SIMULATION OF POTENTIAL INJURIES DUE TO FALL AT THE AIRCRAFT ENTRY DOOR

A Thesis by

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Submitted to Department of Mechanical Engineering
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Wichita State University
in partial fulfillment of
the requirements for the degree of
Master of Science

May 2007

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I have examined the final copy of this thesis for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Master of Science, with a major in Mechanical Engineering.

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We have read this thesis and recommend its acceptance.

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Dr. Kurt A. Soschinske, Committee Member

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Dr. Krishan K. Krishanan, Committee Member

DEDICATION

To my children, my family and my friends

ACKNOWLEDGEMENTS

I would like to express my sincere gratitude to my graduate advisor, Dr. Hamid M. Lankarani, who has been instrumental in guiding me towards the successful completion of this thesis. I would also like to thank Dr. Kurt Soschinske and Dr. Krishan Kirshanan for reviewing my thesis and making valuable suggestions. I am thankful to the National Institute for Aviation Research (NIAR) computational lab and Dr. Geraldo Olivares for supporting my research. I would like to acknowledge the support of my colleagues at my work, especially my manager, in the completion of my studies. I also extend my sincere thanks to my family and all my friends who stood by me at all times.

ABSTRACT

Falls are one of the most common causes of injury among young children and have been recognized as a major cause of death and disability worldwide among children 1-3 years old. A characteristic of injuries among very young children is that aspects of their normal behavior, such as natural curiosity or physiologic development of their motor skills, could be associated with an increased injury risk, especially in non-friendly and new environments.

It has been reported that there have been incidents when children fell from the entry stairs while boarding an airplane. The small size and unpredictable nature of children combined with the size of the gap in the handrails creates risk for children to fall. Falls have long been studied in relation to nursery equipment and playground equipment from heights less than 1-2 meters, but not where the ground is asphalt or concrete. The present study aims to assess fall-related injuries among children on a concrete the airport tarmac.

Since little is known regarding the biomechanics of such falls and injury risk associated with them, computer simulation provides a valuable tool to investigate and predict injury outcomes. The validity of the model is crucial to the reliability of the outcome. In this study, a computer simulation of a child falling from a step of the stair surface onto a hard surface was analyzed for head accelerations using the MADYMO Hybrid III-3 Year Old Child dummy model. There was no crash pulse applied, gravity was used as the fall force.

Automotive and aerospace companies perform tests and computer simulations in order to optimize and design safety devices in their vehicles. This study investigates the fall injury related parameters for the head and neck, as well as the influence of contact friction forces between the dummy and surface, the kinematics of the fall, and on the head and neck acceleration forces. This study may result in an increased study of passenger safety.

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CHAPTER 1

INTRODUCTION

1.1 Background

Fall related deaths are the 6th leading cause of death among the 1-6 year old children, which represents about 2.5% of the total deaths in that age range for the year 2004 [1]. Injuries have been recognized as a major cause of death and disability worldwide and as a factor responsible for a substantial number of deaths among children of all ages 1-3 years old. A characteristic of injuries among very young children is that aspects of their normal behavior, such as the natural curiosity or physiologic development of motor skills, could be associated with an increased injury risk, especially in a non-friendly or new environment. Also, parents or guardians often show insufficient consideration for the risks linked to the surrounding environment. It has been reported that there have been a few incidents when children accompanied by parents/guardians fell from the entry stairs while boarding the airplane.

The type of entry stairs in question are called air-stairs, which are built into the airplane. They are typically used at small airports where boarding tunnels or ground stairs are not available. Airplanes designed with weight in mind typically have the handrails for stairs, that have gaps, near the upper stair near the door for commercial jets, or there are the no hand rails for small airplanes. These type of stairs are generally built into the entry doors of small commuter airplanes. However larger commercial aircraft such as some Business Jets, Airbus A320 and Boeing 737 have these stairs built into the airplane underneath the entry doors. These airstair can be electromechanically extended or retracted by either from inside or outside the airplane. The stairs have sets of folding handrails which are extended manually into the entry doorway for passenger's safety. Fig 1.1 shows such stairs on a commercial jet.

Fig 1.1 Air stairs on a commercial jet, in opening (left) and in opened position (right) [2].

The small size and unpredictable nature of children, combined with the size of the gap in the handrails, creates risk for a child to fall due to an inattentive parent or guardian. Falls are one of the most common causes of injuries among children, and they have long been studied in relation to nursery and playground equipment from heights under a meter. The dummy to ground interaction was studied where the typical ground was a playground. The present study aimed to assess the incidence of fall-related injuries among children on an airport tarmac which is a hard surface such as asphalt or concrete.

Computer simulation provides a valuable tool to investigate falls and predict injury outcomes. With the use of computer simulation, a single characteristic of the fall environment can be altered to investigate its influence on injury potential and outcomes. The validity of the model is crucial to the reliability of the outcomes measured by the simulation. Once an experimentally validated simulation model has been developed, savings can be realized through

reduction in the number of costly experimental tests needed to investigate slight permutations. In this study, a computer simulation of a child falling from a horizontal step of the stair surface onto hard surface such as concrete or asphalt floor was analyzed for head accelerations. For this purpose, computer crash simulations with MADYMO models were performed. The MADYMO Hybrid III-3 Year Old Child dummy model was used. There was no crash pulse applied, instead gravity was used as the fall force.

Automotive and aerospace companies perform tests and computer simulations in order to optimize and design safety devices in their vehicles, including seats for rear impact protection. However, this type of analysis is generally not done by companies due to the low incidence of occurrence. Generally, these tests are performed using only one dummy size which is considered to be more representative. This study involves:

- Investigation of the influence of crash dummy Hybrid III-3 year Old size on the fall injury related parameters.
- Investigation of the influence of contact friction forces during the fall. This is due to the very clear influence of the coefficient of friction between the dummy and surface of fall.

The term 'head injury' encompasses all injuries to the cranial, cervical, and cerebral structures of the head and brain. Head injuries can be the result of falls, violence, or firearms. Many questions surround the functioning of healthy brains, while many more surround the functioning of injured brains. Thus, it is not surprising that head injury ranks among the leading causes of trauma related deaths in the United States [1].

The topic of head injury in relation to vehicle accidents has long been an area of research for biomechanical engineers, medical professionals, crash re-constructionist's, vehicle

manufacturers, and federal regulatory agencies. Over the past few decades this issue has gradually grown in the national consciousness as well. Head injury trends have spurred advancements in head protection such as seatbelt requirements, safer steering wheels, padded dashboards, and most recently, airbags.

This study looked at that the injury can occur to a small child if he/she can manage to slip under the guard rails. This results in an increased understanding of passenger safety.

1.2 Objective

The objective of this research is to analyze the injuries to a 3 year old child's head and neck due to a simulated fall from the top of the stairs.

1.3 Method of Approach

The method of approach for this study is done as following:

- The method utilized is to place the child in a most likely fall scenario with the interfacing structure.
- Apply the type of fall, since there could be an accidental slip, push, or step off. The method that will be looked at is the accidental step off the stair step and to analyze the behavior of the dummy in the most likely fall scenario.
- Aircraft structure and handrails would not play a role in the dummy interaction. The analysis is strictly between the top step and the ground.
- Obtain crash forces at critical joints, measure HIC (Head Injury Criteria), and measure for pass/fail criteria.
- Present results for the head and neck.

It would be reasonable to assume that a 3 year old child, though accompanied by a parent or guardian, is not physically restrained at the top of the step. Due to the location of the hand

rails, if the child's head is angled, it can easily slip under and between the hand rails and airplane structure in the door entering position. Therefore, the most likely scenario then becomes that the child is looking away with its head and body at a slight angle, steps too far to the side, and slides under the hand rail and falls. Gravity is the only force that will be applied to the dummy during the fall.

When investigating a fall or trip, a method of analysis is to correlate the physical evidence with witness or victim accounts. Since there is no physical evidence to start with, only the reasonable assumptions will be used. MADYMO allows us to scientifically analyze fall scenarios and illustrate the results. From a simulation, one can match potential injuries with areas of high loading. Since it is a computer model, the initial fall characteristics can be varied parametrically to compare various fall kinematics. Several fall scenarios will be illustrated. A brief description of scenario will be followed by what issues were examined using MADYMO simulations. This study was carried out using the MADYMO simulation software tool.

CHAPTER 2

MADYMO CRASH SIMULATION CODE

2.1 Introduction to MADYMO

MADYMO (Mathematical Dynamic Modeling) [10]is a user-friendly computer package, which is used to simulate crash situations to a high degree of accuracy and assess injuries sustained by victims. Though originally intended for studying occupant behavior during car crashes, the MADYMO program is sufficiently flexible for analyzing other means of transport. It also allows assessments to be made of the suitability of various restraint systems, including seatbelts and airbags. MADYMO combines in one simulation program the capabilities offered by multibody and finite element techniques.

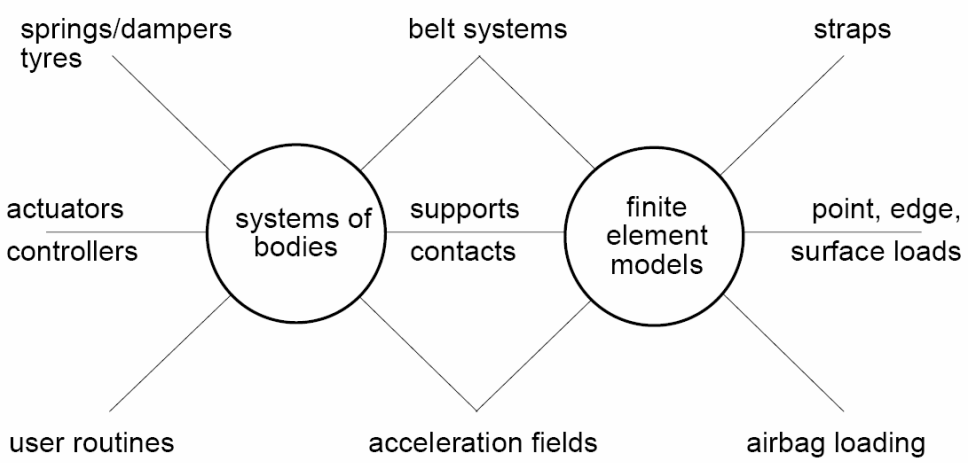

Figure 2.1 MADYMO 3D structures [10].

Multibody elements in MADYMO can be separated into three groups. They are the inertial system, the multibody system, and the null system. The elements used to model the seat, floor and other immovable objects belong to the inertial system. The elements that are affected by acceleration fields and contact forces are included in the multibody system. The

Anthropomorphic Test Device (ATD) is an example of this type of system. A typical crash model consists of an ATD as a multibody system placed in a seat that forms a part of the inertial space. Multibody systems are usually connected by joints. There are essentially two types of joints- kinematic and dynamic joints.

2.1.1 Multibody Systems Modeling

A multibody system is a system of bodies. Any pair of bodies of the same system can be interconnected by a kinematic or a dynamic joint. However bodies belonging to different systems cannot be connected by joints. A kinematic joint restricts the relative motion of the two bodies it connects. In MADYMO 3D, there are 12 types of joints such as spherical joint, translational joint, revolute joint, cylindrical joint, planar joint and universal joint. The way a specific type of kinematic joint restrains the relative motion of two bodies is characteristic for that type of joint. The relative motion allowed by a joint is described by quantities called joint degrees of freedom. The mathematical relationships and Jacobians for each type of joint are described by Nikravesh[4]. The different types of joints and multibodies are seen in Figure 2.2 and Figure 2.3

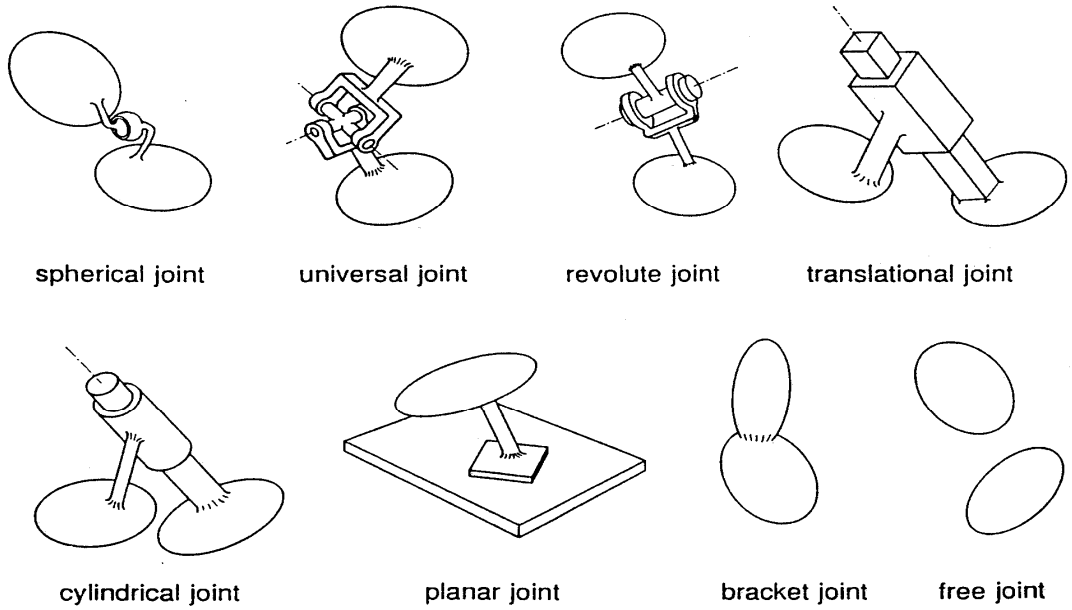

Figure 2.2 Kinematic joints available in MADYMO

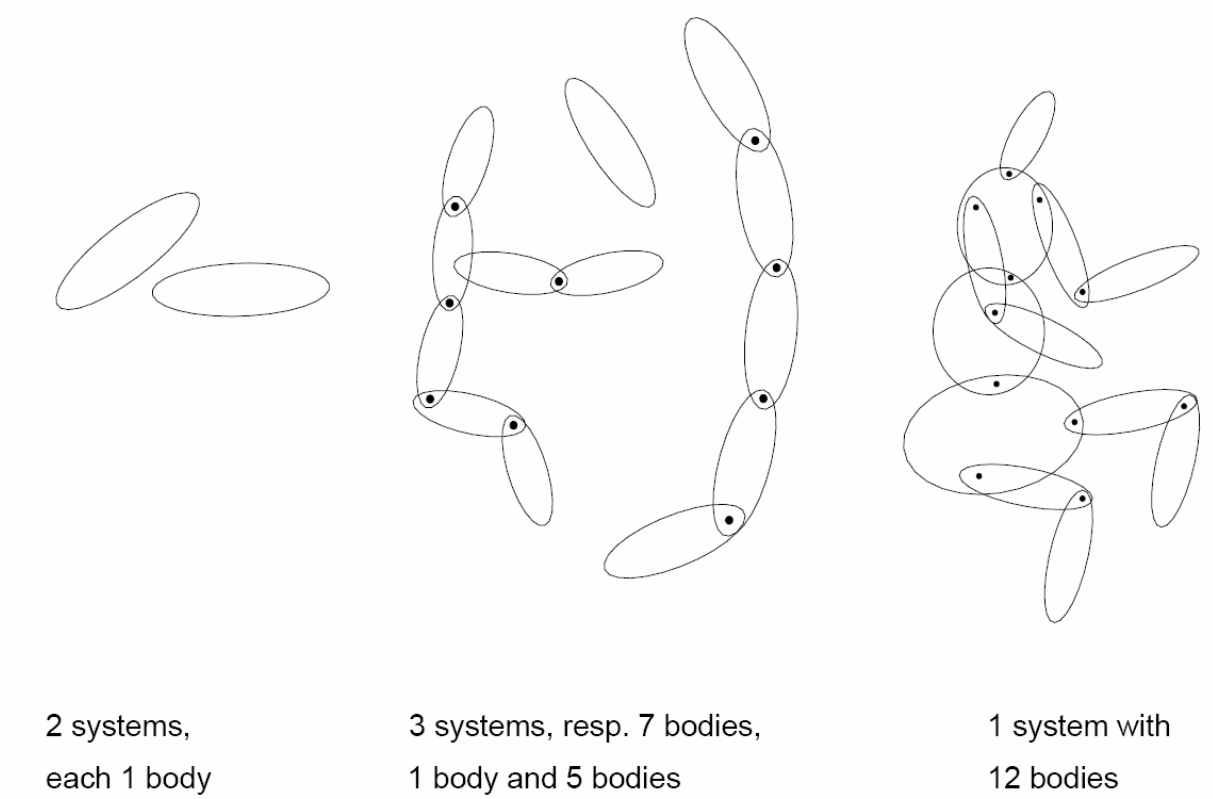

Figure 2.3 Examples of single and multi-body systems with tree structure.

For every type of kinematic joint there corresponds a dynamic joint model. It is a force model that defines the elastic, damping and friction loads that depend on the relative motion in the kinematic joint. For most types of joints, an elastic, damping and friction load can be specified for every joint degree of freedom. The load is either a force or a torque depending on whether the joint degree of freedom is a translation or a rotation, respectively.

2.1.2 Force Interaction

MADYMO offers a set of standard force accelerations and contacts of bodies with each other or their surroundings. These are discussed in the following subsections. Figure 2.3 gives a clear and concise idea of this concept.

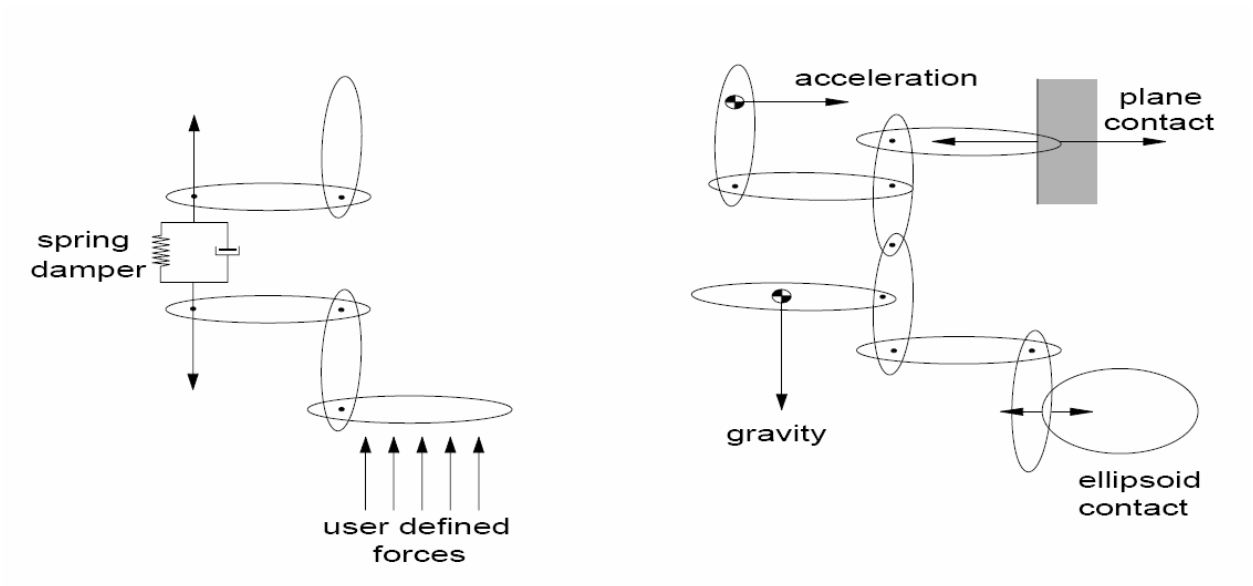

Figure 2.4 Examples of systems of bodies with mass interactions [10].

2.1.3 Acceleration Field

The acceleration field model calculates the forces at the center of gravity of bodies due to a homogeneous acceleration field. An acceleration field does not need to be defined for all bodies or all systems. Usually in an ATD (Anthropomorphic Test Devices) simulation, acceleration is usually imparted to the inertial system and its effects on various multibody systems are studied. Null system motion is used to model pre-simulation motion like the FEM (Finite Element Model) belt fastening process. An example of an acceleration field is shown in Figure 2.4.

2.1.4 Spring Damper

The following spring damper models, Kelvin element and Maxwell element, are used to calculate forces between two bodies within a system. The following are the two main types of spring damper models.

Kelvin Element- This model calculates the forces produced by a spring parallel with a damper. A Kelvin element is a mass-less, uni-axial element without bending or torsion stiffness. The two ends of the element can be attached to arbitrary points of any two bodies. These bodies may be in the same system or in different systems.

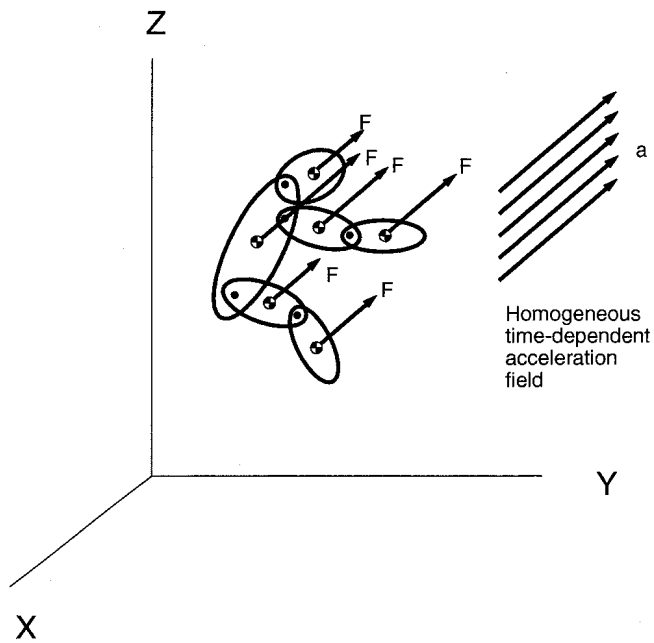

Figure 2.5 A system of bodies in a uniform acceleration field [10].

Maxwell Element- This force model calculates the forces produced by a spring in series with a damper. A Maxwell element is a mass-less, uniaxial element without bending or torsion stiffness. The two ends of the element can be attached to arbitrary points of any two bodies. These bodies may be in the same system or in different systems.

2.1.5 Contact Interaction

In MADYMO, it is possible to model the contact between an ellipsoid and a plane or between two ellipsoids. The resulting contact force is a point force, which consists of elastic, damping and friction components. The elastic force is a user-defined function of the penetration

of surfaces. It depends on the penetration and the force penetration characteristics. A positive value of the force corresponds to a resistive contact force. The damping and friction force depend on the relative velocity of the contacting surfaces. The relative velocity- v between the interacting contact surfaces is defined as the relative velocity at the point P of the two contacting objects. The damping force depends on the normal component of the relative velocity while the frictional force depends on the planar component of the relative velocity. This is seen in Figure 2.6

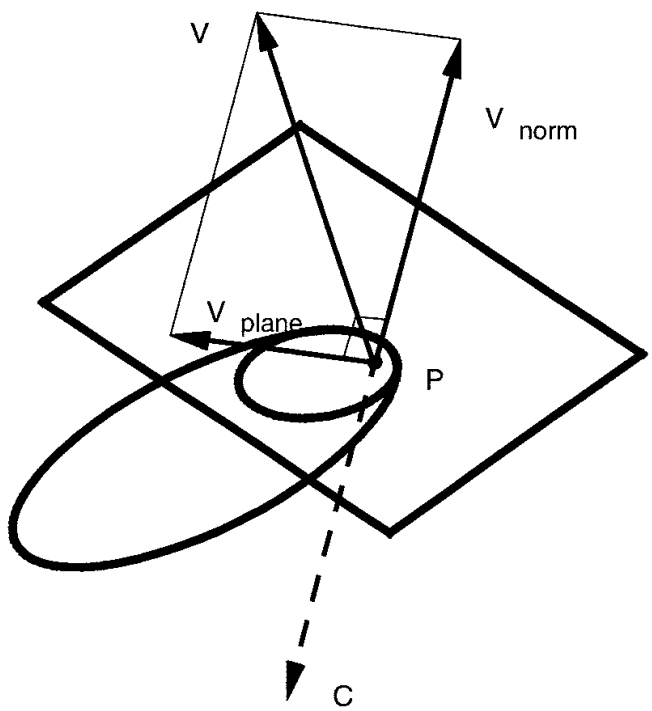

Figure 2.6 the relative velocity v resolved into two components

2.1.6 Inertial Space and Null Systems

Inertial Space: A coordinate system is connected to the inertial space. The origin and the orientation of the coordinate system can be selected arbitrarily. The motion of all systems is

described relative to this coordinate system. Contact surfaces such as planes and ellipsoids, restraint systems and spring damper elements can be attached to the inertial space. Acceleration is applied to the inertial system. The fashion in which this system is set up in an occupant simulation is made clearer in Figure 2.7

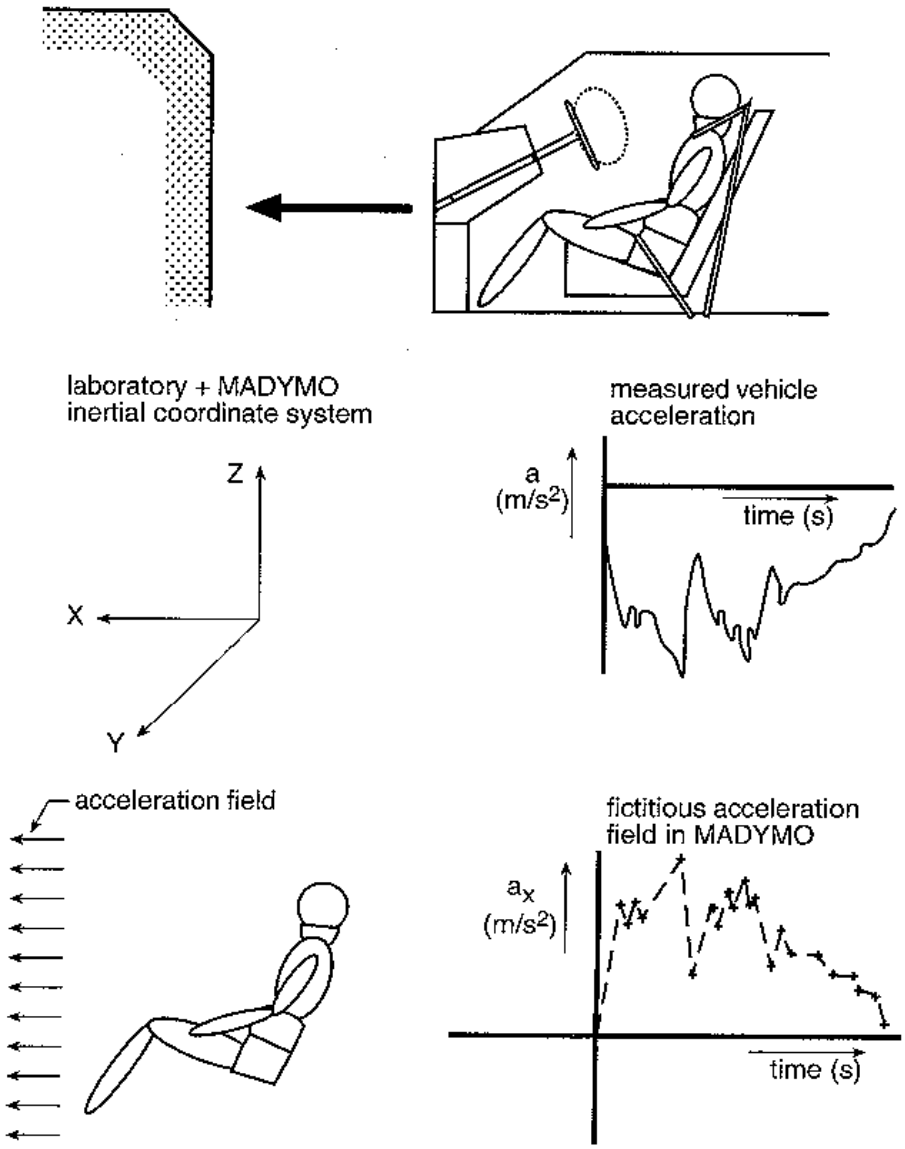

Figure 2.7 Simulation of vehicle deceleration by means of an acceleration field [10].

2.2 MADYMO ATD Databases

Biodynamic simulations are done using well-validated Anthropometric Test Device (ATD) databases. Databases of two-dimensional and three-dimensional ATD models are available in MADYMO. A wide range of TNO ATD's is available. For the purpose of this research, a HYBRID III-3 year old child dummy is used. It consists of 28 bodies, 16 of which are summarized in Table 2.1 below.

Table 2.1 Body parts of a Hybrid III-3 Year old child dummy

Number	**Name**
1	Lower Torso
2	Abdomen
3	Upper Torso
4	Neck
5	Clavicle
6	Lumbar
7	Head
8	Left Upper Arm
9	Right Upper Arm
10	Left Lower Arm
11	Right Lower Arm
12	Femur Left
13	Femur Right
14	Tibia Left
15	Tibia Right
16	Sternum

The geometric data and the inertia properties are based on results from measurements conducted at TNO and at the Wright Patterson Air Force Base in Ohio. The reference joint is chosen in the lower torso. Joint characteristics for the hip, knee, shoulder and head atlas block

attachment are based on measurements of the joint free range of motion. In addition, friction stops are specified for these joints in accordance with the calibration prescribed in ECE regulation 44. The elastic joint characteristics for both neck joints and both the lumbar spine joints have been obtained from static bending measurements. A typical Hybrid III-3 year old dummy is shown below in Figure 2.7.

Table 2.2 Body parts with weights of a Hybrid III 3 Year old child dummy [9]

Weights	Pounds (lbs)	Kilograms
Head	6.00	2.72
Neck	1.74	.79
Upper Torso w/Jacket	15.43	7.00
Upper Arm	.97	.44
Lower Arm w/Hand	1.01	.46
Upper Leg	2.23	1.01
Lower Leg	1.34	.61
Foot	.68	.31
Total Weight	35.65	16.17

Table 2.3 Body parts with dimensions of a Hybrid III-3 Year old child dummy [9]

Dimension	Inches	Centimeters
Head Circumference	20.00	50.80
Head Breath	5.35	13.59
Head Depth	6.89	17.50
Shoulder to Elbow	7.60	19.30
Buttock to Knee length	11.51	29.24
Knee Pivot Height	9.81	24.92
Sitting Height	21.50	54.61
Standing Height	37.2	94.49

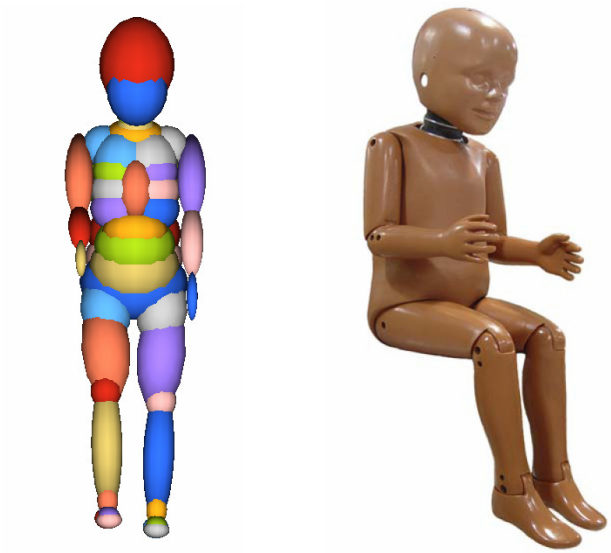

Figure 2.8 Isometric view of a Hybrid III-3 year old child dummy with ellipsoid model (left) and an actual dummy (right).

2.2.1 MADYMO Hybrid III-3 year old Child Dummy Description

The Hybrid III-3 year old dummy is used to evaluate aggressiveness of deploying side and frontal airbags to an out-of-position child of 3 years old. The dummy has recently been included in FMVSS 208 and is the recommended dummy to represent a 3 year old child in the ISO "Out-of-Position" test procedures. The ellipsoid model is based on the standard Hybrid III 50th percentile model. The differences with respect to the standard Hybrid III 50th percentile ellipsoid model are described below. For a description of those parts equivalent to the standard Hybrid III 50th refer to the Hybrid III 50th literature. The finite element model is based on the Hybrid III 3YO ellipsoid model. The differences are described below.

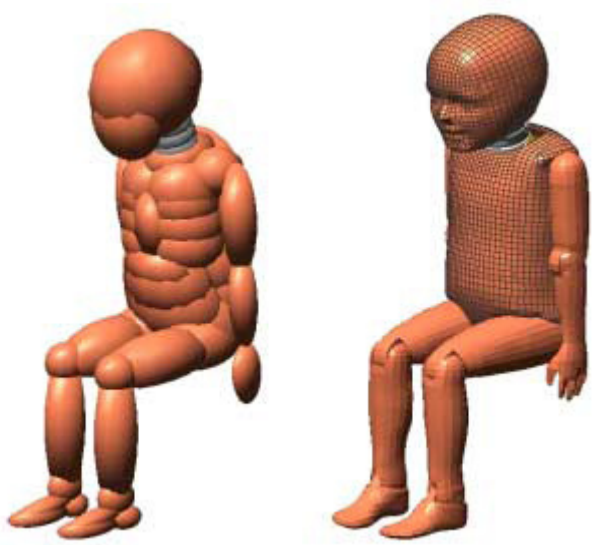

Figure 2.9 Hybrid III-3 year old dummy models; ellipsoid (left) and finite element (right).

Ellipsoid Model

Almost all parts of the Hybrid III 3-year-old dummy are scaled from the ellipsoid model of the Hybrid III 50th percentile dummy. The only exceptions are the thorax/ribs and sternum as described below. The neck and the lumbar spine are similar to the Hybrid III-6 year old dummy.

Thorax/Ribs and Sternum

For the 3YO, a protected joint resistance model is used to model rib displacement. The sternum stop, which limits the sternum to spine box displacement, is modeled by two separate point-restraints.

Finite Element Model

The finite element model has the same multi-body basis as the ellipsoid model. This paragraph describes the differences in the finite element model with respect to the ellipsoid model. The neck column and the lumbar spine are modeled the same as in the ellipsoid model. The arms and the legs have been modeled with facet surfaces.

Head

The skull of the head has been modeled as a single body and attached to it is a solid finite element model of the head skin and the chin insert. The chin insert has been modeled as a compliant part. Contact is defined between the chin insert and the head skin.

Thorax

The multi-body structure of the finite element model is almost the same as the ellipsoid model except that the potentiometer branch has been added. The ribs are modeled with shell elements with one layer of solid elements at the inner side representing the damping material. At the front the ribs are connected to the bib shell elements and the bib is connected to the sternum. The sternum is modeled with solid elements. The spine box is modeled as a rigid body and represented by facet surfaces. The potentiometer components are modeled in the same way as the spine box. The sternum and rib stops are modeled with solid elements. A small facet plane was added to the sternum to improve the contact with the sternum stops.

Abdomen

The abdomen has been modeled using tetrahedron elements and is not supported to a rigid body. Instead, the abdomen is free to move in any direction, restricted only by contact with internal dummy components.

Pelvis

All metal parts of the pelvis are modeled as rigid bodies. The geometry is represented by facet surfaces.

Jacket

The jacket of the Hybrid III-3 year old consists of the complete skin surface of the torso. This means that the skin part of the pelvis is part of the jacket. Generally the jacket contains two layers of solid hexahedron elements over the complete outer surface. The thick parts in the pelvis and the two parts in the shoulders are modeled using tetrahedron elements. The jacket is not supported to any rigid body. The motion of the jacket is only restricted by contact with internal dummy components.

2.3 MADYMO Input file Description

Before carrying out a particular simulation, the user needs to create an input file for MADYMO. In order to carry out a MADYMO simulation, the model has to be described in an input file. The input file completely describes the simulation model. It consists of data on the multibody systems, number of bodies in each system, joints, center of gravity, material behavior, force interactions, airbag data, belt specifications, etc. The final section of the input file deals with the different outputs needed for analysis purposes. The user has to specify the body or bodies for which output is required. Several types of outputs are available such as displacements, velocities, accelerations, forces, etc. The outputs are stored in different data files which can be accessed by various post processing programs for plotting of graphs and visualizations of

multibody simulations. A multibody algorithm is available which yields the second time derivative of the degrees of freedom. Three numerical integration methods are available, namely the Euler method, a fourth order Runga Kutta with a fixed time step and a fifth order Runga Kutta with a variable time step.

CHAPTER 3

HEAD INJURY DEFINITION

Diffuse injuries are the cause for a variety of disruptions of functional communication within brain structures and between the brain and other body areas [7]. A number of different diffuse brain injures have been identified by the medical community thus far. These include concussions, brain swelling, and diffuse axonal injury (DAI). Diffuse damage is often difficult to detect because it is a closed head injury. The injury is the result of motion causing the head to whip forward and back or left and right. The brain is forced to strike the inside of the skull at high velocity. Such violent movement of the brain may lead to the tearing of blood vessels and bruising of the brain. Contact between the soft brain matter and the rough and bony inner skull results in the greatest damage. While a large percentage of diffuse injuries are mild, other brain injuries such as white matter injuries can be quite severe. Moderate diffuse head injury typically results in no loss of consciousness although retrograde and post-traumatic amnesia and contusion may be present. Concussions such as these are the most common diffuse brain injury and have been estimated to constitute 10 percent of all head, neck, and spine injuries. Due to the reversibility of such injuries, sufferers are generally not given medical attention for this ailment. Concussions of greater severity, where immediate loss of consciousness is a factor, are referred to as cerebral concussions.

The severity of head injury from cerebral concussion generally correlates with the period of time unconscious. Typically, those individuals with cerebral concussions suffer periods with loss of consciousness that are reversible and typically do not exceed 24 hours. Individuals with mild concussions generally display the same symptoms as those suffering from cerebral concussions. Ninety-five percent of those suffering from cerebral concussions are considered to

have made a good recovery after one month. Of those remaining, approximately 2 percent exhibit a moderate deficit and another 2 percent exhibit a severe deficit. In a third of all cases, cerebral concussions result in no brain lesions. Researchers have found the remaining cases of cerebral concussion to be associated with vault fracture, contusions, basilar fracture, depressed fracture, and multiple lesions. Thus, recovery oftentimes depends upon factors other than the cerebral concussion, such as associated brain injuries. When the period of lost consciousness exceeds 24 hours, diffuse brain injury becomes life threatening. The sufferer has crossed the threshold from physiological disruption to anatomical disruption (some of which may be non-reversible). Several days of amnesia, moderate memory deficits, and decerebrate posturing (in which the head is tilted back, the spine is hypoextended, and the extremities are rigid) are often results of extended diffuse axonal injury. Full recovery after a prolonged loss of consciousness is much less likely than in cases where loss of consciousness is less than twenty four hours. After one month of unconsciousness due to diffuse axonal injury, only 21 percent of the sufferers show good recovery, 7 percent do not survive, 21 percent remain in a vegetative state, and 50 percent exhibit moderate to severe deficiencies.

The most severe form of diffuse brain injury, diffuse axonal injury (DAI), occurs when the neural axons are stretched and/or injured during the impact of the brain with the inner skull. Detailed examination of injured brains show that the axons in the white matter of both cerebral hemispheres are torn as a result of the blow. Small hemorrhages of the periventricular region, corpus callosum, and superior cerebellar peduncle are not uncommon although they are often unnoticed by computerized topography (CT) scans. Typically, the initial loss of consciousness extends for days and possibly even weeks. The impact causes disruptions in the axons extending through the midbrain into the brainstem. The presence and continuation of abnormal brainstem

signs is what differentiates diffuse axonal injury (DAI) from a serious concussion. In addition to decerebrate posturing (in which the head is tilted back, the spine is hypoextended, and the extremities are rigid) and post-traumatic amnesia (which may last for several weeks), severe memory loss and motor deficits are also symptoms noted in DAI sufferers. Researchers have found that at the end of a one-month period, 55 percent of DAI sufferers are likely to have died, 3 percent are surviving vegetatively, and 9 percent exhibit severe deficits. Brain swelling and cerebral edemas are secondary injuries that can occur in conjunction with concussions or DAI. They can be life threatening because they increase the severity of the primary injury by heightening the pressure within the cranium. Brain swelling is the phenomena whereby there is an increase in intravascular blood within the brain. Research by has revealed that 28 percent of pediatric head-injury patients and 4 percent to 16 percent of adult head injury patients suffer from brain swelling. Adults incur a 33 percent to 50 percent mortality rate from brain swelling while pediatric patients incur a 6 percent mortality rate. Cerebral edema, which is a different condition than brain swelling, occurs when the volume of the brain increases due to tissue fluid.

CHAPTER 4

POTENTIAL INJURY CRITERIA

4.1 Introduction

The automobile industry has developed many injury criteria that can be used to evaluate the potential for occupant injuries in various impact situations. Injuries occur when the stimuli exceed bodily resistance, causing reactions that may range from altered physiology to structural damage and their various biomechanical reactions. The primary injury suffered in transport accidents is mechanical trauma. Injury criteria are developed to address the mechanical responses of the crash test dummies in terms of risk to life or injury to a living human. They are based on the engineering principle that states that the internal responses of a mechanical structure are uniquely governed by the structure's geometry, material properties, and forces and motions applied to its surface. The criteria have been derived from experimental efforts using human surrogates where both measurable engineering parameters and injury consequences are observed, and the most meaningful relationships between forces/motions and resulting injuries are determined using statistical techniques. A number of these injury criteria have been used in the certification of automobile protection. These injury criteria along with FAR 25.562 criteria need to be evaluated to determine if they are useful as a means of evaluating potential injuries in aircraft scenario. A comprehensive list of these potential injury criteria are described in the next sections.

4.2 Federal Aviation Regulations and NHTSA Standards Injury Criteria [15]

Head injury criteria (HIC): Head injury is judged from the amplitude of injury assessment parameters, which are transformed from physical measurements such as force, acceleration, velocity or displacement. The pioneering research of Holbourn resulted in an engineering

interpretation of the mechanism of head injury for blows of short duration or impact. He proposed that injury is proportional to the force multiplied by the time over which it acts.

The most common way the brain can be injured is by acceleration, which usually produces a closed head injury. This is because there is usually no break of the skin or an open wound. This happens when the head suddenly changes its motion. For closed head impact, without skull fracture, a transient force produces a proportional acceleration of the head. Since the acceleration is a measurable quantity on a moving body during impact, the earliest experimental determination of head injury was related to the translational acceleration. Injury is defined as any HIC value exceeding 1000. This criterion was adapted from the FMVSS No. 208 in 1967 [12]. Generally, a maximum window size of 36 ms is used in the automotive industry. Gurdjian subsequently defined the Head Injury Criteria (HIC) as

$$HIC = \left[(t_2 - t_1) \left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t)dt \right\}^{2.5} \right]_{max}$$

Where, a(t) - resultant acceleration of the head center of gravity in G's, and t1, t2 – initial and final integration time, expressed in seconds.

Compressive load: Measured between the pelvis and the lumbar column not to exceed 1,500 lb. (6.67 KN).

Femur load: The axial compressive load in each femur of the ATD shall not exceed 2,250 lb. (10KN).

Strap Loads: Loads in the individual straps must not exceed 1750 lb. (7.8 KN) for pilots.

Restraint Retention: Upper torso restraint strap must remain on the occupant's shoulder during the impact.

However the loads for the HIC value for the 3 year old child are less per NHTSA's FMVS S21.5. Injury criteria for the 49 CFR Part 572, Subpart P 3-year-old child test dummy.

S21.5.1 All portions of the test dummy shall be contained within the outer surfaces of the vehicle passenger compartment.

S21.5.2 Head injury criteria: (a) For any two points in time, t1 and t2, during the event which are separated by not more than a 15 millisecond time interval and where t1 is less than t2, the head injury criterion (HIC15) shall be determined using the resultant head acceleration at the center of gravity of the dummy head, are, expressed as a multiple of g (the acceleration of gravity) and shall be calculated using the expression:

$$HIC = \left[(t_2 - t_1) \left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t)dt \right\}^{2.5} \right]_{max}$$

(b) The maximum calculated HIC15 value shall not exceed 570.

S21.5.3. The resultant acceleration calculated from the output of the thoracic instrumentation shall not exceed 55 g's, except for intervals whose cumulative duration is not more than 3 milliseconds.

S21.5.4 Compression deflection of the sternum relative to the spine, as determined by instrumentation, shall not exceed 34 millimeters (1.3 in).

4.3 Neck Injury [15][16]

In automotive crashes, the loading on the neck due to head contact force is usually a combination of an axial or shear load with bending. Bending loads are almost always present and the degree of axial or shear force is dependent upon the location and direction of the contact force. For impacts near the crown of the head, compressive forces predominate. If the impact is principally in the transverse plane, there is less compression and more shear. Bending can occur

in any direction because impacts can come from any angle around the head. The cervical spine is shown in Figure 4.1. The following injury modes are considered the most prominent: tension-flexion, tension-extension, compression-flexion and compression-extension in the midsagittal plane.

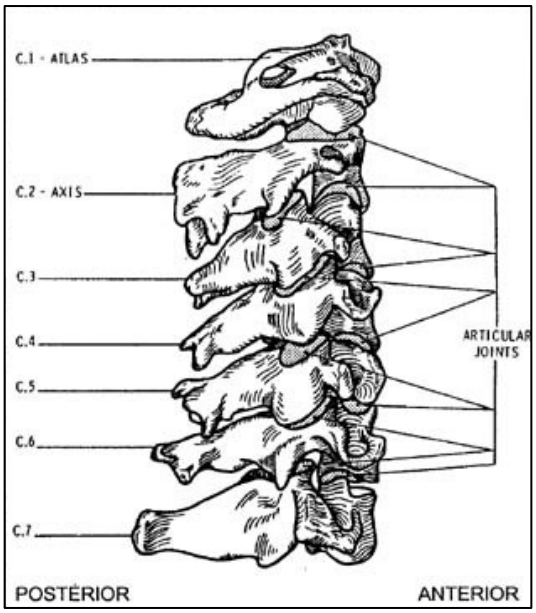

Figure 4.1 Cervical Spine

Tension-Flexion Injuries: Forces resulting from inertial loading of the head-neck system can result in flexion of the cervical spine while it is being subjected to a tensile force. In severe cases atlanto-occipital separation is known to occur.

Tension–Extension Injuries: The most common injury due to combined tension and extension of the cervical spine is the “whiplash” syndrome. A large majority of such injuries involve the soft tissues of the neck and the pain is believed to reside in the joint capsules of the articular facets of the cervical vertebrae. In severe cases, tear drop fractures of the anterior-superior aspect of the vertebral body can occur. Alternately, separation of the anterior aspect of the disc from the vertebral body endplate is known to occur.

Compression-Flexion Injuries: When a force is applied to the posterior-superior quadrant of the head or when a crown impact occurs while the head is in flexion, the neck is subjected to a combined load of axial compression and forward bending. Anterior wedge fractures of vertebral bodies are commonly seen, but with increased load, burst fractures and fracture dislocations of the facets can occur.

Compression-Extension Injuries: Frontal impacts to the head with the neck in extension will cause compression-extension injuries. These involve the fractures of one or more spinous processes and possibly, symmetrical lesions of the pedicles, facets and laminae. If there is a fracture-dislocation, the inferior facet of the upper vertebrae is displaced posteriorly and upward and appears to be more horizontal than on normal x-rays.

Injury tolerance relationships presented by Eppinger in 1982 for neck flexion and extension is summarized in Table [4.1]. for 50% male.

Table 4.1 Injury levels associated with neck extension and flexion for adults.

Extension		**Flexion**	
AIS	**Moment (Nm)**	**AIS**	**Moment (Nm)**
1	M > 47.47	1	M ≤ 61.04
2	M > 61.04	2	M ≤ 189.09
		3	M ≤ 203.45
		4	M > 203.45

4.3.1 Forward Neck Injury Criteria

The current FMVSS No. 208 alternative sled test includes injury criteria for the neck consisting of individual tolerance limits for compression (compression of the neck), tension (force stretching the neck), shear (force perpendicular to the neck column), flexion moment (forward bending of the neck), and extension moment (rearward bending of the neck). Figure 4.2

shows the engineering descriptions of neck loading which causes injury. Figure 4.3 shows the two types of bending motion that can be seen in frontal impacts.

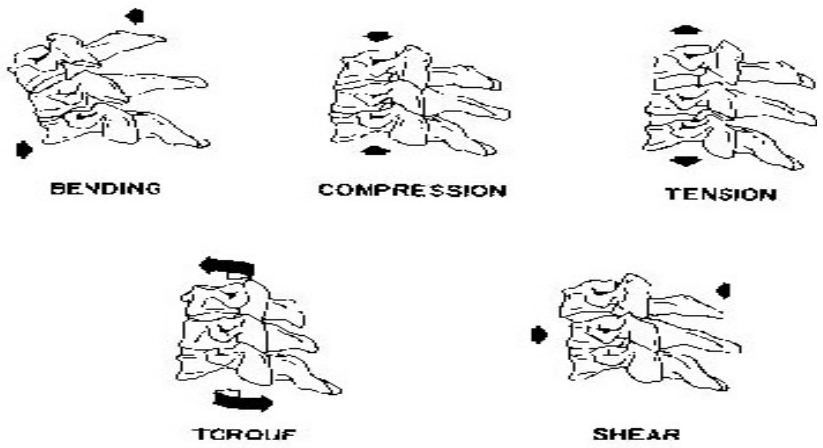

Figure 4.2 Engineering descriptions of neck loading

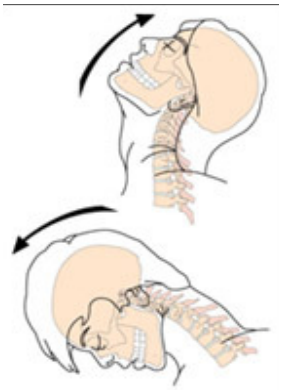

Figure 4.3 Extension (top figure) and Flexion motion of the neck

If axial loads (tension and compression) and bending moments (flexion and extension) are plotted together on a graph, the requirement is that the dummy response must fall within the shaded box, as shown in Figure 4.4.

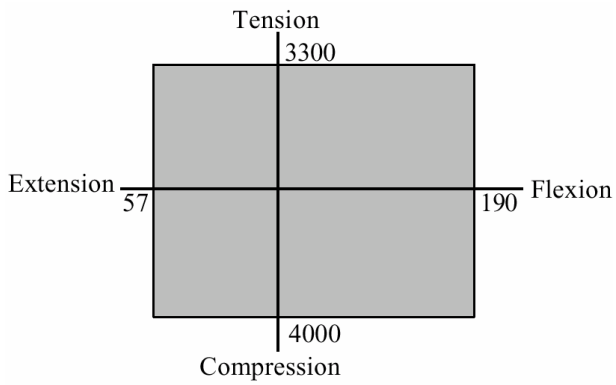

Figure 4.4 Current sled test alternative Neck Injury Criteria

This formulation only takes into account the individual effects of axial loads and bending moments. Later Prasad and Daniel expanded this formulation to consider the linear combinations of loads and moments. According to these formulations the four major classifications on combined neck loading modes; namely tension-extension, tension-flexion, compression-extension, and compression-flexion. The resulting criteria are referred to as Nij, where "ij" represents indices for the four injury mechanisms; namely NTE, NTF, NCE, and NCF. The first index represents the axial load (tension or compression) and the second index represents the sagittal plane bending moment (flexion or extension). The N_{ij} concept was first presented in NHTSA's report on child injury protection [5]. Graphically, the shaded region of the plot in Figure 4.5 shows the region for all four modes of loading which would pass the performance requirements for Nij. The intercept values shown are those proposed for the Hybrid III mid-sized male dummy. The shaded region represents combinations of neck forces and moments that would pass the criteria of Nij 1.0.

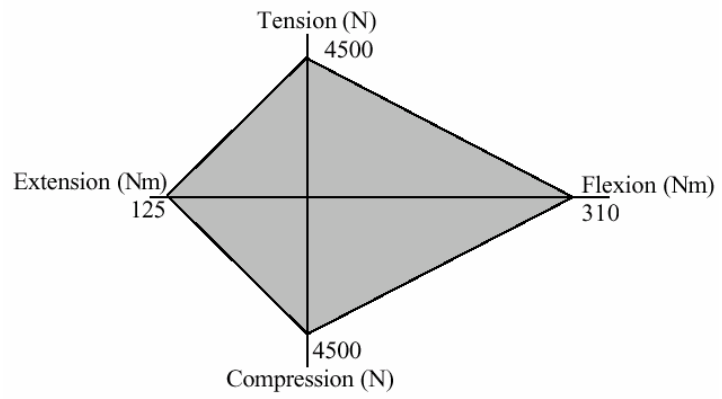

Figure 4.5 Neck Injury Criteria for 50% mid-sized male hybrid III dummy

Since each specific dummy has a unique set of critical intercept values, the subsequent scaling of this plot has been normalized by dividing each semi-axis by its critical intercept value. The resulting plot becomes symmetric about the origin and has maximum allowable values of unity. Graphically, the shaded box shown in Figure 4.6 designates the allowable values of loads and moments represented by this normalized calculation.

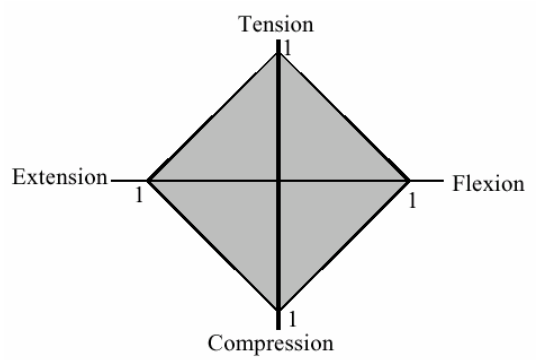

Figure 4.6 Normalized Neck Injury Criteria for all dummies

4.3.2 Calculation of N_{ij}[10]

Loads and moments at each instance in time are normalized with respect to the corresponding critical intercept values defined for tension, compression, extension, and flexion. The normalized flexion and extension moments are added to the normalized axial load to account

for the superposition of load and moment. The proposed neck injury criteria can thus be written as the sum of the normalized loads and moments.

$$N_{ij} = \frac{F_Z}{F_{\text{int}}} + \frac{M_Y}{M_{\text{int}}}$$

Where,

F_Z is the axial load

Fint is the corresponding critical intercept value of load used for normalization

M_Y is the flexion/extension bending moment computed at the occipital condyles

Mint is the corresponding critical intercept value used for normalization

At each instance in time, Fz and My lie in one of the four quadrants shown in Figure 4.7 that correspond to the four loading modes of tension-extension, tension-flexion, compression-flexion, and compression-extension. Nij is computed at each instance in time for only that quadrant where Fz and My lie. For example, if at one instance in time the axial force is +1000 N (*i.e.*, tension) and the bending moment at the occipital condyle is -50 N-m (*i.e.*, extension),

$$N_{TE} = \frac{1000}{4500} + \frac{-50}{-125} = 0.62$$

The maximum *Nij* in time for each of the four loading modes, represented by the four quadrants in Figure 4.8 is computed, from which the maximum Nij for all the four loading modes is determined. The NHTSA neck injury criteria take the form of peak tension and compression axial force limits and combined axial and bending Nij intercepts criteria. The NHTSA N_{ij} intercepts and independent axial force limits are given in Table 4.2 and graphically depicted in Figure 4.7 for the in-position, 50% male (Hybrid III ATD).

Table 4.2 NHTSA Nij Criteria for 50% male

Dummy Size	**Peak Limits**			
50% Male	Tension (N)		Compression (N)	
	4170		4000	
	Nij Intercepts			
50% Male	Tension (N)	Compression (N)	Flexion (Nm)	Extension (Nm)
	6806	6160	310	135

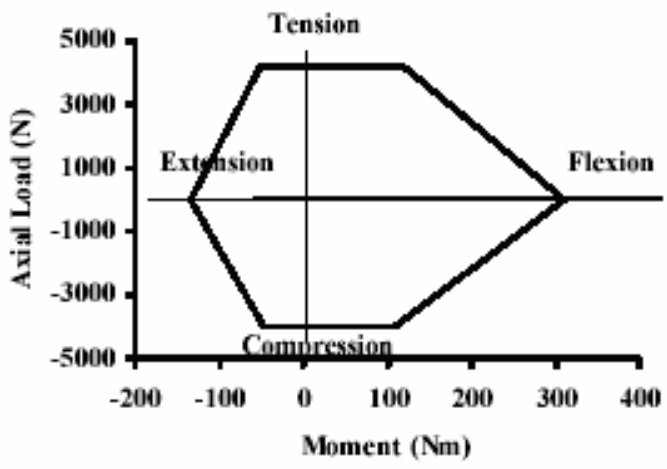

Figure 4.7 NHTSA *Nij* Criteria for 50% Male

The NHTSA Nij criteria were formulated for frontal crash testing and it currently does not consider neck torsional load or lateral bending load in evaluating the potential for neck injury. The NHTSA peak tension and compression axial load limits are considered robust and those limits can readily be used in assessments of the potential of neck injury in certification tests of aircraft seats.

4.3.3 Proposed Side Impact Neck Injury Criteria

In the past, most of the research activities have rightly focused on the more common injury modes that are associated with frontal impacts. Lateral neck injury modes have not had much emphasis.

Aircraft research studies were found to be limited to evaluate the overall dynamic performance characteristics of occupant response. Those studies typically evaluated the potential of occupant injury using the automotive side impact criteria, investigated the potential existence of other injury mechanisms unique to aircraft seats and their installation, and evaluated the performance of various seat/restraint system design concepts. It was found that there are significant differences between the automobile and the aircraft impact environments, injury modes, and seating. A review of head/neck injury modes found that head injury in automobiles, specifically brain injuries, account for more than 60% of the total HARM in the head body region. Consequently much of the past automotive research activities were directed towards enhancing occupant head injury protection. Neck injury (other than whiplash) has not been a dominant occupant injury mode in automobile accidents and thus neck injury research has been limited. The automobile side impact environment may in part minimize the occurrence of serious neck injuries. The lateral velocity change seen in an automobile side impact condition is significantly less than the lateral velocity change that an occupant can see in an aircraft accident. Automotive side impact research tests are thus typically conducted at velocity change levels that are less than those found in the seat dynamic performance standards for aircraft seats. Additionally, the interior design of the automobile may also to some degree minimize the severity of side impact conditions. Consequently little data could be gleaned from automotive side impact research studies that could form the basis of enhanced tolerance limits for lateral neck loading for aircraft applications.

One exception to the above discussion regarding the severity of the automotive impact environment was the Australian Army's TRANSAFE research program. That research program was conducted with a sideward facing seat at vehicle velocity change levels that were consistent with those found in the seat dynamic performance standards for aircraft seats. The TRANSAFE research program did demonstrate the potential for occupant neck injury but it did not establish any new human tolerance limits for lateral neck loading. Human subject tests were conducted at impact conditions (velocity change levels less than 22 ft/sec) that were below the severity level that would cause any serious or permanent injuries. The results of the human subject tests may provide some insight with respect to non-injurious human neck tolerance limits but their usefulness with respect to defining an injurious human neck tolerance limit is limited. A number of automotive cadaver-sled and cadaver-car lateral impact test results were reviewed. It was found that all of the cadaver impact test programs were conducted at velocity change levels consistent with the automotive accident impact environment that is less than the lateral velocity change that an occupant can experience in an aircraft accident. Most of the neck injuries seen in those studies were at an AIS 1 level while one AIS 3 (fracture) and one AIS 5 (spinal cord laceration) were found. A number of other lateral head/neck flexion and impact response studies and NHTSA's improved Nij neck injury criteria were also reviewed. However, none were found to provide any definitive tolerance limits for lateral neck loading.

After reviewing all these data references, Stephen J. Soltis of the Federal Aviation Administration proposed two forms of candidate tolerance limits for lateral neck loading proposed. These limits are based on the maximum measured kinematics and load values found in this review, most of which appear to reach the onset of minor injury and the threshold of serious injury.

1. The first form is based on the kinematics response of the occupant.

- Lateral neck flexion not to exceed 60 degrees. This angle is measured between the head anatomical vertical axis and the mid-sagittal plane of the ATD. Neck flexion was a cited concern by many of the researchers.
- Peak linear acceleration not to exceed 36 G's measured at the C.G of the ATD's head.
- Peak angular acceleration of the ATD's head not to exceed 2600 rad/sec^2.
- Where the head strike with structures or other obstacles occurs the kinematics-based limits are not to be exceeded up to the point of head contact. During head contact HIC not to exceed 1000.

2. The second form of tolerance limits is based on the peak axial loads and moments measured in the neck of the occupant.

- Lateral neck moment (Mx) not to exceed 536 inch-pounds (i.e., 487 inches-pounds increased 10% to account for muscle strength) or 60 Nm measured at the upper neck load cell of an ATD.
- The maximum axial loads not to exceed 940 lbs. force (4170 N) tension and 900 lbs. force (4000 N) compression.
- Nij injury criteria using the NHTSA's intercepts with the above lateral moment limit as shown in Table 4.3.

Table 4.3 Proposed Lateral Load Nij Criteria for adults.

Dummy Size	**Peak Limits**		
50% Male	Tension (N)		Compression (N)
	4170		4000
	Nij Intercepts		
50% Male	Tension (N)	Comp (N)	Lateral Moment (Nm)
	6806	6160	60

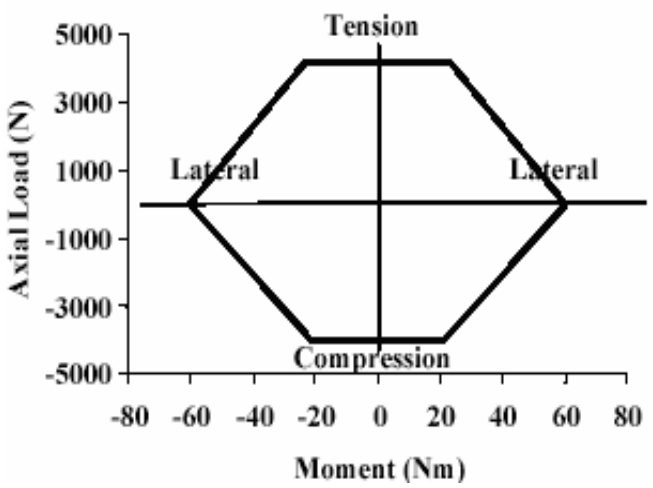

Figure4.8 Proposed Lateral Load *Nij* Criteria

It is believed that these criteria represent thresholds of serious human neck injury. The FAA and NHTSA are initiating research tasks directed to further define tolerance limits for lateral neck loading. The proposed Nij criteria are shown in Figure 4.8. Recent studies have been done in CAMI/FAA with Hybrid III ATD and they have come up with pass/fail neck injury

criteria based on a combination of maximum lateral bending moment (Mx), tensile force (Fz), and shear force (Fy) measured with the Hybrid III neck load cell.

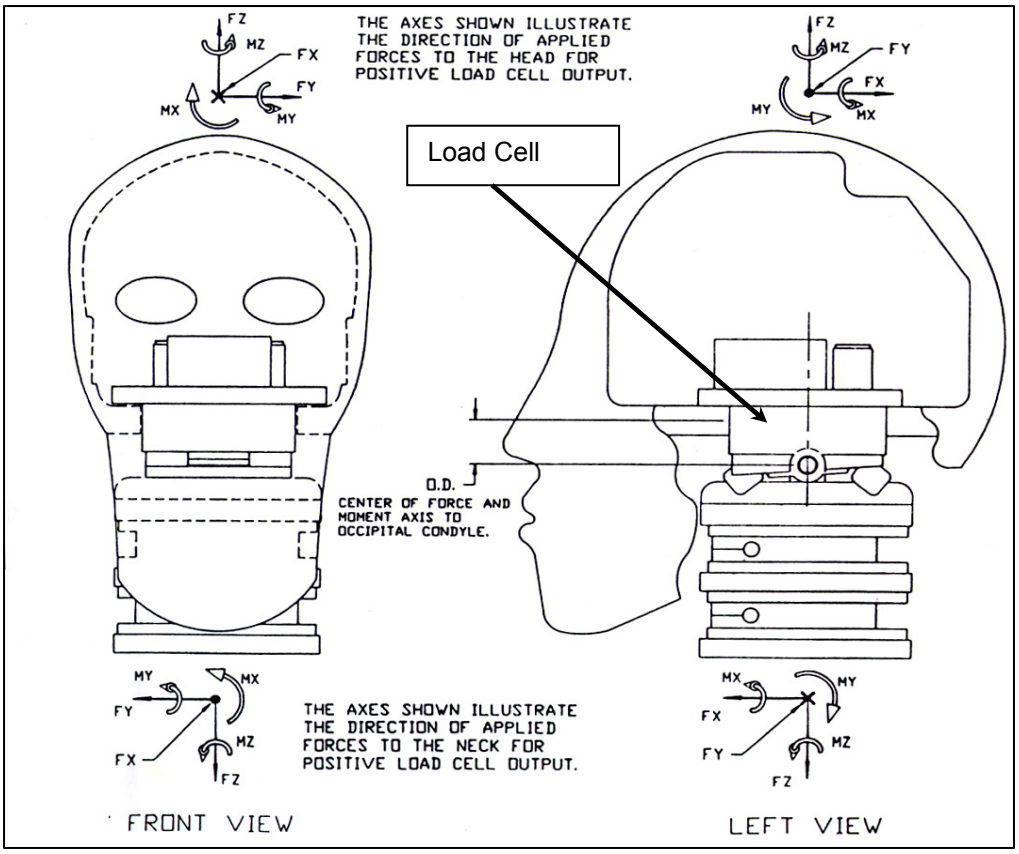

Figure 4.9 Hybrid III Upper Neck Load Cell Orientation

Figure 4.9 explains about the forces and bending moments acting on the human neck. In Hybrid III the axial loads act in the Z direction and the lateral bending moments act about the X-axis. So the Neck injury criteria can be calculated as

$$\text{Nij} = \left(\frac{Mx}{Mxc}\right) + \left(\frac{Fz}{Fzc}\right)$$

where,

Mxc is the corresponding critical intercept used for normalization

Fzc is the corresponding critical intercept used for normalization

The following maximums have been proposed:

Lateral bending moment Mx < 487 inch-lbs

Tensile force Fz < 1530 lbs

Compressive force Fz < 1200 lbs

Lateral shear force Fy < 250 lbs

During the impact response period, the combination of Fz and Mx must be within the following limit:

$$\text{Nij} = \left(\frac{Mx}{487}\right) + \left(\frac{Fz}{1530}\right) < 1.0$$

In this research the above formula has been used to calculate the neck injury criteria and to check whether the injury is within the safe limits.

4.3.4 Neck Injury Criteria for 3 year old child.

Understanding the tolerance level of every body region is crucial for the design engineer. However, human tolerance varies greatly with age and gender [11]. To narrow the range, the tolerance is usually defined for a 50th-percentile, middle-aged male. This means that elderly people, women, and children may be less well protected, and efforts are under way to change government standards to consider tolerance levels of women and children. Information on the responses and tolerances of children is sparse because child cadavers are not readily available. The criterion has been developed by the Federal Motor Vehicle Safety Standards for the 3 year old child.

FMVS states in S21.5.5 Neck injury [12], that when measuring neck injury each of the following injury criteria shall be met. (a) Nij. (1) The shear force (Fx), axial force (Fz), and bending

moment (My) shall be measured by the dummy upper neck load cell for the duration of the crash event as specified in S4.11. Shear force, axial force, and bending moment shall be filtered for Nij purposes. (2) During the event, the axial force (Fz) can be either in tension or compression while the occipital condyle bending moment (Mocy) can be in either flexion or extension. This results in four possible loading conditions for Nij: Tension-extension (Nte), tensionflexion (Ntf), compression-extension (Nce), or compression-flexion (Ncf). (3) When calculating Nij using the equation:

Nij = (Fz / Fzc) + (Mocy / Myc), the critical values, Fzc and Myc, are:

(i) Fzc = 2120 N (477 lbf) when Fz is in tension

(ii) Fzc = 2120 N (477 lbf) when Fz is in compression.

(iii) Myc = 68 Nm (50 lbf-ft) when a flexion moment exists at the occipital condyle

(iv) Myc = 27 Nm (20 lbf-ft) when an extension moment exists at the occipital condyle.

(4) At each point in time, only one of the four loading conditions occurs and the Nij value corresponding to that loading condition is computed and the three remaining loading modes shall be considered a value of zero. The expression for calculating each Nij loading condition is given by:

Nij = (Fz / Fzc) + (Mocy / Myc)

(5) None of the four Nij values shall exceed 1.0 at any time during the event.

Peak tension. Tension force (Fz), measured at the upper neck load cell, shall not exceed 1130 N (254 lbf) at any time.

Peak compression. Compression force (Fz), measured at the upper neck load cell, shall not exceed 1380 N (310 lbf) at any time.

4.4 Abbreviated Injury Scale

The National Automotive Sampling System (NASS) case collection began in 1979. Based on nationally representative sampling units, police accident reports (PARS) are collected from these sites for review and subsequent inclusion in the General Estimates System (GES) or Crashworthiness Data Systems (CDS) databases. The CDS is a representative sample of crashes occurring in the United States. The PAR becomes the basis of FARS crash data. The CDS yields a comprehensive description of the crash events based on the PAR. These cases are then investigated to obtain a complete file on the vehicles involved in the crash, the geometry of the crash location, the interaction of the vehicles with the geometry/location attribute, the demography of the passenger vehicle occupants, and injury mechanisms/patterns, if any exist, for each of the occupants. Injury severity is defined using the Abbreviated Injury Scale (AIS), where the injury severity ranges from AIS 1 (minor) through AIS 6 (maximum or untreatable). The AIS is used to assess risk of fatality [14]. Table 4.4 provides the AIS injury categories [13]:

Table 4.4 AIS Injury Categories

AIS 1	Minor	Light brain injuries with headache, vertigo, no loss of consciousness, light cervical injuries, whiplash, abrasion, contusion
AIS 2	Moderate	Concussion with or without skull fracture, less than 15 minutes unconsciousness, corneal tiny cracks, detachment of retina, face or nose fracture without shifting.
AIS 3	Serious	Concussion with or without skull fracture, more than 15 minutes unconsciousness without severe neurological damages, closed and shifted or impressed skull fracture without unconsciousness or other injury indications in skull, loss of vision, shifted and/or open face bone fracture with antral or orbital implications, cervical fracture without damage of spinal cord.
AIS 4	Severe	Closed and shifted or impressed skull fracture with severe neurological injuries.
AIS 5	Critical	Concussion with or without Skull fracture with more than 12 hours unconsciousness with hemorrhage in skull and/or critical neurological indications
AIS 6	Maximum (untreatable)	death, partly or fully damage of brainstem or upper part of cervical due to pressure or disruption, Fracture and/or wrench of upper part of cervical with injuries of spinal cord.

4.5 Injury and Pass/Fail Criteria [17]

There are no specific FAR's related to this type of injury during boarding. The following reference values are from FMVSS No. 208 and from IIHS's for frontal and side impact crash for a three year old (Hybrid III-three year old child) with an injury risk of 5% for an AIS 4 or greater injury, which mimic this type of injury

Head

HIC 15 ms	570

Neck

Nij	1
<u>Intercepts</u>	
F_{ZC} Tension, F_T (N)	2120
F_{ZC} Compression, Fc (N)	2120
M_{YC} Flexion Moment, M_F (Nm)	68
M_{YC} Extension Moment, M_E (Nm)	27
Peak Tension FZ (N), at upper load cell	1130
Peak compression FZ (N) at upper load cell	1380

CHAPTER 5

FALL INJURY FROM THE AIRCRAFT AIR STAIR TOP STEP

5.1 Introduction

This chapter of the research work deals with the kinematics analysis of the Hybrid III dummy in a standing position at the top of the stair case. Since the height of the dummy is 37.2 inches tall, it would be reasonable to assume that the child would be able to slide underneath the handrail at the top of the step, if the 3 feet high handrail is at around 45 degrees to the ground and the step is horizontal. It would be also reasonable to assume that a three year old child, though accompanied by a parent or guardian, is not physically restrained at the top of the step. Due to the location of the hand rails, if the child's head is slightly angled, it will easily slip between the rails and airplane structure in the door entering position. The child was placed at 30 degrees to the door entering position. Therefore, the most likely scenario then becomes that the child is looking away with its head and body at a slight angle, steps too far to the side, slides under the hand rail and falls.

Friction between the child's foot in contact with the step and the top step plays a key role in the kinematics and the injury response. The common static friction coefficients are given in Table 5.1. The simulations would be carried out in 10 different cases with a change in coefficient of friction in each case, i.e., ranging from 0.1 to 1.0. The kinematics and injury response were different.

5.2 Model Development

The airplane interface data for the airplane entry door and staircase height was taken from the technical specifications [2]. This interface was created for dummy placement and orientation. The step height for the top stair was placed at 2.42 m. The simulated ground and the stair step

were created as rigid bodies using Easy Crash MADYMO Version 5.5. The model used for the work is shown in the figure 5.1.

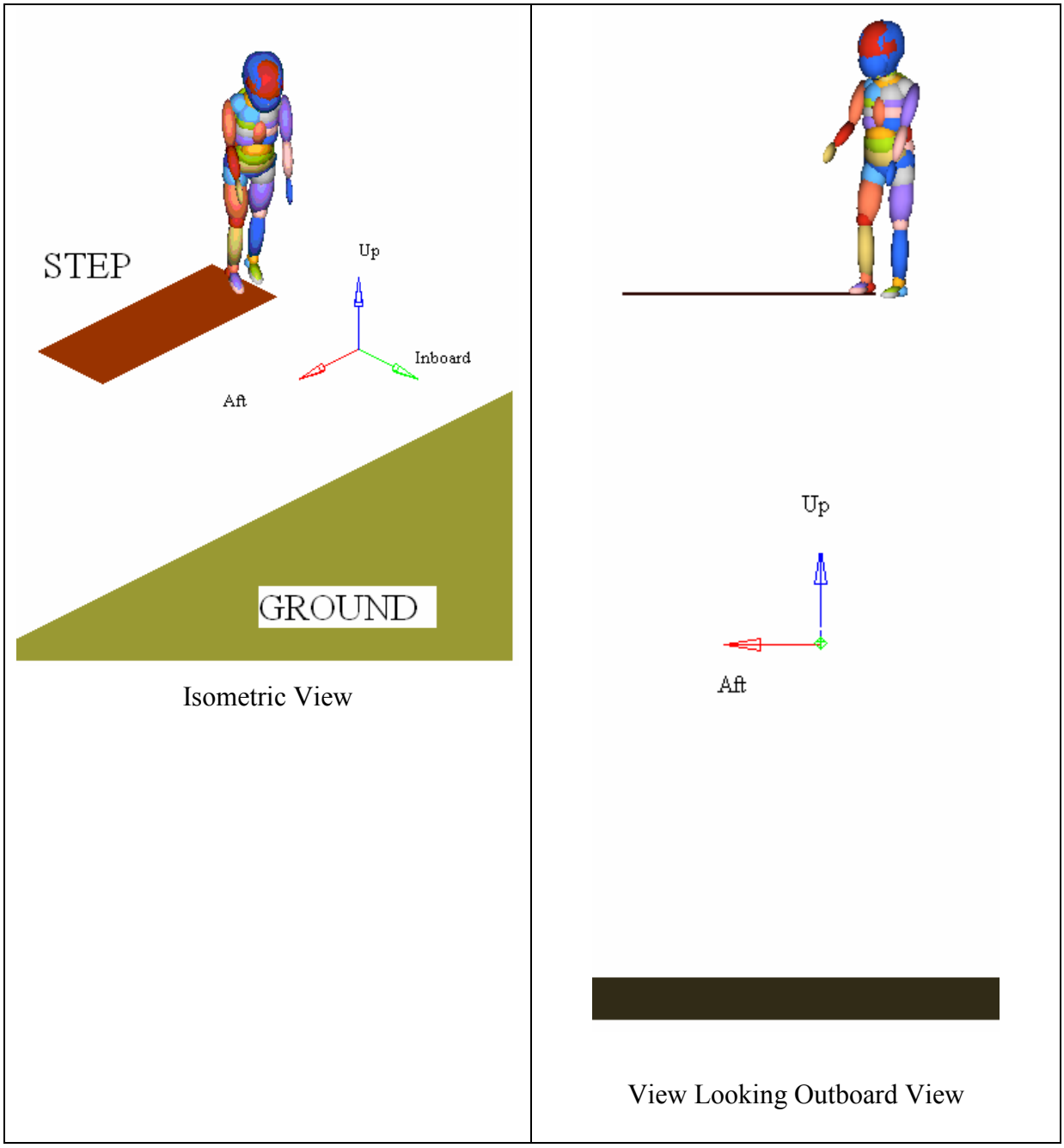

Figure 5.1 Isometric and inboard looking View of the model

Table 5.1 Common Coefficient of Friction between two materials [4].

Material 1	Material 2	Static (Clean)
Leather	Metal (Dry)	.6
Leather	Metal (wet)	.4
Skin	Metal (dry)	.8-1.0
Rubber	Concrete (dry)	1.0
Rubber	Concrete (wet)	.3

5.3 Kinematics of the dummy under gravity free fall, injury response and results.

There was no acceleration pulse given to the dummy. The gravity of 9.81 m/s^2 (1G) was assigned to the dummy. The placement of the dummy was with one foot off the top step of the stairs. Figures 5.2 through 5.31 shows the kinematics results for each different coefficient of friction between the step and the contact foot. The graphs after each simulation show the resultant head acceleration for the dummy and value of the neck injury value Nij.

Table 5.2 shows the values for the Coefficient of friction, HIC15, HIC 36, their corresponding ΔT's (ms) and Nij values for each case. Figure 5.34 through 5.32 show the plots for HIC15, HIC36 and Nij against the coefficient of friction. The HIC 36 value is not included in the pass/fail criteria but is shown for comparison purposes. The Nij value graphs show the maximum value for each case in four different loading conditions, NIJ-NCF (Compression-flexion), NIJ-NTF (Tension-Flexion), NIJ-NCE (Compression-Extension) and NIJ-NTE (Tension-Extension).

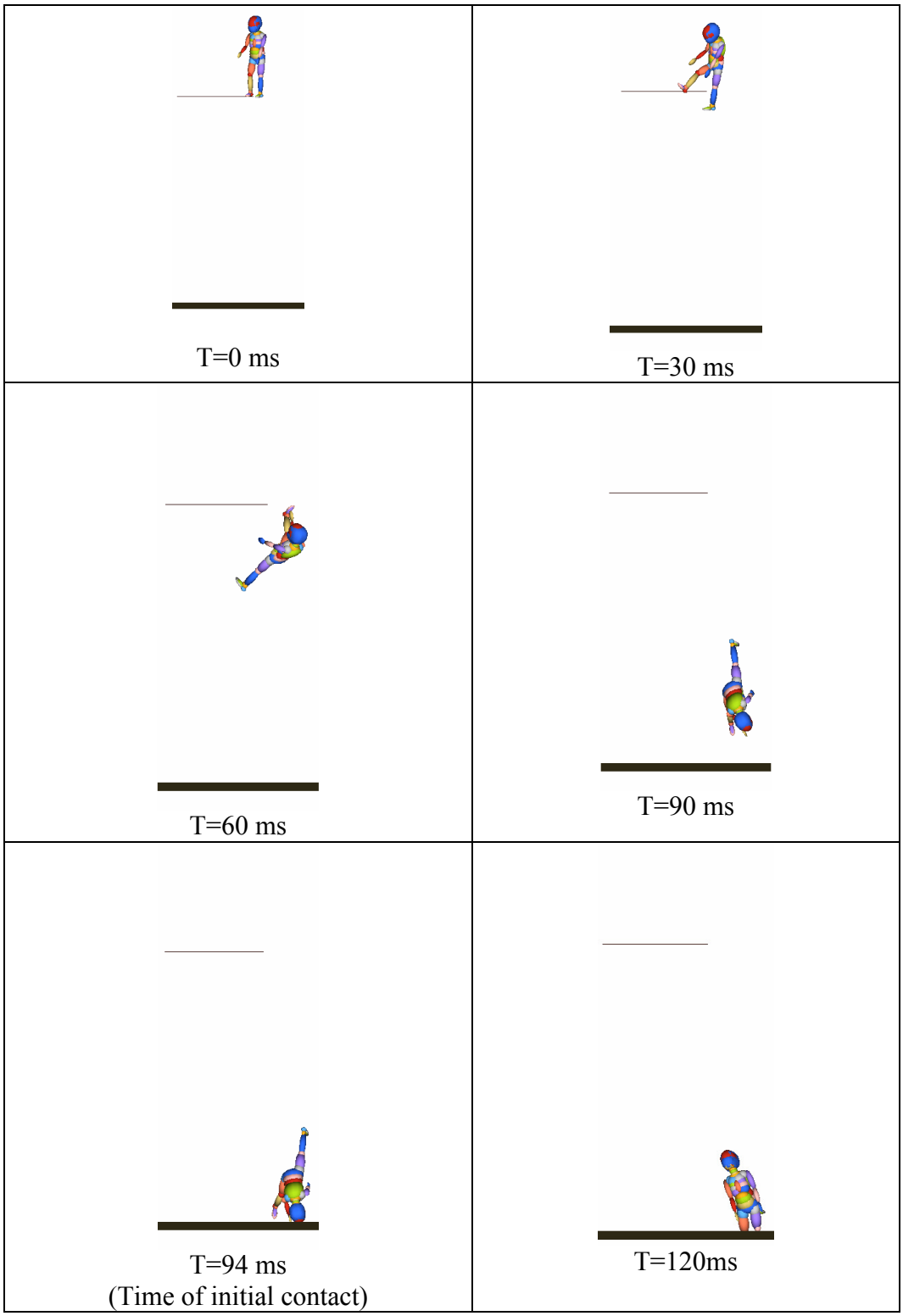

Figure 5.2 Case 1 Kinematics Analysis with 0.1 Coefficient of friction.

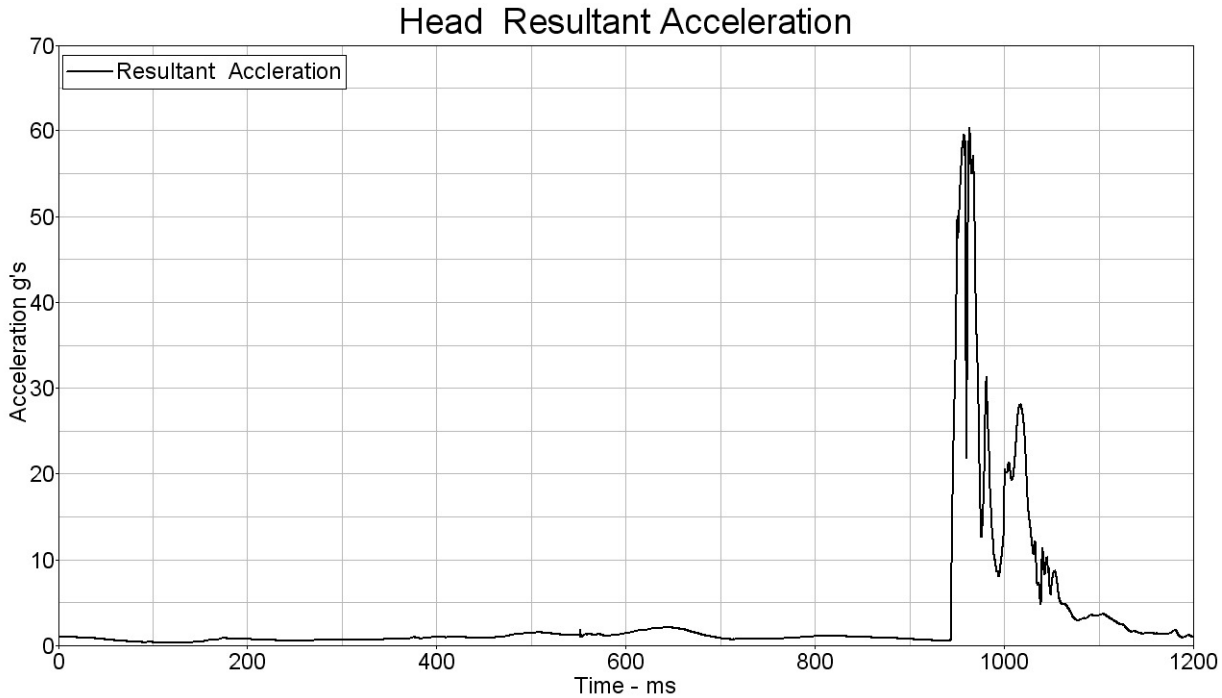

Figure 5.3 Case 1 Resultant Acceleration of the Head with 0.1 Coefficient of friction.

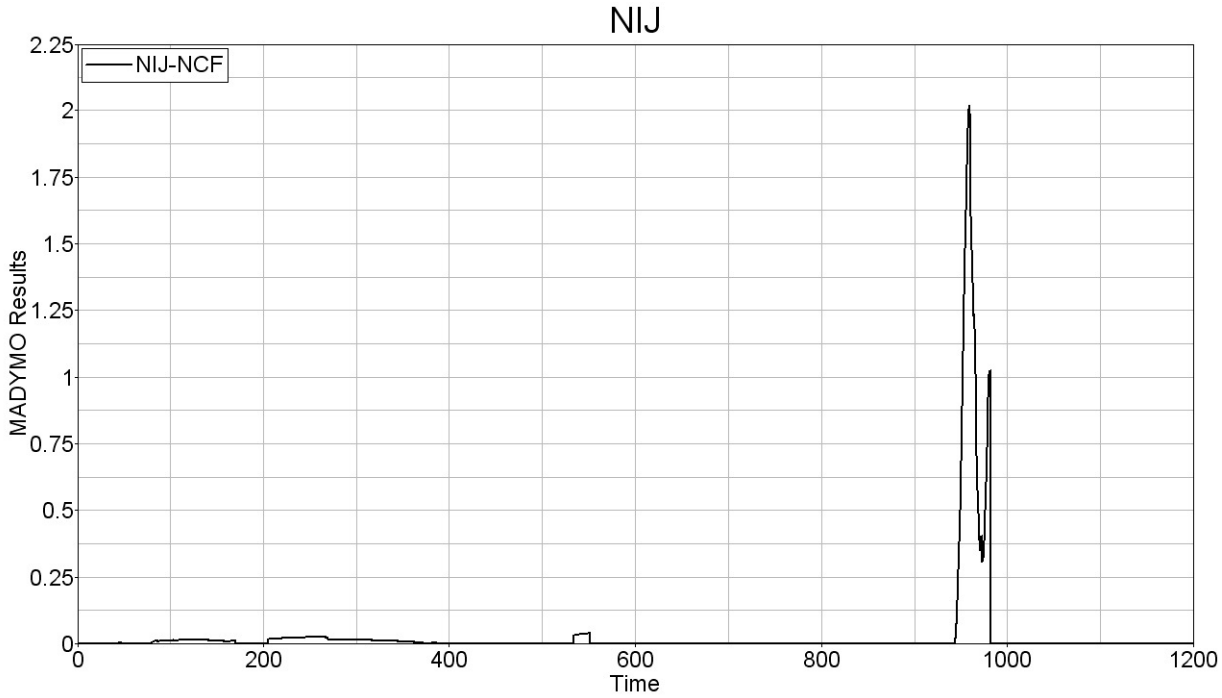

Figure 5.4 Case 1 Neck injury criteria NIJ with the 0.1 Coefficient of friction.

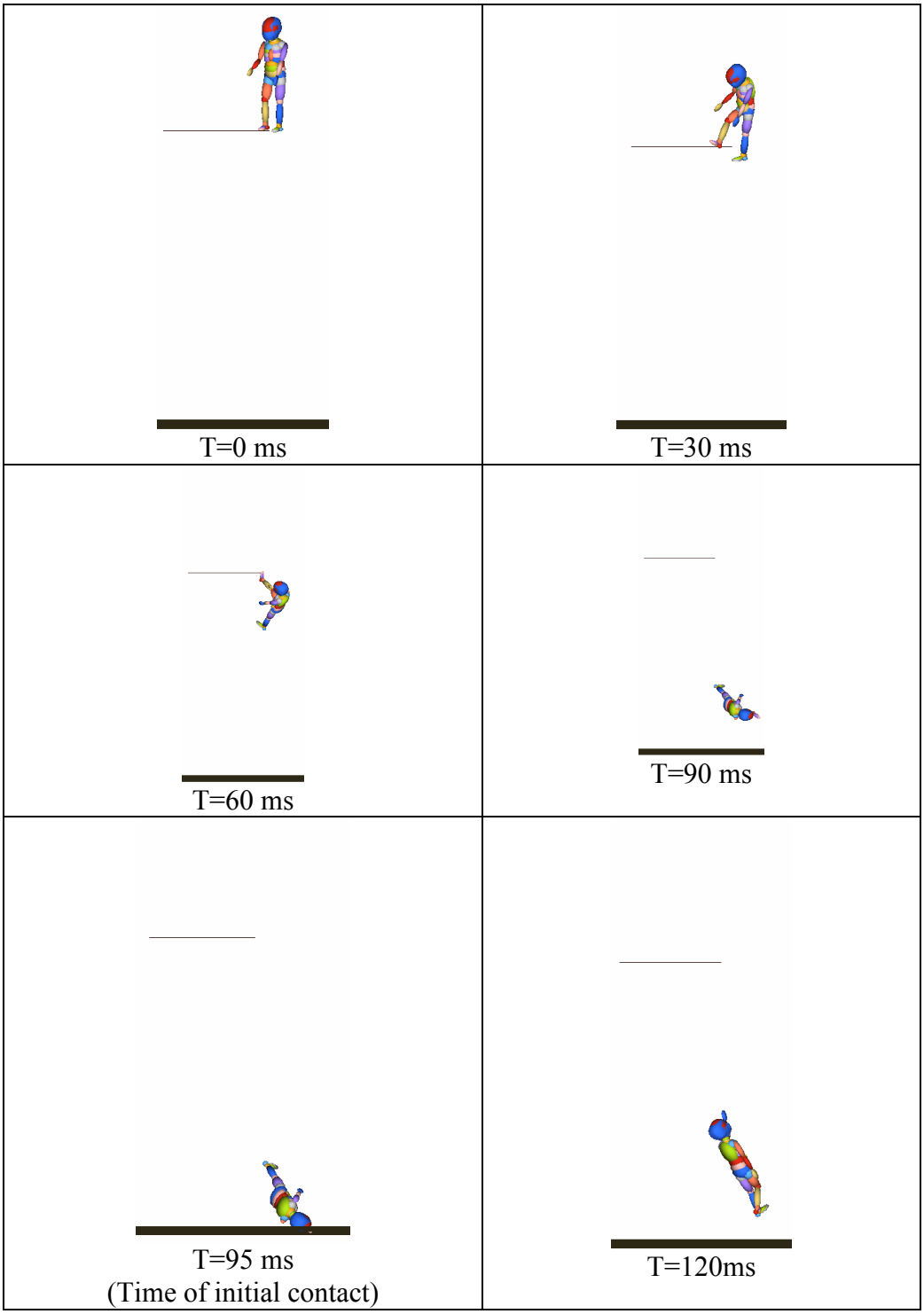

Figure 5.5 Case 2 Kinematics Analysis of with the 0.2 Coefficient of friction.

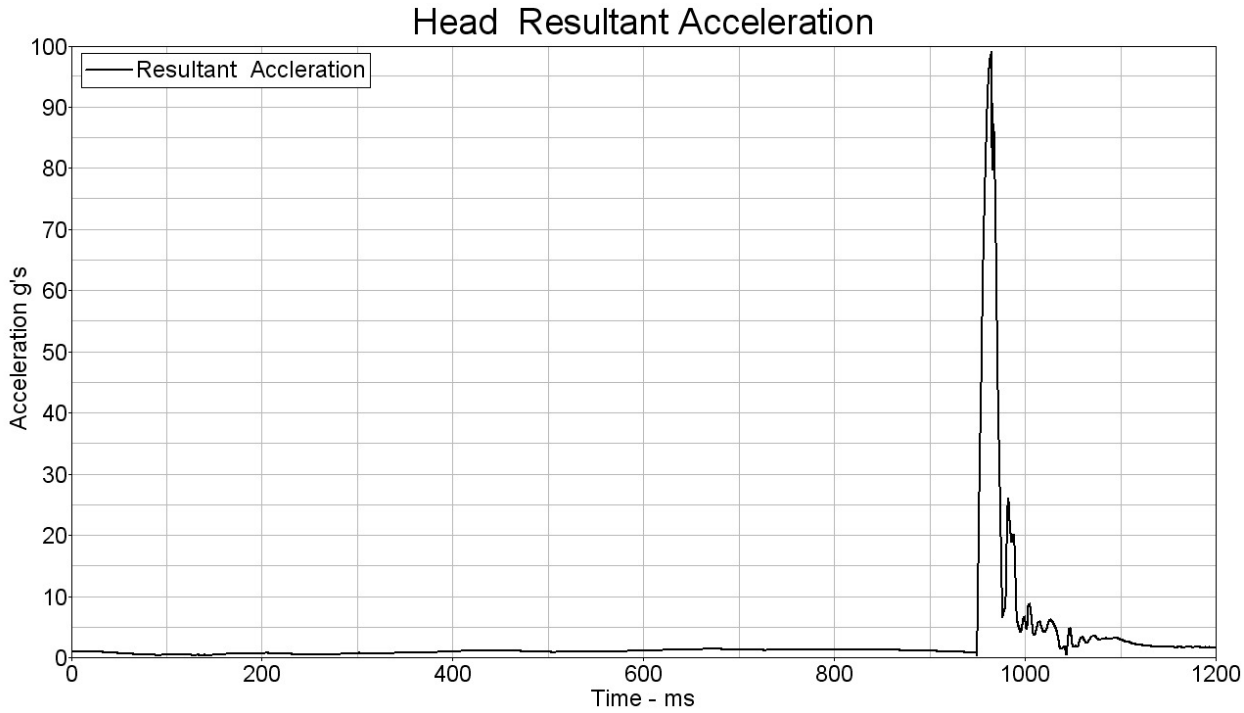

Figure 5.6 Case 2 Resultant Acceleration of the Head with 0.2 Coefficient of friction.

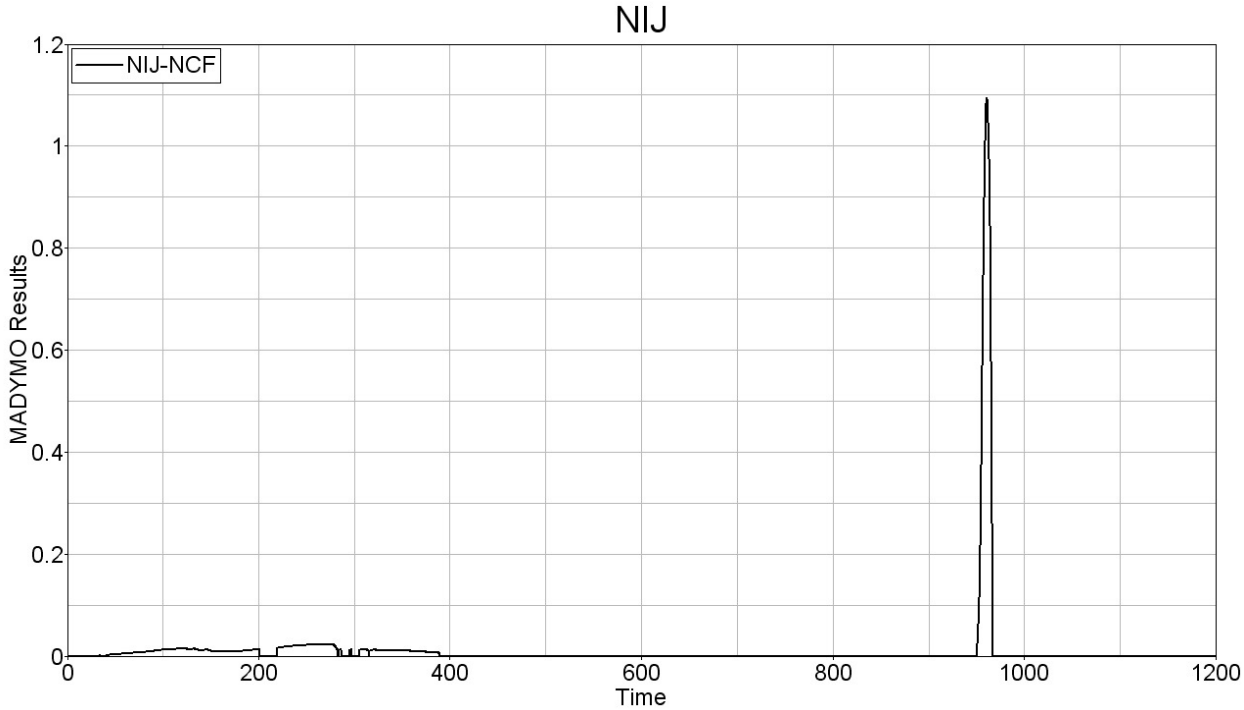

Figure 5.7 Case 2 Neck injury criteria NIJ with 0.2 Coefficient of friction.

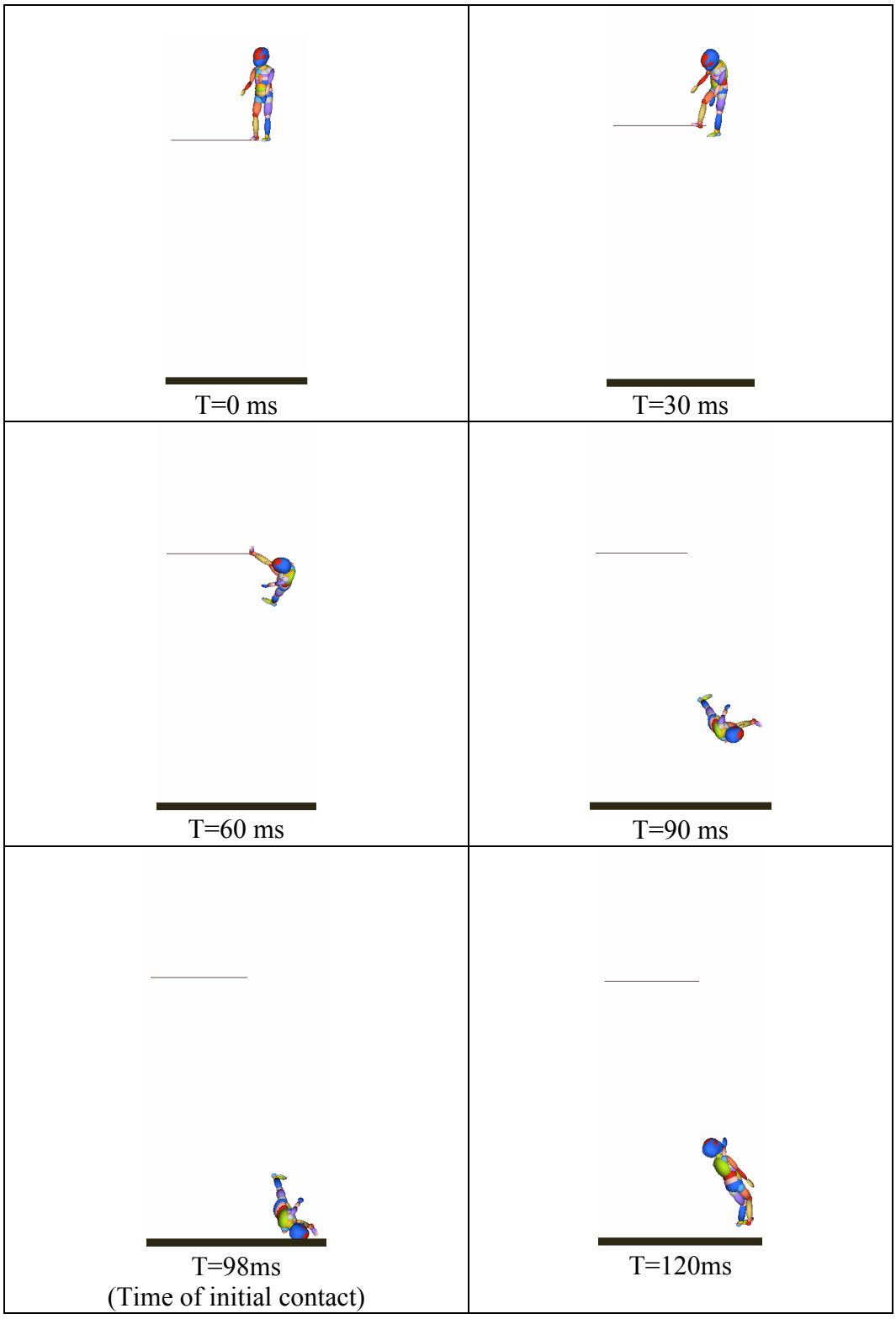

Figure 5.8 Case 3 Kinematics Analysis of with the 0.3 Coefficient of friction.

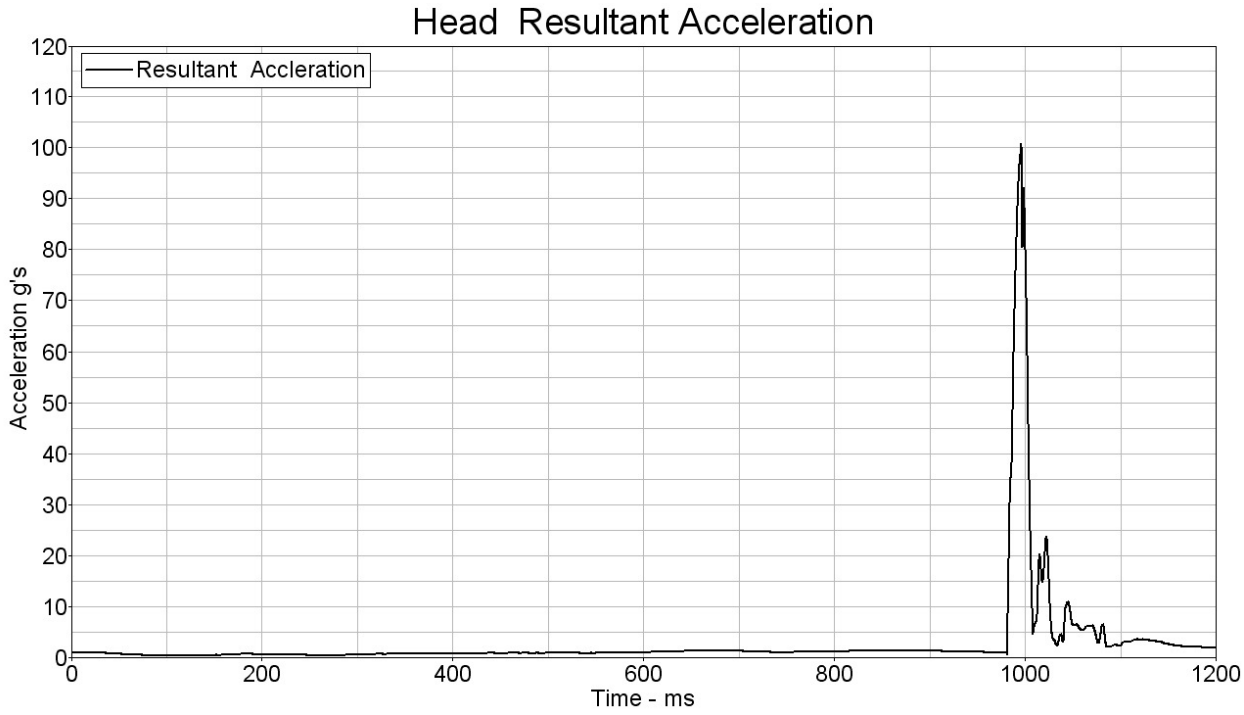

Figure 5.9 CASE 3 Resultant Acceleration of the Head with 0.3 Coefficient of friction.

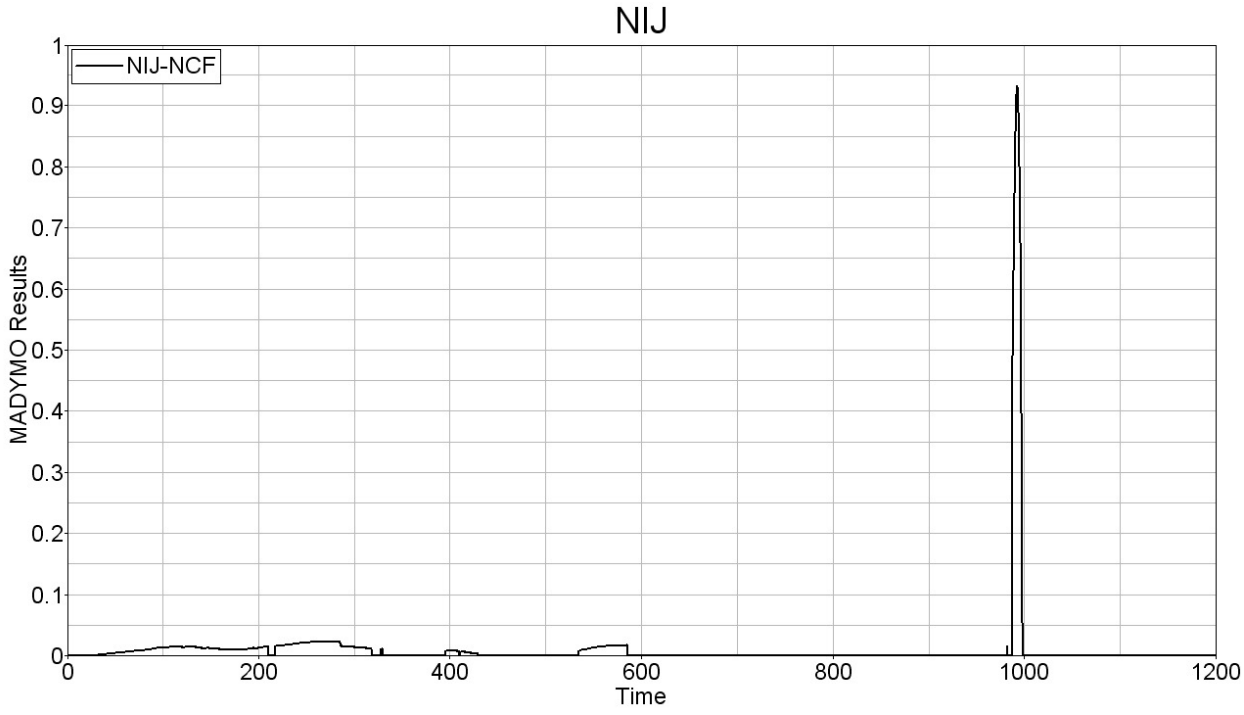

Figure 5.10 Case 3 Neck injury criteria NIJ with 0.3 Coefficient of friction.

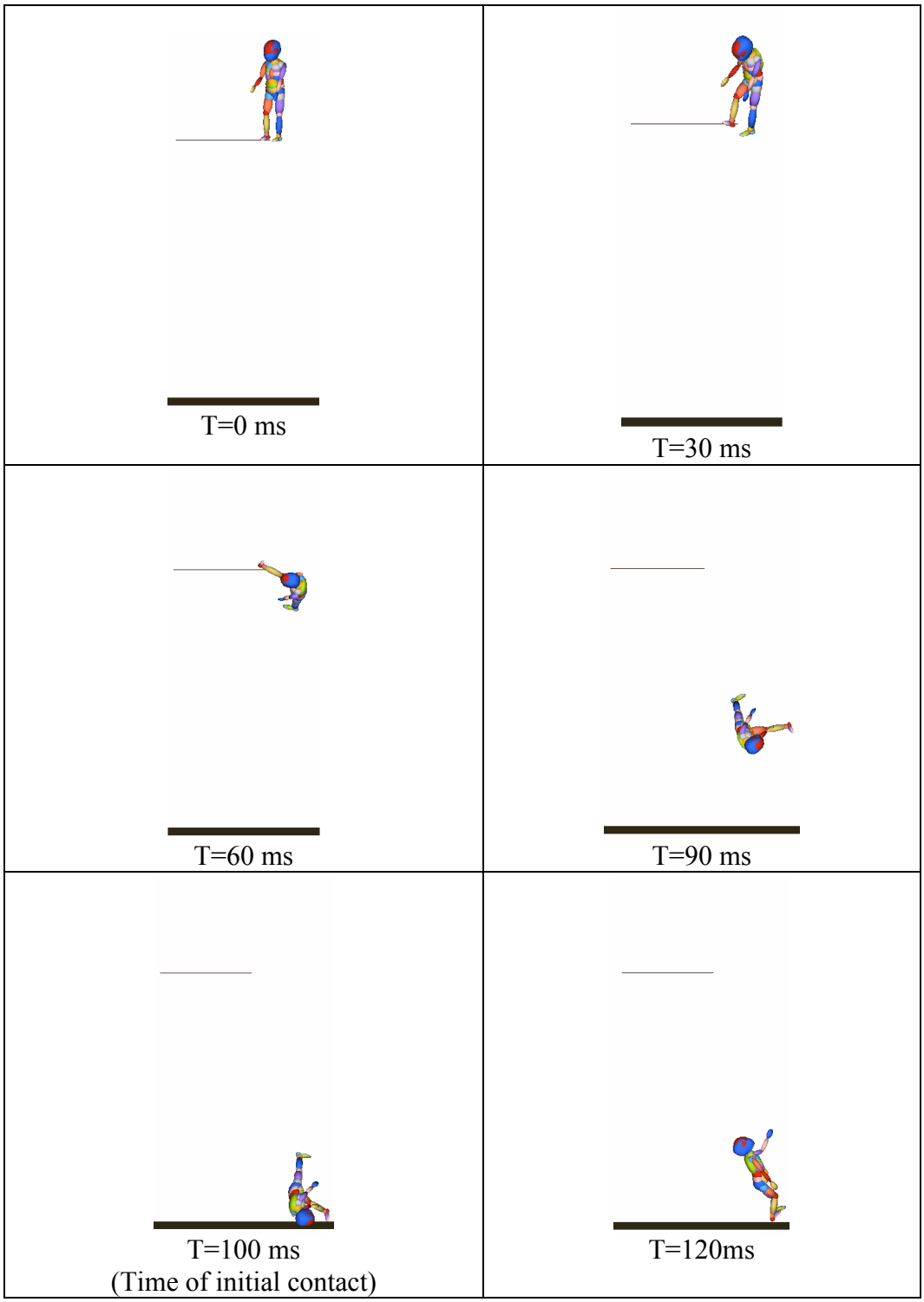

Figure 5.11 Case 4 Kinematics Analysis of with the 0.4 Coefficient of friction.

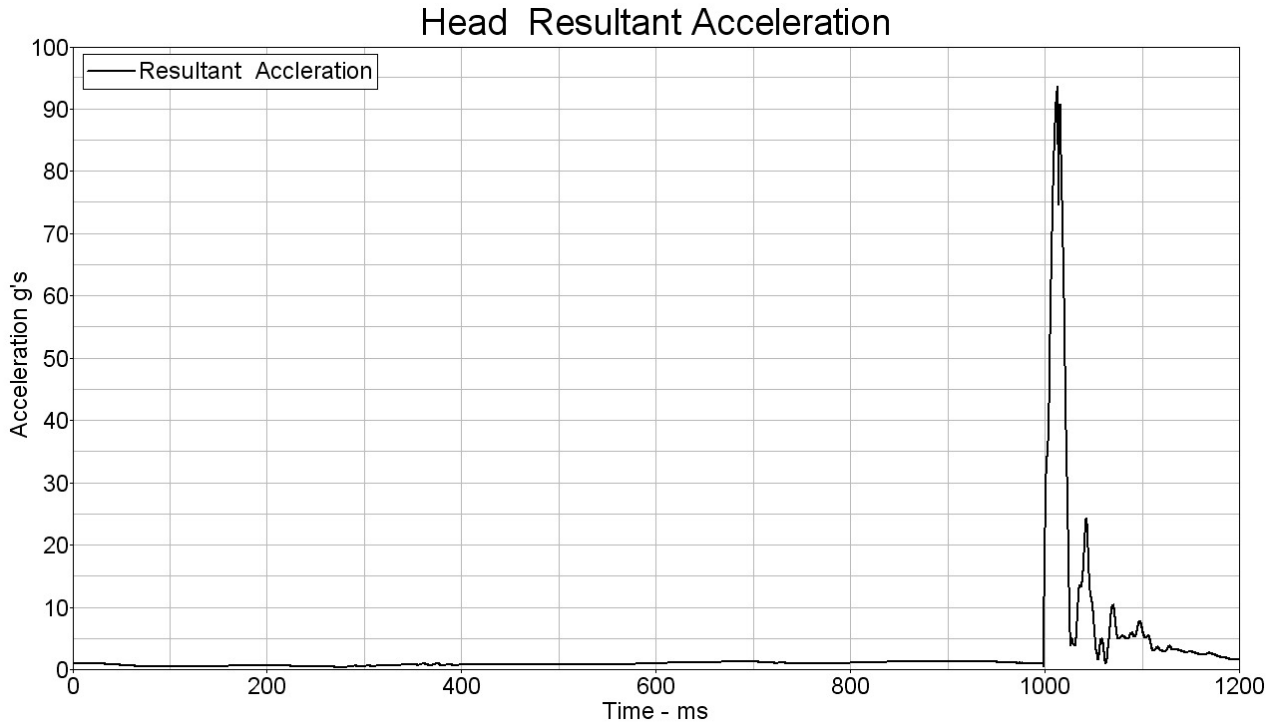

Figure 5.12 Case 4 Resultant Acceleration of the Head with 0.4 Coefficient of friction.

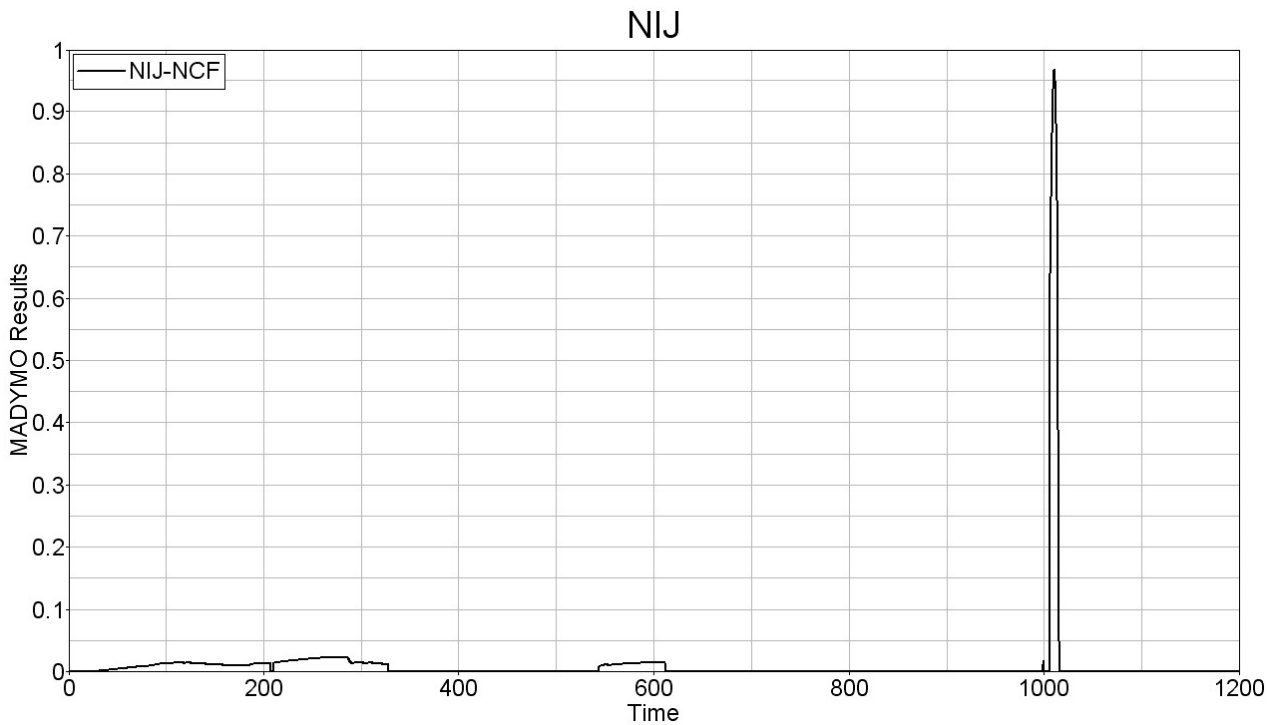

Figure 5.13 Case 4 Neck injury criteria NIJ with 0.4 Coefficient of friction.

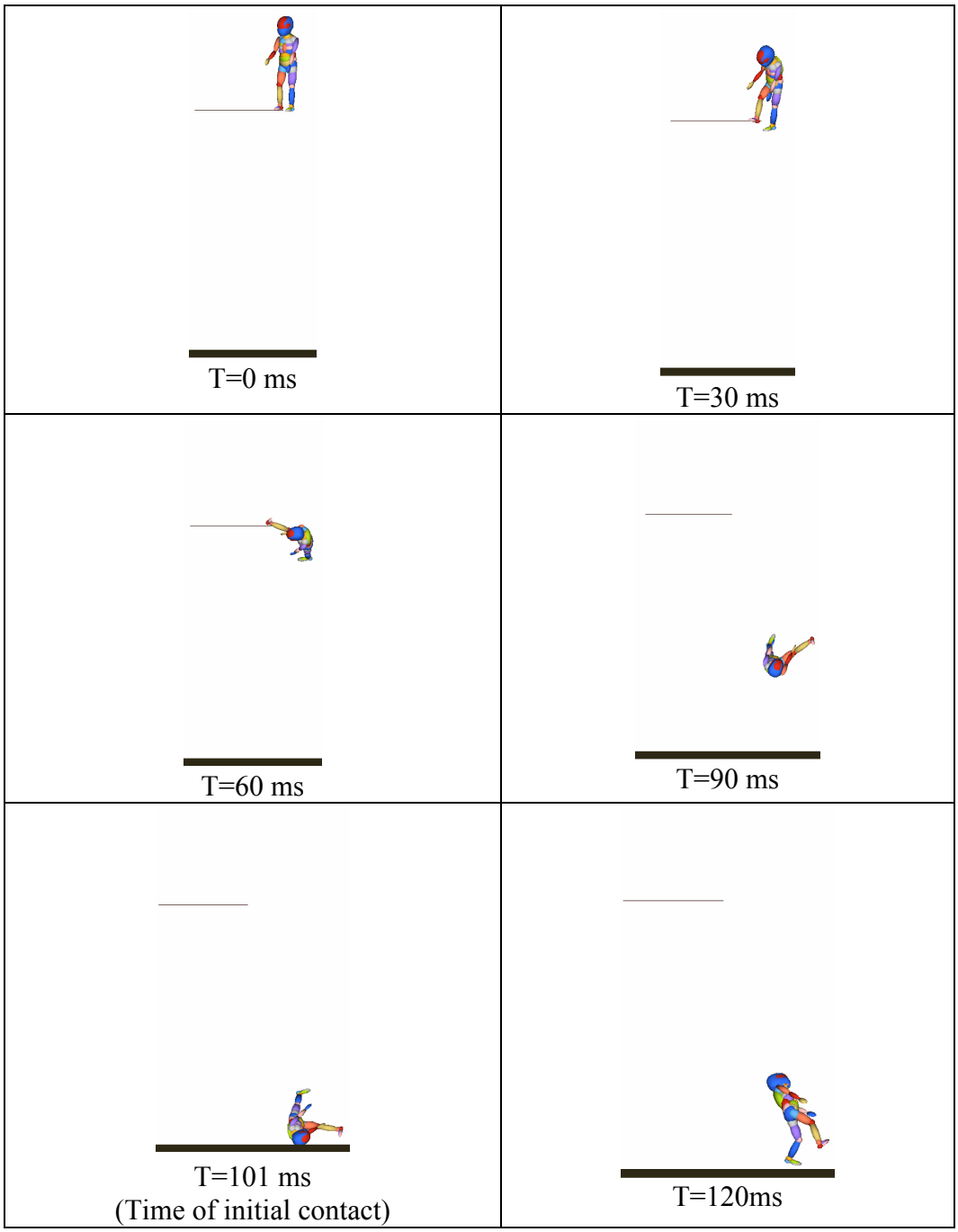

Figure 5.14 Case 5 Kinematics Analysis with the 0.5 Coefficient of friction.

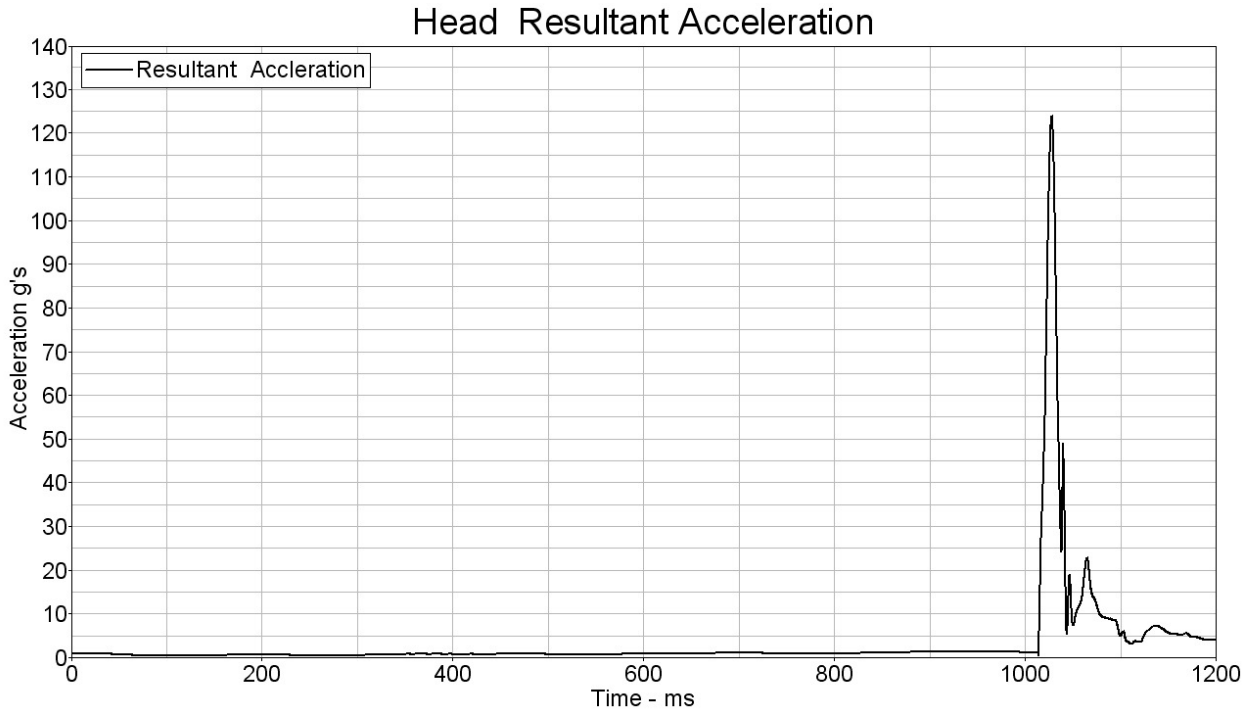

Figure 5.15 Case 5 Resultant Head Acceleration with 0.5 Coefficient of friction.

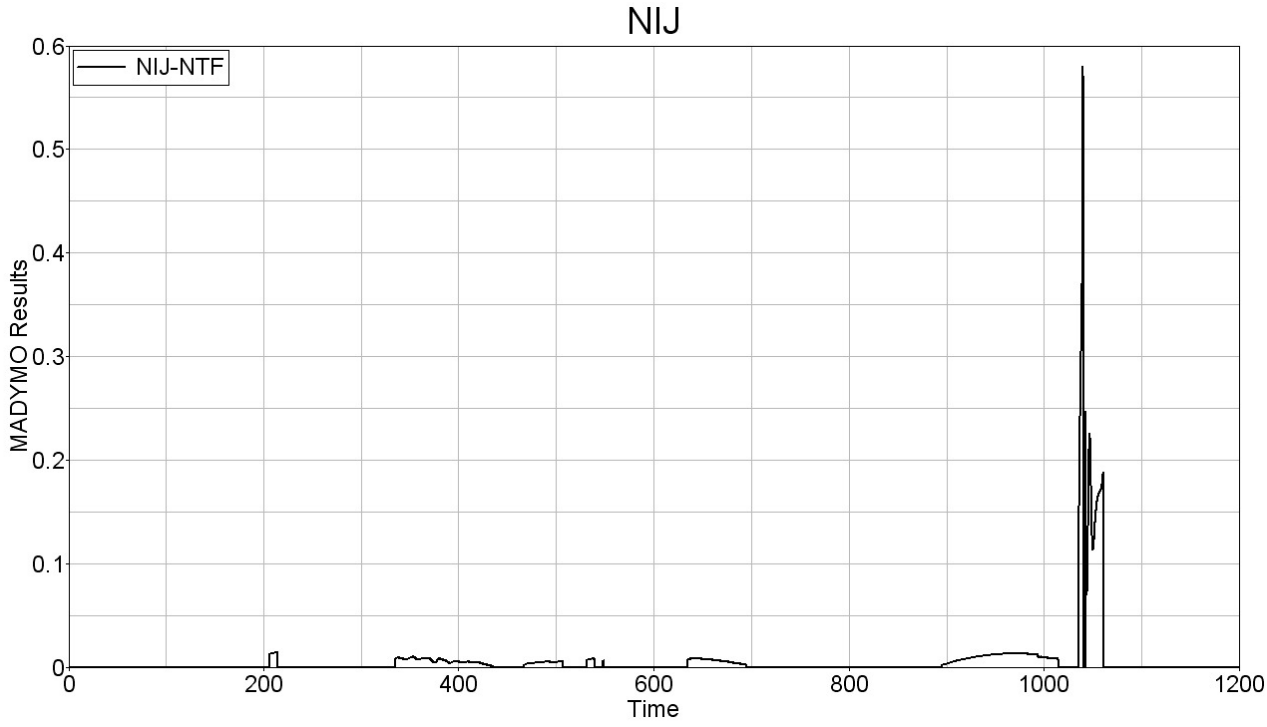

Figure 5.16 Case 5 Resultant Head Acceleration with 0.5 Coefficient of friction.

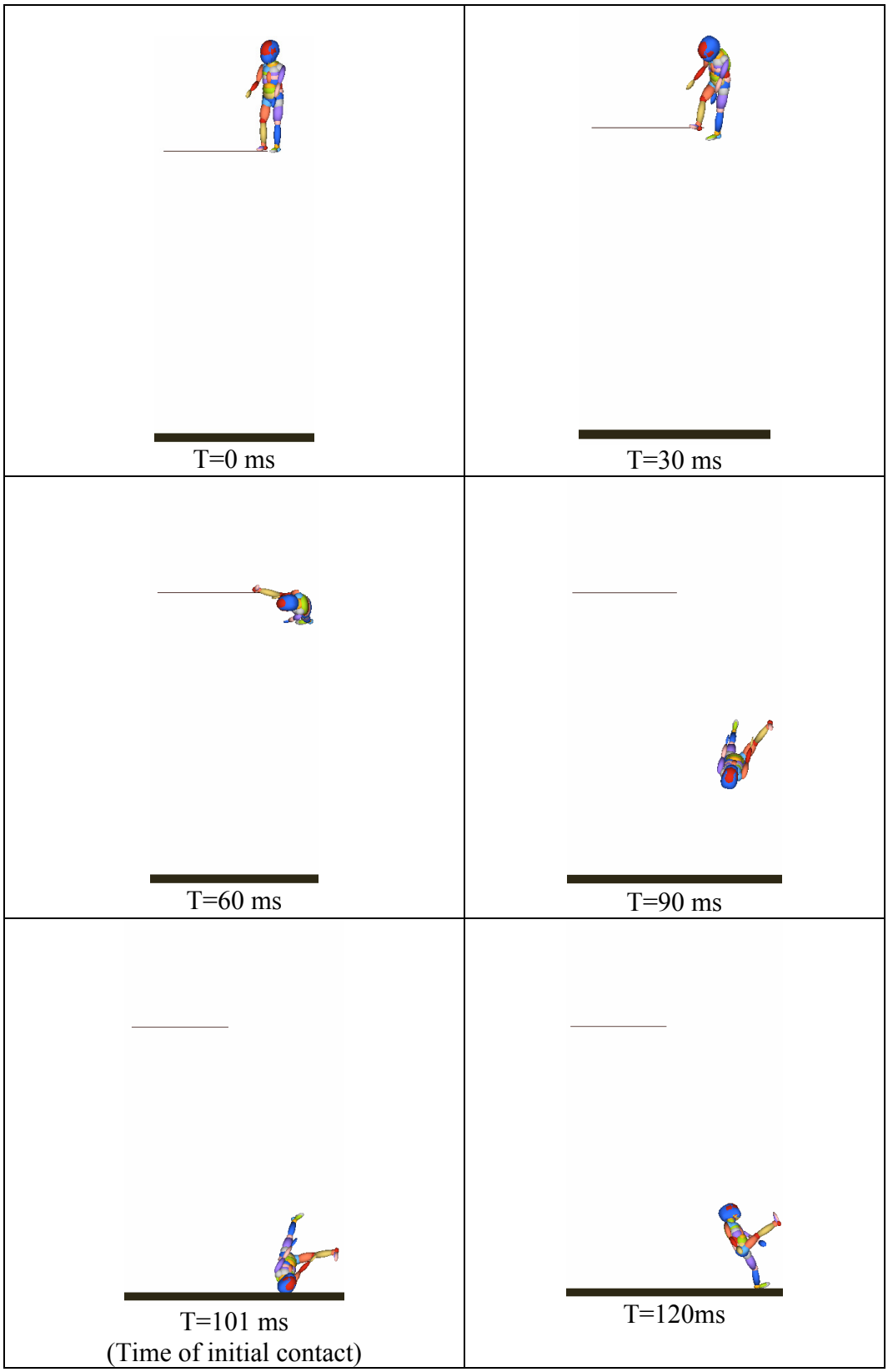

Figure 5.17 Case 6 Kinematics Analysis of with the 0.6 coefficient of friction.

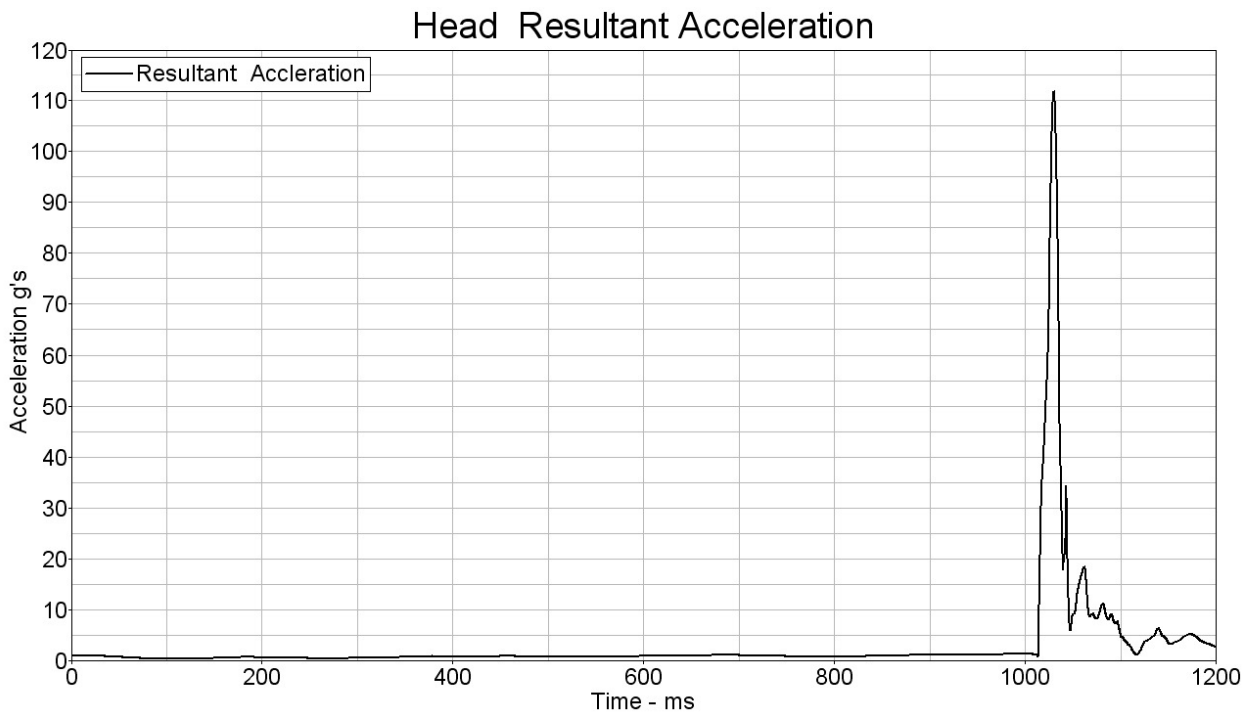

Figure 5.18 Case 6 Resultant Acceleration of the Head with 0.6 Coefficient of friction.

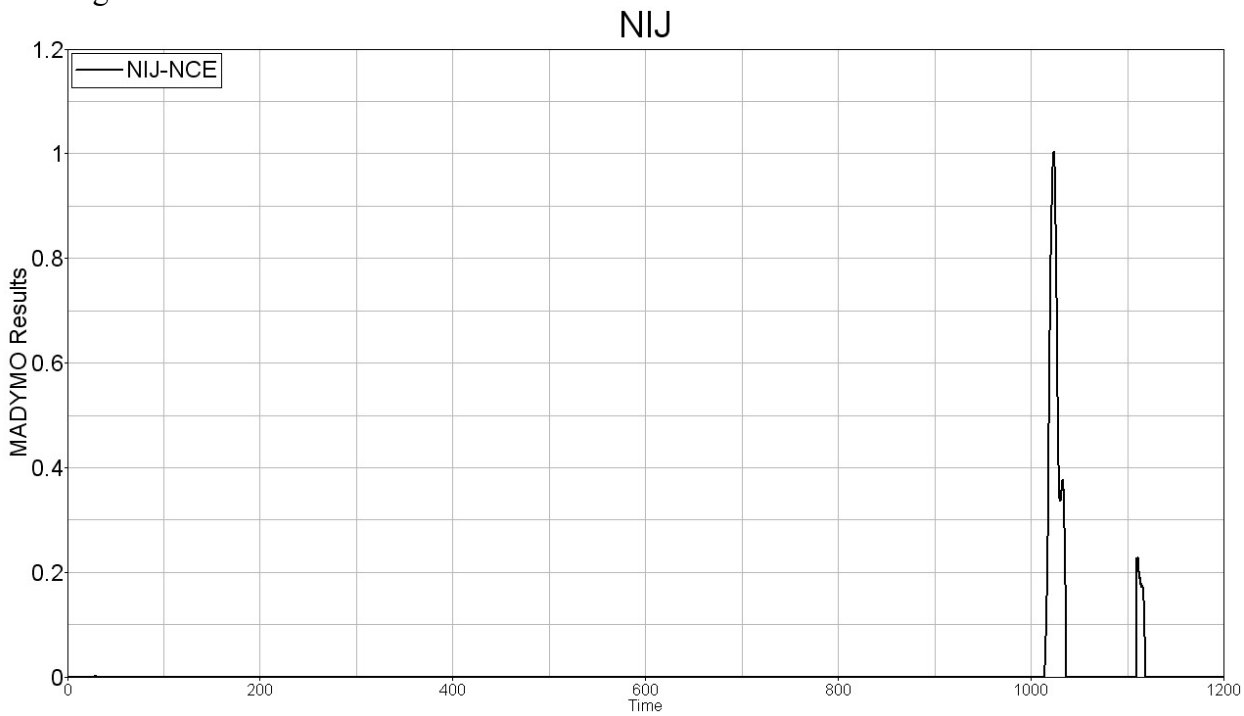

Figure 5.19 Case 6 Neck injury criteria NIJ with 0.6 Coefficient of friction.

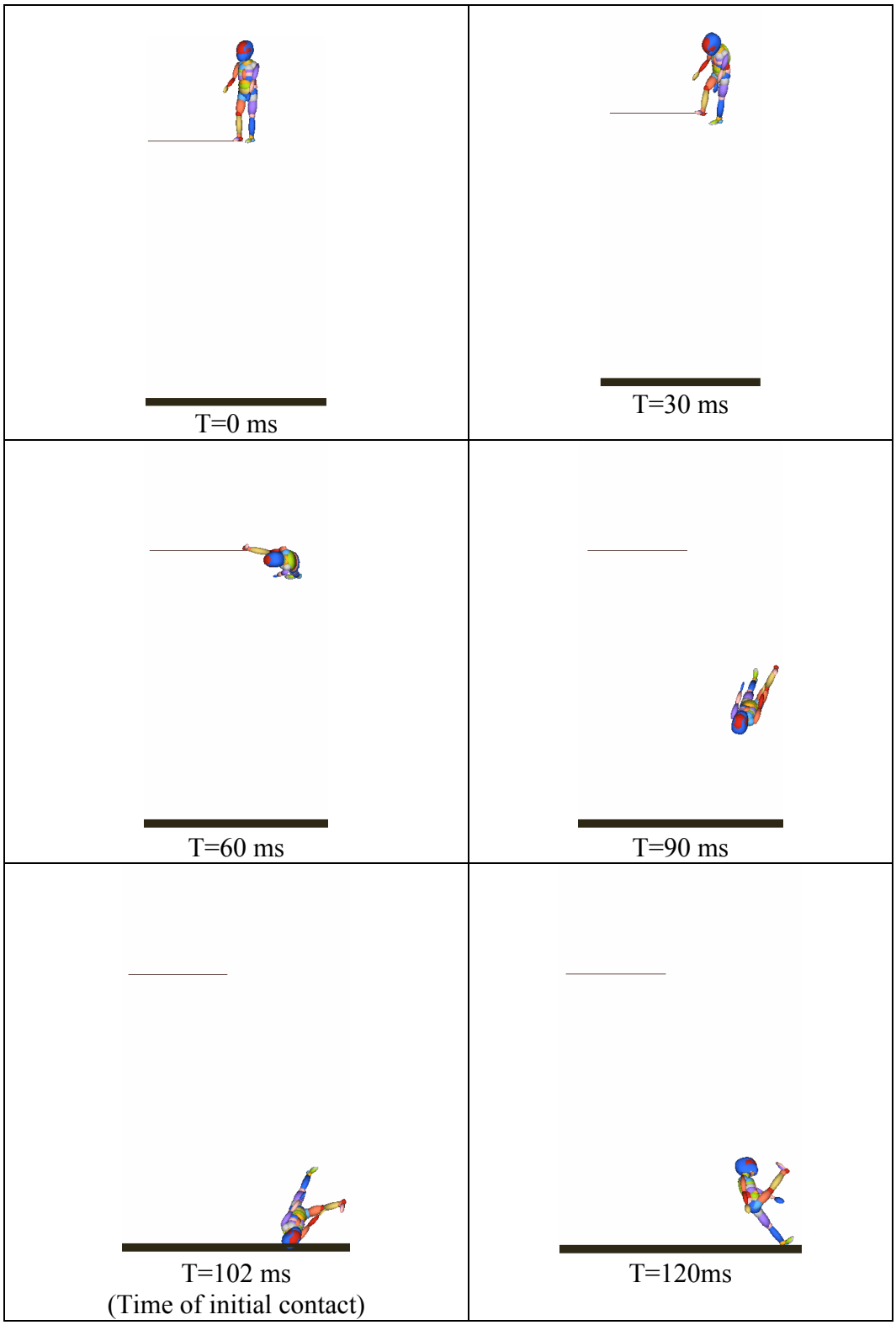

Figure 5.20 Case 7 Kinematics Analysis of with the 0 .7 Coefficient of friction.

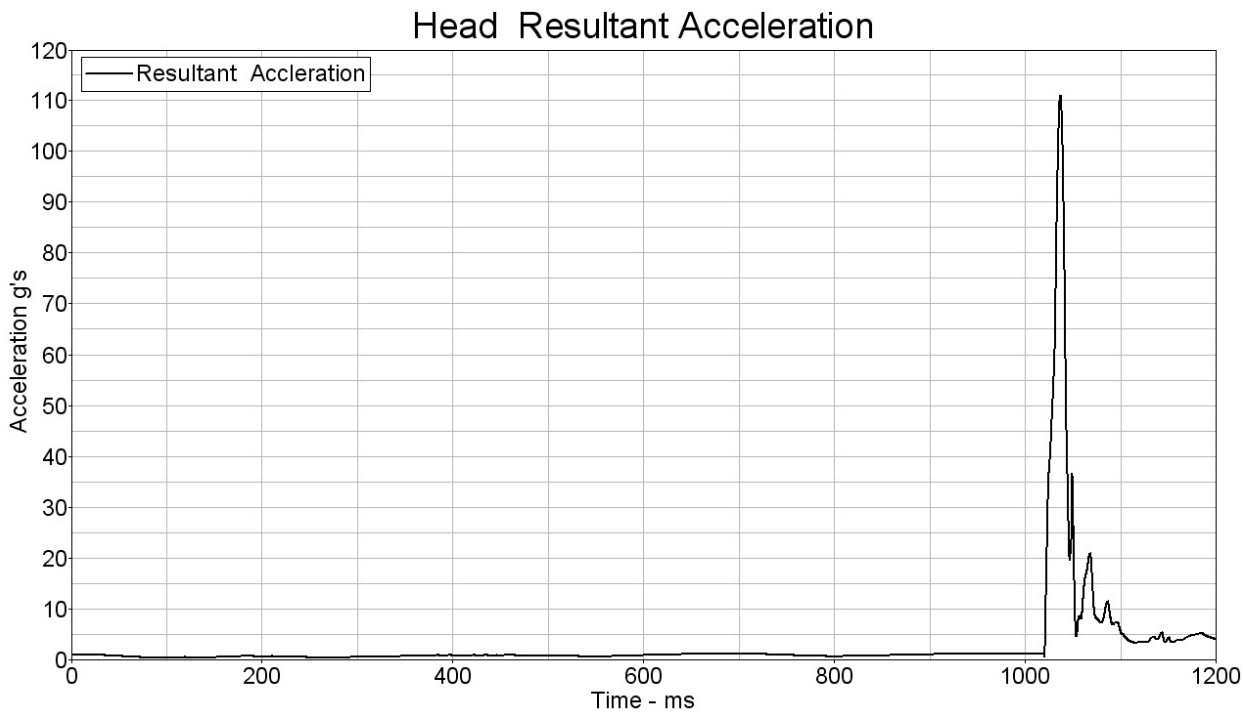

Figure 5.21 Case 7 Resultant Acceleration of the Head with 0.7 Coefficient of friction.

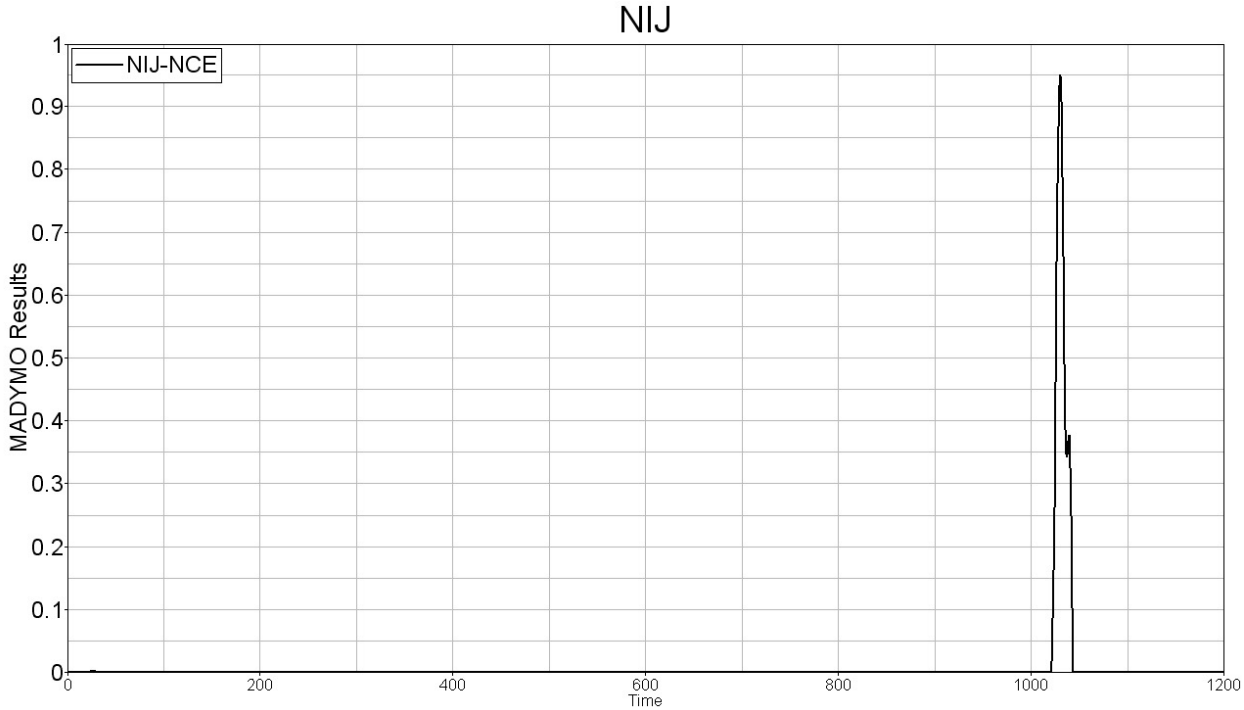

Figure 5.22 Case 7 Neck injury criteria NIJ with 0.7 Coefficient of friction.

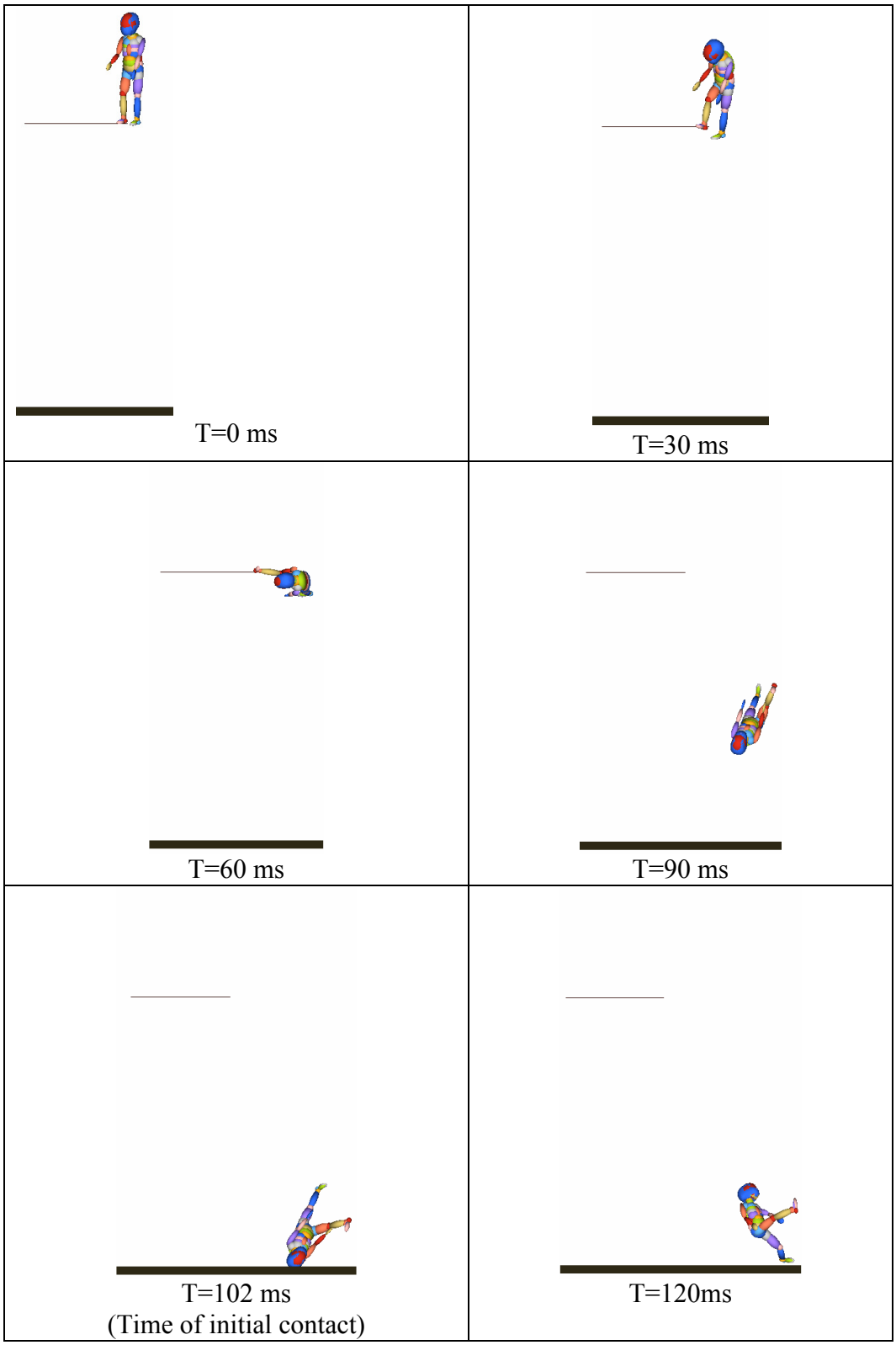

Figure 5.23 Case 8 Kinematics Analysis of with the 0.8 Coefficient of friction.

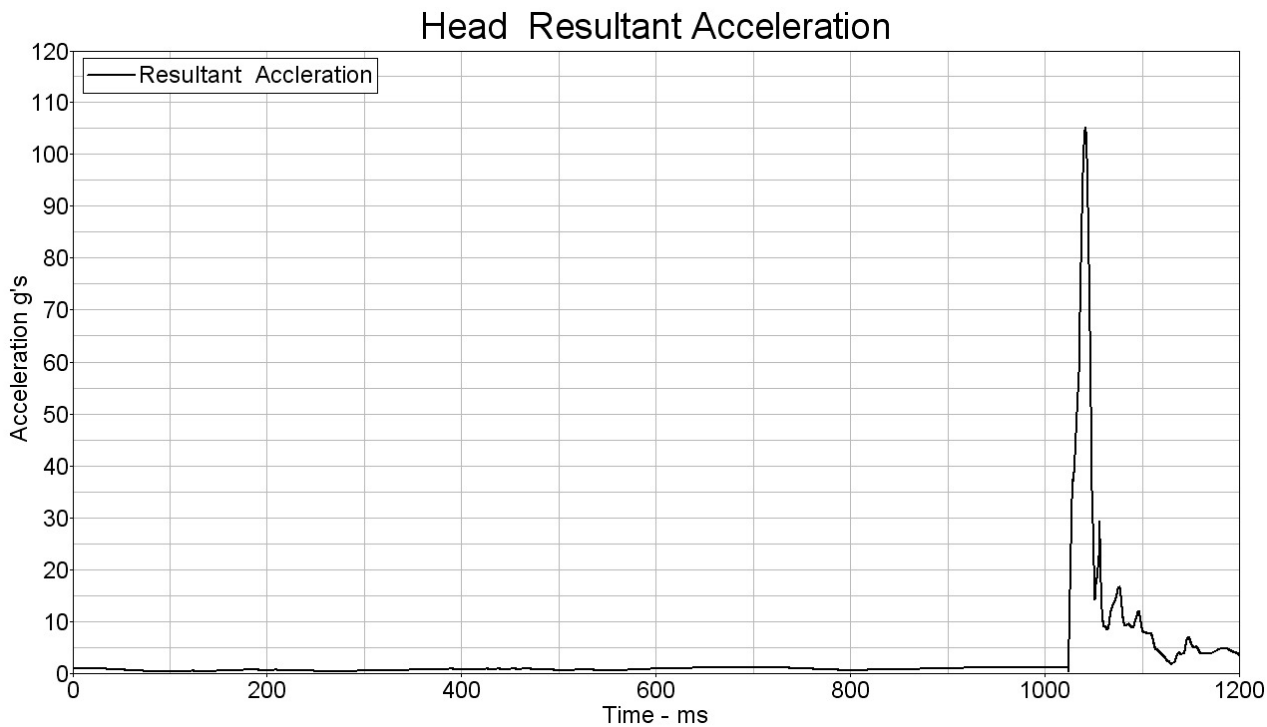

Figure 5.24 Case 8 Resultant Acceleration of the Head with 0.8 Coefficient of friction.

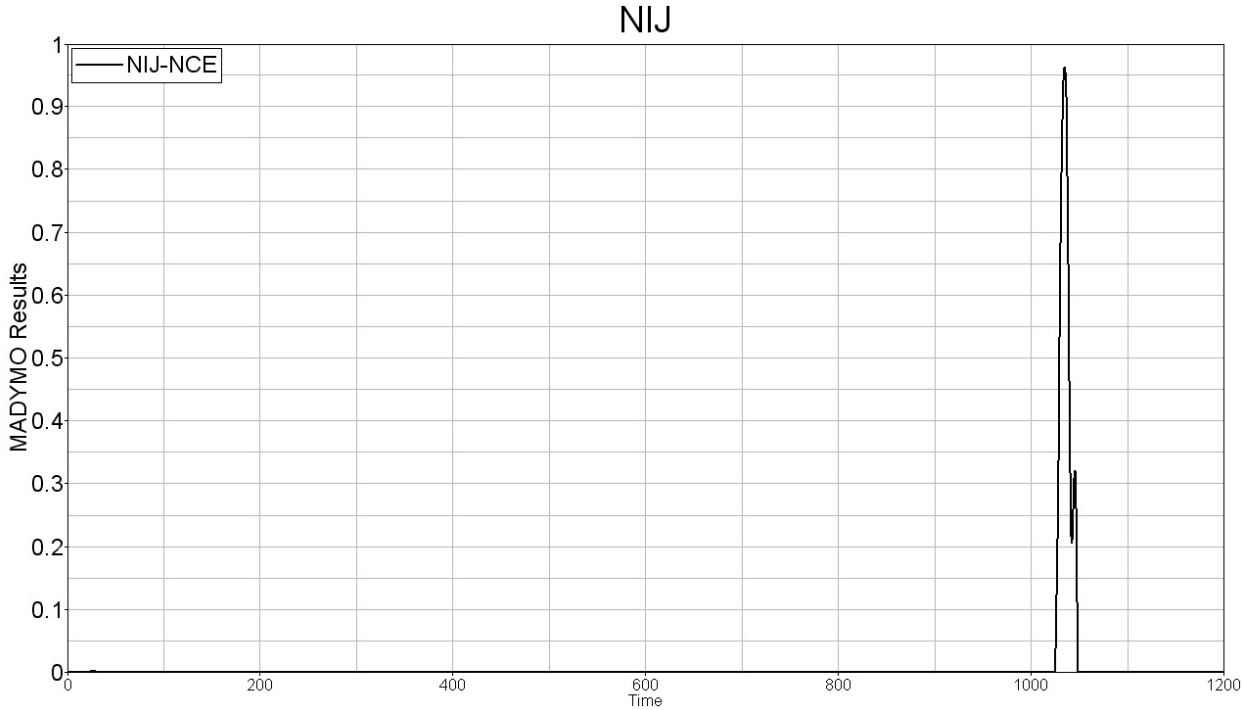

Figure 5.25 Case 8 Neck injury criteria NIJ with 0.8 Coefficient of friction.

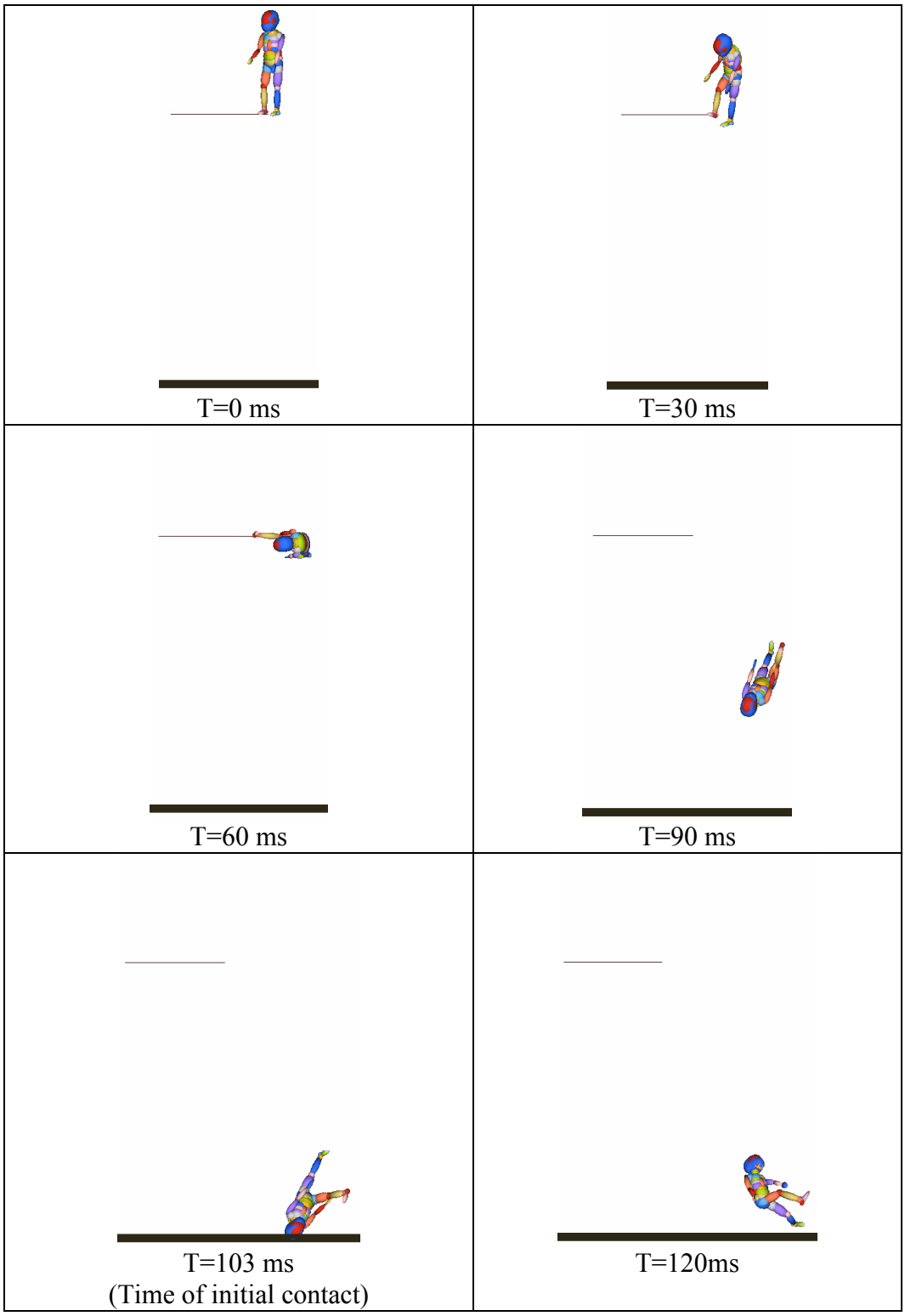

Figure 5.26 Case 9 Kinematics Analysis of with the 0.9 Coefficient of friction.

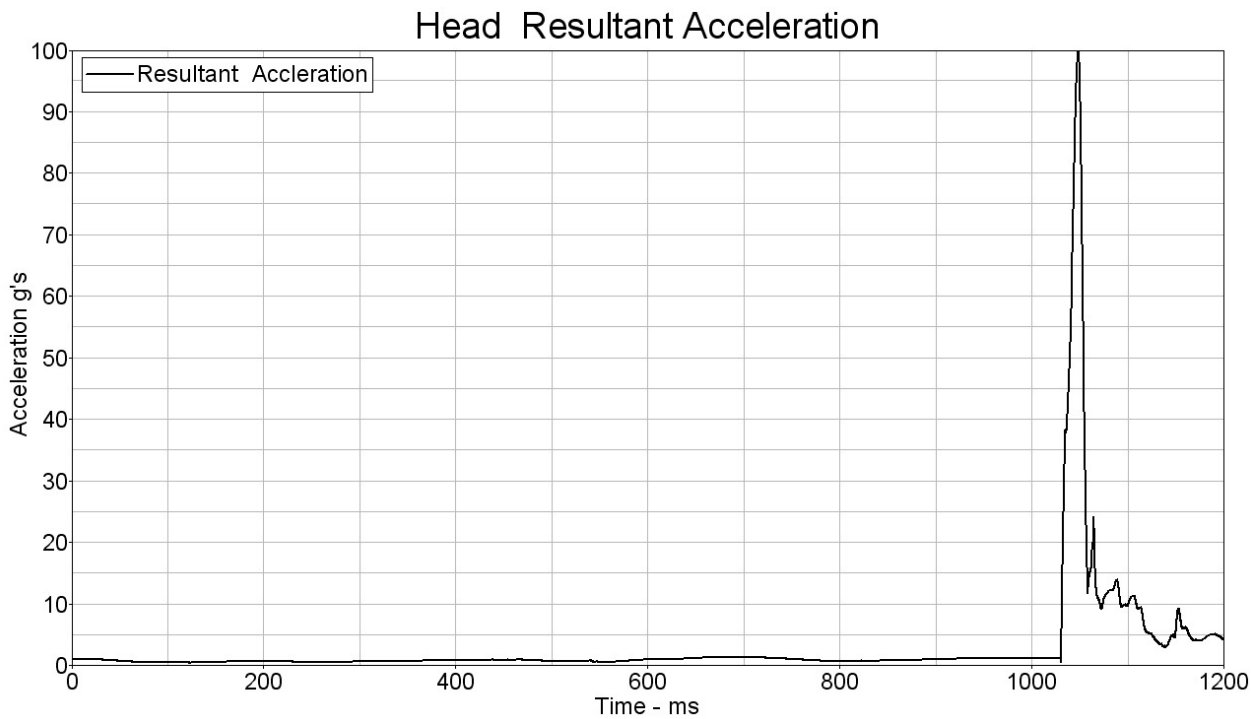

Figure 5.27 Case 9 Resultant Acceleration of the Head with 0.9 Coefficient of friction.

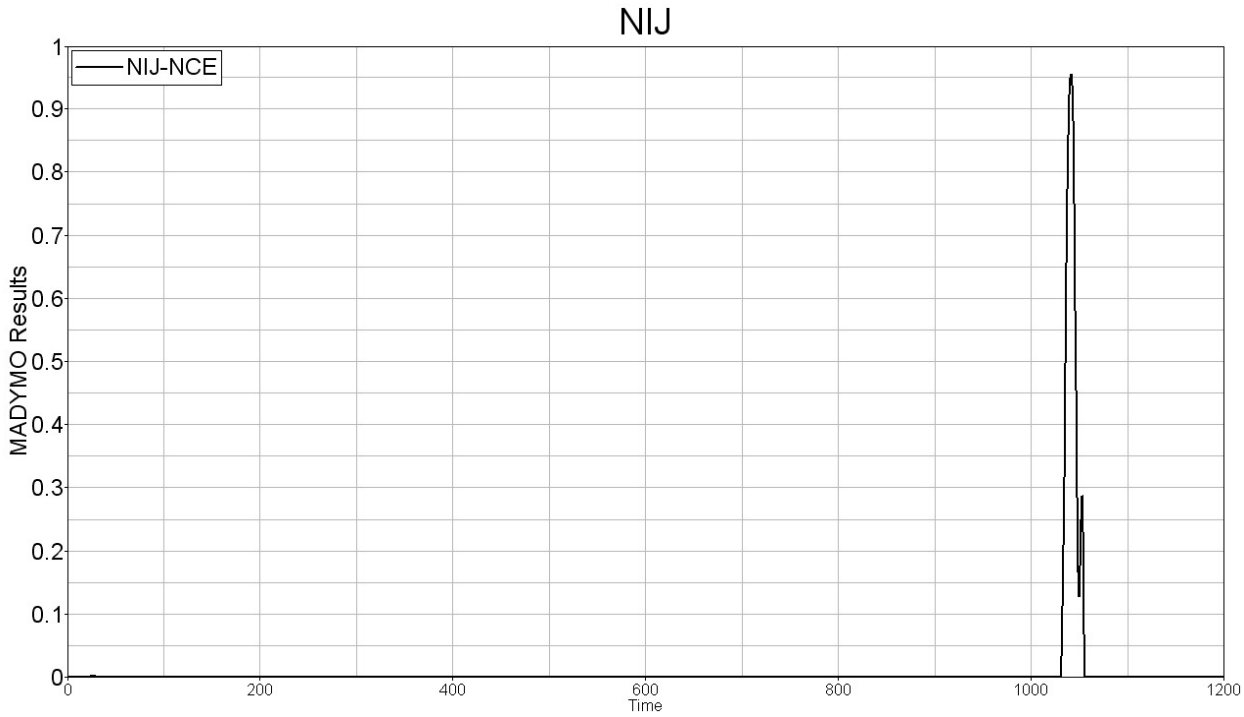

Figure 5.28 Case 9 Neck injury criteria NIJ with 0.9 Coefficient of friction.

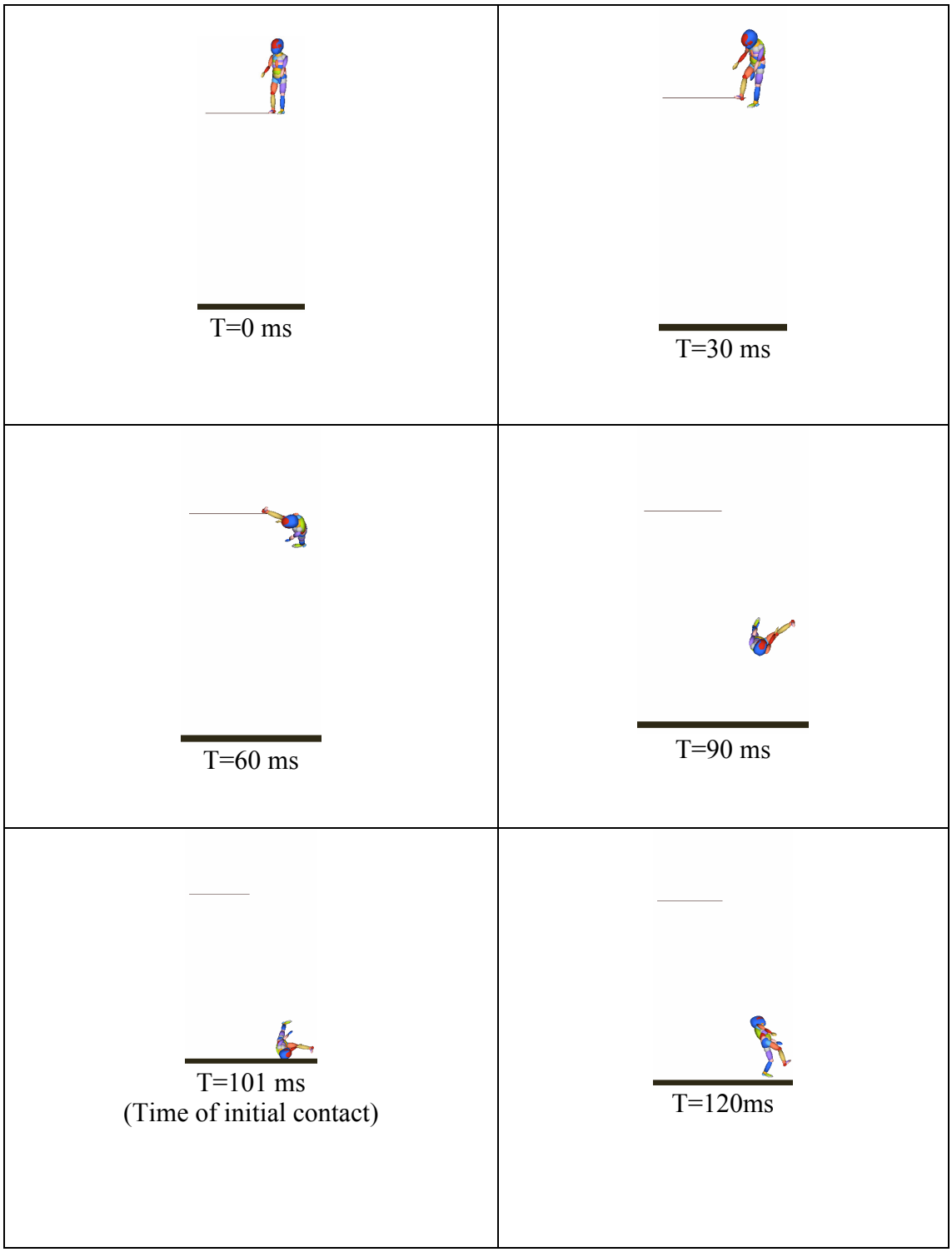

Figure 5.29 Case 10 Kinematics Analysis of with the 1.0 Coefficient of friction.

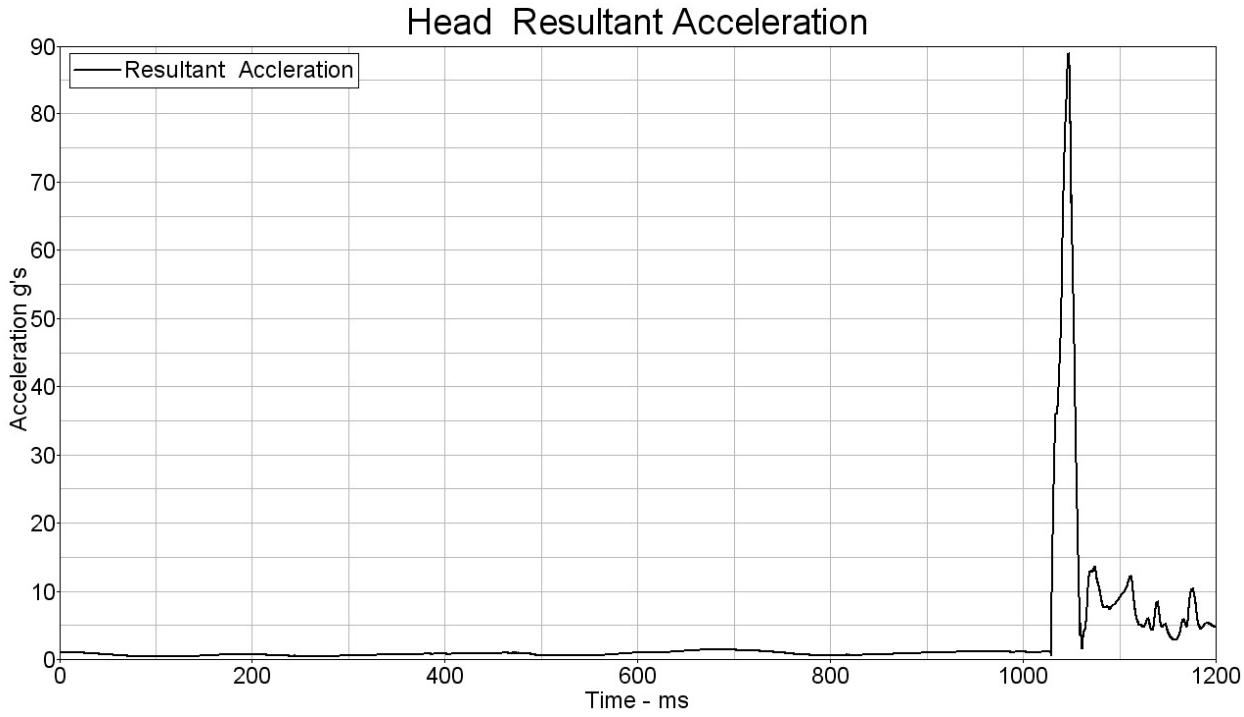

Figure 5.30 Case 10 Resultant Acceleration of the Head with 1.0 Coefficient of friction.

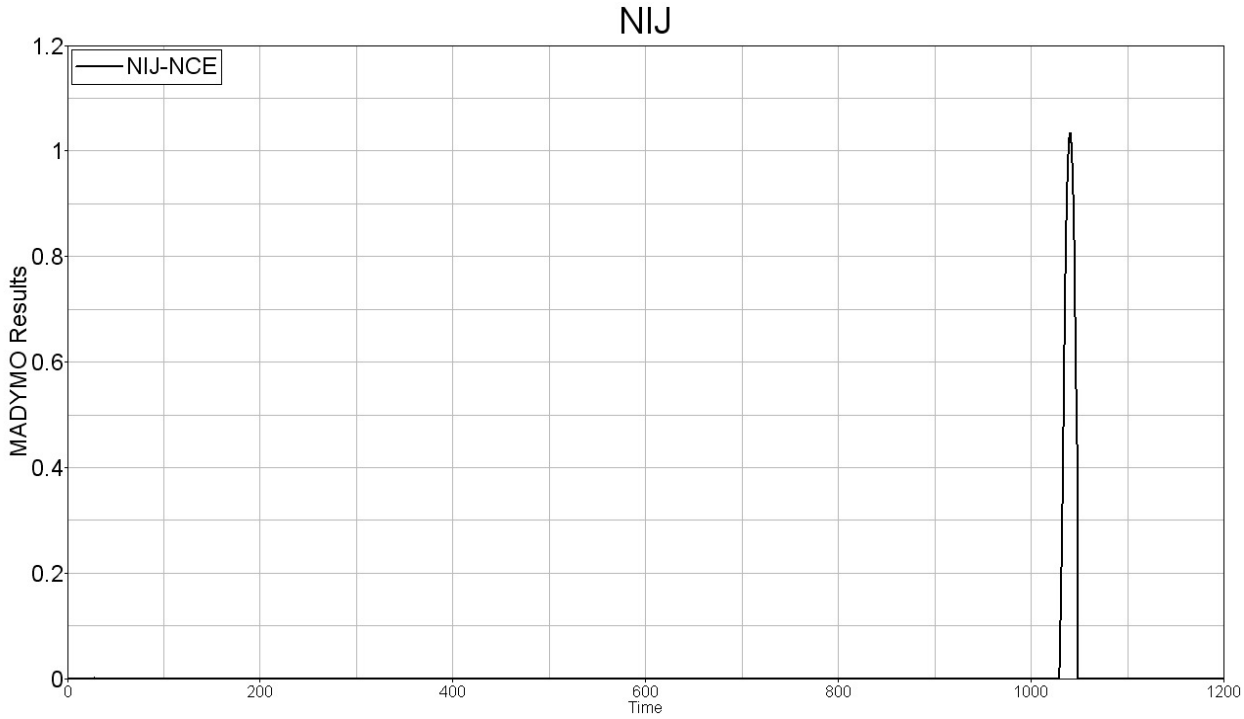

Figure 5.31 Case 10 Neck injury criteria NIJ with 1.0 Coefficient of friction.

Table 5.2 Case 1 thru 10 HIC 15, HIC 36, ΔT's and Nij.

Case Number	Coefficient of Friction	HIC15	ΔT(ms)	HIC36	ΔT(ms)	Nij
1	0.1	309	15	411	25	2.02
2	0.2	951	15	963	16.6	1.09
3	0.3	940	15	948	16.2	0.93
4	0.4	809	15	818	16.7	0.97
5	0.5	1264	12.8	1264	12.8	0.58
6	0.6	968	13.9	968	13.9	1.00
7	0.7	943	13.6	943	13.6	0.95
8	0.8	867	14.9	867	15	0.96
9	0.9	804	15	804	15.7	0.95
10	1	601	14.9	606	17.1	1.03

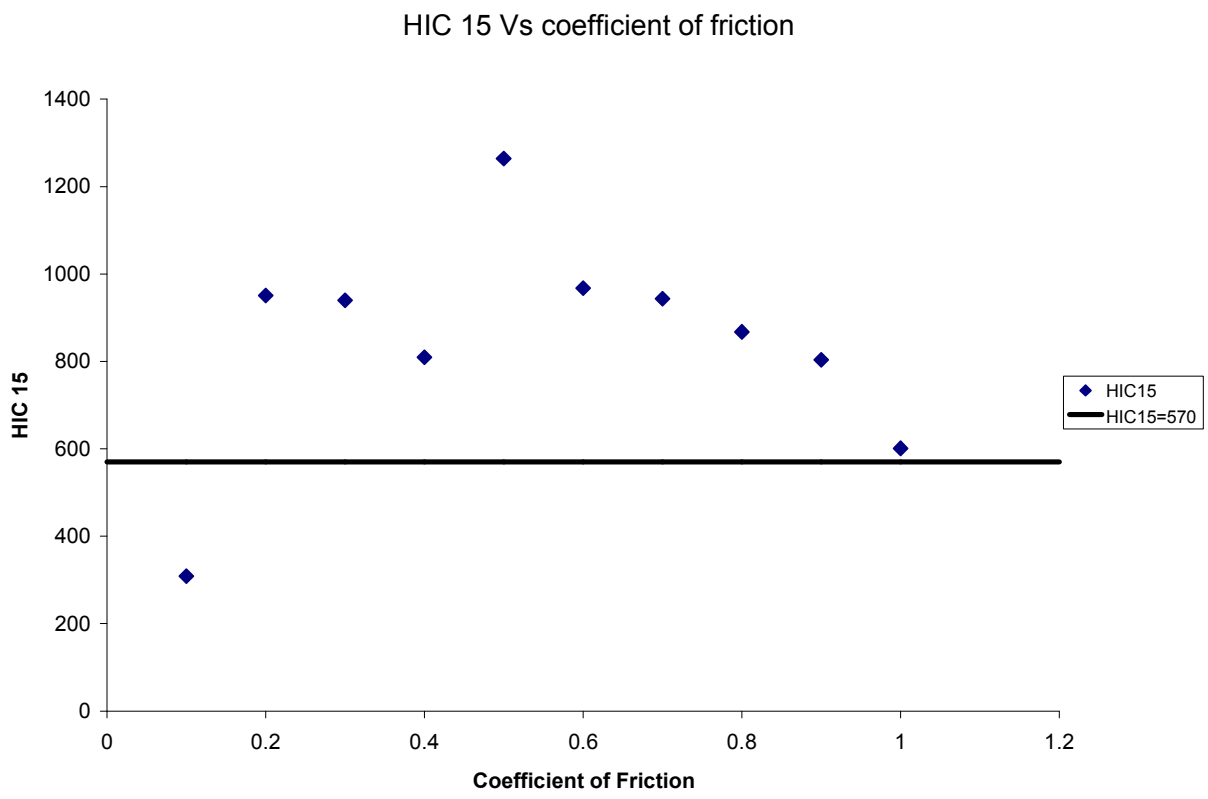

Fig 5.32 HIC15 with Coefficient of friction.

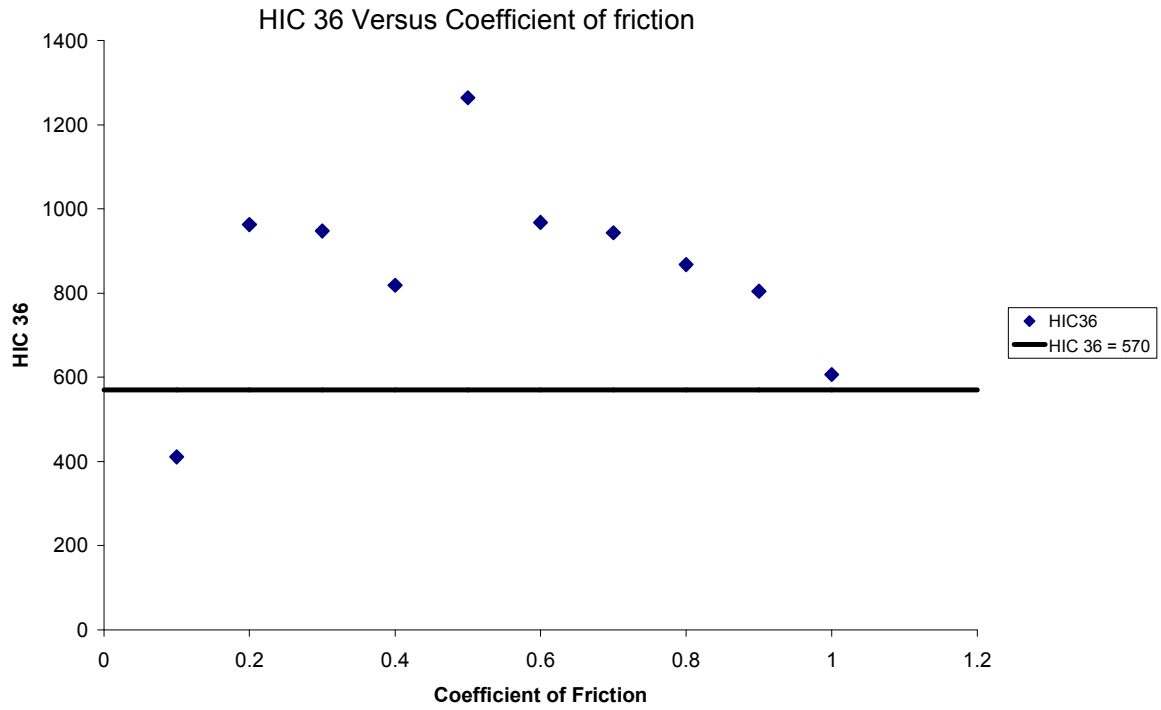

Fig 5.33 HIC36 with Coefficient of friction.

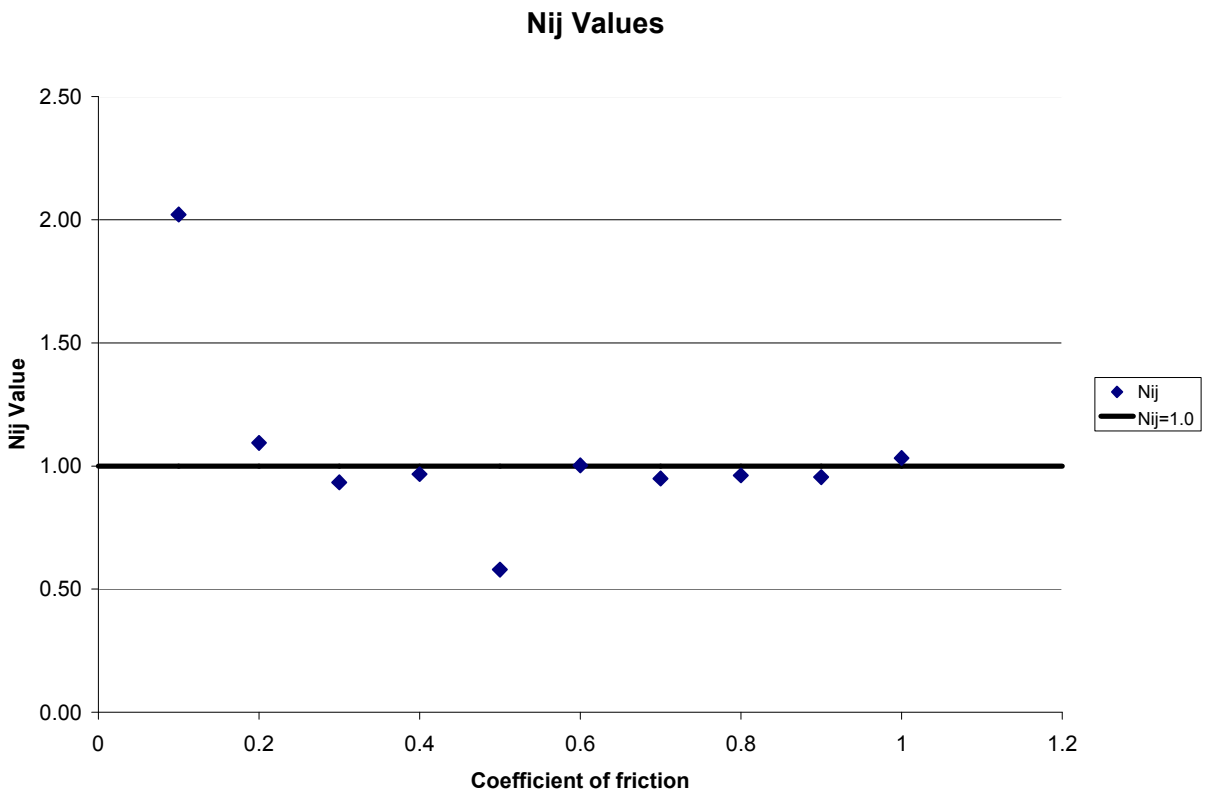

Fig 5.34 Nij with Coefficient of friction.

5.4 Safer fall height for HIC15 and Nij crieteria.

The results from the stair height show the severe likelihood of the injury from the fall due to the values of HIC 15 and Nij exceeding the pass and fail criteria. In cases where the HIC is within the pass limits, the accompanying Nij values will exceed the pass limit or vice-versa. In the interest of finding a height which will pass criteria for both HIC15 and the Nij, the height of the top step was gradually lowered to find a height where both values will pass. The case 5 with .5 friction coefficient was used as the basis for this simulation. The height of the step was gradually lowered and the acceptable height for this to occur was found to be at 1.65m and the same conditions and assumptions.

Height of the step = 1.65m

HIC 15 = 568

Nij =.038

Fig 5.35 shows the kinematics analysis of the fall from and Figure 5.36 thru 5.37 shows the results for the resultant head acceleration and Nij values. The Nij value graphs show the maximum value for NIJ-NTE (Tension-Extension) loading condition.

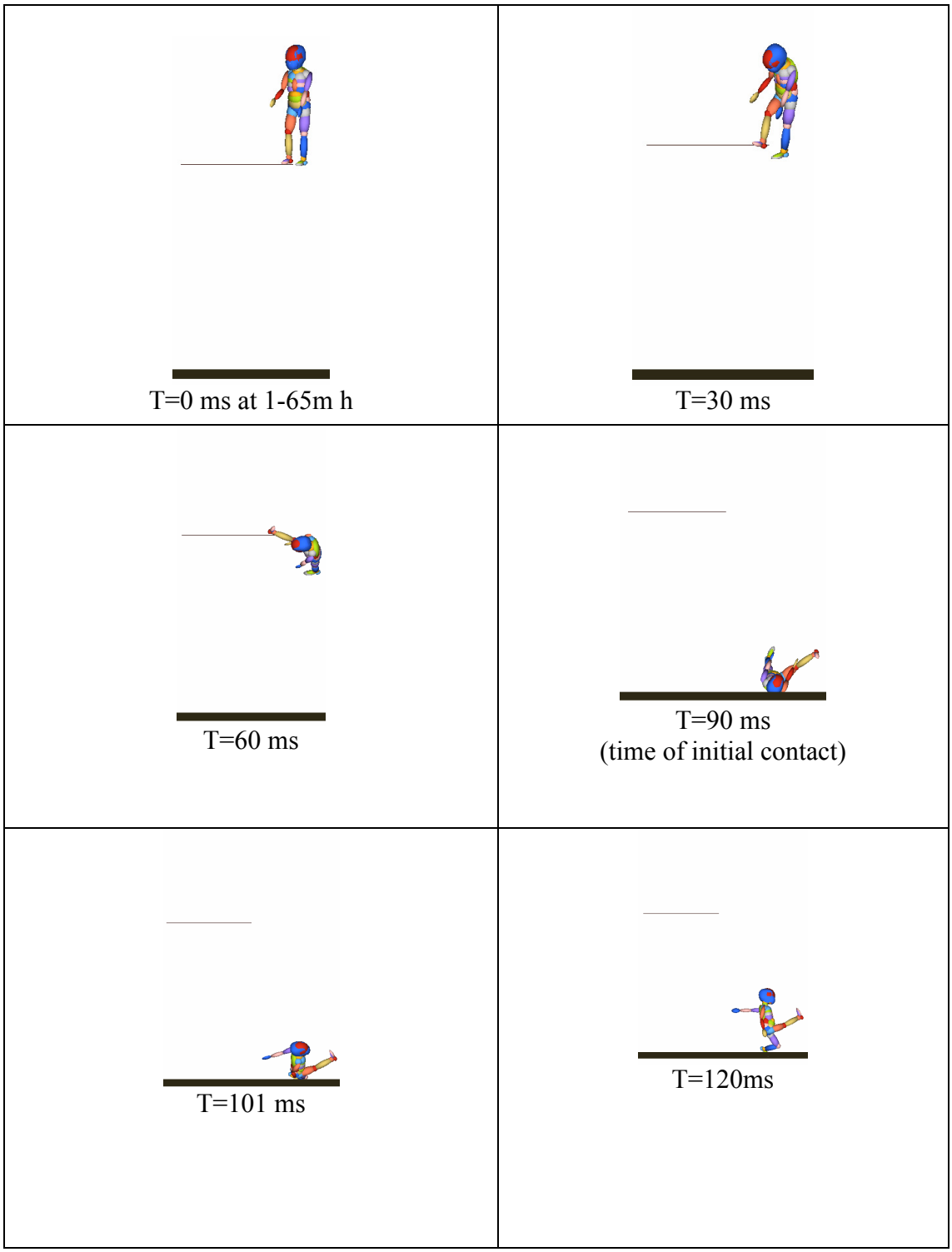

Fig. 5.35 Kinematics Analysis with the 0.5 Coefficient of friction at 1.65 m height.

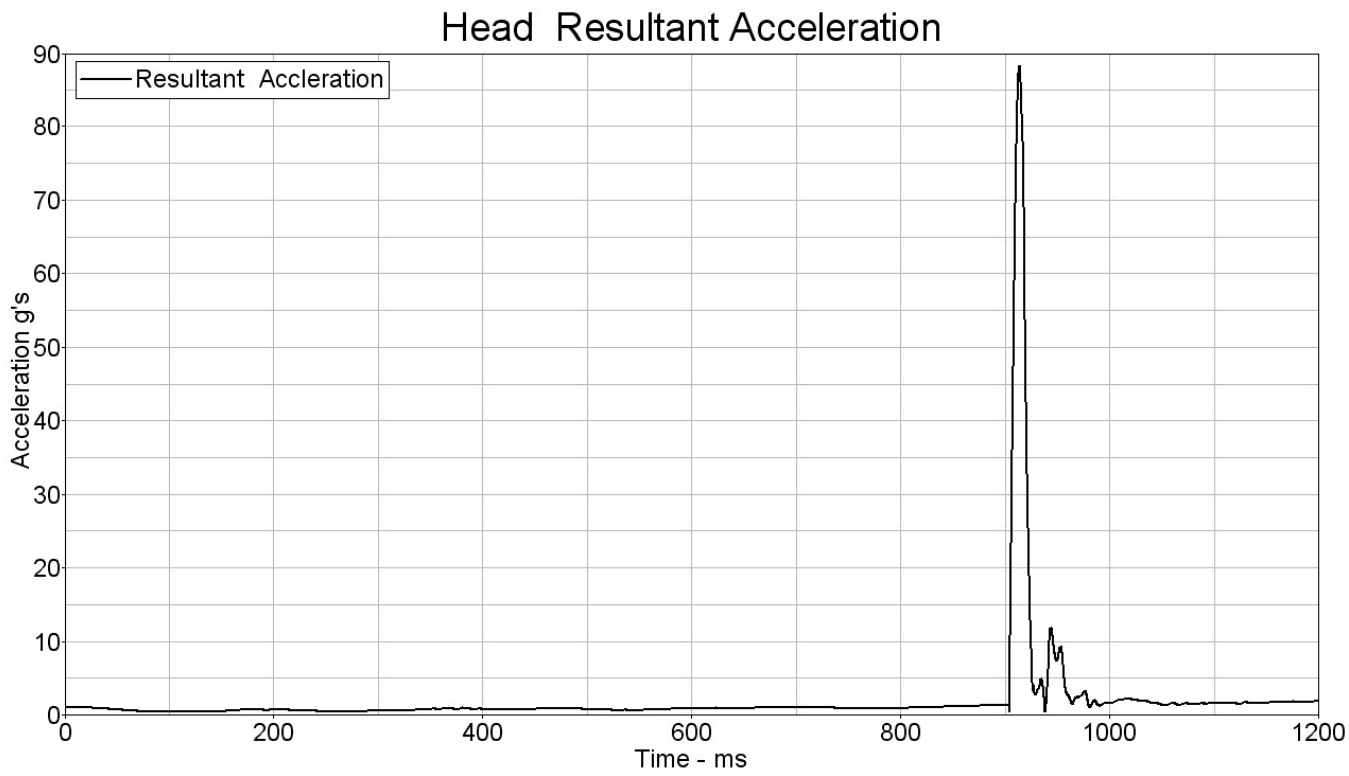

Fig 5.36 Resultant Acceleration of the Head with 0.5 Coefficient of friction at 1.65 m height.

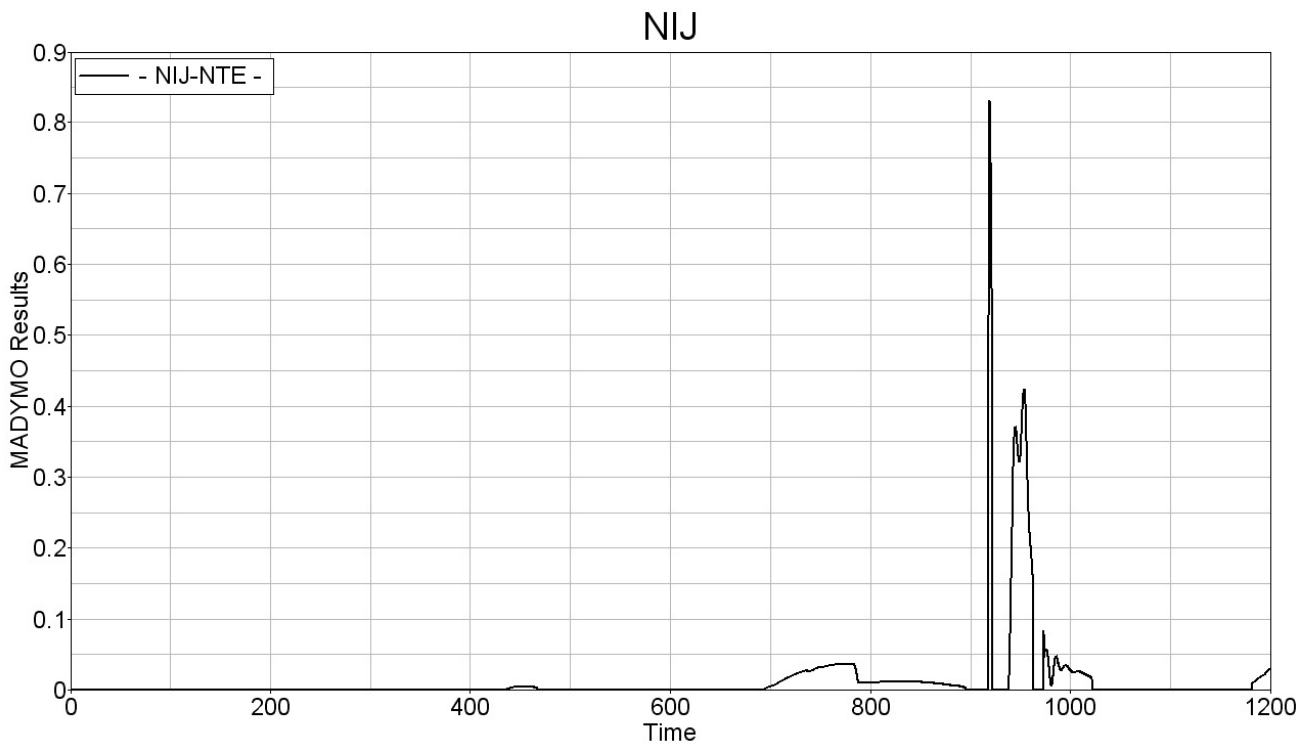

Fig 5.37 Neck injury criteria NIJ with 0.5 Coefficient of friction at 1.65 m height.

CHAPTER 6

CONCLUSIONS AND FUTURE WORK

6.1 Conclusions

The main purpose of this research work was to study the injury response to the human head and neck. This study, although not validated provides reference values for such an injury. The kinematics of the dummy during the fall and injury parameters shows serious injury risk. The following conclusions have been drawn after investigating the injury parameters of the dummy:

1. The HIC 15 value indicates that a serious injury can occur to the child from such a fall, with all the assumptions being correct.
2. The coefficient of friction closer to 1 indicates that injuries can be less severe.
3. The Nij values are higher when friction is low, but as friction increases the Nij stays near desirable levels.
4. This study shows that injuries to the head and the neck depend on the kinematics of the dummy during the fall, which is affected by the initial orientation of the dummy.
5. The coefficient of friction at the top step and the dummy's feet will also affect the kinematics of the fall and the injury results.
6. The injury results will also be affected by the height of the fall to the ground, dummy interface and ground properties.
7. This type of fall, with applied condition, is unsafe for a 3 year old child for heights greater than 1.65m.
8. In conclusion, it can be said that a higher coefficient of friction at the top step decreases the severity of the injury to the head and neck.

6.2 Future Work

The following recommendations can be made out of this work:

1. There is further study that can be to done on different types of dummy orientations and friction coefficients at the ground and the top step of stairs.
2. This study can be validated with a test with actual or similar conditions.
3. There were a lot of assumptions made in this study. Further study could rely on hard data in specific conditions.
4. There is further study that can be done on a child falling under the hand railing from the middle of the stairs.
5. Further work could also be done from different heights, as well as with dummies falling backwards on the staircase.

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