Proposing a 2D Dynamical Model for Investigating the parameters Affecting Whiplash Injuries

Seyed Mohammad Rajaai[i], and Mohammad H Farahani[ii]

ABSTRACT:

This paper proposes a 2D dynamical model for evaluating parameters affecting whiplash. In fact a four segment dynamical model is developed in the sagittal plane for the analysis. The model response is validated using the existing experimental data and is shown to simulate the "S-Shape" and "initial upward ramping" kinematics of the cervical spine and the resulting dynamics observed in human and cadaver experiments. The model is then used to evaluate the effects of parameters such as velocity change between rear vehicle and the target vehicle (Δv), head/head restraint separation (backset) and the awareness of occupant on the whiplash injuries. It is shown that the proposed model can simulate whiplash phenomena very well; therefore it is a suitable alternative for other existing models.

KEY-WORDS:

Whiplash injuries, Dynamic model, Adams, Backset, Velocity change, Awareness

1. INTRODUCTION

The name "Whiplash" derives from the description of the sudden sharp whipping motion of the head and neck, produced at the moment of a traffic accident, particularly subsequent to collisions from the rear-end. Due to rear-end collisions head and neck are first subjected to hyper extension followed by hyper flexion of the neck. This sudden sharp whipping motion can cause pain and also a variety of auditory or visual disorders such as tinnitus, vestibular dysfunction, dizziness, blurry vision and anosmia [1]. These kinds of disorders are not fatal but their cost of insurance is too high. Whiplash injuries therefore represent a substantial societal problem worldwide with associated costs that are estimated at $4.5 -10 billion annually in the USA and at €1 billion annually in Germany [2].

Unlike most automobile-related injuries, the risk of whiplash injury has increased over the last few decades [3, 4, and 5]. In one recent report, the risk of sustaining a whiplash injury with symptoms lasting at least one year was 2.7 times higher in vehicles introduced between 1989 and 1992 than in vehicles introduced between 1981 and 1986 [6]. These findings call for improved methods for testing and assessing protective devices such as seat and car structures. The most essential component when simulating the crash is dummy and data acquisition system that their cost is too high, and further more repeatability of the test would be another important factor. Using computer simulations could be a proper method for avoiding these problems.

Validated mathematical models with high accuracy, when compared with experiments, are useful tools for the simulation of physical tests which could be required a considerably large amount of time and financial resources.

The two modeling techniques which are usually used for the development of mathematical models of humans and dummies are multi body systems (MBS) and finite element modeling (FEM). One of the early mathematical models of whiplash was developed by Martinez and Garcia. Their non-linear model described head motion to be translated relative to the shoulder, causing a shearing action in the neck, and rotation of the head about the top of the cervical spine. The head was given an axis for oscillating relative to the shoulder and an axis for rotation. Both axes were restrained by non-linear springs with hydraulic damping. The deflection of seat back and torso was modeled by a rigid rod connected by a pin with non-linear springs at the hip pivot. Analysis results using this model led to the conclusion that the differences in the head and shoulder accelerations result in neck stretch and shearing deformation which are also affected by seatback tilt, spring stiffness and damping [7].

McKenzie and Williams [8] developed an analytical model in order to investigate the dynamics of the cervical spine during whiplash. They modeled the vertebrae of the

[i] S. M. Rajaai is with the department of Mechanical Engineering, Iran University of Science and Technology, Tehran, Iran (email: s_rajaai@yahoo.com)

[ii] M. H. Farahani is a master student within the department of Mechanical Engineering, Iran University of Science and technology, Tehran, Iran

cervical spine as rigid bodies, and the car seat back and torso as one rigid body pivots at the hip. The vertebral bodies were connected by intervertebral discs that behave as viscoelastic, short, uniform beam segments. Results predicted by the model showed a similarity to the resultant acceleration of the shoulders and the head in the experimental results of Severy, et al. [9].

In simulating rear-end collisions, more recently, Linder [10] developed a comprehensive multi-body mathematical model that was implemented in MADYMO 2D. The model consisted of rigid bodies representing C1 through T1 and the head. Anterior and posterior "strings" that connected the vertebrae modeled a combination of muscles (stiffness and damping) acting on the neck. The vertebral joints contained the stiffness parameters to model the ligaments. The model included the upper portion of a seat back, including a head restraint. The stiffness and damping properties, along with several other parameters were optimized to fit human data. The result of the optimization was a model that closely reproduced the human data. The mathematical model was made to aid in the design of the head and neck of the BioRID which stands for Biofidelic Rear-end Impact Dummy.

2. Research Method

Dynamical model

The model that is introduced in this paper is a MBS dynamical model. This MBS model is a four segment model and consists of three rigid rods connected by four revolute/pin pivots and connected to the head at its center of mass. This model would allow easy calculation of neck loads for the range of impact speeds and other variables. It is showed that the four segment model proposed in this paper reproduces the observed S-shape kinematics of the cervical spine and also the upward ramping of the torso at the beginning of the collision. The model is used to evaluate the effect of some of the parameters such as collision severity, head separation from the head restraint and the occupant's awareness.

According to these results, the model presented in this paper provides an alternative to the more complex models such as that of Linder [10], Mackenzie and Williams [8], and Deng and Goldsmith [11].

The model has four segments and consists of three rigid rods connected by four revolute/pin pivots and connected to the head center of mass. The model combines the body and pin into segments which are connected with rotational springs and dampers at each pivot. The model is constrained to 2D motion in the sagittal plane.

The first pivot was chosen to be at T4-T5 because it is where vertebrae superior to it are unsupported by the seat. Therefore this is where rotation occurs due to forward acceleration of the car seat and after contact to the torso with the seat. This was shown by the x-ray results of Mcconnell et al. [12] and Luan et al [13].

The model developed includes a pivot at C7-T1 level to allow rotation between the thoracic and cervical spine. Another lower cervical pivot (at C5-C6) was added to allow modeling of motion at this level of cervical spine.

The mass of the rigid links was determined by various methods (table 1). The link from T4 to T1 is assumed to have a mass of about one-half of the mass of the thorax. The cervical link masses were determined from the measured values of 50th percentile male individual vertebrae. The link from the Occipital Condylus (OC) pivot to the head center of gravity is mass-less because all the head mass is assumed to be contained at the point representing the head center of gravity.

The lengths of the links were determined by various methods as well (table 1). The length of the thoracic link was calculated by subtracting the vertebral height of T5 from the length of one-half of the thorax height. Lengths of the cervical links were determined from a proportion of vertebrae and disc heights determined from the standard man values.

The moment of inertia was defined as a uniform slender rod for the thoracic rigid link, by either experimentally found values or 50th percentile male values for the head.

Table 1 various properties of the links [14]

Links and head	Mass [Kg]	Length [cm]	Moment of inertia (sagittal plane) [Kg×m^2]
T4-T5 to C7-T1	8.14	10.00	6.78e-3
C7-T1 to C5-C6	0.456	3.75	5.34e-5
C5-C6 to OC	1.02	9.51	7.69e-4
OC to head C.G	0.0	5.03	0.0
Head	5.5	NA	0.035

Initial values of angles in the neural position were determined from radiographs of people in the seated position and other anatomical considerations (table 2).

Table 2 Initial values of angles in the neural position [14]

Link	Initial angle with respect to vertical [deg]
T4-T5 to C7-T1	20
C7-T1 to C5-C6	16
C5-C6 to OC	3
OC to head C.G	33

The relevant stiffness and damping were obtained from dynamic data from Mertz and Patrick [15] and also

Ravani and Garcia [14] (table 3).

Table 3 Stiffness and damping of the pivots [14], [15]

Pivot	Stiffness [Nm/rad]	Damping [Nm-s/rad]
T4-T5	100 flexion 140 extension	2.5
C7-T1	20 flexion 50 extension	0.8
C5-C6	20 flexion 50 extension	0.8
OC	40 flexion 80 extension	2.0

By simulating the rear-end collision by using ADAMS software a 2D dynamical model produced (figure 1). All segments of this model assumed to have rigid behavior. For simulating collision a rectangle was used as the rear car. The produced model is shown in figure 1. The impact was assumed as completely elastic.

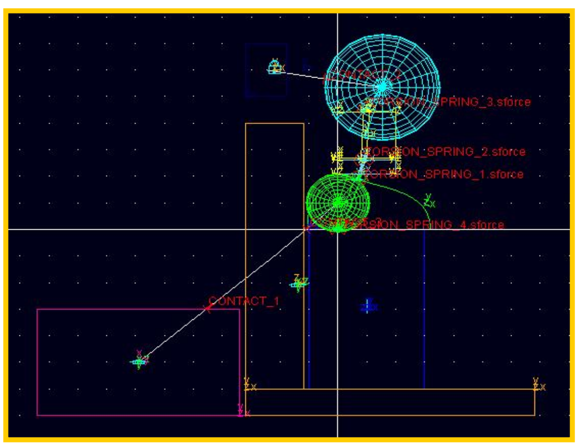

Figure 1 Dynamic model

In order to validate the kinematics of the model, two main factors were investigated. The first factor was S-shape formation of the cervical spine before the neck reaches its full hyper extension. This formation has been showed in figure 2.

The second factor was the upward ramping of the torso at the beginning of the accident. For surveying this matter y-position of head center of gravity was obtained against time for many accident conditions. All results showed that this model can simulate the upward ramping of torso immediately after the accident (figure 3).

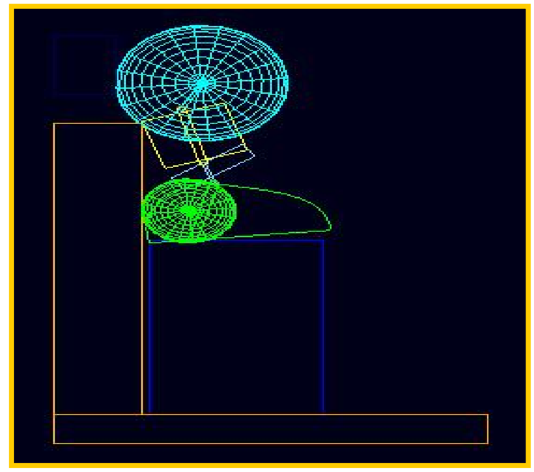

Figure2 S-shape formation of the model

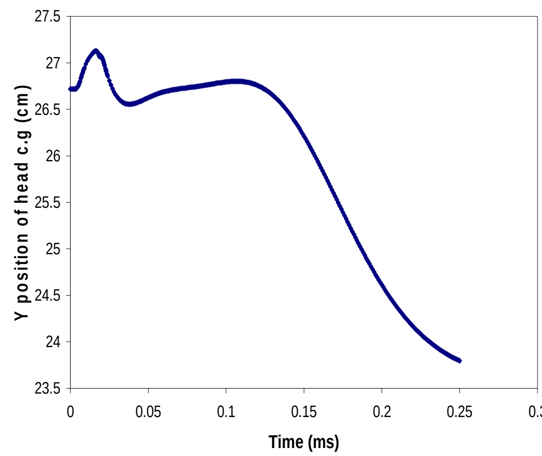

Figure 3 Upward ramping of the torso

The y position of head center of gravity in figure 3 has been obtained from the local coordinates on the shoulder.

3. Results and Discussion

In this section simulations were run using the model. The main purpose of proposing this model was firstly to produce a simple model and then to investigate the parameters affecting whiplash injuries. Therefore parameters such as head/head restraint separation (backset), velocity change between rear vehicle and the target vehicle (ΔV), and awareness of occupant in target vehicle were chosen as variables. In order to validate the model dynamically, some important variables in whiplash phenomena are obtained from the model and compared with the results of other researchers. In all simulations it was assumed that the occupant has fastened the seat belt, $\Delta v = 20 \frac{Km}{hr}$, and backset=8 cm. It has been mentioned in the analysis if some items are variables in special cases. Since the comparison outcomes were satisfactory, parameters affecting whiplash injuries could be investigated by the model. The first variable considered for the analysis, was the head angular acceleration. The relevant results are shown in figure 4. The results obtained by from the simulation have been compared to the results of Ravani and Garcia [14] and also Mertz and Patrick [15] studies.

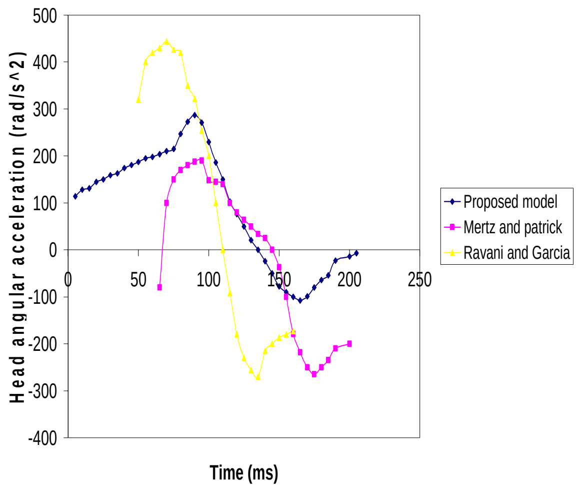

Figure 4 Head angular acceleration

According to figure 4, results of the proposed model are in compliance with the results obtained by the other researchers.

The second variable investigated, is neck moment at OC. The relevant results are represented in figure 5.

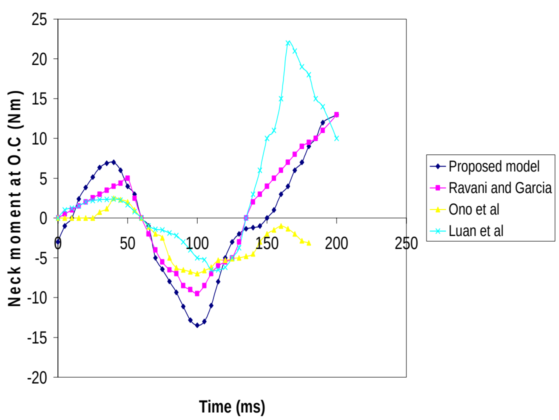

Figure 5 Neck moment at O.C

Other variable investigated, is neck shear force at C5-C6 that its results are illustrated in figure 6.

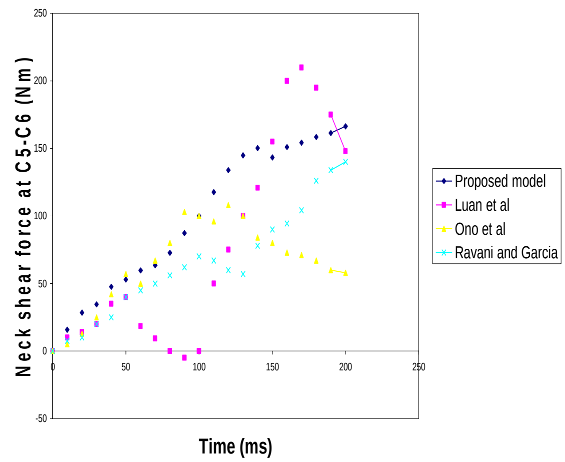

Figure 6 Neck shear force at C5-C6

According to the results shown in figures 4, 5 and 6 and many other results obtained by the model, the proposed model can be validated dynamically. Therefore it was possible to investigate some of the parameters affecting whiplash injuries.

First parameter investigated, is the effect of velocity change between the rear vehicle and target vehicle (Δv) on neck moment C7-T1. Results are shown in figure 7. By increasing Δv neck load will be increased as shown in figure 7.

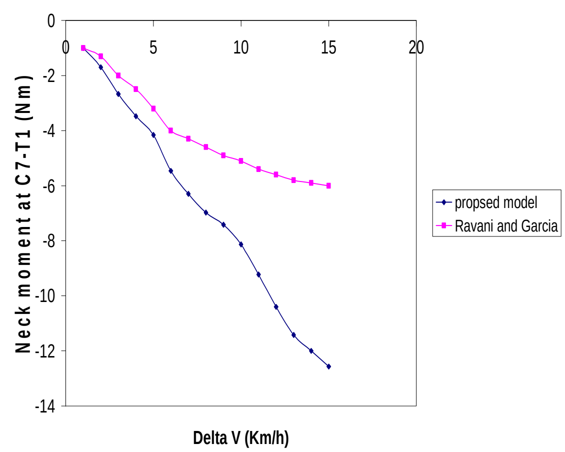

Figure 7 the effect of Δv on neck load

Another parameter investigated by running the simulation is the role of awareness in the amount of neck loads. In order to perform this investigation it is assumed that the stiffness and damping in the joints are more than the normal condition. Initially due to awareness of the occupant theses variables are considered 1.1 times and then 1.2 times. The simulation results are produced in figure 8.

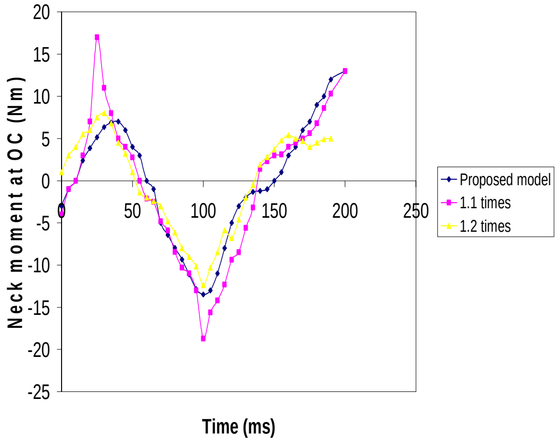

Figure 8 the effect of awareness on neck loads

The model behavior shows that if the occupant be aware of accident the amount of injuries will be decreased as observed in the real life.

Another parameter investigated by running the simulation is the presence of the head restraint. The relevant results are shown in figures 9 and 10.

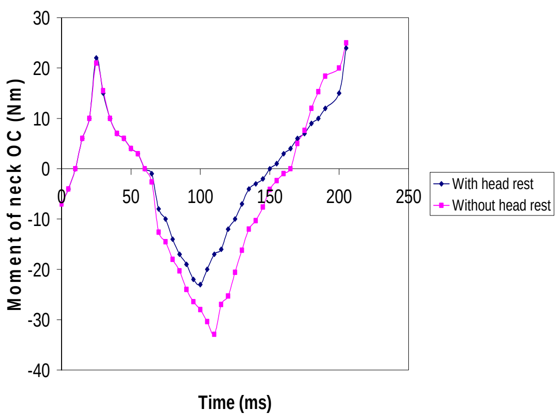

Figure 9 the effect of presence of head restraint on lessening neck loads

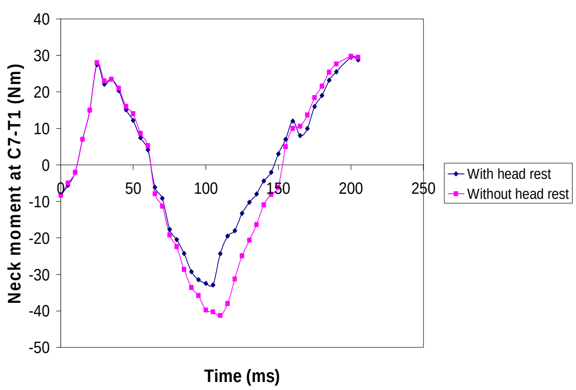

Figure 10 the effect of presence of head restraint on lessening neck loads

According to figures 9 and 10 it is obvious that when the head restraint is removed from the seat more loads are applied to the neck.

The last parameter is the role of backset in the amount of neck loads. The results of running the simulations are shown in figure 11.

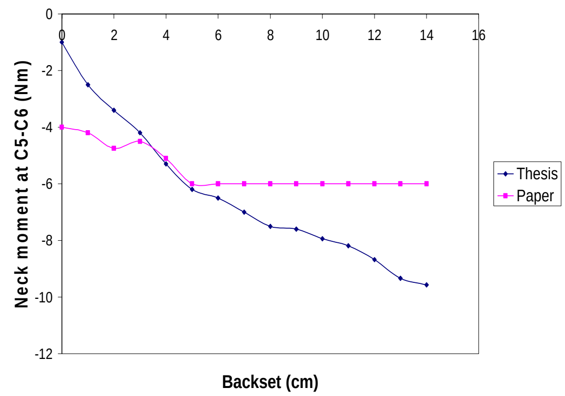

Figure 11 the effect of backset on neck loads

According to the results shown in figure 11 the model behavior shows that by increasing the backset thickness more loads will be applied on the neck.

4. Conclusions

The first objective of this study was to achieve a design of a relatively simple and efficient multi-body linkage model that can be used to simulate the biomechanics of whiplash. It is shown that a two dimensional four segment model in the sagittal plane can capture the peak loads as well as the kinematics of the cervical spine observed in the experimental studies. Data from three experimental studies are used to validate the model. Simulation run with the model are shown to produce peak accelerations and moments as well as the kinematics that agree with other existing experimental data. Much of the data from the model was seen to lie between experimental data from human volunteers and cadavers. This implies that the proposed model simulates the kinematics and dynamics of aware and unaware human subjected to rear-end impact very well. Therefore the first objective is met. Some differences observed are because the conditions of the experiments or simulations of researchers are different from each other.

Finally the model was validated kinematically and dynamically. Therefore it can be used for investigating some parameters affecting whiplash injuries. The results of simulations were compared with the results of other researchers. According to the comparisons the model was found suitable for simulating rear-end collisions. This model is also suitable for investigating various parameters affecting whiplash injuries.

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