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Civil Aircraft Oblique-Facing Seat Research Summary

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12. Abstract

Title 14 of the Code of Federal Regulations defines regulations that are intended to protect aircraft occupants in the event of a survivable civil aircraft crash. The standards for emergency landing conditions (25.562) were developed with a focus primarily on forward-facing seats, although the rules apply to all seats regardless of installation orientation. Requirements have been implemented for aft-facing seats and purely side-facing seats. When airlines began seeking approval to install seats whose orientation was between forward-facing and purely side-facing, it required the use of special conditions since the regulations that existed at the time were insufficient. Beginning in fiscal year 2011, the Civil Aerospace Medical Institute (CAMI) started a project to support the development of an FAA policy for the approval of oblique-facing seats. The project included testing and modeling at CAMI, two grants to the Medical College of Wisconsin (MCW), a grant to Southwest Research Institute, and a partnership with the Center for Child Injury Prevention Studies (CChIPS). Throughout the length of this research project, the results were used in both FAA policy (PS-AIR-25-27) and an industry standard (SAE AS 6316). This report summarizes the research conducted and the main findings.

Anthropomorphic Test Device testing conducted by CAMI showed that significant injuries are possible for occupants seated in oblique-facing seats who are involved in a crash. Based on tests using postmortem human subjects conducted by MCW, an initial tension limit of 1200 lb in the lumbar spine was proposed. Further research led to the development of a combined spinal tension, flexion, and lateral bending criterion (FAA-LL_{tb}). An FAA-LL_{tb} of 1.88 corresponds to a 25% risk of a serious injury (a threshold consistent with other FAA requirements). This limit can be met by existing technology, such as an effective shoulder belt or a well-designed inflatable restraint. An occupant restrained solely by a lap belt who is allowed to flail forward without any upper torso restraint is at high risk of serious injury. Testing of child seats showed that a 3-year-old Anthropomorphic Test Device is safer in a child seat than restrained solely by a lap belt. For a child in a child restraint system (CRS), deactivated inflatable seat belts did not have detrimental effects on the head, neck, and chest metrics examined; however, a deploying inflatable belt may cause damage to the CRS. Future research is recommended to evaluate the risk of injury for occupants who are too tall for child restraints but too short for proper engagement with the fixed shoulder anchorages typically found on aircraft.

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List of Abbreviations

AIS	Abbreviated Injury Scale
AS	Aerospace Standard
ATD	anthropomorphic test device
CAMI	Civil Aerospace Medical Institute
CChIPS	Center for Child Injury Prevention Studies
CFR	Code of Federal Regulations
CRABI	child restraint and airbag interaction [ATD]
CRS	child restraint system
CT	computed tomography
FAA	Federal Aviation Administration
FAA-LL_{tb}	FAA-Lower Lumbar Spine Combined Metric (tension-bending)
FE	finite element
GHBMC	Global Human Body Models Consortium
HBM	Human body model
HIC	Head Injury Criterion
InFO	Information for Operators
MCW	Medical College of Wisconsin
NHTSA	National Highway Traffic Safety Administration
N_{ij}	Neck Injury Criterion
PMHS	postmortem human subject
SwRI	Southwest Research Institute
v-ATD	virtual anthropomorphic test device

BACKGROUND

Title 14 of the Code of Federal Regulations (CFR) defines regulations that are intended to protect aircraft occupants in the event of a survivable civil aircraft crash (General, 14 CFR §25.561; Emergency Landing Dynamic Conditions, 14 CFR §25.562; Seats, Berths, Safety Belts, and Harnesses, 14 CFR §25.785, 2024). These regulations focus primarily on providing occupant protection for the forward-facing and aft-facing seating directions, although the rules apply to all seats regardless of installation orientation. Alternate seating orientations are considered new and novel and require either special conditions or specific Federal Aviation Administration (FAA) policy. Following research into the injury mechanisms of lateral impacts (DeWeese et al., 2012), additional rules were implemented for purely side-facing seats (FAA, 2012). In the early part of the 2010s, airlines expressed interest in installing seats whose orientation was between forward-facing and purely side-facing. These oblique-facing seats were not specifically addressed in the regulations that existed at this time. Starting in 2014, oblique-facing seats were installed using special conditions (e.g., FAA, 2014; FAA, 2015a; FAA, 2015b). Beginning in fiscal year 2011, the Biodynamics Research Team at the Civil Aerospace Medical Institute (CAMI) was tasked with conducting research and providing data to support the development of an FAA policy for approving oblique-facing seats. This research included multiple series of dynamic testing and modeling of anthropomorphic test devices (ATDs) and postmortem human subjects (PMHSs) across 13 years.

Seat orientation is defined as the installation angle of the seat centerline compared to the aircraft centerline (Figure 1). §25.785 specifies that "each occupant of a seat that makes more than an 18-degree angle with the vertical plane containing the airplane centerline must be protected from head injury by a safety belt and an energy absorbing rest that will support the arms, shoulders, head, and spine, or by a safety belt and shoulder harness that will prevent the head from contacting any injurious object." This led to the definition of forward-facing seats to be ± 18° of the aircraft centerline[1]. The FAA has accordingly deemed any seat installed at an angle greater than 18° to be a side-facing seat. These seats have been further subdivided; for instance, there is FAA policy for fully side-facing "i.e., seats oriented in the aircraft with the occupant facing 90 degrees to the direction of aircraft travel" (FAA, 2012) and an SAE International Aerospace Standard (AS) that limits side-facing seats to be "seats which are installed at 90 degrees ± 10 degrees with respect to the longitudinal axis of the aircraft" (SAE International, 2023). Seats between 18° and 80° are generally referred to as oblique-facing, with FAA policy and an industry standard currently limited to seats installed from 18° to 45° (FAA, 2018; SAE International, 2017).

[1] Amendment 25-15 to 14 CFR Part 25, dated October 24, 1967, introduced the requirements for side-facing seats. Amendment 25–20, dated April 23, 1969, changed the definition from sideward-facing seats to a seat that is positioned at more than an 18-degree angle to the vertical plane containing the airplane centerline.

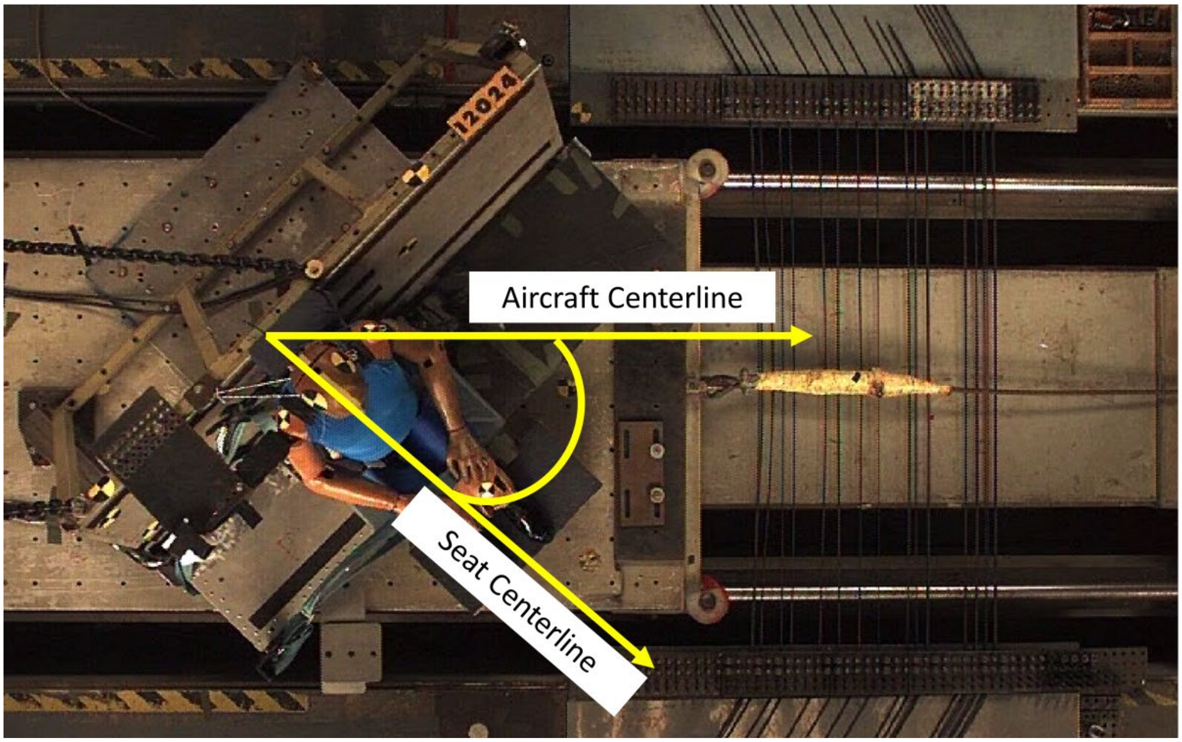

Figure 1: Seat Orientation

At the onset of this research, there was no known research on oblique-facing aircraft seats. The automotive literature does address oblique impacts (e.g., Cavanaugh et al., 1990; Cavanaugh et al., 1993; Maltese et al., 2002; Pintar et al., 2006; Pintar et al., 2007; Viano, 1989; and Yoganandan et al., 2013); however, the automotive environment has relatively lower changes in velocities relative to the §25.562 requirement (Humm et al., 2016b). That literature is split into two basic categories: near-side (e.g., Yoganandan and Pintar, 2008) and far-side (e.g., Pintar et al., 2007; Forman et al., 2013). During a near-side oblique impact, the occupant motion is into the shoulder belt, which greatly limits occupant flail. The door and any side/curtain airbags also contribute to the limited flail. The primary injury concern, at least in terms of injuries not addressed by existing injury criteria, is rib fracture. During far-side oblique impacts, the occupant motion is out of the shoulder belt, allowing for greater torso motion and resulting in additional injury risks, including injuries to internal organs and the spine. Based on discussions with subject matter experts at the Medical College of Wisconsin (MCW), it was determined that the existing literature was insufficient to determine the types of injuries that could occur in aviation oblique-facing seats.

Injury Criteria Review

To protect passengers in a crash, researchers develop injury criteria. These are engineering parameters, such as force or acceleration measured with an ATD, that are related to a risk of injury (typically using injury and no-injury data from PMHSs). Given a relationship between risk of injury and the engineering measurement, a pass/fail limit can be selected based on a chosen level of risk. For instance, many of the injury criteria referenced in FAA requirements have approximately a 25% risk of a serious injury (defined as an Abbreviated Injury Scale (AIS) 3+). Injury criteria are only valid for the environment and loading for which they were derived and are dependent on the specific ATD for which they were developed. Any change to the boundary conditions, the loading, or the ATD will require a reassessment of their applicability to the new exposure (Pellettiere, 2012).

Due to the multi-axis loading of oblique-facing seats, it is possible that injury criteria from both frontal and lateral impacts are useful in evaluating their safety, as well as the possibility that new criteria are needed. At the onset of this research, there were no known injury criteria specifically designed for oblique loading; however, numerous existing criteria that have been applied to other aviation seats may be applicable. This section provides an overview of the existing criteria that are under consideration for the approval of oblique-facing aircraft seats.

Head Injury

The Head Injury Criterion (HIC) was designed for frontal impacts; however, the same calculation and regulatory limit applies for frontal and lateral impacts. For frontal tests, a frontal ATD is used to calculate HIC (e.g., the FAA Hybrid III), and for lateral impacts, a lateral ATD is used (e.g., the ES-2re). The shape of the headforms for the FAA Hybrid III and ES-2re are similar; the main difference is that the calibration of the head acceleration is directional (fore-aft for the FAA Hybrid III and lateral for the ES2-re) and the primary point of contact is different—the center of the forehead in the coronal plane for the FAA Hybrid III and the sagittal plane for the ES-2re (Anthropomorphic Test Devices, 49 CFR §572, 2022). Since the HIC calculation includes the head acceleration in all three principal directions, neither the directional nature of the impact nor the directional nature of the calibration test affects the validity of the calculation. For an impact on the ATD seated in an oblique-facing seat, the HIC calculation and limit are valid as long as the impact is primarily in the direction for which the ATD was calibrated and the impact location is near that of the calibration test. The limitations for ATD calibration for the off-axis angle of an oblique impact are not known. Since the directional nature of impact and calibration tests were considered valid for this research, HIC was used for assessments of head impact injury.

Neck Injury

FAA policy prohibits concentrated loading on the neck in addition to the use of performance-based requirements. For frontal impacts, the Neck Injury Criterion (N_{ij}) evaluates the loads in the neck for tension/compression and flexion/extension (Occupant Crash Protection, 49 CFR §571.208, 2022). There are also limits on pure tension/compression. For oblique loading that results primarily in flexion/extension, N_{ij} should be applicable. For lateral impacts, the FAA developed a separate injury criterion that includes a tension limit that is significantly lower than the frontal tension limit: 405 lb vs. 937 lb (FAA, 2012). This is because the lateral bending of the neck reduces its tolerance to tension. The FAA side-facing seat policy does not define the amount of lateral bending necessary to reduce the neck tolerance, but for purely side-facing seats, it is prudent to assume that the bending pre-condition is met. The level of lateral bending required for the neck tolerance to shift from the forward to the lateral value is unknown. The side-facing policy also includes neck compression, bending, and shear stand-alone limits.

Significant twisting of the neck has been observed in some research and development tests of oblique-facing seats. FAA special conditions include a limit on the amount of head rotation about the neck Z axis (105°). The limit is based on human neck torsional strength data reported in SAE Technical Paper No. 892437 (Myers et al., 1989).

Chest Injury

In forward-facing impacts, shoulder belt loads correlate with both internal injuries and bony fractures (Kuppa, 2004). In side-facing impacts, rib fractures, clavicle fractures, and carotid artery intimal tears were observed (Philippens et al., 2011). Based on data summarized in DeWeese et al. (2012), the shoulder belt load limit for forward-facing seats (1750 lb for a single belt) is also used for side-facing seats (along with a recommendation to measure the belt load with the FAA-Hybrid III instead of the ES-2re). Based on this, it is reasonable to conclude that the belt limit is valid for oblique loading. The side-facing policy also limits the deflection of the thoracic ribs in the ES-2re. These rib potentiometers are linear and may bind; therefore, they produce faulty data at angles less than 60° from the aircraft centerline (National Highway Traffic Safety Administration [NHTSA], 2006).

Femur Injury

14 CFR §25.562 limits the compressive axial force in the femur; however, it is unlikely to be exceeded in a Part 25 certification test (Taylor, DeWeese, & Moorcroft, 2015). The intent of the femur axial force limit in §25.562 is to reduce the chance of leg injuries that could impede egress after an emergency landing (Soltis and Nissley, 1990). The significant lateral orientation of oblique and side-facing seats alters the inertial force vector of the legs during a longitudinal impact, making it very unlikely that the compressive load limit would be exceeded. However, the lateral orientation introduces axial rotation of the femur. Tests of aircraft side-facing seats using PMHSs have sustained serious leg injuries that would not only have impeded egress but could also be life-threatening (Philippens et al., 2011). Based on that testing, the side-facing policy limits the axial rotation of the upper leg (femur) to 35° in either direction from the nominal seated position. The lateral component present in oblique-facing seats may result in axial rotation of the femur and likely carry the same risk as side-facing seats. It is reasonable to apply the femur axial rotation limits to these seat orientations.

Lumbar Spine Injury

The lumbar spine is at risk for a compression fracture during impacts with a significant vertical component. The 1500 lb limit measured with the FAA Hybrid III or equivalent is dependent on the vertical component of the impact and is used as a criterion for forward-facing, aft-facing, and side-facing seats. Therefore, it is appropriate for the range of oblique-facing seats.

Torso Injury

Significant contact between the occupant's back and surrounding hard structure has been observed during rebound in some development tests of oblique-facing seats. Research sponsored by the National Aeronautics and Space Administration (NASA) indicates that concentrated loads in excess of 820 lb applied to the lower back can cause spinal injuries. While direct measurement of spine contact loads is not currently possible, linear acceleration near the torso center of gravity can be measured. The 50th percentile ATD used to test aircraft seats has a torso weight specification of 41.5 lb ± 1.6 lb. One means of estimating the force resulting from contact between the torso and a seat item is to multiply the torso mass by the peak torso acceleration during the contact. Limiting the torso rearward acceleration to 20 g would, in turn, keep the contact forces below the level observed to cause significant injury in current studies. It is reasonable to use a rearward acceleration limit of 20 g as a criterion for avoiding spinal injuries.

Other Injury Concerns

PS-ANM-25-03-R1 contains several other provisions, such as limits on abdominal forces and lateral pelvis compression. These criteria are appropriate for any test using the ES-2re, although as the angle decreases from 90° from the aircraft centerline, it becomes increasingly unlikely that there is sufficient lateral momentum to exceed the limit (Moorcroft, 2013). The existing rules for any seat orientation also contain more qualitative requirements, such as a restriction on body-to-body contact, ensuring the occupant does not translate beyond the edge of the seat bottom, and ensuring the belts remain on the pelvis (for the lap belt) and torso (for a shoulder belt). These requirements would also be valid for oblique-facing seats.

CAMI TESTING

To support this project, the Biodynamics Team completed four separate physical evaluations: a static evaluation of ATDs for potential use in dynamic testing in 2012, lap belt-only dynamic tests at 30° and 45° in 2012, lap and shoulder belt dynamic testing in 2016, and dynamic testing with an inflatable lap belt in 2023 as a potential mitigation strategy (Carroll et al., 2024). All tests were conducted using the 14 CFR §25.562 longitudinal impact (16 g, 90 ms rise time, 44 ft/s velocity change) and run following the SAE J211/1 instrumentation protocols, which follows the SAE J1733 sign conventions (SAE International, 2014; SAE International, 2007).

Static Evaluation of ATDs

Four ATDs were statically evaluated for potential application in oblique-facing seat tests: Hybrid II, FAA Hybrid III, ES-2re, and THOR.

Hybrid II
The Hybrid II is a 50th percentile mid-sized male defined by 49 CFR §572 Subpart B. It has limited instrumentation with accelerometers in the head and pelvis and load cells in the lumbar spine and femur. The concern with the Hybrid II is that it would be unable to assess injuries in the thorax or any loading in the neck due to the limited instrumentation. However, this ATD is fairly robust and was initially used to assess gross occupant kinematics.

FAA-Hybrid III
The FAA-Hybrid III is a mixture of the Subpart B Hybrid II and Subpart E Hybrid III ATDs, as specified in Gowdy et al. (1999). It has an expanded set of instrumentation, which allows for a greater ability to measure potential injury risks. It includes accelerometers in the head and pelvis and load cells in the lumbar spine and femur, similar to the Hybrid II. The FAA-Hybrid III also includes additional load cells in the neck, thorax, and lower leg. Based on the risk of neck injuries seen in side-facing seat tests and the possibility of airbags being used as a mitigation strategy (which require the measurement of neck loads), the FAA Hybrid III is a better option than the Hybrid II for oblique-facing seats that are primarily forward-facing.

ES-2re
The ES-2re ATD is specially designed to evaluate injury in test conditions with significant lateral loading and is defined by 49 CFR §572 Subpart U. This ATD is cited in FAA policy PS-ANM-25-03-R1 for side-facing seats and 49 CFR §571.214 for use in automotive side impact tests. The ES-2re, being designed for purely lateral loading, was excluded from the impact tests due to concerns over permanent damage to rib sliders and the abdomen if any amount of forward flailing occurs. This was particularly a concern for the lap belt-only tests. Its performance instead was evaluated numerically in a separate project (Moorcroft, 2013). NHTSA has tested the ES-2re at angles beyond purely lateral (90°) and concluded that it has acceptable performance up to ±30° from lateral (NHTSA, 2006).

THOR
The THOR is an advanced ATD that has been under development from 2009 through 2020 (Humanetics, n.d.). At the time of the static evaluation, the NT version was current; however, that has been supplanted by the K and M versions. As of this writing, the United States has not

codified the THOR; however, the THOR-50M is required by the European and Australian new car assessment programs (Humanetics, n.d.). The THOR was developed for use in the more compact automotive interior environment, where flail is limited. It has an extensive array of instrumentation, particularly in the thoracic and abdominal regions. While this instrumentation would greatly increase the types of injuries that could be predicted, it was excluded after visual inspection of the internal structures revealed that extensive damage to the instrumentation and the lumbar spine element would likely occur during any test that induced significant lateral or forward flexion at the lumbar. Figure 2 documents those concerns, where the top left photo shows the ATD fully bent over, the top right photo shows a potential wire pinch point, the bottom left photo shows a stress concentration point of the lumbar spine, and the bottom right photo shows abdominal instrumentation interaction with ribs.

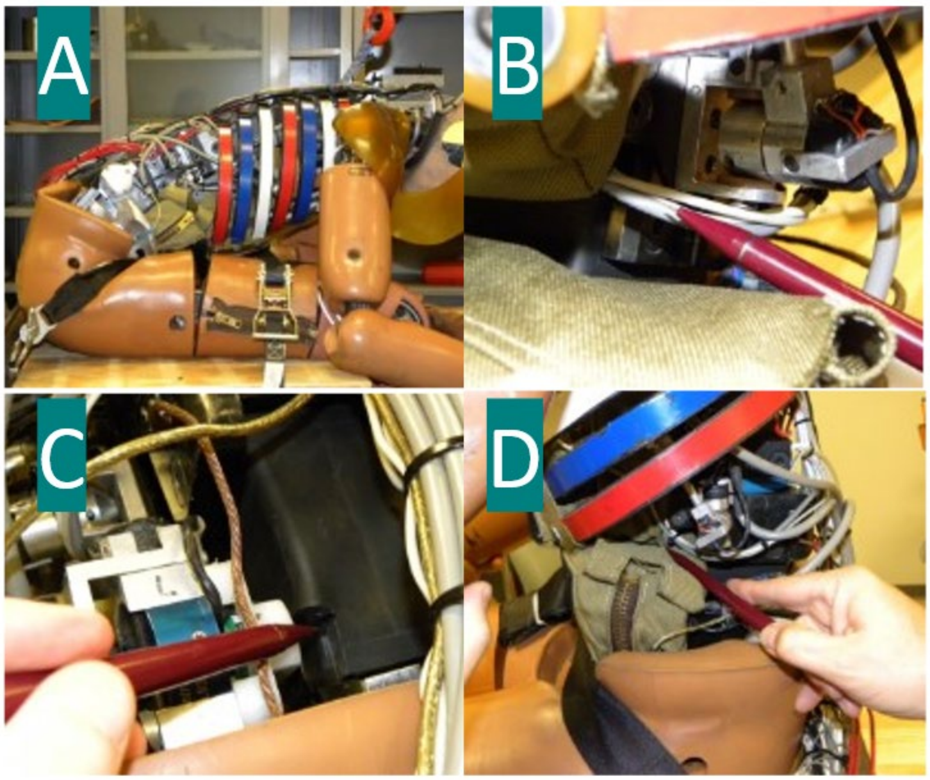

Figure 2: THOR-NT Hardware Interactions

Lap Belt-Only Testing

Based on the static evaluation, only the Hybrid II and FAA-Hybrid III were used for dynamic testing conducted in 2012. Due to its robustness and the availability of spare parts, the initial tests used the Hybrid II to determine gross kinematics. Subsequent tests used the FAA-Hybrid III to gather loads throughout the spine (cervical, thoracic, and lumbar). Based on an informal review of oblique-facing seat designs that were under development at the time of initial test preparation, two configurations that were considered the worst case[2] were tested. The first configuration

[2] Worst-case for the upper body, not necessarily the legs. Significant leg injury was noted in the earlier side-facing research when the legs were allowed to flail. The decision to restrain the legs was made because many designs included features where the legs are contained by the surrounding structure to simplify the PMHS testing, and because the existing leg injury criteria were considered sufficient.

emulates a seat that is rigid from the armrest level down to the floor and open or very flexible above it (left photo in Figure 3). This configuration includes structure that does not provide support to the occupant (Figure 4). The second configuration emulates a scenario where an occupant is seated next to an interior feature that restricts the motion of the legs and pelvis but not the torso (right photo in Figure 3). This configuration includes seats that are wider than typical (allowing significant pelvis motion during an emergency landing) or ones with ineffective armrests. The seat dimensions were based on the results of a survey of the SAE SEAT committee and documented in DeWeese et al. (2007). The first configuration included a conventional lap belt. The second configuration included a conventional lap belt combined with a body-centered lap belt (Figure 5)[3].

Figure 3: Lap Belt-Only Test Setup: Armrest (left) and No Wall (right)

[3] Body-centered lap belts are most commonly found on side-facing seats. The leading belt anchor is close the center of the seat place as opposed to being outside of the hips. This reduces slack in the belt during an emergency landing, which, in turn, reduces the motion of the pelvis.

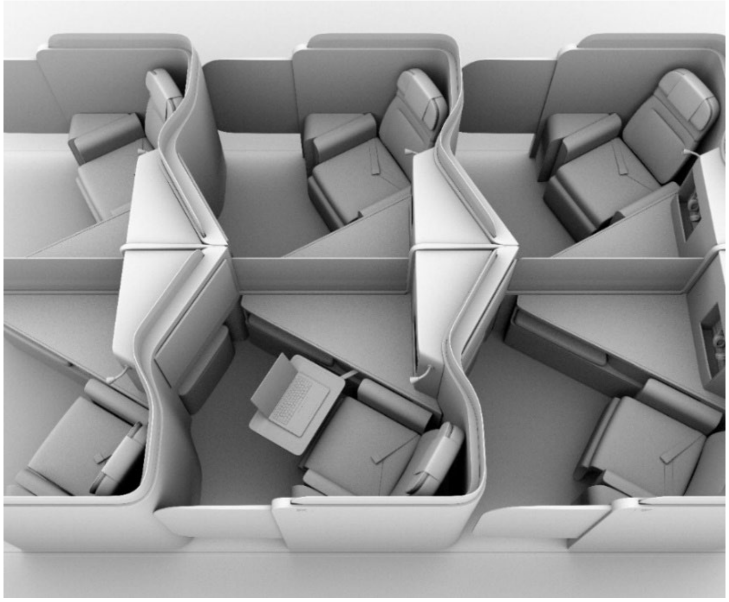

Figure 4: Rendering of Oblique Seats Where the Surrounding Structure May Not Support the Occupant (chesky - stock.adobe.com)

Figure 5: Belt Routing Highlighting Conventional and Body-Centered Lap Belts (Arrows)

Both test configurations produced numerous hardware failures, high neck loads, and high lumbar tension (Carroll et al., 2024). Based on this initial testing, there was concern for potential injuries, including fractures to the sternum, pelvis, and ribs due to contact with a structure, leg fractures due to inertial loading, head and neck injuries due to either direct contact or inertial loading, and injuries to internal organs due to contact with a structure. Based on the results of the initial PMHS testing and data collected by the MCW (discussed below), it seemed likely that conventional lap belt-only seating configurations would not be able to provide an acceptable level of safety.

Lap and Shoulder Belt Testing

Noting the results from the lap belt testing, a follow-on series of tests was conducted in 2016 using the same two configurations from 2012 but including an added shoulder belt across the leading shoulder, which has a fixed attachment adjacent to the lap belt buckle (Figure 6). The shoulder belt was routed through a belt guide that was approximately two inches forward of the ATD centerline, and the anchor was mounted to a rigid structure (Figure 7).

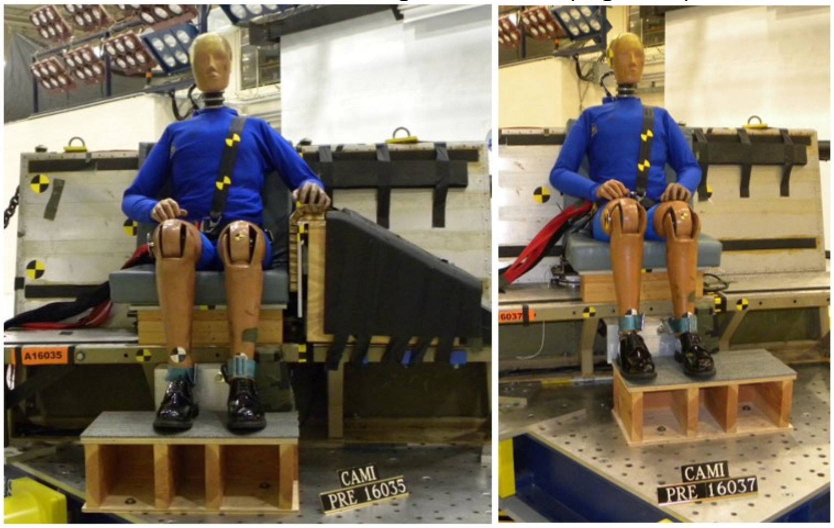

Figure 6: Lap and Shoulder Belt Test Setup: Armrest (left) and No Wall (right)

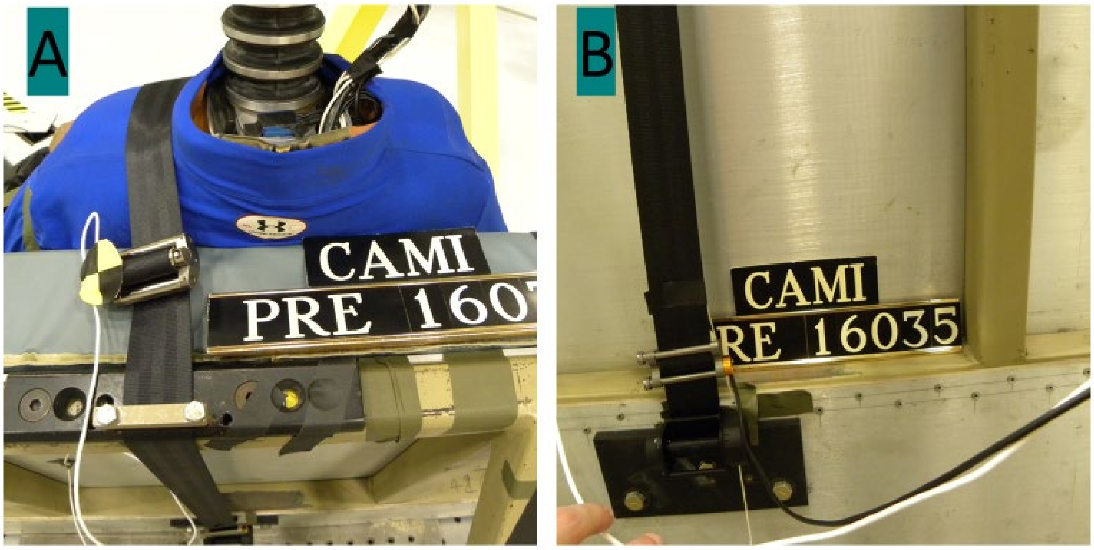

Figure 7: Location of Shoulder Belt Guide Relative to ATD Centerline (A) and Anchor (B)

As expected, the presence of a rigidly mounted shoulder belt greatly reduced the flail of the occupant (Carroll et al., 2024). While the lap belt-only tests produced lumbar tension between 1750 lb and 3100 lb (exceeding the 1200 lb limit)), the lap and shoulder belt tests instead produced compression loads (that remained below the 1500 lb limit). For example, in the 45° armrest configuration, the lumbar load dropped from 2841 lb of tension in test A12025 to 721 lb of compression in test A16035. The reduced flail also limited torso contact with the surrounding structure.

Inflatable Lap Belt Testing

Seeing the large differences between the lap belt-only and lap and shoulder belt tests, the Biodynamics Team tested an alternate restraint approach. A prototype wedge inflatable lap belt was donated by AmSafe, Inc. (Phoenix, AZ). Four tests were run with a 45° seat installation angle and an armrest. The airbag limited torso flail, thus reducing the lumbar tension compared to the lap belt tests (Carroll et al., 2024).

CAMI COMPUTER MODELING

The Biodynamics Team also conducted two modeling and simulation (M&S) projects to support the overall oblique research project. The first M&S project was an evaluation of the ES-2re for impact angles less than pure lateral (90°), and the second evaluated the FAA-Hybrid III at impact angles ranging from 0° to 60°.

Simulated ES-2re

Due to concerns regarding damaging the physical ES-2re ATD when tested outside of its design range, a series of simulations were instead used to evaluate whether the ES-2re would provide responsive results and withstand testing on oblique-facing seats (Moorcroft, 2013). In the simulations, the ES-2re was placed next to a full wall with and without a lap and shoulder belt and exposed to a pure lateral impact (90°) at the 14 CFR §25.562 longitudinal impact. This test matched CAMI test A05065 (DeWeese et al., 2007). The interaction between the virtual ATD (v-ATD) and the wall was similar to the physical ATD from an injury criteria perspective and was mostly conservative (i.e., the simulations produced higher loads than the physical tests). Differences were noted, but the model was deemed to be sufficient for comparing seat orientation and conducting trend analysis of injury criteria and flailing while avoiding unnecessary damage to physical ATDs.

Simulations were run at 90°, 75°, 60°, 45°, and 30° with respect to the aircraft centerline. The ES-2re v-ATD suggests that the injury criteria listed in PS-ANM-25-03-R1 are unlikely to be exceeded for any angle between 30° and 90° when seated next to a padded full wall with a lap and shoulder belt and body-centered lap belt. The ES-2re appears to be capable of avoiding internal damage in this configuration due to a limited range of motion, as shown in the results. Conversely, simulations without a shoulder belt resulted in a significantly larger range of motion, indicating that the ES-2re v-ATD is not capable of flailing in a realistic manner, as shown in the 75° and lower-angle simulations (Figure 8). This is because the abdomen is rigid plastic and contacts the ribs during forward flail, preventing the ES-2re from bending in a human-like manner. Internal damage to the physical ATD's ribs is likely to occur in these configurations.

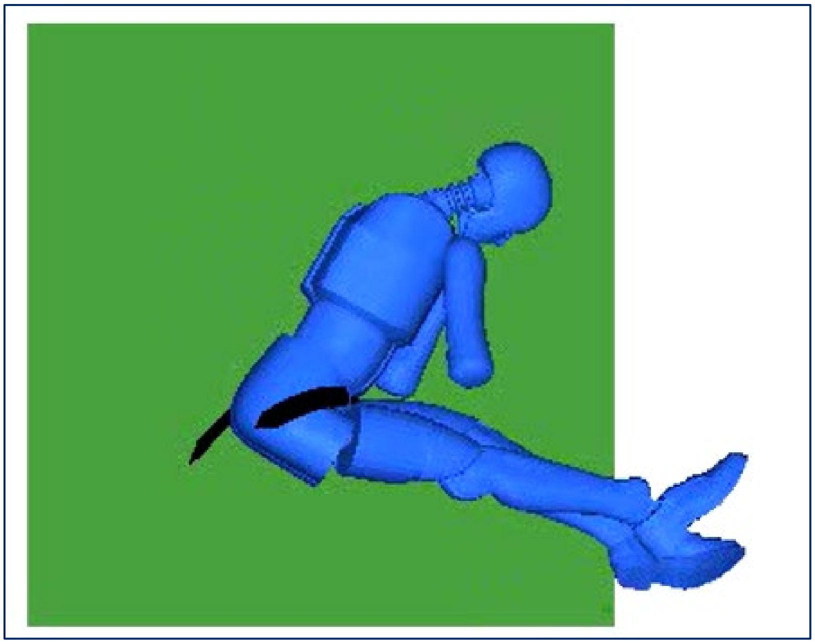

Figure 8: Simulated ES-2re Showing Lack of Bending in the Thoracic Spine

Simulated FAA-Hybrid III

Following the CAMI testing noted above and the results of the first grant to MCW detailed below, the expectation was that certification tests for oblique-facing seats[4] would require the FAA-Hybrid III ATD. Since the aviation seating industry is increasing their use of M&S in certification, the Biodynamics Team performed a series of simulations of the FAA-Hybrid III to determine if the evaluated v-ATD is appropriate for use in these scenarios (Moorcroft, 2021).

The simulations included the Madymo multi-body FAA-Hybrid III model placed in a rigid seat with and without an armrest and restrained with and without a shoulder belt (example shown in Figure 9). The lap belt was a combination of body-centered and conventional lap belts. Simulations were run with seat impact angles of 0°, 30°, 45°, and 60° at two pulse levels for a total of 32 runs. The pulses were the Part 25 longitudinal pulse, which has a symmetric triangular profile rising to 16 g in 90 milliseconds, and a lower-severity impact pulse was achieved by scaling the peak by 60% (9.6 g peak at 90 milliseconds). This lower pulse was used by MCW for the no-injury tests. A subset of eight simulations were compared with the test data to define a baseline of model correlation. One test-simulation pair was defined for the seat with and without an armrest, with and without a shoulder belt, and at 30° and 45°. All these simulations were performed using the 16 g pulse. Overall, the model results were significantly different from the similar physical test results described above. The model under-predicted spinal loads with a shoulder belt and had a different distribution of spinal loads in the lap belt-only configurations. The lack of correlation is to be expected considering that neither the physical nor v-ATDs were designed for this type of off-angle loading (i.e., they were developed with a focus on pure fore-aft loading) and, at the time of the project, the developer of the v-ATD did not have a business case to invest into expanding the correlated range to the physical ATD. The Madymo FAA Hybrid III v-ATD that was used in this project would require further evaluation and possible changes before it could be used to

[4] At least for seats installed from 18° to 45° with respect to the aircraft centerline.

accurately represent a physical ATD in oblique loading configurations. Because of these limitations, the results of this trend analysis are considered preliminary (Moorcroft, 2021).

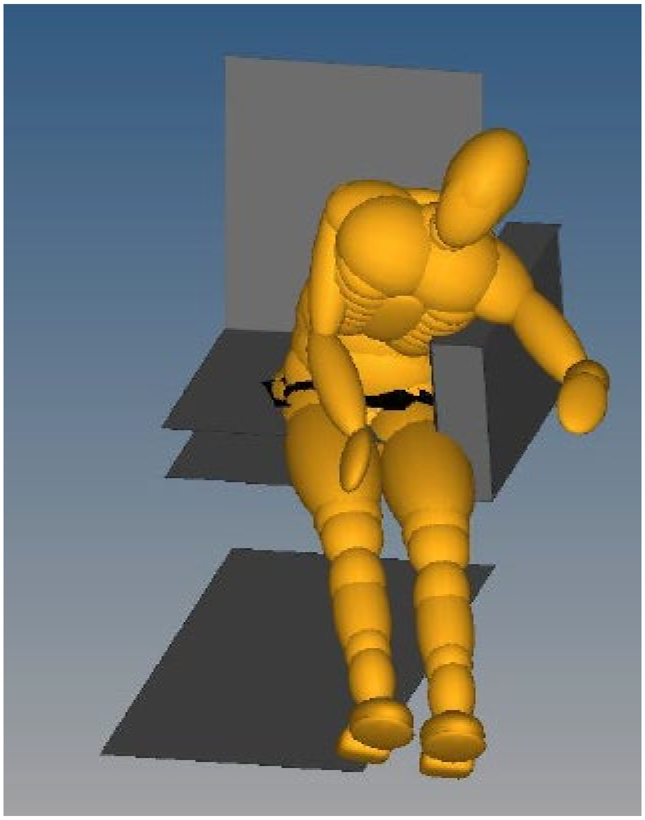

Figure 9: Simulated FAA-Hybrid III: Armrest, Lap Belt-Only, 45°

The full set of 32 simulations was used to evaluate trends based on the configurations and to observe the performance of the model as a function of seat installation angle. The v-ATD model was robust across the range of configurations simulated for this project. No crashes were encountered, and only one simulation produced an apparent numerical instability. The use of a shoulder belt produced compressive spinal loads that did not approach the limit defined in 14 CFR §25.562. For configurations with only a lap belt, the lumbar loads exceeded the 1200 lb tension limit proposed by MCW after their initial testing (as discussed below) for all angles from 0° to 60°. The trend analysis for the lap belt-only configurations suggests that lumbar tension load decreases with impact angle, which, if reproducible, might assist in defining a critical installation angle for certification testing for the current injury criteria. When a shoulder belt was included, the spinal loads became compressive and were generally low. The model with the shoulder belt produced mostly linear results across impact angles, showing the expected trends.

In the 0° orientation, the occupant rolled out of the shoulder belt (Figure 10). During a certification test, this result would not be acceptable. The observed interaction may be due to the location of the shoulder harness to lap belt attachment point (near the ATD centerline), which was designed for purely side-facing seats and not optimized for a forward-facing seat. While current seats on Part 25 aircraft are designed for a single installation angle or a small range of angles, results like this are useful for designers of seats that swivel (such as those that have been proposed for self-driving motor vehicles and advanced air mobility aircraft).

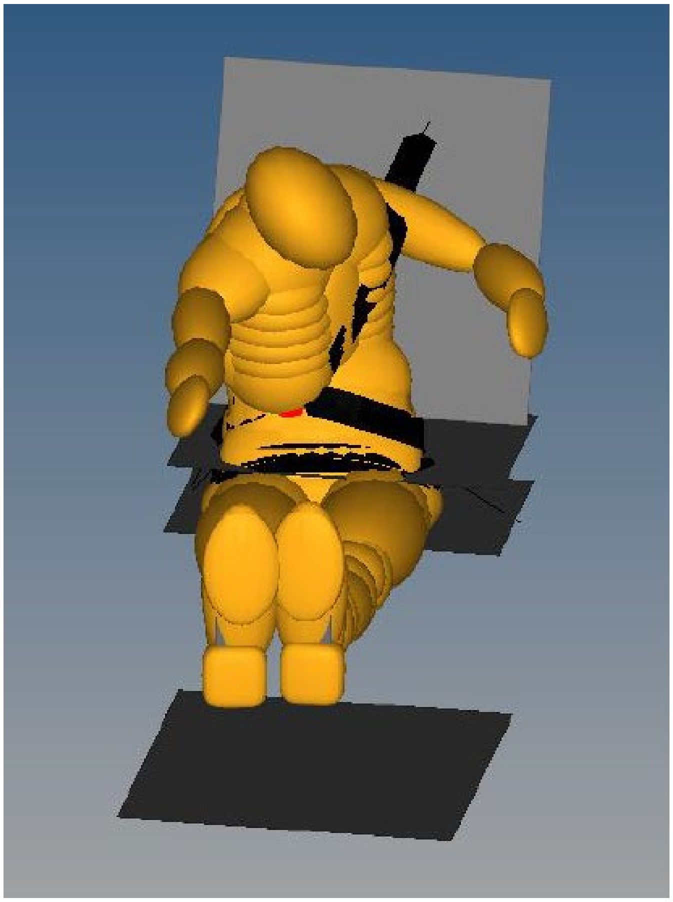

Figure 10: Shoulder Belt Sliding Off Occupant During 0° Simulation[5]

INJURY CRITERIA DEVELOPMENT

MCW was awarded consecutive Aviation Research Grants (13-G-005 and 17-G-002) to investigate occupant safety in obliquely mounted aircraft seat systems (FAA, n.d.). All MCW's human subject research was approved by their Institutional Review Board.

Initial PMHS Testing at MCW

As part of the first grant (13-G-005), four lap belt-only PMHS tests were conducted using seat configurations matching the 2012 CAMI tests, as summarized in Table 1 (Humm et al., 2015a, Humm et al., 2016b). The first test was designed as the worst-case oblique scenario for torso flailing, with an unconstrained torso and fixed pelvis and legs. This represented a condition where the shoulder belt slipped off the shoulder or was ineffective, and the pelvis and lower extremities were restrained by contact with adjacent furniture. Two different lap belt geometries were used. A body-centered belt used one anchor point at the midline of the occupant. This belt wrapped around the right/leading side of the pelvis and was secured to the left seat belt anchor. A second

[5] These multi-body simulations used a force-penetration contact that permits one body to penetrate another while calculating the forces generated from such penetration. This differs from methods that deform the body (such as finite element analysis) and produces visualizations such as this where the pelvis penetrates the seat cushion, and the seat belts penetrate the occupant.

belt, more typical of a standard frontal lap belt restraint, went around the occupant from the left to the right anchor. The second test substituted the body-centered belt with an armrest to constrain pelvis motion. The third test used the two-belt configuration of the first setup but removed the lower extremity restraint to observe the interaction of the thorax and the lower extremity flail. The fourth test was a repeat of the first but at a lower severity pulse. These whole-body PMHS tests served to describe the injuries and occupant kinematics associated with aviation-relevant impact scenarios. In addition to typical sled testing instrumentation and video coverage, each test included functional radiographs and close-up computed tomography (CT) images. An autopsy was conducted by a forensic pathologist at the Milwaukee County Medical Examiner's office to identify and catalog injury. Significant, potentially fatal injuries were observed in all the tests conducted at the full regulatory pulse of 16 g (100% pulse). According to the Abbreviated Injury Scale (AIS), the observed spinal distraction injuries could be rated as AIS 3 injury, which is categorized as a serious injury (Association for the Advancement of Automotive Medicine, 2015). The spinal distraction injuries were estimated to be a combined result of multi-axis bending and tension (Humm et al., 2016b). Schematics of the injury locations are presented for specimen 1 (Figure 11) and specimen 3 (Figure 12). The red marks indicate the approximate location of the injury. To generate the no-injury data needed to derive an injury criterion, a test was run at a lower pulse. The peak acceleration was scaled 60% while keeping the duration constant, resulting in a peak of 9.6 g. No injuries were observed in this test.

Table 1: Configurations for Initial PMHS Tests

Specimen	Angle (°)	Pulse (%)	Conventional Lap Belt	Body-Centered Belt	Armrest	Leg Restraint
1	45	100	Yes	Yes	No	Yes
2	45	100	Yes	No	Yes	Yes
3	30	100	Yes	Yes	No	No
4	45	60	Yes	Yes	No	Yes

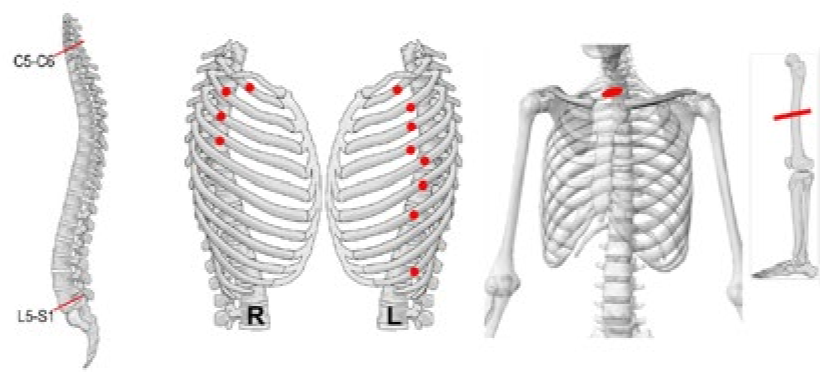

Figure 11: Schematic of Injury Locations for Specimen 1

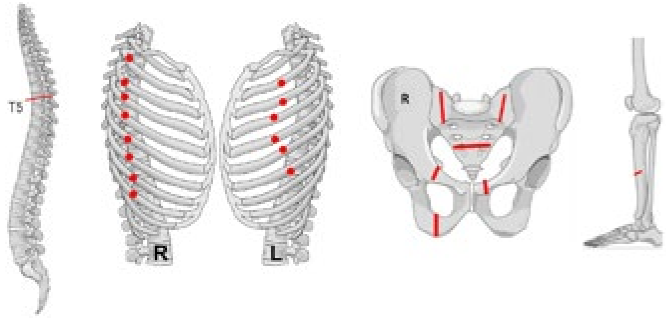

Figure 12: Schematic of Injury Locations for Specimen 3

Concurrent with the PMHS tests, MCW conducted matched-pair testing with the FAA Hybrid III (Humm et al., 2015b). This testing included four different pulse severities scaled by peak acceleration (100%, 75%, 60%, and 50%). By matching a test's initial and boundary conditions, the injuries observed during the PMHS testing can be related to the engineering measures recorded in the ATD testing to ultimately establish a risk curve. The limited number of tests conducted did not provide sufficient data to establish statistical significance. Consequently, a risk curve could not be constructed. Further testing could provide the needed data. However, due to the FAA's need for objective performance requirements for oblique-facing seat special conditions, the data from the 60% tests were used to define a preliminary injury criteria limit of 1200 lb (tension) in the lumbar spine of the FAA Hybrid III. This criterion was adopted in multiple FAA Special Conditions and ultimately proposed in FAA policy and an SAE International Aerospace Standard, as noted in the Background section of this paper.

Grant 13-G-005 also included two PMHS tests, along with accompanying ATD tests, of a front-row (infinite setback) occupant. One test was with a conventional lap belt and one with a y-belt, both using the regulatory 16 g pulse (Humm et al., 2016a). Y-belts are commonly found on seats directly behind a bulkhead. The addition of a second, higher anchor point reduces the forward motion of the occupant (Figure 13). Once again, significant injuries were produced, including bilateral femoral shaft fractures in the lap belt test (Table 2). These tests highlighted that free flail of an occupant at 16 g is likely to be unsafe, regardless of impact angle. This result supported the publication of an FAA Special Airworthiness Information Bulletin (FAA, 2023).

Table 2: Injury Summary for Frontal PMHS Occupants

Test ID	Spine	Ribcage	Pelvis	Extremities
FOC134	T4-T5 Transection	Bilateral rib fractures (1-8) with flail chest	None	Bilateral proximal femoral shaft fractures
FOC139	L5-S1 Transection	Left rib fractures (2-5) with flail chest	None	None

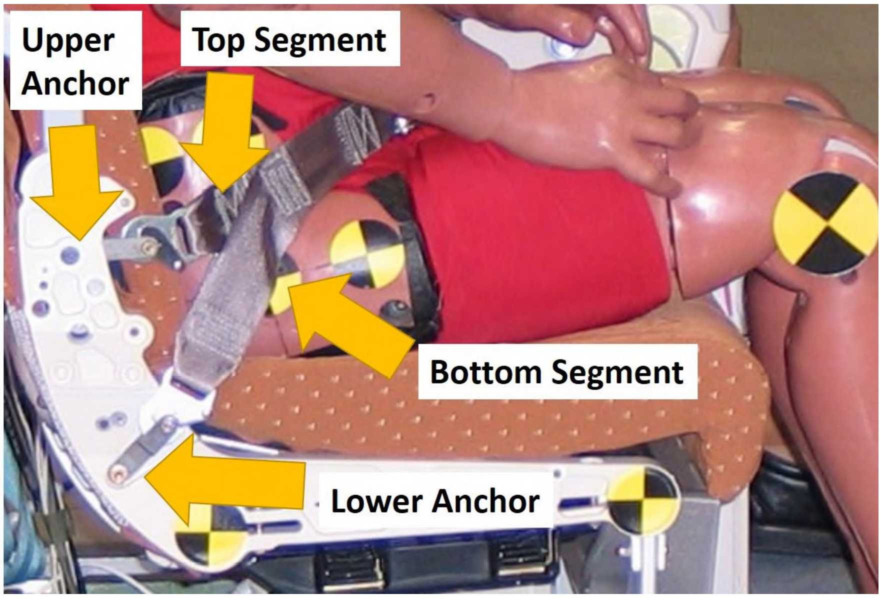

Figure 13: Typical Y-Belt Installed in Aircraft Seat

Lastly, grant 13-G-005 included testing of isolated lumbar spines (T12 to sacrum) to evaluate different loading angles. Isolated spines were chosen for this testing because it was not feasible to use whole-body PMHS tests to collect the data necessary to determine a statistically significant injury criterion. The initial tests demonstrated that sacrum moment decreased as the flexion angle changed from flexion (0°) to lateral bending (90°). These tests were continued as part of a subsequent grant and are further described below.

Follow-on PMHS Testing at MCW

Due to the complex nature of the oblique loading environment, along with time and funding limitations, the first grant to MCW was only able to produce a conservative tension limit (1200 lb). While this limit allowed for active certification projects to proceed, industry and the FAA desired a more robust criterion. As a result, a second grant was awarded to MCW to continue the research (17-G-002).

This grant included whole-body and component-level tests on PMHSs, along with computational modeling and matched paired ATD tests. The isolated component tests were performed to examine the injury mechanisms and determine failure loads that could not be measured in the whole-body tests[6]. The computational study served to determine the effect of pulse severity and seat orientation on the responses and injuries occurring to the spine. Finally, the matched paired FAA-Hybrid III ATD tests served in developing injury criteria for the lower lumbar spine load cell.

Whole-Body Sled Tests

Three additional oblique whole-body tests were conducted, bringing the total to seven. Table 3 defines the test conditions for all seven tests. As seen in Table 4, tests at 70% and 100% of the Part 25 longitudinal pulse produced numerous injuries. Tests at 60% of the regulatory pulse did not produce observable injuries.

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[6] Internal loads cannot be measured in PMHS, and estimating loads in the lumbar spine is difficult due to mass recruitment and accelerometer placement.

Table 3: Configurations for All Whole-Body PMHS Tests

Test ID	Angle (°)	Pulse (%)	Conventional Lap Belt	Body-Centered Lap Belt	Armrest	Leg Restraint
FOC104	45	100	Yes	Yes	No	Yes
FOC105	45	100	Yes	No	Yes	Yes
FOC106	30	100	Yes	Yes	No	No
FOC117	45	60	Yes	Yes	No	Yes
FOC133	45	60	Yes	No	Yes	Yes
FOC150	45	70	Yes	Yes	No	Yes
FOC151	45	70	Yes	No	Yes	Yes

Table 4: Injury Summary for Oblique PMHS Occupants

Test ID	Spine	Ribcage	Pelvis	Extremities
FOC104	C5-C6 Transection, L5-S1 Transection	Right rib fractures (7, 10-12), left rib fractures (9-12)	Left pelvic ala linear fracture	Bilateral proximal femoral shaft fractures
FOC105	C6-C7 Transection, T3-T4 Transection	Right rib fractures (3-11) with flail chest, left rib fracture (1)	None	None
FOC106	T5-T6 Fracture Dislocation	Right rib fractures (2-9), left rib fractures (2-7)	Bilateral and transverse sacral fracture, bilateral pubic ramus fracture	Right tibial shaft fracture
FOC117	None	None	None	None
FOC133	None	None	None	None
FOC150	C6-C7 Dislocation (complete)	Right rib 7 fracture	None	Right femur fracture, proximal diaphysis (simple-displaced)
FOC151	C6-C7 Dislocation (complete), T10-T11 osteophyte/anterior column fracture	Right rib fractures (2-12) with flail chest, sternum fracture (rib 3-4 level)	None	None (simple-displaced)

Isolated Sled Tests

The whole-body sled test at the 100% pulse demonstrated transection injury at L5-S1. Due to the limitations and costs of the PMHS experimental model, component tests were subsequently pursued to continue the work of producing a combined tension-moment injury criterion. Component sled tests were run with the objective of producing a PMHS experimental model that could replicate the injuries observed in the whole-body tests and still measure the local lumbar loads. The primary aim was to determine the loads measured at pulse severities from the no-injury whole-body condition (60%) up to failure. These tests incorporated an intact femur–pelvis–lumbar spine to mimic the interaction between the seat cushion and lap belt with the whole-body sled test. The missing thorax mass from the whole body was replaced with a fixed mass attached to the T12 load cell's superior end. The effective torso mass associated with the whole body was unknown; thus, three different mass magnitudes were examined—9, 18, and 22 lb (Table 5). This mass does not represent the mass of the human torso but rather an estimate of the potential effective mass during dynamic loading (i.e., the amount of mass that pulls on the lumbar column).

Table 5: Configurations for Isolated Sled Tests

Test ID	Pulse (%)	Thorax Mass (lb)	Belt Configuration
FILSC102	30	9	Lap belt body-centered belt
FILSC103	30	18	Lap belt and body-centered belt
FILSC104	30	22	Lap belt and body-centered belt
FILSC105	61	9	Lap belt and body-centered belt
FILSC107	61	22	Lap belt and body-centered belt

The estimated injury from these tests did not match the whole-body test in either the location of injury or the pulse severity. The mechanism of injury in the isolated component sled test experimental model did not match the distraction type (e.g., transection) injury at the L5-S1 level in the whole-body test and was attributed to differences in how the fixed mass loaded the spine. As the isolated sled test experimental model did not replicate the failure test's kinematics and injury mechanism, a new design was necessary. This led to the development of an isolated lumbar column piston test wherein the PMHS posture and loading condition could be more controlled.

Isolated Lumbar Column Piston Test

The previous PMHS testing revealed that the primary load component was tension, but the loading environment produced varied, multi-axis bending. Thus, it was decided to tightly control the bending using isolated lumbar columns in a piston test (Figure 14). The focus of the work was to evaluate the impact of posture (represented via a static bending preload) on the load-displacement response for the whole lumbar spine loaded in axial distraction (Avila et al., 2022). Sub-failure testing was first conducted to assess the impact of posture at physiologic levels of loading. The sequence of non-destructive tests was followed by tests to failure with the spine positioned in either a flexed forward or obliquely bent posture. For the failure testing, the data were only included in the analysis if the injury was similar to the previous whole-body testing (Humm et al., 2015a). Figure 15 shows transection injuries in the lower lumbar spine for a piston test (left) and whole-body sled test (red arrow in right CT scan). There were three tests in the flexed position and five in the oblique position. Failure was determined as the point of maximum tensile force.

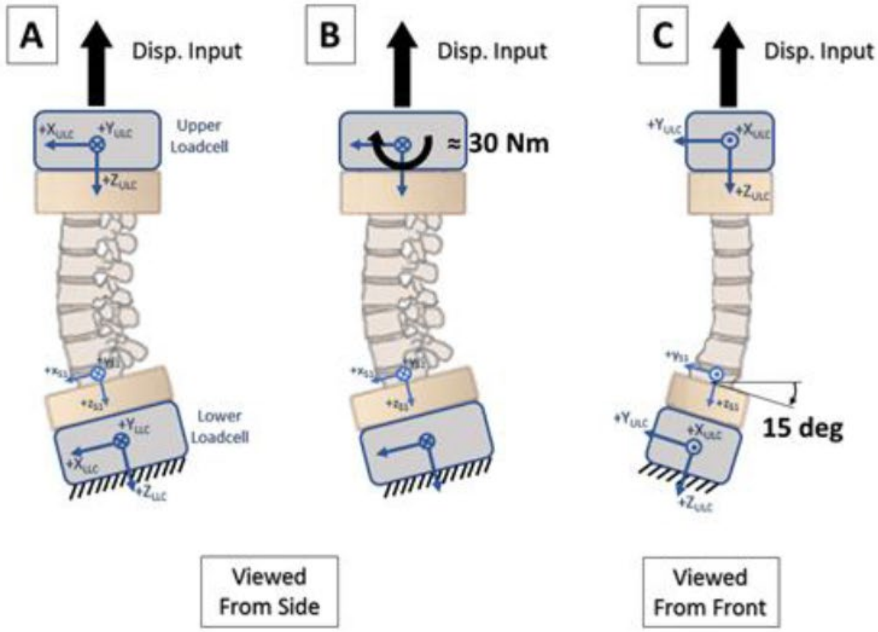

Figure 14: Schematic of Isolated Lumbar Column Piston Test

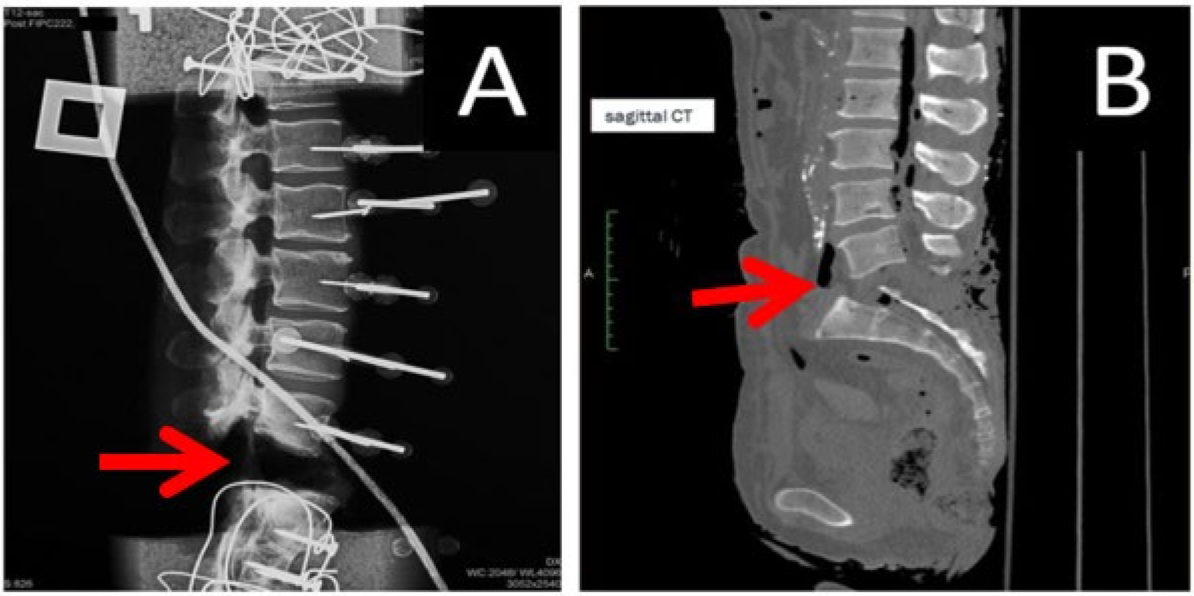

Figure 15: Comparison of Lumbar Transection Injuries from Piston Test (left) and Sled Test (right), Indicated by Red Arrows

Sub-failure testing revealed that deviations from a neutral spine position significantly increased spine loading. The presence of a flexion preload caused statistically significant increases in tensile stiffness, tensile force, and bending moments for similar levels of axial distraction. The addition of a lateral bending preload to an already flexed spine did not significantly alter the tensile response. However, the flexion moment response was significantly affected by the additional postural preload. This work demonstrated that posture has a significant impact on the load response of the lumbar spine and provides an initial understanding of the oblique loading condition. The isolated spine tests recreated the primarily soft-tissue bending-distraction injury produced from whole-body testing (indicated by the red arrows in Figure 15). These failure-inducing tests demonstrated that posture also has a large impact on the failure tolerance of the lumbar spine. Spines exhibited a greatly reduced tolerance for injury in terms of both load and displacement at failure in the oblique position as compared to the flexed position.

Whole-Body Computational Study

To better understand the injury mechanism at a segmental level, computational modeling was performed using a finite element (FE) human body model (HBM) (Somasundaram et al., 2023a). A 75-year-old male Global Human Body Models Consortium (GHBMC) model was selected since the PMHS sled tests used for model validation were completed with male specimens with an average age of 60 years, and the elderly HBM is expected to be more conservative than the standard model, which represents the geometry and material properties of a 26-year-old. The model was evaluated against the longitudinal 16 g PMHS tests previously performed in this project. The HBM was able to replicate the vertebral column transection injuries at the level of the lower lumbar and cervical regions. In addition, the HBM was able to predict the rib fractures observed in the whole-body tests.

The model was used for parametric studies that evaluated pulse severity and seat orientation. The former study demonstrated the significance of the input pulse magnitude on the responses and injury occurrence to the lower lumbar spine. The analysis result also served in defining the pulse severity for the final whole-body PMHS sled tests performed with 70% pulse. The seat orientation parametric study demonstrated that for seat angles ranging from 0° to 90° with respect to the aircraft centerline, the model predicted failure in the lumbosacral region; however, the contribution of the tensile load, lateral bending, and flexion moments varied with respect to change in seat angle.

FAA-Hybrid III Matched Paired Tests

Eighteen FAA-Hybrid III tests were run in an oblique orientation using the same sled as the PMHS tests. The configurations matched the PMHS tests and included sled pulses of 50%, 60%, 75%, and 100% of the peak acceleration regulatory requirement. The loads and moments collected from these tests were used to develop the injury metric.

Eleven FAA-Hybrid III tests were run in a 0° configuration using the 100% pulse. The configurations matched the PMHS tests. For these tests, two different pelvises were used. The standard fixed pelvis and the pedestrian pelvis, which has three separate rubber components (compared to one rubber component for the standard pelvis). The pedestrian pelvis was included because it has more range of motion at the hip joint, which more closely emulates a live person.

FAA-Lower Lumbar Spine Combined Metric

The lumbosacral transection injuries reported in the PMHS tests were observed to be a combined result of multi-axis bending and tension, specifically tension, flexion, and lateral bending moments. Based on this observation, a combined metric similar to N_{ij} was pursued (Somasundaram et al., 2023b). The lower lumbar spine transection injury was estimated to be a result of the occupant's spine flexing at an oblique angle, along with the inertial loading of the upper body creating tension throughout the spine. Hence, critical values for tensile load, flexion, and lateral bending moments were determined based on risk curves (Somasundaram et al., 2023b).

A statistical formulation for parametric survival analysis using Weibull distribution was used for calculating the injury probability curves. The injury status of an ATD test was determined based on injury occurrence in PMHS tests and/or element failure in FE simulations. The data points from the ATD tests with injury status were treated as left censored, and the ATD tests with non-injury status were treated as right censored. The interaction-based force and moment lower lumbar spine injury criterion FAA-LL_{tb} was derived from these data. The formula is given in Equation 1, where FAA-LL represents the lower lumbar load cell of FAA-Hybrid III ATD, and the subscript 'tb' represents tension and bending responses. F_z is the maximum axial tensile load, $F_{z(int)}$ is the critical axial load, M_y is the maximum flexion moment (positive max when following SAE J211/1 sign convention), $M_{y(int)}$ is the critical flexion moment, M_x is the maximum lateral moment (regardless of polarity), and $M_{x(int)}$ is the critical lateral moment. For this calculation, the maximum force and moments are selected irrespective of the time of occurrence.

$$FAA - LL_{tb} = \frac{F_z}{F_{z(int)}} + \frac{M_y}{M_{y(int)}} + \left|\frac{M_x}{M_{x(int)}}\right| \quad (1)$$

To derive the critical intercepts, the force and moments' injury risk curves were evaluated separately. An 85% probability number for each parameter (F_z, M_y, M_x) was used as the critical intercept value for calculation for FAA-LL_{tb} (Table 6). Because the critical intercept value should represent a high probability of injury, the 85% value in the present study was chosen, considering that probability curve values above 85% are highly non-linear. A similar technique has been used in predicting intercept values for the development of lower neck injury criteria (Chirvi. Pintar, & Yoganandan, 2015; Yoganandan, Pintar, & Banerjee, 2017). Two different probability curves were constructed for FAA-LL_{tb}. The first was based on the ATD data with PMHS-defined injury occurrence, and the second was based on the ATD data with FE-predicted failure occurrence. For analysis of peak FAA-LL_{tb}, force time histories, flexion moment time histories, and lateral bending moment time histories were inputted, and peak FAA-LL_{tb} was determined. It is assumed that the presence or absence of a leg restraint boundary condition does not significantly change the intercepts.

Table 6: Critical Intercepts for Combined Metrics

Metric	$F_{z(int)}$ (lb)	$M_{y(int)}$ (in-lb)	$M_{x(int)}$ (in-lb)
FE-based	2832.6	2717.2	3071.2
PMHS-based	3462.1	2531.3	3009.3

The results from survival analysis indicate the FE-based FAA-LL_{tb} risk curve has a tighter corridor, with quality index estimated to be fair at 5% and good at 25% and 50%, respectively (Table 7). In contrast, the PMHS-based ATD metric had unacceptable confidence limits for most of the risk levels. Thus, the FE-based ATD metric was estimated to provide a better risk curve than the PMHS-based ATD metric. A likely reason is the higher sample size used in the development of the FE-based ATD metric.

Table 7: FAA-LLtb Risk and Quality

Risk Level	Mean	95% Confidence Interval (Upper)	95% Confidence Interval (Lower)	Normalized Confidence Interval Size	Quality Index
0.05	1.61	2.30	1.12	0.73	Fair
0.10	1.72	2.27	1.30	0.57	Fair
0.25	1.88	2.27	1.56	0.38	Good
0.50	2.04	2.38	1.75	0.31	Good
0.75	2.17	2.60	1.82	0.36	Good
0.90	2.27	2.83	1.83	0.44	Good
0.95	2.33	2.97	1.83	0.49	Good

FAA-LL$_{tb}$ for CAMI Tests

In the side-facing policy (ANM-25-03-R1), a 25% risk of AIS 3+ injury was selected for the newly developed neck injury criteria. If the same risk level is selected for oblique-facing policy, an FAA-LL$_{tb}$ limit of 1.88 is recommended. Table 8 shows the FAA-LL$_{tb}$ for the 17 tests conducted for this project at CAMI (described above) using the FE-based intercepts listed in Table 6. Values in red italic exceed the proposed limits. The lowest FAA-LL$_{tb}$ in a lap belt-only configuration was 2.51 (FAA-Hybrid III, 45°, no armrest). The lowest lumbar F_z load was 1793 lb in a Hybrid II test of that same configuration. The highest FAA-LL$_{tb}$ in a lap and shoulder belt configuration was 0.8 (FAA-Hybrid III, 30°, with armrest). All the lumbar loads for the lap and shoulder belt tests were in compression, with the highest being -1245 lb. Even factoring in the 95% confidence interval listed in Table 7, all the lap belt-only tests would fail FAA-LL$_{tb}$, while all the lap and shoulder belt tests would pass. For the inflatable lap belt tests, there was a mix of FAA-LL$_{tb}$ values, with two tests producing values above 1.88 and two tests below the limit. The primary difference in the tests was the distance between the ATD and the armrest. All four tests produced lumbar tension above the preliminary limit of 1200 lb.

Table 8: FAA-LL_{tb} for CAMI Tests

Test Number	ATD*	Angle	Belt*	Armrest	Fz (lb)	My (in-lb)	Mx (in-lb)	FAA-LL$_{tb}$
A12021	H2	45°	Lap	Yes	*3078*	2924	3256	*3.22*
A12022	H2	45°	Lap	No	*1793*	3308	2728	*2.74*
A12023	H2	45°	Lap	No	*1969*	3448	4000	*3.27*
A12024	FAA-H3	45°	Lap	No	*2017*	2398	3807	*2.83*
A12025	FAA-H3	45°	Lap	Yes	*2841*	1953	2416	*2.51*
A12026	FAA-H3	30°	Lap	Yes	*2374*	3225	1978	*2.67*
A12027	FAA-H3	30°	Lap	No	*2135*	3323	3107	*2.99*
A16034	FAA-H3	45°	L+S	Yes	173	697	-1245	0.72
A16035	FAA-H3	45°	L+S	Yes	195	665	-946	0.62
A16036	FAA-H3	45°	L+S	No	185	566	-1032	0.61
A16037	FAA-H3	30°	L+S	No	174	806	-1235	0.76
A16038	FAA-H3	30°	L+S	Yes	241	659	-1091	0.68
A16039	FAA-H3	30°	L+S	Yes	220	810	-1292	0.80
A23051	FAA-H3	45°	Inf	Yes	*1533*	1807	2674	*2.08*
A23052	FAA-H3	45°	Inf	Yes	*1738*	1745	2621	*2.11*
A23056	FAA-H3	45°	Inf	Yes	*1434*	1484	-1463	1.53
A23057	FAA-H3	45°	Inf	Yes	*1480*	1656	2094	1.81

* Hybrid II is abbreviated as H2 and FAA-Hybrid III as FAA-H3. Lap and shoulder belt abbreviated as L+S. Inflatable lap belt abbreviated as Inf.

Probabilistic Spine Modeling

In support of the MCW work and to provide the FAA with an example of probabilistic modeling, an Aviation Research Grant was awarded to the Southwest Research Institute (SwRI) titled "Probabilistic Finite Element Modeling of Oblique Impact Loading to the Lumbar Spine" (16-G-001). SwRI developed a probabilistic model of the lumbar spine (T12-Sacrum) to investigate the effect of oblique loading on spinal injury (Figure 16). The model used a range of size, shape, and material properties when generating simulation results (Coogan, 2016). The model was validated based on PMHS experimental testing at different oblique angles. The model showed that increased obliqueness leads to a higher probability of injury, with a 45° oblique angle having the highest system-level injury probability.

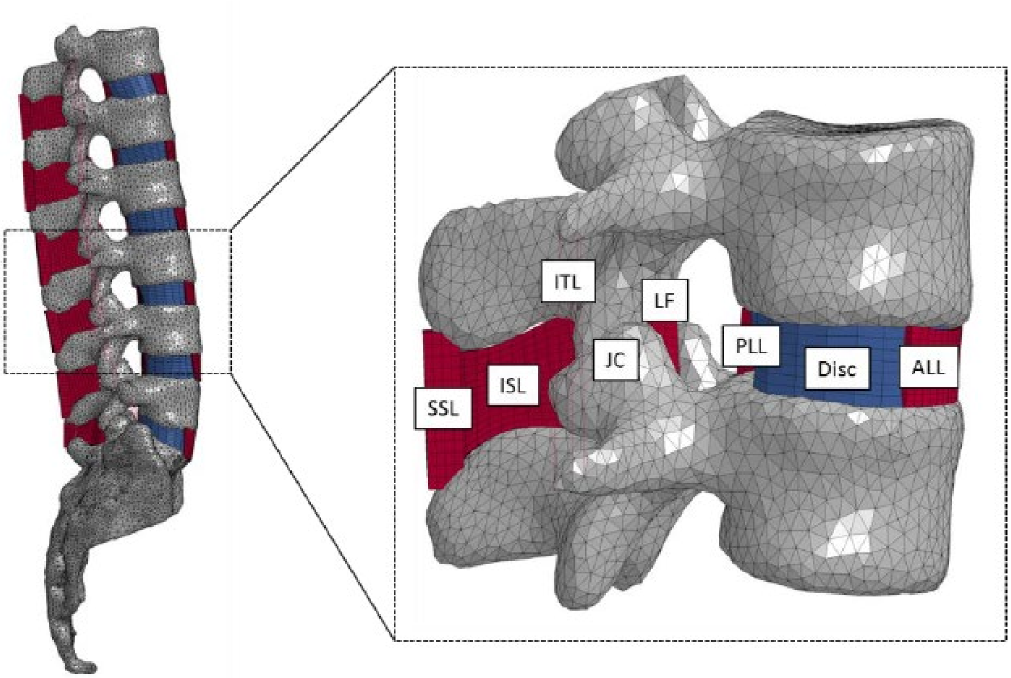

Figure 16: SwRI Probabilistic Lumbar Spine Model

RECOMMENDED INJURY CRITERIA

For seats installed between 18° and 45° with respect to the aircraft centerline, the FAA Hybrid III is recommended for tests used to show compliance with § 25.562. For the combined longitudinal-vertical test, § 25.562 (b)(1), the existing lumbar load compressive limit of 1500 lb is recommended. For the longitudinal test, § 25.562 (b)(2), the following injury criteria are recommended to demonstrate an acceptable level of safety:

- *Head:* Head injury criterion (HIC) less than 1000, as defined in § 25.562(c)(5), unless there is contact with an airbag and no apparent contact with the seat or structure, in which case HIC_{15} (as defined in 49 CFR 571.208) less than 700.
- *Neck:* The neck injury criteria (N_{ij}), as defined in 49 CFR 571.208, below 1.0, peak upper neck force in the z-direction below 937 lb in tension and 899 lb in compression, rotation of the head about the vertical axis limited to 105° relative to the torso, and no concentrated loading on the neck.
- *Torso:* No significant concentrated loading on the occupant's spine in the area between the pelvis and shoulders during ATD impact and rebound. During this type of contact, the interval for any rearward (X direction) acceleration exceeding 20 g must be less than 3 milliseconds, as measured by the thoracic instrumentation specified in 49 CFR part 572, subpart E, and filtered in accordance with SAE Recommended Practice J211/1. No interaction with the armrest or other seat components significantly different than would be expected for a forward-facing seat.
- *Lumbar Spine:* FAA-LLtb below 1.88 using 2832.6 lb, 2717.2 in-lb, and 3071.2 in-lb as the intercepts for F_z, M_y, and M_x, respectively.

- *Pelvis:* No part of the load-bearing portion of the bottom of the ATD pelvis should translate beyond the edges of the bottom seat cushion supporting structure.
- *Femur:* Axial rotation of the upper leg (about the z-axis of the femur per SAE International Recommended Practice J211/1) limited to 35° from the nominal seated position. Evaluation during rebound does not need to be considered, provided that the method of mitigation does not return excessive energy to the legs.

BRACE POSITION

No information was collected that would lead to a recommendation that differs from existing brace position guidelines as documented in Advisory Circular 121-24D (FAA, 2019). If there is a shoulder belt, the occupant should sit upright, place their chins on their sternums, their hands in their laps, and their feet flat on the floor with their knees bent at about 90 degrees. If there is an inflatable lap belt, then the procedures outlined for that restraint should be followed.

CHILD SEAT TESTING

Based on the wording in 14 CFR 25.785, seats installed at greater than 18° typically include a shoulder harness. Accommodating shorter-stature occupants, including children, is difficult in these configurations, particularly because the FAA prohibits booster seats. Because of the challenge, SAE AS 8049/1B limits the occupancy of a side-facing seat to individuals who are 5 ft or taller. In 2015, the FAA issued an Information for Operators (InFO 15011), which prohibited the use of child restraint systems (CRS) in oblique seats (FAA, 2015c). However, airlines have made it known that they wish to allow all passengers to be able to sit in oblique-facing seats.

To evaluate the risk of injury for very young children in oblique-facing seats, the FAA partnered with the Center for Child Injury Prevention Studies (CChIPS). Researchers from The Ohio State University, The Children's Hospital of Philadelphia, the Medical College of Wisconsin, and AmSafe, Inc. worked together with the FAA to conduct the study. They conducted four static inflatable lap belt deployment tests at AmSafe and 42 dynamic tests at CAMI (Mansfield et al., 2024).

Aircraft do not currently have latch or ISOFIX attachment options; therefore, to properly secure a CRS into a seat, the lap belt path must be used. The static tests showed that a CRS can be damaged by the belt inflating within the belt path. The dynamic tests evaluated type of seat belt (conventional lap belt or deactivated inflatable lap belt), impact direction (30° or 45° angle offset from centerline), type of child restraint system (rear-facing (RF), forward-facing (FF), or none), and adjacent structure in the aircraft environment (wall, armrest, or none) (Figure 17). The tests used a 3-year-old Hybrid III ATD (for FF child seats and lap belt only) and a 12-month-old child restraint and airbag interaction (CRABI) ATD (for RF child seats). Tests were run at the 16 g Part 25 longitudinal pulse. The ATDs had instrumentation in their head, neck, and thorax to measure potential injuries to those body regions.

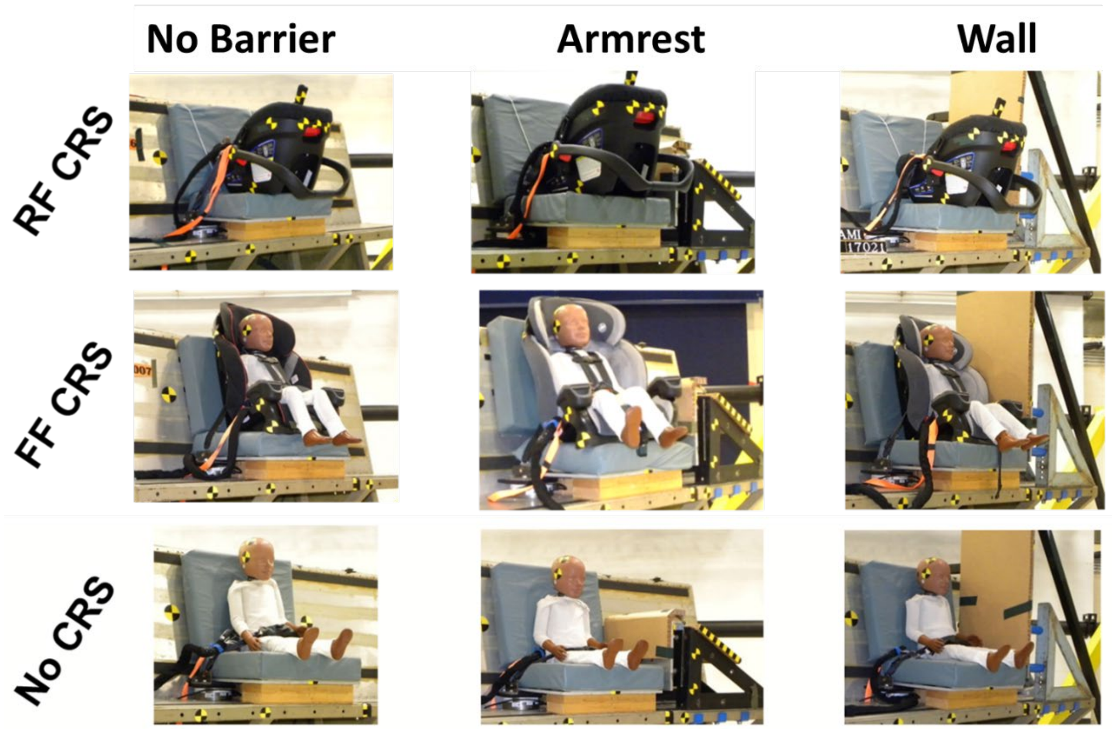

Figure 17: Configurations for Dynamic Tests with Child ATDs

Researchers found that the 3-year-old ATD sitting in a 45° seat without a CRS experienced head strikes that could potentially result in injury, regardless of what structure was adjacent to it (wall, armrest, or nothing). The ATD flexed forward substantially so that in the case with no wall, the ATD's head struck its knee. The tests with the 3-year-old seated in the forward-facing CRS and those with the 12-month-old in the rear-facing CRS were much more likely to result in no injuries to the children. Fewer and less severe interactions occurred with adjacent structures when the 30° offset was used compared to 45°. Deactivated inflatable lap belts did not have detrimental effects on the head, neck, and chest metrics examined. Based on the testing showing that a child in an approved child restraint is safer than in a lap belt alone, the FAA canceled InFO 15011[7], thus allowing for the installation of CRS in oblique seats.

CONCLUSIONS

Beginning in fiscal year 2011, the Biodynamics Research Team at the CAMI was tasked with providing research to support the development of an FAA policy for the approval of oblique-facing seats (e.g., seats installed from 18° to 80° with respect to the aircraft centerline). The project included several iterations of testing and modeling at CAMI, two grants to the MCW to define injury risks and an injury criterion, a grant to SwRI to demonstrate a probabilistic model of the lumbar spine, which, in turn, supported the MCW research, and a partnership with the CChIPS to evaluate the use of child safety seats in oblique-facing aircraft seats. Throughout the length of

[7] The FAA website does not retain full text of canceled Information for Operators messages; however, a copy is available for download at https://img1.wsimg.com/blobby/go/11bcb678-51f9-46c7-8787-345fc1599839/downloads/InFO15011-Child-Restraint-Devices.pdf?ver=1617471492353 as of August 27, 2024.

this research project, the information learned has been used to develop both FAA policy (PS-AIR-25-27) and an industry standard (SAE AS6316).

Early ATD testing conducted by CAMI showed that significant injuries are possible for occupants seated in oblique-facing seats who are involved in a crash. Based on tests using PMHSs conducted by MCW, an initial tension limit of 1200 lb in the lumbar spine was proposed by the FAA. Because this limit was considered conservative and based on a limited number of tests, additional work was funded to generate a more statistically significant injury criterion. This work led to the definition of the FAA-LL_{tb} criterion, which incorporates spinal tension, flexion, and lateral bending. An FAA-LL_{tb} of 1.88 corresponds to a 25% risk of a serious injury (a threshold consistent with other FAA requirements). Because the whole-body development tests were limited to seat installation angles of 30° - 45°, application of this criteria to forward-facing or fully side-facing seats may require additional research. This limit can be met by either the use of an effective shoulder belt or a well-designed inflatable restraint. An occupant restrained solely by a lap belt is at high risk of serious or fatal injury. Testing of child seats shows that a 3-year-old ATD is safer in an approved child restraint than restrained solely by a lap belt. Deactivated inflatable lap belts did not have detrimental effects on the head, neck, and chest metrics examined. However, a deploying inflatable lap belt may cause damage to the CRS by the belt inflating within the CRS belt path; therefore, operators should ensure the inflatable restraint is deactivated upon installation of a child restraint. Future research is recommended to evaluate the risk of injury for occupants who are too tall for child restraints but too short for proper engagement with the fixed shoulder anchorages found on aircraft.

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Appendix A Publications and Presentations

The research work covered by this summary report resulted in multiple publications in peer-reviewed journals and conference proceedings, as well as multiple conference presentations.

Publications

1. Avila, J.G., Humm, J.R., Driesslein, K.G., Moorcroft, D.M., Yoganandan, N., Pintar, F. (2019). *Tensile Injuries of the Isolated Lumbar Spine in Oblique Bending*. Proceedings of 47th NHTSA workshop on Human Subject for Biomechanical Research.
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of Seat Orientation on the Lumbar Spine Response to FAA Longitudinal Crash Pulse: A Parametric Modeling Study. Summer Biomechanics, Bioengineering and Biotranspot Conference (SB3C 2021).
14. Somasundaram, K., Humm, J. R., Yoganandan, N., Moorcroft, D. M., & Pintar, F. A. (2023). *Tension-bending Risk Curves for the ATD Lower Lumbar Spine Subjected to Oblique Impact under FAA Emergency Landing Conditions*. International Journal of Crashworthiness, 28(6), 786–796. https://doi.org/10.1080/13588265.2022.2130611
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Presentations

1. Avila, J., Humm, J., Driesslein, K., Moorcroft, D., Yoganandan, N., Pintar, F. 2019). *Lumbar Spine and Pelvis Injury and Response on Oblique-facing Aircraft Seats – A Preliminary Computational Study*. Proceedings of the Forty-Eighth NHTSA Workshop on Human Subjects for Biomechanical Research. https://www-nrd.nhtsa.dot.gov/database/bio/proceedings/search.asp
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