Human Response Characteristics to Impact Conditions During Spacecraft Takeoff and Impact Landing

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Abstract
Human exposure to local accelerations during strategic spaceflight operations is required for assessing the potential health risks and performance. The objective of this study is to know the relation of human responses to the peak g and duration as well as orientations. This paper discusses an approach for human body modeling and simulation. Multibody dynamics approach is used to model the human body using LifeMOD®. The developed biomechanical models are analyzed for different acceleration pulses ($4g$, $6g$ and $10g$) and seat orientations (20^0, 30^0, and 40^0). A 4-DOF lumped parameter model is studied to compare with multibody model. Interrogation of results indicates the physical significance of the study. From the results obtained, we determine the responses of human body segments to various impact levels and seat orientations. The results obtained using multibody model is in good agreement with the results obtained using lumped parameter model.

1. INTRODUCTION

Human body can tolerate certain levels of impact forces during takeoff and landing approaches of spacecraft, but the high-level impact forces may cause disadvantageous influence on the human body, and even threaten the lives of crew during these spacecraft activities. The human acceleration responses at different segments vary with the level of impacts. Hence musculoskeletal response to the variation of g-force is to be studied to know the behavior of these segments to the specific peaks. Study of the musculoskeletal response helps to determine the responses of human body segments to various impact levels and seat orientations. Approaches such as takeoff and landing of manned spacecraft from and to the earth, involves the activities like ascending and descending followed by large amount of abrupt changes in the accelerations. This makes the crews inside the vehicle to be exposed to the impact forces. The study on the response of the human body to impact conditions has attracted many researchers in this field via experimental studies and mathematical modeling techniques.

Previous experimental studies have demonstrated that the human body can tolerate certain levels of landing impact forces. Stapp et al. [1,2] conducted experiments to know the tolerance limit of human to deceleration. In these experiments 58 human subjects were exposed to acceleration peaks of 10, 15, 20 and $25g$ for short duration from 60 to 130 ms. Results were summarized for 16 body orientations and concluded that all body positions and impact configurations were within voluntarily tolerance limits except the forward facing 45° reclining position at $25.4g$ measured for 60ms, in which compression of soft tissues caused pain and stiffness to thoracic vertebrae for 60 days. Weis et al. [3] conducted 75 experiments. In these experiments 20 different subjects exposed to seven different body orientations had six different impact configurations by means of a vertical drop tower decelerating with a water inertia-piston and cylinder brake. The drop velocity at brake entry ranged from 4.28 to 8.47 m/s, the peak ranged from 13.5 to $26.6g$ and the duration from 56 to 75ms. To study human responses to impact forces Brown [4] conducted 288 experiments on humans to study the effects of the impact forces to the biological systems like musculoskeletal, cardio-respiratory and nervous systems. In this he simulated 24 body orientations for the acceleration peaks from 5.5 to $30.7g$. Observed results indicated that man can endure certain impact forces in different body orientations without significant pain. In brief, the relation of human responses to the peak g, orientations and duration is known and summarized [5]. Zhuang.X [6] analyzed the human responses to impact forces with the help of mechanical models which clearly indicated the loading limit on the human. The vertical drop tower experiments on men, animals and dummies by Wang Yulan, Cheng Zilong et al. [7,8] indicates the human tolerance limitations to deceleration. Liu [9] conducted experiments to determine the hazardous effects of high-level impact forces which have a serious impact of threatening the lives of crew members during emergency landings. Observations indicated that high-level impact forces caused disadvantageous influence on the human body. Bingkun Liu [10] conducted 45 experiments in which 5 young male subjects were voluntarily exposed to the peak from 4 to 10*g* and duration from 50 to 80 ms acceleration pulses at 20° supine angle and the peak 10*g* and duration 50 ms impact at the supine angles from 20^0 to 60^0, The results concluded that acceleration responses showed different properties between chest-back direction and the head-foot direction. As these experimental tests are yet expensive and time consuming. During the past few years a strong increase has been observed in the use of computer simulations due to the fast developments in computer hardware and simulation software. Simulation models are now a days the easier and more economical way to check the effect of design changes even in the earlier stages of the development.

A great effort among the biomechanics researchers in past few years has been devoted to the development of reliable mathematical models of the musculoskeletal system. These models often comprise specific formulations from multibody system dynamics, muscle mechanics and descriptions of musculoskeletal geometry. Specifically, a well-developed human body model helps in understanding injury mechanisms of the bony skeleton and soft tissues/organs of the crew under complex loading conditions in laboratory and real impact testing. The human body models are developed based upon measured or estimated parameter values, representing characteristics of the human body. Although the various human body models differ in many aspects, all are dynamic. The models account for inertial effects by deriving equations of motions for all movable parts, and solving these equations using an iterative method. The mathematical formulations used for these models can be subdivided into: (1) Lumped mass models: The lumped parameter models consider an appropriate mathematical modeling of human body using several rigid bodies, spring and dampers [11,12]. This type of model is simple to analyze and easy to validate with experiments. However, the disadvantage in the limited number of degrees of freedom; (2) Finite element models: In a comprehensive approach of modeling such as Finite element method, detailed information must be available and is quite rough at an early stage. Another disadvantage of this kind of modeling is the great amount of time involved in preparing the model and the computer time required for simulation [13,14]. When many design alternatives have to be investigated a fast simulation model is desired; (3) Multibody models: This type of model is efficient since the motion restrictions between different anatomical segments of model defined as complex kinematic joints, suitable to represent mechanical joints, or as contact/sliding pairs, used to describe realistic human like anatomical joints [15]. Shawn P McGuan [16] developed an active human model to study the multi-axis vibration environment for an off-road heavy vehicle. This study represents an on-going effort into the study of active human response to vibration. Tay Shih Kwang [17] developed a design system that can simulate the kinematic behavior of musculoskeletal forms and generate a human-wheelchair interface. The developed system was accurate and provided effective seating solutions for wheelchair users preventing long-term spinal deformities. Grujicic [18] constructed a rigid-body model of a prototypical adjustable car seat and combined with a public-domain musculoskeletal model of a seated human. The Seated-human/car-seat interactions associated with typical seating postures of the vehicle driver were analyzed. The results summarized indicate various seat adjustments and driver's back supports which had a complex influence on the muscle activation and joint forces.

From the literature is found that not much work has been done in the area of modeling and simulation of human body during spacecraft liftoff and landing. In this study we have used multibody dynamic approach, which helps in understanding the responses of the human body segments for different impact conditions during spacecraft takeoff and landing approaches. Simulation of the biomechanical human body model is the key in this study. The human body is modeled using LifeMOD® software. Since this has the capability to build the musculoskeletal model of varying range of anthropomorphic data of human system. The seat is modeled using MSC ADAMS® software. In multibody dynamics approach the human body parts are connected by tree, loop or chain topologies

[19]. The contact forces are created on the model to provide an interaction between the human model and the seat. This biomechanical model is analyzed for two configurations: (1) Musculoskeletal model for different acceleration peaks at a specified supine angle; (2) Musculoskeletal model for different supine angels at a specified acceleration peak. Through these studies we determine the responses of human body segments to various impact levels and seat orientations. In order to compare the results obtained through the multibody dynamics approach, a lumped parameter model approach is followed to study the responses of the human seated system.

2. METHODOLOGY

The two methods used for the study are multibody dynamics approach and lumped parameter approach.

2.1. Biomechanical modeling and simulation using multibody dynamic approach

Multibody system dynamics is an essential part of computational dynamics - a topic more generally dealing with kinematics and dynamics of rigid and flexible systems, finite elements methods and numerical methods for synthesis [19]. Multibody system dynamics is based on analytical mechanics, and is applied to a wide variety of engineering systems as well as to biomechanical problems. In this method the study of acceleration responses to spacecraft impact conditions and specific body orientations involves the phases as discussed below.

2.1.1. Musculoskeletal Human modeling

This phase includes the development of the biomechanical model. The human body is modeled using the LifeMOD® and the seat by ADAMS(R) environment. The steps in modeling phase involve: (1) Generation of body segments: The first step in the human model creation process is to create the base level segment set. The general properties and dimensions of the 19 segments are created using data from the anthropometric databases. The base level segment set includes: head, neck, upper torso, central torso, lower torso (ilium), right scapular, right upper arm, right lower arm, right hand, left scapular, left upper arm, left lower arm, left hand, right upper leg, right lower leg, right foot, left upper leg, left lower leg, left foot; (2) Creating the joints: The human segments created in the first step are connected together with kinematic joints. The joint consists of a tri-axis hinge and passive or active forces acting on each of the three degrees of freedom. The human body is a highly nonlinear system, hence we use Hybrid III joint [20]. The joint torques generated using the Hybrid III crash dummy model are based on stiffness, damping and friction data measured at the Armstrong Aerospace Medical Research Laboratory, Wright Patterson Air Force Base from the mechanical Hybrid III crash dummy. The Hybrid III strength model is created for the individual joint axis with a user specified scale value. The Hybrid III strength model is based on physical measurements of an actual crash dummy. The strength model consists of nonlinear stiffness, damping and frictional values and also include joint limit stop stiffness with hysteresis; (3) Posing the human model: After creating the segments and placing the joints on the model, we can set the starting posture for the model. We can set angular values for any joint degree-of freedom that we wish to change; (4) Creating the interior seat model: standard ADAMS/View tools are used to create the interior seat model. Simple geometry is used to represent the seat and floor of the compartment. A translational joint is created between this seat and the floor and will be used to provide the acceleration pulse; (5) Posing the Human model to seat model: The posed human model is now moved to the seat model to build the correlation between the two models; (6) Creating the environment contact forces: Contact forces are created on the model to provide an interaction between the human model and the environment. LifeMOD® contact forces utilize a novel contact algorithm for efficient calculation of the reaction of the body segments to impact with the environment.

2.1.2. Analysis to different conditions

Once the modeling of human seat system is complete, the simulation is run for two different conditions: (1) As shown Figure 1, the supine angle is kept constant and the acceleration peaks is changed for a range of *g*-values for a short duration of time; (2) As shown in Figure 2, the acceleration peak *g*-value is kept constant and the supine angle (θ) is varied for range of values and the responses for these two different conditions are analyzed using MSC ADAMS® simulation software.

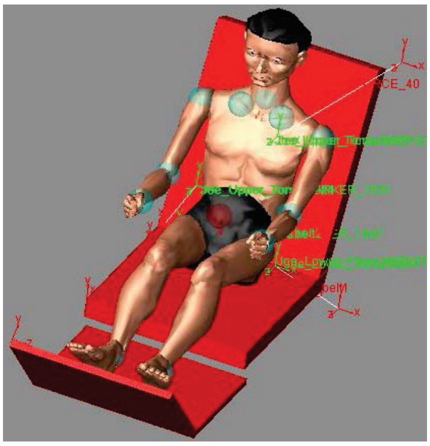

Figure 1. Biomechanical model for varying Acceleration peaks

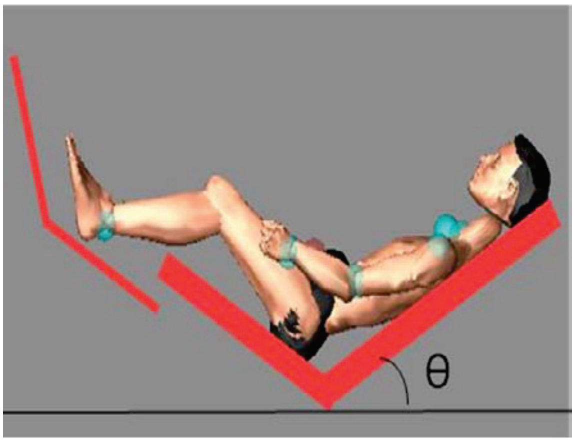

Figure 2. Biomechanical model for varying Supine angle (θ)

2.1.3. Presentation of results

Interrogation of the results is an important step, which furnish guidelines in order to make suitable decisions. With the simulations complete, the results may be reviewed in many ways. One of the best ways to understand model performance is to plot the data and view the animation simultaneously. The results are monitored and responses are assessed on different body segments. Analysis of the biomechanical models for two different conditions determines the responses of human body segments to various impact levels and seat orientations.

2.2. Biomechanical modeling and Simulation using lumped parameter model

The lumped-parameter model is probably one of the most popular analytical methods in the study of biodynamic responses of seated human subjects. The human body in a sitting posture is modelled as a mechanical system that is composed of several rigid bodies interconnected by springs and dampers. In this paper 4-DOF Boileau and Rakheja model [12] is considered for study.

This model consists of four mass segments interconnected by four sets of springs and dampers with a total mass of 55.2 kg. The four masses represent the following four body segments: the head and neck (m_1), the chest and upper torso (m_2), the lower torso (m_3), and the thighs and pelvis in contact with the seat (m_4). The mass due to lower legs and the feet is not included in this representation, assuming their negligible contributions to the biodynamic response of the seated body. The stiffness and damping properties of thighs and pelvis are (k_5) and (c_5), the lower torso are (k_4) and (c_4), upper torso are (k_2, k_3) and (c_2, c_3), and head are (k_1) and (c_1). $x_1, \dot{x}_1, \ddot{x}_1$ are the displacement, velocity and acceleration of head and $x_2, \dot{x}_2, \ddot{x}_2$ for upper torso, $x_3, \dot{x}_3, \ddot{x}_3$ for lower torso, $x_4, \dot{x}_4, \ddot{x}_4$ for thighs, $x_{se}, \dot{x}_{se}$ for seat. The schematic representation of the model is shown in Figure 3 and the biomechanical parameters of the lumped parameter model are listed in Table 1.

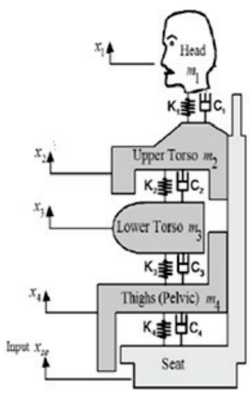

Figure 3. Boileau and Rakheja 4-DOF model

Table 1. Parameters for Boileau and Rakheja 4-DOF model

Mass (Kg)	Damping (N.s/m)	Stiffness (N/m)
$m_1 = 5.31$	$c_1 = 460$	$k_1 = 356370$
$m_2 = 28.49$	$c_2 = 5400$	$k_2 = 208570$
$m_3 = 8.62$	$c_3 = 5190$	$k_3 = 187110$
$m_4 = 12.78$	$c_4 = 2370$	$k_4 = 103480$

The equation of motion for the model is as follows

$$m_1 \ddot{x}_1 = -c_1\left(\dot{x}_1 - \dot{x}_2\right) - k_1\left(x_1 - x_2\right)$$

$$m_2 \ddot{x}_2 = c_1\left(\dot{x}_1 - \dot{x}_2\right) + k_1\left(x_1 - x_2\right) - c_2\left(\dot{x}_2 - \dot{x}_3\right) - k_2\left(x_2 - x_3\right)$$

$$m_3 \ddot{x}_3 = c_2\left(\dot{x}_2 - \dot{x}_3\right) + k_2\left(x_2 - x_3\right) - c_3\left(\dot{x}_3 - \dot{x}_4\right) - k_3\left(x_3 - x_4\right)$$

$$m_4 \ddot{x}_4 = c_3\left(\dot{x}_3 - \dot{x}_4\right) + k_1\left(x_3 - x_4\right) - c_4\left(\dot{x}_4 - \dot{x}_{se}\right) - k_4\left(x_4 - x_{se}\right)$$

3. RESULTS AND DISCUSSION

In this section, we discuss the Simulation results obtained using Multibody model and Lumped parameter model. The dynamic responses of the human body segments are obtained by simulating the biomechanical model for different impact conditions and seat orientations.

3.1. Dynamic responses at different impact levels

(1) Multibody model: In this approach the multibody model is modeled using LifeMOD®. The typical acceleration-time curve measured at the head, shoulder, chest (upper torso) and Ilium (lower torso) for an excitation of 4*g* is shown in Figure 4. The human seated model is subjected to range of acceleration pulses. It shows that the human dynamic responses increase with the rise of impact level, but the increase rates are not the same at different parts of the human body. In order to observe the changes in these body segments responses with impact levels, biomechanical model is subjected to 4*g*, 6*g* and 10*g* excitations and supine angle is kept constant at 90^0. The results tabulated in Table 2 show that the peaks of the human body segments increase with the rise in the impact level.

Table 2. Peak acceleration responses at different *g* levels of impact

Parts	Acceleration pulse (*g*)	Response (*g*)
Head	4	3.75
	6	5.84
	10	10.20
Ilium	4	4.10
	6	6.12
	10	10.54

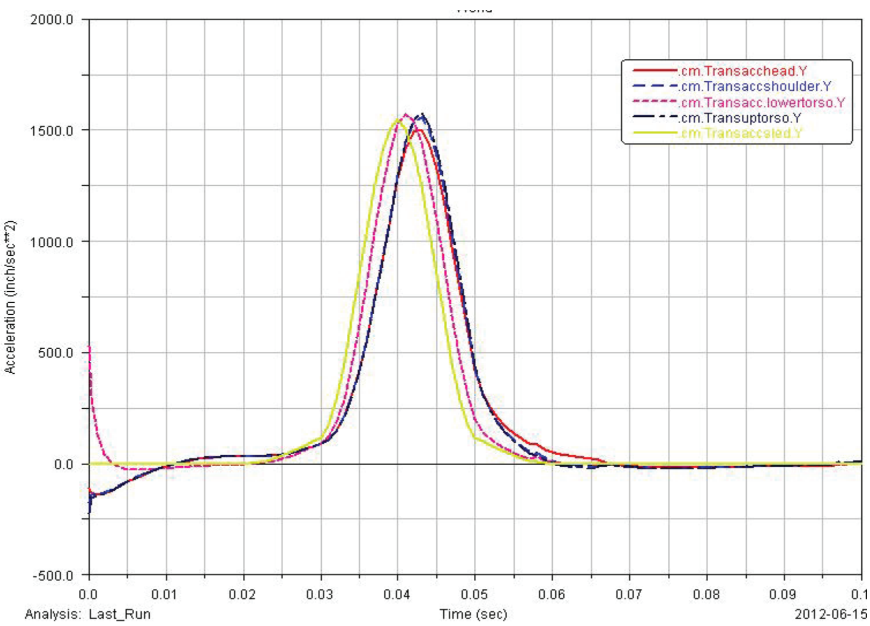

Figure 4: Acceleration response for Impact magnitude of 4*g*.

(2) Lumped Parameter approach: In this approach the 4-DOF Boileau and Rakheja model is modeled using MATLAB/SIMULINK. This model is run for impact levels of 4*g*, 6*g* and 10*g* acceleration pulse, the responses obtained for different body segments are tabulated in Table 3. By observing the values of Table 2 and Table 3, we can infer that the simulation results obtained using LifeMOD$^{(R)}$ software and MATLAB/SIMULINK show a good agreement.

Table 3. Peak acceleration responses in different body Positions

Parts	Acceleration pulse (*g*)	Response (*g*)
Head	4	3.75
	6	5.6
	10	11.95
Ilium	4	3.6
	6	5.4
	10	10.7

3.2. Dynamic responses in different seat orientations

In order to observe the changes in responses of body segments with different supine angles, a biomechanical model shown in Figure 2 is subjected to 10*g* acceleration pulse for different supine angles (θ) ranging from 20^0 to 40^0 is simulated. The results of simulation show the response for different body segments, the values of only the head and ilium are shown in Table 4. It shows that human acceleration responses at different segments vary with the supine angle, and the variation is not the same for different parts of the human body. At a supine angle of 30^0 the segments shows a minimum response. The typical acceleration-time curve for a value of 4*g* is shown in Figure 5.

Table 4. Peak acceleration responses in different body Positions

Parts	Acceleration pulse (g)	Supine angle ϴ=20^0	Supine angle ϴ=30^0	Supine angle ϴ=40^0
Head	10	10.81	10.74	11.12
Ilium	10	10.42	9.79	10.36

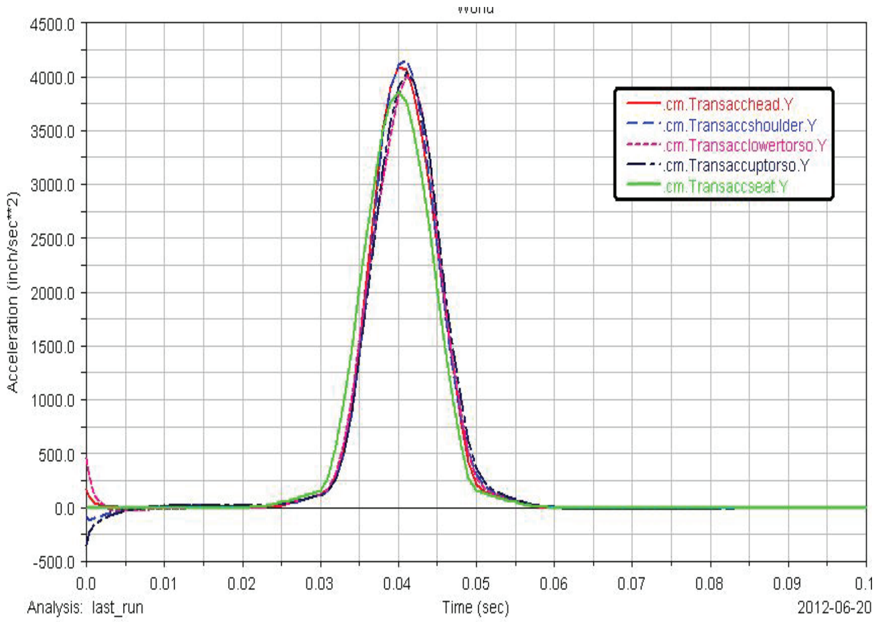

Figure 5: Acceleration responses for a Supine angle of 20^0.

4. CONCLUSIONS

Human body when exposed to various impact levels show dynamic responses which are different at different body segments. There are dynamic overshoots at the body segments, which reflect the viscoelastic behaviour of the human body. According to the results of the simulation, human dynamic responses increase with the rise of impact level, but the increase rates are not the same at different parts of the human body. The reason may be related to the body segment's mechanical properties. For now the simulation performed show a process of modeling using "state of the art" software, as well as opens the possibilities to perform several major task changes and body segments refinements in order to fit more accurate requirements. Analysis performed using multibody dynamics approach and lumped parameter model show a good agreement between the results obtained. In general the region of head experiences more g-level compared to ilium. However this trend changes when the supine angle is varied.

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