

Validation of Detailed Organ Modularity in a Simplified Human Body Model

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Abstract The high degree of computational resources required to simulate finite element human body models has prompted the implementation of a method that incorporates detailed components into a simplified model. Previous studies have demonstrated this ability by incorporating the GHBMC M50-O brain model into the simplified model (GHBMC M50-OS+B), which allowed for localised analysis of the brain at a substantially reduced computational cost. We expand on this concept through the modular incorporation of detailed thoracoabdominal organs into the simplified model (M50-OS+O), comparing force-deflection and organ level response through an array of validation tests. This consisted of a frontal thoracic hub, oblique hub, abdominal bar, and frontal sled simulation. Organ response metrics included 95th percentile strain, internal energy, and organ kinematics and were compared between the M50-O and the M50-OS+O. Generally, the M50-OS+O was able to capture the response of the M50-O in both force-deflection and organ level response, but in one-tenth of the time.

Keywords biomechanics, computational modelling, human modelling, injury, organ

I. INTRODUCTION

Finite element human body models (HBMs) are one of the few tools available to investigate injury mechanisms and tolerance of the human body, particularly with regard to soft tissues and organs [1]. The Global Human Body Model Consortium (GHBMC) is one such model currently utilised for vehicle crash response. The GHBMC is an industry-sponsored and government supported consortium with the aim of developing computational human models for the blunt injury environment.

The GHBMC average male occupant (M50-O v4.4) finite element model was used in this study. The development of this HBM was based on a multi-modality medical image and external anthropometry dataset of a volunteer representing a 50th percentile male in terms of height (174.9 cm) and weight (78.6 ± 0.77 kg), described by [2,3]. The model has since undergone numerous validation simulations at both the regional e.g. [4-7] and full body levels e.g. [8-10]. Additional information on the development of the model can be found in the GHBMC M50 user manual [11].

The significant computational resource requirement to execute the detailed M50-O has motivated the development of a simplified average male occupant model (M50-OS v1.8.4). The M50-OS responds similarly to the M50-O during blunt impact loading scenarios but in a shorter amount of time with less computational requirements. The M50-OS retains the same habitus and rigid bone structures as the detailed model but has reduced mesh density and greatly simplified soft tissue structures. For example, the individually defined organs of the M50-O were homogenised into thoracic and abdominal cavities, and the abdominal muscles were replaced with a single, thick layer of flesh [12]. Previous studies have demonstrated that the M50-OS has the ability to run approximately 32 times faster than the M50-O [13].

Previous studies have demonstrated the ability to modularly incorporate the validated GHBMC M50-O brain model into the simplified model (M50-OS+B), which allowed for localised analysis of the brain at a substantially reduced computational cost [13-15]. We expand on this concept through the modular incorporation of detailed thoracoabdominal organs into the simplified model (M50-OS+O), comparing force-deflection and organ level response through an array of validation tests.

II. METHODS

Detailed Organ Modular Incorporation

The M50-O and M50-OS models were based on the same CAD geometry. Thus, the internal thoracoabdominal cavity morphometry between the two models is identical, but the modelling approach used in each varies significantly. Like its detailed counterpart, the thoracoabdominal cavity of the M50-OS was modelled to tightly fit within the boundaries of the rib cage, spine, pelvic rim, and the surrounding abdominal flesh. Contact definitions are required to hold the cavity in place since nodal connections to these components are not used. This internal structure (Fig. 1B) is defined by a mediastinum region with adjacent lung regions bilaterally and a single abdominal cavity inferiorly, all continuously meshed [12]. This cavity was designed as a placeholder to replicate the force-deflection response of the M50-O, with minimal computational cost. The detailed model on the other hand, explicitly includes meshes of all major thoracoabdominal organs (Fig. 1A); these fit within the same compartment. This similarity of shape and the use of contacts to link the cavity to its surrounding structures, allows for modular inclusion of the detailed model into the simplified (Fig. 1C). More details on the development and validation of the simplified model can be found in [12,15]. In this study, all thoracic and abdominal organs from the detailed GHBMCM50-O v. 4.5 model were incorporated into the M50-OS model to replace the existing homogenised cavity, to create the M50-OS+O.

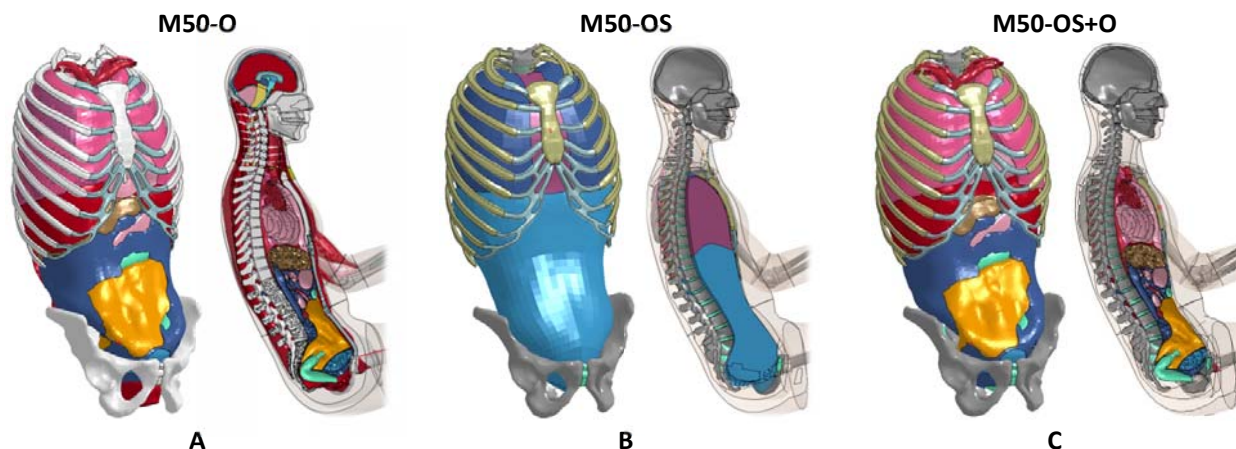


Fig. 1. Comparison of the thoracoabdominal cavities between the M50-O, M50-OS, and M50-OS+O. Isolated views of the cavities within the surrounding bony structures and sagittal cross sections are shown to illustrate differences between the models.

However, intersections between the surrounding flesh and newly incorporated detailed organs were present at regions where the simplified organ cavity underestimated the size of the detailed organs. This spurred a refit of the existing organ cavity to better match the exact geometries of the modular detailed organs, which in turn would expand the flesh and remove the problematic intersections. This simplified model with the refit organ cavity has been tentatively termed the M50-OS v2.0beta and has been referenced as the M50-OS for the remainder of the study. The detailed organs were then incorporated into the M50-OS without these problematic intersections. Additionally, a one way surface to surface tiebreak contact was incorporated into the M50-OS+O between the thoracic organs and the ribcage. This contact was introduced to eliminate the manifestation of dynamic voids during simulation which was observed during motion such as gravity settling. A comparison of the three models in this study can be found in Table I, which includes total number of nodes and elements, as well as the governing time step for each model. The total mass added from this governing time step did not exceed 1% for any of the models.

TABLE I
COMPARISON OF THE M50-O, M50-OS, AND M50-OS+O MODELS IN TERMS OF NUMBER OF NODES, ELEMENTS, AND GOVERNING TIME STEP.

	Nodes	Elements	Governing Time step
<i>M50-O</i>	1,263,445	2,187,575	3.00E-04
<i>M50-OS</i>	296,324	358,649	1.00E-03
<i>M50-OS+O</i>	481,521	643,823	1.00E-03

Simulation Test Setups

All models were simulated on the Distributed Environment for Academic Computing (DEAC) high performance computational cluster at Wake Forest University using 48 CPUs and LS-DYNA R.7.1.2 (LSTC, Livermore, CA, USA). The functionality of this modular organ approach was evaluated through a series of multi-directional hub impacts, and a single frontal sled simulation to replicate that of a common vehicular impact. The force-deflection responses between the M50-O, M50-OS, and M50-OS+O were compared through a test matrix of four hub-style biomechanics impacts, consisting of a 6.7 m/s frontal thoracic hub [16], 6.7 m/s oblique hub [17], and 5.5 m/s abdominal bar [18], as well as an 11.8 m/s frontal sled pulse [19]. A total of 12 simulations were performed.

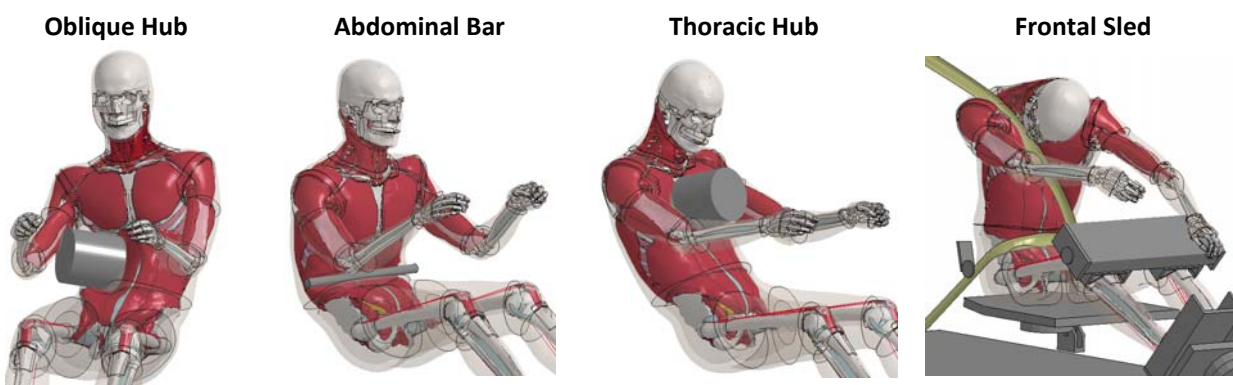


Fig. 2. Visual description of the various simulations used to evaluate the modular organ method. M50-O is shown.

Organ-level response was also compared between the M50-OS+O and the M50-O. This included comparison of peak internal energy, 95th percentile first principle strain, and three-dimensional organ displacement. The organs of interest for comparison were the liver, spleen, pancreas, left and right kidney, and heart.

III. RESULTS

Normalised run times for the various models can be seen in Fig. 3. This is a simulation rate comparison of minutes per real time to calculate one ms of simulation.

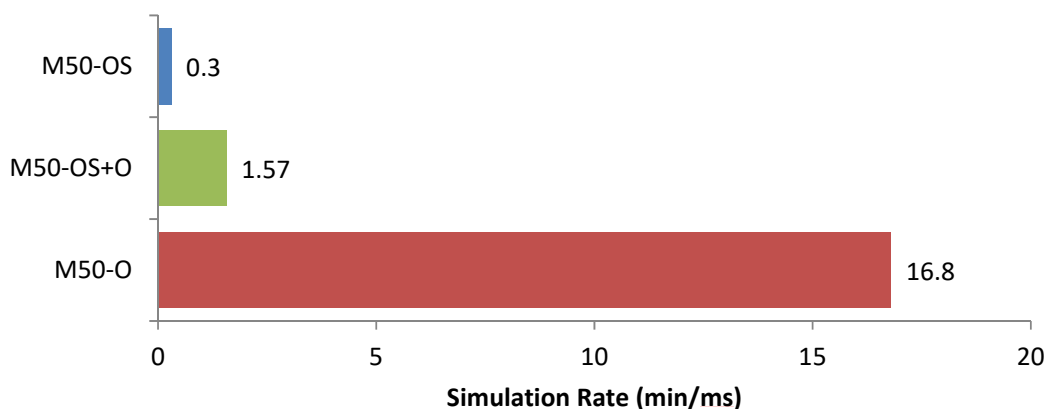


Fig. 3. Simulation rate comparison of the M50-O, M50-OS, and M50-OS+O.

Force-deflection response was compared between the M50-O, M50-OS, and M50-OS+O where experimental corridors were available, shown in Fig. 4. This includes the abdominal bar, frontal chest, and oblique hub. The dotted black lines of the corridors represent the upper, lower, and average PMHS response. Force values are given in kilonewtons (kN) and chest deflection was defined as percentage of the initial chest depth. Measurements used in determining percent deflection varied between the hub impacts. The thoracic hub impact used the distance from the posterior surface of flesh adjacent to the eighth thoracic vertebrae (T8) to the anterior surface of the body normal to T8 [20,21]. Compressible depth for the abdominal bar was defined as the distance from the anterior surface of the third lumbar vertebrae (L3) to the anterior surface of the body normal to L3 [18,22,23]. The off axis nature of the oblique hub made it difficult to quantify this measure, therefore, deflection in centimetres was used in that case. This was defined as the deflection from the point of impact on the right oblique flesh to the flesh opposite along the normal of the impactor [24].

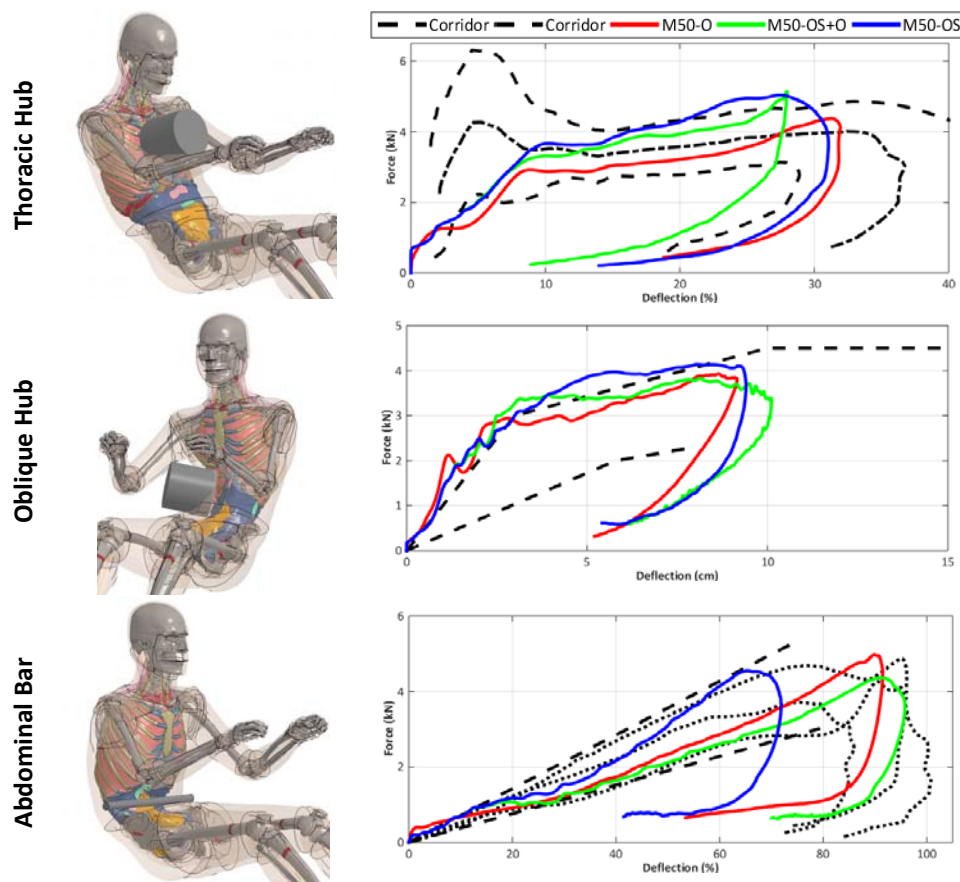


Fig. 4. Force-deflection comparison between the M50-O and M50-OS+O for the abdominal bar and thoracic hub impacts. The M50-OS+O is shown in the pictures on the left. The thoracic hub corridors are from the study of [16-18,21].

Peak displacement, percent chest displacement, and peak forces were compared between the M50-O, M50-OS, and M50-OS+O through the oblique hub, abdominal bar, and thoracic hub impacts, shown in Table II.

TABLE II
COMPARISON OF THE PEAK DISPLACEMENTS AND FORCE OF THE M50-O AND M50-OS+O.

	Thoracic Hub			Ab Hub			Ab Bar		
	M50-O	M50-OS+O	M50-OS	M50-O	M50-OS+O	M50-OS	M50-O	M50-OS+O	M50-OS
Deflection (mm)	76.8	66.4	73.7	92.2	101.7	94.5	117.1	122.6	92.0
Deflection %	31.9	30.0	31.1	~	~		91.5	95.8	71.9
Peak Force (kN)	4.4	5.2	5.0	3.9	3.8	4.1	5.0	4.4	4.6

~ Not calculated due to lack of standardised oblique length.

The risk of injury to thoracic structures during the frontal sled test was quantified using the chest deflection criteria used in [25-27]. The deflection criterion relates the maximum chest deflection to AIS injury risk by risk curve equations. The percentage of AIS3+ was evaluated for the M50-O, M50-OS, and M50-OS+O during the thoracic hub impact and frontal sled and can be seen in Table III.

TABLE III
EVALUATION OF THE M50-O AND M50-OS+O THROUGH THE CHEST DEFLECTION CRITERIA FOR THE FRONTAL SLED PULSE.

Chest Deflection (mm)	M50-O		M50-OS+O		M50-OS		AIS 3+ Injury Probability (Chest Deflection)
	Model Value	P(AIS 3+)	Model Value	P(AIS 3+)	Model Value	P(AIS 3+)	
Thoracic Hub	76.8	74.4%	66.4	57.7%	73.7	70.0%	
Frontal Sled	54.0	33.7%	43.7	16.9%	55.5	36.6%	

Internal energy of the organs of interest were calculated for the M50-O and M50-OS+O. Internal energy is calculated within LS-DYNA by incrementally summing the product of the six components of stress and strain and element volume. A total time history calculation for each organ is obtained by summing the individual element internal energies [28]. The peak organ internal energy values for both models were compared across the three hub impacts, shown in Fig. 5.

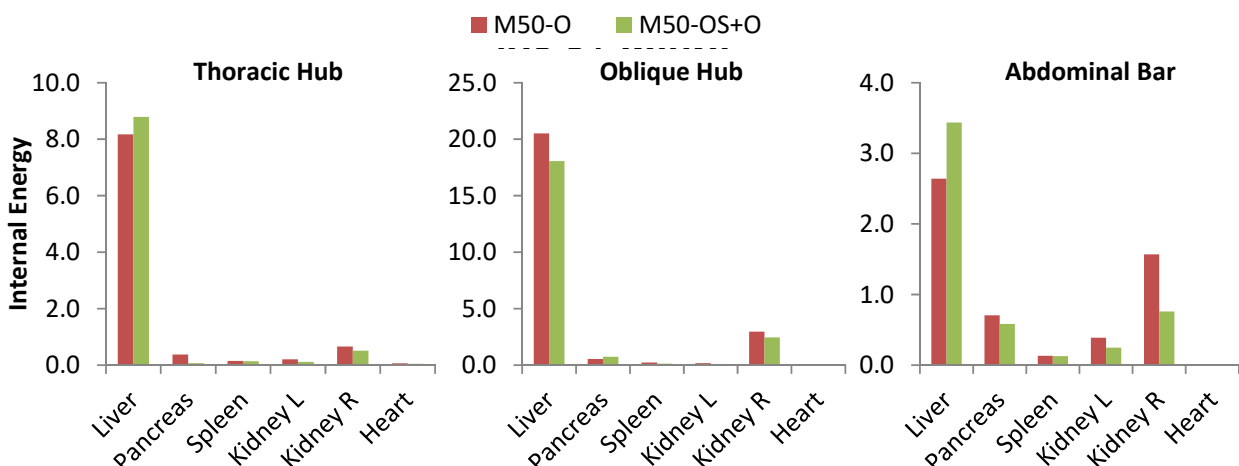


Fig. 5. Graphical comparison of the peak internal energy between the organs of interest during the thoracic hub, oblique hub, and abdominal bar impacts.

Additionally, 95th percentile strain was compared across the same set of organs for the M50-O and M50-

OS+O. The calculation of 95th percentile strain was obtained by ranking all elements of a target organ in increasing order of the peak first principle strain throughout the simulation, and taking the value of the 95th percentile element [29]. A comparison of 95th percentile organ strain between the M50-O and M50-OS+O through the three hub impacts are shown in Fig. 6. The individual 95th percentile strain, as well as a percent difference between the two models, is given for comparison in Table A 2.

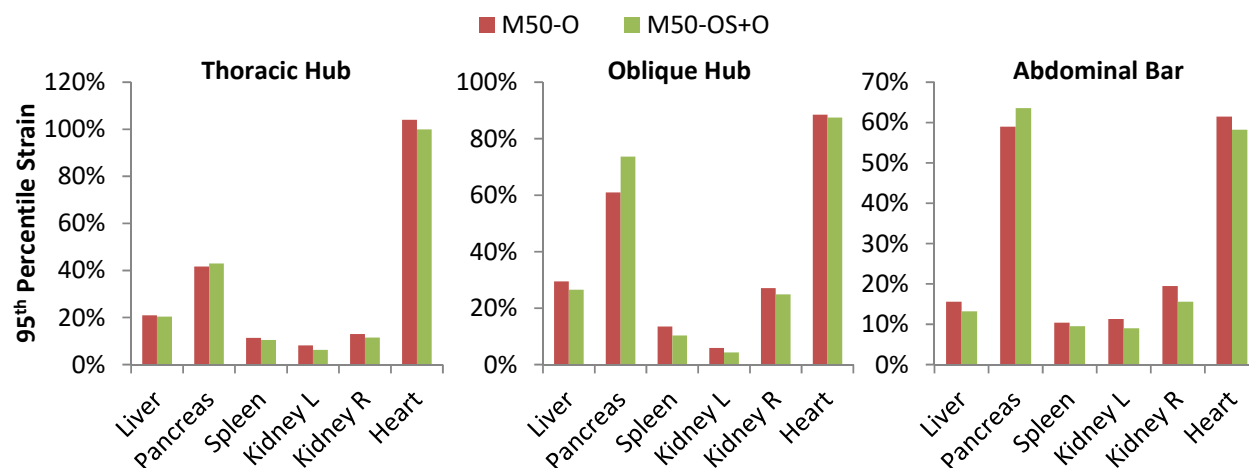


Fig. 6. Graphical comparison of the 95th percentile strain between the organs of interest during the thoracic hub, oblique hub, and abdominal bar impacts.

IV. DISCUSSION

The incorporation of the detailed organs into the M50-OS has shown the ability to obtain abdominal force-deflection response comparable to the experimental data and M50-O response, but with a runtime reduction of ~90%. As we demonstrated, the M50-OS+O had a runtime ratio of 1.57 ms/min vs. 16.8 ms/min of the M50-O. This reduction is achieved through a number of mesh simplification steps and a reduced number of deformable bone components as detailed in previous publications [12,15]. With particular relevance to this study, the rib cage mesh of the M50-OS has 82% fewer elements than the detailed counterpart and the flesh component also was significantly remeshed to have a coarser mesh with 83% fewer elements.

The overall shape of the force-deflection curves improved for the M50-OS+O, most notably during the abdominal bar impact. The results show a mixed effect depending on impact location. The detailed organs are significantly more compliant than the M50-OS organ cavity when looking at the abdominal cavity (as in the bar and oblique impacts). The abdominal bar impact was the only simulation where the M50-O was stiffer than the M50-OS+O. The M50-O has explicitly modelled abdominal muscles which were not included in the M50-OS+O. These muscles with the contact to the flesh and underlying tissue most likely led to the stiffer response compared to the single layer of flesh in the M50-OS+O from this direct impact.

The thoracic impact however showed a marked difference. The decrease in maximum deflection of the thoracic hub impact for the M50-OS+O can be attributed to the differences in the ribs compared to the M50-O. The M50-OS model has a coarser mesh to reduce element count and is not intended to predict rib fracture. We hypothesise that this also plays a role in the differences in chest deflection values. A comparison of the peak values shows that model differences between the M50-OS+O and M50-O for the three hub impacts were <4% for percent deflection, <14% for peak deflection, and <16% for peak force.

It should be noted that when developing the M50-OS model, the goal was to achieve a response as close to the M50-O model as possible. Because explicit organ structures, skeletal muscle, and variation in cortical thickness were not defined in the simplified model, the structural response observed (blue traces, Fig. 4) is a consequence of the modelling approaches selected. These include a greater mesh density than the detailed model, constant cortical shell thickness (0.75 mm), and generalized properties of homogenized organs. The results of this effort was an overall response as close to the corridors as possible. The incorporation of the thoracoabdominal cavity of the detailed model to develop the M50-OS+O appears to have the greatest effect on this tuned response in the thorax cavity, where peak forces remain stable but deflection is reduced by

roughly 5%.

The three models tested have particular components that interact with each particular impactor location. The effect of the insertion of the organs appears to have the greatest effect on the thorax in the anterior-posterior direction, as in the thoracic hub case. The thoracic impact engages a single layer of flesh followed by a reduced-mesh ribcage for both the M50-OS and M50-OS+O. The contents within the ribcage of course vary between these two as one has a simplified cavity and the other has explicitly modelled organs. The M50-O is more complex in this region. It has a layer of flesh, skeletal muscle, and the ribcage mesh is finer than the simplified model. The same organs are within the ribcage for the M50-O and M50-OS+O. Thus there is a trade-off in the simplified model (M50-OS) where the rib cage may be slightly stiffer but the organ cavity is used to achieve a similar response as the detailed model (M50-O). When the detailed organs are inserted into the simplified model (M50-OS+O) the balance is shifted to a more elastic response. In other words, the interplay between these structures is complex and leads to the behaviour observed. This has implications for the use of this model.

The differential in 95th percentile strain comparison, as well as internal energy, varied between test setups and specific organs. Typically, organs exhibited differences at or below 10% for strain. Overall, organ response of the M50-OS+O was consistent across the three hub impacts when compared to the M50-O. The remaining differences between the two models, such as coarser rib and flesh parts are thought to contribute to these remaining differences. Organ displacements were also compared between the two models during the three hub impacts and can be seen in Figure A 1, Figure A 2, and Figure A 3. Generally, the two models were comparable across the three impacts. This further suggests the appropriateness of using the M50-OS+O when analysing organ-level, strain based criteria for injury.

The M50-O and M50-OS closely align in terms chest deflection and thus injury risk. Given the chest deflections in the thoracic hub impact case, both models present a high risk of injury. In fact, both models predict nearly the exact amount of chest deflection permissible within the US NCAP rating for chest injury based on the Hybrid III ATD [30]. These correspond to a greater than 70% risk of AIS 3+ injury. The M50-OS+O model on the other hand displayed less chest compression and thus lower risk of injury. The findings suggest that the incorporation of the modular organs is not advised for calculating correlative injury metrics. Rather, this model should be used for organ based measures exclusively as the strains noted were seen to generally track with the M50-O.

The results from the AIS injury prediction confirms that the M50-OS is adequate at predicting injuries through correlative injury metrics based on kinematics and kinetics. The M50-OS+O provides an extension of the M50-O in that it can be used to study the response of the organs directly within the body and determine strain or internal energy caused by a given insult. As organ based injury metrics develop, this data can be used to potentially provide more localized measures of injury. Currently, this model can be used more efficiently in parametric studies (when compared to the M50-O) to relatively compare strain and other tissue based measures between given simulated events.

V. CONCLUSIONS

Thoracoabdominal soft tissue injuries commonly occur in a vehicular crash yet there are relatively few tools available to study them. Computational models can be used to address this need, but detailed models carry a high computational cost. We have introduced a modular approach to reduce computational cost from 16.8 to 1.57 ms/min simulation time and retain adequate biofidelity in this region. The M50-OS+O is recommended for use in determining organ level strain and other tissue level measures when computational cost conservation is desired. The particular benefit of this approach would compound in cases where large amounts of simulations are required.

VI. ACKNOWLEDGMENT

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VII. REFERENCES

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VIII. APPENDIX

Three dimensional displacement of the organs of interest were obtained by tracking a single node on the organ surface. The chosen node for each organ was kept consistent for both models. Graphical comparisons of these displacements between the M50-O and M50-OS+O are shown in Figure A 1 - Figure A 3.

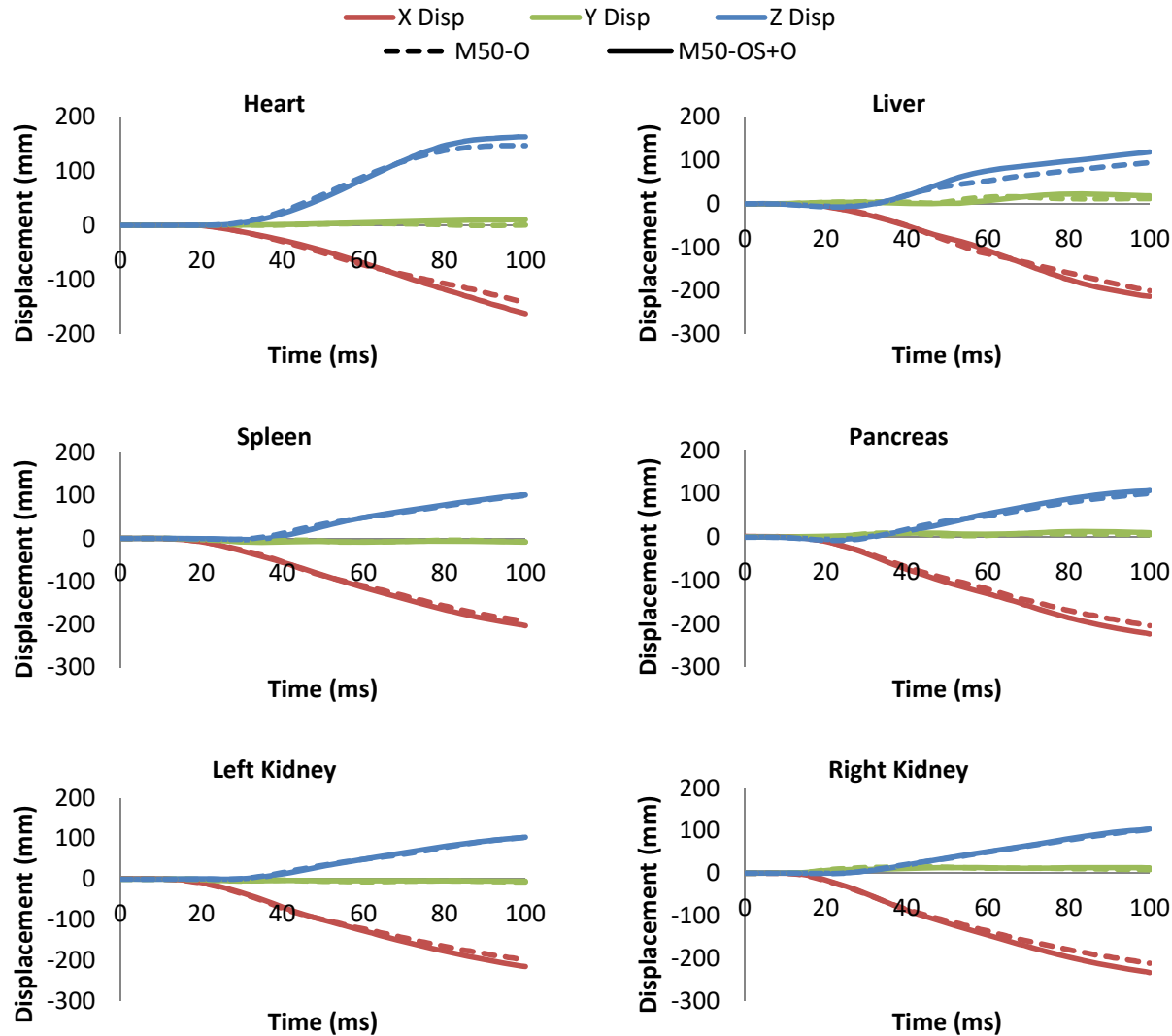


Figure A 1. Organ displacement comparison between the M50-O and M50-OS+O during the abdominal bar impact.

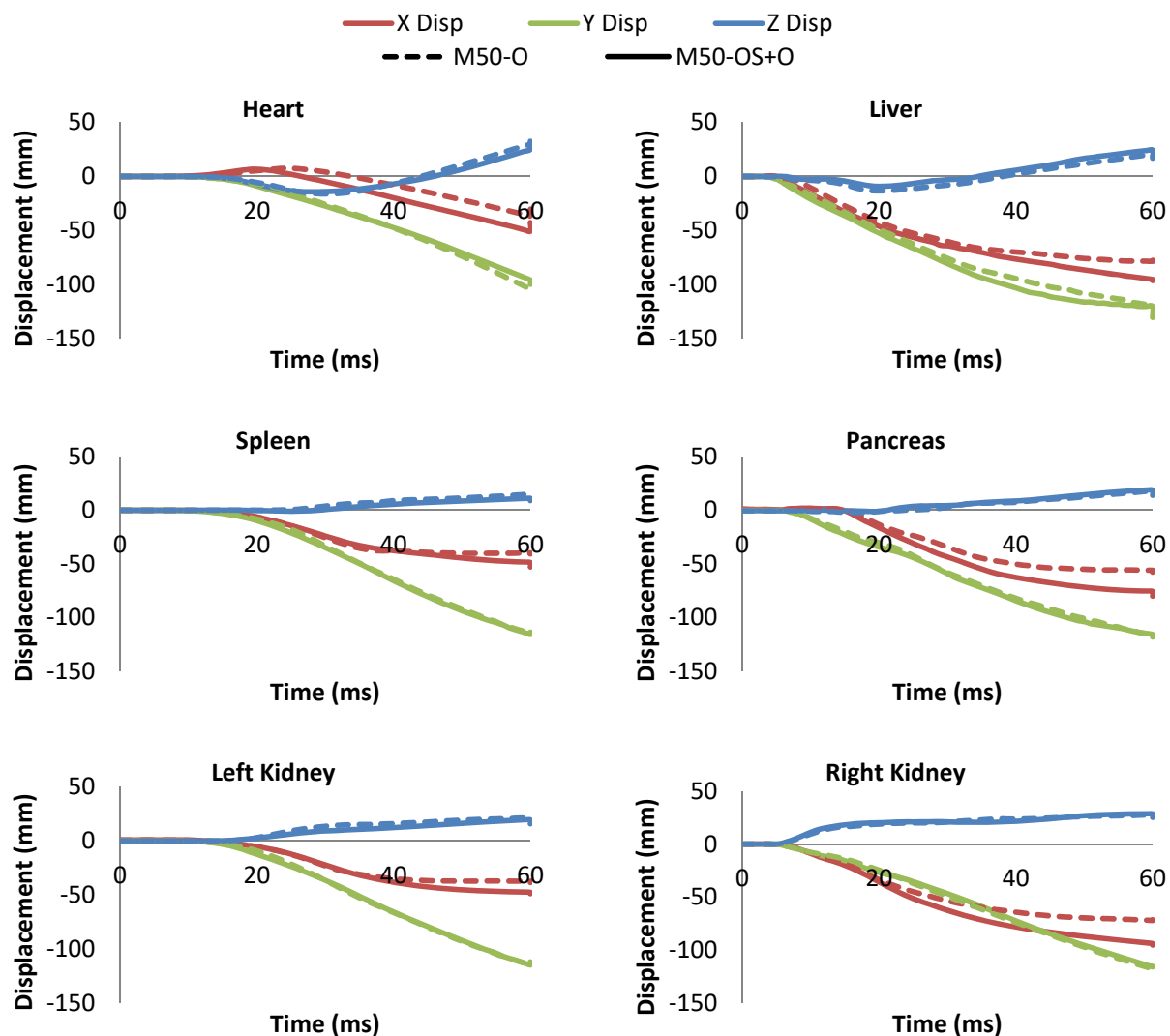


Figure A 2. Organ displacement comparison between the M50-O and M50-OS+O during the oblique hub impact.

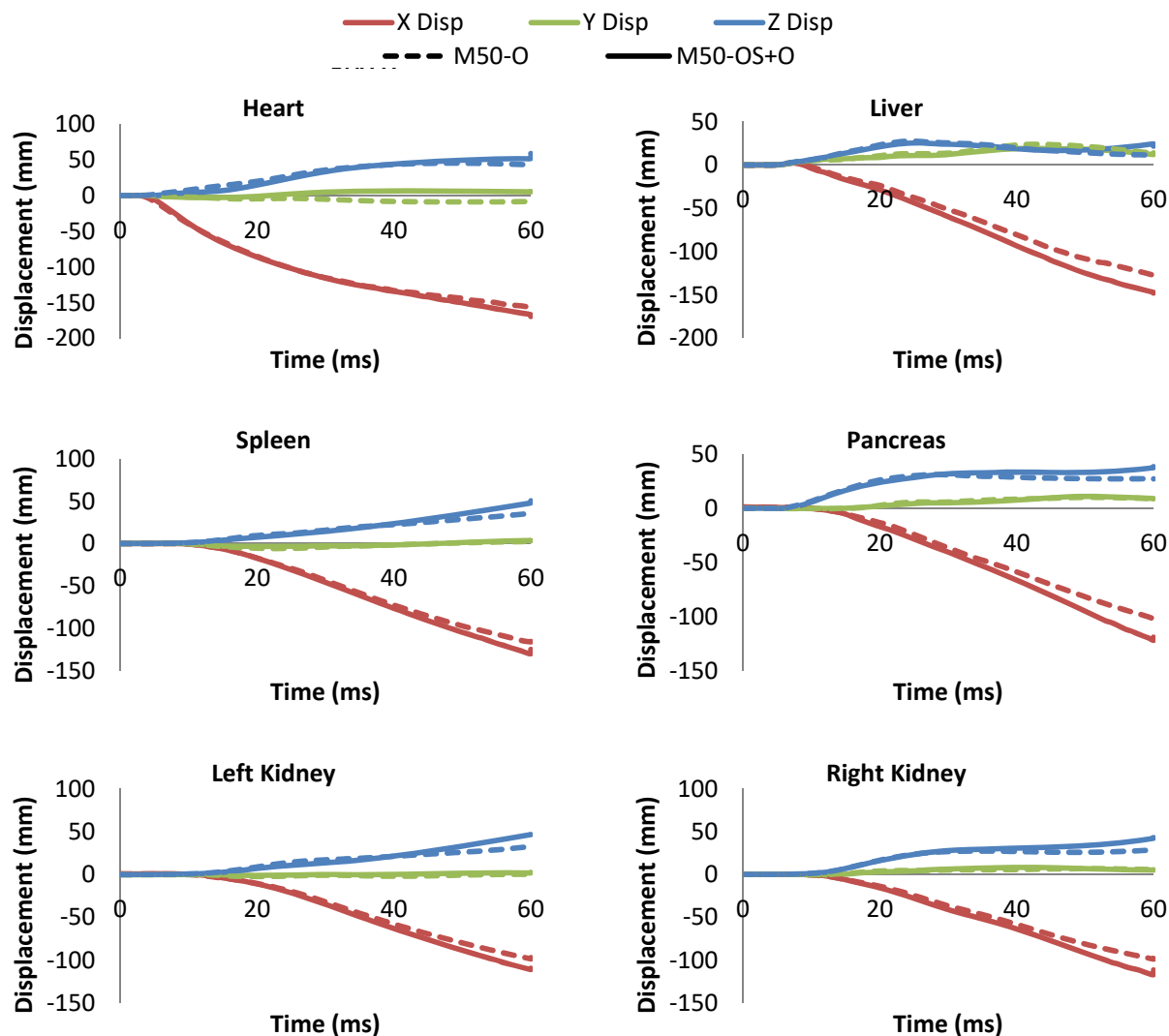


Figure A 3. Organ displacement comparison between the M50-O and M50-OS+O during the thoracic hub impact.

TABLE A 1. COMPARISON OF THE PEAK INTERNAL ENERGY OF EACH ORGAN BETWEEN THE M50-O AND M50-OS+O FOR THE THREE HUB IMPACTS.

Internal Energy (J)	Oblique Hub			Abdominal Bar			Thoracic Hub		
	M50-O	M50-OS+O	Diff	M50-O	M50-OS+O	Diff	M50-O	M50-OS+O	Diff
Liver	20.5085	18.0671	11.9%	2.641	3.4372	30.1%	8.1724	8.7907	7.6%
Pancreas	0.5469	0.7469	36.6%	0.7053	0.583	17.3%	0.3759	0.0702	81.3%
Spleen	0.2294	0.1401	38.9%	0.1316	0.128	2.7%	0.1508	0.1363	9.6%
Kidney_L	0.168	0.104	38.1%	0.3892	0.2468	36.6%	0.2051	0.1135	44.7%
Kidney_R	2.963	2.4561	17.1%	1.5701	0.7584	51.7%	0.6629	0.5110	22.9%
Heart	0.023	0.0155	32.8%	0.0065	0.0061	5.8%	0.0641	0.0473	26.2%

TABLE A 2. COMPARISON OF THE 95TH PERCENTILE FIRST PRINCIPLE STRAIN OF THE M50-O AND M50-OS+O DURING THE THREE HUB IMPACTS.

95th % Strain	Oblique Hub			Abdominal Bar			Thoracic Hub		
	M50-O	M50-OS+O	Diff	M50-O	M50-OS+O	Diff	M50-O	M50-OS+O	Diff
Liver	0.295	0.265	10.3%	0.156	0.132	15.5%	0.209	0.204	2.1%
Pancreas	0.610	0.737	20.7%	0.590	0.636	7.7%	0.417	0.430	3.1%
Spleen	0.135	0.103	23.7%	0.104	0.095	8.1%	0.114	0.105	8.5%
Kidney_L	0.059	0.043	27.0%	0.113	0.090	20.9%	0.082	0.063	23.5%
Kidney_R	0.271	0.249	8.0%	0.195	0.156	20.1%	0.130	0.115	11.5%
Heart	0.885	0.875	1.1%	0.615	0.582	5.3%	1.040	1.000	3.9%