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Biomechanical Response Manual: THOR 5th Percentile Female NHTSA Advanced Frontal Dummy, Revision 2

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This report revises and updates DOT HS 812 370, February 2017, by the same authors.

16. Abstract

The THOR-05F (Test Device for Human Occupant Restraint 5th Percentile Female) is being designed to provide improved biofidelity compared to the Hybrid III 5th Percentile Female, particularly in evaluating head/neck injuries due to air bag deployment, interaction with restraints (e.g., abdominal response in submarining) along with an improved pelvis, knee-thigh-hip, and lower leg. This manual describes the anthropometry and biomechanical response targets which are recommended to assess the THOR-05F anthropomorphic test dummy (ATD). The tests and procedures described here were derived primarily for use by dummy manufacturers during the pre-production design and development process. The tests and procedures are designed so that results may be assessed objectively. Tests are therefore designed to produce results in the form of time-history signals so that an objective quantitative scoring process may be performed.

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Version History

Version	Release Date	
1	February 2017	New manual applicable to THOR-05F female ATD
2	July 2019	Changes by body region: Head: Reference correction; updated scaling factors and response targets; combine head and face sections. Neck: Use volunteer T1 pulse as input to sled tests, with base of head/neck attached directly to sled. Abdomen: Scaling error corrected in lower abdomen rigid bar corridor. Knee-Thigh-Hip, whole body impact: Reduce test speed from 4.9 to 3.5 m/s; added target corridors for impactor force vs. time and pelvis acceleration vs. time. Knee slider impact: Reduce test speed from 2.75 to 2.15 m/s; change allowed probe mass to use existing Hybrid III probe. Leg, axial heel impact: New corridors created using 5th female specimens. Ball of foot impact: Impact speed correction (previously incorrectly reported as 2.0 m/s; should be 5.0 m/s). Full body sled test: New corridors created using 5th female specimens.

1. Introduction / Purpose

Investigations using the FARS and NASS-CDS databases have demonstrated that, in a comparable crash, belted females have higher overall risk of injury and death than belted males (Bose et al., 2011; Kahane, 2013; Parenteau et al., 2013). Differing injury patterns between males and females also suggest differences in restraint interaction and effectiveness. For example, using NASS-CDS data from 1997 to 2011, Parenteau et al. (2013) showed that females have higher risk of belt and air bag sourced injuries, including thoracic and spinal fractures. Kahane (2013) also found that females had a higher percentage of injuries sourced to the air bag in frontal collisions. Short stature females have a higher risk of lower extremity injuries due to seat positioning closer to the steering wheel and knee bolster. Finally, females have higher risk of neck injuries and AIS 2+ abdominal injuries than males (Kahane, 2013). To further evaluate injury sources and mechanisms, a similar data set from the CIREN database was evaluated for restrained, short stature, adult females (height less than 5’4”; weight less than 140 lbs). This analysis showed that thoracic and abdominal injuries were typically due to belt interaction, while neck injuries were often due to interactions of belt and air bag with a close-seated (short-statured) occupant. The available data thus demonstrates that safety concerns for small females differ from those of mid-sized males, supporting the need for development of a small female advanced frontal impact anthropomorphic test dummy (ATD). The THOR-05F (**T**est device for **H**uman **O**ccupant **R**estraint 5th Percentile Female) is being designed to provide improved biofidelity compared to the Hybrid III 5th Percentile Female, particularly in evaluating head/neck injuries due to air bag deployment, interaction with restraints (e.g., abdominal response in submarining) along with an improved pelvis, knee-thigh-hip, and lower leg.

Biofidelity of a test dummy is a measure of the dummy’s ability to mimic a human-like response in a crash environment. An assessment of biofidelity includes, but is not limited to, anthropometry, mass properties, joint properties (e.g., range of motion), and biomechanical response to impact. This manual describes the anthropometry and biomechanical response targets which are recommended to assess the THOR-05F ATD. The tests and procedures described here were derived primarily for use by dummy manufacturers during the pre-production design and development process. The tests and procedures may also be used to verify computer models. The tests and procedures are designed so that results may be assessed objectively. Tests are therefore designed to produce results in the form of time-history signals so that an objective quantitative scoring process, such as BioRanking, may be performed.

We emphasize that the tests described in this manual pertain to impact biofidelity only, and mainly to body segments. NHTSA stresses that a satisfactory assessment based on the procedures herein does not guarantee acceptable biofidelity overall. There are many other important biofidelity characteristics not covered directly by the assessments herein, but are embodied within the design of a fully biofidelic dummy. They include human anthropometry and mass properties, body segmentation and joint properties, and the degree to which soft tissues, ligamentous structures, and musculature are accurately represented in the ATD.

2. Overview of Test Procedures and Assessments

To assess the biomechanical response to impact, a number of tests are described herein and summarized in Table 2.1. The tests and assessments cover the head, neck, thorax, shoulder, abdomen, lumbar spine, and lower extremity. Most of the test procedures follow the recommendations by Lebarbe et al. (2015a, 2015b), who described a comprehensive set of specifications for assessing the biofidelity of an ATD. In addition, assessment procedures used for THOR-50M were considered (Parent, 2017).

Table 2.1. Biofidelity test matrix with test conditions appropriate for the THOR-05F

Body Region	Test	Impact Velocity	Impactor Mass	Impactor Face
Head	Head Impact	2.0 m/s	19.2 kg	152.4 mm disk
	Rigid Bar Face Impact	3.6 ± 0.1 m/s	26.2 kg	Rigid Bar, Diameter 25 mm
	Rigid Disk Face Impact	6.7 ± 0.1 m/s	10.7 kg	152.4 mm disk
Neck	Neck Frontal Flexion Response	15G Sled Acceleration		
	Neck Lateral Flexion Response	7G Sled Acceleration		
	Torsion	500°/sec		
Thorax	Upper Ribcage Central Impact	4.3 ± 0.1 m/s	14.0 kg	152.4 mm disk
	Lower Ribcage Oblique Impact (L & R)	4.3 ± 0.1 m/s	14.0 kg	152.4 mm disk w/pad
Shoulder	Range of Motion/Stiffness Test	-	-	-
Abdomen	Upper Abdomen Dynamic Impact	6.7 ± 0.1 m/s	9.0 kg	Steering Wheel, Diameter 26.7 mm
	Lower Abdomen Dynamic Impact	6.1 ± 0.1 m/s	16.0 kg	Rigid Bar, Diameter 25 mm
	Belt Loading	4 m/s	-	-
Lumbar Spine	Flexion Pendulum Test	2.0 m/s	-	-
Knee-Thigh-Hip	Knee-Thigh-Hip Impact (L & R)	1.2 m/s	250 kg ram	Molded knee interface w/pad
	Whole Body KTH Impact	3.5 m/s	255 kg ram	Padded knee interface
	Knee Slider Impact (L & R)	2.15 m/s	7.26 kg	76.2 mm disk
Leg-Foot-Ankle	Axial Heel Impact (L & R)	3.1 m/s	28.4 kg	Padded Footplate
	Dynamic Dorsiflexion	5.0 m/s	3.0 kg	NHTSA Impactor
	Inversion/Eversion (L & R)	1000°/sec	-	-
Full Body Sled Test		30 km/h with 2 kN force limiting belt		

2.1 Scaling process

Due to lack of original biomechanical data for 5th percentile female occupants, many of the biomechanical response targets for the THOR-05F are based on suitably scaled versions of the response targets of the THOR-50M. The basic assumption in scaling procedures used in normalizing response data for adults is that the mass densities and elastic moduli of human tissue (muscle, bone, etc.) are about the same for all adults, irrespective of size or sex. These assumptions form the basis of equal stress/equal velocity scaling. The basic scaling principles used here have been developed and applied in the design of previous ATDs, including Hybrid III, SID-IIs and WorldSID (Been et al., 2007; Daniel et al., 1995; Irwin et al., 2002; Mertz et al., 1989).

The mass properties, dimensions and scale factors for the THOR-05F were determined from the Anthropometry of Motor Vehicle Occupants (AMVO) study by Schneider et al. (1983), on a wide variety of dimensions, including segment lengths, masses and CG locations; relative joint and landmark positions; and external body contours. These are summarized in Table 2.2. This approach has been used previously (Been et al., 2007). 5th percentile female stature and weight in the current (2008) U.S. population is equivalent (i.e., within 3 mm in height and 3 kg in weight) to the population measured in the AMVO study, giving high confidence that an ATD developed to the AMVO specification will occupy the same percentile relative to the current population (Ebert and Reed, 2013). Seated anthropometric dimensions were targeted where possible (Figure 2.1), because the THOR-05F dummy is being designed solely for seated postures.

Table 2.2. Reference anthropometry for 5th percentile female (from Schneider et al., 1983)

Body Region	Mass (kg)	Center of Gravity Location (mm)		Segment Moment of Inertia (kg-cm^2)	
Head	3.70	Relative to O.C.		I_x	146.2
		x	5	I_y	172.9
		z	59	I_z	131.7
Neck	0.60	Relative to O.C.		I_x	6.1
		x	-17	I_y	9.5
		z	-59	I_z	10.3
Thorax	12.98	Relative to Hip*		I_x	1542.8
		x	-147	I_y	1161.2
		z	238	I_z	1208.6
Abdomen	1.61	Relative to Hip*		I_x	143.5
		x	-82	I_y	101.5
		z	107	I_z	205.7
Pelvis	6.98	Relative to Hip*		I_x	326.2
		x	-76	I_y	282.9
		z	25	I_z	574.2
Upper Arm	1.12	Relative to Elbow†		I_x	50.0
			145	I_y	51.1
				I_z	8.2
Lower Arm (incl. hand)	1.14	Relative to Elbow†		I_x	141.5
			141	I_y	129.4
				I_z	8.3
Upper Leg	5.91	Relative to Hip†		I_x	731.4
			149	I_y	701.0
				I_z	153.9
Lower Leg	2.36	Relative to Knee†		I_x	261.4
			151	I_y	261.9
				I_z	23.1
Foot	0.64	Relative to Heel†		I_x	3.4
			84	I_y	18.4
				I_z	16.6
Total Body	48.2				

**In contoured hardseat representative of typical vehicle seat.*
†Resultant distance.

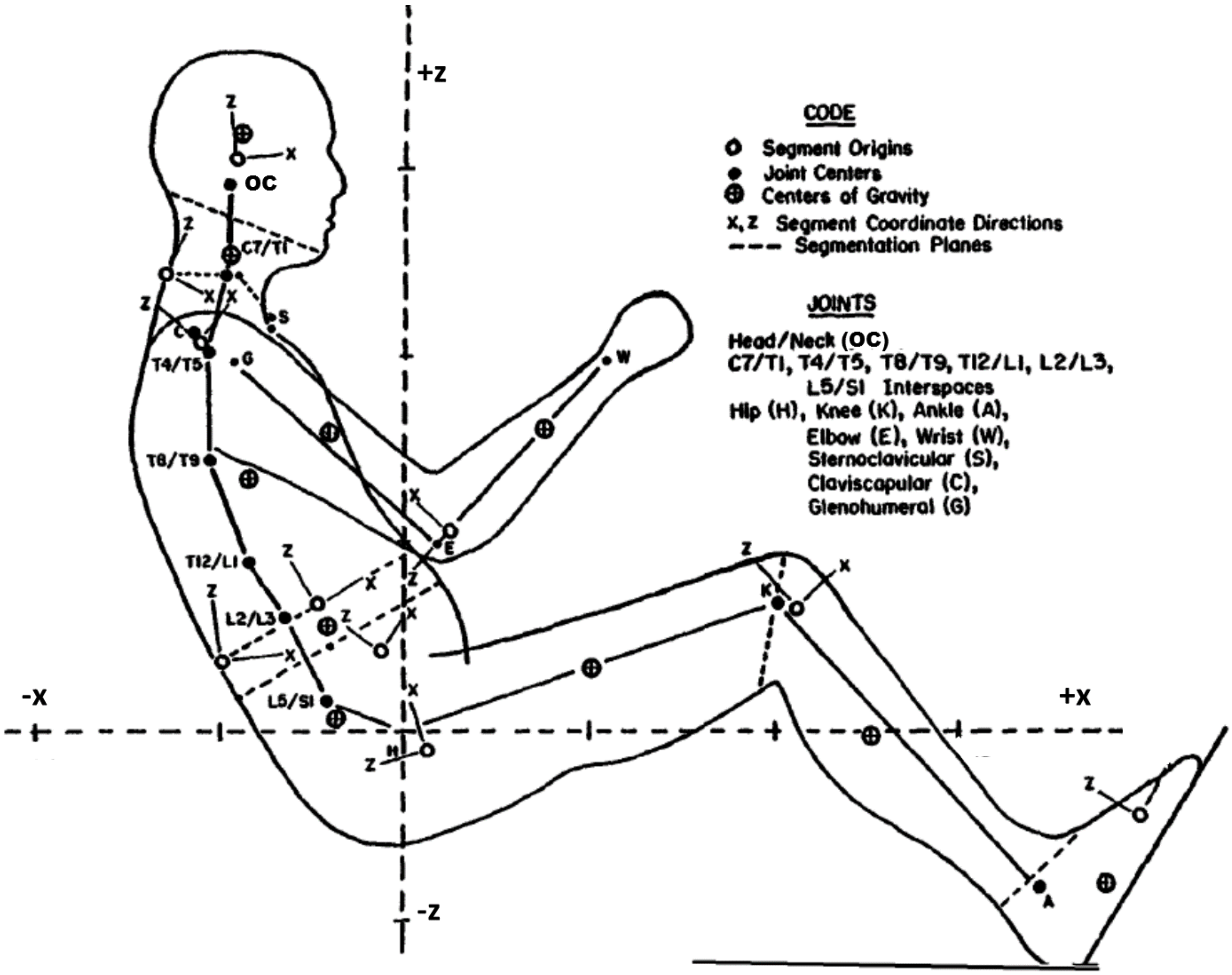

Figure 2.1. Reference posture for anthropometry given in Table 2.2. Figure adapted from Schneider et al. (1983).

2.2 BioRank assessment process

The NHTSA Biofidelity Ranking System (BRS; Rhule et al., 2002, Rhule et al., 2009, Rhule et al., 2013, Rhule et al., 2018) which was established based on the response from post mortem human surrogate (PMHS) will be used as an objective measure of how well the dummy matches the biomechanical response targets outlined here. In order for the BRS to result in meaningful, quantitative comparisons, the human response corridors must consist of a common statistical definition (e.g. the BRS method applies only to corridors that represent the mean ± 1 standard deviation). A BRS < 2.0 means that the dummy can be considered to respond as much like the cadaver corridor as would another subject. For the purposes of this manual, subjective nomenclature to facilitate classification can be used, based on the Biofidelity Scale presented by Rhule et al. (2009) (Table 2.3).

Table 2.3. Biofidelity scale nomenclature

BioRank ($\sqrt{R}$)	Description	Classification
$\sqrt{R} \leq 1$	within one standard deviation of the mean PMHS response	Excellent
$1 < \sqrt{R} \leq 2$	between one and two standard deviations of the mean PMHS response	Good
$2 < \sqrt{R} \leq 3$	between two and three standard deviations of the mean PMHS response	Marginal
$\sqrt{R} > 3$	more than three standard deviations from the mean PMHS response	Poor

Corridors shown herein thus represent the mean ± 1 standard deviation of the human response. For time history corridors, standard deviations of the response metric were taken at each time step. For force-deflection or moment-angle targets, corridors were generated using the elliptical method, as described by Ash et al. (2012). In brief, this consists of calculating the standard deviations in the "vertical" direction (i.e. variation in force for a given deflection) and also the "horizontal" direction (i.e. variation in deflection for a given force). These standard deviations are used as an ellipse associated with that particular point on the mean curve. The upper and lower corridor boundariesare formed using the outer bounds of the ellipses at each point along the mean response curve.

Certain tests do not have suitable corridors for a BioRank scoring, including neck torsion and lumbar spine flexion. These are nevertheless included here for completeness.

3. Head

3.1 Scale Factors

The principle dimensions and scale factors for the head (Table 3.1) were determined from the anthropometric specification by Schneider et al. (AMVO, Schneider et al,1983).

Table 3.1. Scale factors for 5th percentile female head

Measurement	50th Male	5th Female	Scale Factor	
Mass (kg)	44.54	33.70	λ_m	0.82
Head circumference (cm)	57.1	53.4		0.935
Head Length, L (cm)	19.77	18.3	λ_x	0.93
Head Breadth, B (cm)	15.8	14.5	λ_y	0.92
Head Height, H (cm)	23.1	20.0	λ_z	0.87

Historically, for the Hybrid III 5th percentile female dummy, the SID-IIs (which simply utilized the Hybrid III head) and the WorldSID small female, the head was assumed to be geometrically similar to the 50th percentile dummy head. The three dimensional scale factors (λ_x, λ_y, λ_z) were assumed to be equal and had a value of 0.931, which was the ratio of a characteristic length equal to the sum of head circumference, length and breadth (Mertz et al., 1989, Irwin et al., 2002). Anthropometric specifications (Table 3.1) demonstrate that the head of the human 5th percentile female is geometrically similar to the 50th percentile male in the x-y plane but not in height (z axis). To be consistent with the current understanding of 5th percentile female anthropometry, the biofidelity targets will be scaled by $\lambda_x = 0.93$, $\lambda_y = 0.92$ and $\lambda_z = 0.87$.

3.2 Head Impact Test

The head impact response is based on tests by Hodgson and Thomas (1971). In these tests, the cadavers were strapped to a pallet that was free to rotate at the feet, with the head and neck extended over the free end. The pallet was released at a fixed distance from a rigid impact surface. The average peak resultant acceleration was 250g and the average impact velocity was 2.71 m/s, which is equivalent to a free fall height of 376 mm (14.8 in). Hubbard and McLeod (1974) used the data to generate a corridor for the head impact response by using an allowable variation of ±10 percent (i.e. the range was 225-275g). The head drop test and the corresponding corridor became the target for establishing the biofidelity of the GM ATD 502 head (Hubbard, et al., 1974), which was later used as the head for the Hybrid III dummy.

An equivalent procedure was later developed (Melvin, King and Alem, 1988) that equated the impact energy in the head drop tests with the effective impact energy from an impactor test on the forehead of a complete dummy. Due to difficulties separating the mass of the neck from the mass of the head in the THOR dummy (due to neck spring cables), the equivalent procedure was used in developing the THOR-50M head.

As was the case for the Hybrid III 5th percentile dummy, for the THOR-05F, the impact velocity is held constant, in accordance with the equal stress/equal velocity scaling method. The impactor mass is scaled from that for THOR-50M ($m_{p,50}$) to produce an equivalent response in the small female head. The impactor diameter, 152.4 mm, can remain the same because it is much larger than the relative contact area on the head.

$m_{p,5} = \lambda_m m_{p,50} = (0.82)(23.4) = 19.2$ kg

Scaling of Biomechanical Response for Head Impact

The biofidelity target for THOR-50M is a force and time specification. To represent the response of a small female, both the force and time of these response targets must be scaled for the small female. To do so, consider that the head impact condition can be represented as a spring-mass system. The derivation is described in Mertz (1989).

The spring stiffness of the head, k, can be described as a function of Elastic Modulus, E, impact area, A, and head depth, L. Force, F, is then calculated using spring stiffness and depth, x.

$k = E*A/L$
$F = k*x$

Elastic modulus is assumed to be constant. Impact area on the head occurs in the y-z plane, and original head depth is in the x direction. Thus, the resulting scale factors are:

$\lambda_k = \lambda_y \lambda_z / \lambda_x = 0.86$
$\lambda_F = \lambda_k \lambda_x = 0.80$

As noted by Irwin et al. (2002), impact duration scales as: $\lambda_t = \sqrt{\lambda_m/\lambda_k} = 0.98$.

The peak probe force response is given in Table 3.2.

Table 3.2. 5th female scaled targets for peak probe force in head impact

	Upper bound	Lower bound
Probe force (kN)	5.11	4.18
Time (ms)	1.95	2.93

3.3 Rigid Bar Impact

There are two primary facial impact response targets. The first is based on rigid bar impacts performed by Nyquist et al. (1986) and summarized by Melvin and Shee (1989). These tests used a 25 mm diameter bar, with an attached mass of 32 kg or 64 kg, which impacted across the nose and zygoma of unembalmed, seated cadavers. Seven tests, in which only nasal bones were fractured, were included in the generation of the corridor. The average impact velocity was 3.6 m/s (with a range of 2.8 to 4.8 m/s). Because the head was unrestrained, the impact caused the head to translate rearward. The response target is a peak force and time specification that represents the mean ± 1 standard deviation of the cadaver data.

For the THOR-50M, an average impact speed of 3.6 m/s was used, along with a 32 kg impactor. Assuming that the impact velocity is held constant, the response of the THOR-05F face depends on pendulum mass, and the mass and stiffness of the small female head. The pendulum mass is chosen to produce the same percent compression as the 50th male and is scaled according to the mass scale factor.

$m_{p,5} = \lambda_m m_{p,50} = (0.82)(32) = 26.2$ kg

Because the rigid bar impact test was intended to be analogous to steering wheel loading, the diameter of the impactor is not being scaled for the THOR-05F.

Biomechanical Response Target

The response target is a peak force and time of peak force, with scaling factors the same as described in Section 3.2. The scaled corridor is shown in Table 3.3.

Table 3.3. 5th female scaled targets for peak probe force in face rigid bar impact

	Upper bound	Lower bound
Probe force (kN)	2.95	1.80
Time (ms)	6.83	4.69

3.4 Rigid Disk Impact

The second facial impact response target is based on disk impacts performed by Wayne State University and summarized by Melvin and Shee (1989). The test used a 15.2 cm flat disk impactor, with a mass of 13 kg, impacting the face at an average impact velocity of 6.7 m/s. The response target is a peak force and time specification that represents the mean ± 1 standard deviation of the cadaver data.

For the THOR-50M, an impact speed of 6.7 m/s was used, along with a 13 kg impactor. Assuming that the impact velocity is held constant, the response of the THOR-05F face depends on pendulum mass, and the mass and stiffness of the small female head. The pendulum mass is chosen to produce the same percent compression as the 50th percentile male and is scaled according to the mass scale factor. The impactor diameter, 152.4 mm, can remain the same because it is much larger than the relative contact area on the head.

$m_{p,5} = \lambda_m m_{p,50} = (0.82)(13) = 10.7$ kg

Biomechanical Response Target

The response target is a peak force and time of peak force, with scaling factors the same as described in Section 3.2. The scaled corridor is shown in Table 3.4.

Table 3.4. 5th female scaled targets for peak probe force in face disk impact

	Upper bound	Lower bound
Probe force (kN)	6.56	3.52
Time (ms)	4.39	2.64

4. Neck

4.1 Scale Factors

The principle dimensions and scale factors for the neck were determined from the AMVO study (Schneider et al., 1983).

Table 4.1. Scale factors for 5th percentile female neck

Measurement	50th Male	5th Female	Scale Factor		Recommended for THOR-05F
Total body mass (kg)	76.7	46.9	λ_m	0.60	$\lambda_m = 0.60$
Neck mass (kg) (AMVO Table) 5.8)	0.965	0.6	λ_m	0.62	
Neck breadth (cm) (mid)	11.4	9.1	λ_y	0.80	$\lambda_x = \lambda_y = 0.81$ (average)
Neck depth (cm) (mid)	11.5	9.0	λ_x	0.78	
Neck circumference (cm) (mid)	38.3	30.4	λ_x, λ_y	0.79	
Neck breadth (cm) (lower)	12.2	10.4	λ_y	0.85	
Neck depth (cm) (lower)	11.5	9.3	λ_x	0.81	
Neck circumference (cm) (lower)	39.3	32.2	λ_x, λ_y	0.82	
Erect seated height (cm)	90.7	81.3	λ_z	0.896	$\lambda_z = 0.91$
Neck length (anterior) (cm)	8.5	8.1	λ_z	0.95	
Distance between Head/Neck and C7/T1 (cm) joint centers (AMVO Table 5.14)	9.0	11.9	λ_z	0.76	

For the Hybrid III 5th percentile female dummy, mass scaling for the neck was achieved using total body mass (Mertz et al., 1989) and dimensional scaling was achieved using a characteristic length equal to erect seated height. For the WorldSID small female, dimensional scaling was achieved by assuming the necks were geometrically similar and that all dimensions were proportional to the neck circumference ($\lambda_x = \lambda_y = \lambda_z = 0.794$). Neck mass was then calculated using a constant density relationship (Irwin et al., 2002).

To maintain consistency with the scaling for Hybrid III, the total body mass ratio will be used for scaling the response of the THOR-05F. Note that the ratio of 5th percentile female to midsize male neck mass reported in AMVO is similar to the total body mass ratio (Table 4.1).

$\lambda_m = 0.60$

Anthropometric data demonstrates that the neck of the 5th percentile female is geometrically similar to the midsize male in the transverse plane but not in length. For the transverse plane, averaging the scale factors for neck breadth, depth and circumference yields, $\lambda_x = \lambda_y = 0.81$.

If the neck is assumed to behave like a simple cantilevered beam, both neck length and neck circumference contribute to the head/neck motion in a frontal crash. However, neck length is more difficult to measure on live humans than neck circumference and can be defined in different ways. The AMVO study measured anterior neck length (from suprasternale landmark to head-neck junction under the chin), reporting a scaling ratio of 0.95. The distance between estimated joint centers for the head/neck and C7/T1 was also reported in AMVO, with a scaling ratio of 0.76. Given these discrepancies, an alternate approach is to calculate neck length scale factor using the constant density assumption ($\lambda_m = \lambda_x^2\ \lambda_z$).

$\lambda_z = \lambda_m / \lambda_x^2 = 0.91$

The resulting scale factor for λ_z (Table 4.2) is within the range for neck length reported by AMVO (Table 4.1) and is close to the scaling ratio for erect seated height (which was used in scaling the Hybrid III response).

Table 4.2. Scaling for neck length

Dummy	Scale Factor	Basis
Hybrid III 5th	0.896	Erect seated height
World SID 5th	0.794	Assumes $\lambda_x = \lambda_y = \lambda_z$
THOR-05F	0.91*	Assumes $\lambda_z = \lambda_m / \lambda_x^2$

**For reference, neck length scale factor based on AMVO is 0.95.*

4.2 Neck Frontal Flexion Tests

For the THOR-50M, both kinematic and dynamic targets are specified. The kinematic tests were not specified for the Hybrid III and are therefore new for the THOR ATD. The frontal flexion targets arose from the extensive tests conducted on Naval Biodynamics Research Laboratory (NBDL) volunteers by Ewing et al. (1968, 1969, 1973, 1975, 1976, 1977). The peak sled accelerations used in these tests was approximately 15g. These data were later analyzed by Wismans and Spenny (1983, 1984, 1987) and Thunnissen et al. (1995), from which the final form of the corridors were developed. The neck response is primarily prescribed by the time histories of the resultant head acceleration, head rotation angle, neck rotation angle, and the longitudinal and vertical displacements of the head C.G (relative to T1). Comparable volunteer tests were performed in lateral flexion and those tests formed the basis of lateral bending targets for the WorldSID dummies and the THOR-50M. The dynamic targets are in the form of moment-angle corridors and the corridors are based on work by Mertz and Patrick (1971) for flexion and Patrick and Chou (1976) for lateral bending.

A whole dummy sled test most closely mimics the volunteer tests on which the biofidelity targets are based. However, the sled pulse and the restraint system must be consistent with the actual volunteer test. An alternative method for testing the biofidelity of the neck is to simplify the procedure by testing the head and neck assembly only, with the base of the neck being directly attached to a sled apparatus (optionally through a lower neck load cell) that can duplicate the acceleration pulse seen at the T1 location (Figure 4.1).

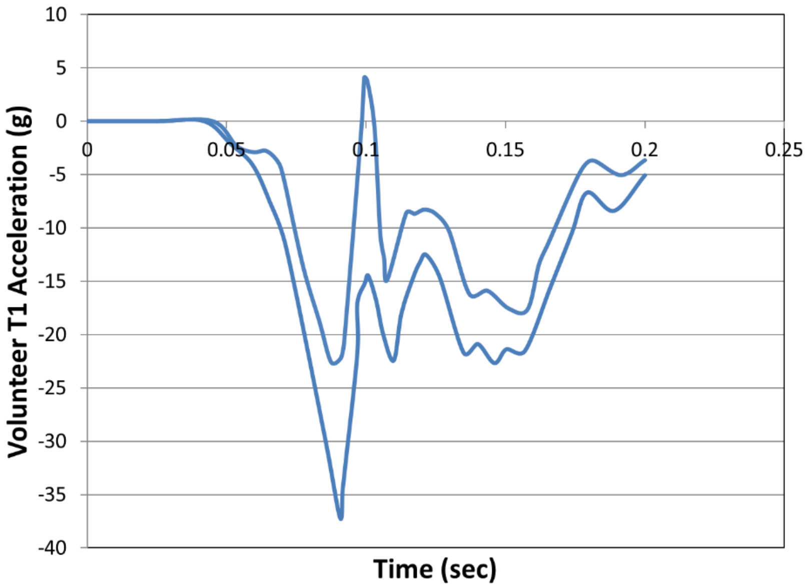

Figure 4.1. T1 acceleration corridor from NBDL volunteers is used as the input to the neck frontal flexion kinematic test, with the base of the neck being directly attached to a sled apparatus.

Scaling of Biomechanical Response

The neck muscles provide the resistive moment to neck bending (Mertz et al., 1989). Assuming that the neck acts as a cantilever beam with a point load, F, applied to the end, the resistive moment, M, can be expressed using a basic bending formula:

$$M = \sigma I / c,$$

where σ is beam stress, c is the distance of the farthest muscle fiber in the neck and I is the moment of inertia ($I = \frac{1}{4}\pi r^4$ for a circular cross section). Since stress is not scaled, per the equal stress/equal velocity scaling method, dimensional analysis demonstrates that the scaling factor for moment, M, simplifies to:

$$\lambda_M = \lambda_x^3 = 0.53$$

Similarly, standard beam analysis provides a formula for the deflection, δ, angular deflection, θ, and bending stiffness, k, of the cantilever beam. In these equations, L is the beam length (neck length), and E is the elastic modulus.

$$\delta = \frac{FL^3}{3EI} = \frac{ML^2}{3EI}$$

$$\theta = \frac{FL^2}{2EI} = \frac{ML}{2EI}$$

$$k = \frac{3EI}{L^3}$$

Again, dimensional analysis demonstrates that the scaling factors for head/neck angle, θ, and neck bending stiffness simplify to:

$\lambda_\theta = \lambda_z / \lambda_x = 1.12$
$\lambda_k = \lambda_x^4 / \lambda_z^3 = 0.57$

Beam deflection, δ, is equivalent to head CG displacement in the x-direction. Head CG displacement in the z-direction is determined by assuming constant strain, ε, between the 50th percentile male and 5th percentile female.

$\lambda_{\delta x} = \lambda_z^2 / \lambda_x = 1.02$
$\lambda_{\delta z} = \lambda_z = 0.91$

To scale the acceleration of the head and time duration of impact, assume that the head/neck obeys the laws of rotational motion for rigid bodies. The resulting scale factors are given by Irwin et al. (2002).

$$\lambda_a = \sqrt{\frac{\lambda_k}{\lambda_m}} = 0.98$$
$$\lambda_t = \sqrt{\frac{\lambda_m}{\lambda_k}} = 1.02$$

The neck response is prescribed by the time histories of the resultant head acceleration, head rotation angle, neck rotation angle, and the longitudinal and vertical displacements of the head CG (relative to T1). In addition, the neck must meet the head lag (neck angle versus head angle) and moment-angle corridors, which are visually assessed. Scaled corridors are shown in Figure 4.2 – Figure 4.7. Figure 4.8 shows the dynamic response corridor (moment versus angle) based on Mertz and Patrick (1971).

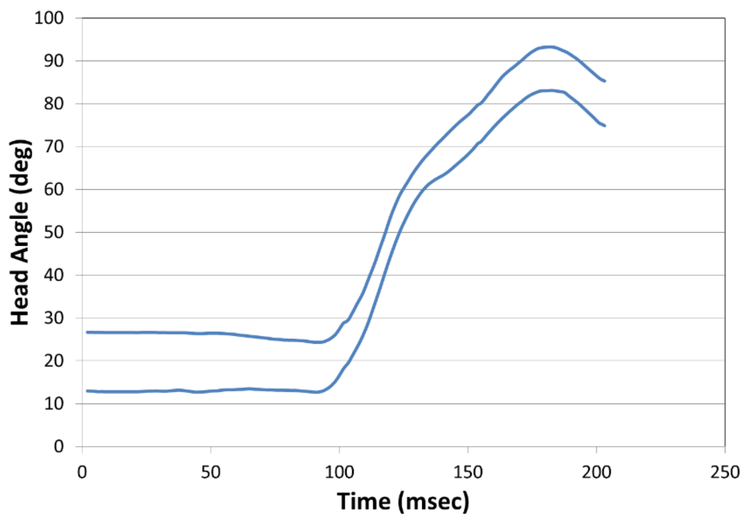

Figure 4.2. Head angle corridor for 5th percentile female.

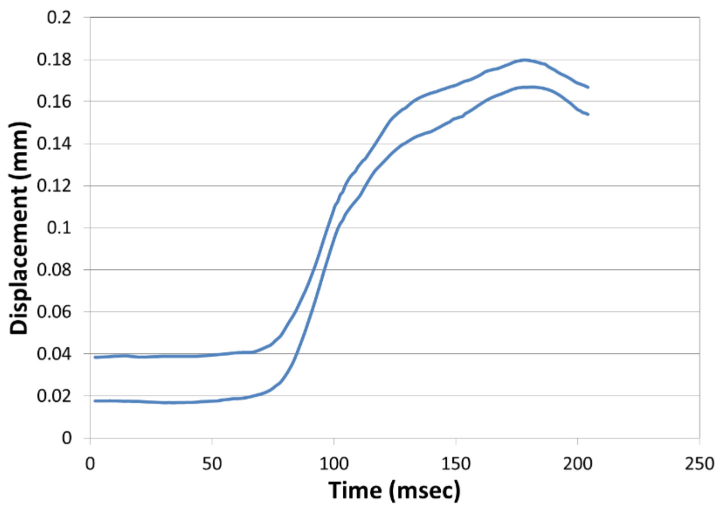

Figure 4.3. Head CG-x displacement corridor for 5th percentile female.

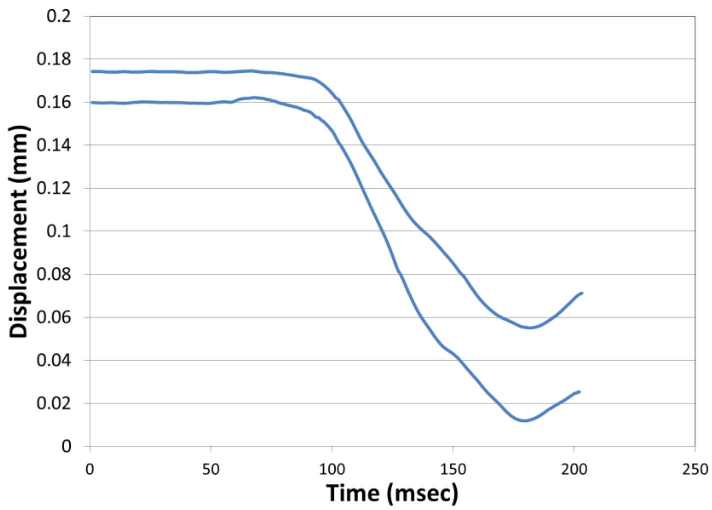

Figure 4.4. Head CG-z displacement corridor for 5th percentile female.

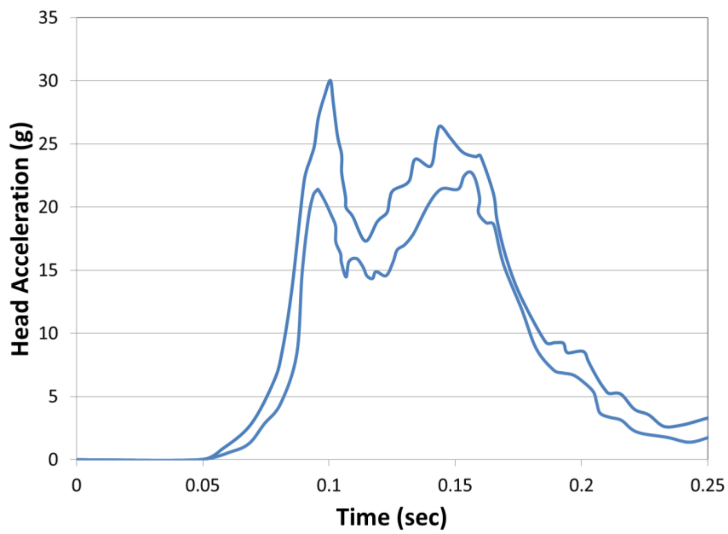

Figure 4.5. Head acceleration corridor for 5th percentile female.

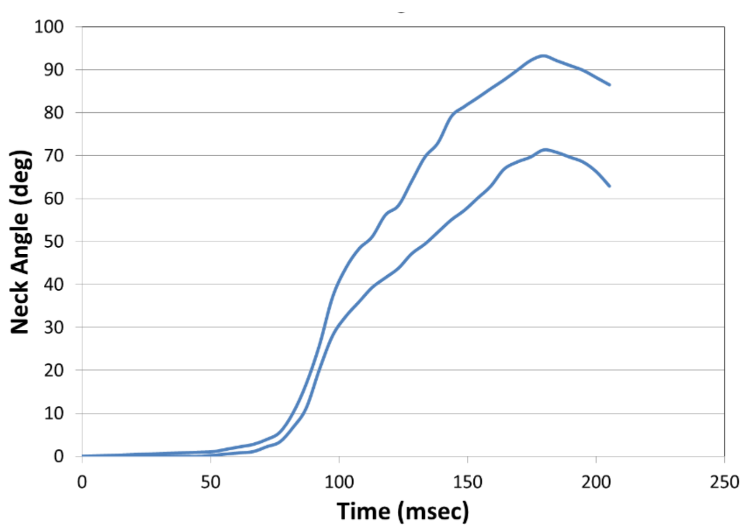

Figure 4.6. Neck angle corridor for 5th percentile female.

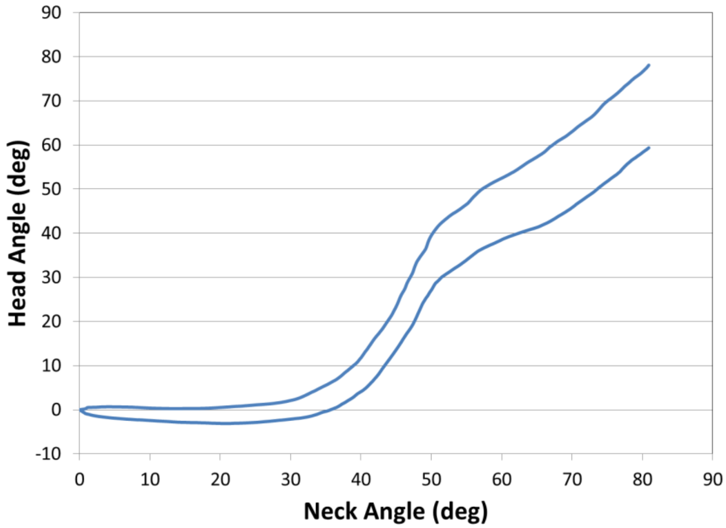

Figure 4.7. Head lag corridor for 5th percentile female.

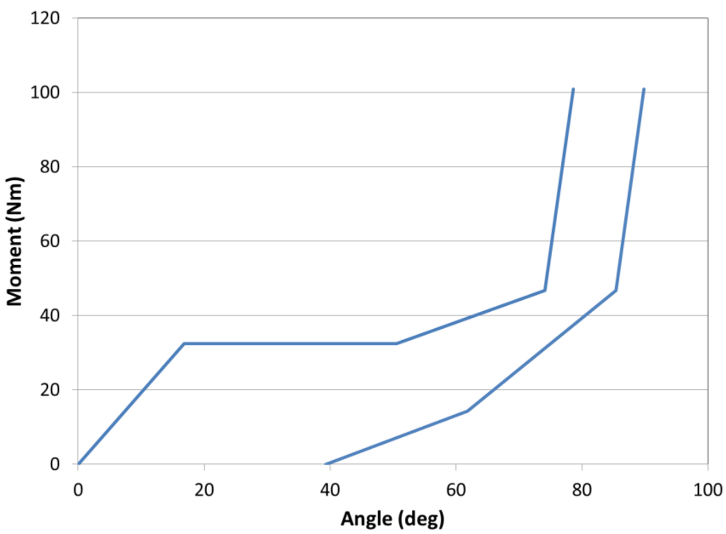

Figure 4.8. Approximate scaled corridor for 5th percentile female moment-angle response.

4.3 Neck Lateral Bending Tests

The lateral flexion targets are based on the same NBDL studies listed in Section 4.2. The peak sled acceleration in these tests was approximately 7g. Again, an alternative method for testing the biofidelity of the neck is to simplify the procedure by testing the head and neck assembly only, with the base of the neck being directly attached to a sled apparatus (optionally through a lower neck load cell) that can duplicate the acceleration pulse seen at the T1 location (Figure 4.9). The T1 acceleration corridor reported in Wismans and Spenny (1983) appeared to be delayed in time relative to the sled acceleration by approximately 50 ms. Therefore, for the corridor shown here, the pulse corridor time was corrected by shifting the pulse forward.

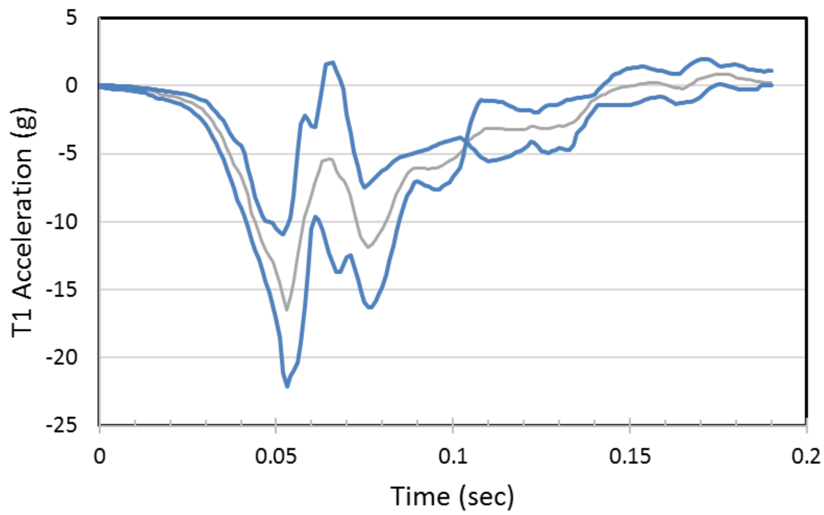

Figure 4.9. Lateral T1 acceleration from NBDL tests used as input to the lateral kinematic test, from Wismans and Spenny (1983).

Scaling of Biomechanical Response

Because $\lambda_x = \lambda_y$, the scale factors described above for frontal flexion (Section 4.1) can also be applied here. The lateral neck response is prescribed by the time histories of the head rotation angle, and the longitudinal and vertical displacements of the head CG (relative to T1). In addition, the neck must meet moment-angle corridors. Scaled corridors are shown in Figure 4.10 through Figure 4.13.

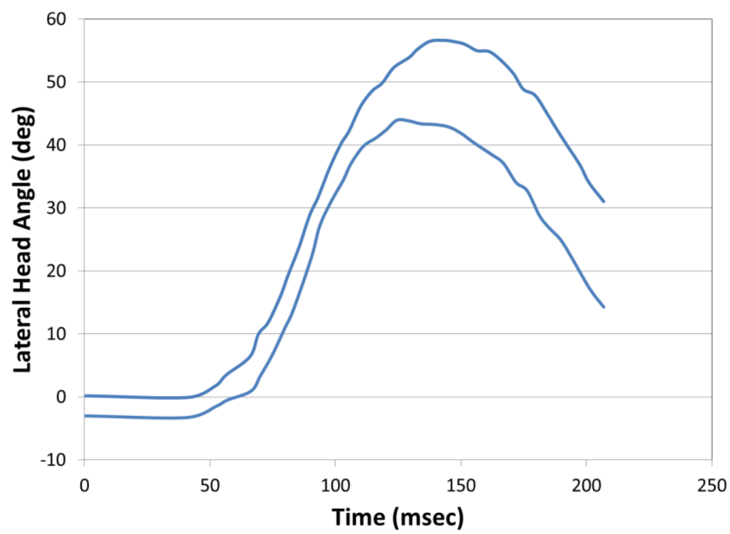

Figure 4.10. Lateral head angle corridors for 5th percentile female.

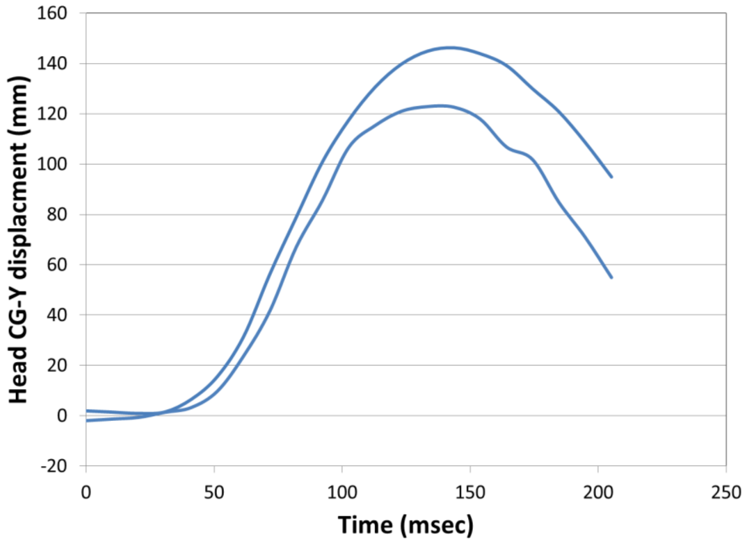

Figure 4.11. Lateral head CG-y displacement corridor for 5th percentile female.

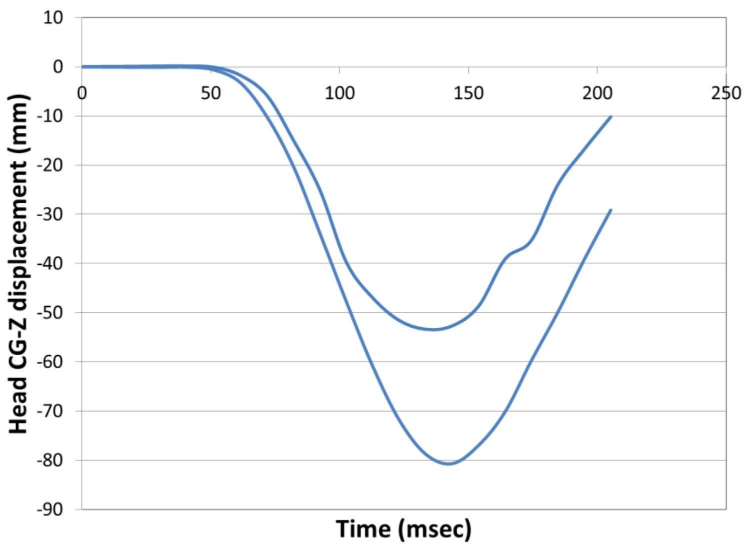

Figure 4.12. Lateral head CG-Z displacement corridor for 5th percentile female.

Dynamic Targets

The dynamic targets in lateral bending were given by Patrick and Chou (1976) and scaled using the scale factors described above for frontal flexion (Section 4.2).

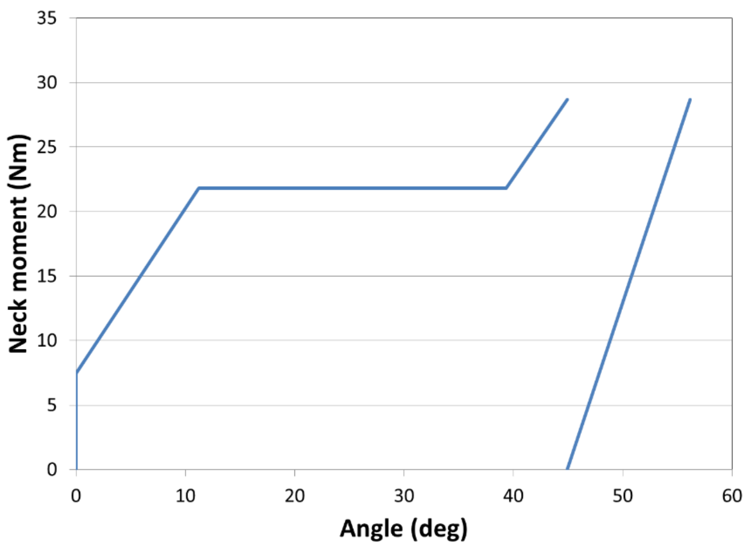

Figure 4.13. Scaled corridor for 5th percentile female lateral flexion moment-angle response.

4.4 Neck Torsion Test

The target for neck torsion is based on work done at Duke University (Myers et a., 1989, Myers et al., 1991) in which a dynamic servocontrolled torsion machine was used to apply pure rotation to the head/neck. Testing was initially done on six intact cervical spine specimens (from base of skull to T1). Tests to failure were conducted at approximately 500 °/sec. No lower cervical spine injuries were observed, so specimens were recast to isolate the lower cervical spine (C2-T1) and a second failure test was performed. The THOR neck is intended to more closely approximate the lower cervical spine in torsion, and therefore it is the results of the second test that are used to define its response. While the torsional response of the human neck is characterized by an initial phase in which the neck rotates almost freely (with no load) followed by a region of approximately constant stiffness, such a design for an ATD would not be practical. Therefore, rather than use the full moment-angle corridor (that would include the low load region), the target for the THOR-05F will be to match the specified moment-angle at injurious levels. For mid-sized male specimens this was approximately 21 ± 5 Nm over 63 ± 18 degrees of axial rotation, and was scaled to the 5th percentile female size. This target is not included in the BioRank calculation.

Scaling of Biomechanical Response

Similar to sagittal plane bending, the neck muscles provide the resistive moment to neck twisting. Therefore, it is assumed that the neck acts as a cantilever beam under axial twist loading, where the torque, T, is a function of shear stress, τ, polar moment of inertia, J ($J = \frac{1}{2}\pi r^4$ for a circular cross section), and c is the distance of the farthest muscle fiber in the neck.

$$T = \tau J / c$$

Similarly, the angle of twist, φ, can be expressed as:

$\varphi = TL/GJ$,

where L is neck length and G is the shear modulus. Since stress and modulus are not scaled, per the equal stress/equal velocity scaling method, dimensional analysis demonstrates that the scaling factors for torque and twist angle, simplify to:

$\lambda_T = \lambda_x^3 = 0.53$
$\lambda_\varphi = \lambda_z / \lambda_x = 1.12$

Table 4.3. Scaled moment-angle targets for THOR-05F in axial torsion

	Moment (Nm)	Twist Angle (deg)
Upper limit (mean + 1 S.D.)	13.8	90.7
Lower limit (mean – 1 S.D.)	8.5	50.4

5. Thorax

5.1 Scale Factors for Thorax

The principle dimensions and scale factors for the thorax (Table 5.1) were determined from the AMVO study (Schneider et al., 1983).

Table 5.1. Dimensional scale factors for 5th percentile female thorax

Measurement	50th Male	5th Female	Scale Factor		Recommended for THOR-05F
Thorax mass (kg) (AMVO, Table 5.8)	22.9	11.9	λ_m	0.52	$\lambda_m = 0.60$
Whole body mass (kg)	76.7	46.9	λ_m	0.60	
Torso depth (cm) (upper)	11.9	9.0	λ_x	0.76	$\lambda_x = 0.76$ $\lambda_y = 0.84$
Depth from T4 to mid-sternum (cm) (AMVO Landmarks # 8 & 19)	19.5	14.9	λ_x	0.76	
Depth from T8 to base of sternum (cm) (AMVO Landmarks # 9 & 20)	22.8	17.4	λ_x	0.76	
Depth from T12 to 10th rib (cm) (AMVO Landmarks #10 & 23)	26.1	19.0	λ_x	0.73	
Chest Breadth (cm) (axilla)	30.4	26.0	λ_y	0.86	
Chest Circumference (cm) (axilla)	103.9	82.4	λ_x, λ_y	0.79	
Chest Breadth (cm) (nipple)	34.9	27.6	λ_y	0.79	
Chest Circumference (cm) (nipple)	101.0	83.3	λ_x, λ_y	0.82	
Erect Seated Height (cm)	91.1	81.2	λ_z	0.89	$\lambda_z = 0.94$
C7 to T12 Height (cm) (AMVO Tables I.3) and I.6)	34.3	32.3	λ_z	0.94	
Cervicale to Trocanterion Height (cm) (hardseat) (AMVO Tables I.2 and I.5)	46.1	46.4	λ_z	1.0	

For the Hybrid III, mass scaling for the thorax was achieved using total body mass (Mertz et al., 1989). Although body segment masses were specified by AMVO based on segment dimensions and estimated densities, Schneider et al. (1983) noted that uncertainty in the estimated density of the partially hollow thorax resulted in uncertainty in the segment mass estimation for the thorax. As such, total body mass is maintained as the relevant dimension for mass scaling of the THOR-05F. In addition, maintaining the prior precedent will allow for use of existing test probes for thoracic impact response.

$\lambda_m = 0.60$

For Hybrid III, dimensional scale factors were determined using a characteristic length (erect seated height) and characteristic mass (total body mass). Scale factors for thorax depth and breadth were assumed to be equal ($\lambda_x = \lambda_y$) and were determined using the constant density relationship (Mertz et al., 1989):

$\lambda_m = \lambda_x^2 \lambda_z$

For a frontal thoracic impact chest depth (λ_x) is the most critical dimension related to thoracic injury. According to the Anthropometry of Motor Vehicle Occupants specification, various measurements of chest depth range from 0.73 to 0.76 (Table 5.1). To be consistent with a current understanding of 5th percentile female anthropometry, the characteristic depth is taken from the measured depth from T8 to base of sternum (based on the landmarks, this is approximately in the local thorax x-axis). Although this specification departs from past precedent, it is believed that it will yield a more accurate and biofidelic corridor for the current dummy. In the Z direction, examination of 5th percentile female anthropometry (Table 5.1) demonstrates that scale factors range from 0.89 to 1.0. Since erect seated height encompasses other body regions (e.g., head, neck and pelvis), the height from C7 to T12 is a more relevant dimension specific to the thorax.

$\lambda_x = 0.76$
$\lambda_z = 0.94$
$\lambda_y = \lambda_m / (\lambda_x \lambda_z) = 0.84$

5.2 Upper Ribcage Central Impact Test

The principal response corridors for the upper ribcage central impact are upper and lower limits for the expected force and deflection during impact, based on rigid disk impacts at 4.3 and 6.7 m/s. These tests were conducted using a seated cadaver, with a force applied in a horizontal direction, centered midsagitally over the fourth costal interspace at the sternum (Kroell et al., 1971). In some tests, internal deflection was measured directly, while in some tests, external deflection was measured and internal deflection corridors were developed using correction factors for the internal/external response (Kroell et al., 1971; Neathery, 1974). Injuries were sustained in about 75 percent of PMHS tested. The primary design target will be to meet the force versus deflection corridor for the 4.3 m/s impact, while the 6.7 m/s impact is secondary (and is used primarily for dummy durability evaluation).

The biofidelity condition for the THOR-50M uses a rigid and flat impactor, with a diameter of 152.4 mm and mass of 23.4 kg. To achieve an equivalent percent compression of the thorax in the 5th percentile female, the impactor mass is scaled by the mass scaling factor, λ_m. Scaling of the impactor mass also accommodates concern that impacting the 5th percentile female dummy using the THOR-50M test conditions would be too severe (i.e., would cause dummy damage). The impactor diameter, 152.4 mm, can remain the same because it engages equivalent structures within the ribcage on both the fiftieth and 5th dummies (i.e., dummy ribs 2-5), consistent with homologous loading.

$m_{p,5} = \lambda_m m_{p,50} = (0.6)\ (23.4\ \text{kg}) = 14.0\ \text{kg}$

Scaling of Biomechanical Response

The scaling of the thoracic corridors has been derived by Mertz et al. (1989). The thorax is represented by a circular ring of radius, r, with an elliptical cross section having a major diameter in the z-direction, h, and a minor diameter in the x-direction, b. The stiffness of the thorax, k, is represented by:

$k = Ehb^3 / r^3$

Since elastic modulus, E, is assumed to remain constant for adult males and females, and both b and r reside in the x-direction, scaling for thoracic stiffness simplifies to:

$\lambda_k = \lambda_z = 0.94$

The scale factor for deflection is given by (Mertz et al., 1989): $\lambda_d = \lambda_x = 0.76$.
The scale factor for force is given by (Mertz et al., 1989): $\lambda_F = \lambda_k \lambda_x = 0.71$.

Again, the primary design target will be to meet the force versus deflection corridor for the 4.3 m/s impact, while the 6.7 m/s impact is used primarily for dummy durability evaluation. Both external and internal deflections are evaluated. Scaled corridors are shown in Figure 5.1– Figure 5.2.

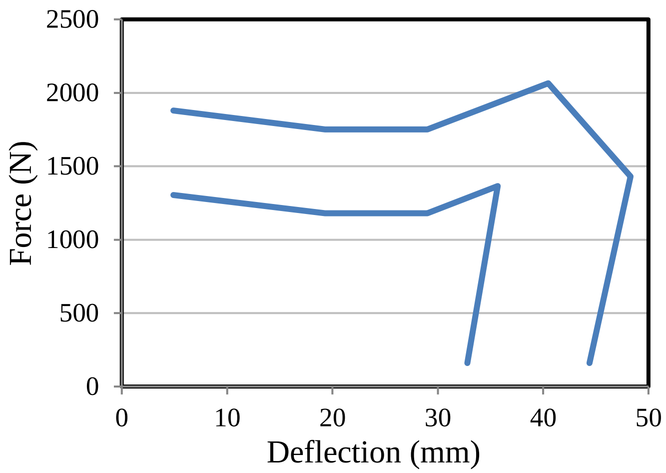

Figure 5.1. The scaled 5th percentile female 4.3 m/s thoracic force versus internal deflection corridor.

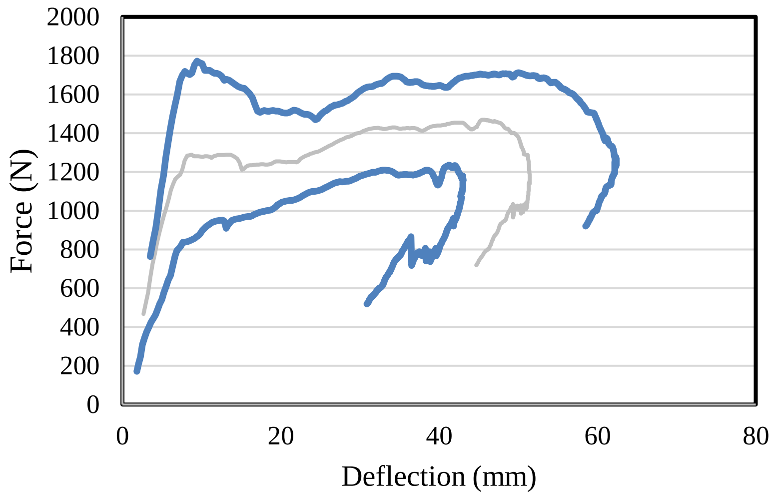

Figure 5.2. The 5th percentile female 4.3 m/s thoracic force versus external deflection corridor (mean curve ± 1 standard deviation).

5.3 Lower Ribcage Oblique Impact Test

This test is based on oblique impacts at the lower ribcage performed by Medical College of Wisconsin (Yoganandan et al., 1997). In these tests, the torso was initially rotated from right to left by 15°, such that the impact occurred on the right antero-lateral thorax (approximately the level of the 8th rib) at a velocity of 4.3 m/s. The weight of the impactor was 23.5 kg. Injuries were sustained in 5 out of 7 PMHS tests. The instrumentation in the tests consisted of a load cell and uniaxial accelerometer attached to the pendulum to measure the impact forces. The external deflection of the thorax was measured with a chest band. The response characteristics of the lower ribcage may be in the form of a force-time corridor and a deflection-time corridor, or a force-deflection corridor.

The impactor used in the THOR-50M oblique biofidelity test is the same as that used in the sternal impact test. For the THOR-05F, the velocity of the test will be consistent with that of the THOR-50M (4.3 m/s) and the impactor mass will match that of the THOR-05F sternal impact test (14.0 kg).

Scaling of Biomechanical Response

The scaling of the biomechanical response to lower thorax oblique impact will use the same scale factors as derived in Section 5.2. Scaled corridors are shown in Figure 5.3 and Figure 5.4.

$\lambda_d = \lambda_x = 0.76$
$\lambda_F = \lambda_k \lambda_x = 0.71$

In addition, time is scaled according to Irwin et al. (2002): $\lambda_t = \sqrt{\lambda_m / \lambda_k} = 0.89$.

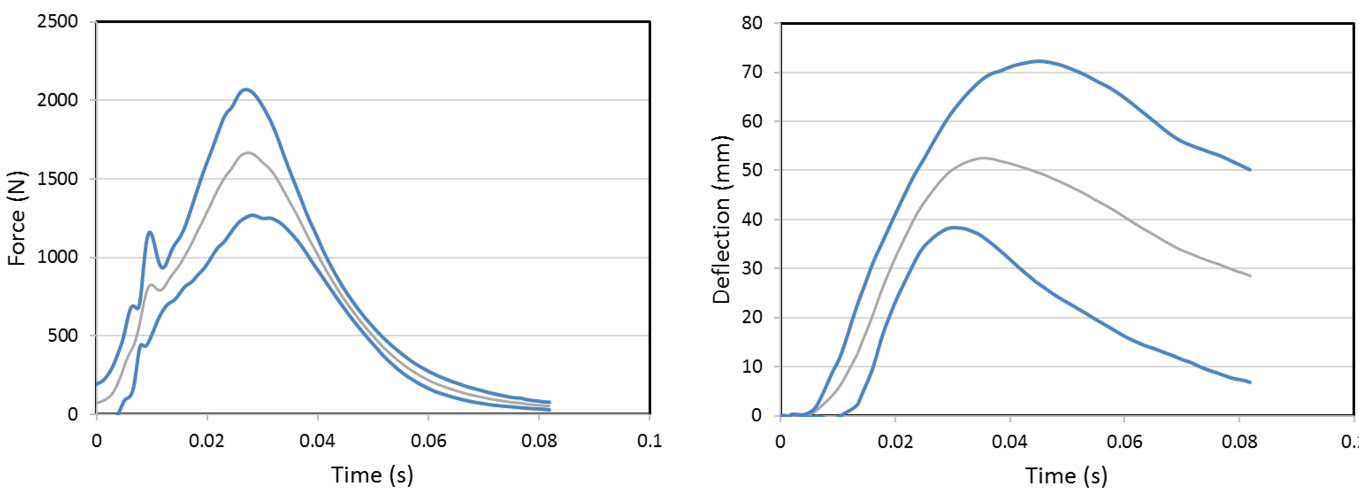

Figure 5.3. Time history corridors for the 5th percentile female in lower thorax oblique impact.

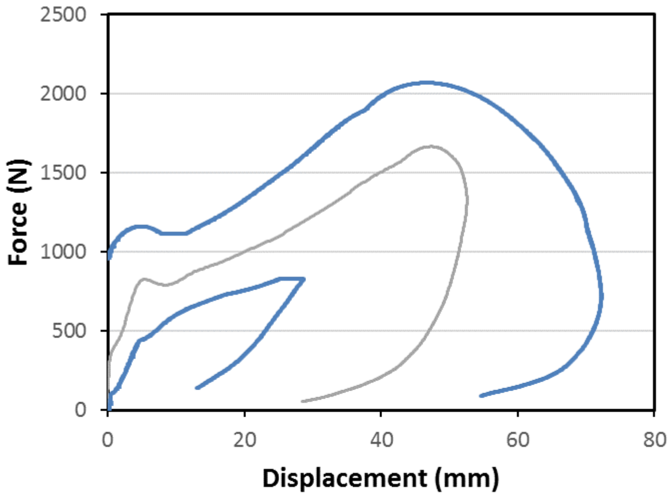

Figure 5.4. The 5th percentile female lower thorax oblique impact corridor for force-deflection, generated using the elliptical method.

6. Shoulder

6.1 Range of Motion/Stiffness Test

The shoulder of the THOR-50M was designed for human-like range of motion (ROM). Human ROM was determined in five male volunteers in three pulling directions, 90° (straight forward), 135° (diagonally upwards), and 170° (upwards) (Tornvall et al., 2005; Tornvall et al., 2007). These pulling angles were chosen to cover the most common angles of the arms with respect to the torso in a frontal collision. Both arms were attached to a steel loading cable by means of a pair of arm brackets, and load was applied in 50 N increments, up to 400 N (200 N per arm). The torso was restrained by a sternum support to prevent torso motion while applying a load to the shoulder. The three-dimensional displacement of the shoulder was determined in each condition using a photo marker placed on the skin surface on the posterior tip of the acromion process.

Shoulder range of motion is not likely to scale geometrically between different sized males and females. Therefore, the male volunteer data will be applied directly to the design of the THOR-05F to ensure that the human-like THOR-50M shoulder design is carried forward into the THOR-05F.

Biomechanical Response

As noted above, the male volunteer data will be applied directly to the design of the THOR-05F. The response is in the form of three dimensional shoulder displacements versus applied pulling force (Figure 6.1).

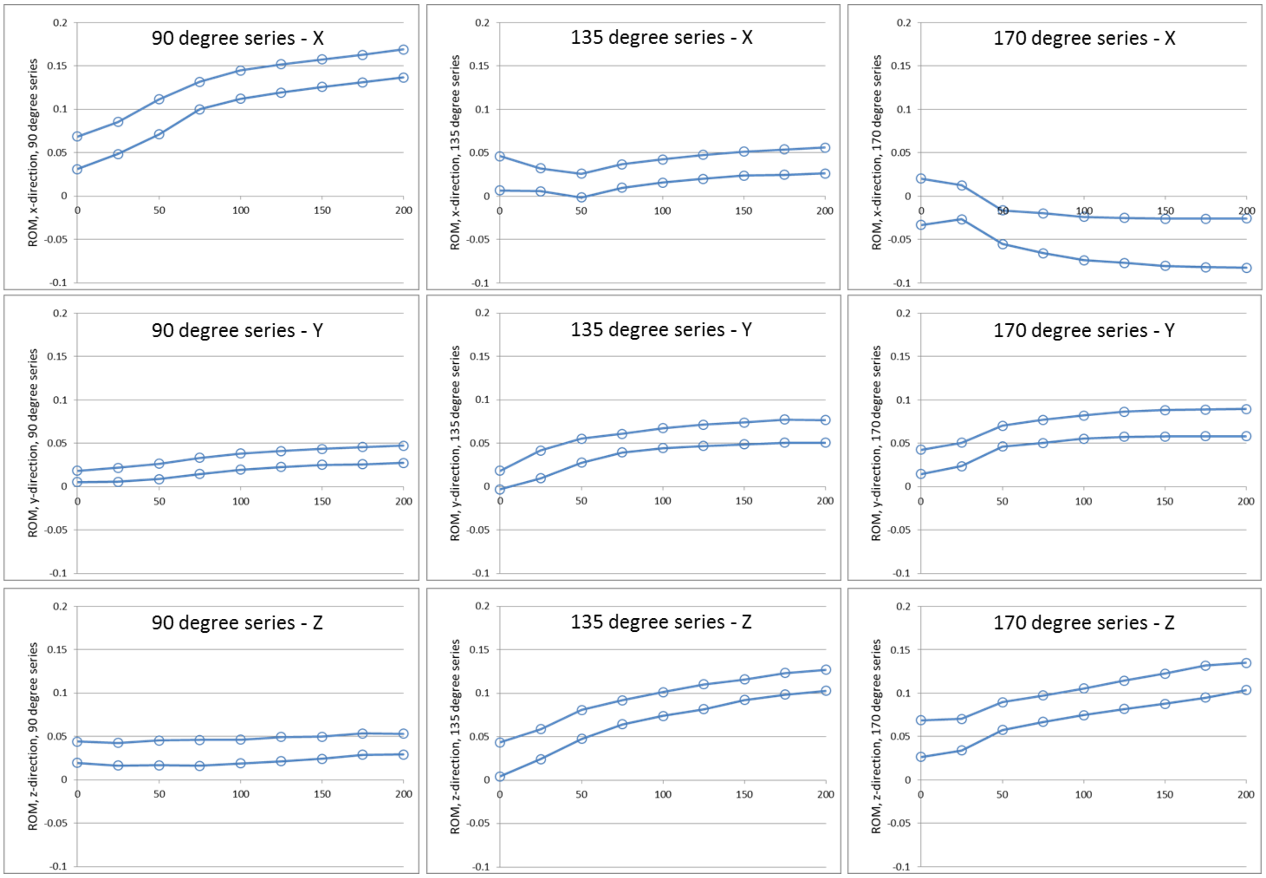

Figure 6.1. Shoulder range of motion corridors from volunteers (mean ± 1 standard deviation) in three pulling configurations (90°, 135°, 170°). Shoulder range of motion was defined by x-, y-, x- displacements of the posterior tip of the acromion.

7. Abdomen

7.1 Scale Factors for Abdomen

The principle dimensions and scale factors for the abdomen (Table 7.1) were determined from the AMVO study by Schneider et al. (1983).

Table 7.1 Scale factors for 5th percentile female abdomen

Measurement	50th Male	5th Female	Scale Factor		Recommended for THOR-05F
Whole body mass (kg) (AMVO, Table I.4)	76.7	46.2	λ_m	0.60	$\lambda_m = 0.50$
Abdomen mass (kg) (AMVO Table 5.8)	2.37	11.6	λ_m	0.68	
Abdominal Breadth (cm)	32.5	27.9	λ_y	0.86	$\lambda_y = 0.86$ $\lambda_x = 0.78$
Abdominal Depth (cm)	26.9	21.0	λ_x	0.78	
Abdominal Circumference (cm)	91.3	75.4	λ_x, λ_y	0.83	
T12 to L5 Distance (cm) (Tables I.3 and I.6)	13.3	9.9	λ_z	0.74	$\lambda_z = 0.74$

Because the Hybrid III had no abdominal measurement capabilities, no abdominal biofidelity targets or scale factors were specified. For the WorldSID small female, abdominal biofidelity targets were scaled according to Irwin et al. (2002). For scaling purposes, the thorax and abdomen were treated as a single segment, using a characteristic length equal to erect seated height ($\lambda_z = 0.895$) and a characteristic mass of either the whole body mass ($\lambda_m = 0.597$) or upper torso mass ($\lambda_m = 0.599$). Scale factors for the transverse plane of the abdomen were assumed to be equal ($\lambda_x = \lambda_y$) and were determined using the constant density relationship (Mertz et al., 1989): $\lambda_m = \lambda_x^2 \lambda_z$.

In frontal impacts, compression (in the local x-direction) is one of the key factors in abdominal injury. Thus, abdominal depth was chosen as the characteristic dimension, rather than erect seated height. Abdominal breadth and height dimensions are also given in Table 7.1.

$\lambda_x = 0.78$
$\lambda_y = 0.86$
$\lambda_z = 0.74$

Using the constant density relationship, the scale factor for abdominal mass can be determined.

$\lambda_m = \lambda_x \lambda_y \lambda_z = 0.50$

Like the thorax (Section 5.1), uncertainty in the estimated density of the partially hollow abdomen may have resulted in uncertainty in the segment mass estimation. Thus, although the resulting mass scale factor differs from the abdominal mass ratio given by AMVO (Table 7.1), scaling the response using dimensional scale factors (which are straightforward to measure and thus are expected to have high accuracy) is expected to yield a biofidelic and achievable corridor.

7.2 Upper Abdomen: Steering Wheel Impact

The response target for the THOR-50M upper abdomen impact is derived from data developed by Nusholtz et al. (1994) based on steering wheel impacts with engagement at the region of L2. The PMHS subjects were seated, with the head and torso supported by a ceiling hoist. Six tests were performed with impact speeds of 3.9 m/s to 10.8 m/s, with an average speed of 8.0 m/s. The average speed of 8.0 m/s has proven to be difficult to practically achieve in dummy labs. Therefore, the test is run at 6.7 m/s.

In order to produce equivalent percent deflection in the 5th percentile female, the impactor mass is scaled from that for THOR-50M ($m_{p,50}$), using the equal stress/equal velocity scaling techniques, by the mass scale factor, λ_m.

$m_{p,5} = \lambda_m \, m_{p,50} = (0.5)(18) = 9.0$

Scaling of Biomechanical Response

The biofidelity target for THOR-50M is a force-deflection response. To represent the response of a small female, both force and deflection must be scaled. To do so, consider that the abdomen impact condition can be represented as a spring-mass system, whereby a rigid pendulum mass, m_p, impacts the abdomen mass, m_a, at a velocity, v.

The stiffness of the abdomen can be written:

$$k = EA/L$$

Elastic modulus, E, is assumed to be constant. Impact area, A, on the abdomen occurs in the y-z plane, and original abdominal depth, L, is in the x direction.

$\lambda_k = \lambda_y \, \lambda_z \,/\, \lambda_x = 0.82$

Peak force can therefore be scaled as: $\lambda_F = \lambda_k \, \lambda_x = \lambda_y \, \lambda_z = 0.64$.

Deflection occurs primarily in the x-direction and can be scaled accordingly: $\lambda_D = \lambda_x = 0.78$.

Impact duration scales as: $\lambda_t = \sqrt{\lambda_m/\lambda_k} = 0.78$.

The corridors for 50th percentile male dummy were re-analyzed by Lebarbe et al. (2015); those are scaled and presented for the 5th percentile female dummy (Figure 7.1, Figure 7.2).

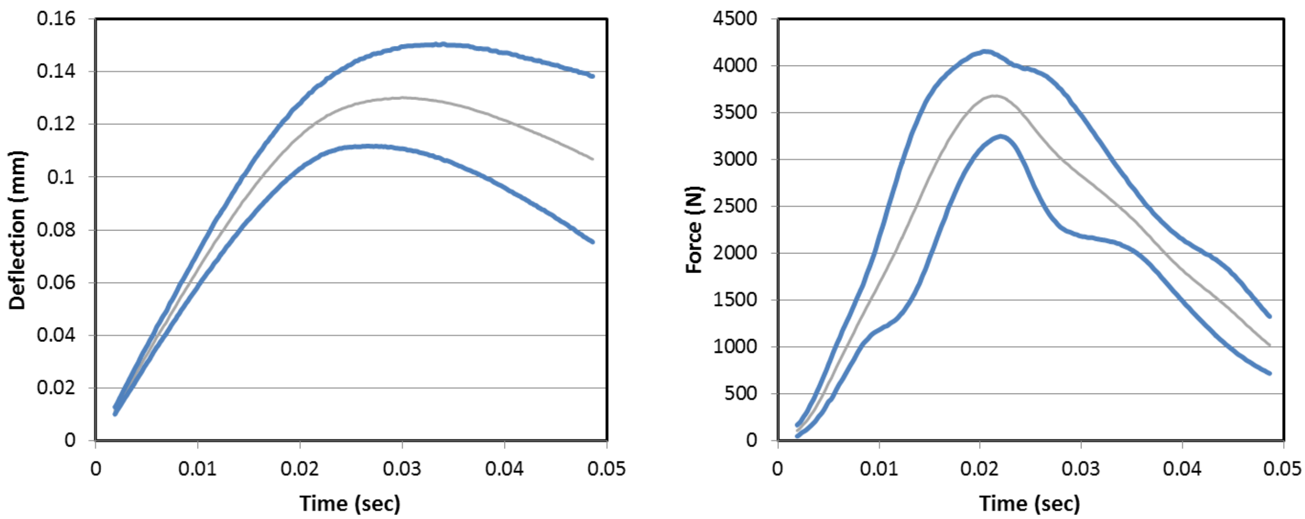

Figure 7.1. Time history corridors (mean ± 1 standard deviation) for upper abdomen steering wheel impact, scaled for the 5th percentile female.

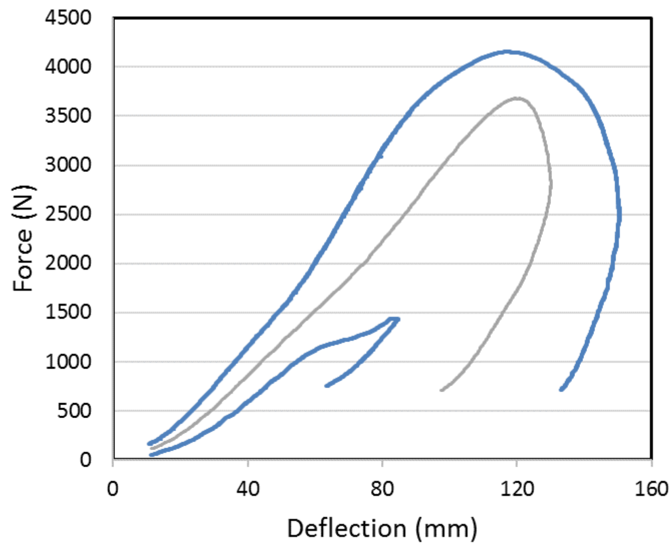

Figure 7.2. Force-deflection corridor (mean ± 1 standard deviation) for upper abdomen steering wheel impact, scaled for the 5th percentile female, generated from elliptical method.

7.3 Lower Abdomen: Rigid Bar Impact

The response targets for lower abdomen impact have been derived from the low severity tests performed by Cavanaugh et al. (1986). The target is in the form of a force versus external deflection corridor and is based on a mean ± 1 standard deviation of the cadaver response. The tests were conducted using a 25 mm diameter rigid bar of length 30 cm and mass 32 kg (i.e., $m_{p,50}$), impacting perpendicularly the abdomen of cadavers at the approximate vertical location of L3 (involving little or no rib contact). Five tests were performed in the speed range of 4.9 to 7.2 m/s with an average impact speed of 6.1 m/s. Only 1 specimen sustained

AIS 3+ injury (liver rupture), however soft tissue injuries are generally unattainable in cadaver tests. Deflection was defined as the difference in horizontal displacement between the impactor and the L3 target.

In order to produce equivalent deflection in the 5th percentile female, the impactor mass is scaled from $m_{p,50}$ by the factor, λ_m.

$m_{p,5} = \lambda_m \, m_{p,50} = (0.5)(32) = 16.0$

Scaling of Biomechanical Response

As described in Section 7.1, the force (λ_F),deflection (λ_D), and impact duration (λ_t) scaling factors are derived using the following formulas:

$\lambda_F = \lambda_y \, \lambda_z = 0.64$
$\lambda_D = \lambda_x = 0.78$
$\lambda_t = \sqrt{\lambda_m/\lambda_k} = 0.78$

The corridors for 50th percentile male dummy were re-analyzed by Lebarbe et al. (2015); those are scaled and presented for the 5th percentile female dummy (Figure 7.3, Figure 7.4).

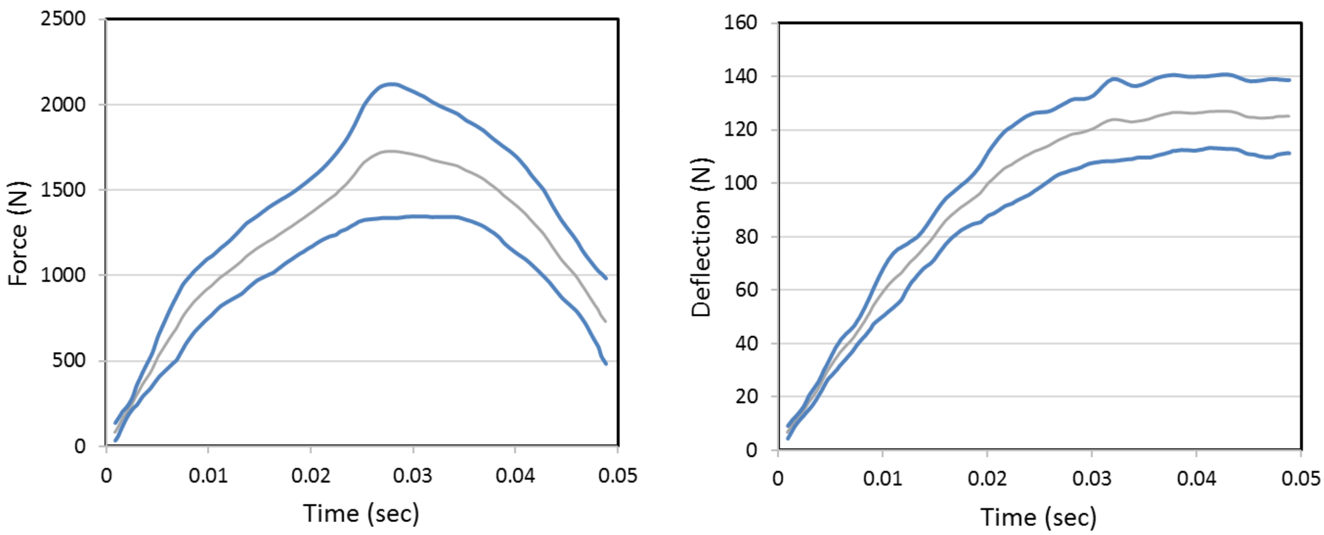

Figure 7.3. 5th percentile female time history corridors (mean ± 1 standard deviation) for lower abdomen rigid bar impact.

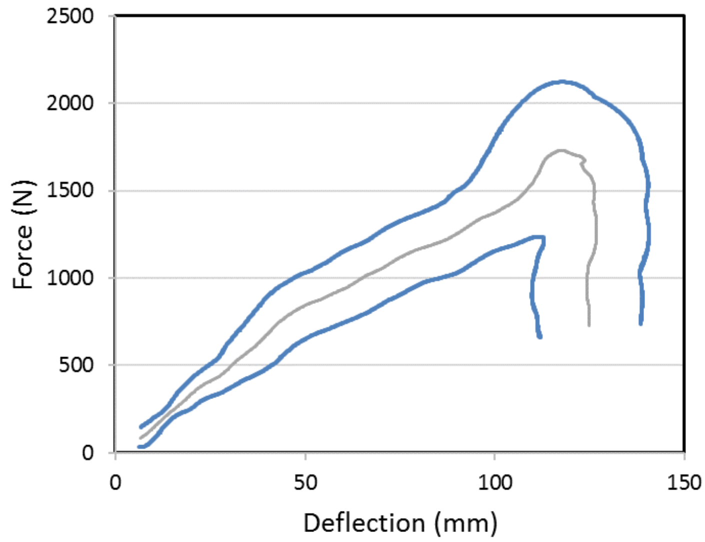

Figure 7.4. 5th percentile female force-deflection corridor (mean ± 1 standard deviation) for lower abdomen rigid bar impact, generated using the elliptical method.

7.4 Abdomen Belt Loading

Although belt loading was not a required test condition for THOR-50M, it is a target for THOR-05F. Belt loading tests replicating submarining conditions have been conducted by Lamielle et al., (2008). In that test series there were five subjects. The test setup consisted of a static rigid seat and back support (fixed back). A 50 mm wide standard seatbelt (elongation of 9% at 10 kN) was initially positioned around the abdomen at the level of the umbilicus (or L2) and oriented horizontally. An initial preload of 20 N was applied to the lap belt. The belt was routed symmetrically rearward on both sides of the subject using a hydraulic cylinder to provide the pull force. The velocity time history is used as the input to the test (Figure 7.5). The response is in the form of a force versus external deflection corridor. Since the input is a velocity-time history, it is not scaled for the 5th percentile female.

Scaling of Biomechanical Response

As described above, force, deflection, and impact duration are scaled as follows:

$\lambda_F = \lambda_y \lambda_z = 0.64$
$\lambda_D = \lambda_x = 0.78$
$\lambda_t = \sqrt{\lambda_m/\lambda_k} = 0.78$

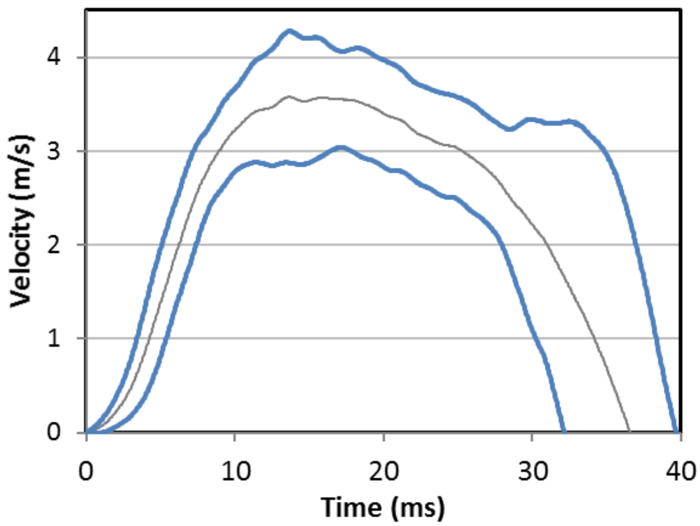

Figure 7.5. Belt strand velocity input for belt pull test (mean ± 1 standard deviation).

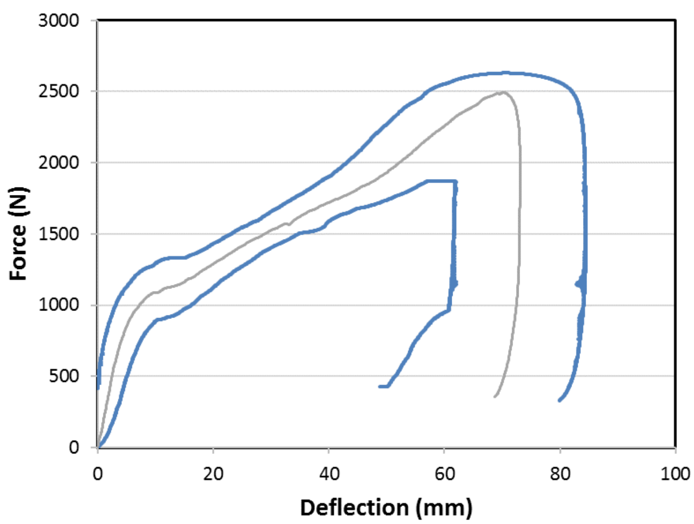

Figure 7.6. Force-deflection corridor for abdomen belt pull test (mean ± 1 standard deviation), up to point of peak force, scaled for 5th percentile female.

8. Lumbar Spine

8.1 Scale Factors for Lumbar Spine

The principle dimensions and scale factors for the lumbar spine are the same as the dimensions used for the abdomen (Table 7.1) and were determined from the AMVO study by Schneider et al. (1983).

$\lambda_x = 0.78$
$\lambda_y = 0.86$
$\lambda_z = 0.74$

8.2 Lumbar Spine Flexion: Pendulum Test

This lumbar flexion target is based on tests conducted in 2015-2016 during evaluation of THOR-50M. The target is a range for peak flexion moment and peak lumbar rotation angle. The purpose of conducting this test for the THOR-05F is to ensure that the flexion response is similar to the THOR-50M. For this test, the full upper thorax is installed on the neck pendulum (49 CFR §572.33(c)3). The head and neck mass are replaced with rigid bar and mass of 2.51 kg. The target velocity of the pendulum is 2.0 m/s and the velocity profile should meet the targets of Table 8.1. Hexcell configuration is 8 cells x 3 cells x 6”. The accelerometer and angular rate sensor (ARS) on the pendulum arm and the thoracic spine load cell flex joint adapter plate are used to define lumbar spine rotation.

Table 8.1. Velocity specification for lumbar spine flexion test

Specification (Mean $\pm$2 S.D.)	Lower	Upper
Velocity at 5 ms (m/s)	0.363	0.441
Velocity at 15 ms (m/s)	1.162	1.331
Velocity at 25 ms (m/s)	1.743	2.005

Scaling of Mechanical Response

To represent the response of a small female ATD, assuming that the lumbar spine acts as a cantilever beam with a point load, F, applied to the end, the resistive moment, M, can be expressed using a basic bending formula:

$$M = \sigma I / c,$$

where σ is beam stress, c is the distance of the farthest muscle fiber in the lumbar region and I is the moment of inertia ($I = \frac{1}{4}\pi ab^3$ for an elliptical cross section, where “a” scales with λ_y and both “b” and “c” scale with λ_x). Since stress is not scaled, per the equal stress/equal velocity scaling method, dimensional analysis demonstrates that the scaling factor for moment, M, simplifies to:

$\lambda_M = \lambda_y \lambda_x^2 = 0.52$

Similarly, standard beam analysis provides a formula for the angular deflection, θ, of the cantilever beam.

$$\theta = \frac{FL^2}{2EI} = \frac{ML}{2EI}$$

Again, dimensional analysis demonstrates that the scaling factor (λ_θ) for lumbar rotation angle, θ, simplifies to:

$\lambda_\theta = \lambda_z / \lambda_x = 0.95$

The target is a range for peak flexion moment and peak lumbar rotation angle (Table 8.2). This target is not included in the BioRank calculation.

Table 8.2. Scaled specifications for THOR-05F lumbar spine flexion test

	Y-axis Moment (Nm)	**Rotation (deg)**
Upper Limit (Mean + 2 S.D.)	70.9	-16.2
Lower Limit (Mean – 2 S.D.)	62.2	-19.4

9. Knee-Thigh-Hip Complex

9.1 Scale Factors for Knee-Thigh-Hip

The principle dimensions and scale factors for the femur were determined from the AMVO study by Schneider et al. (1983).

Table 9.1. Scale factors for 5th percentile female femur

Measurement	50th Male	5th Female	Scale Factor		Recommended for THOR-05F
Thigh mass (kg) (AMVO Table 5.8)	8.61	5.91	λ_m	0.68	$\lambda_m = 0.68$
Troc.-to-Lat. Fem. Condyle (cm) (seated)	44.7	38.1	λ_z	0.85	
Thigh Breadth (cm) (upper)	19.4	17.6	λ_y	0.91	
Thigh Circumference (cm) (upper)	57.9	50.1	λ_x, λ_y	0.87	$\lambda_x = \lambda_y = 0.89$ $\lambda_z = 0.85$
Thigh Breadth (cm) (mid)	15.5	12.5	λ_y	0.81	
Thigh Circumference (cm) (mid)	50.4	42.7	λ_x, λ_y	0.85	
Knee Circumference (cm)	39.2	33.9	λ_x, λ_y	0.86	

Mass scaling for the Hybrid III was accomplished using an “upper leg weight” scale factor of 0.60. Since the thigh mass values specified by Mertz et al. (1989) differ from the specifications provided in the AMVO, the basis for the Hybrid III scale factor is not clear. For the THOR-05F, consistency with 5th percentile female anthropometry was a priority. Thus, despite the departure from past precedent, the thigh mass for the 50th male and 5th female reported in AMVO will be used to scale mass properties for THOR-05F.

$\lambda_m = 0.68$

For the Hybrid III, buttock-to-knee length (in a standing position) was chosen as the characteristic length of the upper leg/femur (between the 5th females and the 50th males i.e., $\lambda_z = 0.88$). Examination of AMVO specifications (Table 9.1) demonstrates that, in a seated posture, femur length (defined by the distance from trochanter to lateral femoral condyle) has a scale factor of 0.85. To maintain consistency with a current understanding of 5th percentile female anthropometry, this dimension is used to scale the femur biofidelity response for the THOR-05F dummy.

$\lambda_z = 0.85$

Historically in dummy design, scale factors for the transverse plane of the femur were assumed to be equal ($\lambda_x = \lambda_y$) (Mertz et al., 1989; Irwin et al., 2002). Based on AMVO specifications, this assumption appears reasonable. Assuming a constant density relationship (Mertz et al., 1989), the transverse plane scale factors are defined by:

$$\lambda_m = \lambda_x^2 \lambda_z$$
$$\lambda_x = \lambda_y = \sqrt{\lambda_m/\lambda_z} = 0.89$$

In the local transverse plane, thigh breadth and thigh circumference scale factors range from 0.81 to 0.91 (Table 9.1). Thus, the calculated value for λ_x and λ_y are considered reasonable in comparision to the dimensional scale factors given by AMVO.

9.2 Knee-Thigh-Hip Complex Impact Test

This target defines the response of the femur (KTH complex) to axial impacts at the knee in a fixed femoral head boundary condition that allows for the characterization of knee/femur responses independent of inertial effects. The biomechanical response is based on the tests by Rupp et al. (2003). They conducted dynamic impacts of the femur using a special test apparatus (Figure 9.1) and an impactor designed to be significantly more massive than the knee-thigh-hip complex. All but one PMHS specimen sustained injury, and three subjects were small stature females. An alternative test apparatus can be used for this test as long as it can be demonstrated it would produce equivalent responses.

The primary target is the force-deflection response of the femur and is based on the mean ± 1 standard deviation of the cadaver response. As long as the impactor mass is substantially greater than the KTH complex, force-deflection response is not expected to change with impactor mass. Furthermore, there is no specification for peak values of force or deflection in this test, nor is there a concern about damaging the 5th percentile female femur because the compressive element allows sufficient stroke to represent a 5th percentile female response. As such, mass scaling of the impactor is not required.

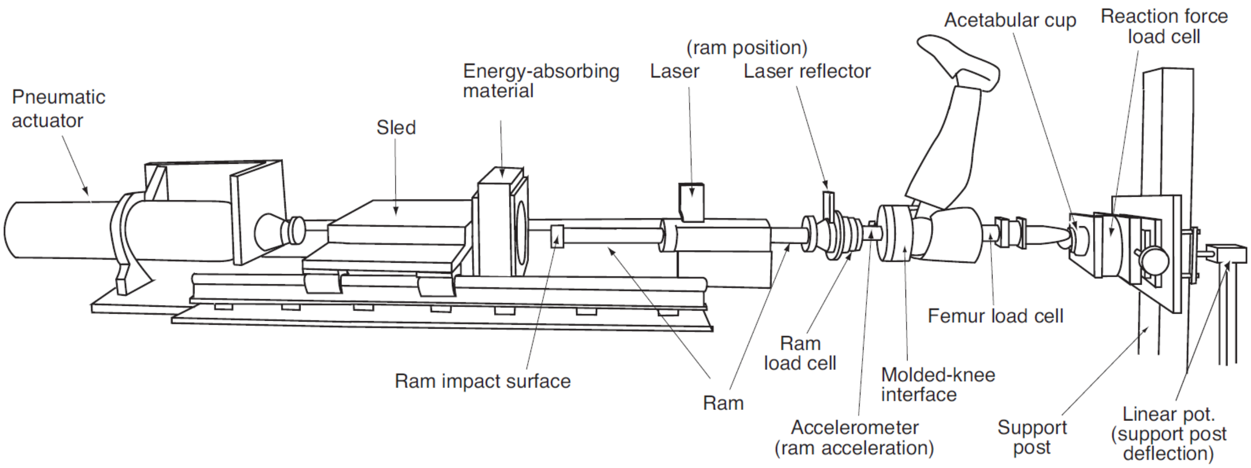

Figure 9.1. Apparatus used for dynamic femur response assessment (courtesy of Rupp et al., 2003).

Scaling of Biomechanical Response

Deflection occurs along the z-axis of the femur (i.e., $\lambda_d = \lambda_z$), and the scaling for the Knee-Thigh-Hip impact test is given by:

$\lambda_k = \lambda_x^2 / \lambda_z = 0.93$
$\lambda_d = \lambda_z = 0.85$
$\lambda_F = \lambda_k \lambda_z = (\lambda_x^2 / \lambda_z) \lambda_z = \lambda_x^2 = 0.79$

Scaled time history corridors and force-deflection corridors are shown in Figure 9.2 and Figure 9.3, respectively.

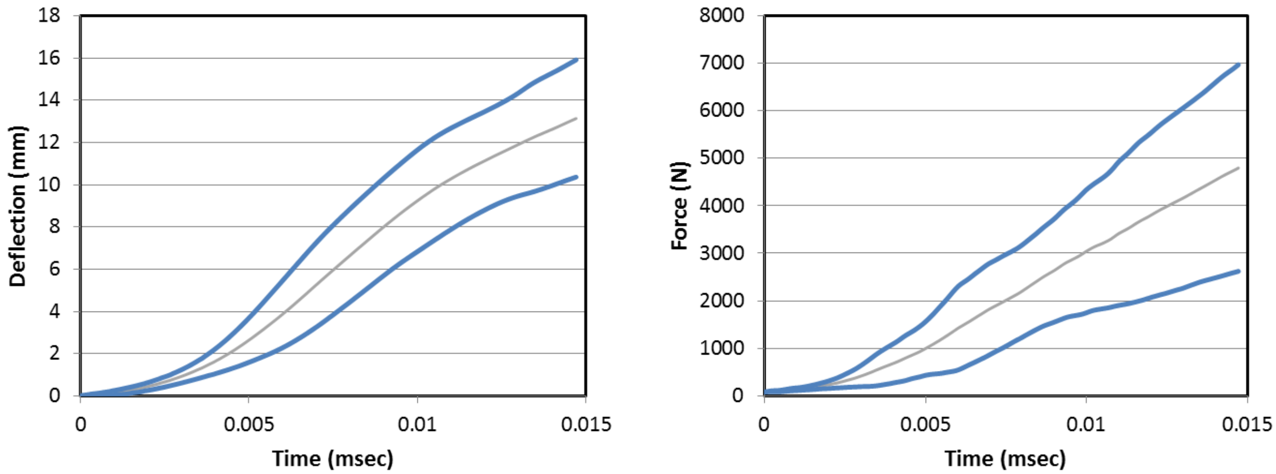

Figure 9.2. 5th percentile female scaled time history corridors for the isolated knee-thigh-hip impact test.

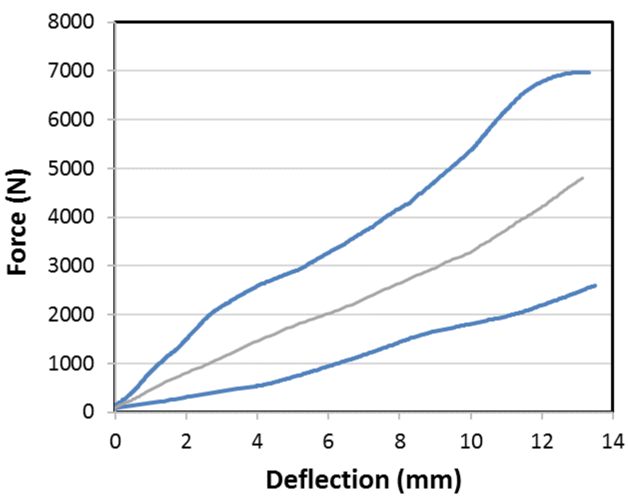

Figure 9.3. 5th percentile female scaled force-deflection corridors for the isolated knee-thigh-hip impact test, generated using the elliptical method.

9.3 Whole Body KTH Test

This target is new for the THOR-05F and defines the response of the whole KTH complex to axial impacts at the knee. This test is based on the series conducted by Rupp et al. (2008). In this series, symmetric loading was applied to the left and right knees of five seated cadavers. A 255 kg ram was pneumatically impacted into the knees at velocities of 1.2 m/s, 3.5 m/s and 4.9 m/s. No specimens sustained any injuries during these tests. Knee impact surfaces were padded to reduce the likelihood of knee injuries and lengthen the duration of knee loading. Force at the hip was estimated by multiplying the anteroposterior acceleration of the femur by the mass between the femur load cell and the acetabulum (about 0.7 kg for the male subjects).

The targets are the impactor force-time response (Figure 9.4), force-time response of the acetabulum (Figure 9.5) and pelvis acceleration-time response (Figure 9.6). These are based on the mean ± 1 standard deviation of the cadaver response. As noted in Section 9.2, as long as the impactor mass is substantially greater than the KTH complex, force-deflection response is not expected to change with impactor mass. Furthermore, there is no specification for peak values of force or deflection in this test, nor is there a concern about damaging the 5th percentile female femur because the compressive element allows sufficient stroke to represent a 5th percentile female response. As such, mass scaling of the impactor is not required. The impact velocity of this test was initially chosen to be 4.9 m/s, because it produced acetabulum loading consistent with provisional injury assessment reference values in THOR-50M. However, in preliminary testing with the THOR-05F, the test fixture sustained damage due to the severity of the loading condition. Thus, the target speed was reduced to the 3.5 m/s condition.

Scaling of Biomechanical Response

Because deflection occurs along the z-axis of the femur ($\lambda_d = \lambda_z$), the scaling for the Whole Body Knee-Thigh-Hip impact test is the same as that described in Section 9.2.

$\lambda_k = \lambda_x^2 / \lambda_z = 0.93$
$\lambda_d = \lambda_z = 0.85$
$\lambda_F = \lambda_k \lambda_z = (\lambda_x^2 / \lambda_z) \lambda_z = \lambda_x^2 = 0.79$
$\lambda_t = \sqrt{\lambda_m / \lambda_k} = 0.86$

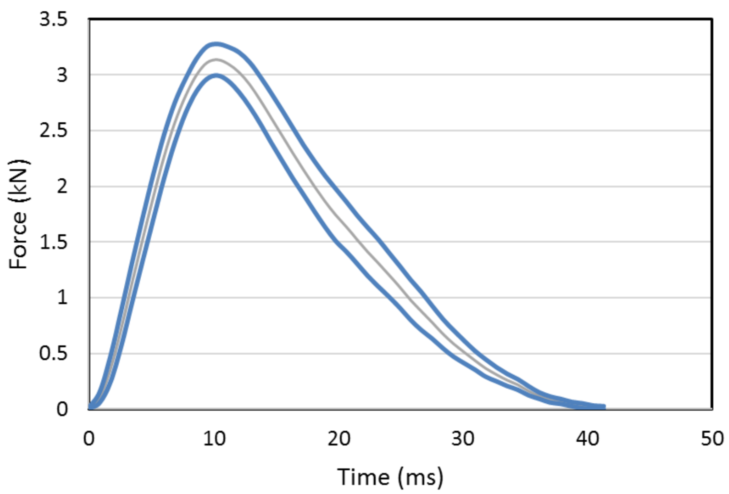

Figure 9.4. 5th percentile female impactor force-time corridor for the whole body KTH test (impact speed 3.5 m/s).

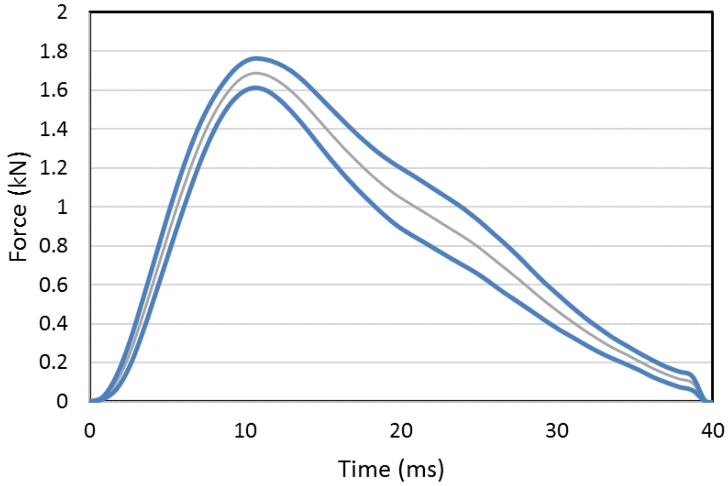

Figure 9.5. 5th percentile female acetabulum force-time corridor for the whole body KTH impact test (impact speed: 3.5 m/s).

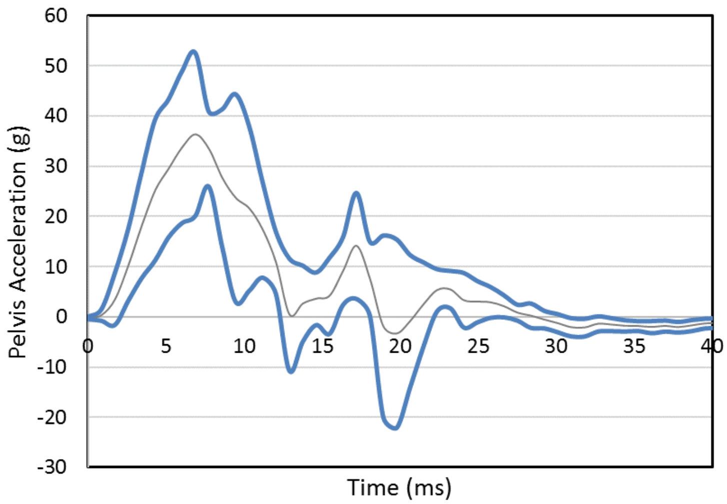

Figure 9.6. 5th percentile female pelvis acceleration-time corridor for the whole body KTH test (impact speed 3.5 m/s).

9.4 Knee Slider Test

The knee response is based on the study of Balasubramanian et al. (2004), who replicated and expanded on earlier data (Viano et al., 1978). Knee specimens were tested at 90° of flexion in an anteroposterior drawer motion of the tibia. Specimens were fixed in place to prevent off-axis loading. Tests were conducted using a servocontrolled INSTRON testing machine with a custom fixture that allowed smooth translation of the tibia rearward with respect to the femur. Specimens were tested with speed range between 1.3 and 2.5 m/s. One option for testing the biofidelity of the THOR-05F knee slider is to use an identical test setup as the PMHS tests with a servocontrolled actuator to translate the dummy tibia rearward.

An alternative method to test knee slider biofidelity is to use the Hybrid III Knee Slider impact Test Procedure (SAE J2856 – Users Manual for the 50th Percentile Male Hybrid III Dummy). The test fixture consists of a rigid test probe and a method of rigidly supporting the knee assembly. A load distribution bracket transmits the impact energy into the slider assembly. The test probe mass for THOR-50M is 12.0 kg ± 0.02 kg, including instrumentation, rigid attachments, and the lower 1/3 of the suspension cable mass. The diameter of the impacting face is 76.2 mm ± 0.3 mm (3.0 in ± 0.01 in) with an edge radius of 0.5 mm (0.02 in).

The original impact velocity was 2.75 ± 0.05 m/s. Preliminary testing with the THOR-05F indicated that the compression rate achieved during a 2.75 m/s impact was well above the compression rate of the PMHS tests. A test velocity of 2.15 m/s was therefore used. This reduced velocity is consistent with the velocity currently being used in THOR-50M qualification tests.

To avoid overloading the knee slider resistive element in the THOR-05F, the pendulum mass is scaled down by the average of the upper and lower leg mass scale factors. This same pendulum scaling procedure was done on the Hybrid III small female (Mertz et al., 1989), although the scale factor was different (as noted in Section 9.1).

$m_{p,5} = \lambda_m m_{p,50} = (0.67)\ (12\ kg) = 8.0\ kg$

The probe used for the Hybrid III 5th ATD is 7.26 kg. The difference in knee force at 13 mm, using a 7.26 kg probe, as compared to an 8.0 kg probe, was estimated to be less than 100 N. Given the minimal difference, the 7.26 kg Hybrid III small female probe can be used.

Scaling of Knee Slider Biomechanical Response

In the human, the anteroposterior drawer motion of the tibia is resisted primarily by the posterior cruciate ligament (PCL). Thus, the resistance to motion is dependent on the size and elastic response of the PCL. Scaling for the Hybrid III knee slider was accomplished by assuming that the length of the PCL is primarily oriented in the x-direction ($\lambda_L = \lambda_x$) and that the cross-section of the ligament lies in the y-z plane ($\lambda_A = \lambda_y\lambda_z$). In fact, the PCL does not elongate purely in either the x- or z- direction relative to either the femur or tibia, and its orientation changes with knee flexion angle. Thus, for the present analysis, the ligament length is assumed to be in the x-z plane and the scale factor is determined from an average of the x- and z- scale factors ($\lambda_L = (\lambda_x + \lambda_z)/2$). For PCL cross-sectional area, knee circumference (Table 9.1) is selected as the characteristic dimension for scaling the knee slider (PCL) response.

$\lambda_x = \lambda_y = 0.865$
$\lambda_z = \lambda_m / (\lambda_x \lambda_y) = 0.90$

The stiffness of the knee slider is given by:

$k = EA/L$

Assuming the elastic modulus is equal for small females and midsize males, scaling for knee stiffness and force can be represented by:

$$\lambda_k = \frac{\lambda_x \lambda_y}{(\lambda_x + \lambda_z)/2} = 0.85$$

$$\lambda_d = \frac{(\lambda_x + \lambda_z)}{2} = 0.88$$

$$\boldsymbol{\lambda_F} = \boldsymbol{\lambda_k \lambda_d} = \frac{\lambda_x \lambda_y (\lambda_x + \lambda_z)/2}{(\lambda_x + \lambda_z)/2} = \boldsymbol{\lambda_x \lambda_y} = 0.75$$

The original biofidelity criteria for the THOR-50M consist of pass-through force corridors at specific levels of displacement. Since that time, full time-history corridors were created from the original PMHS data (Balasubramanian et al. (2004). For these corridors, only the seven subjects who sustained injury by PCL tear were included, because the elongation of the PCL is most similar to the displacement of the ATD knee slider. These 50th percentile male corridors are therefore scaled for the 5th percentile female (Figure 9.7).

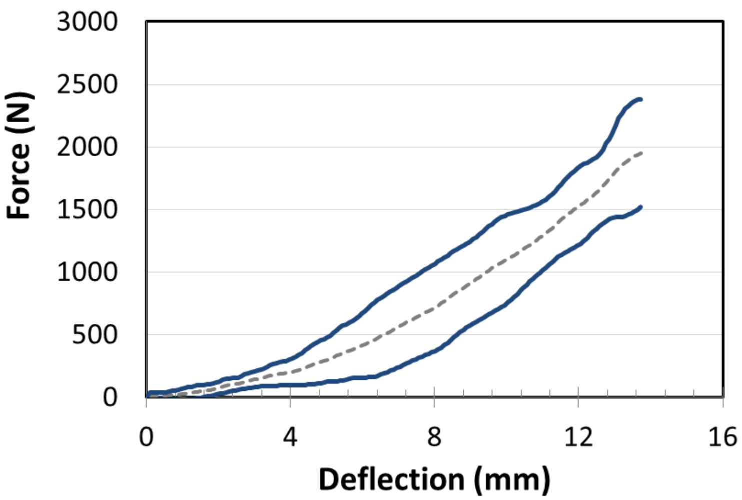

Figure 9.7. 5th percentile female corridor for knee slider response.

10. Lower Extremity (Below Knee)

10.1 Scale Factors for Leg-Foot-Ankle

The principle dimensions and scale factors for the lower extremity (below knee) were determined from an anthropometric specification from the AMVO study (Schneider et al., 1983) and described by Shams et al. (2002).

Table 10.1. Scale factors for 5th percentile female leg

Measurement	**50th Male**	**5th Female**	**Scale Factor**	
Leg mass (kg) (AMVO Table 5.8)	3.59	2.36	λ_m	0.66
Calf Breadth (cm)	11.0	9.4	λ_y	0.85
Calf Depth (cm)	11.8	9.6	λ_x	0.81
Calf Circumference (cm)	37.3	31.5	λ_x, λ_y	0.84
Tibiale to Sphyrion distance (cm) (x-z plane AMVO Tables I.3 and I.6)	38.0	31.9	λ_z	0.84

Table 10.2. Scale factors for 5th percentile female foot and ankle

Measurement	**50th Male**	**5th Female**	**Scale Factor**	
Foot mass (kg) (AMVO Table 5.8)	0.98	0.64	λ_m	0.65
Foot Breadth (cm)	9.6	88.6	λ_y	0.90
Foot Length (cm)	26.4	22.1	λ_x	0.84
Ankle Breadth (cm) (condyles)	7.3	6.3	λ_y	0.86
Ankle Depth (cm) (condyles)	9.4	8.1	λ_x	0.86
Ankle Circumference (cm) (condyles)	26.1	22.0	λ_x, λ_y	0.84

Anthropometric specifications (Table 10.1, Table 10.2) demonstrate that most of the scale factors in x, y, and z are between 0.84 and 0.86. In other words, the 5th percentile female lower extremity is approximately geometrically similar to the fiftieth male lower extremity. Thus, the lower extremity (below knee) biofidelity targets were scaled by a single factor:

$\lambda_x = \lambda_y = \lambda_z = 0.85$ (the average of 0.84 and 0.86)

The appropriate mass scale factor was then determined using the constant density relationship:

$\lambda_m = \lambda_x^3 = 0.61$

10.2 Axial Heel Impact Test

For the THOR-50M and early designs of the THOR-05F lower extremity (previously known as the “FLx”), the response of the lower leg to impact on the plantar surface of the foot was based on the tests conducted at the Medical College of Wisconsin (Kuppa et al., 1998). However, the test procedure used to evaluate dummy biofidelity/certification was inconsistent with the original PMHS test procedure. Newer PMHS axial foot impact data with 5th female specimens is available from Forman et al. (2018) and in the NHTSA Biomechanics database (test numbers 11849-11860). Unlike the earlier tests (Kuppa et al., 1998), in the Forman et al. (2018) tests the musculature and the tibia/fibula natural motion was left intact at the knee joint by severing the leg above the knee. These new tests were conducted both with and without shoes. The proposed corridors are based on only the shod tests.

A test apparatus was constructed to deliver dynamic axial impact loads to the plantar surface of the foot of a cadaver specimen via a compound pendulum. The pendulum struck a padded transfer piston which directed the impact to pure horizontal translation. The leg specimens were placed horizontally in the test rig with the ankle in a neutral orientation and the knee constrained in an adjustable block. The effective mass of the pendulum was 28.4 kg and the impact velocity was 3.1 m/s for all tests.

The time history corridors are shown in Figure 10.1.

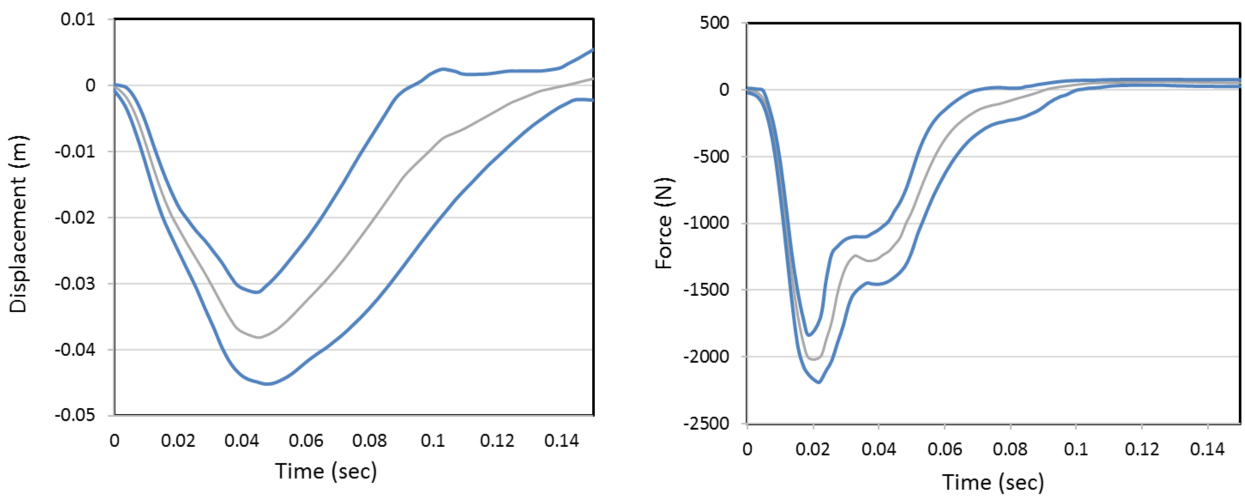

Figure 10.1. 5th percentile female time history corridors for tibia force (left) and footplate intrusion displacement (right).

10.3 Dynamic Dorsiflexion (Ball of Foot Impact)

The current dynamic dorsiflexion response is given by the moment versus angle characteristics at the ankle, when the ball of the foot is impacted. The THOR-50M response was obtained from cadaver tests conducted at Renault and summarized by Crandall et al. (1996). Because the PMHS tests were conducted with bare feet, and the design of the THOR-05F incorporates a molded shoe, experimental work is ongoing to determine the effect of a shoe on dynamic dorsiflexion response.

Scaling of Biomechanical Response

The resistive moment at the ankle is generated by the sum of all of the ligaments/tendons that support the joint. Each ligament generates a moment that can be expressed as a function of its cross-sectional area, A, and moment arm, d.

$M = Fd = \sigma Ad$

Under the assumptions of the equal stress/equal velocity scaling technique, stress is equal for the small female and midsize male. When all dimensions of the lower leg are assumed to scale equally (Section 10.1), the scale factor for ankle moment simplifies to:

$\lambda_M = \lambda_x^3 = 0.61$

Ankle rotation is assumed to take place at a point on the ankle (essentially along a pin axis), implying that ankle rotation angles are not scaled for body size (Shams et al., 2002). In other words, $\lambda_\theta = 1$. The scaled corridor is shown in Figure 10.2.

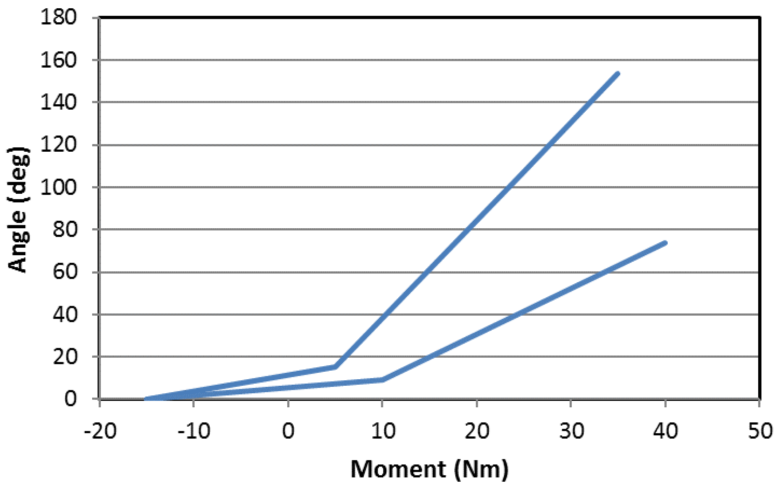

Figure 10.2. Ankle dorsiflexion angle versus moment corridor, scaled for 5th percentile female.

10.4 Inversion/Eversion

Given that THOR-05F will be used primarily in dynamic loading, it is preferable to conduct a dynamic biofidelity test in inversion/eversion. Funk et al. (2002) conducted dynamic inversion/eversion tests on 26 PMHS specimens. A test apparatus was constructed to dynamically (~1000º/s) apply pure moments about the inversion-eversion axis of the foot through the subtalar joint center. It incorporated an adjustable footplate free to rotate about an offset, fixed vertical axis. Rotation of the footplate was driven by a pneumatic impactor striking a guided cam on one side of the plate and controlled by honeycomb crush inside a piston on the other side. PMHS tests were conducted with and without an applied axial preload. For ATD testing, it is recommended to apply pure moment without axial preload. The response corridors are in the form of ankle joint moment versus angle (Figure 10.3 and Figure 10.4).

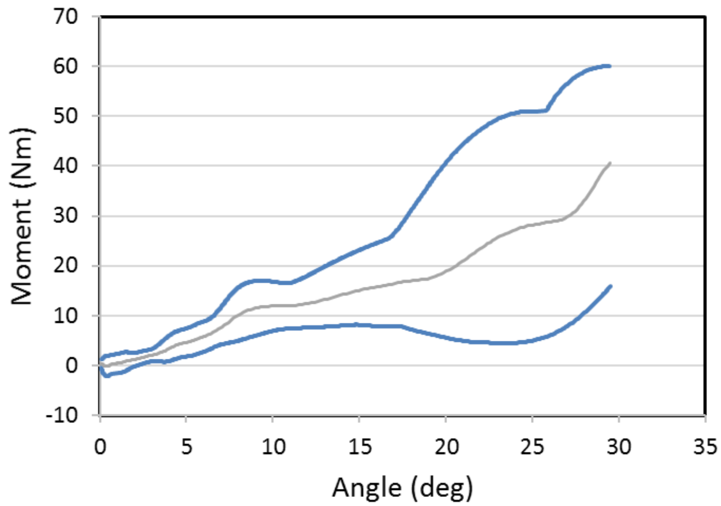

Figure 10.3. Corridor for 5th percentile female ankle in dynamic eversion, generated using elliptical method.

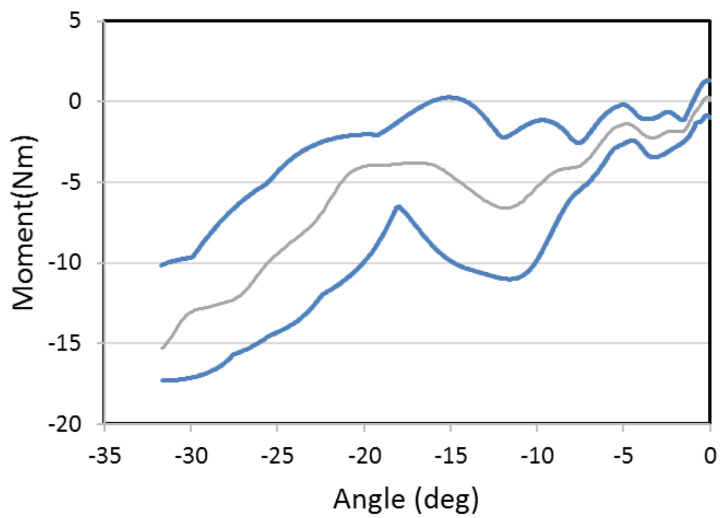

Figure 10.4. Corridor for 5th percentile female ankle in dynamic inersion, generated using elliptical method.

11. Full Body Sled Test

This test procedure was based on a series of ten PMHS tests. The sled velocity was 30 km/hr and belt loads were force-limited to 2 kN (The first five tests are available using NHTSA Biomechanics Database test numbers 11491-11495). In these tests, the subjects were positioned on a rigid planar seat with their torso and head supported by an adjustable matrix of cables to approximate the seated posture of a right front passenger. The restraint system consisted of a custom 3-point shoulder and lap belt with anchor positions approximating those found in a typical mid-size sedan. Prior to the test, the shoulder and lap belt were pre-tensioned to 5 N and 50 N, respectively. Pelvis and lower extremity movements were restricted by a stiff (aluminum) knee bolster adjusted to be in contact with the proximal tibias at the time of impact and by an aluminum footrest with ankle straps. Belt loads were measured with attached load cells. Head and spine kinematics, as well as chest deflection (in four locations: upper right, upper left, lower right, lower left) were measured using VICON motion capture. Corridors for chest deflection, belt loads, fixture reaction forces, and head and spine kinematics were developed and reported in Crandall et al. (2019). Corridors are shown in Figure 11.1 through Figure 11.9.

For comparison to the PMHS data, the THOR-05F can be tested in the same test configuration, using equivalent external instrumentation. Motion tracking targets can be placed on the ATD in the same locations as the PMHS, providing a direct kinematic comparison. For chest deflection, the motion tracking targets on the PMHS were placed in locations equivalent to the THOR-05F internal deflection instrumentation. VICON markers may also be used externally on the chest of the ATD.

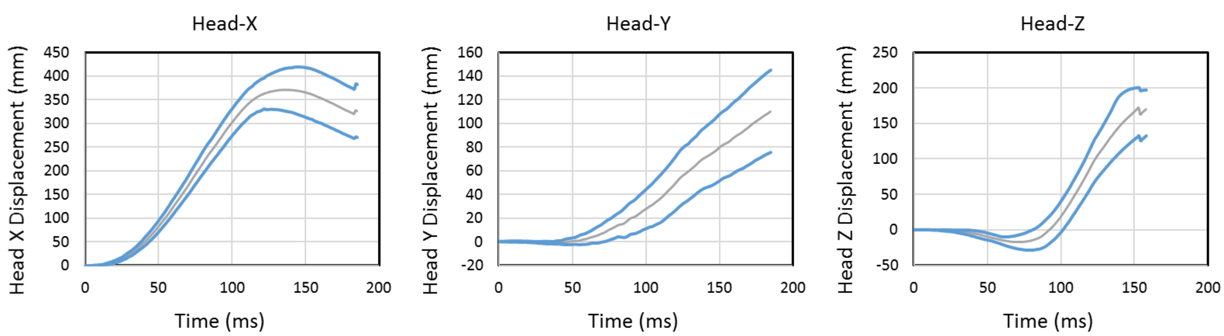

Figure 11.1. Small female sled test: Head kinematics corridors.

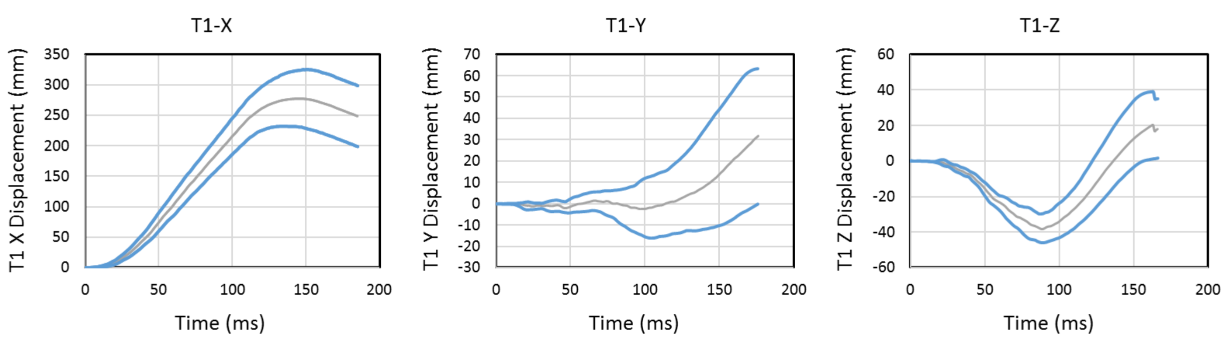

Figure 11.2. Small female sled test: T1 kinematics corridors.

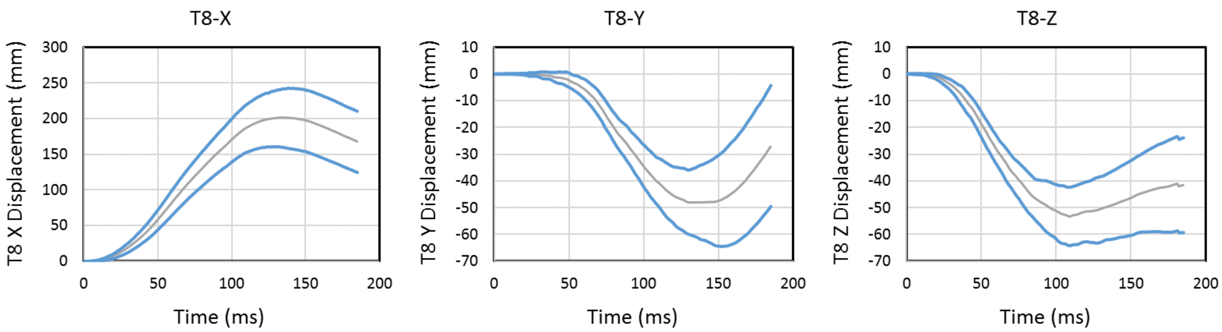

Figure 11.3. Small female sled test: T8 kinematics corridors.

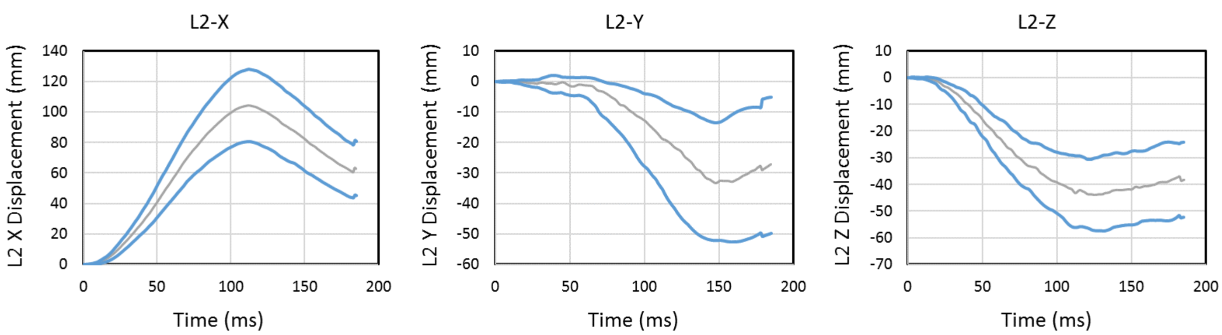

Figure 11.4. Small female sled test: L2 kinematics corridors.

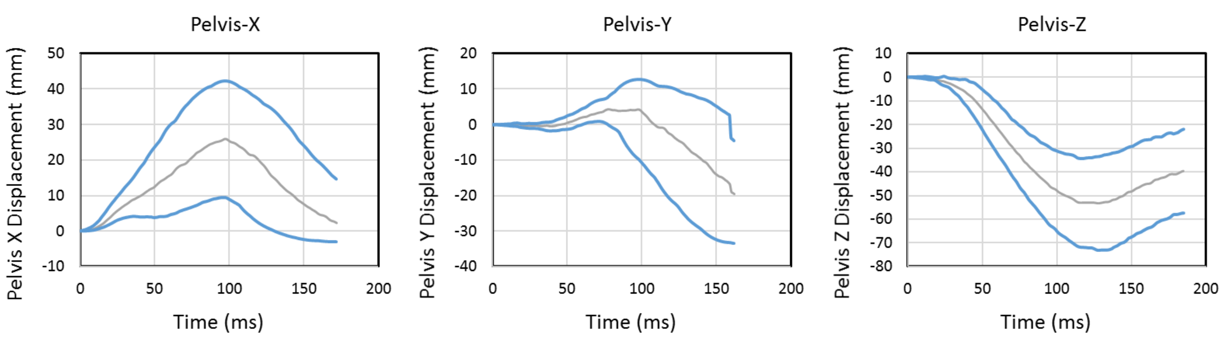

Figure 11.5. Small female sled test: Pelvis kinematics corridors.

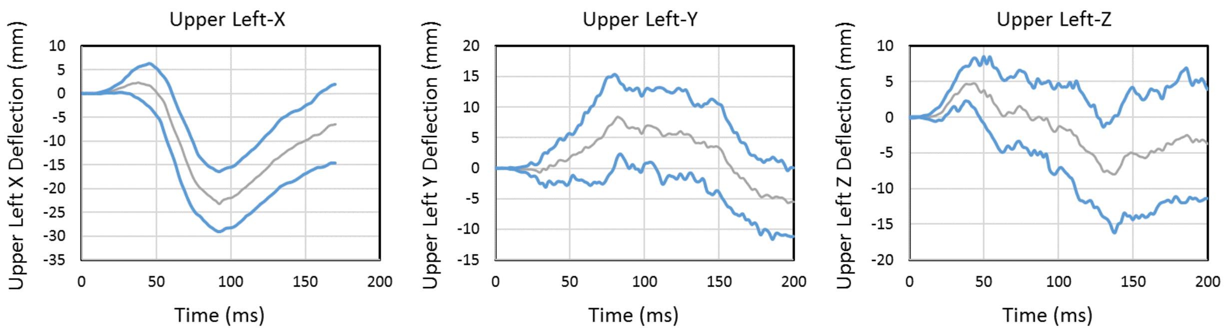

Figure 11.6. Small female sled test: Chest deflection corridors at upper left.

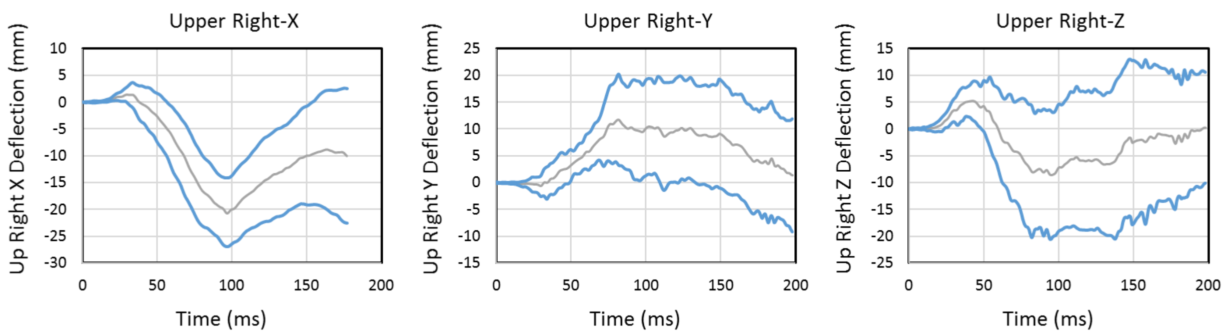

Figure 11.7. Small female sled test: Chest deflection corridors at upper right.

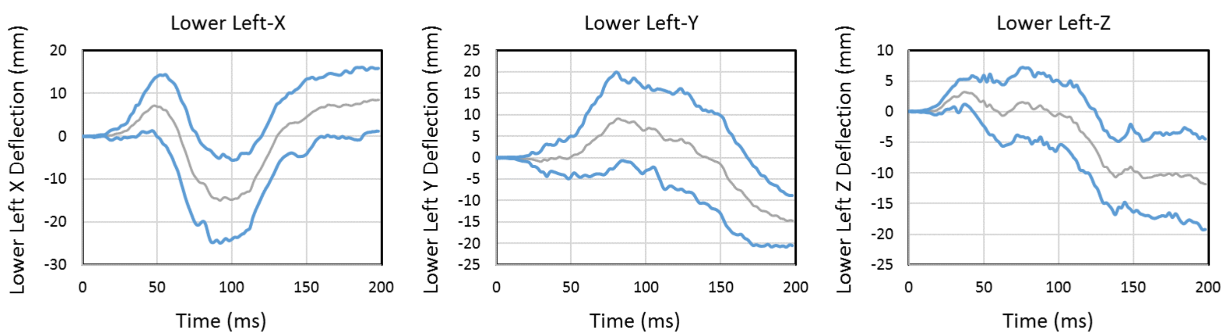

Figure 11.8. Small female sled test: Chest deflection corridors at lower left.

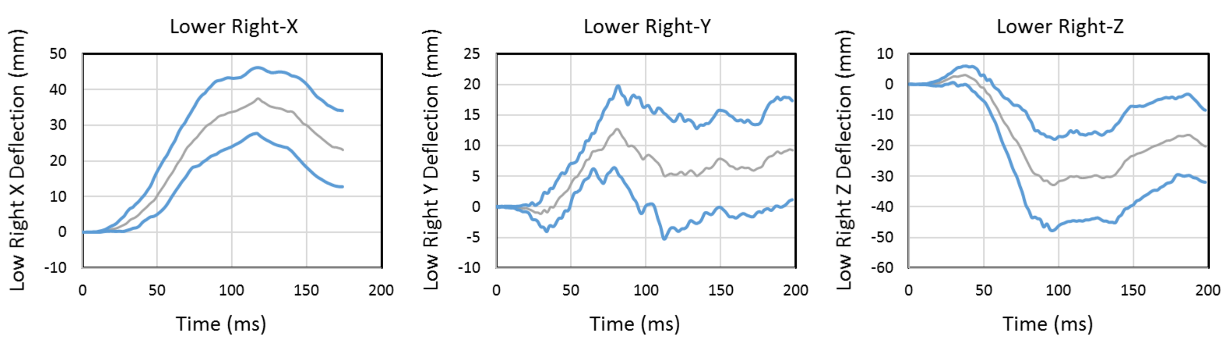

Figure 11.9. Small female sled test: Chest deflection corridors at lower right.

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