A Methodology for Assessing Intrasegmental Kinematics of the Whole Human Spine during Impacts

David Lessley ¹) Greg Shaw ²) Joseph Ash ³) Jeff Crandall ⁴)
1)-4) University of Virginia Center for Applied Biomechanics
4040 Lewis and Clark Drive, Charlottesville, Virginia, 22911, USA (E-mail: lessley@virginia.edu)

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ABSTRACT: Evaluating the biofidelity of spinal response in ATDs and computational models requires additional benchmarking data. The current study presents a kinematic analysis methodology for obtaining intrasegmental kinematics of the spine including elongation, compression, displacement, and anatomical rotations of the spine including flexion, lateral flexion, and rotation. The presented methodology divides the spine into five spine segments selected for detailed kinematic analysis which were specifically the following: head-to-T1, T1-to-T8, T8-to-L2, L2-to-L4, and L4-to-pelvis. The methodology provides a technique to more comprehensively describe the kinematics of the whole human spine than has been performed to date in the impact environment.

KEY WORDS: human engineering, biomechanics, occupant safety

1. Introduction

For restrained vehicle occupants, injuries to the head and thorax are primary sources of morbidity and mortality occurring from motor vehicle accidents (1-5). While increased seat belt usage and advances in restraint design have, on average, substantially improved the injury outcome for restrained occupants, numerous fatalities and debilitating injuries are still occurring nonetheless (6). This is especially true for older occupants who are more susceptible to injury resulting from the restraining forces applied to the torso during a crash (3-5).

Further mitigating injuries to restrained occupants requires a more complete understanding of how the human skeletal system moves and deforms while interacting with the occupant restraint system during a vehicle crash. In particular, the complex motion of the flexible spine provides the foundation of human torso motion during a crash event. Motion of the spine dictates the trajectory of the head as well as thoracic interaction with the restraint system. Optimizing restraint designs requires improving capabilities in predicting injuries associated with the greatest mortality and morbidity, including those to the head and thorax. Improving our injury prediction capabilities, however, requires corresponding improvements to the tools currently used to make such injury predictions which are currently anthropomorphic testing devices (ATDs) and computational models. This ultimately requires additional benchmarking kinematic data for the human spine, including intrasegmental elongations, displacements, and rotations that are that are currently unavailable in the literature.

Spinal response ultimately affects the ability of our current human surrogates to predict seatbelt loading as well as the risk of contact between the head and the interior structures of the vehicle. Consequently, it is imperative that the tools with which we use to assess injury risk be based upon the most detailed data available regarding the motion of the human spine, which in turn dictates the foundational motions of the head, shoulders, and chest. The goal of the current study is to present the results of three frontal sled tests to evaluate an analysis methodology for obtaining intrasegmental kinematics of the spine including elongation, compression, displacement, and also anatomical rotations of the spine including flexion, lateral flexion, and rotation.

2. Method

Three tests (Table 1), which are a subset of those presented by Ref. (7), were selected to for further kinematic analysis to evaluate a methodology for calculating intrasegmental kinematics of the spine during impacts. In the selected tests, three adult male post mortem human surrogates (PMHS) approximating the 50th percentile male anthropometry were tested in a well-controlled 40 km/h frontal impact condition (7-8) that utilized a 14 g deceleration and 3-point restraint. The condition was selected to represent a repeatable pure frontal collision in the severity range of typically conducted regulatory and assessment frontal impact tests (e.g. FMVSS-208 and NCAP) using a standard non-force-limiting 3-point belt. In the reported tests no retractor was used in the shoulder belt in order to maximize repeatability and minimize variation.

Data collected from an optically-based motion capture system during the tests was used to calculate the six degree-of-freedom (6DOF) motion of the head, 1st thoracic vertebrae (T1), 8th thoracic vertebra (T8), 2nd lumbar vertebra (L2), 4th lumbar vertebra (L4), and pelvis of each subject relative to the vehicle buck. Since the tested subjects are a subset of a larger sample that approximated the 50th percentile male (in the US) stature and mass, no scaling was performed on the calculated kinematic results.

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Table 1 PMHS Characteristics.

| TEST No. | Age at Death | Body Mass (kg) | Stature (mm) |
|----------|--------------|----------------|--------------|
| 1358     | 54           | 79             | 1770         |
| 1359     | 49           | 76             | 1840         |
| 1360     | 57           | 64             | 1750         |

2.1. Coordinate Systems and Spine Segments

Using the technique described in detail by Ref. (9), the position and orientation of the head, T1, T8, L2, L4, and pelvis were calculated at one millisecond intervals from the collected motion capture data from the tests. For each anatomical structure selected for measurement (e.g. T1) an anatomically based coordinate system was created using the recommendations presented by Ref. (10). The section of the spine between two adjacent measurement locations represents a selected spine segment for intrasegmental kinematic analysis. Figure 1 illustrates the anatomical coordinate systems, the five spine segments (Head-T1, T1-T8, T8-L2, L2-L4, and L4-Pelvis), and also a cubic spline interpolation passing through the origin of each selected skeletal structure along the spine which was used to represent the 3D shape (or contour) of the PMHS spine at each millisecond during the impact event.

2.2. Elongation and Compression of the Spine

The cubic spline data used to characterize the shape of the spine throughout the impact event were also used to assess the change in length of the spine. This allowed for the calculation of intervertebral elongation and compression occurring within the spine segments located between adjacent measurement locations (Figure 1). Summing the length changes of each individual spine segment provided the changes in total spine length throughout the impact event.

Let the cubic spline interpolation corresponding to a given time during the test event be defined by Equation (1). Referring to Figure 1 the change in length associated with each spine segment can be defined by Equations (2 – 6).

\[
\text{Cubic spline interpolation} = \text{spl}(t)
\]  

\[
\Delta_{\text{Head-T1}}(t) = \int_{\text{Head}}^{\text{T1}} \text{spl}(t, s) ds - \int_{\text{Head}}^{\text{T1}} \text{spl}(t, s) ds
\]  

\[
\Delta_{\text{T1-T8}}(t) = \int_{\text{T1}}^{\text{T8}} \text{spl}(t, s) ds - \int_{\text{T1}}^{\text{T8}} \text{spl}(t, s) ds
\]  

\[
\Delta_{\text{L2-L4}}(t) = \int_{\text{L2}}^{\text{L4}} \text{spl}(t, s) ds - \int_{\text{L2}}^{\text{L4}} \text{spl}(t, s) ds
\]  

\[
\Delta_{\text{L4-Pelvis}}(t) = \int_{\text{L4}}^{\text{Pelvis}} \text{spl}(t, s) ds - \int_{\text{L4}}^{\text{Pelvis}} \text{spl}(t, s) ds
\]  

2.3. Intrasegmental Displacements

Figure 1 illustrates the six measurement locations along the spine and the corresponding five segments spanning the lengths along the spine between the anatomical structures selected for kinematic measurement. These independent and anatomically based coordinate systems allowed for the motion of one skeletal structure to be described relative to another. Intrasegmental displacements were determined using two adjacent anatomical coordinate systems along the length of the spine. For a given spine segment, the displacement of the adjacent superior anatomical coordinate system was described relative to the inferior base coordinate system (e.g. Head relative to T1). Figure 2 depicts two coordinate systems related by a transformation matrix (Equation 7). The position of a given point (e.g. selected vertebral body center) relative to a global system (coordinate system #2) can be determined relative to a local system (coordinate system #1) using Equation 7. The 3D displacement relative to the local anatomical coordinate system is determined by the change in position with time using Equations (8 – 10).

2.4. Intrasegmental Rotations

In addition to the displacements obtained above, anatomically based rotations of the spine were also determined. Specifically these were rotation, flexion, and lateral flexion. Figure 3 illustrates the rotation, flexion, and lateral flexion angles of the head with respect to T1 an example. Other spine segments were analyzed in a similar manner. Using Equation 7, Equation 11 provides the transformation matrix describing the position and orientation of the head with respect to T1. Equations (12 – 17) were used to quantify the anatomical rotation, flexion, and lateral flexion for each spine segment, i, using the superior and inferior endpoints for each segment.
3. Results

The intrasegmental kinematic analysis was successfully carried out on the collected kinematic data from the three selected PMHS tests (Table 1). A summary of the mean peak normalized elongation of the spine by segment is provided in Figure 4. Mean intrasegmental displacements in the X-axis, Y-axis, and Z-axis directions are provided in Figure 5, along with one standard deviation (S.D.) corridors for each of the five analyzed spine segments. Anatomically based intrasegmental rotations of spine are provided for each of the five spine segments in Figure 6 along with one S.D. corridors.

![Diagrams of anatomical rotations and normalized elongation of the spine](image-url)
Fig. 5 Mean intrasegmental displacements by spine segment with one S.D. corridors.
Fig. 6 Mean intrasegmental rotations by spine segment with one S.D. corridors.
4. Discussion

Improving the current understanding of human skeletal motion under impact loading is a crucial step in improving the biofidelity of the human surrogates used to design restraint systems and evaluate injury risk for restrained vehicle occupants. The study represents an integral step in the ongoing effort to address injuries in restrained vehicle occupants by improving the current understanding of spinal motion in a restrained occupant during a collision. The evaluated methodology for calculating intrasegmental kinematics produced well-behaved response data for elongations, displacements, and rotations occurring within selected spine segments. The presented results may represent the most detailed kinematic data that have been published to date regarding motions occurring within specific spine segments of a restrained occupant during a collision.

The results based on these three selected PMHS indicate that the spine undergoes both elongation and compression in each segment of the spine, however, the magnitude of these extensions and compressions were observed to be modest. The whole spine was observed to elongate approximately 3% of its total length. This is an important finding since the risk of head injury in restrained vehicle occupants is related to the maximum displacement of the head relative to the vehicle. The more space that is swept out by the head during a crash event, the more likely it is that the head can reach, and potentially strike, various structures within the interior of the vehicle. Thus, any elongation of the human spine during an impact event provides a mechanism to increase head displacement and associated head injury risk. Importantly, any such elongation present in the human spinal behavior, may be overlooked by human surrogates that do not incorporate this degree of freedom, and thus, the resulting head displacement could be underestimated leading to a corresponding underestimate of head injury risk.

Intrasegmental displacements and rotations along the length of the spine of a restrained occupant during a collision have not yet been published to date, to the authors’ knowledge. The results of the current study indicate that displacement, bending, and rotation of the spine occurs in the sagittal, coronal, and transverse planes. Ultimately, this bending and rotation of the spine affects how the thorax interacts with the restraint system. The presented methodology should be utilized to expand the current dataset to included response data from additional subjects and tests. A larger dataset could be very useful for providing additional kinematic biofidelity targets to evaluate and improve human surrogates used to evaluate restraint systems and predict risk of injury.

5. Conclusion

The current study presents a kinematic analysis methodology for obtaining intrasegmental kinematics of the spine including elongation, compression, displacement, and anatomically-based rotations including flexion, lateral flexion, and rotation. The methodology provides a technique to more comprehensively describe the kinematics of the whole human spine than has been performed to date in the impact environment.

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