Comparison of Pelvis Kinematics and Injuries in Reclined Frontal Impact Sled Tests Between Mid-Size Adult Male and Female PMHS

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Abstract While differences in injury risk between males and females have been highlighted using field data, sex differences in kinematic response and injury tolerance remain understudied in the literature. This study aimed to provide a sex-based comparison of pelvis kinematics, injuries, and lap belt interaction in frontal impacts of reclined occupants. Frontal sled tests were conducted with reclined mid-size adult PMHS (four male and three female). Forward or minimally rearward pelvis pitch (sagittal plane rotation) was observed in all seven subjects. Good engagement of the lap belt with the pelvis was observed; partial submarining was identified in one male test. Maximum forward pelvis displacement of the females exhibited greater variation than that of the males (115-174 mm vs. 139-159 mm). Two male PMHS and two female PMHS sustained iliac wing fractures resulting from belt loading. Due to variations in pelvis geometry compared to male subjects, the H-point positioning target for female subjects needed to be translated forward relative to that of the male subjects to fit the female subjects in the seat. Lap belt anchor points were also moved forward to avoid potential differences in belt–pelvis interaction. Nonetheless, the males and females generally displayed similar injury patterns and pelvis kinematics.

Keywords Pelvis, Female, Sex, Recline, Submarining

I. INTRODUCTION

In a motor vehicle crash (MVC), the pelvis provides an important path through which restraint forces can be transferred to an occupant [1]. As such, the pelvis experiences substantial loading, therefore the lap belt must provide sufficient pelvis restraint while avoiding injuries: both fracture and submarining. Injuries to the pelvis, specifically iliac wing fractures, caused by lap belt loading have been reported in sled testing of post-mortem human subjects (PMHS) since the 1970s [2-7]. Submarining, wherein the lap belt slides over the iliac wing and into the abdomen [4-7], has been documented to be an unfavorable belt engagement condition [8-9], as the lap belt load is transferred from the bony pelvis to the soft abdomen. A higher likelihood of submarining has been suggested to contribute to the increased risk of injury and fatality in reclined occupants, and more recent efforts to properly restrain and mitigate submarining in reclined occupants have been presented in the literature [7][10]. Field data studies have shown that while MVCs with reclined occupants may have low incidence, they lead to higher injury severity [11-12]. Furthermore, the automotive industry suspects that highly automated vehicles may introduce shifts in occupant posture norms, with reclined occupants becoming more common [13-15].

The challenges to lap belt restraint of the pelvis may contribute to known sex differences in injury and fatality risk [16-21]. In particular, after accounting for delta-V, age, stature, BMI, and vehicle model year, female occupants have a greater risk of AIS 2+ injuries to the pelvis than males [19]. The disparity remains even after accounting for more detailed crash characteristics [22]. Though opposite extremes of the population have been considered in recent efforts to examine the whole-body response of reclined occupants [23-24], biomechanical data regarding mid-size females to study the potential cause of these differences in injury risk, especially in an automotive setting, are insufficient. Moreover, the pelvis is often regarded as the most sexually dimorphous bone [25-27]. Sex-based differences in pelvis shape, dimensions, pelvic ring geometry, and acetabular angle are

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well established [28-31]. Despite this stark dichotomy, there has been minimal analysis of the implications that these sex-based differences have in an automotive setting and specifically with regard to pelvis restraint with the lap belt. Further, robust restraint design, especially for reclined postures, should consider pelvis sexual dimorphism. Understanding the effects of sex-based variability on occupant restraint is especially important for the development of virtual testing methodologies, which can include statistical representations of population variances [32-33]. Therefore, this study aimed to compare pelvis kinematics, lap-belt interaction, and injuries between mid-size male and female PMHS in reclined frontal impacts.

II. METHODS

Frontal impact sled tests were conducted on seven adult PMHS (four mid-size male and three mid-size female) using a reverse acceleration sled system (1.4 MN ServoSled®, Seattle Safety, Auburn, WA, USA). PMHS testing procedures followed the ethical guidelines established by the National Highway Traffic Safety Administration (NHTSA) and were reviewed and approved by a biological protocol committee at the Center for Applied Biomechanics as well as the University of Virginia Institutional Review Board – Human Surrogate Use Committee. The PMHS were screened and returned negative results for bloodborne pathogens (i.e. HIV, Hepatitis B and C) and COVID-19 (female subjects only; male subjects were tested prior to the COVID-19 pandemic). The subjects were unembalmed and frozen until testing. Before testing, a full-body computed tomography (CT) scan and dual-energy X-ray absorptiometry (DXA) were used to verify the absence of pre-existing bone injury and to quantify bone mineral density (BMD) (Table I), an indicator of bone quality [34]. T-scores for whole-body BMD are also noted (Table I). The 2011–2012 and 2013–2014 National Health and Nutrition Examination Survey (NHANES) datasets [35] were used to set the target mid-sized female anthropometry, yielding target stature and mass ranges of 155–170 cm and 57–87 kg. All three female subjects had approximately mid-sized female anthropometry and all four male subjects had approximately mid-sized male anthropometry (Table I).

TABLE I
SUBJECT INFORMATION

Test	Subject	Age (years)	Cause of Death	Height (cm)	Weight (kg)	BMD (g/cm^2) [T-score]
S0529	M1	66	Dementia	175	74	1.065 [-1.3]
S0531	M2	72	Sepsis	185	74	1.133 [-0.7]
S0532	M3	25	Gunshot wound	174	75	1.221 [0.2]
S0533	M4	55	Myocardial infarction	180	74	1.009 [-0.8]
S0730	F1	64	End stage liver disease	170	57	0.989 [-0.9]
S0731	F2	58	Rectal Cancer	170	63	1.313 [2.3]
S0732	F3	62	Overdose	163	59	1.004 [-0.8]
Male Avg ± SD		54.5 ± 18.1	—	178.5 ± 4.4	74.3 ± 0.4	1.107 ± 0.08
Female Avg ± SD		61.3 ± 2.5	—	167.7 ± 3.3	59.7 ± 2.5	1.102 ± 0.15
Mid-size Male*		—	—	176	79	—
Mid-size Female*		—	—	163	72	—

*Target

Occupant Environment

Prior studies provide details regarding the male tests [7][36-38]. Specifically, an analysis on the pelvis kinematics and injuries observed in the male tests has previously been published [38], as well as a comparison of the response of the thoracolumbar spine between the male and female subjects [39]. The female tests were aimed to be conducted in a replicate condition to that of the male tests [36] (Fig. A1). A 35 g, 50 km/h sled pulse previously applied in PMHS sled tests to assess submarining risk [5][7] was used (Fig. A2). Subjects were seated in a passenger-side configuration on a semi-rigid seat designed to mimic the behaviour of a real vehicle seat [5-7][23][36-41]. A simulated vehicle seatback in the form of a tether system was used to allow for greater visibility of posterior motion-tracking instrumentation; a drop release mechanism was used to release the tethers at the start of the impact pulse. A three-point prototype seatbelt system consisting of dual lap belt pretensioners, a crash-locking tongue, a shoulder belt pretensioner, and a shoulder belt load-limiter was used. This concept seatbelt system was developed in concurrence with a research focus of this study: reduction of submarining risk in reclined occupants [10]. For each test, the shoulder belt retractor load-limiting factor was set to approximately 3.5 kN. The D-ring was positioned to simulate a seatback-integrated placement: the shoulder belt take-off angle, defined as the angle between the horizontal and the path of the webbing from the shoulder to the D-ring, was approximately 12°. The lap belt webbing was routed as low across the pelvis as possible, with no folding or roping. No knee bolster or instrument panel was present, and the subjects' feet were fastened into an angled foot pan that prohibited any translation or rotation of the feet.

Subject Positioning

Positioning and orientation targets were established from the male tests and utilised for the subsequent female tests. All subjects were reclined to a seatback angle of approximately 50° from the vertical. The torso angle, defined as the angle between the vertical and the line connecting the H-point (approximated by palpation of the greater trochanter) to the acromion (mean ± SD: 46 ± 1 deg), was used as a reference target. The pelvis positioning target that had been established from the male tests required some adjustments. During the first female test, when matching the H-point to the target established in the male tests, the sacrum ended up too rearward, which was significant cause for concern as it indicated a high likelihood for the sacrum to arrest its motion on the rear edge of the seat pan and induce subsequent disruption in pelvis kinematics and/or pelvic injury. Furthermore, this positioning resulted in the posterior pelvis flesh extending over the back edge of the seat pan. Neither of these concerns had been observed in the male subjects. Thus, the H-point target for the females was moved 40 mm in the +X direction of the global coordinate system (Fig. 1) relative to that for the males (Fig. A1) to prevent sacrum interaction with the back edge of the seat pan. As a result, the target H-point X-coordinate for the males and females in the global coordinate system were +130 mm and +170 mm, respectively. The target H-point Z-coordinate was not controlled. The buckle and lap belt retractor anchor points were also moved forward accordingly to allow for consistent lap belt and buckle take-off angles and avoid the potential for differences in belt-to-pelvis interaction. This ensured that the lap belt loaded similar anatomical locations in the females as in the males. Pelvis orientation was defined as the "notch angle", i.e. the angle in the sagittal plane between the horizontal and the line connecting the midpoint of the Anterior Superior Iliac Spines (ASIS) and the midpoint of the Anterior Inferior Iliac Spines (AIIS) (Fig 2.) [41-42]. The notch angle was used as it may better describe the relationship between pelvis loading from restraints and pelvis orientation than other previously defined pelvis angles, such as the Nyquist angle, i.e. the angle in the sagittal plane between the horizontal and the line connecting the midpoint of the ASIS and the pubic symphysis [42-43]. Pelvis orientation ranged from 41 to 47° for the male subjects and 43 to 45° for the female subjects (Table II).

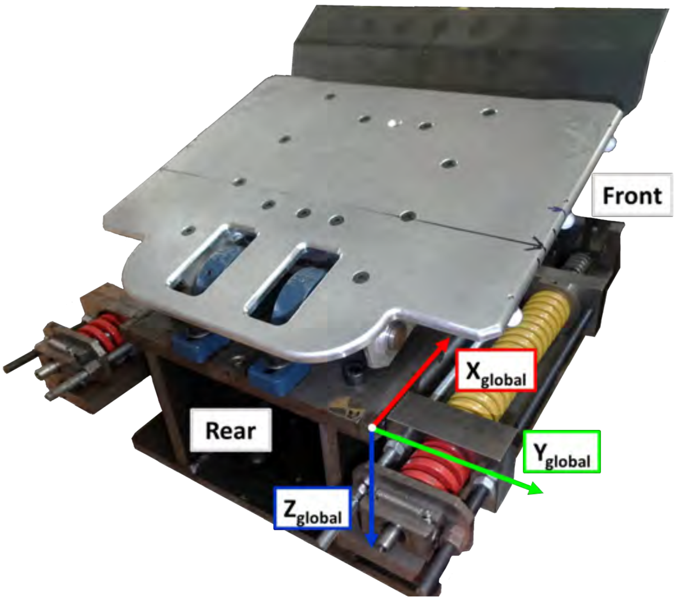


Fig 1. Global coordinate system.

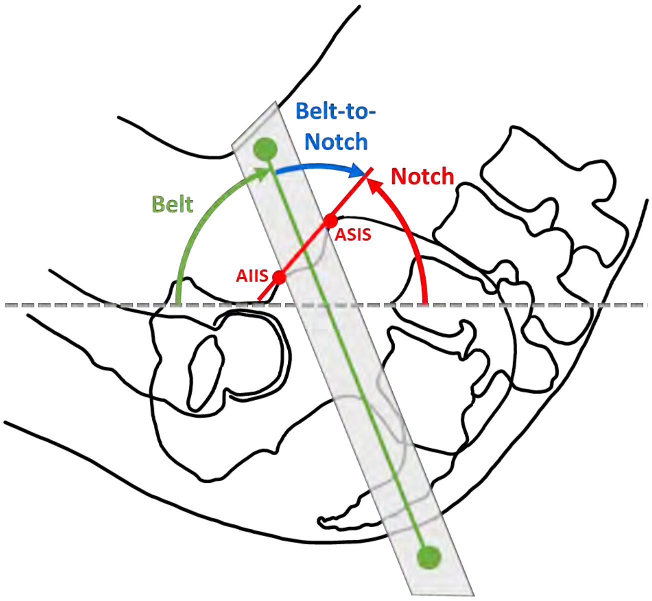


Fig 2. Definition of lap belt, pelvis (Notch), and belt-to-Notch angles.

Instrumentation and Injury Diagnosis

An optoelectronic motion stereophotogrammetric system (Vicon MX^{TM}, Vicon, Centennial, CO, USA) was used to track motions of spherical markers attached to the buck, seat, seatbelt, and subject to track three-dimensional kinematics. A marker plate was rigidly affixed to the pelvis with screws going into the left and right posterior superior iliac spines (PSIS). For the male tests, posterior pelvis soft tissue was clamped between the marker plate and the iliac wings due to the fastening method (Fig. 3). However, the same fastening method was not feasible for the female tests due to the greater amount of posterior pelvis soft tissue that was observed. Consequently, for the female tests, the marker plate was instead clamped to the heads of the screws protruding from the iliac wings (Fig. 3). The markers were then tracked over time and allowed for position and orientation calculations of the iliac wings using rigid body mechanics and coordinate transformations [36][44]. Initial positions of the pelvis, seat pan, and lap belt webbing (reconstructed from motion capture data) were used to determine inferior soft tissue height, defined as the average of the distances from the tip of the coccyx and the ischial tuberosity to the seat pan, and anterior soft tissue depth, defined as the distance between the midpoint of the left and right ASIS and the lap belt at the subject's midline (Fig. A3). Two strain gauge rosettes (C2A-06-062WW-350, Micro-Measurements®, Raleigh, North Carolina, USA) were affixed to the lateral surface of each iliac wing to obtain fracture timing data (Fig. 3). Lap belt forces were measured through a belt tension gauge located between the outboard retractor and the subject's right hip. Post-test CT scans were interpreted, and injuries diagnosed, by a board-certified radiologist. Post-test PMHS dissection supplemented injury analysis.

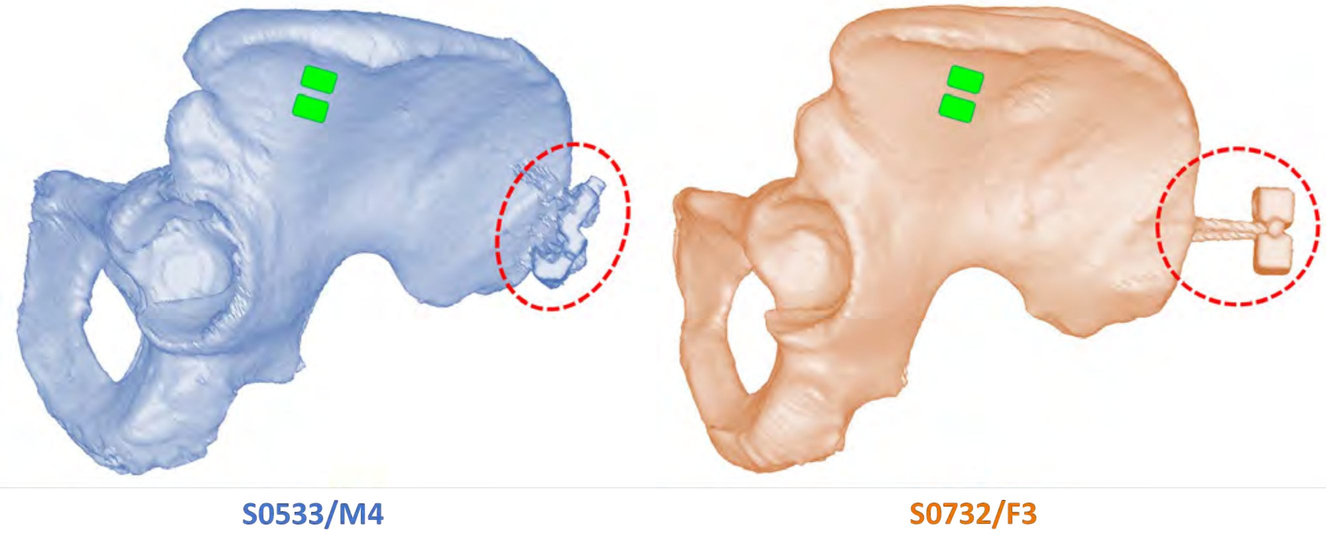
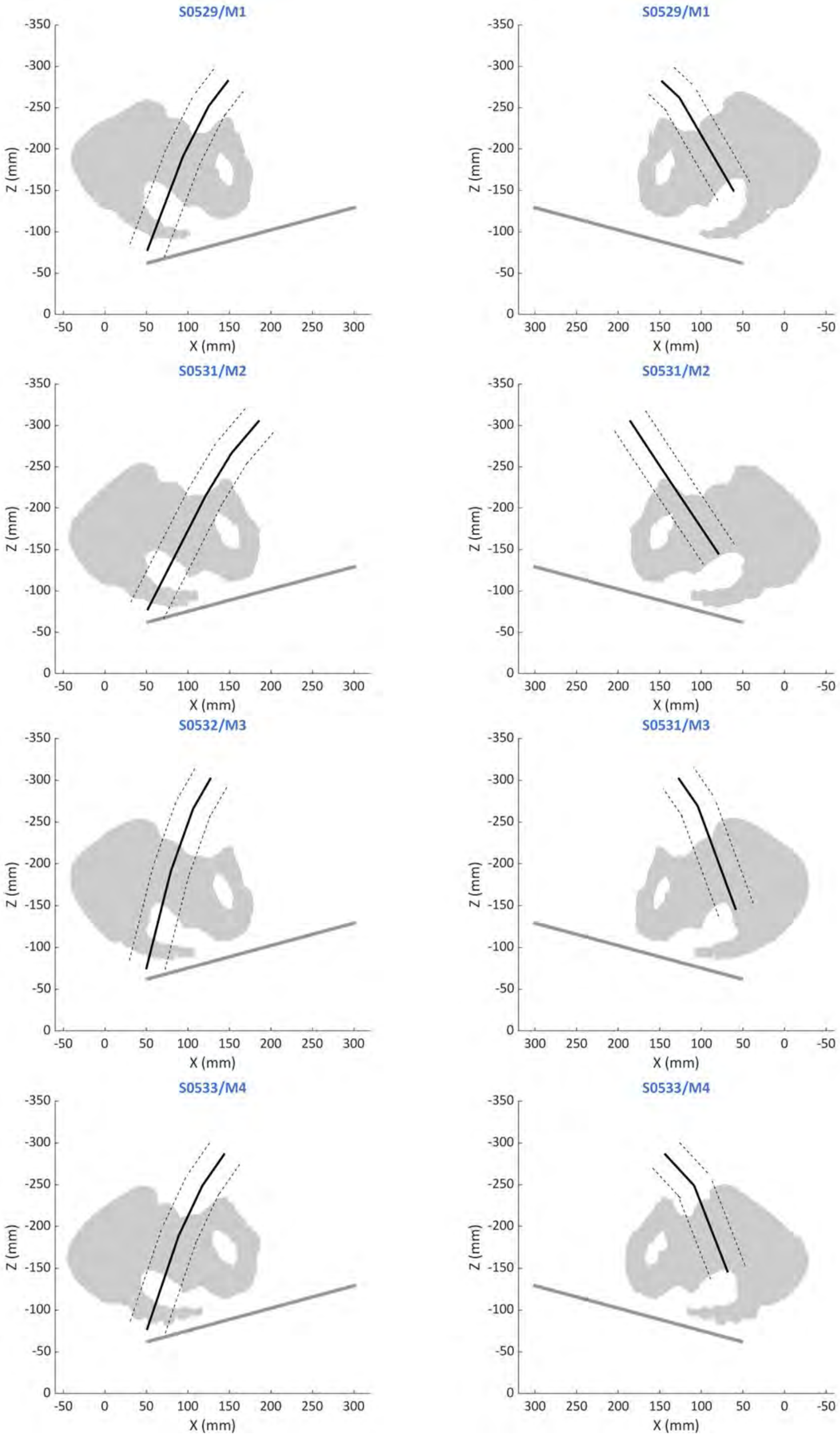


Fig. 3. Alteration to pelvis hardware for motion-tracking array. Left: method for male tests. Right: method for female tests. Approximate locations of strain gauge rosettes are outlined in green.

III. RESULTS

Initial Positioning

Initial positions of the pelvis (iliac wings and sacrum), lap belt, and seat pan were reconstructed using motion capture data (Fig. 4). Though the masses and statures of the females were less than those of the males, pelvis size was comparable across sex (Table II). Sex-based differences in pelvis geometry in the sagittal plane were noted; in their respective pre-test positions: the ischial spine is further forward than the tip of the coccyx for the females, but not for the males (Fig. 5; Fig. A4). Initial pelvis orientation, defined by the Notch angle, was comparable between males (40° to 47°) and females (43° to 45°), as were initial Nyquist angles [43], with angles marginally greater for the male subjects (Table II). Inferior soft tissue heights were generally higher in the females than in the males (Table III). Pre-test belt angles, both inboard (male: 56°–66°; female: 53°–61°) and outboard (male: 60°–71°; female: 56°–63°), were also comparable across sex (Table III). Anterior soft tissue depths were more variable. Two subjects (one male and one female, both approximately 140 mm) showed greater anterior soft tissue depth than the other five subjects (about 80–90 mm) (Table III).


S0529/M1
S0529/M1
S0531/M2
S0531/M2
S0532/M3
S0531/M3
S0533/M4
S0533/M4
Z (mm)
X (mm)
-350
-300
-250
-200
-150
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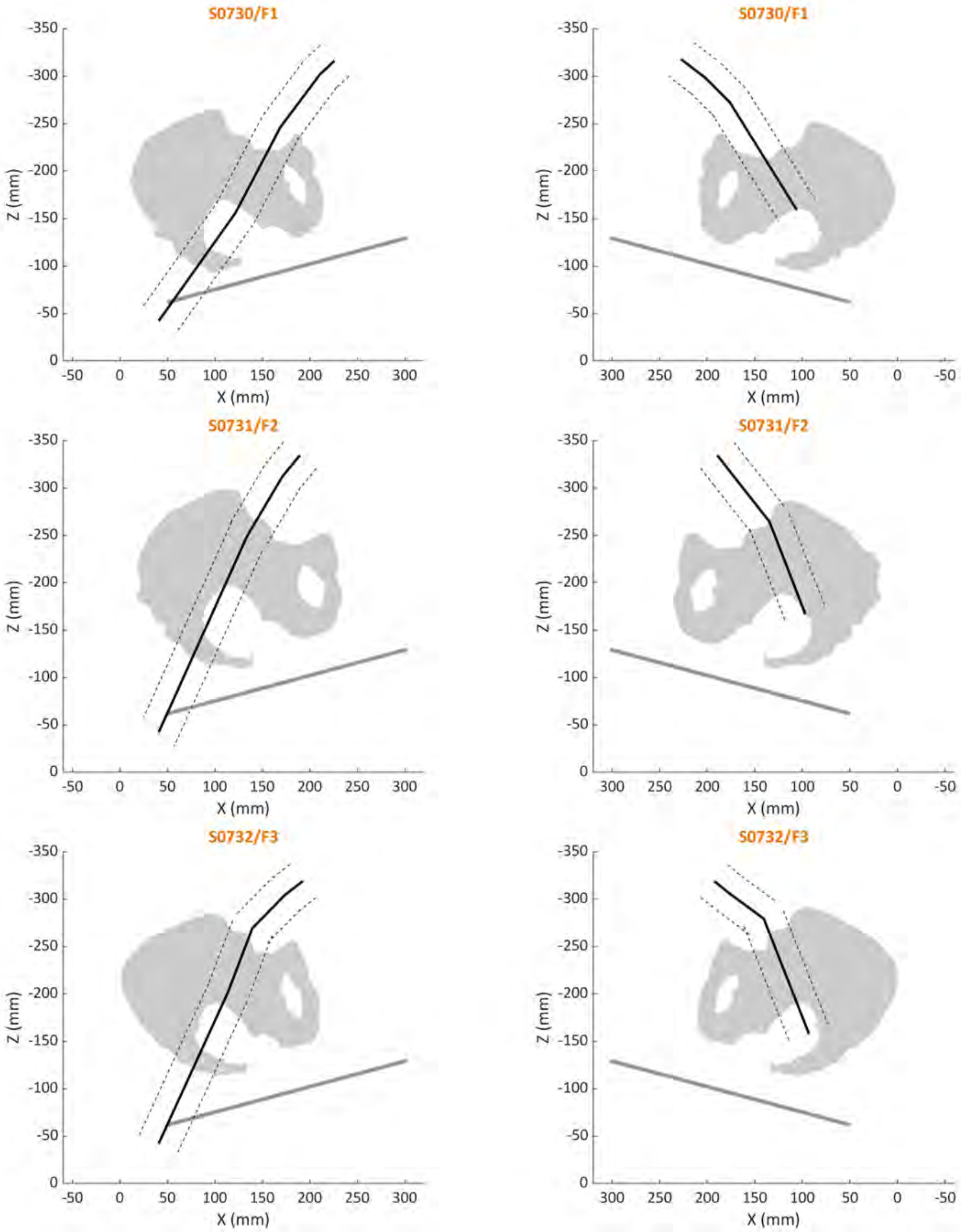
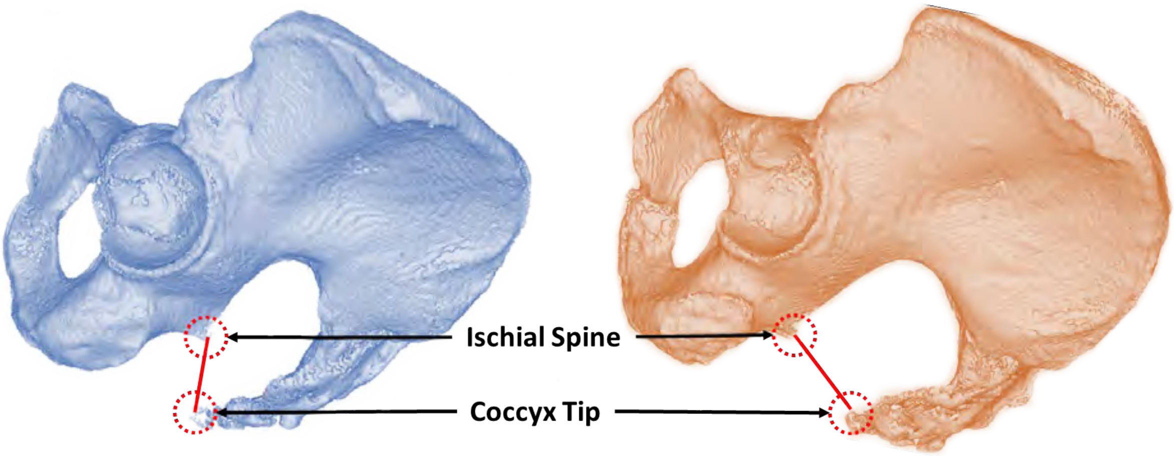


Fig. 4. Initial positions of the pelvis, lap belt, and seat pan in the global coordinate system, as viewed from the outboard (left column) and inboard (right column) sagittal planes. The solid black line connects motion-capture markers placed along the centre (width-wise) of the lap belt. The dashed line approximates the edges of the lap belt. The grey pelvis outline corresponds to each subject's reconstructed pre-test pelvis CT scan.



S0529/M1 **S0730/F1**

Fig. 5. Sex-based difference in pelvis geometry in pre-test position: the ischial spine is further forward to the tip of the coccyx in the females, while the ischial spine is further rearward to the tip of the coccyx in the males.

TABLE II
PELVIS MEASUREMENTS

Test	Subject	Notch Angle (deg)	Nyquist Angle (deg)	Pelvis Depth* (mm)	Pelvis Height* (mm)	Pelvis Width* (mm)
S0529	M1	43	166	220	175	290
S0531	M2	40	168	230	174	302
S0532	M3	44	160	214	170	260
S0533	M4	47	170	227	168	295
S0730	F1	43	167	207	166	247
S0731	F2	45	156	216	184	269
S0732	F3	45	157	209	174	290
Male Average ± SD		43.5 ± 2.5	166 ± 3.7	222.8 ± 6.2	171.8 ± 2.9	286.8 ± 16.0
Female Average ± SD		44.3 ± 0.9	160 ± 5.0	210.7 ± 3.9	174.7 ± 7.4	268.7 ± 17.6

*See Fig. A5 for definitions

TABLE III
LAP-BELT AND SOFT TISSUE MEASUREMENTS

Test	Subject	Inboard Belt Angle (deg)	Outboard Belt Angle (deg)	Inferior Soft Tissue Height (mm)	Anterior Soft Tissue Depth (mm)
S0529	M1	57	65	23	81
S0531	M2	56	60	14	142
S0532	M3	66	71	16	82
S0533	M4	62	66	19	86
S0730	F1	53	56	22	143
S0731	F2	61	63	29	79
S0732	F3	58	61	39	91
Male Average ± SD		60.3 ± 4.0	65.5 ± 3.9	18 ± 3.4	97.8 ± 25.6
Female Average ± SD		57.3 ± 3.3	60.0 ± 2.9	30 ± 7.0	104.3 ± 27.8

Pelvis and Lap Belt Kinematics

Maximum forward pelvis excursions varied within sex and between sex, with male subjects ranging from 140 mm to 165 mm and female subjects ranging from 115 mm to 170 mm (Fig. 6). These peak excursions occurred around 65 to 70 ms in all tests. The male subjects exhibited minimally rearward or forward rotation of the pelvis

(Fig. 7). While the females exhibited a similar response, for two subjects (S0731/F2 and S0732/F3) an oscillatory signal was observed. The pelvis mount of one male (S0531/M2) and one female subject (S0730/F1) interacted with the seat pan at 30 ms and 46 ms, respectively. While this did not affect the subjects' motions and minimally affected measured displacements, the pelvis pitch for these two subjects was truncated.

Lap belt angle (as defined in Fig. 1) measured on the outboard side varied more over time in the male subjects than in the female subjects (Fig. 8). The peak lap belt angle occurs approximately at the same time as peak pelvis forward excursion (60–70 ms). The belt-to-Notch angle (as defined in Fig. 2) provides a representation of how the lap belt interacts with the pelvis (Fig. 9); an angle of 90° indicates that the lap belt is perpendicular to the line connecting the ASIS to the AIIS. As forward pelvis excursion occurs, the lap belt-to-pelvis angle increases, peaking close to 90°.

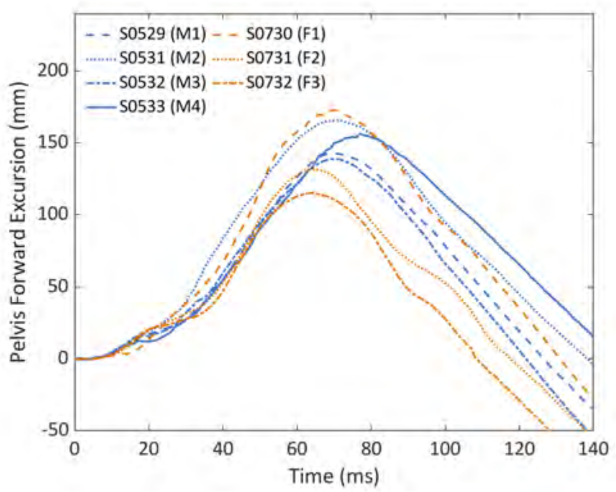


Fig. 6. Pelvis forward excursion time-history.

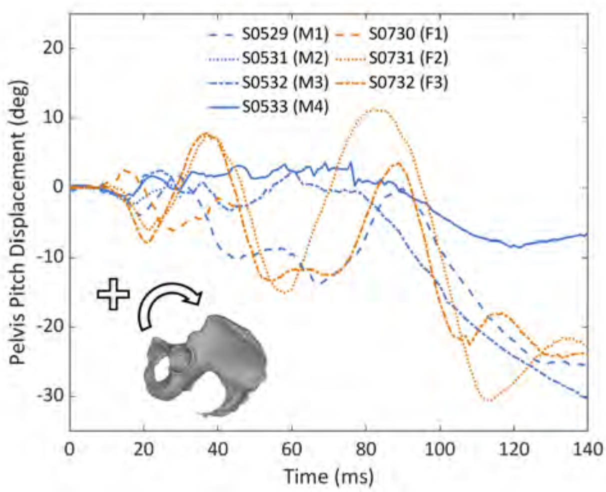


Fig. 7. Pelvis pitch displacement time-history. Signals for S0531/M2 and S0730/F1 were omitted after 30 and 46 ms, respectively, due to pelvis motion-tracking hardware interaction with the seat pan.

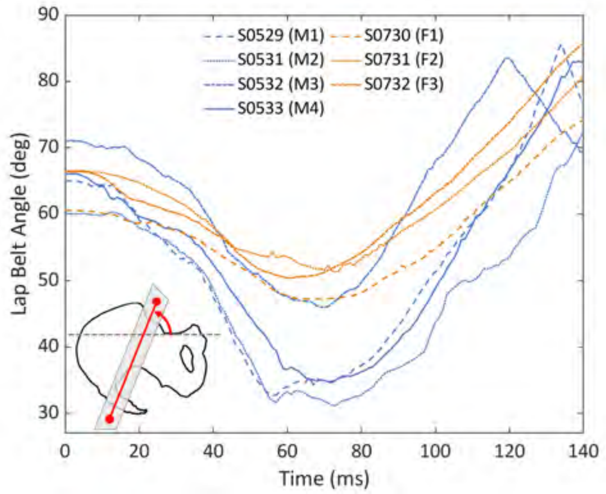


Fig. 8. Outboard lap belt angle time-history.

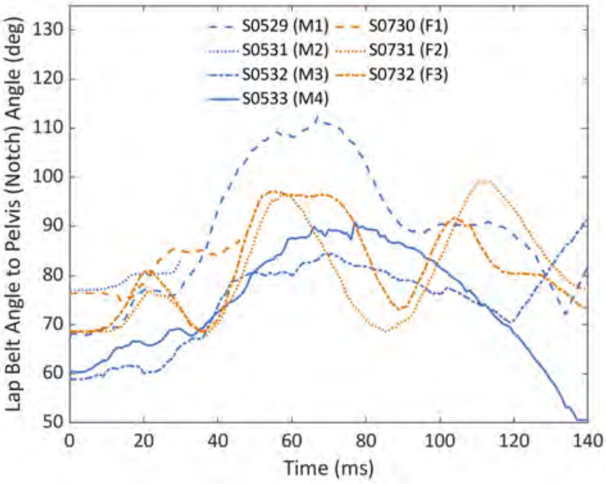


Fig. 9. Belt-to-Notch angle time-history. Signals for S0531/M2 and S0730/F1 were omitted after 30 and 46 ms, respectively (reasoning same as pelvis pitch).

Injuries and Submarining

Injuries to the pelvis were identified from post-test CT scans and confirmed by post-test dissection. Iliac wing fractures near the ASIS were observed in four of the seven subjects (Table IV). For the two male subjects that sustained a fracture, significant comminution of the right iliac wing (outboard/buckle side) was observed [38]. One female subject displayed a fracture on both iliac wings, with significant deformation on the right (similar to the male subjects) and less on the left (Fig. 10). The other female subject displayed minor deformation on the right iliac wing (Fig. 10). Iliac wing surface strain data were used to determine iliac wing fracture timing. For the first and second male tests, fracture timing was determined to be at 63 ms and 56 ms, respectively [38], while for the fracture cases from the female tests the strain data were less conclusive and exact fracture timing could not be determined. Sacrum fractures were observed in all subjects, with fracture level ranging from S2 to S5. Partial submarining on the inboard (buckle) side (left iliac wing) was observed at approximately 60 ms in one of the male subjects (S0533/M4). Submarining was not observed in any of the female tests.

TABLE IV
OBSERVED INJURIES

Test	Subject	Pelvis Fracture	Sacrum Fracture	Submarining
S0529	M1	R	S3-S4	—
S0531	M2	R	S4	—
S0532	M3	—	S5	—
S0533	M4	—	S3	L
S0730	F1	R + L	S2-S3	—
S0731	F2	R	S4-S5	—
S0732	F3	—	S2-S3 + Coccyx Dislocation	—

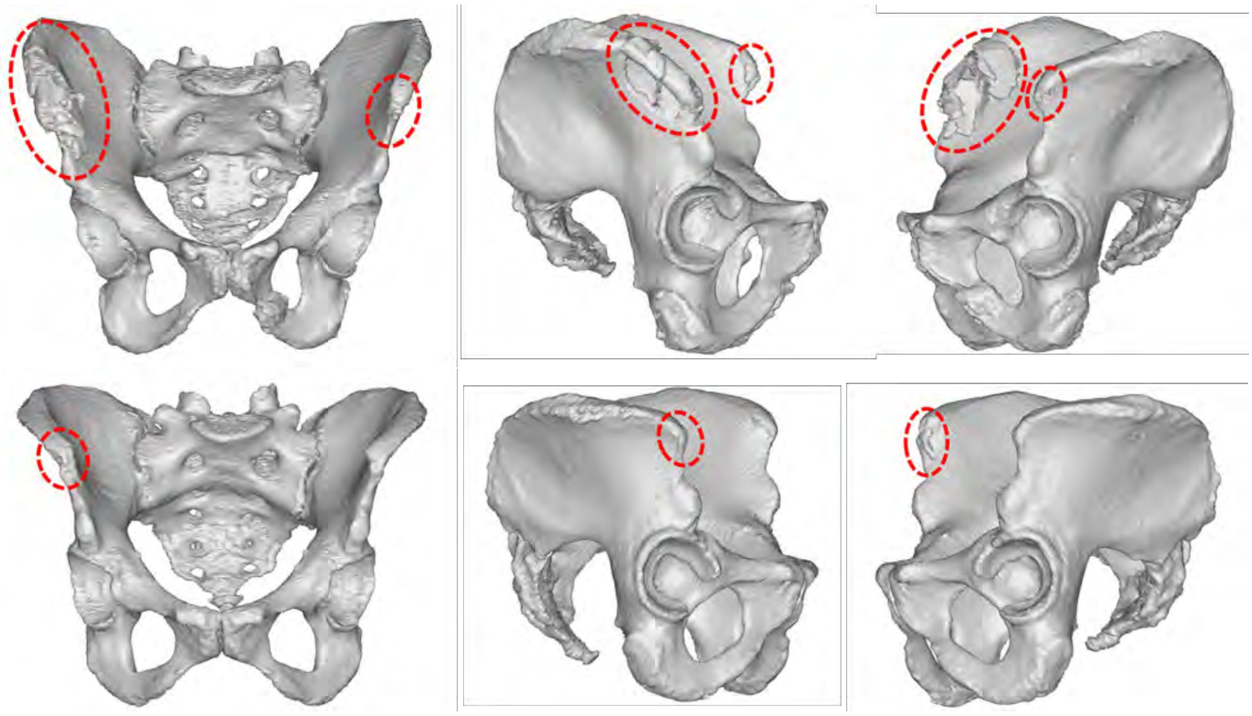


Fig. 10. Reconstructed post-test pelvis CT scans from S0730/F1 (top row) and S0731/F2 (bottom row), showing the fractures at iliac wings near the ASIS landmarks in both subjects. A similar figure for the two fracture cases in the male tests can be found in Richardson et al. [38].

Environment Responses

Generally, maximum lap belt force was greater for the males (7.0 kN to 8.3 kN) than the females (5.8 kN to 7.2 kN) but occurred at approximately a similar time (65 ms) (Fig. A6). One female subject (S0730/F1) exhibited less lap belt force after 30 ms, but the general timing of the loading and unloading phases were similar to the other tests. Angular displacement of the seat pan was greater in the females than in the males, with peak deflections occurring around the same time as maximum lap belt forces (65 ms) (Fig. A6, Fig. A7). Angular displacement of the anti-submarining pan followed a similar trend, with one male subject exhibiting comparable deflection to the females (Fig. A8).

IV. DISCUSSION

This study compared kinematics and injuries of the pelvis between females and males in a reclined frontal impact condition. The data presented provides insight into sex-based responses in occupant kinematics and injuries, which can be used for virtual testing efforts that require validated human body models and vehicle environment models [32]. Prior analysis of reclined mid-size male PMHS showed generally consistent pelvis kinematics (e.g. forward excursion and sagittal plane rotation) among those male subjects, with injuries, namely ASIS fractures and partial submarining, potentially explained by an unfavourable combination of lap belt loading direction and initial pelvis orientation [38]. Richardson *et al.* [38] further proposed identification of restraint design targets that identify a compromise between pelvis injury tolerance and submarining sensitivity for better occupant protection. Overall, male and female kinematics and injuries were indistinguishable, with differences in results possibly attributable to methodological differences between the male and female tests. These findings from the current study further corroborate the suggestion from Richardson *et al.* [38] that restraint design for reclined occupants should focus on providing sufficient lap belt load, directed to reduce the risk of submarining, without causing iliac wing fractures.

Despite a forward shift of the target female H-point position, corresponding shifts to restraint anchor points led to similar initial inboard and outboard belt angles between the females and males. Statistical models of volunteer data show that with a seatback, initial positions of females' pelvises tend to sit further forward than those of males [45]. This suggests not only that the differences observed in pelvis geometry that instigated the forward shift of the H-point target for the female tests may be representative of the greater female population, but also that, with fixed restraint anchor points, females may be subject to shallower initial belt angles than males. Anterior soft tissue depth appeared to play a larger role than sex in initial belt angles; the two subjects (S0531/M2, S0730/F1) with more anterior pelvis soft tissue resulted in lower (shallower) initial belt angles.

Maximum forward pelvis excursion was largely comparable between males and females. Forward rotation of the torso resulting in the shoulders being positioned farther forward than the pelvis are favourable biomechanical conditions for reducing the submarining risk [8]. However, reclined occupants exhibit an unfavourable initial posture; the shoulders start farther rearward to the pelvis. An inherent design goal of the prototype restraint system used was to prevent significant forward excursion of the pelvis, to allow the shoulders to rotate up and over the pelvis. Anterior soft tissue depth appeared to play a greater role in pelvis displacement; the two subjects with more anterior pelvis soft tissue (S0531/M2, S0730/F1) resulted in the two greatest forward pelvis excursions. Even with dual lap belt pretensioners, there likely was still more soft tissue that was able to compress post-pretensioning in those two subjects. The third highest case of forward pelvis excursion came from the subject that experienced partial submarining (S0533/M4), which is to be expected as submarining prevents the lap belt from constraining forward pelvis motion. Of the remaining four tests with comparable soft tissue depth and no observed submarining, the females exhibited marginally (approximately 15 mm) less forward pelvis excursion than the males. This may be due to the initial position of the H-point; as the females' H-points were positioned farther forward, it is likely that the females' thigh tissue and pelvises engaged with the anti-submarining pan earlier, which would restrict forward movement. Greater deflections of the anti-submarining pan for the female subjects lend credence to this theory (Fig. A8). Another possibility is the lower mass of the females resulting in less energy required to restrain the subject. The similarities between males and females in pelvis forward excursion suggest that the concept seatbelt system used was similarly effective in minimising forward pelvis excursion in both males and females.

Similar to the male subjects, the female subjects exhibited minimal to no rotation of the pelvis. Forward rotation of the pelvis is more favourable for reducing submarining risk than rearward rotation as the lap belt is more likely to engage with the iliac wing with forward rotation. A key distinction in pelvis pitch between the males and females is a more oscillatory response in the females' signals, especially in S0731/F2 and S0732/F3. Examining pelvis pitch concurrently with seat pan deflection provides an insight into how the females' pelvises (specifically S0731/F2 and S0732/F3) rotate throughout the impact pulse (Fig. 13). From 0 ms to 20 ms (Fig. 13A), both the pelvis and seat pan rotate forward as a result of pretensioning. From 20 ms to 35 ms (Fig. 13B), seat pan rotation plateaus, and rearward rotation of the pelvis is observed as a result of axial compression in the spine that is applied to the posterior side of the pelvis. From 35 ms to 55 ms (Fig. 13C), the seat pan continues to rotate forward, and the females' pelvises also pitch forward. Just as the pelvis begins to rebound, around 55 ms (Fig. 13D), the seat pan begins to rotate rearward, and a similar rotation of the pelvis is observed.

To further clarify the cause(s) of the oscillatory behaviour in pelvis rotation, the authors intend to conduct a computational study using human body models aimed to examine the sensitivity of pelvis motion to pelvis bony and soft tissue geometry. Typically, experimental data need not be validated, but because the pelvis rotation data cannot be fully validated at this time, further analysis of the pelvis rotation data is needed to corroborate the unexpected oscillatory nature of those signals. While the alteration to the pelvis mount design made to accommodate the females' pelvis soft tissue geometry (Fig. 3) may have induced soft tissue interaction with the motion-tracking hardware, there was no indication that the motion-tracking hardware for the pelvis was loosely fastened in the female tests, and as such, soft tissue interaction was deemed to be negligible. On the other hand, the presence or absence of soft tissue about the pelvis, especially in the space between the pelvis and the seat pan, may have played a role in instigating the oscillatory responses. Inferior soft tissue was greatest in the two subjects (S0731/F2: 39 mm, S0732/F3: 29 mm) that exhibited the most oscillatory responses, and next greatest in the two other subjects (S0529/M1: 23 mm, S0730/F1: 22 mm) that exhibited oscillatory responses. The differences in pelvis pitch responses may even be attributable to the observed fundamental variations in pelvis geometry (Fig. 5) (e.g. variations in the point about which the pelvis rotates). These above listed potential causes, in conjunction with seat pan rotation, may have brought about the oscillatory nature of some rotation signals, and should be considered as parameters in the computational sensitivity study.

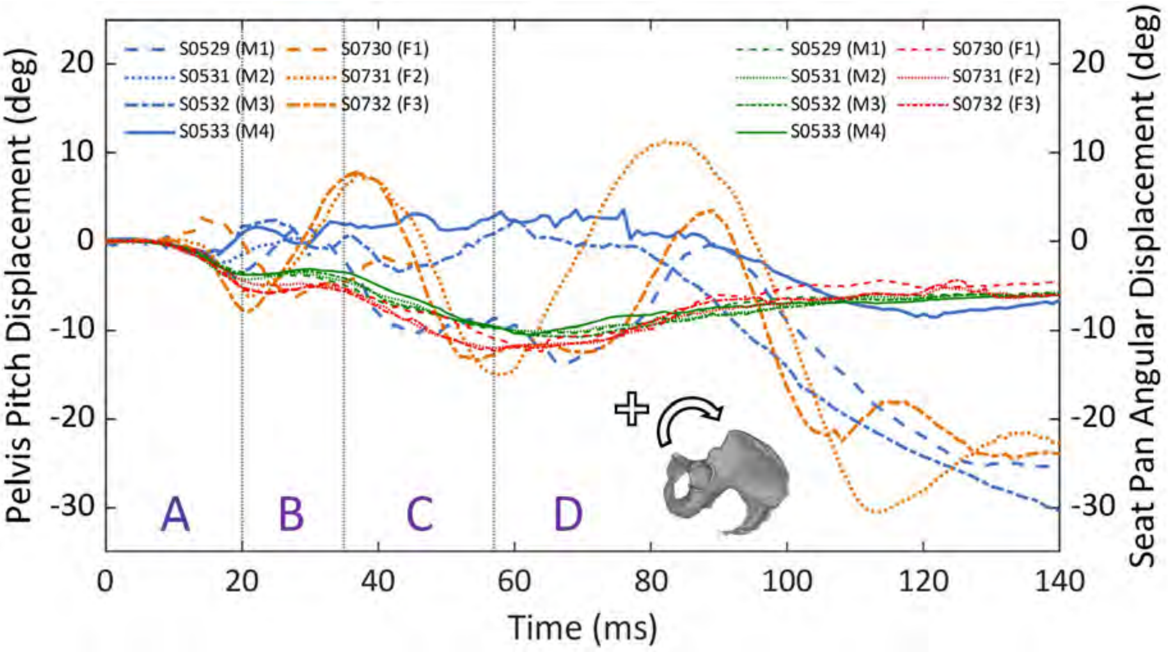


Fig. 13. Pelvis pitch displacement time-history (male: blue; female: orange) overlayed on seat pan angular displacement time-history (male: green; female: red). Forward rotation is negative for both pelvis pitch and seat pan angular displacement.

The female subjects experienced less change in outboard lap belt angle over time than the male subjects. These differences in lap belt angle time-histories may be explained by the forward shift of the H-point target in the females; more of the thigh and pelvis likely engaged with the anti-submarining pan earlier. The lap belt-to-pelvis angle time-history shows that the lap belt engaged with the pelvis similarly between the males and females throughout the duration of the test. These results are consistent with those from a study that aimed to replicate lap belt loading conditions from a sled test in a non-impact dynamic environment [40]. Future work should investigate lap belt angle time-histories with the same fore-aft H-point target for both males and females. Similarities between males and females in pelvis rotation, lap belt angle time-histories, and lap belt to pelvis angle time-histories all suggest that lap belt to pelvis interaction are comparable across sex.

Similar injury patterns were observed between the male and female tests. Four of the seven subjects (two male and two female) sustained iliac wing fractures, with three sustaining just a right iliac wing fracture and one sustaining bilateral iliac wing fractures. The more significant fractures all occurring on the outboard side may be a function of the restraint geometry. As noted by Richardson *et al.* [38] a combination of a forward-pitched pelvis and a horizontally oriented lap belt angle may have increased the effective load of the lap belt on the pelvis. The peak lap belt loads for the two male fracture cases, S0529/M1 and S0531/M2, were 7.8 kN and 6.6 kN, respectively. Despite being subject to the highest lap-belt force (8.5 kN), S0532/M3 was potentially uninjured due to this subject's young age. The remaining male subject, S0533/M4, had a comparable peak lap-belt force (7.6 kN) to the two male fracture cases, but the partial submarining on the left iliac wing may have prevented a fracture since some of the lap belt load was transmitted to the abdominal soft tissues. The peak lap belt loads for the two female fracture cases, S0730/F1 and S0732/F2, were 5.8 kN and 6.9 kN, respectively. The female non-fracture case, S0732/F3, was subject to the highest lap belt load among the females (7.2 kN), likely due to the absence of a fracture. As noted by Richardson *et al.* [35], a high lap belt force may be achieved if a subject does not sustain a fracture, since a fracture will reduce peak lap belt force.

The time of fracture, as far as could be determined, did not vary by sex. Fracture timing for S0529/M1 and S0531/M2 was determined to be near peak lap belt load and peak forward excursion, at approximately 60 ms [38]. For S0731/F2, due to the minimal nature of the fracture, it was difficult to identify fracture timing from strain gauge data. For S0730/F1, strain gauge data provided no clear indication of fracture timing, but as lap belt force was less than the other tests, it was less straightforward to pinpoint fracture timing. Several timepoints along the lap belt force time-history can be noted as potential fracture times: plateaus in belt load near 42 ms and 45 ms may be indicators of fracture. It is unlikely that the fracture occurred during pretensioning, as lap belt loads are consistent with other tests until 40 ms. Both within and between sex, a clear correlation between incidence of pelvis fracture and BMD could not be established.

All subjects sustained sacrum fractures; previous studies with both upright and reclined occupants have noted a similar phenomenon [5][40]. However, this may be an artefact of the rigid nature of the surface of the seat pan, which was consistent across both aforementioned studies. It is unknown exactly when these sacrum fractures occurred, but the high internal axial forces experienced during initial forward excursion [39] may contribute to the injury mechanism. Future research should examine the effect of seat padding and seat surface stiffness on the occurrence of sacrum fractures.

Environment responses were also comparable between males and females. Females exhibited less peak lap belt loads, but this may be explained by the lighter masses of the female subjects. Greater comminution of the right iliac wing may explain why one male subject (S0531/M2) and one female subject (S0730/F1) showed lower lap belt loads than the rest of the males and females, respectively. The forward shift in H-point target for the female tests likely caused greater deflection of the seat pan. The load imparted on the seat pan, caused by the inertia of the subject’s body, was applied further away from the seat pan’s axis of rotation, resulting in a greater moment despite the lighter masses of the females. Deflections of the anti-submarining pan were generally greater in the females, again likely due to the forward shift of the H-point. The two subjects with the greatest anti-submarining pan deflection (S0531/M2, S0730/F1) exhibited the most forward pelvis excursion, which was likely because they also possessed the greatest amount of anterior soft tissue depth.

The results of this study further highlight, as noted by Richardson *et al.* [38], the need for future research into the tolerance of the pelvis, specifically near the ASIS where the lap belt load is applied, as a function of extrinsic restraint design factors (e.g. belt angle, belt load, recline angle, seat compliance). Lap belt load limiters have been presented as a potential modification to the current restraint system design that would alleviate concerns regarding iliac wing fractures and have been applied with varying success [5][40]. It should be noted that the studies utilising lap belt load-limiters were conducted with mid-size male PMHS, and though the results of this study suggest that there are minimal sex-based differences in pelvis kinematics and injuries of reclined occupants, future research should consider applying lap belt load-limiting to mid-size female PMHS.

To the authors’ knowledge, the current study is one of few to analyse the kinematic and injury response of adult mid-size female PMHS in an automotive setting, and one of the first to directly compare pelvis differences between mid-size females and mid-size males. Further, this study contributes to the literature data regarding the response of reclined PMHS in a highly automated vehicle environment.

V. CONCLUSIONS

A sex-based comparative analysis of pelvis kinematics and injuries of reclined occupants in a highly automated vehicle environment was performed. Modifications to the target H-point and restraint anchor positions in the female tests were necessitated due to differences in pelvis geometry and soft tissue thickness between the male and female subjects observed prior to testing. Generally, similar kinematic and injury responses between the males and females were evident, with the H-point target modification likely accounting for minor discrepancies. The seat-integrated, submarining-mitigating restraint system was successful in protecting both female and male reclined occupants from submarining. Future research foci should be centered around optimising restraints to successfully reduce the risk of submarining while concurrently preventing iliac wing fractures.

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VIII. APPENDIX

APPENDIX A

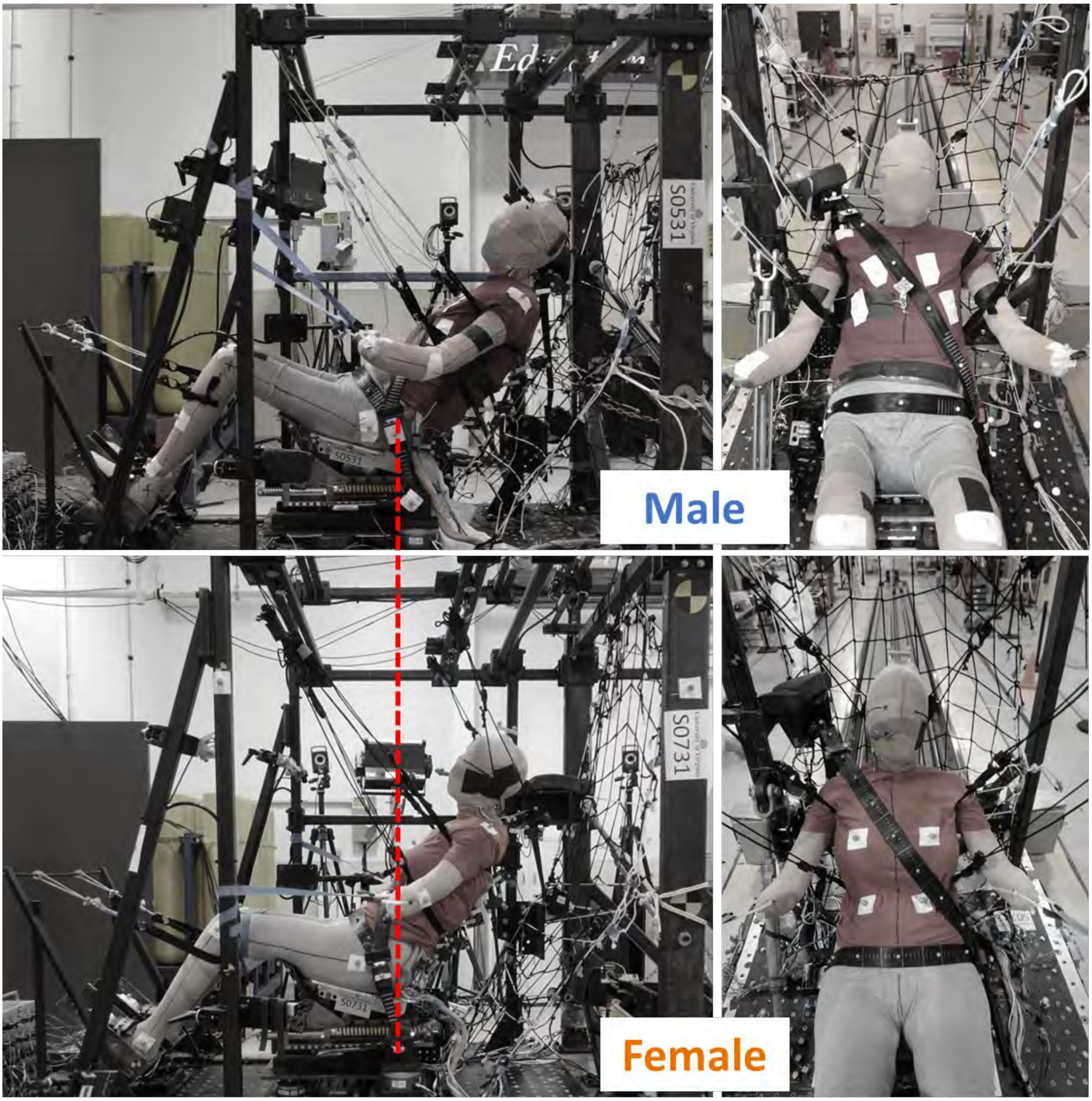

Fig. A1. Photos of test setup from male tests (top) and female tests (bottom). Dotted red line outlines the forward shift in H-point target and restraint anchor points for the female tests.

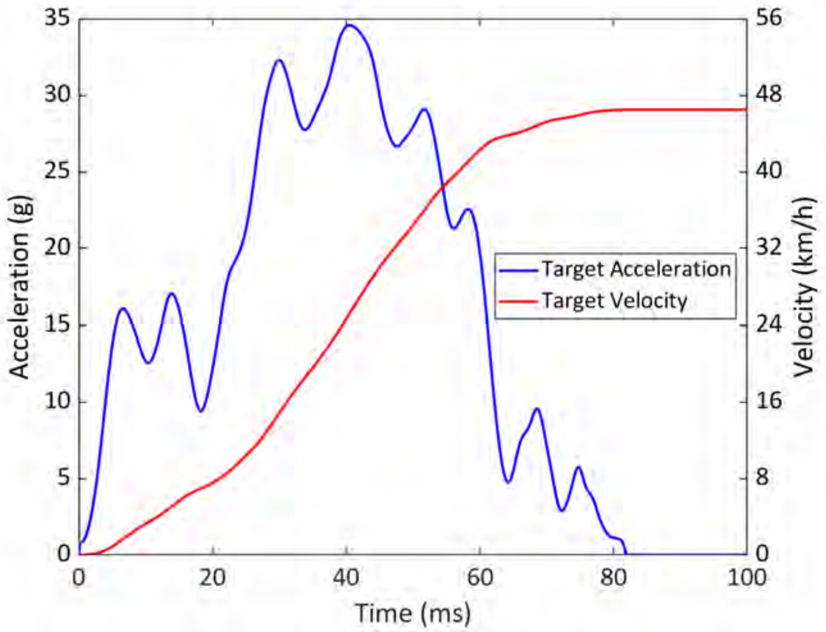

Fig. A2. Target acceleration and velocity of sled system used in this test series.

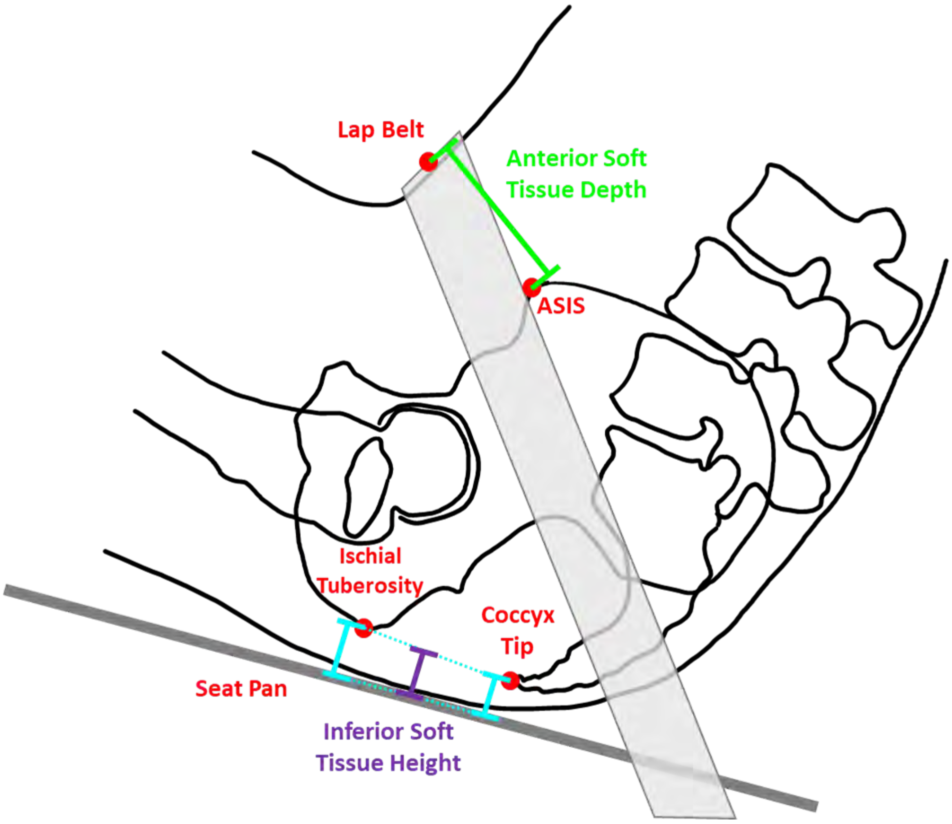

Fig. A3. Definitions for pelvis anterior soft tissue depth and inferior soft tissue height.

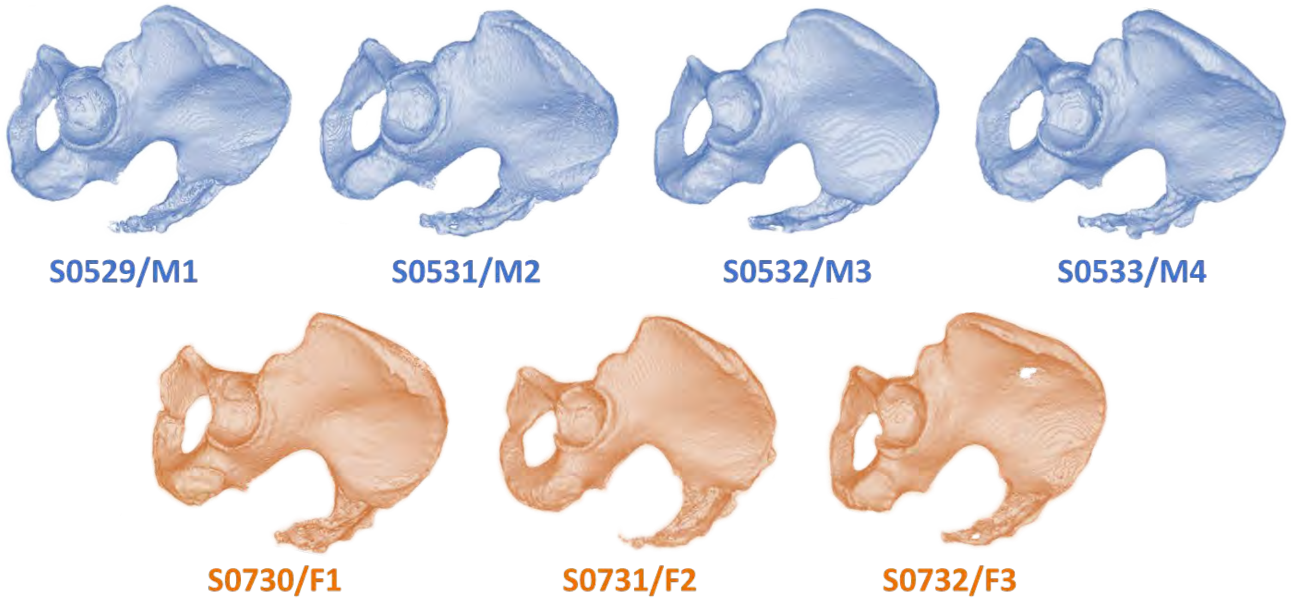

Fig. A4. Sex-based difference in pelvis geometry in pre-test position; the ischial spine is further forward to the tip of the coccyx in the females, while the ischial spine is further rearward to the tip of the coccyx in the males.

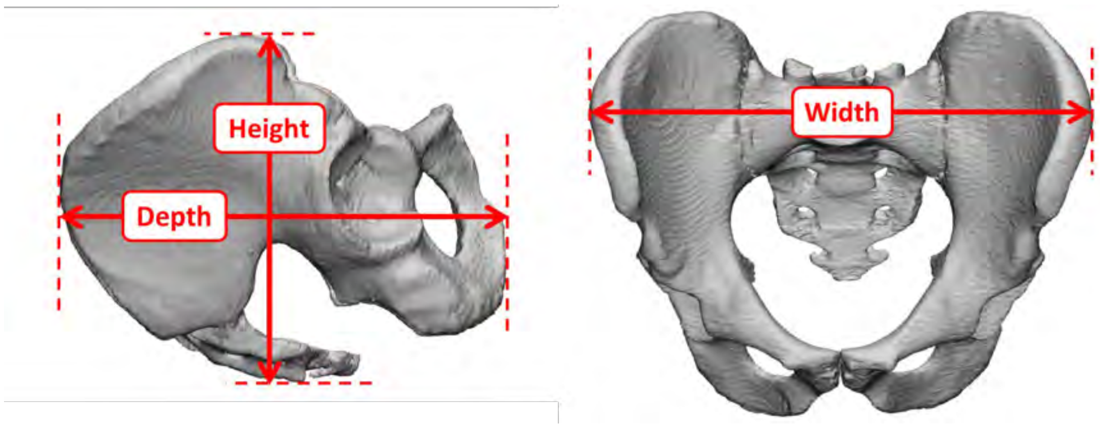

Fig. A5. Definitions of pelvis depth, height, and width. Depth and height were determined from the sagittal view of the pelvis in its pre-test position. Width was determined from the axial view of the pelvis in its pre-test position.

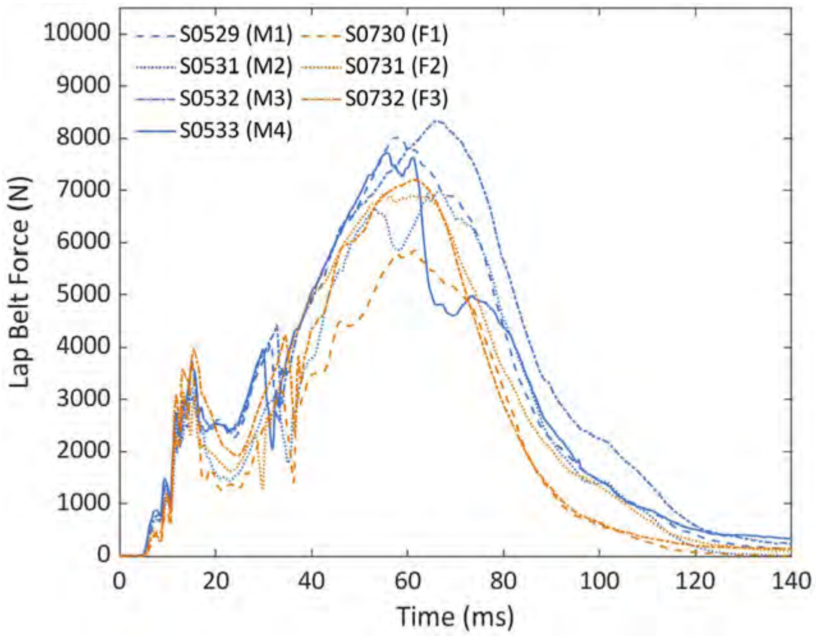

Fig. A6. Lap belt force time-history (CFC 600).

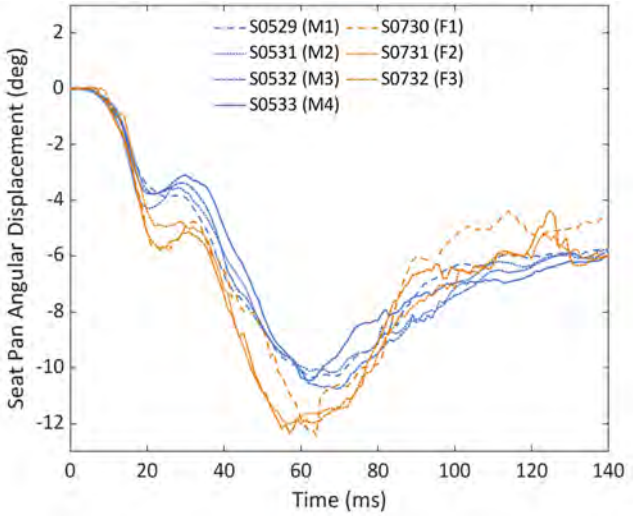

Fig. A7. Seat pan angular displacement time-history.

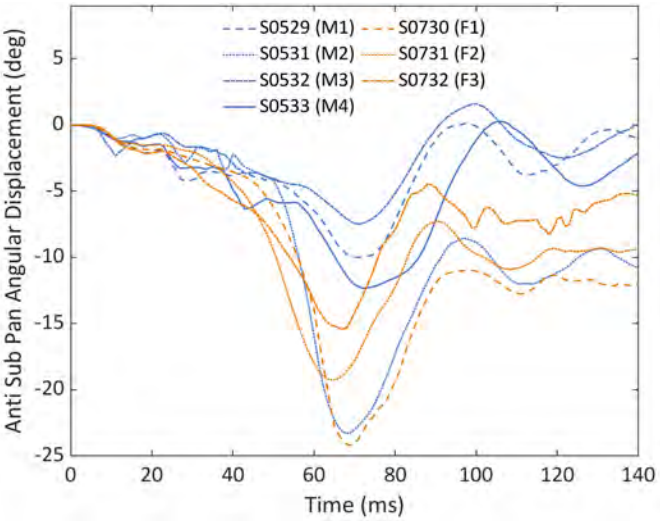

Fig. A8. Anti-submarining pan angular displacement time-history.