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Hybrid III Manikin Lumbar Spine Loading Under Vertical Impact

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ABSTRACT

Introduction:

Clinical investigations have attributed lumbar spine injuries in combat to the vertical vector. Injury prevention strategies include the determination of spine biomechanics under this vector and developing/evaluating physical devices for use in live fire and evaluation-type tests to enhance Warfighter safety. While biological models have replicated theater injuries in the laboratory, matched-pair tests with physical devices are needed for standardized tests. The objective of this investigation is to determine the responses of the widely used Hybrid III lumbar spine under the vertical impact-loading vector.

Materials and Methods:

Our custom vertical accelerator device was used in the study. The manikin spinal column was mounted between the inferior and superior six-axis load cells, and the impact was delivered to the inferior end. The first group of tests consisted of matched-pair repeatability tests, second group consisted of adding matched-pair tests to this first group to determine the response characteristics, and the third group consisted of repeating the earlier two groups by changing the effective torso mass from 12 to 16 kg. Peak axial, shear, and resultant forces at the two ends of the spine were obtained.

Results:

The first group of 12 repeatability tests showed that the mean difference in the axial force between two tests at the same velocity across the entire range of inputs was <3% at both ends. In the second group, at the inferior end, the axial and shear forces ranged from 4.9-25.2 kN to 0.7-3.0 kN. Shear forces accounted for a mean of $11 \pm 6\%$ and $12 \pm 4\%$ of axial forces at the two ends. In the third group of tests with increased torso mass, repeatability tests showed that the mean difference in the axial force between the two tests at the same velocity across the entire range of inputs was <2% at both ends. At the inferior end, the axial and shear forces ranged from 5.7-28.7 kN to 0.6-3.4 kN. Shear forces accounted for a mean of $11 \pm 8\%$ and $9 \pm 3\%$ of axial forces across all tests at the inferior and superior ends. Other data including plots of axial and shear forces at the superior and inferior ends across tested velocities of the spine are given in the paper.

Conclusions:

The Hybrid III lumbar spine when subjected to vertical impact simulating underbody blast levels showed that the impact is transmitted via the axial loading mechanism. This finding paralleled the results of axial force predominance over shear forces and axial loading injuries to human spines. Axial forces increased with increasing velocity suggesting the possibility of developing injury assessment risk curves, i.e., the manikin spine does not saturate, and its response is not a step function. It is possible to associate probability values for different force magnitudes. A similar conclusion was found to be true for both magnitudes of added effective torso mass at the superior end of the manikin spinal column. Additional matched-pair tests are needed to develop injury criteria for the Hybrid III male and female lumbar spines.

INTRODUCTION

Retrospective studies of combat activities such as those from underbody blast-loading events from improvised explosive devices have reported trauma to the lumbar spine in clinical and other literatures.[1–5] Impact loading from the seat structure via pelvis of the Service Member is transmitted to the vertebral column during these events. Clinical investigations based on radiographically documented fractures have attributed spinal injuries to the vertical load vector traversing along the longitudinal axis of the column.[2,6–8] Injury prevention and mitigation strategies include the determination of the biomechanics of the lumbar spine under this vector and developing and evaluating anthropomorphic test devices, also known as manikins or dummies in military and automotive studies. Biological models are needed to replicate injuries in the laboratory environment so that field observations can be replicated and used to develop injury criteria in the form of injury assessment risk curves for dummies. Different types of biomechanical studies have been conducted using lumbar spines excised from human cadaver subjects. Controlled constant force rate tests using a material testing machine have been conducted using lumbar columns.[9] In the cited study, the piston applied the axial force from the caudal end, and transmitted loads were recorded at the cephalad end of the specimen. Posttest computed tomography and necropsy were

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used to assess injuries. In other studies, lumbar spinal columns have been dropped from a predetermined height on to a force plate, and impact loads were recorded at the inferior end of the preparation via a six-axis load cell.[10–12] Pre- and post-test X-rays, computed tomography, and autopsy were used for injury identification and comparison with field outcomes. Tests have also been conducted by loading the lumbar spine in a manner similar to those encountered in combat, i.e., dynamic accelerative loading from the caudal to the cephalad axis was applied using a vertical accelerator device.[13–15] The vertical accelerator allowed spinal vertebrae to sustain accelerative loading, and in these studies, specimens were aligned in a seated-soldier posture, confirmed by pretest radiographs, and injuries were documented using posttest computed tomography and autopsy.[16] A majority of injuries occurred to the vertebral body at one or more spinal levels, identified as compression or wedge fractures, and axial loading was found to be the principal mechanism of injury to the lumbar spinal column. The peak axial force was associated with the injury.[17] These studies can be used to conduct similar tests with manikins.

The next step is to conduct matched-pair tests with a manikin and derive tolerance criteria using injury outcomes from human cadaver lumbar spinal column tests. Such efforts have been carried out by the military for the development of the Warrior Injury Assessment Manikin. This manikin is currently used in Live Fire Testing and Evaluation tests by the U.S. Military. Other widely available manikins include the Hybrid III family of dummies, and they were originally developed for automotive safety applications.[18] This manikin is still being used around the world as a standard test device for automotive and other crashworthiness studies to advance human safety. While it includes military scenarios, matched-pair studies are needed to confirm the effectiveness of the manikin for applications to the combat and military-relevant impact vertical loading vector. Thus, the objective of this study is to subject the Hybrid III manikin lumbar spine (termed hereafter as manikin) to the same input conditions as human cadaver tests, use the same vertical accelerator that recorded human biomechanical responses and spinal injuries, and evaluate its response potential for its validated use and under vertical impacts.

METHODS

Vertical Accelerator

The custom vertical accelerator device (Fig. 1) that was used to conduct human cadaver spinal column tests was also used to perform Hybrid III lumbar spine tests.[15] An impact-delivering and an impact-absorbing section were the two principal components of the device. The impact-delivery section consisted of a stanchion that was fixed to the wall of the laboratory. It had a custom cart assembly that projected as an outrigger from the wall. The stanchion was designed to accommodate weights on its platform so that the chosen weights can be dropped on to the pulse shaping material/padding attached to one end of a lever arm that was fixed to the custom-designed laboratory floor. A metal plate was attached to the other end of the level arm. The size of the metal plate matched the inferior end of the fixation of the manikin lumbar spinal column. The impact-absorbing component was free-standing, i.e., it was not fixed to the wall or floor. It consisted of a moveable stanchion to position the manikin lumbar spinal column. The stanchion consisted of a frame with two vertical fixtures and a cart assembly that comprised of a metal base to which the lower end of the manikin spine was rested upon and hinged to maintain a vertical orientation during application of the upward force vector. The cart translated vertically upward following the thrust from the lever arm plate of the stanchion of the impact-delivering component. Upon releasing the desired weight from a predetermined height to achieve a chosen impact velocity profile, the dropped weight contacted the lever arm and applied the dynamic loading across the fulcrum and along the vertical direction to the manikin lumbar spinal column, thus inducing the impact loading to the system.

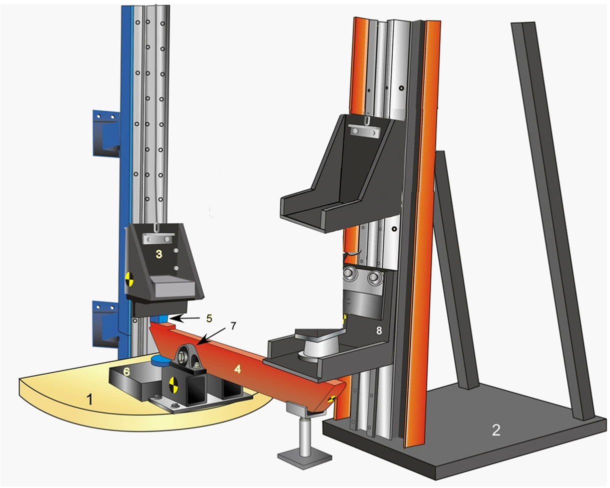

FIGURE 1. Schematic of the vertical accelerator device.[15]

Manikin Spine

The 50th percentile male Hybrid III lumbar spine was used to conduct impact tests, matching previously conducted male human cadaver tests. The spinal column was mounted between the inferior and superior six-axis load cells (model M3944, Sunrise Instruments, Shanghai, China). The two load cells were orthogonal to the platform of the vertical accelerator device. The superior load cell was mounted to the upper carriage, and the inferior load cell was mounted to the lower carriage of the impact-absorbing component of the accelerator device. This arrangement allowed for unconstrained inferior-to-superior motion along a vertical track. As described, the impact-delivering component of the device applied the vertical impact pulse with varying velocities.

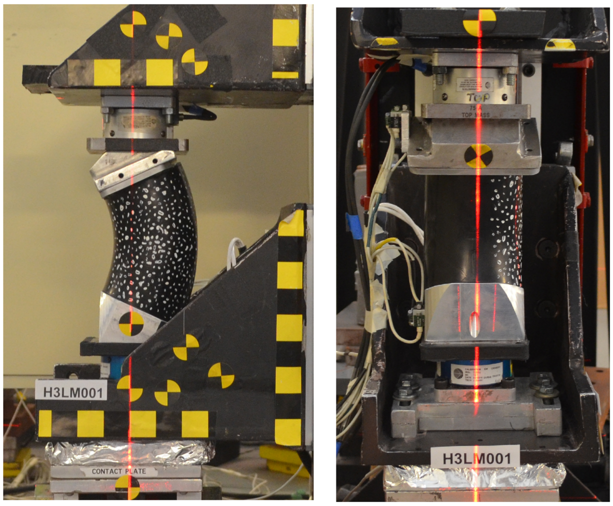

FIGURE 2. Hybrid III lumbar spine attached to the vertical accelerator platform and connected to the load cells at the inferior and superior levels of the manikin spine (left).

Sensors and Data

Two pairs of accelerometers (model 6DX, Denton Inc., Troy, MI) were attached to the superior and inferior mounting blocks on the manikin lumbar spine. Data from the superior and inferior spine load cells and accelerometers were gathered using a digital data acquisition system (DTS Inc., Seal Beach, CA). The sampling rate was 100 kHz, and a 4-pole 1 kHz Butterworth filter was used to process the signals. The input impact velocity was computed by integrating the acceleration-time signals at the inferior end. The gathered time signals from both load cells were transformed to the interface between the metal and rubber at the superior and inferior ends of the manikin spine (Fig. 2). Mass compensation was applied on a temporal basis, and the maximum compressive (axial/vertical), antero-posterior (lateral) shear, and resultant forces (square root of the sum of the squares of the axial and shear force-time histories) at the superior and inferior ends were calculated. Shear factor was defined as the ratio of the shear force to the axial force. These variables were used to demonstrate the effectiveness of the dummy for underbody loading in military events, via repeatability tests and evaluation of axial and shear force data.

Impact Loading

The same initial/boundary conditions used in human cadaver tests were used to conduct matched-pair tests with the manikin lumbar spine. Three groups of tests were conducted. In the first group, the response of the manikin spine was checked for its repeatability by conducting 12 repeat tests (six pairs), and each were with matched-pair conditions that used an effective torso mass of 12 kg in cadaver tests. To account for the relaxation of the rubber in the manikin spine, at least 30 min was allowed between two consecutive tests. In the second group of tests, after ensuring the repeatability of the manikin spine, additional experiments were conducted (no repeated tests) to determine the matched-pair response of the manikin spine under the vertical impact loading vector. In the third group, the effective mass of the torso was increased from 12 to 16 kg, and tests were repeated. As stated, peak axial and shear and resultant forces were obtained at the superior and inferior ends of the spine, and data from all tests were pooled in the analysis.

RESULTS

The first group of repeatability tests conducted with an effective mass of 12 kg on the superior end showed that the manikin responses are repeatable as defined by the morphology of the axial force-time histories and peak magnitudes. The average difference in the axial force between the two tests at the same velocity across the entire range was 2.6% at the inferior end and 1.9% at the superior end. The manikin response was thus found to be repeatable for vertical impact loading.

In the second group, six tests from the previous repeatability tests group were used in addition to six new tests, i.e., 12 total tests. There were five tests in the velocity range 2 to 4 m/s, four in the 4 to 6 m/s, and three in the 6 to 8 m/s. At the inferior end, axial, shear, and resultant forces ranged from 4.9 to 25.2 kN, 0.7 to 3.0 kN, and 5.0 to 25.2 kN, respectively (Supplementary Table A1). At the superior end, axial, shear, and resultant forces ranged from 4.9 to 25.2 kN, 0.6 to 2.6 kN, and 4.5 to 25.3 kN, respectively. Shear forces accounted for a mean of $11 \pm 6\%$ and $12 \pm 4\%$ (shear factor) of axial forces across all tests at the inferior and superior ends. Figure 3 shows the progression of the peak forces with increasing velocity for the inferior axial and shear forces, and superior axial and shear forces. Axial forces predominated over shear forces at both levels and at all velocities. Furthermore, shear forces were consistent and remained below 4 kN, while axial forces had a larger range of up to approximately 25 kN at the highest velocity. In addition, a somewhat increasing trend in the axial force was observed beyond 4 m/s velocity. These observations were true for both inferior and superior forces.

In the third group of tests with increased in torso mass from 12 to 16 kg, like in the first group, a total of 12 (six duplicate) tests were conducted at the same condition to examine the repeatability of the manikin spine. The average difference in the axial force between the two tests at the same velocity across the entire range was less than 2% at the inferior end and less than 1% at the superior end. The manikin response was thus found to be repeatable for vertical impact loading for the increased torso mass condition. In the subsequent tests, like the second group, additional non-repeat tests were conducted, resulting in a total of 12 tests. At the inferior end, axial, shear, and resultant forces ranged from 5.7 to 28.7 kN, 0.6 to 3.4 kN, and 5.7 to 28.7 kN, respectively

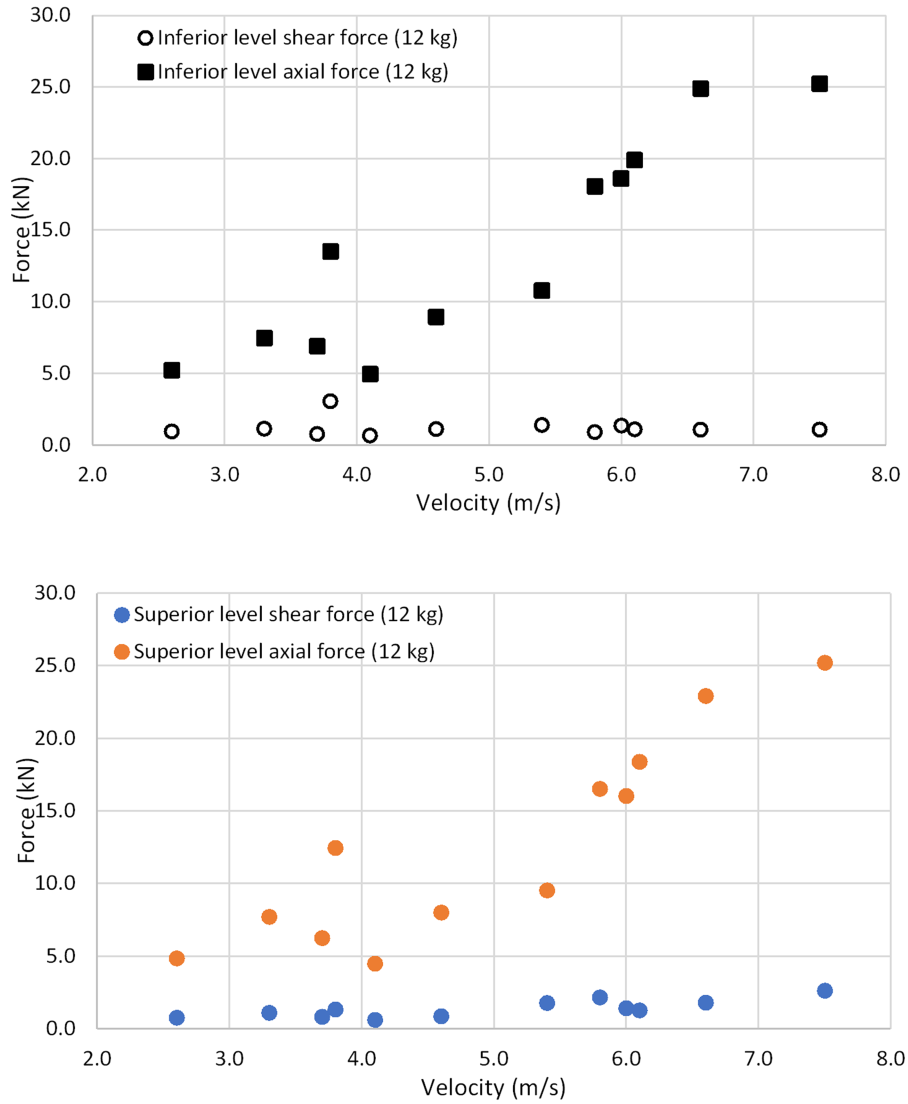

FIGURE 3. Plot of axial and shear forces at the inferior and superior ends of the spine for 12 kg tests.

(Supplementary Table A2). At the superior end, axial, shear, and resultant forces ranged from 5.2 to 27.8 kN, 0.6 to 1.6 kN, and 5.2 to 27.8 kN, respectively. Shear forces accounted for a mean of 11 ± 8% and 9 ± 3% of axial forces across all tests for the inferior and superior ends. Note that the resultant forces were almost the same as axial force indicating the relatively small contribution from the shear force. Figure 4 shows the progression of the peak forces with increasing velocity for the axial and shear forces at the two ends of the spinal column for the 16 kg tests. Similar to the 12 kg tests, axial forces predominated over shear forces, and shear forces remained below 5 kN across all velocities; however, axial forces demonstrated a larger range of up to approximately 30 kN, and an increasing trend in the axial force was apparent velocities greater than 4 m/s. These results show that the response of the manikin spine is similar for both 12 kg and 16 kg conditions.

DISCUSSION

As stated in the introduction, the objective of the present study was to determine the suitability of the Hybrid III lumbar spine under vertical impact loading as applied to combat scenarios. The Hybrid III manikin was selected in this study because of its wide use, as it has been time-tested since the late 1970 under a variety of scenarios including military testing.[18,19] While it was originally developed to serve as an anthropomorphic testing device for unbelted occupants in frontal impacts without airbags, the same manikin continues to be used in modern environments, in which vehicles have mandatory airbags and advanced belt systems with pretensioners and load limiters. Likewise, although the manikin was designed only to mimic flexion-extension response of the head-neck at that time, the same manikin has been used under compressive loading scenarios such as rollovers. It is also used to evaluate head-supported mass.[20–22] Thus, a history exists for the

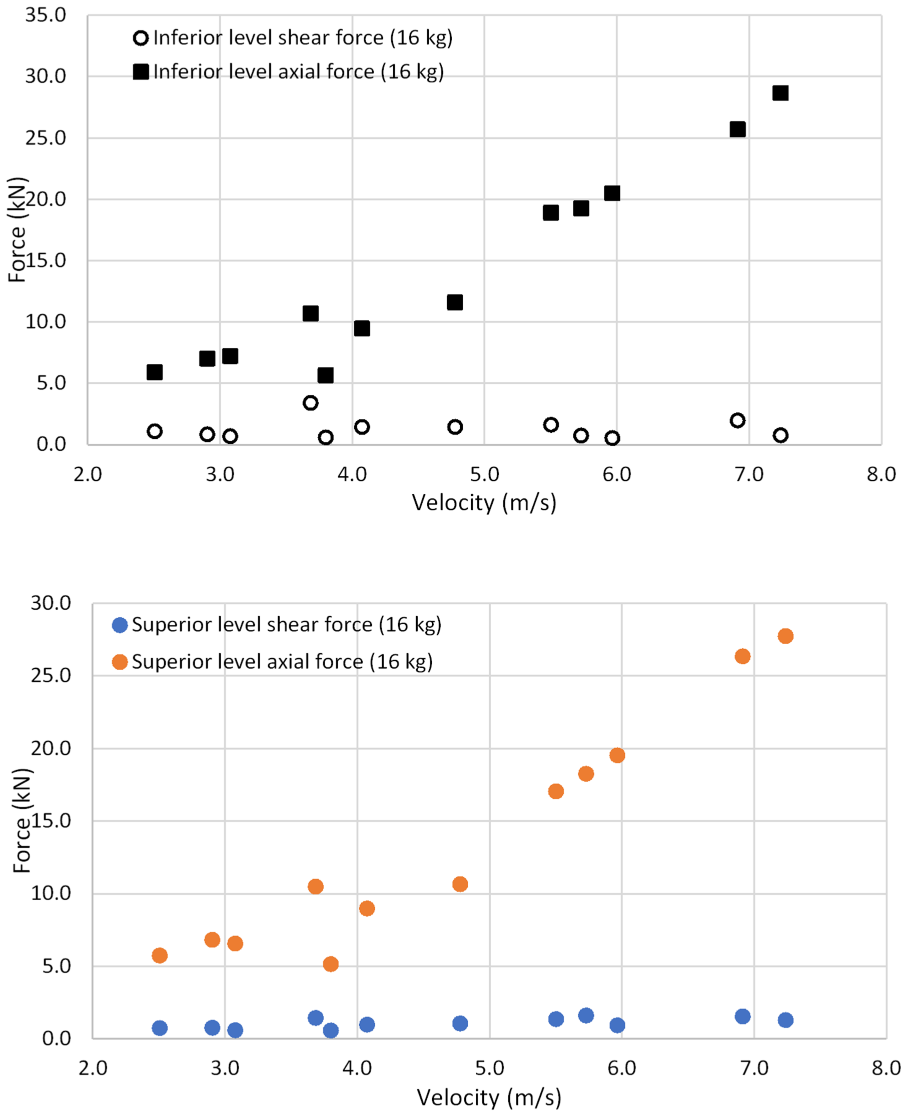

FIGURE 4. Plot of axial and shear forces at the inferior and superior ends of the spine for 16 kg tests.

use of this manikin in other scenarios. Military vehicle crashworthiness studies have also used this device for Warfighter safety.

The present study was conducted using the Hybrid III spine, although the manikin was not designed for vertical impacts: its lumbar spine may be a suitable component to sense injury in this loading mode. The current experimental design involved using matched-pair testing conditions with human cadavers and using the same impact delivering vertical accelerator device so that the manikin can be assessed, and injury assessment risk curves can be developed. It should be noted that data from matched pair human cadaver tests included in this study were generated and authorized for release by the Warrior Injury Assessment Manikin Project. The process for manikin use needed an initial check to ensure its suitability for vertical impacts and was the objective in the first group of tests. The repeatability tests encompassed a wide range of velocities used in earlier human cadaver tests and used the same effective torso mass.[17] Peak axial forces at the inferior and superior ends of the spine were used to determine the repeatability because the expected response is predominantly along the inferior to superior direction. Because results showed excellent repeatability (within 6% across all data on an individual test basis) for both locations, the manikin lumbar spine was deemed suitable to be used as a testing device under the vertical impact loading vector. A higher level of repeatability was found in tests with increased effective mass.

Instead of using the recorded data from the load cells, forces were translated to the locations equivalent to the L5-S1 and T12-L1 intervertebral disc levels in the human cadaver. This is because cadaver forces were transformed in earlier studies to these locations. This process ensured matched-pair testing and analysis of data with the manikin. Data revealed that axial forces predominate over shear forces, as demonstrated by the magnitude of the resultant forces and, also, by the shear factor (Supplementary Tables A1 and A2). This phenomenon was true for both superior and inferior forces and for both effective mass tests. This shows that the manikin responds and transmits the vertical loading along the length of the spinal column, without considerable shear forces, implying axial mechanism of load transfer. Human cadaver spinal column vertical loading tests also revealed the same mechanism of impact transfer: in this mode, axial forces predominated over shear forces like the manikin, and injuries were attributed to the axial mechanism.[14,17] Thus, both surrogates responded similarly to the vertical impact loading.

It should be noted that the human cadaver is frangible, i.e., forces exceeding tolerance result in injuries such as fractures. A desired design feature of any manikin is infrangibility because repeated tests are expected with the physical surrogate. The infrangible feature makes the manikin spine stiffer compared to the human spine. Thus, at greater impact velocities, while the biological model saturates or shows a plateaue/decreased response due to fracture, forces in the manikin continue to increase, suggesting that the isoenrgy approach for risk curves/tolerances is not optimal with matched-pair testing. The greatest forces occurring with the highest velocity support this observation. It should be noted that one test at 3.8 m/s for both 12 kg and 16 kg tests showed relatively larger increase in forces compared to the others around this velocity, and this was attributed to the shortest time to peak of the velocity parameter. This implies that there exists an interaction of the time and velocity variables in the response of the manikin spine, and it is also shown to be true in human cadaver studies wherein shorter times tends to fracture the components in proximity to impact while longer times tend to injury components farther away from the loading site, i.e., mass recruitment effect.[23,24] This combined effect needs further study in the manikin; however, the peak axial and resultant forces at the superior and inferior levels were similar for both series of tests (12 kg and 16 kg effective torso mass) at equivalent velocities. This shows that the forces at the inferior end of the manikin lumbar spine monotonically increases with impact velocity and does not saturate, and it is also true for the superior end forces. These observations suggest that injury assessment risk curves (IARCs) from this manikin will not be a step function. Additional matched-pair studies are needed to develop such curves.

As expected, spinal forces at the two ends for both series of tests followed a similar pattern, although individual magnitudes were different. The magnitude differences between the two series of tests explain the role of the added effective mass on the top of the spine. Axial force changes were within 25% at both ends of the spine, while shear force changes were greater. While the percentage change in the forces was larger for the shear component, shear force magnitudes (Figs. 3 and 4) were considerably less than the axial forces. Because axial forces predominate under the vertical impact loading vector, any shear force effect that may account for flexion-extension bending moments will have minimal effect on injury metric. In other words, like the human cadaver spinal columns that responded with axial-loading mechanism of injury, the Hybrid III manikin lumbar spine also responded with the same mechanism. This suggests that this manikin mimics the human responses and force-based injury criteria metric applicable to the human cadaver spine can also be used for the Hybrid III dummy, albeit with different magnitudes.

Limitations/Future Research Requirements

Because of the parallelism between human cadaver tests from the previous study that showed the overriding influence of the axial over shear forces and the current manikin study, uniaxial force-based injury assessment risk curves (IARCs) without interaction from off-axis forces or bending moments are appropriate as the injury criteria. Additional tests are, however, necessary to cover all the tested human cadaver experiments and alternate mechanisms of injury. For those mechanisms, additional matched-pair tests should be conducted to develop IARCs for the Hybrid III dummy. This is a future study. Because all human cadaver tests were conducted with male specimens, the Hybrid III male spine was used in this series of tests. To develop IARCs for the Hybrid III female lumbar spine, the abovementioned tests should be repeated. Along the same vein, a similar study can be conducted with other devices such as the FAA-H3 spine to compare across different manikins representing different curvatures of the spinal column and demographics. It should be noted that the recently formed Human Factors Medicine Panel (HFM-379) by the North Atlantic Treaty Organization is interested in manikin testing and IARCs. Because lumbar spine responses of men and women Warfighters are not directly scalable, it would be necessary to test female spinal columns, develop human injury assessment probability curves, and use those data to develop IARCs for female manikins. In other words, female human cadaver and manikin matched-pair tests are needed to provide a comprehensive set of IARCs to advance safety to women and men Warfighters. While one test for both 12 kg and 16 kg responded with greater forces, because of the interaction of the time-to-peak and peak velocity, additional tests covering a large spectrum of both variables and grouping based on their paired combinations are needed. These are considered as future studies.

CONCLUSIONS

The Hybrid III lumbar spinal column subjected to vertical impact loading simulating underbody blast levels showed that the impact is transmitted via the axial-loading mechanism. This finding paralleled the results of axial force predominance over shear forces and axial-loading injuries to the spine from human cadaver studies. Shear forces were considerably lower than axial and resultant forces at either ends of the dummy spine. Axial forces increased with increasing velocity suggesting the possibility of developing injury assessment risk curves using this metric, i.e., the manikin spine does not saturate, and its response is not a step function. It is possible to associate probability values for different force magnitudes. A similar conclusion was found to be true for both magnitudes of added effective torso mass at the superior end of the manikin spinal column. Additional matched-pair tests are needed to develop injury criteria for the Hybrid III lumbar spine.

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INSTITUTIONAL REVIEW BOARD (HUMAN SUBJECTS)

Not applicable.

INSTITUTIONAL ANIMAL CARE AND USE COMMITTEE (IACUC)

Not applicable.

CONFLICT OF INTEREST AND FINANCIAL DISCLOSURE

None declared.

INDIVIDUAL AUTHOR CONTRIBUTION STATEMENT

N.Y. conceptualized, designed, and supervised the study including data analysis, prepared the manuscript, and edited based on comments from authors. J.M. developed test procedures, conducted manikin experiments, processed biomechanical data, and participated in technical discussions. T.W. and N.F. participated in technical and data discussions and reviewed the manuscript.

SUPPLEMENTARY MATERIAL

Supplementary material is available at *Military Medicine* online.

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CONFLICT OF INTEREST STATEMENT

None declared.

DATA AVAILABILITY

The data underlying this article may be shared on reasonable request to the corresponding author.

REFERENCES

1. Rankin IA, Nguyen TT, Carpanen D, et al: A new understanding of the mechanism of injury to the pelvis and lower limbs in blast. Front Bioeng Biotechnol 2020; 8: 960. 10.3389/fbioe.2020.00960
2. Danelson K, Watkins L, Hendricks J, et al: Analysis of the frequency and mechanism of injury to warfighters in the under-body blast environment. Stapp Car Crash J 2018; 62: 489–513. 10.4271/2018-22-0014
3. Schoenfeld AJ, Dunn JC, Belmont PJ: Pelvic, spinal and extremity wounds among combat-specific personnel serving in Iraq and Afghanistan (2003-2011): a new paradigm in military musculoskeletal medicine. Injury 2013; 44(12): 1866–70. 10.1016/j.injury.2013.08.001
4. Spurrier E, Gibb I, Masouros S, et al: Identifying spinal injury patterns in underbody blast to develop mechanistic hypotheses. Spine (Phila Pa 1976) 2016; 41(5): E268–75. 10.1097/BRS.0000000000001213
5. Ramasamy A, Harrisson SE, Clasper JC, et al: Injuries from roadside improvised explosive devices. J Trauma 2008; 65(4): 910–4. 10.1097/TA.0b013e3181848cf6
6. Patzkowski JC, Blair JA, Schoenfeld AJ, et al: Multiple associated injuries are common with spine fractures during war. Spine J 2012; 12(9): 791–7. 10.1016/j.spinee.2011.10.001
7. Vasquez KB, Brozoski FT, Logsdon KP, et al: Retrospective analysis of injuries in underbody blast events: 2007-2010. Mil Med 2018; 183(suppl_1): 347–52. 10.1093/milmed/usx216
8. Loftis KL, Mazuchowski EL, Clouser MC, et al: Prominent injury types in vehicle underbody blast. Mil Med 2019; 184(Suppl 1): 261–4. 10.1093/milmed/usy322
9. Ortiz-Paparoni M, Op ‘t Eynde J, Kait J, et al: The human lumbar spine during high-rate under seat loading: a combined metric injury criteria. Ann Biomed Eng 2021; 49(11): 3018–30. 10.1007/s10439-021-02823-x
10. Stemper BD, Chirvi S, Doan N, et al: Biomechanical tolerance of whole lumbar spines in straightened posture subjected to axial acceleration. J Orthop Res 2018; 36(6): 1747–56. 10.1002/jor.23826
11. Stemper BD, Storvik SG, Yoganandan N, et al: A new PMHS model for lumbar spine injuries during vertical acceleration. J Biomech Eng 2011; 133(8): 081002. 10.1115/1.4004655
12. Yoganandan N, Arun MW, Stemper BD, et al: Biomechanics of human thoracolumbar spinal column trauma from vertical impact loading. Ann Adv Automot Med 2013; 57: 155–66.
13. Yoganandan N, Moore J, DeVogel N, et al: Human lumbar spinal column injury criteria from vertical loading at the base: applications to military environments. J Mech Behav Biomed Mater 2020; 105(1): 103690. 10.1016/j.jmbbm.2020.103690
14. Yoganandan N, Moore J, Pintar FA, et al: Role of disc area and trabecular bone density on lumbar spinal column fracture risk curves under vertical impact. J Biomech 2018; 72(1): 90–8. 10.1016/j.jbiomech.2018.02.030
15. Yoganandan N, Pintar FA, Schlick M, et al: Vertical accelerator device to apply loads simulating blast environments in the military to human surrogates. J Biomech 2015; 48(12): 3534–8. 10.1016/j.jbiomech.2015.06.008
16. Reed MP: Development of anthropometric specifications for the warrior injury assessment manikin. University of Michigan Transportation Research Institute. UMTRI-2013-382013. 1–38.
17. Yoganandan N, DeVogel N, Moore J, et al: Human lumbar spine responses from vertical loading: ranking of forces via brier score metrics and injury risk curves. Ann Biomed Eng 2020; 48(1): 79–91. 10.1007/s10439-019-02363-5
18. Backaitis SH: *Hybrid III: The First Human-like Crash Test Dummy*. Society of Automotive Engineers; 1994.
19. FMVSS-208: FMVSS 208: 49Code of Federal Regulations: 571.214. US Government Printing Office; 2001.
20. Begonia MT, Pintar FA, Yoganandan N: Comparison of NOCSAE head kinematics using the Hybrid III and EuroSID-2 necks. J Biomech 2018; 80(10): 37–44. 10.1016/j.jbiomech.2018.08.018
21. Begonia M, Rooks T, Pintar FA, et al: Development of a methodology for simulating complex head impacts with the

advanced combat helmet. Mil Med 2019; 184(Suppl 1): 237–44. 10.1093/milmed/usy282

22. Jadischke R, Viano DC, McCarthy J, et al: The effects of helmet weight on Hybrid III head and neck responses by comparing unhelmeted and helmeted impacts. J Biomech Eng 2016; 138(10): 101008. 10.1115/1.4034306

23. Yoganandan N, Moore J, Humm JR, et al: Human pelvis injury risk curves from underbody blast impact. BMJ Mil Health 2023; 169(5): 436–42. 10.1136/bmjmilitary-2021-001863

24. Yoganandan N, Moore J, Arun MW, et al: Dynamic responses of intact post mortem human surrogates from inferior-to-superior loading at the pelvis. Stapp Car Crash J 2014; 58: 123–43. 10.4271/2014-22-0005