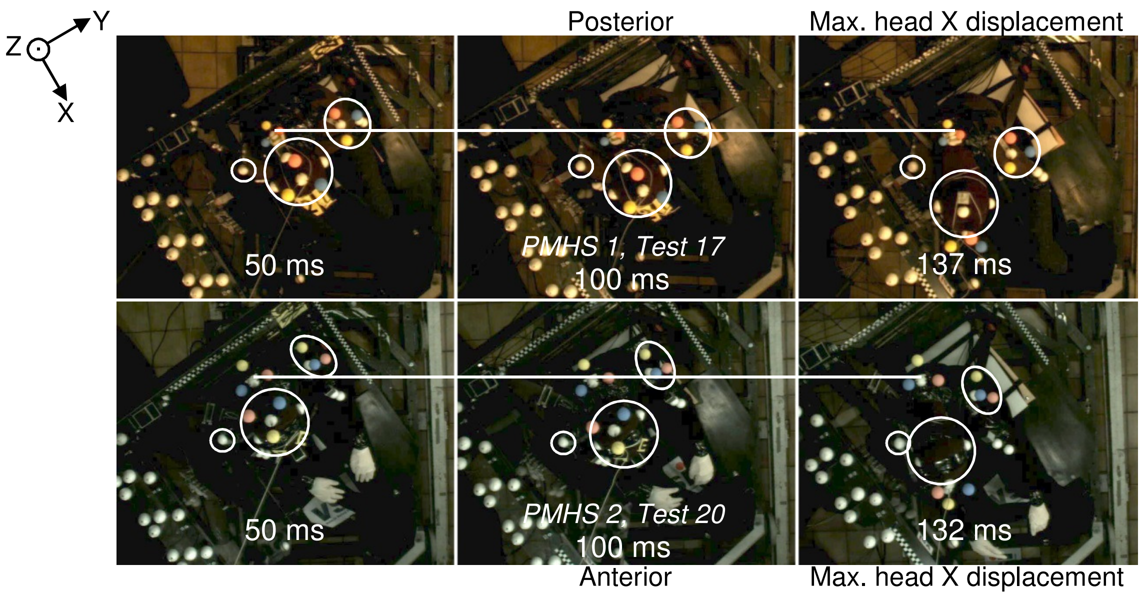


Figure E-5 Top view of PMHSs 1 and 2 in the 30° near-side test set-up at 50 ms, 100 ms and at maximum anterior head CG displacement (X direction) in relation to the sled. The horizontal line in each row of photos is set at T1 level. The middle circle in each photo is the head position. The smaller white circles and ellipses represent the shoulder marker. (Test 25 with PMHS 3 is not represented in the figure due to top camera failure), (adopted from Törnvall et al., 2008)

The loaded shoulder and arm of PMHS 3 made contact with the simulated door after approximately 115 ms (during the loading phase), in contrast to those of PMHSs 1 and 2 which did not make contact with the simulated car door. The test subjects reached their maximum anterior head CG displacement at 137 ms (PMHS 1), 132 ms (PMHS 2), and 129 ms (PMHS 3, Table B-2 and Figure E-9, 13). The head of PMHS 1 made contact with the right arm which, in turn, made contact with the simulated car door at 140 ms. The 3-D markers mounted at the top of the heads of PMHSs 2 and 3 did make contact with the simulated car door after approximately 160 ms and 155 ms, respectively.

The loaded shoulder of PMHS 1 moved anterior to T1 (

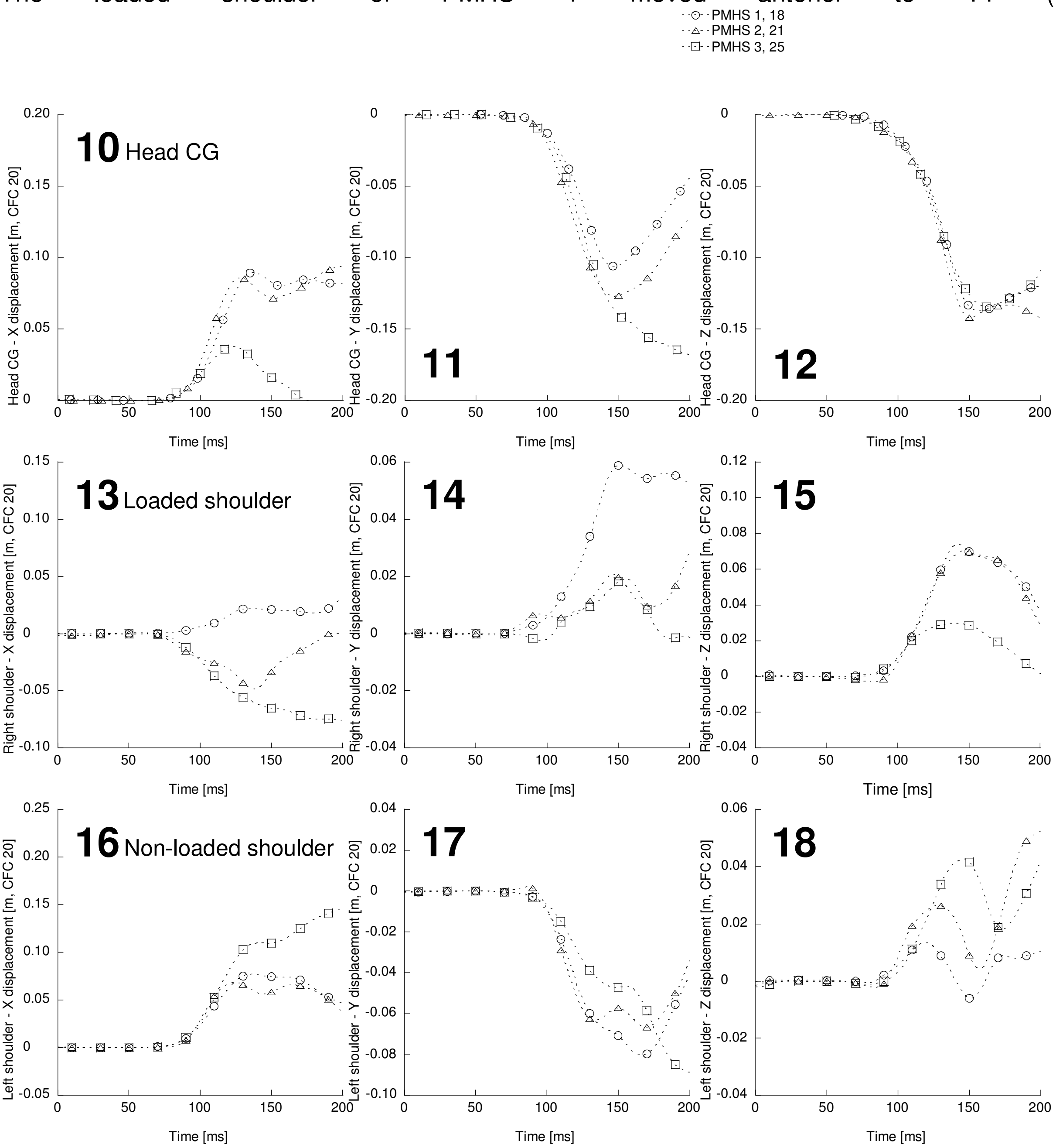


Figure E-12, 13). However, the loaded shoulders of PMHSs 2 and 3 moved posterior to T1. The loaded shoulders of all test subjects moved superior to T1 (

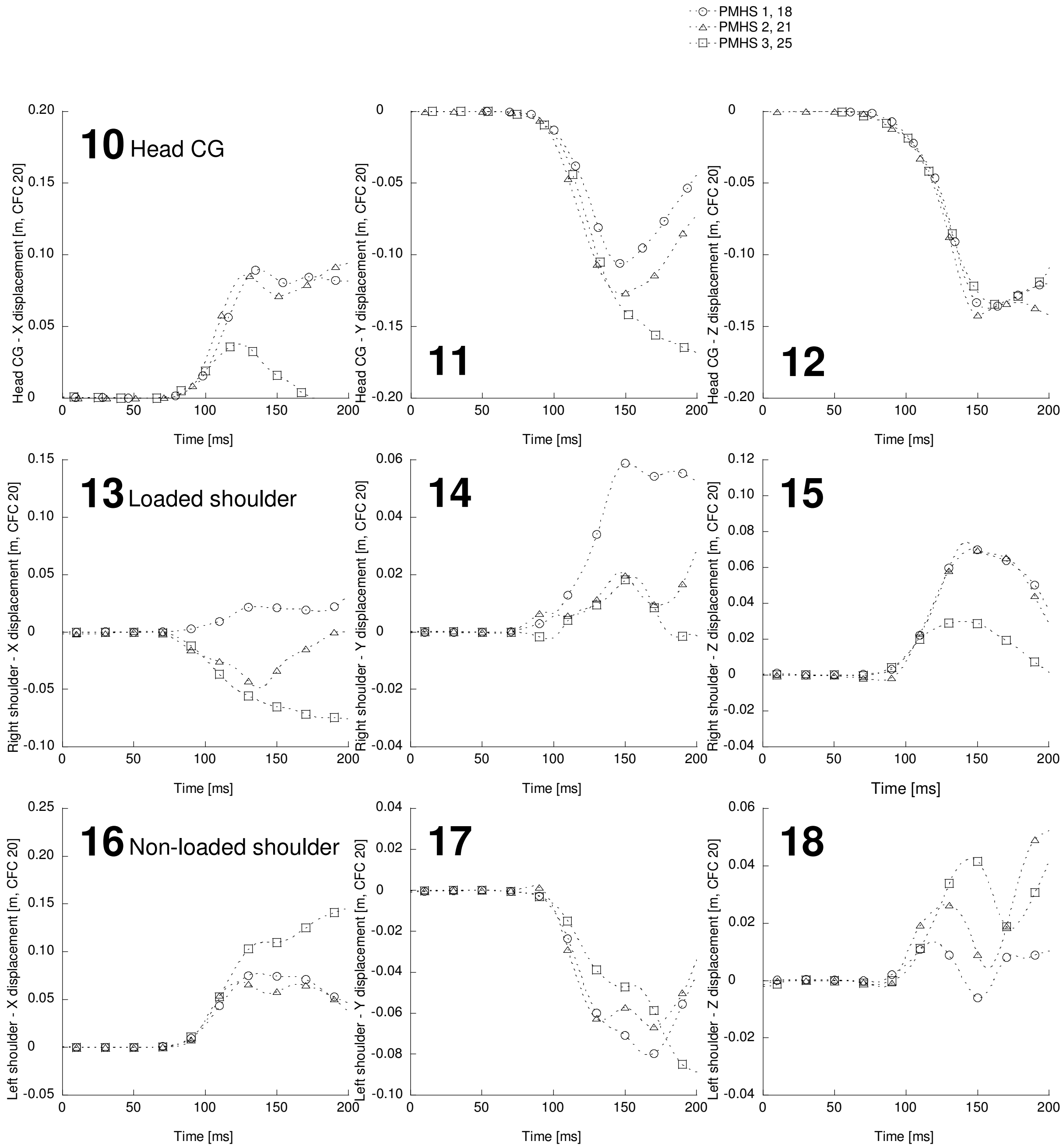


Figure E-12, 15). The non-loaded shoulder moved anterior to T1 for all three test subjects (

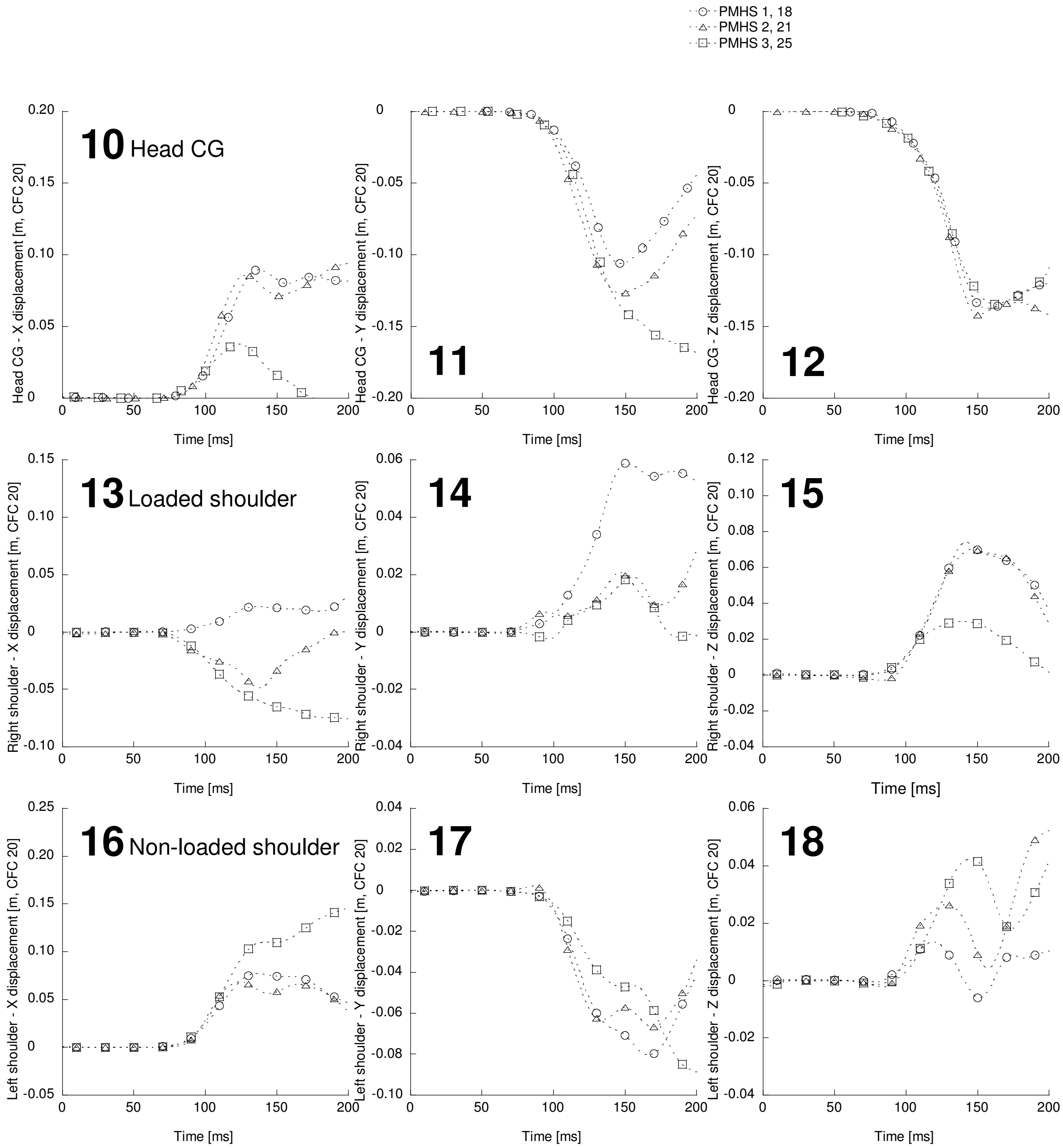


Figure E-12, 16). Moreover, the non-loaded shoulder moved superior to T1, initially, and then inferior to T1 for all test subjects (

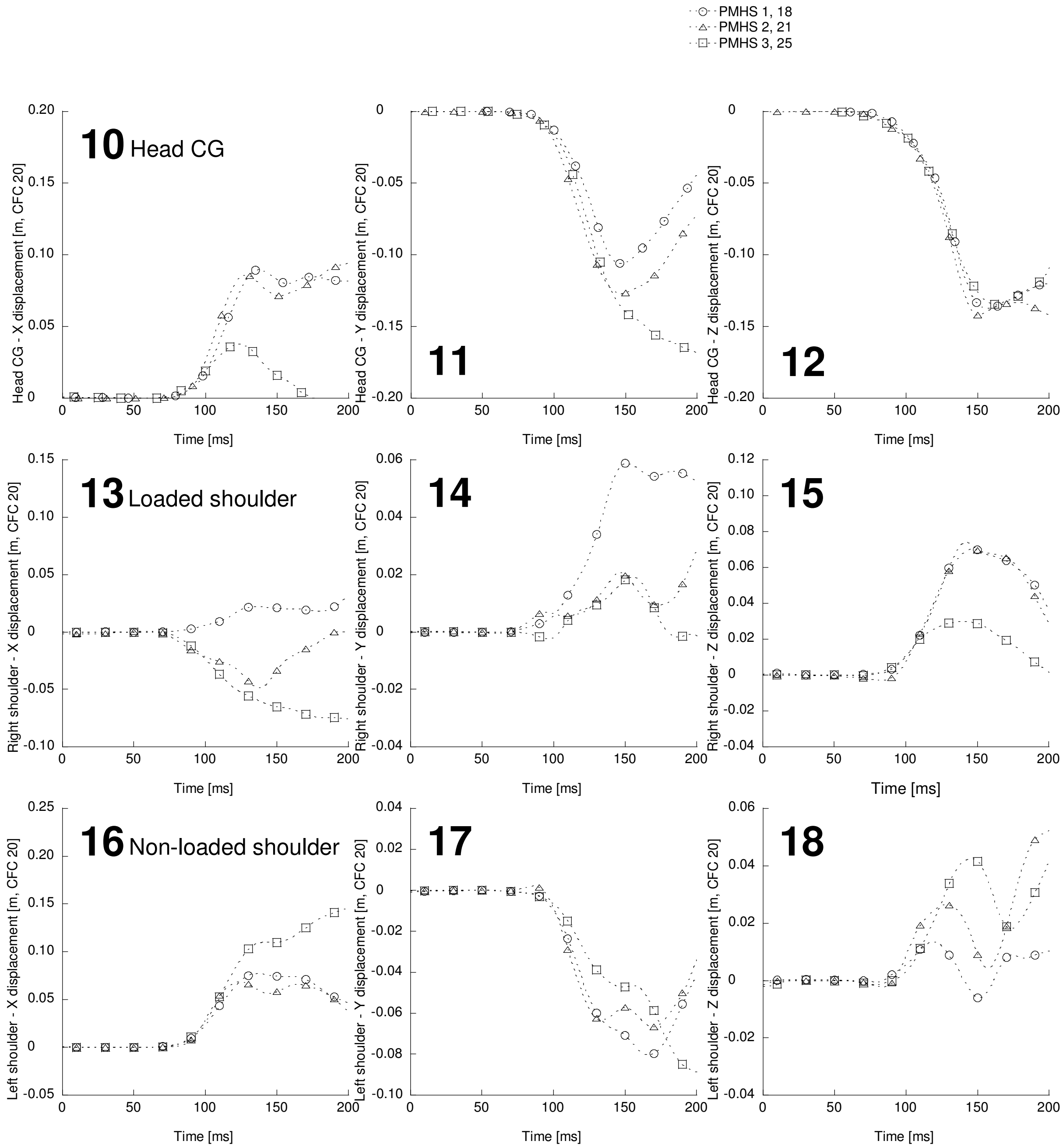


Figure E-12, 18).

E.3.3 0° Full-Frontal Sled Tests

The high-speed video data show that the test subjects' T1 and head CG markers moved forward, in the reverse direction to that of the sled acceleration (Figure E-6 and Figure E-7). The shoulder belt restrained the loaded shoulder. As a consequence, the torso of all test subjects rotated around their T1 Z axis (this was noted by visual inspection of high speed video data). When the shoulder belt restrained the test subjects, their heads rotated forward, in the reverse direction to that of the sled acceleration (Figure E-7, 1–2).

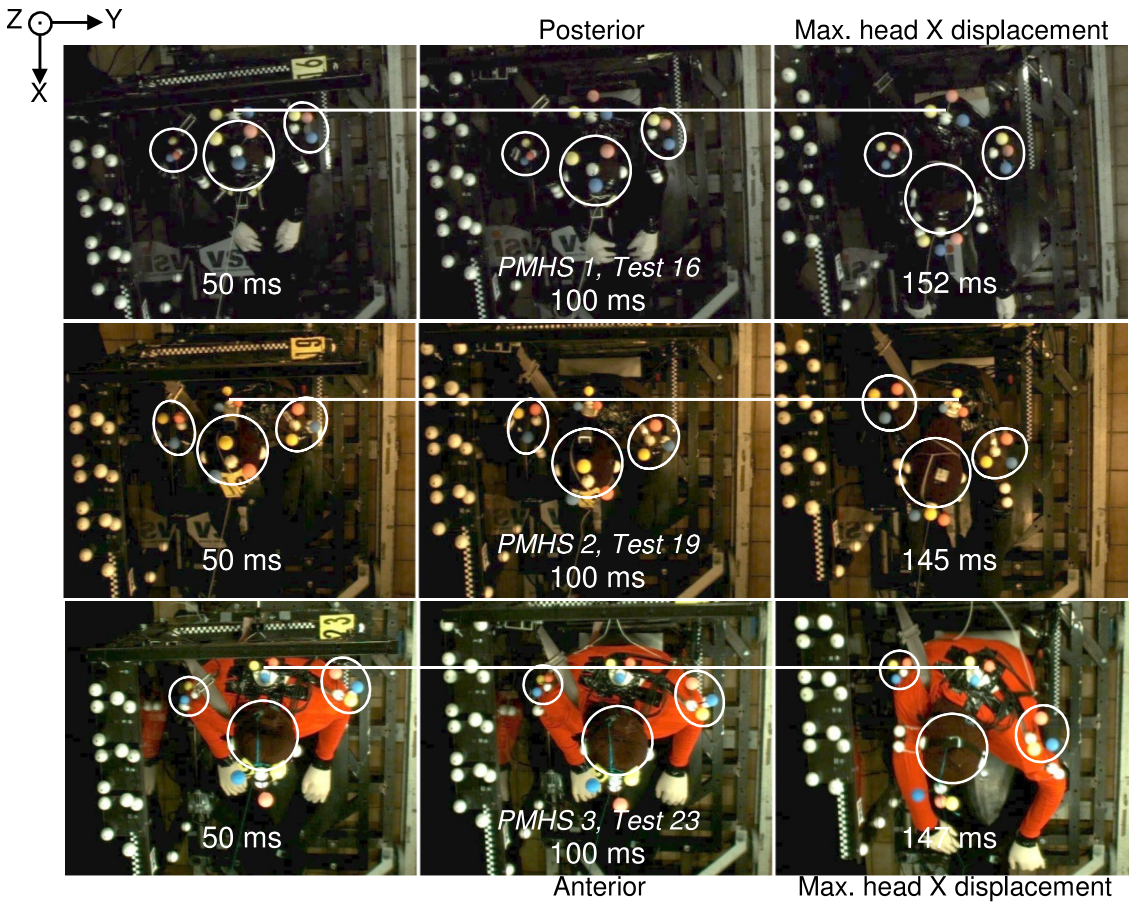


Figure E-6 Top view of the test subjects in the full-frontal test set-up at 50 ms, 100 ms and at maximum anterior head CG displacement (X direction) in relation to the sled. The horizontal line in each row of photos is set at T1 level. The middle circle in each photo is the head position. The smaller white circles and ellipses represent the shoulder markers. (Test 22 with PMHS 3 is not represented in the figure due to top camera failure), (adopted from Törnvall et al., 2008)

The test subjects reached their maximum anterior head CG displacement at 152 ms (PMHS 1), 145 ms (PMHS 2) and 147 ms (PMHS 3, Table B-2 and (Figure E-10, 25). The loaded shoulder of all three subjects moved posterior to T1 (

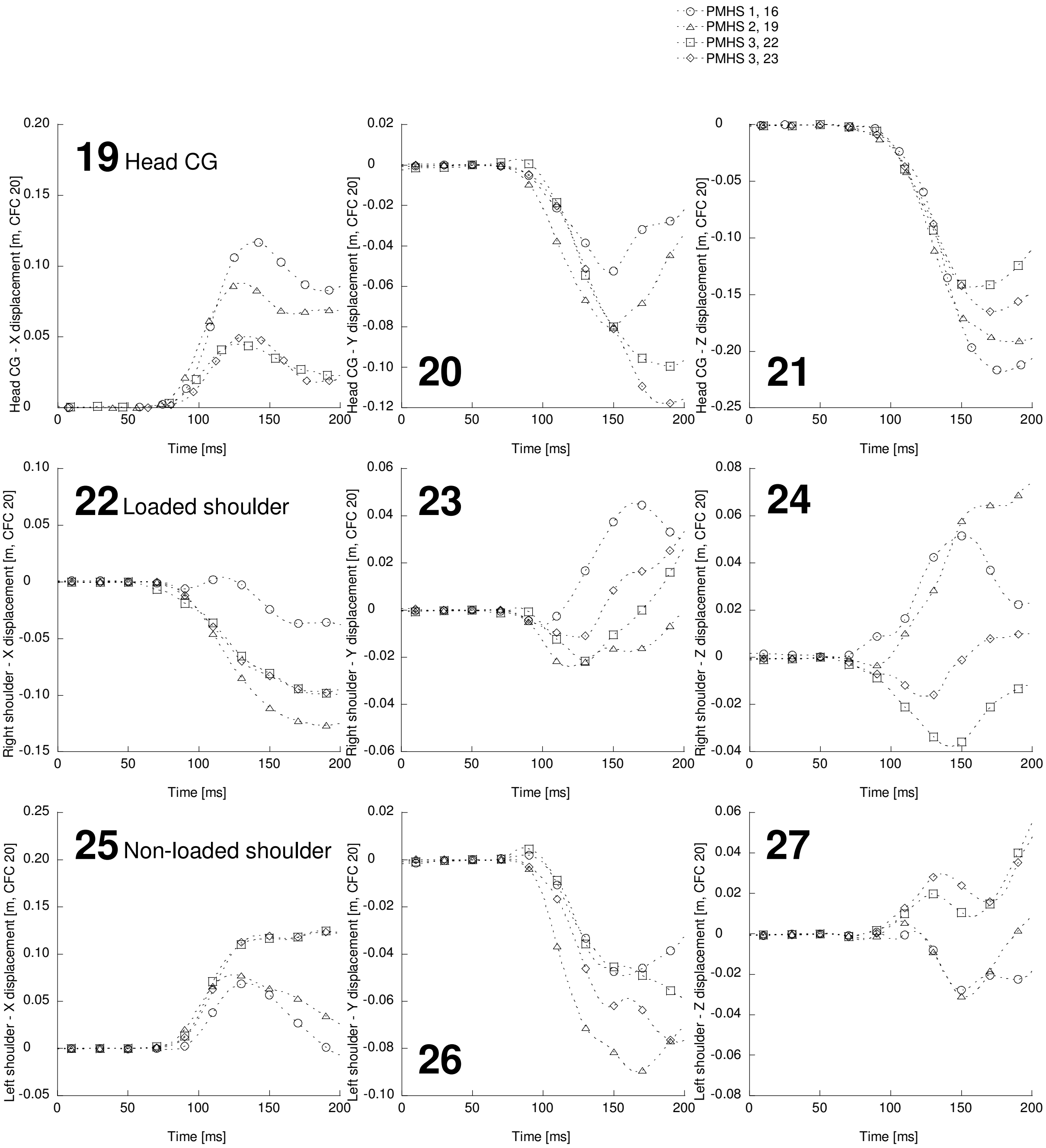


Figure E-13, 22). The loaded shoulder of PMHS 1 did not move as much posterior as those of PMHSs 2 and 3. The non-loaded shoulder moved anterior to T1 for all three test subjects (

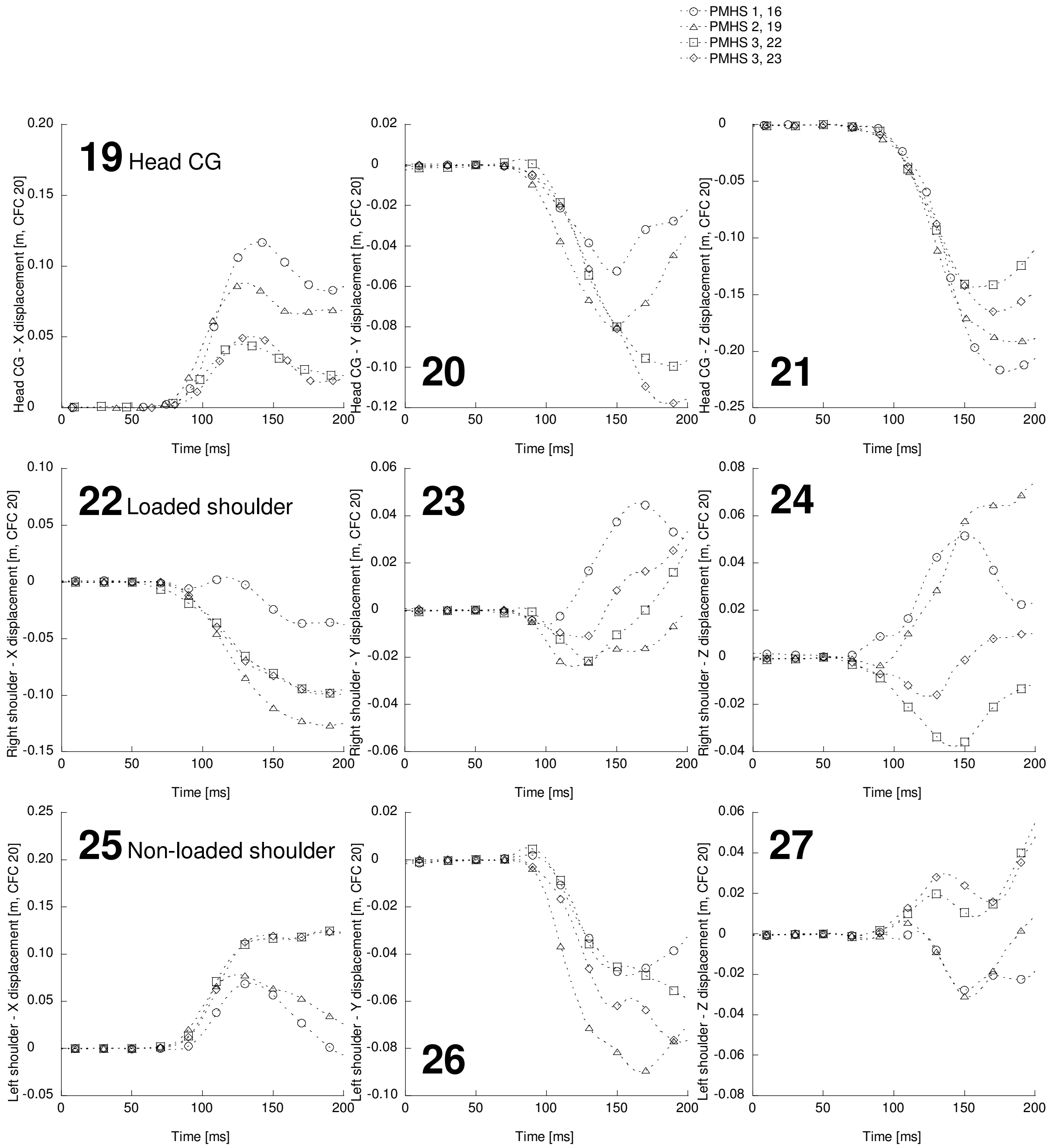


Figure E-13, 25). There were contradictory results regarding the superior-inferior motion of the shoulders. The loaded shoulders of PMHSs 1 and 2 moved superior to T1 in contrast to the shoulder of PMHS 3 (

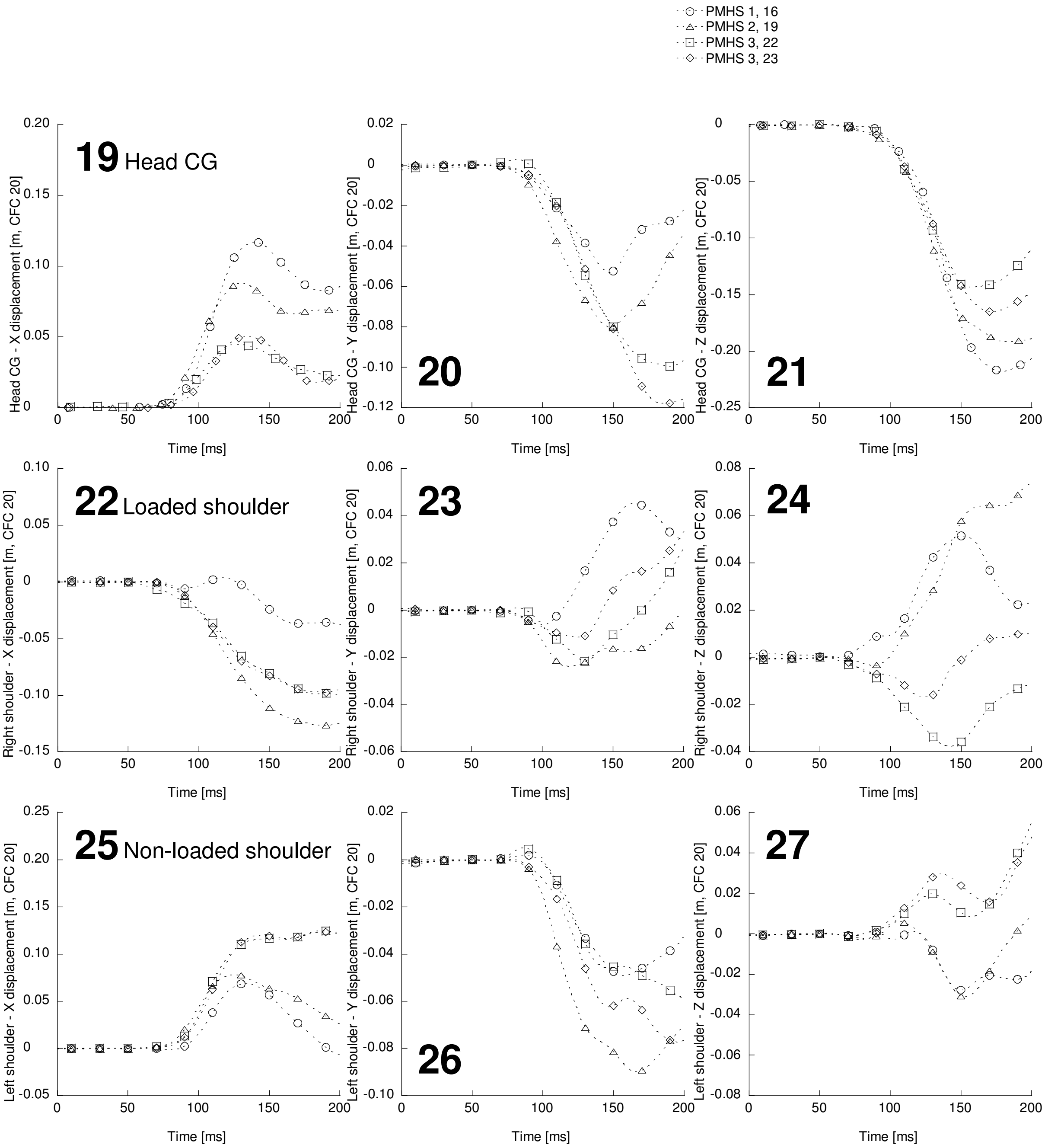


Figure E-13, 24). Moreover, the non-loaded shoulders of PMHSs 1 and 2 moved inferior to T1 in contrast to the non-loaded shoulder of PMHS 3 (

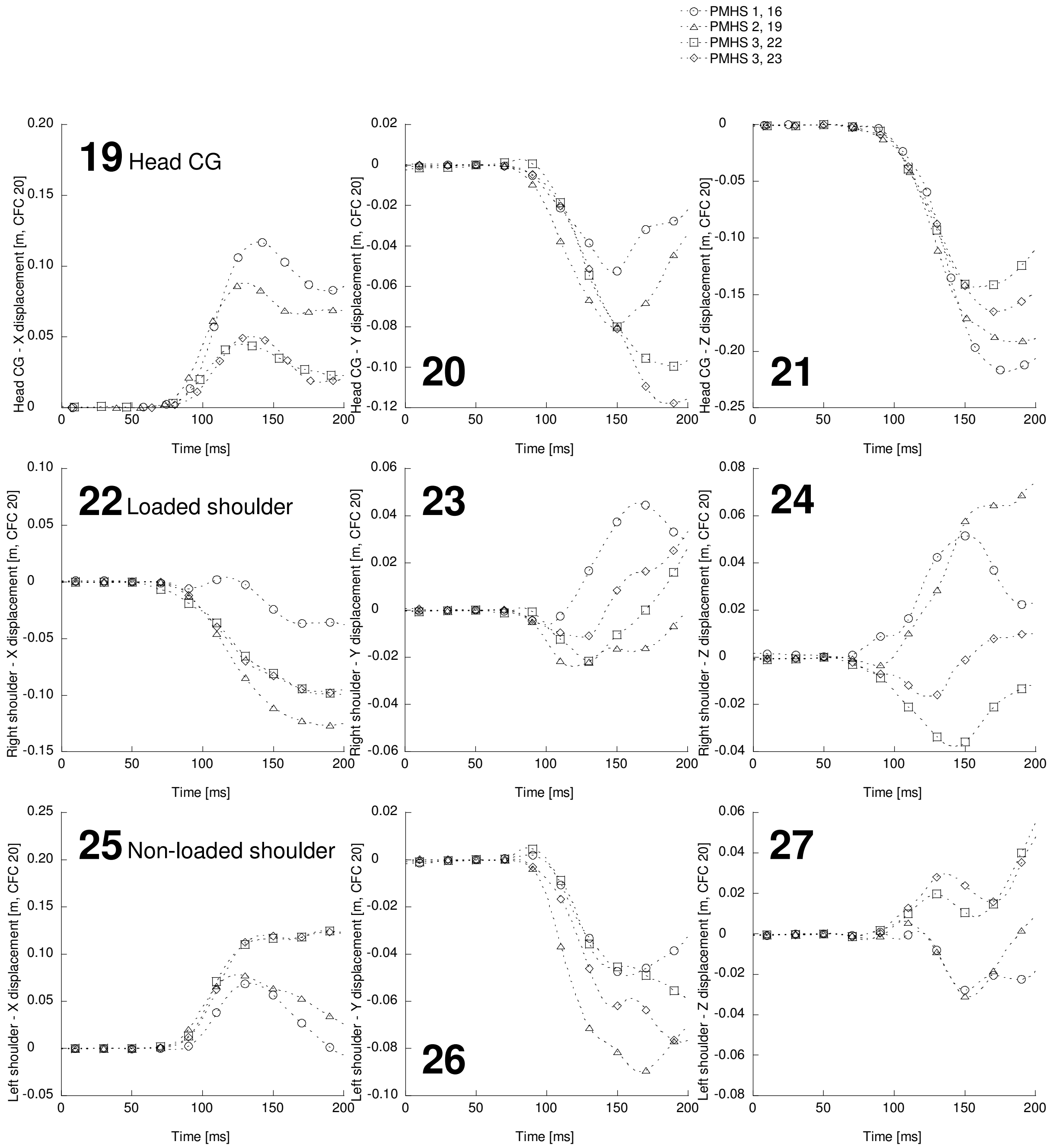


Figure E-13, 27).

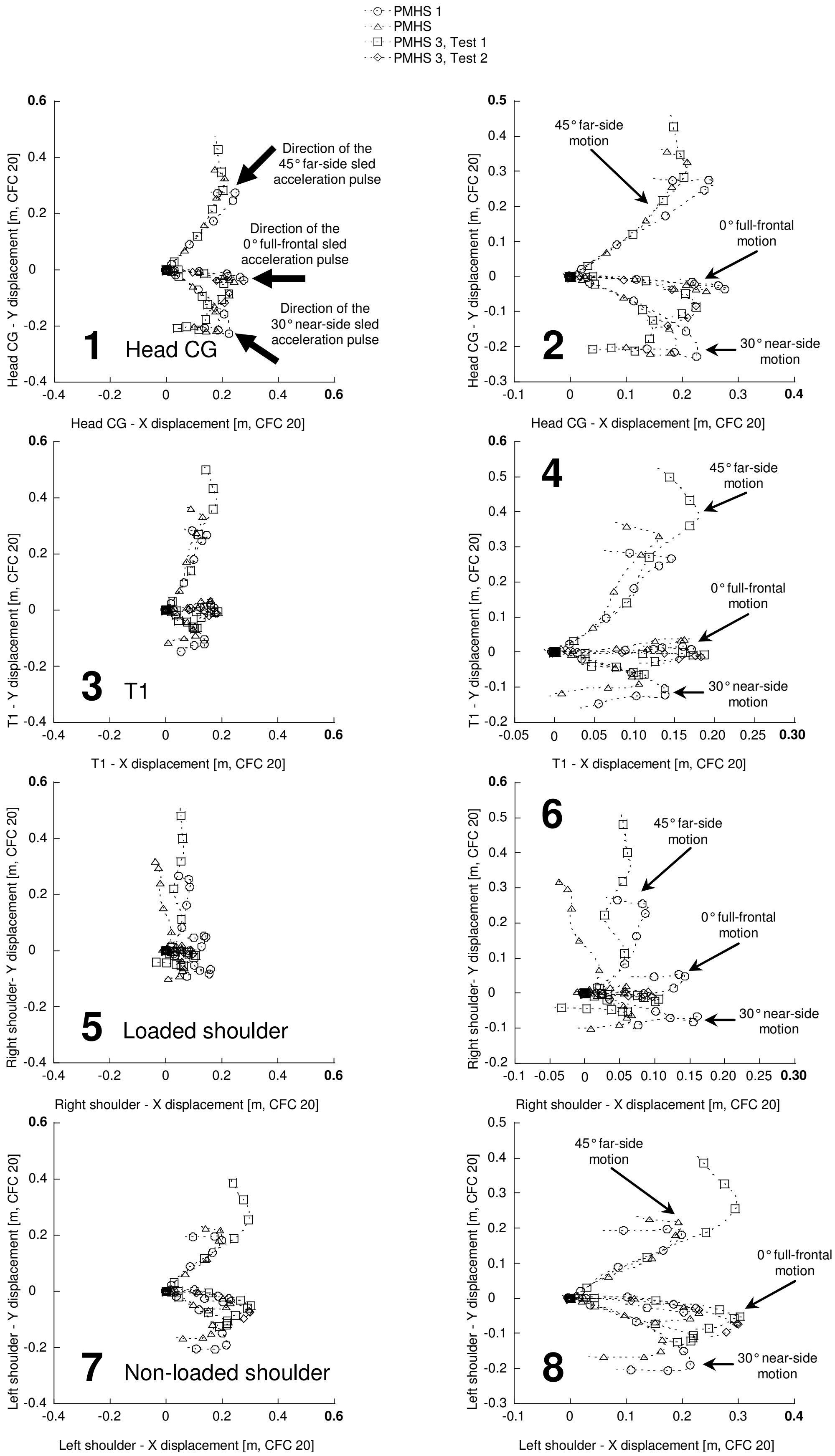


Figure E-7 Head, T1 and shoulders trajectories in 45° far-side, 30° near-side and 0° full-frontal (adopted from Törnvall et al., 2008)

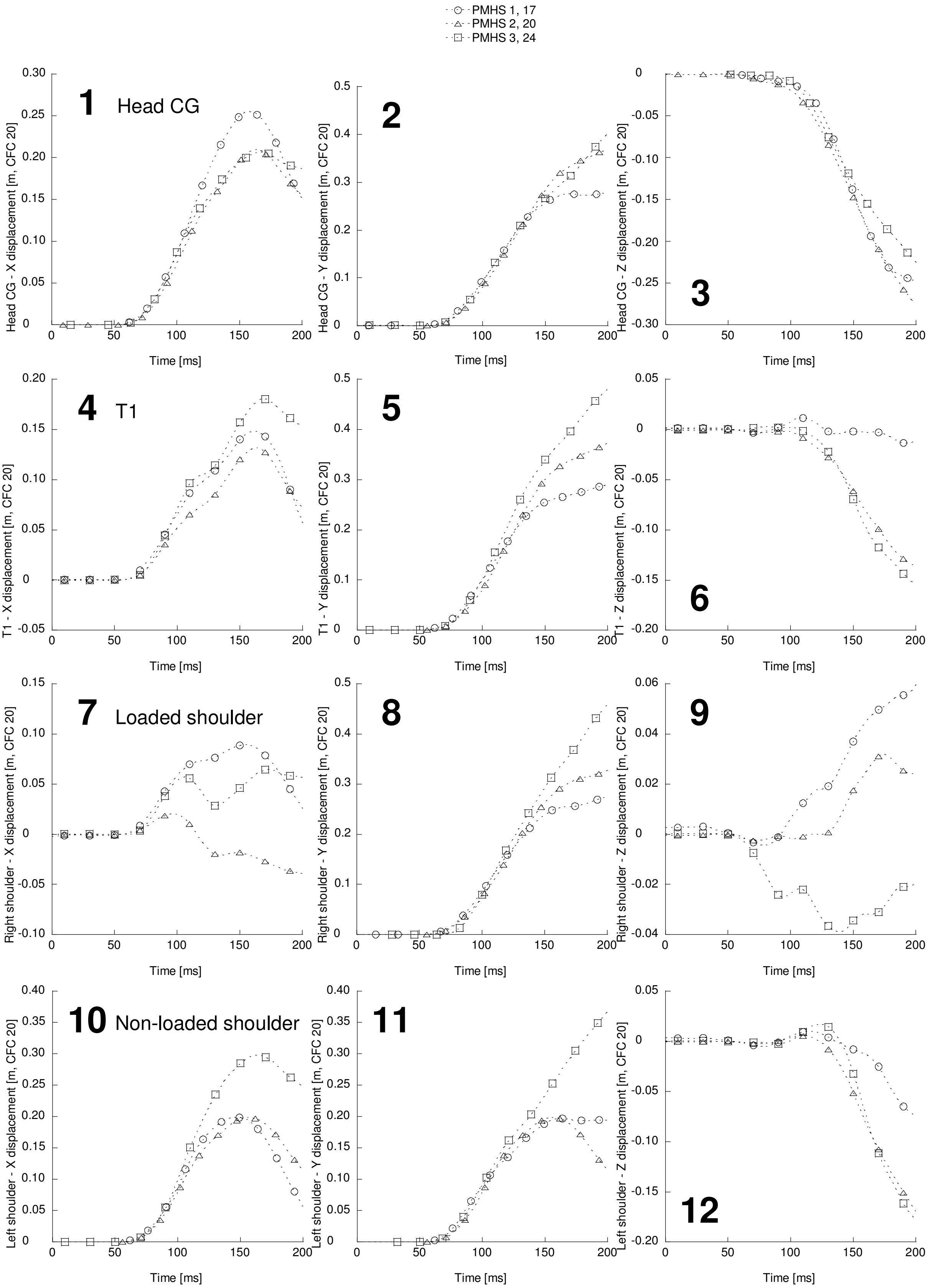


Figure E-8 Head, T1 and shoulders displacements relative the sled in 45° far-side (adopted from Törnvall et al., 2008)

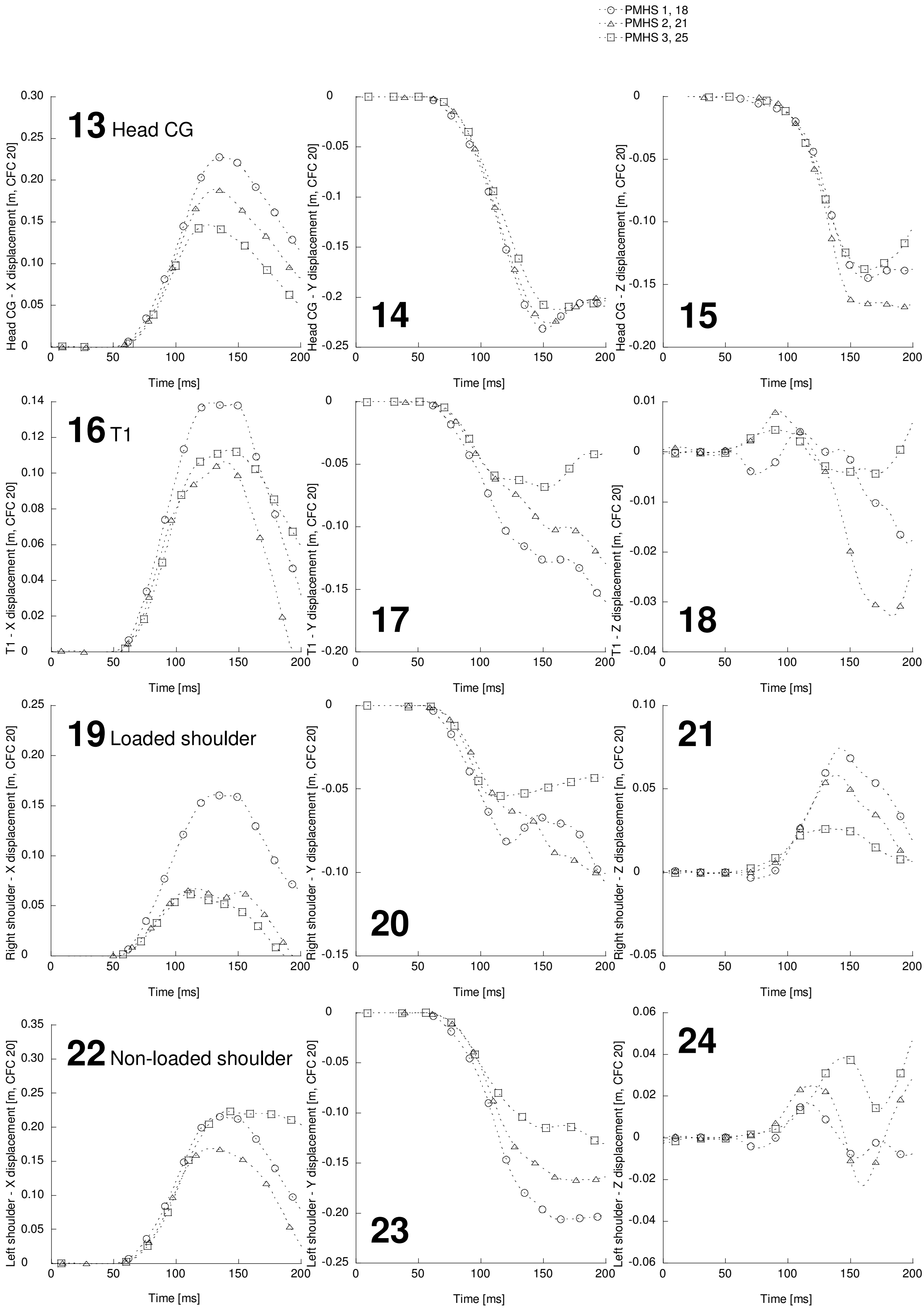


Figure E-9 Head, T1 and shoulders displacements relative the sled in 30° near-side (adopted from Törnvall et al., 2008)

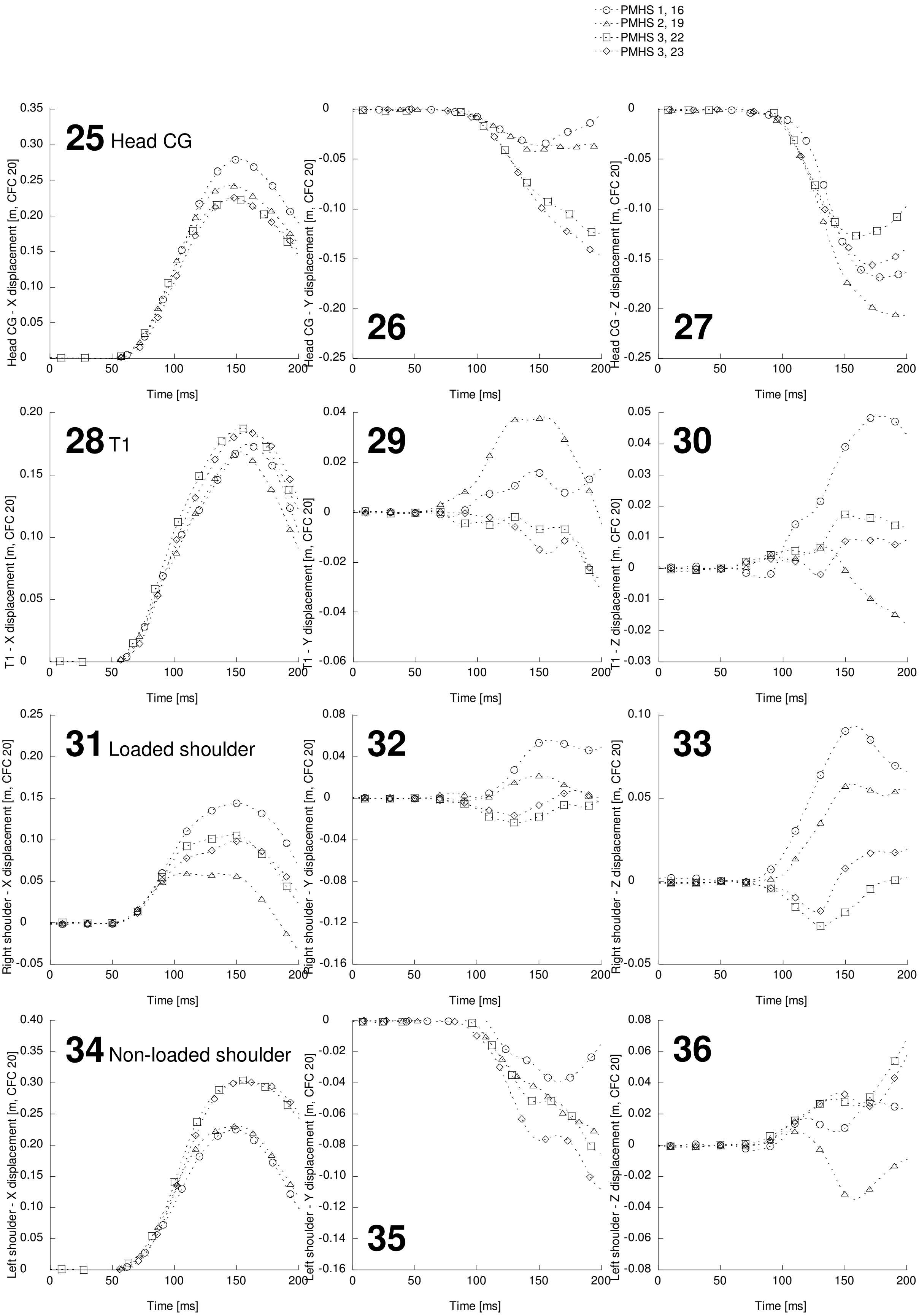


Figure E-10 Head, T1 and shoulders displacements relative the sled in 0° full-frontal F3 (adopted from Törnvall et al., 2008)

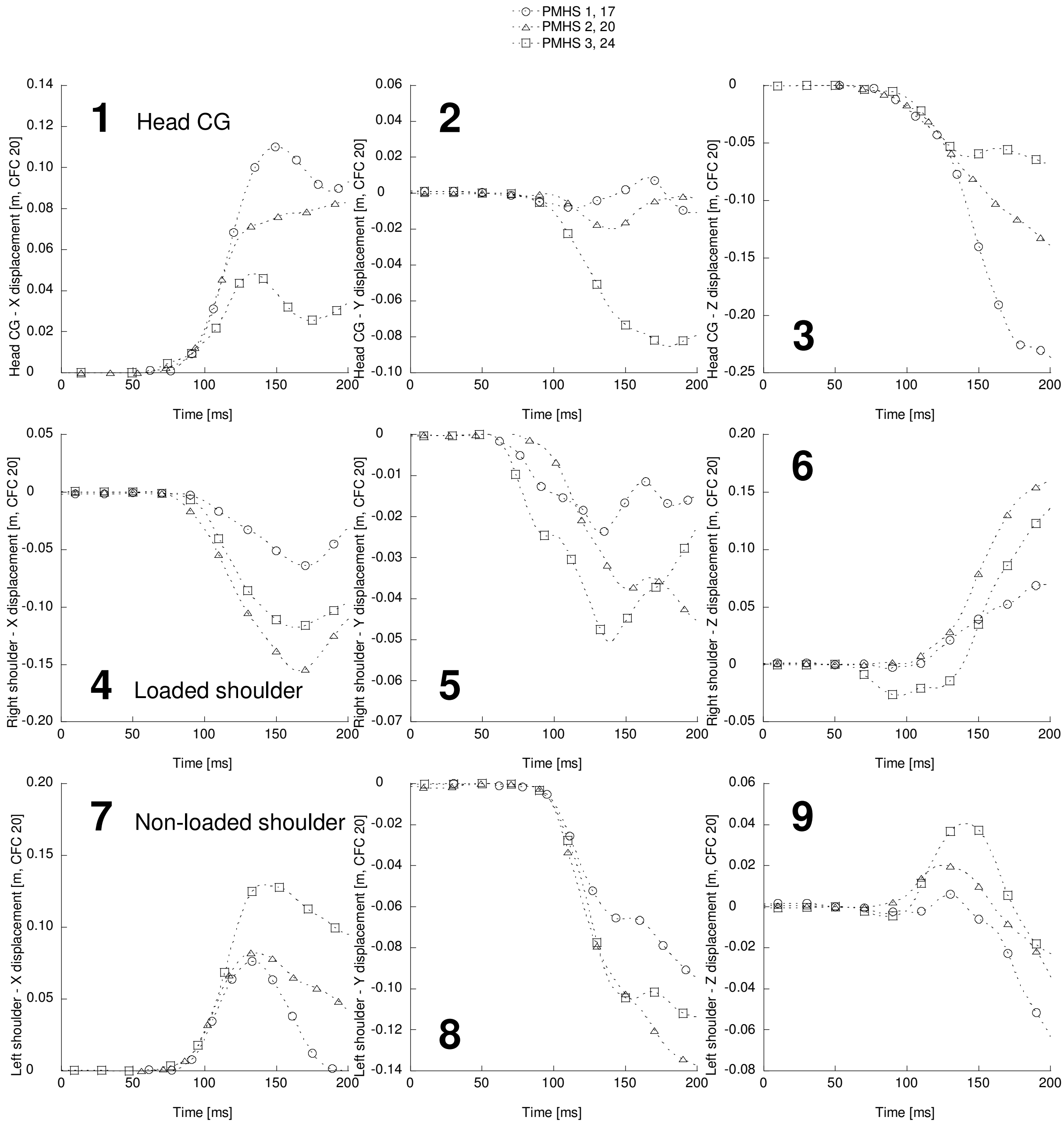


Figure E-11 Head and shoulders displacements relative T1 in 45° far-side (adopted from Törnvall et al., 2008)

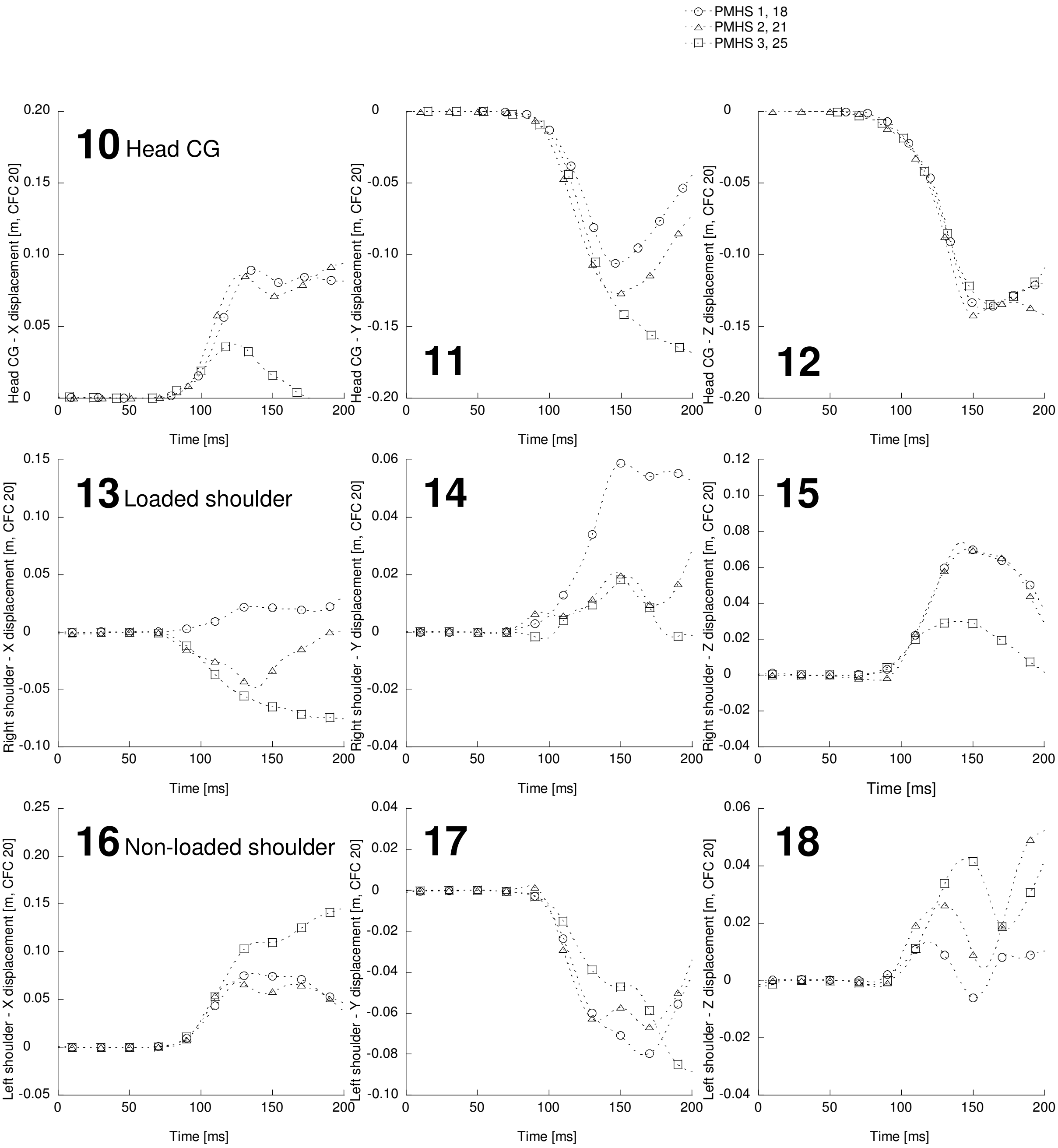


Figure E-12 Head and shoulders displacements relative T1 in 30° near-side (adopted from Törnvall et al., 2008)

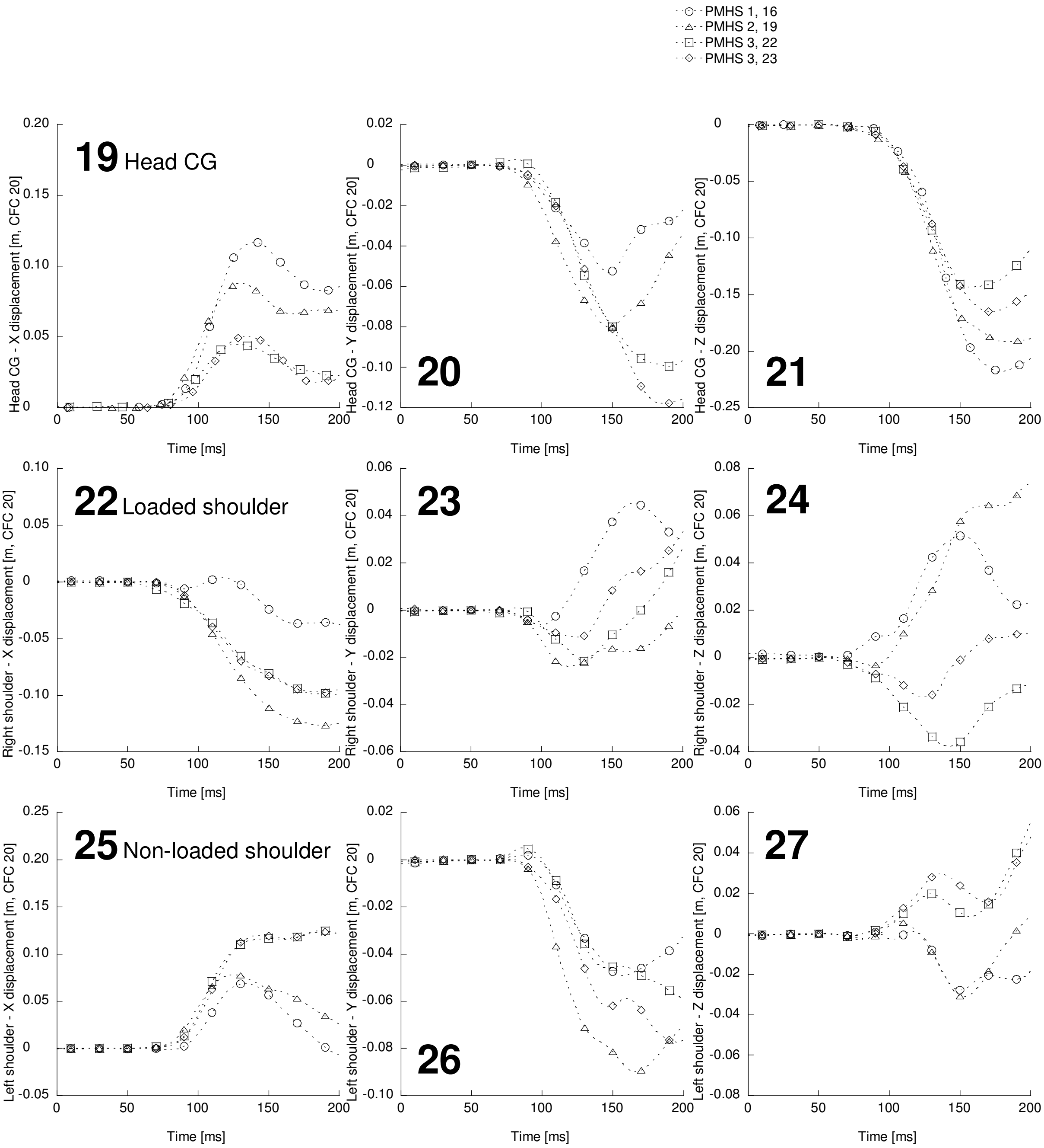


Figure E-13 Head and shoulders displacements relative T1 in 0° full-frontal (adopted from Törnvall et al., 2008)

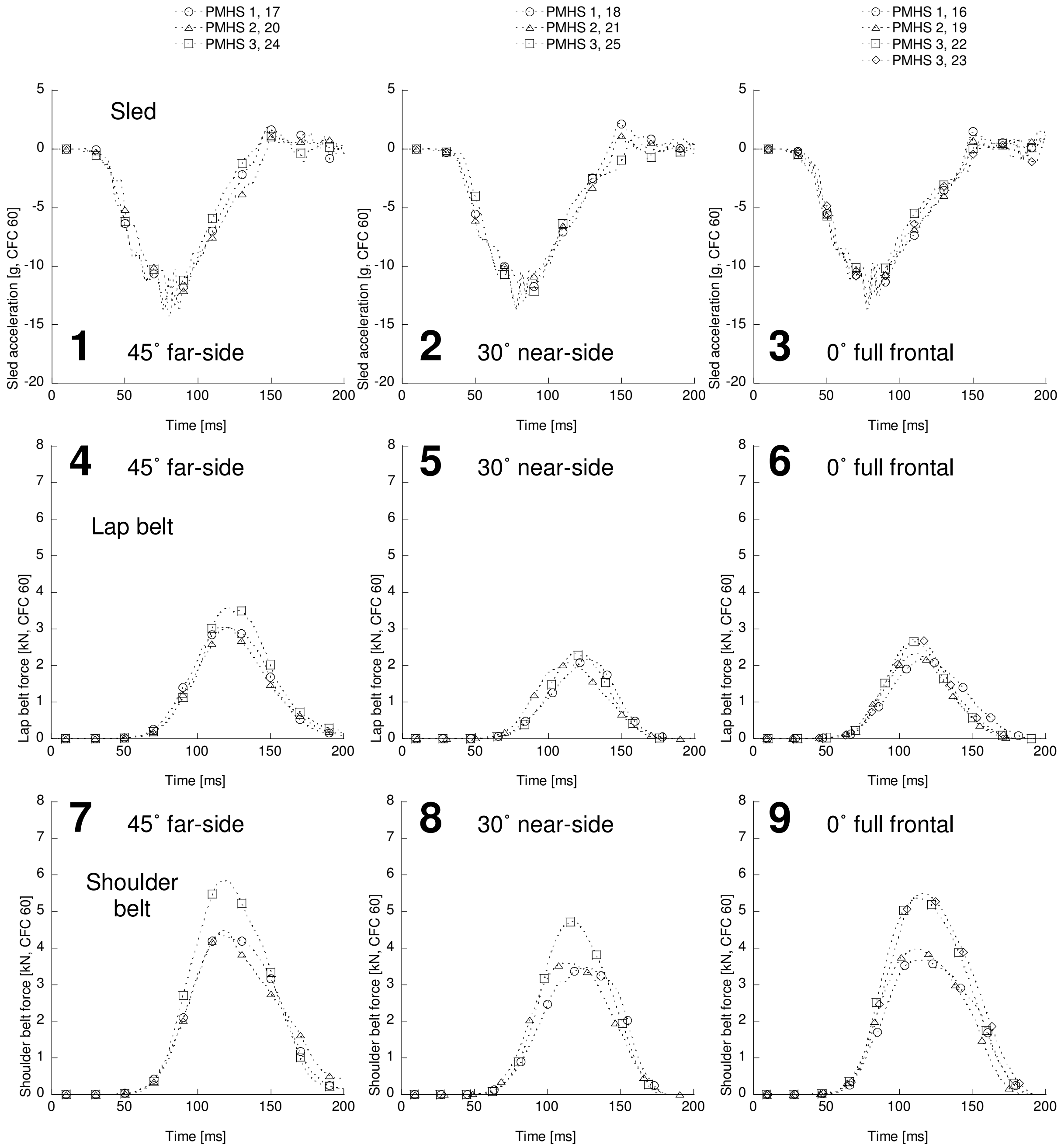


Figure E-14 Sled acceleration and belt forces in 45° far-side, 30° near-side and 0° full-frontal (adopted from Törnvall et al., 2008)

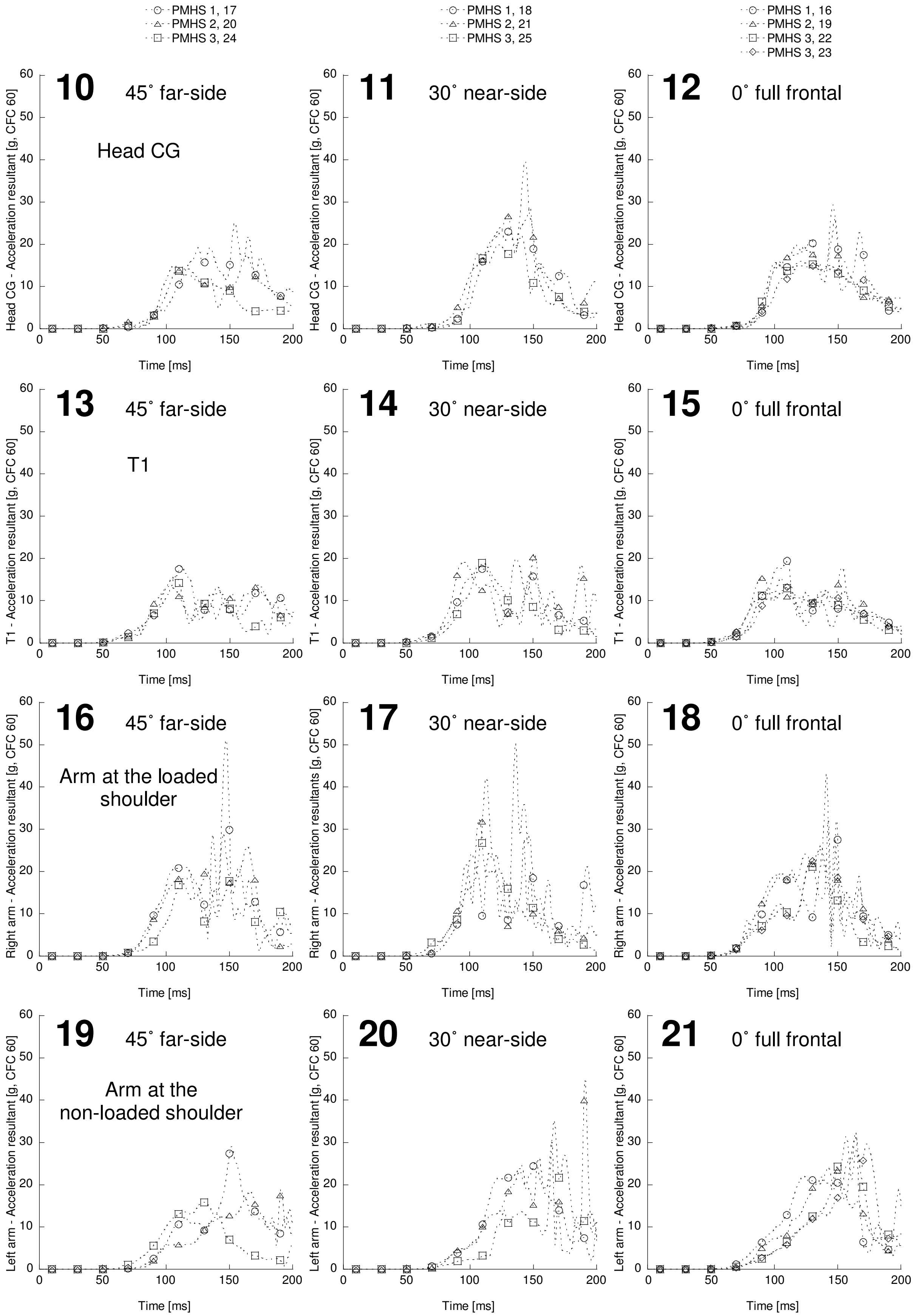


Figure E-15 Head, T1 and arm acceleration in 45° far-side, 30° near-side and 0° full-frontal (adopted from Törnvall et al., 2008)

Appendix F **Cavanaugh *et al.* (1988)**

Recommended Engineering Guideline:
Relative chest deflection
Table-top tests

The information of the table top test performed by Cavanaugh et al 1988 were obtained from the papers of Cavanaugh et al 1988, Schneider et al 1989, Schneider et al 1992, and Shaw et al 2005.

F.1 Methods

Chests of PMHS in supine position were loaded statically using a 50mm x 100mm (2”x4”) rigid plate while measuring the chest deflection at 9 points (including the indentor displacement), three of them on sternum (upper, mid, and lower position). Points measured on ribs were positioned approximately at 76 mm from the middle line at 2nd, 5th, and 8th rib.

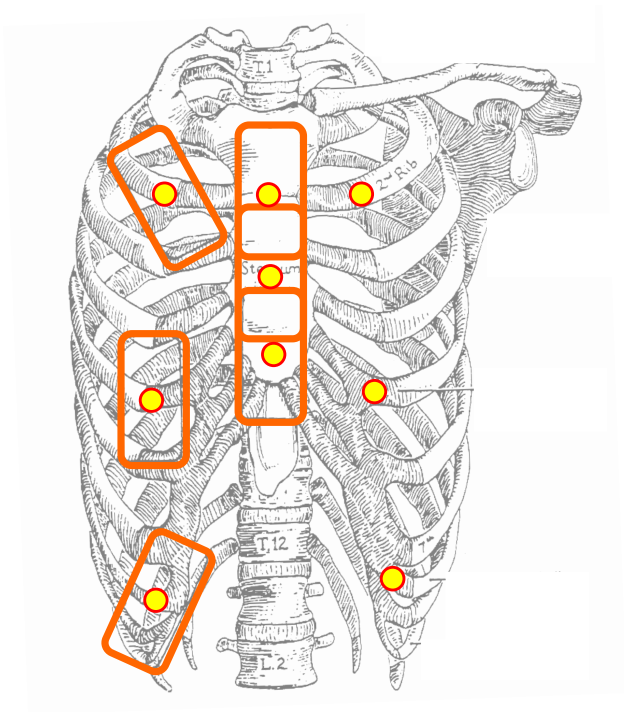


Figure F-1: Regions of static loading with 5 cm x 10 cm surface and deflection measurement used by Cavanaugh (adopted from Cavanaugh et al. 1988)

Tests were designed to measure the deflection of the adjacent ribs to the loading location (i.e. coupling effects), and the regional stiffness at locations where seatbelt could press the torso. In tests, the coupling magnitude was expressed in relative deflection with respect at point pressed for the gimballed rectangular indentor (25.4mm).

Points to fix the chest were two types: In one case, for an aluminium bar in a structure closed to the spine, and in other case, besides the aluminium bar, ribs rest bilaterally at 7 cm (2.8”) from the medium line approximately.

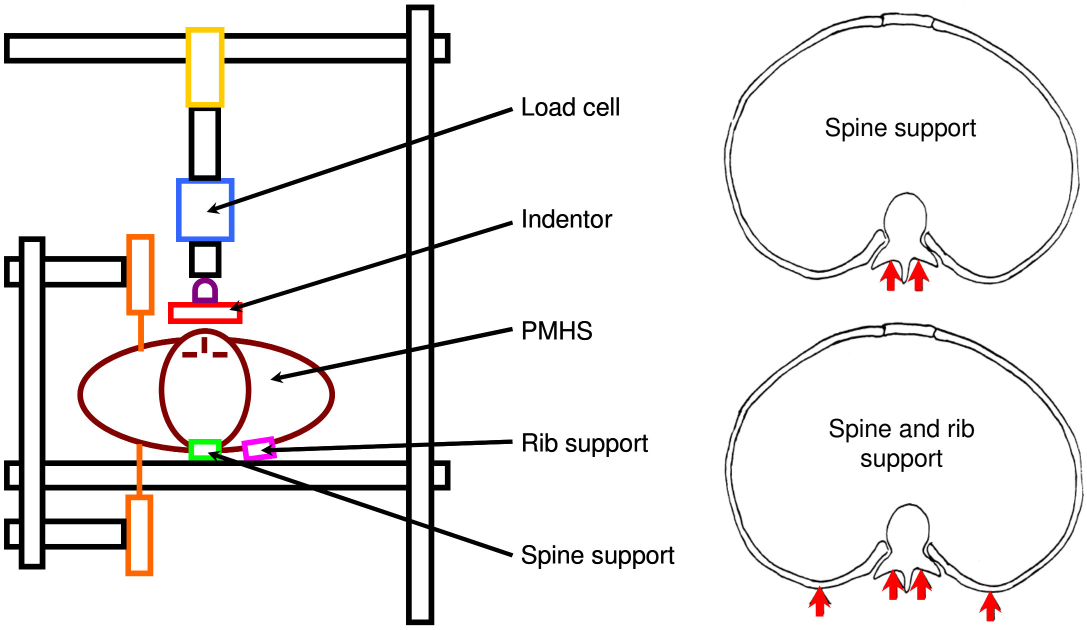


Figure F-2: Test setup for static loading tests

Loads were applied on upper, mid, and lower sternum, and the ribs at upper, mid, and lower regions. Test with load on sternum were measured with two test setups described above, while tests with loads on ribs were measured only with the second test setup (spine and ribs supported). When the tests were performed with the spine support only, the displacements on anterior ribs were registered. By contrast, when the ribs were supported as well the anterior rib displacement was not measured (did not allow it).

Loading rates ranged from 1.7 mm/s to 102 mm/s (0.067 in/s to 4 in/s), and the displacement was normally 25mm (1 inch).

X deflection was measured. The X axis definition is according with SAE J211 (sternum to spine).

Two subjects were tested. A brief description of them is included below:

Table F-1: General information of the cadaveric surrogate data of Cavanaugh et al 1988 tests

Test	PMHS No.	Gender	Age	Height	Weight
AATD2; test 1-10	986	Male	29	1.73	70.3
AATD5; test 1-30	115	Male	57	1.79	57.6

F.2 Results

The data obtained by Cavanaugh et al 1988 are showed below. These data are taken from Schneider et al. 1989, Schneider et al. 1992, and Shaw et al. 2005 (these papers have more information than the original paper that it has been used for verifying the results shown below).

Table F-2: 1.7mm/s. Max deflection 25 mm (1.0 in relative). Ribs supported bilaterally (Schneider et al 1989).

Load	Coupling								
	Right Ribs			Sternum			Left Ribs		
	R2	R5	R8	SU	SM	SL	L2	L5	L8
R2	1.00	0.22	0.05	0.31	-	0.24	0.13	0.00	0.01
R5	0.08	1.00	0.40	0.04	-	0.49	0.01	0.09	0.12
R8	0.00	0.42	1.00	0.00	-	0.42	0.00	0.11	0.14
SU	0.5	0.2	0.07	1.0	-	0.35	0.5	0.2	0.07
SM	0.37	0.30	0.16	-	1.0	-	0.37	0.30	0.16
SL	0.20	0.50	0.55	0.16	-	1.0	0.20	0.50	0.55

Table F-3: Quasi-static stiffness values at 25 mm deflection obtained by Cavanaugh using a 50mm x 100mm (2" x 4") rigid loading plate (Schneider et al 1992).

Cadaver	Stiffness (N/mm)					
	Sternum			Right Ribs		
	SU	SM	SL	R2	R5	R8
1	12.3	10.6	11.4	7.3	8.4	5.2
2	11.4	10.6	5.9	5.6	5.4	3.4
3	11.7	8.6	7.4	7.0	5.1	3.9
Mean	11.8	9.93	8.23	6.63	6.33	4.17

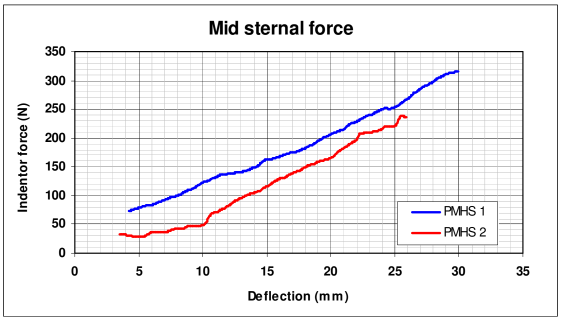


Figure F-3: Mid-sternal force results for two cadavers tested by Cavanaugh et al 1988 (Shaw et al 2005)

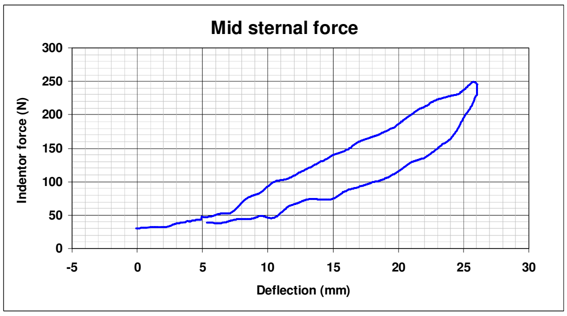


Figure F-4: Force-deflection curve of AATD5, Run 1A, Cadaver #115, loaded at mid-sternum with 25mm (1”) stroke at 1.7 mm/s (0.067 in/s) (Schneider et al 1989)

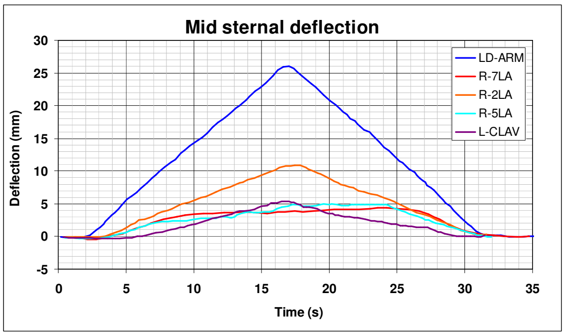


Figure F-5: Deflection-time history at mid-sternal load arm, second, fifth and seventh ribs and left clavicle (Schneider et al 1989)

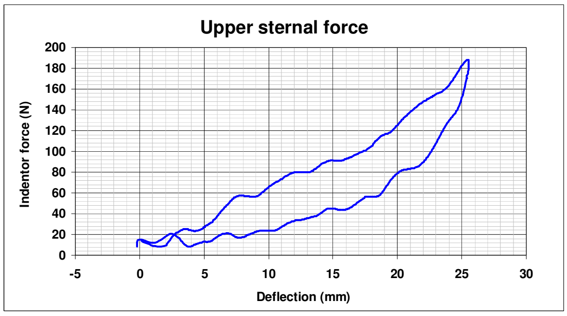


Figure F-6: Force-deflection curve of AATD5, Run 5, Cadaver #114, loaded at upper sternum with 25mm (1”) stroke at 1.7 mm/s (0.067 in/s) (Schneider et al 1989)

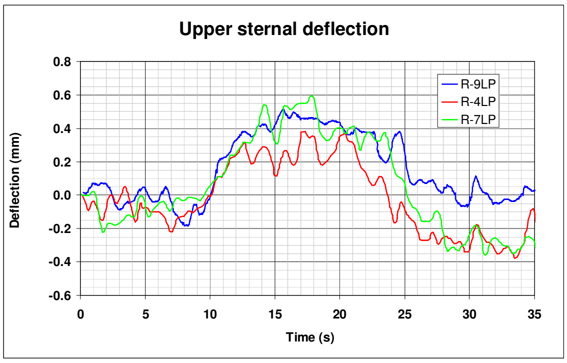


Figure F-7: Deflection-time history at fourth, seventh and ninth left ribs, posteriorly (Schneider et al 1989)

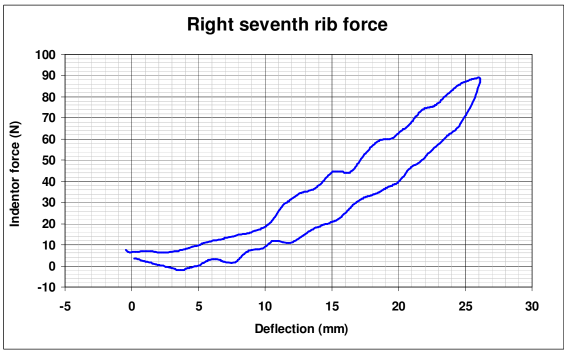


Figure F-8: Force-deflection curve of AATD5, Run 17, Cadaver #115, loaded at right seventh rib with 25 mm (1 in) stroke at 1.7 mm/s (0.067 in/s) (Schneider et al 1989)

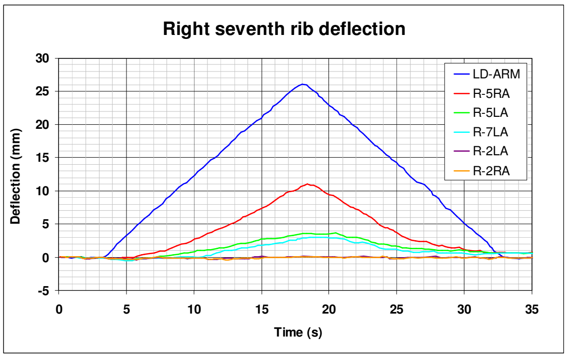


Figure F-9: Deflection-time history at load arm (right seventh rib), right second rib, right fifth rib, and left second, fifth and seventh ribs, anteriorly (Schneider et al 1989)

F.3 Additional information for replication of tests with dummies

Dummy should be tested without its jacket. A spine support and a rib support will be used under the dummy, with a target distance of 7cm bilaterally from the mid line.

Points to measure will be equivalent (dummies have normally not the same number of ribs than humans).

The stroke of the tests will be at a speed of 1.7 mm/s and 102 mm/s with a stroke of 25 mm.

Appendix G **Shaw et al. (2007) table top**

Dynamic and quasi-static tests were performed on five not embalmed post-mortem human subjects (PMHS). The used indentors were designed in order to have a similar contact section that the seat belt's section.

G.1 Description of the PMHS tests

A perforated steel plate was bolted to the spine and pelvis. Wood screws were placed bilaterally through ten vertebrae, in the range T1 to L5, and the posterior iliac crests. The steel plate was joined to a 19 mm plywood sheet. In this way, the spinal curvature in the sagittal plane was eliminated and prevents any other undesired movement.

Two sizes of aluminium indentors were used: 62 mm wide x 62 mm tall (2" x 2") and 62 mm wide x 113 mm tall (2" x 4"). They were attached to the load cell through a spherical joint which allows 15 degrees of rotation in order to adapt to chest shape.

The indentor was rotated 30 degrees (15 degrees in some case) with respect the longitudinal chest line. Three millimetres thick natural rubber sheets were adhered under the indentor, measuring 58 mm x 58 mm and 58 mm x 105 mm.

Indentor load and three-dimensional deflection were recorded in order to measure the force-deflection. The loading tests were performed at three locations, which coincide with the path of the shoulder belt, and were loaded quasi-statically and dynamically (1 m/s) with an indentor mounted on a 6-axis load cell.

At least the three-dimensional deflection from eight up to nine points was recorded. The fractures were identified during autopsy and the X-Y distance from the centre of the sternum was measured.

Speeds were 1.7 mm/s for the quasi-static tests, and 1000 mm/s for the dynamic tests.

Indentor displacement: All subjects were "preconditioned" applying 10 one second sinusoidal deflections with 12 mm peak load in the mid sternum. Non injurious displacements were reached from 18 mm to 30 mm, while there were injuries with greater displacements (up to 80 mm). The distances from 18 mm to 30 mm were based on earlier work of Cavanaugh et al. (1988) who show that there was no fracture at 25.4 mm, but 50.8 mm made fractures, confirmed with the study by Eckert et al. (2000) who reported that 40 mm deflection was non injurious in similar tests with denuded thoraces.

To align the indentor with the rib cage contour a 25 ± 5N preload is allowed.

The centre of the indentor was placed at the following points: sternum at rib 5 (S5), the left side on rib 3 (L3) at the costochondral junction and the right side on rib 6 (R6) at the costochondral junction. These points are shown on the next figure. The points, where the displacement was measured by Vicon cameras, are also shown.

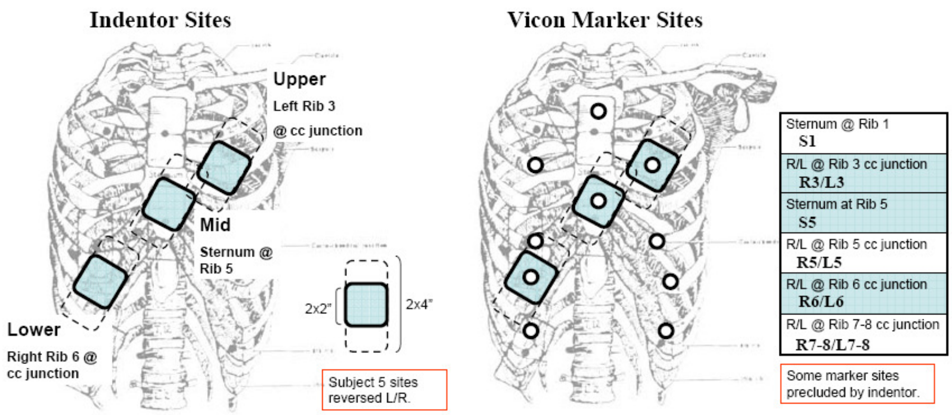


Figure G-1. Indentor and Vicon marker sites. From Shaw et al. (2007)

X, Y and Z axis definition is according with SAE J211. The data recorded are:

- Forces and moments recorded by the 6 axis load cell.
- Indentor displacement along the X-axis (potentiometer).
- Indentor and chest three axis displacement (eight Vicon cameras).

The potentiometer and load cell data were recorded with a sampling rate of 10000 Hz for the dynamic tests and 200 Hz for the quasi-static tests. High speed cameras (Vicon) were used with 1000 fps for the dynamic tests and 50 fps for the quasi-static tests.

G.1.1 Review of the Available Test Data

A total of five subjects were tested. A brief description of them is showed below:

Table G-1: Summary PMHS

Test	PMHS No.	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
1	343	Male	72	66	1.80	20.4	15
2	342	Male	75	73	1.83	21.8	10
3	320	Male	48	68	1.68	24.1	4
4	319	Male	52	77	1.79	24.0	17
5	203	Male	67	77	1.70	26.6	15

Table G-2: Tests performed in Shaw et al. (2007)

Indentor	Site	Speed		Deflection	PMHS				
		Quasi-static	Dynamic		1	2	3	4	5
2x2	Lower		x	Low	x	x	x	x	x
	Mid		x	Low	x	x	x	x	x
	Upper		x	Low	x	x	x	x	x
	Lower	x		Low		x	x	x	x
	Mid	x		Low		x	x	x	x
	Upper	x		Low		x	x	x	x
2x4	Lower		x	Low	x				
	Mid		x	Low	x				
	Upper		x	Low	x				
	Lower		x	To failure	x				
	Mid			To failure				x	x
	Upper		x	To failure		x	x		
2x4 (-15º)	Lower		x	Low	x				
2x4 (15º)	Lower		x	Low	x				

G.1.2 Normalization of data

The results were normalized to the "standard anthropometry" of the 50th male (standard mass 75 kg) using the process of Eppinger et al. (1984). This process creates a variable scale λ which is based on comparison of mass between the standard mass with respect to the mass of the subject:

$$\lambda = \left({}^{75}\!/_{M_i} \right)^{1/3}$$

Data obtained in tests were scaled with this equation for length and loads signals. “s” is the scaling, and “i”, the initial one.

Length: $L_s = \lambda \times L_i$

Force: $F_s = \lambda^2 \times F_i$

The graphs contain the L_s values, although the λ parameter is practically 1 and there were no significantly difference between L_s and L_i.

For the quasi-static tests, force-deflection values were mass scaled and were filtered with a CFC 1000. In contrast, for the dynamic tests, the force-deflection values were scaled and mass compensated, and then filtered with a CFC 180. Deflection data was filtered with a CFC 1000.

The load cell Y and Z axis have not got a large influence in the resultant force (the average resultant force was 6% higher than the X), so results reflect only the X axis force.

Given that the marks were mounted 9 mm above the target measuring point, the deformation and rotation of the bones during the tests were taken into account and they were calculated on the data obtained by the Vicon cameras using the finite element model human, H-model (Handbook of Numerical Analysis, 2004). All the data obtained were presented on three-dimensional graphic with relative values with respect to the depth of the indentor (maximum depth in any case). Some Vicon markers could not be measured because it was necessary to position the indentor, or the cameras lost the position of these markers.

G.1.3 Injury data

The autopsy found the following fractures:

Table G-3: Fractures of the subjects after the tets

Fracture type	Subject 1	2	3	4	5	Total
Cartilage	4	1	1	2	2	10
Mono-cortical		3		9		12
Bi-cortical non displaced	11	4			10	25
Bi-cortical displaced		2	2			4
Other			1	6	3	10
Total	15	10	4	17	15	61

Next, a briefly description of the fractures is shown:

- Cartilage: Cartilage fractures.
- Mono-cortical: Disruption of the outer or inner rib surface.
- Bi-cortical Non Displaced: Disruption of both the outer and inner surfaces of the rib.
- Bi-cortical Displaced: Complete separation of the rib.
- Other: Disruption of the sterno-costal joint and incomplete rib fractures in which there is no cortical disruption but damage internal to the rib.

G.1.4 Force – Deflection data retrieval

The values have been obtained directly from Shaw et al 2007 due to the raw data of these configurations are not available by the authors of this appendix. Therefore the graphs have been digitalized to obtain a series of points. Of course, the quality of the data depends on the quality of the graphs represented in the paper.

A noteworthy aspect is the process taken to identify each of the tests. Shaw et al 2007 shows two types of force – displacement response: the same subject with different locations and the same location of the indentor loading with different surrogates. These graphs have not uniquely identified each curve. The two types of graphs were overlapped to identify correctly each of the curves to later make the digitalization process

G.1.5 Response corridor development

Once identified and digitized curves has been done the mean and standard deviation for each of the locations tested. Due to the maximum stroke of the tests are different, the mean curve has abrupt jumps when each test reach its maximum deflection registered. As well as the mean curve, the standard deviation can produce very wide corridors and have abrupt jumps. These changes of level only occur when a test reaches its maximum displacement and does not intervene in the calculation of the mean and standard deviation. To identify these features, it has been identified with a colour code which tests were used for the calculation of the corridor along the measured displacement.

G.2 Dummy requirements

G.2.1 Force –deflection data

Below are the graphs of force – deflection for each configuration with the following clarifications:

- S* corresponds with the surrogate tested (from S1 to S5).
- The mean curve in its final section coincides with the largest displacement test (there is only one test to make the mean curve).
- In the corridor, the legend for each section shows the number of subjects used to calculate the corridor. The last section, marked in white colour, indicates the test with greater displacement.

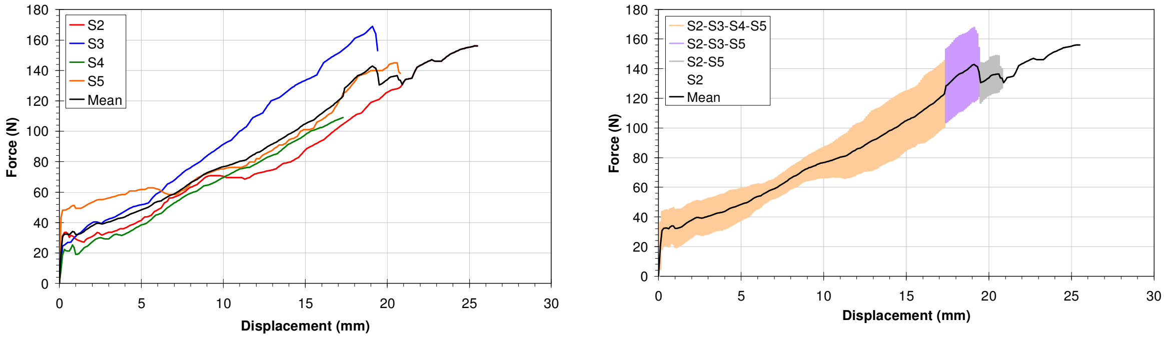


Figure G-2: Upper quasistatic. Left: all subjects. Right: mean and corridors

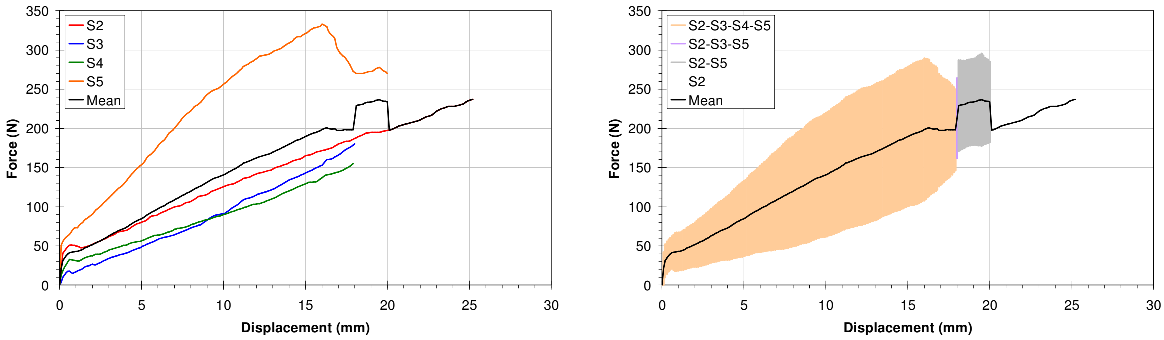


Figure G-3: Mid quasistatic. Left: all subjects. Right: mean and corridors

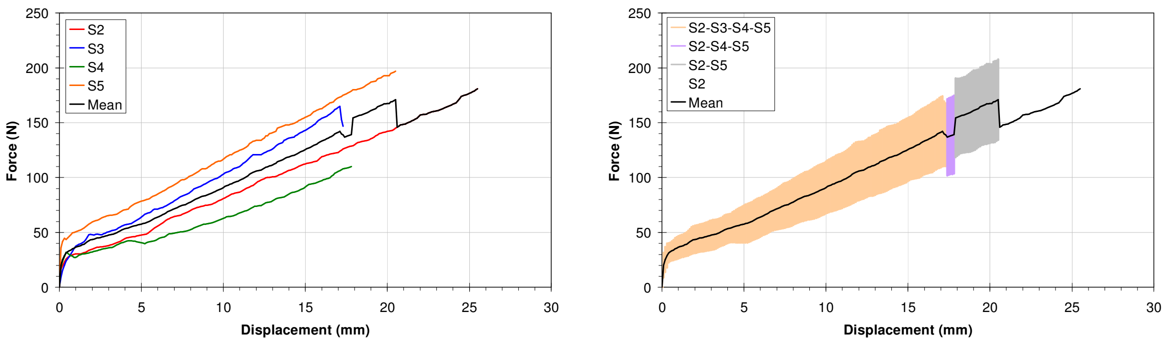


Figure G-4: Lower quasistatic. Left: all subjects. Right: mean and corridors

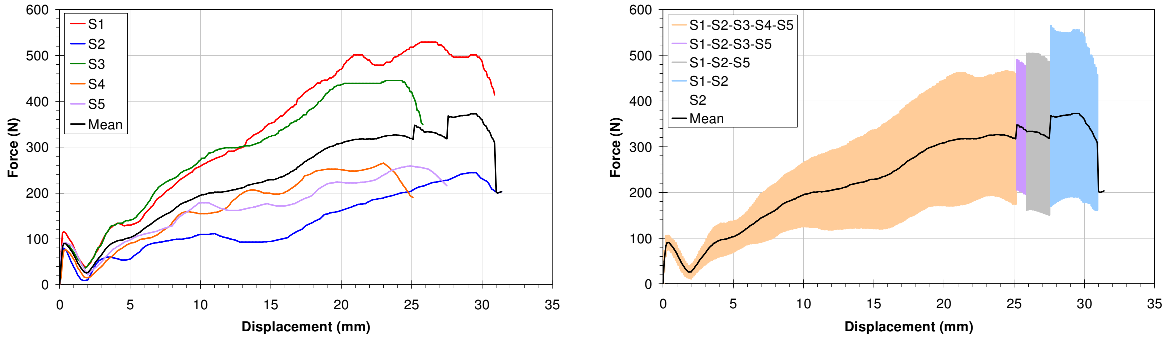


Figure G-5: Upper dynamic. Left: all subjects. Right: mean and corridors

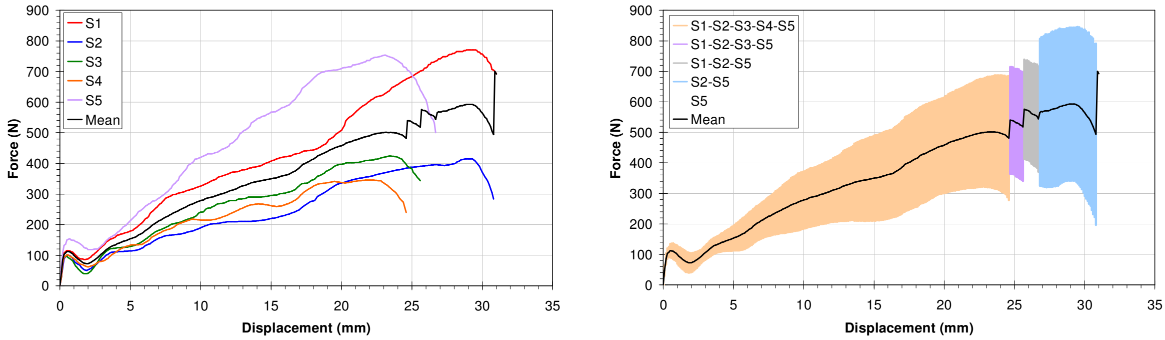


Figure G-6: Mid dynamic. Left: all subjects. Right: mean and corridors

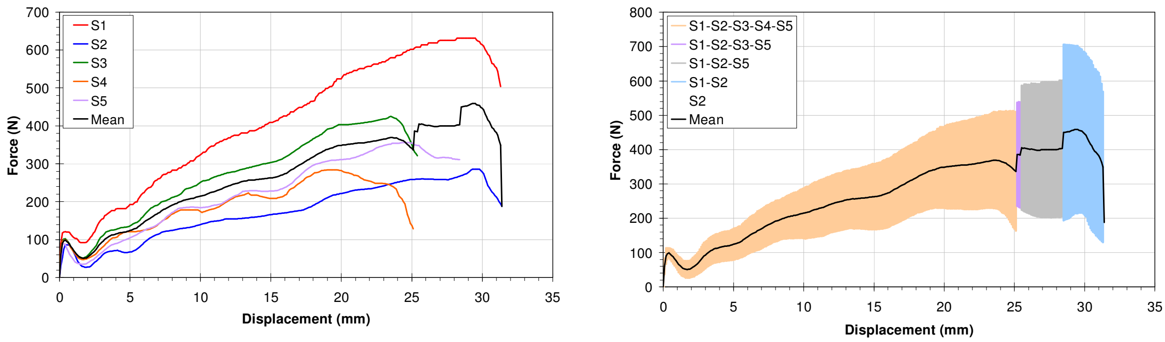
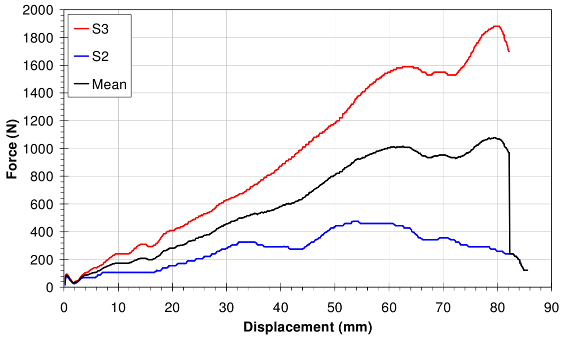


Figure G-7: Lower dynamic. Left: all subjects. Right: mean and corridors

The dynamic to failure tests data are scarce and have scattered values (only one subject in the lower position, and two subjects in the mid and upper positions). Moreover, these tests are considered a very severe for the THOR dummy (these tests reach up to 85 mm of chest deflection). These data with their corridors are shown for future research and shouldn't be included for the biofidelity requirements. In case of the need to be performed these tests, the approval of the consortium will be needed.



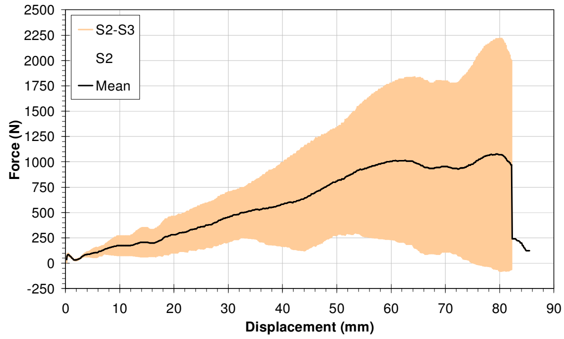
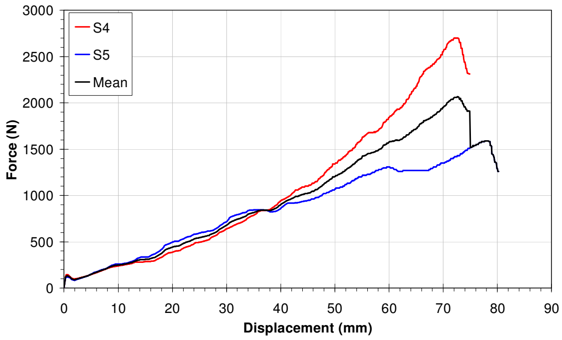


Figure G-8: Upper dynamic to failure. Left: all subjects. Right: mean and corridors



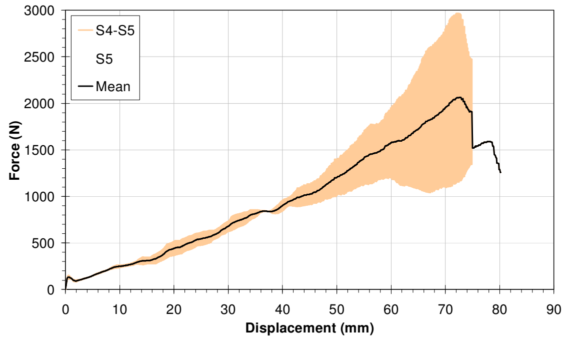


Figure G-9: Mid dynamic to failure. Left: all subjects. Right: mean and corridors

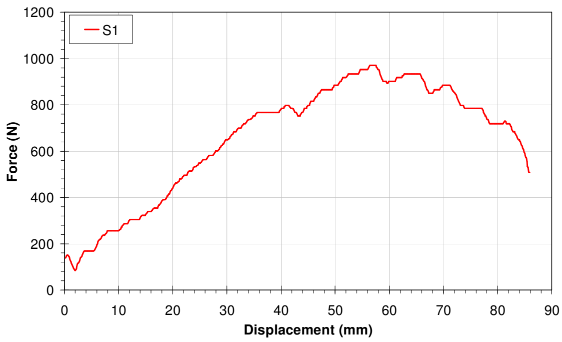


Figure G-10: Low dynamic to failure.

The authors of this appendix have performed an analysis of indentor size and orientation effects and concluded that there is not too much difference in the force-deflection response (conclusion also obtained by Shaw). Therefore, these tests will be not part of the biofidelity criteria (due to their influence is limited and there are not too many tests).

The 2x4 inches indentor was used only in the dynamic test to failure and for checking the orientation effects (± 15 degrees). These tests haven't got defined biofidelity requirements for the THOR dummy (dynamic test to failure are too severe for the THOR dummy and the orientation and size effects haven't got too many influence).

G.2.2 Stiffness data

The chest stiffness has been obtained at different locations from the force – displacement data. Stiffness is reported in N/mm and was calculated by dividing the recorded indentor force at 15mm by the 15mm deflection. Stiffness has been calculated for both quasi-static tests, and non-injury dynamic tests.

In the case of quasi-static tests, the stiffness has been taken from the subjects 2 to 5 due to this test configuration was not performed with the subject 1. For the dynamic tests, the stiffness has been calculated from all the subjects (5 surrogates).

Table G-4: Stiffness data (data from Shaw et al 2007)

	Position	Mean	SD
Quasistatic	Upper	7.0	1.3
	Mid	12.8	6.2
	Lower	8.3	2.1
Dynamic	Upper	15.3	7.1
	Mid	23.3	9.2
	Lower	17.5	6.3

To check and verify that the digitized data of the graphs are quite accurate, the mean and the standard deviation have be recalculated using data from Shaw et al 2007 to add a decimal place. The following table shows the data from Shaw et al 2007, the obtained by the authors from digitized graphs, and their difference.

Table G-5: Data validation

		Shaw 2007		Digitalized		Difference	
	Position	Mean	SD	Mean	SD	Mean	SD
Quasistatic	Upper	6.98	1.33	7.00	1.32	-0.02	0.01
	Mid	12.76	6.20	12.64	5.92	0.12	0.28
	Lower	8.30	2.06	8.36	1.93	-0.06	0.13
Dynamic	Upper	15.28	7.14	15.25	7.14	0.03	0.00
	Mid	23.32	9.17	23.35	9.31	-0.03	-0.14
	Lower	17.52	6.27	17.53	6.36	-0.01	-0.09

G.2.3 Coupling data

Shaw et al 2007 shows the coupling recorded in the dynamic tests (which have similar results respect to the quasi-static test) and dynamic tests to failure. These results are represents as follow: the indentor displacement, the maximum, is labelled a "0.0"; a site that recorded deflection would be labelled "1.0". In the tables, the "0" values are marked in red to highlight the indentor position.

The coupling matrix obtained from the dynamic test data are shown below. For these tests (with a maximum stroke of 1 inch), the coupling matrix are calculated at the peak indentor deflection and measurements of the rest of the rest of the markers.

Table G-6: Values for coupling response to the indentor position (dynamic test).

Position	L3	L5	L6	L7-8	S1	S5	R3	R5	R6	R7-8
Upper	0	0.7	0.7	0.8	0.7	0.6	0.9	0.9	0.9	0.9
Mid	0.6	0.5	0.4	0.5	0.8	0	0.7	0.5	0.4	0.5
Lower	1	0.8	0.7	0.8	1	0.5	0.8	0.1	0	0.2

The coupling data results obtained from the dynamic test to failure are represents in the Table G-7. In this case, due to the larger stroke, the coupling response are calculated at three distances are taken. These distances are taken at 20 and 30% of the chest depth and the last measure at the failure deflection. In the next table, as well as mentioned before, the indentor position are marked in red. The average values have a yellow background, unless the average value cannot be calculated (because there is only one subject) that have an orange background.

Table G-7: Values for coupling response, in dynamic tests to failure, to the indentor position.

Position	Subject	Depth	L3	L5	L6	L7-8	S1	S5	R3	R5	R6	R7-8
Lower	Subject 1	20%	0.9	0.8	-	0.6	0.9	0.6	0.6	-	0	-
		30%	0.9	0.8	-	0.7	0.8	0.5	0.5	-	0	-
		Failure	0.9	0.9	-	0.7	0.8	0.6	0.5	-	0	-
Upper	Subject 2	20%	0	0.9	0.8	0.9	0.9	0.8	0.9	0.9	0.9	1
		30%	0	0.9	0.9	0.9	0.9	0.9	0.9	0.9	1	1
		Failure	0	0.9	0.9	1	0.9	0.9	1	1	1	1
	Subject 3	20%	0	0.4	0.6	0.8	0.8	-	0.9	0.9	0.9	1
		30%	0	0.5	0.6	0.8	0.7	-	0.9	0.9	1	1.1
		Failure	0	0.7	0.8	0.9	0.8	-	0.9	1	1	1.1
	Mean 2-3	20%	0	0.65	0.7	0.85	0.85	0.8	0.9	0.9	0.9	1
		30%	0	0.7	0.75	0.85	0.8	0.9	0.9	0.9	1	1.05
		Failure	0	0.8	0.85	0.95	0.85	0.9	0.95	1	1	1.05
Mid	Subject 4	20%	0.5	0.7	0.4	0.6	0.8	0	0.7	0.6	0.3	0.5
		30%	0.4	0.6	0.3	0.5	0.8	0	0.6	0.5	0.3	0.5
		Failure	0.3	0.6	0.4	-	0.8	0	0.6	0.5	0.3	-
	Subject 5	20%	0.6	0.5	0.4	0.6	0.4	0	0.5	0.6	0.4	0.5
		30%	0.6	0.6	0.4	0.5	-	0	0.4	0.6	0.4	0.6
		Failure	0.6	0.7	0.4	0.5	-	0	0.3	0.6	0.3	0.6
	Mean 4-5	20%	0.55	0.6	0.4	0.6	0.6	0	0.6	0.6	0.35	0.5
		30%	0.5	0.6	0.35	0.5	0.8	0	0.5	0.55	0.35	0.55
		Failure	0.45	0.65	0.4	0.5	0.8	0	0.45	0.55	0.3	0.6

G.2.4 Recommendations for replication of these tests

The test should preferably be performed without the dummy jacket.

The spine should be anchored rigidly to the foundation; the aim should be to immobilization of the spine without immobilizing the ribs.

The indentors should be of 62 x 62 mm (and 62 x 113 mm if it is necessary reproduce these tests), with a 3 mm thick natural rubber sheet with dimensions of 58 x 58 mm (and 58 x 105 mm). Above this plate, there should be a spherical joint that allows the rotation about the X axis (indentor direction) and an inclination to avoid adaptation to the chest.

To measure the forces on the indentor, a load cell located just above the ball joint should be used.

The points to be measured shall be the equivalent to those studied in Shaw et al. (2007).

Appendix H **Cesari and Bouquet (1990/94) / Riordain *et al.* (1991)**

Recommended Engineering Guideline:
Relative chest deflection
Table-top tests

As described above, L'Abbé *et al.* (1982) reported on a series of tests that included dynamic and static seatbelt loading tests to examine the thoracic deflection characteristics of human volunteers. Further table-top PMHS tests were undertaken by Cesari and Bouquet, 1990, Riordain *et al.*, 1991 and Cesari and Bouquet, 1994, based on the test set-up originally used by L'Abbé *et al.*, 1982. In each case, the human subjects were lying supine on a rigid table with the legs in a sitting position. They were loaded by a diagonal seatbelt passing from the left clavicle down to the lower right ribs. The belt was centred on the sternum and was at an angle of 36° to the mid-sagittal plane. Vertical chest compression was measured at 10 points on the chest, including a mid-clavicle position, and lateral chest compression was measured at one point. Dynamic chest loading was applied through an impact mechanism that had a pre-load and a pendulum or impactor striker.

The L'Abbé volunteer tests are not considered here for biofidelity requirements because the test severity was very low and the data is not available in a suitable form for defining biofidelity requirements. The PMHS tests identified in Cesari and Bouquet, 1990, Riordain *et al.*, 1991 and Cesari and Bouquet, 1994 included tests at three loading configurations:

- Low-velocity, low mass (~3 $m.s^{-1}$ and 22.4 kg)
- Low-velocity, high mass (~3 $m.s^{-1}$ and 76.1 kg)
- High-velocity, low mass (~7-9 $m.s^{-1}$ and 22.4 kg)

In total 34 tests were undertaken with PMHS. Replication of the relevant tests is intended to be used to assess the dummy deformation pattern to diagonal belt loading. Therefore, these data have been assessed and a sub-set has been used to define relative compression requirements for each vertical compression measurement relative to the compression at the mid-sternum.

H.1 Review of the Available Test Data

IFSTTAR have provided the THORAX project with the data from the PMHS test series' in electronic form. The test database includes multiple tests with some subjects, but these have all been excluded except for two subjects who received low-velocity, low-mass loading who did not sustain any fractures in the first impact. This leaves the dataset shown in Table H-1 to Table H-3.

Table H-1 Test matrix for low-speed, low-mass experiments conducted by INRETS (Cesari and Bouquet [1990] (first three tests) and Cesari and Bouquet [1994] (last four tests))

Test	PMHS No.	Test number with the same PMHS	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
THC61	K	Assumed No 1	1 Diagonal belt	M	72	53	1,83	16	0
THC64	L*	2		M	71	41	1,70	14	0
THC68	M*	2		M	40	56	1,83	17	0
THC76	Q	1		F	64	49	1,64	18	0
THC78	R	1		M	43	54	1,86	16	0
THC90	S	1		M	67	67	1,80	21	0
THC92	T	1		M	63	56	1,76	18	0

* The tested PMHS had been exposed to a low-speed and low-mass impact before this test.

Table H-2 Test matrix for high-speed, low-mass experiments conducted by INRETS (Cesari and Bouquet, 1990)

Test	PMHS No.	Test number with the same PMHS	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
THC11	A	1	1 Diagonal belt	F	47	92,5	1,70	32	8
THC12	B	1		F	17	58,5	1,64	22	0
THC13	C	Assumed No 1		F	86	43	1,60	17	17
THC14	D	1		M	69	82	1,73	27	16
THC15	E	1		M	60	69	1,77	22	3
THC16	F	1		M	59	62	1,70	21	4
THC17	G	1		M	71	75	1,77	24	7

Table H-3 Test matrix for low-speed, high-mass experiments conducted by INRETS (Cesari and Bouquet, 1990 (first six tests) and Cesari and Bouquet, 1994 (last five tests))

Test	PMHS No.	Test number with the same PMHS	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF
THC18	H	1	1 Diagonal belt	M	67	47	1,74	16	6
THC19	I	1		F	83	43	1,55	18	4
THC20	J	1		M	70	63	1,60	25	18

A series of statistical tests were undertaken in order to determine whether:

- The data from the three test conditions can be combined in to a single set of requirements
- The subjects with BMI outside the acceptable BMI range of 17-27 defined in section 4.2 have a different response to those subjects within the acceptable BMI range, or whether consideration could be given to combining the data from all of the subjects.

Question: Are the relative deformation of the {low-velocity, high-mass}, {high-velocity, low-mass} and {low-velocity, low-mass} experiments significantly different?

A repeated measures ANOVA was used because more than one measurement has been taken from each subject. A standard ANOVA is not applicable in these cases as one of the main assumptions is violated: the outcomes are not independent.

Variables included:

- Within factors (repeated): Chest compression measurement location
- Between factors: Impactor mass; Impactor speed; Subject sex
- Covariates: BMI; Subject stature (m); Subject age; Subject mass (kg)

Conclusions:
The results of the ANOVA show that the mass of the pendulum does not affect the relative deformation outcome. However, the velocity of the pendulum has a highly significant effect ($p<0.001$). Experiments with a high velocity result in significantly lower relative deformation. The effect of the subject covariates age, stature, BMI and mass are statistically not significant at the standard 5% level.

Question: In the low-velocity, low-mass experiment do the relative deformations vary by BMI group?

Conclusions:
Across all parts, an overall ANOVA shows that there is no significant difference between relative deformation for the two BMI groups. Individual t-tests show that there is insufficient data to be able to detect significant differences between the two BMI groups for any individual part.

Given that impactor mass did not affect the relative deformation outcome, consideration could be given to combining the low-velocity, low-mass and the low-velocity, high-mass data sets. The influence of BMI for this combined group was assessed.

Question: In the low-velocity, low-mass and low-velocity, high-mass experiments do the relative deformations vary by BMI group?

Conclusions:
Over all points the ANOVA shows that the difference of relative deformation for the two BMI groups is approaching significance ($p<0.10$).

The velocity for the high-velocity tests exceeds that defined in the inclusion criteria in Section 4.2, so these tests will not be considered further here. The statistical analysis showed clearly that the low and high mass, low-velocity data could be combined. It also showed that the BMI groups could be combined, although this result was less conclusive. Therefore the requirements for all subjects and for only those subjects with suitable BMI will be given.

H.2 Description of the Test Set-up

Schematics of the test set-up used for volunteer tests by L'Abbé at al., (1982), and for PMHS tests by Cesari and Bouquet are shown in those papers. In all cases the subjects were lying supine on a rigid table with the legs in a sitting position.

The subjects were loaded by a diagonal seatbelt passing from the left clavicle down to the lower right ribs. The seat-belt should be routed through the table top around low-friction guides. Figure H-1 shows the low-friction guides for the belt (orange circles). Though the distance between these guides is not known, it was inferred from trial set-ups with Hybrid dummies to be about 600 mm at the level of the table surface. The angle with respect to the dummy orientation is specified as being 36°. Unlike Figure H-1, the set-up should avoid having a belt buckle close to either of the belt guides.

Figure H-1 also indicates that there is no need with dummy tests to include a box under the legs. Instead the legs can be removed.

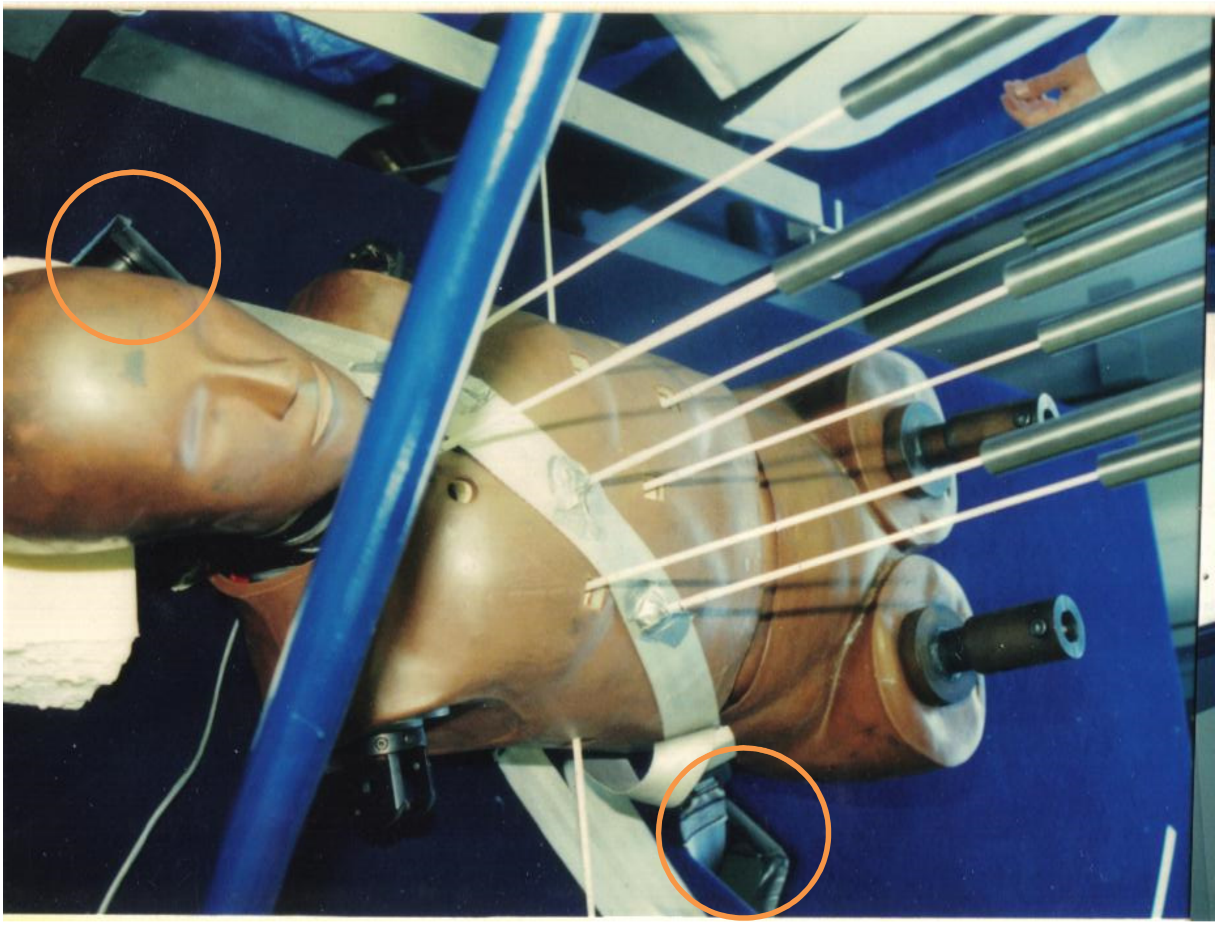


Figure H-1: Test set-up from a previous installation of the apparatus, showing belt guides through the table circled in orange

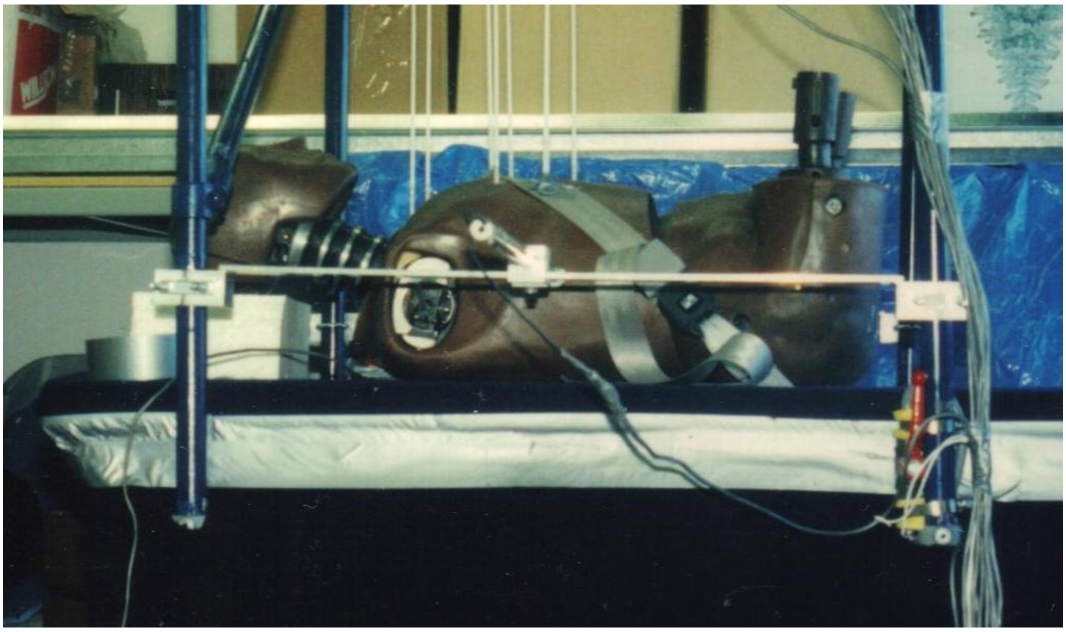


Figure H-2: Test set-up from a previous installation of the apparatus, showing belt lie relative to the dummy torso

The Hybrid III dummy requires padding under the head and upper thoracic spine in order to achieve a fully-upright (or fully supine) position. This support is defined for the Hybrid III dummy on the UNECE web site at the link below.
http://www.unece.org/trans/doc/2009/wp29/Part-572-Subpart-B-Texts-572.5-572.11.pdf
The spacers provide 43.2 mm support behind the head and 6.35 mm behind the spine box.

However, in this test configuration, the dummy jacket is to be worn. This will add depth to the torso but not to the pelvis. As a result, spacers will be needed under the pelvis and head instead of the spine box and head. The size of spacer has been determined to keep the 3° reference angle at the rear of the dummy. Blocks of 24 and 50 mm should be used under the pelvis and head, respectively.

Equivalent spacers will be needed with the THOR. In this case they should provide 12 mm under the pelvis, 5 to 8 mm over a 30 mm wedge under the thorax back plate and 43 mm under the head.

The level of pre-load on the pulling cable is undefined in the reference sources. Therefore, any reasonable implementation of some pre-load to remove slack from the belt system can be considered acceptable.

H.3 Loading mechanism

The two ends of the seat belt passed through the table over low friction supports and were attached to a horizontal rod. This rod/bar was pulled down by a cable, around a system of pulleys, to a suspended catcher plate. The movement of the rod was activated by a dynamic impactor. The force at each end of the belt was measured with a load cell before connection to the rod.

Dynamic belt loading was provided by a 22.4 or 76.1 kg pendulum impactor, dropped from heights up to 40 cm.

The impactor consisted of a rigid tube which was propelled by rubber springs into a plate attached to the cable pulling on the belt. According to Riordain *et al.* (1991), the impactor was in free-flight momentarily before contacting the load transfer plate. It was during this brief period that the impact speed was measured.

H.4 Measurement sites

Chest compression was originally measured with long LVDTs (linear variable differential transformers) mounted to a frame above the subject. These were attached directly to the ribs or clavicle, or to the seat-belt if this masked the attachment point. Therefore points 1, 6, and 8 (see Figure H-5) should be belt-attached, while all the other points should be attached directly to the dummy. This will ensure that the belt-attached measurements include the influence of the torso flesh as represented by the bib and suit on the dummy, while the rib attachment points will be directly reproduced. Any instrumentation that provides measurements equivalent to those made with the LVDTs may be considered suitable. It should be noted that the position of a dummy's ribs are unlikely to match exactly those of the standard human subject. Therefore, care should be taken to ensure that the measurement points on the dummy are as close as possible to those defined for the human.

L'Abbé *et al.* (1982) noted that the measurement points included the *'centres of the right and left clavicles'* and the *'3rd, 5th and 7th rib bilaterally lined up below the mid-clavicle'*. Also, *'All but the mid-clavicle deflectometer supports were permanently fixed in space to provide a constant deflection measurement angle from one test subject to the next. Clavicle deflectometers were not permanently affixed because of slight variances in subject anatomies.'*

The UMTRI drawings (Schneider et al., 1983) can be used to define the mid-clavicle positions relative to the H-point of a standard human subject, and from these the rib and sternum measurement positions can also be defined. This process is described below:

The UMTRI data were rotated into a horizontal plane, to simulate a subject in a horizontal table-top test rig.

Assuming that the clavicle location is best described by UMTRI points 33 and 62, then, relative to the H-point, the mid-clavicle position is (473.7, ± 95.5, 5.6) mm in the coordinate system (superior-inferior, lateral, anterior-posterior).

Furthermore, assuming the clavicle is level with rib 1, and using the anterior mid-line of rib 10 as the lower edge of the rib cage (roughly) in line with the lateral position of the C&B measurement points; and assuming equal spacing between the ribs along this lateral position, we get measurement points (relative to the H-point, in the plane of the table top / laboratory co-ordinate system) as shown in Table H 3.

Table H-4 Estimated measurement points relative to the H-point of the dummy

	X	Y
Mid clavicle	473.7	95.5
3rd rib	390.5	95.5
5th rib	327.0	95.5
7th rib	263.6	95.5
Upper sternum	453.9	0
Mid sternum	391.6	0
Lower sternum	336.2	0

H.5 Hybrid III tests

As mentioned in the main body of the report, biofidelity data sets have been selected, where possible, that offer Hybrid III test data as well as human subject data. The purpose of this is to offer some validation that the test set-up has been recreated appropriately. These table-top tests have such Hybrid III data available. Therefore, it is expected that any replications of the set-up use the Hybrid III data for assessment of the set-up before the principal biofidelity test work. Knowing that the set-up is similar will then allow comparison with the PMHS and volunteer results from previous versions.

The following figure (Figure H-3) shows the combined belt force results from tests performed at the Biokinetics facility (L'Abbé *et al.*, 1982). Any new set-up should confirm that the input force is about the correct level compared with the previous data.

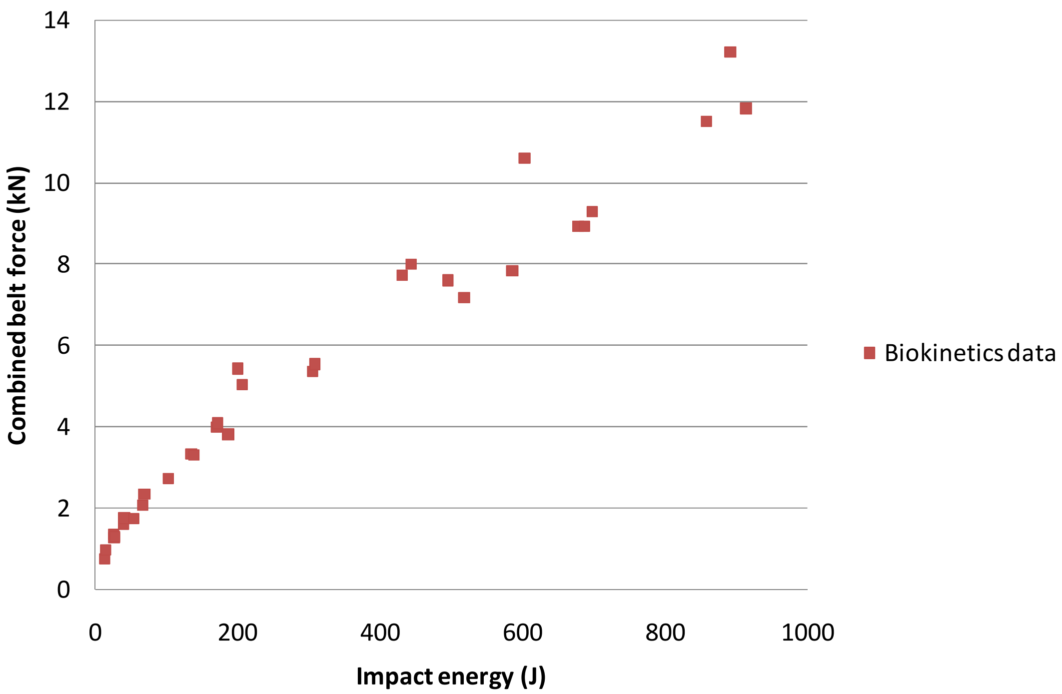


Figure H-3: Previous belt force results from Hybrid III tests at Biokinetics (as described by L'Abbé et al. 1982)

A graph, similar to Figure H-3, is shown in Figure H-4Figure H-4:. This time the y-axis represents the internal chest deflection measurement taken from the Hybrid III dummy. These data should also be used to give additional confidence that the loading of the thorax by the belt is similar when considering the historic set-up compared with any replication.

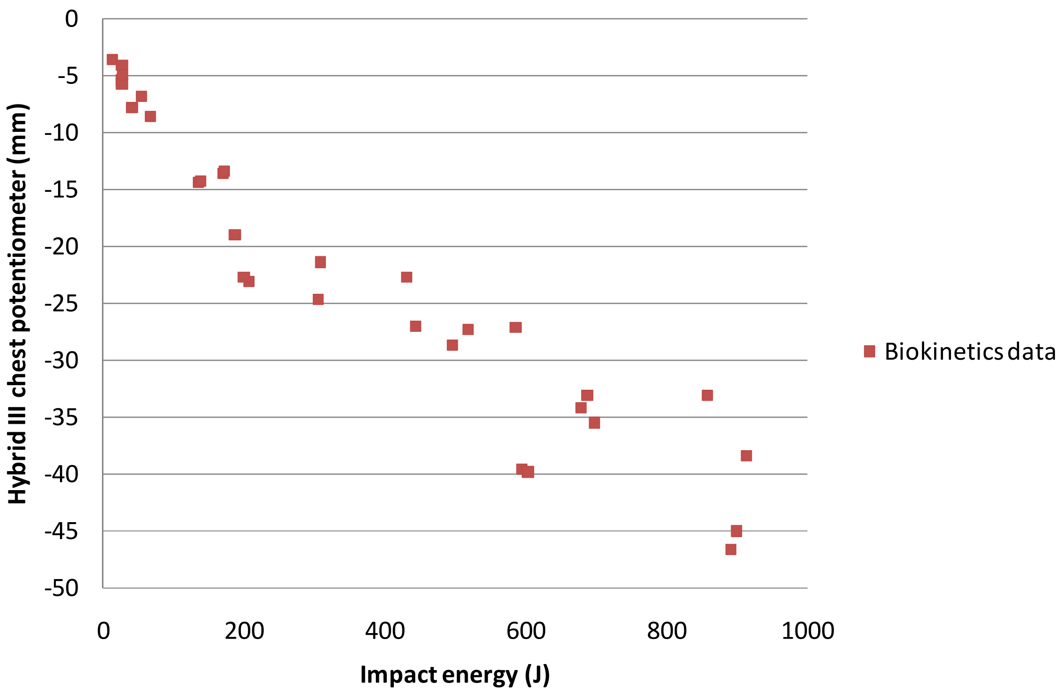


Figure H-4: Internal Hybrid III chest deflection from the previous Biokinetics tests (as described by (L'Abbé et al., 1982)

The following figure (Figure H-6) shows the external measurements taken with the Hybrid III dummy across the thorax. The values depicted are the peak displacement measured with the LVDT at that position from historic tests (L'Abbé at al., 1982 and Cesari and Bouquet 1990)). The order of the points is the same as shown here, in Figure H-5.

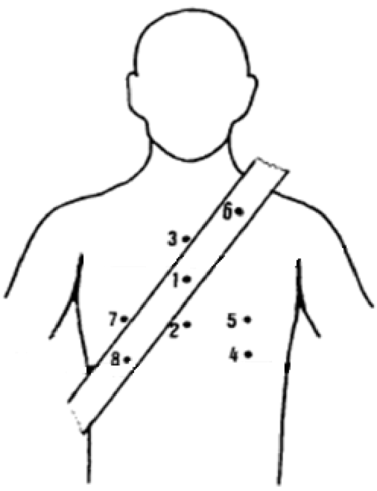


Figure H-5: Numbering system for external chest deflection points

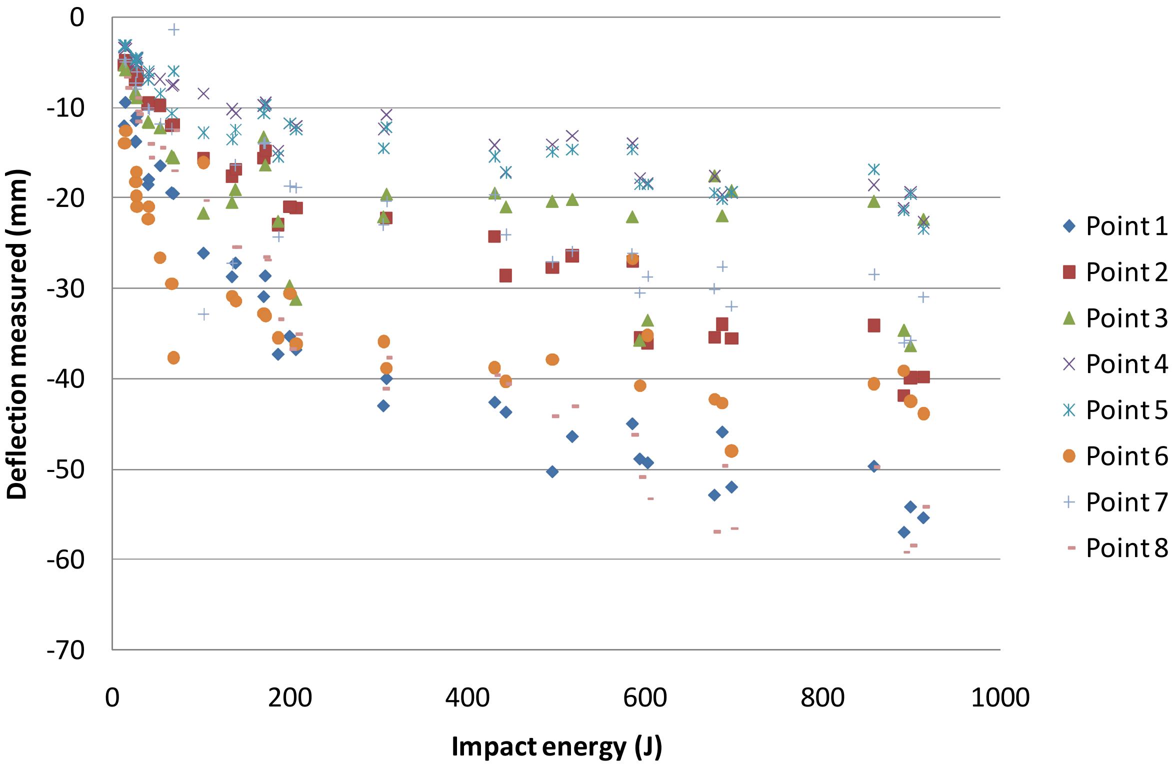


Figure H-6: External Hybrid III chest deflection from previous results of tests (as described by (L'Abbé et al, 1982) and (Cesari and Bouquet 1990)

When related to the mid-sternum measurement, the Hybrid III external measurements are normalised as in Figure H-7. This represents a method whereby the required measurements for the biofidelity assessment can be compared directly with previous results. Tests with the Hybrid III should therefore give confidence that the set-up can indeed be used to assess the thorax biofidelity of a new dummy

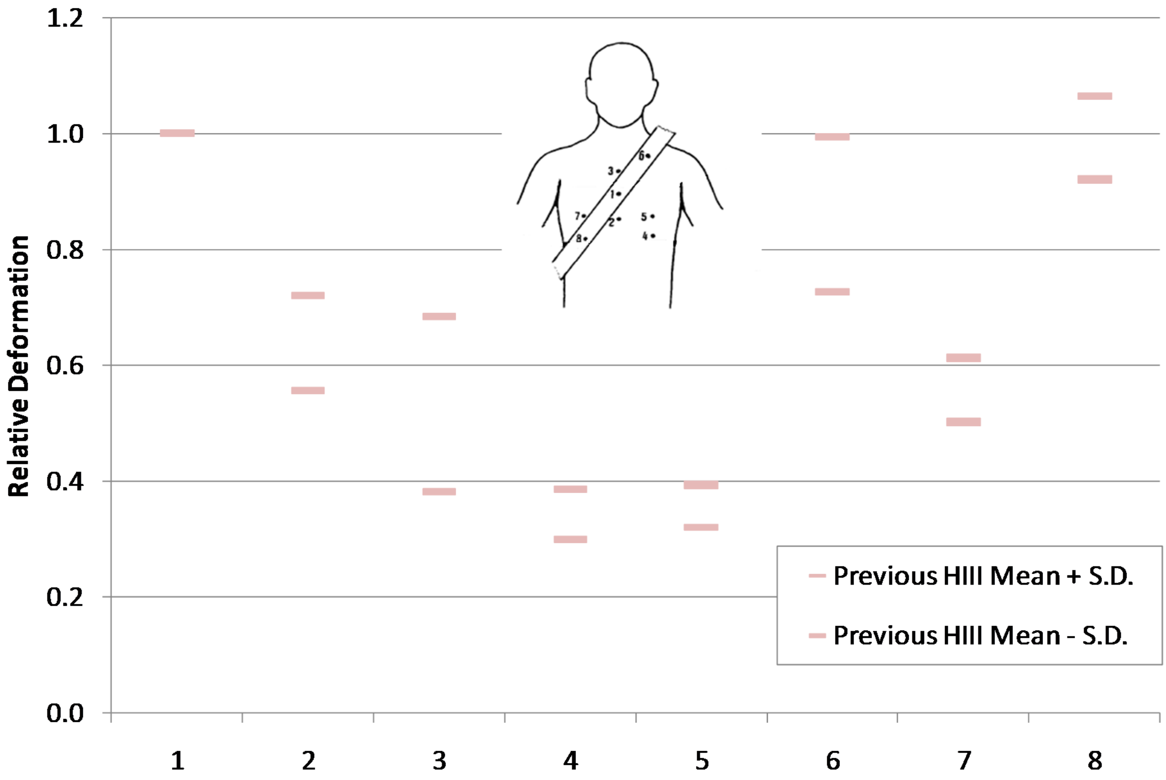


Figure H-7: Comparison of mean peak Hybrid III deflection results relative to the mid-sternum measurement

H.6 Dummy requirements

The normalised chest compression biofidelity requirements are shown in TAB. Given that the normalised compression at each point was not significantly related to subject age, height and mass, the same requirements are defined for all dummy sizes.

Table H-5 Normalised thorax compression biofidelity requirements for Cesari and Bouquet table-top test condition

Right Side		Sternum		Left Side	
Clavicle (11)	0.30 – 0.46	Upper (3)	0.5 – 0.83	Clavicle (6)	0.51 – 0.89
Rib 5 (7)	0.99 – 1.22	Mid (1)	1.0	Rib 5 (5)	0.07 – 0.45
Rib 7 (8)	1.01 – 1.34	Lower (2)	0.84 – 1.18	Rib 7 (4)	-0.06 – 0.32
Rib 8 (10)	-0.45 – 0.14				

H.7 Comparison with GESAC 2005 Biofidelity Requirements

Table H-6 GESAC normalised thorax compression biofidelity requirements for Cesari and Bouquet table-top test condition GESAC, 2005

Right Side		Sternum		Left Side	
Clavicle (11)	0.3 – 0.5	Upper (3)	0.5 – 0.9	Clavicle (6)	0.5 – 0.9
Rib 5 (7)	0.8 – 1.2	Mid (1)	1.0	Rib 5 (5)	0.1 – 0.3
Rib 7 (8)	1.0 – 1.4	Lower (2)	0.8 – 1.1	Rib 7 (4)	0.1 – 0.3
Rib 8 (10)	-0.3 – -0.5				

Appendix I **Kent *et al.* (2004)**

Recommended Engineering Guideline:
Relative chest deflection
Table-top tests

Kent *et al.*, 2004 carried out 67 tests on 15 un-embalmed PMHS lying supine on a rigid bench by either a two-point diagonal belt, a hub, a pair of two-point diagonal belts (in a crossed configuration), or a distributed load in random order (Table A-17). The posterior boundary condition was a rigid flat plate on which the subject was laid. The subject was free to move on the plate and the spinal curvature was not controlled other than by the flat plate interface. Prior to testing the subjects' pulmonary systems were pressurised to typical mean full-inspiration volume immediately prior to testing. The airway remained occluded throughout loading. A high-speed material testing machine applied the load at rate of 1.0 m/s. A load cell measured the reaction force on the PMHS back support. Mid-sternal chest deflection was obtained from string potentiometers attached to the belt, band or hub. All PMHS were tested five times, the first four times up to non-injurious levels with the four different loading cases. The fifth test was injurious and this loading case varied between the specimens.

Kent calculated force-deflection corridors for each load case using whole body mass and modulus scaling factors for the reaction force. The mid-sternal chest deflection and the reaction force at the PMHS back were used to calculate thorax stiffness.

For additional information on test set up please see Kent et al (2004). The results are to be used for engineering, e.g. to assess stiffness differences when different restraints are used to load the chest. The response corridors recommended to be used are presented in Kent et al. (2004).

Table I-7 Subject data for the PMHS included in the experiments by Kent *et al.*, 2004

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	
	176	Multiple test using different loadings conditions	F	85	58	1,57	24	
	182		F	80	65	1,57	26	
	177		F	79	48	1,61	19	
	155a		F	71	54	1,66	20	
	173		F	67	57	1,62	22	
	~~147~~		~~F~~	~~63~~	~~45~~	~~1,61~~	~~17~~	
	186		F	58	61	1,78	19	
	157		F	55	74	1,68	26	
	189		M	79	57	1,59	23	
	190		M	79	73	1,73	24	
	170		M	75	65	1,78	21	
	178		M	73	81	1,82	24	
	~~188~~		~~M~~	~~71~~	~~85~~	~~1,73~~	~~28~~	
	145		M	54	88	1,92	24	
	~~187~~		~~M~~	~~54~~	~~113~~	~~1,78~~	~~36~~	

Appendix J **Rouhana *et al.* (2003)**

Potential Biofidelity Requirement:
Absolute chest deflection
Sled tests

Rouhana *et al*., 2003 carried out sled tests to study the effect of load distribution, by introducing four-point seat-belts, on the number of rib fractures.

J.1 Summary of the PMHS and dummy tests

In these tests eight PMHS were included (Table A-8, Figure J-8) of which six were restrained by a four-point seat-belt system. The seat used was rigid but covered with foam and trim from a production bucket seat, and the locations of the anchorage points were specified in the paper. Some type of force limitation for the upper anchorage point in combination with restricted pelvis was adopted that allowed large torso forward rotations.
Matching Hybrid III and THOR dummy test data is available.

Single-point sternum compression measurements using a trans-thoracic rod technique data may be made available.

The male PMHS in the four-point configuration were all slightly above average weight (although within the range specified for the inclusion criteria), while the two females varied (BMI 18 and 26). In six of the tests (tests 209 to 222 in Table A-8), the lungs were inflated three times prior to testing and partially inflated at the time of the test. No vascular repressurisation was performed.

Table J-8 Test matrix for the tests conducted jointly by UMTRI and Ford (Rouhana *et al.*, 2003)

Test	PMHS No.	Restraint	Gender	Age	Mass (kg)	Stature (m)	BMI	NRF	Delta-v (km/h)
208	206	3pt SB	M	75	72	1,75	24	32	40
209	474		M	72	82	1,78	26	16	40
210	853	4pt FL + PT belt	M	75	81	1,80	25	12	40
217	247		M	41	82	1,75	27	0	40
218	639		M	60	91	1,83	27	3	40
221	683		F	69	42	1,52	18	12	40
222	657		F	79	59	1,52	26	3	40

Figure J-8 Rouhana test setup with a Hybrid III 5F dummy. Top and bottom left; 3pt belt system, bottom right 4-pt belt (photos from the preparation of the tests used in the reproduction of the tests by Rouhana et al. 2003)

J.2 Biofidelity requirements

Thorax requirements are pending at the time this report was compiled. Peak chest displacement deformations are available in Rouhana et al. (2003).

Appendix K **Shaw *et al.* (2009)**

Potential Biofidelity Requirement:
Absolute chest deflection and shoulder kinematics
Sled tests

As described in Appendix A, University of Virginia carried out frontal sled tests with restrained PMHS at 40 km/h. These are referred to as Shaw *et al.*, 2009 and have provided the nickname Gold Standard series 1. Most of the eight PMHSs included in the study were close to 50th percentile male, which makes the test series useful in defining biofidelity requirements for a 50th percentile male dummy.

In brief, the test subjects were restrained on a rigid planar seat by bilateral rigid knee bolsters, pelvic blocks, and a custom three-point diagonal and lap belt. The seat and belt restraint geometry represented, approximately, that of the right front passenger position of a standard US saloon. Neither the pelvis nor the diagonal belt included a retractor. Pelvis and lower extremity motions were blocked by a rigid knee bolster (adjusted to be in contact with the knees at T0), bilateral posterior pelvic blocks and by a footrest with ankle straps.

Test conditions facilitated tracking of spherical markers on the torso by a 16 camera motion tracking system. Video analysis provided three-dimensional trajectories of multiple skeletal sites on the torso relative to the spine. Kinematics of the head, spine and the belted and the unbelted shoulder have been provided. These can be used to check that the general kinematics of the dummy are representative of the tests, and therefore that the input to the thorax from the restraint system is likely to be representative.

A subset of these tests were first reported by Crandall (2008a) and later another subset by Shaw et al. (2009a), chest displacements for all eight PMHs in Shaw et al. (2009b) In addition, information on the test set-up and responses measured were reported in Crandall et al. (2012), Lessley et al. (2012) and in Ash et al. (2012b). Additional analysis of chest displacements and belt loads were carried out by Lebarbé (2011). While the belt loads presented in Ash et al. (2012a) were non-normalized, Lebarbé found that the belt load response corridors were narrower when the data was normalized; for this reason normalized chest displacements and belt loads were included in his report. In this report both normalized and non-normalized data is presented.

K.1 Review of the Available Test Data

University of Virginia has provided the THORAX project with data from eight adult male PMHS, approximating the 50th percentile male anthropometry, sled tests. A single test was carried out with each subject. The tests were carried out in a well-controlled environment that was designed to simulate a full frontal collision and to generate human frontal impact response data for the evaluation and development of human surrogates. In these tests an optically-based motion capture system was used to describe the skeletal motion of the head, acromion, spine and the pelvis of each subject relative to the vehicle buck and multiple point chest compression, i.e. chest relative spine deformations. The dataset is presented in Table **K**-**1**.

Table K-1 Test matrix for the sled experiments conducted by University of Virginia (Shaw et al. 2009).

Test	PMHS No.	Age	Mass (kg)	Stature (m)	BMI	NRF	Cause of Death
1294	411	76	70	1,78	22	7	Pancreatic Cancer
1295	403	47	68	1,77	22	27	Coronary Artery Disease
1358	425	54	79	1,77	25	15	CVA and Atrial Fibulation
1359	426	49	76	1,84	22	9	Lung Cancer
1360	428	57	64	1,75	21	5	Neoplasm of Brain
1378	443	72	81	1,84	24	9	Cancer
1379	433	40	88	1,79	27	10	Cardiovascular Disease
1380	441	37	78	1,80	24	2	Seizure Disorder

* Belt system comprised two separate belt webbings for the lap and the shoulder.

K.2 Description of the Test Set-up

PMHS that were non-ambulant for an extended period prior to death were excluded the study. Similarly, subjects with bony pathology in the thorax as determined from pre-test CT scans were excluded the study. All PMHSs that were selected for testing were preserved by freezing.

After installation of instrumentation and targets for motion capture, the PMHSs were seated on a rigid planar seat and an adjustable matrix of cables supported the torso and head (**Error! Reference source not found.**). A lap and a diagonal belt were joined near the left hip in a location that approximates the position of a stalk-mounted buckle. The diagonal and lap belts were constructed of 48 mm-wide restraint webbing (Narricut, International twill pattern 13195, 6 to 8 % elongation, 6000 lbf (26.7 kN) minimum tensile strength). The belts were replaced after each test. Pelvis and lower extremity movements were restricted by a stiff knee bolster, positioned to be in contact with the lower leg at the location of proximal tibias at the time of impact, and bilateral posterior pelvic blocks that limited any rearward motion of the pelvis, and two footrests. The latter were fitted ankle straps.

Small sled engineering drawings are provided in Figure K-1. Larger drawings are available in the test report (Crandall 2008a).

K.3 Pre-test PMHS posture

The average pre-test PMHS H-point was slightly rear of the H-point as defined for the Hybrid III. The pre-test PMHS sternal angle relative a vertical line ranged from 16 to 24 degrees; average 22 degrees. The upper diagonal belt anchor location was adjusted; the angle from the anchor to the top of the shoulder ranged, measured in the sagittal plane from the upper anchor to the shoulder relative a horizontal line, from 24 to 29 degrees; average 27 degrees.

The belt routing varied slightly between subjects (See Figure Shaw et al. 2009b); the belt angle across the chest relative to the vertical was reported to range from 45 to 55 degrees, the under chin to top of belt on centre line was reported to have been between 95 to 148 mm for THOR NT and THOR SD1 dummies tested in the Gold Standard conditions (Crandall 2008b)

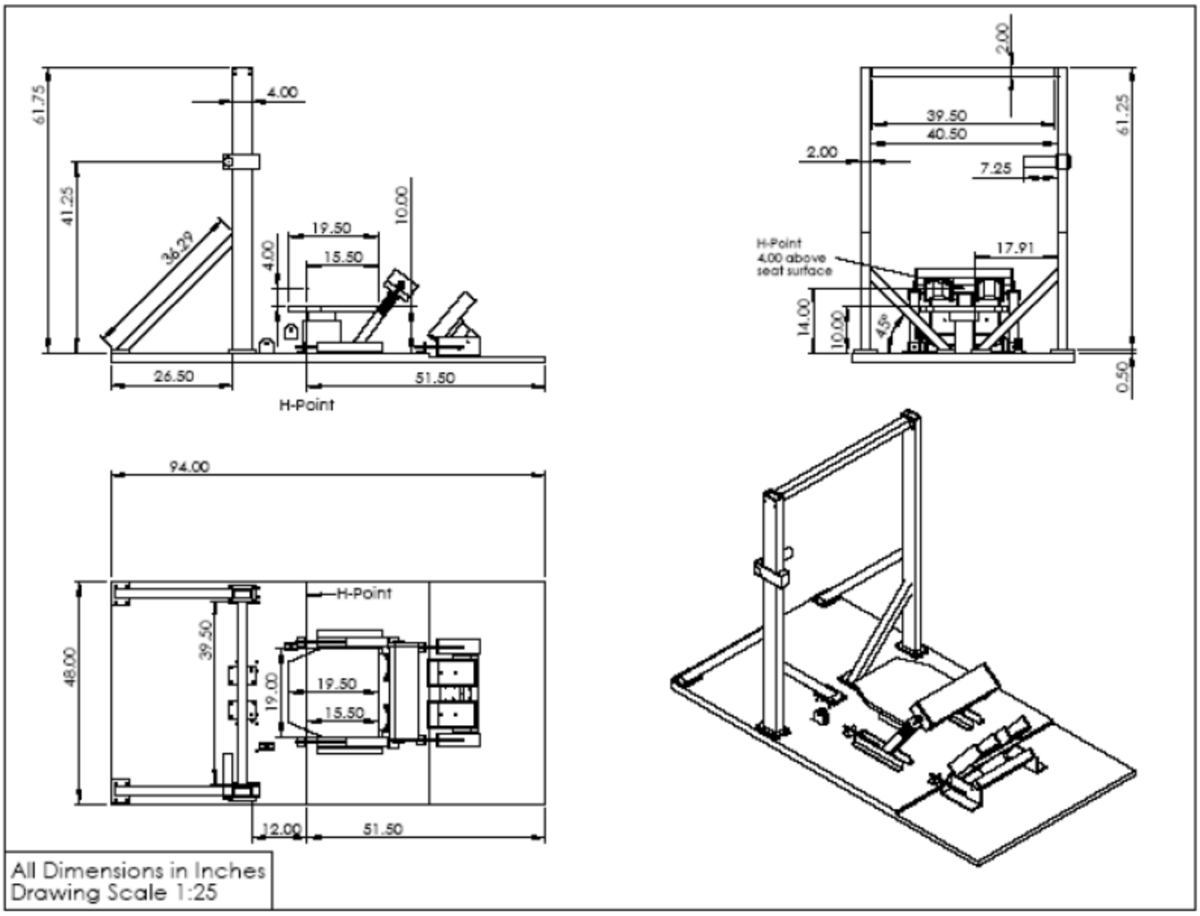


Figure K-1: Sled drawings (Crandall 2008a)

K.4 Measurements

Interaction loads

Instrumentation used in these PMHS tests to measure interaction loads include 6-axis load cells at the seat, left and right knee bolster, and footrest (Figure K-2). After the data was collected the effect of the acceleration of the mass attached to the load cells was removed through the process of inertial compensation (Ash et al. 2012).

The tensions in the upper and lower part of the diagonal belt as well as the lap belt were also recorded. The diagonal belt tension in the upper region was measured near the D-ring while the diagonal belt tension in the lower region was measured near the joining location of the lap and diagonal belt. The lap belt tension was measured near the right anchoring location. Interaction loads selected for inclusion in this requirement report are presented in Table K-2.

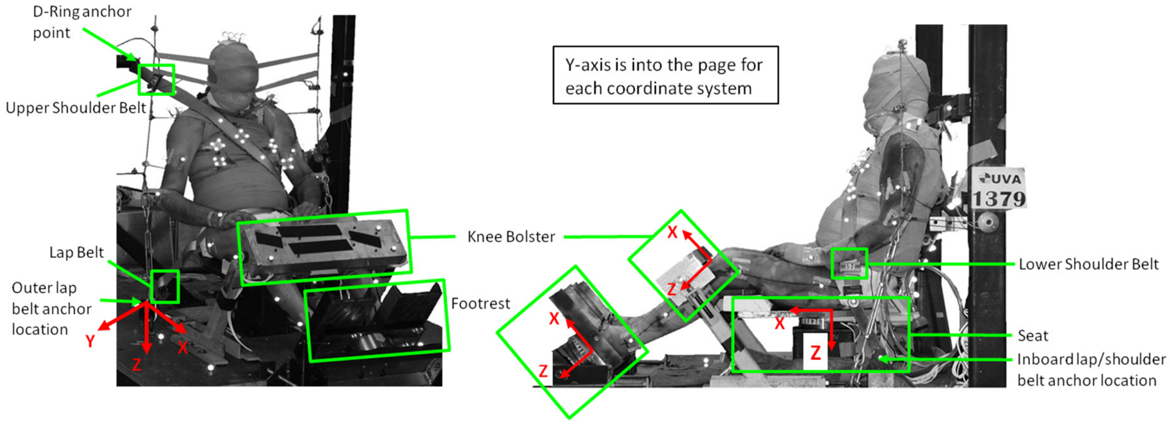


Figure K-2: Schematics of the instrumentation used in the Shaw 2009 (Ash et al. 2012a).

PMHS instrumentation

In the tests several accelerometer and angular rate sensor clusters were attached to the PMHSs. However, instrumentation data are not selected for inclusion in the requirements.

Table K-2: Shaw 2009 subject measurements provided in this biofidelity requirement report.

	Location	Instrument /coordinate system used	Measurement channels included in the requirements
Instrument data	Seat, Right and Left Knee bolster, Foot rest forces	Tri-axial external loads	Fx, Fy, Fz
	Diagonal, upper and lower, and lap seat belt forces	Belt force gauge	
Vicon data	Chest; ribs 4 and 7-8	Relative T8	Dx, Dy, Dz
	Head centre	Relative sled	Dx, Dy, Dz,
	T1, T8, L2 centre	Relative sled	Dx, Dy, Dz,
	Right and left acromion	Relative sled	Dx, Dy, Dz,
	Pelvis centre	Relative sled	Dx, Dy, Dz,
	Head centre	Relative T1	Dx, Dy, Dz, angular disp. around x-, y-, and z-axes
	T1 centre	Relative T8	Angular disp. around y-, and z-axes
	T8 centre	Relative sled	Angular disp. around y-, and z-axes

Data acquisition and processing

All instrument data were collected at 10,000 Hz and hardware-filtered to 3000 Hz. In addition post-processing included filtering to SAE J211-prescribed filter classes. After the data was collected the effect of the acceleration of the mass attached to the load cells was removed through the process of inertial compensation. This was accomplished using a test with no subject conducted in order to record the intertially generated forces and moments acting on the affected load cell coordinate system axes (mass compensation for forces: X-axis for the seat load cell and X and Y-axes for the left and right knee bolster load cells and the footrest load cell; mass compensation for moments: Y-axis for the seat, left and right knee bolster and footrest load cells). These measured forces and moments with no occupant were then subtracted from the tests conducted with the PMHS to remove the forces and moments generated by the acceleration of the testing hardware. In general data from instrument data from tests with eight subjects were available.

The data were truncated to -10 to 250 ms for presentation.

Displacement data
Motions were captured at 1000 Hz using a optoelectric stereophotogrammetric system that consisted of 16 Vicon MX™ cameras. Data were processed to provide 6D displacements of the underlying anatomical structures. The head marker cluster motion was transformed to the head centre. The spine marker cluster motions were transformed to the representative vertebra bodies. The shoulder marker cluster motions were transformed to the right and left acromion. The pelvis marker cluster motion was transformed to the pelvis centre. A separate study was carried out to relate these anatomical structures to structures in the THOR NT dummy; for additional detail on this please see Parent et al. (2012). Both displacements relative the sled and intra-segmental kinematics of the spine were provided (Shaw et al. 2009a, Shaw et al. 2009b and Lessley 2012). For the displacement relative the sled data from eight subjects were made available while for intra-segmental kinematics data from three subjects were made available. Selected displacement data used in these test to capture head, spine, pelvis and acromion displacements are presented in Table K-2.

Coordinate systems
All data, for which there is no special description, are provided in SAE coordinate system (positive axes: x forward, y to the right, z down.

Development of response requirements
Digital response corridors were provided the THORAX project. These corridors were developed using average and one standard deviation.

K.5 Sled type and sled acceleration

Sled acceleration corridors from the eight PMHS tests (Shaw et al. 2009b) are shown in Figure K-3.

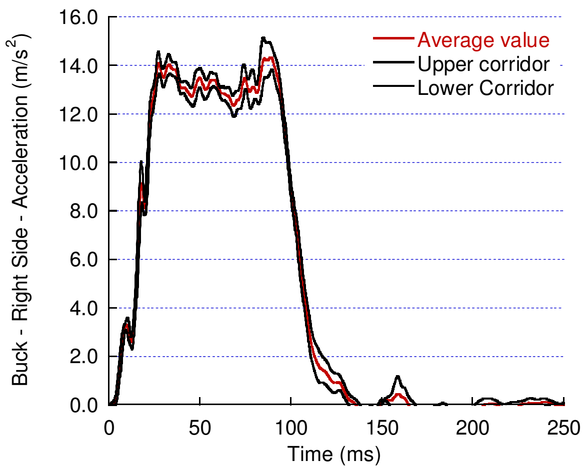


Figure K-3 Shaw 2009 40 km/h sled pulse

K.6 Normalization

The original PMHS data were not scaled for PMHS size or age. The chest deformations relative to T8 were normalized by Lebarbé (2011) for the requirements for ACEA/ISO.

K.7 Dummy requirements

The biofidelity requirements for the thorax are being developed by the ACEA/ISO Frontal Impact Biofidelity Task Force; draft requirements are added to this report. These requirements include belt forces and multiple chest deformations relative to T8. In addition to the requirements developed by the ACEA/ISO, head, T1,T8, L2, L4, pelvis and acromion displacements have been provided by the Center for Applied Biomechanics UVa through various publications (Ash et al. 2012b and Crandall et al. 2012).

The ACEA/ISO requirements were scaled using a method presented by Lebarbé (2011). In this report scaled chest deformations relative to T8 were included whereas all other data were included non-scaled. The effects of scaling on the chest deformations (average value and corridors) were however microscopic; all Shaw 2009 data can be considered to be non-scaled. These requirements are defined in Figure K-4 to K-17.

Included for assessment of the reproducibility of the original tests are foot rest forces, knee bolster forces and moments, and seat forces (Ash et al. 2012a), see Figure K-18-20.

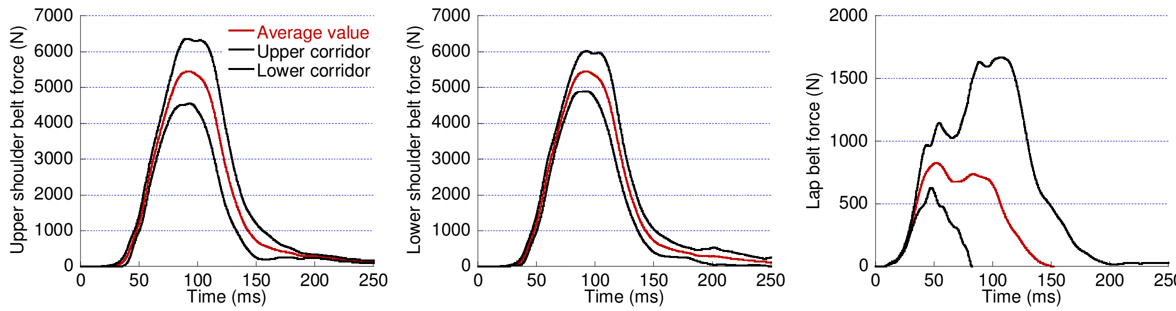


Figure K-4: Belt forces

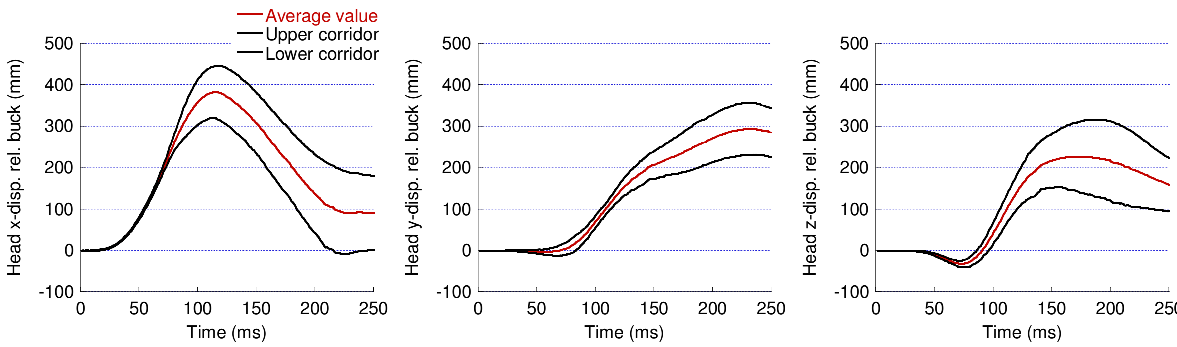


Figure K-5: Head displacement in the x-, y-, and z-axes

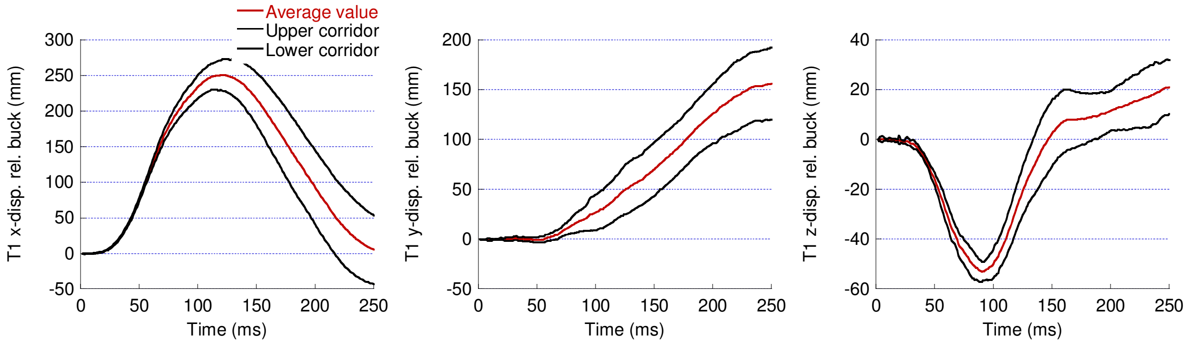


Figure K-6: T1 displacement in the x-, y-, and z-axes

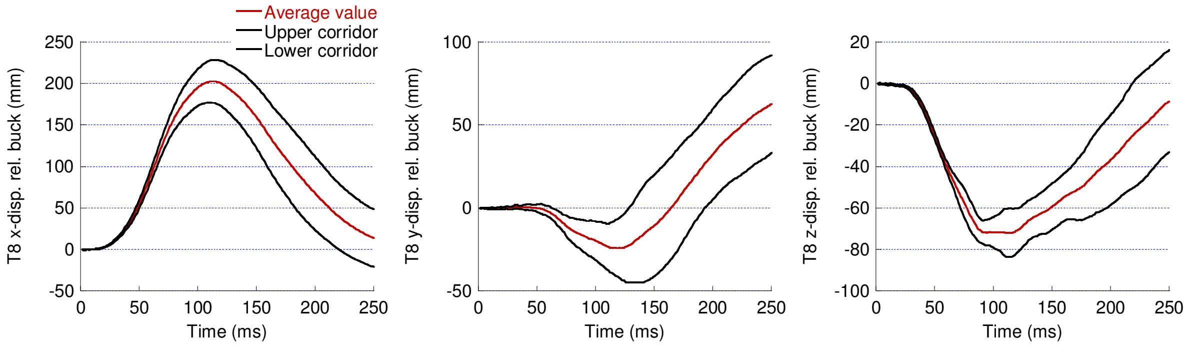


Figure K-7: T8 displacement in the x-, y-, and z-axes

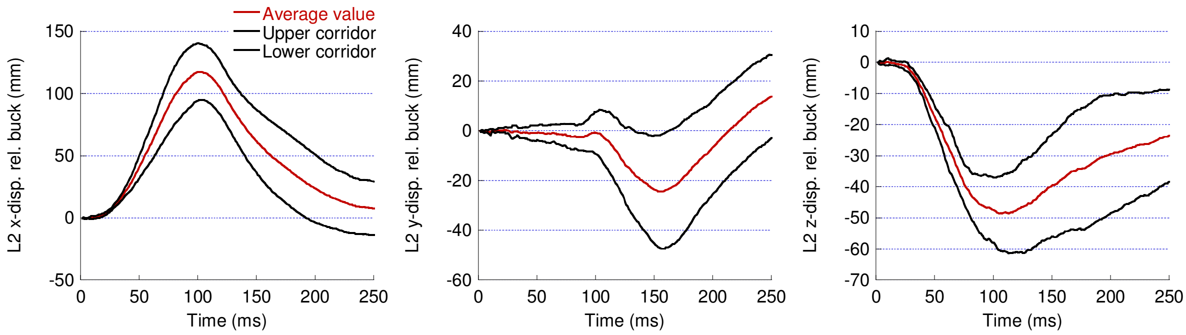


Figure K-8: L2 displacement in the x-, y-, and z-axes

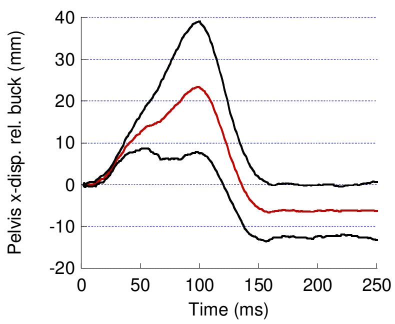


Figure K-9: Pelvis displacement in the x-axes

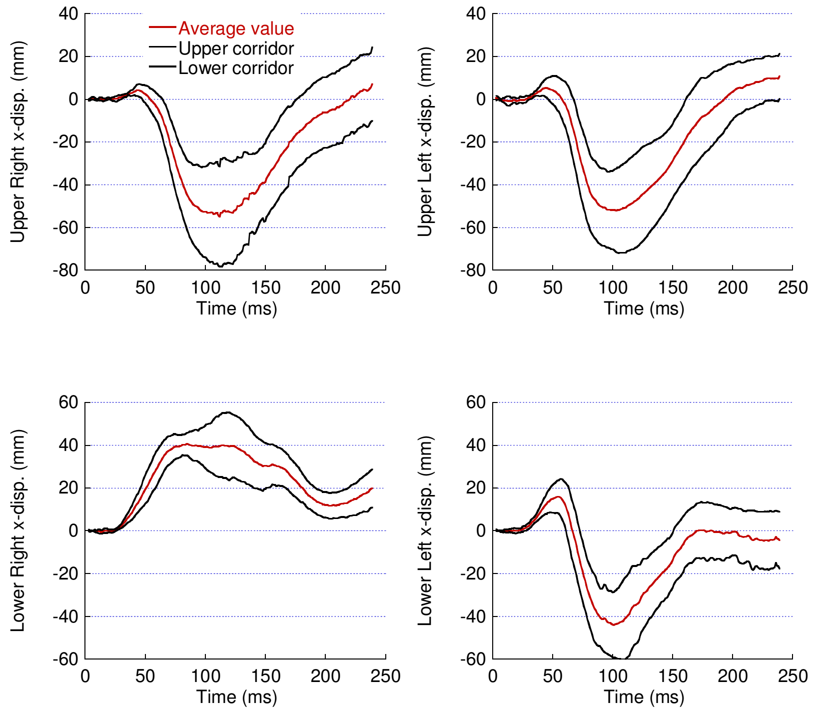


Figure K-10: Chest displacement in the x-axes

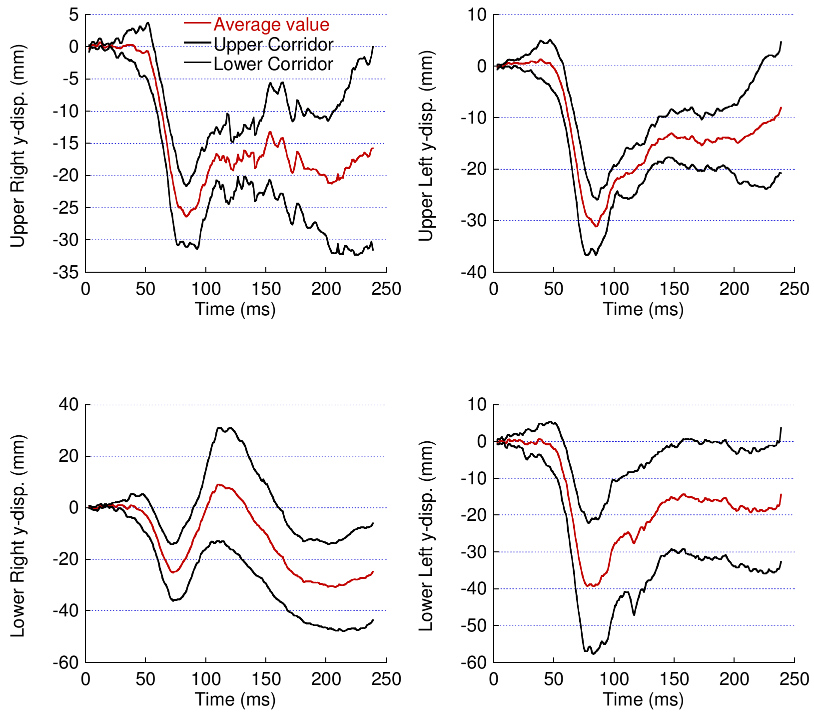


Figure K-11: Chest displacement in the y -axes

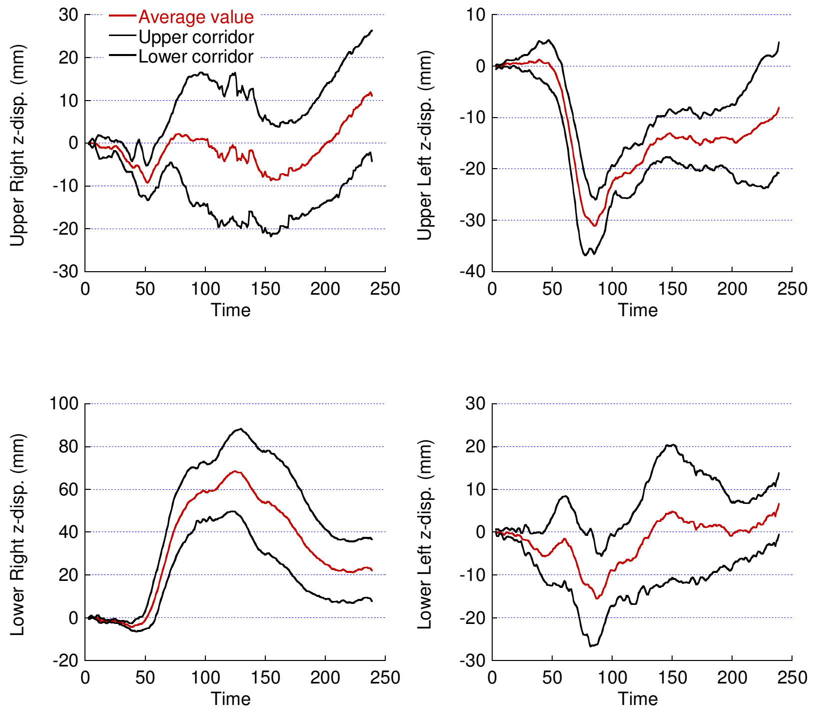


Figure K12: Chest displacement in the z-axes

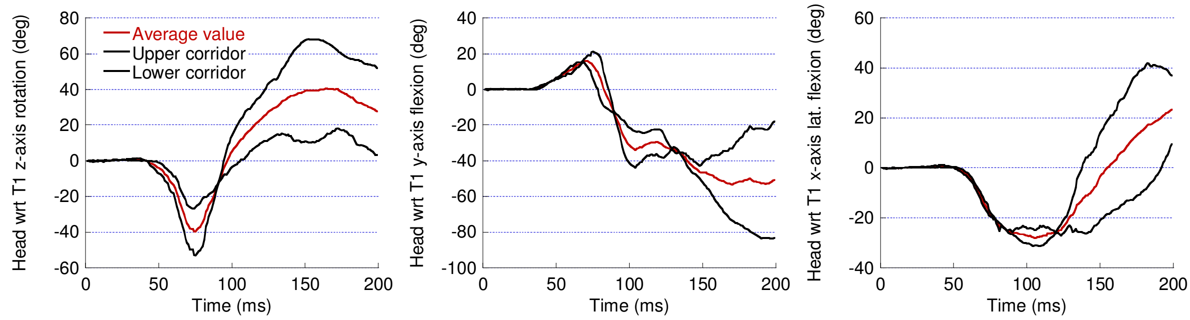


Figure K-13: Head relative T1 angular displacement around the z- (rotation), z- (flexion), and x- (lateral flexion) axes Shaw 2009 (No. of subjects = 3).

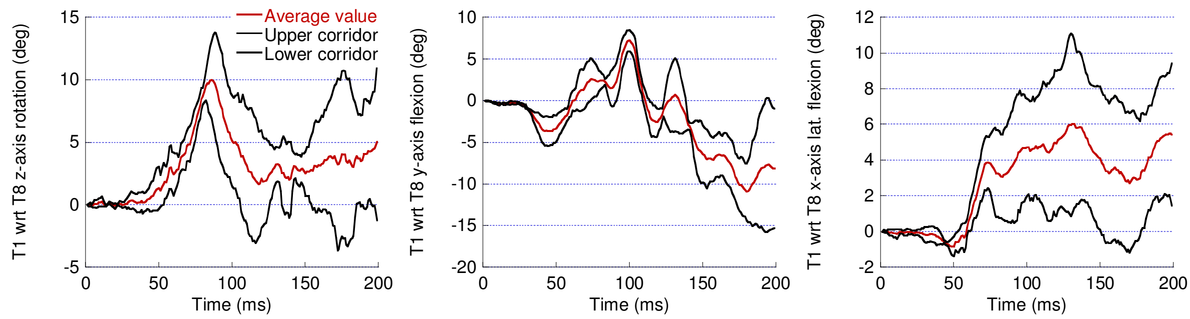


Figure K-14: T1 relative T8 angular displacement around the x-, y-, and z-axes

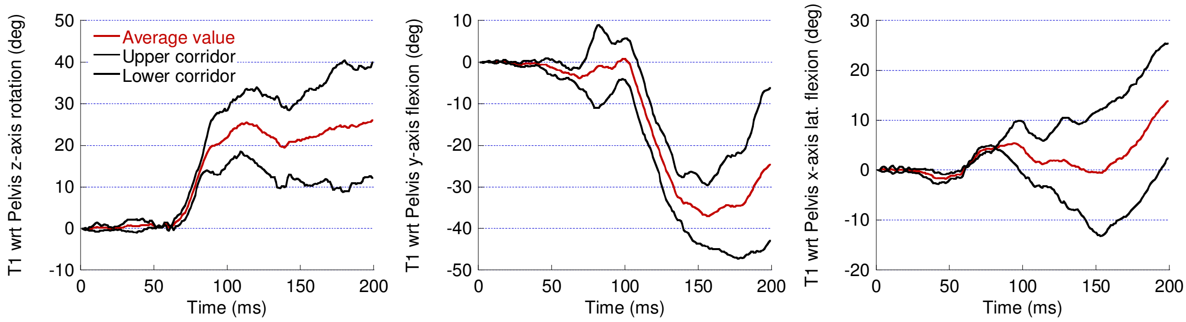


Figure K-15: T1 relative Pelvis angular displacement around the x-, y-, and z-axes

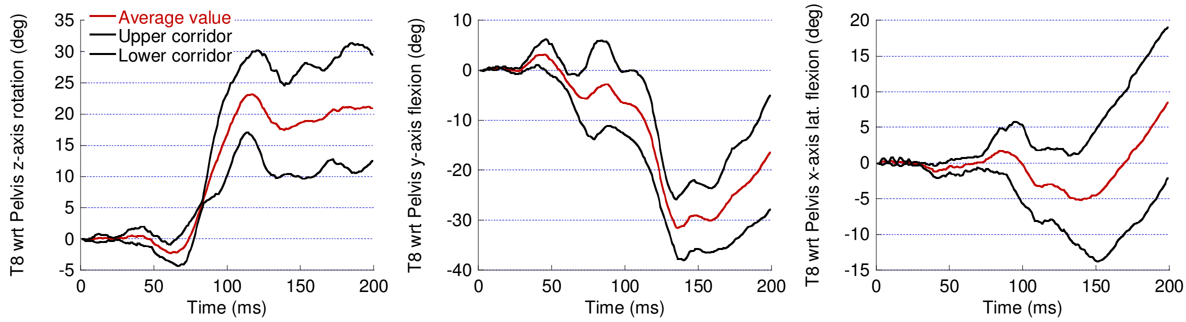


Figure K-16: T8 relative Pelvis angular displacement around the x-, y-, and z-axes

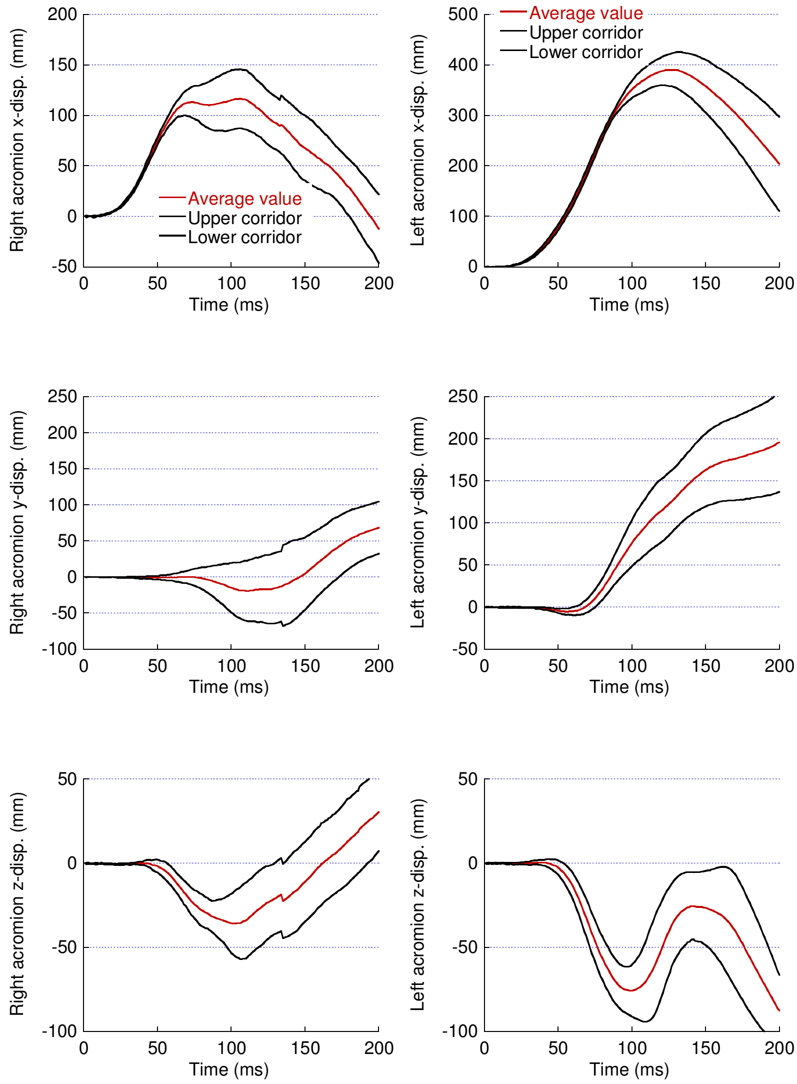


Figure K-17: X- Y and Z-axis displacements for the left and right acromion

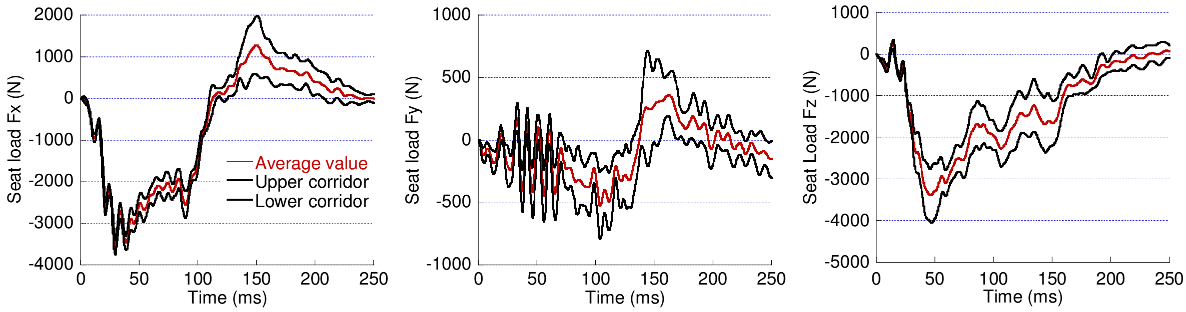


Figure K-18: Seat forces

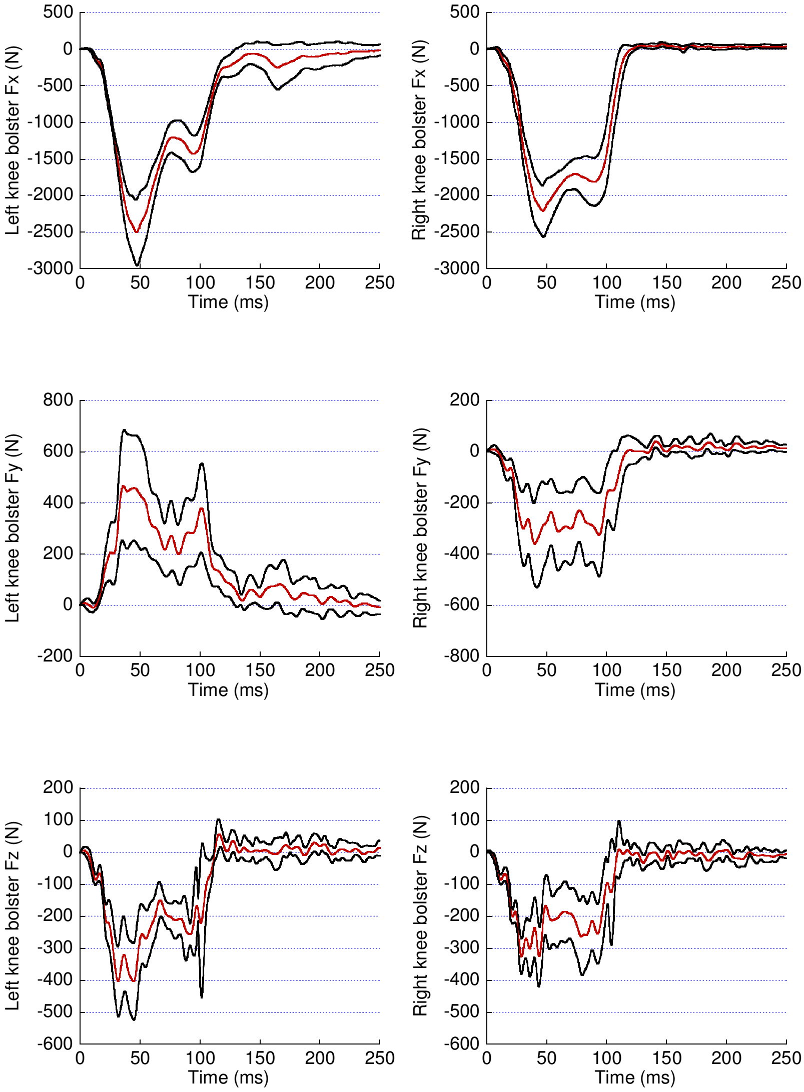


Figure K-19: Knee bolster forces

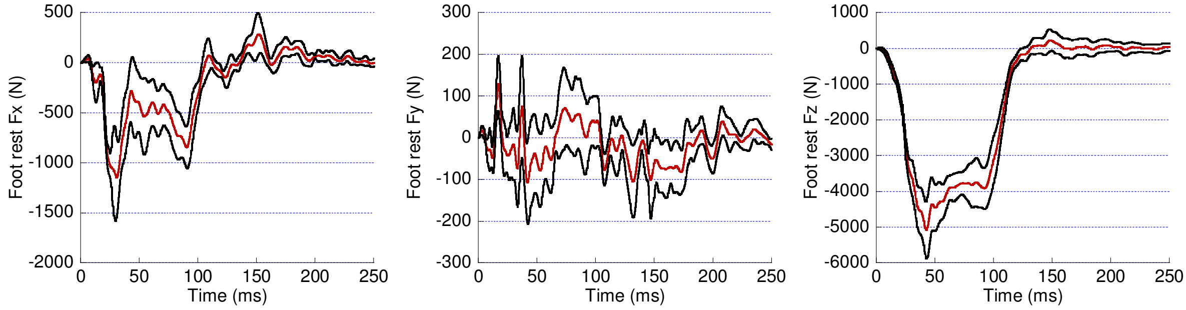


Figure K-20: Footrest forces