Research on head-neck injuries of pilots during emergency ejection from the aircraft

Run-Zi Song¹, Shuang-Fu Suo¹, Xiao-Hong Jia¹, Yue Liu¹ and Song-Yang Liu²

¹Department of Mechanical Engineering, Tsinghua University, Beijing 100084, China
²Air Force Aeronautical Medical Institute, Beijing 100089, China

Abstract. This paper established a multi-body dynamics model of head-neck using LifeMod. To explore the damage of pilots’ head-neck during the ejection process, we used the established model to do simulation combined with the seat and helmet multi-body model. The head-neck model was validated through the volunteer frontal-collision and rear-collision tests, whose head acceleration, displacement curve and response time are in good agreement with the experimental data. The simulation showed that the head-neck is the most vulnerable to injury, when the head lean forward to the limit position and collide with headrest due to airflow. The results of various simulation calculations implied the head-neck protective device should be installed to avoid pilots’ neck damage in the ejection.

1. Introduction

Ejection lifesaving is the only way to save pilots’ lives in the fighter plane accident[1]. However, in the emergency ejection, pilots will be easily affected by impulsive overweight and airflow[2]. And then pilot’s head-neck moves quickly relative to the body, which can lead to neck injury or even death[3] of pilots. Wearing night vision goggles and other equipment increased the weight of helmet, so that it increases the probability of damage [4]. According to the US Air Force statistics[5], the total incidence of head, neck and cervical spine injuries are very high, accounting for 29% of the total ejection. ACES-5 seats[6] in the United States and US16E seats in the United Kingdom have been equipped with head-neck protective devices. Consequently, to provide a theoretical basis in this field, it is necessary to study the mechanism of head-neck damage during ejection.

The research on head-neck injury mechanism includes biological experiment[7], physical model[8] and mathematical model. Computer simulation has become an indispensable method of research[9]. Mathematical models can be divided into three categories: lumped mass model, multi-rigid-body dynamic model and finite element model[10]. Among them, the multi-rigid-body model is simple to construct, fast to calculate and can be accurately simulated for head-neck kinematics and dynamics[11]. Huston et al. established 3-rigid body and 9-rigid body models[13-14] of human head-neck based on Kane’s method[12]; Y.C.Deng established a 10-rigid-body model[15] to study the dynamic characteristics of collision process; M. de Jager established a head-neck overall model and a detailed model[16-17]; Horst refined the muscle of Jager’s model[18]; at present, the biomechanical simulation software commonly used in human multi-body dynamics simulation includes SIMM, LifeMod and Anybody, etc[19]. LifeMod is a human modeling plug-in unit based on Adams, which can simulate with other mechanical structures built by Adams. Zhang Jing used LifeMod to build 5-vertebral head-neck model and validated them[20]; Liu Lan et al. used LifeMod to build a model of type III dummy[22].

Based on the biomechanical simulation software LifeMod, the head-neck model of human body is established in this paper. By simulating the frontal and rear collisions of the model, we get the
dynamic responses of head-neck. Comparing with the experimental data, we found that the model has good biological fidelity so that produced a desirable conclusion. Besides, we develop multi-body dynamic model of human-seat to explore the mechanism of head-neck injuries in different ejection conditions with the help of Adams.

2. Establishment of the head-neck multibody dynamics model

2.1 Establishment process of human head-neck model

Based on the anatomical parameters of pilots in Human dimensions of Chinese male pilot population GJB 4856-2003[21], the multi-body dynamics model of head-neck was built for 50-percentile pilots. Model parameters: height 1704 mm, weight 68 Kg, head mass 5.5 Kg, moment of inertia {314, 304, 190} kg∙cm²; head movement limit: head-neck flexion 60°, backward extension 70°, left and right torsion 80°.

We set up the human head-neck and upper body model based on Adams biomechanical simulation plug-in unit LifeMod, as shown in Figure 1. In the coordinate system of the model, the X axis corresponds to the sagittal rotation axis, the Y axis corresponds to the parallel median coronal axis, and the Z axis corresponds to the lead hammer axis[22]. The model is constructed by rigid vertebrae and viscoelastic muscles and joints, mainly including skull, cervical spine, scapula and upper trunk four parts. The cervical spine is partially refined into seven vertebrae (C1-C7). The model consists of eight joints. Each vertebrae of the cervical spine are connected by joints. The cervical spine C1 is linked with the skull by joints; the cervical spine C7 is connected with the sternum T1 by joints. The model included eight muscles (sternocleidomastoid, scalenus, splenius cervicis and splenius capitis), which play a major role in head and neck movement, and only provide tension. The upper part of the body including the thoracic spine and so on plays a major role in fixation.

Each joint has three rotational degrees of freedom. The rotational angle is defined according to the range of motion of every cervical vertebra [23], so that the range of motion is limited. The head-neck model can complete the basic movements of human head-neck such as bending, stretching, lateral bending and rotation [24].

Figure 2 reveals the simulation process[19]. Firstly, the basic human body model is established, simplified and redefined for the need. Joints and muscles are added to the model; environmental constraints are set; contact forces are added between bones and environment. Adjust the model to the desired attitude, run simulation, solve, obtain data, and further optimize the model.

2.2 Simulation and verification of the human head-neck model

At present, to verify the accuracy of the model, most head-neck models are compared with the US Naval Laboratory’s frontal-collision test or Chalmers University’s rear-collision test. To evaluate the accuracy of this model, we analyse the data between head-neck model simulation and frontal-rear collision tests.
2.2.1 The simulation and comparison of frontal-collision test. Ewing et al. carried out the frontal-collision trolley test [25-26] in the U.S. Naval Laboratory. Ten volunteers were selected to be strapped to the pulley seat, and the front collision, rear collision and side collision with different accelerations were carried out. The head dynamic responses were recorded.

The head and neck model is placed in the gravity field of 1g. The parts of upper trunk are fixed, and only the degree of freedom of movement in the X direction is retained. Head-neck can rotate relative to thoracic spine T1. The velocity of thoracic spine T1 in 15g acceleration frontal-collision test is taken as the boundary condition. The X-direction acceleration and Z-direction displacement data of the model can be obtained by simulation. Compared with the experimental data, the validity of the model can be confirmed if the simulation results fall within the scope of the experimental data.

The head acceleration curve and displacement curve are obtained by simulation. Compared with the experimental results, the simulation curve basically falls within the range of the experimental curves, and the neck movement process fits the actual movement process well. The X-direction acceleration curve is in good agreement with the test curve, and the Z-direction displacement peak value is slightly smaller than the test value. Reasons for the small peak displacement of Z-direction: 1. The Z-direction motion of the head-neck model’s upper trunk was limited, but the trunk of the human body was not fixed absolutely in the experiment. 2. The model is built according to China’s 50 percentile male pilot, so ethnicity and height are different from those tested. 3. The model does not construct all muscles, ligaments, intervertebral discs and other soft tissues, so there are some deviations.

2.2.2 The simulation and comparison of frontal-collision test. Davidsson et al. carried out the rear-collision trolley test at Charles University [27]. A total of 13 volunteers were selected, who were bound by seat belts on the seat of the trolley. The trolley was impacted by a 560 kg trolley. The head dynamic responses of the volunteers were recorded.
The boundary conditions of the model simulation are the displacement of thoracic spine T1 in Z and X directions, and the other boundary conditions are same as those in the simulation of frontal-collision.

![Figure 7. Displacement curve of T1 in Z direction.](image)

![Figure 8. Displacement curve of T1 in X direction.](image)

The acceleration curve of head in X direction obtained by simulation basically falls within the test range. At 0.13ms-0.17ms, the head displacement in X direction is slightly smaller than the test value, and the head movement process is basically consistent with the experimental process.

![Figure 9. The process of head-neck movement.](image)

![Figure 10. Acceleration curve of head relative to T1 in X direction.](image)

![Figure 11. Displacement curve of head relative to T1 in X direction.](image)

Generally speaking, the simulation data of the model fits the data of the frontal and rear collision tests. This implies that the model has a very good degree of authenticity, which can reflect the multi-body dynamic characteristics of the head-neck accurately.

3. Establishment of the human-seat multibody model and exploration of head-neck injury during ejection

3.1 Force analysis of ejection lifesaving
The ejection process is divided into seven stages respectively [26]: preparation, start-up, channel clearance, escape from aircraft, stable deceleration, human-seat separation and stable landing, just as shown in Figure 13. From the beginning of ejection to the seat leaving the aircraft, the human body mainly bears the overload in the direction of +Gz [27]. The ejection seat will be blown by strong airflow after leaving the cabin, which will cause brake overload –Gx. The high-speed airflow flowing through the helmet surface will bring more than 1800N aerodynamic lift [28], and the face-to-face airflow will bring great aerodynamic drag [29]. The combined action of ejection overload and airflow will lead pilots’ head to bend forward quickly, followed by jerked up and back, so that the head will impact headrest [28]. The cervical spine will injury in this process.

Before 0.5s, the seat will be propelled by the ejection gun and rocket pack, and the pilot will be subjected to a large longitudinal overload +Gz. Before 0.22s when ignition of the rocket pack, pilot’s head-neck will be exposed to the outside of the cabin and blown by airflow [29]. That is, within 0.5 seconds, the head and neck of the human body will be affected by the combined effect of ejection overload and airflow, and the airflow is the strongest when the seat just leaves the cabin [29], during which the head-neck is most vulnerable to injury. In this study, we adopt the human-seat model to simulate the ejection condition during the period of 0.5s in the beginning of ejection, so as to analyse the head-neck injury in this process.

3.2 Establishment of human-seat mode and application of boundary conditions
The established human head-neck model is put on the ejection seat model with the helmet model on. The pilots’ helmet model is built according to the standard pilots’ helmet whose measured mass is 1.73 kg. The coordinate system of the human-seat model is the same as that of the head-neck model.

Figure 12. Displacement curve of T1 in Z direction.
Figure 13. Displacement curve of T1 in X direction.
Figure 14. The human-seat model.
The head-neck injuries under three different ejection conditions are explored by using the human-seat model. The curve of horizontal velocity and horizontal overload $-G_x$ with time after the seat leaves the cabin is shown in Fig. 15/16\textsuperscript{[29]}. The three ejection states are shown as follows:

1. height: $H = 1000\text{m}$, airspeed: $V = 1000\text{km/h}$;
2. height: $H = 1000\text{m}$, airspeed: $V = 600\text{km/h}$;
3. height: $H = 8000\text{m}$, airspeed: $V = 1000\text{km/h}$.

![Figure 15. Airspeed.](image1)

![Figure 16. The change of $-G_x$ with time.](image2)

In the first 0.5 seconds of ejection process, the human body will be affected by ejection overload, aerodynamic drag, aerodynamic lift and aerodynamic overload. When the ejection overload is 20g, the endurance limit of human body is 0.2s\textsuperscript{[6]}. The time to reach the peak value of overload is about 0.15s\textsuperscript{[29]}. Assume that the ejection overload peak value is 20g and the peak duration time is 0.15s. The expression of aerodynamic drag\textsuperscript{[27]} is as follows:

$$F = \frac{1}{2} \rho V^2 C_x A_x$$

where $C_x$ is aerodynamic coefficient; $A_x$ is flat plate area; $\rho$ is air density; $V$ is relative velocity of air. Noting the airflow resistance begins to exert effect at 0.15s, the aerodynamic resistance yields according to (1), as shown in Figure 17 (b). Suppose the aerodynamic lift starts at 0.15s, which is depicted in Figure 17 (c). Besides, we get the overload curve\textsuperscript{[29]} (Figure 17 (d)) under the case aerodynamic overload starts exert effect at 0.1s.

![Figure 17. Force on pilots’ head during ejection.](image3)
3.3 Human-seat model simulation results

The model produces the movement process of human head-neck by simulation, which is presented in Figure 18. The head-neck of human body is subjected to ejection overload in the beginning, then rapidly tilts forward, and reaches the extreme position of the forward tilt at about 0.2s. Next, under the action of airflow, the head-neck is swiftly jerked backward and upward, collides with the headrest at about 0.3s. After the collision, the head-neck tilts forward again. The simulated motion process corresponds with the real ejection process. The relative velocity and acceleration between C1 and T1 were obtained, as shown in Figure 19/20.

3.4 The conclusion of research on head-neck injury in emergency ejection

Based on the neck damage criterion NIC \[^{[30]}\], we adopt the NIC formula to explore the neck damage, namely,

\[
NIC = a_{\text{relative}} * 0.2 + V_{\text{relative}}^2
\]  

(2)

where \(a_{\text{relative}}\) is the relative acceleration between the first thoracic vertebra T1 and the first cervical vertebra C1 in the horizontal direction, \(V_{\text{relative}}\) is the relative acceleration between T1 and C1 in the horizontal direction, 0.2 is length in m. The critical value of NIC is 15 m/s\(^2\) and damage occurs once the value of NIC exceeds 15 m/s\(^2\). According to the research, there is a 50% risk of neck injury when the peak value of NIC reaches 15 m/s\(^2\); when the NIC value achieves 32 m/s\(^2\), there is a 100% risk of neck injury. Figure 21 characterizes the NIC values calculated in detail.
By observing the NIC curves in Figure 21, we arrive at the conclusion:

1. There are two peaks in each curve. The first peak stands for the head bending forward to the limit position, and the second peak represents the head bumping against the headrest. The injury can easily occur when the head moves to these two positions.

2. In the mentioned ejection cases, the NIC value is bigger than 15 m/s² when the head bending forward to the limit position, which indicates that neck injury risk is beyond 50%.

3. Head collides with headrest is mainly caused by airflow. Different flying altitudes and speeds affect the size of airflow impact force, and have a big influence on head-neck injury. When the head impact the headrest, in the situation of H = 1000 m and V = 1000 km/h, the NIC value exceeds 46 m/s², it will result damage risk in 100% according to the criterion; Under the circumstance H = 8000 m and V = 1000 km/h, the NIC value is 22.8 m/s², the damage risk is 50%; for H = 8000 m, V = 1000 km/h, the neck damage risk is less than 50%.

4. Among these three different ejection situations, the NIC value is the largest when H = 1000 m and V = 1000 km/h, implying that pilots’ head-neck is most vulnerable to injury at low altitude and high airspeed, especially when head collides with headrest.

5. Head-neck can be fixed for protection during ejection. The head-neck injuries are more likely to occur when head impact headrest. Therefore, we can fix the head-neck in the forward bending position to reduce the probability of injury.

4. Conclusion

In this study, by virtue of LifeMod, we develop a head-neck model of human body, including rigid vertebrae, viscoelastic muscles and joints. Comparisons of the data between the model simulation and the frontal-rear collision tests have been carried out to validate the reliability of the model.

Based on the head-neck model of human body, a human-seat model is built on Adams platform. For three different ejection environment parameters, the computer simulates the head-neck motion response. The head first bends forward quickly, then flings back fleetly and collides with headrest. We calculate NIC values by simulating results. These values indicate two dangerous locations under the limit of head-neck flexion and the head colliding with the headrest, especially at low altitude and high airspeed. On the basis of these research, we conclude it is the essential to install head-neck protective devices on ejection seat.

References

[1] Chen D, Wang J, Wu W and Chen L 2006 Ejection seat test techniques in a high-speed wind tunnel J. Aircraft 43 1593-96.

[2] Teng Y Y 1992 Windblast injury and its protection of head-neck in emergency ejection J. Space Medicine & Medical Engineering 4 302-307.

[3] Chen X and Yuan X G 2000 Human-machine--environmental system of engineering physiology (Beijing University of Aeronautics and Astronautics Press) pp 30-59.
[4] Parr J C, Miller M E, Pellettiere J A and Erich R A 2013 Neck injury criteria formulation and injury risk curves for the ejection environment: a pilot study J. Aviat. Space Environ Med. 84 1240-48.

[5] Anton, D 1985 The clinical and biomedical evaluation of trauma and fatalities associated with aircrew ejection and crash. A Working Group Report. AD-152 350.

[6] Su B J 2014 Introduction to modern rocket ejection escape technology (Beijing: Aviation industry press) pp 8-178.

[7] Jr P P, Wipasuramonton P and Begeman P 1999 A three-dimensional finite element model of the human Arm Stapp Car Crash Conf. Pro. 688-90.

[8] Li X R and Long H J 2008 J. Space Medicine & Medical Engineering 2 187-191.

[9] Yao J F 2002 Establishment and verification of human neck finite element model (Master dissertation, College of Mechanical and Vehicle Engineering, Hunan University).

[10] Du D Y 2014 The human neck modeling and research of neck injuries in vehicle rear-end impact (Master dissertation, College of Automotive Engineering, Jilin University).

[11] Xiao Z 2007 Study on the protection of Passengers’ neck in automobile rear crash (Doctoral dissertation, College of Mechanical and Vehicle Engineering, Hunan University).

[12] Tien C S and Huston R L 1985 Biodynamic modelling of the head/neck system Publication of Society of Automotive 8 573-78.

[13] Tien C S and Huston R L 1987 Numerical advances in gross-motion simulations of head/neck dynamics J. Biomechanical Engineering 109 163.

[14] Liu Y Z, Pan Z K and Ge X S 2014 Dynamics of multibody systems (Higher Education Press vol 2) pp 50-68.

[15] Deng Y C and Goldsmith W 1987 Response of a human head/neck/upper-torso replica to dynamic loading--II. Analytical/numerical model J. Biomechanics 20 487-97.

[16] De Jager M, Sauren A, Thunnissen J and Wismans J 1994 A three-dimensional head-neck model J. Passenger case 103 1060-76.

[17] De Jager M, Sauren A, Thunnissen J and Wismans J 1996 A global and a detailed mathematical model for head-neck dynamics J. Passenger Cars 105 6.

[18] Van der Horst M J, Thunnissen J G M, Happee R and Wismans 1997 The influence of muscle activity on head-neck response during impact R.SAE Technical Paper 973346.

[19] Ma N and Xiao L Y 2011 Chinese Journal of Rehabilitation Medicine 26 538-542.

[20] Liu F, Liu G, Zeng Q R and Sun L2010 Machinery Design and Manufacture 4 255-257.

[21] The PLA general armaments department GJB 4856-2003 Human dimensions of Chinese male pilot population.

[22] Liu B S and Wang X W 2014 Aviation Anthropometry (Beijing Aerospace University Press) pp 20-48.

[23] White, Augustus A , White A A 1978 Clinical biomechanics of the spine M. Biomechanics in Orthopedics.

[24] Huelke D F and Nusholtz G S 1986 Cervical spine biomechanics: a review of the literature J. Orthopaedic Research 4 232.

[25] Ewing C. L, Thomas D. J, Lustick L, et al 1976 J. Society of Automotive Engineers 20 3-41.

[26] Ewing C L, Thomas D J, Beeler G W, et al. 1968 Dynamic Response of the Head and Neck of the Living Human to—G x Impact Acceleration R. SAE Technical Paper.

[27] Feng W C 2014 Aviation emergency rescue system (National Defense Industry Press) pp 6-36.

[28] Mastrolia, B 2012 U.S. Patent No. 8,191,830. Washington, DC: U.S. Patent and Trademark Office.

[29] Yu L 2015 Aircraft lifesaving and life support technology (National Defense Industry Press) pp 3-88.

[30] Boström O, Svensson M Y, Aldman B, et al. 1996 A new neck injury criterion candidate-based on injury findings in the cervical spinal ganglia after experimental neck extension trauma Proc. 1996 Int. Ircobi Conf. on Biomechanics of Impact (Ireland) pp 123-136.