Comparing the Frontal Impact Responses of the VIVA+ Average Female and SAFER Average Male Human Body Models in a Generic Seat

Erik Svenning[1], Krystoffer Mroz[2], Torbjörn Johansen[1], Nils Lubbe[2], Johan Iraeus[3]

[1]DYNAmore Nordic AB
[2]Autoliv Research
[3]Chalmers University of Technology

1 Abstract

The VIVA+ 50F average female Human Body Model (HBM), currently in early beta status, was compared to the SAFER average male HBM Version 9 with the aim of investigating differences between females and males in terms of kinematics and injury assessment in frontal impacts. The VIVA+ HBM is under development within the research project VIRTUAL and will be released as open source during the summer 2022.

The comparison between the HBMs was carried out using LS-DYNA version R9.3.1 in a generic sled test interior consisting of a semi-rigid seat, a footwell and a pretensioned three-point belt system with 3.5 kN load limiter. The HBMs were positioned in a pre-simulation using spring and damper elements attached to target points estimated using an automobile driving posture prediction model from the literature. Subsequently, the HBMs were subjected to a frontal crash corresponding to an initial velocity of 50 km/h.

Occupant kinematics were analyzed by comparing head, chest, and pelvis kinematics between the two models. Additionally, HIC15, rib peak strains, and upper neck, lumbar spine and pelvis anterior superior iliac spine (ASIS) resultant forces were compared between the two models. The largest differences between SAFER HBM and the beta version of VIVA+ 50F were found for rib peak strains, where VIVA+ predicts higher strain than SAFER HBM, and lumbar spine forces, where VIVA+ predicts lower forces than SAFER HBM. Furthermore, higher neck forces and ASIS forces were predicted by SAFER HBM compared to VIVA+.

2 Introduction

Traffic safety has improved greatly over the last decades, but road traffic crashes are still among the most common causes of death [1]. To further improve traffic safety, a valuable tool is virtual testing using Human Body Models (HBMs) that can predict injury outcomes in complex crashes. Hence, there is an increasing need for biofidelic HBMs that accurately model car occupants and vulnerable road users. These HBMs also need to cover the diversity of the human population with varying anthropometry, age, sex and health status. To meet this need, a number of HBMs with different anthropometry have been developed, e.g., the Total Human Model for Safety (THUMS) [2], the SAFER HBM [3][4], the Global Human Body Models Consortium (GHBMC) models [5][6], and some are under development, such as the Virtual Vehicle-safety Assessment (VIVA+) models [7][8].

In a study by Forman et al. [9], it was found that females are at greater risk of AIS 2+ and AIS 3+ injuries compared to males when exposed to comparable crashes. The greatest differences in injury risk were observed in the lower extremities and ankles, followed by AIS3+ rib fracture injury and abdomen injuries. Due to these differences, it is of interest to compare the response of HBMs representing males and females. In the present work, we therefore compare the SAFER HBM version 9 average male to a beta version of the VIVA+ 50F average female HBM.

The SAFER HBM version 9 is a finite element model of an adult average male which was developed to improve the understanding of impact response and injury mechanisms in humans. The SAFER HBM, originally developed from version 3 of the Total Human Model for Safety (THUMS) [2], has been updated with new head, neck, and rib cage models [3], as well as more detailed lumbar spine [4]. The capability of the HBM model to predict kinematics and rib fractures in the upright posture has been demonstrated [10], as has its capability to predict whole-body kinematics in the reclined posture [11]. In this study, the SAFER HBM was modified by increasing the friction between the pelvis bone and the pelvis soft tissue from 0.2 to 0.6 to improve the correlation of the model's pelvis rotations with those of the PMHS tests. The SAFER HBM was also compared to other HBMs in pelvis and lumbar spine kinematics and loading for reclined postures in [12].

The VIVA+ HBMs are a set of HBMs currently being developed within the VIRTUAL project [7] with the purpose to provide open source HBMs suitable for safety assessment for occupants as well as for vulnerable road users. The different HBMs represent seated and standing adult average females and average males, whereby the seated average female HBM (VIVA+ 50F) is considered in the present work. The VIVA+ HBM is partly based on the VIVA HBM [13], but has substantially improved biofidelity. The VIVA+ HBM is still under development and validation is on-going [14]. It should be noted that the model is currently not validated for any of the injury metrics used in the current study.

In the present work, the first objective was to evaluate the beta version of the VIVA+ average female HBM in terms of robustness, whole-body occupant kinematics and injury assessment in frontal crashes. The second objective was to identify differences in kinematics and loading between two different HBMs comparing the responses from the beta version of the VIVA+ HBM to those of the SAFER HBM.

3 Methods

3.1 Method overview

The responses of the HBMs are evaluated in a generic sled test interior (similar to the interior in [15]) consisting of a semi-rigid seat, a pretensioned and 3.5 kN load-limited seat-back mounted three-point belt system, and a footwell. All simulations are performed with LS-DYNA R9.3.1, mpp, single precision. Autoliv models of the pretensioner, retractor, and webbing material, which closely matched their mechanical counterparts, were used for the model of the belt system.

3.2 Positioning

To position the HBMs, target values for suitably chosen control points were computed and the HBMs were pulled into position in a pre-simulation. In the present work, the posture prediction model in Park et al. [16] was employed, with the input data listed in Table 1 to compute target positions for the head center eye, head tragion, vertebrae junctions C7-T1, T12-L1, and L5-S1, and the hip, knee, and ankle. The SAFER HBM has been previously positioned according to Park's data with good results [17], whereas the positioning of VIVA+ is less well established. In the following, we will therefore focus on the positioning of VIVA+. More precisely, the target points for VIVA+ are computed according to the female driver posture-prediction model in Table 5 in [16].

The initial postures of the HBMs are compared to the target points in Figure 1. For VIVA+, we note that the pelvis angle differs substantially between the initial position of VIVA+ and the target position; the computed target posture has a very upright pelvis position. Comparing different posture prediction models is, however, beyond the scope of the current work. Therefore, we choose to use the computed target points without modifications. Nevertheless, the target posture for the average female deserves further investigation.

When positioning the HBMs, it was noted that the anthropometry for VIVA+ does not seem to fully match the anthropometry for average females according to [16]: VIVA+ seems to have a slightly longer upper body and slightly shorter upper legs. This deviation will, of course, affect the possibility to closely match all target positions between the HBM and the posture prediction model. This is a challenge that will most likely increase as HBM models become more diverse in the future: as more diversity is covered by HBMs, we will more frequently encounter situations where the anthropometry of the HBM does not fully match the anthropometry of the posture prediction model. Therefore, a positioning strategy is needed that can handle an HBM that is e.g. taller or has higher body mass index (BMI) than the reference, without introducing unphysical deformations during the pre-simulation. More precisely, we need a positioning strategy that avoids unphysical compression or tension of the spine to make the HBM match the target points of the shorter reference. In the same way, the positioning strategy needs to avoid unphysical stretching of the legs to make the HBM match the target points of the longer legs of the reference. To address this challenge, we consider two positioning strategies for VIVA+:

1. All target points are included in the pre-simulation and fixed in all directions. In the following, this pre-simulation alternative is denoted “All points”, since all target points are fixed in all directions.
2. All target points except center eye are included in the simulation. Some target points are free to move in some directions as shown in Table 2. The basic idea here is to choose the free directions in such a way that unphysical stretching or compression of the HBM is avoided. In the following, this pre-simulation alternative is denoted “Relaxed z”, since the upper body target points are relaxed in z, thereby allowing the HBM back to move upwards or downwards in order to avoid unphysical extension or compression of the spine. In the same way, the knees are free to move in order to avoid unphysical stretching of the legs.

The HBMs are positioned by means of pre-simulations with spring-damper elements attached to the control points. The node positions from the pre-simulation are used as initial positions in the sled test. Initial strains and stresses from the pre-simulation are not included in the sled test.

Parameter	SAFER HBM	VIVA+
Sex	M	F
Age	50	50
Stature (mm)	1760	1620
Mass (kg)	79.9	62.1
BMI	25.8	24.0
SHS	0.52	0.52
L6re (mm)	-24.0	-24.0
H30 (mm)	341.0	341.0

Table 1: Input data for target point calculation (SHS = sitting height divided by stature, H30 = seat height, L6re = steering wheel center relative to its nominal position. Parameter definitions from [16]).

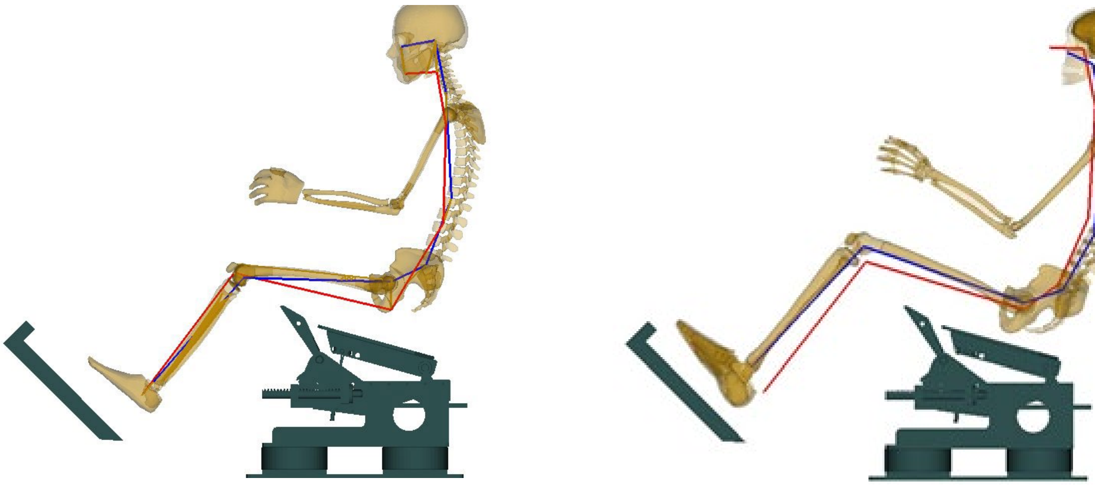

Fig.1: Comparison between initial posture and target posture for VIVA+ (left) and SAFER HBM (right). The solid lines show the initial positions (blue) and target positions (red) of the attachment points for the spring-damper elements. (Note that the center eye and tragion attachment points differ slightly from the target points calculated according to [16].)

Point	Center eye	Tragion	C7-T1	T12-L1	L5-S1	HP	Knee	Ankle
X	Free	Locked	Locked	Locked	Locked	Locked	Free	Free
Z	Free	Free	Free	Free	Free	Locked	Locked	Locked

Table 2: Constraints applied for target points in the pre-simulation for VIVA+ denoted “Relaxed Z”.

3.3 Sled test with generic interior

An FE model of the semi-rigid seat used in the PMHS tests conducted in [18] was used in the present work. The seat model, which is shown in Figure 2, consists of two adjustable plates: the seat and submarining pans. Their geometry and stiffness response were configured to match a vehicle seat. The seat model was validated by comparing moment-rotation responses under static loading against reference data for both front and rear-seat configurations in [11]. In the current study, the front configuration was used with a 15.5 deg seat pan angle, 30 deg submarining pan angle, 128 N/mm seat pan side spring stiffness, 379 N/mm seat pan center spring stiffness, and 132 N/mm submarining pan spring stiffness.

The semi-rigid seat, seat belt, and HBMs were subjected to a full-frontal, 50 km/h 35-g pulse. The crash pulse, which corresponds to the pulse used in several previous PMHS tests [18], is shown in Figure 3.

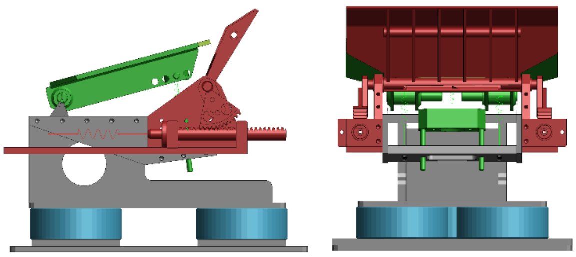

Fig.2: Model of the semi-rigid seat (front seat configuration) with the seat pan in green and the submarining pan adjustable arrangement in red. From [11].

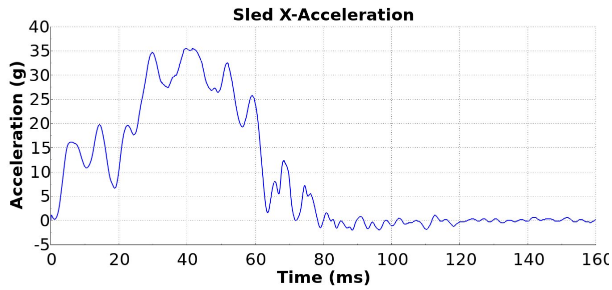

Fig.3: Crash pulse 50 km/h.

3.4 Result Parameters

As output from the simulations, we monitor displacements, accelerations, and section forces. While most measures are straight-forward to extract and need no explanation, we want to highlight a few injury measures.

In the SAFER HBM, upper neck tension forces were measured in the C1 vertebra using cross-sectional force measurements with respect to a local coordinate system in the center of the vertebra: Figure 4. The cross-section included the cortical and spongy bones, neck skin muscles, and ligaments. Pelvis resultant forces were measured using cross section force measurements in the anterior superior iliac spine (ASIS) with respect to a local coordinate system. Lumbar spine forces and moments were measured in the L1 to L5 vertebrae using cross section force-moment measurements with respect to a local coordinate system in the center of each vertebra body (including ligaments).

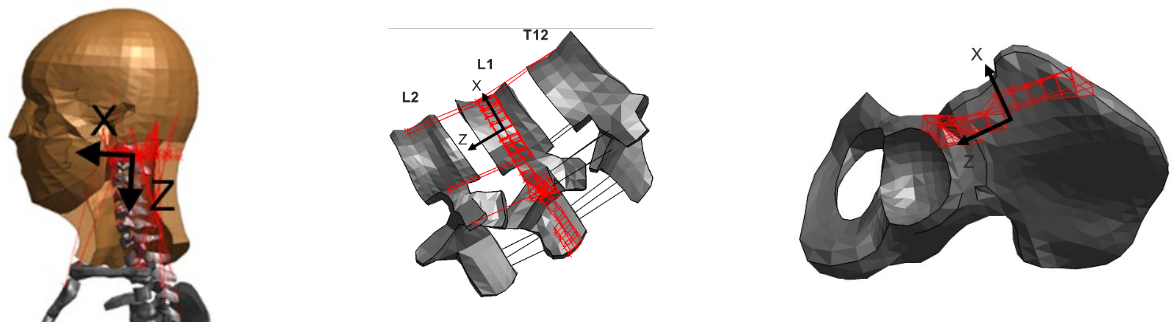

Fig.4: SAFER HBM cross section definitions (in red) for the upper neck, lumbar spine vertebra L1 (left) and pelvis ASIS (right). Intervertebral discs not shown for the lumbar spine.

In the VIVA+ HBM, the upper neck and pelvis ASIS forces were measured using cross-sectional force measurements in the same way as for the SAFER HBM. The lumbar spine forces were measured using **`*DATABASE_HISTORY_BEAM`** for the beams connecting the rigid vertebrae of the lumbar spine.

For the ribs, peak strains were measured in the cortical bone of each rib. The reason for choosing this strain measure is that it can be used to predict AIS1+ and AIS2+ rib fracture risks [20][21]. The rib strains are evaluated in the current study, whereas evaluation of rib fracture risks is left as future work.

4 Results

4.1 Positioning

The HBMs are positioned by running a pre-simulation for 400 ms with spring-damper elements attached to the chosen control points. Figure 5 shows the pre-simulation results with all points for VIVA+, relaxed z for VIVA+, and all points for SAFER HBM. A comparison between the postures of the VIVA+ and SAFER HBMs is shown in Figure 6. The control points for SAFER HBM are close to the points predicted with Park's model. The good match is a result of the close agreement between the anthropometry of the SAFER HBM and the reference. For VIVA+, larger discrepancies can be seen, especially for L5-S1. Pre-simulation with some points relaxed in z gives much closer agreement for L5-S1. This approach avoids unphysical compression of the spine, at the price of substantially different foot position and different z-location for the head.

As can also be seen in Figure 5, the pelvis angle differs between VIVA+ and Safer HBM as well as between the two VIVA+ pre-simulations. For VIVA+, the pelvis angle was estimated to 55 degrees for the pre-simulation with all points and 16 degrees for the pre-simulation with relaxed z. The SAFER HBM pelvis angle was estimated to 43 degrees. We note that the pelvis angles of 43 degrees and 55 degrees are within the range of angles measured in [22], whereas the angle of 16 degrees is not.

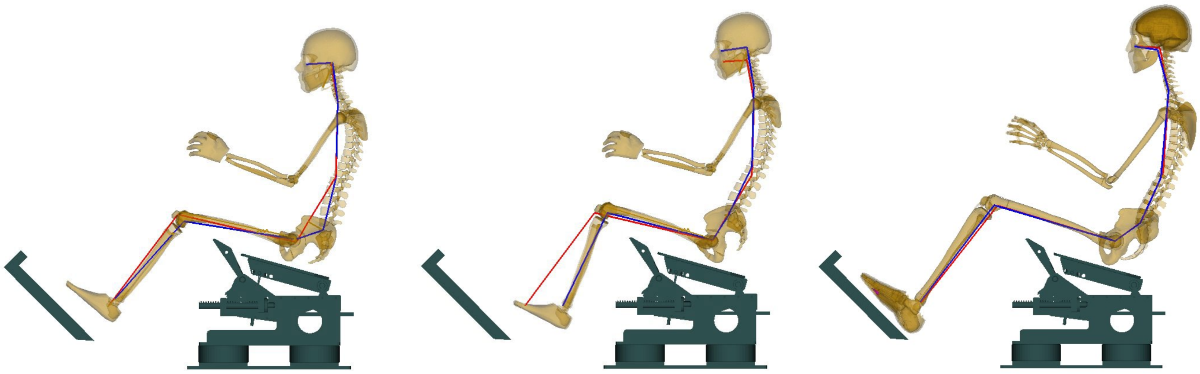

Fig.5: Pre-simulation results for VIVA+ with all points (left), VIVA+ with Relaxed Z (center), and SAFER HBM (right).

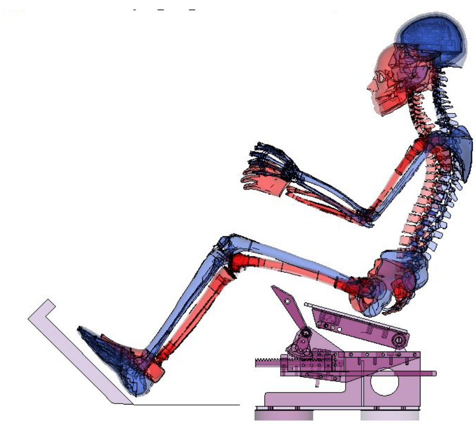

Fig.6: Comparison between VIVA+ with all points (red) and SAFER HBM (blue) after pre-simulation.

4.2 Sled test

Sled test simulations are performed for VIVA+ and SAFER HBM as described previously. For VIVA+, two sets of simulations are performed with the occupant positions from the two different pre-simulations as input. The HBM motion is shown in Figure 7 and Figure 8. We note that the motion of the arms differs between Safer HBM and VIVA+: The arms of SAFER HBM are fully straight at around 80 ms, whereas the arms of VIVA+ remain partly bent at the elbow during the whole simulation.

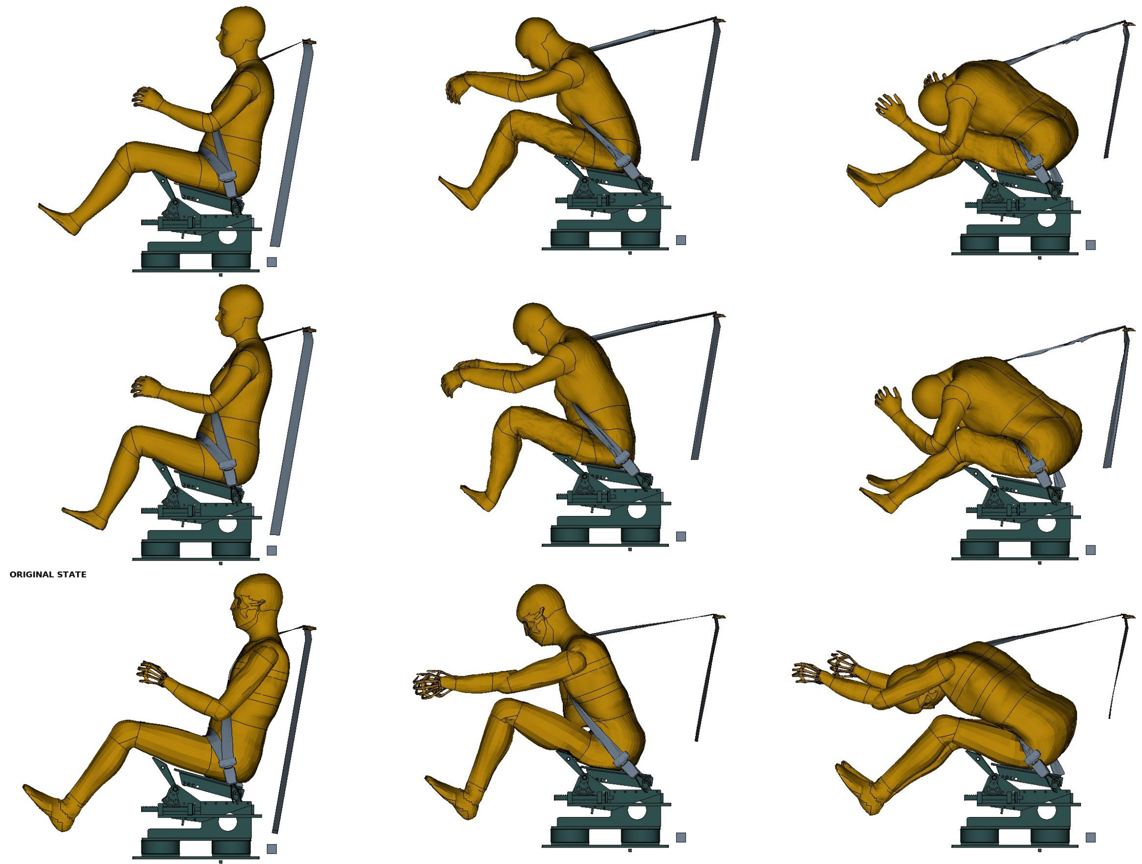

Fig.7: HBM motion for VIVA+ with all points presim (top), VIVA+ with relaxed z presim (center), and SAFER HBM (bottom) at 0, 80 and 130 ms.

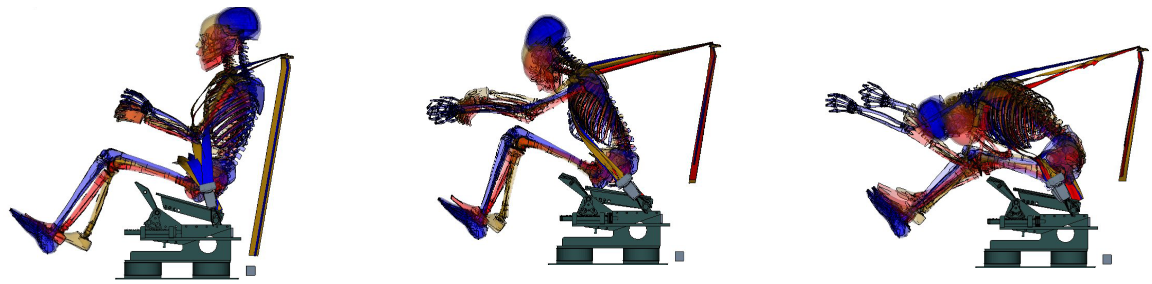

Fig.8: Comparison of the kinematics for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (gold) at 0, 80 and 130 ms.

The head trajectory is shown in Figure 9. SAFER HBM follows a wider path than VIVA+ due to the larger sitting height of SAFER HBM. Figure 10 shows the pelvis x-displacement and Figure 11 shows the pelvis angle in degrees. Note that the angle shows the total angle and not the angular displacement.

Figure 12 shows the head acceleration measured in local coordinate system for VIVA+ and SAFER HBM. The HIC15 are 201 for SAFER HBM, 320 for VIVA+ with all points pre-simulation, and 263 for VIVA+ with relaxed z pre-simulation.

Figure 13 shows the forces for the lumbar spine vertebrae. As can be seen, VIVA+ consistently predicts forces of lower magnitude than SAFER HBM. Figure 14 shows the neck and ASIS cross section forces. For the ASIS forces, we note that the magnitude of the initial peak is around 3-4 kN for SAFER HBM, whereas it is only around 2 kN for VIVA+. For completeness, forces and moments for the femur and tibia are shown in the appendix in Figures 18, 19, 20 and 21.

The belt forces are shown in Figure 15. The belt forces are very similar between SAFER HBM and VIVA+ for the first 80 ms, whereas the belt forces decrease more rapidly for VIVA+ than for SAFER HBM during the later stage of the simulation.

The rib cortical bone peak strains predicted for SAFER HBM and VIVA+ are shown in Figure 16. Maximum strains of 1.8 % was measured in the SAFER HBM, 3.9 % in the VIVA+ All points and 4.3 % in the VIVA+ relaxed z. The peak strains were measured in the anterior part of right rib 3 for the SAFER HBM and in the anterior part of right rib 5 for both VIVA+ All points and VIVA+ relaxed z HBMs, see Figure 17.

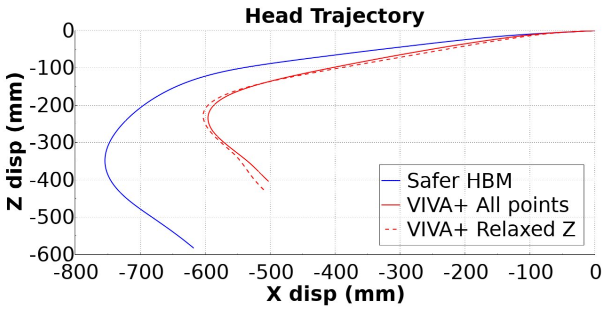

Fig.9: Head trajectory for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (dashed red).

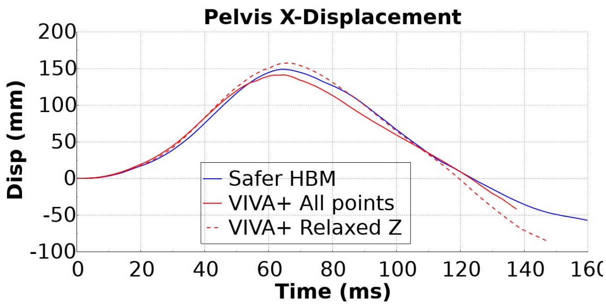

Fig.10: Pelvis X-displacement for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (dashed red).

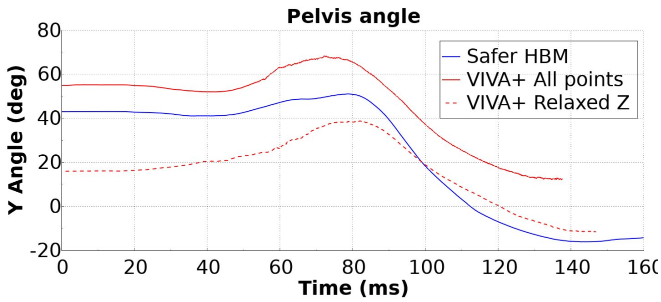

Fig.11: Pelvis angle for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (dashed red).

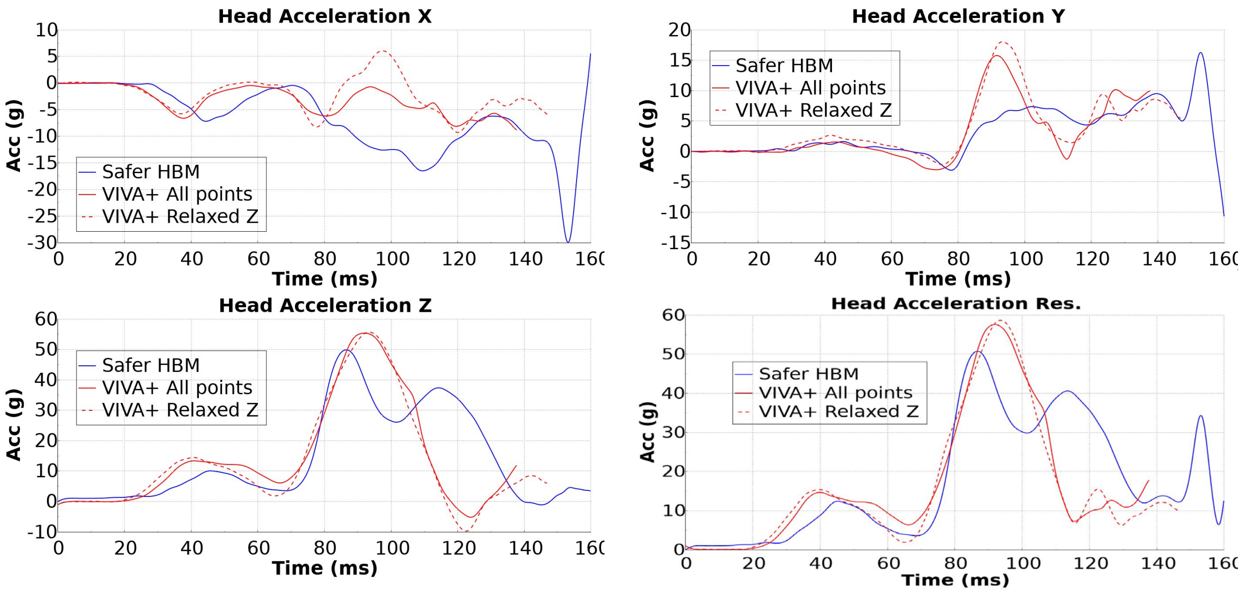

Fig.12: Head acceleration for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (dashed red). The components shown are local x (top left), local y (top right), local z (bottom left) and magnitude (bottom right).

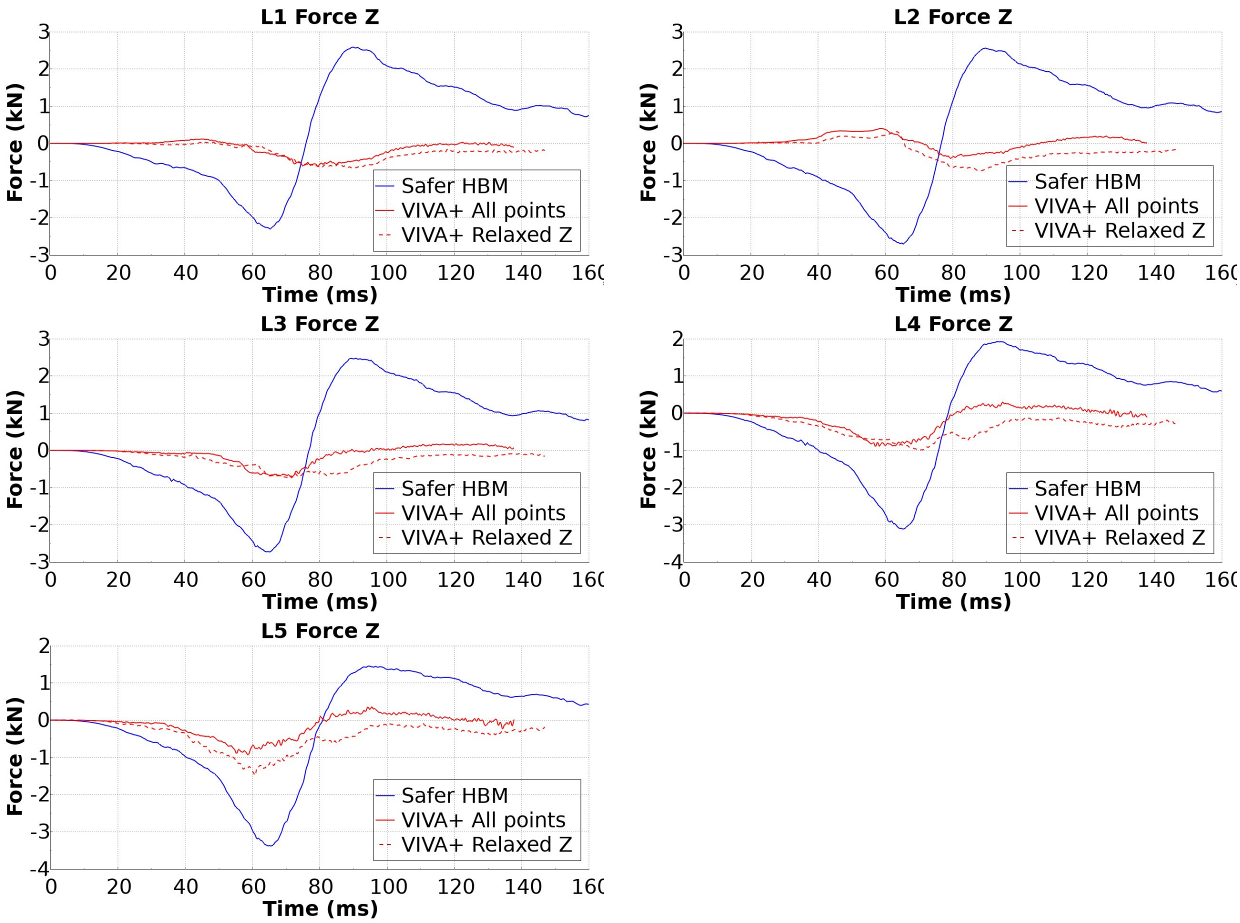

Fig.13: Lumbar spine forces for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (dashed red).

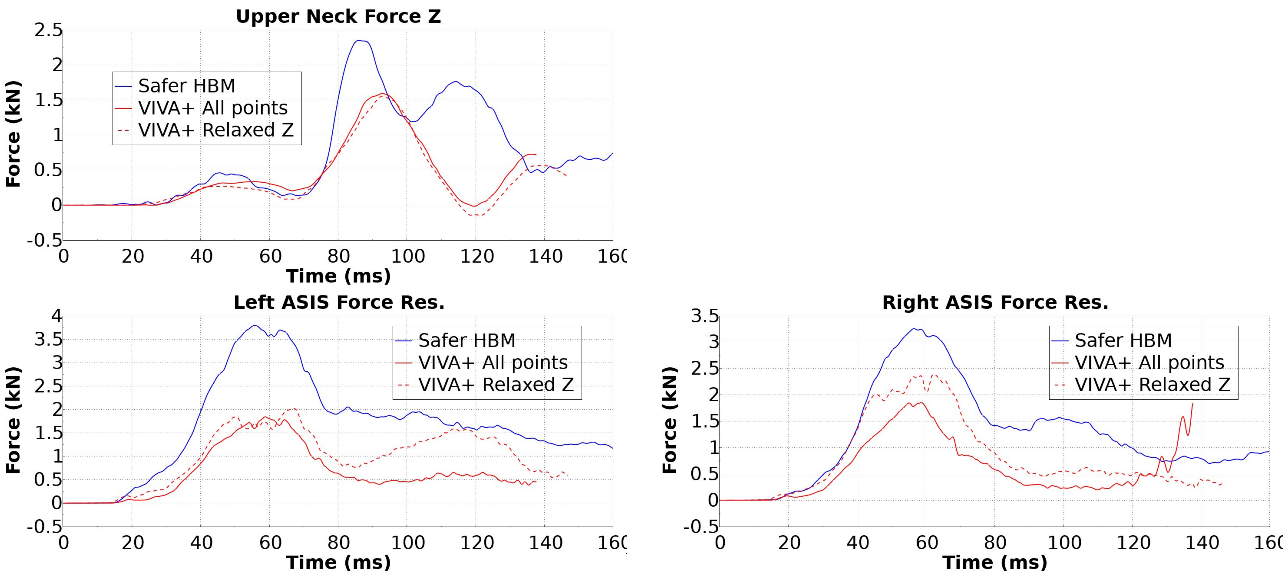

Fig.14: Upper neck and ASIS forces for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (dashed red).

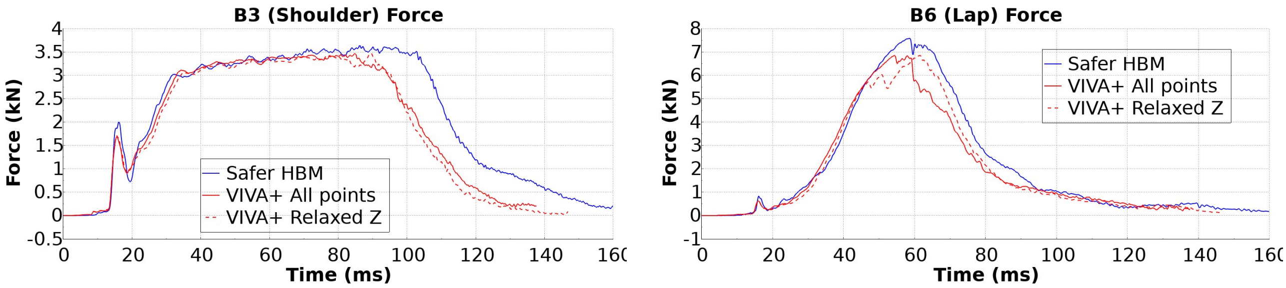

Fig.15: Belt forces for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (dashed red).

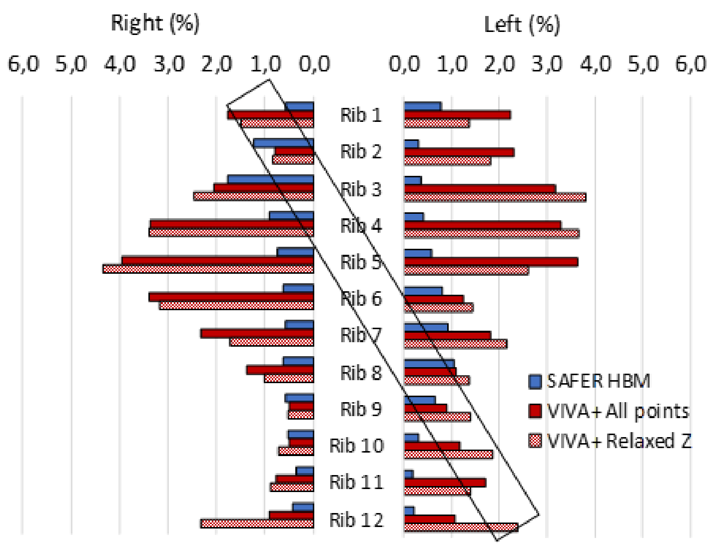

Fig.16: Rib cortical bone peak strains for SAFER HBM, VIVA+ with all points, and VIVA+ with relaxed z.

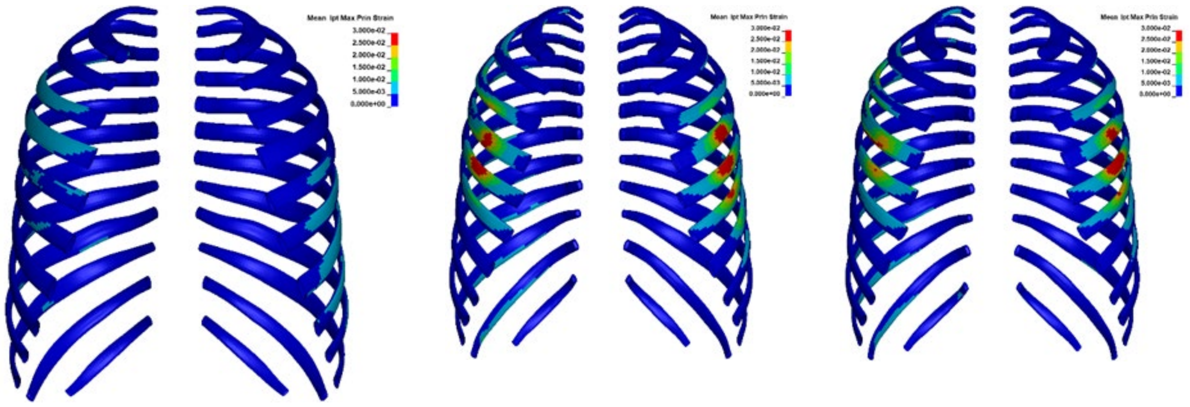

Fig.17: Rib cortical bone strain distributions (frontal view) for the SAFER HBM at 71 ms (left), VIVA+ with all points at 96 ms (middle), and VIVA+ with relaxed z at 96 ms (right).

5 Discussion

For the positioning, pre-simulation with spring-damper elements was successfully applied for both HBMs. However, positioning was more challenging for the VIVA+ HBM compared to the SAFER HBM for mainly two reasons. First, the SAFER average male HBM closely matches the anthropometry for average males in [16], and, hence, all control points can be pulled into the desired target position without introducing unphysical stretching of the HBM. This is, in contrast, not true for the VIVA+ HBM. The anthropometry of VIVA+ 50F differs in some respects from the anthropometry for average females in [16]. Therefore, unphysical stretching would occur in the VIVA+ HBM if all control points would be pulled into the desired target position. As a remedy, we propose to constrain spine and head control points only in x to avoid unphysical compression/stretching of the spine. Similarly, we propose to relax some constraints on the legs to avoid unphysical stretching/compression of the legs. The proposed approach avoids unphysical deformations of the HBM, but of course at the cost of larger distance between some of the HBM control points and the corresponding target points. Second, the posture prediction model employed in the present work predicts a posture leading to a very upright pelvis angle for the average female. This issue deserves further attention, since our work only considered one posture prediction model. There are other posture prediction models, such as the joint-angle-prediction model described in [16], that might be better to handle anthropometric differences. Comparing different posture prediction models is, however, beyond the scope of the present work.

Regarding the crash test simulation, several significant differences were identified between the SAFER HBM and the VIVA+ 50F HBM.
For the lumbar spine forces, very large differences are seen between SAFER HBM and VIVA+. While recognizing the HBM anthropometries and initial pelvis angles, the largest part of the differences are most likely due to the modeling approaches, which deserve further attention.
For the pelvis ASIS forces, higher values are seen for SAFER HBM than for VIVA+. Due to the high level of similarity of the belt forces for VIVA+ and SAFER HBM, we would expect a higher level of similarity also for the ASIS forces. One reason for the observed differences is probably that the rather stiff soft tissue in VIVA+ takes load that would otherwise go through the ASIS.
Finally, the large difference in predicted rib cortical bone peak strains between SAFER HBM and VIVA+ needs to be investigated.
In general, the VIVA+ model was found robust in most body segments except for the soft tissue surrounding the pelvis, which showed an instability on the interior side and limited the VIVA+ simulation run times to 138 and 147 ms.

6 Conclusion

The beta version of the VIVA+ 50F average female HBM and the SAFER average male HBM were compared in a generic sled test interior subjected to a 35 g full frontal crash pulse. Significant differences were identified in the lumbar spine, pelvis ASIS and chest responses between the two HBMs. Further development of the VIVA+ 50F average female HBM is needed to fully use the model for understanding the differences in injury risks between males and females.

7 Acknowledgment

We would like to thank the VIRTUAL project for giving us the opportunity to test the beta version of the VIVA+ 50F model.

8 References

[1] https://ourworldindata.org/causes-of-death

[2] Iwamoto, M., Kisanuki, Y., Watanabe, I., Furusu, K., Miki, K.: "Development of a Finite Element Model of the Total Human Model for Safety (THUMS) and Application to Injury Reconstruction", Proceedings of IRCOBI Conference, Munich, Germany, 2002.

[3] Iraeus, J., and Pipkorn, B.: "Development and Validation of a Generic Finite Element Ribcage to be used for Strain based Fracture Prediction", Proceedings of the IRCOBI Conference, Florence, Italy, 2019.

[4] Östh, J., Bohman, K., Jakobsson, L.: "Evaluation of Kinematics and Restraint Interaction when Repositioning a Driver from a Reclined to Upright Position Prior to Frontal Impact using Active Human Body Model Simulations", Proceedings of the IRCOBI Conference, Munich, Germany, 2020.

[5] Global Human Body Models Consortium (GHBMC), https://www.elemance.com/models/

[6] Gayzik, F.S., Moreno, D.P., Vavalle, N.A., Rhyne, A.C., Stitzel, J.D.: Development of a Full Human Body Finite Element Model for Blunt Injury Prediction Utilizing a Multi-Modality Medical Imaging Protocol, IRCOBI, 2012.

[7] https://projectvirtual.eu/

[8] John, J.D., Iraeus, J., Svensson, M.Y.: A Framework for Continuous Integration in Human Body Finite Element Model Lineup, Proceedings of the XXVIII Congress of the International Society of Biomechanics (ISB), Stockholm, Sweden, 2021.

[9] Forman, J., Poplin, G.S., Shaw, C.G., McMurry, T. L., Schmidt, K., Ash, J. and Sunnevang, C.: : "Automobile injury trends in the contemporary fleet: Belted occupants in frontal collisions", Traffic Injury Prevention, DOI: 10.1080/15389588.2019.1630825, 2019.

[10] Pipkorn, B., Iraeus, J., Björklund, M., Bunketorp, O. et al.: "Multi-Scale Validation of a Rib Fracture Prediction Method for Human Body Models", Proceedings of the IRCOBI Conference, Florence, Italy, 2019.

[11] Mroz, K., Östling, M., Richardson, R., Kerrigan, J. et al.: "Effect of Seat and Seat Belt characteristics on the Lumbar Spine and Pelvis Loading of the SAFER Human Body Model in reclined Postures", Proceedings of the IRCOBI Conference, Munich, Germany, 2020.

[12] Gepner, B.D., Draper, D. et al.: Comparison of Human Body Models in Frontal Crashes with Reclined Seatback. Proceedings of the IRCOBI Conference, Florence, Italy, 2019.

[13] Östh, J., Medoza-Vazquez, M., Sato, F., Svensson, M.Y., Linder, A. and Brolin, K.: "A Female Head-neck Model for Rear Impact Simulations" Journal of Biomechanics 51: 49–56, 2017.

[14] https://projectvirtual.eu/wp-content/uploads/2020/11/2020-11-20_Klug_et_al_VIVAVRU.pdf

[15] OSCCAR Project Deliverable D2.5 "Validation and Demonstration of Advanced Passenger Protection Principles", 2021, www.osccarproject.eu, (In press).

[16] Park, J., Ebert, S.M., Reed, M.P.: "Statistical Models for Predicting Automobile Driving Postures for Men and Women Including Effects of Age", Human Factors, DOI: 10.1177/0018720815610249, 2016.

[17] Mroz, K., Östling, M., Klug, C., Höschele, P. and Lubbe, N.: "Supplementing Future Occupant Safety Assessments with Severe Intersection Crashes Selected Using the SAFER Human Body Model." SAE International Journal of Transportation Safety – Special Issue on Occupant Protection & Crashworthiness for ADS-Equipped Vehicles, USA, 2021. In Press.

[18] Uriot, J., Potier, P., et al.: Reference PMHS Sled Tests to Assess Submarining. Stapp Car Crash Journal. Vol 59, 2015.

[19] Forman, J.L., Kent, R.W., Mroz, K., Pipkorn, B. et al.: "Predicting Rib Fracture Risk with Whole-Body Finite Element Models: Development and Preliminary Evaluation of a Probabilistic Analytical Framework", 56th AAAM Conference, October 14-17, Vol 56, October 2012.

[20] Iraeus, J.: "Stochastic finite element simulations of real life frontal crashes: With emphasis on chest injury mechanisms in near-side oblique loading conditions", PhD Thesis, Faculty of Medicine, Department of Surgical and Perioperative Sciences, Surgery, Umeå University, Sweden, 2015.

[21] Iraeus, J., & Lindquist, M. (2020). Analysis of minimum pulse shape information needed for accurate chest injury prediction in real life frontal crashes. International Journal of Crashworthiness. doi:https://doi.org/10.1080/13588265.2020.1769004
[22] Reed, M. P., Ebert, S. M.: "Effects of Recline on Passenger Posture and Belt Fit". University of Michigan, Ann Arbor, Transportation Research Institute, 2018.

9 Appendix

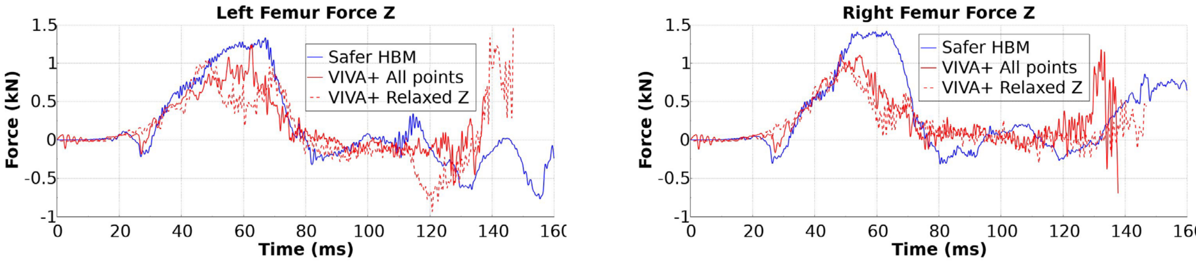

Fig.18: Axial forces on left and right femur for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (dashed red).

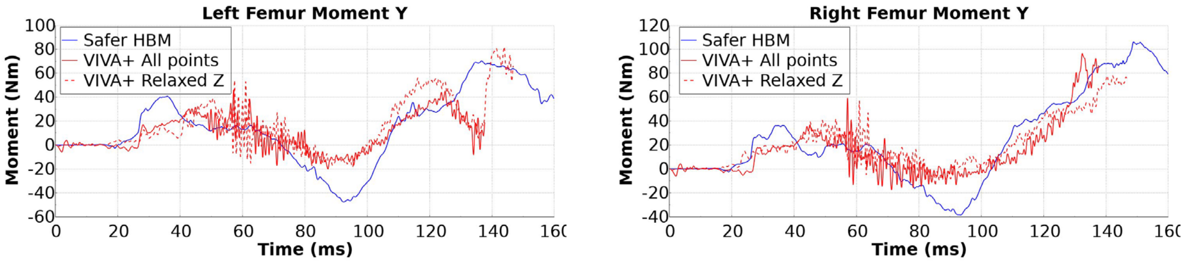

Fig.19: Flexion moments on left and right femur for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (dashed red).

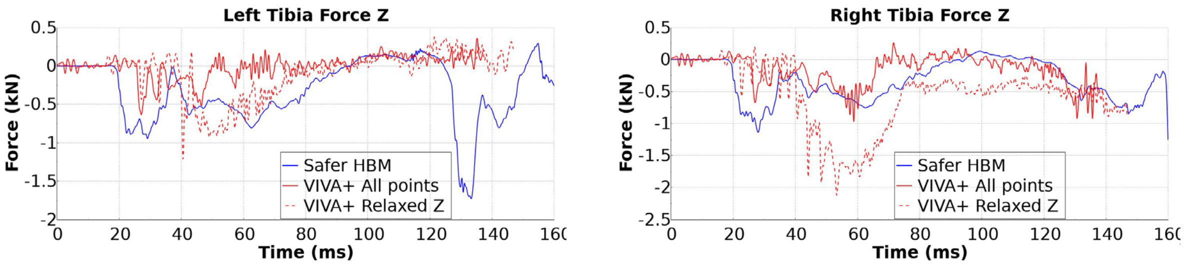

Fig.20: Axial forces on left and right tibia for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (dashed red).

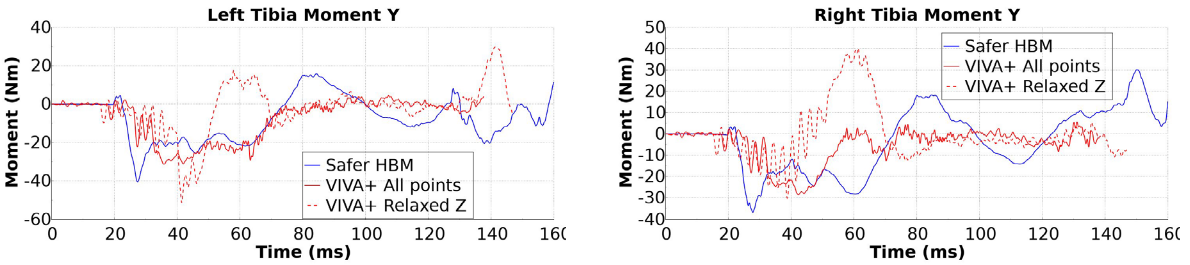

Fig.21: Flexion moment on left and right tibia for SAFER HBM (blue), VIVA+ with all points presim (red), and VIVA+ with relaxed z presim (dashed red).