

Assessment of Seat Belt Anchor Position on Rear Seat Occupant Kinematics in Frontal Impact Using a
Validated THOR-5F Finite Element Model

by

Patrick Thangarajah

A thesis

presented to the University of Waterloo

in fulfillment of the

thesis requirement for the degree of

Master of Applied Science

in

Mechanical and Mechatronics Engineering

Waterloo, Ontario, Canada, 2026

© Patrick Thangarajah 2026

Author's Declaration

This thesis consists of material all of which I authored or co-authored: see Statement of Contributions included in the thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Statement of Contributions

Some of the content in this thesis has previously been published in a conference proceeding:

Thangarajah, P., Watson, B., Tang, K., Tylko, S., & Cronin, D. (2025). Analysis of Rear Seat Shoulder Belt Anchor Position on Kinematics of Small Stature Female in Frontal Impact. 20th Injury Biomechanics Symposium, Ohio State University. <https://ibrc.osu.edu/wp-content/uploads/2025/05/Thangarajah-20th-IBS-Paper.pdf>

PT: Manuscript writing, Methodology, Simulations, Data analysis, Visualization

BW: Manuscript review and editing, Methodology

KT: Manuscript review and editing, Physical testing

ST: Manuscript review and editing, Physical testing, Resources

DC: Manuscript review and editing, Supervision, Methodology, Project conceptualization

Abstract

Rear seat occupant safety has been identified as an area of concern in frontal impacts, with small stature females experiencing severe injuries more frequently, including a higher rate of thoracic injuries. Seat belts, the primary restraint for rear seat occupants, have been identified as contributors to thoracic injury, with studies finding that belt geometry and belt anchor locations influence occupant response.

A new small stature female Anthropometric Testing Device (ATD) to be used in vehicle testing is the THOR-5F ATD, designed with improved anthropometry and updated measurement capabilities, including four chest deflection sensors.

In this study, a Finite Element (FE) model of the THOR-5F was validated against physical testing of the THOR-5F ATD in a Rigid Bench test configuration with a 23 g frontal crash pulse, by comparing belt interaction, iliac forces, pelvis accelerations, chest deflections, and head accelerations. The THOR-5F FE model was also assessed on a simplified version of an existing bench with a pivoting seat (Semi-Rigid Bench) and a 4-Point 2-Belt system, and the results were compared with those from a 9 g sled test reported in the literature. In both bench configurations, the overall THOR-5F FE response matched the reported physical response well, with an average ISO 18571 rating of 0.71, corresponding to a “Fair” grade. Physical testing of the THOR-5F ATD also showed bottoming out of the lower-right chest deflection sensor, which is currently under investigation as the THOR-5F design is being refined.

Parametric studies were conducted to explore the effects of shoulder and lap belt anchor positioning in the fore/aft (X), lateral (Y), and vertical (Z) directions on THOR-5F FE peak chest deflections and peak resultant pelvis accelerations. The shoulder belt anchor X and Y-positions had a large effect on the overall peak chest deflection in both bench configurations, with rearward and inboard anchor locations reducing deflection, and the lowest deflections approximately 16% lower than the average. The lap belt anchor position had a smaller effect than the shoulder belt anchor on chest deflections. Based on preliminary chest deflection injury risk curves, rearward and inboard shoulder belt anchor positions resulted in lower corresponding AIS 3+ injury risks. However, the risks remained very high

across all anchor configurations in both benches, indicating a need to implement restraint technologies, such as load limiters, in addition to favorable anchor positions.

The lap belt anchor position had a larger effect on peak pelvis acceleration than the shoulder belt anchor position. The X, Y, and Z-positions of the lap belt anchor influenced the peak resultant pelvis acceleration in the Rigid Bench, with rearward, upward, and inboard locations reducing pelvis accelerations. The Y-position of the lap belt anchors and the YZ interaction had the largest effect on pelvis acceleration, indicating a complex interaction between the anchors and the Semi-Rigid Bench components.

This study demonstrated that the Rigid and Semi-Rigid Bench FE models could replicate physical occupant and restraint responses, and that anchor position affected the chest deflection and pelvis acceleration of the THOR-5F FE. Future research should focus on belt material properties, vehicle fleet-specific anchor positions, the implementation of load limiters, and the combined interactions between shoulder and lap belt anchor positions.

Acknowledgements

Many people have enriched my master's experience, and I am grateful for their support.

I would first like to express my sincere gratitude to my supervisor, Professor Duane Cronin, for providing me with the opportunity to conduct research in the IMMC group, as well as for his continuous guidance and support throughout my master's. I have learned a lot under his supervision, and I truly believe he is one of the best supervisors one can have.

I would also like to thank Transport Canada for this project and the PMG Technologies staff for conducting the physical testing and providing the physical test data. I would like to thank Suzanne Tylko and Kathy Tang for providing helpful insights and advice throughout my time working on this project.

I would also like to thank Brock Watson for his mentorship and thorough advice. It was an amazing experience to learn a lot about FE modelling practices and the various vehicle safety topics from you. I would like to thank the IMMC group for their advice and experience, and for the many great memories they have given me throughout my time in the group.

I would like to acknowledge ATD Models GmbH for their timely provision of the THOR-5F FE model, and the Digital Research Alliance of Canada for access for computing, which enabled me to conduct simulations.

I would also like to acknowledge the Canada Research Chairs Program and the Natural Sciences and Engineering Research Council of Canada for their financial support.

Finally, I would like to thank my family. The boundless and continuous support of my dad, Jude; my mom, Pamini; my aunt, Dharshini; my sister, Maryanne; and my brother, Thomas, was instrumental in keeping me motivated throughout my thesis. I am ever grateful for their presence from the beginning.

Dedication

To my family.

Table of Contents

Author's Declaration	ii
Statement of Contributions.....	iii
Abstract	iv
Acknowledgements	vi
Dedication	vii
List of Figures	xiii
List of Tables.....	xxi
List of Abbreviations.....	xxiii
Chapter 1 Introduction.....	1
1.1 Research Motivation.....	1
1.2 Research Objectives	3
1.3 Thesis Organization.....	4
Chapter 2 Background.....	5
2.1 Rear Seat Occupant Safety in Frontal Impact	5
2.2 Seat Belts and Occupant Restraint	7
2.3 Small Stature Female Human Surrogates.....	9
2.3.1 THOR-5F Physical ATD and Biofidelity.....	10
2.3.2 ATD Injury Criteria and Injury Risk Metrics.....	13
2.4 Safety Regulations and Evaluation for Rear Seat Occupants.....	17
2.5 Sled Test Benches Representing the Rear Seat Environment	20
2.5.1 Federal Motor Vehicle Safety Standard (FMVSS) 213 Sled and Bench.....	21
2.5.2 Physical Semi-Rigid Bench.....	21

2.6 Finite Element Modelling in Occupant Safety Testing	23
2.6.1 Sled FE Models	24
2.6.2 Seat Belt FE Models	26
2.6.3 FE Modelling of ATDs.....	28
2.6.4 FE Model Validation	28
2.7 Previous Rear Seat Anchor Location Studies.....	31
2.7.1 Anchor Location Surveys	31
2.7.2 Effects of Shoulder Belt Anchor/D-ring Positioning on Occupant Response	32
2.7.3 Effects of Lap Belt Anchor Positioning on Occupant Response	33
Chapter 3 Methods	34
3.1 Overview of Methods	34
3.2 Rigid Bench and Sled Testing at Transport Canada.....	35
3.2.1 Rigid Bench Geometry and Sled Specifications.....	35
3.2.2 Physical THOR-5F ATD Testing.....	40
3.3 Integration of THOR-5F FE Model with the Rigid Bench and Validation with Experimental Tests	41
3.3.1 Rigid Bench FE Model.....	41
3.3.2 Positioning of THOR-5F FE Model on the FE Rigid Bench.....	44
3.3.3 Implementation of Seat Belts in FE Rigid Bench Model	47
3.3.4 Validation of THOR-5F in FE Rigid Bench.....	50
3.4 Integration of THOR-5F FE Model with the Semi-Rigid Bench and Assessment Using Reported Experimental Test Data	52
3.4.1 Semi-Rigid Bench FE Model	53

3.4.2 Positioning of THOR-5F FE in Semi-Rigid Bench FE Model.....	57
3.4.3 Implementation of Seat Belts in the Semi-Rigid Bench FE Model.....	59
3.4.4 Assessment of THOR-5F FE Model in the Semi-Rigid Bench.....	61
3.4.5 Implementation of 3-Point Seat Belt in Semi-Rigid Bench FE Model	61
3.5 Parametric Study of Belt Anchor Locations and THOR-5F FE Kinematic Responses.....	62
3.5.1 Shoulder Belt Anchor Position Investigation	63
3.5.2 Lap Belt Anchor Position Investigation	65
3.5.3 Calculation of Main and Interaction Effects.....	67
Chapter 4 Results.....	70
4.1 Physical Rigid Bench Testing at Transport Canada	70
4.2 Integration of THOR-5F FE Model with the Rigid Bench and Validation with Experimental Tests	73
4.2.1 Rigid Bench FE Model.....	73
4.2.2 Positioning of THOR-5F FE in Rigid Bench	74
4.2.3 Implementation of Belts in Rigid Bench FE Model	75
4.2.4 Validation of THOR-5F FE Model in Rigid Bench	76
4.3 Integration of THOR-5F FE Model with the Semi-Rigid Bench and Assessment Using Reported Experimental Test Data	80
4.3.1 Semi-Rigid Bench FE Model	80
4.3.2 Positioning of THOR-5F FE in the Semi-Rigid Bench Model.....	81
4.3.3 Implementation of Four-Point, Two-Belt Seat Belt System in the Semi-Rigid Bench FE Model.....	82
4.3.4 Assessment of THOR-5F FE in Semi-Rigid Bench	84
4.3.5 Implementation of 3-Point Belt System in Semi-Rigid Bench Model	87

4.4 Parametric Study of Belt Anchor Locations and THOR-5F FE Kinematic Responses.....	89
4.4.1 Screening of Non-characteristic Kinematics	89
4.4.2 Shoulder Belt Anchor Investigation for Rigid Bench.....	91
4.4.3 Lap Belt Anchor Investigation for Rigid Bench.....	96
4.4.4 Shoulder Belt Anchor Investigation for Semi-Rigid Bench.....	101
4.4.5 Lap Belt Anchor Investigation for Semi-Rigid Bench	105
Chapter 5 Discussion.....	111
5.1 Validation of THOR-5F FE Model in Rigid Bench	111
5.2 Assessment of THOR-5F FE Model in Semi-Rigid Bench.....	113
5.2.1 Implementation of 3-Point Belt System from the Rigid Bench.....	116
5.3 Parametric Study of Belt Anchor Location and THOR-5F FE Peak Chest Deflection.....	118
5.3.1 Rigid Bench Shoulder Belt Anchor DOE.....	118
5.3.2 Rigid Bench Lap Belt Anchor DOE.....	121
5.3.3 Semi-Rigid Bench Shoulder Belt Anchor DOE	123
5.3.4 Semi-Rigid Bench Lap Belt Anchor DOE	125
5.3.5 Chest Deflection Regions	127
5.3.6 Chest Deflection and Injury Risk	128
5.4 Belt Anchor Location and Peak Resultant Pelvis Acceleration.....	129
5.4.1 Rigid Bench Shoulder Belt Anchor DOE.....	129
5.4.2 Rigid Bench Lap Belt Anchor DOE.....	131
5.4.3 Semi-Rigid Bench Shoulder Belt Anchor DOE	133
5.4.4 Semi-Rigid Bench Lap Belt Anchor DOE	135
5.5 Comparison of Rigid Bench and Semi-Rigid Bench.....	137

5.6 Anchor Locations and FMVSS/CMVSS Regulations	139
5.7 Seat Belt Anchor Recommendations for the Small Statured Female	141
5.8 Limitations.....	142
Chapter 6 Conclusion	146
Letter of Copyright Permission	149
References	150

List of Figures

Figure 2.1: Comparison of male and female injuries and fatalities in the rear seat. Reproduced from Jermakian et al. (2019).	6
Figure 2.2: THOR-5F ATD seated in a testing bench. Photo taken by Transport Canada.....	11
Figure 2.3: Preliminary AIS 3+ injury risk curves for sternum deflection in the X direction, the X-Y resultant direction and the X-Y-Z resultant direction.....	17
Figure 2.4: Acceptable seat belt anchor locations from FMVSS/CMVSS 210 standards, adapted to the THOR-5F in Rigid Bench described in Section 3.2. The green regions indicate the acceptable seat belt anchor locations (Transport Canada, n.d.; U.S. Department of Transportation, n.d.).	20
Figure 2.5: Semi-Rigid Bench configuration with HIII-50M positioned (Uriot et al., 2015). Picture used with permission from the Stapp Association.....	22
Figure 2.6: (A) Isometric view and (B) side view of the ENOP sled model with belt system (Lebarbé et al., 2023). Note: Knee support not shown.	26
Figure 2.7: FE model of 3-point belt with 1D elements for connection to belt anchors and slings and 2D elements for occupant contact.....	27
Figure 2.8: Component ratings considered for the overall ISO 18571 score, consisting of a slope rating, magnitude rating, phase rating and corridor rating. The red curve is the model while the black curve is the experiment. The green range defines the inner corridor, while the yellow area defines the outer corridor.	30
Figure 3.1: Overview of methods for (a) Rigid Bench and (b) Semi-Rigid Bench configurations with THOR-5F ATD FE model.....	34
Figure 3.2: 3D scan of Rigid Bench used in physical tests. (A) Isometric view of physical Rigid Bench with shoulder and lap belt anchor assemblies identified, where the red circle represents the origin (0,0,0) and the green circle represents the seat belt anchor. (B) Side view of Rigid Bench with retractor.	36
Figure 3.3: Dimensions of Rigid Bench used in physical tests with THOR-5F ATD: (A) Front view and (B) side-view. All linear dimensions are in millimeters.	37
Figure 3.4: Seat belt retractor (Tang et al., 2019) used in the Rigid Bench test setup. Image provided by TC/PMG.	38
Figure 3.5: Retractor force-pullout curve extracted from video data of sled tests in a different study.	39

Figure 3.6: (A) Physical Rigid Bench test setup with THOR-5F on the acceleration sled and (B) sled acceleration pulse used in physical tests.....	40
Figure 3.7: FE model of the Rigid Bench used in sled testing with the THOR-5F. The setup consisted of shoulder and lap belt anchor assemblies, a raised platform, seat pan, retractor, and seatback.	42
Figure 3.8: (A) Lap belt anchor assembly used in the Rigid Bench model, with spherical joint, seat belt webbing guide and anchor attachment. (B) Shoulder belt anchor assembly used in the Rigid Bench model, with D-ring, revolute joint and anchor attachment. Anchor location indicated by red circle.	43
Figure 3.9: Overview of the positioning pre-simulation, where tensioned cables moved the THOR-5F FE model from a (A) start position to a (B) target position.	44
Figure 3.10: Overview of positioning process. (A) Wireframe from FAROArm measurements of average physical THOR-5F ATD position, used for positioning THOR-5F FE model. (B) Overlay of THOR-5F FE model (gray) on wireframe. (C) Overlay of the THOR-5F FE model (gray) on 3D scan of physical THOR-5F ATD (red). (D) Marionette method with THOR-5F starting configuration, with grey nodes representing target position coordinates. (E) Final THOR-5F FE position after marionette method.	46
Figure 3.11: Seat belt routing process using THOR-5F FE jacket model in LS-PrePost V4.10. (A) Nodes, corresponding to the anchor locations (red circles), were created for the lap and shoulder belt ends. Orange intermediate nodes were selected along the jacket surface to define the belt path. The nodes were then used to create (B) shoulder belts and (C) lap belts.	48
Figure 3.12: Belt model implemented in the Rigid Bench model. The right lap belt anchor contains a slipring to connect the shoulder and lap belts.....	49
Figure 3.13: (A) Retractor force-pullout test data from TC/PMG (orange and blue) and retractor-pullout curve implemented in Rigid Bench FE model. (B) Force-strain curve of belts used in THOR-5F FE model.	49
Figure 3.14: Three sled acceleration pulses from physical tests, with the average pulse shown in red.	50
Figure 3.15: Validation metric locations of the THOR-5F in the Rigid Bench. (A) Seat belt measurement locations. (B) THOR-5F measurements used in validation.	51
Figure 3.16: Semi-Rigid Bench FE model used in THOR-5F FE assessment simulations.....	54

Figure 3.17: Semi-Rigid Bench model dimensions. (A) Front view rotated 20° about the Y-axis to show the bench. (B) Side view of the bench with dimensions. (C) Dimensions of the anti-submarining pan. All measurements are in mm. The red circle represents the origin (0, 0, 0), the green dot represents the seat pan spring and revolute joint, and the blue dot represents the anti-submarining pan spring and revolute joint. *Dimensions adjusted to match Richard et al. (2022).	55
Figure 3.18: Spring stiffnesses of seat pan and anti-submarining pan in the Semi-Rigid Bench (Richard et al., 2022).	56
Figure 3.19: Tilt sensors and points used to position the THOR-5F FE to match with Richard et al. (2022) Configuration 1 position. The upper and lower lap belt edges were used to position the lap belt to match the Configuration 1 position.	57
Figure 3.20: Positioning pre-simulation process for the THOR-5F in the Semi-Rigid Bench, where cables moved the THOR-5F FE model from a (A) start position to a (B) target position. The Semi-Rigid Bench seat pan and anti-submarining pan were constrained from deforming to simplify the positioning process.	58
Figure 3.21: Positioning of seatback from (A) standard position of rigid plane in Richard et al. bench test to (B) behind the THOR-5F jacket with minimal spacing and without node interference.	59
Figure 3.22: Seat belt force-strain curves for the original belt model used in the Rigid Bench model and the modified belt in the Semi-Rigid Bench model. The slope of the modified belt in the higher strain region (>0.06) was reduced to match the physical seat belt force-strain response reported by Richard et al. (2022).	60
Figure 3.23: Acceleration pulse used in Configuration 1 (non-submarining) sled tests from Richard et al. (2022).	61
Figure 3.24: Process of changing belt configuration from the (A) 4-point 2-belt configuration to the (B) 3-point belt configuration.	62
Figure 3.25: Shoulder belt anchor locations simulated with Rigid Bench (top) and Semi-Rigid Bench (bottom).	64
Figure 3.26: THOR-5F in Rigid Bench (top) and Semi-Rigid Bench (bottom) with (A) inboard lap anchor positions simulated with each configuration (B) outboard lap belt anchor simulated. *Anchors 10 and 19 were unable to be simulated due to pelvis interference.	66

Figure 3.27: Calculation of the X main effect of shoulder belt anchor positioning on chest deflection. (A) Each anchor position at the maximum ($X = 1$) and minimum ($X = -1$) is simulated, and the peak deflection is obtained from the recorded trace. (B) The peaks are averaged for each X position, and the difference between the averages is the main effect.	68
Figure 4.1: Response traces from three physical tests conducted by TC/PMG with the THOR-5F in the Rigid Bench.	72
Figure 4.2: Overlay of Rigid Bench FE model (blue) on 3D scan of physical Rigid Bench (gray) with (A) isometric view and (B) side view of overlay.	73
Figure 4.3: Definition of seat belt measurement points on the THOR-5F FE in the Rigid Bench for comparison to the physical test.	75
Figure 4.4: Images of the THOR-5F physical ATD (A–C) and THOR-5F FE model (D–E) at synchronized times $t = 12$ ms, 85 ms and 125 ms (*Video time offset by -50 ms and simulation time offset by -100 ms to synchronize test images (A–C) with model (D–F)). At 12 ms, the FE model posture matched the physical ATD. At 85 ms and 125 ms, lower Z-rotation was observed in the FE model than in the physical ATD.	77
Figure 4.5: Responses of THOR-5F FE model and physical THOR-5F ATD in the Rigid Bench.	79
Figure 4.6: Overlay of Semi-Rigid Bench FE model (blue) on ENOP bench model (gray) with (A) Isometric view and (B) Oblique lateral view. The ENOP bench model included a foam cushion layer on the seat back, which was not present in the Semi-Rigid Bench model.	81
Figure 4.7: Definition of lap belt measurement points on the THOR-5F FE model in the Semi-Rigid Bench. The midpoints ($Y = 0$) of the upper and lower lap belt edges were used to compare with the lap belt position in the physical tests (Table 4.6).	83
Figure 4.8: Images of the THOR-5F physical ATD (A–C) and THOR-5F FE model (D–F) at synchronized times $t = 0$ ms, 75 ms, and 120 ms. (*Simulation time offset by -50 ms to synchronize model (D–F) with test images (A–C)). Physical test images from Richard et al. (2022) reproduced with permission from the Stapp Association.	84
Figure 4.9: Responses of THOR-5F FE model in the Semi-Rigid Bench with the 4-point, 2-belt configuration compared to the physical test data by Richard et al. (2022).	86

Figure 4.10: Responses of THOR-5F FE model in the Semi-Rigid Bench with the 4-point, 2-belt configuration and the 3-point belt configuration compared to the physical test data from Richard et al. (2022).	88
Figure 4.11: Shoulder belt migration from shoulder to the abdomen during simulation of Position 21 in the (A) Rigid Bench and (B) Semi-Rigid Bench.....	90
Figure 4.12: Kinematic response modes resulting from lap belt anchor variations. Submarining occurred for multiple lap belt anchor configurations in both benches, with Position 1 (top row) shown as an example. Forward pitch was observed for a different set of lap belt anchor configurations in both benches, with Position 3 (bottom row) shown as an example.....	91
Figure 4.13: THOR-5F responses for shoulder belt anchor positions in the Rigid Bench. (A) Overall peak chest deflection for simulated anchor locations (B) Peak pelvis resultant accelerations for simulated anchor locations. Marker numbers correspond to the DOE position. *Indicates configuration removed from analysis.....	92
Figure 4.14: THOR-5F peak chest sensor deflections for shoulder belt anchor locations in the Rigid Bench: (A) left upper sensor, (B) right upper sensor, (C) left lower sensor, and (D) right lower sensor. Marker numbers correspond to DOE position. *Indicates configurations removed from analysis.....	93
Figure 4.15: Main and interaction effects on peak chest deflection for the Rigid Bench shoulder belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).	94
Figure 4.16: Main and interaction effects on peak pelvis resultant acceleration for the Rigid Bench shoulder belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).	96
Figure 4.17: THOR-5F responses for outboard lap belt anchor positions in the Rigid Bench. (A) Overall peak chest deflection for simulated outboard lap belt anchor locations (B) Peak pelvis resultant accelerations for simulated outboard lap belt anchor locations. Marker numbers correspond to the DOE position. *Indicates configurations removed from the analysis; ⊗ indicates configurations that were not simulated.....	97
Figure 4.18: THOR-5F peak chest sensor deflections for lap belt anchor locations in the Rigid Bench: (A) left upper sensor, (B) right upper sensor, (C) left lower sensor, and (D) right lower sensor. Marker numbers correspond to DOE position. *Indicates configurations removed from the analysis; ⊗ indicates configurations that were not simulated.	98

Figure 4.19: Main and interaction effects on peak chest deflection for the Rigid Bench lap belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).	99
Figure 4.20: Main and interaction effects on peak pelvis resultant acceleration for the Rigid Bench lap belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).	100
Figure 4.21: THOR-5F responses for shoulder belt anchor positions in the Semi-Rigid Bench. (A) Overall peak chest deflection for simulated anchor locations (B) Peak pelvis resultant accelerations for simulated anchor locations. Marker numbers correspond to the DOE position. *Indicates configuration removed from analysis.	101
Figure 4.22: THOR-5F peak chest sensor deflections for shoulder belt anchor locations in the Semi-Rigid Bench: (A) left upper sensor, (B) right upper sensor, (C) left lower sensor, and (D) right lower sensor. Marker numbers correspond to DOE position. *Indicates configurations removed from analysis.	102
Figure 4.23: Main and interaction effects on peak chest deflection for the Semi-Rigid Bench shoulder belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).	103
Figure 4.24: Main and interaction effects on peak pelvis resultant acceleration for the Semi-Rigid Bench shoulder belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).	105
Figure 4.25: THOR-5F responses for outboard lap belt anchor positions in the Semi-Rigid Bench. (A) Overall peak chest deflection for simulated outboard lap belt anchor locations (B) Peak pelvis resultant accelerations for simulated outboard lap belt anchor locations. Marker numbers correspond to the DOE position. *Indicates configurations removed from the analysis; ⊗ indicates configurations that were not simulated.	106
Figure 4.26: THOR-5F peak chest sensor deflections for lap belt anchor locations in the Semi-Rigid Bench: (A) left upper sensor, (B) right upper sensor, (C) left lower sensor, and (D) right lower sensor. Marker numbers correspond to DOE position. *Indicates configurations removed from analysis.	107
Figure 4.27: Main and interaction effects on peak chest deflection for the Semi-Rigid Bench lap belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).	109
Figure 4.28: Main and interaction effects on peak pelvis resultant acceleration for the Semi-Rigid Bench lap belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).	110

Figure 5.1: THOR-5F at 105 ms. (A) Top view of physical ATD. (B) Side view of physical ATD. (C) Top view of FE model. (D) Side view of FE model. The reduced torso rotation of the THOR-5F FE model, shown by the angle indicator (dashed green lines), resulted in higher head accelerations at the end of the crash pulse.	112
Figure 5.2: THOR 5F FE iliac load cell with belt-induced compressive loading and sliding between front and back components. The iliac load cell beam records the beam axial force between the components as the iliac force.	115
Figure 5.3: Comparison of the 4-point, 2-belt configuration and the 3-point belt configuration in the Semi-Rigid Bench. (A) Lower shoulder belt fit in the 3-point belt configuration (light gray) compared to the 4-point, 2-belt configuration (black). (B) Front view and (C) side view of the 4-point, 2-belt kinematics to the 3-point belt.	117
Figure 5.4: Images of the THOR-5F in the Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum chest deflection shoulder belt anchor configuration, (C,D) midpoint shoulder belt anchor configuration and (E,F) maximum chest deflection shoulder belt anchor configuration. B,D,F are the side view images of each configuration.	119
Figure 5.5: Images of the THOR-5F in the Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum deflection lap belt anchor configuration, (C,D) midpoint deflection lap belt anchor configuration and (E,F) maximum deflection lap belt anchor configuration. B,D,F are the side view images of each configuration.	122
Figure 5.6: Images of the THOR-5F in the Semi-Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum deflection shoulder belt anchor configuration, (C,D) midpoint deflection shoulder belt anchor configuration and (E,F) maximum deflection shoulder belt anchor configuration. B,D,F are the side view images of each configuration.	124
Figure 5.7: Images of the THOR-5F in the Semi-Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum deflection lap belt anchor configuration, (C,D) midpoint lap belt anchor configuration, and (E,F) maximum deflection lap belt anchor configuration. B,D,F are the side view images of each configuration.	126
Figure 5.8: Preliminary AIS 3+ injury risk curves with seat belt anchor DOE configurations.	129
Figure 5.9: Images of the THOR-5F in the Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum resultant pelvis acceleration shoulder belt anchor configuration, (C,D) midpoint	

shoulder belt anchor configuration, and (E,F) maximum resultant pelvis acceleration shoulder belt anchor configuration. B,D,F are the side view images of each configuration.....	130
Figure 5.10: Images of the THOR-5F in the Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum resultant pelvis acceleration lap belt anchor configuration, (C,D) midpoint lap belt anchor configuration, and (E,F) maximum resultant pelvis acceleration lap belt anchor configuration. B,D,F are the side view images of each configuration.	132
Figure 5.11: Images of the THOR-5F in the Semi-Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum resultant pelvis acceleration shoulder belt anchor configuration, (C,D) midpoint shoulder belt anchor configuration, and (E,F) maximum resultant pelvis acceleration shoulder belt anchor configuration. B,D,F are the side view images of each configuration.....	134
Figure 5.12: Images of the THOR-5F in the Semi-Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum resultant pelvis acceleration lap belt anchor configuration, (C,D) midpoint lap belt anchor configuration, and (E,F) maximum resultant pelvis acceleration lap belt anchor configuration. B,D,F are the side view images of each configuration.....	136
Figure 5.13: Motion of THOR-5F FE at the lowest pelvis resultant acceleration in the (A) Rigid Bench configuration and (B) Semi-Rigid Bench configuration. The Rigid Bench configuration only enabled forward (+X) translation of the pelvis, while the Semi-Rigid Bench configuration enabled the forward (+X) and lower (+Z) motion of the pelvis, which led to a higher pelvis resultant acceleration.	138
Figure 5.14: Side view of Rigid Bench with shoulder belt anchor X and Z-positions investigated and the acceptable anchor range defined by the CMVSS and FMVSS 210 Standards (Transport Canada, n.d.; U.S. Department of Transportation, n.d.). Locations within the red circle represent simulated locations outside the acceptable range.	140
Figure 5.15: Side view of Semi-Rigid Bench with lap belt anchor X and Z-positions investigated, and the acceptable anchor range defined by the CMVSS and FMVSS 210 Standard (Transport Canada, n.d.; U.S. Department of Transportation, n.d.). Locations within red circles represent simulated locations outside the acceptable range.	141
Figure 5.16: Shoulder belt interaction at maximum excursion ($t = 105$ ms) for (A) physical Rigid Bench test and (B) FE simulation of Rigid Bench.	144

List of Tables

Table 2.1: Maxims for good occupant restraint. Adapted from Eppinger (2002).	8
Table 2.2: Biofidelity rankings, associated scores, and statistical interpretation (Wang et al., 2020). 12	
Table 2.3: Abbreviated Injury Scale (AIS) code, associated severity and thoracic injury example for each code (adapted from the Association for the Advancement of Automotive Medicine; Cavanaugh & Yoganandan, 2015).	14
Table 2.4: Injury metrics for the HIII-5F ATD in the FMVSS/CMVSS 208 standard for frontal impact (National Highway Traffic Safety Administration, 2026a).	15
Table 2.5: Injury metrics for the HIII-5F ATD in the Euro NCAP evaluation for frontal impact (Euro NCAP, 2025).	18
Table 2.6: Injury metrics for the HIII-5F ATD in the IIHS evaluation for frontal impact (Insurance Institute for Highway Safety, 2025).	19
Table 2.7: Geometry and material cards of three different belt system models.	28
Table 2.8: ISO ratings and corresponding grades (International Organization for Standardization, 2024).	31
Table 3.1: Rigid Bench seat belt anchor locations reported with respect to origin (red circle in Figure 3.2A).	37
Table 3.2: Material properties of seatback, seat pan, sled floor, and seat belt anchors of the Rigid Bench FE model.	41
Table 3.3: Geometry and material card of Rigid Bench FE model seat belt.	49
Table 3.4: Metrics used for THOR-5F FE model validation and corresponding SAE filters.	52
Table 3.5: Semi-Rigid Bench belt anchor locations from origin.	56
Table 3.6: Shoulder belt anchor locations (from origin) for minimum, maximum and midpoint configurations.	64
Table 3.7: Shoulder belt anchor location effect simulations corresponding to Figure 3.25.	65
Table 3.8: Lap belt anchor location effect simulations corresponding to Figure 3.26.	67
Table 3.9: Lap belt anchor locations (from origin) for minimum, maximum and midpoint configurations for outboard location.	67
Table 4.1: Maximum difference between FAROArm measurements for ATD positioning landmarks for the three physical tests.	71

Table 4.2: Distance between THOR-5F FE model and wireframe location for select (distinguishable) nodes after positioning simulation.....	74
Table 4.3: Distance between THOR-5F FE model and average physical ATD belt position.	76
Table 4.4: ISO 18571 ratings and grades for THOR-5F FE model and physical THOR-5F ATD in the Rigid Bench.....	80
Table 4.5: Comparison of THOR-5F FE position to Richard et al. (2022) Test A11 and A12 measurements.	82
Table 4.6: Distance between THOR-5F FE model and physical ATD belt position.....	83
Table 4.7: ISO 18571 ratings and grades for THOR-5F FE model in Semi-Rigid Bench compared to Richard et al. (2022) Tests A11 and A12. The table contains the ratings for each test along with the average ratings and associated grades.	87
Table 5.1: Difference of H-point from average of physical test measurements for different settling methods.	143

List of Abbreviations

AIS: Abbreviated Injury Scale

ATD: Anthropomorphic Test Device

CMVSS: Canadian Motor Vehicle Safety Standard

DOF: Degrees of Freedom

ENOP: Enable New Occupant Seating Positions

FE: Finite Element

FMVSS: Federal Motor Vehicle Safety Standard

Hybrid III 5th: Hybrid III 5th percentile female

Hybrid III 50th: Hybrid III 50th percentile male

NCAP: New Car Assessment Program

THOR-5F: Test Device for Human Occupant Restraint 5th percentile female

THOR-50M: Test Device for Human Occupant Restraint 50th percentile male

TC/PMG: Transport Canada/PMG Technologies

Chapter 1

Introduction

1.1 Research Motivation

While vehicle safety has steadily improved over time, frontal impacts still account for a large portion of vehicle fatalities. Out of all collision types, frontal impacts accounted for 60% of passenger vehicle deaths in 2023 in the United States, while in Canada, frontal impacts represented the most fatal multi-vehicle collision mode (Insurance Institute for Highway Safety, n.d.; Transport Canada, 2025). Historically, the rear seat of vehicles was considered to be the safest position in frontal impact conditions, compared to the front seat (Mayrose & Priya, 2008). However, recent evidence has suggested that rear seats are becoming relatively less safe compared to the front seat, with restrained occupants facing a higher risk of fatality (Durbin et al., 2015) and increased exposure with higher rear seat occupancy rates associated with ride-sharing transportation (Nemire, 2017). In addition, female occupants have been identified as having a higher risk of injury in the rear seat environment (Jermakian et al., 2019) compared to males.

The lower rear seat occupant safety compared to the front seat has been attributed in part to the focus on advancing safety technologies in the front seat, due to the higher emphasis on front seat occupant safety in regulatory testing (Durbin et al., 2015). In automotive safety testing, occupant surrogates, including physical anthropomorphic test devices (ATDs) from different manufacturers, are used to assess the performance of automobile safety systems (Crandall et al., 2011; H. J. Mertz & Irwin, 2015; Schmitt et al., 2025). In addition to physical testing, computational modelling of ATDs provides several advantages, including the ability to evaluate the effects of design parameter changes and reduce costs associated with physical testing and to assess new ATDs in a variety of controlled impact scenarios (Yang & Chou, 2015).

The frontal crash tests defined by Canadian (CMVSS 208) and United States (FMVSS 208) standards evaluate the small female Hybrid III 5th percentile ATD (HIII-5F) only in the driver and first-row passenger seats (National Highway Traffic Safety Administration, 2026a). However, the HIII-5F ATD is being implemented in the rear seat in consumer test programs worldwide (Tang et al., 2022). Research using the HIII-5F in the rear seat indicated that the belt placement, anchor geometry, and crash pulse severity strongly influenced ATD kinematics (Tylko & Bussi res, 2012), while additional

studies have found that belt positioning affected chest deflections (Edwards et al., 2022). Several computational studies using ATD models in rear seat environments, including those of the HIII 5F, the Hybrid III 50th percentile male (HIII-50M) and the Test Device for Human Occupant Restraint 50th percentile male (THOR-50M), have also found that belt anchor location influenced occupant response (Hu, Wu, Klinich, et al., 2013; Meng et al., 2025; Youn et al., 2015).

While the HIII-5F is currently used in regulatory safety testing (Boyle et al., 2024), there are recognized limitations of the surrogate. Rear-seat studies from Transport Canada identified that chest deflection measurement is influenced by the seat belt motion (Tylko & Bussi res, 2012). In addition, the HIII-5F thorax was stiffer than a human thorax (Carroll et al., 2021), and the HIII-5F was derived by scaling down the HIII-50M instead of using female-specific dimensions (Linder & Svensson, 2019; Tang et al., 2022).

The 5th percentile female THOR ATD (THOR-5F) is a modern ATD designed for improved biofidelity and updated instrumentation compared to the HIII-5F (Lee et al., 2020; Wang et al., 2017). One study conducted physical sled testing with the THOR-5F and the HIII-5F in 2 different front seat belt anchor configurations, representing the locations of 3 different vehicles (Eggers et al., 2023), finding that different shoulder belt routings influenced chest deflection; however, it did not evaluate rear seat configurations or isolate the effects of seat belt anchor positioning on small stature female occupant response. Additionally, based on the available published literature, limited work has examined the effect of belt anchor positioning on the THOR-5F in a controlled rear seat environment.

Given the simpler nature of the rear seat restraint system with the seat belt as the primary restraint (Mizuno et al., 2018), the rear seat environment could provide a useful platform for evaluating different human surrogates, such as the THOR-5F, as well as restraint parameters such as anchor location. However, full-vehicle crash testing may be costly and complex (Smyth & Smith, 2007) and may introduce variability associated with vehicle seat properties. Sled testing, comprising a seat with an occupant surrogate and restraint system on an acceleration sled, is a repeatable method for evaluating vehicle safety systems (Smyth & Smith, 2007). Rear seat safety sled testing bench configurations include a rigid seat pan configuration (e.g., Luet et al., 2012), which prioritizes repeatability and reproducibility, and a semi-rigid seat pan configuration (e.g., Uriot et al., 2015), which aims to better represent the rear seat environment. Consequently, the use of a simplified testing

bench may enable a controlled investigation of THOR-5F occupant responses while isolating the effects of specific restraint parameters, such as seat-belt anchor geometry, on rear seat occupant response.

1.2 Research Objectives

The overall aim of this study was to develop and validate FE models of two rear seat bench sled tests and to use these models to determine the effects of shoulder and lap belt anchor positioning on the kinematics of a small stature female rear seat occupant during a frontal impact. To achieve this aim, three objectives were defined:

Objective 1: Develop a FE model of the Transport Canada Rigid Bench and validate a THOR-5F FE model in the controlled environment through kinetic and kinematic comparisons to physical sled tests with matched initial conditions and boundary conditions. The Rigid Bench FE model was developed to match the physical bench designed at Transport Canada, and the THOR-5F FE was positioned on the bench to match the pre-test sled test position. After implementing an open-source FE seat belt model, the crash pulse used in the physical Rigid Bench tests at Transport Canada was applied, and the resulting THOR-5F responses were compared to the physical test time history responses.

Objective 2: Develop a simplified FE model of a previously published physical Semi-Rigid Bench (Richard et al., 2022) in a rear seat configuration, including a 4-point 2-belt system, and validate the THOR-5F FE model with published physical test responses. The THOR-5F FE was positioned to match the test position described in the paper, and one rear seat configuration was simulated, including the 4-point 2-belt system. The responses were compared with the physical test data to validate the THOR-5F FE on the Semi-Rigid Bench. Then, the bench model was updated to include a standard 3-point seat belt system for comparison between the Rigid and Semi-Rigid Benches.

Objective 3: Determine the effect of seat belt anchor positioning on the THOR-5F FE kinematics and injury-related metrics in both the Rigid Bench and Semi-Rigid Bench, specifically on chest deflection and pelvis acceleration. A 3^3 design-of-experiments (DOE) matrix of seat belt anchor locations was defined from a study that examined multiple rear seat anchor locations, and from the matrix, all physically feasible anchor locations were simulated. From the anchor location simulations, the peak chest deflection and peak resultant pelvis acceleration were compared. Additionally, the effects of

belt anchor positioning in the X, Y, and Z directions were quantified, and the potential influence of seat pan deflection on the effect of belt anchor locations was investigated.

1.3 Thesis Organization

Chapter 2 presents background information on rear seat occupant injury sources and characteristics, current safety standards and seat belt geometries, and an overview of the THOR-5F anthropomorphic test devices and crash test benches.

The next chapter (Chapter 3) details the physical testing undertaken at Transport Canada and the FE modelling approach corresponding to the physical tests. Chapter 3 also covers the process of creating a simplified FE model of a Semi-Rigid Bench and integrating the THOR-5F FE model on the bench. The FE model was assessed by comparing kinetic and kinematic responses with a published study that used the physical Semi-Rigid Bench with a 4-point 2-belt system and the physical THOR-5F ATD. Additionally, the chapter discusses the belt anchor study, in which, for both benches, the belt anchor position varied in 3 dimensions to determine the effect of anchor locations for each component (X, Y, Z). To capture a wide range of rear seat belt anchor locations representative of the vehicle fleet, the anchor locations were defined based on the minimum and maximum values reported in a survey of vehicle rear seat geometry (Aram & Rockwell, 2012), with the midpoint between these values included, resulting in 3 levels for each dimension. The following chapter (Chapter 4) presents the results from the methods described in Chapter 3.

Chapter 5 discusses the results, compares the Rigid and Semi-Rigid Benches, evaluates the effects of seat belt anchor locations on the THOR-5F response, and explores the mechanisms driving the observed responses. This chapter also identifies seat belt anchor locations associated with more favorable and less favorable occupant responses. Additionally, the chapter describes the current regulations for the rear seat anchor locations and their relation to the study findings and discusses the study limitations. Chapter 6 provides a conclusion based on the study's results and discussion and offers recommendations for effective belt anchor positioning.

Chapter 2

Background

2.1 Rear Seat Occupant Safety in Frontal Impact

Frontal impacts continue to account for a large portion of vehicle fatalities, representing 60% of passenger vehicle deaths in 2023 in the United States, while in Canada, the head-on collision represented the most fatal multi-vehicle collision mode (Insurance Institute for Highway Safety, n.d.; Transport Canada, 2025).

Previous studies have found that rear seat occupants historically experienced a lower risk of fatality relative to front seat occupants. An analysis of crash data from the National Highway Traffic Safety Administration (NHTSA) Fatality Analysis Reporting System (FARS), for passenger vehicle crashes from 1990 and 2001 reported the risk of death for rear seat occupants was, on average, 21% lower compared to front seat passengers (Smith & Cummings, 2006). Another study looking at fatal crashes between 2000 and 2003 found that the chances of surviving for the second-row occupants were 29.1% higher than those of the front row occupants, with the rear middle seat having a 25% higher chance of survival compared to all other seats (Mayrose & Priya, 2008).

More recent studies have reported a shifting trend, with the rear seat becoming relatively less safe than the front seat. A 2015 study analyzing two automotive crash databases, the National Automotive Sampling System Crashworthiness Data System (NASS-CDS) and FARS, found that vehicle models newer than 2007 have a 46% higher risk of fatality for rear seat occupants than for front-row occupants (Durbin et al., 2015; Jermakian et al., 2019). The primary cause of the increase in relative safety of the front seat compared to the rear is the implementation of safety technologies.

Technologies such as load limiters, pretensioners and airbags were implemented primarily in the front seat and less consistently in the rear seat; a Consumer Reports rating in 2020 found that less than 40% of current vehicles had load limiters and pretensioners in the rear seat (Barry, 2020; Edwards et al., 2022). Additionally, the stiffness of vehicles has increased (Durbin et al., 2015; Locey et al., 2012), resulting in higher crash pulse severity for the rear seat (Esfahani et al., 2011). In addition to the rear seat being considered relatively less safe, rear seat occupancy rates have been rising, driven by the emergence of ridesharing, including taxis. In a study looking at the rear seat occupancy rates in San

Francisco and Las Vegas, 81% of ride share passengers and 91% of taxi passengers use rear seats compared to the front seat (Nemire, 2017).

Of the occupants injured in frontal impacts, female occupants were found to be at higher risk of sustaining serious injury or fatalities. A 2019 study looking at the injuries and fatalities of belted rear seat occupants found that female occupants were at higher risk than male occupants (Figure 2.1) (Jermakian et al., 2019). A study conducted in 2011 found that females were 47% more likely than males to sustain a serious injury when exposed to a comparable crash, which is potentially due to the bias in safety system testing towards male occupants (Bose et al., 2011). The commonly observed injuries for rear seat occupants in frontal impact were to the head, from contact with the vehicle interior or intrusion into the vehicle, or the chest from seat belt loading (Jermakian et al., 2019).

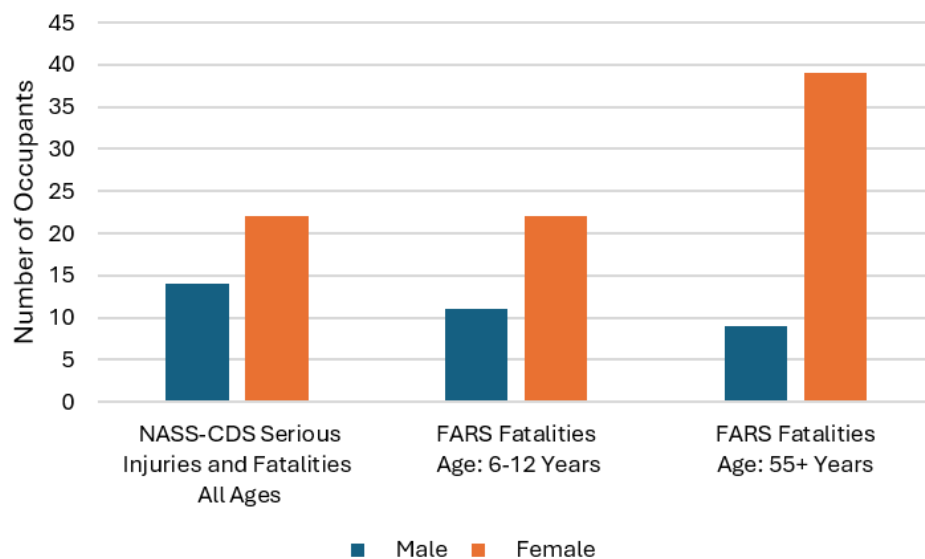


Figure 2.1: Comparison of male and female injuries and fatalities in the rear seat. Reproduced from Jermakian et al. (2019).

2.2 Seat Belts and Occupant Restraint

In the 1930s, physicians advocated seat belts for injury mitigation, and starting in 1950, lap belts were introduced in American vehicle models. In the 1970s, legislation on seat belt use was emerging globally (Agarwal et al., 2021; Eastman, 1981; Gantz & Henkle, 2002). To mitigate the risk of abdominal injury from early lap belt designs, Griswold and Haven (1955) introduced an early lap-and-shoulder belt configuration, with a buckle in the middle (Agarwal et al., 2021; Griswold & De, 1955). The modern-day 3-point seat belt was invented by Nils Bohlin in 1958 (Luet et al., 2012), which had the buckle on the side, allowing the occupant to secure the belt with one hand (Agarwal et al., 2021). The locking retractor, which followed the introduction of the 3-point belt, allowed the belt to loosen and retract for occupant comfort, but constrained it when a sudden load was applied by the occupant during vehicle deceleration (Agarwal et al., 2021; Ryan, 1963).

Another safety technology that enhanced occupant safety is the load limiter, which was introduced in the mid-1990s when airbags were common (Agarwal et al., 2021; Brumbelow et al., 2007). The first variations of the load limiter had a stitched fold in the seat belt webbing, which tore and unfolded at the load limit (Agarwal et al., 2021; Kahane, 2013). The modern load limiter, located in the retractor, limits the force from the belt by allowing pay-out of belt webbing at a threshold load, deforming a torsion bar (Håland, 2006).

The pre-tensioner was also introduced in 1983 enabled the occupant to be comfortable during normal use while also enabling proper restraint fit during impact (Håland, 2006; Mitzkus & Eyraier, 1984). The pretensioners were pyrotechnic; upon electronic sensor detection, a gas was generated to create higher pressure, resulting in the belt pulling in. Modern-day seat belt systems commonly use load limiters and pyrotechnic pretensioners at the retractor, and for the lap belt, pretensioners can be found in the buckle (Agarwal et al., 2021).

The seat belt should be as effective as possible and follow the maxims of good restraint defined by Eppinger (2002) (Table 2.1). The implementation of pretensioners and load limiters helps the restraint systems adhere to the maxims. Pretensioners, when activated by a sensor, quickly retract the belts during the onset of a crash to remove slack in the belt and allows the belt to act earlier and over a larger time, reducing peak belt forces. Load limiters increase the distance over which a force is acting on the occupant, which results in lower energy being absorbed by the occupant (Eppinger, 2002). Seat

belt systems in the rear seat do not commonly include load limiters or pre-tensioners (Edwards et al., 2022), which may result in higher peak belt forces for rear seat occupants.

Table 2.1: Maxims for good occupant restraint. Adapted from Eppinger (2002).

Maxim	Reason
Extend the duration of the restraint force on occupant	Reduces the rate of the restraint force application, resulting in lower peak forces experienced by the occupant
Increase the distance over which the restraint forces act on the occupant relative to the ground	Increases the extraction of kinetic energy through work, resulting in less energy being absorbed by the occupant
Apply as much restraint force possible, as early as possible	Maximizes kinetic energy extracted because the body translates more over the ground during the early phase of the crash; thus, work and energy extraction are the greatest. Early in the crash pulse is when the occupant translates the most; early application of restraint forces increases energy dissipated through work and reduces the energy absorbed by the body
Limit body articulation of occupant, minimize deformation rates and localized deformation during restraints	Reduces body articulation, high deformation rates and local deformations, which are associated with injury mechanisms
Distribute loading from restraint over as large a contact area as possible	Distributing loads over a larger contact area reduces localized stress and surface pressures, lowering the risk of deformation-based injury
Distribute restraint loads through the bony structures while minimizing loading of soft tissues	Loads skeletal structures such as the pelvis femur, thorax, shoulder and head, which are more capable of experiencing higher loading with minimal deformation, while soft tissues are susceptible to injury under restraint forces

Seat belts, particularly the shoulder belts, were identified as the most common source of injuries in rear seat occupants in frontal impacts, leading to chest and abdominal trauma (Beck et al., 2016; Jermakian et al., 2019). Jermakian et al. (2019) found that serious chest injuries occurred in 22 out of 36 injured occupants and were attributed to shoulder belt loading, and that the specific chest injury

varied with the age of the occupant, with the younger occupants primarily experiencing lung contusions while the older occupants were more likely to experience rib fractures and injuries to the heart and major vessels, indicating the age-dependence of thoracic injury tolerance.

The absence of airbags and seat belt technologies, such as pretensioners and load limiters, in the rear seat has potentially resulted in a stronger influence of belt anchor location on injury, with the shoulder belt anchor position affecting the upper body kinematics of rear seat occupants (Meng et al., 2025; Youn et al., 2015). Previous studies investigating the effects of belt anchor location on occupant kinematics are detailed in Section 2.7.

2.3 Small Stature Female Human Surrogates

Testing for vehicle occupant safety can involve physical surrogates or numerical models. Physical surrogates for testing are post-mortem human subjects (PMHS), human volunteers, animals and animal cadavers, and the Anthropomorphic Testing devices (ATDs) (Carpanen et al., 2016; Crandall et al., 2011). PMHS were crucial in creating injury risk curves and provide the advantage of anatomical accuracy; however, a significant drawback is the availability of PMHS that meet the criteria of a test (Carpanen et al., 2016). The drawbacks for human volunteers are the inability to expose subjects to injury-inducing loading and, for animal testing, differences in anatomy and physiology (Carpanen et al., 2016).

ATDs represent the human body when evaluating occupant safety (H. J. Mertz & Irwin, 2015) and simulate the kinematics and kinetics of the human body's response in impact. ATDs are also fitted with instruments to measure forces, deflections, and accelerations experienced in frontal crashes, allowing for the indication of potential injury using injury criteria while also being durable for repeatable experiments (Carpanen et al., 2016). Initially developed for air force ejection testing, the ATDs were adopted for the frontal impact crash tests. The Hybrid II, developed in 1972, was a commonly used ATD to assess belt systems and mimicked the mass and size of the 50th percentile adult male but lacked instrumentation and was too stiff. The Hybrid III 50th percentile male ATD (HIII-50M) was developed to address these deficiencies, with more human-like response for the head and neck, with more instrumentation in the head, neck, upper torso, pelvis and spine, and replaced the Hybrid II in crash testing for frontal crash, becoming the primary regulatory ATD for crash testing in 1997. The Hybrid III 5th percentile (HIII-5F) female ATD was developed in 1987, with the size and

lengths of the ATD matching those of the 5th percentile female in the US population. However, aside from size, the HIII-5F was identical to the HIII-50M, lacking anatomical accuracy (Linder & Svensson, 2019; Tang et al., 2022).

As a result of the HIII ATDs' limited ability to accurately represent and measure chest deflections, the Test Device for Human Occupant Restraint (THOR) 50th percentile male (THOR-50M) was developed starting in the early 1990s, and had more human-like anatomy, with a two segment thoracic spine, a more anatomically accurate rib cage, and passive muscle force simulations for the neck and ankles (H. J. Mertz & Irwin, 2015). The improved thoracic design and implementation of advanced instrumentation, such as three-dimensional deformation measurement of the rib cage, have provided improved capability to assess thoracic injury risk (National Highway Traffic Safety Administration, 2023).

2.3.1 THOR-5F Physical ATD and Biofidelity

A 5th percentile female version of the THOR ATD (THOR-5F) had initially been designed in 2003 (McDonald et al., 2003) but had not been updated until 2015 (Wang et al., 2017). The THOR-5F (Figure 2.2) had notable anatomical and instrumentation updates from the HIII-5F, including a combined skull and facial load cell mounting plate, a redesigned clavicle designed for shoulder belt contact and engagement based on the statistical shape model of the female clavicle (Lu & Untaroiu, 2013; Wang et al., 2017). The THOR-5F sternum and breast were integrated to address variances in testing due to breast location, resulting in a more anatomically accurate representation and the pelvis design was based on a female-specific pelvis bone model (Klein et al., 2016; Wang et al., 2017). The updates to the THOR-5F provided anatomical accuracy that was missing in the HIII-5F. Additionally, the THOR-5F can record up to 174 channels, which is over 100 channels more than the HIII-5F instrumentation (Humanetics Innovative Solutions, n.d.; Shah et al., 2017). The THOR-5F features four Infra-Red Telescoping Rods for the Assessment of Chest Compression (IRTRACC) sensors, which allow for more accurate measurement than the potentiometer used in the HIII-5F.



Figure 2.2: THOR-5F ATD seated in a testing bench. Photo taken by Transport Canada.

The THOR-5F ATD has undergone design updates throughout the past few years, from its initial design described in 2017. Wang et al. (2018) evaluated the biofidelity of the THOR-5F ATD body regions using the third version of the BioRank scoring system (Table 2.2) from Rhule et al. (2018) and found the overall biofidelity of the ATD to be 1.28, within the “good” range. The head and shoulder biofidelity were excellent, while the neck, thorax, abdomen, pelvis and lower extremity were in the “good” range. An updated version of the ATD design, from a package in September 2019 recorded an overall biofidelity of 1.52, corresponding to a “good” rating, with only the shoulder scoring an “excellent” rating while the head, thorax, Knee-thigh-hip and lower extremity were rated as “good”. The neck and abdomen both scored above 2.0, which was categorized as “marginal” (Wang et al., 2020).

Table 2.2: Biofidelity rankings, associated scores, and statistical interpretation (Wang et al., 2020).

Biofidelity	BioRank score (B)	Statistical interpretation
Excellent	$B \leq 1.0$	Within 1 standard deviation of the mean
Good	$1.0 < B \leq 2.0$	Between 1–2 standard deviations of the mean
Marginal	$2.0 < B \leq 3.0$	Between 2–3 standard deviations of the mean
Poor	$B > 3.0$	Outside 3 standard deviations of the mean

The THOR-5F biofidelity was evaluated through subsequent sled tests in a simplified frontal impact sled configuration (“gold standard” condition), which was conducted by The Ohio State University and the University of Virginia (Wang et al., 2021). Using the NHTSA’s updated method for evaluating biofidelity, the overall BioRank score was 0.89, corresponding to “excellent” biofidelity. The biofidelity ranged from 0.67–1.0 for the head, spine, thorax and pelvis, corresponding to “excellent” biofidelity, while the belt load scored 1.23, which was rated as “good” biofidelity. An additional study investigated the THOR-5F abdominal loading, finding that the THOR-5F ATD showed good biofidelity overall, with a BRS score of 1.84, however, the THOR-5F showed a softer initial stiffness in all loading conditions compared to the PMHS and a rapid increase of force when compressed beyond 30% abdomen depth (Ramachandra et al., 2023).

A study by Richard et al. (2022) investigated the biofidelity of the THOR-5F ATD in the sled setup discussed in Section 2.5.2, with three configurations: Configuration 1 representing the standard rear seat, and Configurations 2 and 3 representing the submarining configurations. From the submarining configurations (Configurations 2 and 3), the THOR-5F demonstrated similar submarining behavior to the PMHS overall, with slightly later submarining times and similar pelvis excursion. In the first configuration (non-submarining), the pelvis excursions of the ATD were higher, which was explained by the difference in flesh thickness between the THOR-5F ATD and the PMHS. Additionally, the lap belt loads and pelvis kinematics were generally similar to those of the PMHS, with slight delays in restraint engagement and kinematic response, potentially due to anthropometric differences between the PMHS and ATD, as outlined by the author.

Somasundaram et al. (2023) also investigated the THOR-5F biofidelity seated in upright and reclined positions in a sled buck containing a semi-rigid seat and anti-submarining pan from Uriot et al. (2015), with a seat integrated 3-point belt. The configuration with the THOR-5F underwent two crash pulses: a 4.5 g and a 9.5 g crash pulse. The THOR-5F biofidelity was evaluated using an updated version of the BRS (Hagedorn et al., 2022). In the upright configuration, the THOR recorded average BRS scores of 0.86 for the low crash pulse and 1.16 for the higher magnitude crash pulse, corresponding to “excellent” and “good” biofidelity, respectively (Somasundaram et al., 2023). The head, spine, pelvis and seat belts, similar to other THOR-5F evaluations, had “excellent” biofidelity in the low crash pulse; however, the average scores of the head, pelvis and belts were within “good” biofidelity (1.01–1.39) for the high crash pulse.

The THOR-5F is still undergoing design updates to improve durability (National Highway Traffic Safety Administration, 2026b). Durability concerns with the THOR-5F have been identified from rear seat sled testing with a 40 g crash pulse, in which severe rib deformation and IR-TRACC failures were observed, indicating a need to improve durability while maintaining biofidelity (Hutter et al., 2024). Another design focus is the THOR-5F ankle, which is less stiff than the PMHS ankle. The current release of the THOR-5F, as of October 2025, is the Standard Build Level C (Cellbond, n.d.; Humanetics, 2024), and the THOR-5F ATDs used in this study are Standard Build Level B (SBL-B).

2.3.2 ATD Injury Criteria and Injury Risk Metrics

For ATDs to effectively represent human occupant response in crash testing, the relationship between the ATD measurements and the resulting injury risk is crucial, as ATDs output a parametric response rather than direct injury information (Xu et al., 2018). Injury criteria relate a physical response parameter with the probability of an injury through a specific injury mechanism, and are based on experimental studies and empirical data (Schmitt et al., 2025).

The Abbreviated Injury Scale (AIS) is a standard method for classifying injuries by body region and severity in automotive injury research (“Abbreviated Injury Scale (AIS),” n.d.). Injuries are ranked from 1 to 6, with 1 being a minor injury and 6 being a maximal injury (Table 2.3).

Table 2.3: Abbreviated Injury Scale (AIS) code, associated severity and thoracic injury example for each code (adapted from the Association for the Advancement of Automotive Medicine; Cavanaugh & Yoganandan, 2015).

AIS code	Injury severity	Injury
1	Minor	1 rib fracture
2	Moderate	2 rib fractures, sternum fracture
3	Serious	3+ rib fractures, unilateral flail chest with 3–5 flail ribs
4	Severe	Unilateral flail chest with >5 ribs
5	Critical	Bilateral flail chest
6	Maximal (Untreatable)	Aortic laceration with hemorrhage not confined to mediastinum

With injury severity defined using AIS, injury risk curves are developed to establish a statistical relationship between the loads measured by the ATD and the probability of injury at a given severity level (Xu et al., 2018). The injury curves are developed from statistical relationships between injury occurrence and the injury criterion experienced by the body when under external load from impact (Petitjean et al., 2015). Injury risk curves are developed using PMHS responses to a known stimulus, such as acceleration, force and the recording of any failure in the test samples (Prasad, 2015). From the injury risk curve, an Injury Assessment Reference Value (IARV) was selected such that if the measured value did not exceed the IARV, then the risk of the corresponding injury would be considered “unlikely”, which was defined as less than 5% (H. J. Mertz, 2002). The IARVs for the HIII-5F are ATD-specific and are scaled from the HIII-50M based on size and mass (H. Mertz et al., 1989).

The regulatory standards for frontal impacts in the United States and Canada (FMVSS/CMVSS 208) also define required thresholds for vehicle manufacturers, with the defined thresholds also based on PMHS and animal testing (Prasad, 2015). For the HIII-5F in frontal impact the head, neck, thorax and femur are evaluated (Table 2.4) (National Highway Traffic Safety Administration, 2026a).

Table 2.4: Injury metrics for the HIII-5F ATD in the FMVSS/CMVSS 208 standard for frontal impact (National Highway Traffic Safety Administration, 2026a).

Metric	Threshold
Head injury criterion (HIC_{15})	700
Neck injury criterion (N_{ij})	1
Neck axial force (F_z) [kN]	-2.62 (compression), 2.52 (tension)
Thorax acceleration [g]	60
Thorax deflection [mm]	52
Femur axial force [kN]	6.805

The injury risk curves for the THOR-5F are still being developed, with a need for matched PMHS testing (Boyle et al., 2024). NHTSA is currently conducting THOR-5F tests in a simplified sled test configuration that was used to record PMHS and HIII-5F data, and a complex configuration with advanced seat belts and airbags (National Highway Traffic Safety Administration, 2026b).

2.3.2.1 Thoracic Injuries and Injury Metrics

Thoracic injuries are grouped into skeletal (e.g. rib fractures) and soft tissue injuries (e.g. lung contusion). Soft tissue thoracic injuries may involve the respiratory system, heart and aorta (Schmitt et al., 2025). Rib fractures are, in general, the most frequent type of AIS 3+ chest injuries (Leport et al., 2011) and may also lead to hemothorax or pneumothorax (Cavanaugh & Yoganandan, 2015). The thorax was observed to be more vulnerable to injury under concentrated belt loading, which results in thoracic compression (Schmitt et al., 2025).

Rear seat occupants have experienced both skeletal and soft tissue injuries, including rib fractures, lung contusions, hemothorax, pneumothorax, and heart or vessel damage, with 20 of the 22 occupants' chest injuries due to shoulder belt loading (Jermakian et al., 2019). While children generally suffered soft tissue injuries, such as lung contusions, older occupants more frequently sustained skeletal injuries, such as rib fractures.

Arbogast et al. (2012), examining the Crash Injury Research and Engineering Network (CIREN) database, found that multiple and bilateral rib fractures were more common among adult (25+ years) occupants. The injured adults were generally involved in lower severity impacts, but all sustained complex rib fractures, with 7 of the 8 experiencing multiple rib fractures, and 4 of the 8 bilateral. Belt compression was identified as the primary injury mechanism. Additional injuries that were observed included diaphragm laceration, sternum fracture, and mediastinal hematoma.

Lung contusion is generally considered to be rate dependent, and the occurrence of rib fractures, along with hemothorax and pneumothorax are considered to be primarily deflection dependent (Cavanaugh & Yoganandan, 2015). In the frontal impact regulation (FMVSS/CMVSS 208), the thoracic injury criteria for the HIII-5F include a maximum chest deflection of 52 mm and a maximum thoracic acceleration of 60 g over a 3 ms cumulative interval (National Highway Traffic Safety Administration, 2026a). The limits were derived from injury risk curves from human surrogate testing (Bandak et al., 1999). The regulatory chest deflection limit was based on the 5% AIS 4+ injury risk value, instead of the AIS 3+ injury risk value. The reason for using AIS 4 was that AIS 3 was generally associated with rib fractures, which children rarely experienced, and AIS 4+ covered both rib fractures and soft tissue injuries (Bandak et al., 1999). A THOR-5F injury risk criterion is currently under consideration (National Highway Traffic Safety Administration, 2026b). The maximum deflection across the 4 chest sensors is used in the Euro NCAP frontal test procedures for the THOR-50M (Euro NCAP, 2025).

NHTSA developed preliminary THOR-5F thoracic injury risk functions using matched-pair small stature female PMHS and THOR-5F sled tests. The predictor variables investigated were the maximum deflection across the four THOR-5F chest sensors in the X direction, X-Y resultant direction, and X-Y-Z resultant direction (Lee & Craig, 2021). The data from the matched-pair tests were then used to generate AIS 3+ thoracic injury risk functions based on the respective chest deflection metrics (Figure 2.3).

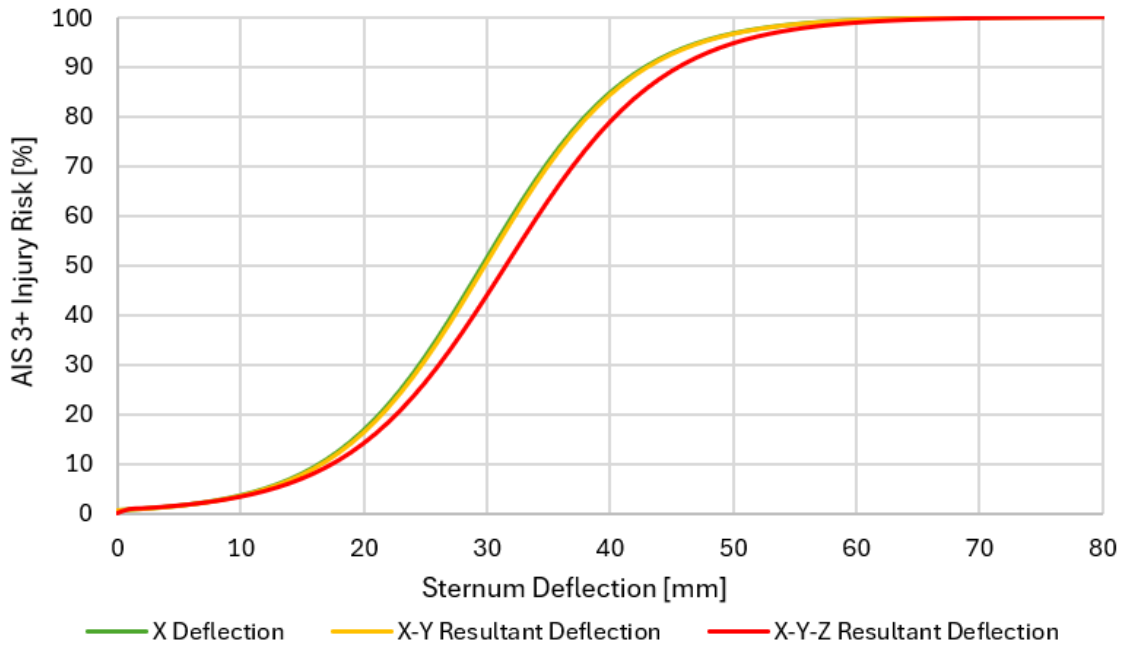


Figure 2.3: Preliminary AIS 3+ injury risk curves for sternum deflection in the X direction, the X-Y resultant direction and the X-Y-Z resultant direction.

2.4 Safety Regulations and Evaluation for Rear Seat Occupants

Despite numerous tests to evaluate front seat occupant safety in frontal impacts, there is limited regulation for the injury assessment of rear seat occupants (Jermakian et al., 2019). While the regulations do not consider the rear seat occupants in frontal crash testing standards, consumer tests, which evaluate and compare vehicles to encourage manufacturers to design beyond the minimum requirements (Schmitt et al., 2025), have started to implement or consider implementing rear seat adult occupants in its evaluation.

The National Highway Traffic Safety Administration (NHTSA), the governing body of automobile safety, began considering improvements to the New Car Assessment Program (NCAP), with rear seat occupant safety among them. Additionally, several NCAP programs worldwide have begun to implement the HIII-5F in the rear seat for frontal impact tests (Tang et al., 2022). In the European

version of the NCAP (Euro NCAP), a HIII-5F is placed in the passenger side of the rear seat for the Full Width Deformable Barrier (FWDB) test with a 56 km/hr pulse, with metrics from the head, neck, chest, abdomen, knee, femur, pelvis and lower extremities evaluated (Table 2.5). The standard also uses head excursion to modify the score for the head region (Euro NCAP, 2025).

Table 2.5: Injury metrics for the HIII-5F ATD in the Euro NCAP evaluation for frontal impact (Euro NCAP, 2025).

Criterion	Value/Range	Cap
Head injury criterion (HIC_{15})	500–700	700
Head resultant acceleration ($A_{res-3ms}$) [g]	72–80	80
Neck shear force ($F_{x, shear}$) [kN]	1.2–2.0	NA
Neck tension force ($F_{Z, tension}$) [kN]	1.70–2.6	NA
Neck extension moment ($M_{Yextension}$) [Nm]	36–49	NA
Chest compression ($D_{chest\ compression}$) [mm]	22–42	42
Viscous criterion ($V_{viscous\ criterion}$) [m/s]	0.5–1.0	1.0
Femur force (F_{femur}) [kN]	2.6–6.2	NA
Knee displacement (D_{Knee}) [mm]	6–15	NA
Tibia index (I_{tibia})	1.3	NA
Tibia force (F_{tibia}) [kN]	8.0	NA

The Insurance Institute for Highway Safety (IIHS), a nonprofit scientific organization dedicated to vehicle safety, has initiated the process of integrating second-row occupant protection for scoring and awarding vehicles. An additional HIII-5F ATD is placed in the rear seat of the vehicle in a moderate overlap frontal crash test, in which the vehicle collides with a barrier with 40% overlap at 60 km/hr (Insurance Institute for Highway Safety, 2025; Lam et al., 2014). The testing considers the injury risk measured by the ATD in the crash, chest injury risk in the rear seat, and intrusion in the occupant compartment at specific locations (Insurance Institute for Highway Safety, 2025). The testing considers head acceleration measurements, neck force measurements for tension and compression,

and chest index, which accounts for both chest deflection and shoulder belt positioning (Table 2.6). The shoulder belt tension and femur axial compression are also measured. Head excursion, head contact with the seatback, submarining, and the belt positioning to cause belt-neck contact are also evaluated for demerit scores.

Table 2.6: Injury metrics for the HIII-5F ATD in the IIHS evaluation for frontal impact (Insurance Institute for Highway Safety, 2025).

Metric	Good	Acceptable	Marginal	Poor
Head injury criterion (HIC ₁₅) (only with contacts)	560	700	840	>840
Head resultant acceleration from contact [g]	>70: Downgrade scores for head and neck region			
Neck injury criterion (N _{ij}) (only with contacts)	0.80	1	1.20	>1.20
Neck axial tension [kN]	2.0	2.4	2.8	>2.8
Neck compression force [kN]	2.0	2.5	3.0	>3.0
Chest index	35	40	45	>45
Shoulder belt tension [kN]	< 5.9	NA	<5.9	NA
Femur axial compression [kN]	4.9	6.2	7.4	>7.4

While consumer evaluation has progressed to include the rear seat occupant in testing, most of the rear seat standards in the United States and Canada are based on static criteria, such as the location of the seat belt anchor points and the seat belt assemblies (Durbin et al., 2015). According to the Federal Motor Vehicle Safety Standard (FMVSS) 210 in America, and its Canadian equivalent, the Canadian Motor Vehicle Safety Standard (CMVSS) 210, the belt anchor locations for the upper anchor must be within a region that is defined by measuring the SAE manakin in the seat (Transport Canada, n.d.; U.S. Department of Transportation, n.d.). From the shoulder reference point of the SAE manakin, the upper anchor (D-ring) should be 40 degrees below or 80 degrees above the horizontal at a point 151.8 mm above the shoulder reference point (Figure 2.4). For the lower anchor point, the angle between the horizontal and the H-point must be between 75 and 30 degrees.

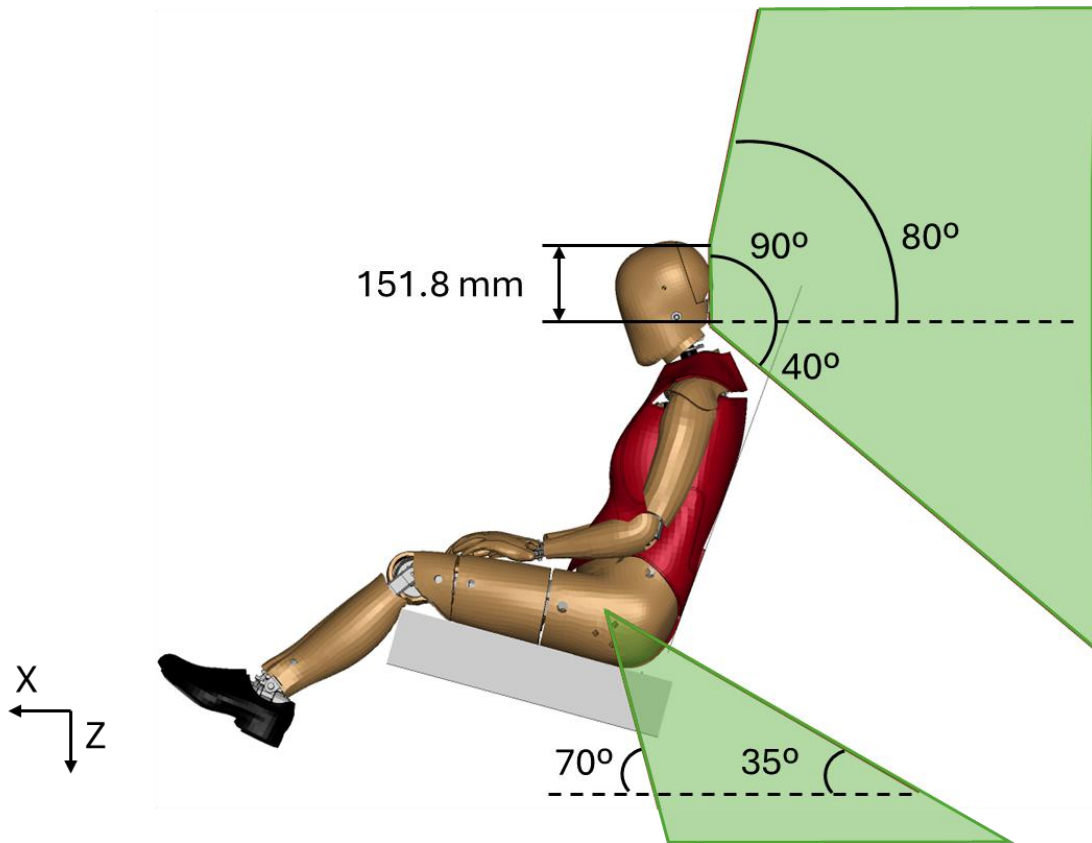


Figure 2.4: Acceptable seat belt anchor locations from FMVSS/CMVSS 210 standards, adapted to the THOR-5F in Rigid Bench described in Section 3.2. The green regions indicate the acceptable seat belt anchor locations (Transport Canada, n.d.; U.S. Department of Transportation, n.d.).

2.5 Sled Test Benches Representing the Rear Seat Environment

Sled testing is a type of automotive safety testing used to evaluate vehicle component or restraint system performance by applying a controlled acceleration or deceleration pulse to a sled and the attached vehicle components (Schmitt et al., 2025). Sled tests are repeatable and less costly than full-scale vehicle crash testing, and allow components of the test rig to be reused (Schmitt et al., 2025; Smyth & Smith, 2007).

Several types of sled-mounted test setups have been reported in the literature, including setups described as bucks or benches. Examples of sled-mounted setups include bucks with production vehicle seats (e.g. Dehner et al., 2007), a buck with a rigid seat (e.g. Luet et al., 2012), a bench with a foam seat (Wietholter et al., 2021), and a bench with a semi-rigid seat, referred to as the Semi-Rigid Bench (Uriot et al., 2015). Tests conducted in production seats were found to be inconsistent, whereas rigid and semi-rigid seats enable more repeatable and reproducible testing (Luet et al., 2012; Uriot et al., 2015). The rigid seat pan has no articulation, whereas the physical Semi-Rigid Bench uses a spring-loaded seat pan to enable articulation.

2.5.1 Federal Motor Vehicle Safety Standard (FMVSS) 213 Sled and Bench

The FMVSS 213 test bench is an example of a standard rear seat bench used in regulatory safety testing, specifically for child restraint testing (Wietholter et al., 2021). The test bench was initially based on the 1974 Chevrolet Impala bench seat, then upgraded in 2003 by incorporating child ATDs, associated injury criteria and making it more representative of rear seats in the fleet at the time (Wietholter et al., 2021). However, not all of the recommended modifications were implemented. From regulation updates, the bench was further modified to match rear seat geometry, as measured in a report by Aram & Rockwell (2012), in which rear seat geometry, including belt anchor locations, seat pan lengths, and seat back geometries, was measured. The bench has seat cushions, shoulder belt, and lap belt anchorages and tether anchorages (Wietholter et al., 2021).

2.5.2 Physical Semi-Rigid Bench

While benches with rigid seat pans provided a stable and consistent testing environment, one concern was that the kinematic response of an occupant may not represent occupant reaction in a rear seat of a vehicle (Uriot et al., 2015). The physical Semi-Rigid Bench (Figure 2.5) was introduced in a study by Uriot et al. (2015), in which the authors examined the submarining ability of ATDs in a reproducible setup. The notable difference between rigid seat benches and the physical Semi-Rigid Bench is the articulation of the bench seat pan.

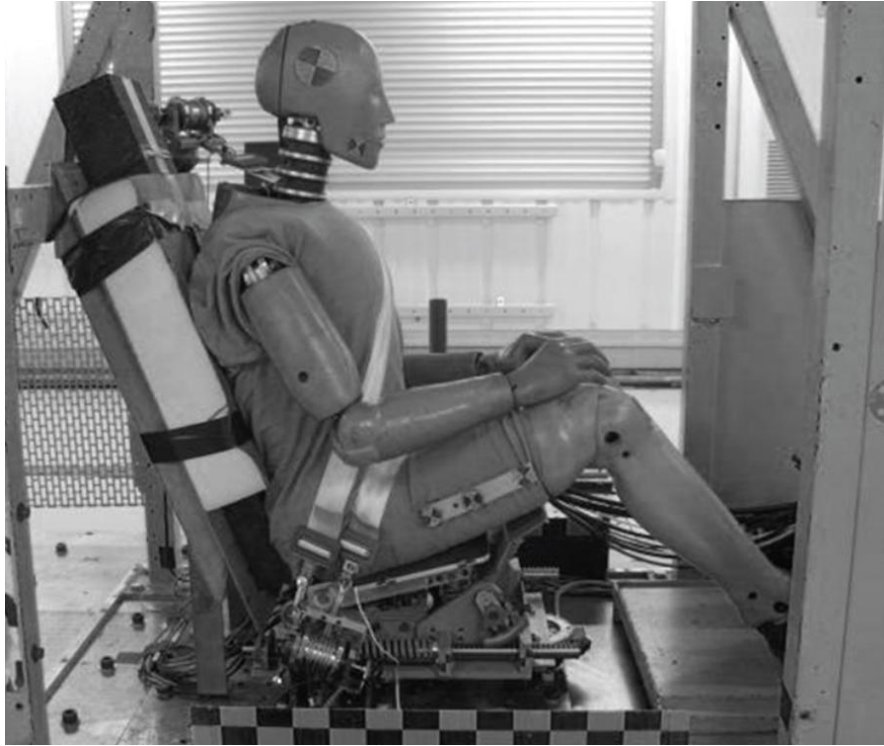


**Figure 2.5: Semi-Rigid Bench configuration with HIII-50M positioned (Uriot et al., 2015).
Picture used with permission from the Stapp Association.**

The Semi-Rigid Bench consisted of a rigid seat pan, with its stiffness controlled by two springs located underneath the plate, with blocking systems to manage rebounding. The backrest consisted of a rigid plane fixed to a rigid frame and covered at the higher thoracic level by foam and at the lower pelvis level by a polypropylene pad. The bench could be configured to emulate a front or rear seat by adjusting the spring stiffness, articulation position, and the belt anchor positions.

The physical Semi-Rigid Bench was also used to evaluate the THOR-5F response in submarining compared to PMHS data in a previous study. The study, conducted by Richard et al. (2022), conducted 6 tests with the THOR-5F placed in the test configuration described by Trosseille et al. (2018) . The data measured by the THOR-5F was then compared to the PMHS data from Trosseille et al. (2018) to qualitatively assess the biofidelity of the THOR-5F. In the study, the bench was set to 3 configurations, with the first configuration being the standard rear seat configuration, with the anti-

submarining pan being lowered from and rotated to avoid contact with the ATD, while the second and third configurations featured modifications to the anchor locations to favor submarining. The deceleration pulse of the sled was set to 9 g for the first configuration, while the deceleration for the second and third configurations was set to 15 g.

2.6 Finite Element Modelling in Occupant Safety Testing

In addition to physical testing, computational models are applied in areas of safety engineering, such as vehicle crashworthiness design, and can complement existing data (Schmitt et al., 2025).

Computational modelling, in addition to providing an efficient and cost effective analysis of occupant safety (Kan et al., 2003), enables the changing of design parameters to find trends and optimize designs (Yang & Chou, 2015). Computational models have become a recognized tool in vehicle safety engineering, as validated models can provide accurate and predictive assessments (Schöneburg et al., 2026). Models have been developed for many purposes in occupant safety, including occupant kinematics studies and restraint system development, such as airbags and seat belts (Yang & Chou, 2015).

Computational models require three key inputs: boundary conditions (e.g. force), geometry and material properties (Schmitt et al., 2025). There are two common approaches to computational modelling: the multi-body system approach (e.g. MADYMO) based on rigid body dynamics and the finite element method. While the multi-body system approach condenses structures such as vehicle components, resulting in an efficient analysis, the finite element method allows for detailed response analysis, including complex geometry and material nonlinearity (Schmitt et al., 2025).

Finite element methods for vehicle structural modelling emerged in the automotive industry during the 1980s and expanded significantly through the 1990s and early 2000s as computational power increased. As computational capabilities have increased, FE models have become widely adopted in automotive safety applications (Yang & Chou, 2015). Finite element models are formulated by discretizing the geometry into a mesh consisting of elements and nodes. Boundary value problems are then solved to determine the nodal displacements from which the element stresses and strains can be obtained (Noorumar et al., 2022). The FE method is typically implemented using two approaches: the explicit formulation, which solves dynamic response over small time increments, and the implicit formulation, an approach grounded on force equilibrium (Schmitt et al., 2025). For impact

simulations, which typically involve short durations and complex contact interactions between components, explicit formulations are more suitable (Schöneburg et al., 2026).

Computational models undergo several stages before application, including model conceptualization, development, verification, and validation (Schmitt et al., 2025). Verification assesses that the model has been implemented correctly and solves the governing equations as intended, while validation compares the model outputs to an independent dataset to evaluate the accuracy of the real-world representation (Ray et al., 2011; Schmitt et al., 2025).

2.6.1 Sled FE Models

Several studies have used FE sled models to validate ATD models, evaluate or compare ATD responses, explore restraint and seat parameters, and optimize safety design. Maatouki et al. (2018) used a simplified sled model with a rigid seat to validate a THOR-50M FE model, and Teng et al. (2007) created a frontal sled and evaluated four different HIII-50M FE models, with varying levels of rigid material defined. In a project investigating the methods to assess thoracic injury risk to elderly occupants, Eggers et al. (2017) developed a frontal occupant sled test setup and a matching FE model. The sled model included an airbag, a rigid seat, and a three-point seat belt, and was validated by comparing the THOR-50M FE model response with paired physical tests. Kang et al. (2018) defined restraint system parameters, including belt material properties and a belt buckle friction coefficient, from controlled testing, and used a HIII-50M FE model in a rigid seat sled model to compare restraint system response with physical tests.

FE rear seat sled models also include models of production seats and simplified rear seat environments. NHTSA currently has a publicly available Honda Odyssey production rear seat model, which underwent testing in multiple configurations with a THOR-50M FE model, including a frontal test (Bridges et al., 2022). Meng et al. (2022) created FE models based on geometric data from 3D scans of rear seats and assessed the seat models based on the occupant injury criteria recorded by THOR-50M and HIII-50M FE models, with the simulation results then used to select seats for physical testing. Meng et al. (2025) also developed a simplified FE model of a 2018 sedan rear seat, validated the model against physical sled test data, and used the model to examine parameters such as seatback angle and anchor locations, identifying optimal rear seat parameters for the HIII-50M and THOR-50M ATD models.

Sled models representing the fleet rear seats or corresponding to physical rear seat sleds have also been explored. Youn et al. (2015) created a rear seat frontal impact sled model based on the UN R44 sled, modelled polyurethane foam for the FE seat model, and included a front seat. The model was then used to evaluate the response of the HIII-5F and HIII-50M with different vehicle-specific anchor configurations. The FMVSS 213 sled bench has also been modelled and used to evaluate child restraint performance. One study used an FE model of the FMVSS 213 sled for the parametric evaluation of booster seats for children (Miller et al., 2021). Another study used an FMVSS 213 sled bench FE model to evaluate the model performance of a child ATD model in different booster seats (Challa et al., 2025).

2.6.1.1 Open-Source Semi-Rigid Bench FE Model (ENOP Bench)

The physical Semi-Rigid Bench seat pan was also modelled in the open-source Enable New Occupant Seating positions (ENOP) seat FE model (Figure 2.6) (Lebarbé et al., 2023). Industry partners and academia created the ENOP project to understand the effects of non-conventional seating positions, such as reclined seating, on occupant kinematics and potential new injury patterns (Östling et al., 2022). The ENOP sled included a semi-rigid seat pan, a sled floor, a seat back, foot and knee supports, and a seat belt system. The seat pan model within the ENOP project sled model matched the seat pan developed by Uriot et al. (2015), consisting of articulated seat pan and anti-submarining pan assembly, which included the anti-submarining pan rack and stoppers.

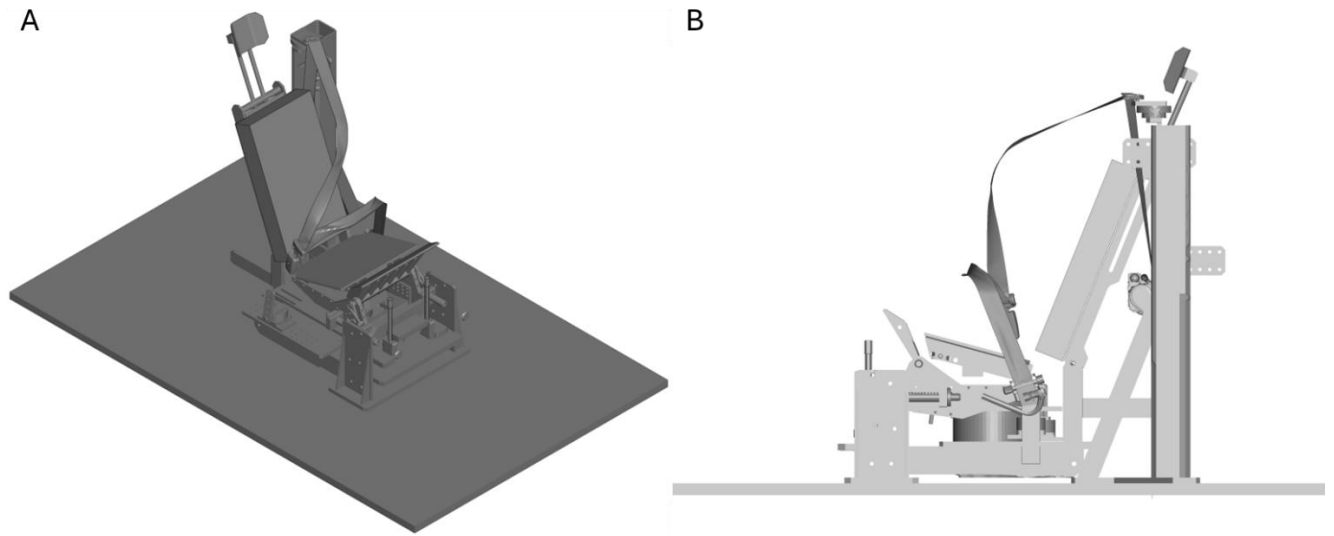


Figure 2.6: (A) Isometric view and (B) side view of the ENOP sled model with belt system (Lebarbé et al., 2023). Note: Knee support not shown.

2.6.2 Seat Belt FE Models

The FE models of seat belts were initially modelled as 1D elements. Then the hybrid model (Figure 2.7) was introduced, with 2D elements for the belt fabric and 1D elements to use the slipping function in LS-DYNA, which allows the belts to interact with a user-defined slipping (Dahlgren et al., 2020). The 1D seat belt elements connect the 2D seat belt elements to the slipping and seat belt anchors.

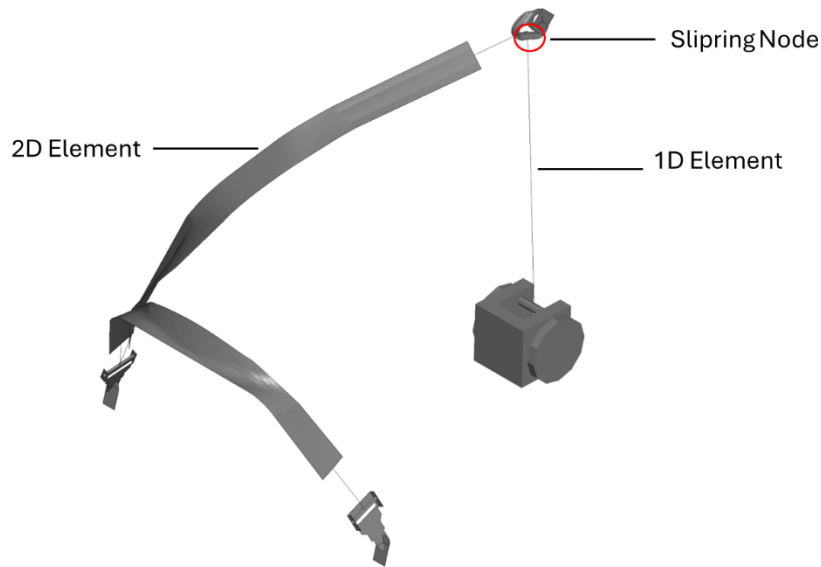


Figure 2.7: FE model of 3-point belt with 1D elements for connection to belt anchors and slirings and 2D elements for occupant contact.

The slip rings allow continuous sliding of the belt through a change in angle. The user defines the belt elements that meet at the slip ring node, and the belt elements must have a common node with the slip ring node. The elements pass through the defined slipring node. The amount of slip is determined by the ratio of the forces on the elements on each side of the slipring node. The angle between the elements and the defined coefficient of friction determines the ratio of forces (Livermore Software Technology, 2021).

The introduction of 2D belts was intended to improve interaction between the FE occupant model and the seat belt model routing around D-rings. Initially, the 2D belt elements were shell elements, but the recent introduction of a new material model (*MAT_SEATBELT_2D) allowed for a more accurate belt bending representation (Dahlgren et al., 2020).

Across literature, several seat belt models have been developed and made available, including a model from an LSTC belting tutorial, the Baudrit belt (Baudrit et al., 1999), and an open-source Autoliv belt model (Gepner et al., 2022) (Table 2.7). One of the most recently developed seat belt

models, the belt model for the ENOP sled (Section 2.4.2.1), used 2D seat belt elements with simplified webbing material, which was the average of the commonly used webbings in Autoliv.

Table 2.7: Geometry and material cards of three different belt system models.

Belt model	Belt element type	Belt width [mm]	Elements across width	Belt thickness [mm]	LS-DYNA material card
LSTC	Tria shells	50	4	1	*MAT_FABRIC
Baudrit	Quad shells	46	4	1.2	*MAT_SIMPLIFIED_JOHNSON_COOK
Autoliv	2D seat belt	47	8	1.2	*MAT_SEATBELT

2.6.3 FE Modelling of ATDs

Early ATD FE models were simplified or consisted of a single component due to computational constraints. However, current ATD models include detailed geometry with identical composition to their physical counterparts (Yang & Chou, 2015). The general process for developing an ATD FE model starts by creating the geometry of each part in CAD or from physical data, treating these parts as an assembly or component, and finally assembling the entire ATD model. The components and subsystem models are then validated through calibration of ATD parts, such as pendulum testing, and the full model is validated using sled or crash testing data (Yang & Chou, 2015). Current FE models of adult ATDs include the HIII-5F, HIII-50M, HIII-95M and the THOR-50M (Humanetics, n.d.). The THOR-5F FE model is currently being developed by companies such as Humanetics and ATD models GmbH (ATD Models GmbH, n.d.; Humanetics, 2024).

2.6.4 FE Model Validation

FE models of ATDs are validated by comparisons to physical tests (Xu et al., 2018). The assessment or validation of an FE model must be quantitative, and the initial validation should be with simple boundary conditions or impact scenarios (Schmitt et al., 2025). Recent model validation approaches include a model response comparison with experimental corridors, along with cross-correlation

between the model and experimental response (Schmitt et al., 2025). Cross-correlation compares two responses and provides rating on similarity based on the size, shape, and phase shift of the response.

The ISO 18571 standard is one of the approaches used to quantify the validation of a model (Schmitt et al., 2025). The standard is used to assess the correlation of two signals and to provide a quantified metric for the validation of computer models compared to physical testing through comparison of time history signals (International Organization for Standardization, 2024). The overall rating consists of four component ratings (Figure 2.8): the slope (E_s), magnitude (E_M), phase shift (E_p), and corridor ratings (Z).

The phase score is used to evaluate the phase differences between time histories, while the slope score evaluates the difference in topology between the signals and is calculated from time-shifted histories (International Organization for Standardization, 2024). The magnitude score measures the difference in amplitude between the two time histories when adjusted for any phase differences. The corridor score measures the deviation through two sets of corridors, an inner corridor and an outer corridor. If the curve is within the inner corridor of the reference curve, then the score will be 1, and if the curve is outside the inner corridor, the score between 0 and 1 is given, with lower scores as the curve deviates towards the outer corridor bounds. If the curve is outside the outer bounds, the given score is 0.

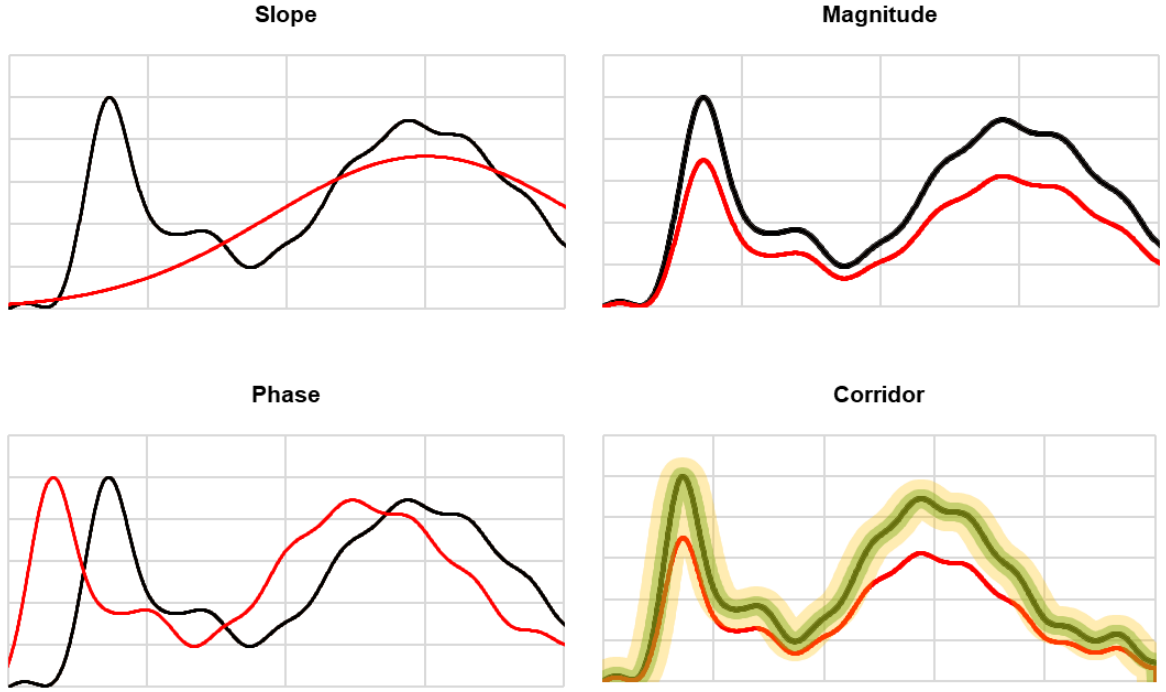


Figure 2.8: Component ratings considered for the overall ISO 18571 score, consisting of a slope rating, magnitude rating, phase rating and corridor rating. The red curve is the model while the black curve is the experiment. The green range defines the inner corridor, while the yellow area defines the outer corridor.

Each of the individual ratings ranges from 0 to 1, with 0 indicating no correlation between the two tested curves and 1 indicating perfect correlation between the two curves (International Organization for Standardization, 2024). The ratings are then weighted and summed according to the following equation (Equation 2.1), where w is the weight for each component, and R is the overall rating. All weights must add to 1.

$$R = w_Z Z + w_P E_P + w_M E_M + w_S E_S \quad (\text{Equation 2.1})$$

Higher values of R indicate a stronger match between the simulated and experimental response, with different rating grades for each rating range (Table 2.8).

Table 2.8: ISO ratings and corresponding grades (International Organization for Standardization, 2024).

Rating (R)	Grade
$0.94 < R \leq 1.00$	Excellent
$0.80 < R \leq 0.94$	Good
$0.58 < R \leq 0.80$	Fair
$0 \leq R \leq 0.58$	Poor

2.7 Previous Rear Seat Anchor Location Studies

2.7.1 Anchor Location Surveys

As part of a study to upgrade the FMVSS 213 sled to represent the current vehicle fleet, Aram and Rockwell (2012) measured the rear seat geometry of 24 2010 model year vehicles. The measurements included the shoulder belt and lap belt anchor distances from a point on the seat geometry measuring fixture (SGMF), which consisted of two wooden blocks, each measuring 600 mm x 88 mm x 38 mm, and a 76.2 mm (3 inch) hinge. When measuring the anchor locations, the distance from the intersection of the two wooded blocks at the hinge, with the SGMF positioned at the centerline of the seat, was taken. The measurements included the fore-aft, lateral and vertical positions of the shoulder belt anchorages, as well as the spacing in the lap belt anchor points. The study found wide variation in the upper anchor location, with the shoulder belt anchor located on the package shelf or side point. In contrast, for the center rear seat, the shoulder belt anchor locations were also found on the seat, roof and rear wall. Reed and Hamilton (2013) also investigated seat belt anchor locations, examining 83 passenger cars and light trucks using the SAE J829 (manikin) H-point procedure and coordinating measurement methods, finding that the shoulder belt and lap belt anchor locations varied widely within the accepted regulatory range. Youn et al. (2015) also reported that anchorage positions in the rear seat of a vehicle varied widely, even outside the regulatory acceptable ranges.

2.7.2 Effects of Shoulder Belt Anchor/D-ring Positioning on Occupant Response

Numerous studies have investigated the effects of shoulder belt anchor positioning on occupant kinematics using physical ATDs or FE models. A study examining the effects of belt routing on HIII-5F and THOR-5F chest deflections in contemporary vehicles found that belt routing closer to the neck reduced chest deflection. In contrast, for the THOR-5F, reduced chest deflection was observed at the maximum region, but increased chest deflection in another region (Eggers et al., 2023). Another study conducted in 2022 utilized physical frontal impact sled tests to assess the influence of the shoulder belt anchor (D-ring) vertical position on the sternum deflection in the HIII 5th percentile (HIII-5F) Anthropomorphic Testing Device (ATD), finding that higher vertical positions resulted in lower sternum deflections (Edwards et al., 2022).

Additionally, several studies have investigated the effects of D-ring locations on rear seat kinematics using various computational models of ATDs. Hu, Wu, Klinich et al. (2013) used the MADYMO models of a HIII-M50 a 12-year-old ATD, and a rear-facing infant in a CRS in sled tests to determine the effect of D-ring position on the head and knee excursions, along with torso rotations, finding that moving the D-ring forward (+X) and outboard (-Y) resulted in higher head excursion and torso angle for both the 6-year old and the M50, however a lower (-Z) D-ring was found to reduce head excursion for children while a higher D-ring was found to reduce head excursion for adults. For the adult ATDs, the optimal D-ring position was higher and rearward. Meng et al. (2025) also conducted a study with the HIII-M50 and the THOR-M50 FE models where a simplified rear seat model was created with seat foam and looked at the effects of modifying various seat parameters including lap belt and shoulder belt anchor locations, finding that that the location of the D-ring in Y and Z had the largest effect on chest injury of the HIII while the X direction had the largest effect for the THOR model. For both the HIII and THOR-50M, the optimal locations for the D-ring were narrow, high and rearward.

Youn et al. (2015) conducted computational simulations with the HIII-5F and HIII-50M multibody models with seat belt anchor locations for 28 vehicles, finding that the D-ring strongly affects the upper body motion, and for head injury and chest deflection, the fore-aft (X) location of the upper anchor location was the most dominant factor for both ATDs.

2.7.3 Effects of Lap Belt Anchor Positioning on Occupant Response

In addition to investigating the response to shoulder belt anchor positioning, several studies have examined the effect of lap belt anchorages on occupant response. Research at UMTRI found that for older children, rearward and higher lower anchorages reduced lower body excursion except during submarining (Hu et al., 2012; Hu, Wu, Reed, et al., 2013; Klinich et al., 2011; Reed & Ebert-Hamilton, 2013).

Uriot et al. (2006) used a hydraulic setup to dynamically rotate lap belt anchorages under constant lap belt tension and consistent pelvis positioning for a PMHS, THOR-NT, and HIII-50M. The study found that submarining occurred at a higher lap belt angle for the HIII-50M and THOR-NT than for the PMHS (Uriot et al., 2006).

Hu, Wu, Klinich et al. (2013) found that for the HIII-M50, the optimal lap belt position for reducing head and knee excursion was rearward and at a moderate height, whereas for the child ATD, the optimal position was a higher, forward lap belt. Meng et al. (2025) found that wider and higher seat belt anchorage locations resulted in lower overall occupant injury risk for the HIII-50M and the THOR-50M.

Chapter 3

Methods

3.1 Overview of Methods

The overall aim of this study was to use computational sled, occupant, and restraint system models to quantify the effect of shoulder and lap belt anchor position on the kinematics of a small stature female ATD in a frontal impact environment (Figure 3.1). The frontal impact environments investigated included the Rigid Bench developed by Transport Canada and PMG Technologies (TC/PMG) and the Semi-Rigid Bench reported in the literature (Richard et al. 2022). Computational models of each bench were developed, and a small stature female ATD finite element model (THOR-5F FE, Standard Build Level B, Humanetics, Farmington Hills, MI, USA) was positioned on each bench and restrained with a seat belt. The ATD FE model response was compared to physical test data, and then the model was used in a parametric study with varying restraint anchor points to quantify the effect of changing anchor position.

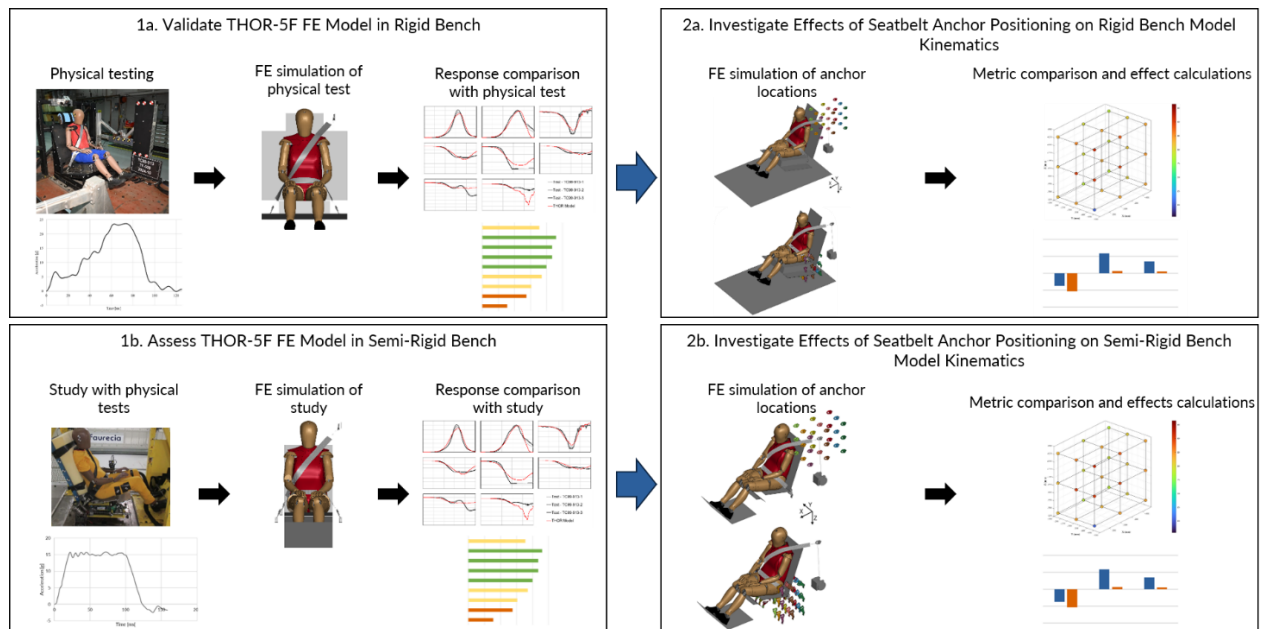


Figure 3.1: Overview of methods for (a) Rigid Bench and (b) Semi-Rigid Bench configurations with THOR-5F ATD FE model.

3.2 Rigid Bench and Sled Testing at Transport Canada

3.2.1 Rigid Bench Geometry and Sled Specifications

The Rigid Bench design, physical ATD positioning, and testing described in this section were conducted by TC/PMG. The Rigid Bench was designed as a representative but simplified rear seat environment to enable the comparison of physical and numerical human surrogates. A 3D scan of the Rigid Bench (Figure 3.2) was provided by TC/PMG using a MetraSCAN 3D scanner (Creaform, Lévis, QC, CA) with a mesh resolution of 0.1 mm and a measurement resolution of 0.025 mm. The 3D scan and CAD drawings of the Rigid Bench were also provided by TC/PMG.

The Rigid Bench comprised a seat pan, a seat back, a retractor, and shoulder and lap belt anchor assemblies. The shoulder belt anchor assembly consisted of the D-ring, which rotated about the shoulder belt anchor. The left (fixed) side and right (buckle) side lap belt anchor assemblies each included a seat belt webbing guide connected to the bench using welded chain links to provide freedom of motion. The Rigid Bench also included a raised platform (Figure 3.2B) to vertically offset the THOR-5F from the lap belt anchors, as would occur in a typical vehicle rear seat configuration. The structural components of the Rigid Bench were constructed from steel.

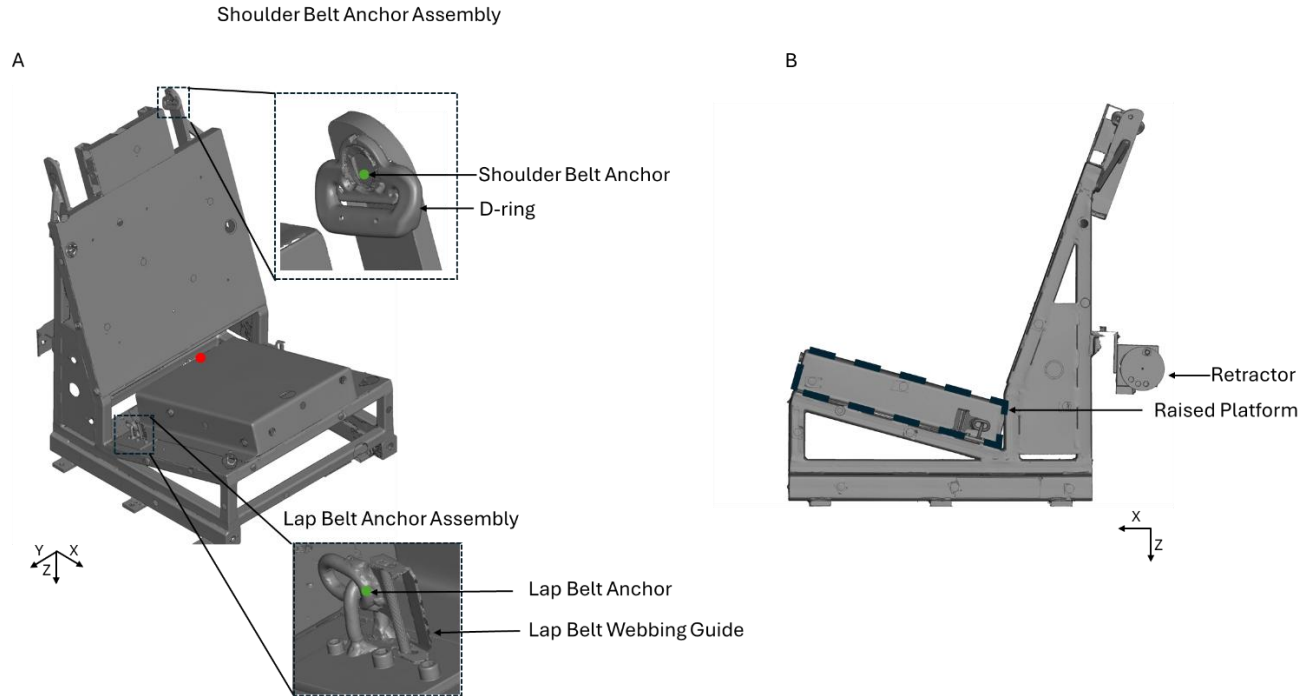


Figure 3.2: 3D scan of Rigid Bench used in physical tests. (A) Isometric view of physical Rigid Bench with shoulder and lap belt anchor assemblies identified, where the red circle represents the origin (0,0,0) and the green circle represents the seat belt anchor. (B) Side view of Rigid Bench with retractor.

The SAE coordinate system (+X defined as forward, +Y defined as toward the right side of the Rigid Bench, +Z defined as downward) was used (Figure 3.2), with the origin located at the mid-width of the raised platform surface rear edge (red circle in Figure 3.2A). The seat pan, seatback lengths and angles of the Rigid Bench (Figure 3.3) were based on the FMVSS 213 bench, which was designed to represent an average rear seat configuration (Aram & Rockwell, 2012). The shoulder and lap belt anchor locations of the Rigid Bench (Table 3.1) were based on the FMVSS 213 bench, intended to represent an average rear seat configuration (Aram & Rockwell, 2012). The seat belt anchor locations remained consistent across the experimental tests.

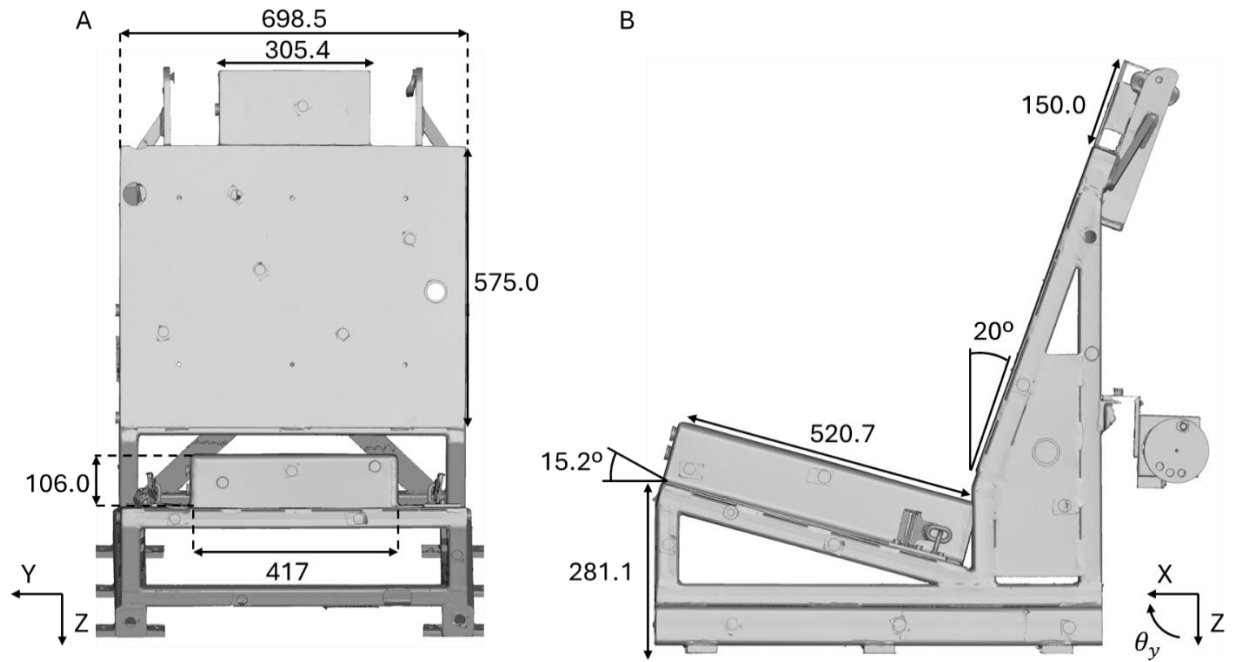


Figure 3.3: Dimensions of Rigid Bench used in physical tests with THOR-5F ATD: (A) Front view and (B) side-view. All linear dimensions are in millimeters.

Table 3.1: Rigid Bench seat belt anchor locations reported with respect to origin (red circle in Figure 3.2A).

Anchor	X [mm]	Y [mm]	Z [mm]
Shoulder belt anchor	-316.9	-262.0	-699.0
Right lap belt anchor	45.2	286.6	62.0
Left belt anchor	45.2	-286.6	62.0

The seat belt retractor used in the Rigid Bench (Figure 3.4) was a custom-built retracting device with a base that housed a rotating shaft to wind the belt webbing and a locking pin (Tang et al., 2019). When the shaft rotated, it released the belt webbing, and the locking pin moved horizontally to a

position where the retractor locked. Once locked, a small amount of pullout could occur under seat belt tension generated during ATD loading.

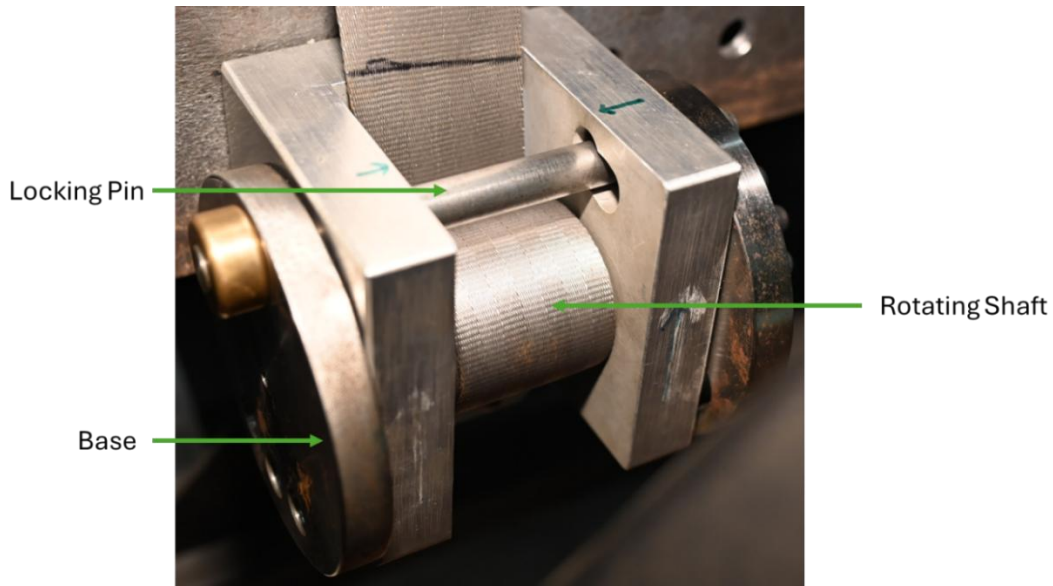


Figure 3.4: Seat belt retractor (Tang et al., 2019) used in the Rigid Bench test setup. Image provided by TC/PMG.

The retractor pullout as a function of time was measured from a video recording of a previous sled test conducted by TC/PMG on an earlier bench version. The seat belt webbing was marked at 25.4 mm (1 in) intervals to allow pullout to be measured from the video. The corresponding retractor force time history was measured using a load cell on the shoulder belt. The retractor pullout and force were then time synchronized to generate the force-pullout curves for the retractor (Figure 3.5), which were used to define the retractor response in the FE model of the Rigid Bench.

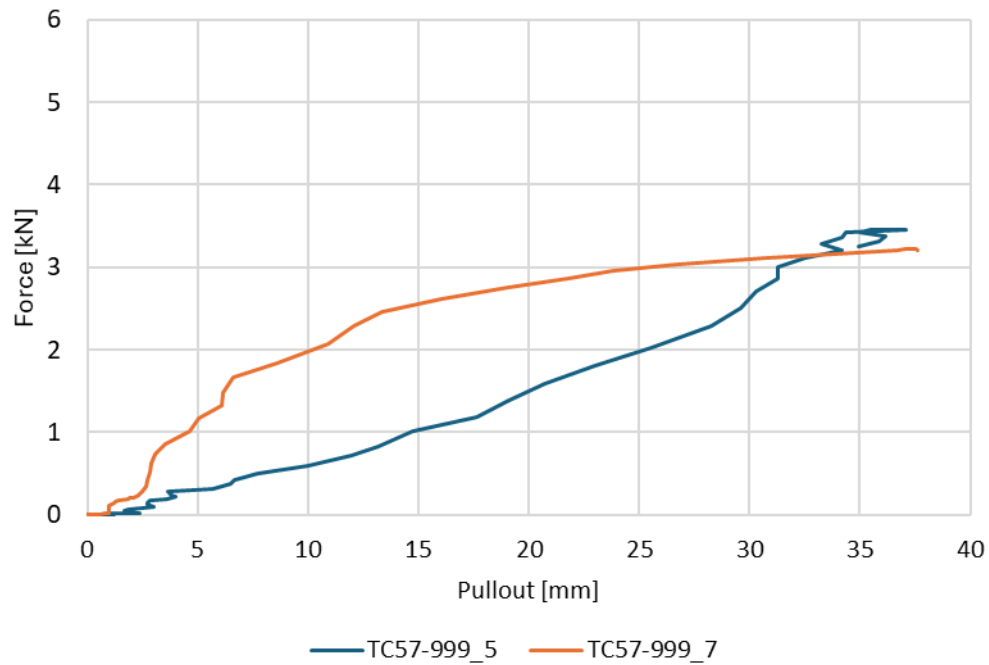


Figure 3.5: Retractor force-pullout curve extracted from video data of sled tests in a different study.

The Rigid Bench was mounted on an acceleration sled at the TC/PMG facility in Blainville, Quebec, with the base of the bench rigidly attached to the sled bed (Figure 3.6A). The acceleration sled was programmed to a 23 g peak acceleration crash pulse (Figure 3.6B), taken from measurements at the center of gravity (CG) of a vehicle (2015 Chevrolet Equinox) in a 40% frontal offset vehicle-vehicle crash test (Tang et al., 2021).

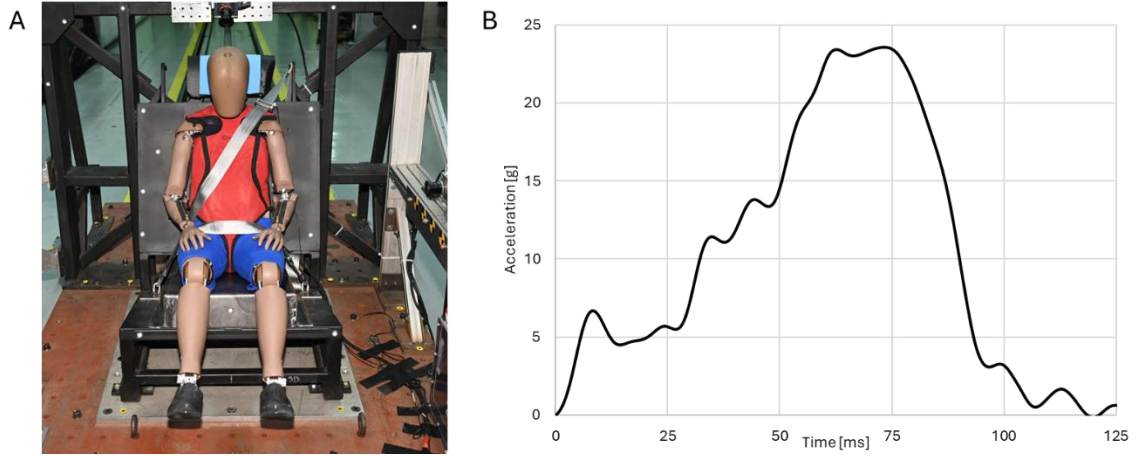


Figure 3.6: (A) Physical Rigid Bench test setup with THOR-5F on the acceleration sled and (B) sled acceleration pulse used in physical tests.

3.2.2 Physical THOR-5F ATD Testing

Physical testing was conducted with the Rigid Bench at the TC/PMG facility (Motor Vehicle Test Center, Blainville, QC), in support of this project, and TC/PMG provided the test results. Three sled tests were conducted with the THOR-5F (Standard Build Level B, Humanetics, Farmington Hills, MI, USA) positioned on the Rigid Bench (Figure 3.6A). For each test, the ATD was positioned and measured using a FAROArm 3D metrology system, with measurements recorded with a resolution of 1 mm. After the THOR-5F was positioned in the Rigid Bench, the seat belts were routed, and the ATD with the seat belts was scanned using the MetraSCAN 3D scanner described in Section 3.2.1.

From the physical tests, TC/PMG provided all available data from the THOR-5F ATD components loaded during the sled tests, consisting of 77 channels of kinematic and kinetic time-series data from the head, neck, thorax, and pelvis instrumentation. TC/PMG provided 8 channels of sled acceleration time histories, and 2 channels from the seat belt, corresponding to the lap and shoulder belt forces. All signals were sampled at 20 kHz and filtered according to the SAE J211 standard.

3.3 Integration of THOR-5F FE Model with the Rigid Bench and Validation with Experimental Tests

3.3.1 Rigid Bench FE Model

A model of the Rigid Bench was created from the CAD drawings and 3D scan (Figure 3.3) provided by TC/PMG and using a rigid material with steel-defined material properties (Table 3.2) for contact purposes. The FE model included a seat pan, seatback, retractor, shoulder and lap belt anchors, and the raised platform to offset the seated THOR-5F from the lap belt anchors (Figure 3.7). The Rigid Bench seat pan, seat back and sled base all used quadrilateral shell elements.

Table 3.2: Material properties of seatback, seat pan, sled floor, and seat belt anchors of the Rigid Bench FE model.

Material type	Steel
Density [kg/m³]	7850
Young's modulus [GPa]	210
Poisson's ratio	0.3

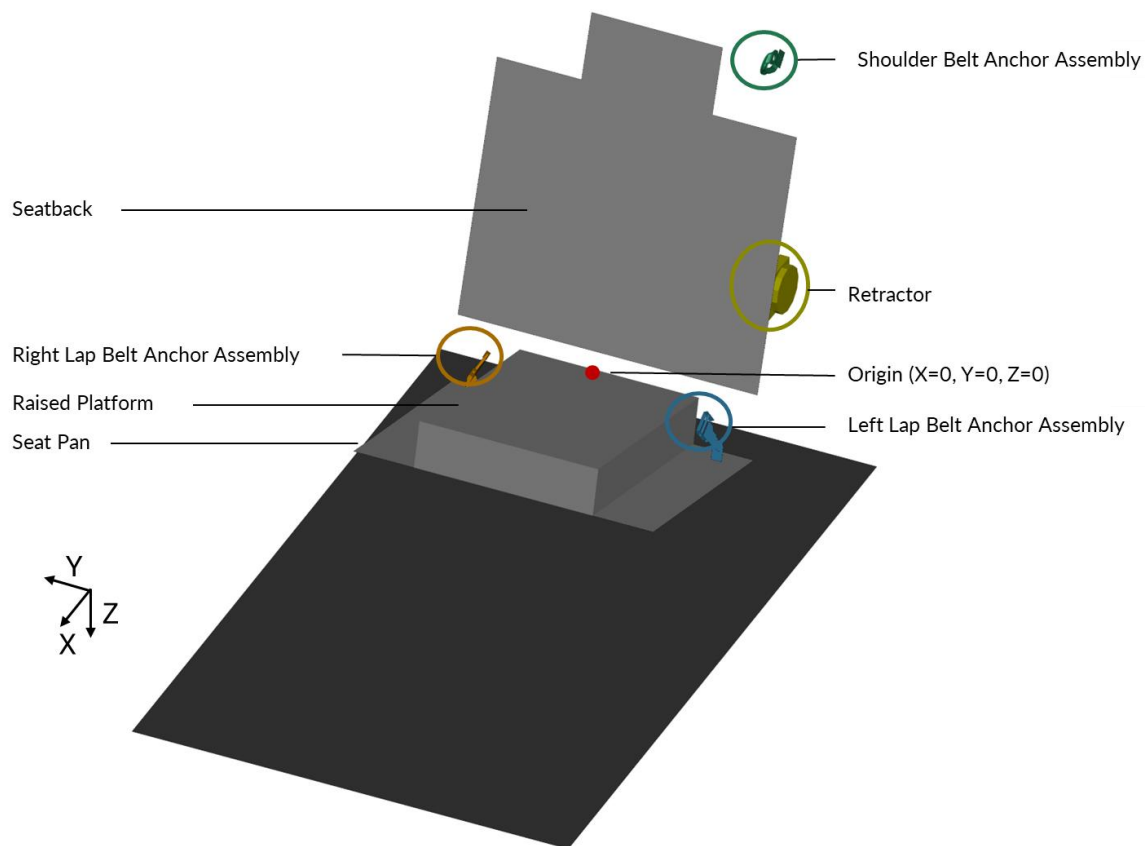


Figure 3.7: FE model of the Rigid Bench used in sled testing with the THOR-5F. The setup consisted of shoulder and lap belt anchor assemblies, a raised platform, seat pan, retractor, and seatback.

In the FE model, each lap belt anchor assembly consisted of the lap belt anchor attachment, a spherical joint and a seat belt webbing guide (Figure 3.8A). The lap belt anchor was defined as nodes on the lap belt anchor attachments, about which a spherical joint beam was created. The spherical joint connected the lap belt anchor to the seat belt webbing guide. The lap belt anchor attachment was fully constrained to prevent translation or rotation, while the spherical joint beam allowed rotation in the 3 degrees of freedom (DOF), enabling the seat belt webbing guide to rotate about the lap belt anchor, consistent with the physical Rigid Bench. The right lap belt webbing guide also included a slipping to connect the lap belt to the shoulder belt.

The shoulder belt anchor assembly consisted of a shoulder belt anchor attachment, a revolute joint and a seat belt webbing guide (D-ring). The shoulder belt anchor was defined as a node on the anchor attachment, about which a revolute joint was created. The revolute joint connected the shoulder belt anchor to the D-ring. The shoulder belt anchor was fully constrained to the sled base to prevent translation or rotation, and the revolute joint allowed 1 DOF rotation about the Y-axis, enabling the D-ring to rotate about the shoulder belt anchor, consistent with the physical Rigid Bench. A slip ring was defined at the D-ring to allow the belt to move through the D-ring to the retractor.

The shoulder and lap belt anchor assemblies were modelled using quadrilateral shell elements and were positioned to match the anchor locations of the physical Rigid Bench (Table 3.1).

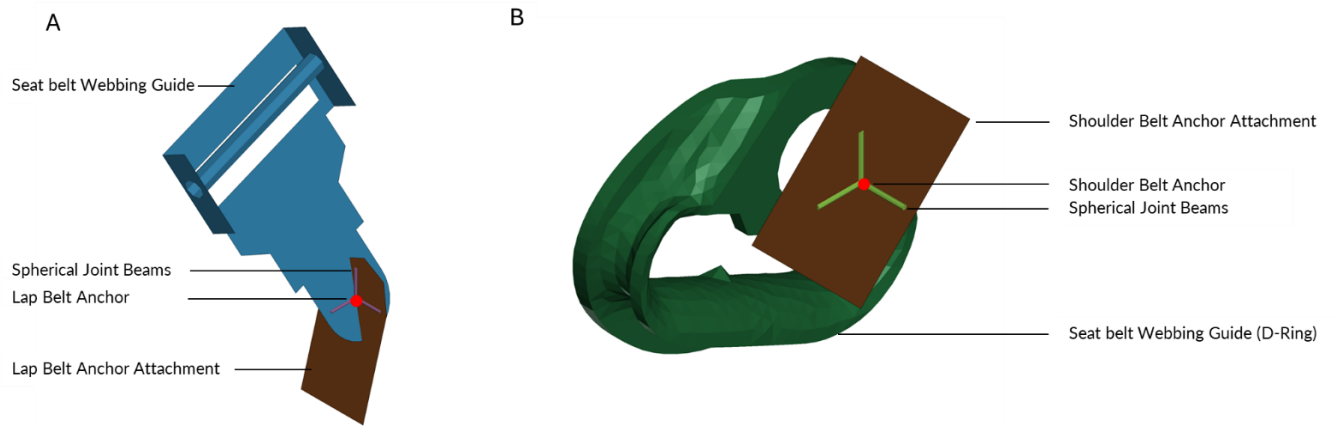


Figure 3.8: (A) Lap belt anchor assembly used in the Rigid Bench model, with spherical joint, seat belt webbing guide and anchor attachment. (B) Shoulder belt anchor assembly used in the Rigid Bench model, with D-ring, revolute joint and anchor attachment. Anchor location indicated by red circle.

3.3.2 Positioning of THOR-5F FE Model on the FE Rigid Bench

The THOR-5F FE model (ATD Models GmbH) positioning was completed through a pre-simulation recommended by the developer (ATD Models GmbH), which applied a “marionette” (Asadi & Karlsson, 2023) positioning approach in which 86 4 kN-tensioned cables pulled the model to the defined target position corresponding to the physical test position (Figure 3.9). The target position was the desired position of the THOR-5F FE at the end of the simulation, defined by nodal coordinates corresponding to the physical test position.

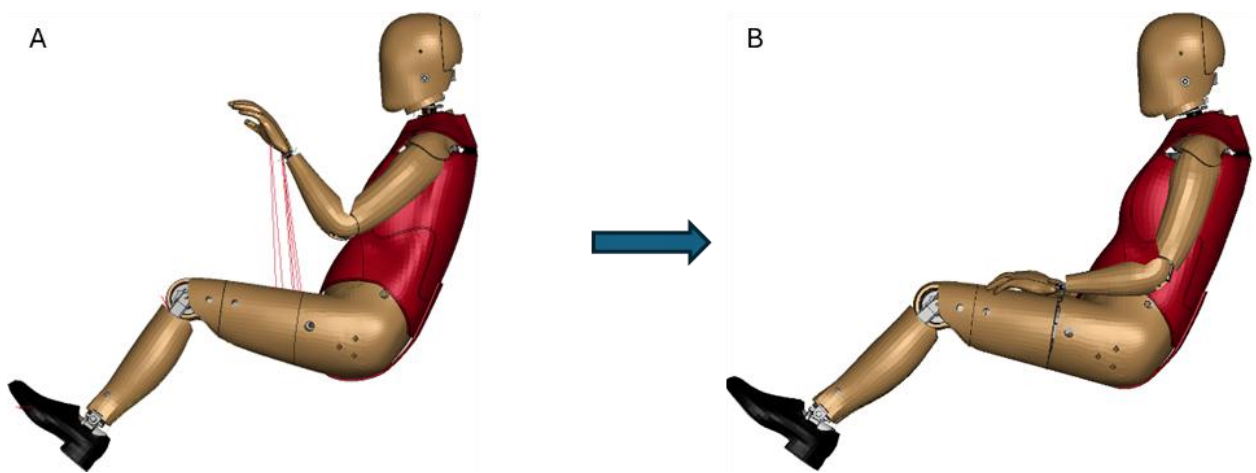


Figure 3.9: Overview of the positioning pre-simulation, where tensioned cables moved the THOR-5F FE model from a (A) start position to a (B) target position.

The marionette method required a predefined target position corresponding to the physical test position. The following procedure was used to define the target position.

1. A wireframe was created from the FAROArm measurements of 18 predefined landmarks from the average physical test position (Figure 3.10A).
2. The THOR-5F FE was overlaid on the wireframe using a commercial preprocessor (LS-PrePost v4.10) and translated and rotated using the preprocessor ATD positioning module to align the model with the wireframe at identifiable landmarks, generating an initial target position (Figure 3.10B). The target position was further refined to match the physical test by overlaying the 3D scans of the physical test positions and adjusting the THOR-5F FE model joint angles to improve alignment between the FE model and the 3D scan (Figure 3.10C).
3. After aligning with the 3D scan, the nodal coordinates of the THOR-5F FE were used to define the target position. The nodes from the target position were then used as the end of the cable elements for the marionette positioning simulation (Figure 3.10D,E).

After the marionette positioning simulation, the position of the THOR-5F FE was compared with the physical test wireframe at identifiable landmarks: head target, H-point, ankle bolt, shoulder bolt, elbow bolt, wrist bolt.

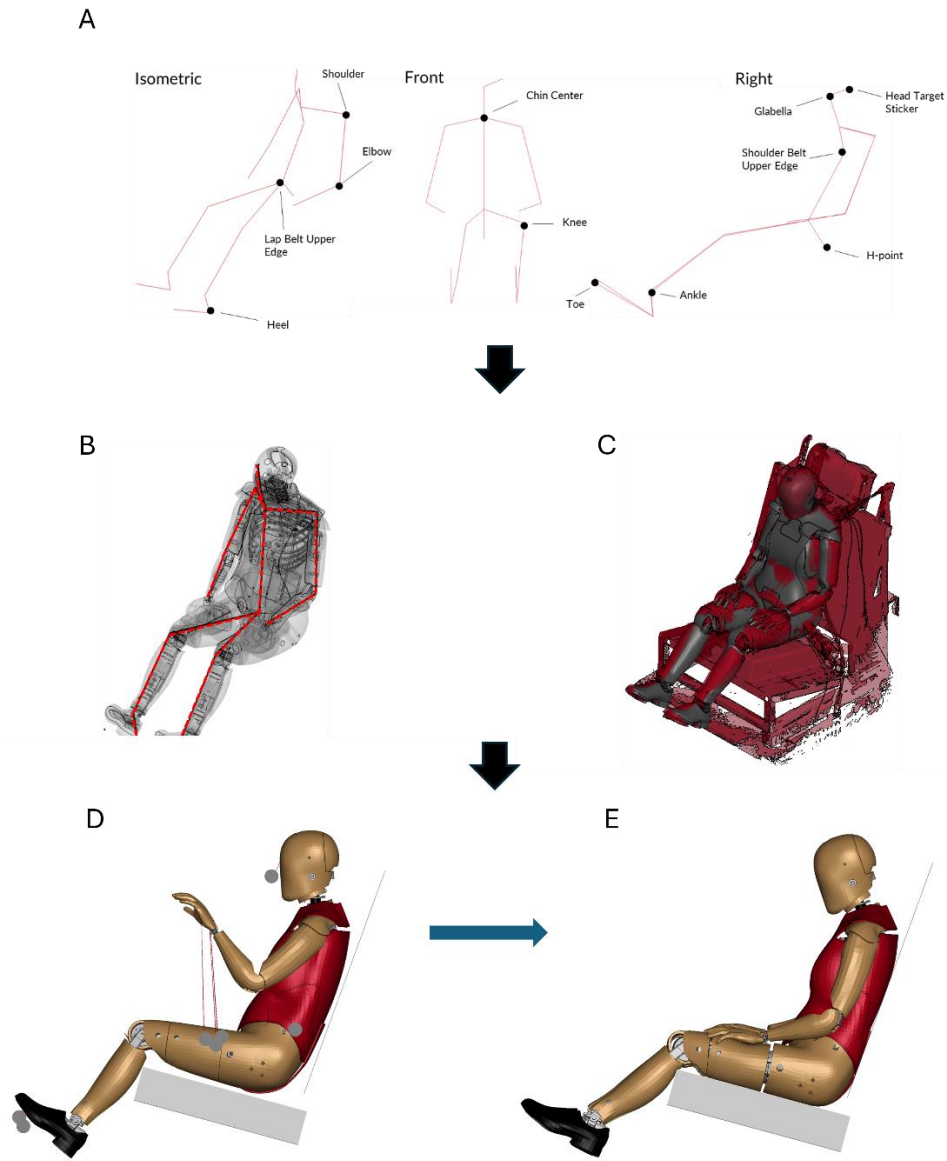


Figure 3.10: Overview of positioning process. (A) Wireframe from FAROArm measurements of average physical THOR-5F ATD position, used for positioning THOR-5F FE model. (B) Overlay of THOR-5F FE model (gray) on wireframe. (C) Overlay of the THOR-5F FE model (gray) on 3D scan of physical THOR-5F ATD (red). (D) Marionette method with THOR-5F starting configuration, with grey nodes representing target position coordinates. (E) Final THOR-5F FE position after marionette method.

3.3.3 Implementation of Seat Belts in FE Rigid Bench Model

After the ATD positioning simulation, the seat belt model was integrated and routed around the THOR-5F FE using a commercial preprocessor (LS-PrePost 4.10). Seat belt routing was performed by selecting two nodes at the belt anchor points (red nodes in Figure 3.11A) and an intermediate node on the THOR-5F FE jacket, close to the lateral midline of the torso (orange nodes in Figure 3.11B,C). The selected intermediate node was located at the center of the upper chest for the shoulder belt (Figure 3.11B), and the lateral center of the upper pelvis for the lap belt (Figure 3.11C). The “Belt Fit” command was used to generate a belt path between the anchor nodes that passed through the intermediate guide node along the jacket surface.

The seat belt geometry was subsequently shortened using the “Stretch” command, which removed excess seat belt length and improved conformity to the jacket surface. After each routing operation, the seat belts were manually inspected for any warped elements or interference with the THOR-5F jacket. If interference or warped elements were observed, the node selection was modified (through either moving the intermediate node or adding additional nodes on the jacket), and the belt fitting process was repeated until a smooth, non-interfering seat belt configuration was achieved.

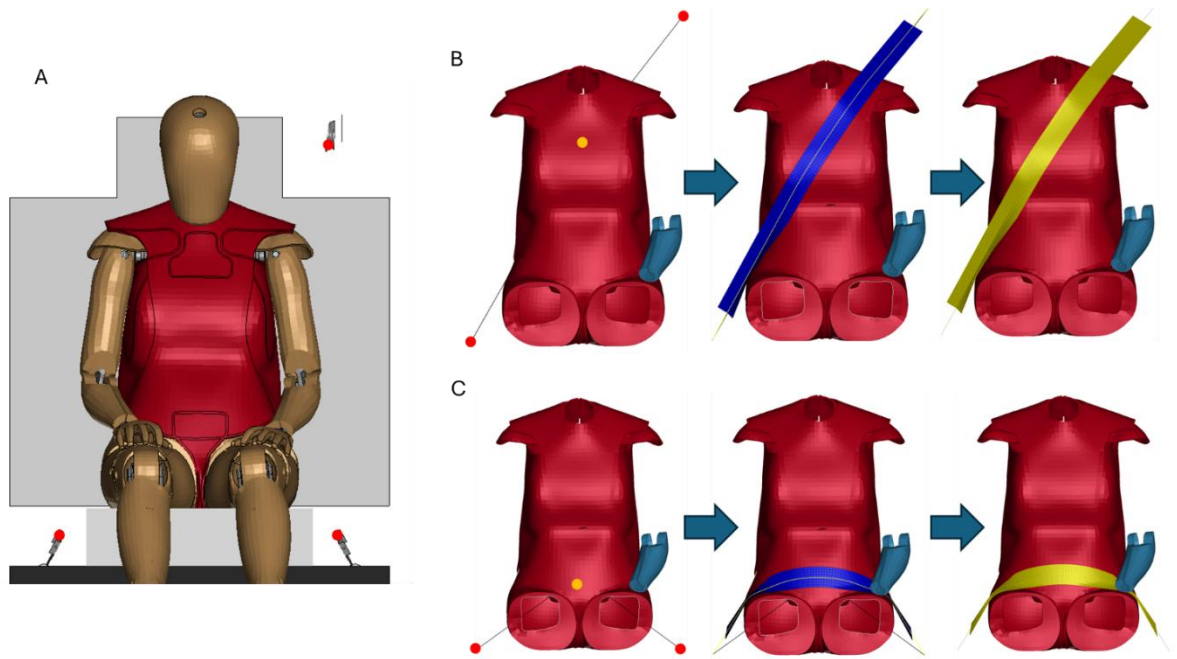


Figure 3.11: Seat belt routing process using THOR-5F FE jacket model in LS-PrePost V4.10.
(A) Nodes, corresponding to the anchor locations (red circles), were created for the lap and shoulder belt ends. Orange intermediate nodes were selected along the jacket surface to define the belt path. The nodes were then used to create (B) shoulder belts and (C) lap belts.

The belt system integrated into the Rigid Bench model (Figure 3.12) used the material properties from the ENOP project seat belt model (Lebarbé et al., 2023), as it was a contemporary FE seat belt model with current belt properties. While the ENOP seat belt model consisted of 2D elements, the model used in this study used both 1D seat belt elements and 1.2 mm-thick 2D seat belt elements to simplify contact (Table 3.3). The belt width was modified to match TC/PMG measurements (50 mm), and the retractor pullout force was updated to a linear curve from 0 to the intersection of the two measured retractor curves (Figure 3.13). The seat belt FE model force-strain curve was implemented from the ENOP belt model (Figure 3.13B). The 2D seat belt elements were connected to the anchors and sliprings using 1D seat belt elements. During a seat belt slack removal process, a pre-crash force was applied at the retractor to remove slack and match the initial belt tensions measured prior to physical testing.

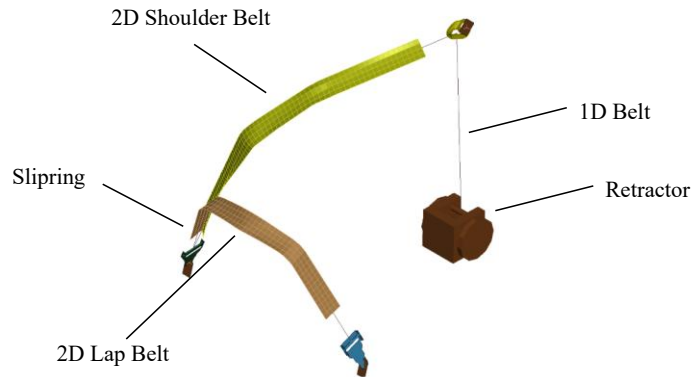


Figure 3.12: Belt model implemented in the Rigid Bench model. The right lap belt anchor contains a slipring to connect the shoulder and lap belts.

Table 3.3: Geometry and material card of Rigid Bench FE model seat belt.

Belt element type	2D seat belt, 1D seat belt
Belt width [mm]	50
Elements across width	8
Belt thickness [mm]	1.2
LS-DYNA material card	*MAT SEATBELT

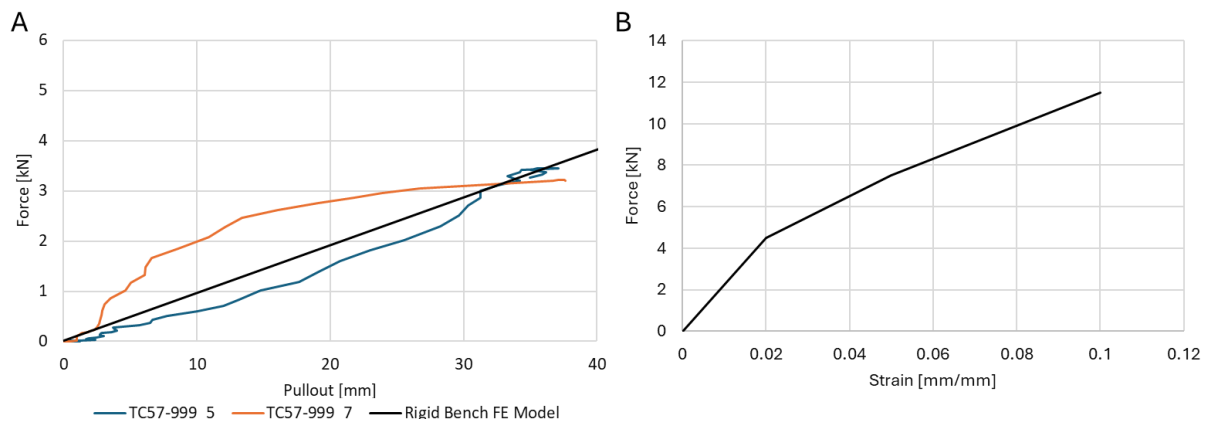


Figure 3.13: (A) Retractor force-pullout test data from TC/PMG (orange and blue) and retractor-pullout curve implemented in Rigid Bench FE model. (B) Force-strain curve of belts used in THOR-5F FE model.

3.3.4 Validation of THOR-5F in FE Rigid Bench

Following positioning the THOR-5F on the Rigid Bench model and implementing the seat belt model, the crash acceleration pulse was applied to the model. The simulation included 100 ms of seat belt slack removal prior to the onset of the crash pulse. The contact friction between the THOR-5F FE and the Rigid Bench was defined with a coefficient of 0.28, and the contact friction between the THOR-5F FE and the belt was set to 0.3, with both values being taken from the ENOP model (Lebarbé et al., 2023). Three sled acceleration pulses (Figure 3.14), measured from the Rigid Bench sled tests, were provided by TC/PMG. The average of the three crash pulses was calculated using ARCGen (Hartlen & Cronin, 2022). For the simulations, the first 125 ms of the crash acceleration pulse was applied following the 100 ms slack removal, resulting in a total simulation time of 225 ms. The simulation was terminated at 225 ms, as this duration captured the primary THOR-5F FE interaction with the restraint, including seat belt loading and forward occupant motion. Beyond 225 ms, the crash pulse magnitude was close to zero, and the occupant response was primarily in the rebound phase. Limiting the simulation duration to 225 ms captured the occupant response of interest while also reducing computational time.

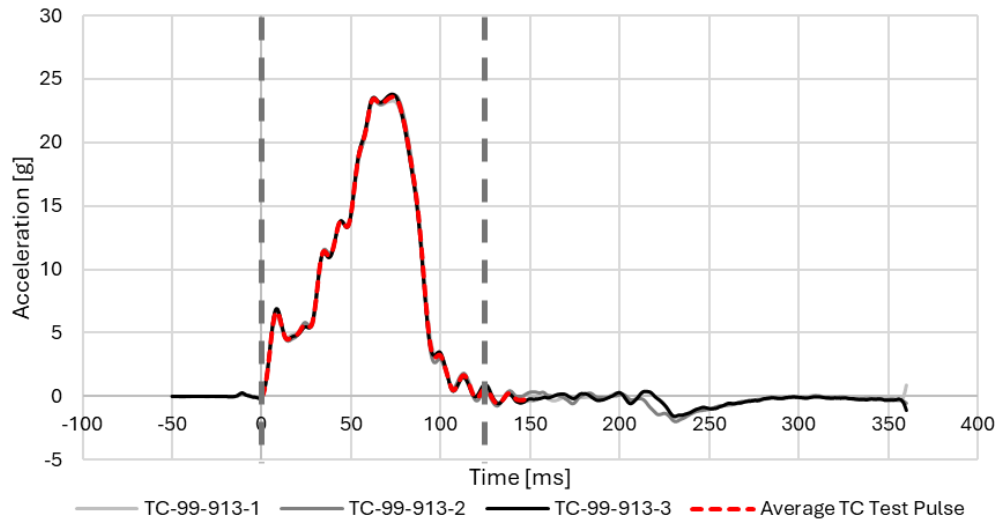


Figure 3.14: Three sled acceleration pulses from physical tests, with the average pulse shown in red.

The THOR-5F in the Rigid Bench FE model was solved using a commercial explicit finite element solver (LS-DYNA R12.2.1, MPP, single-precision) on a parallel computing platform (32 Cores, Intel 6972P processor at 2.4 GHz, 348MB cache). The following channels were used to compare the FE model with the physical test: head resultant acceleration, 1D chest sensor deflections, pelvis resultant acceleration, iliac loads, and belt loads (Figure 3.15). These channels were selected based on their availability across the physical tests, as well as their relevance to rear seat occupant injury regions (e.g. chest deflection), occupant kinetics and kinematics, and belt-occupant interaction. The selected channels also overlapped with those reported in Richard et al. (2022), which were used for the Semi-Rigid Bench assessment described in Section 3.4.4. The simulation results were filtered using SAE J211 recommended filters (Table 3.4) and then compared with physical test results by calculating scores according to ISO 18571.

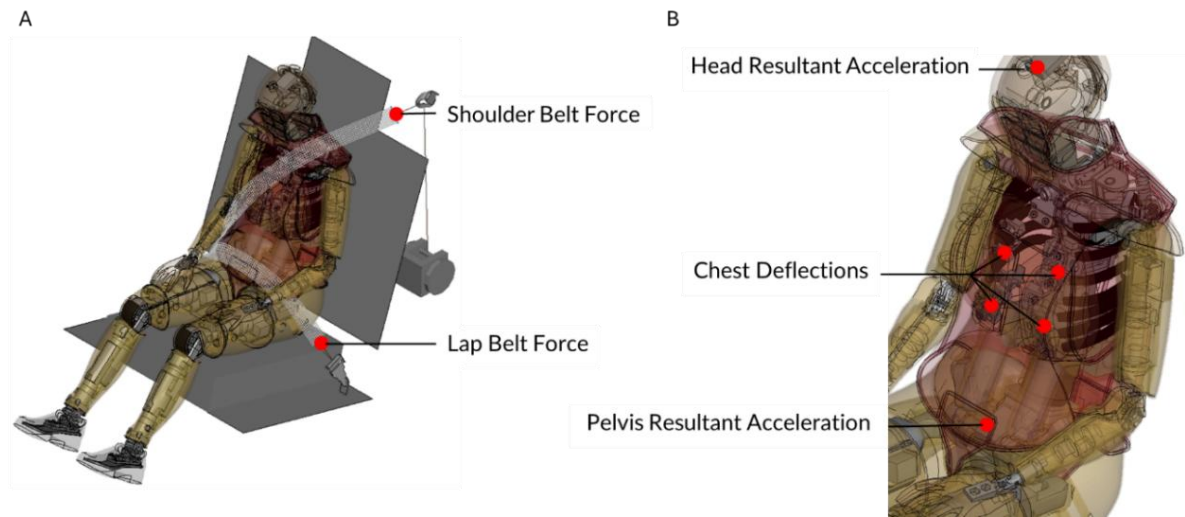


Figure 3.15: Validation metric locations of the THOR-5F in the Rigid Bench. (A) Seat belt measurement locations. (B) THOR-5F measurements used in validation.

Table 3.4: Metrics used for THOR-5F FE model validation and corresponding SAE filters.

Metric	Filter
Head resultant acceleration	SAE 1000
Chest left lower deflection	SAE 600
Chest left upper deflection	SAE 600
Chest right lower deflection	SAE 600
Chest right upper deflection	SAE 600
Left iliac force	SAE 600
Right iliac force	SAE 600
Pelvis resultant acceleration	SAE 180
Shoulder belt load	SAE 1000
Lap belt load	SAE 1000

3.4 Integration of THOR-5F FE Model with the Semi-Rigid Bench and Assessment Using Reported Experimental Test Data

The THOR-5F FE was simulated in the Semi-Rigid Bench model to investigate the potential effects of seat deformation on ATD response. The first step was to develop a Semi-Rigid Bench FE model including seat belt anchor positions, seat pan angle, and anti-submarining pan position, matching the physical sled described in the literature. Although a publicly available model of the Semi-Rigid Bench existed (Lebarbé et al., 2023), it was not used due to its higher level of complexity, which would have resulted in increased model setup time and computational cost without providing additional benefit for this study. The next step was to validate the THOR-5F FE in the Semi-Rigid Bench. The THOR-5F FE was simulated to match the first (non-submarining) configuration in the study, as this configuration provided anchor locations and seat stiffness that most accurately represented the rear seat configuration (Richard et al., 2022).

3.4.1 Semi-Rigid Bench FE Model

The Semi-Rigid Bench model was created as a simplified version of the ENOP bench described in Section 2.4.2. The simplified model was used because it required less computational time and model setup, while still capturing key features such as belt anchor geometry, seat pan deflection, and footrest interaction. The reduced model complexity enabled simple and efficient anchor location adjustment and seat belt routing for the ATD model. The Semi-Rigid Bench FE model included the main components of the physical Semi-Rigid Bench designed by Uriot et al. (2015) and used by Richard et al. (2022), including a seat pan, anti-submarining ramp, seatback and footrest (Figure 3.16).

Similar to the physical Semi-Rigid Bench, the model also included separate shoulder and lap seat belts, retractors on the shoulder and left and right lap belt anchor assemblies to remove slack, an additional right-side anchor assembly, and no slings. The anchor assemblies were identical to the ones modelled in Section 3.3.1.

For the shoulder belt, the retractor was applied at the location corresponding to the reported shoulder belt anchor position from the physical test and implemented at the upper shoulder belt webbing guide (D-ring). The seat belt-D-ring interaction point was positioned at the reported shoulder belt anchor location, and the D-ring was fixed in all DOF. The D-ring was used as the retractor, as it originally revolved about the Y-axis only. In contrast, the lap belt webbing guides were modelled with 3 DOF rotation, which allowed them to align with the occupant during the crash pulse. The retractor was at the seat belt webbing guide. The lap belt anchor locations were therefore kept at the physical lap belt anchor locations.

The Semi-Rigid Bench model consisted of 1 mm shell elements for the anchors, seatback, seat pan and footrest. Beam elements were used for the revolute joints of the anchors, and discrete elements were used for the springs of the anti-submarining pan and the seat pan. The foam used in the seatback of the physical sled was not included in the simplified model due to the lack of material property data in the literature and to reduce the potential for numerical instability in simulations.

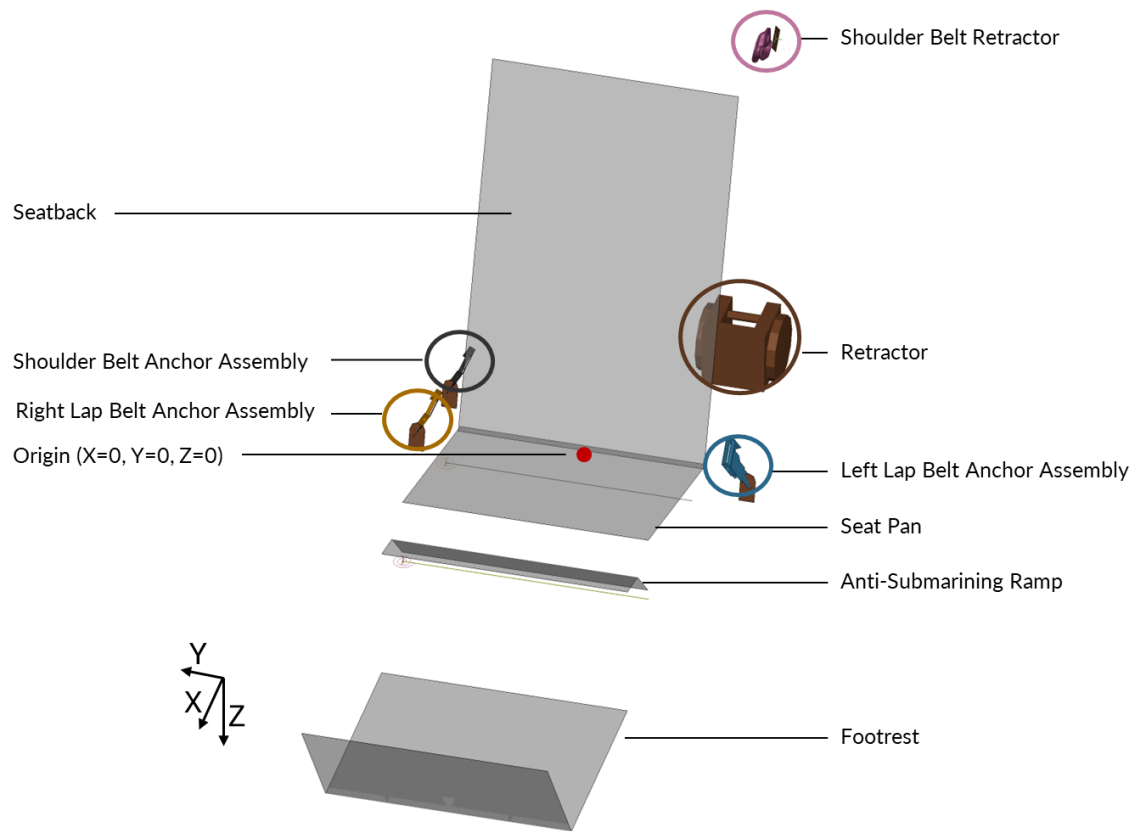


Figure 3.16: Semi-Rigid Bench FE model used in THOR-5F FE assessment simulations.

The geometry of the Semi-Rigid Bench FE model was defined using representative rear seat dimensions (Figure 3.17). The seatback height was based on the upgraded representative bench assembly detailed in a report by Wietholter et al. (Wietholter et al., 2021), and the bench width was the average lateral distance between lap belt anchors from the same report (Figure 3.17A). To match the study by Richard et al. (2022), the kinematically relevant features, such as the seatback angle, seat pan angle and position, anti-submarining ramp geometry, and footrest ramp position, were set to the reported geometry. The seat pan length was extended rearward to intersect with the seatback to simplify the geometry and eliminate gaps between the components. The seatback height in the FE model was 74 mm shorter than the physical Semi-Rigid Bench described by Richard et al. (2022). The modelled seatback provided adequate posterior support during the seat belt slack removal prior to

the crash pulse; therefore, extending the seatback beyond the region contacted by the THOR-5F FE would not affect kinematic responses.

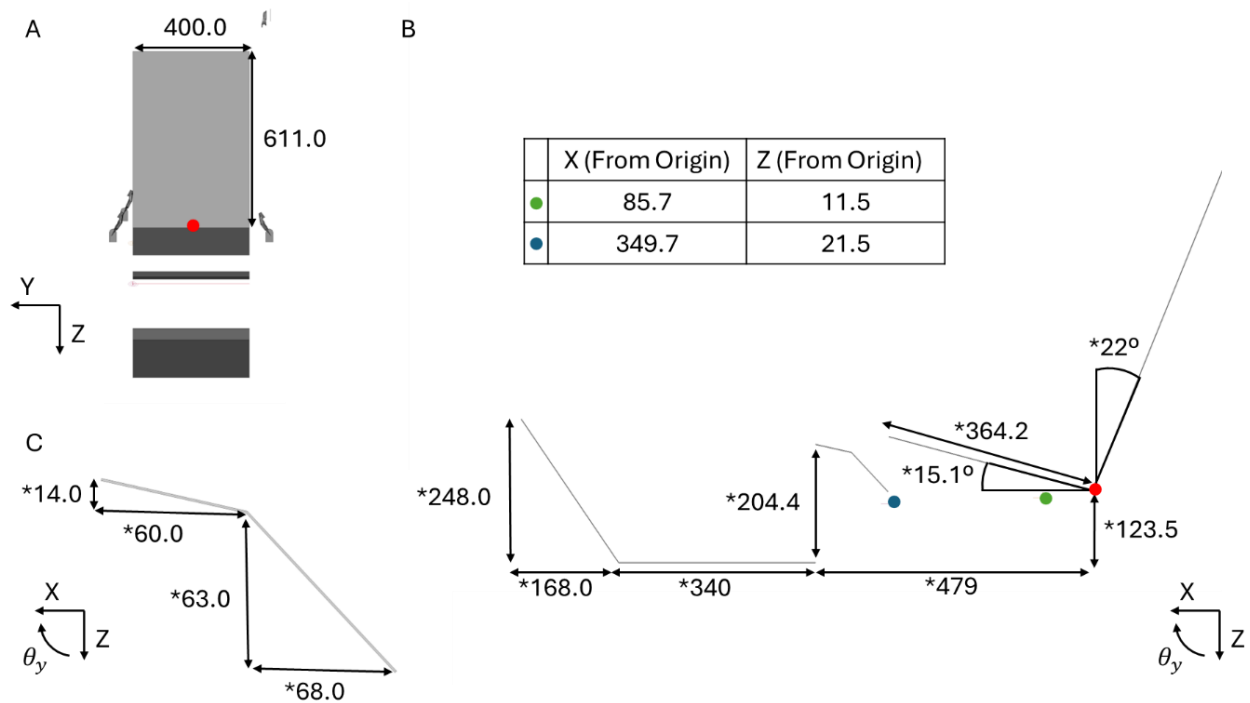


Figure 3.17: Semi-Rigid Bench model dimensions. (A) Front view rotated 20° about the Y-axis to show the bench. (B) Side view of the bench with dimensions. (C) Dimensions of the anti-submarining pan. All measurements are in mm. The red circle represents the origin (0, 0, 0), the green dot represents the seat pan spring and revolute joint, and the blue dot represents the anti-submarining pan spring and revolute joint. *Dimensions adjusted to match Richard et al. (2022).

The Semi-Rigid Bench anchors were positioned to match the non-submarining configuration in the study by Richard et al. (Table 3.5). The bench had the same material properties as the Rigid Bench (Table 3.1), with springs that enabled the anti-submarining ramp and the seat pan to deform (Figure

3.18). The Semi-Rigid Bench seatback did not include foam properties, as they were not available in the literature, and excluding foam reduced the risk of numerical instability when running the model.

Table 3.5: Semi-Rigid Bench belt anchor locations from origin.

Anchor	X [mm]	Y [mm]	Z [mm]
Shoulder belt retractor	-169.3	-248.0	-716.5
Shoulder belt anchor	-51.3	230.0	-35.5
Right lap belt anchor	3.7	269.0	16.5
Left lap belt anchor	3.7	-269.0	16.5

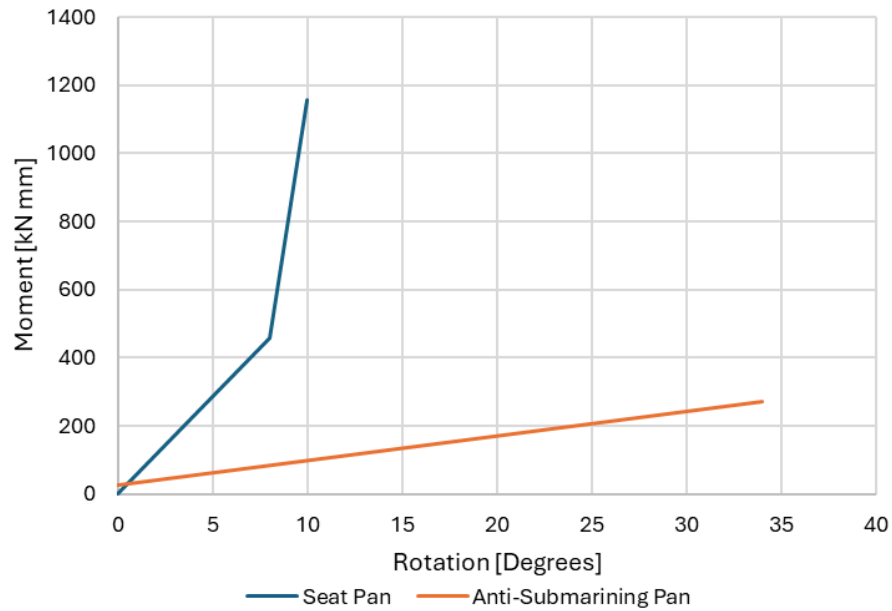


Figure 3.18: Spring stiffnesses of seat pan and anti-submarining pan in the Semi-Rigid Bench (Richard et al., 2022).

3.4.2 Positioning of THOR-5F FE in Semi-Rigid Bench FE Model

The THOR-5F FE was positioned to match the non-submarining configuration (Configuration 1) from the study conducted by Richard et al. (2022), using the available measurements from the study (Figure 3.19). The positioned model was used to generate target node positions, which were achieved in a pre-simulation described in Section 3.1.3 (Figure 3.20).

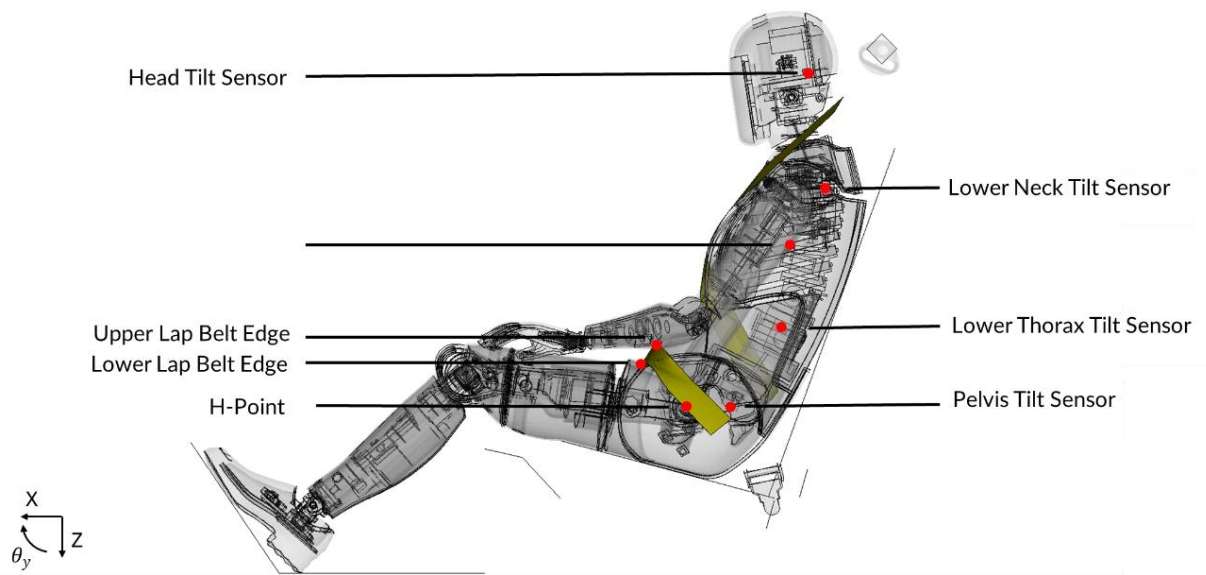


Figure 3.19: Tilt sensors and points used to position the THOR-5F FE to match with Richard et al. (2022) Configuration 1 position. The upper and lower lap belt edges were used to position the lap belt to match the Configuration 1 position.

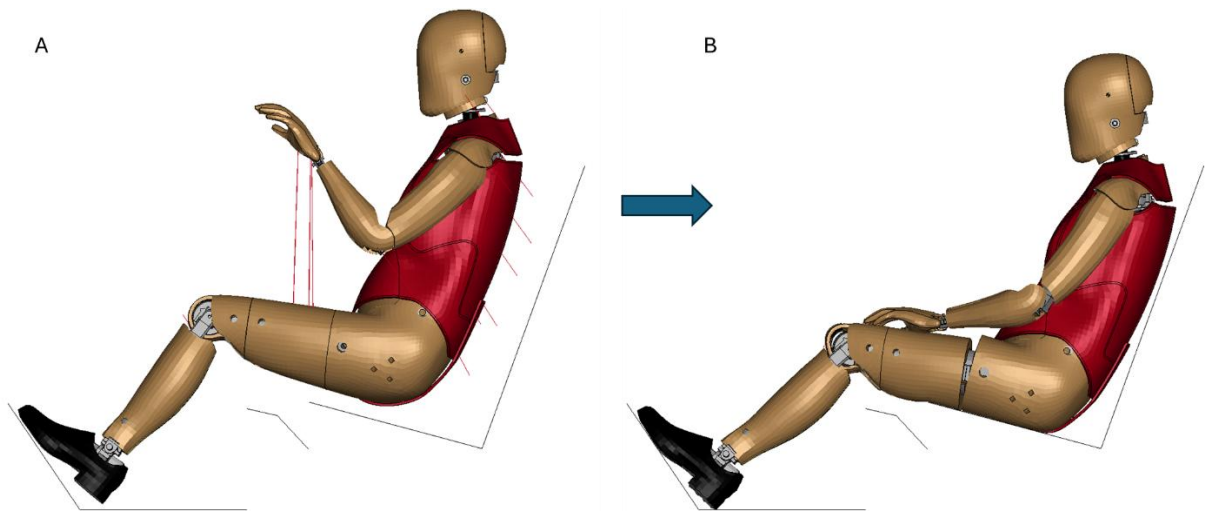


Figure 3.20: Positioning pre-simulation process for the THOR-5F in the Semi-Rigid Bench, where cables moved the THOR-5F FE model from a (A) start position to a (B) target position. The Semi-Rigid Bench seat pan and anti-submarining pan were constrained from deforming to simplify the positioning process.

After the THOR-5F FE was positioned and prior to the application of the crash pulse, the Semi-Rigid Bench seatback was translated forward in the X direction toward the THOR-5F FE to achieve minimal spacing between the seatback and the THOR-5F FE jacket without node interference (Figure 3.21). This translation was applied to replicate the posterior support from the physical Semi-Rigid Bench foam seatback following occupant settling.

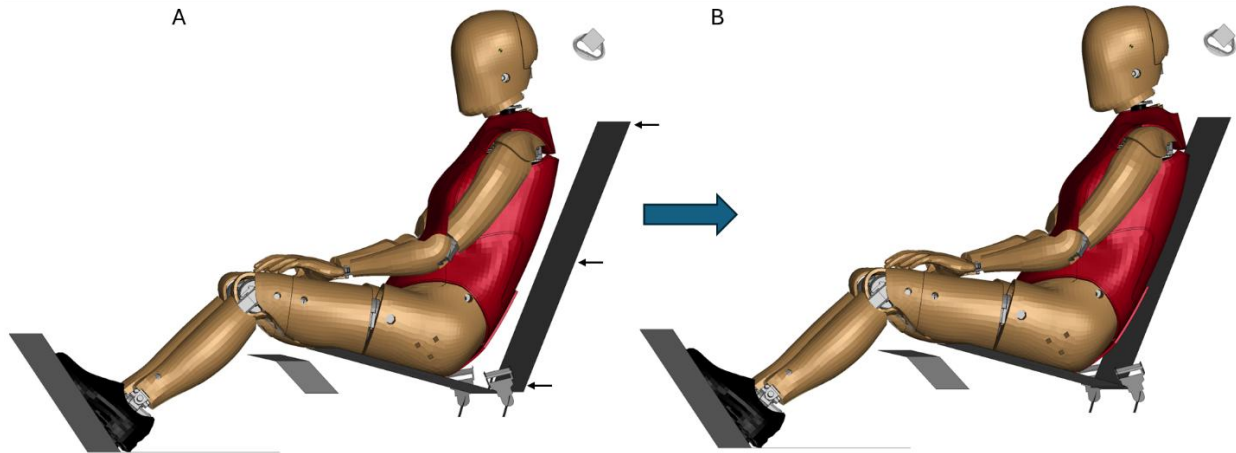


Figure 3.21: Positioning of seatback from (A) standard position of rigid plane in Richard et al. bench test to (B) behind the THOR-5F jacket with minimal spacing and without node interference.

3.4.3 Implementation of Seat Belts in the Semi-Rigid Bench FE Model

The seat belts used in the Semi-Rigid Bench model had the same properties as those in the Rigid Bench model, but with a width of 45 mm, as in Richard et al. (2022). The slope of the seat belt force-strain curve at higher strain levels was reduced to match the physical seat belt force-strain curve reported in the literature (Figure 3.22). Additionally, the seat belt FE model was modified to align with the literature, with separate lap and shoulder belts, four anchor points with anchor assemblies described in Section 3.1.1, no load limiter, and a pretensioner. The updated seat belt model used in the Semi-Rigid Bench was referred to as the 4-point 2-belt model.

A retractor was implemented in the shoulder belt to match the test and in the lap belt to remove slack from the belt while ensuring 50 N tension at each side, as confirmed by the study authors.

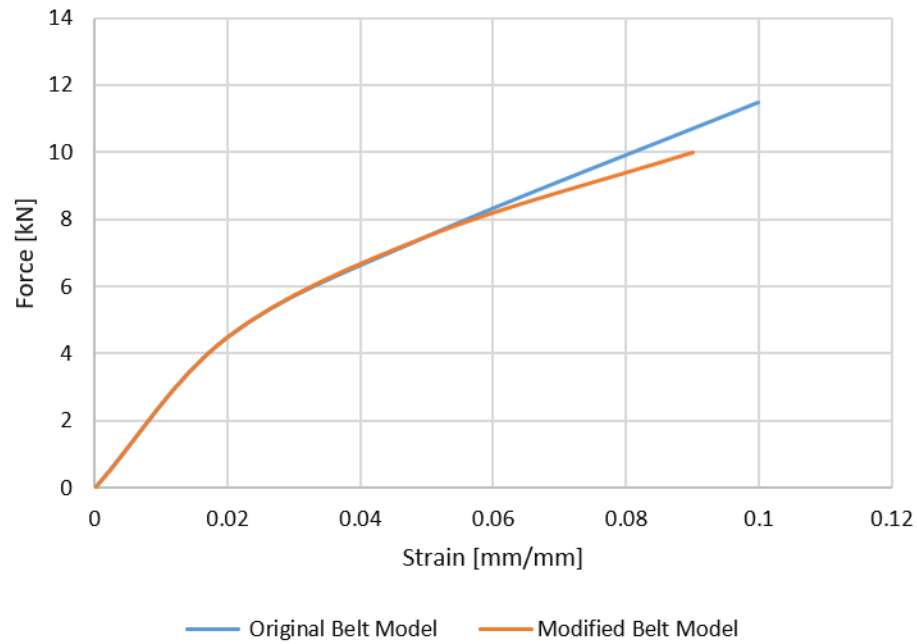


Figure 3.22: Seat belt force-strain curves for the original belt model used in the Rigid Bench model and the modified belt in the Semi-Rigid Bench model. The slope of the modified belt in the higher strain region (>0.06) was reduced to match the physical seat belt force-strain response reported by Richard et al. (2022).

In the model, the retractor was set to lock immediately after the crash pulse was initiated, preventing any pullout, which matched the description of the physical test setup. The retractor pullout to match the belt forces to the timing of the physical sled was then determined through an incremental retractor pullout study, in which the validation simulation was run with increasing retractor pullout by 5 mm. The retractor pullout with the closest lap belt, shoulder belt peak and pelvis acceleration to the physical tests was then chosen for the final assessment simulation.

3.4.4 Assessment of THOR-5F FE Model in the Semi-Rigid Bench

After positioning, the 4-point 2-belt seat belt model was integrated into the Semi-Rigid Bench model, and the crash pulse from literature (Richard et al., 2022) was applied (Figure 3.23).

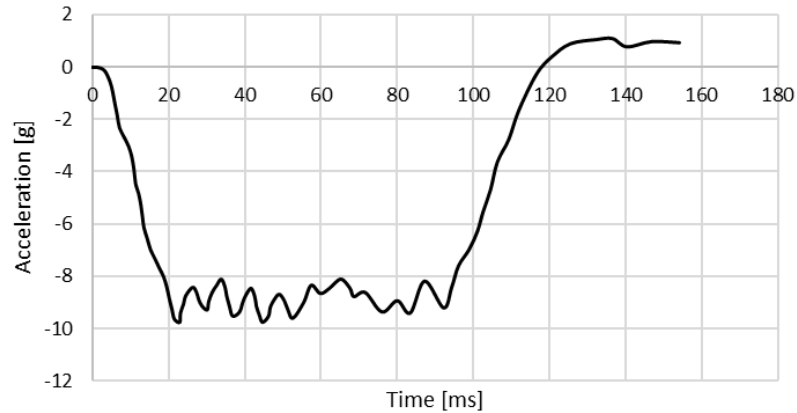


Figure 3.23: Acceleration pulse used in Configuration 1 (non-submarining) sled tests from Richard et al. (2022).

The response traces from the physical tests were digitized using WebPlotDigitizer v5.2 (Rohatgi, n.d.), a free web-based plot digitizing tool, and used to compare the simulation to the physical test data. For consistency with the Rigid Bench validation process and the availability of data from the study, the following simulation measurements were recorded and filtered using the SAE J211 filter values (Table 3.4): head resultant acceleration, 1D chest sensor deflections, pelvis resultant acceleration, iliac forces, and belt loads. As only two response curves were available for each measurement, the ISO 18571 cross-correlation scores were calculated for each test curve and averaged across both curves.

3.4.5 Implementation of 3-Point Seat Belt in Semi-Rigid Bench FE Model

After assessing the Semi-Rigid Bench model using the 4-point, 2-belt configuration, the next step was to implement the 3-point seat belt system in the Rigid Bench model (Figure 3.24). The 3-point belt was used for the current study in place of the 4-point 2-belt setup to create consistency between the

Rigid Bench and Semi-Rigid Bench models, and to represent belt configurations found in modern vehicles. The lap belt anchor locations remained the same, while the lower anchor for the shoulder belt was removed, and the shoulder belt was routed from the shoulder belt anchor to the right-side lap belt anchor.

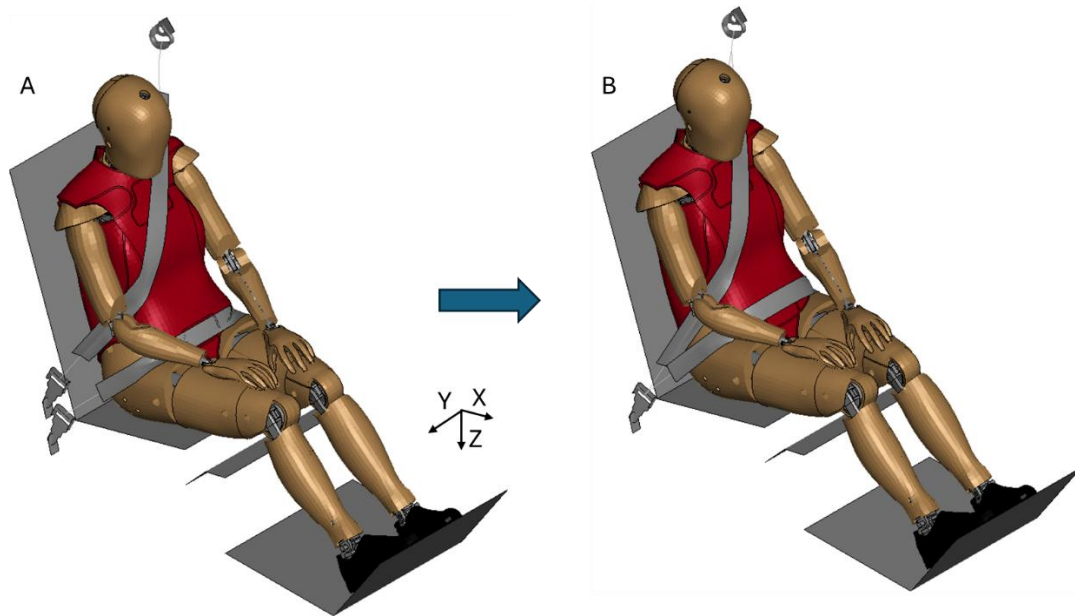


Figure 3.24: Process of changing belt configuration from the (A) 4-point 2-belt configuration to the (B) 3-point belt configuration.

3.5 Parametric Study of Belt Anchor Locations and THOR-5F FE Kinematic Responses

A Design of Experiment (DOE) approach (Montgomery, 2017) was used to test the influence of shoulder and lap belt anchor locations on ATD kinematics. A 3-factor, 3-level (3^3) full factorial experiment was designed using minimum, midpoint, and maximum values from a study that measured the X, Y, Z distances (SAE J211 coordinates) from the shoulder and lap belt anchors to a fixed reference point for 24 vehicles, including multi-purpose vehicles, passenger cars, and trucks (Aram & Rockwell, 2012). The midpoint value was used rather than the mean value to ensure even spacing of factor levels, as the midpoint corresponded to the geometric center of the range. These

anchor value combinations were simulated to examine the responses at the location combination, which were then used to identify the main and interactive effects (Montgomery, 2017). Main effects refer to how shifting a factor (e.g., moving the seat belt anchor in X) influences the overall response. In contrast, interaction effects consider how changing another factor impacts the effect of a variable on the response (e.g., the influence of X on chest deflection at high Y versus low Y).

3.5.1 Shoulder Belt Anchor Position Investigation

Twenty-seven simulations were conducted with the shoulder belt anchor positioned at parametrically defined combinations of maximum, minimum, and midpoint locations in X (fore-aft), Y (lateral), and Z (vertical) locations, based on the reported data from Aram & Rockwell (2012). The lap belt anchor locations were held constant at the midpoint of the reported maximum and minimum lap belt anchor values for all simulations. The full-factorial design was used to quantify the sensitivity of the THOR-5F FE kinematic response to shoulder belt anchor placement effects on the Rigid and Semi-Rigid Benches (Figure 3.25, Tables 3.6–3.7).

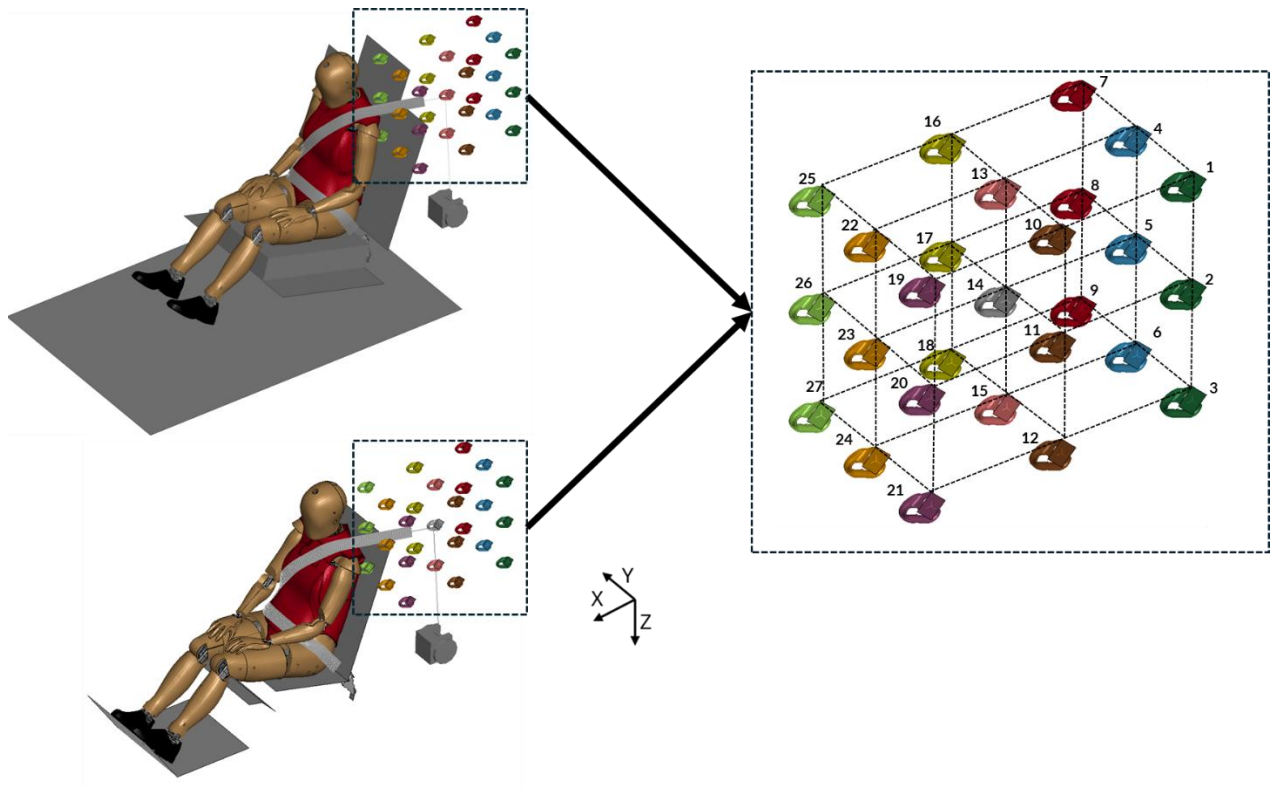


Figure 3.25: Shoulder belt anchor locations simulated with Rigid Bench (top) and Semi-Rigid Bench (bottom).

Table 3.6: Shoulder belt anchor locations (from origin) for minimum, maximum and midpoint configurations.

Configuration	X [mm]	Y [mm]	Z [mm]
MIN	-579.5	-457	-863.5
MID	-348.3	-318	-672
MAX	-117	-179	-480.5

Table 3.7: Shoulder belt anchor location effect simulations corresponding to Figure 3.25.

Simulation	X	Y	Z		X	Y	Z		X	Y	Z
1	MIN	MIN	MIN	10	MID	MIN	MIN	19	MAX	MIN	MIN
2	MIN	MIN	MID	11	MID	MIN	MID	20	MAX	MIN	MID
3	MIN	MIN	MAX	12	MID	MIN	MAX	21	MAX	MIN	MAX
4	MIN	MID	MIN	13	MID	MID	MIN	22	MAX	MID	MIN
5	MIN	MID	MID	14	MID	MID	MID	23	MAX	MID	MID
6	MIN	MID	MAX	15	MID	MID	MAX	24	MAX	MID	MAX
7	MIN	MAX	MIN	16	MID	MAX	MIN	25	MAX	MAX	MIN
8	MIN	MAX	MID	17	MID	MAX	MID	26	MAX	MAX	MID
9	MIN	MAX	MAX	18	MID	MAX	MAX	27	MAX	MAX	MAX

After setting each shoulder belt anchor location, the seat belts were routed around THOR-5F FE model using LS-PrePost V4.10 and simulated with the 23g crash pulse. In some configurations, the 2D seat belt segment would interact with the D-ring during the slack removal phase. The interaction of the 2D belt with the D-ring in these configurations was prevented by slightly increasing the adjacent 1D belt length. The kinematic measurements investigated included the peak value of deflection-time response from each of the four 1D chest sensors (upper right, upper left, lower right, lower left), as well as the overall peak chest deflection, defined as the maximum value of the four sensors. The peak resultant pelvis acceleration was also investigated. Peak chest deflection was examined because thoracic deformation is a primary injury assessment metric in frontal crash tests. In NCAP testing using the THOR-50M, the peak chest deflection of the four sensors is recorded (Brumbelow et al., 2022). Additionally, the thorax is a major injury region in rear seat occupants (Beck et al., 2016). Pelvis resultant acceleration was used as a scalar measure of lower-body kinematics and an indicator of seat belt engagement timing.

3.5.2 Lap Belt Anchor Position Investigation

The lap belt anchor positions were also investigated for both the Rigid Bench and the Semi-Rigid Bench. The lap belt anchors were modified in the X, Y, and Z directions at the maximum, minimum, and midpoint lap belt anchor locations (Figure 3.26, Tables 3.8–3.9). The lateral lap belt locations were based on the maximum and minimum lateral distances between the anchors, and the anchors were set up symmetrically to maintain simplicity. Since lap belt anchor Positions 10 and 19 were

inside the THOR-5F pelvis, they were excluded from the analysis. The lap belt anchor positions were labelled as maximum, midpoint and minimum based on the inboard anchor position and the SAE coordinate system. The inboard lap belt positioning enabled a straightforward and intuitive understanding of the maximum distance between the lap belt anchors. For each simulation, the shoulder belt anchor position was held constant at the midpoint of the reported max and min values from Aram and Rockwell (2012). Each lap belt anchor configuration required re-routing of the shoulder and the lap belt, as the inboard lap belt anchor connected the two belts, influencing the overall belt path.

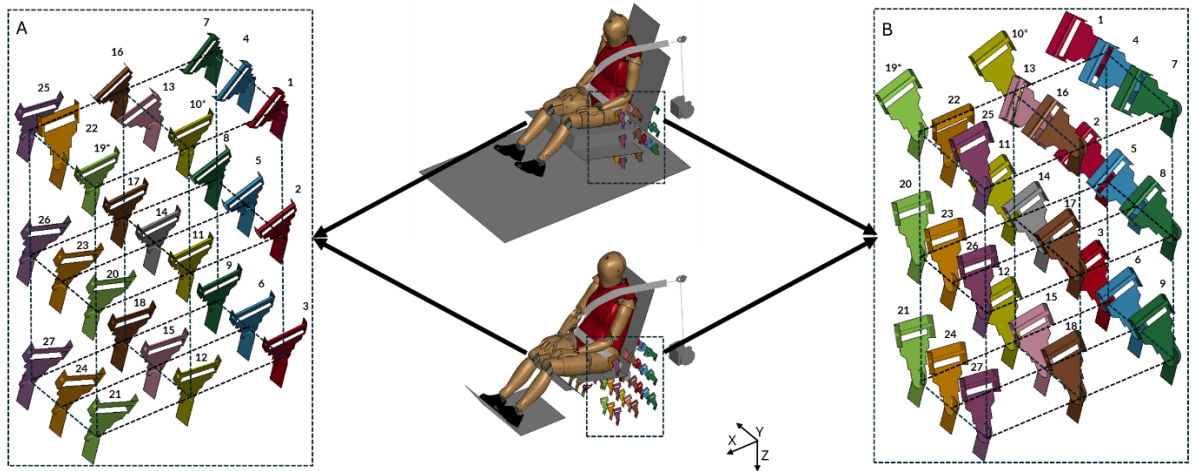


Figure 3.26: THOR-5F in Rigid Bench (top) and Semi-Rigid Bench (bottom) with (A) inboard lap anchor positions simulated with each configuration (B) outboard lap belt anchor simulated. *Anchors 10 and 19 were unable to be simulated due to pelvis interference.

Table 3.8: Lap belt anchor location effect simulations corresponding to Figure 3.26.

Simulation	X	Y	Z		X	Y	Z		X	Y	Z
1	MIN	MIN	MIN	10	MID	MIN	MIN	19	MAX	MIN	MIN
2	MIN	MIN	MID	11	MID	MIN	MID	20	MAX	MIN	MID
3	MIN	MIN	MAX	12	MID	MIN	MAX	21	MAX	MIN	MAX
4	MIN	MID	MIN	13	MID	MID	MIN	22	MAX	MID	MIN
5	MIN	MID	MID	14	MID	MID	MID	23	MAX	MID	MID
6	MIN	MID	MAX	15	MID	MID	MAX	24	MAX	MID	MAX
7	MIN	MAX	MIN	16	MID	MAX	MIN	25	MAX	MAX	MIN
8	MIN	MAX	MID	17	MID	MAX	MID	26	MAX	MAX	MID
9	MIN	MAX	MAX	18	MID	MAX	MAX	27	MAX	MAX	MAX

Table 3.9: Lap belt anchor locations (from origin) for minimum, maximum and midpoint configurations for outboard location.

Configuration	X [mm]	Inboard Y [mm]	Outboard Y [mm]	Z [mm]
MIN	-176.6	116	-116	-116
MID	-34.1	186.8	-186.8	69.1
MAX	108.4	257.5	-257.5	256.6

3.5.3 Calculation of Main and Interaction Effects

The calculation of main and interaction effects was based on two key responses: the pelvis resultant acceleration and the peak chest deflection. Five peak chest deflection metrics were defined for each anchor location: the peak value at each of the four chest sensor locations and an overall peak chest deflection, defined as the maximum value across all sensors. The X,Y,Z coordinates of the shoulder and lap belt anchors were treated as quantitative factors. Each factor was coded numerically as -1 (minimum), 0 (midpoint), 1 (maximum), consistent with the standard three-level factorial design practice (Montgomery, 2017).

The effects were defined as the difference between the average responses at the maximum and minimum coded levels of the anchor position factors (X, Y, Z). For example, the main effect of X on chest deflection was defined using the difference between the average responses of the simulations

with $X = 1$ and $X = -1$ (Figure 3.27). The peak deflection at each simulation at $X = 1$ and $X = -1$ was recorded (Figure 3.27A). Then peak values were averaged across all simulations at $X = 1$ and $X = -1$ (Figure 3.27B), and the difference between the average at $X = 1$ and $X = -1$ was used to calculate the effect. Interaction effects were calculated in a similar manner by comparing the combined influence of factors. For example, the XZ interaction effect was calculated as the difference between the average responses when X and Z were at the same level ($X*Z = 1$) and when they were at opposite levels ($X*Z = -1$).

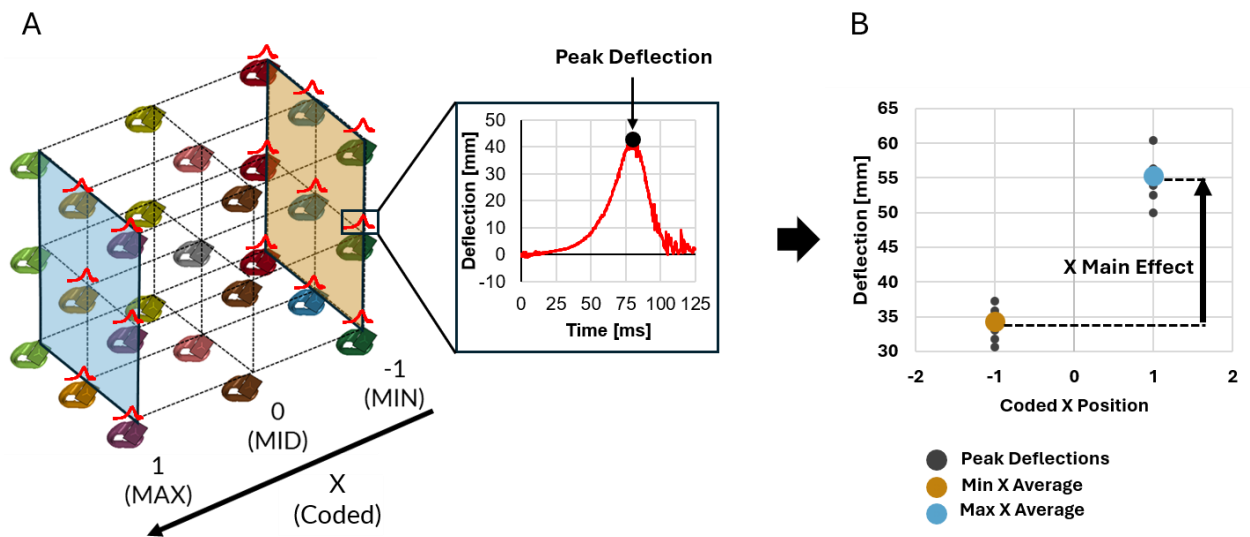


Figure 3.27: Calculation of the X main effect of shoulder belt anchor positioning on chest deflection. (A) Each anchor position at the maximum ($X = 1$) and minimum ($X = -1$) is simulated, and the peak deflection is obtained from the recorded trace. (B) The peaks are averaged for each X position, and the difference between the averages is the main effect.

Several configurations were excluded from the analysis due to uncharacteristic occupant response (e.g. submarining, seat belt slipping from the shoulder) or infeasible anchor geometries (e.g. anchor located in the pelvis region of the THOR-5F FE). To account for the excluded configurations, the remaining simulations were analyzed using multiple linear regression with interaction terms based on the least squares method. The fitted model was then used to capture the underlying trends in the

remaining simulations, estimate the expected response at the excluded configurations, and quantify the main and interaction effects. This approach enabled an unbalanced DOE analysis resulting from missing runs (Montgomery, 2017).

A first-order linear model, including main effects and interaction terms (Equation 3.1), was fit to the data for each response to ensure that the models did not overfit the smaller sample sizes. In the equation, β represented the regression coefficient of each effect or interaction, and β_0 represented the mean response at the coded center. The regression model was used to quantify the main and interaction effects.

$$\text{Response} = \beta_0 + \beta_1 X + \beta_2 Y + \beta_3 Z + \beta_4 XY + \beta_5 XZ + \beta_6 YZ + \beta_7 XYZ \quad \text{(Equation 3.1)}$$

After establishing the linear model, an analysis of variance (ANOVA) was performed to evaluate the statistical significance of the model terms. All regression and ANOVA analyses were performed using JMP Student Edition Version 19.0.1 (JMP Statistical Discovery LLC, Cary, NC, USA). The R^2 value was used to assess how each model fit the data, and the F-test p-value was used to determine the significance of effects ($p < 0.05$).

Chapter 4

Results

4.1 Physical Rigid Bench Testing at Transport Canada

The physical THOR-5F ATD was positioned in the Rigid Bench with the same target position for the three test repeats. The FAROArm 3D coordinate measurements for the belts and key anatomical landmarks of the ATD, including the H-point, glabella, and shoulder bolts, were all within 10 mm of each other between the three tests (Table 4.1). The H-point measurements had a maximum difference of 9 mm in the Z direction. The largest discrepancies in positioning were found in the left ankle and foot: the ankle front bolt showed a maximum difference of 18 mm in Y, the heel of the left foot a maximum difference of 14 mm in Y, and the toe a maximum difference of 14 mm in Z. The differences in leg positions were less relevant, as the ATD measurements focused more on the pelvis and upper body.

Table 4.1: Maximum difference between FAROArm measurements for ATD positioning landmarks for the three physical tests.

FAROArm Measurement	Maximum difference between test positions [mm]		
	X [mm]	Y [mm]	Z [mm]
Head target sticker, right side	3	4	4
H-Point, on tool, right side	5	3	9
Left knee, interior side	1	1	5
Left ankle front bolt, center of the bolt head	2	18	4
Heel of left foot	9	14	11
Left toe, foremost point	5	6	14
Right knee, exterior side	6	4	5
Right ankle front bolt, center of the bolt head	3	6	2
Heel of right foot	6	8	4
Right toe, foremost point	6	5	5
Glabella (Nose, at the top between the eyes)	6	2	3
Chin center	2	1	0
Yaw reference point, right side	6	0	4
Shoulder belt, upper edge, at y=0	3	1	6
Shoulder belt, lower edge, at y=0	2	1	6
Lap belt, upper edge, at y=0	3	0	2
Lap belt, lower edge, at y=0	3	1	4
Shoulder front bolt, left shoulder, center of the head or tail of the bolt	9	5	3
Left arm exterior elbow bolt, center of the bolt head	3	5	4
Left arm wrist interior bolt, center of the bolt head	2	5	4
Shoulder front bolt, right shoulder, center of the head or tail of the bolt	1	4	4
Right arm exterior elbow bolt, center of the bolt head	3	3	3
Right arm wrist interior bolt, center of the bolt head	1	1	3
Shoulder front bolt, right shoulder, center of the head or tail of the bolt	1	4	4
Right arm exterior elbow bolt, center of the bolt head	3	3	3
Right arm wrist interior bolt, center of the bolt head	1	1	3

The kinematic and kinetic responses of the physical ATD for the 23 g crash pulse showed low variability across all three tests for chest deflections, resultant acceleration, and shoulder belt loads (Figure 4.1B,C,D,E,F,I). For all tests, the right lower chest deflection plateaued at approximately 60 mm deflection. Slight differences in measurements were observed in the head resultant acceleration

and the lap belt force (Figure 4.1A,H,I,J). TC 99-913-3 reported a 7% higher peak lap belt force compared to the next highest test. Additionally, TC 99-913-3 showed a 5.4% higher absolute peak head resultant acceleration, with the initial peak occurring 1.2 ms later than the next highest test.

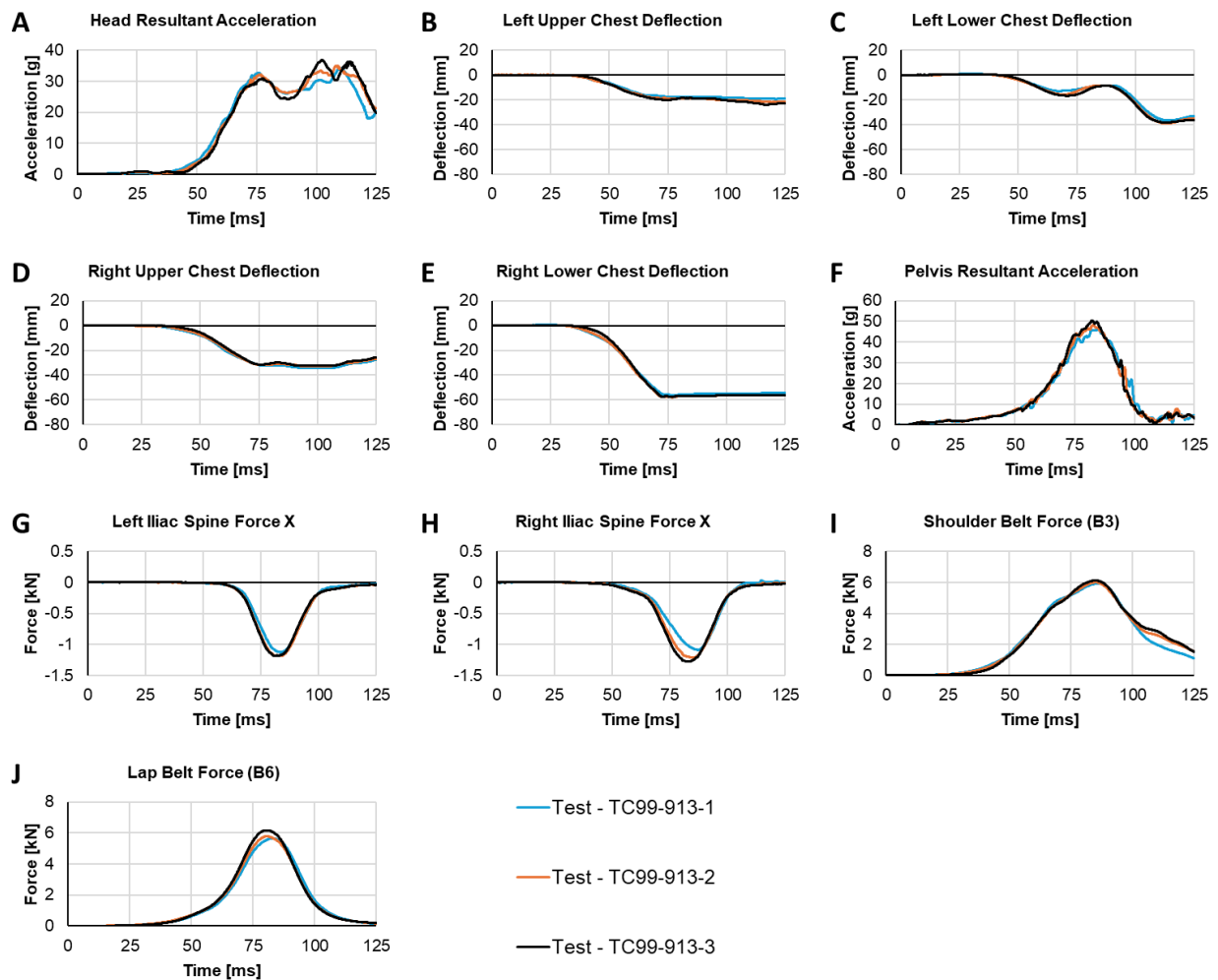


Figure 4.1: Response traces from three physical tests conducted by TC/PMG with the THOR-5F in the Rigid Bench.

4.2 Integration of THOR-5F FE Model with the Rigid Bench and Validation with Experimental Tests

4.2.1 Rigid Bench FE Model

The Rigid Bench FE model matched the physical Rigid Bench geometrically (Figure 4.2). The lengths and widths of the seat pan, seatback, and raised platform matched the physical measurements, with minor differences attributed to scan variability. The seatback and seat pan angles, seat belt anchor locations, and retractor location were defined from physical measurements and were identical to the physical Rigid Bench.

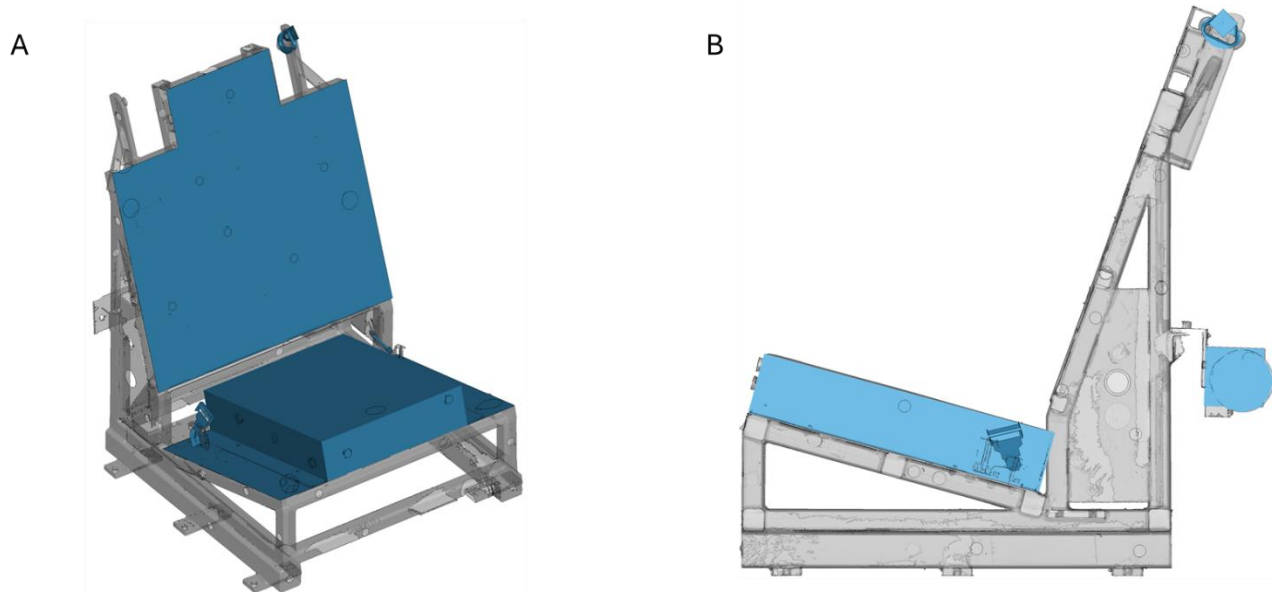


Figure 4.2: Overlay of Rigid Bench FE model (blue) on 3D scan of physical Rigid Bench (gray) with (A) isometric view and (B) side view of overlay.

4.2.2 Positioning of THOR-5F FE in Rigid Bench

The marionette method positioned the THOR-5F FE such that the H-point was within 0.1 mm of the average physical test H-point. Following positioning, the seat belt slack removal was applied prior to the crash pulse, resulting in minor positional adjustments of the THOR-5F FE relative to the target position (Table 4.2). After the seat belt slack removal and at the onset of the crash pulse, the H-point of the THOR-5F FE was 1.2 mm forward, 0.1 mm inboard and 2 mm lower than the average of all three physical test measurements. The largest discrepancy between the FE model position and the average physical test position was measured at the head target, with a -12.9 mm difference in the Z-direction. All other measured locations, including the shoulders, elbows, wrists and arms, were within 10 mm of the physical test position, indicating good overall agreement between the FE model and the physical test positions.

Table 4.2: Distance between THOR-5F FE model and wireframe location for select (distinguishable) nodes after positioning simulation.

Distance of THOR-5F FE from average of measured values after pre-sim			
Measurement	X [mm]	Y [mm]	Z [mm]
Head, target sticker, right side	1.0	-2.0	-12.9
H-Point	1.2	0.1	-2.1
Left ankle front bolt	-7.1	-1.1	-7.9
Right ankle front bolt	1.2	-1.4	-5.1
Shoulder front bolt, left shoulder	-2.6	-4.9	-5.2
Left arm exterior elbow bolt	-8.3	2.6	-4.7
Left arm wrist interior bolt	0.8	0.5	-1.1
Shoulder front bolt, right shoulder	-3.8	1.4	-2.8
Right arm exterior elbow bolt	-6.8	-1.6	-2.3
Right arm wrist interior bolt	-0.5	-1.0	-1.6

4.2.3 Implementation of Belts in Rigid Bench FE Model

The belts generated for the THOR-5F FE on the Rigid Bench exhibited a smooth geometric fit to the ATD, with no visible warping or abnormal elements, and no interference with the THOR-5F jacket. Following seat belt fitting and prior to the crash pulse, the slack in the seat belt was removed. The simulation remained numerically stable, with no error terminations or instabilities, resulting in improved belt conformity and contact with the THOR-5F FE and closer alignment with the physical test belt position. After belt slack removal, the distance between the THOR-5F FE model and the belt position in the physical ATD model at the midpoint ($Y = 0$) was within 20 mm in X and 10 mm in Z (Table 4.3). The lap belt upper and lower edges at the midpoint were farther rearward and lower than the physical test belt (Figure 4.3), while the shoulder belts were farther forward and lower. The lower edge of the FE shoulder belt had the largest difference, as it was 17 mm forward in X.

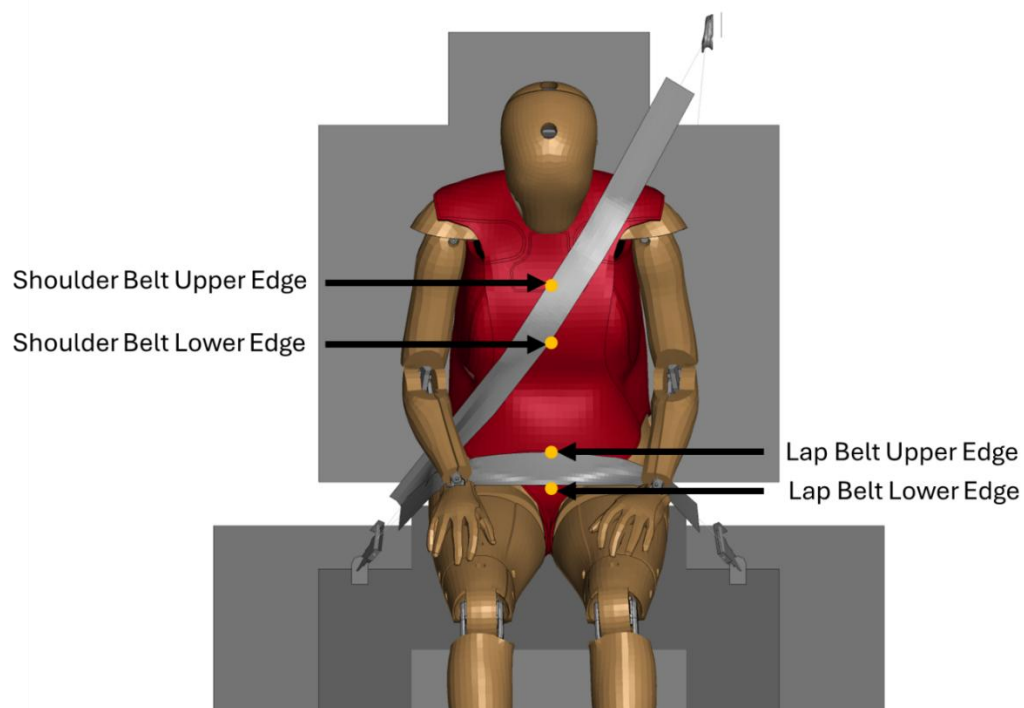


Figure 4.3: Definition of seat belt measurement points on the THOR-5F FE in the Rigid Bench for comparison to the physical test.

Table 4.3: Distance between THOR-5F FE model and average physical ATD belt position.

Distance of belt point from average of measured values			
Measurement	X [mm]	Y [mm]	Z [mm]
Shoulder belt, upper edge, at y = 0	1.7	0.0	-7.2
Shoulder belt, lower edge, at y = 0	17.0	0.0	-3.6
Lap belt, upper edge, at y = 0	-8.7	0.0	-7.0
Lap belt, lower edge, at y = 0	-1.8	0.0	-9.5

4.2.4 Validation of THOR-5F FE Model in Rigid Bench

After completion of the Rigid Bench simulation, a qualitative comparison of the physical test video and model response was conducted (Figure 4.4). From the initial excursion (Figure 4.4A,D) to approximately 70 ms, the seat belt placement and occupant posture in the simulation matched the physical test, with both belts contacting similar regions of the THOR-5F shoulder in the FE model and the physical ATD. After approximately 70 ms, the differences in the rotational behavior of the THOR-5F FE and the physical ATD became apparent. The physical ATD exhibited more leftward rotation than the THOR-5F FE, while the THOR-5F FE maintained forward motion with minimal Z-rotation. As a result of the increased Z-rotation in the physical THOR-5F, the physical belt migrated to the outboard edge of the THOR-5F shoulder, while the simulation belt remained in a central position. At approximately 85 ms (Figure 4.4B,E), corresponding to peak seat belt loads, and at the end of the simulation (Figure 4.4C,F), the forward torso excursion remained similar between the FE model and the physical ATD, with differences in torso Z-rotation.

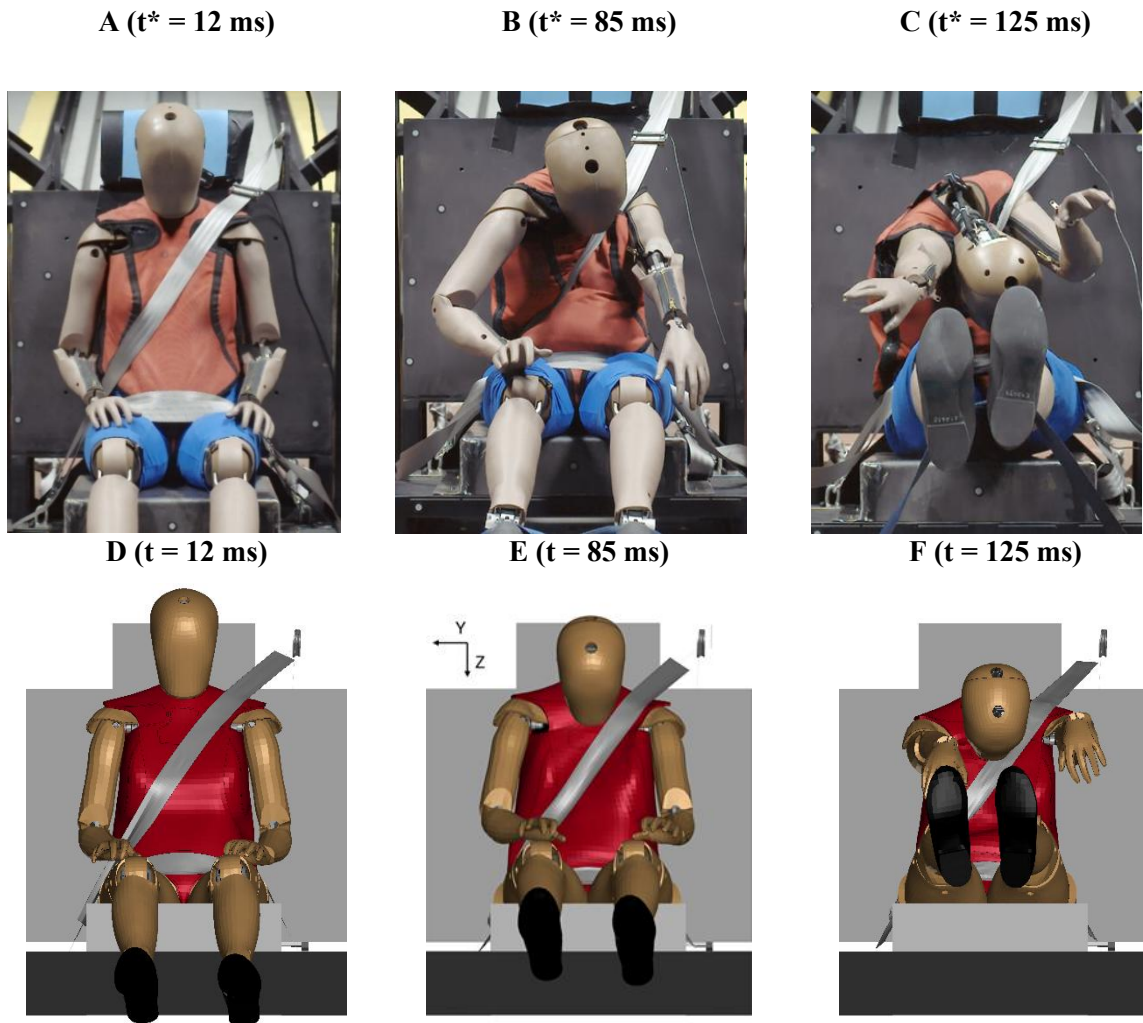


Figure 4.4: Images of the THOR-5F physical ATD (A–C) and THOR-5F FE model (D–E) at synchronized times $t = 12$ ms, 85 ms and 125 ms (*Video time offset by -50 ms and simulation time offset by -100 ms to synchronize test images (A–C) with model (D–F)). At 12 ms, the FE model posture matched the physical ATD. At 85 ms and 125 ms, lower Z-rotation was observed in the FE model than in the physical ATD.

Additionally, the kinematic and kinetic responses measured by the FE model were compared to the physical ATD test data (Figure 4.5). The shoulder belt, lap belt, and peak pelvis resultant acceleration (Figure 4.5F,I,J) of the THOR-5F FE model had similar magnitudes and timing to those of the

THOR-5F physical ATD, indicating accurate interaction of the model with the belt and lower body kinematics. From the kinematic traces, it was observed that there were notable differences between simulation and experiments in the upper body, especially in the head (Figure 4.5A), which experienced nearly 30% higher peak accelerations after matching the rise in acceleration.

The chest deflections in the model (Figure 4.5B,C,D,E) had similar rise and peak timing, with the upper left and right chest in the FE model measuring chest deflections up to 20% higher than the physical test, while the lower right chest deflection was approximately 15% lower than the physical test. The lower left chest deflection of the THOR-5F FE (Figure 4.5C) was approximately 50% lower than the physical test, after having a similar rise in deflection. One of the largest differences between the THOR-5F FE and physical tests was observed in iliac spine force (Figure 4.5G,H): the THOR-5F FE measured 40% of the force measured in the physical tests, despite similar peak timing.

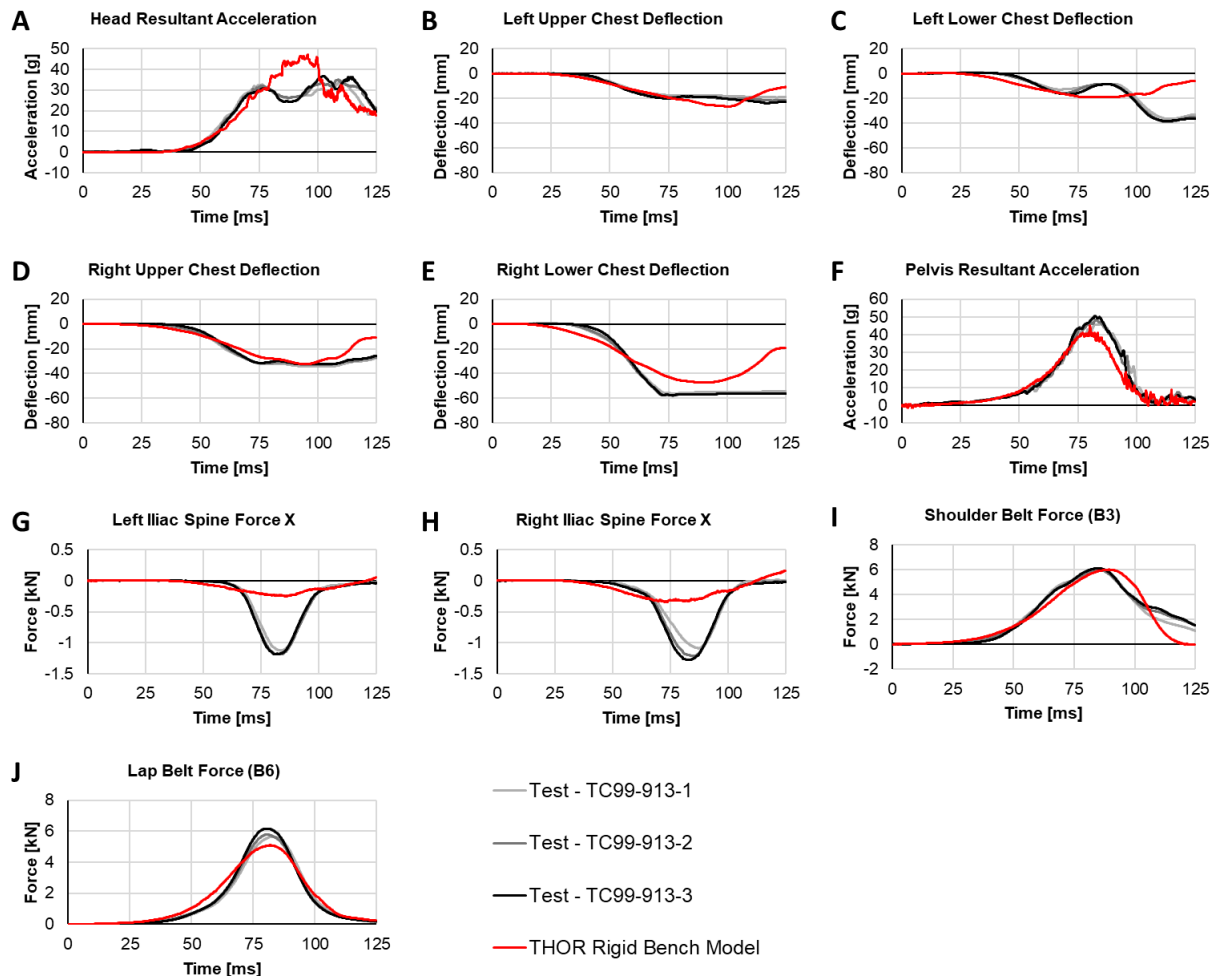


Figure 4.5: Responses of THOR-5F FE model and physical THOR-5F ATD in the Rigid Bench.

The THOR-5F FE Rigid Bench simulation was validated using the experimental tests and quantified through ISO 18571 cross-correlation ratings (Table 4.4). Good agreement was observed for the seat belt loads and the resultant acceleration of the pelvis. Several kinematic measurements, including the right chest deflection, the left upper chest deflection, and the head resultant acceleration, received Fair ratings, indicating acceptable agreement in timing and magnitude between the physical tests and the simulation. The left lower chest deflection and iliac loads received poor ratings, indicating areas of discrepancy between the model and the physical test. The overall ISO 18571 rating of 0.71 indicated the model as “Fair”, and despite certain localized discrepancies, the THOR-5F Rigid Bench FE model

generally reproduced the key kinematic and kinetic responses, supporting its use for further simulations and analysis.

Table 4.4: ISO 18571 ratings and grades for THOR-5F FE model and physical THOR-5F ATD in the Rigid Bench.

Response	Rating	Grade
Head resultant acceleration	0.69	Fair
Chest left lower deflection	0.44	Poor
Chest left upper deflection	0.69	Fair
Chest right lower deflection	0.62	Fair
Chest right upper deflection	0.78	Fair
Pelvis resultant acceleration	0.88	Good
Left iliac load	0.61	Fair
Right iliac load	0.58	Fair
Shoulder belt load	0.86	Good
Lap belt load	0.91	Good
Average	0.71	Fair

4.3 Integration of THOR-5F FE Model with the Semi-Rigid Bench and Assessment Using Reported Experimental Test Data

4.3.1 Semi-Rigid Bench FE Model

The components of the Semi-Rigid Bench consisted of the seatback, anti-submarining pan, seat pan, footrest, and seat belt anchor assemblies. The Semi-Rigid Bench seatback angle was identical to the ENOP bench FE model and the physical Semi-Rigid Bench described in Richard et al. (2022) (Figure 4.6); however, after positioning the THOR-5F FE, the seatback was translated approximately 25 mm forward in the X direction. The Semi-Rigid Bench seatback differed in length by +64 mm relative to the ENOP FE model and -74 mm relative to the physical Semi-Rigid Bench. The seat pan was 20 mm

wider than the Rigid Bench. Additionally, while the physical Semi-Rigid Bench and ENOP FE model also had a foam cushion on the seatback, the Semi-Rigid Bench model had a rigid seatback.

The geometry of the key components, such as the seat pan length and angle, footrest position, anti-submarining ramp angle, and the seat belt anchor locations, was defined using the reported values from Richard et al. (2022) and therefore matched the physical ENOP bench for these components.

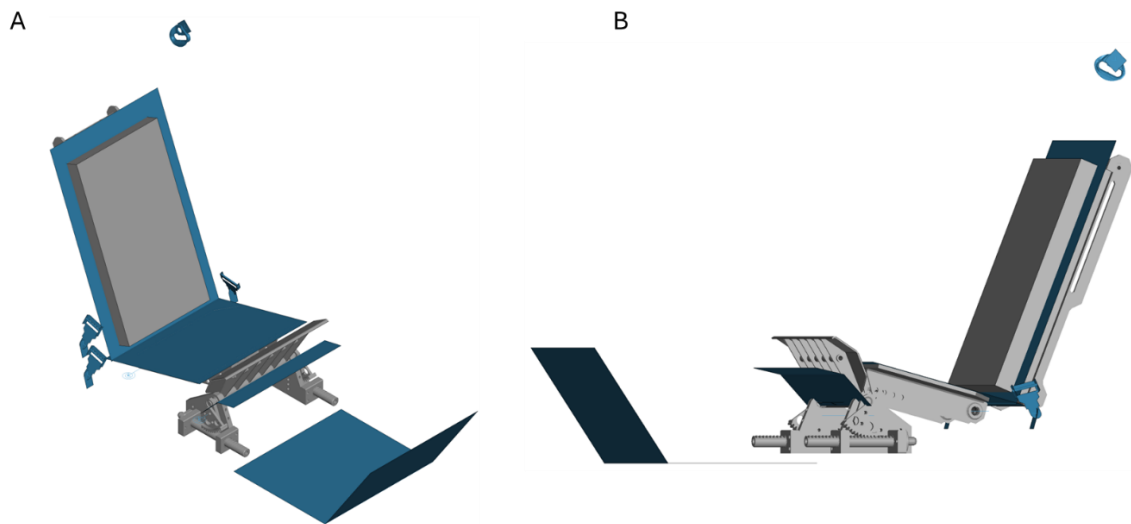


Figure 4.6: Overlay of Semi-Rigid Bench FE model (blue) on ENOP bench model (gray) with (A) Isometric view and (B) Oblique lateral view. The ENOP bench model included a foam cushion layer on the seat back, which was not present in the Semi-Rigid Bench model.

4.3.2 Positioning of THOR-5F FE in the Semi-Rigid Bench Model

The THOR-5F FE position, following marionette positioning and a 50 ms seat belt slack removal period, differed from the THOR-5F physical ATD position in the Richard et al. (2022) tests. Compared to the average physical ATD H-point measurement, the THOR-5F FE model was 2.6 mm rearward and 7.1 mm lower (Table 4.5). The differences in the angular measurements were generally small, with all angles being approximately within 1° of the physical test position. The largest

deviation in angular measurement was the head angle, with the THOR-5F FE having a head angle 1.3° higher than the average of the tests.

Table 4.5: Comparison of THOR-5F FE position to Richard et al. (2022) Test A11 and A12 measurements.

Measurement	Difference between simulation and average of Richard et al. (2022) test values
Head angle [°]	1.3
Lower neck angle [°]	1.0
Mid thorax angle [°]	0.5
Lower thorax angle [°]	0.4
Pelvis (sacrum) angle [°]	0.7
H-point X [mm]	-2.6
H-point Z [mm]	-7.1

4.3.3 Implementation of Four-Point, Two-Belt Seat Belt System in the Semi-Rigid Bench FE Model

The lap belt positioning in the FE model was further rearward and lower than that measured in the physical tests, with the largest discrepancy of 11.3 mm observed at the midpoint of the upper edge of the lap belt (Figure 4.7). All other measurements were within 10 mm of the physical test measurements (Table 4.6).

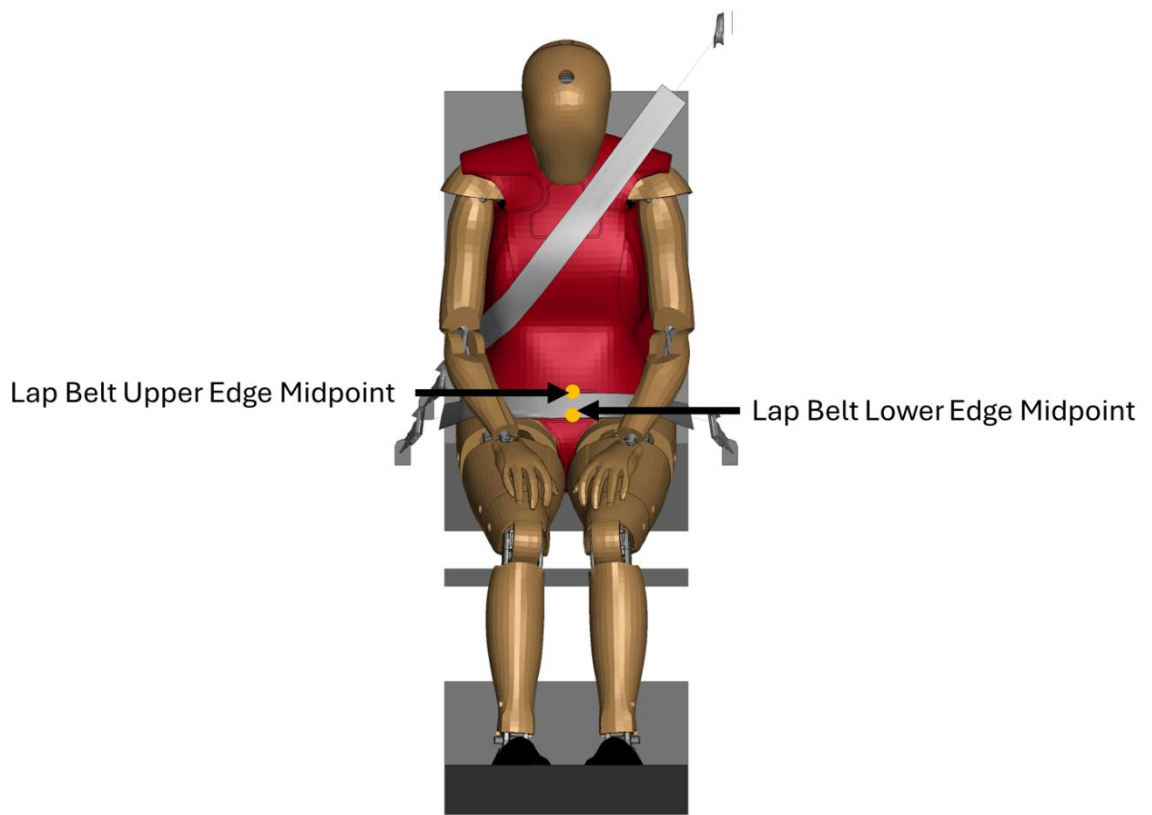


Figure 4.7: Definition of lap belt measurement points on the THOR-5F FE model in the Semi-Rigid Bench. The midpoints ($Y=0$) of the upper and lower lap belt edges were used to compare with the lap belt position in the physical tests (Table 4.6).

Table 4.6: Distance between THOR-5F FE model and physical ATD belt position.

Distance of belt point from average of test values			
Measurement	X [mm]	Y [mm]	Z [mm]
Lap belt, upper edge, at $y = 0$	-9.4	0	-11.1
Lap belt, lower edge, at $y = 0$	-6.9	0	-8.9

The seat belt retractor implemented in the Semi-Rigid Bench FE model incorporated an additional 30 mm of available pullout compared to the zero pullout reported in Richard et al. (2022), which

accounted for potential unintentional slack in the seat belt system. The additional pullout resulted in seat belt engagement timing matching the physical test data, with shoulder belt forces peaking around 75 ms and lap belt forces peaking at approximately 80 ms.

4.3.4 Assessment of THOR-5F FE in Semi-Rigid Bench

After simulating with the Semi-Rigid Bench model, the kinematics of the THOR-5F FE were compared with the physical test at times $t = 0, 75$, and 120 ms (Figure 4.8). The THOR-5F FE generally matched the kinematics of the physical ATD, with similar forward motion at 75 ms. At 120 ms, the THOR-5F FE showed minor differences in kinematics compared to the physical test, with a slightly less forward pitch and reduced arm extension; however, the forward excursion of the pelvis and lower body matched the physical test.

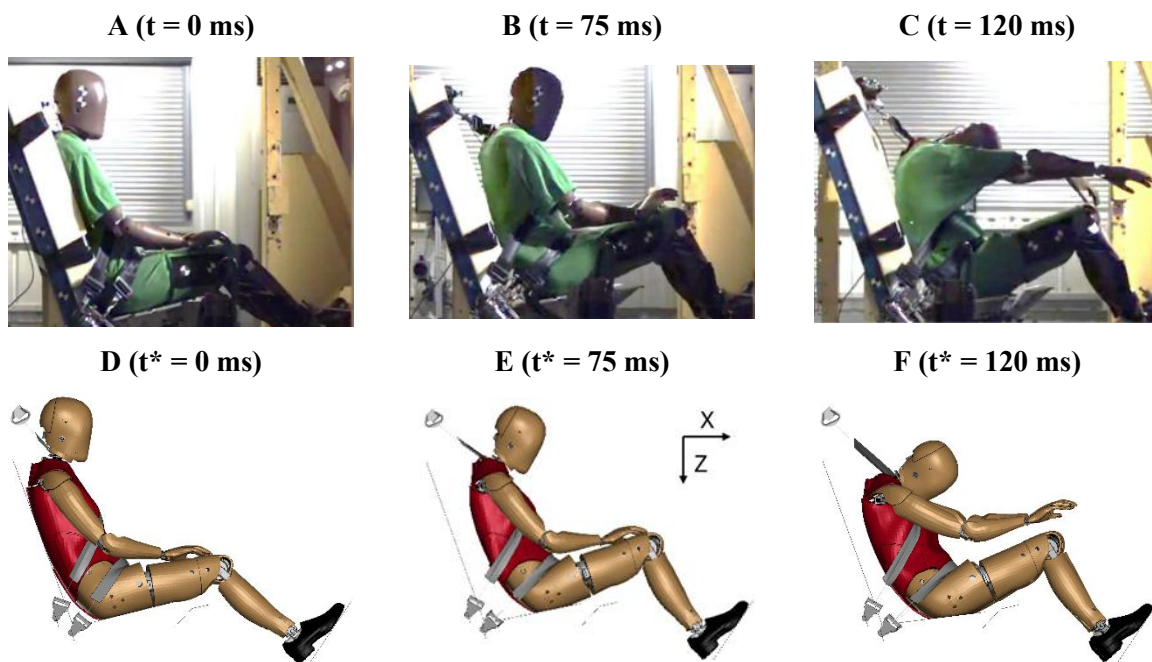


Figure 4.8: Images of the THOR-5F physical ATD (A–C) and THOR-5F FE model (D–F) at synchronized times $t = 0$ ms, 75 ms, and 120 ms. (*Simulation time offset by -50 ms to synchronize model (D–F) with test images (A–C)). Physical test images from Richard et al. (2022) reproduced with permission from the Stapp Association.

In addition to visual comparisons of kinematics, test traces of kinetics and kinematics were also compared with digitized traces from the physical tests. The shoulder belt load and peak pelvis resultant accelerations (Figure 4.9F,I) showed close agreement in timing and magnitude to the physical tests, indicating the simulation captured the loading in the physical test. The lap belt force (Figure 4.9J) was slightly higher than the physical test, with a peak magnitude approximately 10–18% higher than physical test measurements.

The iliac spine forces (Figure 4.9G,H) of the THOR-5F FE had similar timing to the physical test, with substantial differences in magnitude and signal polarity. The FE model sensor force recorded rising positive values, whereas the test measurements were negative and returned to 0 near the end of the test. The left and right iliac forces were underpredicted by approximately 90% relative to the physical test peaks. The head resultant acceleration peak (Figure 4.9A) was 15% lower than the physical tests and occurred within 2.5 ms of the physical test peaks. The chest deflection timing varied by region; relative to the physical tests, the left upper chest deflection (Figure 4.9B) was delayed by approximately 8 ms, the left lower deflection (Figure 4.9C) was within 1 ms, the right upper deflection (Figure 4.9D) was delayed by 17 ms, and the right lower deflection (Figure 4.9E) was delayed by 15 ms. Comparing the magnitudes, the chest deflections were within 14%, 46%, 12% and 17% of the physical test responses for the left upper, left lower, right upper and right lower regions, respectively.

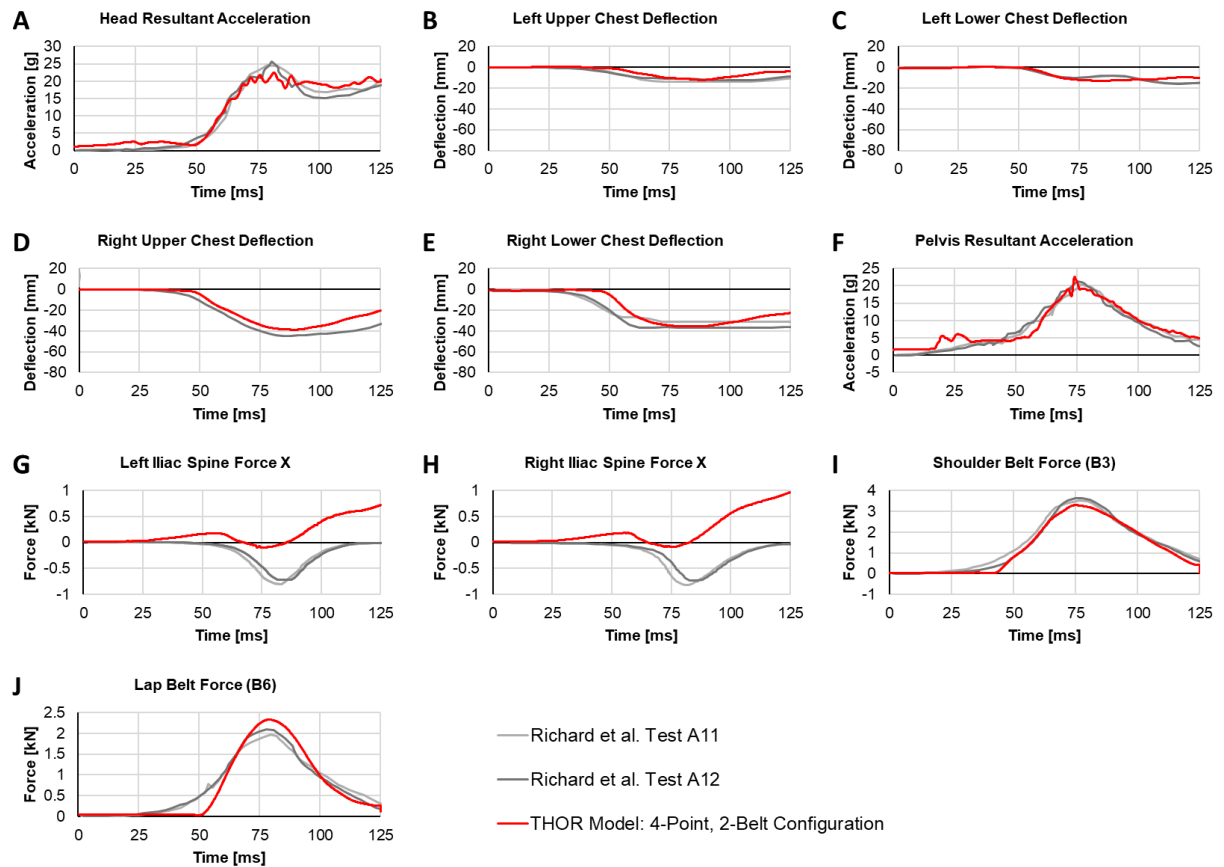


Figure 4.9: Responses of THOR-5F FE model in the Semi-Rigid Bench with the 4-point, 2-belt configuration compared to the physical test data by Richard et al. (2022).

The comparison of the THOR-5F FE in the Semi-Rigid Bench model to the experimental tests was quantified by calculating ISO 18571 cross-correlation scores for two physical tests, then averaging the results (Table 4.7). The head resultant acceleration, chest right upper deflection, pelvis resultant acceleration, shoulder belt load and lap belt load were rated as good, indicating strong agreement between the model and the physical test data. The chest left lower, left upper and right lower deflections were rated as fair, with the THOR-5F FE model tracking the physical tests with slight deviations. The iliac spine loads were rated as poor, with the THOR-5F FE having a different response trend than the physical test. Overall, the average rating of the model was 0.71, which corresponded to a “Fair” grade. Despite differences in overall iliac load measurements, the model

tracked the kinematics and kinetics of the physical ATD tests, demonstrating its appropriateness for the DOE simulations.

Table 4.7: ISO 18571 ratings and grades for THOR-5F FE model in Semi-Rigid Bench compared to Richard et al. (2022) Tests A11 and A12. The table contains the ratings for each test along with the average ratings and associated grades.

Response	Test A11 rating	Test A12 rating	Average rating	Grades
Head resultant acceleration	0.87	0.85	0.86	Good
Chest left lower deflection	0.73	0.67	0.70	Fair
Chest left upper deflection	0.65	0.69	0.67	Fair
Chest right lower deflection	0.68	0.72	0.70	Fair
Chest right upper deflection	0.76	0.81	0.79	Fair
Pelvis resultant acceleration	0.92	0.87	0.9	Good
Left iliac load	0.34	0.47	0.41	Poor
Right iliac load	0.4	0.33	0.37	Poor
Shoulder belt load	0.91	0.94	0.93	Good
Lap belt load	0.80	0.81	0.81	Good
Average	0.71	0.72	0.71	Fair

4.3.5 Implementation of 3-Point Belt System in Semi-Rigid Bench Model

The 3-point seat belt configuration from the Rigid Bench was implemented in the Semi-Rigid Bench for comparison between the loading conditions. When the updated Semi-Rigid Bench with the 3-point belt configuration was run using the crash pulse from Richard et al. (2022), the shoulder belt peak force (Figure 4.10G) was 27% lower and occurred approximately 3 ms later than in the 4-point 2-belt configuration. The lap belt peak force (Figure 4.10H) was 22% lower and occurred approximately 10 ms earlier. The differences in the force measurements were due to the belt engaging earlier and experiencing more pullout.

Despite differences in seat belt force timing and magnitude, the chest deflections (Figure 4.8B-E) were only moderately lower in magnitude (approximately 60–80% of peak values observed in the 4-point 2-belt configuration) with similar timing. The resultant head acceleration (Figure 4.8A) also exhibited a similar magnitude as the physical tests, with the peak value within 12% of the physical test, despite a more gradual rise in acceleration and a 24 ms delayed peak. Although the 3-point belt configuration exhibited a different loading response, the overall kinematic patterns remained similar, indicating that the Semi-Rigid Bench model with the 3-point belt configuration was suitable for the DOE study.

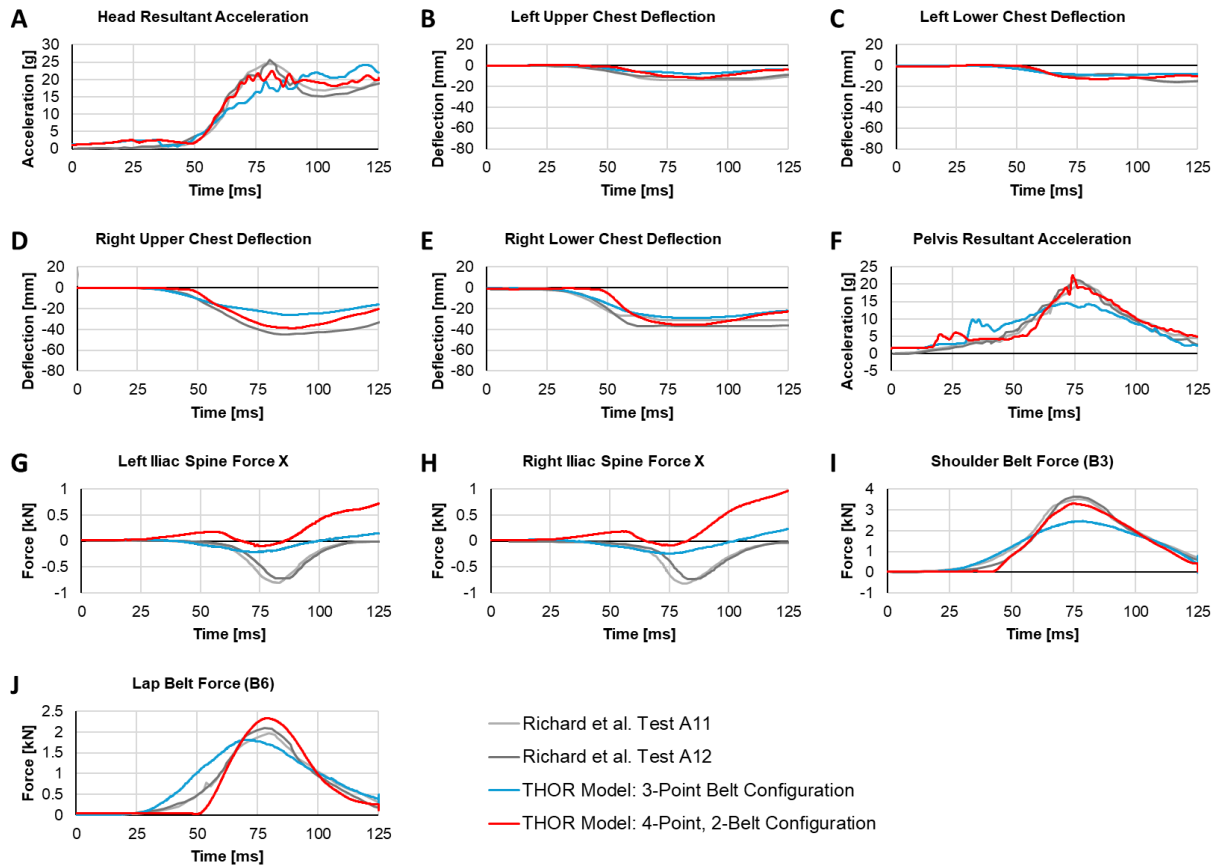


Figure 4.10: Responses of THOR-5F FE model in the Semi-Rigid Bench with the 4-point, 2-belt configuration and the 3-point belt configuration compared to the physical test data from Richard et al. (2022).

4.4 Parametric Study of Belt Anchor Locations and THOR-5F FE Kinematic Responses

Fifty-two (27 for shoulder belt anchor and 25 for lap belt anchor) seat belt anchor configurations were simulated in each of the Semi-Rigid Bench and the Rigid Benches, resulting in a total of 108 simulations. Each simulation was run to completion with LS-DYNA v12.2.1 with 32 cores, requiring approximately 1.5 hours per run. For each DOE configuration, the filtered peak chest deflections and peak pelvis resultant accelerations were extracted and used to compare the effect of anchor positions. For each simulated configuration, the peak chest deflections from the 4 IR-TRACC sensors and peak pelvis resultant acceleration were measured using post-processing software (LS-PrePost 4.10). The responses were filtered according to SAE J211 (Table 3.4). From the peak responses in each configuration, a linear model with interactions was fitted to determine the main and interaction effects using a statistical software package (JMP Student Edition Version 19.0.1).

4.4.1 Screening of Non-characteristic Kinematics

Most shoulder belt anchor positions resulted in the shoulder belt restraining the THOR-5F as intended. However, three positions (Positions 24, 21, and 12), in which the shoulder belt anchor was outboard and low, resulted in a shoulder belt fit that failed to remain on the upper torso of the THOR-5F during the acceleration pulse, causing the belt to migrate downward to the lower abdomen (Figure 4.11). The belt restraint, when sliding off the shoulder, was uncharacteristic of a properly functioning restraint system and was not included in any of the analyses.

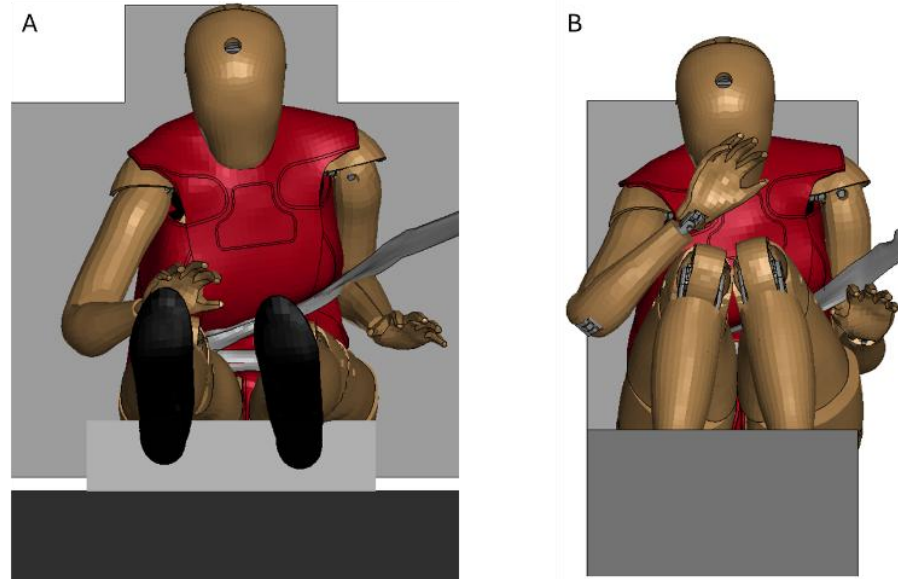


Figure 4.11: Shoulder belt migration from shoulder to the abdomen during simulation of Position 21 in the (A) Rigid Bench and (B) Semi-Rigid Bench.

The lap belt configurations resulted in two distinct kinematics observed between all simulations: submarining (Figure 4.12A,C), or a forward pitch (Figure 4.12B,D). Submarining occurred when the lap belt anchors were at the highest position (Z_{MIN}), resulting in undesirable kinematics for a restrained rear seat occupant. To maintain focus on functional restraint performance, the submarining cases were excluded from the analysis.

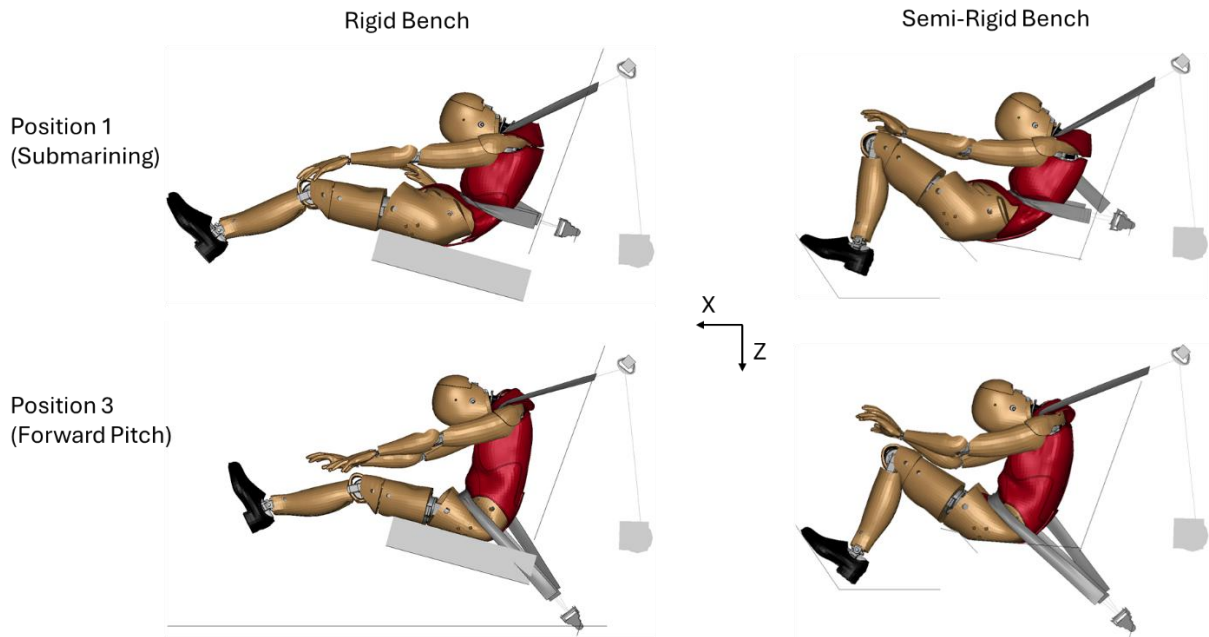


Figure 4.12: Kinematic response modes resulting from lap belt anchor variations. Submarining occurred for multiple lap belt anchor configurations in both benches, with Position 1 (top row) shown as an example. Forward pitch was observed for a different set of lap belt anchor configurations in both benches, with Position 3 (bottom row) shown as an example.

4.4.2 Shoulder Belt Anchor Investigation for Rigid Bench

The overall chest deflection peaks ranged from 42 mm to 63 mm (Figure 4.13A), with an average of 51 mm. The average deflection was 0.3 mm more than the midpoint configuration (Position 14), which exhibited a peak average chest deflection of 50.6 mm. The highest chest deflection relative to the average (+24%) occurred at Position 19 (X_{MIN} , Y_{MIN} , Z_{MIN}), while the lowest deflection relative to the average (-17%) was recorded at Position 7 (X_{MIN} , Y_{MAX} , Z_{MIN}).

The peak pelvis resultant accelerations ranged from 36.4 g to 49 g (Figure 4.13B), with an average of 42.6 g, 0.3 g higher than at the midpoint (Position 14) configuration. The highest peak pelvis resultant accelerations occurred at the forward and lower anchor (+X, +Z) locations, with Position 27 (X_{MAX} ,

Y_{MAX} , Z_{MAX}) having the highest acceleration compared to the average (+15 %). The lowest acceleration relative to the average (-14%) was observed in Position 9 (X_{MIN} , Y_{MIN} , Z_{MAX}).

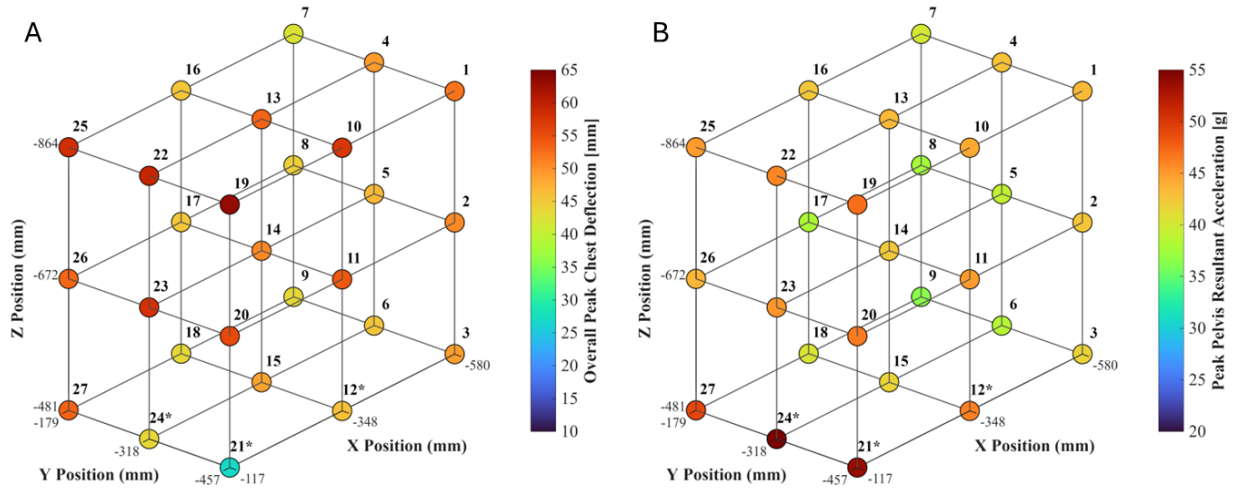


Figure 4.13: THOR-5F responses for shoulder belt anchor positions in the Rigid Bench. (A) Overall peak chest deflection for simulated anchor locations (B) Peak pelvis resultant accelerations for simulated anchor locations. Marker numbers correspond to the DOE position. *Indicates configuration removed from analysis.

The right lower chest sensor (Figure 4.14D) generally measured the highest chest deflections, accounting for 22 of 24 overall peak chest deflections. The two exceptions were Positions 20 and 22, which recorded the highest deflection at the left lower and right upper sensors, respectively. The left lower sensor (Figure 4.14C) generally measured lower deflections. Chest deflections increased at more outboard (-Y) and forward (+X) anchor locations for the left chest sensors and the right lower chest sensor (Figure 4.14A,C,D), while the right upper sensor showed increased deflections at higher (-Z) and inboard (+Y) anchor locations (Figure 4.14B).

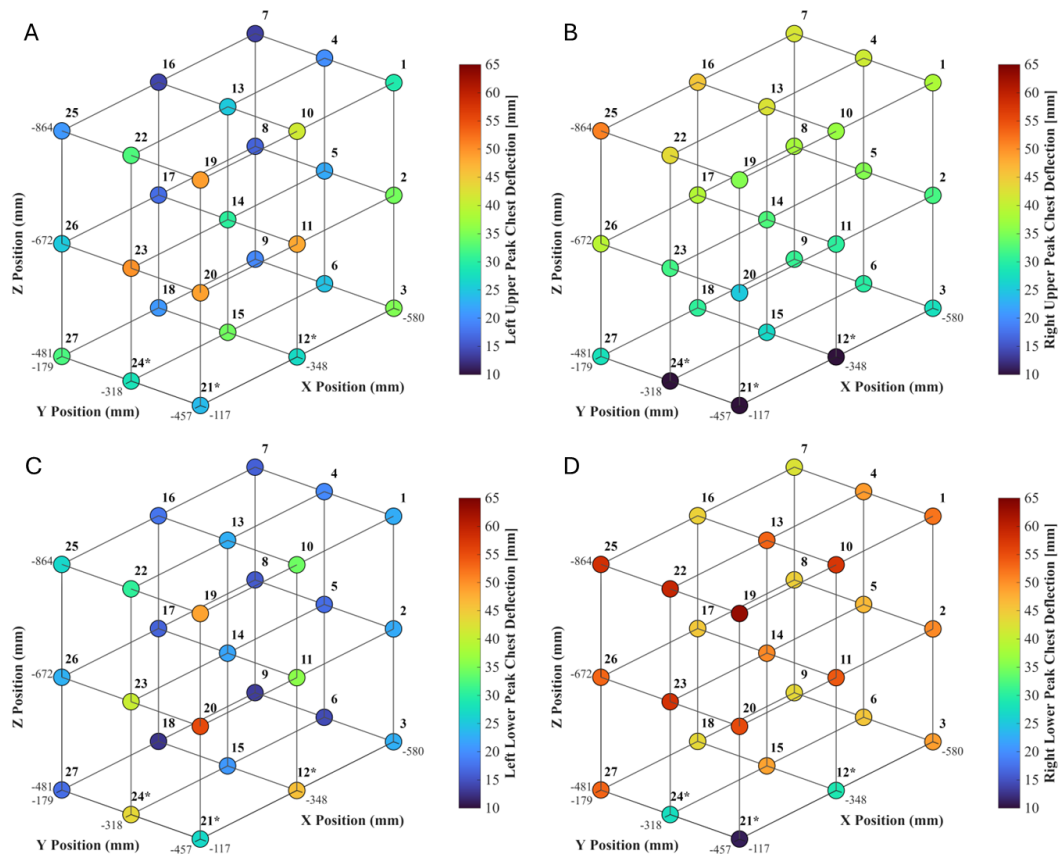


Figure 4.14: THOR-5F peak chest sensor deflections for shoulder belt anchor locations in the Rigid Bench: (A) left upper sensor, (B) right upper sensor, (C) left lower sensor, and (D) right lower sensor. Marker numbers correspond to DOE position. *Indicates configurations removed from analysis.

4.4.2.1 Shoulder Belt Anchor Design of Experiment (DOE) Effects Calculations for Peak Chest Deflection

The linear regression model explained 92% of the variance in overall peak chest deflections on the Rigid Bench model, with an R^2 value of 0.92. The effects analysis showed that the overall peak chest deflection on the Rigid Bench was significantly influenced by all three main effects: the shoulder belt X, Y, and Z-directions, and one interaction effect: XZ (Figure 4.15). The X-direction had the largest effect and was positive; moving the shoulder belt anchor X-position from minimum to maximum (e.g.

from Position 1 to 19) increased the deflection by an average of 9.5 mm. The Y-direction effect on chest deflection was the second highest main effect and negative, as moving the anchor from minimum to maximum (e.g. Position 1 to 7) decreased the peak deflection by 6.5 mm. The Z-direction effect on peak chest deflection was also negative; moving the anchor from minimum to maximum in the Z-direction (e.g. Position 1 to 3) reduced chest deflection by an average of 4.5 mm. The XZ interaction effect was negative, indicating that the reduction in chest deflection associated with Z-direction was greater at X_{MAX} than X_{MIN} by 3.16 mm.

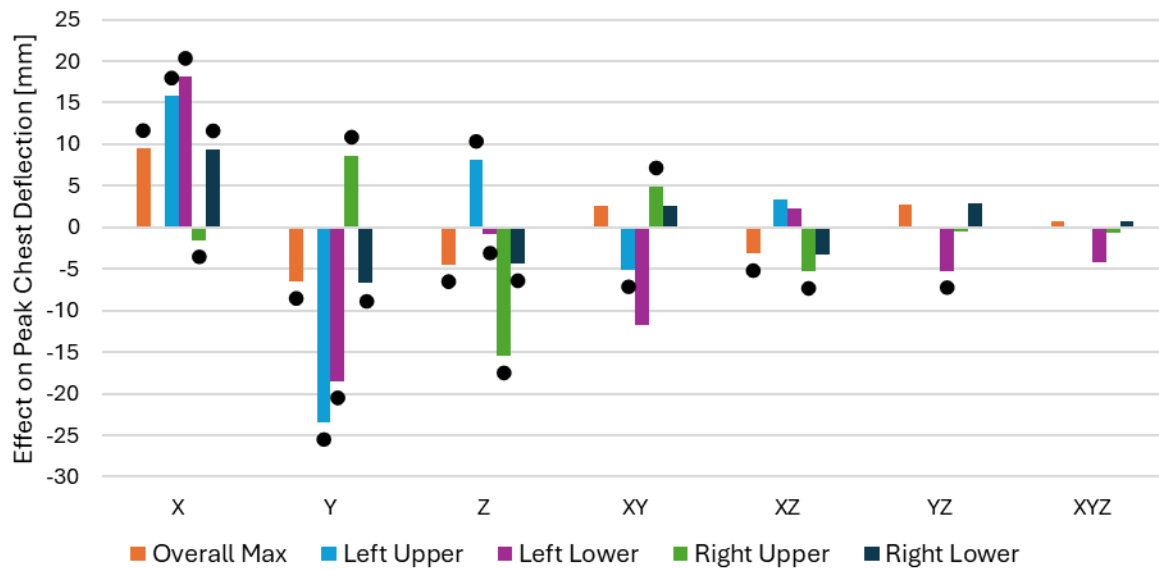


Figure 4.15: Main and interaction effects on peak chest deflection for the Rigid Bench shoulder belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).

The regression models for the individual chest regions ranged from 0.91 to 0.99. The lower right chest sensor exhibited consistent trends compared to the overall peak chest deflection for the main effects for magnitude and significance. The left side sensors showed larger effect magnitudes for the X and Y-direction main effects, while maintaining the same signs as the overall peak chest deflection. In contrast, the upper right chest showed signs with a lower X-direction magnitude and a slightly

larger Y magnitude compared to the overall peak chest deflection. All chest sensors except for the upper left chest sensors showed the same sign for the Z-direction effect.

Interaction effects varied with the sensor location. The upper right chest matched the overall peak deflection, with both showing a significant XZ interaction. Both upper chest sensors exhibited a significant XY effect, while the lower left chest sensor showed a significant YZ interaction.

4.4.2.2 Shoulder Belt Anchor Design of Experiment (DOE) Effects Calculations for Peak Pelvis Resultant Acceleration

The fit of the multiple regression model to the peak pelvis resultant accelerations on the Rigid Bench had an R^2 value of 0.88. The effects analysis for peak pelvis resultant acceleration showed that the X,Y main effects and the XZ interaction effects were significant (Figure 4.16). The X-direction had the largest effect magnitude, as moving the shoulder belt anchor forward (+X) from minimum to maximum (e.g. Position 1 to 19) increased the peak acceleration by an average of 5.6 g. The Y-direction had the second largest effect, followed by the XZ interaction effect. Moving the shoulder belt anchor inboard (+Y) from minimum to maximum (e.g. Position 1 to 7) decreased the peak acceleration by an average of 3.2 g. The significant XZ interaction effect indicated that the effect of the fore-aft (X) shoulder belt anchor position depended on the vertical (Z) position. When moving the shoulder belt anchor downward (+Z) from minimum to maximum, the increase in peak pelvis acceleration from moving the anchor forward (+X) became larger by approximately 2.19 g.

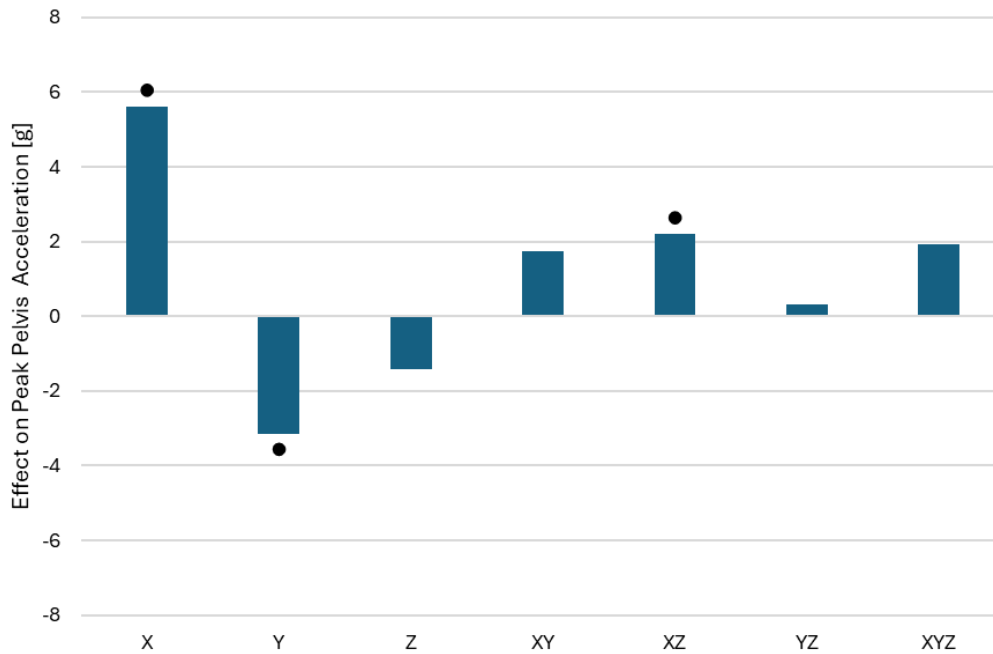


Figure 4.16: Main and interaction effects on peak pelvis resultant acceleration for the Rigid Bench shoulder belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).

4.4.3 Lap Belt Anchor Investigation for Rigid Bench

The average chest deflection peaks ranged from 44.3 mm to 53.1 mm (Figure 4.17A), with an average of 49.2 mm, 1.4 mm lower than the midpoint configuration (Position 14) deflection of 50.6 mm. The largest chest deflection relative to the average (+8%) occurred at Position 23 (X_{MAX} , Y_{MID} , Z_{MID}). The lowest chest deflection relative to the average (-10%) was recorded at Position 5 (X_{MIN} , Y_{MID} , Z_{MID}).

The peak pelvis resultant accelerations ranged from 30.8 g to 54.8 g (Figure 4.17B), with an average of 43.6 g, 1.3 g lower than the midpoint configuration measurement (42.3 g). The highest peak pelvis resultant accelerations occurred at the forward and midpoint anchor locations, with Position 20 (X_{MAX} , Y_{MIN} , Z_{MID}) having the largest highest acceleration compared to the average (+25.8%). The lowest acceleration relative to the average (-29.3%) was observed in Position 25 (X_{MAX} , Y_{MAX} , Z_{MIN}).

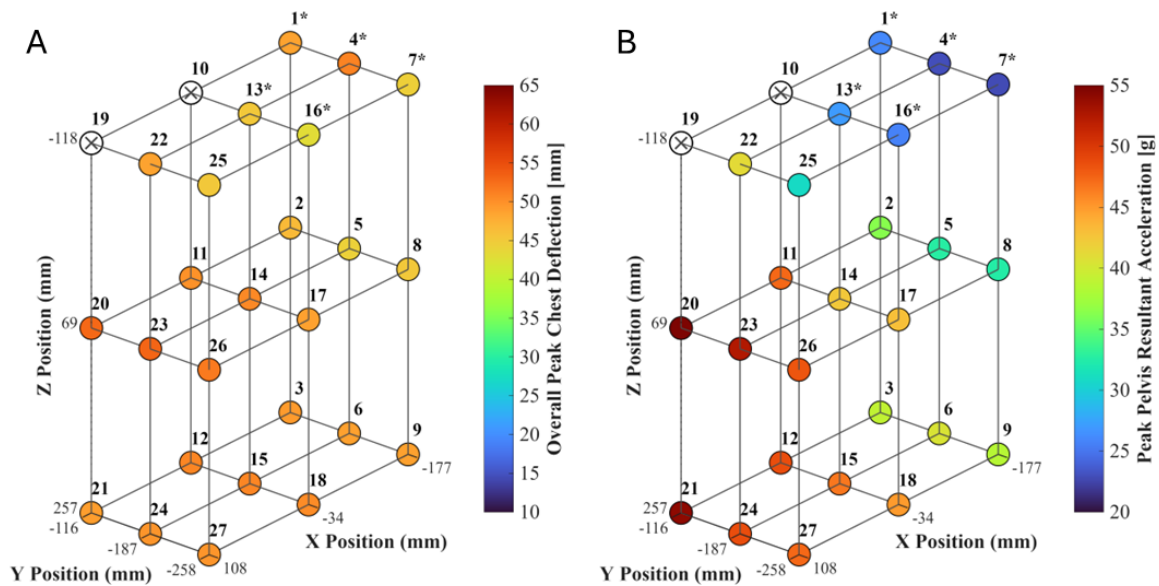


Figure 4.17: THOR-5F responses for outboard lap belt anchor positions in the Rigid Bench. (A) Overall peak chest deflection for simulated outboard lap belt anchor locations (B) Peak pelvis resultant accelerations for simulated outboard lap belt anchor locations. Marker numbers correspond to the DOE position. *Indicates configurations removed from the analysis; ⊗ indicates configurations that were not simulated.

The right lower chest sensor (Figure 4.18D) generally recorded the highest chest deflections, accounting for all 22 overall peak chest deflections, and showed increased deflections at more forward (+X) anchor locations. The left-side chest sensors (Figure 4.18A,C) showed a similar trend, although with lower magnitudes. The right upper sensor (Figure 4.18B) exhibited the opposite trend, with higher deflections at more rearward (-X) and higher (-Z) lap belt anchor locations.

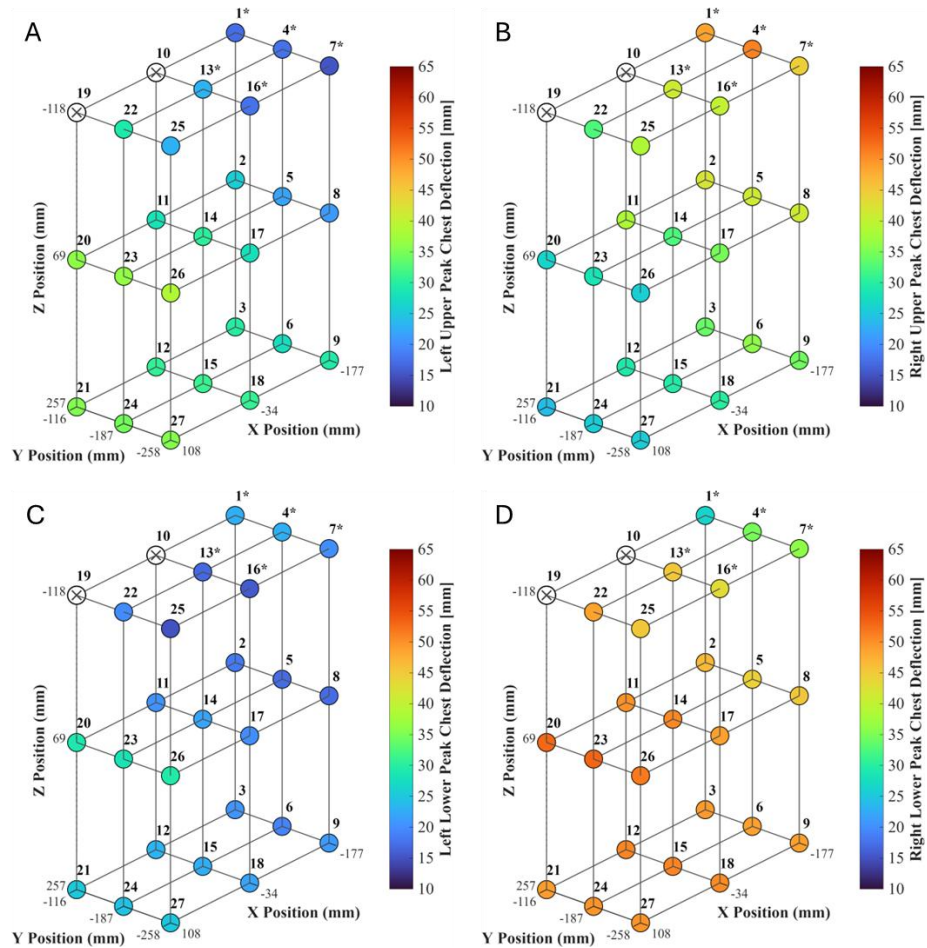


Figure 4.18: THOR-5F peak chest sensor deflections for lap belt anchor locations in the Rigid Bench: (A) left upper sensor, (B) right upper sensor, (C) left lower sensor, and (D) right lower sensor. Marker numbers correspond to DOE position. *Indicates configurations removed from the analysis; ⊗ indicates configurations that were not simulated.

4.4.3.1 Lap Belt Anchor Design of Experiment (DOE) Effects Calculations for Peak Chest Deflection

The linear model explained the peak overall chest deflection on the Rigid Bench model with an R^2 value of 0.67. The effect analysis showed that the only significant effects were the X and Z-positions of the anchor (Figure 4.19). Moving the anchor forward (+X) from minimum to maximum (e.g.

Position 9 to 27) increased deflections by an average of 3.7 mm. Moving the anchor positions downward from minimum to maximum (e.g. Position 25 to 27) increased deflections by an average of 3.0 mm.

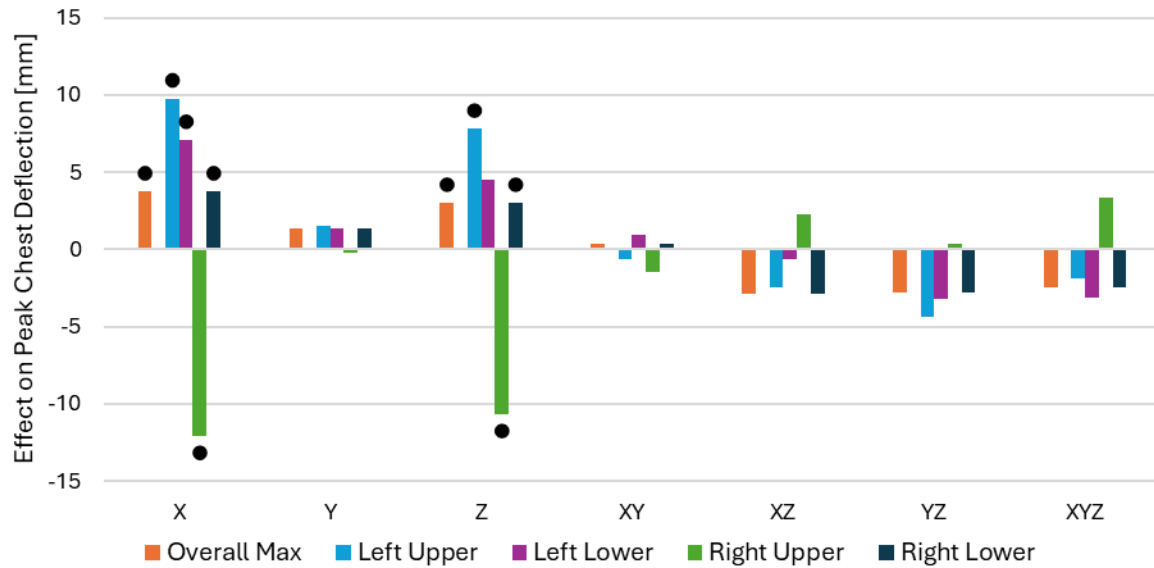


Figure 4.19: Main and interaction effects on peak chest deflection for the Rigid Bench lap belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).

For the individual chest sensors, the regression models showed R^2 values ranging from 0.67 to 0.92, indicating variability in model fit across sensor locations. The upper chest sensors showed higher R^2 (0.8–0.92), while the lower chest sensors showed lower values (0.67).

For the main effects, the lower right and both left-side sensors showed consistent trends with the overall peak chest deflection, while the right upper sensor exhibited effects with opposite signs. The lower right sensor showed a similar magnitude to the overall response, while the other three sensors exhibited larger magnitudes. None of the interaction effects were significant across the sensors.

4.4.3.2 Lap Belt Anchor Design of Experiment (DOE) Effects Calculations for Peak Pelvis Resultant Acceleration

The linear regression model fit the peak pelvis resultant accelerations on the Rigid Bench model DOE with an R^2 value of 0.86. The effect analysis showed that all three main effects (X, Y, and Z) were significant (Figure 4.20). The largest effect was the X-direction, as moving the lap belt anchors forward from minimum to maximum (e.g. Position 8 to 26) increased the peak pelvis resultant acceleration by an average of 13.1 g. The Z-direction had a positive effect on the peak pelvis resultant acceleration, as moving the anchors downward from minimum to maximum (e.g. Position 25 to Position 27) increased peak pelvis resultant accelerations by an average of 8.9 g. The Y-direction also had a positive effect on the peak pelvis resultant acceleration; moving the lap belt anchors from a minimum distance between anchors to a maximum lateral distance between anchors (e.g. Position 2 to Position 8) resulted in an increase in the pelvis acceleration by an average of 6 g.

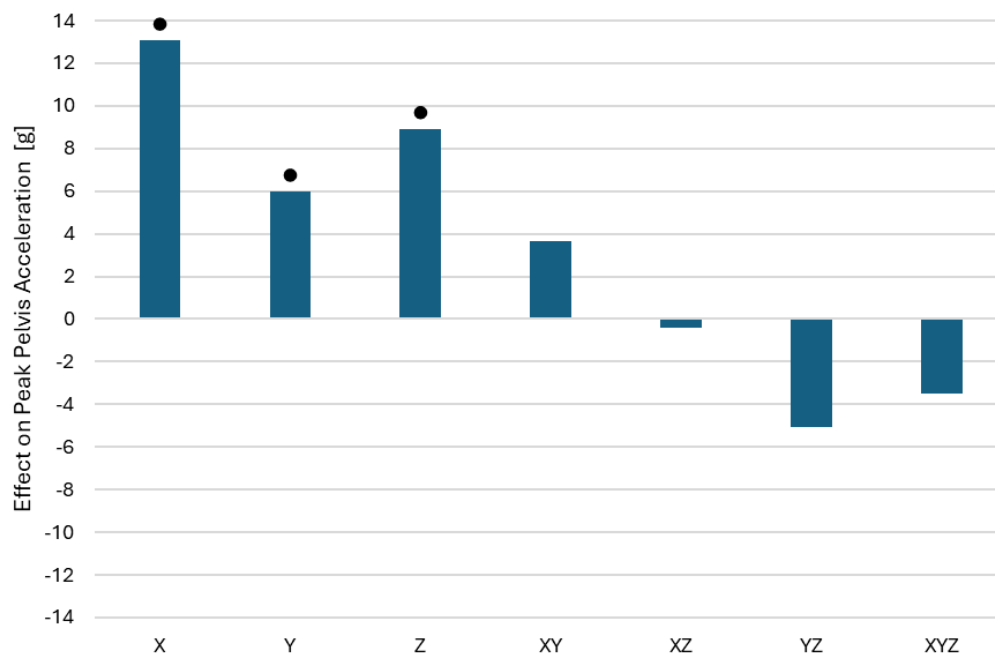


Figure 4.20: Main and interaction effects on peak pelvis resultant acceleration for the Rigid Bench lap belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).

4.4.4 Shoulder Belt Anchor Investigation for Semi-Rigid Bench

The peak overall chest deflections (Figure 4.21A) for the shoulder belt anchor DOE in the Semi-Rigid Bench ranged from 35.0 mm to 49.6 mm, with an average of 41.7 mm. The average was 0.4 mm less than the midpoint (Position 14) deflection of 42.0 mm. The anchors positioned inboard and rearward (-X, +Y) exhibited lower chest deflections, with Position 8 (X_{MIN} , Y_{MAX} , Z_{MID}) showing the lowest deflection relative to the average (-16%). The forward anchor positions (+X) resulted in the highest chest deflection values, with Position 22 (X_{MAX} , Y_{MID} , Z_{MIN}) showing the largest chest deflection relative to the average (+19%).

The peak pelvis resultant accelerations (Figure 4.21B) ranged from 34.8 g to 45.6 g, with an average of 39.8 g, which was 2.2 g higher than the midpoint (Position 14) measurement of 37.6 g. The lower peak pelvis resultant accelerations were found in the inboard and rearward (-X, +Y) shoulder belt anchor locations, with the lowest peak pelvis resultant acceleration relative to the average (-13 %) located at Position 8 (X_{MIN} , Y_{MAX} , Z_{MID}). Position 27 (X_{MAX} , Y_{MAX} , Z_{MAX}) experienced the highest peak pelvis resultant acceleration relative to the average (+15 %).

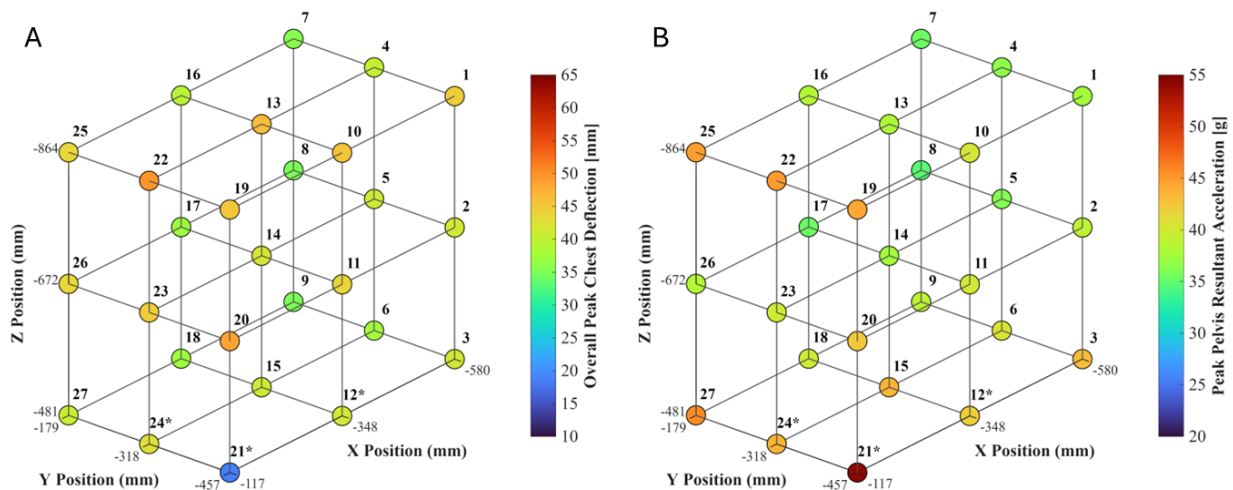


Figure 4.21: THOR-5F responses for shoulder belt anchor positions in the Semi-Rigid Bench.

(A) Overall peak chest deflection for simulated anchor locations (B) Peak pelvis resultant accelerations for simulated anchor locations. Marker numbers correspond to the DOE position.

***Indicates configuration removed from analysis.**

Higher chest deflections were generally observed at the right-side chest sensors (Figure 4.22B,D), while lower chest deflections were generally recorded at the left lower chest sensor (Figure 4.22C). The sensor recording the overall peak chest deflection varied across configurations, occurring most frequently at the right upper chest sensor (17 configurations), followed by the right lower (5) and left upper (2). The left-side chest sensors (Figure 4.22A,C) exhibited higher deflections at forward (+X) and outboard (+Y) anchor locations, with greater variation than the right-side sensors. The right lower chest sensor recorded higher deflections at more forward (+X) and outboard (+Y) anchor locations, while the right upper chest sensor recorded higher deflections at higher (+Z) anchor locations.

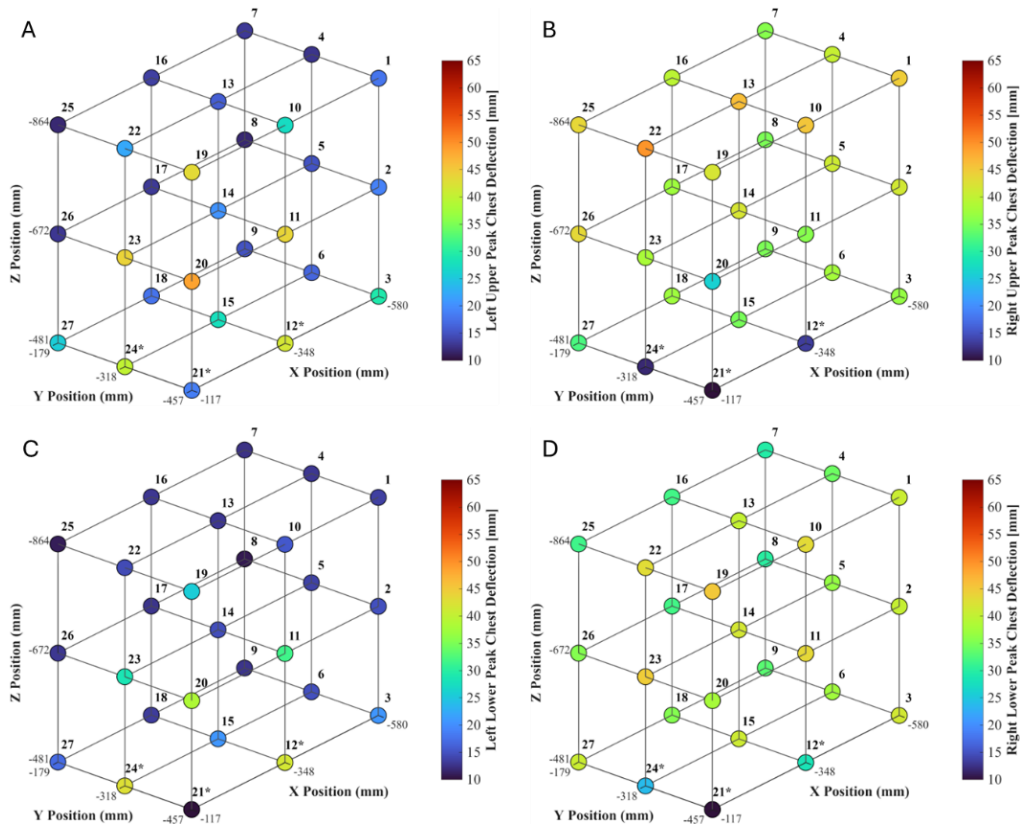


Figure 4.22: THOR-5F peak chest sensor deflections for shoulder belt anchor locations in the Semi-Rigid Bench: (A) left upper sensor, (B) right upper sensor, (C) left lower sensor, and (D) right lower sensor. Marker numbers correspond to DOE position. *Indicates configurations removed from analysis.

4.4.4.1 Shoulder Belt Anchor Design of Experiment (DOE) Effects Calculations for Peak Chest Deflection

The linear model fit the Semi-Rigid Bench model with an R^2 value of 0.89. The effect analysis indicated that the main effects (X,Y,Z) were all significant (Figure 4.23). Moving the shoulder belt anchor inboard from the minimum to maximum in the X-direction (e.g. from Position 1 to 19) decreased deflections by an average of 6.3 mm, while moving the anchor from minimum to maximum in the Y-direction (e.g. Position 1 to 7) decreased chest deflections by an average of 6.2 mm. Additionally, the Z main effect was negative, indicating that the chest deflections decreased by approximately 2.33 mm when the anchors were moved downwards from Z_{MIN} to Z_{MAX} (e.g. Position 1 to 3).

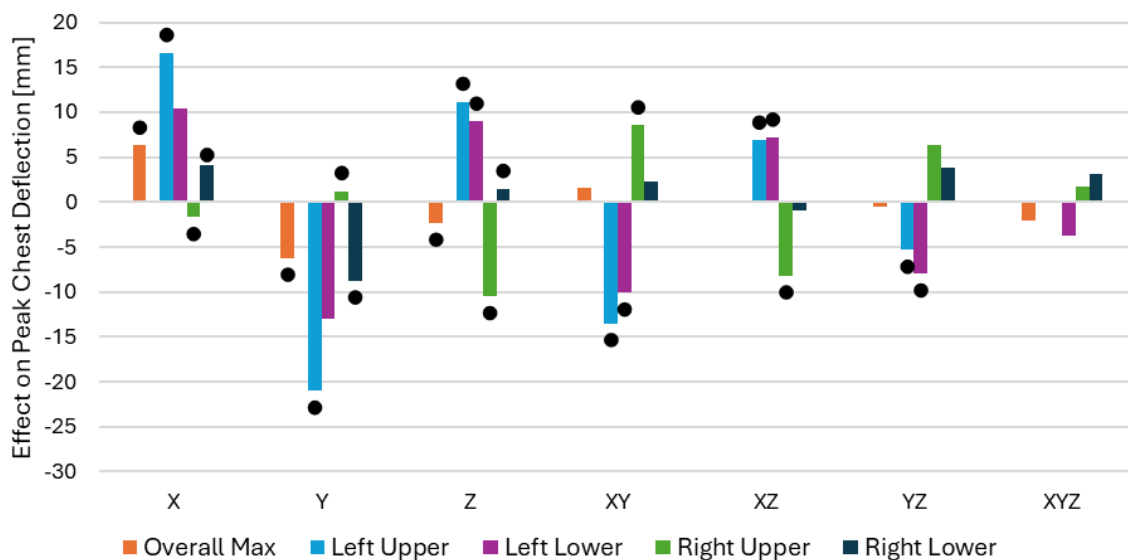


Figure 4.23: Main and interaction effects on peak chest deflection for the Semi-Rigid Bench shoulder belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).

The regression models for the individual chest sensors had R^2 values ranging from 0.78 to 0.91. The upper left, lower left and lower right chest sensors exhibited larger effect magnitudes compared to the overall peak chest deflection. The trends of these sensors were generally consistent with the overall

peak chest deflections in terms of significance and sign, with X and Y-direction producing the largest effects. However, for the Z-direction, both left chest sensors showed effects in the opposite sign compared to the overall peak deflection. The upper right chest sensor consistently exhibited opposite trends compared to the overall peak chest deflection for all effects except for the Z main effect and the XY interaction effect.

Overall, the left-side chest sensors showed sensitivity to all two-way interaction effects. The upper right sensor also exhibited significance for the XY and XZ interaction effects, although the effects were of the opposite sign compared to the other sensors.

4.4.4.2 Shoulder Belt Anchor Design of Experiment (DOE) Effects Calculations for Peak Pelvis Resultant Acceleration

For the Semi-Rigid Bench, the regression model fit showed an R^2 of 0.66. The effects analysis (Figure 4.24) indicated that the only significant effect on the peak pelvis resultant acceleration was the X-direction. When the shoulder belt anchor was moved forward from minimum to maximum (e.g. Position 1 to 19), the peak pelvis resultant acceleration increased by an average of 4.9 g.

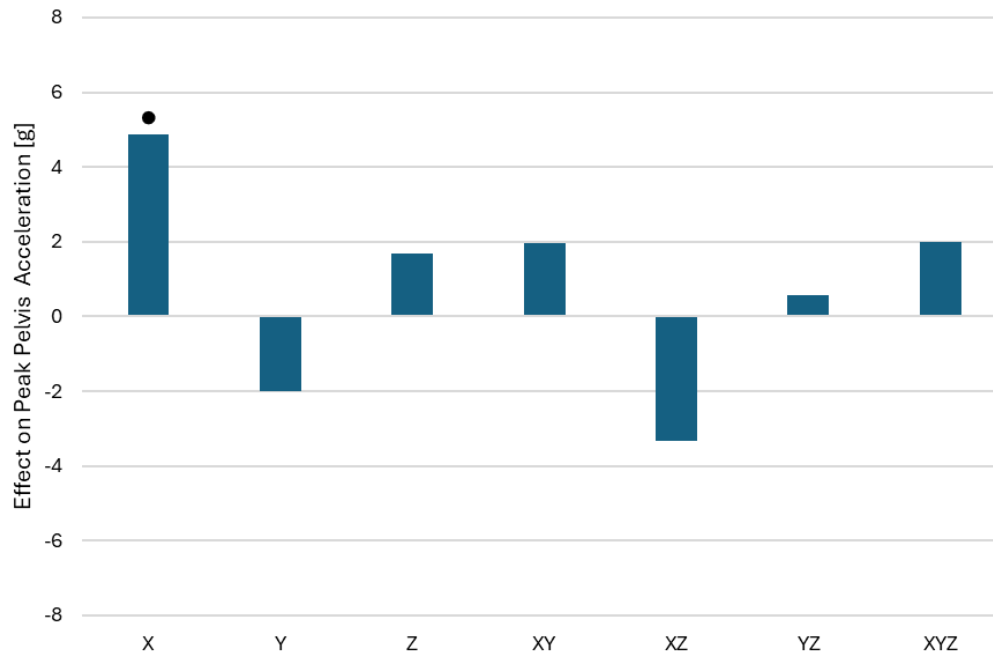


Figure 4.24: Main and interaction effects on peak pelvis resultant acceleration for the Semi-Rigid Bench shoulder belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).

4.4.5 Lap Belt Anchor Investigation for Semi-Rigid Bench

The average chest deflection peaks ranged from 40.3 mm to 44.6 mm (Figure 4.25A), with an average of 42.4 mm and a midpoint (Position 14) deflection of 42.0 mm. The highest deflection relative to the average (+5%) was at Position 2 (X_{MIN} , Y_{MIN} , Z_{MID}). The lowest chest deflection relative to the average (-5%) occurred at Position 21 (X_{MAX} , Y_{MIN} , Z_{MAX}). The peak pelvis resultant acceleration ranged from 31.9 g to 55.7 g (Figure 4.25B). The average was 40.4 g, 2.8 g higher than the midpoint configuration (Position 14). The lowest peak pelvis resultant acceleration relative to the average (-21 %) was at Position 5 (X_{MID} , Y_{MID} , Z_{MID}). The highest peak pelvis resultant acceleration compared to the average (+38 %) was at Position 2 (X_{MIN} , Y_{MIN} , Z_{MID}). The extreme values of peak pelvis resultant acceleration occurred at the midpoint location.

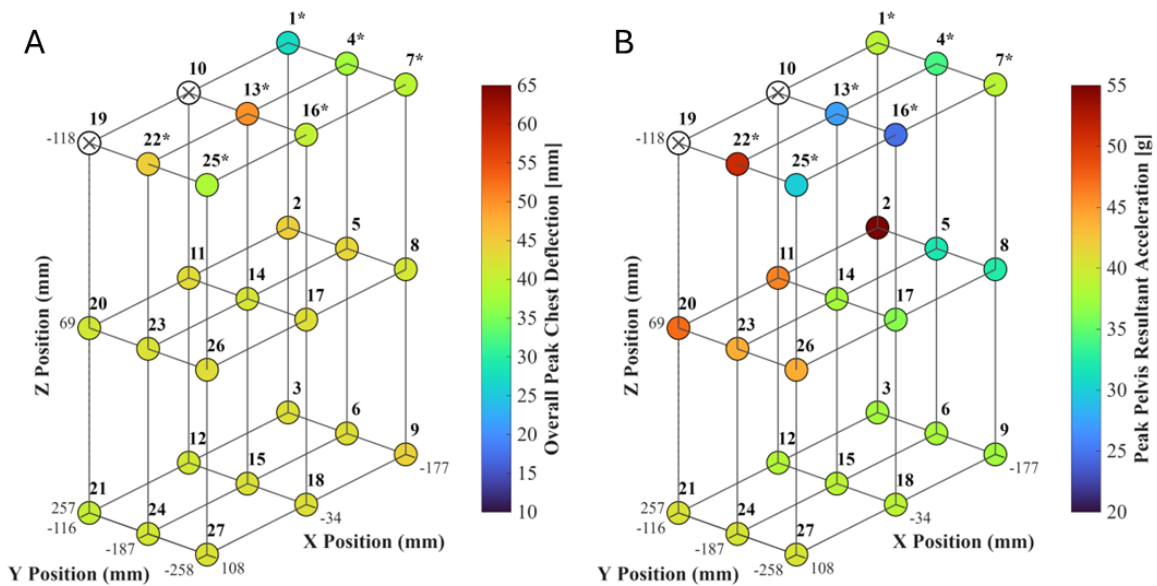


Figure 4.25: THOR-5F responses for outboard lap belt anchor positions in the Semi-Rigid Bench. (A) Overall peak chest deflection for simulated outboard lap belt anchor locations (B) Peak pelvis resultant accelerations for simulated outboard lap belt anchor locations. Marker numbers correspond to the DOE position. *Indicates configurations removed from the analysis; ⊗ indicates configurations that were not simulated.

The right-side chest sensors (Figure 4.26B,D) generally exhibited higher deflections than the left side sensors (Figure 4.26A,C), with the right upper and lower sensors each recording the overall peak chest deflections in 9 of the 18 analyzed configurations. The right lower and both left-side chest sensors showed lower deflections at rearward (-X) lap belt anchor locations. The right upper chest sensor exhibited an opposite trend, with rearward (-X) anchor locations producing slightly higher chest deflections.

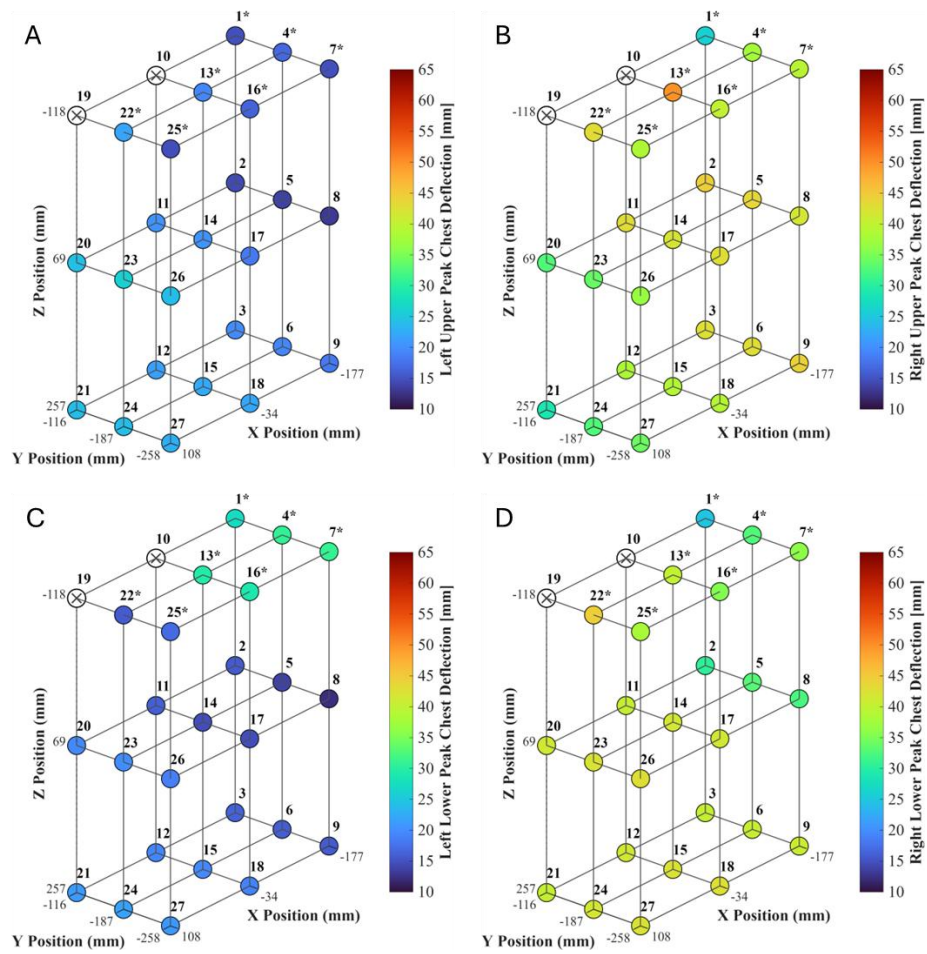


Figure 4.26: THOR-5F peak chest sensor deflections for lap belt anchor locations in the Semi-Rigid Bench: (A) left upper sensor, (B) right upper sensor, (C) left lower sensor, and (D) right lower sensor. Marker numbers correspond to DOE position. *Indicates configurations removed from analysis.

4.4.5.1 Lap Belt Anchor Design of Experiment (DOE) Effects Calculations for Peak Chest Deflection

The regression model for the overall peak chest deflection measurements using the Semi-Rigid Bench model had an R^2 value of 0.92. The effects analysis identified multiple significant terms, including the main effects of X and Z, as well as the XY, YZ and XYZ interactions (Figure 4.27). The X-direction effect was negative; moving the lap belt anchors from minimum to maximum X-position (e.g. Position 8 to 26) resulted in an overall peak chest deflection decrease of 1.4 mm on average. The Z-direction also showed a negative effect, with moving the lap belt anchors downwards from minimum to maximum (e.g. Position 25 to 27) reducing chest deflections by an average of 0.9 mm.

Among the interaction terms, the negative YZ interaction effect had the largest effect magnitude, indicating that the influence of the vertical (Z) anchor position was dependent on the lateral (Y) position of the lap belt anchors. When the anchors were moved to Y_{MAX} , the Z effect in reducing chest deflection was amplified by approximately 2.11 mm compared to the Z effect at Y_{MIN} . The XY interaction was also large and significant, as the magnitude of the X effect was 1.2 mm higher at Y_{MAX} compared to Y_{MIN} . A significant three-way interaction effect (1.82 mm) was observed, showing that the combined influence of the three directions contributed to the differences in peak chest deflections beyond the main effects and the two-way interaction effects.

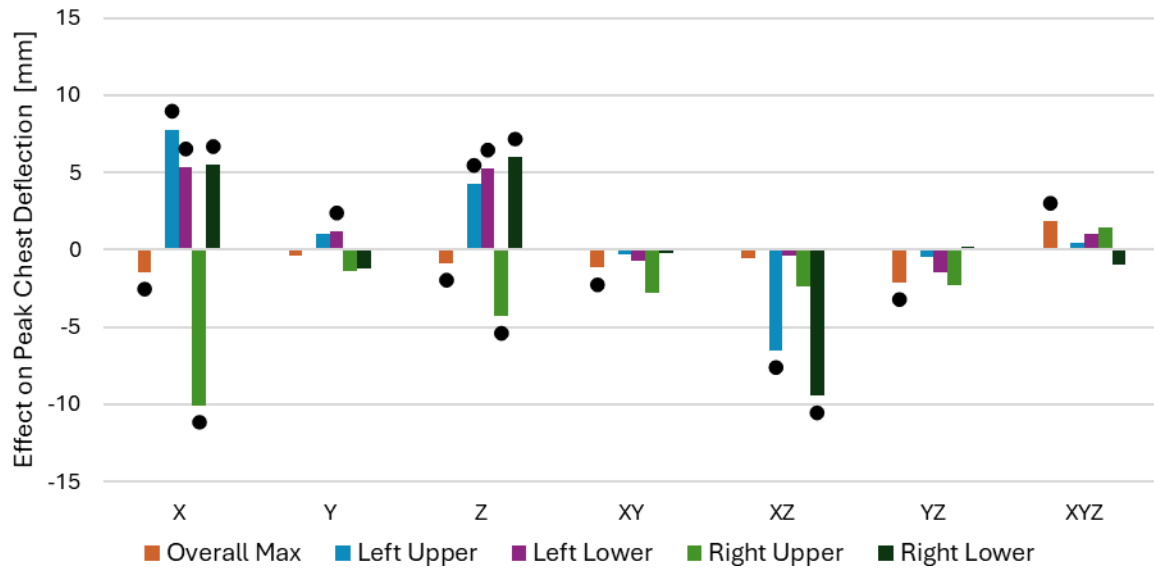


Figure 4.27: Main and interaction effects on peak chest deflection for the Semi-Rigid Bench lap belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).

The regression models for the individual chest sensors ranged from 0.84 to 0.97, with the lowest R^2 value observed at the lower right chest. For the main effects, the individual sensors generally exhibited larger effect magnitudes relative to the overall peak chest deflection, although the sign varied depending on the sensor location. The upper right chest had the same sign as the overall deflection, whereas the other three sensors exhibited effects with opposite signs. The Y-direction (lateral) main effect was only significant for the lower left sensor. For the interaction effects, a limited number of significant terms were observed at the individual sensor level. The XZ interaction was significant for the upper left and the lower right sensors.

4.4.5.2 Lap Belt Anchor Design of Experiment (DOE) Effects Calculations for Peak Pelvis Resultant Acceleration

The linear regression model fit the peak pelvis resultant accelerations on the Semi-Rigid Bench model with an R^2 value of 0.76. The effects analysis for the Semi-Rigid Bench lap DOE identified the Y-direction main effect and the YZ interaction effect as significant (Figure 4.28). The effect of Y was negative; as the anchor was moved laterally (+Y) from minimum to maximum (e.g. Position 2 to 8), the peak acceleration decreased by an average of 6 g. The YZ interaction effect was positive, indicating that the effect of lateral (Y) anchor position was dependent on the vertical (Z) position. When the anchors were moved downward (+Z) from Z_{MIN} to Z_{MAX} , the magnitude of the Y-effect on peak pelvis resultant acceleration was reduced by 12.1 g.

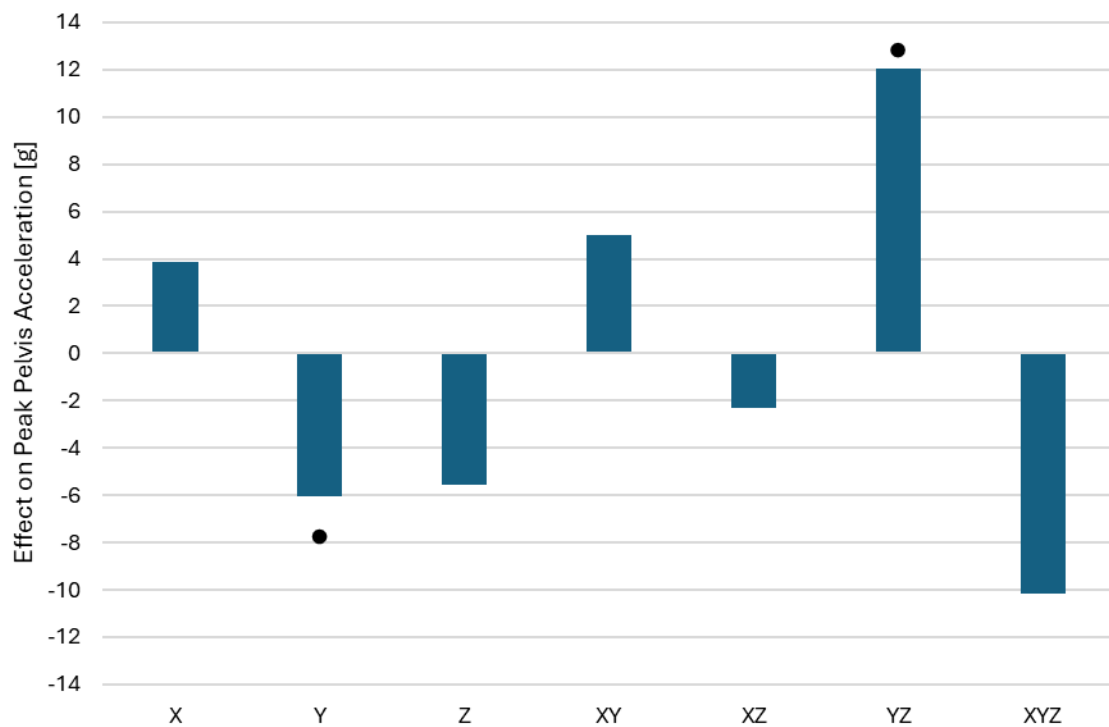


Figure 4.28: Main and interaction effects on peak pelvis resultant acceleration for the Semi-Rigid Bench lap belt anchor DOE. Effects marked with a dot (•) are statistically significant ($p < 0.05$).

Chapter 5

Discussion

5.1 Validation of THOR-5F FE Model in Rigid Bench

The design and development of the THOR-5F physical ATD is ongoing, with injury criteria and IARVs still under development (National Highway Traffic Safety Administration, 2026b). The physical ATD used in the current study was Standard Build Level B (SBL-B) and has since been updated to Standard Build Level C (SBL-C) as of October 2025 (Cellbond, n.d.; Humanetics, 2024). The THOR-5F FE ATD model was a beta version (version D00.02) from February 20, 2023. The validation provided with the model included component-level validation for the head, neck, abdomen and lower extremities; however, sled validation was not included. The validation of the THOR-5F FE model in the Rigid Bench showed an ISO 18571 grade of “Good” for belt loads and pelvis accelerations, demonstrating close alignment between the THOR-5F FE model and physical ATD in bench and belt interactions. The chest deflections showed moderate agreement, as the right-side deflections and left upper deflection were rated as “Fair”; however, the left lower deflection was rated as “Poor”, which is attributed to kinematic differences between the model and the test.

While the kinematics were generally consistent between the model and physical ATD, differences were observed at maximum excursion (105 ms). In the physical test, the shoulder belt migrated laterally outwards on the shoulder of the THOR-5F ATD, and the torso rotated about the Z-axis with minimal pelvis Z-rotation (Figure 5.1A,B), which was not replicated in the model (Figure 5.1C,D). The lack of torso Z-rotation in the model resulted in abrupt loading at the end of the excursion, leading to higher head X-accelerations and decreased left lower chest deflections.

Additionally, despite the close agreement between lap belt loading and the pelvis resultant accelerations and the “Fair” ISO 18571 grade, the iliac load cells recorded approximately 1 kN lower peak forces than the physical testing. The model response was investigated and was observed to have minimal Y-rotation of the pelvis in the simulation (Figure 5.1D) compared to the physical test (Figure 5.1B). The FE model instead showed greater forward pelvis translation, causing the belt to translate below the iliac load cell and reducing engagement with the sensor.

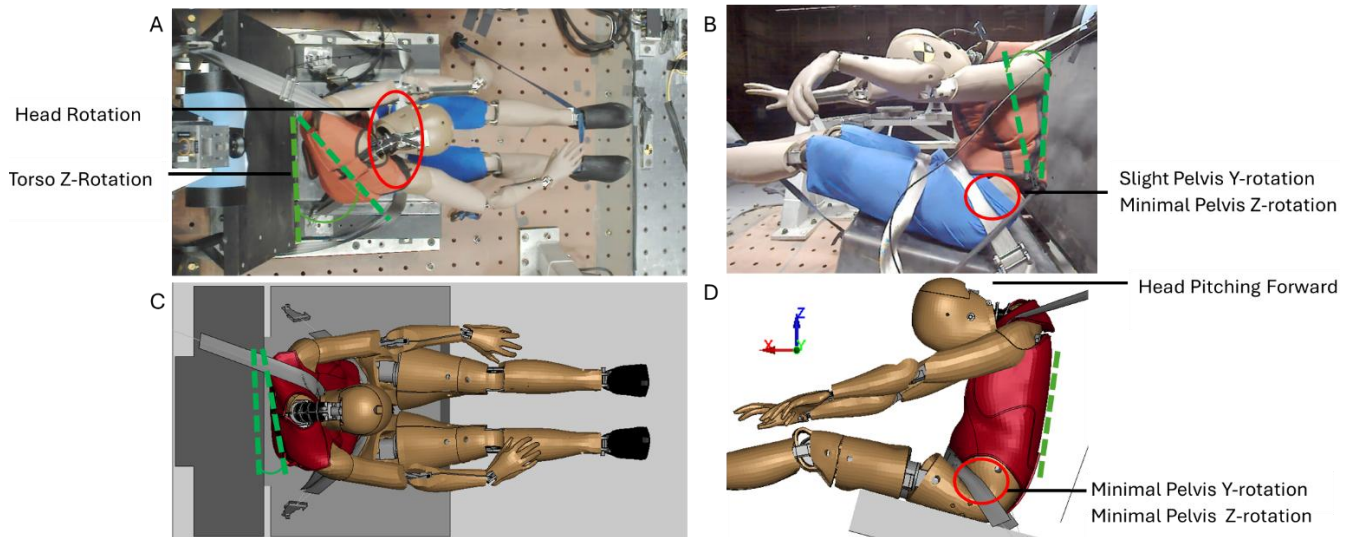


Figure 5.1: THOR-5F at 105 ms. (A) Top view of physical ATD. (B) Side view of physical ATD. (C) Top view of FE model. (D) Side view of FE model. The reduced torso rotation of the THOR-5F FE model, shown by the angle indicator (dashed green lines), resulted in higher head accelerations at the end of the crash pulse.

The reduced torso Z-rotation and pelvis Y-rotation in the FE model were investigated by reviewing the baseline model and in a separate study with varying belt routing and occupant initial position. No substantial changes to the kinematics were observed when varying belt routing and initial position. It is hypothesized that the reduced rotation in the model may stem from a higher rotation stiffness, potentially in the lumbar spine in the model, and is a topic for further investigation. The commercial ATD model used in this study was encrypted and did not allow for changes to the model. However, the close match between the FE model and the physical ATD in the regions of interest (belt loads, thoracic deflection, and pelvis X-acceleration) and at relevant earlier time points indicated that the model could provide guidance on the effect of D-ring positions.

5.2 Assessment of THOR-5F FE Model in Semi-Rigid Bench

The initial assessment of the THOR-5F FE on the Semi-Rigid Bench with immediate retractor locking revealed substantial discrepancies in shoulder and lap belt loads compared with the physical tests by Richard et al. (2022). The belt loads from the simulation reached the peak value approximately 10 ms earlier compared to the physical tests. The belt loads were approximately 0.5 kN (15%) lower in the model. The resultant accelerations in the simulation also peaked 10 ms (13%) earlier than in the physical test and underpredicted the peak magnitude by more than 5 g (24%). The earlier timing of the belt load peaks, along with the lower belt loads and pelvis accelerations, indicated that the belts in the FE model engaged significantly earlier than in the physical test, which was interpreted as a difference in restraint system compliance between the physical test setup and the FE model.

To account for the compliance not captured by the initial FE belt model, a series of simulations with 5 mm pullout increments added to the retractor were conducted, with the identified pullout amount based on how closely the lap belt and shoulder belt peaks matched the physical test. From the simulations, 30 mm was identified as the optimal added pullout amount, and after implementing the pullout, the timing and magnitude of the shoulder belt and lap belt load peaks more closely matched those observed in the physical test by Richard et al. (2022). The agreement in the peak magnitudes and timings of the refined belt system loads was further quantified by ISO 18571 “Good” grades, corresponding to numerical ratings above 0.80 and indicating strong agreement between the calibrated simulation and the physical test.

The 30 mm addition to pullout represented physical restraint effects that were not captured by the FE model, such as belt-path settling around the webbing guides, hardware compliance, retractor webbing-on-spool tightening, and seat deformation. The 30 mm value is within reported real-world seat belt slack ranges, including observed slack ranging from 25–50 mm in some vehicle configurations (Cicccone & Wells, 1988), with slack coming from sources like seat belt fit or clothing (Soni et al., 2023). Soni et al. (2023), citing Bacon (1989), also stated that webbing tightening in a locked retractor can contribute to additional slack. Additionally, many vehicle seat belt assemblies tested by Wietholter & Wetli (2021) measured more than 100 mm of seat belt elongation from buckle

stalk movement. Considering the reported values in literature, the 30 mm of additional compliance was considered reasonable.

The THOR-5F FE head resultant acceleration also closely matched the physical tests with a “Good” ISO 18571 grade, as in both cases, the head motion was primarily a forward Y-rotation. Although peak chest deflection timing was similar, the rise in the FE model deflection occurred later, with a steeper slope than the physical test. The timing difference was most distinctly displayed in the right lower chest deflection, as the loading of the chest only occurred when the shoulder belt engaged the thorax of the THOR-5F FE.

The gradual rise in the physical test chest deflection was consistent with distributed seat belt system compliance not captured in the FE model, such as friction in the webbing guides, which led to a gradual shoulder belt loading. Friction within the seat belt system had also been identified in another model, the ENOP sled model, which noted that guide and tongue friction coefficients for the seat belts can vary broadly and should be measured in testing (Valdano, 2023). However, the chest deflection peaks from the FE model for the left upper, left lower, and right upper chest areas tracked well with the physical test data, showing similar timing and slightly lower magnitudes, corresponding to “Fair” ISO 18571 ratings. The reduced chest deflection in the FE model at the end of the crash pulse, not observed in the physical test measurements, coincided with the shoulder belt sliding upwards on the chest, as the THOR-5F FE did not pitch into the belt to the same extent as in the physical test. However, the difference in response occurred after the primary belt loading phase and maximum ATD excursion, during the rebound phase, and was therefore outside the response interval of interest for model assessment.

The timing and magnitude of the resultant pelvis acceleration showed close correspondence with the physical tests, indicated by the “Good” ISO 18571 rating. Despite the strong agreement in pelvis acceleration and lap belt loading, the iliac forces measured in the THOR-5F FE differed from the physical test measurements, characterized by “Poor” ISO 18571 cross-correlation ratings. Visual inspection of the simulation displayed the lap belt engaging the location of the iliac sensors, so the discrepancies between the physical test and simulation iliac forces were investigated further by observing the load cell behavior during the pulse. The two load cell components showed relative sliding under belt-induced compression (Figure 5.2). Since the iliac force was recorded from the axial

force in the beam between the components, the relative sliding altered the load path through the beam, resulting in reduced measured compressive forces and increased measured tensile forces.

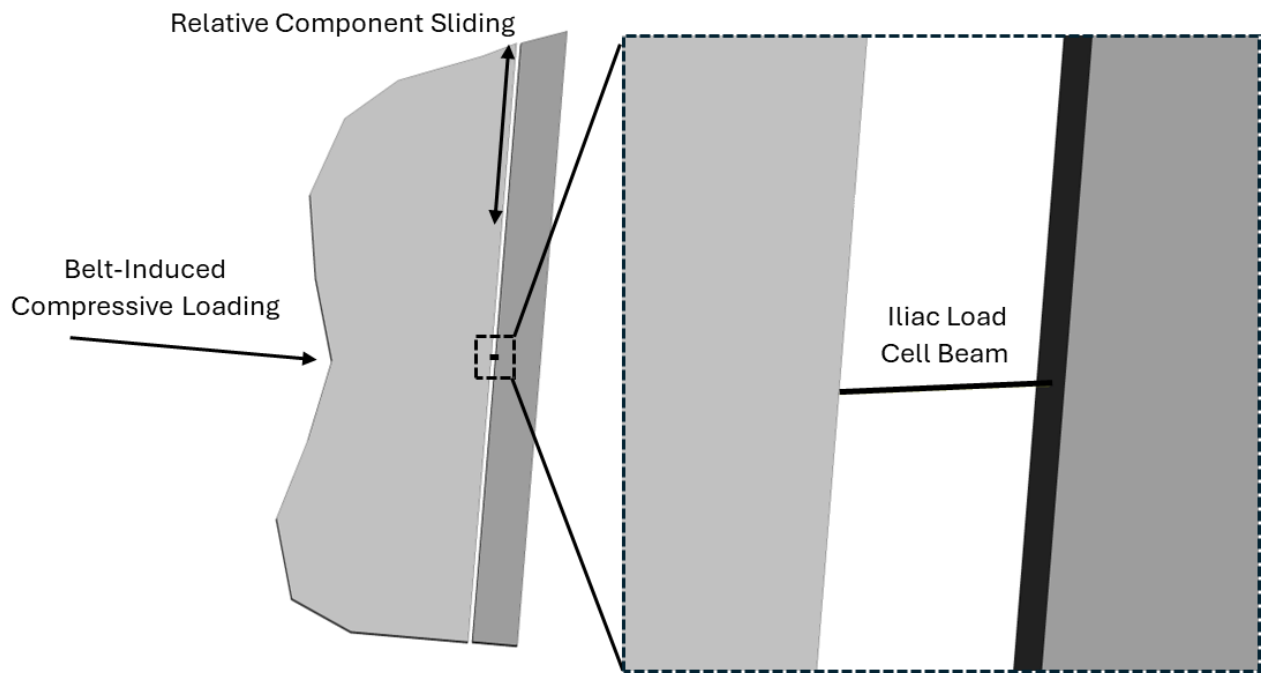


Figure 5.2: THOR 5F FE iliac load cell with belt-induced compressive loading and sliding between front and back components. The iliac load cell beam records the beam axial force between the components as the iliac force.

Overall, the THOR-5F FE in the Semi-Rigid Bench model successfully captured the interaction with the bench and belts in the physical test, as well as the key kinematic responses of the physical ATD, supporting the model's use in the parametric belt anchor analysis.

5.2.1 Implementation of 3-Point Belt System from the Rigid Bench

The implementation of the 3-point belt configuration was undertaken so that results from the Semi-Rigid Bench could be compared to the Rigid Bench. Changing from a 4-point, 2-belt configuration to a 3-point belt configuration in the Semi-Rigid Bench resulted in lower and earlier lap and shoulder belt load peaks, caused by earlier belt engagement, load sharing between belts through the slipring and the Rigid Bench retractor providing less resistance to pullout. The combined effects of the 3-point belt configuration and the addition of the Rigid Bench retractor also resulted in a lower peak pelvis resultant acceleration, while the head resultant acceleration displayed a delayed rise to a lower peak. Chest deflections for the 3-point belt configuration had lower magnitudes due to lower shoulder belt load but displayed similar timing.

Additionally, the slipring of the 3-point belt was located lower than the right side shoulder belt anchor of the 4-point, 2-belt system. As a result of the different anchor location, the shoulder belt angle shifted downward and farther from the upper right chest deflection sensor (Figure 5.3A), further lowering the peak chest deflection measurements. Despite the differences in chest deflection magnitude, the overall occupant kinematics were similar between the 4-point, 2-belt configuration and the 3-point belt configuration (Figure 5.3B,C). Accounting for the routing difference and the retractor allowing progressive belt pullout rather than immediate locking, the Semi-Rigid Bench model with the 3-point belt configuration produced lower response magnitudes. Given the need for consistency with the Rigid Bench simulations, and because the 3-point belt system was more representative of contemporary rear seat belts, the Semi-Rigid Bench with the 3-point belt system was used for the parametric belt anchor location study.

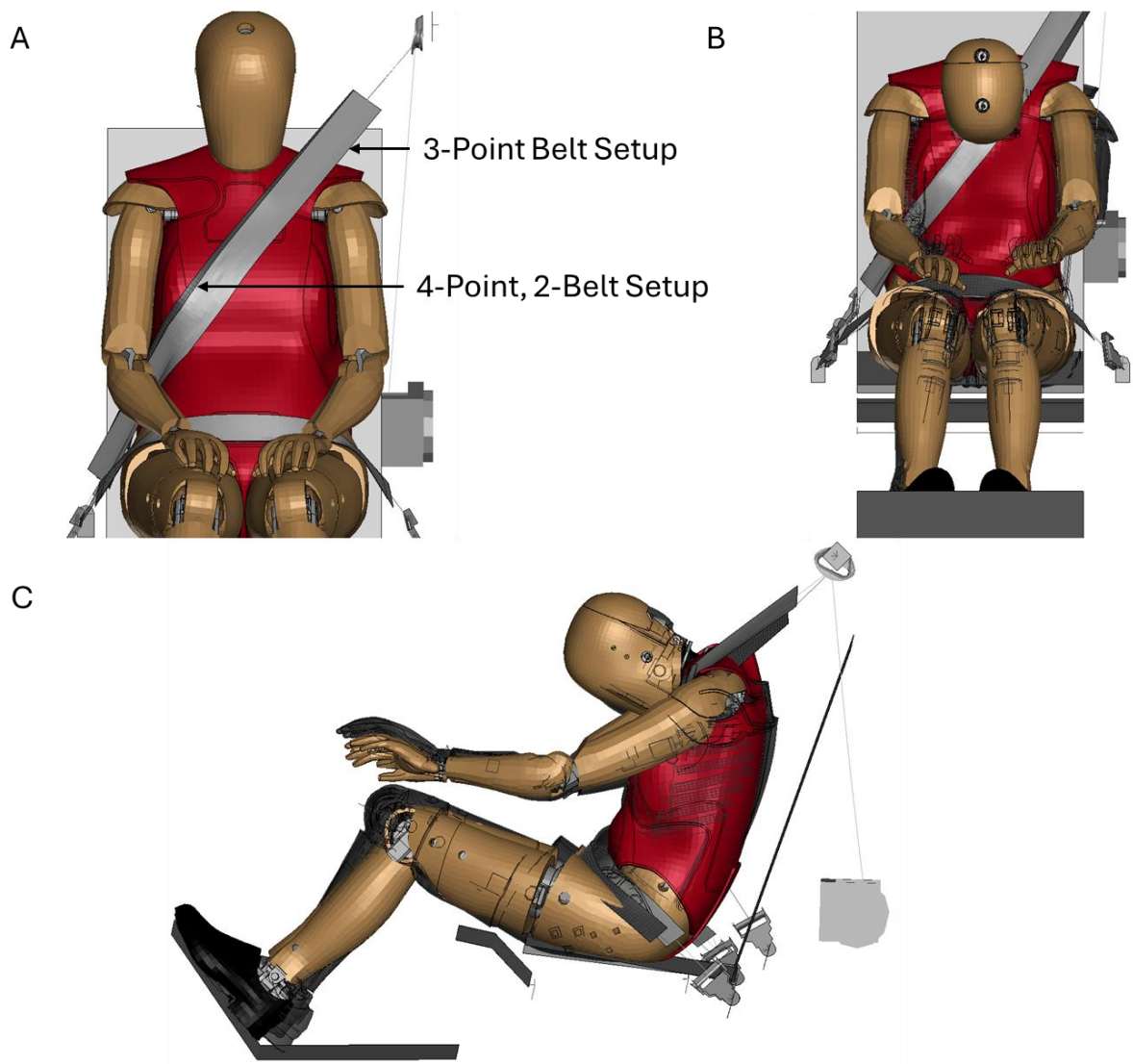


Figure 5.3: Comparison of the 4-point, 2-belt configuration and the 3-point belt configuration in the Semi-Rigid Bench. (A) Lower shoulder belt fit in the 3-point belt configuration (light gray) compared to the 4-point, 2-belt configuration (black). (B) Front view and (C) side view of the 4-point, 2-belt kinematics to the 3-point belt.

5.3 Parametric Study of Belt Anchor Location and THOR-5F FE Peak Chest Deflection

5.3.1 Rigid Bench Shoulder Belt Anchor DOE

For the Rigid Bench, when moving the shoulder belt anchor, the overall peak chest deflection ranged from 42 mm to 63 mm. Based on the preliminary injury risk curves from Lee and Craig (2021), the peak deflections correspond to AIS 3+ injury risks of 89% to 99%. The lowest overall peak chest deflection occurred when the shoulder belt anchor was furthest rearward, inboard and highest in the vertical direction. The shoulder belt anchor location caused the belt to fit more medially across the upper thorax region (Figure 5.4A,B), increasing load transfer through the shoulder region compared to the midpoint configuration (Figure 5.4C,D). The increased loading on the shoulder resulted in less load being transferred to the chest, resulting in lower chest deflections. However, the belt path was closer to the neck, causing potential neck-shoulder belt interaction.

The shoulder belt anchor location with the highest peak chest deflection occurred at the most forward, outboard, and highest vertical position. The shoulder belt anchor location caused the belt to route closer to the edge of the shoulder and delayed belt engagement with the THOR-5F torso, resulting in greater forward torso excursion and higher overall peak chest deflection (Figure 5.4E,F) compared to the midpoint configuration.

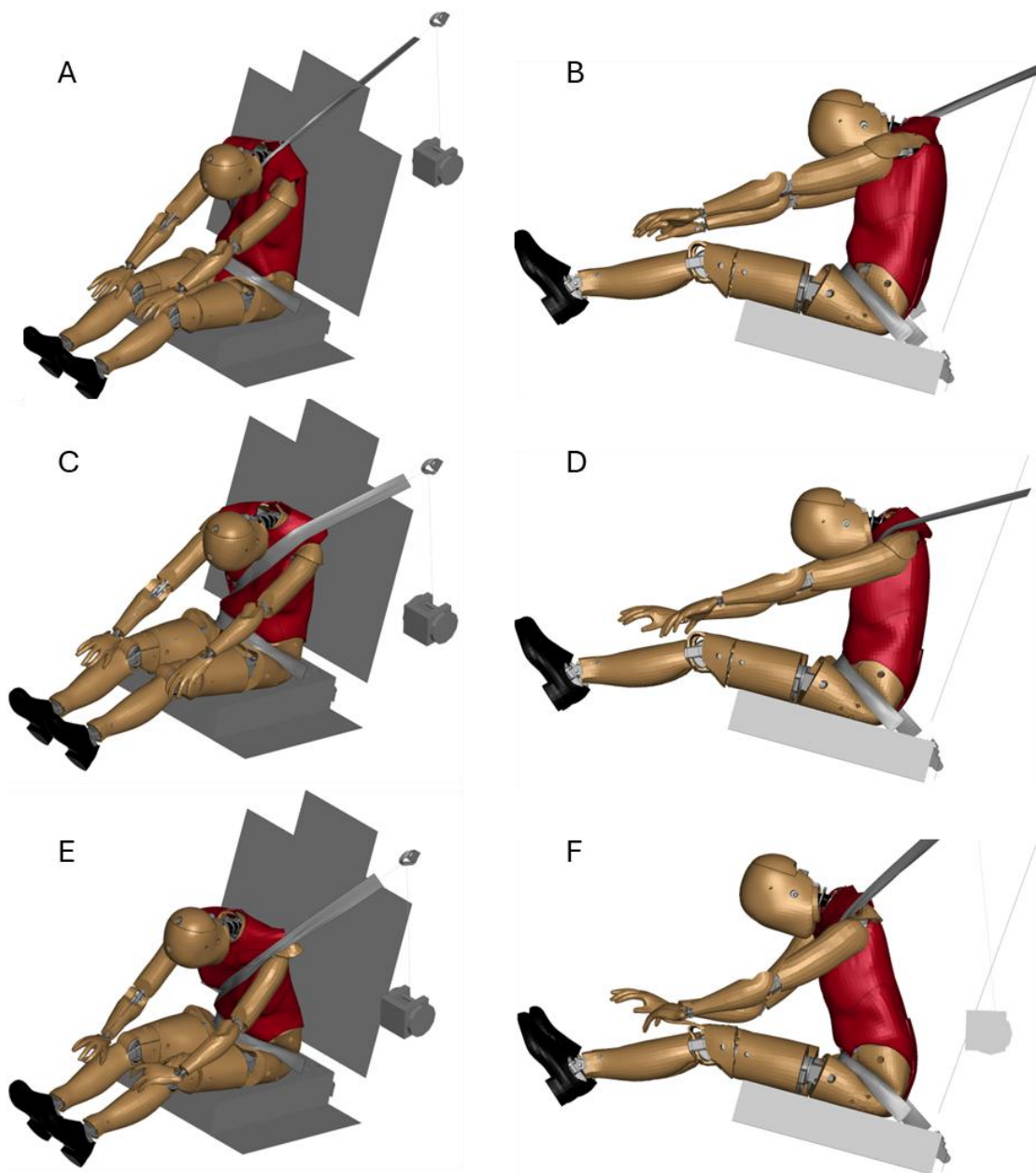


Figure 5.4: Images of the THOR-5F in the Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum chest deflection shoulder belt anchor configuration, (C,D) midpoint shoulder belt anchor configuration and (E,F) maximum chest deflection shoulder belt anchor configuration. B,D,F are the side view images of each configuration.

The shoulder belt anchor locations corresponding to the maximum and minimum chest deflections further demonstrate the main effects of shoulder belt anchor position on the overall peak chest deflection of the THOR-5F FE on the Rigid Bench. The X-position of the shoulder belt anchor was the dominant contributor to the overall peak chest deflection, followed by the Y-position. The positive X-position effect indicated that moving the shoulder belt anchor forward increased the overall peak chest deflection. Forward shoulder belt anchor positions caused the shoulder belt to engage the THOR-5F FE later in the crash pulse, after greater forward excursion, resulting in greater loading of the chest.

The Y-position had a negative effect; moving the anchor inboard reduced the overall peak chest deflection. Inboard shoulder belt anchor positions increased shoulder contact with the belt and reduced loading of the chest. The Z-position had the next largest effect and was also negative, indicating that lowering the shoulder belt anchor reduced the overall peak chest deflection. Lower shoulder belt anchor positions enabled the belt to wrap around the shoulder more effectively, resulting in better load path transfer to the shoulder.

Although the XZ interaction was smaller in magnitude than the main effects, it was also significant. The negative XZ interaction indicated that lowering the shoulder belt anchor reduced the overall peak chest deflection to a greater extent at forward X-positions than at rearward X-positions. The interaction suggests that the shoulder belt fit around the shoulder became increasingly important at forward shoulder belt anchor positions, where the greater forward excursion increased the sensitivity of the response to anchor height. At the forward anchor locations, the lower shoulder belt anchor likely helps to restrain the torso earlier and prevent excessive excursions or high belt loads leading to increased deflection. However, this effect was weaker than the main effects and should be interpreted as secondary to the main effects.

5.3.2 Rigid Bench Lap Belt Anchor DOE

The lap belt anchor positioning for the Rigid Bench resulted in the overall peak chest deflections ranging from 44.3 mm to 53.1 mm, which corresponded to an AIS 3+ injury risk range of 92% to 98% based on the preliminary curves from Lee and Craig (2021). The lowest deflection occurred when the lap belt anchors were at the highest vertical position, largest lateral distance, and forward fore-aft position. The low deflection may be a result of more near-submarining behavior (Figure 5.5A,B), with the pelvis slightly sliding under the lap belt and the torso leaning rearward compared to the midpoint belt configuration (Figure 5.5C,D). The highest peak chest deflection occurred at the forward lap belt anchor location, midpoint height and midpoint lateral distance. The position of the lap belt anchor altered the lap belt angle, resulting in later engagement with the THOR-5F FE pelvis and a more downward-directed restraint force, resulting in a more forward pitch (Figure 5.5E,F). The forward pitch, combined with the shoulder belt migrating to the edge of the shoulder, increased load concentration on the chest, thereby increasing chest deflections. The maximum and minimum chest deflection configurations both having forward (X) anchors indicated that the kinematic mode (forward pitch versus near-submarining) was a contributor to the chest deflection.

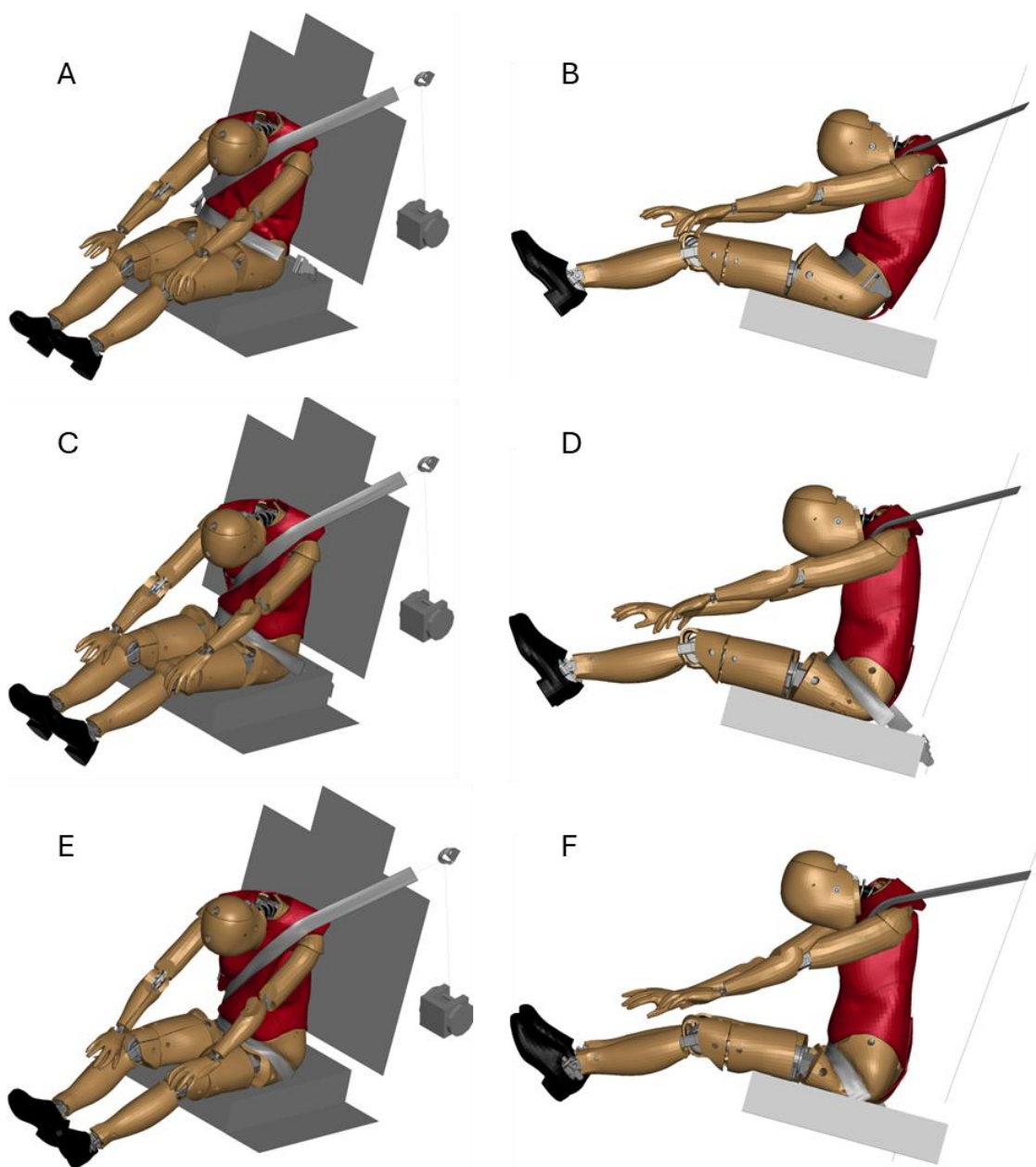


Figure 5.5: Images of the THOR-5F in the Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum deflection lap belt anchor configuration, (C,D) midpoint deflection lap belt anchor configuration and (E,F) maximum deflection lap belt anchor configuration. B,D,F are the side view images of each configuration.

The kinematics observed at the maximum and minimum chest deflection did not fully align with the linear model effects trends, indicating that multiple kinematic modes influenced the chest deflection response. While the maximum and minimum deflections occurred at the midpoint, the DOE analysis identified the X-position and Z-position of the lap belt anchors as the dominant effects. The X-position was the largest and was positive; moving the lap belt anchor forward resulted in higher overall peak chest deflections. The forward anchor positions resulted in delayed restraint of the THOR-5F until later in the crash pulse, when more excursion occurred and increased reliance on the shoulder belt loading, resulting in more loading of the chest. The Z-position was also positive, moving the anchors downward increased chest deflection, as the restraint of the pelvis was more directed towards downwards motion than forwards motion, causing a forward pitch and increasing shoulder belt loading on the chest.

5.3.3 Semi-Rigid Bench Shoulder Belt Anchor DOE

The shoulder belt anchor positions in the Semi-Rigid Bench resulted in overall peak chest deflections ranging from approximately 35 mm to 50 mm, corresponding to AIS 3+ injury risks of 70% to 96% based on the data from Lee and Craig (2021). The lowest peak chest deflection occurred when the shoulder belt anchor was the furthest rearward, inboard, and midpoint height position. At this shoulder belt anchor location, the shoulder belt was positioned on the shoulder closer to the neck (Figure 5.6A,B), increasing engagement over the clavicular region and reducing loading of the thorax compared to the midpoint configuration (Figure 5.6C,D). Additionally, the rotation of the seat pan caused the shoulder belt path to shift upward on the torso from the chest sensors, contributing to lower chest deflections in the minimum chest deflection configuration. The maximum overall peak chest deflection was observed when the shoulder belt anchor was positioned at the most forward, midpoint lateral and highest location, in which the THOR-5F FE torso translated farther forward before belt engagement prevented further motion (Figure 5.6E,F). The seat pan deflection further lowered the THOR-5F, moving the shoulder belt closer to the IR-TRACC sensors and increasing chest deflection measurements.

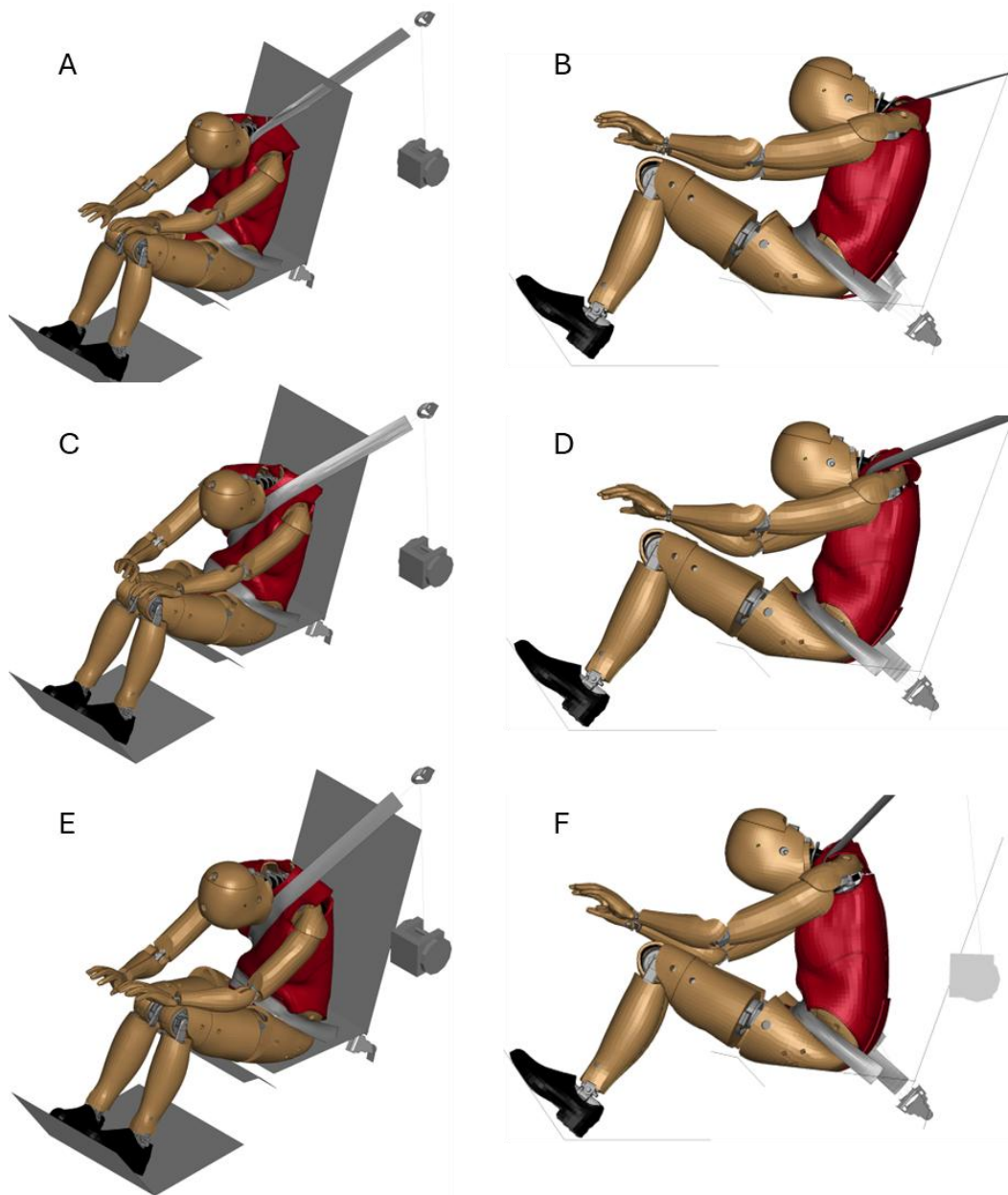


Figure 5.6: Images of the THOR-5F in the Semi-Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum deflection shoulder belt anchor configuration, (C,D) midpoint deflection shoulder belt anchor configuration and (E,F) maximum deflection shoulder belt anchor configuration. B,D,F are the side view images of each configuration.

The effects of shoulder belt anchor location reflected trends observed at the configurations with the maximum and minimum overall peak chest deflection. The X, Y, and Z-position main effects of the shoulder belt anchor were significant. The largest effects were the X and Y-positions, with the positive X effect slightly larger in magnitude than the negative Y effect. The positive X effect indicated that moving the shoulder belt anchor forward increased chest deflections, as it enabled the THOR-5F FE to translate farther forward before shoulder belt engagement, resulting in higher chest loading. The negative Y effect showed that moving the shoulder belt anchor inboard reduced the overall peak chest deflection by improving shoulder belt contact with the THOR-5F FE shoulder, allowing a greater proportion of the restraint load to be transferred through the shoulder rather than the chest. The Z effect was much smaller and negative, indicating that moving the shoulder belt anchor downward reduced the peak chest deflection. The lower anchor position improved belt wrapping about the shoulder, also allowing for more load to be transferred through the shoulder, thereby reducing chest loading and peak deflection.

5.3.4 Semi-Rigid Bench Lap Belt Anchor DOE

Compared with the shoulder belt anchor positions, the lap belt anchor positions had a smaller, but still notable, influence on chest deflection. The overall peak chest deflections ranged from 40 mm to 45 mm, which corresponded to AIS 3+ injury risks ranging from 85% to 92% based on the data from Lee and Craig (2021).

The lowest chest deflection occurred with the lap belt anchors located at the most forward position, smallest lateral distance and lowest vertical position. In this configuration, the lap belt engaged the pelvis and restrained the occupant motion primarily to forward translation, with minimal torso pitch (Figure 5.7A,B) compared to the midpoint configuration (Figure 5.7C,D). Although the increased translation and downward pelvis motion caused the shoulder belt to migrate outboard to the edge of the THOR-5F FE, the lap belt contributed more to occupant restraint, resulting in reduced shoulder belt loading and lower thoracic compression. The highest chest deflection was observed when the lap belt anchors were at the rearmost position, smallest lateral distance, and midpoint vertical position. The lap belt position restrained further pelvis translation, causing the torso to pitch forward into the shoulder belt (Figure 5.7E,F), which increased loading of the chest by the shoulder belt and resulted in greater deflection.

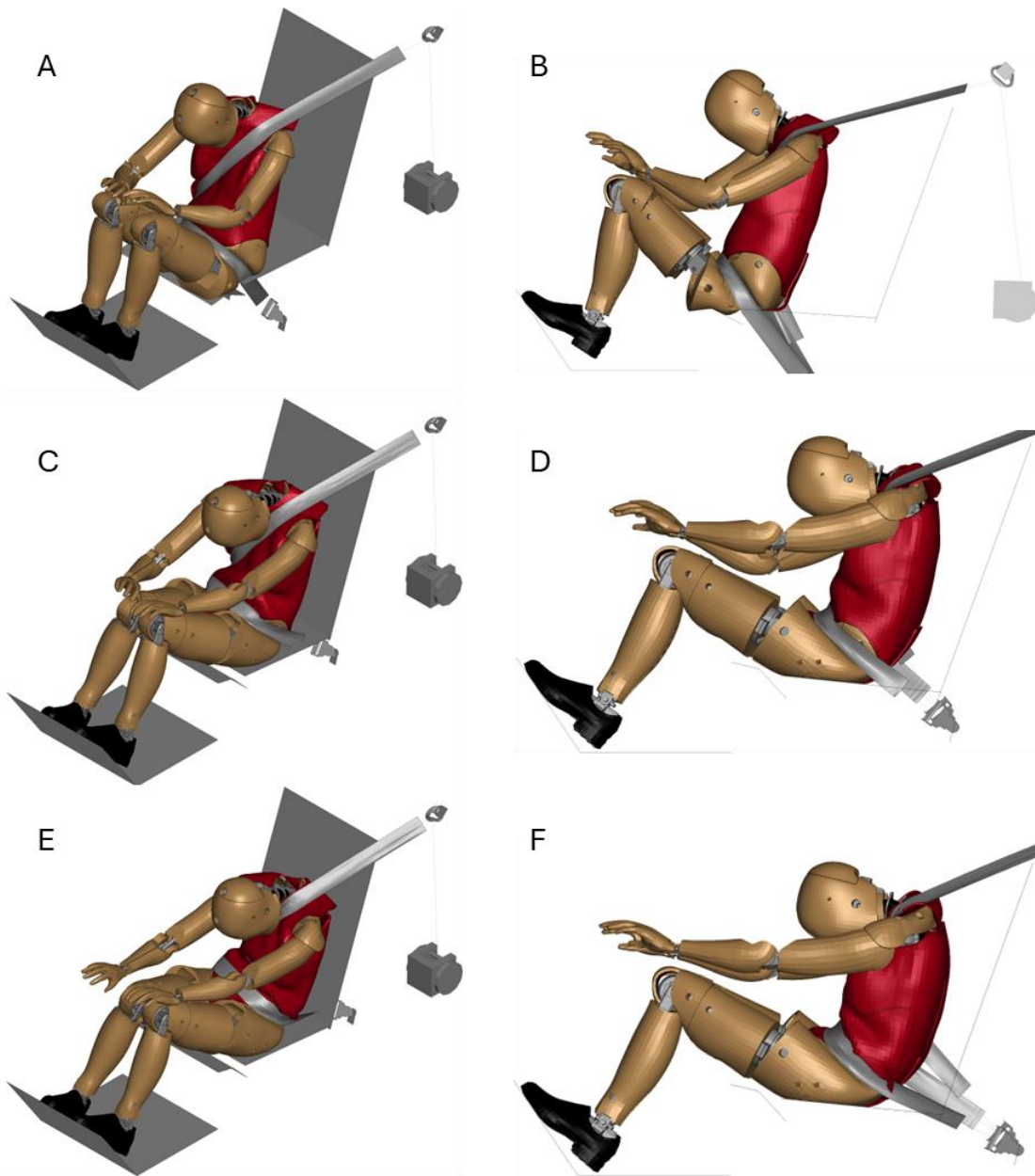


Figure 5.7: Images of the THOR-5F in the Semi-Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum deflection lap belt anchor configuration, (C,D) midpoint lap belt anchor configuration, and (E,F) maximum deflection lap belt anchor configuration. B,D,F are the side view images of each configuration.

The effects analysis of lap belt anchor locations on the overall peak chest deflection indicated a more complex relationship than that observed for the shoulder belt anchor DOE, with smaller effect magnitudes and more significant interactions. Generally, the peak chest deflection was minimized when the lap belt controlled pelvis motion without turning the pelvis into a pivot point for torso pitch. The significant effects on overall peak chest deflection were the X- and Z-position main effects and the XY, YZ, and XYZ interactions.

The largest effect was the negative YZ effect; at the downward Z-positions, the increased lateral spacing reduced the lap belt wrap around the pelvis, thereby reducing excessive pelvis constraint while allowing controlled motion, which resulted in less forward pitch and chest loading. The next largest effect was the positive XYZ effect, which indicated that the combined influence of the three anchor position components was not equal to the sum of their independent effects, with certain combinations promoting controlled translation and others promoting greater torso pitch. For example, at a rearward and downward lap belt anchor location, increased lateral spacing created a belt geometry that did not over-constrain the pelvis, which resulted in reduced torso pitch and lower peak chest deflection.

The next largest effect was the negative X effect; as the anchors were moved forward, the lap belt allowed more forward translation, which reduced forward pitch into the shoulder belt and decreased the overall peak chest deflection on average. The negative XY effect suggested that, as the belt anchors were moved forward, greater lateral spacing of the anchors would reduce the lap belt wrap around the pelvis and therefore reduce forward pitch. The Z effect was the smallest significant effect; moving the anchors downward reduced chest deflection through improved restraint alignment with downward pelvis motion. All lap belt anchor position components influenced the lap belt angle and the belt's ability to restrain the pelvis without excessively restricting pelvis motion and increasing torso pitch.

5.3.5 Chest Deflection Regions

The individual chest sensor deflections generally reflected similar trends to the overall peak chest deflection for all DOEs, while also providing additional insight into the locations of chest loading. For the Rigid Bench, overall peak deflections were most often located at the right lower chest sensor, indicating a consistent belt load path with loading concentrated in the right lower region. As a result

of the consistent belt loading, the right lower chest sensor exhibited similar trends and magnitudes to the overall peak chest deflection for the DOE. The right upper chest sensor generally exhibited opposite effects, as the shoulder belt path was usually closer to the sensor region but farther away in configurations that caused higher overall peak chest deflections. Different significant interaction terms further indicated that the anchor location altered the local belt loading patterns.

For the Semi-Rigid Bench, the peak overall chest deflections were located at a mix of upper and lower chest regions. The inconsistent locations of peak deflections were also reflected by differences in significant main and interaction effects. Higher peak responses recorded by the upper chest regions were likely due to the deflecting seat pan, which resulted in the THOR-5F translating downward, shifting the shoulder belt loading region upward. Additionally, the belt paths were different due to the downward and forward motion of the THOR-5F, resulting in different chest regions recording the overall peak chest deflections.

Across all DOEs, the individual sensors exhibited larger effect magnitudes than the overall chest deflection. The larger effect magnitudes occurred because each sensor captured localized deflections, which were dependent on the routing of the belt and anchor configuration. In comparison, the overall chest response represented the maximum response of the four sensors, which masked the localized sensitivity.

5.3.6 Chest Deflection and Injury Risk

All anchor locations for the Rigid and Semi-Rigid Bench recorded overall peak chest deflections corresponding to AIS 3+ injury risks above 50% according to the curves by Lee and Craig (2021) (Figure 5.8). The Rigid Bench anchor positions all resulted in AIS 3+ injury risks exceeding 80%. The lower chest deflections in the Semi-Rigid Bench configurations resulted in lower AIS 3+ injury risks, with some shoulder belt anchor configurations having less than 80% AIS 3+ injury risk. Rearward and inboard anchor locations corresponded to the lower risks, while the outboard and forward anchor locations corresponded to higher risks. Although the injury risks were generally high, the range in injury risk across shoulder belt anchor positions indicates the potential to combine seat belt anchor positioning with technologies such as load limiters to further reduce chest deflections and injury risk.

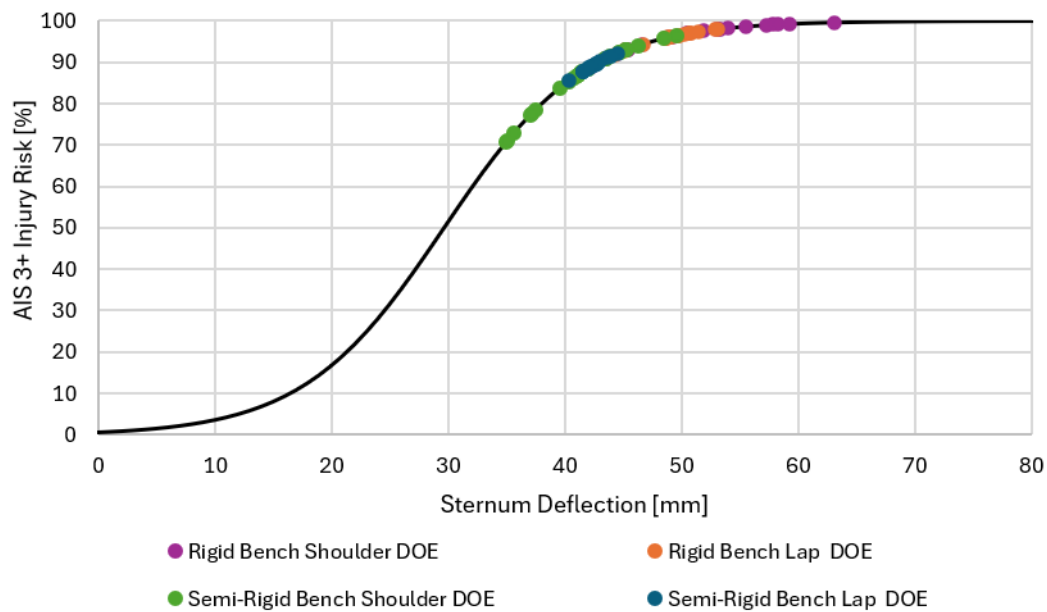


Figure 5.8: Preliminary AIS 3+ injury risk curves with seat belt anchor DOE configurations.

5.4 Belt Anchor Location and Peak Resultant Pelvis Acceleration

5.4.1 Rigid Bench Shoulder Belt Anchor DOE

The shoulder belt anchor positions for the Rigid Bench resulted in peak pelvis resultant accelerations ranging from 36 g to 49 g. The lowest pelvis acceleration occurred when the shoulder belt anchor was at its lowest and most rearward location, which produced a belt fit that engaged the torso early and limited forward excursion of the THOR-5F FE (Figure 5.9A,B) compared to the midpoint (Figure 5.9C,D). The highest pelvis resultant acceleration occurred when the shoulder belt anchor was at its lowest, inboard and furthest forward. The shoulder belt anchor location created a belt fit that resulted in late engagement with the THOR-5F torso until after substantial forward motion (Figure 5.9E,F).

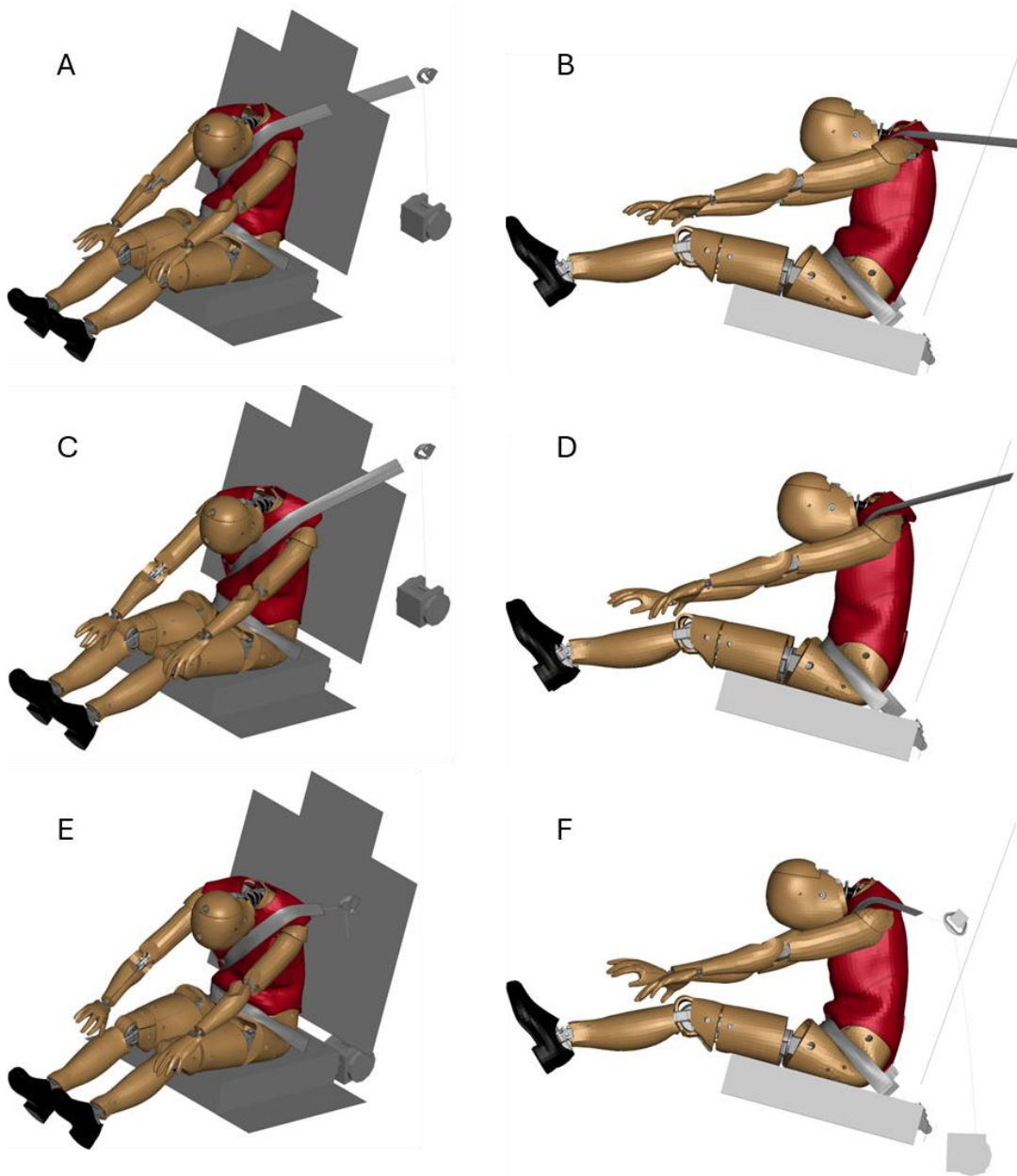


Figure 5.9: Images of the THOR-5F in the Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum resultant pelvis acceleration shoulder belt anchor configuration, (C,D) midpoint shoulder belt anchor configuration, and (E,F) maximum resultant pelvis acceleration shoulder belt anchor configuration. B,D,F are the side view images of each configuration.

The effects analysis clearly illustrated the trends observed in the maximum and minimum peak pelvis resultant accelerations. The X and Y-positions of the shoulder belt anchor were significant, with the X-position having the largest effect magnitude, approximately 1.8 times higher than the next-largest effect magnitude, the Y-position. Moving the shoulder belt anchor rearward and inboard resulted in lower pelvis acceleration through earlier belt engagement and increased belt contact with the shoulder. The XZ interaction had the smallest effect and was positive, with downward anchor position resulting in slightly reduced peak pelvis accelerations, as it improved the belt engagement and contact with the torso of the THOR-5F.

5.4.2 Rigid Bench Lap Belt Anchor DOE

The lap belt anchor position had a greater influence on pelvis acceleration, as it directly controlled lap belt engagement with the THOR-5F pelvis, producing a wider range of peak resultant accelerations (31 g to 55 g). The lowest pelvis acceleration occurred at position 25, similar to the chest deflections, where the lap belt anchors were at the highest Z-position, most forward X-position, with the largest lateral distance between the anchors. At this lap belt anchor location, the belt slightly slipped upward on the pelvis, allowing the THOR-5F to partially slide under the belt with reduced resultant acceleration (Figure 5.10A,B). Compared to the midpoint motion (Figure 5.10C,D), the kinematics of the lowest pelvis resultant acceleration were like the initial stages of submarining but did not progress to full submarining. The highest pelvis acceleration occurred at position 20, where the anchors were at midpoint Z, most forward location, with the minimum lateral spacing between the lap belt anchors, when the belt allowed significant forward translation before engaging the THOR-5F pelvis (Figure 5.10E,F).

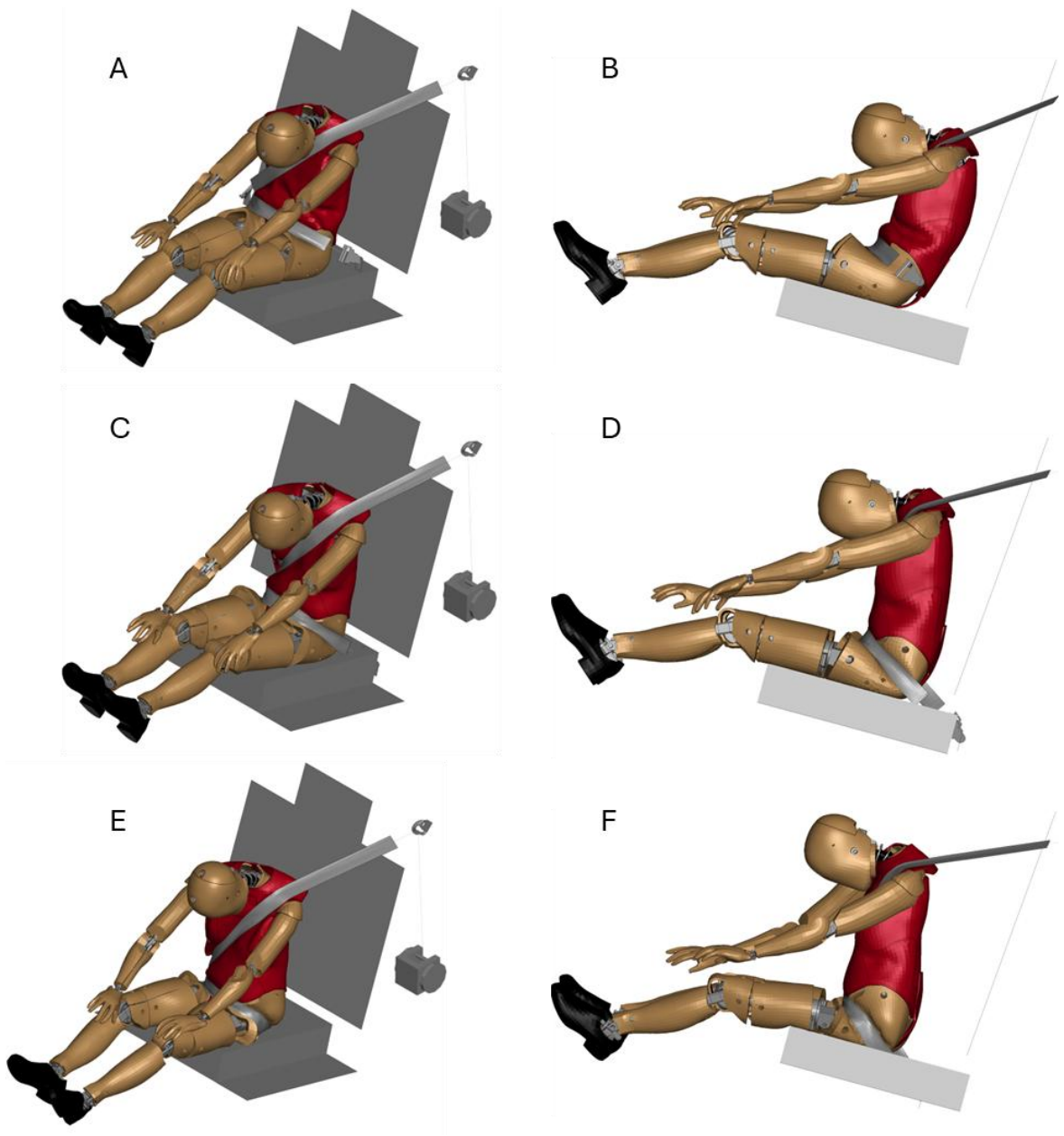


Figure 5.10: Images of the THOR-5F in the Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum resultant pelvis acceleration lap belt anchor configuration, (C,D) midpoint lap belt anchor configuration, and (E,F) maximum resultant pelvis acceleration lap belt anchor configuration. B,D,F are the side view images of each configuration.

The effects analysis of the lap belt anchor locations revealed the X, Y, and Z effects as significant, with the X effect having the largest magnitude, approximately 1.5 times higher than the Z effect magnitude, which was the second largest effect. Moving the lap belt anchor rearward substantially reduced the peak pelvis resultant acceleration by enabling earlier lap belt engagement and more gradual load transfer. The downward anchor positioning (increased Z-position) resulted in higher accelerations, as it resulted in the lap belt giving downward-directed restraint (Figure 5.10F) rather than a more rearward-directed restraint. Increasing the lateral distance between lap belt anchors resulted in increased pelvis acceleration to a lesser extent, as it reduced effective belt-pelvis contact and delayed engagement during the crash pulse.

5.4.3 Semi-Rigid Bench Shoulder Belt Anchor DOE

The shoulder belt anchor position simulations for the Semi-Rigid Bench resulted in peak pelvis accelerations ranging from 35 g to 46 g. The lowest peak pelvis acceleration occurred at the same position as the lowest chest deflection when the shoulder belt anchor was furthest rearward, with inboard lateral (Y) and height (Z) positions. In this configuration, the belt fit had better contact with the THOR-5F upper torso for early engagement (Figure 5.11A,B) than in the midpoint configuration (Figure 5.11C,D). The highest peak pelvis acceleration occurred at Position 27, when the shoulder belt anchor was at the most inboard, forward and lowest position. The belt fit resulted in less shoulder belt engagement with the THOR-5F torso and caused the pelvis to move forward substantially before the belt engaged (Figure 5.11E,F).

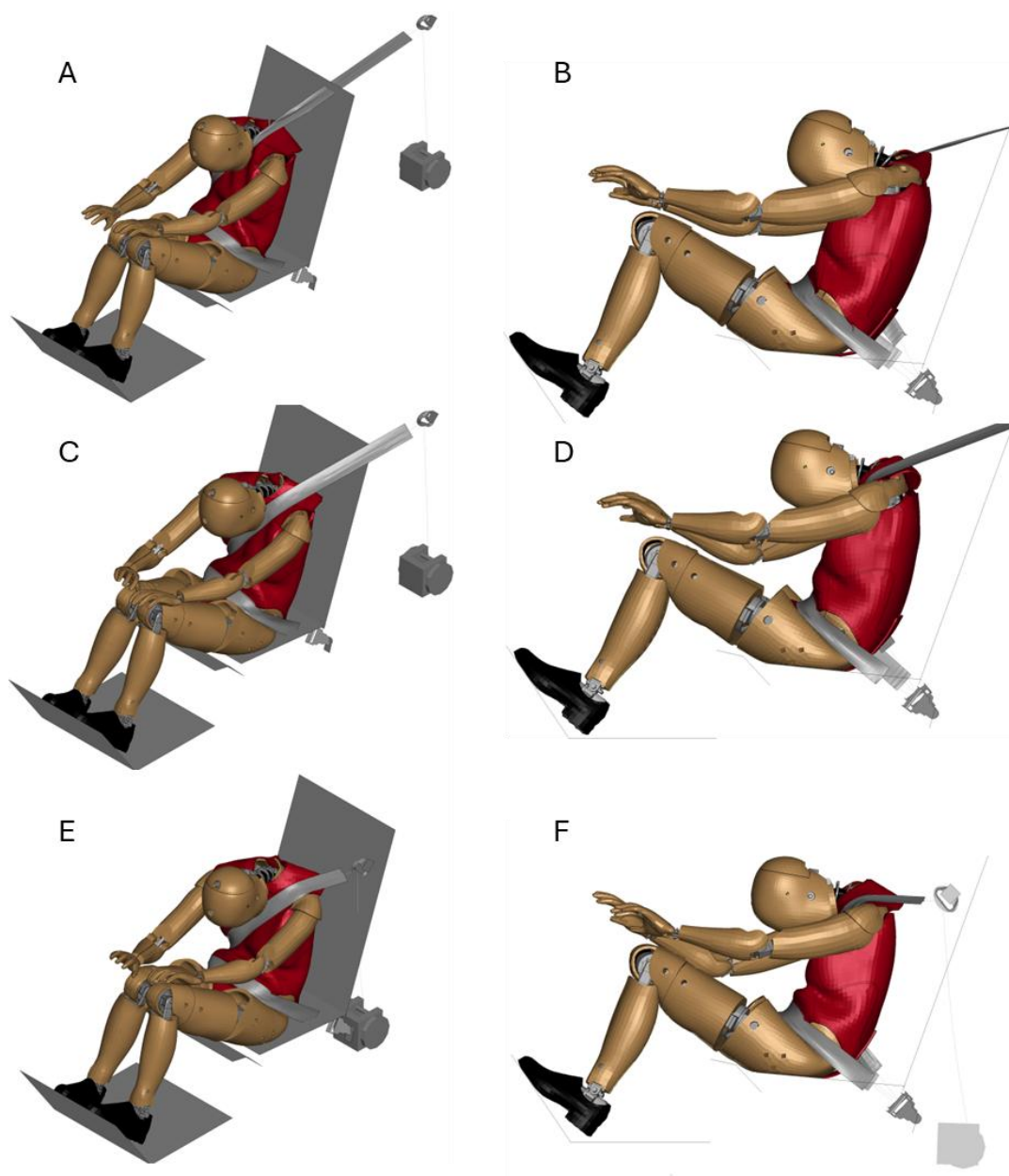


Figure 5.11: Images of the THOR-5F in the Semi-Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum resultant pelvis acceleration shoulder belt anchor configuration, (C,D) midpoint shoulder belt anchor configuration, and (E,F) maximum resultant pelvis acceleration shoulder belt anchor configuration. B,D,F are the side view images of each configuration.

The effects analysis of the shoulder belt anchor in the Semi-Rigid Bench identified one statistically significant factor: the fore-aft (X) position. The positive effect of the X-position meant that moving the shoulder belt anchor forward increased pelvis acceleration, while moving the shoulder belt anchor rearward reduced peak pelvis resultant acceleration. When moving the belt anchor rearward, the belt engaged at an earlier point in the crash pulse, resulting in more gradual load transfer and energy absorption.

5.4.4 Semi-Rigid Bench Lap Belt Anchor DOE

The lap belt anchor location in the Semi-Rigid Bench produced a wider range of resultant pelvis accelerations (32 g to 56 g) compared to the shoulder belt anchor, due to the lap belt directly loading the pelvis. The lowest peak pelvis acceleration was observed at the furthest rearward fore-aft (X) position, average inboard (Y) position and average vertical (Z) position. At this configuration, the pelvis was able to slightly slip under the lap belt, approaching but not fully entering a submarining motion (Figure 5.12A,B). The motion of the THOR-5F FE at the lowest peak resultant pelvis acceleration contrasted with the kinematics at the midpoint (Figure 5.12C, D) and the highest peak pelvis resultant acceleration (Figure 5.12E,F). The highest peak pelvis resultant acceleration occurred when the lap belt anchors were positioned at the most rearward fore/aft (X) position, the smallest lateral (Y) space between the anchors, and the average vertical (Z) position. The belt fit at this anchor configuration, restricted pelvis rotation and increased the concentration of the loading on the lap belt, resulting in a higher peak pelvis acceleration. While the side views and the kinematics seem similar between the max and min positions, the restraint mechanics were different. The maximum pelvis acceleration case involved less effective pelvic engagement, resulting in further excursion before restraint, while the minimum position resulted in the belt wrapping around the pelvis, enabling earlier contact and gradual decelerations.

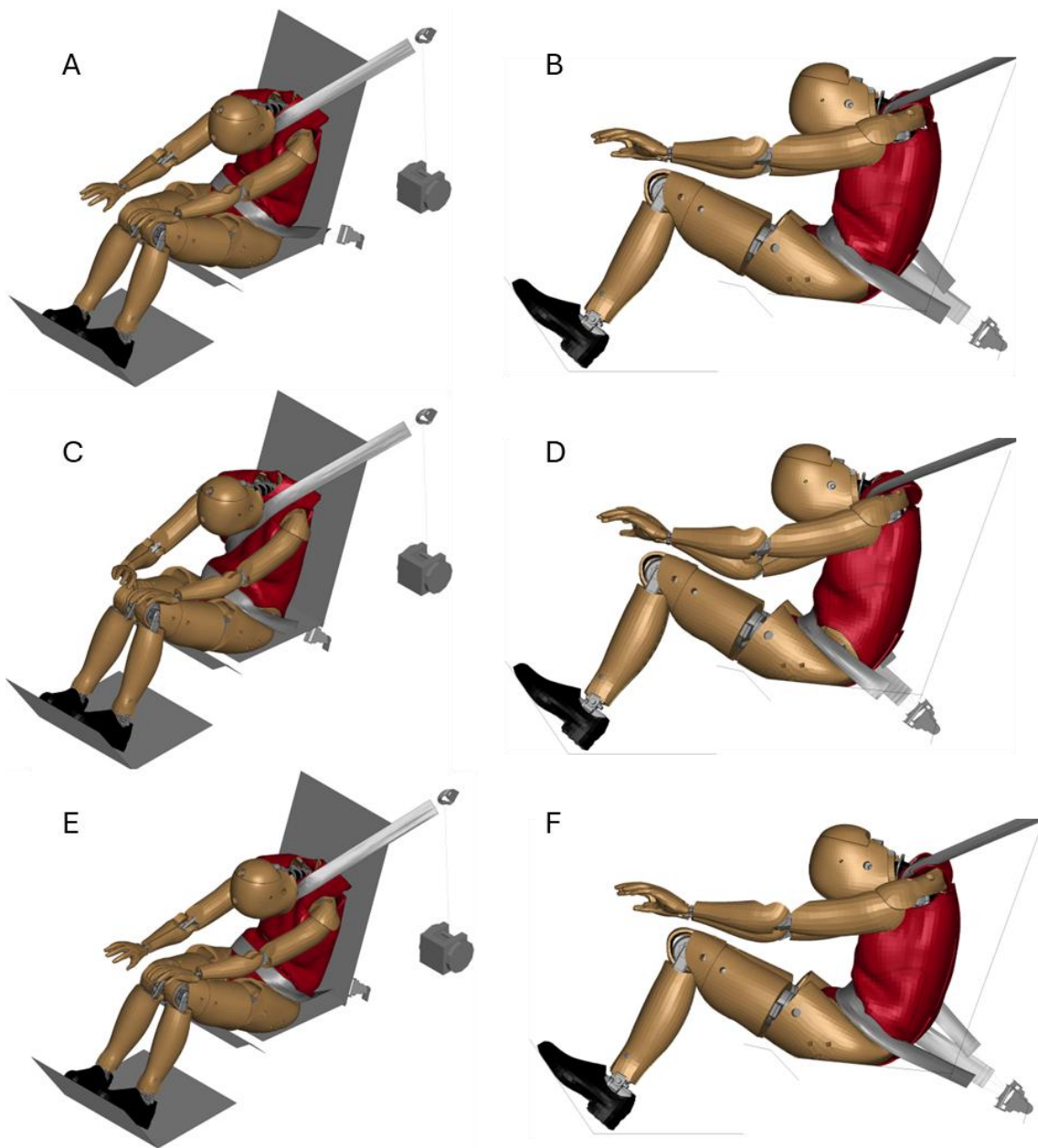


Figure 5.12: Images of the THOR-5F in the Semi-Rigid Bench model at maximum excursion ($t = 105$ ms), with (A,B) minimum resultant pelvis acceleration lap belt anchor configuration, (C,D) midpoint lap belt anchor configuration, and (E,F) maximum resultant pelvis acceleration lap belt anchor configuration. B,D,F are the side view images of each configuration.

The effects analysis identified two statistically significant effects for peak resultant pelvis acceleration: the Y-position and the YZ interaction. The Y-position effect was negative, indicating that reducing the lateral spacing decreased peak pelvis resultant acceleration, as observed in the anchor configurations corresponding to the maximum and minimum pelvis accelerations. However, the YZ interaction was positive and approximately twice the magnitude of the Y main effect, indicating that the effect of the anchor's lateral position on the pelvis acceleration depended on the anchor's height. When the anchor was positioned higher vertically, increased lateral spacing reduced pelvis acceleration due to the shift in response modes from forward translation to a slight rotation about the lap belt, a near-submarining motion. When the lap belt anchor is positioned lower, the effect of moving Y outboard increased pelvis acceleration, as it reduced belt wrap around the pelvis, resulting in less effective restraint.

5.5 Comparison of Rigid Bench and Semi-Rigid Bench

The Rigid Bench and Semi-Rigid Bench exhibited several similar overall trends in how the belt anchor locations influenced the THOR-5F FE kinematics. However, several kinematic differences were observed due to the rotating seat pan and contact with the footrest in the Semi-Rigid Bench. For the shoulder belt anchor positions, both benches showed that rearward and inboard positioning reduced the overall peak chest deflection, with the X and Y-positions having the largest effects. Additionally, the THOR-5F FE showed larger chest deflection magnitudes on the Rigid Bench than on the Semi-Rigid Bench. The higher deflections indicated that footrest support and a deflecting seat pan altered the belt loading and belt position relative to the occupant, reducing thoracic loading during the simulation. The X-position effect was larger than the Y-position effect in both benches' DOEs; however, the Rigid Bench showed a larger difference between the two effect magnitudes. For both benches, the Z-position was significant and negative, although the Rigid Bench also included significant negative XZ interaction. The interaction effect suggested a greater dependence on the shoulder belt to reduce forward motion due to the lack of a footrest and a deflecting seat pan.

When observing the effect of shoulder belt anchor positioning on the peak pelvis acceleration, the footrest contact also influenced the peak pelvis acceleration for the Semi-Rigid Bench, resulting in a lower maximum peak pelvis resultant acceleration. Additionally, the Semi-Rigid Bench identified only the X-position as significant, whereas the Rigid Bench identified the Y-position and the XZ

interaction as significant. The seat pan deflection in the Semi-Rigid Bench produced a pelvis motion comprising forward and downward components, compared to the Rigid Bench, which primarily produced a forward motion (Figure 5.13).

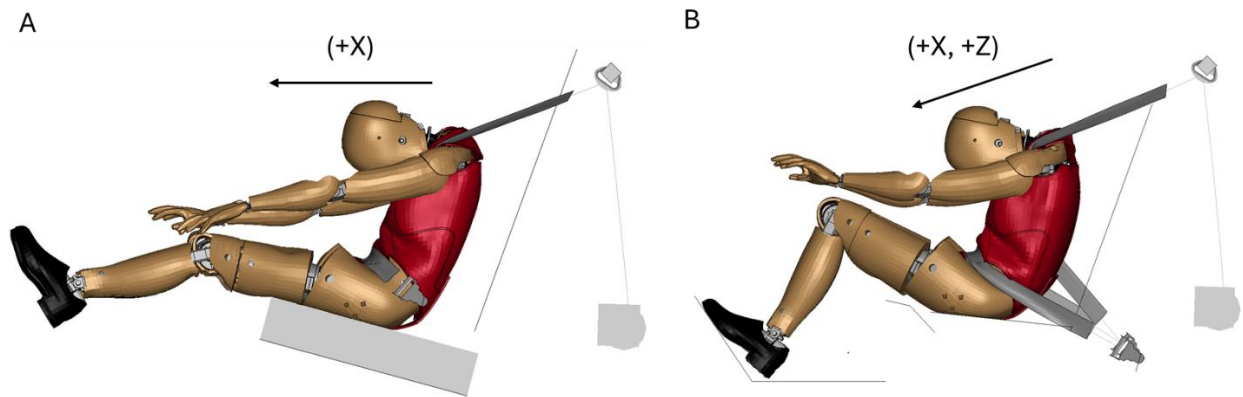


Figure 5.13: Motion of THOR-5F FE at the lowest pelvis resultant acceleration in the (A) Rigid Bench configuration and (B) Semi-Rigid Bench configuration. The Rigid Bench configuration only enabled forward (+X) translation of the pelvis, while the Semi-Rigid Bench configuration enabled the forward (+X) and lower (+Z) motion of the pelvis, which led to a higher pelvis resultant acceleration.

For the lap belt anchor positions and chest deflection response, the Rigid Bench identified the X and Z-positions as the only statistically significant effects, whereas the Semi-Rigid Bench also showed multiple interaction effects with larger magnitudes. The significant interactions for the Semi-Rigid Bench were likely caused by seat pan rotation and footrest contact, which made certain lap belt angles effective in restraining pelvic movement without causing a forward torso pitch into the shoulder belt. The seat pan deflection enabled more downward motion of the THOR-5F FE and reduced the peak chest deflections compared to the Rigid Bench configurations. As a result of the deflecting seat pan and the footrest, the lap belt response depended on the combined position of the anchors in the Semi-Rigid Bench, whereas the Rigid Bench response was more dependent on the individual anchor locations.

Both benches exhibited a significant Y-position effect for peak pelvis acceleration; however, the effect was positive for the Rigid Bench and negative for the Semi-Rigid Bench. Additionally, the Rigid Bench also showed the X and Z-positions as significant, while the Semi-Rigid Bench indicated a positive YZ effect as significant. Kinematically, the effects analysis indicated that in the Rigid Bench, wider lateral spacing between anchors increased peak pelvis acceleration through reduced belt engagement. In contrast, in the Semi-Rigid Bench, wider spacing at upward anchor locations reduced pelvis accelerations, likely due to improved pelvic engagement combined with the seat pan deflection, resulting in a slight rotation of the pelvis around the lap belt instead of translation. The positive YZ interaction indicated that moving anchors downward created a greater vertical restraint to reduce pelvis rotation; in this case, the increased lateral spacing increased the pelvis acceleration through less effective belt contact.

5.6 Anchor Locations and FMVSS/CMVSS Regulations

For the shoulder belt anchor position, the X-Z configuration had some extreme anchor position combinations that were outside the acceptable design space defined by the FMVSS/CMVSS 210 standard (Transport Canada, n.d.; U.S. Department of Transportation, n.d.), with the lowest (Z_{MAX}) and most forward (X_{MAX}) positions not accepted (Figure 5.14). All other anchor locations were within the acceptable anchor range, including the highest (Z_{MIN}) and the most forward (X_{MAX}), which resulted in the highest chest loading and pelvis accelerations due to late belt engagement. Positions 21 and 24, in which the shoulder belt slid from the shoulder of the THOR-5F and rendered the restraint ineffective, were also outside the acceptable range of belt anchors. However, the other anchor location where the belt slid off the shoulder (Position 12) remained within the acceptable anchor range. Additionally, the acceptable anchor range only considers the X and Z-positions and does not include the Y-positions, which had a large effect on reducing chest deflection for both benches. The importance of Y-position in reducing chest deflection, together with its influence on the belt fit resulting in neck contact or slipping off the shoulder, shows that compliance with the regulation alone does not guarantee an optimal shoulder belt fit for the small stature female occupant.

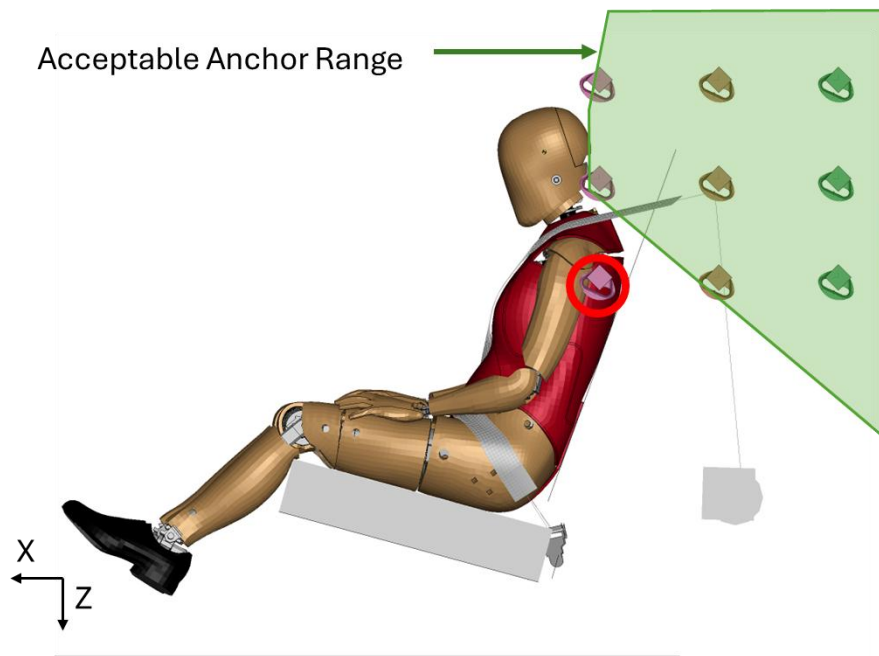


Figure 5.14: Side view of Rigid Bench with shoulder belt anchor X and Z-positions investigated and the acceptable anchor range defined by the CMVSS and FMVSS 210 Standards (Transport Canada, n.d.; U.S. Department of Transportation, n.d.). Locations within the red circle represent simulated locations outside the acceptable range.

For the lap belt anchor positions, all combinations with the most forward (X_{MAX}) and highest (Z_{MIN}) positions were outside the FMVSS/CMVSS 210 acceptable range (Figure 5.15) (Transport Canada, n.d.; U.S. Department of Transportation, n.d.). The anchor position combinations outside the acceptable range caused submarining at the highest vertical positions (Z_{MIN}). Additionally, the forward anchors allowed more forward excursion, resulting in lower chest deflections but increasing the risk of poor lap belt engagement with the pelvis. The lap belt anchors outside the acceptable region illustrated how extreme lap belt anchor positioning influences occupant kinematics and restraint effectiveness.

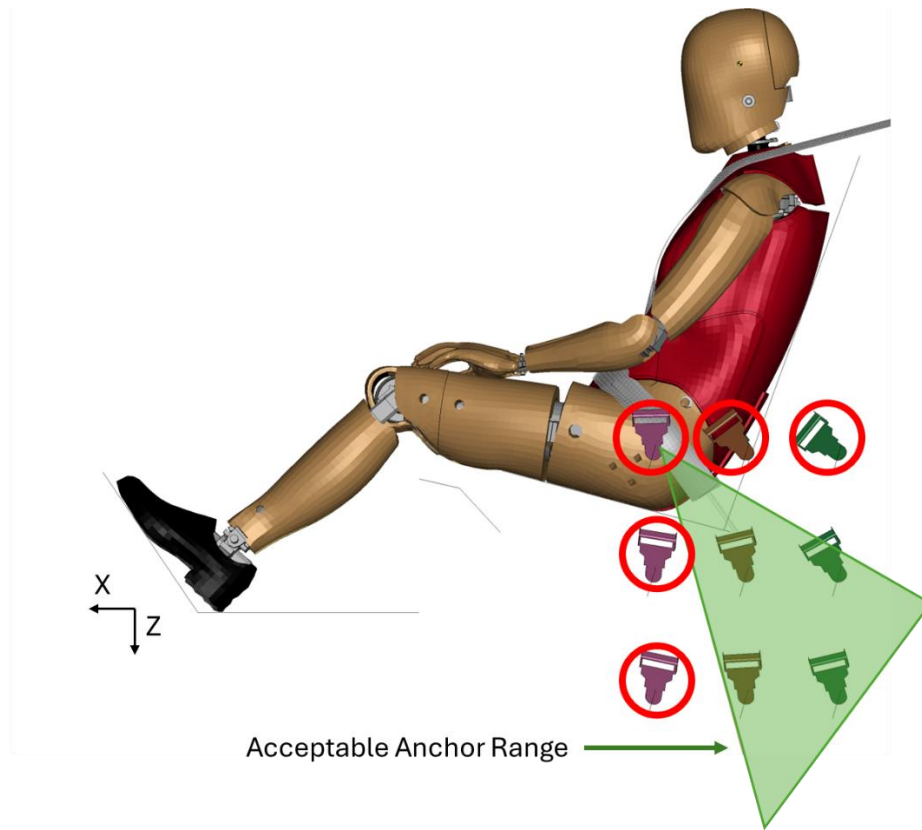


Figure 5.15: Side view of Semi-Rigid Bench with lap belt anchor X and Z-positions investigated, and the acceptable anchor range defined by the CMVSS and FMVSS 210 Standard (Transport Canada, n.d.; U.S. Department of Transportation, n.d.). Locations within red circles represent simulated locations outside the acceptable range.

5.7 Seat Belt Anchor Recommendations for the Small Statured Female

Based on the results of the shoulder belt anchor parametric study, the favorable anchor location for the THOR-5F in both benches was rearward in X, and inboard in Y, while avoiding extreme inboard positions that would cause belt loading on the neck. Additionally, the lower positioning of the shoulder belt anchor was also a contributor to reduced injury risks through lower chest deflection. This location (lower Z, inboard Y, rearward X) aligns with the maxims of effective restraint design (Eppinger, 2002), as the rearward shoulder belt anchor location produced early shoulder belt

engagement for more gradual deceleration and loading. The inboard position also adhered to the maxim of applying restraint forces to the bony anatomy, instead of the softer tissues of the chest. Additionally, the anchor location would remain within the FMVSS/CMVSS-defined acceptable anchor range.

The most beneficial lap belt anchor location depended on the bench configuration. For the Rigid Bench, the effects analysis indicated that moving the lap belt anchors rearward reduced chest deflection and peak pelvis acceleration through earlier lap belt engagement. Additionally, moving the anchors downward in Z further reduced pelvis acceleration by providing a stable belt-pelvis contact interface for the THOR-5F FE.

The favorable anchor location for the Semi-Rigid Bench differed due to the coupled forward (+X) and downward (+Z) motion, along with foot contact with the footrest. Based on the dominant interaction effects, the most beneficial lap belt anchor configuration was at a forward position within the regulations, with a lower vertical position and outboard lap belt anchor positioning (maximum lateral distance). The resulting lap belt angle would be better suited for the coupled X–Z motion of the THOR-5F FE, providing sufficient restraint and pelvis engagement without over-constraining and promoting torso pitch, while adhering to the FMVSS/CMVSS 210 standard.

In addition to minimizing the metrics of interest (peak overall chest deflection and pelvis resultant acceleration), the favorable shoulder and lap belt anchor locations also align with general biomechanical principles governing effective occupant restraint.

5.8 Limitations

There were several limitations in the current study, including the positioning of the THOR-5F FE model and the seat belt implementation.

One observed drawback with the physical THOR-5F ATD was the bottoming out of the right lower chest deflection sensor, which was observed in both the TC/PMG testing and the physical Semi-Rigid Bench testing (Richard et al., 2022). The bottoming out occurred when the sensor traces plateaued at a certain deflection, which may have resulted from interactions with other components. A potential cause for the sensor bottoming out is contact between the right lower IR-TRACC and the spine box, which was reported by NHTSA in a design update (Hutter et al., 2024). Although the bottoming out

behavior was not observed in the THOR-5F FE, it may have implications for injury assessment metrics or regulatory criteria that rely on the maximum overall chest deflection.

The positioning of the THOR-5F FE model required the use of 4 kN beam elements to align the FE model to the average physical ATD position. While the approach enabled the H-point of the THOR-5F FE to match the physical test measurements, the positioning process required relatively large forces. To investigate the reason for high forces, two additional simulations were run: one with 1 kN beams and one with gravity settling (Table 5.1). In both cases, the H-point was much further forward and higher than the physical test point, suggesting that the potentially stiff pelvis foam in the FE model may necessitate higher positioning forces to replicate the physical ATD seated posture.

Table 5.1: Difference of H-point from average of physical test measurements for different settling methods.

Settling method	X [mm]	Y [mm]	Z [mm]
Gravity settling	22.9	0	18.6
1 kN positioning beams	8.6	1.3	8.6
4 kN positioning beams	0	0	0

Additionally, stresses and foam compression in the THOR-5F FE were not translated from the positioning simulation to the sled acceleration simulation. The lack of stress transfer may have accounted for slightly different initial posture and motion; however, the differences did not lead to a large deviation in overall kinematics. The lack of stress transfer also resulted in the belt slack removal slightly shifting the position of the THOR-5F FE during the Rigid Bench simulations, resulting in a difference of approximately 1–2 mm in H-point positioning. Despite the differences in positioning, the pelvis acceleration response and belt loading trends remained consistent with the physical tests.

The belt system also exhibited higher stiffness than the physical test belt, resulting in reduced conformity to the THOR-5F jacket. The higher stiffness in the FE model compared to the physical test was evident during the validation crash pulse (Figure 5.16), where the physical belt demonstrated greater compliance and folding compared to the FE model. Future work should investigate the belt

material properties to replicate the physical belt fit stiffness and fit. Despite the slight forward positioning of the lower belt edge, the belt-occupant interaction between the THOR-5F FE and the FE model of the open-source belt matched the physical test in terms of belt loads and kinematic response.

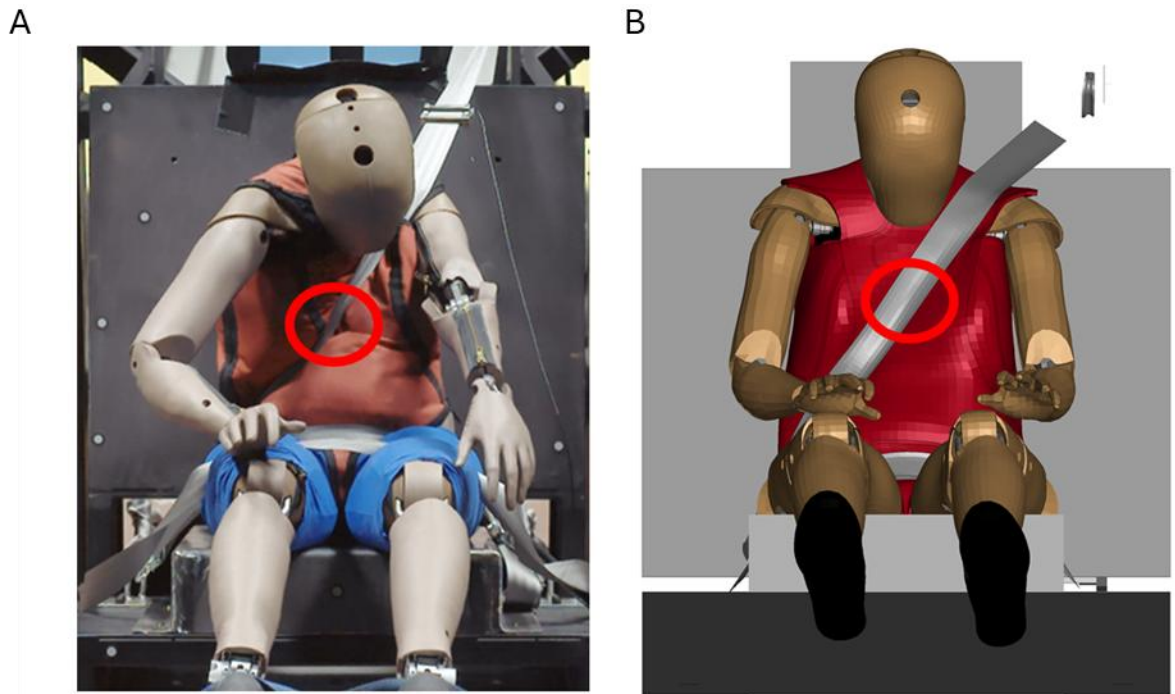


Figure 5.16: Shoulder belt interaction at maximum excursion ($t = 105$ ms) for (A) physical Rigid Bench test and (B) FE simulation of Rigid Bench.

While not significantly affecting the parameters of interest in the present study, the THOR-5F FE model demonstrated lower thorax Z-rotation relative to the physical ATD. The lower thorax Z-rotation could influence the THOR-5F interaction with the shoulder belt, resulting in different loading trends from physical testing. The belt system used in the Semi-Rigid Bench had a simplified belt model, which resulted in a sudden rise in shoulder and lap belt loads compared to the physical test. Despite differences in torso rotation, the magnitudes of the THOR-5F responses generally matched those of the physical test. The iliac load cells were also different from the physical tests for both the

Rigid Bench and the Semi-Rigid Bench. The differences were despite similar lap belt loads and pelvis acceleration, indicating an internal interference in the iliac sensor readings, potentially from the abdomen pressure sensor.

Another limitation was that the shoulder and lap belt anchor positions represented maximum and minimum locations reported in the literature, and specific combinations considered may not represent contemporary vehicle mounting locations. Several anchor locations were outside the range of acceptable anchor locations (Section 5.6). In addition, the DOEs contained removed configurations, and the linear models with interactions were used to estimate the trends despite the missing points. The statistical model approach assumed smooth behavior across the location range and may not account for any non-linear responses or differing trends in the regions extrapolated. As an exploratory study, the method was appropriate for identifying the general positioning trends; however, future work could focus on optimization within realistic and recommended anchor ranges.

The only kinematic metrics considered in this study were chest deflection and pelvis resultant acceleration, which were selected because they directly reflected the restraint of the shoulder belt on the upper torso and the lap belt on the lower torso. Including commonly investigated other injury metrics, such as head resultant acceleration, knee excursion, may provide a more comprehensive overview of whole body kinematics; however, the findings may lead to complications in modelling and effects analysis. Additionally, the interactions between the shoulder belt anchor positioning and lap belt anchor positioning were not explored, as the potential DOE would result in shoulder and lap belt anchors with unstable and unrealistic belt fits. Future studies could examine interactions between the shoulder and lap belt anchors within the acceptable anchor region or across belt geometries in the current fleet.

Chapter 6

Conclusion

The THOR-5F FE in the Rigid Bench was validated through a paired comparison with physical tests conducted on the Rigid Bench at TC/PMG. The model showed good agreement with the physical tests for belt loads and pelvis accelerations. Despite differences in torso rotation, the THOR-5F FE adequately captured physical ATD responses during the Rigid Bench test. Additionally, the THOR-5F FE in the Semi-Rigid Bench model was assessed through comparison with published physical test data from Richard et al. (2022). After belt pullout calibration, the model captured the belt interactions and the key ATD responses observed in the physical tests. A plateau in lower chest deflection due to the sensor bottoming out was also observed during physical testing on both benches, which has been identified as the THOR-5F design undergoes updates. Overall, the ATD in both FE bench models generally reproduced the primary occupant kinematics and restraint responses observed in their corresponding physical tests.

Both the Rigid and Semi-Rigid Bench models were used in parametric studies evaluating the fore/aft (X), lateral (Y), and vertical (Z) positions of the shoulder and lap belt anchors, using peak chest deflection and pelvis acceleration as response metrics. Structural differences between the benches, such as the footrest and deflecting seat pan, altered occupant kinematics and sensitivity to anchor positioning, with the Rigid Bench allowing greater forward excursion and the Semi-Rigid Bench introducing additional downward motion.

Most individual chest sensors showed trends similar to the overall peak chest deflections, with localized differences due to belt routing, indicating consistency in the overarching trends across the different sensor locations. For the shoulder belt anchors, both benches indicated that the X and Y-positions had a large influence on the peak overall chest deflection of the THOR-5F FE. Rearward and inboard shoulder belt anchor positions generally reduced chest deflections through improved belt engagement, with downward positions also providing a reduction through improved belt contact with the shoulder.

The lap belt anchor position had a smaller effect than the shoulder belt anchor on chest deflections. For the Rigid Bench, the X and Z-positions of the lap belt anchor had the largest influence on the

chest deflection. Rearward and upward locations resulted in lower chest deflections, with the anchor location providing greater restraint from forward motions and transferring the load from the shoulder belt. For the Semi-Rigid Bench, the X, Y, and Z-positions all influenced peak overall chest deflections, with the effects analysis indicating a balance between effective restraint and over-restraint causing a forward pitch of the occupant. The forward, downward, and outboard anchor position was identified as a favorable configuration for reducing chest deflection during restraint, as it provided restraint from high excursions into the shoulder belt without promoting a forward pitch into the shoulder belt.

All overall peak chest deflections resulted in high (70+%) AIS 3+ injury risk ranges, which emphasizes the importance of additional restraint technologies, such as load limiters, in addition to improved anchor geometry. Some anchor locations led to undesirable restraint interactions (e.g., belt slipping off the shoulder, submarining), underscoring the need for careful anchor placement. Current belt anchor regulations (FMVSS/CMVSS 210) define acceptable fore/aft and vertical regions; however, this study indicates that lateral anchor positioning is also important and should be considered.

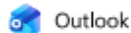
The shoulder belt anchor positioning had a smaller but notable effect on the peak pelvis acceleration, with rearward X positions reducing acceleration through earlier engagement with the THOR-5F FE torso in both benches. Additionally, for the Rigid Bench, the Y-position and the XZ interaction effects were also considerable due to the primarily forward motion in the Rigid Bench, which enabled the belt to have improved engagement about the shoulder when moving the anchor in these directions.

By comparison, the lap belt anchor had a much larger effect on the peak resultant pelvis acceleration, and reductions in acceleration were associated with either effective restraint or a near-submarining response mode. The X, Y, and Z-positions of the lap belt anchor influenced the peak resultant pelvis acceleration in the Rigid Bench. Excluding submarining or near-submarining motion, the rearward, upward, and inboard locations reduced pelvis accelerations through earlier engagement and increased contact with the pelvis and restraint in the X direction. For the Semi-Rigid Bench, the large YZ interaction effect indicated a complex interaction between the deflecting seat pan and the anchor locations, which resulted in the THOR-5F FE response occurring in two modes: a forward translation or a near-submarining motion. When considering the forward motion case, the favorable anchor

position restrained the pelvis with inboard and downward anchor positions, resulting in increased belt contact with the pelvis and more effective restraint.

Future studies should investigate the effects of belt material properties, vehicle fleet-specific anchor locations, interactions between lap and shoulder belt anchor positions, and the implementation of restraint technologies such as load limiters. Overall, the results from this study demonstrated that belt anchor positions substantially influenced the THOR-5F thorax and pelvis responses. The findings from this study demonstrate that validated simplified bench models can reproduce physical occupant and restraint responses, enable the investigation of parameters such as seat belt anchor position, and support the development of rear seat occupant safety.

Letter of Copyright Permission



RE: Permission Request to Reuse Figures from Stapp Car Crash Journal Article

From: Stapp Journal <stappjournal@mcw.edu>
Date: Wed 5/20/2026 12:27 PM
To: Patrick Thangarajah <[REDACTED]>
Cc: Stapp Journal <stappjournal@mcw.edu>

Hello Patrick –

Thank you for your patience while your request was reviewed. There were some delays as our EIC was out of the country for a conference. The request has now been approved. Please see the permissions below:

We are pleased to grant non-exclusive permission to reproduce Figures 3 and 15 from paper titled “THOR-05F Response in Sled Tests Inducing Submarining and Comparison with PMHS Response Corridors” in Stapp Car Crash Journal vol. 65 for use in Patrick Thangarajah’s Master’s thesis. Please ensure proper credit is given to the original publication, such as specifying “Figure used by permission of Stapp Association” in the paper.

If you need anything further, please let us know.

Thanks -

JESSICA ALONSO
Editorial Assistant
Stapp Journal, Editor’s Office
Upcoming OOO: May 22 - 25

From: Patrick Thangarajah <[REDACTED]>
Sent: Tuesday, May 12, 2026 2:50 PM
To: Stapp Journal <stappjournal@mcw.edu>
Subject: Permission Request to Reuse Figures from Stapp Car Crash Journal Article

ATTENTION: This email originated from a sender outside of MCW. Use caution when clicking on links or opening attachments.

Hello,

My name is Patrick Thangarajah, and I am a Master’s student at the University of Waterloo. I am currently writing my Master’s thesis and would like to request permission to reuse figures from the following Stapp Car Crash Journal publication:

Richard O, Lebarbé M, Uriot J, Trosseille X, Petit P, Wang ZJ, Lee E. *THOR-05F Response in Sled Tests Inducing Submarining and Comparison with PMHS Response Corridors*. Stapp Car Crash Journal. 2021;65:91–138.
<https://doi.org/10.4271/2021-22-0005>

I would like to use the images in Figures 3 and 15 for inclusion in my thesis, with full citation to the original source. I attempted to obtain permission through the Copyright Clearance Center, but was advised to contact the publisher or authors directly.

Please let me know if this is possible, or if there are any additional conditions required.

Thank you,
Patrick Thangarajah

References

- Abbreviated Injury Scale (AIS). (n.d.). *Association for the Advancement of Automotive Medicine*. Retrieved December 19, 2025, from <https://www.aaam.org/education-resource-center/public-position-statements/abbreviated-injury-scale-ais-position-statement/>
- Agarwal, G., Kidambi, N., & Lange, R. (2021). *Seat Belts: A Review of Technological Milestones, Regulatory Advancements, and Anticipated Future Trajectories*. SAE International. <https://doi.org/10.4271/2021-01-5097>
- Aram, M., & Rockwell, T. (2012). *Vehicle rear seat study*. National Highway Traffic Safety Administration, Washington, DC.
- Arbogast, K. B., Locey, C. M., & Zonfrillo, M. R. (2012). Differences in Thoracic Injury Causation Patterns Between Seat Belt Restrained Children and Adults. *Annals of Advances in Automotive Medicine / Annual Scientific Conference*, 56, 213–221.
- Asadi, K., & Karlsson, K. (2023). *Automation of ATD positioning for crash analysis simulation*. ATD Models GmbH. (n.d.). *ATD-TF05*. Retrieved May 6, 2026, from <https://www.atd-models.com/en/products/thor/atd-tf05.html>
- Bandak, F., Eppinger, R., Haffner, M., Khaewpong, N., Kuppa, S., Maltese, M., Nguyen, T., Saul, R., Sun, E., & Takhounts, E. (1999). *Development of improved injury criteria for the assessment of advanced automotive restraint systems: II*.
- Barry, K. (2020, September 3). *Making Back Seats Safer*. Consumer Reports. <https://www.consumerreports.org/car-safety/making-car-back-seats-safer/>
- Beck, B., Bilston, L. E., & Brown, J. (2016). Injury patterns of rear seat occupants in frontal impact: An in-depth crash investigation study. *Injury Prevention: Journal of the International Society for Child and Adolescent Injury Prevention*, 22(3), 165–170. <https://doi.org/10.1136/injuryprev-2015-041715>
- Bose, D., Segui-Gomez, M., & Crandall, J. R. (2011). Vulnerability of Female Drivers Involved in Motor Vehicle Crashes: An Analysis of US Population at Risk. *American Journal of Public Health*, 101(12), 2368–2373. <https://doi.org/10.2105/AJPH.2011.300275>
- Boyle, D. M., Albert, D. L., Hardy, W. N., & Kemper, A. R. (2024). Evaluating the Responses of the THOR and Hybrid III Small Female ATDs during Frontal Sled Tests. *SAE International Journal of Transportation Safety*, 12(3), 325–354.
- Bridges, W., Ganesan, V., Barki, G., Jayakumar, P., Davies, J., & Umashankar, S. K. M. (2022). *Integrated Seat Belt System Model Development*.
- Brumbelow, M. L., Baker, B. C., & Nolan, J. M. (2007). Effects of seat belt load limiters on driver fatalities in frontal crashes of passenger cars. *Proceedings of the International Technical Conference on the Enhanced Safety of Vehicles*, (07–0067).
- Brumbelow, M. L., Jermakian, J. S., & Arbelaez, R. A. (2022). Predicting real-world thoracic injury using THOR and hybrid III crash tests. *Proceedings of the International Research Council on the Biomechanics on Injury (IRCOBI) Conference*, 264–287.

- Carpanen, D., Masouros, S., & Newell, N. (2016). Surrogates of Human Injury. In A. M. J. Bull, J. Clasper, & P. F. Mahoney (Eds.), *Blast Injury Science and Engineering: A Guide for Clinicians and Researchers* (pp. 189–198). Springer International Publishing. https://doi.org/10.1007/978-3-319-21867-0_16
- Carroll, J., Kruse, D., Mishra, E., Mroz, K., & Lubbe, N. (2021). Comparison of Hybrid III and THOR 5th percentile female dummies in frontal crash tests. *Proceedings of the International Research Council on the Biomechanics on Injury (IRCOBI) Conference*.
- Cavanaugh, J. M., & Yoganandan, N. (2015). Thorax Injury Biomechanics. In N. Yoganandan, A. M. Nahum, & J. W. Melvin (Eds.), *Accidental Injury: Biomechanics and Prevention* (pp. 331–372). Springer. https://doi.org/10.1007/978-1-4939-1732-7_13
- Cellbond. (n.d.). *THOR-5F*. Cellbond.Com. Retrieved May 1, 2026, from <https://www.cellbond.com/products/atds/frontal-impact/thor-5f/>
- Challa, B. N. S. A., Yang, P., Carlson, M., Suntay, B., Stammen, J., & Noll, S. (2025). Development of a Dynamic Nonlinear Finite Element Model of the Large Omnidirectional Child Crash Test Dummy. *SAE International Journal of Advances and Current Practices in Mobility*, 7(1), 430–447. <https://doi.org/10.4271/2024-01-2509>
- Ciccone, M. A., & Wells, J. K. (1988). Improper Shoulder Belt Use by Maryland Drivers. *Human Factors*, 30(3), 359–366. <https://doi.org/10.1177/001872088803000309>
- Crandall, J. R., Bose, D., Forman, J., Untaroiu, C. D., Arregui-Dalmases, C., Shaw, C. G., & Kerrigan, J. (2011). Human surrogates for injury biomechanics research. *Clinical Anatomy*, 24(3), 362–371.
- Dahlgren, M., Vishwanatha, A., Soni, A., Engstrand, K., Forsberg, J., & Yeh, I. (2020). *Belt Modelling in LSDYNA*. 16th International LS-DYNA User's Conference. Available: <https://www.dynalook.com/conferences/16th-international-ls-dynaconference/aerospace-01-2/t1-2-b-aerospace-148.pdf>
- Dehner, C., Elbel, M., Schick, S., Walz, F., Hell, W., & Kramer, M. (2007). Risk of injury of the cervical spine in sled tests in female volunteers. *Clinical Biomechanics*, 22(6), 615–622. <https://doi.org/10.1016/j.clinbiomech.2007.02.004>
- Durbin, D. R., Jermakian, J. S., Kallan, M. J., McCartt, A. T., Arbogast, K. B., Zonfrillo, M. R., & Myers, R. K. (2015). Rear seat safety: Variation in protection by occupant, crash and vehicle characteristics. *Accident Analysis & Prevention*, 80, 185–192. <https://doi.org/10.1016/j.aap.2015.04.006>
- Eastman, J. W. (1981). “DOCTORS’ORDERS”: THE AMERICAN MEDICAL PROFESSION AND THE ORIGINS OF AUTOMOBILE DESIGN FOR CRASH PROTECTION, 1930-1955. *Bulletin of the History of Medicine*, 55(3), 407–424.
- Edwards, M. A., Jagtap, S. R., & Jermakian, J. S. (2022). Sensitivity of Established and Alternative Rear Occupant Thoracic Injury Metrics to Seat Belt Design Variables in Frontal Impacts. *Proceedings of the International Research Council on the Biomechanics on Injury (IRCOBI) Conference*.

- Eggers, A., Ott, J., Pipkorn, B., Brase, D., Mroz, K., Valdes, F. L., Hynd, D., & Peldschus, S. (2017). A new generic frontal occupant sled test set-up developed within the EU-project SENIORS. *Proceedings of the Conference on the Enhanced Safety of Vehicles (ESV)*.
- Eggers, A., Schießler, M., Ott, J., Langner, T., & Wisch, M. (2023). Sensitivity of Chest Deflection Measurements in Thor-5F and Hybrid III Small Female Dummies to Different Seat and Belt Settings. *Proceedings of the Conference on the Enhanced Safety of Vehicles (ESV)*, (23–0301).
- Eppinger, R. (2002). Occupant Restraint Systems. In A. M. Nahum & J. W. Melvin (Eds.), *Accidental Injury: Biomechanics and Prevention* (pp. 187–197). Springer New York.
https://doi.org/10.1007/978-0-387-21787-1_8
- Esfahani, E. S., Marzougui, D., Digges, K., & Kan, C. D. (Steve). (2011). Effect of increase in weight and stiffness of vehicles on the safety of rear seat occupants. *International Journal of Crashworthiness*, 16(3), 309–318. <https://doi.org/10.1080/13588265.2011.566474>
- Euro NCAP. (2025). *Crash Protection: Frontal Impact Protocol* (Version 1.1).
- Exploring advanced restraint technologies for improving safety equity for frontal occupants using diverse human body models. (2025). *Traffic Injury Prevention*, 26(1), S31–S39.
<https://doi.org/10.1080/15389588.2025.2510571>
- Gantz, T., & Henkle, G. (2002). Seatbelts: Current issues. *Prevention Institute*.
- Gepner, B. D., Perez-Rapela, D., Forman, J. L., Ostling, M., Pipkorn, B., & Kerrigan, J. R. (2022). Evaluation of GHBM, THUMS and SAFER human body models in frontal impacts in reclined postures. *Proceedings of the International Research Council on the Biomechanics on Injury (IRCOBI) Conference*, 116–143.
- Griswold, I. R. W., & De, H. H. (1955). *Combination shoulder and lap safety belts*.
- Hagedorn, A., Stammen, J., Ramachandra, R., Rhule, H., Thomas, C., Suntay, B., Kang, Y.-S., Kwon, H. J., Moorhouse, K., & Iv, J. H. B. (2022). Biofidelity Evaluation of THOR-50M in Rear-Facing Seating Configurations Using an Updated Biofidelity Ranking System. *SAE International Journal of Transportation Safety*, 10(2), 291–375. <https://doi.org/10.4271/09-10-02-0013>
- Håland, Y. (2006). The evolution of the three point seat belt from yesterday to tomorrow. *Proceedings of the International Research Council on the Biomechanics on Injury (IRCOBI) Conference*, 3–15.
- Hartlen, D. C., & Cronin, D. S. (2022). Arc-length re-parametrization and signal registration to determine a characteristic average and statistical response corridors of biomechanical data. *Frontiers in Bioengineering and Biotechnology*, 10, 843148.
- Hu, J., Klinich, K. D., Reed, M. P., Kokkolaras, M., & Rupp, J. D. (2012). Development and validation of a modified Hybrid-III six-year-old dummy model for simulating submarining in motor-vehicle crashes. *Medical Engineering & Physics*, 34(5), 541–551.

- Hu, J., Wu, J., Klinich, K. D., Reed, M. P., Rupp, J. D., & Cao, L. (2013). Optimizing the Rear Seat Environment for Older Children, Adults, and Infants. *Traffic Injury Prevention*, 14(sup1), S13–S22. <https://doi.org/10.1080/15389588.2013.796043>
- Hu, J., Wu, J., Reed, M. P., Klinich, K. D., & Cao, L. (2013). Rear seat restraint system optimization for older children in frontal crashes. *Traffic Injury Prevention*, 14(6), 614–622.
- Humanetics. (n.d.). *Frontal Impact ATD Virtual Models*. Retrieved April 30, 2026, from <https://www.humaneticsgroup.com/products/virtual-models/frontal-impact-atd-virtual-models>
- Humanetics. (2024). *Humanetics CAE updates: January 2024*. Humanetics. https://www.humaneticsgroup.com/sites/default/files/2024-01/humanetics_cae_updates_january-2024_2.pdf
- Humanetics Innovative Solutions. (n.d.). *THOR 5th Percentile Female Frontal Crash Test Dummy (THOR-5F)* (Product Brochure). Humanetics Innovative Solutions. Retrieved <https://www.humaneticsgroup.com/products/anthropomorphic-test-devices/frontal-impact/thor-5f>
- Hutter, E., Moorhouse, K., Carlson, M., & Hagedorn, A. (2024). *NHTSA's THOR-05F Design Update* (Nos. 16180-NSR-231211–006). National Highway Traffic Safety Administration; Transportation Research Center, Inc. https://www.nhtsa.gov/sites/nhtsa.gov/files/2024-02/16180-NSR-231211-006_SAE_NHTSAs%20THOR-05F%20Design%20Update-tag.pdf
- Insurance Institute for Highway Safety. (n.d.). *Fatality Facts 2023: Passenger vehicle occupants*. IIHS-HLDI Crash Testing and Highway Safety. Retrieved December 21, 2025, from <https://www.iihs.org/research-areas/fatality-statistics/detail/passenger-vehicle-occupants>
- Insurance Institute for Highway Safety. (2025). *Moderate Overlap Frontal Crashworthiness Evaluation 2.0 Crash Test Protocol* (Version II).
- International Organization for Standardization. (2024). *Road vehicles—Objective rating for non-ambiguous signals* (Technical Specification ISO/TS 18571:2024). <https://www.iso.org/standard/85791.html>
- Jermakian, J., Edwards, M., Fein, S., & Maltese, M. R. (2019). Factors contributing to serious and fatal injuries in belted rear seat occupants in frontal crashes. *Traffic Injury Prevention*, 20(sup1), S84–S91. <https://doi.org/10.1080/15389588.2019.1601182>
- Kahane, C. J. (2013). *Effectiveness of pretensioners and load limiters for enhancing fatality reduction by seat belts* (DOT HS-811-835; p. 835). National Highway Traffic Safety Administration.
- Kan, C.-D., Marzougui, D., & Bedewi, N. E. (2003). *Development of a 50th percentile Hybrid III dummy model*. 22–23.
- Kang, S., Chen, C., Guha, S., Paladugu, M., Ramasamy, M. S., Gade, L., & Zhu, F. (2018). *LS-DYNA® belted occupant model*. 15th International LS-DYNA Users Conference.
- Klein, K. F., Reed, M. P., & Rupp, J. D. (2016). *Development of geometric specifications for a small female anthropomorphic test device pelvis*. University of Michigan, Ann Arbor, Transportation Research Institute. <https://hdl.handle.net/2027.42/117574>

- Klinich, K. D., Reed, M. P., Ebert, S. M., & Rupp, J. (2011). *Effect of realistic vehicle seats, cushion length, and lap belt geometry on child ATD kinematics*. University of Michigan, Ann Arbor, Transportation Research Institute.
- Lam, J., Dennis, N. J., Dix, J., Lambrecht, M., & Ootani, R. (2014). A Comparison of the NHTSA Research Offset Oblique and Small Overlap Impact Tests and the IIHS Moderate and Small Overlap Tests. *SAE International Journal of Transportation Safety*, 2(2), 268–278.
- Lebarbé, M., Valdés, F. L., Valdano, M., Petit, P., Marshall, W., Lemmen, P., Östling, M., Masuda, M., Richard, O., & Beillas, P. (2023). The ENOP project (Enable New Occupant Seating Positions). In *Fahrzeugsicherheit* (Vol. 2426, pp. 159–176).
- Lee, E., & Craig, M. (2021). *Thoracic Injury Criteria Considerations for the THOR 5th ATD*. 49th International Workshop on Human Subjects for Biomechanical Research.
- Lee, E., Parent, D., Craig, M., McFadden, J., & Moorhouse, K. (2020). *Biomechanical response manual: THOR 5th percentile female NHTSA advanced frontal dummy, Revision 2*.
- Leport, T., Baudrit, P., Potier, P., Trosseille, X., Lecuyer, E., & Vallancien, G. (2011). *Study of Rib Fracture Mechanisms Based on the Rib Strain Profiles in Side and Forward Oblique Impact*. The Stapp Association. 55th Stapp Car Crash Conference. <https://doi.org/10.4271/2011-22-0009>
- Linder, A., & Svensson, M. Y. (2019). Road safety: The average male as a norm in vehicle occupant crash safety assessment. *Interdisciplinary Science Reviews*, 44(2), 140–153. <https://doi.org/10.1080/03080188.2019.1603870>
- Livermore Software Technology. (2021). *LS-DYNA® Keyword User's Manual, Volume I*. (Revision r:14186). https://www.dynasupport.com/manuals/ls-dyna-manuals/ls-dyna_manual_volume_i_r13.pdf
- Locey, C. M., Garcia-Espana, J. F., Toh, A., Belwadi, A., Arbogast, K. B., & Maltese, M. R. (2012). Homogenization of vehicle fleet frontal crash pulses from 2000-2010. *Ann. Adv. Automot. Med.*, 56, 299–311.
- Lu, Y.-C., & Untaroiu, C. D. (2013). Statistical shape analysis of clavicular cortical bone with applications to the development of mean and boundary shape models. *Computer Methods and Programs in Biomedicine*, 111(3), 613–628. <https://doi.org/10.1016/j.cmpb.2013.05.017>
- Luet, C., Trosseille, X., Drazétic, P., Potier, P., & Vallancien, G. (2012). Kinematics and dynamics of the pelvis in the process of submarining using PMHS sled tests. *Stapp Car Crash Journal*, 56, 411–442. <https://doi.org/10.4271/2012-22-0011>
- Maatouki, I., Fu, S., & Zhou, Z. (2018). *Latest FE model development of THOR-50M crash test dummy*. 15th International LS-DYNA Users Conference.
- Mayrose, J., & Priya, A. (2008). The safest seat: Effect of seating position on occupant mortality. *Journal of Safety Research*, 39(4), 433–436. <https://doi.org/10.1016/j.jsr.2008.06.003>
- McDonald, J. P., Shams, T., Rangarajan, N., Beach, D., Huang, T.-J., Freemire, J., Artis, M., Wang, Y., & Haffner, M. (2003). *Design and Development of a Thor-Based Small Female Crash*

- Test Dummy*. The Stapp Association. 47th Stapp Car Crash Conference (2003).
<https://doi.org/10.4271/2003-22-0024>
- Meng, Y., Yates, K., Dahiya, A., & Untaroiu, C. D. (2025). Effects of rear seat design on occupant injury risk during frontal crashes: A parametric study using finite element analysis. *Mechanics Based Design of Structures and Machines*, 53(1), 753–769.
- Meng, Y., Yates, K., & Untaroiu, C. D. (2022). Numerical Investigation of the Performance of Current Vehicle Rear Seats Using Finite Element Analysis. *SAE International Journal of Transportation Safety*, 10(2), 377–401. <https://doi.org/10.4271/09-10-02-0014>
- Mertz, H., Irwin, A., Melvin, J., Stanaker, R., & Beebe, Ms. (1989). *Size, weight and biomechanical impact response requirements for adult size small female and large male dummies* (Nos. 0148–7191). SAE Technical Paper.
- Mertz, H. J. (2002). Injury Risk Assessments Based on Dummy Responses. In A. M. Nahum & J. W. Melvin (Eds.), *Accidental Injury: Biomechanics and Prevention* (pp. 89–102). Springer.
https://doi.org/10.1007/978-0-387-21787-1_5
- Mertz, H. J., & Irwin, A. L. (2015). Anthropomorphic Test Devices and Injury Risk Assessments. In N. Yoganandan, A. M. Nahum, & J. W. Melvin (Eds.), *Accidental Injury: Biomechanics and Prevention* (pp. 83–112). Springer. https://doi.org/10.1007/978-1-4939-1732-7_4
- Miller, M., Perez-Rapela, D., Gepner, B., Edwards, M., Jermakian, J., & Forman, J. (2021). *A methodology for large-scale parametric evaluation of child booster seats*. 1–2.
- Mitzkus, J. E., & Eyrainer, H. (1984). Three-Point Belt Improvements for Increased Occupant Protection. *SAE Transactions*, 93, 1056–1064.
- Mizuno, K., Yoshida, R., Nakajima, Y., Tanaka, Y., Ishigaki, R., Hosokawa, N., & Hitosugi, M. (2018). *The Effects of Inboard Shoulder Belt and Lap Belt Loadings on Chest Deflection*. The Stapp Association. 62nd Stapp Car Crash Conference. <https://doi.org/10.4271/2018-22-0002>
- Montgomery, D. C. (2017). *Design and analysis of experiments*. John Wiley & sons.
- National Highway Traffic Safety Administration. (2023). *49 CFR Part 572: Anthropomorphic Test Devices*. U.S. Department of Transportation. <https://www.regulations.gov/docket/NHTSA-2023-0031>
- National Highway Traffic Safety Administration. (2026a). *Occupant Crash Protection* (208).
<https://www.ecfr.gov/current/title-49/subtitle-B/chapter-V/part-571/subpart-B/section-571.208>
- National Highway Traffic Safety Administration. (2026b). *Report to Congress: Progress with THOR-05F, WorldSID-05F, and Female Crash Safety Research*.
<https://www.nhtsa.gov/sites/nhtsa.gov/files/2026-01/report-congress-2026-thor-05f-worldsid-05F-female-crash-safety.pdf>
- Nemire, K. (2017). Seat belt use by adult rear seat passengers in private passenger, taxi, and rideshare vehicles. *Proceedings of the Human Factors and Ergonomics Society Annual Meeting*, 61(1), 1644–1648. <https://doi.org/10.1177/1541931213601896>

- Noorsumar, G., Rogovchenko, S., Robbersmyr, K. G., & Vysochinskiy, D. (2022). Mathematical models for assessment of vehicle crashworthiness: A review. *International Journal of Crashworthiness*, 27(5), 1545–1559. <https://doi.org/10.1080/13588265.2021.1929760>
- Östling, M., Lebarbe, M., Compigne, S., Masuda, M., Beillas, P., Eggers, A., Petit, P., Lemmen, P., Richard, O., & Lopez-Valdes, F. J. (2022). In Preparation for Enable New Occupant Seating Positions: The ENOP Project. *Proceedings of the International Research Council on the Biomechanics on Injury (IRCOBI) Conference, 2022*, 2p.
- Petitjean, A., Trosseille, X., Yoganandan, N., & Pintar, F. A. (2015). Normalization and Scaling for Human Response Corridors and Development of Injury Risk Curves. In N. Yoganandan, A. M. Nahum, & J. W. Melvin (Eds.), *Accidental Injury: Biomechanics and Prevention* (pp. 769–792). Springer. https://doi.org/10.1007/978-1-4939-1732-7_26
- Prasad, P. (2015). Injury Criteria and Motor Vehicle Regulations. In N. Yoganandan, A. M. Nahum, & J. W. Melvin (Eds.), *Accidental Injury: Biomechanics and Prevention* (pp. 793–809). Springer. https://doi.org/10.1007/978-1-4939-1732-7_27
- Ramachandra, R., Stammen, J., Kang, Y.-S., Hutter, E., Watkins, L., & Moorhouse, K. (2023). Abdominal biofidelity assessment of 5th percentile female ATD responses relative to recently developed belt and bar loading corridors. *Traffic Injury Prevention*, 24(sup1), S16–S22. <https://doi.org/10.1080/15389588.2022.2160199>
- Ray, M. H., Mongiardini, M., Plaxico, C. A., & Anghileri, M. (2011). *Procedures for verification and validation of computer simulations used for roadside safety applications* (Issue NCHRP Project 22-24).
- Reed, M., & Ebert-Hamilton, S. (2013). Distribution of belt anchorage locations in the second row of passenger cars and light trucks. *SAE International Journal of Transportation Safety*, 1(1), 25–31.
- Rhule, H., Stricklin, J., Moorhouse, K., & Donnelly, B. (2018). *Improvements to NHTSA's Biofidelity Ranking System and Application to the Evaluation of the THOR 5th Female Dummy*. IRCOBI Conference Proceedings.
- Richard, O., Lebarbé, M., Uriot, J., Trosseille, X., Petit, P., Wang, Z. J., Lee, E., Richard, O., Lebarbé, M., Uriot, J., Trosseille, X., Petit, P., Wang, Z. J., & Lee, E. (2022, May 20). *THOR-05F Response in Sled Tests Inducing Submarining and Comparison with PMHS Response Corridors*. 65th Stapp Car Crash Conference. <https://doi.org/10.4271/2021-22-0005>
- Rohatgi, A. (n.d.). *WebPlotDigitizer* (Version 5.2) [Computer software]. Retrieved <https://automeris.io>
- Ryan, J. J. (1963). *Safety Seat Belt*. <https://patents.google.com/patent/US3074761A>
- Schmitt, K.-U., Cronin, D. S., Morrison Iii, B., Callaghan, J. P., & Muser, M. H. (2025). *Trauma Biomechanics: An Introduction to Injury Biomechanics*. Springer Nature Switzerland. <https://doi.org/10.1007/978-3-032-00357-7>
- Schöneburg, R., Franz, U., Haufe, A., Kramer, F., & Zlocki, A. (2026). Integral Safety in the Vehicle Development Process. In R. Schöneburg (Ed.), *Integral Vehicle Safety: Biomechanics –*

- Accident Prevention – Occupant Protection – Sensors—Vehicle Safety in the Development Process* (pp. 529–636). Springer Fachmedien. https://doi.org/10.1007/978-3-658-48100-1_8
- Shah, C., Feng, Z., Hu, G., Jagadish, R., Wang, J., Jagadish, M., Prasad, R., Shelat, C., & Mohan, P. (2017). Development of the THOR-5F crash test dummy FE model and synergy with design support. *Proceedings of the Conference on the Enhanced Safety of Vehicles (ESV)*. 25th International Technical Conference on the Enhanced Safety of Vehicles (ESV).
- Smith, K. M., & Cummings, P. (2006). Passenger seating position and the risk of passenger death in traffic crashes: A matched cohort study. *Injury Prevention*, 12(2), 83–86. <https://doi.org/10.1136/ip.2005.010306>
- Smyth, B. J., & Smith, J. W. (2007). *Developing a Sled Test from Crash Test Data*. SAE International. SAE World Congress & Exhibition. <https://doi.org/10.4271/2007-01-0711>
- Somasundaram, K., Hauschild, H., Driesslein, K., Pintar, F., Richardson, R., & Parent, D. (2023). THOR-05F biofidelity evaluation in reclined and upright seated postures subjected to frontal crash pulses. *Accident Analysis & Prevention*, 191, 107185. <https://doi.org/10.1016/j.aap.2023.107185>
- Soni, A., Schilling, S., & Lucht, A. (2023). *Investigating the Effects of Pre-Pretensioning on Occupant Kinematics and Abdominal Organ Response during Pre-Crash and In-Crash*.
- Tang, K., Tylko, S., & Bussi eres, A. (2021). Responses of the Hybrid III 5th percentile ATD tested outside of regulatory protocols. *Proceedings of the International Research Council on Biomechanics of Injury (IRCOBI) Conference*.
- Tang, K., Tylko, S., & Bussi eres, A. (2022). Comparison of the THOR 5th to the Hybrid III 5th in Full Scale Frontal Crash Tests. *Proceedings of the International Research Council on Biomechanics of Injury (IRCOBI) Conference*, 439–451.
- Tang, K., Tylkoe, S., & Giguere Bussieres, A. (2019). *Modification of the belt system assembly on the test bench to better simulate the response observed in passenger car crashes*. 17th International Conference Protection of Children in Cars, December, Munich, Germany.
- Teng, T.-L., Lan, H.-W., & Yang, S.-M. (2007). Comparative Studies of Finite Element Model of Frontal Impact Dummy. *SAE Transactions*, 116, 2055–2061.
- Transport Canada. (n.d.). *Canada Motor Vehicle Safety Standard (CMVSS) No. 210 – Seat Belt Anchorages*. Government of Canada. Retrieved <https://tc.canada.ca/en/corporate-services/acts-regulations/list-regulations/motor-vehicle-safety-regulations-crc-c-1038>
- Transport Canada. (2025). *National Collisions Database Online 1.0* [Dataset]. <https://wwwapps2.tc.gc.ca/Saf-Sec-Sur/7/NCDB-BNDC/p.aspx?c=100-0-0&l=en>
- Trosseille, X., Petit, P., Uriot, J., Potier, P., Baudrit, P., Richard, O., Compigne, S., Masuda, M., & Douard, R. (2018). *Reference PMHS Sled Tests to Assess Submarining of the Small Female*. The Stapp Association. 62nd Stapp Car Crash Conference. <https://doi.org/10.4271/2018-22-0003>
- Tylko, S., & Bussi eres, A. (2012). *Responses of the Hybrid III 5th female and 10-year-old ATD seated in the rear seats of passenger vehicles in frontal crash tests*. p565-579.

- Uriot, J., Baudrit, P., Potier, P., Trosseille, X., Petit, P., Guillemot, H., Guérin, L., & Vallancien, G. (2006). *Investigations on the belt-to-pelvis interaction in case of submarining*. SAE Technical Paper.
- Uriot, J., Potier, P., Baudrit, P., Trosseille, X., Petit, P., Richard, O., Compigne, S., Masuda, M., & Douard, R. (2015). Reference PMHS Sled Tests to Assess Submarining. *Stapp Car Crash Journal*, 59, 203–223. <https://doi.org/10.4271/2015-22-0008>
- U.S. Department of Transportation. (n.d.). *Federal Motor Vehicle Safety Standard No. 210 – Seat Belt Assembly Anchorages*. National Highway Traffic Safety Administration, Washington, DC. Retrieved <https://www.ecfr.gov/current/title-49/subtitle-B/chapter-V/part-571/subpart-B/section-571.210>
- Valdano, M. (2023). *ENOP Sled model User guide* [User Manual]. MOBIOS Lab, Comillas Pontifical University. https://gitlab.com/piper-project.org/misc_models/-/blob/master/Environment/ENOP_ENV_v1.0/230602_ENOP_Sled_model_manual.pdf
- Wang, Z. J., McInnis, J., Benfant, L., Feng, Z., & Lee, E. (2017). *THOR 5th percentile female ATD design*. Proceedings of 25th International Technical Conference on the Enhanced Safety of Vehicles.
- Wang, Z. J., Below, J., Loeber, B., & Greenlees, B. (2020). *Biofidelity report of the THOR 5th percentile anthropomorphic test device* (DOT HS 812 853). National Highway Traffic Safety Administration.
- Wang, Z. J., Lee, E., Bolte IV, J., Below, J., Loeber, B., Ramachandra, R., Greenlees, B., & Guck, D. (2018). *Biofidelity evaluation of THOR 5th percentile female ATD*. Proceedings of IRCOBI conference.
- Wang, Z. J., Ramachandra, R., Bolte, J. H., & Lee, E. (2021). THOR 5th Female Biofidelity Evaluation in Gold Standard Sled Test Condition. *Proceedings of the International Research Council on the Biomechanics on Injury (IRCOBI) Conference, 2021-September*, 157–176. <https://ohiostate.elsevierpure.com/en/publications/thor-5supthsup-female-biofidelity-evaluation-in-gold-standard-sle/>
- Wietholter, K., Loudon, A. E., & Echemendia, C. (2021). *Development of a Representative Seat Assembly for FMVSS No. 213* (DOT HS 813 096). National Highway Traffic Safety Administration.
- Wietholter, K., & Wetli, A. (2021). *Seat Belt Assembly Tensile Test Procedure Development* (DOT HS 812 925). National Highway Traffic Safety Administration.
- Xu, T., Sheng, X., Zhang, T., Liu, H., Liang, X., & Ding, A. (2018). Development and Validation of Dummies and Human Models Used in Crash Test. *Applied Bionics and Biomechanics*, 2018, 3832850. <https://doi.org/10.1155/2018/3832850>
- Yang, K. H., & Chou, C. C. (2015). Mathematical Models, Computer Aided Design, and Occupant Safety. In N. Yoganandan, A. M. Nahum, & J. W. Melvin (Eds.), *Accidental Injury: Biomechanics and Prevention* (pp. 143–182). Springer New York. https://doi.org/10.1007/978-1-4939-1732-7_6

Youn, Y., Kim, H., Park, J., Lee, J., & Kim, S. (2015). Investigation of rear seat occupant potential injury risk base on seat belt configurations. *Proceedings of the Conference on the Enhanced Safety of Vehicles (ESV)*, (15–0443).