

Research Paper

Influence of Driver-Steering Wheel Distance on Multi-Body-Region Injury Outcomes in Frontal Crashes Using Finite Element Analysis

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Abstract

The objective of this paper was to investigate the influence of driver seating distance on injury outcome. The study employed controlled finite element analysis using a virtual dummy and a mid-sized sedan to examine the isolated effect of driver seating distance on injury severity. Three simulations were conducted by varying the seating distance from the closest position to the farthest position (corresponding horizontal positions from chest to steering distances 308 mm, 358 mm, and 408 mm, respectively), while maintaining an identical frontal crash into a rigid wall. Injury severity was analyzed using established biomechanical injury criteria, including Head Injury Criterion (HIC₁₅) for head trauma, maximum rib deflection ($R_{\max}$) and Principal Component score for chest injury, anterior superior iliac spine (ASIS) and acetabulum forces for pelvic injury, and Tibia Index for lower extremity injury. The simulation results revealed that the driver's seating distance influenced the selected body regions in a distinct pattern. Contrary to the conventional assumption, the intermediate position produced the highest head injury risk (HIC₁₅ = 824.5), compared to the closed position (HIC₁₅ = 693.4) and the farthest position (HIC₁₅ = 749.9). In the chest, the closest position ($R_{\max}$ = 87.2mm) contributed 6.6% higher rib deflection than the farthest position ($R_{\max}$ = 81.8mm). Conversely, pelvic loading (ASIS force) increased by 9.1% with increased seating distance (from the closest to the farthest position). Lower extremity injury risk was highest in the closest position due to early knee-dashboard contact. These results indicate that the distance between the driver and the steering wheel affects the injury risk to the occupant in a distinct manner rather than uniformly increasing or decreasing overall injury severity. The optimal seating position varies by body region due to differences in restraint system interaction timing, load distribution patterns, and contact mechanics, emphasizing the need for position-specific injury mitigation strategies.

Keywords: Frontal crash simulation; Occupant protection; Seating position; Injury biomechanics; THOR dummy

1. Introduction

Despite advanced safety systems, including multistage airbag configurations, collision avoidance systems, and autonomous emergency braking systems, road traffic accidents remain a major concern worldwide [1], [2]. Records show that each year, road crashes result in about 1.3 million deaths and 20 to 50 million non-fatal

injuries worldwide [3] due to vehicle collisions. A previous investigation revealed that 10% of drivers experienced an Abbreviated Injury Scale (AIS) 3+ injury severity at a velocity of 15.5 m/s (56 km/h) in a head-on crash [4]. These statistics indicate the need for further research on the safety system. However, developing effective interventions requires understanding how



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specific parameters interact to influence the severity of injuries.

The severity of crash-related injuries depends on multiple interacting factors, including the type of collision, the individual characteristics of the occupant, the vehicle's structural design, and the occupant's seating position [5]. Additionally, dynamic physiological parameters such as muscle bracing, instantaneous posture, and anthropometry significantly influence occupant kinematics and their interaction with vehicle restraint systems during a collision [6]. Among these factors, occupant anthropometry has received particular attention in literature and provided important baseline understanding.

Extensive research has demonstrated that occupant anthropometry has a significant influence on injury patterns across various body regions. For instance, drivers with a height of around 160-165 cm were more vulnerable to head, face, and neck injuries due to their shorter seating distance from the steering wheel [7]. Similarly, the study in [8] indicated that height, gender, and age of the occupant influence the lower extremity injury pattern. While anthropometric characteristics establish baseline sensitivity to injury, occupant positioning during impact represents an equally important and more controllable factor in reducing injury severity.

Beyond anthropometric factors, previous research has revealed critical relationships between occupant positioning and injury mechanisms across multiple body regions. For example, Ye *et al.* [9] analyzed fracture mechanisms and parameters affecting thoracolumbar spine responses, using reconstructed data from 11 motor vehicle crashes, and found that seat track position and seat back angle strongly influenced spine response. At the same time, Leledakis *et al.* [5] demonstrated that thoracic injury patterns were affected by spinal posture in frontal crashes. Likewise, seating postures increased the risk of submarining (where the occupant slips under the seatbelt during a crash), resulting in altered occupant kinematics during collisions [10]. Moreover, the driver's hand position during the pre-crash can alter the injury pattern, emphasizing the importance of considering these factors when analyzing the effects of driver-to-steering wheel distance on injury patterns [11]. These positioning effects on

injury patterns are basically mediated through the contact loading mechanisms that occur when occupants interact with vehicle interior components during impact.

Contact loading patterns between occupant and vehicle interior, which fundamentally determine injury outcomes in frontal crashes. The occupant's seating distance influences these loading patterns, as distance affects both the timing and magnitude of contact forces during impact. For instance, Xiao *et al.* [12] found that knee-to-dashboard clearance directly correlates with the severity of lower extremity injuries, suggesting that increased clearance reduces knee injuries through altered contact dynamics. Similarly, research has demonstrated that proximity to airbag deployment zones increases injury risk to the head, neck, thorax, and upper extremities due to high-magnitude forces during initial airbag inflation [13]. These distance-injury relationships across multiple body regions underscore that driver-to-steering wheel distance represents a uniquely critical parameter that governs contact timing, airbag interaction effectiveness, and restraint system performance [14].

Despite the recognized importance of the relationship between driver-to-steering wheel distance and occupant injury outcome, current research approaches have significant methodological limitations that prevent a clear understanding of these mechanisms. Most studies have been conducted based on epidemiological surveys, post-mortem analyses, or real-world crash reconstructions, which fundamentally involve complex combined variables that limit the isolation of specific causal factors. These studies have limited controllability over variables such as crash severity, restraint system configuration, and occupant characteristics simultaneously. Furthermore, current studies, based on simulation, have analyzed the broad effects of seating position, such as reclined position, without isolating the driver-to-steering wheel distance as the primary variable of interest. However, the multiple combined factors of those investigations make it more difficult to initiate a clear relationship between the driver position and injury outcome, which is an important consideration in developing a new safety strategy, representing a notable gap in the existing studies.

Essential research is needed to investigate the isolated influence of driver-to-steering wheel distance on occupant injury patterns, enabling targeted safety interventions and optimized restraint system design.

This study quantifies the isolated influence of driver-to-steering wheel distance on occupant injury risk in frontal crashes, using the Test Device for Human Occupant Restraint (THOR) - 50th percentile male (50M) anthropomorphic test device in standardized crash scenarios. This systematic approach establishes the relationship between the driver's position and injury outcome, varying the distance from the driver to the steering wheel as the sole variable while maintaining identical crash conditions, vehicle configurations, and restraint system parameters. It enables precise assessment of relationships between drivers to steering wheel distance and injury patterns across multiple body regions, providing valuable insights for improving occupant safety system design.

2. Methods

This study was conducted entirely through computational finite element simulation using validated models (no physical crash tests were performed) to systematically investigate the isolated influence of driver-to-steering wheel distance on occupant injury outcomes, while maintaining all variables constant except the primary factor of interest. A flow chart outlines the overall experimental steps and setup, including the four main phases as illustrated in [Figure 1](#). In Phase 1, a validated finite element (FE) model of the THOR-50M anthropomorphic test device and a vehicle model were prepared, with the dummy response benchmarked against National Highway Traffic Safety Administration (NHTSA) data and the vehicle model verified against literature results. The second phase defined the experimental design by varying the driver-to-steering wheel distance at three levels (308 mm, 358 mm, and 408 mm) while maintaining standardized control variables, including a seat back angle, hand position, standard restraint system, and frontal crash pulse. In Phase 3 of the present work involved executing three crash simulations corresponding to the seating positions, from which sensor data such as head acceleration, rib deflection, pelvic (anterior superior iliac spine and acetabular) forces, and tibia/femur forces were extracted. Finally, Phase 4

focused on injury assessment and comparative analysis, where injury criteria including Head Injury Criteria, Peak acceleration, rib deflection (R_{max}), Principal Component (PC) score, anterior superior iliac spine (ASIS) force, acetabular force, and Tibia Index were evaluated, followed by application of injury risk functions and statistical comparison to identify distance-dependent effects on occupant injury patterns.

2.1. Experimental Design Setup

2.1.1. Simulation framework

The simulations were conducted using LS-Dyna, integrated into ANSYS Workbench, a widely used finite element analysis (FEA) software. The Test Device for Human Occupant Restraint (THOR) -50M (50th percentile male) was selected due to its sophisticated bio fidelity and detailed anatomical representation [\[15\]](#). The THOR-50M model was produced at the University of Virginia's Center for Applied Biomechanics and verified through correlation with physical crash-test datasets reported by the National Highway Traffic Safety Administration (NHTSA) [\[16\]](#), [\[17\]](#), demonstrating accurate injury prediction capabilities across multiple body regions including head acceleration, thoracic deflection, pelvic loading, and lower extremity responses. The dummy model includes pre-defined virtual sensor locations that correspond to instrumentation positions in the physical THOR dummy, enabling extraction of injury-relevant data (accelerations, forces, moments, deflections) comparable to experimental crash test measurements. The Toyota Camry 2012 Finite Element (FE) model, developed with 2 million elements through reverse engineering at the Center for Collision Safety and Analysis (CCSA), was used as a vehicle FE model for simulations. To ensure realistic structural deformation patterns and occupant compartment behavior, the model was validated against the literature [\[18\]](#), [\[19\]](#).

2.1.2. Seating configurations

Three seating configurations were set up for the simulations: Simulation 1 (closest position), with a 308 mm chest-to-steering-wheel distance; Simulation 2 (intermediate position), with a 358 mm distance; and Simulation 3 (farthest position), with a 408 mm distance. [Table 1](#) indicated all the measurements of the dummy's position, as illustrated in [Figure 2](#). The distance between the dummy's nose and the center of the steering wheel,

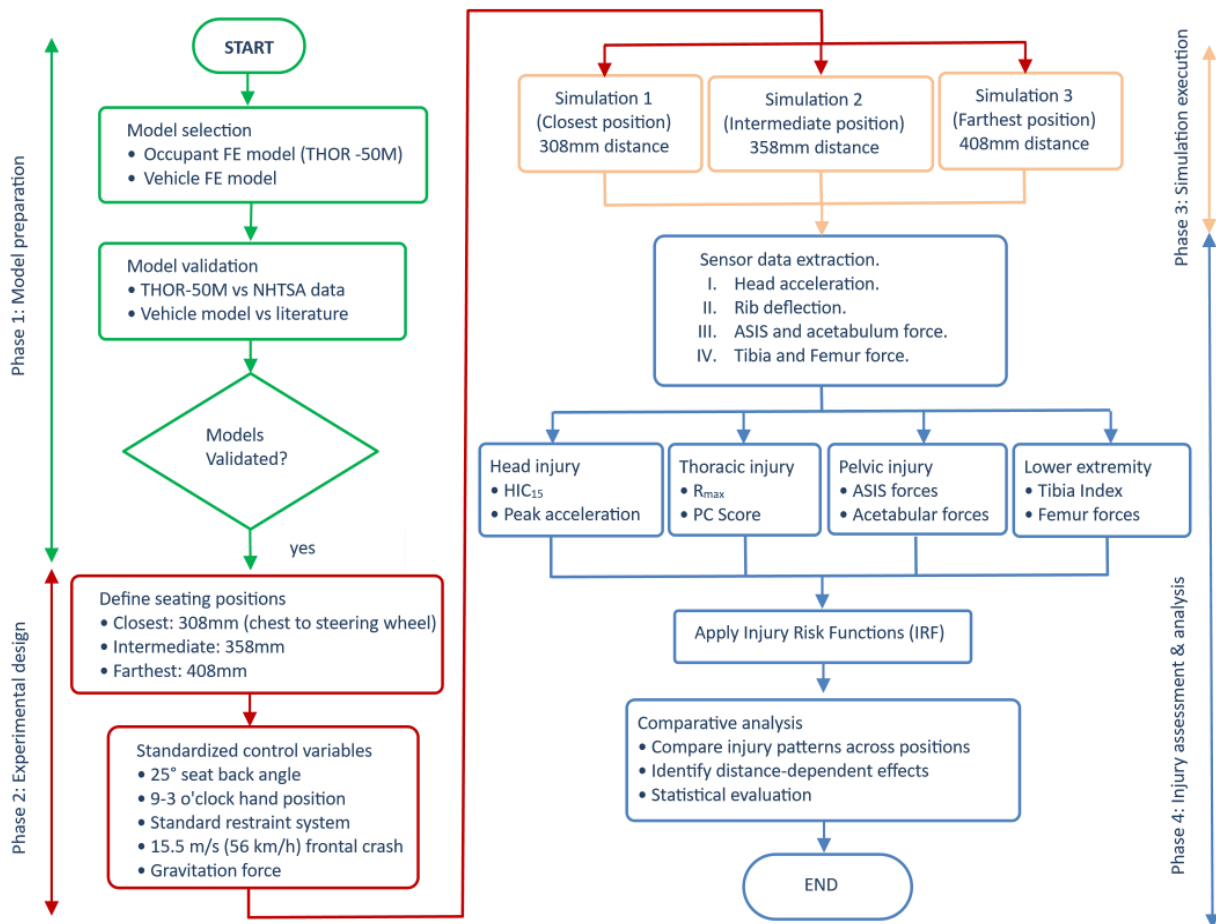


Figure 1. Outline of overall experimental setup

Table 1. Dummy positions setup for the simulations

Position (simulation number)	Distance between head and steering wheel (a) (mm)	Distance between chest and steering wheel (b) (mm)	Distance between dash panel and knee (c) (mm)
1	489	308	134
2	530	358	184
3	572	408	234

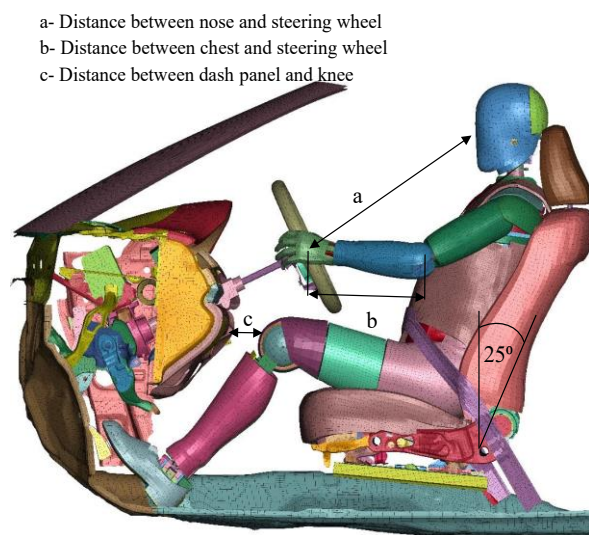


Figure 2. THOR-50M dummy seating configurations

the distance between chest and steering wheel, and the distance between dash panel, and knee were indicated as a, b, and c, respectively, as in Table 1. The distances from the chest to the steering wheel and from the knee to the dash panel were measured as horizontal distances [7].

These distances (308 mm, 358 mm, 408 mm) were selected based on real-world driving position data and safety guidelines. For instance, Gomez *et al.* [13] analyzed data from 890 drivers classified into three stature groups: under 163 cm, 163–178 cm, and over 178 cm and reported that chest-to-wheel distance range was from 305 mm to 480 mm. Importantly, the study established that a minimum distance of approximately 305 mm is necessary to avoid airbag-induced injuries during

deployment, which provides the basis for selecting 308 mm as the closest position. Welsh *et al.* [7] performed finite element simulations using distances of 265 mm, 320 mm, and 375 mm for small-stature drivers (≈ 160 cm), and Kim *et al.* [20] reported chest to steering wheel distances of 353 mm and 361 mm in physical crash tests.

The maximum distance of 408 mm was determined as the farthest position at which the dummy could maintain proper reach to both the steering wheel and pedals (The THOR-50M dummy used in this study has a height of 177 cm, placing it within the upper range of the 163–178 cm category from Gomez *et al.* [13]). The intermediate distance of 358 mm represents the midpoint between these safety-constrained minimum and ergonomically-constrained maximum positions. This distance range captures the full spectrum of feasible seating positions—from the minimum safe distance to the maximum ergonomic reach—thereby representing the complete envelope of driver-steering wheel interactions that occur in real-world frontal crashes.

2.1.3. Standardized control variables and boundary conditions

Initial Conditions- The entire vehicle structure was assigned an initial velocity of 15.5 m/s (56 km/h) in the forward direction [21], according to the standardized crash protocol (FMVSS 208). For all configurations, the THOR dummy was positioned in the driver's seat in a standard driving posture (mid-track positioning), which is the ideal driving position [22], as illustrated in Figure 2, ensuring consistency in the initial setup. The position was adjusted iteratively to ensure biomechanically accurate alignment with the vehicle interior geometry. It was found that the most common hand positions were the 10 and 2 o'clock positions or the 9 and 3 o'clock positions [11], [23]. According to these studies, 9 and 3 o'clock postures were selected as hand postures for all three simulations. The feet were positioned on the accelerator pedal and the brake pedal to replicate a realistic pre-crash posture. Based on previous investigations that used 20–30 degrees as the seat back angle, the seat back angle was adjusted to 25 degrees from the vertical axis [5], [24].

Contact and Constraint Boundary Conditions- The rigid barrier was fully constrained with zero degrees of freedom, and vehicle front nodes contacted the wall with a friction coefficient of 0.3, following FMVSS 208 configuration. The three-point seat belt system was implemented with realistic anchor geometry fixed to the vehicle body, maintained consistently across all simulations. The seat belts were modelled using one-dimensional (1D) elements for tension and two-dimensional (2D) elements for occupant interaction [25]. A pre-tensioner with 4 kN peak force (active 0–150 ms) and a load limiter were integrated to simulate belt tightening and prevent excessive loading [25]. Dummy-to-vehicle contact interactions were defined using penalty-based contact algorithms (AUTOMATIC_SURFACE_TO_SURFACE) with friction coefficient of 0.3 for seat surfaces, ensuring realistic occupant kinematics.

Mechanical Boundary Conditions- Ensuring a realistic vehicle approach, the vehicle model was created without artificial single-point constraints. All the joint nodes were modelled as constraint joints to replicate real-world mechanical behaviour. Spot welds were implemented as tied contacts with vehicle surfaces. Gravitational acceleration of 9.81 m/s^2 was applied in the negative z-direction throughout the simulation to ensure a realistic dummy posture, weight distribution, and belt tension.

Impact and Termination Conditions- The full-scale frontal crash simulation followed FMVSS 208 protocol, with the vehicle impacting a rigid barrier at 15.5 m/s (56 km/h) [21], generating a realistic crash pulse without prescribed deceleration. Simulation duration was 150 ms, capturing primary impact, restraint system engagement, maximum occupant excursion, and initial rebound phases. Output data for accelerations, forces, moments, and deflections were recorded at 10 kHz sampling frequency for detailed injury criteria assessment.

These boundary conditions are validated in automotive crash simulation research [8], [9], [10], [12]. The identical application of these conditions across all three seating configurations ensured that seating distance remained the sole variable influencing injury outcomes.

2.2. Injury assessment methodology

Injury criteria (IC) are a biomechanical index utilized to evaluate the severity of injuries. Kinematic and kinetic responses, measured in the crash test dummy, are used to derive injury criteria. In this study, ICs were extracted from the THOR-50M dummy to evaluate the potential trauma in the head, chest, pelvis, and lower limbs, which were interpreted using the corresponding injury risk function (IRFs), used to measure the probability of injury risk at various levels, defined by the Abbreviated Injury Scale (AIS) [4]. Figure 3 illustrated this process by mapping each targeted body region to its respective biomechanical criteria and the underlying virtual sensors data, where head injuries were assessed using Head Injury Criteria (HIC₁₅) from head acceleration, chest injuries were quantified using Principal Component (PC) score and R_{max} from chest acceleration and rib deflection, pelvic injuries were determined from anterior superior iliac spine (ASIS) and acetabulum forces, and lower extremity injuries were evaluated using the Tibia Index derived from tibia and femur forces.

2.2.1. Head injury assessment method

Two primary criteria were established to assess the severity of the head injury: the Head Injury Criteria (HIC) and the resultant peak acceleration of the center of gravity (CG) [11], [26]. According to the National Highway Traffic Safety Administration (NHTSA), HIC₁₅ (15 ms evaluation time window) was selected as the primary metric with a threshold of 700, proposed by the Automobile Manufacturers Alliance [26], [27]. Eq. (1) and Eq. (2) were used to compute the HIC value (Eq. (1)) [27] and injury probability (Eq. (2)) [28].

$$HIC = \max \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) dt \right]^{2.5} (t_2 - t_1) \quad (1)$$

$$\Pr(AIS3) = 1 - \exp \left(\frac{-9 \cdot HIC_{15}^{2.64}}{10^9} \right) \quad (2)$$

where $a(t)$ was the resultant linear acceleration of the head, measured in units of gravitational acceleration (g), and t_1 to t_2 denoted the time interval (in seconds) over which the HIC value reaches its maximum. Head acceleration was extracted from a virtual accelerometer placed at the center of gravity of the THOR dummy as shown in Figure 4.

2.2.2. Chest injury assessment method

To evaluate the chest injury, maximum Rib deflection (R_{max}) and Principal Component (PC) score were used, which illustrated a high correlation with chest injury patterns [15], [25], [29]. Four virtual sensors (IR-TRACC) were attached to the ribs of the dummy to obtain the R_{max} value. In this study, PC scores were calculated using the Eq. (3) [23], [28], [29], and injury risk functions of R_{max} and PC scores were calculated using the Eq. (4) and Eq. (5), respectively [23], [28], [29].

$$\frac{UP_{tot}}{35.88} + \frac{LOW_{tot}}{29.95} + \frac{UP_{dif}}{19.5} + \frac{LOW_{dif}}{23.54} \quad (3)$$

$$1 - e^{-\left(\frac{R_{max}}{58.183}\right)^{2.977}} \quad (4)$$

$$1 - e^{-\left(\frac{PC_{score}}{8.568}\right)^{3.31}} \quad (5)$$

where, UL- the Upper Left, UR- Upper Right, LL- Lower Left, and LR -Lower Right of IR-TRACC. $UP_{tot} = \max (UL+UR)$, $LOW_{tot} = \max (LL+LR)$, $UP_{dif} = \max (|UR-UL|)$, $LOW_{dif} = \max (|LL-LR|)$. Evaluated Thoracic IC and IRFs for the THOR-50M Dummy Representing a 40-Year-Old Occupant.

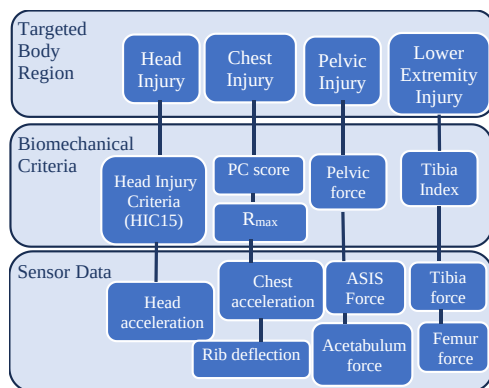


Figure 3. Injury mechanism

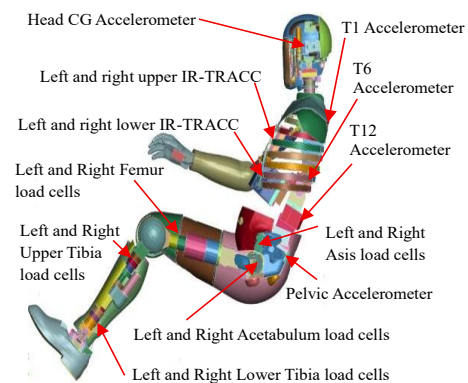


Figure 4. THOR-50M dummy virtual sensor and measurement locations

2.2.3. Pelvic injury assessment method

Pelvic injury criteria were measured by force responses, captured by the bilateral acetabulum virtual load cell and anterior superior iliac spine (ASIS) virtual load cells, mounted in the THOR-50M dummy. These components were employed to record the triaxial forces at the hip joint and capture the load, transmitted through the pelvic region [30], [31]. The ASIS load cells provide additional insight into lap belt loading and the potential for submarining effects, which are critical contributors to pelvic trauma during high-speed impact [30]. For analyzing the injury risk, the peak force of the acetabulum and ASIS were used.

2.2.4. Lower extremity injury assessment method

This study approached a method that widely accepts biomechanical criteria, primarily focusing on the Tibia Index (TI), to analyze lower extremity injury. Both axial forces and the resultant bending moment were employed to assess the likelihood of tibial shaft fractures [12]. To extract the data, virtual load cells were placed in the left and right femurs and upper tibia for both legs, which were identified as the critical locations of the THOR-50M dummy model [8], as illustrated in Figure 3. The Tibia Index was computed individually for each tibia, using the Equation 6 [8], [12].

$$TI = \frac{F}{F_c} + \frac{M}{M_c} \quad (6)$$

where F is the measured axial compressive force and M is the resultant bending moment, combining M_x and M_y . Critical threshold values used for injury prediction were $F_c = 12 \text{ kN}$ and $M_c = 240 \text{ Nm}$ [8], [12].

3. Results and Discussion

The vehicle's exterior deformation pattern remained identical across all three simulations. At the initial contact ($t = 0 \text{ ms}$), the vehicle structure remains unchanged, with all components in their original position. Deformations became visible at 30 ms, especially in the front bumper and hood area, and by 90 ms, the front-end collapse reached its maximum extent with a substantial shortening of the vehicle length. At 120 ms to 150 ms, the structural deformation stabilized, indicating that the main energy absorption phase has been completed and the vehicle has reached its final

deformed state. This progressive structural change was illustrated in Table 2.

3.1. Head Injury Criteria

Across all three simulations, a similar head motion patterns were observed and characterized by distinct phases of kinematic response. The head was relatively stable during the initial 30 ms at approximately the velocity of 15.5 m/s (56 km/h). Over the next 30 ms, the head began to move forward relative to the torso. The significant deceleration phase happened during the 50 to 80 ms period, during which the head velocity decreased from 15.5 m/s to near zero over around 25 ms (in each simulation), followed by rebound oscillations of $\pm 2\text{-}3 \text{ m/s}$ from 80-120 ms.

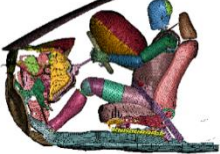
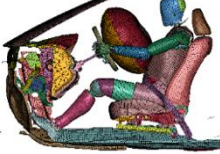
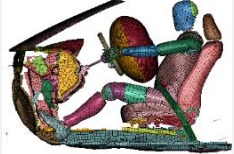
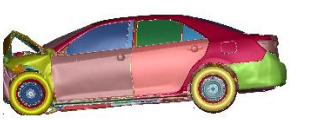
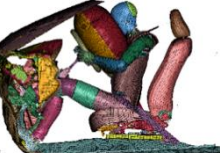
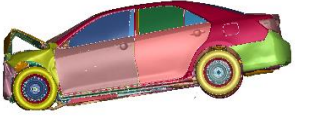
3.1.1. Comparative head injury metrics analysis

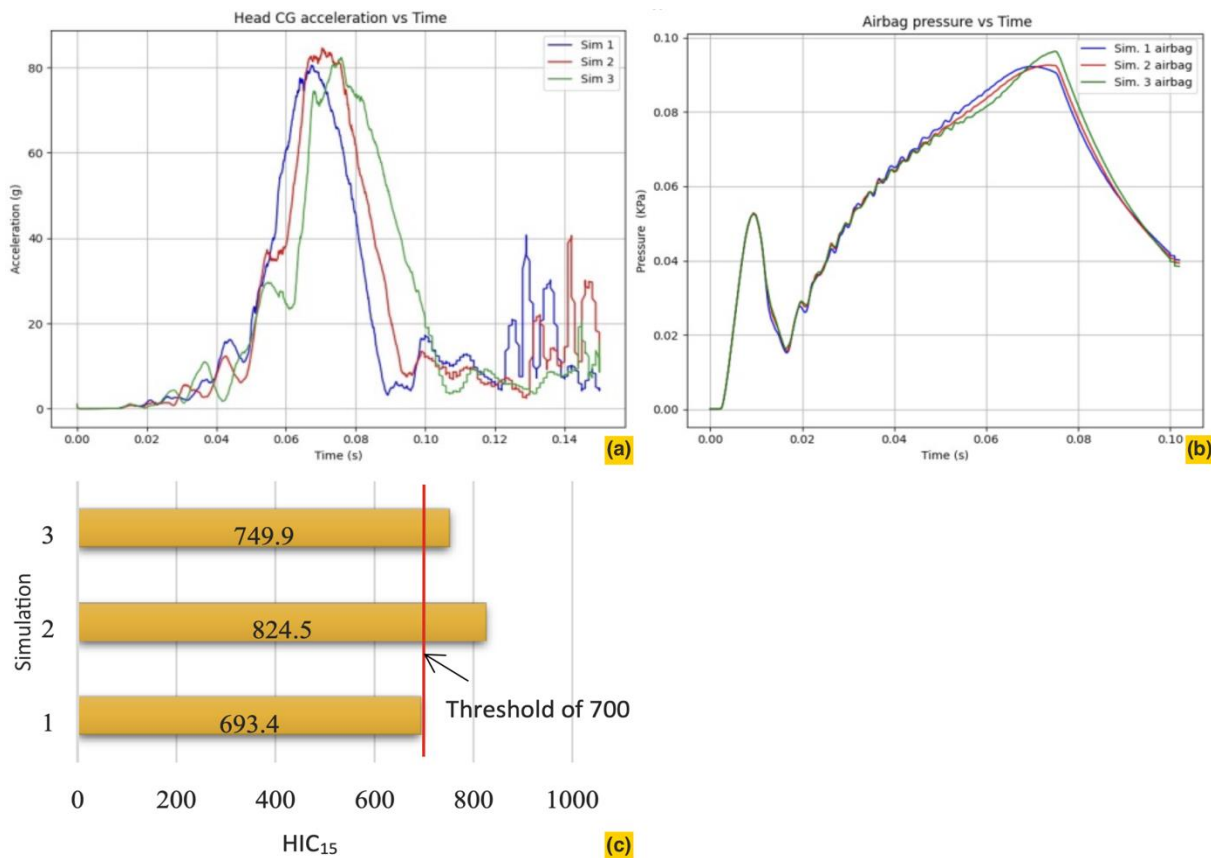
Simulation results demonstrated a noticeable variation in Head Injury Criteria (HIC_{15}) by changing seating distance, even with closely matching peak accelerations. The highest peak acceleration (85 g) was produced by Simulation 2 (intermediate position) with an HIC_{15} value of 824.5, contradicting conventional safety assumptions that proximity to the steering wheel correlates directly with injury severity. In contrast, the lowest injury severity occurred in Simulation 1 (closest position) with peak acceleration value of 81.9 g and HIC_{15} value of 693.4, which was the only HIC_{15} value below the threshold of 700 [26], [27], illustrated in Figure 5a and Figure 5c. Concurrently, Simulation 3 (farthest position) generated an intermediate value for both HIC_{15} value (749.9) and peak acceleration (83g). Despite the minimal peak acceleration difference (3.1g - lowest to highest peak acceleration), HIC_{15} showed 31% noticeable variation, further supported by the injury probability value, with Simulation 1 demonstrating 24.8% probability of AIS 3+ injury severity, Simulation 2 showing 36.2% probability, and Simulation 3 exhibiting 29.5% probability, demonstrating a non-linear relationship between seating distance and head injury risk.

3.1.2. Head-restrain system interaction

The mechanism behind these unexpected injury patterns lies in head-airbag interaction dynamics. In Simulation 1, the head acceleration reached a peak at 67 ms, corresponding to an airbag pressure of approximately 91 kPa and an

Table 2. Simulation results

Time	Vehicle model	Simulation 1 (Dummy)	Simulation 2 (Dummy)	Simulation 3 (Dummy)
30 ms				
60 ms				
90 ms				
120 ms				

**Figure 5.** Head injury analysis results; (a) Head C.G. acceleration, (b) Airbag pressure, (c) HIC₁₅ value

airbag stiffness of 28.2 N/mm. Following this, the airbag pressure was slightly reduced (average deflation rate: 0.12 kPa/ms) to approximately 90

kPa (airbag stiffness reduced to 23.1 N/mm) by 75 ms, at which point the airbag began to deflate. Conversely, in Simulation 2, the peak

acceleration occurred at 70 ms (airbag pressure: 92 kPa, airbag stiffness: 29.7 N/mm), with rising airbag pressure (average inflation rate: 0.13 kPa/ms) to 92.6 kPa at 75 ms, as shown in [Figure 5b](#). This prolonged exposure to the increasing pressure and higher stiffness likely contributed to the elevated HIC_{15} value in Simulation 2. Meanwhile, the pressure decreased with an 18% reduction in stiffness (28.2 N/mm to 23.1 N/mm), following peak acceleration in Simulation 1, which led to a reduction in HIC_{15} value. Even though Simulation 3 resulted in high peak airbag pressure, the airbag's mean effective pressure (during the head deceleration phase) was 85.7 kPa, which was lower than the mean effective pressure of Simulation 2 (88 kPa). Similarly, the stiffness of the airbag (28.0 N/mm) at the peak acceleration was 5% lower than that of the Simulation 2 airbag stiffness (29.7 N/mm) at its peak acceleration. The lower pressure and reduced stiffness resulted in a significantly lower head injury in Simulation 3 compared to Simulation 2. This demonstrates that seating distance creates a non-linear injury relationship governed by airbag contact synchronization rather than direct proximity effects.

These insights highlight that head injury severity is highly sensitive to the contact pressure of the airbag, which varies with the timing of head–airbag interaction relative to the airbag's inflation. The results aligned with established safety research, which notes that “timing of events is critical when investigating the use of restraint mechanisms” [7]. Current airbag deployment strategies, which assumed optimal performance with standardized timing protocols, may require adaptive algorithms to accommodate variable seating configurations and minimize injury risk across different driver positions.

3.2. Chest Injury Criteria

Across all three simulations, the occupant's torso remained relatively stable, with the approximate speed of 15 m/s until around 20 ms to 35 ms. Initial rib compression began between 20 ms and 40 ms due to the forward motion of chest relative to the vehicle structure. The peak rib compression occurred during the 60–80 ms at maximum occupant deceleration. Following this peak compression, the chest rebounded between 75 ms and 90 ms.

3.2.1. Anatomical distribution and load redistribution analysis

These temporal patterns produced varying deflection magnitudes depending on seating position. Simulation 1 (closest position) and 2 (intermediate position) resulted in the highest peak rib deflection (87.2 mm at 67 ms) and intermediate peak rib deflection (85.1 mm at 65 ms) in the upper right rib, respectively, as illustrated in [Figure 6a](#) and [Figure 6b](#). In contrast, Simulation 3 (farthest position) generated the lowest peak rib deflection with 81.8 mm (at 78.9 ms) in the lower right rib, indicating that the peak deflection shifted to the lower right rib.

The upper right rib exhibited greater rib deflection due to shoulder belt alignment and direct exposure to airbag loading, compared to the upper left rib, which experiences deflection primarily from airbag pressure. Meanwhile, the lower left rib illustrated the lowest value among all four ribs, and the right lower rib was primarily subjected to shoulder belt exposure. However, load distribution progressively shifted towards the lower ribs with increasing seating distance, which increased the seatbelt angle at the floor anchor point. The increased belt angle resulted in higher Y and Z-components of seatbelt force, directed toward the lower chest region, while simultaneously reducing the X-component of the force. This reduction in X-component of the force contributed to lower X-displacement in the upper right ribs, where X-displacement is the dominant deformation mechanism, compared to Y and Z displacement. The Enhanced Y and Z components of the seatbelt force produced higher displacement in the lower right ribs, resulting in greater deflection in the lower right ribs, as shown in [Figure 6c](#). Thus, increasing seating distance systematically redistributes chest loading from upper to lower rib regions through altered seatbelt geometry.

3.2.2. Injury risk assessment

To quantify the injury implications of these deflection patterns, maximum Rib deflection (R_{max}) was used as the primary metric; Simulation 1, Simulation 2, and Simulation 3 corresponded to 96.5%, 95.5%, and 93.7% probability of sustaining AIS 3+ chest injury, respectively, reducing the injury risk by 2.8% as the seating distance increased. Additionally, Principal Component (PC) scores revealed distinct patterns of injury outcomes. Simulations 1 and 2 sustained the highest

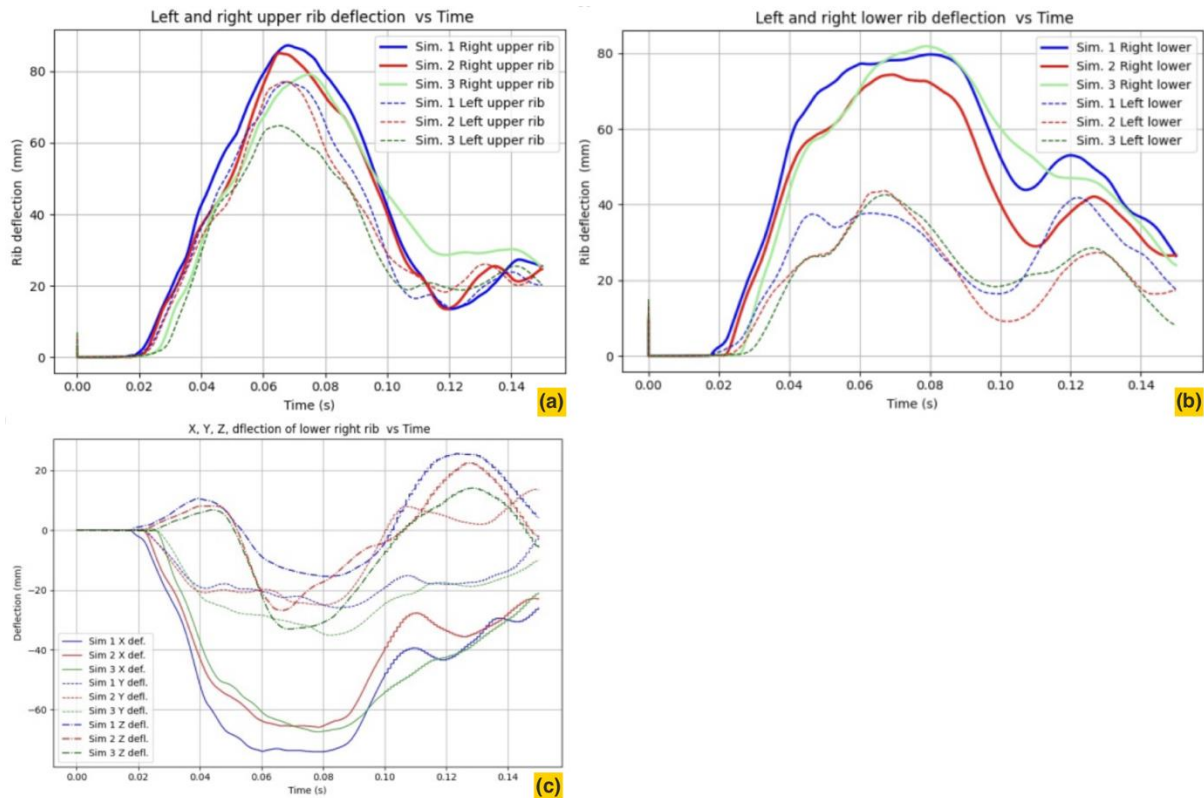


Figure 6. Chest injury analysis results; (a) Left, and right upper rib deflection, (b) Left and right lower rib deflection, (c) X, Y, Z component of right lower rib deflection

and lowest PC scores of 10.8 (88.1% AIS 3+ injury probability) and 10.2 (82.9% probability), respectively, and Simulation 3 scored an intermediate value of 10.6 (86.5% probability).

Despite the intermediate peak deflection value, Simulation 2 produced the lowest PC score, by a 3.6% lower AIS 3+ injury probability than Simulation 3. This PC scores variation occurred, because PC scores capture multi-regional loading patterns, while maximum rib deflection measures peak single-location deformation. In Simulation 3, primarily high deflection of the lower right rib and force redistribution increased the PC score, despite lower peak deflection. These findings illustrated the multi-dimensional nature of chest injury mechanisms, where peak deflection and multi-regional loading patterns respond differently to seating position variations.

3.2.3. Chest-restraint system interaction

These injury variations can be attributed to differences in interactions between the chest-restraint system and seating positions. In all simulations, the chest interacted with airbags during the inflation phase. However, Simulation 1 experienced the airbag's expanding phase due to

early engagement (at around 20 ms), resulting in dynamic loading forces from the expanding airbag system that increased the peak deflection. Additionally, the occupant's forward motion caused deformation of the steering column, which applied further pressure to the chest and increased rib deflection. Conversely, the chest began to engage with the airbag at around 40 ms in Simulation 3, experiencing partial exposure to both the inflation and deflation phases, which allowed for more gradual occupant deceleration and reduced dynamic load. This gradual deceleration and increased distance also minimized steering column deformation, resulting in decreased peak deflection values. Simulation 2, with an intermediate engagement timing of 30 ms and moderate steering column deformation, yielded peak deflection values that fell between the two extremes. The timing differences in airbag engagement described above directly explain the non-linear injury patterns observed in both R_{max} and PC scores.

These findings demonstrate that seating distance influences injury risk non-linearly, with different injury metrics yielding varying risk rankings depending on the specific injury

mechanism evaluated. However, aligning with previous investigations [24], these results suggested that current fixed-point seatbelt anchor systems and uniform airbag deployment strategies may not optimally accommodate varying seating distances. The findings indicated that both seatbelt anchor points and airbag deployment timing could benefit from adaptive configurations that adjust based on occupant seating position.

3.3. Pelvic and Lower Extremity Injury Analysis

Similar kinematic patterns were observed in the upper body region, initiating the forward transition motion at around 20 ms. During 60-90 ms, the peak deceleration occurred, and at around 90 ms, the pelvic and the lower extremities began to rebound due to the occupant's deceleration.

3.3.1. Pelvic force analysis

Significant variations of the left anterior superior iliac spine (ASIS) forces were demonstrated across all simulations. The highest peak force was recorded in Simulation 2

(intermediate position, 9.2 kN), followed by Simulation 3 (farthest position) at 8.84 kN, and Simulation 1 (closest position) producing the lowest peak of 8.1 kN, as shown in Figure 7a. Corresponding right ASIS forces were 6.10 kN, 6.63 kN, and 5.61 kN, respectively, and left acetabulum forces (Figure 7b) for simulations 1, 2, and 3 were 2.79 kN, 3.0 kN, and 3.5 kN, respectively. It was found that increasing the seating position (simulation 1 to 3) raised the peak left ASIS force and acetabulum force by 9.1% and 25%, respectively, illustrating a load shift towards the Pelvic region with increased seating distance.

Load Distribution Mechanisms- Reflecting the combined effect of the femur loading and seatbelt force, both tensile (due to lower extremity inertia) and compressive forces were recorded in the acetabulum load cells [30]. In every simulation, ASIS forces began to rise at approximately 40 ms, coinciding with engagement of the lap belt (Figure 7c). ASIS forces were purely compressive, directly tracking seatbelt forces. Notably, Simulation 2 received a higher left ASIS force than Simulation 3, despite a

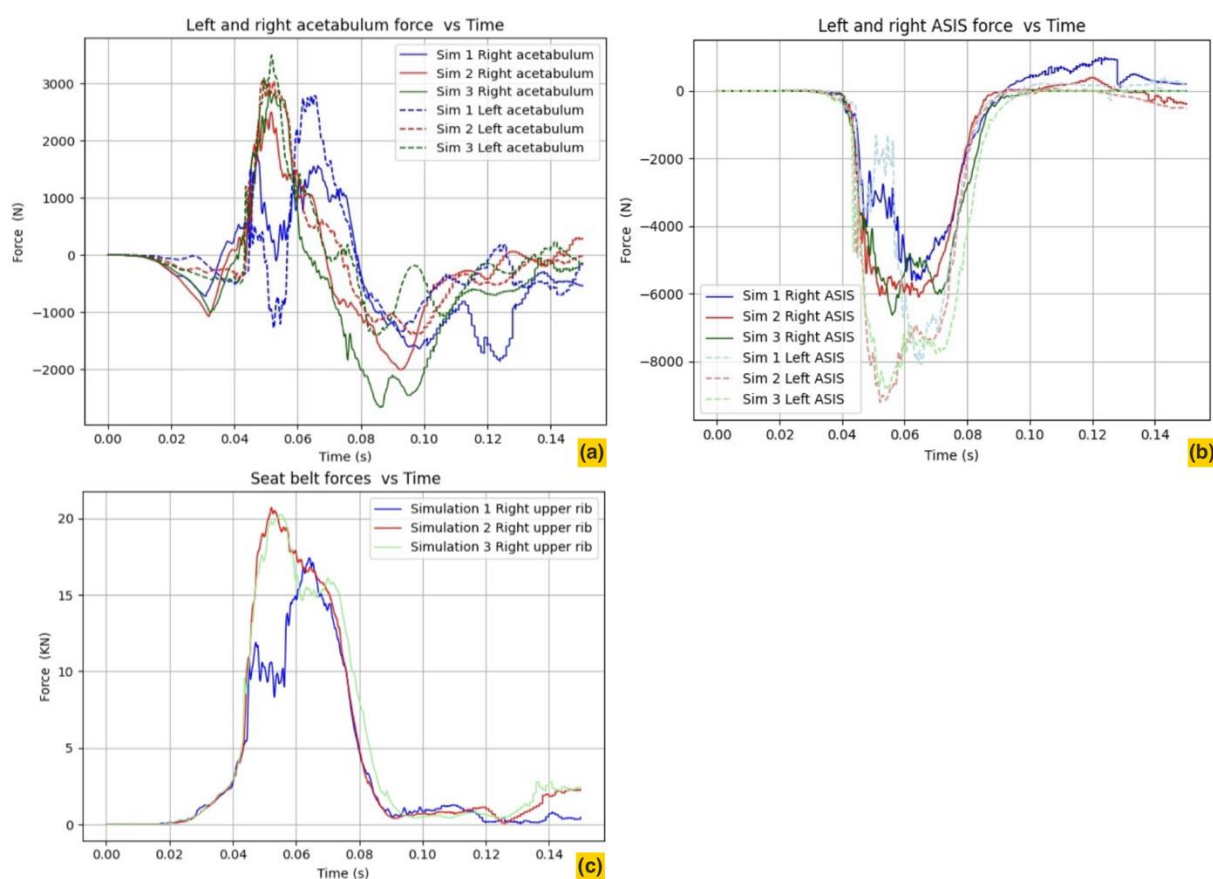


Figure 7. Pelvic injury analysis results; (a) Left, and right acetabulum force, (b) Left and right ASIS force, (c) Seatbelt force

lower right side ASIS force compared to Simulation 3. These complex force patterns result from the interplay between seating distance and occupant deceleration mechanisms.

Seating Position Effects on Pelvic Injury Patterns- These complex force patterns result from varying seating distances that influence the pattern of injuries. Early and hard contact (at 50 ms) of the knee with the front panel created compression loads on the femurs, which helped decelerate the occupant, reducing the loads on the ASIS (8.1 kN) and the acetabulum (2.79 kN) in Simulation 1. In contrast, in Simulations 2 and 3, the knee generated late (55-60 ms) and light contact forces with the front panel, allowing the lap belt to absorb a larger portion of the occupant's deceleration. This progressive load transfer to the pelvic region resulted in 9.1% and 25.4% increases in ASIS and acetabulum forces, respectively, from closest to farthest position (Simulation 1 to Simulation 3). These findings illustrate the inverse relationship between knee contact effectiveness and pelvic loading across seating positions.

3.3.2. Lower extremity injury metrics analysis

Force Distribution- The tibia force response pattern showed an inverse linear relationship with seating distance. The highest (1.12 kN) and the lowest (772 N) tibial axial forces were produced by Simulations 1 and 3 with tibia index (TI) of 0.32 and 0.36, respectively, represented in Figure 8a. Meanwhile, the intermediate tibial axial force (787 N) and the lowest tibia index (0.21) resulted in Simulation 2. All tibia index values remained below the critical threshold of 1.0 [12], indicating low fracture probability. Statistics revealed that increasing the seating distance

reduced the tibial axial force by 31.1%. The peak Femur force (Compressive) at the knee impact was 710 N in Simulation 1, as illustrated in Figure 8b. Simulations 2 and 3 resulted in tensile forces of 607 N and 715 N, respectively, on the femur at the time, indicating an extremely low impact on the knee. However, this force reduction pattern does not directly translate to injury risk due to the complex interplay of multiple loading components.

Seating Position Effects on Lower Extremity Injury Patterns- The force variations described above reflected distinct injury mechanisms that vary with seating position. Simulation 1 produced the highest femur force (710 N) and tibial axial force (1.12 kN), due to the early hard contact generated by a low decelerated velocity (compared to a higher velocity), which aligns with previous investigation results [12]. In Simulation 2, the intermediate tibial axial force and the lowest bending moment (36 Nm) were obtained. Due to its extremely low impact on the knee, as illustrated by the femur force (tensile force), and optimal loading distribution, it resulted in the lowest tibia index (TI). Meanwhile, Simulation 3 resulted in the lowest tibia axial force but the highest bending moment (73 Nm), due to higher knee flexion [12], which created longer arm moments and amplified bending stresses with a highest tibia index (TI). These competing mechanisms, characterized by reduced axial forces versus increased bending moments, result in non-linear injury risk patterns across seating positions.

Tibia Index Analysis and Injury Risk Assessment- Due to competing biomechanical factors, a non-linear tibia index (TI) pattern was

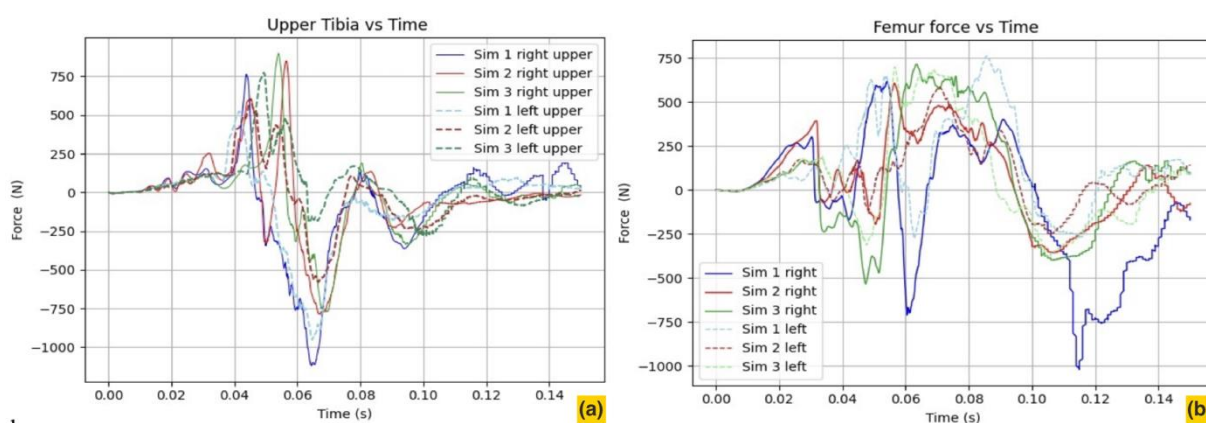


Figure 8. Lower Extremity injury analysis results: (a) Left, and upper tibia force; (b) Left and right femur force

exhibited. A 34.4 % TI reduction was observed from simulation 1 to 2. TI increased by 71.4% from simulation 2 to 3 due to the dramatic escalation of bending moment component (30.2% higher bending moment compared to Simulation 1) with increased knee flexion angle in the farthest position (Simulation 3), despite lower axial tibia force.

These findings demonstrate that seat-track position critically influences the distribution of impact load between the hip and lower extremities. The pelvic-to-lower extremity force ratio (ASIS: tibia axial force) increased from 7.2:1 (Simulation 1) to 11.4:1 (Simulation 3). Closer positioning shifts the load into the femur and tibia, while reducing pelvic forces, whereas distant positioning transfers more energy into the hip joint and increases tibial bending moments due to extreme tibial flexion angles. Confirming the finding, a previous investigation reported that the Seat-track position should be considered an active tuning parameter for optimizing both pelvic and lower-extremity injury mitigation strategies in automotive safety design [8]. This demonstrates that optimal seating position requires balancing trade-offs between different injury mechanisms rather than simply minimizing individual force components.

3.4. Limitations

This study presents several limitations that should be acknowledged. The investigation utilized only the THOR-50M (50th percentile male) anthropomorphic test device, limiting generalizability to other body sizes, genders, and age groups. Real-world crash scenarios involve variable impact angles, speeds, and environmental conditions, whereas this study employed idealized frontal rigid wall crashes at a single velocity (56 km/h). The finite element model validation, while based on established protocols, inherently contains uncertainties in material properties and contact algorithms that may influence injury predictions. Statistical significance testing was not performed due to single simulation runs per configuration, preventing assessment of result variability and confidence intervals. The injury analysis focused on four primary body regions (head, thorax, pelvis, lower extremities), potentially overlooking injuries to the neck, and upper extremities.

4. Conclusion

This study was conducted to investigate the relationship between the driver to steering wheel distance and the injury severity of selected body regions, using a full-frontal collision at 15.5 m/s (56 km/h). Controlled Finite element analysis was utilized in this research with the help of the THOR-50M dummy. By isolating seating distance as the only variable, this research demonstrated that injury outcomes vary significantly with changes in occupant positioning. For the Head Injury risk, all three simulations produced much closer results. The highest HIC value was received from Simulation 2 (mid position) due to prolonged exposure to airbag inflation pressure, emphasizing the critical role of airbag contact timing. To represent the chest injury severity, rib deflection and PC score were selected. It was noted that a closer driving position increased the rib deflection due to early and forceful interaction with the deploying airbag. Meanwhile, increased seating distance shifted the load distribution toward the lower chest region due to changes in seatbelt geometry and anchor point angles, resulting in higher PC score in farthest position. Pelvic and lower extremity responses revealed that seating distance affects the load distribution between the lap belt and the limbs. According to the result, a lower distance increased the force on the tibia and femur due to the early and hard contact of the knee with the dashboard, while reducing pelvic forces. Conversely, farther distances allowed greater excursion and higher belt loading at the pelvis, with elevated ASIS and acetabular forces. The findings highlight that moderate changes in driver seating position can alter occupant kinematics and injury risk profiles across multiple body regions.

4.1. Design Implications and Recommendations

(1) Adaptive airbag deployment systems: Airbag deployment timing should be adjusted based on detected occupant seating position. (2) Intelligent seating position guidance: Driver assistance systems should provide real-time feedback on optimal seating positions balancing multiple injury mechanisms across body regions. (3) Enhanced position detection: Modern vehicles should integrate sensors to detect driver-to-steering wheel distance for adaptive restraint system tuning. These findings provide

foundational understanding for the development of position-adaptive safety systems that could reduce injury severity across diverse driver seating preferences, with future research needed to validate these concepts through experimental testing and system prototyping.

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Author's declaration

Authors' contributions and responsibilities

P.A.K.P.I, M.Z.H. The authors made substantial contributions to the conception and design of the study. **P.A.K.P.I** The authors took responsibility for data analysis, interpretation and discussion of results. **M.Z.H., J.K.** The authors read and approved the final manuscript.

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Competing interests

The authors declare no competing interest.

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References

- [1] M. I. Sulaiman *et al.*, "Non-Driving Related Activities Inside an Automated Vehicle Among Malaysia Passengers," *Automotive Experiences*, vol. 6, no. 3, pp. 452–465, Nov. 2023, doi: 10.31603/ae.9152.
- [2] A. I. Petrov and A. V. Pistsov, "Training and Applying Artificial Neural Networks in Traffic Light Control: Improving the Management and Safety of Road Traffic in Tyumen (Russia)," *Automotive Experiences*, vol. 6, no. 3, pp. 528–550, Nov. 2023, doi: 10.31603/ae.10025.
- [3] World Health Organization, *Global status report on road safety 2023*. 2023.
- [4] N. Lubbe, H. Jeppsson, S. Sternlund, and A. Morando, "Injury risk curves to guide safe speed limits on Swedish roads using German crash data supplemented with estimated non-injury crashes," *Accident Analysis & Prevention*, vol. 202, p. 107586, Jul. 2024, doi: 10.1016/j.aap.2024.107586.
- [5] A. Leledakis, J. Östh, J. Davidsson, and L. Jakobsson, "The influence of car passengers' sitting postures in intersection crashes," *Accident Analysis & Prevention*, vol. 157, p. 106170, Jul. 2021, doi: 10.1016/j.aap.2021.106170.
- [6] D. Bose, J. R. Crandall, C. D. Untaroiu, and E. H. Maslen, "Influence of pre-collision occupant parameters on injury outcome in a frontal collision," *Accident Analysis & Prevention*, vol. 42, no. 4, pp. 1398–1407, Jul. 2010, doi: 10.1016/j.aap.2010.03.004.
- [7] R. Welsh, L. Clift, A. Morris, S. Cook, and J. Watson, *An assessment of cars for small drivers*. 2003.
- [8] X. Ye *et al.*, "Numerical investigation of driver lower extremity injuries in finite element frontal crash reconstruction," *Traffic Injury Prevention*, vol. 19, no. sup1, pp. S21–S28, Feb. 2018, doi: 10.1080/15389588.2017.1376051.
- [9] X. Ye, J. P. Gaewsky, D. A. Jones, L. E. Miller, J. D. Stitzel, and A. A. Weaver, "Computational modeling and analysis of thoracolumbar spine fractures in frontal crash reconstruction," *Traffic Injury Prevention*, vol. 19, no. sup2, pp. S32–S39, Dec. 2018, doi: 10.1080/15389588.2018.1498090.
- [10] E. Mishra and N. Lubbe, "Assessing injury

- risks of reclined occupants in a frontal crash preceded by braking with varied seatbelt designs using the SAFER Human Body Model," *Traffic Injury Prevention*, vol. 25, no. 3, pp. 445–453, Apr. 2024, doi: 10.1080/15389588.2024.2318414.
- [11] B. Jonsson, "Hand Position on Steering Wheel During Driving," *Traffic Injury Prevention*, vol. 12, no. 2, pp. 187–190, Mar. 2011, doi: 10.1080/15389588.2010.533722.
- [12] S. Xiao, S. You, T. Tian, J. Wu, and H. Zhang, "Investigation of lower limb injury under different contact stiffness for drivers during frontal crash," *Acta of Bioengineering and Biomechanics*, vol. 24, Jan. 2022, doi: 10.37190/ABB-02057-2022-02.
- [13] M. Segui-Gomez, J. Levy, H. Roman, K. M. Thompson, K. McCabe, and J. D. Graham, "Driver distance from the steering wheel: perception and objective measurement," *American Journal of Public Health*, vol. 89, no. 7, pp. 1109–1111, Jul. 1999, doi: 10.2105/AJPH.89.7.1109.
- [14] J. Forman *et al.*, "Automobile injury trends in the contemporary fleet: Belted occupants in frontal collisions," *Traffic Injury Prevention*, vol. 20, no. 6, pp. 607–612, Aug. 2019, doi: 10.1080/15389588.2019.1630825.
- [15] L. Watkins, W. Millis, and H. Kwon, "THOR-50M In-Dummy Data Acquisition System Evaluation (Report No. DOT HS 813 618)," 2024.
- [16] M. B. Panzer, S. Guidice, and J. R. Crandall, "Evaluation of the NHTSA THOR finite element model," 2015.
- [17] K. Maehara, Y. Dokko, and K. Ohhashi, "A Study on the Method to Reduce Thoracic Injury in Frontal Crash using Elderly Human and THOR FE Models," in *27th International Technical Conference on the Enhanced Safety of Vehicles (ESV)*, 2023.
- [18] Center for Collision Safety and Analysis, "Development & Validation of a Finite Element Model for the 2012 Toyota Camry Passenger Sedan," *Tech Summary*, no. May, pp. 1–5, 2016, doi: 10.13021/G8N88.
- [19] D. Marzougui, D. Brown, H. K. Park, C. D. Kan, and K. S. Opiela, "Development & Validation of a Finite Element Model for a Mid-Sized Passenger Sedan," in *13th International LS-DYNA Users Conference*, 2014, pp. 1–10.
- [20] A. J. Golman, K. A. Danelson, L. E. Miller, and J. D. Stitzel, "Injury prediction in a side impact crash using human body model simulation," *Accident Analysis & Prevention*, vol. 64, pp. 1–8, 2014, doi: 10.1016/j.aap.2013.10.026.
- [21] A. Hault-Dubrulle, F. Robache, M.-P. Pacaux, and H. Morvan, "Determination of pre-impact occupant postures and analysis of consequences on injury outcome. Part I: A driving simulator study," *Accident Analysis & Prevention*, vol. 43, no. 1, pp. 66–74, Jan. 2011, doi: 10.1016/j.aap.2010.07.012.
- [22] G. P. Siegmund, D. D. Chimich, B. E. Heinrichs, A. L. DeMarco, and J. R. Brault, "Variations in Occupant Response with Seat Belt Slack and Anchor Location During Moderate Frontal Impacts," *Traffic Injury Prevention*, vol. 6, no. 1, pp. 38–43, Mar. 2005, doi: 10.1080/15389580590903159.
- [23] Xuewei Shi, Sen Xiao, and Yanchao Qie, "Analysis of chest injury characteristics in frontal vehicle collision under three typical overlaps," *Proceedings of the Institution of Mechanical Engineers, Part D: Journal of Automobile Engineering*, vol. 235, no. 2–3, pp. 580–588, Sep. 2020, doi: 10.1177/0954407020957482.
- [24] W. T. Hollowell, H. C. Gabler, S. L. Stucki, S. Summers, and N. James R. Hackney, *Review of potential test procedures FMVSS No. 208*. 1998.
- [25] H. J. Mertz, A. L. Irwin, and P. Prasad, "Biomechanical and Scaling Basis for Frontal and Side Impact Injury Assessment Reference Values," in *Stapp car crash journal*, Nov. 2016. doi: 10.4271/2016-22-0018.
- [26] R. Eppinger *et al.*, "Development of Improved Injury Criteria for the Assessment of Advanced Automotive Restraint Systems - II By," in *National Highway Traffic Safety Administration*, Washington, DC, 1999.
- [27] L. Chybowski and W. Przetakiewicz, "Estimation of the Probability of Head Injury at a Given Abbreviated Injury Scale Level by Means of a Fuction of Head Injury Criterion," *System Safety: Human - Technical Facility -*

- Environment*, vol. 2, no. 1, pp. 91–99, Mar. 2020, doi: 10.2478/czoto-2020-0012.
- [28] M. Nylund and N. Olofsson, “Rib fracture injury risk function assessment for the THOR-50M using population-based finite element crash reconstructions,” Chalmers University Of Technology, 2022.
- [29] Z. Gao, J. Cheng, R. McCoy, X. Li, R. El-Jawahri, and S. Barbat, *Acetabulum Injury Investigation of Proposed US-NCAP in OI Mode*. SAE Technical Paper, 2018. doi: 10.4271/2018-01-0538.
- [30] J. Uriot, P. Potier, P. Baudrit, X. Trosseille, O. Richard, and R. Douard, “Comparison of HII, HIII and THOR dummy responses with respect to PMHS sled tests,” IRCOBI Conference 2015, 2015, pp. 435–451.
- [31] P. Li, Y. Lei, J. Liao, D. Zhang, X. Dong, and T. Zhang, “Study of AEB and active seat belt on driver injury in vehicle–vehicle frontal oblique crash,” *Scientific Reports*, vol. 13, no. 1, p. 22621, 2023, doi: 10.1038/s41598-023-48729-7.