# Frontal Loading Parametric Sensitivity Analysis on the 50th Percentile Male HBM Connect™

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***Abstract*** This study aims to analyze the sensitivity of the 50th percentile male finite element human body model HBM Connect™, which will be referred to hereafter as HBMC50M, under frontal crash loading. Five selected boundary conditions and restraint system parameters – pelvis position, belt path, crash severity, load limiter force, and pretensioner time-to-fire (TTF) – were varied at two levels (one higher and one lower) within the realistic ranges chosen for the total of 11 simulations, including the baseline. The responses were assessed for the head injury predictors (HIC, BrIC, DAMAGE) and thorax injury predictors (rib strain and anthropometric test device (ATD) analogous chest deflection) metrics. The HBMC50M exhibited head injury predictors varying within a 10% range compared to the baseline. The maximum principal rib strains and ATD analogous chest deflections varied under 30% and 10% compared to the baseline, respectively. For all simulated cases, the absolute maximum principal rib strains remained below 1%, indicating no increased rib fracture risk, which consequently is a function of the injury risk curve. The HBMC50M demonstrated lower sensitivity for the head and thorax injury predictors compared to recently published data for the Total Human Model for Safety (THUMS). The study also highlights the outlook of strain-based injury metrics in the context of product development under the virtual testing framework.

***Keywords*** HBM Connect™, virtual testing, sensitivity, thorax injury, head injury.

## I. INTRODUCTION

Virtual testing (VT) is becoming increasingly important in the field of vehicle safety, with a focus on enabling the integration of human body models (HBMs) together with anthropometric test device (ATD) models to enhance the realism and robustness of safety assessments [1]. This approach supports the development of advanced occupant protection systems and robust vehicle designs.

HBMs have primarily been used as research tools so far, and their applications in product development remain limited. As HBMs are new to product development, engineers lack enough experience implementing them in a practical application, unlike ATDs. The human body model for virtual testing (HBM4VT) group is working on a qualification process to assess the biofidelity of multiple HBMs in harmonised ways, fostering greater confidence in their adoption. One crucial aspect in leveraging HBMs in virtual testing and product development is to perform sensitivity analyses to assess how small variations in boundary conditions and restraint system parameters influence HBM responses. By providing detailed insights into model sensitivities, these analyses could build confidence within the vehicle design community and guide the optimisation of restraint systems for consistent, superior performance in various crash conditions.

Furukawa and Eggers [2] conducted a sensitivity analysis of the Total Human Model for Safety (THUMS) AM50 (American Male 50th percentile), Version 7.0, under the frontal loading condition. The normalised variation of injury probability reached 72% and 122% in the head injury criterion (HIC15) and strain-based AIS3+ rib fracture risk predictions, respectively. This emphasises the importance of sensitivity analysis with other available HBMs.

The HBM Connect™ family of HBMs (developed by Humanetics Group) includes the 50th percentile male (HBMC50M) and the 5th percentile female (HBMC05F) anthropometries, with a focus on virtual testing and product development. This paper analyses the sensitivity of HBMC50M V1.1 as part of [2]'s sensitivity analyses in the same frontal loading condition.

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## II. METHODS

Furukawa and Eggers [2] employed the vehicle environment model consisting of a generic rigid seat setup, which was validated against sled tests conducted at Bundesanstalt für Straßenwesen (BASt). The setup included a three-point seatbelt system equipped with a pretensioner and load limiter, a pre-deployed driver airbag, and a foam block representing knee bolster.

The HBMC50M was positioned on the rigid seat, as illustrated in Fig. 1(a), with the seatbelt routing performed using PRIMER (Ove-Arup). The crash simulation was conducted using a 50 kmph Mobile Progressive Deformable Barrier (MPDP) generic pulse [3] (Fig. 1(d)), representing as the baseline case.

To evaluate model sensitivity, five parameters were selected for variation: three related to boundary conditions (occupant initial pelvis position, belt path, crash severity), and two associated with the restraint system (load limiter force and pretensioner time-to-fire (TTF)). Each parameter was varied across two levels (upper and lower limits) based on ranges derived from the European New Car Assessment Program (Euro NCAP) MPDB protocol for realistic scatter observed in sled testing. In total, 11 simulations were performed, comprising one baseline and 10 with parameter variations (five parameters at two levels each). Figure 1 provides a detailed illustration of the baseline setup and all parameter variations. All the simulations were performed with LS-DYNA (Ansys Inc.) single precision explicit solver (mpp971_s_R9.3).

To analyse the responses, three parameters were used to evaluate head injury: (1) Head Injury Criterion 15 milliseconds (HIC15), (2) Brain Rotational Injury Criterion (BrIC), and (3) Diffuse Axonal Multi-Axis General Evaluation (DAMAGE). For chest loading, the maximum principal strain (MPS) in the ribs was analysed, and the MPS values were further used to estimate the probability of AIS3+ rib fracture risk based on the method proposed by Larsson *et al*. [4]. Chest deflections were calculated using measurement springs defined in the HBMC50M analogous to the HIII-50M chest pot and four IR-TRACCS in THOR-50M ATDs (see Fig. A1 in the Appendix). These outputs were compared for all 11 simulations to analyse the influence of parameter variations on the HBMC50M responses.

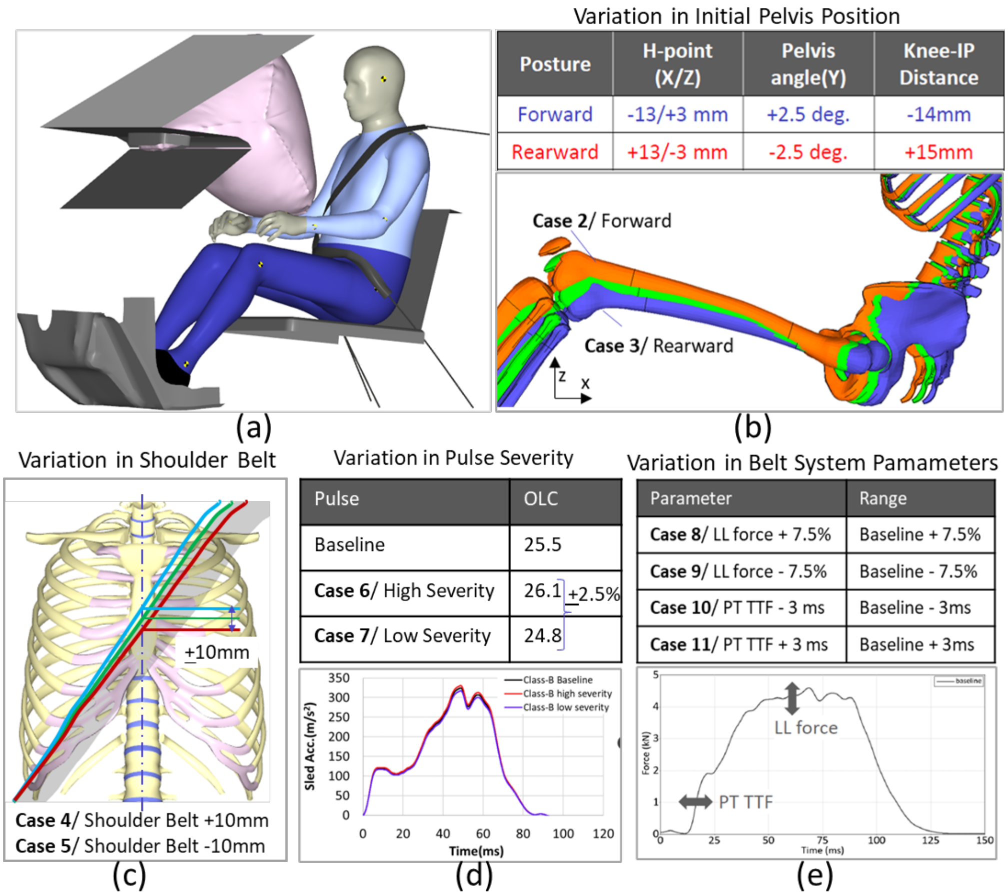
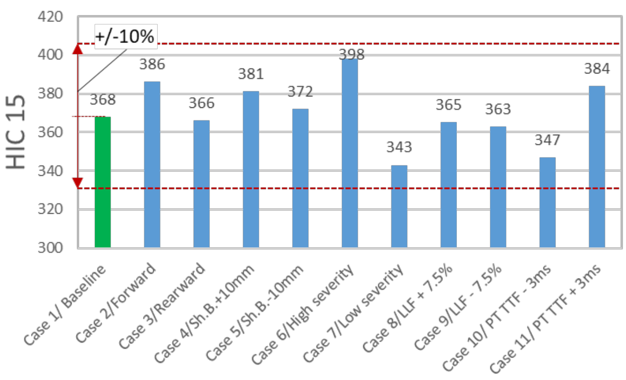


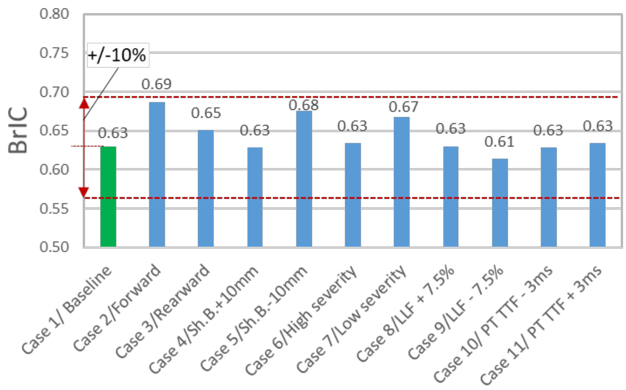

Fig. 1. (a) Simulation setup showing HBMC50M in the sled environment, (b) variation for the pelvis position, (c) variation of the shoulder-belt path, (d) variation of the pulse severity, and (e) variation of the load limiter level and pre-tensioner TTF.

## III. RESULTS

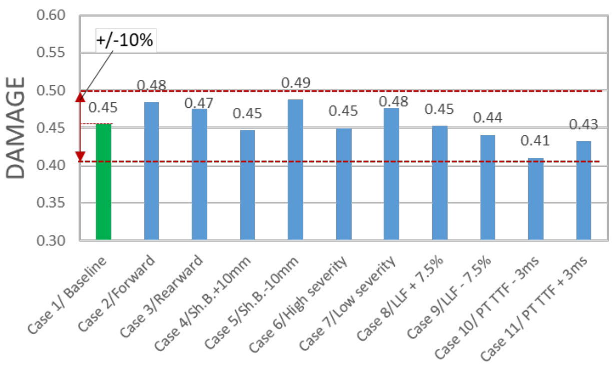
All simulations were completed successfully without any numerical issues. A comparison of head injury metrics across all simulated cases relative to the baseline case is presented in Fig. 2. All head injury metrics (HIC, BrIC, and DAMAGE) exhibited variations within 10% of the baseline case. The baseline HIC was 368, with values ranging from 343 to 398 across simulations. Similarly, the baseline BrIC was 0.63, varying between 0.61 and 0.69, while the DAMAGE, with a baseline value of 0.45, ranged from 0.41 to 0.49.




(a)




(b)




(c)

Fig. 2. Comparison of head injury predictors across all simulated cases: (a) HIC 15, (b) BrIC, and (c) DAMAGE calculated in HBMC50M.

Figure 3 (a–e) compares MPS in the ribs from HBMC50M across all simulations to the baseline. In general, rib MPS trends for most simulated variations were consistent with the baseline. However, notable deviations were observed in rib R1, where, compared to the baseline, the MPS was approximately 50% higher in case 4 (higher shoulder-belt position), and cases 6 and 7 (variations in the pulse severity). Despite these deviations, the elevated strains were primarily confined to ribs along the belt path, and the absolute MPS values remained consistently below 1%. The MPS values were further analysed to estimate the probability of AIS3+ rib fracture risk using the methodology employed by [4]. Since all MPS values were below 1%, the calculated AIS3+ fracture risk was zero for all cases.

Figure 3(f) presents the chest deflection in HBMC50M at the sternum analogous to HIII-50M chest pot for all cases. At the sternum of HBMC50M, the baseline deflection value was 56.5 mm, and it ranged between 54.4 mm and 58.2 mm (within 5%) across all simulations. The chest deflections at the other four locations analogous to the THOR-50M IR-TRACC deflections are presented in the Appendix, Fig. B1. Notable deviations were observed at the lower left spring (Fig. B1) in simulations involving load limiter variations (case 8 and case 9). Compared to the baseline value of 30.3 mm, an increase in load limiter force resulted in a 14% higher deflection (34.4 mm), while a decrease in load limiter force reduced the deflection by 13% (26.5 mm).

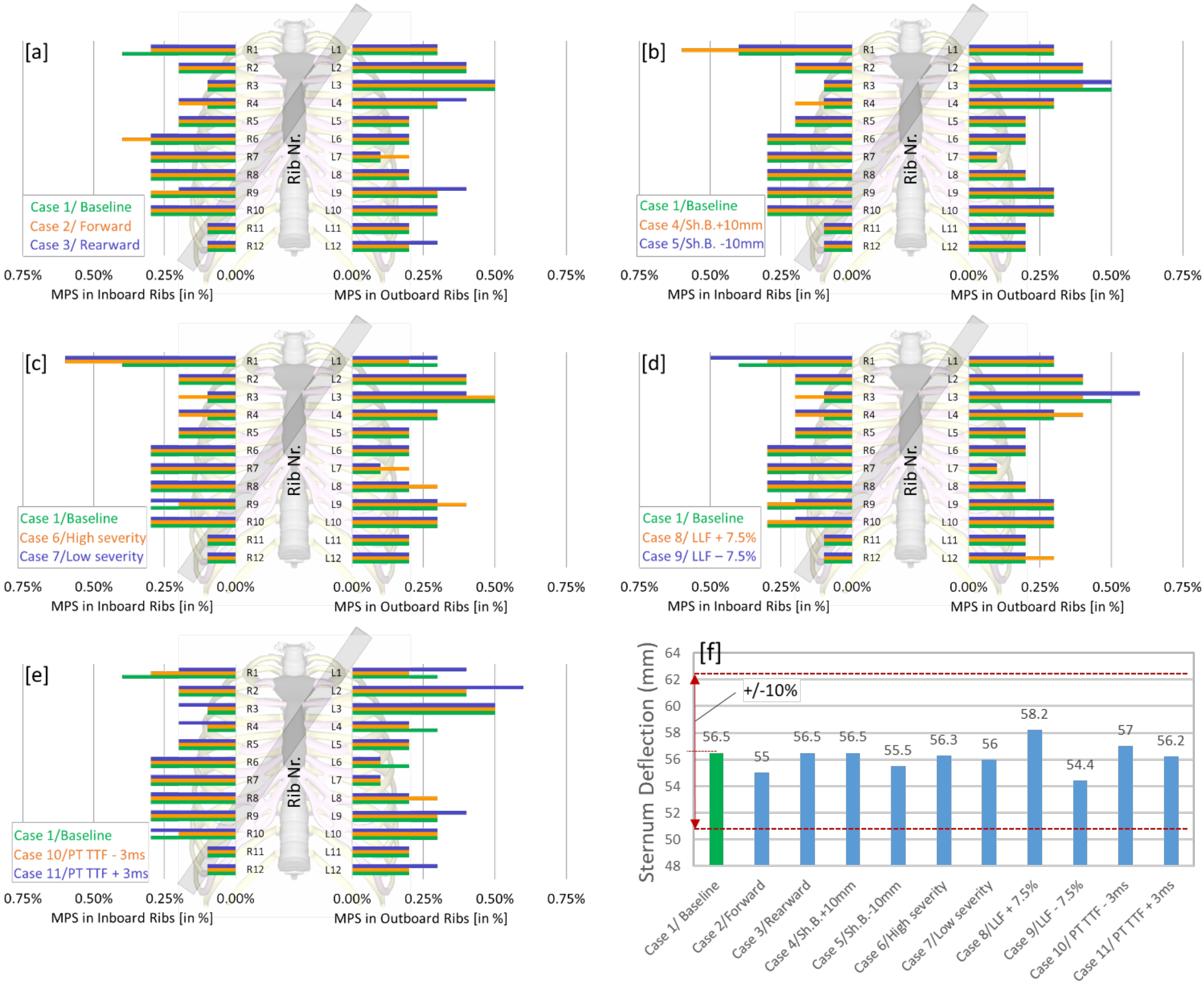



Fig. 3. Comparison of the rib MPS in all simulated cases with the baseline case, categorised as follows: (a) variations of pelvis position, (b) variation of shoulder-belt path, (c) variation of pulse severity, (d) variations of load limiter force levels, (e) variation of pretensioner activation times (TTFs), and (f) comparison of HIII-50M analogous sternum deflection for all the cases with the HBMC50M.

## IV. DISCUSSION

The HBMC50M demonstrated lower sensitivity across all head injury predictors (HIC15, BrIC and DAMAGE) for the simulated variations in the current study. The observed differences were within 10% of the baseline values. While calculating the normalised variation for HIC15, the maximum variation was 15% for HBMC50M (see Fig. C1, Appendix). Therefore, the HBMC50M response is considered robust, emphasising its reliability under the tested conditions.

For chest loading, represented by the MPS in the ribs, the overall strain distribution remained consistent with the baseline case across all parameter variations. While the classical belt-loading pattern on the ribs was evident, additional effects from airbag and the upper legs during forward movement were observed, especially in the lower outboard ribs. Most parameter variations resulted in MPS differences within 30% of the baseline. Despite these differences, the absolute MPS values stayed well below 1% in all cases, indicating no increase in AIS3+ rib fracture risk compared to the baseline. This resulted in zero normalised variation for HBMC50M. Direct comparisons of MPS between the current study to Furukawa and Eggers [2] were limited due to the unavailability of rib MPS data for THUMS.

The most notable deviations in MPS were observed in the non-belted side upper rib R1. Under both high and low pulse severity conditions, rib R1 exhibited a 50% higher strain compared to the baseline (0.6% vs 0.4% in the baseline). Initially, this was thought to be a non-physical artifact, potentially caused by strain concentration near the sternum or spinal joints because rib R1 was not directly loaded by the seatbelt. However, further investigation revealed that the affected element was at the mid-section of rib R1. The analysis suggested that rib R1 experienced a jerking motion induced by surrounding connective tissues, leading to increase in localised strains. This observation highlights the complexities of using strain-based metrics for chest injury assessment. The chest loading response could easily be influenced by the intricate interaction of ribs with surrounding tissues, which can alter load paths and affect local strain distribution. These complexities must be accounted for in interpreting strain-based injury assessment with the HBMs.

There are a few thought-provoking considerations from a product development perspective about the application of strain-based injury assessment for rib fracture risk prediction.

1) Averaging Strain Values: instead of relying solely on the MPS of a single element, averaging strain over a cluster of elements within a defined area may provide a more realistic outcome and rule out potential outliers.
2) Adopting Higher Variation Margins: recognising that rib strain variation can be inherently influenced by complex biomechanical interactions and adopting higher variation margins for rib strain could improve the reliability of the analysis if these variations are not detrimental in altering rib fracture risk prediction.
3) Accounting for Strain Direction: bone fractures are primarily caused by tensile strains. Differentiating between tensile and compressive strains and incorporating triaxiality (a measure of the stress state) into the analysis may yield further improvements.
4) Using Chest Deflection-Based Metrics: metrics such as those used in ATD evaluations, which rely on chest deflection instead of strain, can offer a less sensitive and more comprehensive measure of chest injury.

The study must be further expanded to include multiple load cases and configurations (such as frontal, side, far-side) to overcome the limitation of analysis based on only one frontal load-case. The head injury analysis includes kinematic-based predictors for the current study only with respect to HBMC50M. For the rib injury risk, the HBMC50M uses the Larsson [4] methodology without calibration to have a model-specific adaptation of injury risk function. Such function is primarily expected to shift in strain threshold more than the shape change of risk function. It may be important to further investigate rib fracture predictions in load-cases where MPS values are closer to rib fracture thresholds based on calibrated risk function.

## V. CONCLUSION

The HBMC50M demonstrated better robustness and reliability with significantly lower sensitivity across all head injury metrics and chest loading assessments. Similar analysis must be conducted across multiple HBMs, and an engineering judgement in terms of expected variation range must be applied to identify the applicability of a specific HBM in virtual testing and product development. Additionally, the study highlights the outlook on strain-based injury metrics, emphasising the need for further investigation before implementation to virtual testing framework.

## VI. ACKNOWLEDGEMENTS

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## VII. REFERENCES

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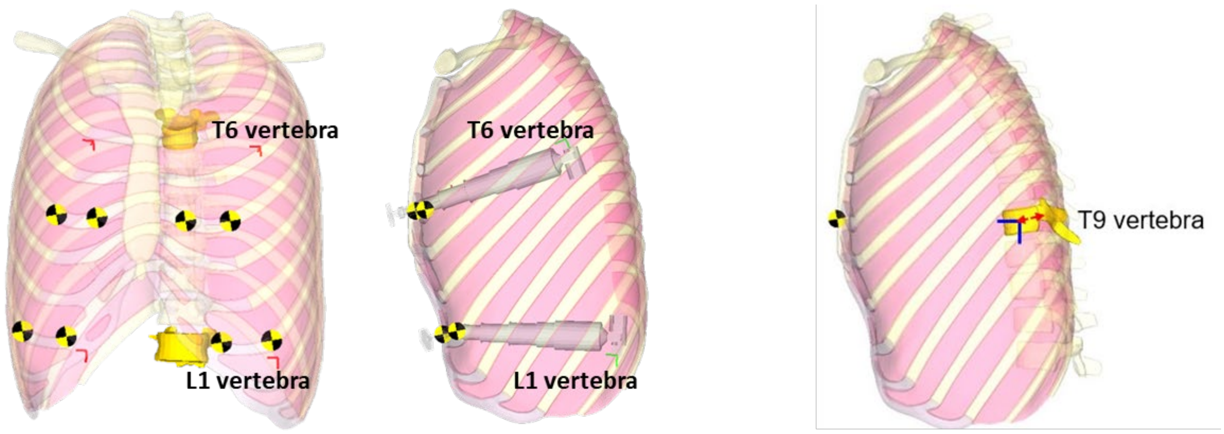
## VIII. APPENDIX

### Appendix A: Details of ATD analogous chest deflection measurements in HBMC50M




(a) (b)

Fig. A1. Chest deflection measurement in HBMC50M: (a) at four locations (upper left, upper right, lower left and lower right) to match IR-TRACC locations in THOR50M, and (b) at sternum location to match to HIII-50 chest pot measurement.

### Appendix B: Chest deflection in HBMC50M at four locations (upper left, upper right, lower left and lower right) across all simulated cases

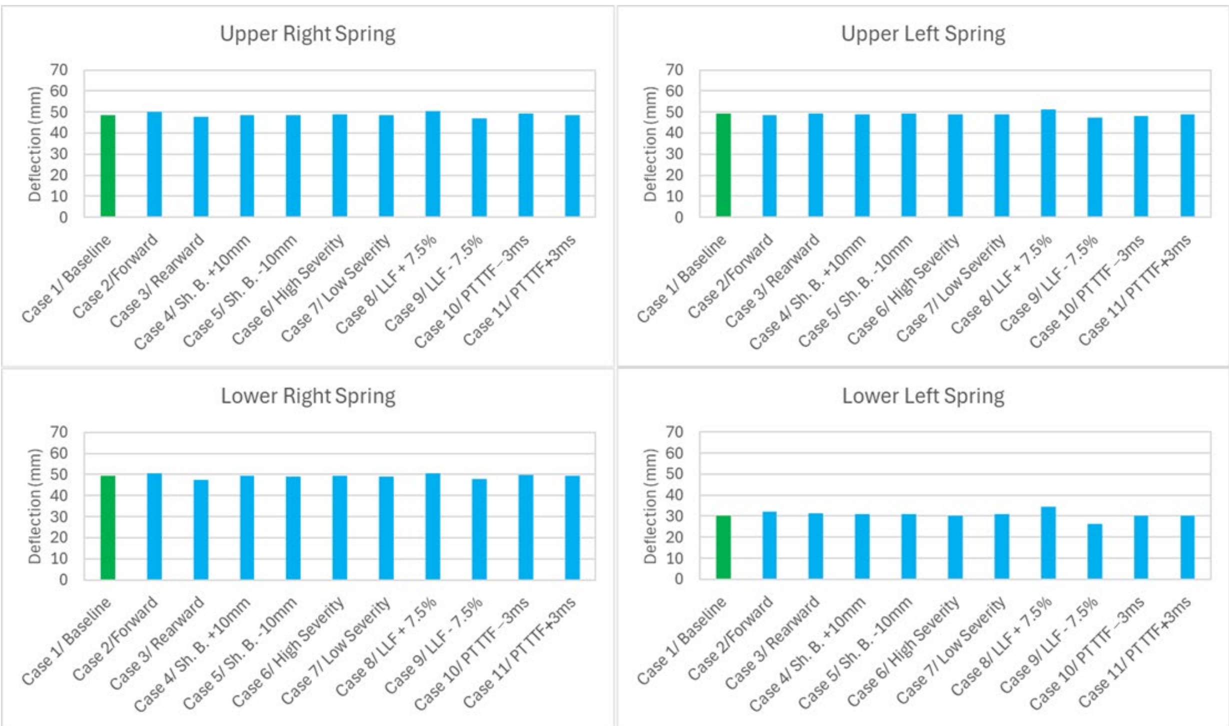



Fig. B1. Comparison of chest deflection in HBMC50M across all simulated cases with the baseline case. These deflections from four locations correspond to the four IR-TRACC locations in THOR50M.

**Appendix C: Comparison of normalised variations for HIC15 from HBMC50M for all cases**

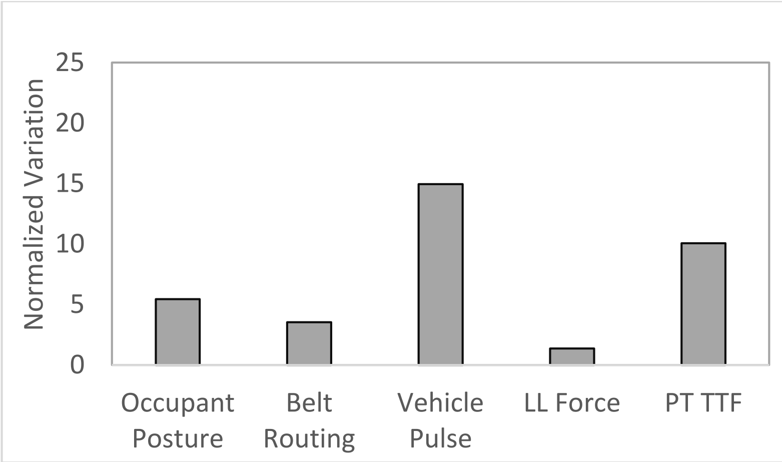



Fig. C1. Comparison of normalised variations for HIC15 from HBMC50M for all simulated cases.