Kinetics of Small Female PMHS in OEM vs. Spring-Controlled Seats.

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Abstract

This study evaluated whether a mechanically defined spring-controlled (SC) seat could serve as a reasonable proxy for an original equipment manufacturer (OEM) driver-seat environment in small-female post-mortem human surrogate frontal-impact testing. Data from two previously conducted frontal sled-test series were comparatively analysed. Each series included three small-female PMHS tested under the same 50 km/h crash pulse using a three-point restraint with a 2.5 kN load limiter. The first series included tests with an original OEM seat with full compartment interaction, whereas the second series used an isolated SC seat with defined front-seat stiffness and anti-submarining support. Pelvis motion and restraint loading were analysed up to the time of knee-bolster contact in the first series to isolate early seat-governed response. Knee-bolster contact in the first series occurred at approximately 40–50 ms. Before this time, the OEM seat showed greater forward pelvis excursion and a tendency toward earlier pelvis rotation, whereas the SC seat showed higher lap-belt loading. Shoulder-belt loading was similar between series. These findings indicate that seat boundary conditions influence early lower-torso response under frontal loading. The SC seat provided a useful and repeatable boundary condition for mechanistic study, but it did not fully reproduce the broader system-level behavior of the SC-seat and environment.

Keywords Frontal impact, OEM seats, Pelvis kinematics, Spring-controlled seats, mall female.

I. INTRODUCTION

Occupant pelvis interaction with the seat and restraint system plays an important role in frontal-impact kinematics and in the progression of lower torso loading [1–3]. During a crash event, pelvis motion is shaped by the combined influence of seat support, lap-belt engagement, and subsequent interaction with forward interior structures [2–5]. These factors affect occupant excursion, load transfer through the lower torso, and the potential for adverse belt–pelvis interaction, including submarining-related mechanisms [1][4][6]. This is important to understand as vehicle interiors continue to evolve. In seating concepts associated with automated and autonomous vehicles, occupant interaction with the surrounding compartment may differ from that of a conventional driver environment. In such conditions, unlike in a traditional passenger vehicle where the frontal compartment may contribute to constraining the occupant, the seat and restraint system may play an increasingly important role in controlling occupant excursion and lower torso motion before contact with other interior structures. For small-stature occupants in particular, pelvis response is of interest because anthropometry, seated posture, and belt fit may alter how restraint forces are transferred through the pelvis and lower abdomen[6–8].

Because the pelvis is supported directly by the seat, comparison of pelvis kinetics between different seat environments provides a logical starting point for evaluating how boundary conditions influence occupant response. A controlled comparison between an original equipment manufacturer (OEM) driver-seat environment and a simplified environment in which occupant motion is constrained primarily by the seat and belt system, without a frontal compartment, is therefore warranted to better understand pelvis kinetics. Recent studies [3–9] have used spring-controlled (SC) seat or otherwise simplified seat environments with mechanically defined support characteristics to investigate occupant kinematics, belt–pelvis interaction, and submarining-related response under controlled restraint conditions, including test conditions motivated by future or non-traditional vehicle interiors. Under similar crash-pulse and belt-restraint conditions, it was expected that pelvis response in the OEM and SC configurations would be broadly similar, particularly because the SC seat stiffness was intended to approximate a representative front-OEM seat.

The two test series examined in the present study provided such an opportunity. The first series used an U.S. OEM driver-seat environment with the full driver compartment. The second used a SC seat with stiffness representative of a frontal driver-seat support condition from Uriot study. Although the two configurations were

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originally developed for different experimental purposes, both were conducted using small-female surrogates under the same frontal crash pulse and three-point belt restraint conditions, allowing comparative assessment of pelvis motion and lower torso loading. The objective of this retrospective analysis was to investigate whether pelvis response in small-female PMHS was comparable between an OEM driver-seat environment and a spring-controlled seat under the same frontal crash pulse and nominal belt-restraint condition. We hypothesized that the kinetic and kinematic responses would be broadly similar between the two seat configurations, particularly before substantial interaction with other restraint or forward-compartment components.

II. METHODS

Data from two previously conducted frontal sled-test series were comparatively analysed to examine occupant response in OEM and spring-controlled (SC) seating configurations. Each series included three small-female PMHS tested under a 50 km/h crash pulse. The OEM cohort was tested in a Toyota Yaris driver-seat buck with the production seat, three-point restraint, steering wheel, knee bolster, and frontal airbag. The spring-controlled (SC) cohort was tested in a generic automotive buck with a semi-rigid spring-controlled seat, anti-submarining ramp, adjustable B-pillar, open seatback, and knee bolster. In the SC configuration, the seat pan and ramp were independently tuned to simulate front-seat stiffness representative of a European OEM seat, and the restraint system included a production-type retractor with a 2.5 kN load limiter. The OEM configuration represented a contemporary production driver-seat environment, whereas the SC configuration provided a simplified and controlled seat-boundary condition relevant to autonomous-vehicle-type seating studies. The test protocol for both configurations have been described previously [8–9]; a brief summary is provided here for completeness. For the OEM configuration, previous publications described near-side and far-side procedures using the same overall test methodology, whereas the present study considered the frontal-impact dataset. For the SC configuration, previous work reported low-speed (15 km/h) and moderate-speed (32 km/h) responses, whereas the present study examined the 50 km/h dataset. Accordingly, the data analysed here have not been published previously in the comparative context presented in this paper. Although selected injury outcomes and pelvis-response metrics from the spring-controlled cohort have been used previously for development of an iliac-wing fracture risk curve and for evaluation of submarining indicators, the nominal-posture, 50 km/h small-female response corridors from this cohort have not been previously reported. The OEM cohort response and injury outcomes are reported here for the first time. Therefore, the present study provides the first comparative analysis of small-female PMHS pelvis kinetics, restraint loading, and injury outcomes between the OEM driver-seat environment and the spring-controlled seat condition.

PMHS Positioning

In both cohorts, three small-female PMHS were positioned in a nominal (25°deg seat back) driving posture for frontal impact testing (Fig. 1). For the OEM series, the driver seat was first placed fully rearward and fully reclined to permit specimen installation. The PMHS was aligned with the seat centreline and seated by applying forces at the pelvis and knees until the sacrum and torso contacted the seatback. The specimen was then settled into the seat cushion with symmetric shoulder loading. Knee spacing was set to 165 mm between patellar centres, and the right foot was aligned with the centre of the accelerator pedal. The seat was subsequently advanced to the first detent, or as far forward as possible while maintaining at least 5 mm clearance from the knee bolster. The steering wheel was positioned at mid-adjustment, and the seatback was set to the nominal NCAP position, defined as the sixth detent rearward of full upright. Head posture was adjusted to place the Frankfurt plane parallel to the laboratory x-y plane using a digital inclinometer, and masking tape routed to the head-restraint posts was used to maintain pre-test posture. The seatbelt was routed over the thorax and pelvis, and padding was placed between the arm and seatback to position the arms at approximately 45° relative to vertical.

For the spring-controlled series, positioning was based on target sagittal-plane head, thoracic, and pelvic segment angles calculated from seatback angle and specimen BMI (10). The PMHS was lifted into the buck and aligned such that the midsagittal plane matched the seat centreline, and the greater trochanter approximated the designed H-point. Positioning proceeded sequentially from pelvis to thorax to head using coordinate measuring machine measurements and CT-based anatomic landmarks. The femurs were placed symmetrically on the seat pan and anti-submarining ramp, and knee spacing was set according to the target anthropometric condition. The feet were placed flat on the floor, the tibiae were adjusted to approximately 135° knee flexion, and the arms were positioned at about 45° from vertical. The dual seatback straps served as the primary means of setting torso posture, while an adjustable lower-spine sling controlled lumbar and lower-thoracic spine curvature to achieve the target thoracic angle. A head-positioning fixture and arm support wires were used during setup and inertially released at sled onset. The D-ring was adjusted to the target shoulder-belt geometry, the three-point belt was routed across the ASIS region and thorax. The knee bolster was positioned one femur length anterior to the knee.

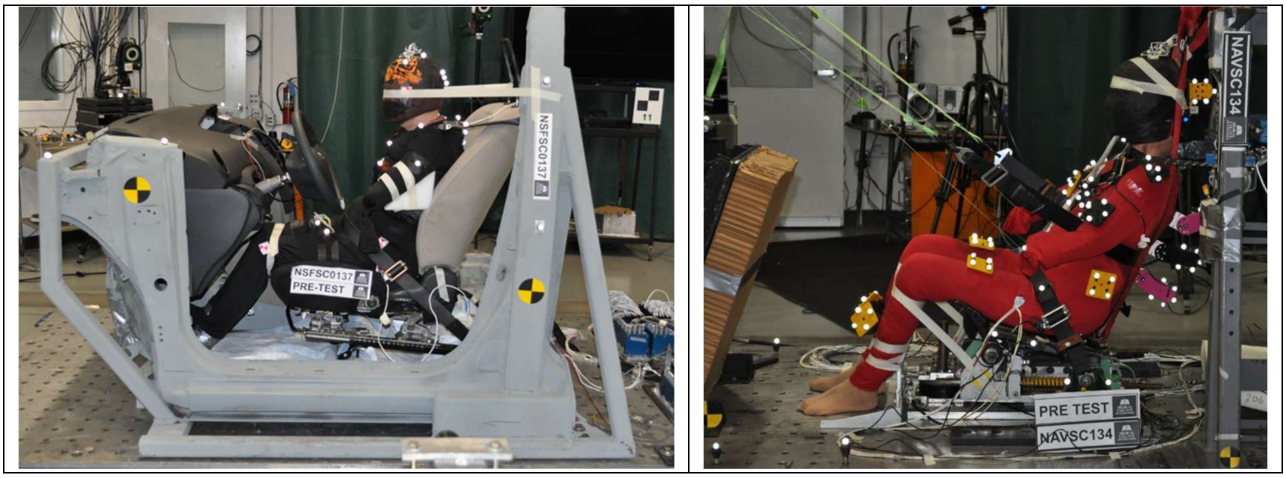

Fig. 1. Test setup for test with OEM seat (left) and SC seat (right).

Instrumentation

PMHS instrumentation differed somewhat between the two series, but both included inertial sensing at the head, upper torso, and pelvis together with optical tracking. In the OEM series, the PMHS were instrumented with a head-mounted nine-accelerometer package (NAP) and three angular rate sensors, 6DOF sensors at T1, T8, L2, and sacrum, and two 59-gauge chestbands placed circumferentially at the rib-4 and rib-7 levels. Optical marker clusters were placed on the dorsal head and spine. Pre-test CT scans were used to identify anatomic fiducials and instrumentation locations, including head, spinal, and pelvic landmarks.

In the SC seat series, more extensive instrumentation was used. A posterior surgical approach allowed placement of specimen-specific PLA-mounted 6DOF sensors at T1, T8, T12, and L4, with anchor-point interfaces for motion-capture registration. A tetrahedral NAP with three angular rate sensors was mounted to the posterior cranium. Bilateral 6DOF sensors were mounted at the iliac wings, femurs, tibiae, and arms. Strain gauges were applied bilaterally to the clavicles and ribs 4–7. Optical marker clusters were placed at the posterior head, dorsal spine, shoulders, sternum, arms, femurs, tibiae, and selected rib locations. Pre-test instrumented CT scans were then used to determine three-dimensional locations of anatomic fiducials, sensor blocks, retroreflective targets, and anchor points for subsequent kinematic reconstruction.

Buck instrumentation differed because of the setup. In the OEM series, the restraint system included two uniaxial load cells in the lap belt and two uniaxial load cells in the shoulder belt, together with a timed output module for pretensioner and airbag firing; sled acceleration was measured with a uniaxial accelerometer. In the spring-controlled series, the buck was instrumented more comprehensively, including six-axis load cells, triaxial accelerometers, and angular potentiometers at the seat pan and anti-submarining plate; triaxial load cells and accelerometers at the belt anchors and D-ring/retractor; and uniaxial load cells in the lap and shoulder belt segments.

Data processing

Analogue sensor signals were collected at 20 kHz and sign conventions were in accordance with SAE J211, and three-dimensional occupant motion was recorded at 1000 Hz using a 27-camera motion-capture system. High-speed video was acquired at 1000 fps. For both test series, sensor and kinematic data were processed relative to specimen-specific anatomic reference frames derived from pre-test instrumented CT scans, digitised fiducials, and motion-capture target locations.

For the head, spine, and pelvis, a 4 × 4 orientation matrix was calculated to relate each sensor block to the local anatomy. Angular accelerations were obtained from the measured angular velocity signals, and sensor time-history data were transformed into the local anatomic coordinate system. Linear accelerations at the anatomic origin were then computed from the measured sensor accelerations and rotational terms. In the OEM series, this processing was applied to the head, T1, T8, L2, and sacral instrumentation; in the spring-controlled series, the same approach was applied to the head, T1, T8, T12, L4, pelvis, and extremity instrumentation.

Injury Assessment

Post-test injury assessment was performed using gross examination, radiographic imaging, computed tomography, and autopsy. Imaging findings were reviewed together with dissection results to identify and localise skeletal and soft-tissue injuries, with additional focused dissections conducted in regions of interest, including the spine and pelvis.

Data Analysis

Given the limited sample size and the exploratory nature of the proposed analysis, comparisons between the OEM and SC cohorts were primarily descriptive rather than inferential. Analysis considered specimen-level and cohort-level response time histories, including occupant kinematics, belt loading, and seat response where available. Comparisons focused on the timing, magnitude, and directional trends of the measured responses, with selected discrete metrics used to support interpretation of the time-history data. For the present analysis, only response measures from the head, T1, pelvis, and restraint system were considered, as these were most relevant to the study objectives; the full set of recorded test data is available through the NHTSA biomechanical database.

III. RESULTS

Specimen Demographics and Test Overview

Three small-female PMHS were tested in each cohort under the same 50 km/h frontal sled pulse using either an OEM driver-seat buck or a spring-controlled (SC). Specimen demographic information is summarized in Table 1.

Table 1. Specimen demographic.

Test ID	Age (years)	Height (mm)	Weight (kg)
SC-134	76	1595	50.9
SC-135	82	1582	42.7
SC-136	86	1590	42.3
Avg	81.3 ± 5.0	1589 ± 6.6	45.3 ± 4.9
OEM-137	85	1549	57.2
OEM-139	86	1621	49.4
OEM-141	58	1605	37.7
Avg	76.0 ± 16.0	1592 ± 3.8	48.1 ± 9.8

Global Excursion Response

Full time-history responses for the OEM and SC cohorts are presented in Appendix Fig. A1 and A2, respectively. Comparison of the complete traces showed that the two cohorts diverged over the full event. Relative to the SC cohort, the OEM cohort showed lower lap-belt loads, more modest upper-body rotational response, and a more attenuated pelvis response during the later phase of the event. The SC cohort, by comparison, exhibited larger and more sustained head and T1 acceleration and angular rate excursions, and substantially higher lap-belt loading.

Belt-loading response provided one of the clearest distinctions between seat conditions. Lap-belt loads were substantially lower in the OEM cohort (approximately 1.55 kN) than in the SC cohort (approximately 5 kN), while shoulder-belt loads were of similar overall magnitude between cohorts (Fig. A1 and A2). This difference is consistent with the absence of a knee bolster in the SC setup to provide an alternative load path or unloading mechanism.

The divergence between cohorts became more apparent after knee-bolster engagement in the OEM configuration. After this point, occupant motion in the OEM cohort was influenced not only by seat and belt interaction but also by the broader driver-compartment restraint environment. In the full-event traces, pelvis angular-rate response in the OEM cohort showed limited continued progression after knee-bolster contact, whereas the SC cohort did not exhibit a comparable attenuation, consistent with continued forward loading in the absence of production-type forward-compartment interaction. Differences in head and upper-torso response were likewise more evident in the complete traces than in the early-time responses, consistent with the later timing of airbag interaction relative to initial belt engagement in the OEM setup.

Taken together, these findings indicated that the OEM and SC seating conditions were not comparable over the complete event. Accordingly, the kinematic and belt-loading data were subsequently re-examined within a time interval limited to knee-bolster contact to enable a more direct comparison of seat-dominated response.

Early-phase Pelvis Sensor Response and Kinematics

To better isolate the interval before substantial forward-compartment interaction, the pre-knee-bolster-contact plots were limited to the first 60 ms. In the OEM cohort, knee-bolster engagement occurred at approximately 40–50 ms. This early interval provided the clearest basis for examining pelvis motion while it was governed primarily by seat boundary conditions, before the loading path was substantially altered by knee-bolster and later airbag interaction.

This time-limited focus does not reduce the relevance of the later response; instead, it helps distinguish seat-driven pelvis behavior from the subsequent system-level effects of the full driver-compartment environment. Knee-bolster engagement is therefore central to interpretation of the overall findings and helps explain why occupant response diverged later in the event, even when early pelvis motion appeared more comparable.

Pelvis response differed qualitatively between cohorts during the early loading phase (Fig. 2). In the OEM cohort, there appeared to be a tendency for earlier pelvis rotation prior to knee-bolster interaction, whereas the SC cohort did not show the same degree of early rotational response. This pattern may indicate that pelvis motion in the OEM seat was influenced by local seat interaction before substantial forward compartment engagement. However, this trend should be interpreted cautiously because of the limited sample size and the inconsistency observed across some pelvis-side measurements.

Pelvis displacement traces further illustrated cohort-level differences in early pelvis motion (Fig. 3). In the x-direction, the OEM cohort showed greater forward pelvis excursion by the end of the 0–60 ms interval, with all three tests exhibiting a similar monotonic increase in displacement after approximately 30–35 ms. In contrast, the SC cohort showed a smaller x-displacement magnitude over the same interval, although the overall time course remained similar. In the z-direction, pelvis displacement responses were more comparable in overall magnitude, but the OEM cohort exhibited greater inter-test variability, whereas the SC cohort showed a more uniform progression across the available traces. Together, the acceleration, angular-rate, and displacement responses indicate that the two seat environments produced different early pelvis motion patterns, with the OEM cohort tending toward greater forward excursion and the SC cohort showing a more constrained translational response during the first 60 ms.

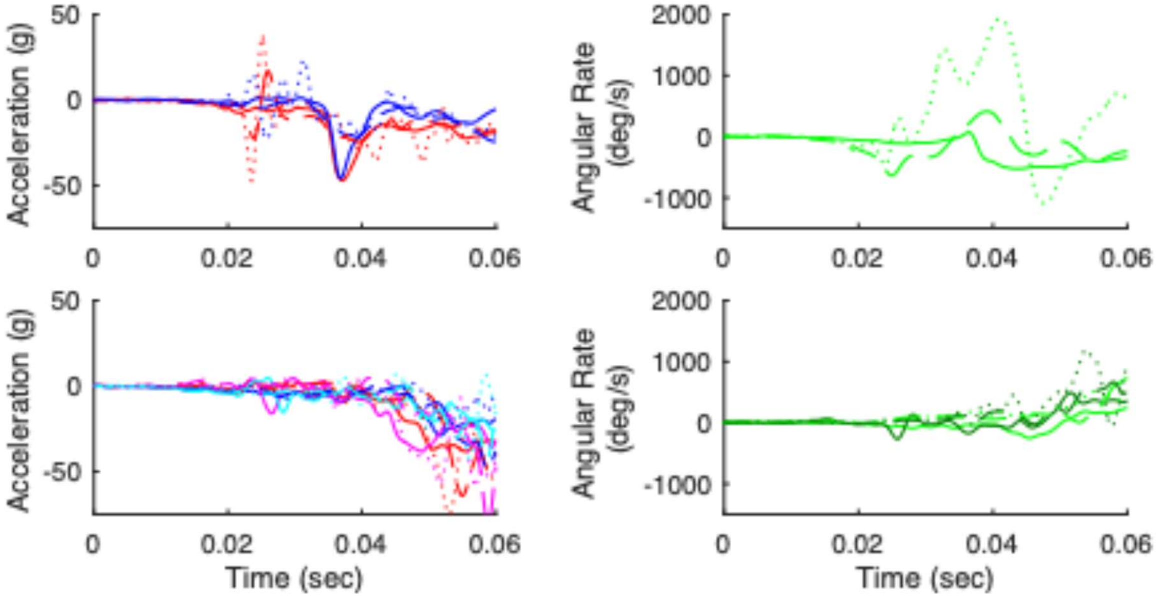

Fig. 2: Pelvis Accelerations (left) and angular rates (right) for the OEM-cohort (upper) and SC cohort (bottom). Blue and red lines are x- and z-accelerations. For the SC cohort the left and right ASIS-mounted instrumentation are shown in different hue (seen in color, x acceleration is cyan and z acceleration is magenta). Solid, dashed, and dotted lines are the three OEM tests.

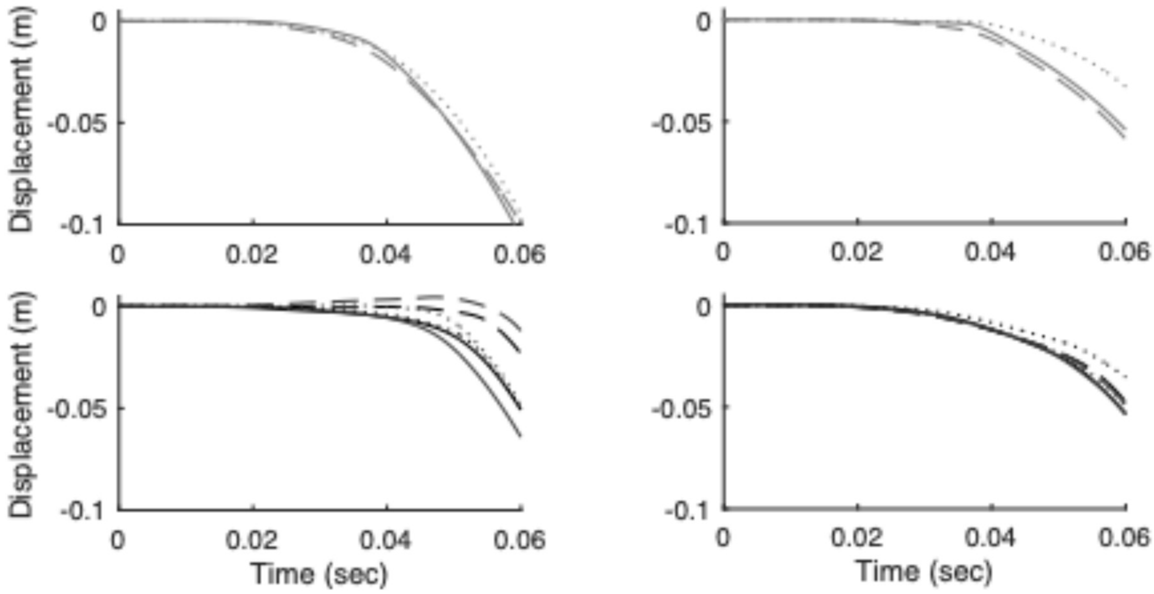

Fig. 3: Pelvis x-displacement (left) and z-displacement (right) for the OEM-cohort (upper) and SC cohort (bottom).

Early-phase Lap- and Shoulder-belt Loading

In the pre-knee bolster contact comparison, lap-belt loading between cohorts was comparable initially but diverged around the timing of knee-bolster engagement in the OEM cohort (Fig. 4).

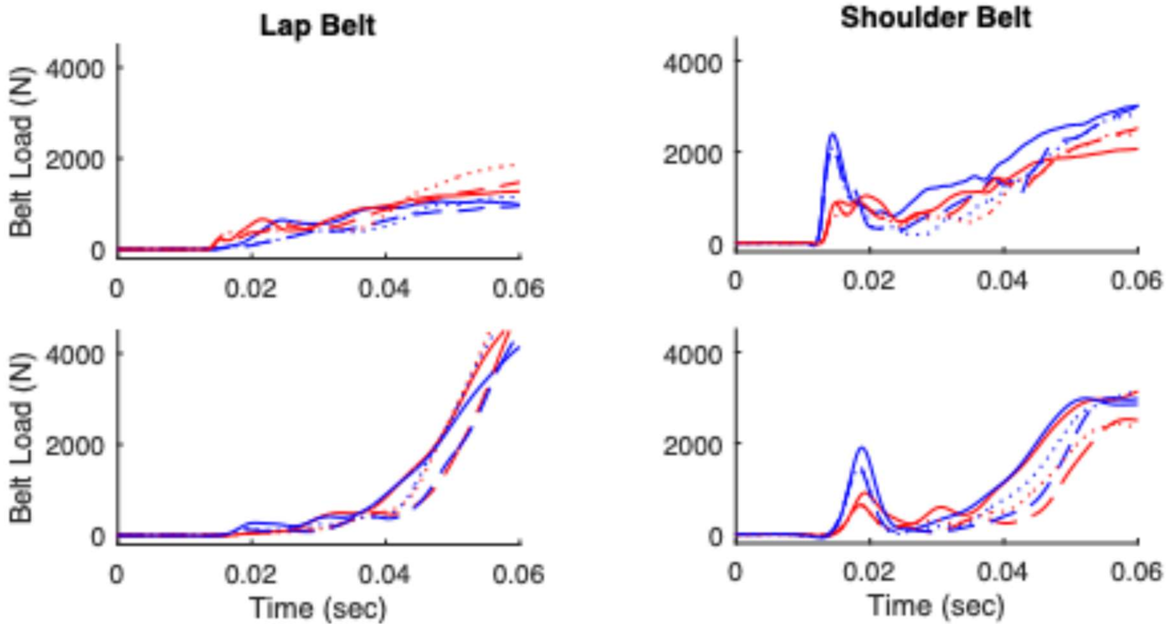

Fig. 4: Lap (left) and shoulder (Right) belt loads for the OEM-cohort (upper) and SC cohort (bottom). Blue and red lines are x- and z-accelerations. Solid, dashed, and dotted lines are the three OEM tests.

In contrast, shoulder-belt loads were nearly the same (~3.2 kN) between cohorts despite the presence of the frontal airbag in the OEM tests. While an unloading effect of the airbag on shoulder-belt loading was expected, that pattern was not evident in the present data. Both cohorts also showed evidence of shoulder-belt pretensioner firing at approximately 20 ms in the early-time plots, and this event corresponded temporally with the onset of early T1 acceleration response.

Together, these observations indicate that the principal lower-torso loading difference between the two seat environments was reflected in lap-belt loading rather than shoulder-belt loading, with divergence emerging at approximately the same time that the OEM occupant began interacting with the forward compartment.

Early-phase Head and Upper Torso Response

Upper torso and head response were strongly affected by the additional frontal restraint environment in the OEM cohort (Fig. 5 and Fig. 6). The airbag reduced T1 and head rotations and head z-acceleration, whereas its effect on x-acceleration appeared smaller. The pre-knee bolster traces also showed T1 accelerations shortly after the apparent pretensioner firing event at approximately 20 ms (Figure 5), suggesting that the initial upper-torso response was influenced by shoulder-belt loading in both cohorts before the OEM airbag produced a clearer separation in head and thoracic response.

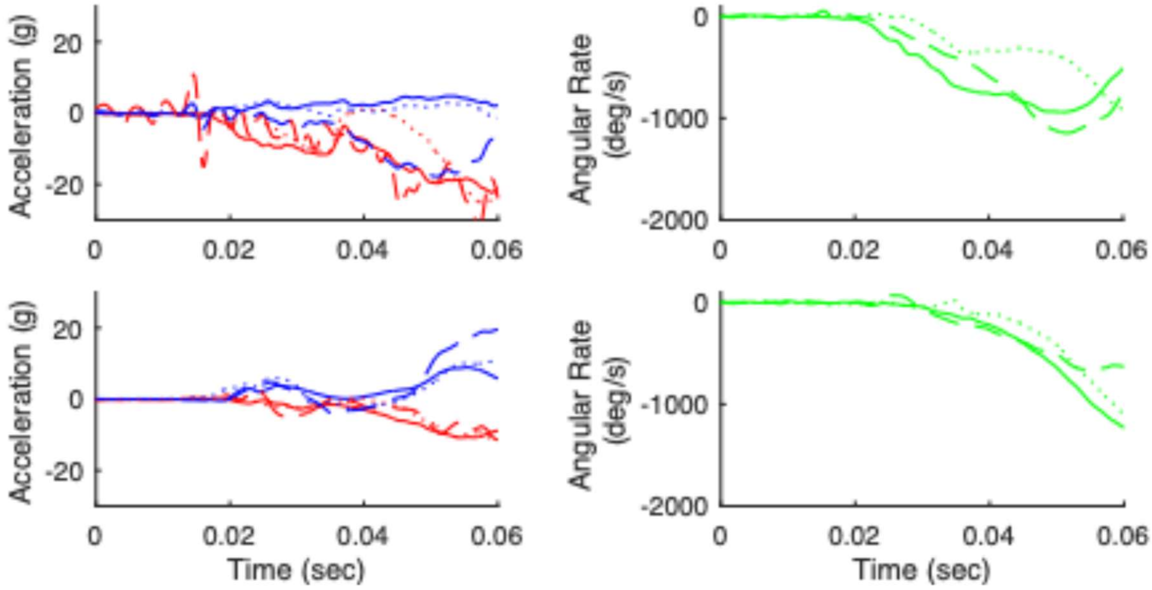

Fig. 5: Head Accelerations (left) and angular rates (right) for the OEM-cohort (upper) and SC cohort (bottom). Blue and red lines are x- and z-accelerations. Solid, dashed, and dotted lines are the three OEM tests.

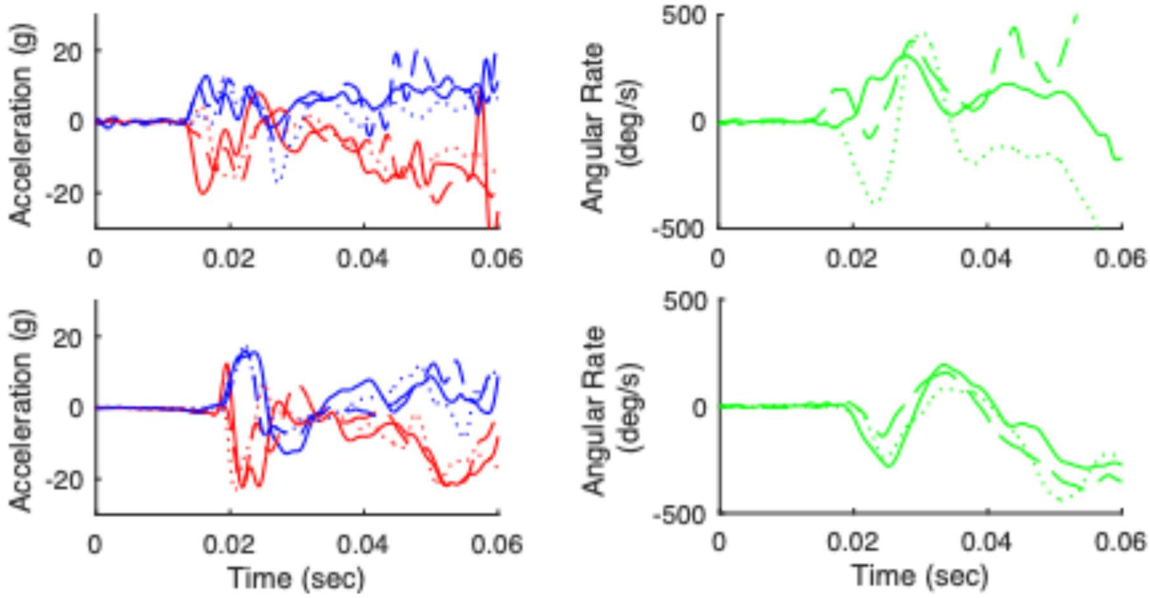

Fig. 6: T1 Accelerations (left) and angular rates (right) for the OEM-cohort (upper) and Spring control cohort (bottom). Blue and red lines are x- and z-accelerations. Solid, dashed, and dotted lines are the three OEM tests.

Post-test Injury Observations

In the OEM cohort, injuries in the OEM cohort were limited to thoracic fractures; no pelvic or lumbar spine fractures were identified. In contrast, in the SC cohort all three tests resulted in bilateral rib fractures and sternal body fracture. Two SC specimens sustained pelvic fractures, and one sustained a lumbar spine injury.

The pelvic and lumbar spine fractures observed in the SC cohort were associated with the later phase of the event, occurring closer to the time of peak lap-belt load at approximately 80 ms, rather than during the early pelvis-response interval considered in the present analysis. This timing suggests that these injuries were more closely related to belt loading than to the initial seat-dominated pelvis motion. The injury implications of this later loading phase have been discussed in detail in prior publications [11].

IV. DISCUSSION

The principal finding of this study was that the spring-controlled seat reproduced the general sequence of early frontal occupant loading but did not fully reproduce the pelvis kinematics and lower-torso loading observed in the OEM driver-seat environment. Thus, the spring-controlled seat should be interpreted as a controlled, mechanically defined boundary condition for mechanistic investigation rather than as a direct proxy for the tested OEM seat. This distinction is important because the value of a simplified seat depends on the intended use: repeatable isolation of seat–occupant interaction versus representation of a specific production seating environment.

The present study began as a comparison of occupant response between an OEM driver-seat environment and a spring-controlled seat configuration under similar frontal loading conditions. In this limited sample, shoulder-belt loading was similar in magnitude between the OEM and spring-controlled cohorts, and bilateral rib fractures were observed in all specimens from both cohorts. However, the full time-history response was not the same across configurations. In particular, pelvis and lower-torso response diverged over the course of the event, and it became evident that the OEM setup introduced knee-bolster interaction that altered the subsequent response, whereas the spring-controlled configuration progressed without the same compartment influence. This difference in the later event shifted the interpretation of the data: rather than attributing the full response primarily to seat characteristics, the more meaningful comparison was the early loading phase, before knee-bolster engagement became a dominant factor.

Effect of Seat Characteristics on Early Pelvis Kinematics

Within that early interval, the results suggest that seat boundary conditions influenced pelvis kinematics. The OEM configuration showed greater pelvic rotation, whereas the spring-controlled seat produced greater forward pelvis excursion and higher lap-belt loads. Because this phase preceded substantial knee-bolster involvement, pelvis motion was governed more directly by interaction with the seat pan, ramp/support surfaces, and the developing lap-belt engagement. This early phase is biomechanically important because it establishes the initial trajectory of the pelvis and lower torso, which can in turn influence subsequent excursion and belt-to-pelvis interaction[12–14].

Differences between seat conditions likely reflected the combined influence of seat pan compliance, local support conditions, ramp characteristics, and overall boundary geometry. In the spring-controlled configuration, the seat pan and anti-submarining ramp were designed to provide a prescribed mechanical response, making the pelvis-seat interaction more directly interpretable in terms of stiffness and support. In contrast, the OEM seat likely distributed loading through a more complex structural and cushion system, with local deformation patterns influenced by production seat construction and surrounding compartment geometry. These distinctions are consistent with prior reclined-seat work showing that pelvis excursion and submarining sensitivity depend not only on belt restraint but also on seat support, recline geometry, and local boundary conditions [15–17]. In addition, our prior work by Somasundaram *et al.* (18) demonstrated clear differences in pelvis kinematics as a function of seat stiffness, further supporting the interpretation that the differences observed in the present study were influenced, at least in part, by the mechanical response of the seat system.

The present findings therefore reinforce the concept that early pelvis kinematics should not be interpreted solely as a function of belt restraint. Rather, the seat itself is an active component of the restraint environment, particularly for small occupants whose pelvis position, seated posture, and lower torso geometry may increase sensitivity to local seat-pan support and ramp interaction. From a design perspective, this suggests that seat characteristics may influence how effectively the restraint system establishes favorable pelvis engagement before forward excursion progresses [7][14][19].

The comparison between the OEM and spring-controlled seats should therefore be viewed with appropriate caution. The spring-controlled seat reproduced a general frontal loading sequence and remains useful as a simplified and mechanically characterized test condition. However, the present results also indicate that equivalence between a simplified seat and a production seat cannot be assumed on the basis of global loading alone. Production-seat behavior reflects a broader system-level response that includes cushion and structure deformation as well as interaction with surrounding vehicle components. For this reason, the spring-controlled seat is best interpreted as a controlled approximation of seat support rather than a full surrogate for the OEM occupant environment [3][6].

The spring-controlled seat represented the occupant support condition using two prescribed linear stiffness components, corresponding to the initial seat-pan response and the anti-submarining-bar response. These stiffness characteristics were derived from a European OEM seat condition reported by Uriot *et al.*[5]. The present results therefore should not be interpreted as evidence that production-seat behavior cannot be represented using such a simplified model, but rather that the specific tuning used here did not fully reproduce the response of the U.S. OEM condition tested in this study. A simplified seat remains valuable for controlled mechanistic investigation, but correspondence to a production seat depends on how well the prescribed stiffness characteristics and resulting local and global response are matched. Future work should examine whether tuning the SC springs to a specific U.S. OEM seat can improve this correspondence.

Implications for Experimental Modeling and Occupant Protection

The findings have implications both for experimental biomechanics and for restraint-system development. From an experimental perspective, the spring-controlled seat offers a well-defined boundary condition that can improve repeatability and facilitate interpretation of seat-occupant interaction. This is particularly valuable for model development and validation, where simplified but mechanically characterised test conditions are often preferable to complex production environments that are more difficult to parameterise. A tunable seat system also provides the opportunity to vary stiffness systematically and examine how seat response affects pelvis trajectory, belt interaction, and potential submarining-related mechanisms [3][7][18].

At the same time, the study highlights the need to consider seat design as an integral component of occupant protection. Frontal restraint performance is often discussed in terms of belt geometry, load limiter behavior, airbags, and hard-contact structures, but prior work has shown that downward pelvis excursion and lap-belt engagement are central to submarining risk, particularly in non-traditional or reclined seating environments. The Newman *et al.* [20]further suggests that vehicle seats may allow larger pelvis downward excursions than simplified semi-rigid seats, reinforcing the importance of representative seat boundary conditions when interpreting restraint performance.

These considerations are especially relevant for populations that may not be well represented by conventional evaluation approaches. Small-female occupants may be more sensitive to differences in seat-pan contour, support stiffness, and local geometry because of their body size, pelvis shape, and seated posture. Accordingly, improved characterisation of seat effects may help support more inclusive restraint assessment and guide development of seat and restraint systems that better manage early pelvis motion across a wider range of occupant sizes [7][8][19].

The present test environment used a simplified laboratory seat boundary condition and should not be interpreted as representing the full protective capability of an optimized production/OEM seat system. Prior computational and experimental studies have shown that seat pan geometry, seat stiffness, seat-track/load-limiting characteristics, and belt-system design substantially affect pelvis restraint, submarining tendency, and lumbar spine loading in reclined frontal impacts. For example, Mroz et al. [21] demonstrated that combined seat-track, seat, and lap-belt load-limiting reduced lumbar spine compression force and flexion moment, as well as pelvis ASIS loading, in reclined occupant simulations. Similarly, Grébonval et al. [22] showed that seat pan, and pelvis angles influenced occupant kinematics and predicted injury risk. Therefore, while the present findings identify potential pelvis and lumbar injury mechanisms in small female occupants under the tested conditions, production/OEM seats with optimized energy management and occupant support may reduce these injury risks. Future studies should directly compare simplified laboratory seats with representative OEM seat systems to quantify their relative influence on pelvis and lumbar injury outcomes.

The spring-controlled seat evaluated here may also be useful as a modeling-oriented surrogate condition. Its defined stiffness can represent a specific class of seat behavior and, because it is adjustable, it offers a practical framework for future efforts aimed at matching different production-seat responses. In that sense, the system is not only an experimental simplification but also a potentially useful bridge between PMHS testing and computational parametric studies.

Limitations and Future Work

Several limitations should be considered when interpreting the present findings. First, the sample size was limited, as is typical in PMHS studies, and inter-specimen variability could not be separated fully from test-condition effects. Second, although the study compared two seat environments under the same frontal pulse, the OEM and spring-controlled configurations differed in more than seat stiffness alone. The configurations also differed in local support conditions, geometry, and surrounding compartment features, particularly the presence of the knee bolster in the OEM setup. Accordingly, the observed differences should not be attributed to stiffness alone.

An additional limitation is that the later occupant response reflected interaction among multiple restraint and compartment components. For this reason, the present study is most appropriately interpreted as a comparison of early pelvis response under differing seat boundary conditions, rather than as a complete evaluation of full-body restraint-system performance. The findings are also specific to the tested crash pulse, restraint configuration, and small-female PMHS condition, and therefore may not generalise directly to other impact severities, seating postures, or occupant sizes.

The spring-controlled seat should also be interpreted within the context of its simplifying assumptions. In this study, it represented occupant support using two prescribed linear stiffness components corresponding to the initial seat-pan response and the anti-submarining-bar response. Those characteristics were derived from a European OEM seat condition reported by Uriot *et al.*[4]. The present results therefore do not demonstrate that production-seat behavior cannot be represented using such a simplified approach; rather, they indicate that the specific tuning used here did not fully reproduce the response of the U.S. OEM condition tested in this study. Future work should examine whether tuning spring-controlled seat characteristics to a specific OEM seat can improve correspondence in both global and local response.

Further work is also needed to evaluate how sensitive early pelvis kinematics are to seat stiffness, support geometry, and local deformation patterns. Systematic testing across tunable seat conditions, combined with ATD and human body model simulations, would help identify which aspects of the early restraint environment are most important for reproducing production-seat behavior and for improving prediction of small-female occupant response. Such studies would help clarify which aspects of pelvis motion are most sensitive to stiffness, geometry, and local support conditions[3][6][20].

Extension of this work to ATD and finite-element human body model evaluation would also be valuable, particularly for assessing whether current surrogates can reproduce the seat-sensitive pelvis kinematics observed in small-female PMHS. More broadly, combining controlled seat experiments with computational modeling could provide a useful framework for identifying design parameters that improve early pelvis control and reduce the likelihood of unfavorable restraint interaction.

V. CONCLUSION

This study showed that seat boundary conditions influenced small-female PMHS pelvis response during the early phase of frontal impact, before substantial forward compartment interaction in the OEM configuration. Relative to the spring-controlled seat, the OEM seat exhibited greater early pelvis rotation and greater forward pelvis excursion, whereas the spring-controlled seat produced higher lap-belt loading. In contrast, shoulder-belt loading was similar between cohorts, indicating that the principal lower-torso difference between seat conditions was expressed through pelvis motion and lap-belt response rather than through the shoulder belt.

These findings support the interpretation that the seat is an active component of the early restraint environment and can influence the initial trajectory of the pelvis before knee-bolster engagement. The spring-controlled seat reproduced the general sequence of early frontal motion and provided a useful, well-defined boundary condition for mechanistic study, but it did not fully replicate the broader system-level behavior of the OEM seat. The new finding from this comparison is that a spring-controlled seat tuned to represent a general front-seat stiffness condition may still produce different pelvis motion and lap-belt loading than a specific OEM driver-seat environment. Therefore, simplified tunable seats are valuable for controlled mechanistic studies, but their use as OEM proxies requires condition-specific validation against the production seat and restraint environment of interest. Overall, the results highlight the importance of representative seat boundary conditions when evaluating early pelvis kinematics in small-female occupants and support the use of tunable simplified seats for future PMHS, ATD, and computational modeling studies.

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VIII. APPENDIX

Table A1. Injury summary

	Test ID	MAIS	AIS	Injury description
Tests with spring controlled seat	SC-134	4	450203.3 450213.4 450804.2 650632.2	**Ribs:** 2-9 (L) **Ribs:** 2-10 (R) **Sternum: B**ody fx at level of ribs 2-3 **Spine:** L4 compression fx (minor)
	SC-135	3	450203.3 450804.2 856151.2	**Ribs:** 2-9 (L) **Ribs:** 2-9 (R) **Sternum: B**ody fx at level of ribs 2-3 **Pelvis:** Lt Ilium fx: ALA, comminuted. Sacral ALA f, bilateral non-displaced
	SC-136	4	450203.3 450804.2 856162.4	**Ribs:** 1-5 (L) **Ribs:** 2-8 (R) **Sternum: B**ody fx at level of rib3 **Pelvis:** Lt Ilium fx: ALA, comminuted/open fx Sacral Body + ALA fx, bilateral, displaced. Superior pubic ramus fx, bilateral
Tests with OEM seat	OEM-137	3	450203.3 450804.2 650234.3 650224.2 650432.2	**Ribs:** 1-3,5-7,11 (L) **Ribs:** 2,3,5-8,11 (R) Sternum: Body fx C7: Vertebral body fx C7 Vertebral lamina fx rt. T6: Vertebral body fx
	OEM-139	3	450203.3 450213.3 650432.2	**Ribs:** 1-5 (L) **Ribs:** 2-7 (R) T1 Vertebral Body fx.
	OEM-141	3	450203.3	**Ribs:** 3, 5, 7-9 (L) **Ribs:** 3-7 (R)

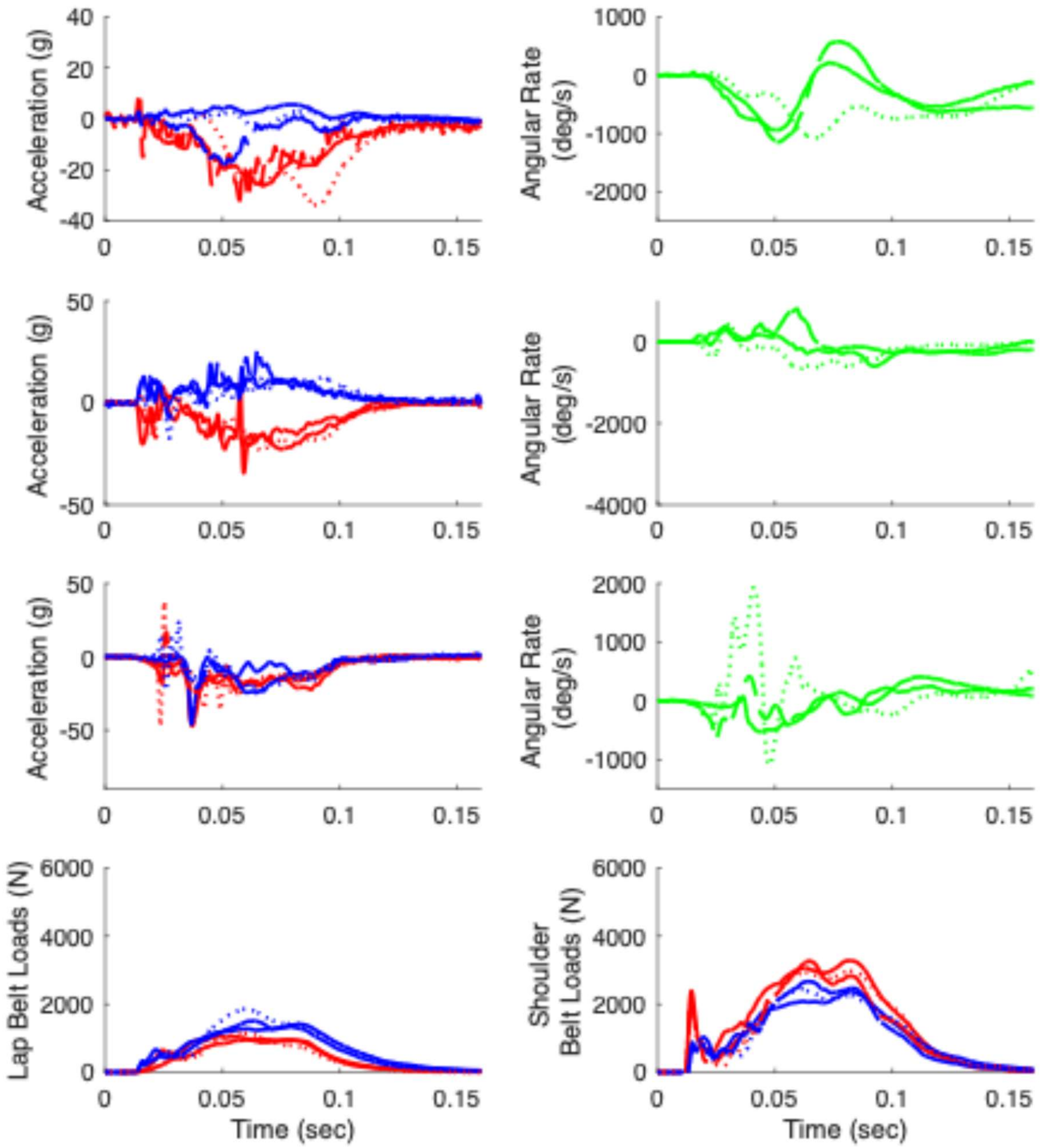

Fig. A1: OEM-Cohort tests with the full time-trace. Line styles (solid, dashed, and dotted) show the three tests. For the Head and T1 acceleration plots, red is x-acceleration and blue is z-acceleration. For the pelvis acceleration, red and magenta are the left and right ASIS x-acceleration while blue and cyan are the left and right ASIS z-acceleration. For the belt loads, the red lines are the right lap/shoulder belt and blue are the left lap/shoulder belt.

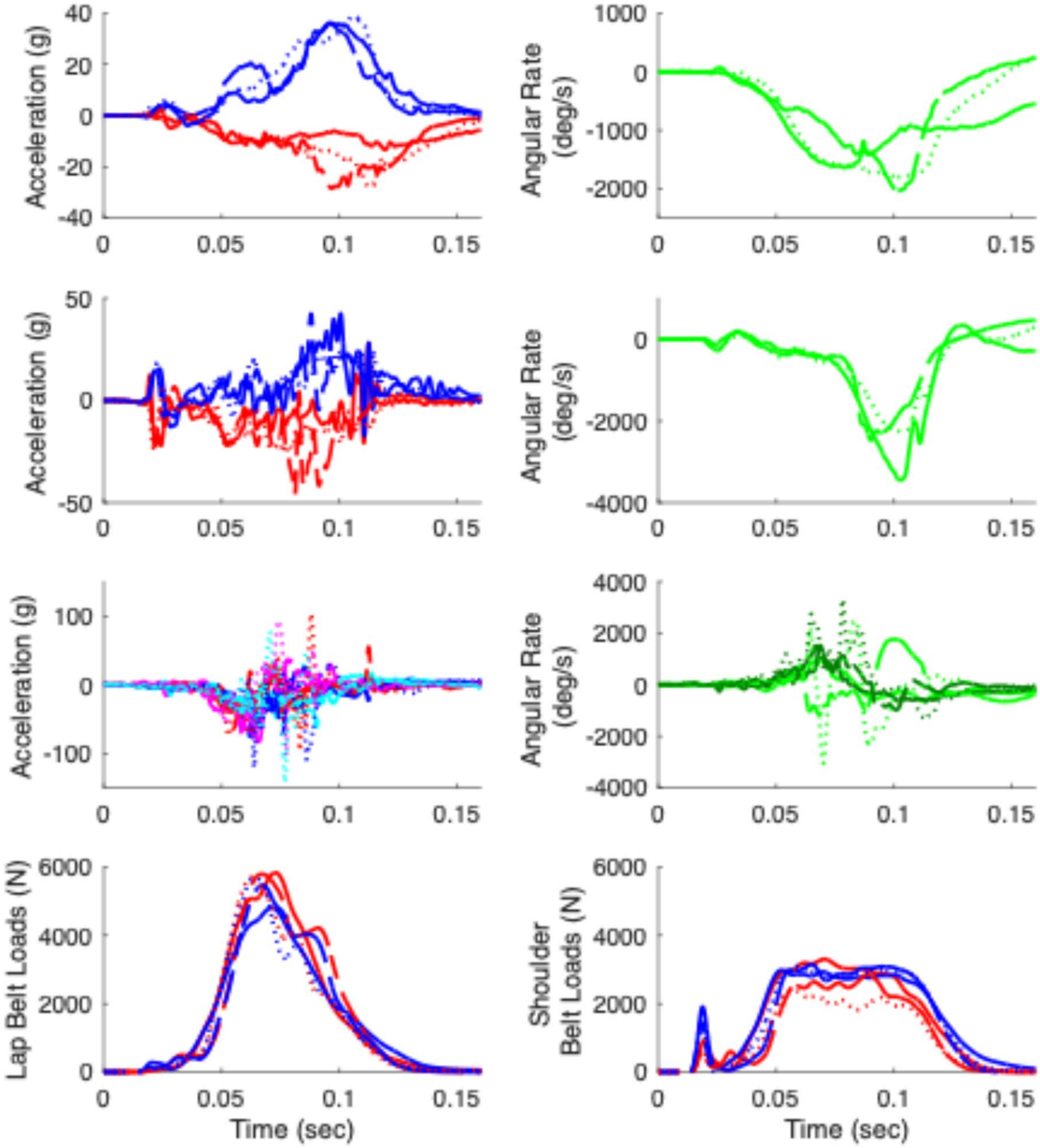

Fig. A2: SC-Cohort tests with the full time-trace. Line styles (solid, dashed, and dotted) show the three tests in order (solid is always the lowest test number and dotted is the highest). For the Head and T1 acceleration plots, red is x-acceleration and blue is z-acceleration. For the pelvis acceleration, red and magenta are the left and right ASIS x-acceleration while blue and cyan are the left and right ASIS z-acceleration. For pelvis angular rate, the green is the ASIS Left y-rotation and the brown is the ASIS right y-rotation. For the belt loads, the red lines are the right lap/shoulder belt and blue are the left lap/shoulder belt.