Sensitivity of Human Body Model Response Relative to the Lumbar Spine and Pelvic Tissue Formulation

B.D. Gepner, J. Toczyski, K. Rawska, D. Moreau, J.R. Kerrigan

Abstract

With the introduction of Level 3 Autonomous Driving Systems, an ability to maintain the appropriate level of protection for reclined occupants will become highly relevant. Virtual tools, such as human body models offer an attractive tool to support the rapid shift in vehicle development and restraint system design. However, previous research has shown that, due to differences in lumbar spine and pelvis flesh formulations, the available human body models, while valid in upright scenarios, differ in their responses in recline postures. The goal of this study is to use an updated lumbar spine and different pelvis flesh formulations to evaluate how these changes affect the response of reclined occupants in frontal crashes. An updated simplified lumbar spine was developed based on recently released GHBMC update. The updated simplified lumbar spine model was imported into previously utilised GHBMC simplified and detailed models, and an alternative pelvis flesh attachment was developed creating a total of four modified human body models. These models were evaluated in the sled environment representative of recent post-mortem human subject recline test series. The pelvis-flesh attachment model had a substantial effect on the submarining outcome across all evaluated models. With the unified updated simplified lumbar spine model both human body models showed differences in kinematic response and recorded lumbar forces.

Keywords: HBM, PMHS, recline, restraint, submarining,

I. INTRODUCTION

The introduction of advanced driver assistance systems (ADAS) is changing the nature of personal transportation. These systems, when used as intended, have the capability to improve vehicle safety and reduce the number and severity of vehicle crashes [1-3]. It is expected that in the near future, ADAS functionality will allow occupants to ride without constantly interfacing with the vehicle's controls, effectively resulting in a Level 3 automated driving system (ADS) [4-6]. With increased automation, the vehicle manufacturers will be enticed to offer new interior configurations, along with new seating arrangement. The passengers may choose to rest during extended travel by reclining their seats and moving it away from the instrument/entertainment panel [7-8]. However, these seating choices may challenge the ability of current vehicle safety systems to adequately protect the occupants in case the vehicle crashes.

Previous computational studies have suggested that the current restraint systems (3 point belt with B-pillar mounted D-ring), may no longer be adequate for the application in the ADS environment [9 - 11]. In [9] authors have shown that, for reclined occupants, belt with B-pillar mounted D-ring leads to a delayed torso engagement and increase in forward excursion of the pelvis. This was mitigated by the use of a belt system with a seatback integrated D-ring. However, several studies have shown that the increase in the seatback recline angle leads to an increased risk of submarining, i.e., the lap belt loads the abdominal soft tissues, after passing over the iliac crest of the pelvis without engaging, or after disengaging, the pelvis. This effect was observed for systems with both seatback integrated belts and lap belt pre-tensioning [12-14]. Early knee engagement, through the use of knee bolster or knee airbags was shown to be an effective countermeasure for preventing submarining [12 - 13][15], however these systems may no longer be present in the ADS interior. This shows that there is a need for a new restraint system that could effectively protect occupants in the ADS relevant environment.

Traditionally, restraint systems are developed using anthropomorphic test devices (ATDs), or crash test

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dummies, designed for specific loading direction, i.e., front, side, oblique. However, no ATD was developed for evaluation of occupant-restraint interaction in the reclined environment. Human body models (HBMs), such as the Total Human Model for Safety (THUMS) developed by Toyota or the Global Human Body Model Consortium (GHBMC), offer an attractive tool to support the design and evaluation process. HBMs have a potential to provide an omnidirectional response applicable for a variety of loading conditions. However, similar to the ATDs, HBMs have limitations and remain relevant only within their validation regime. Currently no HBMs has been validated for the recline scenario given the lack of published and relevant post-mortem human subject (PMHS) reference data.

In a recent study [16], the authors compared the performance of available HBMs outside their validation regime in the frontal, reclined scenario, utilising a new prototype restraint system. Four different 50th percentile male HBMs were compared, the GHBMC simplified occupant model (GHBMC-S) version 1.8.4, the GHBMC detailed model (GHBMC-D) version 4.5, THUMS SAFER model version 9 (THUMS-S), and THUMS version 5 model (THUMS-V5) in a sled environment matching the one used in the recent reclined PMHS test series [17]. The simulation outcome varied between specific models including belt placement, pelvis engagement/submarining, and pelvis and lumbar spine kinematics. The authors identified that the lumbar spine and pelvis flesh model definitions were most likely the source of the observed discrepancies. The GHBMC models utilized a simplified discrete formulation using zero length six degree of freedom (6DOF) beams with either linear (GHBMC-S) or bilinear (GHBMC-D) moment/angle, force/displacement definitions. THUMS models, on the other hand, utilized an anatomical lumbar spine with explicitly modeled anatomical features such as ligaments and intervertebral discs. Additionally, the GHBMC-D, which submarined, was the only model which used a continuous pelvis flesh attachment model, whereas all other models featured a sliding contact pelvis-flesh definition.

New lumbar spine definition became available in the recently released GHBMC-D version 5, following the modelling principals used in the GHBMC cervical spine which were described previously in [18-19]. New definition features a ligamentous, contact based, anatomical lumbar spine (ALS) (Fig. 1ab). The new formulation addressed several limitations of a discrete, simplified lumbar spine (SLS) model used previously in the GHBMC-S and GHBMC-D models. It replaced zero length 6DOF beams with appropriate anatomical structures, included definition of deformable vertebrae, ligamentous structures and was validated against several PMHS studies available in the literature [20-26]. The new ALS model is likely to improve the HBM response in recline scenario, however, no such formulation is currently available for the simplified GHBMC lumbar spine model.

The goal of this study was to evaluate the effect of lumbar spine and pelvis flesh formulation onto the HBM response in the recline scenario. To maintain the comparability with results of the previous study [16], this study proposes to leverage the extensive validation of the newly available ALS model, by replicating its performance in the previously utilized GHBMC models. Thus, the first specific goal was to use the new ALS model to develop a new updated SLS model definition that could be used with the previously utilised HBMs. The second goal of this study was to evaluate both GHBMC-S and GHBMC-D model response sensitivity to the lumbar spine, and pelvis flesh definition in the frontal, reclined scenario, matching a previous modelling study and recent PMHS test series. The information from this study may be used inform the future modelling approach, and component bio-fidelity/validation experiment design.

II. METHODS

The first goal of this study was to develop an updated simplified lumbar spine model (USLS) that could utilise the discrete SLS formulation, but at the same time match the ALS model performance. Once complete the obsolete SLS model was replaced by the USLS definition and updated GHBMC models were exercised in the sled environment matching the recent PMHS test series.

Lumbar Spine Model Development

First, T12-S1 sections of the ALS and SLS models were extracted from their respective HBMs. The ALS model was extracted from the GHBMC-D v5.1.1, and two SLS models were extracted from GHBMC-D v4.5 and GHBMC-S v1.8.4., respectively. In order to develop the USLS model the ALS model was subdivided into a three vertebrae segments. Zero length 6DOF beams, with zero stiffness, were copied into the three vertebrae models and attached to respective vertebral bodies through *CONSTRAINED_INTERPOLATE keyword. These beams were used as sensors, allowing for measurement of relative vertebra displacement in the local 6DOF beam coordinate

system. In all segments the bottom vertebral body was constrained in space (*BOUNDARY_SPC_SET on the bottom endplate), while the top vertebra was loaded through *BOUNDARY_PRESCRIBED_MOTION in all nine unique directions (+x, -x, ±y, +z, -z, ±rx, +ry, -ry, ±rz) (Fig. 1c). All segments were subjected to linear deformation input within the expected deformation range applicable to the reclined occupant environment (±10mm translation/±20 deg rotation, Appendix A). The boundary force along with the recorded 6DOF beam displacement was used to construct the set of new performance curves to be used as an input for the 6DOF beam definition in the USLS model. The performance curves developed through this effort are shown in Appendix B. Next, all T12-S1 spine models were exercised in all unique directions to compare and evaluate the performance of each model. For this effort the sacrum was fixed in space (*BOUNDARY_SPC_SET on the entire sacrum) and the *LOAD (linear load 0-500N and 0-40Nm, Appendix A) keyword was applied to T12 vertebra (Fig. 1b).

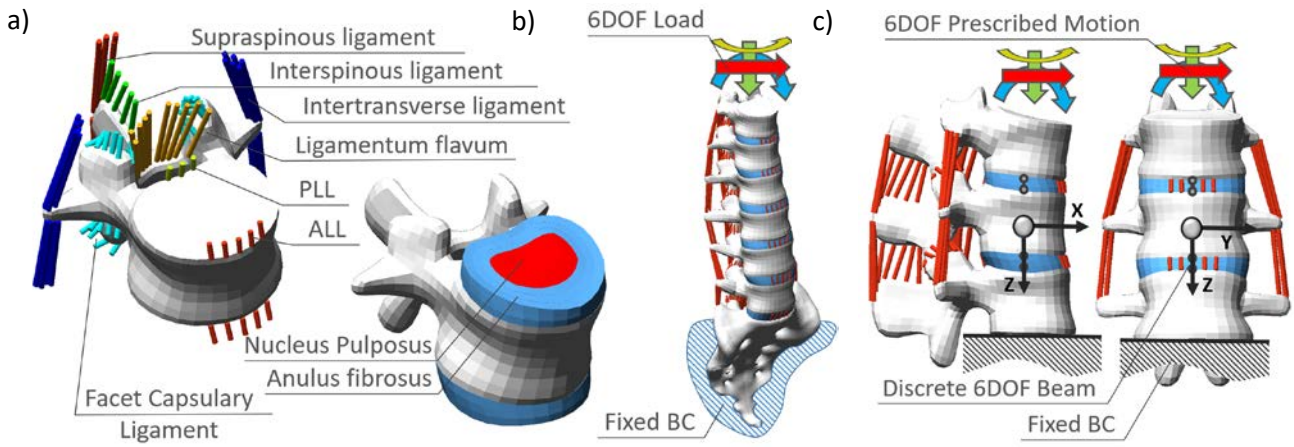

Fig. 1. a) GHBMC v5 ALS model, b) whole spine 6DOF evaluation setup, c) Three vertebrae test segments used for development of USLS performance curves.

HBM Sled Simulations

The boundary conditions as well as occupant positioning used in this study were matched with the previously published research, and will be only briefly summarised. For details please refer to the original publication [16]. The original simulation environment was developed following the setup used in the recently completed reclined, full body, PMHS test series [17], which a used semi-rigid simplified seat, a 50 deg. torso recline angle, a 50 km/h full frontal rigid barrier pulse and a prototype seatback integrated 3-point restraint system. The semi-rigid seat was based on a design developed by [27] and previously used in several PMHS studies [27-28]. The stiffness of the springs was set to represent a front seat configuration. The Finite Element (FE) model of the semi-rigid seat built by LAB/CEESAR in cooperation with PDB was further improved by Autoliv Research (Fig. 2).

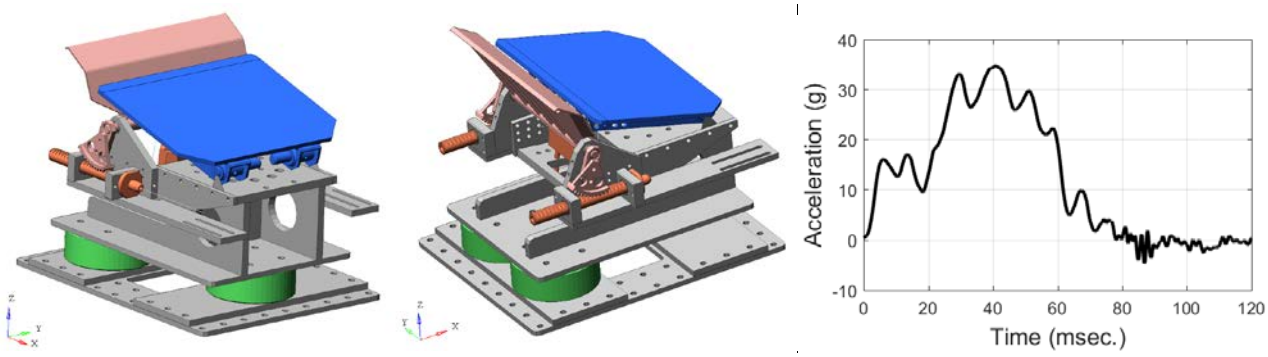

Fig. 2. FE model of LAB Seat developed by LAB/CEESAR in cooperation with PDB (left). Full frontal rigid barrier pulse 50 km/h (right) [16].

The belt system used in this study consisted of a 3-point belt equipped with dual lap-belt pretensioners and a shoulder-belt retractor pretensioner. The belt system included a crash locking tongue and a shoulder-belt load limiter of 3.5 kN. The FE model of the restraint system, built by Autoliv Research, was developed using shell seatbelt element and 2D slip-rings formulation to facilitate stable and unobstructed belt payout.

HBMs

Two 50th percentile male HBMs, the GHBMC-S version 1.8.4, the GHBMC-D version 4.5, were updated with the newly developed USLS model and utilised in this study. Since the GHBMC-S and GHBMC-D models utilised the zero length 6DOF modeling approach the USLS model was implemented by updating the old performance functions in the existing models. Additionally, the pelvis flesh attachment formulation was modified for each model to evaluate its effect on submarining outcome. For GHBMC-D model the flesh, including subcutaneous adipose tissue as well as surrounding muscle tissue, was detached from the pelvis. This was done by releasing node connectivity for any flesh/muscle part sharing nodes with the pelvis and deleting *CONTACT_TIED definitions attaching muscles to the pelvis. Following the modeling approach of GHBMC-S a *CONTACT_SURFACE_TO_SURFACE with zero friction coefficient was added to facilitate sliding boundary condition (GHBMC-DF). For GHBMC-S the flesh parts (GHBMC-S lacks separate parts for muscles) was attached to the pelvis through nodal constraint (*CONSTRAINED_EXTRA_NODES) (GHBMC-SF). Both GHBMC models were originally modified from their release configuration by, modifying contacts, and changes in hourglass definition to improve model robustness in severe loading conditions. For details please see [14]

Positioning

To facilitate the direct comparison with previously published research this study used occupant and belt positions from [16]. Initially no detailed occupant positioning information was available from [17] and all HBMs were adjusted to match the approximated PMHS positions. Since the 50th male GHBMC models share specific anthropometry the exact match between positions was achieved (Fig. 3). For details on positioning please see [16].

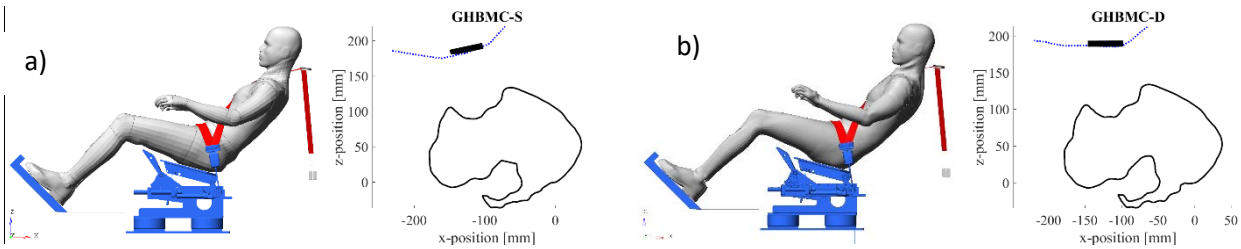

Fig. 3. Initial HBM position and belt position at the centreline relative to the occupant's pelvis. The outline of occupant flesh shown. a) GHBMC-S, b) GHBMC-D.

III. RESULTS

USLS Model Evaluation

Comparing the load applied to T12 and its resultant displacement, the newly developed USLS model matched the response of the detailed ALS model. The best agreement between the models was shown for anterior shear and flexion loads, and the worst for compression and torsion. Both SLS models differed substantially from each other and from both ALS and USLS models (Fig. 4).

HBM Simulation Outcome

All four HBM simulations were scheduled for 120 msec. runtime, however GHBMC-DF terminated prematurely at 73msec. Nevertheless, these results provided satisfactory basis for the model performance comparison.

Kinematic

Both the GHBMC-S and GHBMCC-SF showed similar kinematics throughout the simulation (Fig. 5ab). For both models the lap belt remained engaged with the pelvis during the loading phase, however during the unloading phase (90 msec.) in the GHBMC-SF the belt moved up and partially slipped over the anterior superior iliac spine (ASIS) and into the abdomen (Fig. 5b). The GHBMC-D differed from all other models. In this case the model showed pelvis submarining under the lap belt at 61msec, increased forward pelvis motion, pelvis engagement with the anti-submarining ramp, and limited upper extremity motion (Fig. 4c). The GHBMC-DF terminated prematurely at 73msec., however at that time the pelvis was engaged with the belt, distinguishing it from the GHBMC-D model.

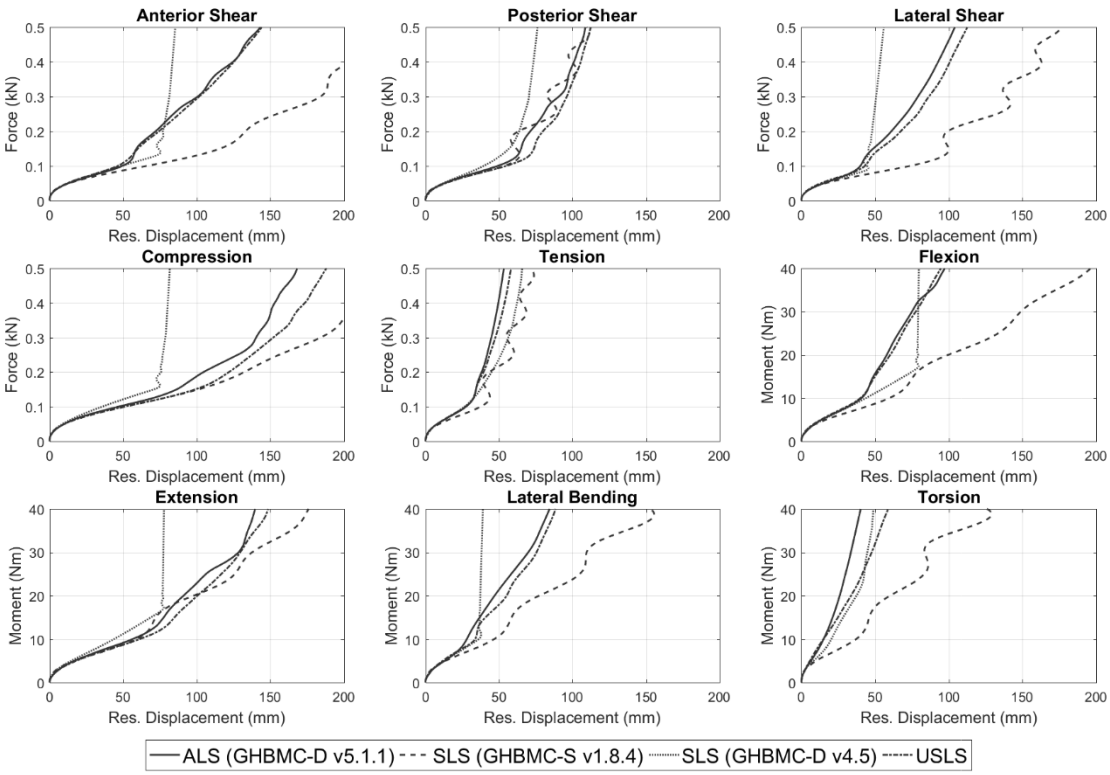

Fig. 4. T12 load vs. resultant displacement for all T12-S1 models for ALS, USLS, and both SLS models.

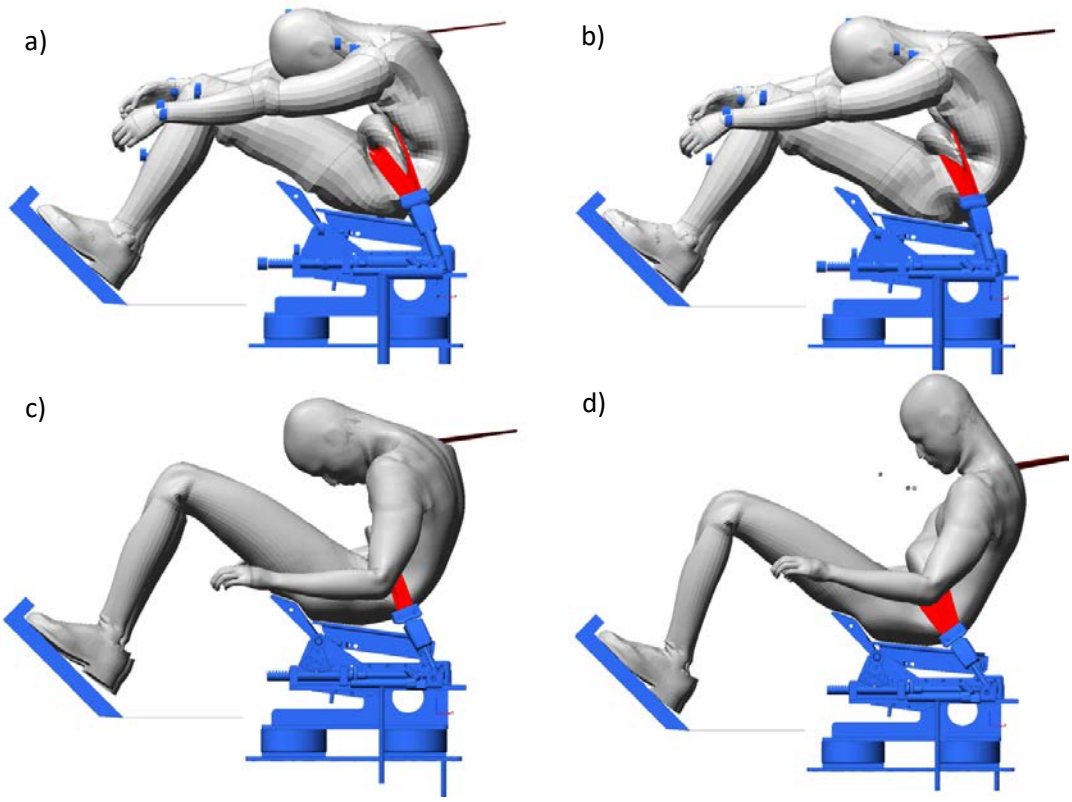

Fig. 5. Still frames extracted from the simulation at 100 msec. a) GHBMC-S, b) GHBMC-SF, c) GHBMC-D and d) GHBMC-DF (73 msec).

Belt Forces

The recorded belt forces were comparable across all models with the exception of the GHBMC-D model. In this case the load measured on both sides of the lap belt showed a sharp drop at 60 msec. associated with the loss of engagement of the pelvis. Additionally, the GHBMC-D model experienced reduction in the shoulder buckle

force past 60 msec. All models showed a peak lap belt force at 7 kN for the buckle side and about 8 kN on the anchor side (Fig. 6).

Belt Engagement

The models with the detached pelvis flesh (GHBMC-S and GHBMC-DF) showed good pelvis-lap belt engagement (Fig. 7). In these cases, the lap belt pre-tensioners were more effective in moving the belt below the ASIS. The belt remained in front of the ASIS throughout the simulation, effectively limiting pelvis forward motion. The GHBMC-SF showed good initial belt engagement, however past the peak of the lap belt force (90 msec.) the belt partially slipped up relative to the pelvis and intruded onto the abdomen. In case of the GHBMC-D model the lap belt initially engaged the pelvis at the ASIS, however at peak lap belt force (60 msec.) it slipped off, passed over the iliac crest and penetrated into the abdomen (Fig. 7).

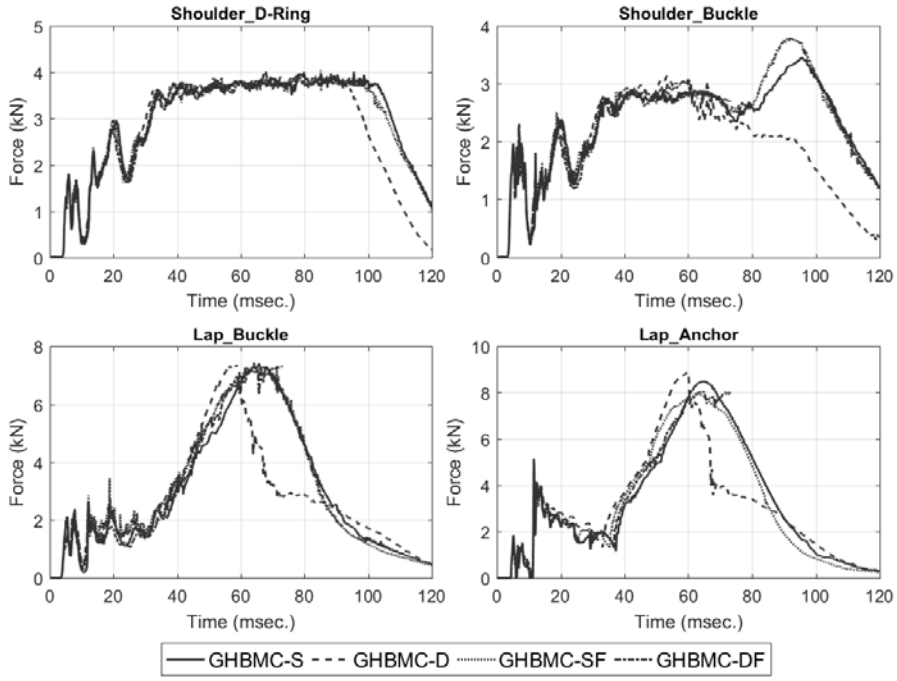

Fig. 6. Cross-sectional belt forces measured for each HBM.

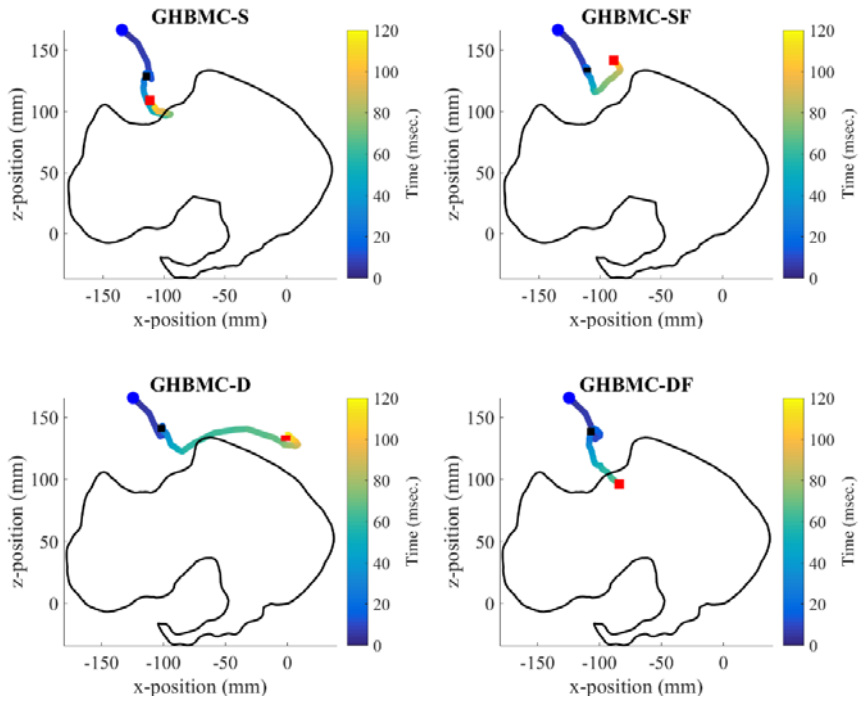

Fig. 7. Trajectory of the lap belt recorded in the pelvis reference frame. For each case a point at the bottom of the lap belt, directly in front of the left iliac wing is shown. Blue circle depicts the belt initial, and red square its final position. Black square depicts a point (20msec.) after both lap belt pretensiones were fully deployed.

Lumbar Spine Kinematics

The lumbar spine response varied across all evaluated HBMs. All models showed initial lumbar spine alignment, however past 30 msec. the GHBMC-S and GHBMC-SF models transitioned into flexion. The lumbar spine response remained comparable between these two models throughout the simulation. For the GHBMC-D and GHBMC-DF models the flexion transition was not observed until 50msec. (Fig. 8). The detailed models showed similar response until 60 msec., where the GHBMC-D model submarined under the lap belt. At 70 msec. both models showed diverging response and the GHBMC-DF model error terminated at 73msec.

All models showed substantial posterior rotation of the pelvis, which further amplified lumbar spine flexion. The rotation was the lowest for models with the flesh attached to the pelvis (GHBMC-D and GHBMC-SF), and the largest for models with flesh detached from the pelvis (GHBMC-S and GHBMC-DF). The GHBMC-D model, showed large pelvis forward excursion which was a consequence of the model submarining under the lap belt (Fig. 9).

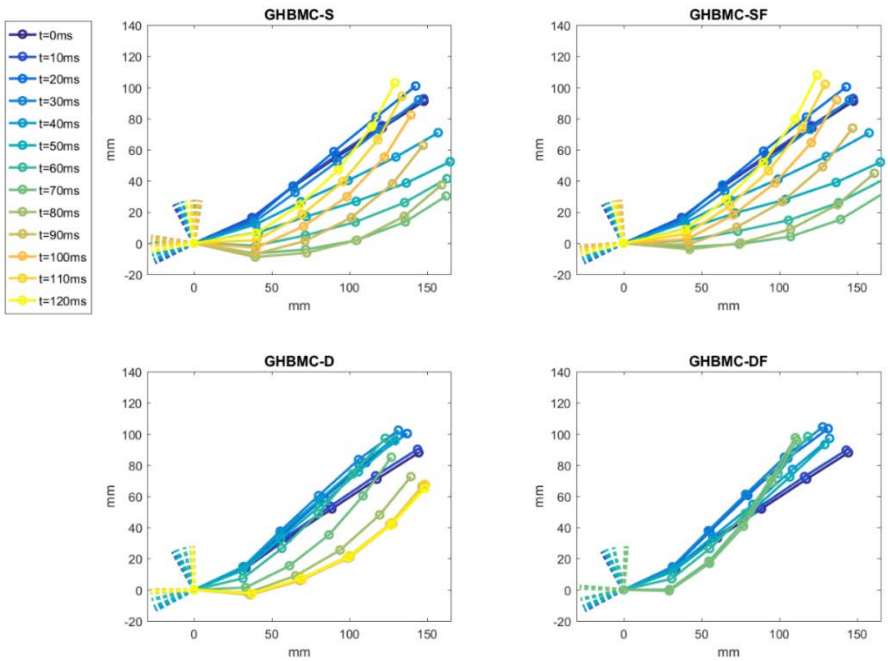

Fig. 8. Lumbar spine (Pelvis - L1) trajectories across all evaluated HBM. All responses were measured relative to the pelvis origin [16]. Each vertebra is shown as a dot, and X and Z axis of the pelvis coordinate system are shown as dashed lines.

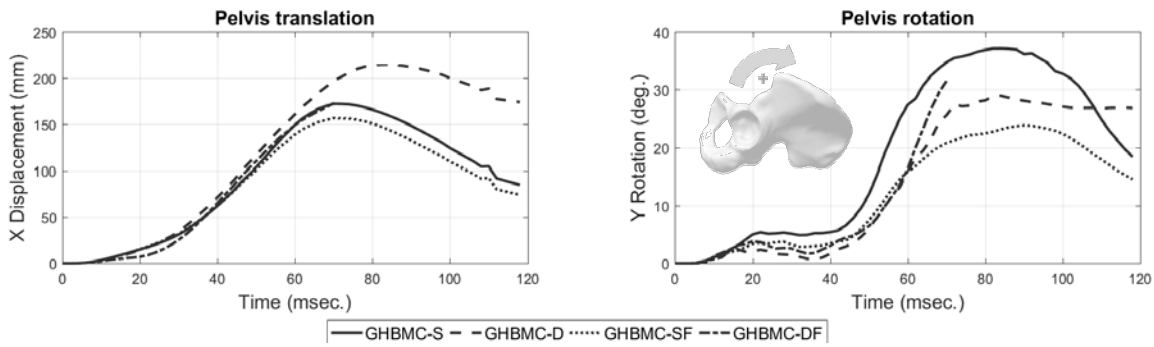

Fig. 9. Pelvis translation relative to the sled in the longitudinal direction (left), and pelvis rotation in the sagittal plane (right). Comparison across all four evaluated models.

Lumbar Spine Forces

All models showed substantial lumbar anterior-posterior shear (Fx), compression force (Fz) and flexion moment (My), and negligible forces in all other directions (Fig. 10, Fig. 11, Fig. 12, and Fig. 13). The GHBMC-D model should not be compared with other models past 60msec. when submarining occurred. The simplified models showed lower peak Fz (2kN vs. 4kN), but comparable My (~75Nm). Interestingly the detailed and simplified models showed inverted polarity of Fx throughout the simulation. The simplified models showed highest My in either the T12-L1 or L1-L2 joints, whereas, prior to submarining detailed models showed highest My in the L5-Pelvis joint.

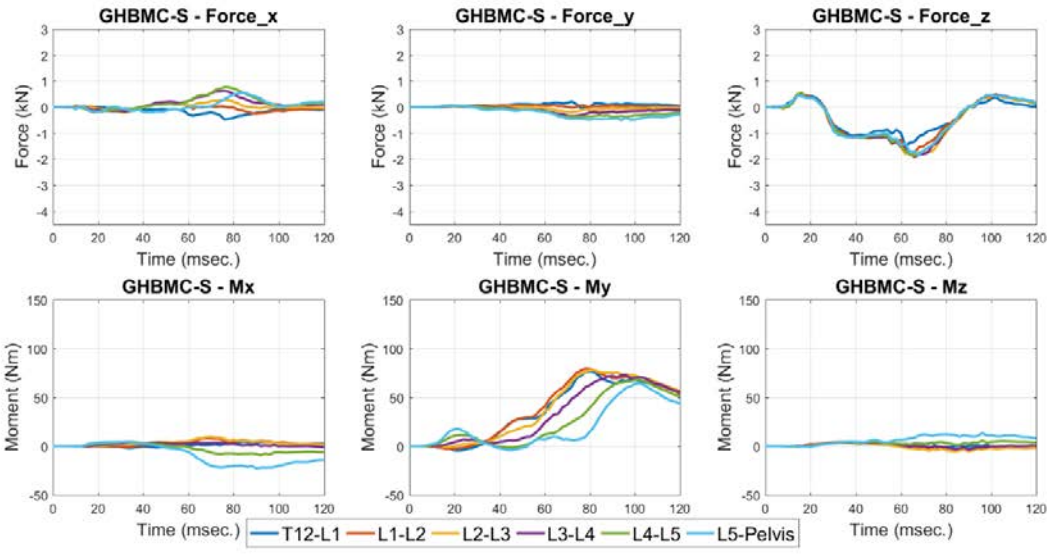

Fig. 10. Lumbar spine forces recorded for the GHBMC-S model.

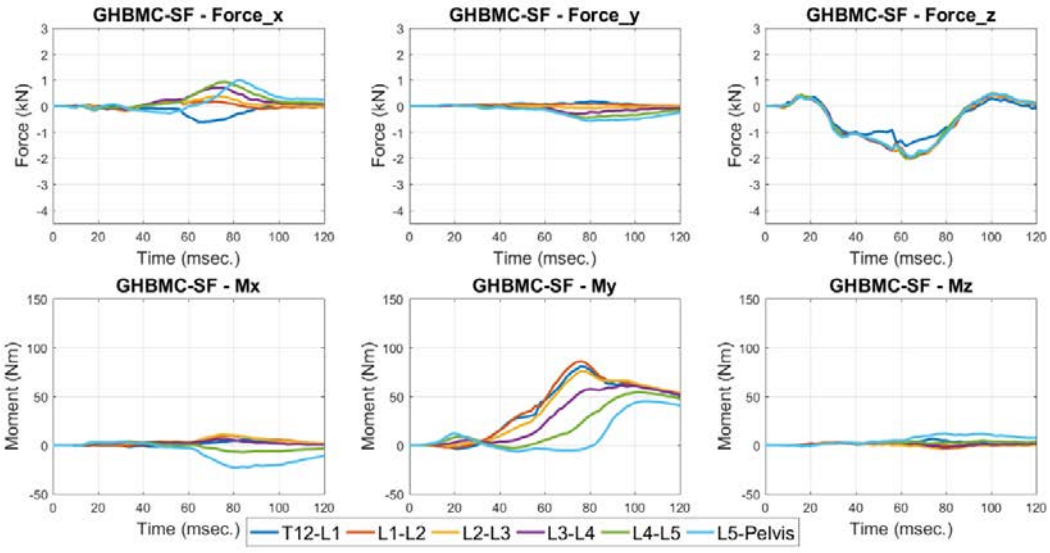

Fig. 11. Lumbar spine forces recorded for the GHBMC-SF model.

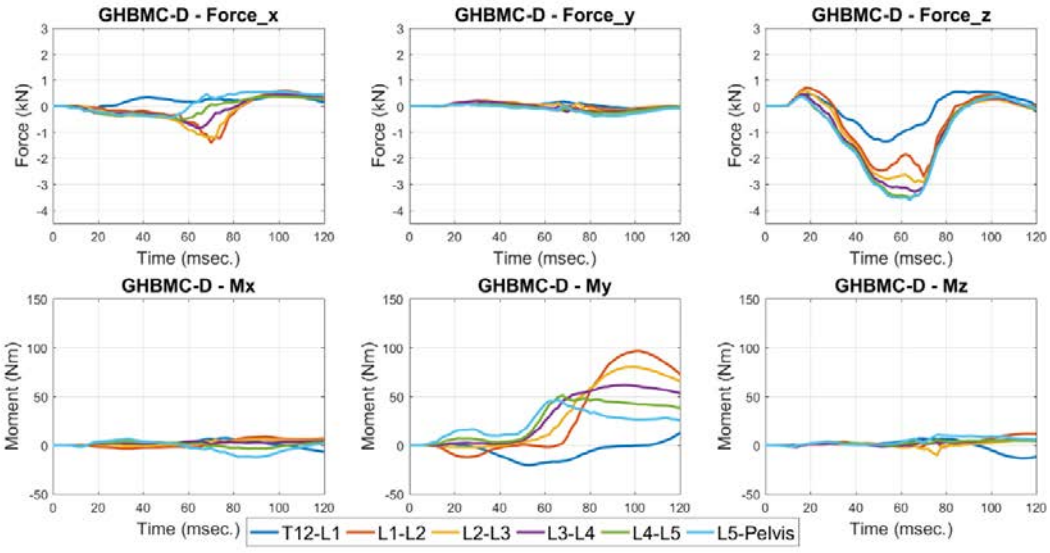

Fig. 12. Lumbar spine forces recorded for the GHBMC-D model with submarining occuring at 60 msec.

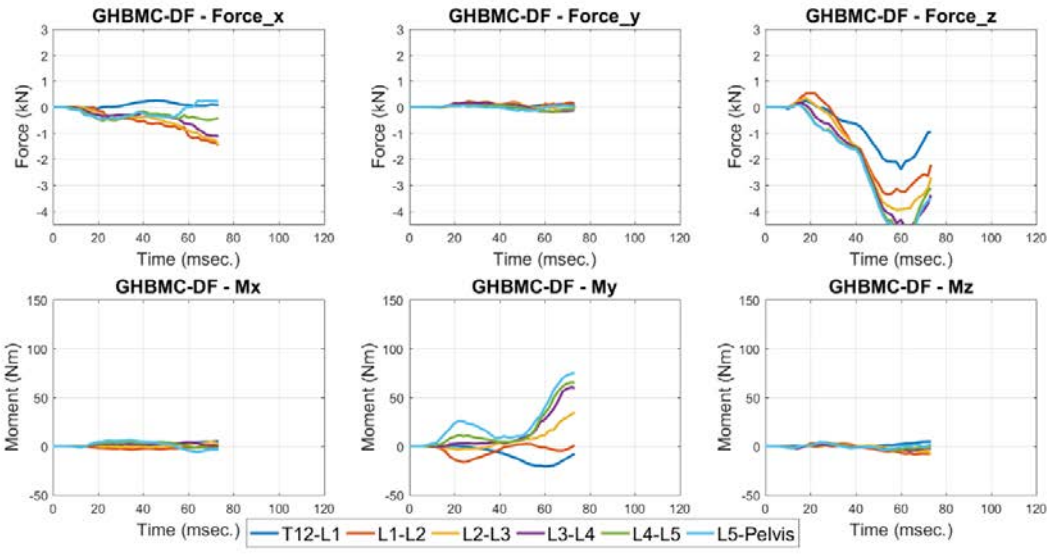

Fig. 13. Lumbar spine forces recorded for the GHBMC-DF model.

IV. DISCUSSION

This paper describes the process of development and evaluation of an updated simplified lumbar spine (USLS) needed for the implementation with the existing simplified GHBMC model. The USLS model was constructed to match the performance of the more complex and anatomically representative lumbar spine (ALS) developed for the new implementation of the detailed GHBMC v5 model. The USLS model showed good agreement with the ALS model in all evaluated directions, whereas previously used SLS models showed large discrepancies from ALS and from each other. The differences in respective SLS models were observed before and documented in [16]. The loading directions relevant for the frontal loading directions, flexion and anterior shear, showed especially good agreement between USLS and ALS. Some discrepancy was observed for compression and torsion. The difference in compression was a consequence of lumbar spine buckling under compressive load and progressing into considerable extension. Under torsion, all models deformed out of plane. Since the SLS model defines stiffness independently in each direction, it is particularly challenged by combined (e.g. compression with extension, out of plane motion) loading, which may result in a softer response for complex loads. Nevertheless, The USLS model showed much better agreement with the ALS model then previously utilized SLS models.

Next, the USLS model along with the pelvis flesh modifications was implemented into two HBMs, previously evaluated in the reclined environment, to evaluate their effect onto the whole body response in a reclined scenario. The results of the sled test indicate that even with modified pelvis flesh and updated and unified lumbar spine, substantial differences remain between performance of the evaluated GHBMC-S and GHBMC-D models. Both HBMs showed differing responses when tested with matching boundary conditions.

The GHBMC-S and GHBMC-DF showed good belt-pelvis engagement throughout the entire simulation. In these cases, the lap belt pre-tensioners were effective in moving the lap belt down, and placing it low relative to ASIS. In the other two models, after pre-tensioning, the belt settled further away from the ASIS when compared to their respective counterparts. In the GHBMC-SF model the lap belt also engaged the pelvis, until 90 msec., when it partially slipped off the left iliac wing. The GHBMC-D model lost pelvis engagement at peak lap belt load (60 msec.), when the belt intruded into the abdomen (Fig. 7).

The fundamental difference between the models with divergent submarining outcome, is within their flesh definition and its attachment to surrounding tissues. Both models that showed good lap belt engagement were characterised by a tissue model detached from the pelvis. In these HBMs the flesh was allowed to move on top of the pubic bone with a sliding style contact. The potential for pelvis-to-flesh sliding, although not anatomically correct, facilitates the low shear boundary condition and mitigates the effect of high flesh stiffness previously documented [28]. This in turn improved the effectiveness of dual lap belt pre-tensioners, which when fully deployed (20 msec.), were effective in moving the belt lower below the ASIS (Fig. 7). However, due to the continuous definition of the pelvis-flesh construct, the pre-tensioner effectiveness was limited in case of the GHBMC-D and GHBMC-SF models. This led to the higher position of the lap belt, and consequently submarining.

It needs to be noted that, while a sliding flesh-pelvis is not anatomically correct, it might be the necessary short term solution for a representative pelvis flesh interaction given the current challenges of soft tissue modeling with lagrangian formulation [29]. A similar finding has been indicated before in several studies [13][16], however this is the first study that looked at direct comparison with matching HBMs. It needs to be noted that recently released GHBMC v5 model features a new softer flesh material model that could address these discrepancies.

A previous study [16] has identified substantial differences between the THUMS and GHBMC family models in lumbar spine kinematic response. While the THUMS models showed little backward pelvis rotation and their lumbar spine remained aligned up until 60 msec. before progressing into flexion, the GHBMC models showed substantial pelvis and lower lumbar backward rotation. It was speculated that the simplified lumbar spine representation in both the GHBMC models might be responsible for these differences. The new USLS model showed substantial increase in flexion stiffness when compared with the previously used GHBMC-S model (Fig. 4). However, with the updated lumbar spine, all GHBMC models showed backward pelvis rotation that resulted in horizontal alignment of the lower lumbar spine. Interestingly, the models with the flesh detached from the pelvis experienced the largest backward pelvis rotation (Fig. 8 and Fig. 9). However, recently published results from the reclined PMHS test series suggest that subjects either maintained their initial pelvis orientation throughout the test or rotated forward [30]. Similarly, the lumbar spine was shown to maintain its initial alignment before progressing into flexion [31]. All variations of investigated GHBMC models showed opposite pelvis and lumbar spine motion.

In [16] authors compared recorded lumbar spine forces between four different HBMs in a matching recline sled scenario. While the THUMS models were the most similar both the GHBMC-S and GHBMC-D showed substantial differences in recorded forces. The authors again speculated that the simplified lumbar spine representation, with different performance curves in both GHBMC models might be responsible for these differences. The unified USLS model addressed some of the observed issues, such as discrepancies in flexion moment, however some differences remained. First, the simplified and detailed models differed in the amount of compression force carried by the lumbar spine. With the unified lumbar spine, the simplified models recorded a peak compression of 2kN, while detailed models recorded about 4kN in peak force. Similarly, while the peak flexion moment remained comparable (75Nm) across all models, the location of the recorded moment differed. The simplified models recorded peak moment in the T12-L2 segment, the detailed model (before submarining) showed the highest recorded moment in the L4-S1 segment (Fig. 10, Fig. 11, Fig. 12 and Fig. 13). This could be explained by either, large differences in lumbar spine alignment at peak force (Fig. 8), different connectivity of tissue models surrounding the lumbar spine, or could indicate an existence of alternative load path that could alter the force transmitted by the lumbar spine.

V. CONCLUSIONS

The following conclusions can be drawn from this study. A new updated simplified lumbar spine, representative of more complex anatomical lumbar spine model was developed for use with simplified GHBMC model. Updated lumbar spine model showed little effect on lumbar kinematic response in recline scenario. Pelvis-flesh attachment model had a substantial effect on the submarining outcome across all evaluated models. With the updated and unified lumbar spine the detailed and simplified models showed substantial differences in recorded forces indicating that other modeling assumptions should be investigated to identify the source of differed responses. Finally, all GHBMC models experienced posterior pelvis rotation which was inconsistent with the recently published PMHS data.

VI. ACKNOWLEDGEMENT

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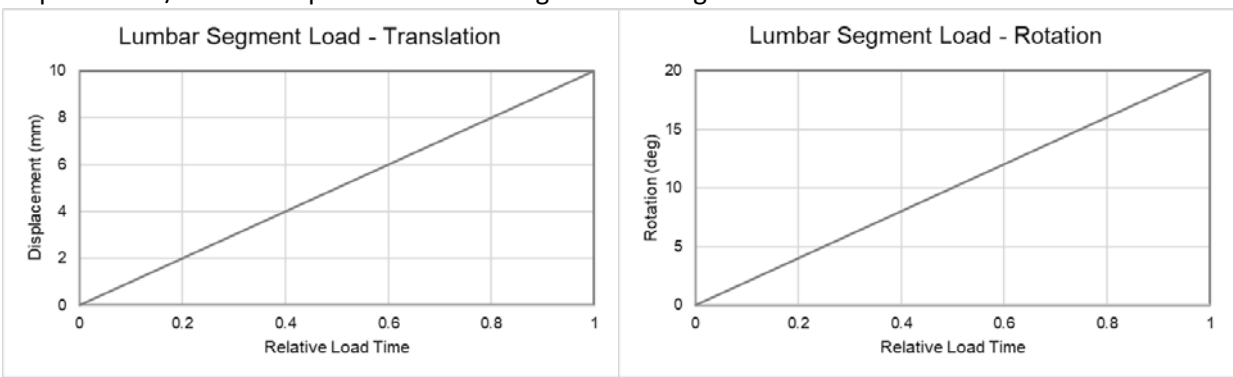
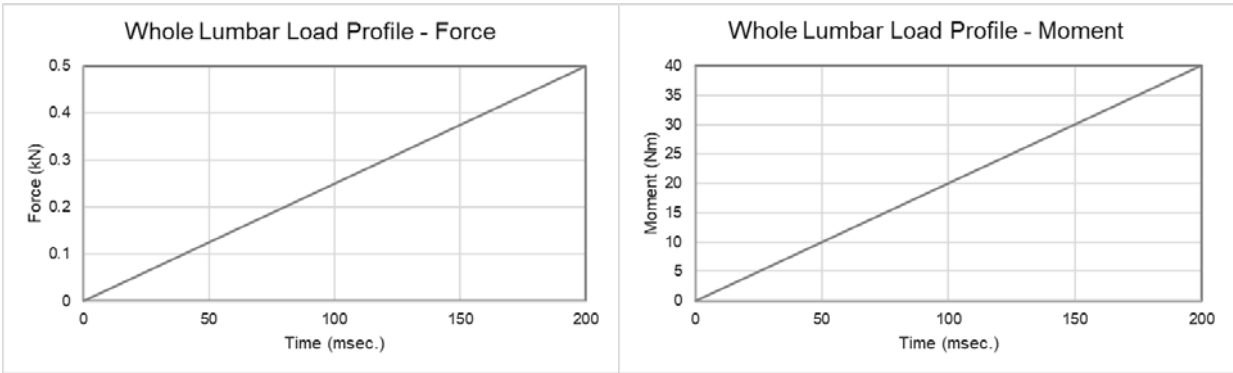
VIII. APPENDIX

APPENDIX A

Displacement/Rotation input used for ALS segment loading.



Force/Moment input used for whole spine (T12-S1) evaluation.



APPENDIX B

Performance curves developed for the USLS model.