A MULTI-BODY MODEL OF THE WHOLE HUMAN SPINE FOR WHIPLASH INVESTIGATIONS

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Paper Number 07-0437

ABSTRACT

This paper presents whiplash simulations and analyses under various impact conditions and acceleration levels by employing a rigorously validated biofidelic multi-body (MB) model of the whole human spine. The novel MB model possesses highly advanced material properties such as viscoelastic behaviour, active-passive muscles, and geometric nonlinearities. Validation is carried out comparing the motion segment responses, the MB model responses for frontal and lateral impacts, the vertical loading results, and the responses of thoracolumbar region in rear-end impact. The model successfully reproduces the characteristic motion of the head and neck when subjected to rear-end crash scenarios. Whiplash simulations involve not only the responses of the ligamentous spine model, but also predictions of the model with active/passive musculature. The MB model simulation results and model predictions such as head translations and rotations, muscle and ligament forces, and intervertebral angles show good agreement with experiments. The study is limited to presenting the kinematics and kinetics of the cervical spine. The biofidelic whole human spine model proves to be a highly capable and versatile platform to simulate various traumatic whiplash injury situations.

INTRODUCTION

Multi-body/discrete parameter models possess the potential to simulate the kinematics and kinetics of the human spine, both entirely and partially. Multi-body models have advantages such as less complexity, less demand on computational power, and relatively simpler validation requirements when compared to FE models. Williams and Belytschko [1] constructed a three-dimensional human cervical spine model for impact simulation, which included a special facet element which allows the model to simulate both lateral and frontal plane motions. In a similar but an advanced manner, van Lopik and Acar [2] generated and validated a three dimensional multi-body model of the human head and neck using the dynamic simulation package MSC.visualNastran 4D. The model of the head-neck complex involves rigid bodies representing the head and 7 vertebrae of the neck interconnected by linear viscoelastic disc elements, nonlinear viscoelastic ligaments, frictionless facet joints and contractile muscle elements describing both passive and active muscle behaviour. Using a different approach, the emphasis is more on the lumbar spine in Jaeger and Luttmann’s work [3]. Monheit and Badler [4] constructed a kinematic model of the human spine and torso based on the anatomy of the physical vertebrae and discs, range of movement of each vertebra, and effect of the surrounding ligaments and muscles. Broman et al. [5] generated a model of the lumbar spine, pelvis and buttocks to observe transmission of vibrations from the seat to L3 in the sitting posture. De Zee et al. [6] built a multi-body human spine model partially, in which only the lumbar spine part was completed, consisting of seven rigid segments as pelvis, the five lumbar vertebrae and a thoracic part, where the joints between each vertebra set of two was modelled as a three degrees-of-freedom spherical joint. Ishikawa et al. [7] developed a musculoskeletal dynamic multi-body spine model in order to perform Functional Electrical Stimulation (FES) effectively as well as to simulate spinal motion and analyse stress distributions within the vertebrae. The muscles were incorporated to the skeletal model by using 3D analysis software MSC.visualNastran 4D.

This paper reports a rigorously validated biofidelic multi-body (MB) model of the whole human spine including whiplash simulations and analyses under various impact conditions and acceleration levels. The main advantage associated with the model lies in incorporating the whole spinal components such as vertebrae, ribs, muscles, intervertebral discs, and ligaments, which helps to simulate and validate more realistically. The MB model devoid of muscles is validated against Panjabi and colleagues’ experiments conducted using a bench-top trauma sled and isolated cervical spine specimens [8, 9]. These studies used cadaveric cervical spine specimens devoid of all non-ligamentous soft tissues fixed to a bench top sled device where an acceleration pulse is applied to the base of the specimen to reproduce whiplash trauma. These tests constitute an alternative to experiments using volunteers or whole body cadavers. They have been successfully used for developing computational models that simulate whiplash trauma and provide valuable insights into the complex events and interactions that cause injuries to the cervical spine [10].

METHOD

The MB model utilised in the simulations is recently developed by the authors [11, 12]. The model embodies highly advanced material properties such as viscoelastic behaviour, active-passive muscles, and geometric nonlinearities.

Multi-Body Model Characteristics

The multi-body model developed is constructed by employing a similar methodology to the cervical spine multi-body model of van Lopik and Acar [10]. The vertebrae were modelled as rigid bodies, interconnected by linear viscoelastic intervertebral disc elements, nonlinear viscoelastic ligaments and contractile muscle elements possessing both passive and active behaviour. The dynamic simulation package *MSC.visualNastran 4D 2001* is utilised as the computational medium.

The multi-body model of the whole human spine incorporates four essential elements: the vertebrae, the muscles, the ligaments and the intervertebral discs. The solid model of the upright erect human spine constituted the basis for the developed multi-body model, in which the geometrical surfaces that defined realistic anatomical dimensions of the spinal parts are entirely constructed from CT scans by Van Sint Jan [13] at the University of Brussels, Belgium, and stored into the software, Data Manager of Multimod project. These solid bodies not only include the essential parts of the vertebrae, but also accommodate other selected skeletal parts such as the head, the ribs, the clavicles, the scapulae, and the iliacs. The CT-scanned segments of the human skeletal parts are combined to form the whole solid model as shown in Figure 1.

Muscles are incorporated into the model as contractile muscle elements possessing both passive and active behaviour. The most essential muscle groups, such as fascicles of the erector spinae and multifidus, are integrated. Necessary geometric and morphologic features such as the origins, insertions and dimensions are taken from various studies in the literature [14-16]. In MSC.visualNastran 4D, linear actuator element is used to incorporate the muscles, governed by an external software, Virtual Muscle v.3.1.5 of Alfred E. Mann Institute at the University of Southern California [16] that runs within Matlab/Simulink and communicates with MSC.visualNastran 4D at each incremental step.

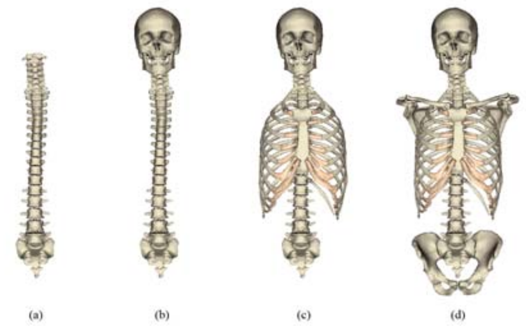

Figure 1. The solid model of the human spine as the basis of the multi-body model: (a) the entire spinal column, (b) with the head, (c) with the head and the ribs, and (d) with the head, the ribs, the clavicles, the scapulae, and the iliacs.

The ligaments in the present model are chosen as nonlinear viscoelastic ligaments. All six common types of ligaments are introduced to the model, which are ALL (anterior longitudinal ligament), PLL (posterior longitudinal ligament), LF (ligament flavum), JC (joint capsules), ISL (interspinous ligament) and SSL (supraspinous ligament) as depicted in Figure 2. The necessary biomechanical properties of human spine ligaments are taken from the literature [17, 18].

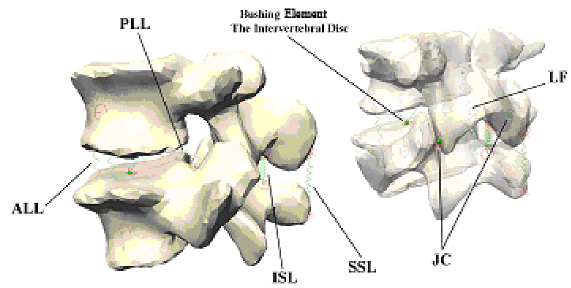

Figure 2. Ligaments and the intervertebral disc in the multi-body model.

Intervertebral discs are modelled as bushing elements in MSC.visualNastran 4D as illustrated in Figure 2. All translational and rotational degrees of freedom are allowed in a bushing element, but they are restricted through spring-damper relationships. The intervertebral discs are located at the centre of the space between the upper and lower end plates of adjacent vertebrae at a fixed distance relative to the centre of the upper vertebrae. There are no discs between the axis, atlas and occiput. Material properties of the disc for the model are collected from the studies in the literature [18-21].

All the constituting elements of the whole human spine are integrated to form the MB model as shown in Figure 3.

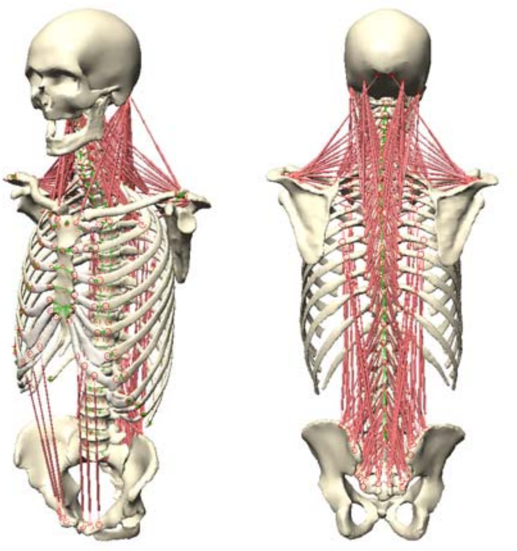

Figure 3. Oblique and rear views of the multi-body model of the whole human spine.

Multi-Body Model Validation

The validation of the multi-body model is conducted rigorously by comparing motion segment responses in the cervical spine, MB model responses in the cervical spine for frontal and lateral impacts, vertical loading for cervical spine, and MB model responses of thoracic and lumbar regions in rear-end impact. In 15g frontal and 7g lateral impact cases, the model is validated against well-known NBDL data [22]. In Figures 4 and 5, typical results from frontal and lateral impact cases are shown, respectively. C3C4 displacements and rotations under certain loading are provided in Figure 6 as an example for validation by using motion segment responses.

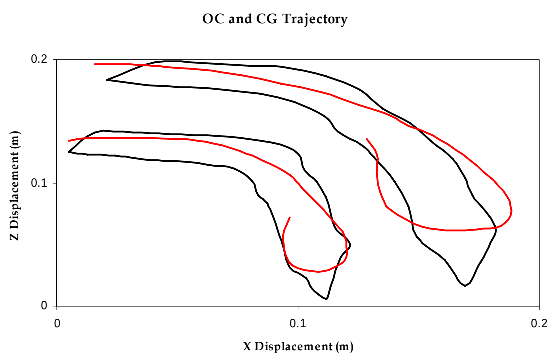

Figure 4. Head occiput and head centre of gravity trajectories in the horizontal (X) and vertical (Z) planes (OC lower, CG upper graph).

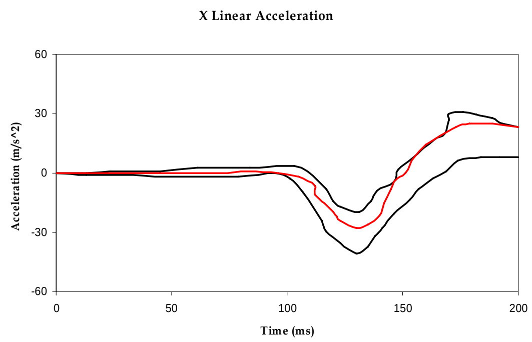

Figure 5. x linear acceleration of head centre of gravity vs. time in lateral impact.

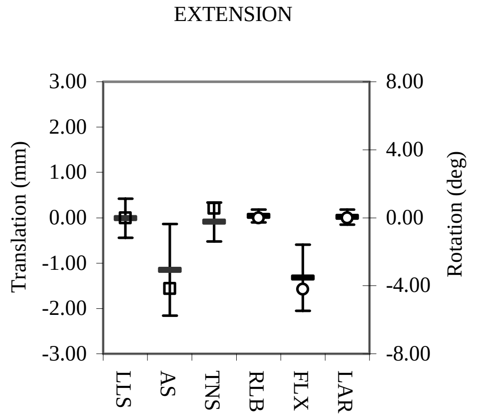

Figure 6. Displacements of model motion segment C3C4 in response to applied rotational load of 1.8 Nm for extension shown against the experimental results [23]. Resulting displacements are shown along the vertical axis, translations (□) on the left, rotations (○) on the right. Anterior shear (+AS), posterior shear (-AS), left lateral shear (+LLS), right lateral shear (-LLS), tension (+TNS), compression (-TNS), right lateral bending (+RLB), left lateral bending (-RLB), flexion (+FLX), extension (-FLX), left axial rotation (+LAR) and right axial rotation (-LAR).

ANALYSIS & RESULTS

The MB model is used to simulate the experimental conditions of Panjabi and co-workers [8, 9], who utilised a bench-top sled to simulate whiplash trauma on ligamentous human cadaveric cervical spine specimens which were without the muscle tissue and mounted to the sled at T1. The whiplash trauma input in the horizontal direction was introduced as the profile of the sled acceleration-time curve to the base of the specimen represented. The acceleration input was a triangular pulse with duration of 105ms and peak accelerations of 2.5g, 4.5g. 6.5g and 8.5g (1g = 9.8m/s^2).

In the MB simulation, all muscles are deactivated. The motion of T1 is constrained so only translation along the x-axis was allowed. No gravitational effect is taken into consideration at this stage. The acceleration profiles are triangular with the same 105ms duration and corresponding peak accelerations as depicted in Figure 7.

The resulting head rotations and translations are compared against the results for the 8.5g trauma class.

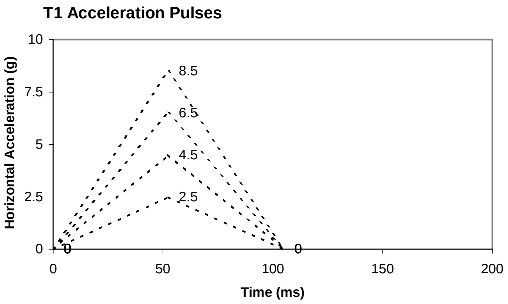

Figure 7. T1 acceleration profiles used as input to the cervical spine model.

The response of the ligamentous spine model to the 8.5g trauma acceleration is provided schematically in Figure 8. In the resulting head-neck motion a characteristic S-shaped curvature of the neck with lower level hyperextension and upper level flexion and subsequent C-shaped curvature with extension at all levels of the entire cervical spine are observed.

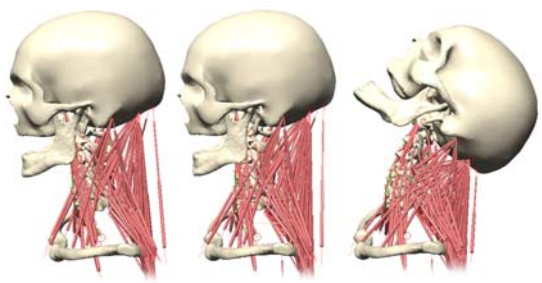

Figure 8. Response of the model to 8.5g whiplash acceleration for 0, 60, 120 ms (respectively, from left to right). Muscles were deactivated completely.

The head rotation, and head vertical and horizontal translations for the 8.5g case compared with the experimental results [24] is provided in Figure 9. The model shows a similar response to the cadaveric spine specimen, where head rotation follows a similar pattern but with a higher peak value. The MB model returns back slightly slower than is seen with the spine specimen soon after the maximum rotation and maximum posterior translation of the head. The vertical displacement of the head reaches a peak of around 6 cm below the initial height and shows good agreement with the experimental results.

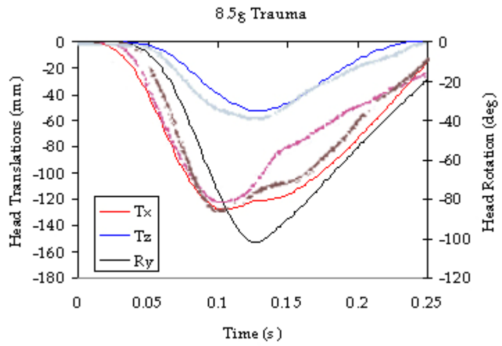

Figure 9. Model head translations and rotations for 8.5 g case compared to experimental values (which are shaded with similar colour).

During the acceleration part of the impulse, the head translates posteriorly and inferiorly with respect to T1 as the spine extends. Around 60 ms time period, the development of the characteristic S-shaped curvature of the cervical spine is observed. The vertebral rotation graphs in Figure 10 depicts that during this time period the upper levels of the spine (C0-C3) are flexed while the lower levels (C5-T1) are extended as observed from the experimental results. In the 75-100 ms time period, the upper vertebrae of the model change from flexion to extension as the whole model becomes more and more extended into a C-shaped curvature as also observed in the experiments. Maximum extension of the head and neck is reached at approximately 130 ms, slightly later than the experimental results. In the later stages of trauma the head rebounds almost completely to its initial starting configuration.

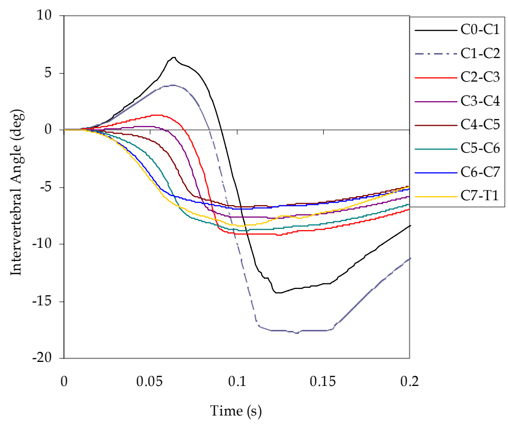

Figure 10. Intervertebral rotations at 8.5g impact.

Model predictions for head translations and rotations are provided in Figures 11-13, From which it can be observed that the more severe the impact, the greater the rotations and translations are.

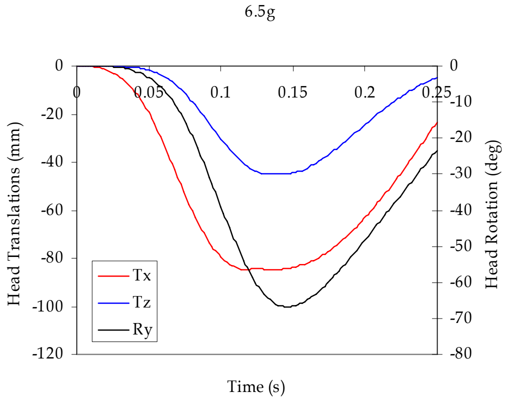

Figure 11. Model head translations and rotations for 6.5 g case.

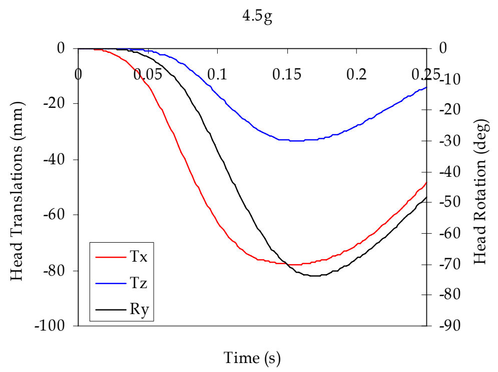

Figure 12. Model head translations and rotations for 4.5 g case.

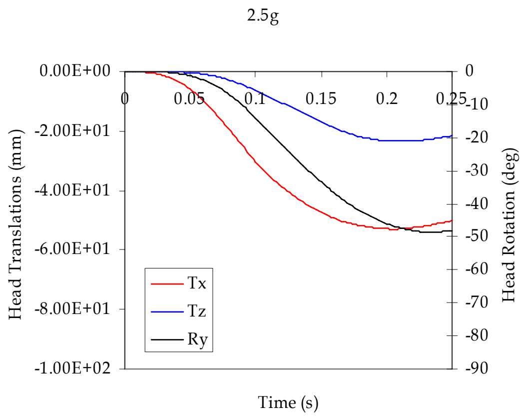

Figure 13. Model head translations and rotations for 2.5 g case.

The maximum intervertebral rotations of the model for the four cases simulated are presented in Figures 14-18. For the C2-C3 level of the cervical spine, the graph (Fig. 14) show that although the upper levels are initially forced into flexion in the model, the levels of flexion experienced are slightly smaller than the experimental values, which may be an indication of the model being slightly stiff in flexion in these areas. The levels of extension experienced in the later stages of impact show better agreement with the experimental data. Figures 15-18 show the maximum intervertebral extension rotations experienced by the lower five levels of the spine model. From the results, it seems that generally level C6-C7 appears to be too stiff when compared to the experimental results.

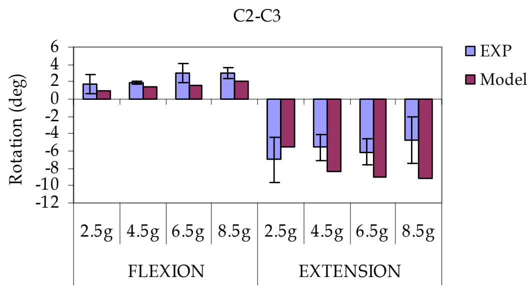

Figure 14. Maximum intervertebral angles achieved for C2-C3.

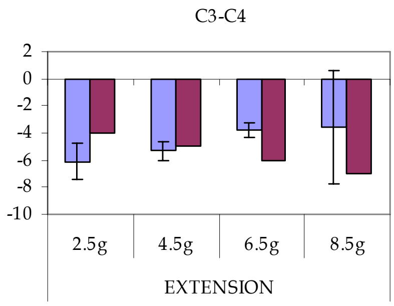

Figure 15. Maximum intervertebral angles achieved for C3-C4.

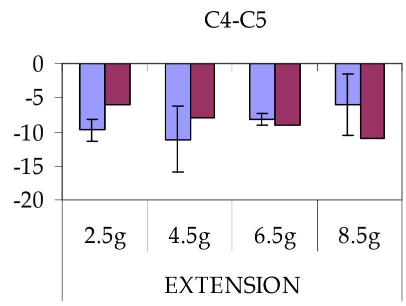

Figure 16. Maximum intervertebral angles achieved for C4-C5.

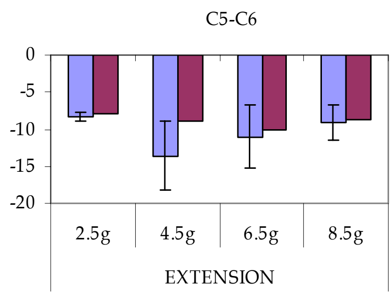

Figure 17. Maximum intervertebral angles achieved for C5-C6.

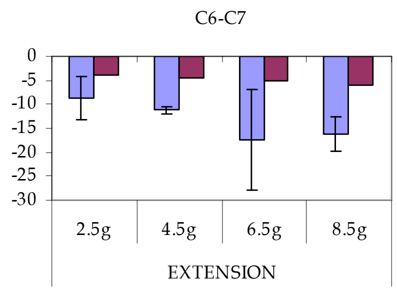

Figure 18. Maximum intervertebral angles achieved for C6-C7.

In order to investigate the effects of musculature, the muscles are incorporated into the model in two ways; as passive musculature and as active musculature. The comparison of model predictions for maximum intervertebral disc forces in tension and compression cases for 8.5g acceleration are tabulated in Table 1.

Table 1. Comparison of maximum disc forces in tension and compression for 8.5g acceleration case.

Force (N)	C2-C3	C3-C4	C4-C5	C5-C6	C6-C7
Maximum tension					
No muscles	95	62	25	2	0
Passive	223	132	64	31	0
Active	103	47	13	3	0
Maximum compression					
No muscles	224	273	281	279	265
Passive	257	303	294	310	289
Active	394	417	432	423	402

In maximum tension predictions for 8.5g case, the inclusion of passive musculature significantly increases the values at each level. Within each case, the forces decrease from C2-C3 to C6-C7. Inclusion of active musculature normalises the values towards no-muscle case, which appears to be more realistic in the light of the experimental data. In maximum compression predictions, active muscles seem to have a considerable effect on the maximum values with relatively higher magnitudes, whereas both no muscle and passive muscle cases exhibit similar values.

CONCLUSIONS

This study shows that the MB model of the whole human spine can be used to simulate a ligamentous cervical spine undergoing whiplash trauma. The MB model devoid of muscles is validated against test results, while most of the simulation results and model predictions showed good agreement with experiments. The model can successfully reproduce the characteristic motion of the head and neck when subjected to rear-end impact. The differential movement between the head and T1 causes initial flexion in the upper joints as the head translates backward, without rotation, relative to T1.

In the most severe impact case of 8.5g, the head rotations and displacements show reasonably good agreement with experimental data, particularly in following the trends of the empirical graphs. However, the model predictions yield slightly larger values than the experimental results.

The inclusion of the muscles into the model does not significantly alter the head and cervical spine rotations. However, the forces occurring at intervertebral levels are considerably affected due to muscle tensioning. It could be concluded from the model predictions with active musculature that an initially unaware occupant would not be affected in terms of cervical spine kinematics, but would be influenced via the varying loads within the soft tissues such as intervertebral discs.

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