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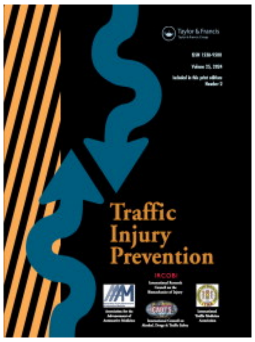

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Assessing injury risks of reclined occupants in a frontal crash preceded by braking with varied seatbelt designs using the SAFER Human Body Model

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ABSTRACT

Objective: This study investigated the effects of different seatbelt geometries and load-limiting levels on the kinematics and injury risks of a reclined occupant during a whole-sequence frontal crash scenario, using simulations with the Active SAFER Human Body Model (Active SHBM).
Methods: The Active SHBM was positioned in a reclined position (50°) on a semi-rigid seat model. A whole-sequence frontal crash scenario, an 11 m/s^2 Automated Emergency Braking (AEB) phase followed by a frontal crash at 50 km/h, was simulated. The seatbelt geometry was varied using either a B-pillar-integrated (BPI) or Belt-in-seat (BIS) design. The shoulder belt load-limiting level of the BPI seatbelt was also varied to achieve either similar shoulder belt forces (BPI_Lower_LL) or comparable upper body displacements (BPI_Higher_LL) to the BIS seatbelt. Kinematics of different body regions and seatbelt forces were compared. The risks of sustaining a mild traumatic brain injury (mTBI), two or more fractured ribs (NFR2+), and lumbar spine vertebral fractures were also compared.
Results: During the pre-crash phase, head, first thoracic vertebra, and first lumbar vertebra displacements were greater with the BPI seatbelt than with the BIS, mainly due to the lack of initial contact between the torso and the seatbelt. Pelvis pre-crash displacements, however, remained consistent across seatbelt types. In the in-crash phase, variations in shoulder belt forces were directly influenced by the different load-limiting levels of the shoulder belt. The mTBI (around 20%) and NFR2+ (around 70–100%) risks were amplified with BPI seatbelts, especially at higher load-limiting force. However, the BPI design demonstrated reduced lumbar spine fracture risks (from 30% to 1%).
Conclusions: The BIS seatbelt appears promising, as seen with the reduced mTBI and NFR2+ risks, for ensuring the protection of reclined occupants in frontal crashes. However, additional solutions, such as lap belt load limiting, should be considered to reduce lumbar spine loading.

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Human body model; reclined occupant; safety; automated vehicles; out-of-position

Introduction

Automated Vehicles (AVs) are anticipated to promote shifts in traditional occupant seating positions. This necessitates innovations in vehicle interiors to accommodate preferences for flexible seating configurations (Matsushita et al. 2019). One notable trend is the adoption of reclined seating postures, allowing occupants to engage in diverse activities, from relaxation to work (Östling and Larsson 2019). However, these seating configurations pose new challenges for occupant safety.

Traditionally, restraint systems such as seatbelts have primarily been evaluated for upright seating positions. Prevailing consumer testing requirements and regulations similarly focus on upright seating positions. In reclined seating postures, the biomechanical responses during crashes vary considerably. The rearward rotation of the pelvis in reclined seating postures increases the risk of submarining, where the anterior posterior iliac spine (ASIS) slides under the lap belt (Dissanaike et al. 2008; Leung et al. 1982). Such a loading scenario may result in abdominal organ or lumbar spine injuries (Poplin et al. 2015). As the distance between the occupant and the instrument panel may increase in AVs, conventional submarining countermeasures relying on knee interaction to restrain the pelvis may not be possible.

Interestingly, no substantial risk of submarining has been observed in many recent reclined Postmortem Human Subjects (PMHS) studies (Baudrit et al. 2023; Richardson et al. 2019; Shin et al. 2023; Somasundaram et al. 2023a, 2023b; Umale et al. 2022; Yoganandan et al. 2023). The absence of submarining in some of these studies (Baudrit et al. 2023; Richardson et al. 2019; Shin et al. 2023) can be attributed to the use of double lap belt pretensioning which intends to improve belt-to-pelvis coupling (Håland and

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Nilson 1991; Page et al. 2012; Richard et al. 2015; Richardson et al. 2019). Further influential factors for submarining, such as reclined angle, lap belt angle, and center of gravity of the pelvis (Boyle et al. 2019; Górniak et al. 2022; Grébonval et al. 2021) have not been fully explored in PMHS testing.

Moreover, the use of a B-pillar integrated seatbelt (BPI) in a reclined orientation poses another set of challenges as it is optimized for upright postures. Specifically, the torso may go out-of-contact with the seatbelt, thus introducing belt slack and reducing seatbelt effectiveness (Forman et al. 2019). It also leads to increased upper body frontal displacements (Izumiyama et al. 2022). As an alternative, seat-integrated seatbelts or Belt-in-seat seatbelts (BIS) are designed to maintain consistent contact with the occupant, regardless of changes in seatback orientation (Matsushita et al. 2019). Previous research indicates that BIS seatbelts in reclined seating postures result in earlier torso engagement and decreased head movement compared to BPI seatbelts (Forman et al. 2019). Any detailed injury metrics were, however, not studied (Forman et al. 2019). Given that the BPI seatbelt remains the prevalent standard seatbelt design in modern vehicles, a detailed analysis comparing the effects of the BPI and BIS seatbelts on the injury risks in reclined crash scenarios is needed.

Adding to the complexity, pre-crash interventions like Automated Emergency Braking (AEB) can alter the occupant posture and kinematics going into a crash, if a crash is not avoided (Schoeneburg et al. 2011). About half of the crashes in the real world are preceded by evasive actions such as braking and steering (Ejima et al. 2009; Scanlon et al. 2015; Stockman 2016). The muscular response and kinematics of occupants in complicated whole-sequence crashes, that is a crash preceded by pre-crash maneuvers, can be simulated using Active Human Body Models (Matsuda et al. 2018; Östh et al. 2022; Saito et al. 2016). In addition, they can be used to predict injuries during a crash. The Active SAFER HBM (SHBM) has been validated to predict occupant kinematics and muscular response during pre-crash maneuvers for an upright posture (Larsson et al. 2019; Ólafsdóttir et al. 2019; Östh et al. 2015). It has also been used to simulate whole-sequence crashes with a reclined posture (Östh et al. 2020).

Given this context, this study aimed to quantify the effects of BPI and BIS seatbelts on the injury risks of reclined occupants in a whole-sequence crash scenario. The seatbelt forces and occupant kinematics measured were compared to explain the differences in injury risks.

Methods

Finite element (FE) simulations were performed using the explicit solver LS-DYNA double-precision version R9.3.1 (LSTC, Livermore, CA, USA). The Active SHBM version 10.0 (Pipkorn et al. 2023) was used to represent a reclined passenger occupant sitting away from the instrument panel in an AV. The passive SHBM has been validated to predict the kinematics of a reclined seated occupant in a frontal crash (Gepner et al. 2022; Mroz et al. 2020b). The SHBM, developed to represent the anthropometry of mid-size male dummies (stature 175.3 cm and weight 77.3 kg) as defined by Schneider et al. (Schneider et al. 1983), measures 175 cm in height and 77 kg in weight. It employs 1D Hill-type elements to model the active muscles of the cervical and lumbar regions, as well as the upper arms, following a closed-loop control system (Larsson et al. 2019). To maintain the occupant posture, it uses an angular position feedback control system. The Active SHBM has also been used for whole-sequence pre-crash and crash simulations previously for both upright and reclined occupants (Mishra et al. 2022; Östh et al. 2022; Östh et al. 2020).

The simulation setup was based on previously published PMHS studies and subsequent simulation work (Gepner et al. 2022; Mroz et al. 2020b; Richardson et al. 2019). A simplified semi-rigid seat model was included comprising two adjustable plates: the seat pan and the submarining pan. The seat configurations, such as the pan angles and spring stiffnesses, were the same as those used by Mroz et al. (Mroz et al. 2020b), configured to represent a standard vehicle seat. Additionally, generic models for a rigid headrest and backrest were integrated considering the pre-crash phase, as recommended in the OSCCAR project (Klein et al. 2019).

A separate positioning simulation of 400 ms was run to position the SHBM at a 50° recline relative to the vertical axis using the Marionette method (Poulard et al. 2015), as shown in Figure 1. Adjustments were made to the pelvis and torso angles to align them with the PMHS data. The pelvis angle was measured to be 77.5° (target 75.2°) between the ASIS and pubic symphysis to the vertical in the sagittal plane. Subsequently, the torso angle was modified until the origins of the first (T1), eighth (T8), and eleventh (T11) thoracic vertebrae and first (L1) and third (L3) lumbar

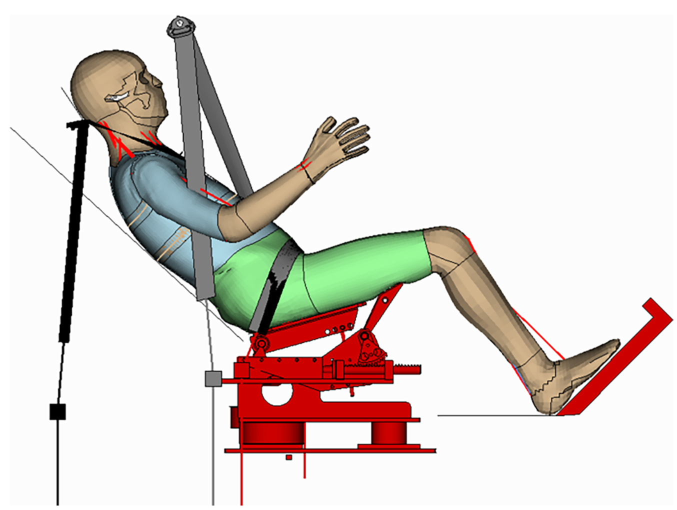

Figure 1. The Active SHBM seated in a reclined posture, 50° from the vertical axis, on the semi-rigid seat model with two different seatbelt geometries: BIS (black belt textile) and BPI (gray belt textile). Note: the BIS and BPI simulations were run separately. They are shown in the same figure only for comparison. For the BPI seatbelt geometry, the right arm of the HBM is outboard of the vertical portion of the 2D seatbelt. As this is not reasonable, no contact was defined between the 2D seatbelt and the arm to avoid problems with the initial configuration.

vertebrae approximately matched (within 20 mm in x and 30 mm in z directions, respectively) the positioning targets obtained from the PMHS tests (Gepner et al. 2022; Mroz et al. 2020b). The cervical spine and head positions were then adjusted to match the head origin and angle target (head origin within 11 mm in x-direction and 6 mm in z-direction, head y-angle within 0.1°).

The restraint system model used comprised a 3-point seatbelt with specific features: a shoulder belt retractor pretensioner, dual lap belt pretensioners designed to avoid submarining, and a crash locking tongue to mitigate webbing transfer from the shoulder belt to the lap belt. Separate simulations were run based on the D-ring placements: either on the seatback (BIS seatbelt geometry) or on the B-pillar (BPI seatbelt geometry). Prior studies informed the D-ring location for BIS (Gepner et al. 2022; Mroz et al. 2020b) and BPI (Östling et al. 2017). For the BIS simulation, a shoulder belt load limiter (LL) of 3.5 kN was used. In contrast, for the BPI seatbelt geometry, simulations considered a lower LL (approximately 3 kN). This LL ensured a shoulder belt force comparable to the BIS simulation as the force in the seatbelt webbing depends on the wrapping angle around the D-ring. An additional BPI simulation employed a higher LL (approximately 8 kN) to achieve occupant upper body displacements comparable to those in BIS simulation, accounting for the greater frontal displacements with BPI owing to delayed HBM to seatbelt contact. A combination of 1-dimensional (1D) and 2-dimensional (2D) elements was used to model the seatbelt. Specifically, the 1D elements connected the seatbelt to the retractor and the pretensioners, whereas the 2D seatbelt, 48 mm with six elements across its width, was used for the webbing. A *MAT_SEATBELT material model with a defined force-strain relationship giving 9% elongation at a tensile force of approximately 10 kN was used. The slipring friction coefficient at the D-ring was 0.18. At the buckle, a friction coefficient of 0.13 was used until 40 ms into the crash, and then it was increased to 0.35 to model the crash locking tongue. The seatbelt system model (with the corresponding parameters) has been used in previous studies to simulate reclined HBMs in crash-only simulations (Gepner et al. 2022; Mroz et al. 2020b). The belt routing and anchorage points were defined on the basis of the 3D position measurements from previous PMHS tests (Richardson et al. 2019).

A whole-sequence crash scenario spanning 1360 ms, and including AEB followed by a full-frontal impact, was simulated. It consisted of a 300 ms model initialization phase, specifically for the Active SHBM, to initialize muscle activations. It was followed by an 11 m/s^2 pre-crash AEB phase lasting 900 ms. Maximum braking was achieved after a 200 ms delay and 200 ms ramp-up time (Mishra et al. 2023). The maximum braking level used was comparable to the values reported from analysis of consumer testing in the US (Mahdinia et al. 2022). The final phase of the whole-sequence scenario was a full-frontal crash at 50 km/h (resulting in a peak deceleration of 35 g) for 160 ms. The crash pulse used was the same as that used in previous PMHS tests (Richardson et al. 2019). The entire scenario is illustrated in Figure A1 in the Appendix. Importantly, the muscle activation levels achieved at the end of the pre-crash phase were retained and remained constant during the crash phase (Östh et al. 2022).

Frontal nodal displacement-time histories of the head center of gravity, T1 and L1 vertebrae origins, and the pelvis (posterior superior iliac spine) were extracted relative to the sled motion and compared. Pelvis rotational displacements (y-rotation) were also extracted and compared. The time histories of the magnitude of the shoulder belt force (measured at a cross-section between the D-ring and the shoulder), buckle force (measured on the buckle attachment), and lap belt force (measured at a cross-section near the end of the 2D lap belt) were also compared. The effects of the different seatbelt geometries and load limiters on potential injury risks to the head, thorax, and lumbar spine were evaluated. To this end, risks for mild traumatic brain injury (mTBI), two or more fractured ribs (NFR2+) in a 65-year-old male, and lumbar spine vertebra fracture (L1 fracture risk) in a 65-year-old male were assessed. The SHBM has been validated to predict these risks—mTBI based on the maximal principal strain in the brain tissue (Fahlstedt et al. 2022), NFR2+ based on the peak first principal strains in the cortical bone of each rib (Larsson et al. 2021), and the lumbar spine vertebra fracture risk based on the combined compression-flexion loading of the lumbar vertebrae (Tushak et al. 2023). For the lumbar spine, only the L1 fracture risk was considered because the L1 vertebra demonstrated the highest loading in all simulations. Time histories of the L1 vertebra loads as well as the SHBM to seat pan contact force were also compared.

Results

Figure 2 (a–c) compares the shoulder belt, buckle, and lap belt forces from all three simulations during the crash phase (the forces across the entire timeline, including pre-crash braking, are available in the Appendix Figure A2 (a–c)). As intended, the BPI_Lower_LL simulation produced a shoulder belt force comparable to that of the BIS simulation. Conversely, the shoulder belt force measured in the BPI_Higher_LL simulation exceeded its counterpart in the BIS simulation, a result of the intention to equate the upper body displacement with BIS. Correspondingly, the buckle force measured followed the same trend. The lap belt force of the BIS seatbelt geometry, however, surpassed that of the BPI by approximately 1 kN.

Figure 3 (a–d) compares the head, T1, L1, and pelvis frontal displacements across all three simulations along the entire whole-sequence scenario. Tracing the sequence from the onset shows negligible SHBM frontal movement until 500 ms, when the AEB ramp-up started. Between 500 and 1200 ms, SHBM responses due to the AEB pulse were observed. The head displacement with BPI exceeded that of BIS by approximately 170 mm, moving approximately 130 mm with BIS compared to 300 mm with BPI. Peak T1 displacements during the AEB phase were recorded at 90 mm for BIS and 230 mm for BPI. L1 displacements were

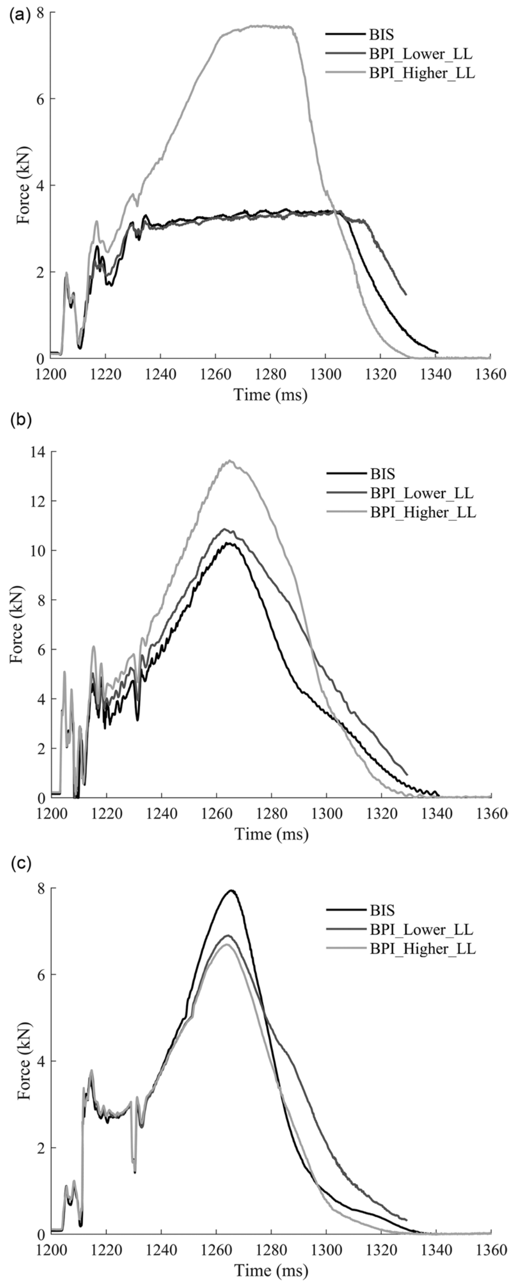

Figure 2. Seatbelt forces from the (a) shoulder belt, (b) buckle attachment, and (c) lap belt only for the crash phase of the simulation.

recorded at 15 mm for BIS and 50 mm for BPI. Pre-crash pelvis displacements showed (Figure 3(d)) minimal variance across seatbelt geometries, although notable differences in pelvis y-rotations were evident (Figure 4). BIS induced a rearward rotation of approximately 3.6°, while BPI resulted in a forward rotation nearing 6° during the pre-crash phase. A comparison of the postures of the Active SHBM at the start of the crash phase, at maximum L1 compression, and at peak head excursions for the three different seatbelts is shown in Appendix Figure A3 (a–i).

In the crash phase, peak upper body (head and T1) displacements appeared similar between BIS and BPI_Higher_LL, as intended, while they were approximately 160 mm greater with BPI_Lower_LL. L1 peak displacement was lowest with BIS (154 mm). It increased by 50 mm with BPI_Higher_LL, and 115 mm with BPI_Lower_LL. Pelvis displacements under both BPI scenarios remained nearly identical but were greater by 40 mm with BIS. Pelvis y-rotations during the crash phase initially depicted a rearward trend (around 3–4°), transitioning to a forward rotation during the rebound phase. The most pronounced rebound rotation (about 62°) was with BPI_Lower_LL, while BPI_Higher_LL and BIS had rotations of 30° and 40°, respectively. No submarining was observed in any simulation.

The mTBI risk was very low (1%) with BIS. Switching to BPI increased this to 21%, irrespective of high or low load limiting. BIS simulations predicted no NFR2+ risk, whereas the BPI_Lower_LL simulation estimated the risk at 73% (Figure 5). Elevating the load limiting with BPI_Higher_LL pushed it to 100%. L1 fracture risks associated with BIS were calculated at 30%, correlating with a compressive force of 3.4 kN and a flexion moment of 90 Nm on the L1 vertebra. With the BPI design, it decreased to a mere 1%. Comparing the time histories of the loads on the L1 vertebra, as shown in Appendix Figure A4 (a, b), a consistent pattern was noted: the peak compression force and the peak flexion moment in all three cases occurred around the same time frame, approximately 45–60 ms into the crash. This timing coincided with when the lumbar spine-pelvis region of the SHBM was loading the seat pan (Appendix Figure A4 (c)).

Discussion

The current study used the capabilities of FE simulations with the Active SHBM to investigate the implications of different seatbelt geometries and load-limiting levels on the kinematics and injury risks of a reclined occupant in an AV during a whole-sequence crash scenario.

Comparing the different seatbelt designs in the pre-crash phase, head, T1, and L1 had greater displacements with the BPI seatbelt compared to the BIS. This was attributed to the initial lack of contact between the torso of the SHBM and the diagonal seatbelt in the BPI design, permitting the SHBM upper body to move unrestricted until contact was established. Although pelvis pre-crash displacements were consistent across the seatbelt designs due to the unchanged lap belt design and routing, pelvis y-rotation varied notably during the pre-crash phase. The BIS resulted in a slightly (3.6°) rearward rotation, whereas the BPI caused a forward rotation (6°), likely due to the more pronounced frontal displacement and forward rotation of the torso with the BPI as the pelvis is attached to the upper body via the spine. Importantly, the pre-crash displacements were consistent between BPI load-limiting levels because the load limiter only works during the in-crash phase.

For the in-crash phase, we intentionally set the load-limiting level with the BPI seatbelt to achieve either a similar shoulder belt force (BPI_Lower_LL) or upper body displacement (BPI_Higher_LL) as observed with the BIS seatbelt. The resulting and expected variations in the shoulder belt forces affected other biomechanical outcomes. BPI_Lower_LL resulted in greater displacements for the head, T1, and L1 than both BIS and BPI_Higher_LL.

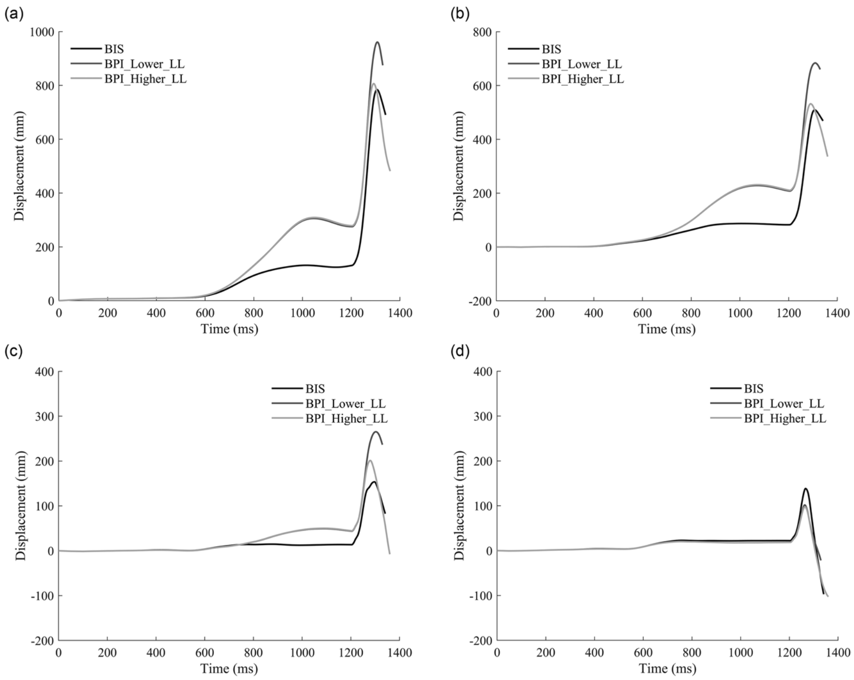

Figure 3. Frontal displacements of the (a) head, (b) T1, (c) L1, and (d) pelvis across the entire timeline of the whole-sequence frontal crash.

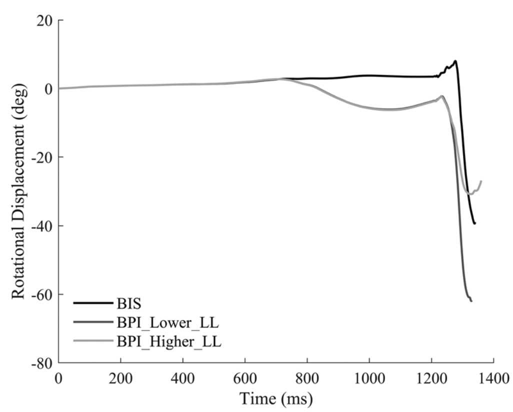

Figure 4. The pelvis y-rotations across the entire timeline of the whole-sequence frontal crash.

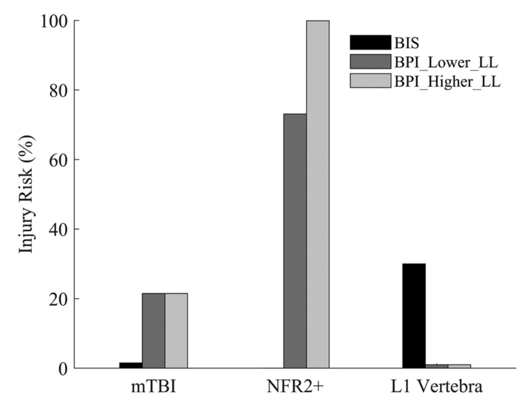

Figure 5. Comparison of mTBI, NFR2+, and L1 vertebra fracture risks.

This underlines the importance of carefully selecting load-limiting values, especially when considering alternative seatbelt geometries.

Injury risk assessments revealed substantially increased risks for mTBI and NFR2+ with the BPI seatbelt, particularly at the higher load-limiting level. The delayed engagement of the SHBM torso with the BPI shoulder belt contributed to increased head forward excursion and rotational acceleration, elevating mTBI risks. With regard to NFR2+ risks, the BIS seatbelt predicted no risk of rib fractures, a result consistent with previous SHBM v10 simulation findings with the same setup (Mroz et al. 2023). While better belt engagement in terms of earlier coupling with the BIS might naturally lower rib fracture risks, predicting zero risk for a 65-year-old may also reflect model limitations. Comparison with previous recline PMHS tests (Richardson et al. 2020) highlights an under-prediction of rib fracture risks in older individuals by the SHBM, which is a known limitation of the SHBM (Larsson 2023). On the other hand, the NFR2+ risk increased substantially for BPI. This was likely due to the different loading patterns of the seatbelt on the chest. With BIS, the belt tended to shift toward the neck and hence loaded the chest differently compared to BPI. Consequently, the strains measured in the ribs were higher in the BPI configuration, particularly on the right side, resulting in higher NFR2+ risk. When load limiting was increased with BPI, the NFR2+ risk further increased due to higher forces on the chest.

Interestingly, while BPI elevated mTBI and NFR2+ risks, it substantially reduced lumbar spine loading and thus the vertebral (L1) fracture risk. Consistent with PMHS test findings (Richardson et al. 2020), L1 bore the maximum load of all lumbar vertebrae in all cases. Previous studies have indicated that a reclined posture combined with a BIS seatbelt can exacerbate lumbar spine loading (Boyle et al. 2019; Forman et al. 2019). When the occupant is in a front-facing reclined position in a frontal crash, the lumbar spine is in the direction of the loading leading to more compression of the lumbar spine. In addition, with the BIS seatbelt, as shown by Forman et al. (Forman et al. 2019), earlier belt engagement coupled with a more horizontal shoulder belt angle results in decreased forward rotation of the upper body, further increasing the lumbar spine loading. The lumbar spine loading increased in proportion to the SHBM loading the seat pan. With the BIS design, the contact force between the SHBM and the seat pan was approximately 4 kN higher, leading to substantially increased lumbar spine loading (about 3.4 kN in compression and 90 Nm in flexion) resulting in a 30% risk of L1 fracture. Conversely, in the BPI designs, the reduced SHBM loading on the seat pan led to lower lumbar spine loading (about 1.2 kN in compression and 45 Nm in flexion), translating to a negligible risk of L1 fracture. Additionally, a relatively high tensile load of approximately 2.6 kN was observed in the L1 vertebra with the BPI_Higher_LL design during the rebound phase of the crash. However, this was not factored into the fracture prediction, as the SHBM's lumbar spine fracture prediction is based solely on compression-flexion loading (Tushak et al. 2023).

Addressing the influence of the pre-crash maneuver and kinematics on the crash response, one might expect substantial out-of-position of the upper body. However, the BIS belt restrained the upper body effectively. In contrast, the BPI belt permitted substantial upper body movement until it engaged with the belt, resulting in a head displacement of about 300 mm and the T1 displacement of about 230 mm solely during the pre-crash phase. Such pronounced displacements might account for the elevated injury risks to the brain and the ribs associated with the BPI seatbelt. It is crucial to note that no injury risk was predicted solely during the pre-crash phase itself, but the pre-crash HBM response does influence the subsequent crash response.

These findings illustrate that the biomechanical effects of different seatbelt configurations are multifaceted and may not universally lead to safer or riskier outcomes across all injury metrics. Although a BIS seatbelt may help solve some restraint challenges posed by reclined seating postures, lumbar spine loading may remain a concern. An additional restraint design may be needed to reduce the loading on the lumbar spine. There could be different solutions for different interiors. For roomier interiors (as expected in future AVs), it has been shown previously that adding load limiting to the lap belt can considerably reduce the risk of lumbar spine vertebrae fracture (Östling and Lubbe 2022). Thus, a BIS seatbelt geometry combined with lap belt load limiting could enhance safety for reclined occupants in frontal crashes. Such a solution was also shown to be beneficial for upright seating postures (Östling and Lubbe 2022). For traditional interiors, similar to those in current standard passenger cars, the role of the frontal airbag becomes an important consideration. Given that frontal airbags are primarily evaluated for protecting occupants in upright positions, their effectiveness in restraining the head of a reclined occupant could be less optimal. The expected interaction between the occupant and the airbag in reclined seating postures would likely differ, primarily due to delayed coupling. New airbag designs, such as the dual shoulder airbag integrated within the seat, might be essential. These airbags could engage the upper body earlier, mitigating the kinetic energy before the head contacts the airbag, thereby enhancing safety in reclined seating positions (Matsushita et al. 2019).

Moving beyond purely biomechanical perspectives, it is also important to discuss user preferences. Some might feel safer sitting reclined with a BIS seatbelt compared to a BPI seatbelt, given that the BIS aligns with both the seat and occupant's posture. While it is predicted that occupants prefer to be seated in a reclined seating posture in AVs, there are limited user insights on it, particularly regarding perceived safety. Future work should delve deeper into user insights regarding various occupant postures and restraint systems.

This study has several limitations. The simulations, although advanced, represent specific scenarios, and real-world crashes can present several variables unaccounted for in this study. The BIS geometry used does not represent an actual BIS as it was not integrated into the seatback, which was also missing as a generic seat model was used. It is possible that the results in a production vehicle seat with a real seatback integrated D-ring could be different. The seat position and the BPI D-ring anchorage location were also not changed. It is possible that the relative position of the occupant and the BPI D-ring anchorage may affect the results. However, it is expected that the fundamental differences between BIS and BPI seatbelts for reclined occupants will also persist for production vehicle seats, as shown by Forman et al. (2019). Moreover, consistent with the PMHS test setup (Richardson et al. 2020), the current simulations included a footrest, although the seat position was described as away from the instrument panel. Previous research employing the same model setup showed only minor differences in occupant kinematics and loadings with or without footrest (Mroz et al. 2020a). However, the absence of a footrest can influence submarining in some cases (Östh et al. 2020), depending on the seat model used. It should also be noted that the current evaluation was confined to scenarios involving braking followed by a frontal crash using a single pulse. The results could differ in situations such as intersection crashes, which involve oblique and far side kinematics and are expected to be common in AVs (Mroz et al. 2022). Prior research has indicated increased chest loadings and a higher risk of belt slippage in such intersection crashes (Mroz et al. 2022).

Regarding the occupant model, the Active SHBM has been validated for predicting occupant kinematics of upright occupants during braking. Its prediction for reclined occupants is an extrapolation of its validation for upright occupants. Detailed reclined volunteer kinematic and muscle activation data during braking are needed to further validate the Active SHBM. If the same study were conducted using the Passive SHBM, greater frontal displacements, especially during the pre-crash phase, might be anticipated. This is because the active muscle control in the Active SHBM tries to maintain the initial position of the model (reclined in this case), a response not mirrored by the Passive SHBM. A previous study comparing the Passive and Active SHBM in combined braking and crash scenarios indicated lower frontal displacements and injury risks with the Active SHBM, primarily due to the muscle activations generating muscle forces (Östh et al. 2020). However, the specific effect of muscle activations on injury risks remains to be validated. Additionally, while the SHBM has been validated for predicting the kinematics of reclined occupants in frontal crashes, its injury prediction validation for reclined occupants remains unestablished. Two of the three injury predictions used in this study, mTBI and NFR2+, are based on tissue-based criteria such as strains which are supposed to be omnidirectional. The current study assumes a direct correlation of the injury prediction values obtained from the SHBM with the injuries sustained in reclined PMHS studies; however, the model has not been specifically validated to predict injuries under the same injury inducing conditions. Nevertheless, the insights obtained from this study offer a valuable basis for future investigations. Moreover, the current study accounted for only one type of occupant anthropometry, posture, shape, muscle control, and muscle activation timing. The positioning targets for the reclined seating posture were also based on PMHS studies and not actual vehicle occupants. It is possible that the specific body angles may not be representative of real-life occupant postures. There could also be large variance in occupant reclined seating postures and body shapes, affecting the interaction with the restraint systems, which was not considered in this study. Future studies should evaluate how varying these factors might influence the results. A more technical detail, the BIS and BPI_Lower_LL simulations terminated 20 and 30 ms before the planned termination time, respectively. Nevertheless, given that the peak frontal displacements and maximum shoulder belt forces were already realized by that time, it was deemed acceptable to use the results for subsequent analyses.

In conclusion, as AVs continue to evolve, pre-crash maneuvers such as AEB become common, and occupant postures diversify, it becomes vital to critically evaluate and adapt restraint systems. This study elucidates the differences in injury risks based on the seatbelt design, with specific emphasis on the D-ring location and shoulder belt load limiting, for reclined occupants. The BIS seatbelt facilitates improved torso engagement, effectively restraining the upper body during pre-crash braking and minimizing the mTBI and NFR2+ risks during the crash. However, it also increases the lumbar spine loading. Incorporating solutions like lap belt load limiting alongside a BIS seatbelt is likely to offer enhanced protection to reclined occupants during a whole-sequence frontal crash.

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Disclosure statement

The authors work at Autoliv Research, located in Vårgårda, Sweden. Autoliv Research is part of Autoliv (https://www.autoliv.com), a company that develops, manufactures, and sells for example protective safety systems to car manufacturers. Results from this study may impact how Autoliv choose to develop their products.

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