Sensitivity of Lumbar Spine Loadings to Different Seat Characteristics and Human Body Model Boundary Conditions

María González-García, Jens Weber, Steffen Peldschus, Andre Eggers

Abstract With the introduction of human body models in virtual testing and the future objective of assessing occupant loading during crash events, the required level of model quality of the occupant environment becomes a relevant question. This study analyses the sensitivity of human body model loadings, with focus on the lumbar spine, to parameter variations with respect to the stiffness of the seat model, the boundary conditions and the occupant posture. Variations in seat stiffness resulted in a maximum deviation of 14% in the compressive force and 22% in the bending moment at L1 level for the configuration with the largest reduction in the cross-beam thickness of the seat. The next highest deviations due to seat stiffness variations were 5% in compressive force at L2 and 14% in bending moment at L1. In contrast, variations in boundary conditions led to substantially larger effects, with deviations up to 39% in compressive force at L1 and 81% in bending moment at L1. These results indicate that variations in occupant posture and initial boundary conditions have a greater influence on lumbar spine loading than moderate changes in seat structural stiffness. This underlines the importance of carefully defining initial conditions in virtual safety assessments.

Keywords HBM positioning, Seat structure, Virtual testing.

I. INTRODUCTION

The prevailing injury assessment methods applied during vehicle safety development process and finally by legal authorities and consumer rating organisations employ standardised injury metrics for different body regions, mainly using crash test dummies. However, these methods have evolved from global kinematic evaluations toward increasingly detailed analyses of internal load paths and tissue-level responses [1-2], supported by the growing application and improvement of finite element Human Body Models (HBMs). An even greater increase in their use has yet started following their introduction into virtual testing assessment procedures [3-5]. Although these models enable detailed analysis of load transfer within the body and injury mechanisms, their predictions remain sensitive to modelling assumptions and boundary conditions when applied in a vehicle model. Parameters influencing the contact with the vehicle interior and safety systems, as well as the initial HBM posture, can affect the resulting injury outcomes. HBMs provide the opportunity to complement current safety assessments based on dummies, leading to a more robust evaluation of occupant safety.

Although lumbar spine injuries do not frequently occur – accounting for approximately 5% of all vehicle crashes [6] - they are often associated with chronic pain and long-term functional limitations [7]. Hence, the analysis of lumbar spine loadings remains relevant, and their investigation gains importance with the potential allowance of reclined seating postures in autonomous driving. Previous PMHS studies focusing on lumbar spine injuries [8-9] used rigid or semi-rigid seat structures. Therefore, the transfer of the findings to deformable vehicle seats remains uncertain. Additionally, it has been hypothesised that the assessment of spinal loading may require a reasonably validated seat model. Thus, for an injury assessment in virtual testing an additional qualification confirming a sufficient validation level of the seat might be required [10]. The interaction between the HBM and the seat represents the initial link in the transmission of loads from the pelvis to the lumbar spine. It is therefore essential to ensure not only that the HBM itself is properly validated, but also that the interaction of the HBM with the environment is accurately modelled. Furthermore, previous studies with HBMs [11-12] have shown that variations in HBM positioning procedure and posture can influence the predicted injury outcome.

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Previous researchers [13-14] have studied the sensitivity of the lumbar spine to modifications in the geometry of vertebral bodies, posture and soft tissue material properties – ligaments and intervertebral discs (IVDs) – in component tests under dynamic loading with the THUMS and GHBMC models, respectively. Rieger *et al.* (2024) [13] reported that while morphing the vertebral bodies to subject-specific geometry had an influence on the pressure of the trabecular bone, the major effect was accounted when considering subject-specific spinal curvature. On the other hand, Tushak *et al.* (2024) showed minor effect of subject-specific vertebral body geometrical changes compared to application of the GHBMC baseline geometry in the moment-angle measurements. The scaling of the soft tissue stiffness had a greater role in this latter investigation.

Tang *et al.* (2020) [15] performed a numerical analysis to assess which factors influence lumbar spine forces using HIII mid-size male dummy. Among the parameters that were varied are: position of anti-submarining bar in the seat design, coefficient of friction in the contact between seat and pelvis, pulse severity and posture. Position of the anti-submarining bar was the most statistically significant factor, and while decreasing the coefficient of friction reduced the lumbar spine forces, this parameter did not show a statistical significance.

The present study aims to quantify the sensitivity of lumbar spine loading in a Human Body Model to variations of the following factors: seat model characteristics, HBM–seat interaction, HBM posture and other initial boundary conditions. The objective is to identify the most sensitive factors influencing lumbar spine loading and, consequently, being potentially relevant for injury risk assessment. Ultimately, the findings are intended to highlight the factors that could be considered and prioritised when defining the required qualification level for virtual testing and related processes.

II. METHODS

The sensitivity of the lumbar spine loadings to different boundary conditions was investigated in this study applying the 50th percentile male THUMS v4.1 VW Group model. This Human Body Model includes modifications in the lumbar spine modelling, such as the cortical thickness, according to Yadav *et al.* (2026) [16].

A total of 31 simulations were conducted including seat parameter changes as well as variations in the initial boundary conditions of the HBM. All the simulations were computed with the solver Virtual Performance Solution (VPS) (ESI Group, Keysight, Bagneux, France) version 2022.06. All of them were run with single precision and 240 CPUs. The baseline simulation was repeated three times to evaluate repeatability and potential scatter of the simulation results.

The HBM was positioned and seated on the driver seat in a vehicle interior representing a modern passenger vehicle in a pre-simulation by applying gravity. The seat represents a conventional vehicle seat with the seatback allowing for an upright seating posture. The seat model was previously validated against the corresponding hardware. The safety system includes a frontal airbag and a three-point belt with pyrotechnical shoulder belt pretensioner and load limiter. The Moving Progressive Deformable Barrier (MPDB) generic pulse from the European New Car Assessment Programme (Euro NCAP) [17] was applied in all the simulations. Additionally, for selected simulations the configuration was repeated with the generic acceleration curve scaled by 50% to evaluate the sensitivity of the spine loading under lower accelerations.

The complete simulation matrix is defined in Appendix A. These modifications include variations in structural seat components (Fig. 1), namely the thickness of the seat pan steel structure and the front seat cross-beam. The seat cross-beam thickness was varied by ±20% and ±40% (simulation IDs 01-04 in Appendix A), whereas the seat pan thickness was varied by ±20% (simulation IDs 05-06 in Appendix A). Additionally, the influence of the compressive stiffness of the seat-cushion foam was evaluated by varying it by ±20% and ±40% (simulation IDs 07 to 10 in Appendix A). Furthermore, different friction coefficients were defined for the contact interaction between the seat and the HBM, ranging from 0.2 to 0.4 in increments of 0.05 (simulation IDs 11-14 in Appendix A). A friction coefficient of 0.3 was used in the baseline simulation.

To investigate the influence of pre-stresses and pre-strains introduced during the positioning procedure on the sensitivity of lumbar spine loading, four configurations were analysed: (1) no pre-stresses, (2) pre-stresses applied to the seat foam only, (3) pre-stresses applied to the seat foam and HBM contact regions (i.e., thorax and leg soft tissues), and (4) pre-stresses applied to the seat foam, HBM contact regions, and the intervertebral discs (IVDs). Configurations 1, 2, and 4 correspond to simulation IDs 15–17 in reverse order in Appendix A, with configuration 3 representing the baseline simulation. Besides the consideration of pre-stresses, the influence of including a 100 ms settling phase with gravity loading prior to the impact pulse was also analysed (simulation 18 in Appendix A). In this latter condition, the skeleton is rigid and fixed in space, with the exception of the spine and pelvis which are allowed to move. In this way, the deformation of the soft tissues due to the gravity was taken into account, similar as in the definition of the HBM certification load cases in CP550.

The baseline simulation includes pre-stresses to the seat foam as well as to the thorax and leg soft tissues of the HBM and does not include any settling phase prior to the impact pulse.

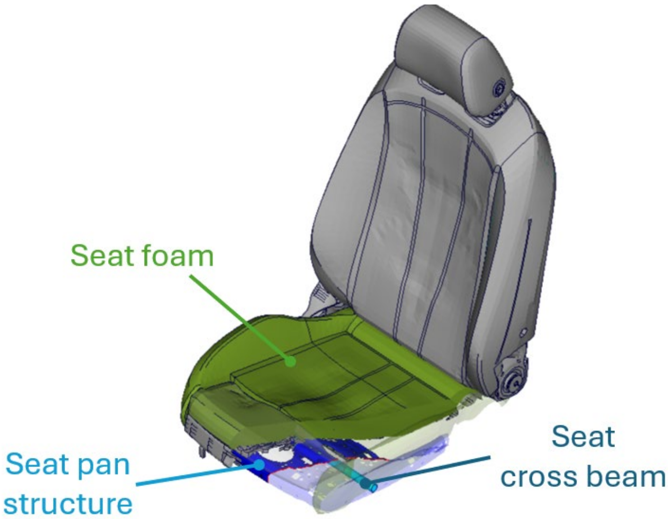

Fig. 1. Seat assembly showing the components that are modified in the sensitivity study: seat foam, seat cross-beam and seat pan structure.

Lastly, the posture and position of the HBM on the seat were varied. The pelvis angle of the baseline posture was rotated by ±2°, ±5°, and ±8° (simulation IDs 19-24 in Appendix A). The baseline pelvis angle in the sagittal plane is 121.5°, defined as the angle between the line connecting the midpoint of the left and right ASIS to the pubic crest and the anterior horizontal [18]. In a pre-simulation step, the skeleton of the seated HBM was defined as a rigid body, with the exception of the lumbar spine and the pelvic bones, and fixed in space. This procedure was used to maintain the position between configurations as consistent as possible while controlling only the pelvic angle. For this purpose, a prescribed sagittal rotation of the pelvis about the acetabular joints was applied according to the abovementioned angles, while only the lumbar spine was allowed to deform to accommodate the pelvic rotation. To account for variations in the HBM seating position, the model was translated to six different locations relative to the baseline configuration. These included ±10 mm and ±20 mm translations in the X-direction (simulation IDs 25-28 in Appendix A), as well as −10 mm and -20 mm translations in the Z-direction (simulation IDs 29-30 in Appendix A).

Compressive forces and bending moments were computed at each vertebral level of the lumbar spine. These metrics were selected for the assessment instead of injury risk in order to enable comparison of the loading across different HBMs. The cross-sections defined at each vertebral body included the vertebral body and the spinous process. For the baseline simulation, which was repeated three times, the average forces and moments from the three runs were used for comparison with the different configurations.

To ensure comparable loading among the different variations, the seat belt forces were compared at the shoulder (B3) and lap belt (B5 and B6). Lastly, the Head Injury Criterion (HIC) and the AIS3+ rib injury risk [19] – since head and thorax are the most frequently injured body regions in frontal crashes – were computed for those configurations not showing any sensitivity of the lumbar spine loading.

III. RESULTS

In the following sub-sections, the results are presented in three categories of parameter modifications: seat characteristics, boundary conditions and HBM posture, followed by a comparison of the sensitivity across all configurations.

Effect of Seat Characteristic Variations

The variation in the thickness of the seat cross-beam of 40% decrease showed a minor influence on the forces and moments, however the -20% to +40% variation did not show almost any effect (Fig. 2). The maximum variation in compressive force with respect to the baseline is 0.14 kN and 4.0 Nm in bending moment.

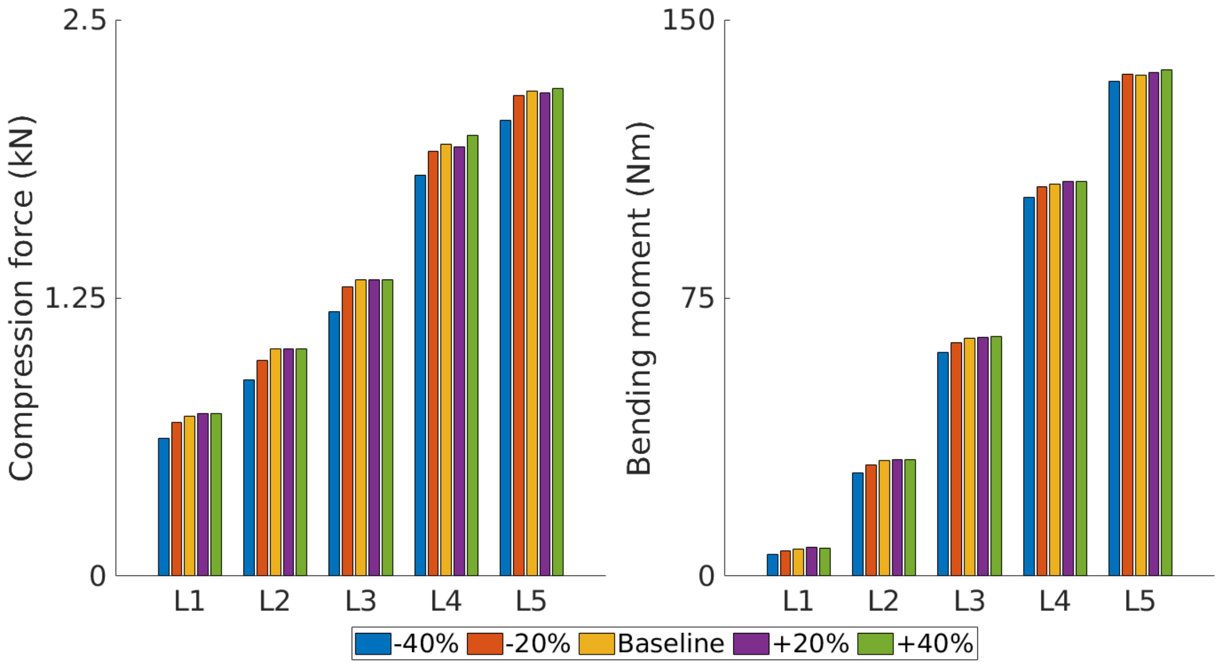

Fig. 2. Lumbar spine forces and moments for modifications to the seat cross-beam thickness.

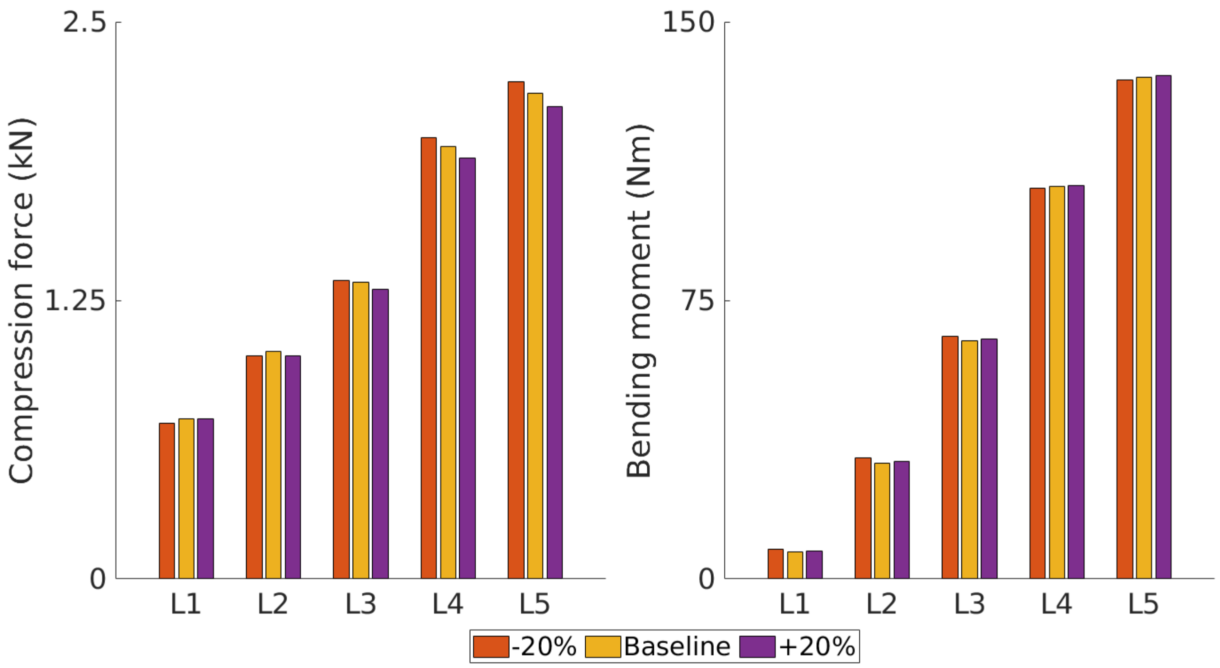

Fig. 3. Lumbar spine forces and moments for modifications to the seat pan structure thickness.

As shown in Figure 3, an increase in the thickness of the seat pan structure by 20% shows a slight decrease in the compressive forces at L4 and L5 level of 0.05 kN, and it has no significant effect on the bending moments. Similarly, a decrease in the thickness by 20% results in a maximum increase of the forces of 0.05 kN at L4 and L5 level and no major influence on the bending moments.

Lastly, Figure 4 shows the effect of the foam stiffness on the loading of the lumbar spine. The variation in the stiffness of ±20% and ±%40 shows almost no influence on the compressive forces of L1 to L3 levels. This variation has a higher influence on L4 and L5, however, the maximum increment in the force with respect to the baseline is 0.04 kN at L5. On the other hand, the results show that with decreasing foam stiffness the bending moment increases.

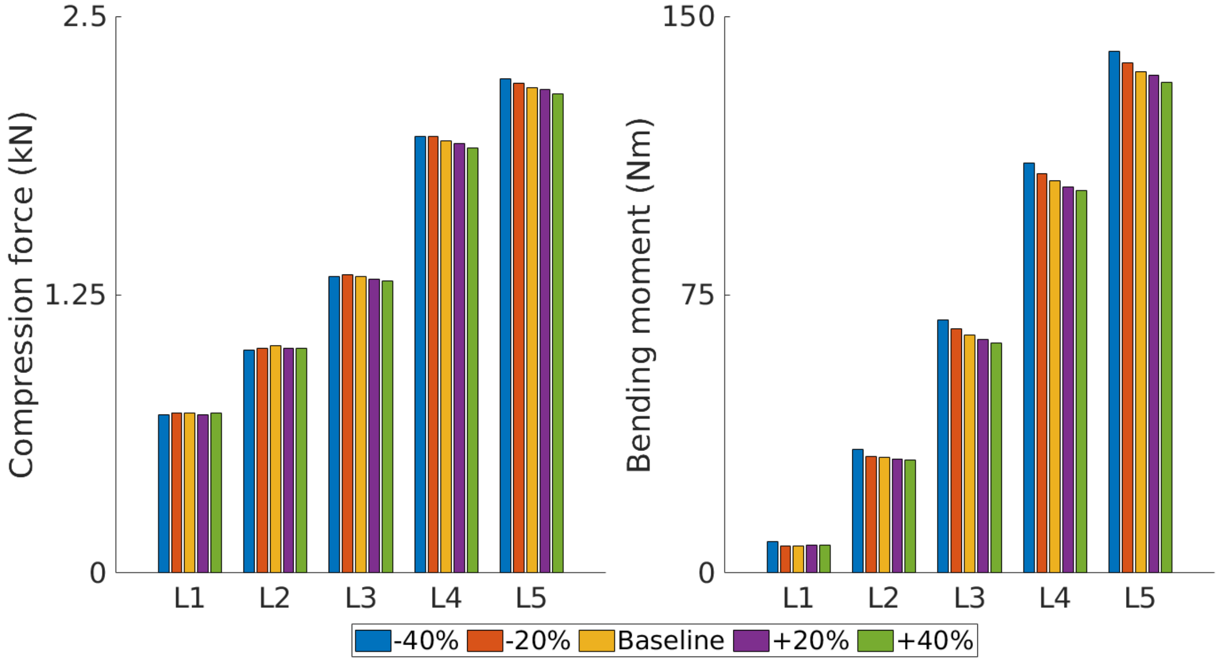

Fig. 4. Lumbar spine forces and moments for modifications to the stiffness of the seat-cushion foam.

Effect of Boundary Conditions

The friction coefficient was varied from 0.2 to 0.4 in increments of 0.05. The results from these variations on the lumbar spine loadings are shown in Figure 5.

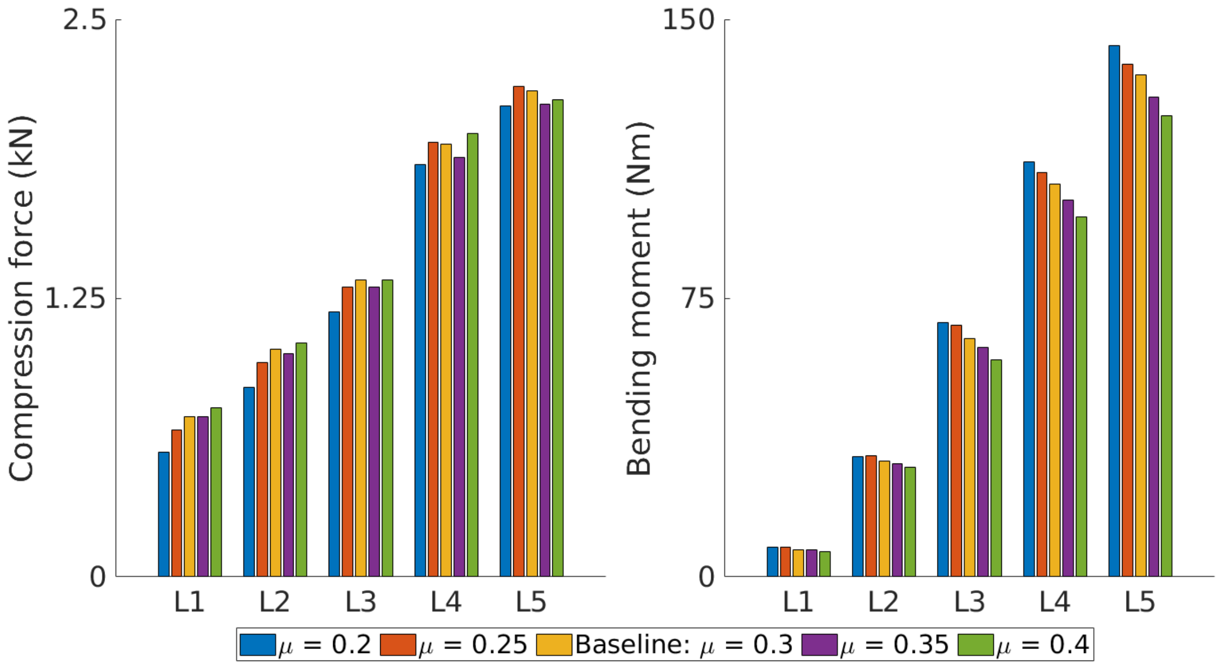

Fig. 5. Lumbar spine forces and moments for variations in the friction coefficient.

Increasing the coefficient of friction shows a tendency to increase the lumbar spine forces at L1 to L3 levels. At L4 and L5 the tendency is less clear. The largest variation in force with respect to the baseline occurs at L2 with a difference of 0.17 kN. In contrast, increasing the friction coefficient decreases the bending moments, especially in the L3 to L5 lumbar segments. The bending moment at L5 varies by 19.1 Nm from μ 0.4 to 0.2.

Furthermore, the baseline model (METRIC 2 in Fig. 6) includes initial pre-stresses from the positioning process in the seat foam and the regions of the HBM in contact with the seat, i.e. the legs and back. The inclusion of pre-stresses in the IVDs of the spine (METRIC 1 in Fig. 6) leads to an increase in the compressive forces and bending moments. The maximum increase in force is at L5 level of 0.04 kN and in bending moment of 3.2 Nm. The configuration with only initial pre-stresses in the foam (METRIC 3 in Fig. 6) shows similar results with respect to the baseline simulation. Finally, the simulation without any pre-stress either in the seat foam or HBM (No METRIC in Fig. 6) results in an increase of the compressive force, with a maximum increase of 0.12 kN with respect to the baseline at L2 level. The bending moments increase for L1 to L3 and remain almost unvaried for L4-L5.

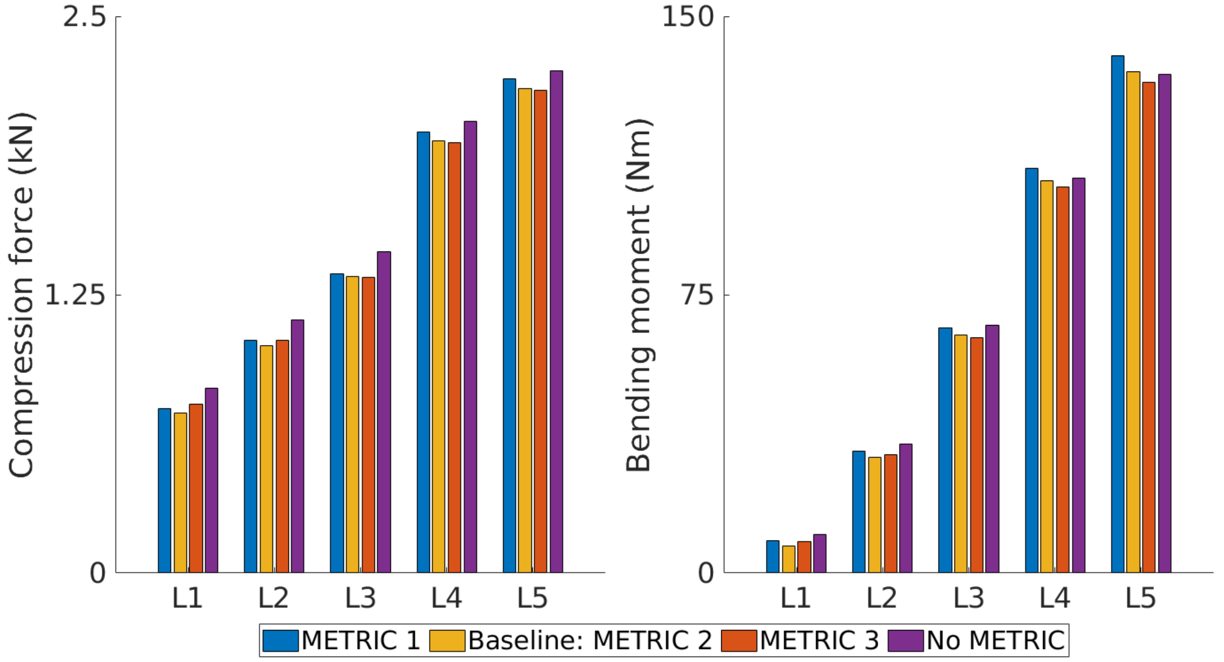

Fig. 6. Lumbar spine forces and moments relative to the baseline simulation for different configurations accounting for pre-stresses from the positioning process.

Figure 7 displays the comparison between the two simulations with and without settling phase prior to the impact pulse. The results show a decrease in the maximum compressive forces when a settling phase is considered. The maximum decrease in the forces was observed at L2 vertebral level of 0.31 kN. The bending moments were comparable between both simulations but for the L5 vertebral body. For this latter, the bending moment increased by 7.5 Nm when including the settling phase.

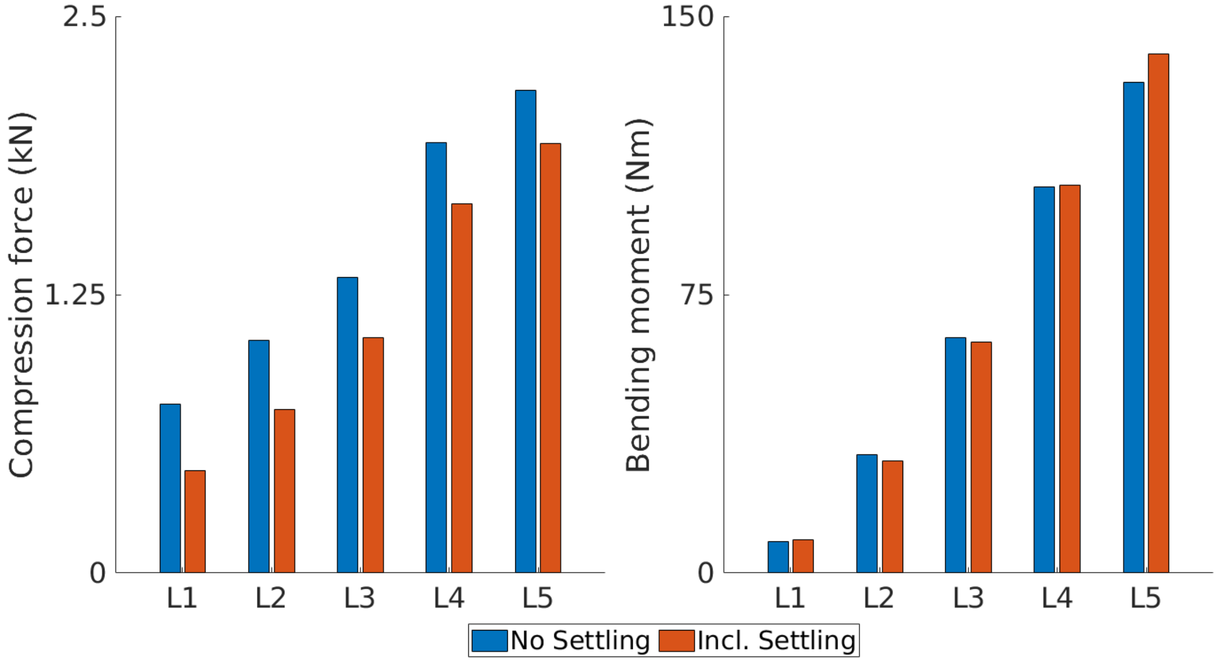

Fig. 7. Lumbar spine forces and moments for a simulation including and a simulation not including a settling phase prior to the pulse.

Effect of Human Body Model Posture

The following three figures show the influence on the lumbar spine loading of the posture of the HBM, including rotation of the pelvis and position on the seat. There is a tendency to higher compressive forces and bending moments with forward rotation of the pelvis – i.e. negative rotation. The largest increase in force of 0.16 kN occurs at L2 level and the largest increase in bending moment of 24.5 Nm occurs at L5 level, both for the posture with a pelvis rotation of -8° (Fig. 8). In contrast, both the compressive forces and bending moments tend to decrease for backward rotation of the pelvis – i.e. positive rotation. The maximum decrease in forces is 0.18 kN at L2 and the maximum decrease in moments is 14.3 Nm at L4, both for the posture with 8° pelvic rotation. The forces and moments remain almost invariant for a rotation of the pelvis of ±2° compared to the baseline simulation.

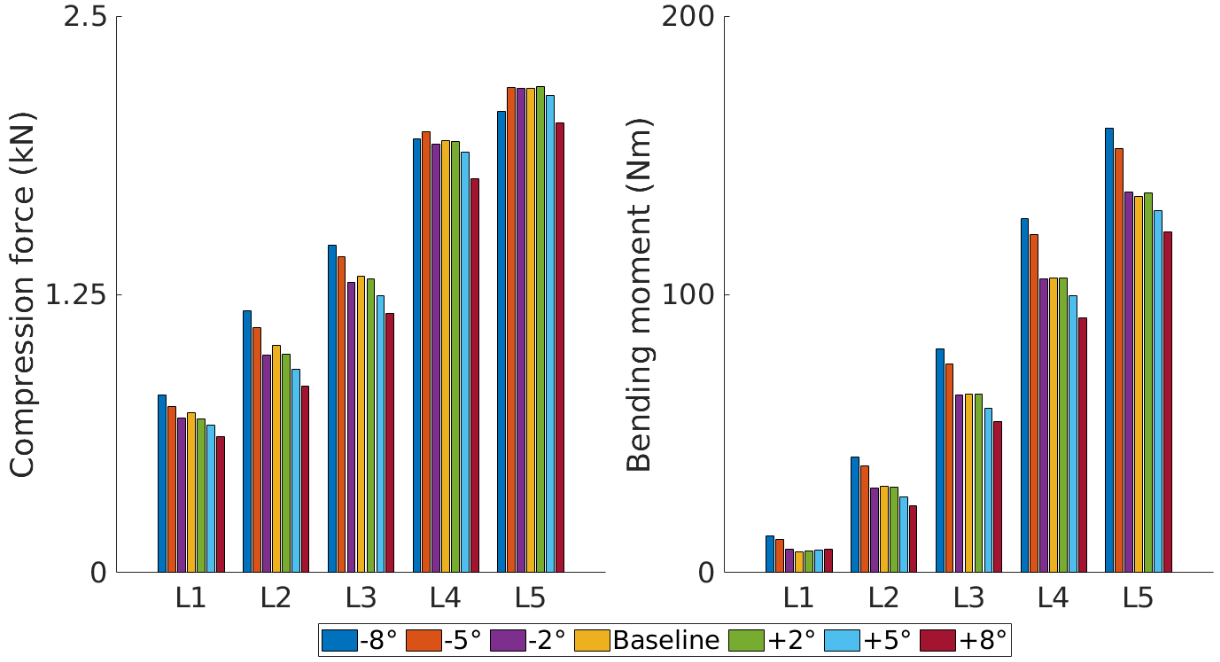

Fig. 8. Lumbar spine forces and moments considering different initial pelvis angles. In the legend: negative rotation indicates forward rotation, and positive rotation indicates backward rotation of the pelvis.

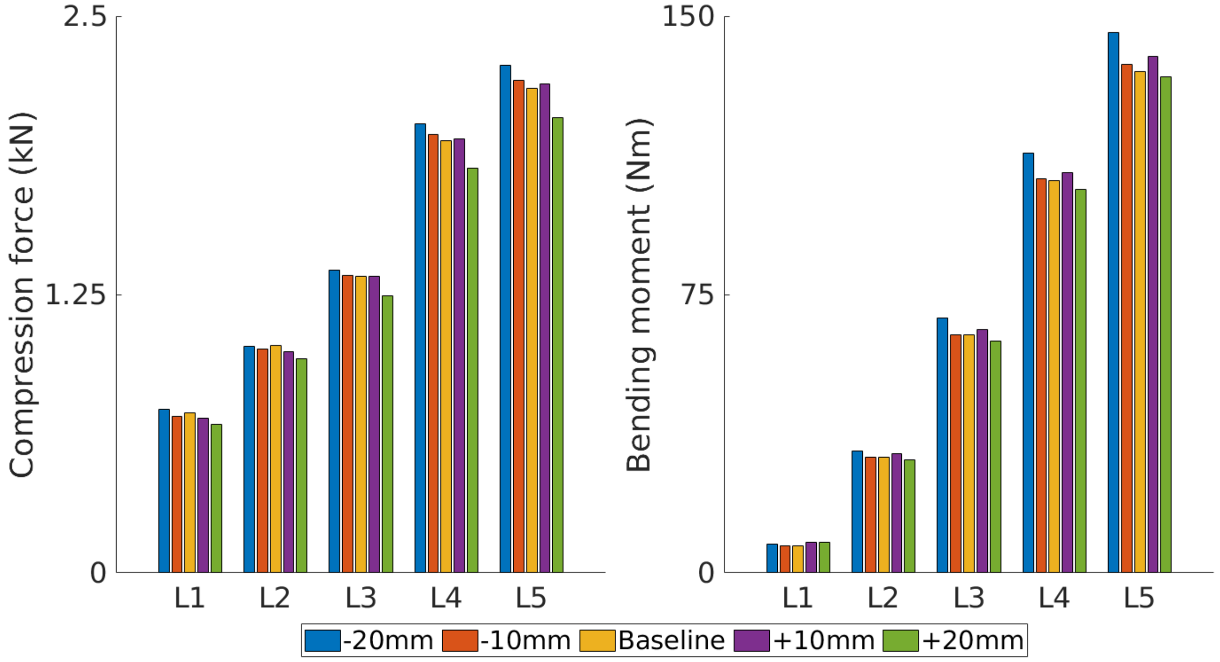

Fig. 9. Lumbar spine forces and moments for different HBM positions along the seat in the X-direction. In the legend: negative values indicate rearward translation of the HBM on the seat, and positive values indicate forward translation.

Figures 9 and 10 show the effect of the position of the HBM on the seat along the X-direction and Z-direction, respectively. There is a tendency for a decrease in both compressive forces and bending moments from the most backward position (-20 mm) to the most forward position (+20 mm) on the seat. However, the bending moments slightly increase for the +10 mm position with respect to the baseline and the +20 mm positions (Fig. 9). The downward position on the seat shows an effect on the compressive forces at all the lumbar spine vertebral levels, while it only has an influence on L4-L5 levels for the bending moment (Fig. 10). There is a decrease in the forces between 0.22 to 0.29 kN for the most downward position compared to the baseline. The bending moment increases for the most downward posture at L4 by 4.0 Nm and at L5 by 11.8 Nm.

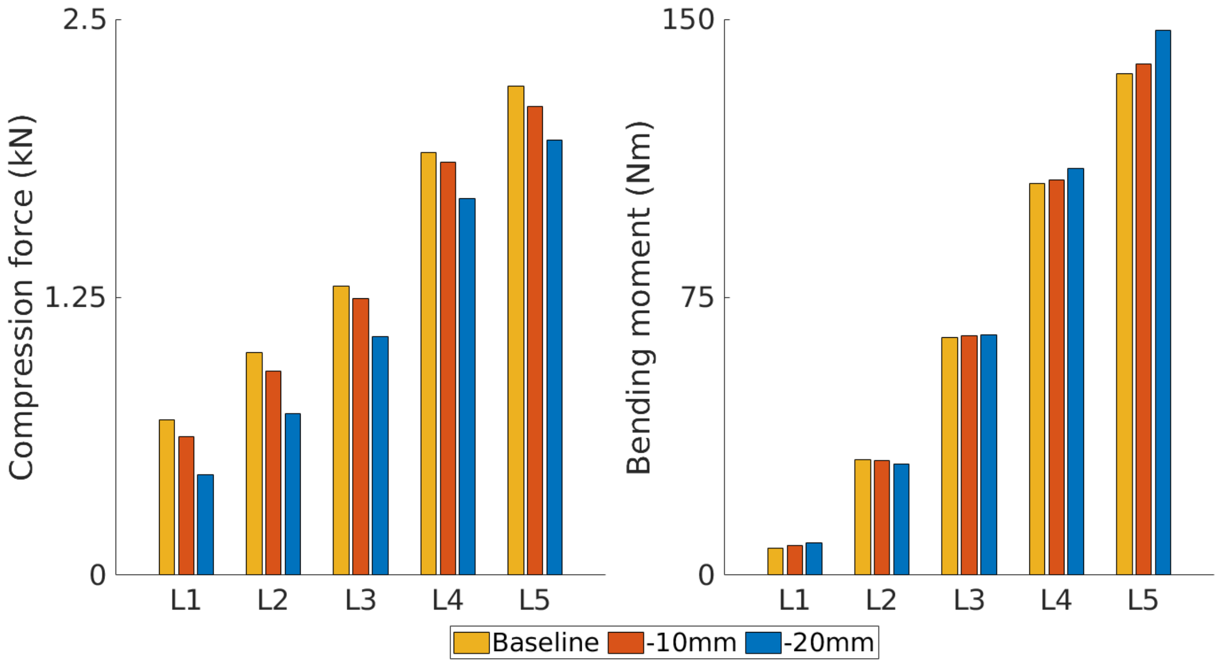

Fig. 10. Lumbar spine forces and moments for different HBM positions on the seat in the Z-direction. In the legend: negative values indicate downward motion of the HBM on the seat.

Sensitivity analysis of all the parameters

As shown in Figures 2 to 10, the compressive forces and bending moments increase from L1 to L5. In Table 1 and Table 2, the forces and moments are reported relative to the baseline simulation and expressed as percentages; these results are hereafter referred to as sensitivity results. Due to the large difference in the load magnitude between vertebral levels, these percentage values should only be compared between configurations at the same vertebral level and not across different levels.

The highest increase in compressive forces compared to the baseline was found in the simulations without pre-stresses and in those with the forward pelvic rotation of 5° and 8°. A decrease in compressive forces was observed in the simulations including a settling phase, as well as in the simulations with lower HBM positions on the seat. Lower sensitivity, while still relevant, was observed for the HBM with a backward pelvic rotation of 8° and for the simulation with a 40% reduction in the seat cross-beam thickness. Lastly, the loading of the spine also showed sensitivity to the friction coefficient, decreasing the lumbar compressive forces for lower friction coefficient values.

Table 1. Sensitivity of the compression forces in the lumbar spine to the different variations relative to the baseline simulation (%)

PARAMETER VARIATIONS		COMPRESSION FORCES (%)				
		L1	L2	L3	L4	L5
Thickness seat cross beam	-40%	86%	86%	89%	93%	94%
	-20%	96%	95%	98%	98%	99%
	+20%	101%	100%	100%	99%	100%
	+40%	101%	100%	100%	102%	101%
Thickness seat pan structure	-20%	97%	98%	101%	102%	102%
	+20%	100%	98%	98%	97%	97%
Stiffness seat cushion foam	-40%	99%	98%	100%	101%	102%
	-20%	100%	99%	101%	101%	101%
	+20%	99%	99%	99%	99%	100%
	+40%	100%	99%	98%	98%	99%
Friction coefficient	0.2	78%	83%	89%	95%	97%
	0.25	92%	94%	98%	101%	101%
	0.35	100%	98%	98%	97%	97%
	0.4	106%	103%	100%	103%	98%
Pre-stresses	IVD+HBM+Seat	103%	102%	101%	102%	102%
	Seat	105%	102%	100%	100%	100%
	None	115%	111%	108%	104%	104%
Settling phase	Yes	61%	70%	80%	86%	89%
Rotation of the pelvis	-8°	111%	115%	111%	100%	95%
	-5°	104%	108%	107%	102%	100%
	-2°	97%	96%	98%	99%	100%
	+2°	96%	96%	99%	100%	100%
	+5°	92%	90%	93%	97%	99%
	+8°	85%	82%	88%	91%	93%
HBM Position X-direction	-20 mm	102%	100%	102%	104%	105%
	-10 mm	98%	99%	101%	101%	102%
	+10 mm	97%	97%	100%	100%	101%
	+20 mm	93%	94%	94%	94%	94%
HBM Position Z-direction	-10 mm	87%	90%	93%	96%	97%
	- 20 mm	63%	71%	81%	87%	90%

The bending moment measurements showed a strong sensitivity to the pelvis angle. The highest and lowest values were obtained for forward and backward rotation of the pelvis by 5° and 8°, respectively (Table 2). In addition, the reduction of 40% thickness of the seat cross-beam significantly reduced the bending moment, especially at L1 and L2, compared to the other configurations.

Similarly to the analysis of the forces, not including any pre-stresses significantly increased the bending moments at L1 and L2 levels. Lastly, the bending moments also showed sensitivity to modifications to the coefficient of friction, increasing the loading with a reduction in this parameter.

Table 2. Sensitivity of the bending moments in the lumbar spine to the different variations relative to the baseline simulation (%)

PARAMETER VARIATIONS		BENDING MOMENTS (%)				
		L1	L2	L3	L4	L5
Thickness seat cross beam	-40%	78%	89%	94%	97%	99%
	-20%	92%	96%	98%	99%	100%
	+20%	103%	101%	100%	101%	101%
	+40%	101%	101%	101%	101%	101%
Thickness seat pan structure	-20%	109%	105%	102%	99%	99%
	+20%	102%	102%	101%	100%	100%
Stiffness seat cushion foam	-40%	114%	108%	106%	104%	104%
	-20%	99%	101%	103%	102%	102%
	+20%	100%	99%	98%	98%	99%
	+40%	103%	98%	97%	97%	98%
Friction coefficient	0.2	107%	104%	107%	106%	106%
	0.25	110%	105%	105%	103%	102%
	0.35	98%	98%	96%	96%	95%
	0.4	92%	95%	91%	92%	92%
Pre-stresses	IVD+HBM+Seat	118%	106%	103%	103%	103%
	Seat	114%	102%	99%	98%	98%
	None	140%	112%	104%	101%	99%
Settling phase	Yes	106%	95%	98%	100%	106%
Rotation of the pelvis	-8°	181%	134%	125%	120%	118%
	-5°	162%	124%	117%	115%	113%
	-2°	114%	98%	99%	100%	101%
	+2°	107%	99%	100%	100%	101%
	+5°	112%	88%	92%	94%	96%
	+8°	114%	77%	85%	87%	91%
HBM Position X-direction	-20 mm	105%	105%	107%	107%	108%
	-10 mm	100%	100%	100%	100%	101%
	+10 mm	110%	103%	102%	102%	103%
	+20 mm	111%	98%	97%	98%	99%
HBM Position Z-direction	-10 mm	110%	100%	101%	101%	102%
	- 20 mm	117%	97%	101%	104%	109%

The results from the repetition of the baseline simulation, which was run three times with identical settings, showed some scatter in lumbar spine loading. The largest deviations were found for the compressive forces at L2 and the bending moments at L1, with deviations of up to 3%.

The shoulder belt forces, measured at B3, were repeatable across all the configurations, with maximum deviation of 1%. The lap belt forces were also repeatable across the configurations, however in this case, there were some exceptions. There were deviations up to 8% due to the changes in the posture and seating position, as well as due to the variations in the friction coefficient.

Lastly, HIC and the AIS3+ rib injury risk were analysed with focus on the configurations where the lumbar spine did show low sensitivity in the loading. HIC values were comparable across all configurations, ranging from 222 to 293, which are well below the injurious threshold. The AIS3+ rib injury risk did, however, vary in some configurations. Modifications to the selected seat characteristics had only a minor effect on the rib injury risk, with a maximum deviation of 4% across the corresponding configurations. This deviation is within the scatter of the rib injury risk observed in the three repetitions of the baseline simulation and is therefore not considered relevant. In contrast, the position of the HBM along the X-direction on the seat influenced the AIS3+ rib injury risk. Although this parameter was not among the configurations with the largest influence on lumbar spine loading, a more forward position of the HBM resulted in an increased rib injury risk.

IV. DISCUSSION

The present study aimed to quantify the sensitivity of lumbar spine loading in an HBM. To this end, a series of simulations was performed in which seat structural properties, contact parameters, HBM posture, and initial positioning on the seat were systematically varied. The resulting lumbar spine forces and moments were analysed to evaluate the relative influence of these modelling assumptions. The results obtained were clustered into three categories of parameter modifications: the seat characteristics, the boundary conditions and the HBM posture.

With respect to the modifications concerning the seat characteristics, it turned out that variations of ±20% in the thickness of the seat pan structure and seat cross-beam or in the stiffness of the seat-cushion foam resulted in less than ±5% deviation for the compressive forces and a maximum of ±9% for bending moments compared to the baseline of the respective vertebral body. The largest influence, as far as the seat structure modifications are concerned, was found for the 40% thickness reduction of the seat cross-beam for the bending moment at L1 of 22%.

A larger influence compared to the modifications of the seat structure was found for the variation of the boundary conditions. In this category the most significant influence is observed in the configuration including a settling phase of the HBM compared to the configuration where it is not. While bending moment variations are within ±6% only, over all lumbar vertebrae, the compressive forces decreased by up to 30% for L2, representing one of the most substantial absolute deviations in this study. Only the lowering of the HBM by 20 mm in Z-direction resulted in similar large deviations for the compressive forces in the lumbar vertebrae. Interestingly, after the settling phase, the HBM moves approximately 15 mm downward, thus the pelvis position of both models is comparable. Nonetheless, the curvature of the spine differs, which is assumed to be one reason for the difference in the bending moments. In the virtual testing protocol CP551 applying HBMs [5], a maximum tolerance in the initial Z-position of the H-point is defined as ±13.7 mm. Thus, solely the 10 mm downward movement variation falls within the defined tolerance [5]. In this configuration, the results show a maximum reduction of 13% in the compressive force at L1 and a maximum increase of 10% in the bending moment at L1. In terms of absolute values, the maximum compressive force reduction is 0.1 kN at L2 and the maximum increase in bending moment is 2.7 Nm at L5.

Besides this, the variation of the friction coefficient showed a significant influence. While variations of ±0.05 from the baseline friction coefficient of 0.3 resulted in compressive forces and bending moments within ±10%, the largest variations of ±0.1 led to a more significant reduction in compressive forces, down by 22% at L1.

Finally, variations in seating posture and the pre-stresses resulting from the positioning process led to considerable deviations in lumbar spine loading. The greatest deviation in the bending moments among all the configurations was found for the ±8° and ±5° rotations of the pelvis. The maximum increase was 81% at L1 for the forward rotation of the pelvis and the maximum decrease occurred at L2 with a reduction of 23% for the backward rotation. The compressive forces showed a similar pattern for the rotation of the pelvis, increasing up to 15% for the forward rotation and decreasing by 18% for the backward rotation. It is worth mentioning that submarining was not detected in any simulation of the study. Lumbar spine loading did not show a strong sensitivity to the position of the HBM along the X-axis. However, bending moments tended to increase in the most rearward position, most likely due to greater flexion of the lumbar spine resulting from the increased distance to the airbag. The position variations along the X-direction on the seat fall within the H-point tolerance in [5] of ±27.6 mm.

The sensitivity of the lumbar spine loading has been previously investigated, however mostly focused on the modelling of the lumbar spine itself, including modifications to the geometry or material properties [13-14]. To the author's knowledge, there is no thorough investigation on which external factors influence the loading of the lumbar spine, besides an increase in the seatback angle and applying a semi-rigid seat [20]. In addition, Tang *et al.* [15] investigated the influence of different design parameters on the spine loading using a HIII crash dummy. In contrast to the modification of seat design features reported in [15], such as geometric variations of an anti-submarining bar, the present study instead examines the influence of seat model stiffness variations. Furthermore, the objective of this study is not to optimise the structural seat components to reduce the lumbar spine loading in the HBM, but to evaluate the sensitivity of lumbar spine loading to smaller (±20%) and larger (±40%) changes in seat component stiffness. Finally, this work - in the framework of the IMPROVA project - will provide a valuable contribution to the definition of the required level of seat model validation for occupant safety analysis and virtual testing by identifying the parameters that most strongly influence the potential injury outcome. Ultimately, a higher level of modelling detail and accuracy would be required for the components most relevant to occupant loading, while less influencing components, such as the seat foam stiffness according to this study, may require a lower level of validation effort.

These aspects become even more important when future driving scenarios are considered. The introduction of advanced driver assistance systems, and subsequently, autonomous vehicles is expected to motivate drivers to adopt more reclined postures compared to today's upright seating postures. In the event of a crash, it is anticipated that the loads acting on the spine will be higher compared to those experienced in current upright seating postures [8]. These evaluations might need even higher requirements in terms of seat model validation quality. Nevertheless, it is important to note that the findings of this study should not be extrapolated to such altered conditions without further examination. Additionally, a specific seat model was applied, differences in seat structure design or level of modelling detail of the seat – e.g. generic design - could lead to different results. The present study did not consider any variation to represent reclined postures, as the seat design for such purposes is expected to differ beyond a simple rotation of the seatback. The associated restraint systems would also likely change, for example through the use of integrated seat belts, which could alter the loading of the lumbar spine. Therefore, such configurations were excluded in the simulation matrix.

One limitation of the present sensitivity study is that it is based on results derived from a single HBM, namely the 50th percentile male THUMS 4.1 VW Group in VPS. Results and conclusions drawn from this analysis might vary when applying other anthropometries or HBMs. The sensitivity observed in the variation of the posture supports this assumption, given that other HBMs are derived from different individuals, thus with different initial spinal alignments and pelvic angles. In addition, different anthropometries may result in different seat deformation modes. Variations in body mass index (BMI) may alter the regions of the seat that are primarily loaded, eventually changing the parameters most relevant for occupant loading and injury assessment. Furthermore, only a single frontal load case is considered in this study. However, in order to avoid the potential oversight of dependencies on the pulse, its amplitude was reduced by 50% and the corresponding results were analysed. The results obtained and reported in the present study have been confirmed when using the reduced pulse.

The results are presented as absolute values and as percentages relative to the baseline simulation. Although an injury risk curve for the lumbar spine based on experiments has previously been developed by Tushak *et al.* (2023) [21], it has been demonstrated for other body regions that injury risk curves should be adjusted to be model-specific [1]. Hence, the relevance of the results with respect to injury risk probability was not estimated due to the absence of an appropriate injury risk curve specific to the THUMS AM50 v4.1 VW Group model in VPS. Future work should therefore assess the most relevant parameters based on injury risk outcomes. Nonetheless, this investigation highlights the parameters to which lumbar spine loading shows the greatest sensitivity, and the largest variations may indicate the variables most relevant for injury risk. Additionally, some variations of the parameters showed a sensitivity within the scatter observed in the repeated baseline simulation (< 3%), hence these parameters are considered as irrelevant in the context of this investigation.

V. CONCLUSION

The findings of this study provide an insight into the sensitivity of the lumbar spine loadings by the variation of seat model characteristics, HBM to seat interactions, HBM posture and initial boundary conditions. The results obtained are intended to contribute to the basis for future discussions on their implications on injury risk assessment in the lumbar spine using HBMs and for prioritising the most relevant parameters to be controlled within virtual testing qualification processes.

Overall, the results suggest that variations in occupant posture and initial boundary conditions have a greater influence on lumbar spine loading than moderate modifications to the structural stiffness of the seat model. This highlights the importance of carefully defining the initial occupant position when performing virtual safety assessments.

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VII. REFERENCES

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VIII. APPENDIX A

Table A1. Simulation matrix with parameter modifications relative to the baseline simulation

Simulation ID	Pre-stresses from positioning			Settling phase	Seat charachteristics			Friction	HBM Posture		
	Seat	HBM Soft tissue	HBM IVD		Thickness of front cross beam	Thickness of steel seat pan stucture	Compressive stiffness cushion foam		Pelvis angle	X-Position	Z-Position
Baseline	X	X		No	Base	Base	Base	0.3	Base	Base	Base
01	X	X		No	+40%	Base	Base	0.3	Base	Base	Base
02	X	X		No	+20%	Base	Base	0.3	Base	Base	Base
03	X	X		No	-20%	Base	Base	0.3	Base	Base	Base
04	X	X		No	-40%	Base	Base	0.3	Base	Base	Base
05	X	X		No	Base	+20%	Base	0.3	Base	Base	Base
06	X	X		No	Base	-20%	Base	0.3	Base	Base	Base
07	X	X		No	Base	Base	+40%	0.3	Base	Base	Base
08	X	X		No	Base	Base	+20%	0.3	Base	Base	Base
09	X	X		No	Base	Base	-20%	0.3	Base	Base	Base
10	X	X		No	Base	Base	-40%	0.3	Base	Base	Base
11	X	X		No	Base	Base	Base	0.2	Base	Base	Base
12	X	X		No	Base	Base	Base	0.25	Base	Base	Base
13	X	X		No	Base	Base	Base	0.35	Base	Base	Base
14	X	X		No	Base	Base	Base	0.4	Base	Base	Base
15	X	X	X	No	Base	Base	Base	0.3	Base	Base	Base
16	X			No	Base	Base	Base	0.3	Base	Base	Base
17				No	Base	Base	Base	0.3	Base	Base	Base
18	X			Yes	Base	Base	Base	0.3	Base	Base	Base
19	X	X		No	Base	Base	Base	0.3	+8°	Base	Base
20	X	X		No	Base	Base	Base	0.3	+5°	Base	Base
21	X	X		No	Base	Base	Base	0.3	+2°	Base	Base
22	X	X		No	Base	Base	Base	0.3	-2°	Base	Base
23	X	X		No	Base	Base	Base	0.3	-5°	Base	Base
24	X	X		No	Base	Base	Base	0.3	-8°	Base	Base
25	X	X		No	Base	Base	Base	0.3	Base	+20 mm	Base
26	X	X		No	Base	Base	Base	0.3	Base	+10 mm	Base
27	X	X		No	Base	Base	Base	0.3	Base	-10 mm	Base
28	X	X		No	Base	Base	Base	0.3	Base	-20 mm	Base
29	X	X		No	Base	Base	Base	0.3	Base	Base	-10 mm
30	X	X		No	Base	Base	Base	0.3	Base	Base	-20 mm