A Study of Occupant Dynamics
in Various Crash Accident Scenarios

by
Ioannis Harizopoulos

Submitted to the Department of Electrical Engineering and Computer Science
in Partial Fulfillment of the Requirements for the Degree of
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Author

Department of Electrical Engineering and Computer Science
May 23, 1997

Certified by
B.C. Lesieutre
Thesis Supervisor

Accepted by
Arthur C. Smith
Chairman, Department Committee on Graduate Theses
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Abstract

This thesis discusses occupant dynamics during various car crash scenarios. The study is comprised of four core experiments which investigate in some depth various elements associated with the injury and behavior of an occupant in a car crash. Topics covered include: occupant distance and occupant linear velocity versus injury, advantages and disadvantages of variable airbag deployment time, and a comparative analysis of a freebody displacement prediction algorithm versus a multibody computer simulation. The MADYMO finite element and multibody computer simulation module was used as the means to evaluate occupant, airbag and structure performance and as the basis for identifying and analyzing key parameters that affect occupant dynamics and occupant injury. The findings of the study are used to propose a smart airbag system which promises to reduce inflation-induced injuries associated with airbag deployment.

Thesis Supervisor: Bernard Lesieutre
Title: Assistant Professor
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My transition from analog circuit design into the world of occupant dynamics and car accidents was one of the biggest challenges. I would have never accomplished this work if it were not for Professor Lesieutre who decided to supervise an unusual thesis, and Antonis Eleftheriou from the Physics department who helped revive my rusty physics skills.

In addition, I would like to acknowledge roommates, friends and members of the MIT Pillota club for their support and encouragement.

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Chapter 1

Introduction

1.1 Background

Recently, there has been an intensive effort within the automotive industry to improve the protection level of occupants throughout vehicle accidents. These efforts stem from increasingly demanding federal standards for occupant protection, emerging technologies which allow for more advanced restraint systems and an increasing public awareness of the dangers associated with car crashes.

Much controversy surrounds the use of passive inflatory restraining systems (airbags). Field experience indicates that children, small occupants and infants in rear facing infant seats can find themselves dangerously close to the airbag at the time of deployment. In the past
three years, 36 fatalities in car accidents are attributed to airbag inflation. Federal Standards on the other hand, require that all US passenger cars are equipped with a passive restraint system for both passenger and driver. In an attempt to reduce occupant exposure to inflation induced injuries, the National Highway Traffic Safety Administration (NHTSA) has placed a stringent set of requirements on the automotive industry to produce an occupant friendly passive restraint system.

Subsequently, this has unveiled a set of new challenges for automotive systems designers. In order to develop an occupant friendly restraint system, one has to have a sound understanding of occupant behavior throughout the course of an accident. This study attempts to provide such a foundation by identifying the injury contribution of various accident related parameters. The injury is calculated based on the Head Injury Criterion and the 3ms G Injury Criterion described in Appendix I. The severity of the injury is assessed by the Federal Motor Vehicle Safety Standard (FMVSS) 208 for frontal collisions. By identifying and evaluating sources of injury, an automotive designer can take advantage of the restraining capabilities of various vehicle parameters (angle of the toe pan, windshield) while designing advanced restraint systems to reduce the injury contribution due to other accident parameters (vehicle speed, driver-steering wheel impact etc.).

This study concentrates entirely on frontal passenger car crashes. Frontal crashes account for up to 8,000 fatalities and 120,000 moderate to critical injuries (i.e. injuries of injury severity index larger than 2) and constitute by far the most important cause of occupant injury. The main dynamic performance requirement in FMVSS 208 involves successful crash testing into a rigid barrier with a 50th percentile adult dummy at all speeds up to 48 kilometers per hour (30 mph) at all angles between perpendicular and 30 degrees to either side of perpendicular. The tests can be run both with the dummy being unbelted or belted. “Successful” crash testing requires that the dummy chest deceleration is below 60 G's, the dummy Head Injury Criterion (HIC) is below 1,000 and the dummy femur loads are below 10,000 Newtons. This is why the majority of the simulations in this study involve an angle perpendicular at the rigid obstacle and a vehicle speed of 49 km/h.

Occupant behavior and injury contribution from vehicle-driver and airbag-driver interactions are evaluated by means of computer
The subject of occupant performance in car crash accidents is very broad and very complex. To maintain focus, this study concentrates on four very specific topics. The analysis of these four fairly narrow topics unveils several findings about occupant performance and injury in frontal crashes. Some of these aspects of occupant performance are well known within the automotive industry, while others are completely original. These findings are used to propose a series of restraint systems which promise to reduce driver injury in frontal crashes. The four main topics along with their respective motivations are presented below.

**Evaluation and comparative analysis of the freebody algorithm and MADYMO**

Airbag system designers need information about the displacement of the occupant in the first few milliseconds of a crash accident. Several key decisions are based on this information such as triggering time, gas flow, airbag size etc. Many systems designers use the freebody model for occupant displacement as a worst case algorithm to assess occupant position as a function of time. The freebody displacement is calculated by integrating twice the filtered acceleration pulse of a car crash. While it is an effective method, it does not provide information about the position of various body parts (i.e. head, chest etc.). As designers seek to optimize current airbag deployment algorithms and design new restraint systems, they are interested in more accurate predictions of occupant displacement and relative head and chest displacement. In this study, the freebody algorithm is compared to the MADYMO output of a more elaborate model which incorporates friction and bio-mechanics. The behavior of the head and the chest are analyzed and the freebody algorithm is evaluated.

**Correlation between driver linear impact velocity and upper body injury**

In their efforts to design “smart” airbags, several system designers are faced with questions regarding upper body injury and the velocity of the driver. This experiment will attempt to answer how occupant size, occupant velocity and the vehicle interior affect upper body injury.
Correlation between driver distance and upper body injury

Most airbag related fatalities occur because occupants find themselves dangerously close to the airbag at the time of deployment. Previous work involving computer simulation and live animals indicate that distances of less than four inches from the airbag are almost always fatal when airbag deployment occurs, while distances of more than twelve inches are mostly non-fatal. It seems intuitive if the occupant is less than four inches away from the airbag (the steering wheel in the case of the driver) there should be no deployment. However, how much injury will the driver suffer if there is no deployment? How much injury will he or she suffer if the distance is six or eight inches and there is no deployment? The goal of this experiment is to correlate driver distance and upper body injury and answer questions along these lines. The analysis will attempt to define a range of occupant distances for which non deployment will yield acceptable levels of injury. It will also attempt to determine how distance and injury depend on a number of accident related parameters.

Quantitative analysis of the benefits of variable airbag deployment time.

A general rule used to determine the trigger time for an airbag is the “t_{125 \text{mm}} - 30” rule. Airbag triggering has to be determined by the time it takes for the occupant to move 125mm minus the time it takes for the airbag to inflate, which is around 30ms. Depending on the structure and the size of the vehicle, the trigger time can range from 6ms to 30ms from the beginning of the crash. The Sensing and Diagnostic Module (SDM) --which is responsible for the deployment decision-- requires some time in order to identify the severity of the crash. This places severe limitations on the trigger time of the airbag. As radar based anticipatory sensing becomes available, systems designers have the option of detecting the severity of an accident before actual contact occurs. This introduces the possibility of earlier deployment times. This experiment will attempt to quantify the benefits of varying the deployment time in terms of the driver’s upper body injury. We hope to establish a correlation --if any-- between the deployment time and the injury sustained.
To further reduce complexity and maintain focus, all simulations are concentrated on the unbelted driver of a typical mid-size passenger car. To cover a range of drivers, a 5th percentile female and a 50th and 95th percentile male dummies are used. Most simulations are subjected to an acceleration pulse of a 49 km/h frontal collision with a rigid barrier.

1.2 Organization of the thesis

- Chapter 2 discusses the modeling techniques used throughout the study. It presents background information about MADYMO and the way it handles kinematics, surface interactions, forces etc.
- Chapter 3 presents the setup of the various simulations. It discusses the goals of the experiments and the motivations behind each experiment setup.
- Chapter 4 contains some of the simulation results.
- Chapter 5 presents a detailed analysis of the simulation results organized by experiment. In many cases, physics models are used to substantiate the observations of the simulations. Several conclusions are reached and errors are discussed.
- Chapter 6 uses some of the key findings of the analysis to propose a series of applications which promise to reduce driver injury in vehicle crashes.
- Chapter 7 outlines the conclusions of the study.
- Chapter 8 summarizes the methods, results and analysis of the study and provides suggestions for future work.
- Appendix I contains a brief description of the injury criteria used throughout the study.
- Appendix II lists a typical MADYMO file which implements the models presented in chapters 2 and 3.
Chapter 2

Methods

The results of this study rely heavily on the simulation capabilities of MADYMO and the tools used in conjunction with it. Therefore, an elaborate description of the simulation techniques is presented below.

2.1 MADYMO

MADYMO (MAthematical DYnamic MOdel) is an industry standard computer package developed by TNO, The Netherlands, which is used to simulate crash situations to a high degree of accuracy and to assess injuries sustained by potential victims. MADYMO was developed originally for studying occupant behavior during car crashes making it the most suitable solution for a cost effective and fast assessment of various
crash conditions. Although it is available in both 2D and 3D versions, the study utilized exclusively the 3D version. MADYMO combines in one simulation program the capabilities offered by multibody (for the simulation of the gross motion of systems of bodies connected by complicated kinematic joints) and finite element techniques (for the simulation of structural behavior).

The multibody algorithm yields the second derivatives of the degrees of freedom in explicit form. The number of computer operations is linear in the number of bodies in case all joints have the same number of degrees of freedom. This leads to an efficient algorithm for large systems of bodies. At the start of the integration the initial state of the systems of bodies has to be specified (initial conditions).

The finite element method divides the actual continuum into finite volumes, surfaces or line segments. The continuum is then analyzed as a complex system, composed of relatively simple elements where continuity should be ensured along all boundaries between elements. These elements are interconnected at a discrete number of points, the nodes. The initial nodal positions and velocities, the nodes corresponding to each element, the connectivity, as well as the element properties, i.e. the material behavior, must be specified at the start of the simulation.

The way the interaction between bodies and finite elements is modeled, allows the use of different time integration methods for the equations of motion of the finite element part and the multibody part. All used integration methods are conditionally stable and therefore limit the time step that can be used. To increase the efficiency of the entire analysis the finite element module is being sub-cycled with respect to the multibody module using a different constant time step for each module.

MADYMO offers a set of standard force models e.g. for belts, airbags and contacts of bodies with each other or with their surroundings. To create a MADYMO input data file the user first selects the number of multibody systems and finite element structures to be included in the simulation model. For instance, a simulation model can consist of one multibody system for a dummy, one for a deformable steering column and one for a child restraint system, and finite element structures for the driver, passenger airbag and the kneebolster. For crash dummies, standard databases are available. The characteristics of the dummies
used throughout this study are listed in Appendix I. Next, for each of
the multibody systems, the number of bodies and their configuration and
for each structure, the finite element mesh, the element types and the
material properties must be specified.

An input data file is then set up which specifies the configuration, the
mass distribution and the general properties of the multibody systems
(joint characteristics) and the finite element structures.

The acceleration field model calculates the forces at the centers of
gavity of bodies or finite elements due to a homogenous acceleration
field. This model is particularity useful for the simulation of the
acceleration forces on a vehicle occupant during an impact. It is not
necessary to apply the acceleration field to all bodies.

Planes and ellipsoids can be attached to a body to represent its shape.
These planes and ellipsoids are also used to model contact with other
bodies or with finite elements. The contact surfaces are of major
importance in the description of the interaction of the occupant with
the vehicle interior. The elastic contact forces, including hysteresis,
are a function of the penetration of the contact surfaces. In addition
to elastic forces, damping and friction can be specified.

Three types of massless spring-damper elements are available. The
Kelvin element is an uniaxial element which simulates a spring parallel
with a damper. The Maxwell element is an uniaxial element which
simulates a spring and a damper in series.

The final section of the input file deals with the output required from
the simulation. The output generated by MADYMO is specified through a
set of output control parameters. A large number of standard output
parameters is available, such as accelerations, forces, torques and
kinematic data. MADYMO offers in addition to standard output
quantities, the possibility to calculate injury criteria like femur and
tibia loads, Head Injury Criterion (HIC), Gadd Severity Index (GSI),
Thoracic Trauma Index (TTI) and Viscous Injury Response (VC). (See
Appendix I)

Results of the simulation are stored in a number of output files, which
are accessible by postprocessing programs. Programs are available for
the visualization of the kinematics, time histories and cross plots. A sample MADYMO data input file is included in Appendix II.

![Figure 2.1.1. Example of a MADYMO generated crash sequence](image)

Once a given crash situation has been modeled with the MADYMO package, it is relatively straightforward for users to determine how the scale of potential injuries can be reduced by introducing special safety features or by changing certain design parameters. This makes the MADYMO package an extremely useful tool for enhancing vehicle safety.

**Numerical Integration methods for the equations of motion**

The equations of motion form a system of coupled non-linear second order differential equations. These equations can be written as:

\[ q = g(q,\dot{q},t) \]

with initial values \( q_0 \) and \( \dot{q}_0 \).

\( q \) is a column matrix with the generalized coordinates, the joint degrees of freedom; \( \dot{q} \) and \( \ddot{q} \) are the first and second time derivatives of the generalized coordinates. The column matrix \( q \) contains \( m \) elements, corresponding with the \( m \) degrees of freedom of the model.
The equations are solved numerically. Three methods are available:

1. Modified Euler method with a fixed time step;
2. Runge-Kutta method with a fixed time step;

These are one-step explicit methods, i.e. solution at a time point \( t_{n+1} \) can be written explicitly in terms of the solution at the preceding time point \( t_n \). For most problems the error in the solution will reduce when the time step is decreased. In case one of the fixed time step methods is used the accuracy of the solution varies with different time steps.

In this study the Runge-Kutta method was used; the system of \( m \) second order differential equations is reduced to \( 2m \) first order differential equations. Introduce the column matrix \( x \) defined by:

\[
\begin{bmatrix}
\dot{q} \\
q
\end{bmatrix}
\]  

(2)

Using this substitution, equation (1) becomes:

\[
\begin{bmatrix}
\dot{q} \\
q
\end{bmatrix} = \begin{bmatrix}
g(q, q, t) \\
\dot{q}
\end{bmatrix} = f(x, t)
\]

(3)

with initial condition

\[
\begin{bmatrix}
\dot{q} \\
q
\end{bmatrix}(t_0) = \begin{bmatrix}
q_o \\
\dot{q}_o
\end{bmatrix}
\]

(4)

Equation (3) is integrated, resulting in solutions for \( q \) and \( \dot{q} \) in the next time point.

From the Runge-Kutta methods available the fourth order Runge-Kutta method with fixed step was preferred due to its accuracy, need for less computation power and its simplicity. The fourth order solution of equation (3) at the time point \( t_{n+1} = t_n + t_s \) can be written as:
\[
x_{n+1} = x_n + \frac{1}{6} t_s (k_1 + 2k_2 + 2k_3 + k_4)
\]

where \( t_s \) = fixed integration time step and

\[
k_1 = f(t_n, x_n)
\]
\[
k_2 = f(t_n + \frac{1}{2} t_s, x_n + \frac{1}{2} t_s k_1)
\]
\[
k_3 = f(t_n + \frac{1}{2} t_s, x_n + \frac{1}{2} t_s k_2)
\]
\[
k_4 = f(t_n + t_s, x_n + t_s k_3)
\]

The Elastic Force

As we will see later, the way MADYMO handles and depicts surface interactions can often be the source of error in our simulation. Therefore it is important that we examine the mathematical mechanisms behind the process.

An elastic force \( F_e \) is generated if an ellipsoid penetrates a plane or another ellipsoid, provided that the interaction is defined as a possible contact. The elastic force depends on the penetration and the force-penetration characteristics.

The force penetration characteristics used for the elastic contact force calculation are defined analogous to the specified contact interactions. If the force penetration characteristics of both interacting objects are combined by MADYMO to form one resultant characteristic, the two interacting surfaces can be considered as two springs in series.

Force penetration characteristics are defined by means of the function option available in MADYMO. A positive value of the force corresponds to a resistive contact force. In addition, hysteresis and dynamic amplification can be defined. Separate function characteristics for loading and unloading were entered.
In order to define damping and friction forces a reference plane is introduced. In the case of plane-ellipsoid interaction, the reference plane is parallel to the contact plane. For the ellipsoid-ellipsoid interactions the reference plane is parallel to the tangent planes.

The relative velocity $\Delta V$ between the interacting contact surfaces is defined as the relative velocity at the point P of the two contacting objects. This velocity vector is resolved in two components: a component $\Delta V_{\text{plane}}$ in the reference plane and a component $\Delta V_{\text{norm}}$ normal to this plane.

The damping force $F_d$ is defined as:

$$F_d = C_d \cdot |\Delta V_{\text{norm}}|$$

$$C_d = C_{ld} \cdot (\Delta V_{\text{norm}}) \cdot C_{2d}(F_e)$$

where $C_d$ is the positive damping coefficient, which is defined as the product of a function of $\Delta V_{\text{norm}}$ and a function of the elastic force $F_e$. In the case of increasing penetration (loading) the damping force is added to the elastic force. If the penetration decreases (unloading) the damping force counteracts the elastic force. Since contact forces are resistive forces no contact forces are applied during unloading if the damping force exceeds the elastic force.
In addition to the damping force, a dry friction force $F_f$ can be specified. This friction force acts in the reference plane in the direction opposite to the relative velocity component $\Delta V_{\text{plane}}$.

$$F_f = C \cdot f(|F_e + F_d|) \cdot |F_e + F_d|$$

where $f(|F_e + F_d|)$ is a Coulomb friction coefficient and $C$ a so-called ramp function. This ramp function varies between 0 and 1 as a function of the relative velocity $\Delta V_{\text{plane}}$. The ramp function has been introduced in order to avoid vibrations induced by dry friction. The friction coefficient can be defined as a function of the magnitude of the normal force.

**Hysteresis**

Elastic properties for joints, springs, contacts, belts and restraints are defined by means of functions. Energy dissipation in these force-interaction models can be described by means of hysteresis. MADYMO offers three different hysteresis modes. In this study the simplest one was used. The hysteresis model requires the specification of the following:
- a loading curve \( y_1(x) \)
- an unloading curve \( y_u(x) \)
- a hysteresis slope \( s_l \)
- an elastic limit \( x_e \).

The slope parameter \( s_l \) (i.e. \( \Delta y/\Delta x \) in Figure 2.1.5) defines a linear function between the loading and the unloading curve. The same slope is used for positive and negative values of the deformation \( x \).

![Figure 2.1.5. Parameters for description of hysteresis model](image)

The hysteresis behavior for \( x > 0 \) is as follows:

- If the independent variable \( x \) reaches a maximum value (\( x_{\text{max}} \)) and the elastic limit \( x_e \) has been exceeded, unloading will take place along the hysteresis slope \( s_l \) until the unloading curve is reached.
- The unloading then proceeds downward along the unloading curve.
- A reloading will first follow the unloading curve in opposite direction until the point where the unloading curve was first entered.
- Reloading then continues upward along the hysteresis slope \( s_l \) until the loading curve is reached again.
- Further loading beyond \( x_{\text{max}} \) will follow the loading curve until a new \( x_{\text{max}} \). A subsequent sequence of loading and unloading will occur in the same way as just described.
For $x<0$ the hysteresis behavior is similar, except that hysteresis calculations are carried out if $x$ reaches a minimum value ($x_{\text{min}}$) and the elastic limit $x_e$ has been exceeded in the negative $x$-direction.
Chapter three discusses the experimental setup used to achieve the goals outlined in the introduction of this study. Initially, we present issues that are common throughout the study, such as the modeling of the vehicle, the dummies, the airbag and the conditioning of the acceleration profile. Then we proceed to examine separately each simulation setup and the specific considerations and assumptions associated with each experiment. A sample MADYMO input file is included in Appendix II.
3.1 Vehicle Model

The vehicle interior dimensions were taken from a mid-size passenger car. These interior dimensions were transformed to MADYMO contact planes. The model in Figure 3.1 shows the driver side of the car while the passenger side looks similar. For reasons of simplicity and illustration, only the relevant elements of the interior of the car were included.

Since the relative positions of the seat, the steering wheel, the dashboard, the instrument panel and the floor are constant they were rigidly mounted in the inertial space. To achieve further simplification and ease of illustration, the driver and passenger seat were replaced by a bench while the pedals were omitted. The lower part of the seat consists of two planes representing the seat cushion and the seat ramp. The seat is also connected to the floor with a Maxwell element which allows a small range of motion. Seat motion is typical in car crash accidents and the spring damper combination was empirically specified to model this displacement during high impact crashes.
All the surfaces were given the appropriate hysteresis functions and friction coefficients. The loading functions and friction coefficients of the contact planes were determined on the basis of component test and are typical for the materials used in the automotive industry. The loading and unloading functions of the surfaces are summarized in Figure 3.1. The same vehicle model was used for all four experiments in this study.

Figure 3.1. The loading functions (Force vs. distance) of the various interior surfaces

3.2 Dummy Models

Throughout the study, dummies were used as the experiment setup commanded. The 5th percentile female Hybrid III model was used to simulate a typical small female, the 95th percentile male Hybrid III
model to simulate a large typical male and a 50th percentile male Part 572 to cover the typical adult occupant. These are well proven, validated models used widely in the industry. They were taken from existing dummy libraries provided by TNO.

![Image](image_url)

**Figure 3.2.1. Example of TNO crash dummies:**
(a) 50th percentile male (b) 5th percentile female

### 3.3 Airbag Models

The driver side airbag is modeled as two parallel circular planes, using a triangular constant strain membrane element. The material behavior of the airbag fabric is modeled with a linear elastic isotropic material model. The planes are joined at the edges. In the initial configuration, the circular planes coincide. The model of the driver side airbag consists of 1024 elements. The number of degrees of freedom of the finite element airbag model is 1542. The front and back plane of the airbag are connected to each other with four straps. These straps are included in the model as massless springs between the proper nodes of the finite element model. Leakage of gas through the exhaust orifices has been taken into account. It should be noted that when the model is used for the simulation, the unfolding effects which occur in the early stages of the inflation process of the airbag are not taken into account. However, as long as the occupant interacts with the airbag after unfolding only, these effects are not expected to influence the simulation results significantly. The reason for avoiding the simulation of early stage inflation process stems from MADYMO's well documented difficulty to model accurately occupant-airbag interactions.
during these stages and from the increased computation power that would have been necessary. The airbag properties are summarized in Table 3.3.1. The temperature and mass flow rates of the inflowing gas used as input for the simulations are shown in Figure 3.3.1 while a two dimensional picture of the airbag is included in Figure 3.3.2.

![Inflator gas temperature and mass flux](image)

**Figure 3.3.1.** Temperature and mass flow rates for the airbag model

![Example of a MADYMO generated driver airbag](image)

**Figure 3.3.2.** Example of a MADYMO generated driver airbag
### Table 3.3.1. Summary of airbag properties

<table>
<thead>
<tr>
<th>PARAMETER</th>
<th>DRIVER AIRBAG</th>
</tr>
</thead>
<tbody>
<tr>
<td>fabric porosity</td>
<td>0.0</td>
</tr>
<tr>
<td>fabric density</td>
<td>823.8 [kg m⁻³]</td>
</tr>
<tr>
<td>fabric thickness</td>
<td>0.4 [mm]</td>
</tr>
<tr>
<td>trigger time</td>
<td>12 [ms]</td>
</tr>
<tr>
<td>Young's modulus</td>
<td>6.0x10⁷ [N m⁻²]</td>
</tr>
<tr>
<td>Poisson ratio</td>
<td>0.3</td>
</tr>
</tbody>
</table>

#### 3.4 Acceleration Profile

The acceleration field --used where applicable-- is the acceleration pulse of a real world crash. An accelerometer was placed in the front of similar car as the one in the model and then a crash test was conducted at a controlled speed. The output of the accelerometer was connected to an Analog to Digital converter and was sampled at 10Khz. The unfiltered acceleration profile for a mid severity crash of a mid-size vehicle with a rigid obstacle is shown below.

![Figure 3.4.1. Unfiltered longitudinal acceleration profile for a 49 km/h frontal collision of a midsize passenger car with a rigid obstacle](image)
This is the acceleration seen by the chassis of the car and contains many frequencies associated with the structure of the particular vehicle. For simulation purposes the signal can be reasonably filtered and still yield realistic results. This is due to the fact that the human body acts as a low pass filter to the high frequencies contained in the acceleration pulse. The various body parts tend to resist sudden changes in motion due to their mass, inertia and biomechanical properties. Therefore, they are completely unaffected by many of the sharp peaks and dips in the acceleration profile. For simplicity, ease of computation, and higher degree of accuracy the actual acceleration profile inputted in MADYM0 was filtered using the standard SAE Class 1000 filtering:

\[ a[n] = \sum_{n}^{n+4} \frac{f[n]}{4} \]

![Filtered acceleration profile](image)

**Figure 3.4.2. Filtered acceleration profile**

### 3.5 Experiment 1

To conduct a comparative analysis of the freebody occupant displacement algorithm and the MADYM0 simulation, the vehicle model described above was used.

A 50th percentile unrestrained male driver was placed in normal driving position inside the vehicle model shown below.
The dummy was left in the vehicle with the gravitational field acting on it so that it is balanced until the interactions of the various body parts with the vehicle surfaces reached steady state. This was necessary in order to achieve more accurate and realistic results. Since the experiment focuses in the displacement of the occupant during the early stages of the crash, minute motion due to a potentially imbalanced original position can yield considerably distorted results. Once the balanced position of the dummy was determined, the vehicle was subjected to a real world acceleration profile of a mid-size passenger sedan engaging into a mid-severity frontal crash with a rigid obstacle (Figure 3.4.2).

The MADYMO data file was instructed to track points at the sternum and the head of the occupant and generate data about their displacement as a function of time.

As discussed earlier, the freebody algorithm is based on the fact that

\[ d(x) = \int \int a(x) dt \]

where \( d(x) \) is the displacement and \( a(x) \) the acceleration of the (free) body. The acceleration profile of Figure 3.4.2 was integrated twice to yield the displacement. The two curves were set side by side for comparison. (Figure 4.1.1)
3.6 Experiment 2

In the second experiment we attempt to establish a correlation between the driver’s linear impact velocity and upper body injury as well as investigate the various parameters that affect it. As linear impact velocity we define the velocity of the driver at the time of impact with an interior vehicle surface. In a frontal collision the linear impact velocity is a one dimensional vector along the x-axis.

To cover a range of occupants, a 5th percentile female, and a 50th and 95th percentile male dummies were placed in various driving positions inside the vehicle model of Figure 3.1. All drivers were unrestrained (i.e. no seat belt or airbag). In order to maintain control of the driver’s linear impact velocity, the only acceleration field acting on the system was gravity. Subsequently, the dummy was given an initial linear uniform velocity along the x-axis equal to the desired impact velocity.

In order to investigate how various vehicle variables affect injury at the time of impact, we designed a set of simulations focusing on the effects of the following accident related parameters:

- seat design and position
- occupant size
- steering wheel position
- roof
- kneebolster angle
- toe-pan angle
- driver position.

While keeping all variables constant, one was designated as the “free” variable and was modified in order to understand its contribution to overall driver behavior. Initially the "normal" driving position was established for all three dummies as the seat setting which allowed the pedals and the steering wheel to be reached comfortably (no stretching, leaning etc.). Then the dummies were subjected to initial velocities ranging from 2 m/sec to 30 m/sec in order to cover a wide range of crash scenarios. Indicatively, a linear velocity of 2m/sec corresponds to a very soft impact (i.e. hard braking) while a velocity of 30 m/sec
corresponds to an extremely severe collision (100 mph). Finally, MADYMO produced animated sequences and upper body injury assessment which were analyzed to produce a series of conclusions.

Figure 3.6.1 shows the various vehicle setups used to investigate the accident related parameters described above.

![Setup Diagram](image)

Figure 3.6.1. setup to investigate the effects of
(a) the roof, (b) the toe pan (c) the windshield (d) the steering wheel
in occupant behavior during crashes

### 3.7 Experiment 3

In the third experiment we used MADYMO in an attempt to establish a correlation between the distance of the driver from the steering wheel and upper body injury. A 5th percentile female, and a 50th and 95th percentile male dummies were placed in the driver position inside the vehicle of Figure 3.1. All drivers were unrestrained (i.e. no seat belt or airbag). To investigate the correlation between driver distance and injury, we took special care to cover a variety of realistic driving positions. After considering the degrees of freedom of a typical driver seat and observed the driving habits of many real world drivers, we compiled about twenty different driving positions for each occupant.
Figure 3.7.1 shows six different positions for a 5th percentile female driver.

![Six different driving positions for a 5th percentile female driver](image)

The distance of the occupant from the steering wheel is defined as the combined distance from the center of the head and the center of the steering wheel and the center of the chest from the steering wheel. (Figure 3.7.2)

![Distance between driver and steering wheel](image)

The vehicle was subjected to a real world acceleration pulse of a mid-size passenger sedan engaging into a mid-severity frontal crash with a rigid obstacle (Figure 3.5.1). The MADYMO data file was instructed to track the acceleration of the sternum and the head and assess the
sustained injury. The HIC and the chest G's were plotted as a function of the distance for each driver.

We also considered the effect of the angle of the steering column to the overall upper body injury. To accomplish this we set $\theta$ (Figure 3.7.3) to be the free variable while we kept the driver distance constant. We set $\theta$ to 26 degrees --typical value for most vehicles-- 17 and 0 degrees and we recorded the changes in injury.

![Figure 3.7.3. Angle $\theta$ of the steering wheel](image)

To further understand upper body injury as a function of driver distance we designed a series of simulations with the chest distance as the free variable while the head distance was kept constant and vice versa.

### 3.8 Experiment 4

To investigate the effects of varying the airbag deployment time, a 50th percentile Part 572 dummy was placed unrestrained in the driver's seat in a normal driving position. The vehicle model is identical to the one in Figure 3.1 with the addition of a driver's airbag mounted on the steering wheel. The airbag was configured with straps so that its deployment behavior resembled in a high degree the deployment of a real driver airbag.
The vehicle was then subjected to a real world acceleration profile of a mid-size passenger sedan engaging into a mid-severity frontal crash with a rigid obstacle. (Figure 3.4.2)

![Diagram](image)

Figure 3.8.1. Setup to investigate effects of variable airbag deployment times

The deployment time of the airbag was set to be the free variable. Normally, for a car like the one in the simulation, a typical airbag trigger time is $t_0 = 15$ msec from the beginning of the crash. The input data file was set to trigger the airbag at $-20$ msec, $-10$, $-5$, $-2$, $0$, $+5$ $+10$ msec with respect to the normal deployment time $(t_0)$. The Head Injury Criterion and the chest 3ms g’s were recorded and plotted against the trigger time. In order to examine the effects of variable airbag deployment on smaller occupants, a series of simulations were setup with a 5th percentile female Hybrid III dummy in normal driving position and variable trigger times. Care was taken to place the drivers in a proper driving position as an out of position driver would introduce a set of new variables associated with the airbag-driver interaction and injury. In order to minimize computer simulation time and to improve the reliability of the model, the airbag was placed fully unfolded and deflated on the steering wheel. At the time of deployment the gas flow was turned on. Since the interaction between driver and airbag are expected to occur at its final inflation stage this convention does not introduce another variable to the experiment.
In all simulations, MADYMO was instructed to produce an animated sequence of the simulation in order to study the behavior of the occupant during the accident. In many cases comparative sequences were produced, where the two animation sequences were superimposed in order to examine the differences produced by the change of a free variable. A series of sequences were recorded on video tape.
Chapter 4

Results

In this chapter we present briefly the results of the four experiments. Various parameters are plotted and several kinematic sequences are included to help illustrate key concepts in the analysis to follow. In total, we ran more than 600 simulations in order to study a series of accident related parameters.

4.1 Experiment 1

The following plot shows the displacement of the freebody and the MADYMO modeled sternum and head as functions of time. The crash occurs at $t=0$ while occupant impact occurs at $t=120$ msec.
4.2 Experiment 2

The following sets of curves show the correlation between the Head Injury Criterion and the driver linear impact velocity for various accident related parameters. As a reference model we considered the normal driving position of a 95th percentile male with typical vehicle interior parameters ($\theta_{\text{steering wheel}}=27$ deg, $\theta_{\text{toe-pan}}=130$ deg etc.) For the steeper and leaner toe-pan we "pushed" the toe pan backward and forward 10cm respectively.
Much like Figure 4.2.1, Figure 4.2.2 shows the correlation between chest injury and driver impact velocity for various accident related parameters.

Figure 4.2.1. HIC vs. Occupant Linear Impact Velocity for a 95th percentile male for various impact parameters

Figure 4.2.2. Chest 3ms g’s vs. Occupant Linear Impact Velocity for a 95th percentile male for various impact parameters
4.3 Experiment 3

The following graphs summarize various correlations between upper body injury and driver distance.

Figure 4.3.1. HIC vs. head distance from the steering wheel for a 5th percentile female and various steering wheel angles
Figure 4.3.2. HIC vs. chest distance from the steering wheel for a 5th percentile female and various steering wheel angles

Figure 4.3.3. Chest 3ms acceleration vs. distance from the steering wheel for a 5th percentile female

4.4 Experiment 4

The results of our attempt to quantify the benefits of variable deployment time are listed below. Figures 4.4.1 and 4.4.2 show upper body injury for various deployment time offsets, while figures 4.4.3 and 4.4.4 illustrate the kinematics of earlier and later deployment times.
Figure 4.4.1. HIC and Chest g’s vs. airbag deployment time for a 50th percentile male

Figure 4.4.2. HIC and Chest g’s vs. airbag deployment time for a 5th percentile female
Figure 4.4.3. Kinematics for late airbag deployment time ($t=+15$ msec) for a 50th percentile male

Figure 4.4.4. Kinematics for early airbag deployment time ($t=-20$ msec) for a 50th percentile male
Chapter 5

Analysis

In Chapter five we present a detailed analysis of the simulation results organized by experiment. We discuss the simulation output, analyze any errors, investigate what seem to be inconsistencies and reach several conclusions. In many cases, physically-based models and intuition are used to substantiate the observations of the simulations.

5.1 Experiment 1

As expected, there was a difference between the displacement predicted by the freebody algorithm and the results provided by MADYMO. Looking at Figure 4.1.1 in Chapter 4 one can distinguish three different curves. The freebody displacement and the MADYMO produced head and sternum
displacements are virtually identical during the first fifteen milliseconds of the crash. The displacement is very small, on the order of half of a centimeter. This is due to the typical acceleration profile observed in frontal car crashes (Figure 3.4.1). During the first few milliseconds of the crash the acceleration pulse is fairly small due to the energy absorption of the outer structure of the car (bumpers, plastic trim etc.).

At about 25 msec, there is a clear differentiation between the freebody and the MADYMO results. The slope of the freebody curve is steeper than the slopes of both the head and sternum curves. This is largely due to two factors. First, there is friction developed between the occupant’s feet and the floor and --most important-- between the occupant’s lower body and the surface of the seat. Second, the various occupant body parts have inertia which by definition resists any changes in motion. In the first few milliseconds however, the most important factor affecting occupant behavior is the friction. There are three kinds of friction: static, critical and kinetic:

\[ F_{\text{static}} \leq F_{\text{kin}} \leq F_{\text{crit}} \]

In the first 15 msec there is no motion due to the static friction. The maximum value of the static friction is reached at about 20 msec after which the body experiences the kinetic friction.

Moving along the time axis one can distinguish an increasing differentiation between the sternum and head displacement curves. The data indicates that the head is lagging with respect to the sternum. The observed motion has its roots in the bio-mechanics associated with a human body. In the case of the 50th percentile 572 dummy, the mass of the sternum and the head are 17 and 4 kilograms respectively. The sternum is part of the thorax which is connected to the lower body via the spine joint and to the head via the neck joint. Both those joints are modeled by the flexion-torsion joint model. While biomechanics and joint analysis are beyond the scope of this study, one can think of a joint as a spherical linkage with a damping coefficient (Figure 5.1.1). Thus the joint dampens the energy transferred from the sternum to the head and allows for relative motion freedom between the two body parts.
At this point one might argue that since both the head and the sternum are subjected to the same acceleration field, they should follow the same motion. This would be largely true if it were not for the friction experienced by the lower body. The upper legs and the pelvis tend to remain stationary and that induces a rotary motion for the upper body via the spine spherical joint. The head, due to its inertia, resists to the rotational motion of the sternum and it lags. The joints introduce intricate dynamics that are beyond the scope of the study. For the set of conditions of this experiment the relative displacement \((D_{\text{sternum}} - D_{\text{head}})\) increases as the time increases and the head lags further behind. The head is also engaged in a rotational motion due to the neck joint. The observed behavior can be explained qualitatively by looking at Figure 5.1.2. The dummy can be modeled as a system of bodies as shown below:

Bodies L, S and H are defined as the rods AB, BC and CD shown in the figure above. Body L represents the dummy's lower body, S the chest and
H the head. All bodies are connected via cylindrical joints and can rotate freely around the pivot points A, B and C. As the system of bodies starts moving the friction force F, developed between the seat and the lower body effectively slows down L and S starts rotating as shown. Because of its inertia, H resists the motion and it "falls back." A more rigorous treatment of essentially the same phenomenon is presented in the analysis of Experiment two, where we examine how pelvis deceleration affects head motion.

As Figure 4.1.1 indicates, the freebody displacement curve crosses the sternum displacement curve at t=75msec. While such a result might seem erroneous at a first glance, a closer look at how the acceleration profile affects the physical bodies reveals a possible explanation. Suppose that a body part of the dummy is subjected to the acceleration a(t) (Figure 3.7.1). If x(t) is its displacement then:

\[ -\frac{k}{m}x + \frac{b}{m}\dot{x} = a(t) \]  (1)

where k and b are constants reflecting the physical properties of the body and m is its mass. The applied acceleration a(t) can be also expressed as:

\[ a(t) = \int a(\omega) e^{i\omega t} d\omega \]

To solve (1) we can try \( x = Ae^{i\omega t} \). Then for a single frequency \( \omega \) we have:

\[-A\omega^2 + A\frac{k}{m} + i\omega A\frac{b}{m} = a(\omega) \iff |A|^2((\frac{k}{m} - \omega^2)^2 + (\frac{b}{m}\omega)^2) = |a(\omega)|^2\]

and \(|A|\) is:

\[ |A| = \frac{|a(\omega)|}{\sqrt{(\frac{k}{m} - \omega^2)^2 + (\frac{b}{m}\omega)^2}} \]  (2)

We can define \( a_{\infty}(t) \) to be the acceleration experienced by the body (a body part of the dummy). Then from (1) and (2):
Equation (3) shows the relation between a single frequency component of the applied acceleration field \(a(\omega)\) and the acceleration of the body \(a_{\text{body}}(\omega)\). Equivalently, in the time domain we have:

\[
a_{\text{body}}(t) = \int a_{\text{body}}(\omega)e^{i\omega t} dt = \int a(\omega)f(\omega)e^{i\omega t} dt
\]  

Equation (4) indicates that the acceleration experienced by the body is a weighted average of the applied acceleration. In our experiment, the applied acceleration curve experiences a peak at 73 msec immediately followed by a trough and then by another peak (Figure 3.5.2). Apparently, the physical properties of the sternum average out the trough. Therefore the acceleration of the body is greater than the applied acceleration for the duration of the trough which explains the crossing between the displacement curves of the freebody and the sternum.

An additional source of error can be attributed to the way MADYMO treats surface interactions. MADYMO depicts elastic forces between two surfaces as penetration of one into the other. In the current model, the upper surface of the seat is not quite horizontal but tilted upwards. (Figure 3.5.3) Consequently there is a series of intricate dynamics introduced by the seat-dummy interactions which affect the motion of the dummy.

In conclusion, one might argue that the freebody algorithm is a crude description of a worst case scenario. Emphasis should be placed on appropriately conditioning the acceleration profile. For airbag deployment considerations the times involved from the beginning of the crash until full airbag deployment are in the order of 60 msec. In our results, at 60 msec the difference between the freebody prediction and the more realistic MADYMO output is about 7cm or 2.75 inches. Depending on the severity of the crash and the size of the occupant, that
difference may increase or decrease. The freebody algorithm serves well airbag system designers who design for the worst.

In addition to its value as a comparative analysis, Figure 4.1.1 provides a good illustration of the kind of displacement one should realistically expect from an occupant in the course of a frontal crash accident.

5.2 Experiment 2

The correlation between the driver's linear impact velocity and upper body injury for a 95th percentile Hybrid III male dummy is shown in Figures 4.2.1 and 4.2.2. As linear impact velocity we defined the velocity of the driver at the time of impact with an interior vehicle surface. This velocity is taken with respect to the vehicle and in a frontal collision it is a one-dimensional vector along the x-axis. Indicatively, for a 49 km/h frontal collision with a rigid barrier, typical values of the driver's velocity at the time of impact range from 8m/sec to 12 m/sec depending on a variety of factors including initial position, weight, seat design, vehicle interior etc.

As expected, the injury increases as the linear velocity of the occupant increases. From a first look at the data, it is made apparent that there is a clear correlation between occupant velocity and injury. A logarithmic plot of Figure 4.2.1 reveals that the HIC relates to the velocity in a second or third degree fashion depending on a multitude of variables, which we will examine below. Using the slope of the semi-log plot (assuming the default is 2) we will be able to quantify the effects of several variables associated with occupant dynamics.

![Figure 5.2.1. A logarithmic plot of HIC vs. occupant linear velocity](image)
One of the most interesting revelations of this study is the contribution of the toe pan in the upper body injury. As the size of the dummy increases, the effect gets more pronounced. For the 95th percentile Hybrid III dummy, the free variable was set to be the angle of the toe pan (Figure 5.2.2a). It is immediately apparent that for a given set of conditions (velocity, occupant initial position etc.) a leaner toe pan results in lower upper body injury (Figure 5.2.2b). In the extreme case where the toe pan is omitted (θ=0 deg.) the results are very dramatic in comparison.

![Diagram of toe pan angle](image)

![Graph of HIC vs. velocity for various toe pan angles](image)

Figure 5.2.2.  
(a) Toe pan angle  
(b) HIC vs. velocity for various toe pan angles

The analysis produced by MADYMO suggests that there is a substantial coupling between the lower and upper body. We can postulate that fast pelvis stops result in increased head injury and proceed into using simplified two-dimensional physical models to substantiate our assumption.

The dummy can be modeled once again as a system of rods shown in Figure 5.2.3. For ease of reference let us call body A the rod from a to b, body B the rod from b to c and C from c to d. Then A represents the dummy's lower body, B is the chest the length of which is l, and C is the head. All bodies are connected via cylindrical joints and can rotate freely around the pivot points b and c. Since we are not concerned with small changes in the position of the head we could treat the neck as a
rigid joint and thus simplify our analysis. To model a pelvis stop, body A is decelerating with an acceleration \( a \) and eventually stops. We will prove that the deceleration of body A increases the velocity of point c and consequently its acceleration. Since injury depends on acceleration it is expected to increase as well.

![Simplified model to analyze the effects of fast pelvis stops](image)

Figure 5.2.3 Simplified model to analyze the effects of fast pelvis stops

The velocity \( v(x,y) \) of point c is:

\[
\begin{align*}
v &= \begin{cases} 
  x' &= x_{pivot} - r \sin \phi \frac{d\phi}{dt} \\
y' &= r \cos \phi \frac{d\phi}{dt}
\end{cases}
\end{align*}
\]

where \( y' \) and \( x' \) are the time derivatives of the coordinates of point c and \( x_{pivot} \) is the velocity of the pivot point b. The angular momentum of body B is:

\[
L = \frac{mI}{l} \int_0^t (r \times v) \, dr = \ldots = \frac{mI}{l} \int_0^t \left( x_{pivot} r \cos \phi \frac{d\phi}{dt} - x_{pivot} r \sin \phi + r^2 \frac{d\phi}{dt} \right) \, dr
\]

Evaluating the integral and simplifying yields:

\[
L = \frac{ml}{2} \left[ \left( x_{pivot} \cos \phi \frac{d\phi}{dt} - x_{pivot} \sin \phi \right) + \frac{2l}{3} \frac{d^2\phi}{dt^2} \right]
\]

Due to conservation of angular momentum:

\[
\frac{dL}{dt} = 0 \iff \frac{dx_{pivot}}{dt} \cos \phi \frac{d\phi}{dt} - x_{pivot} \sin \phi \left(\frac{d\phi}{dt}\right)^2 - \frac{dx_{pivot}}{dt} \cos \phi \frac{d\phi}{dt} + x_{pivot} \cos \phi \frac{d^2\phi}{dt^2} - x_{pivot} \sin \phi \frac{d^2\phi}{dt^2} - \frac{2l}{3} \frac{d^3\phi}{dt^3} = 0
\]
using \(-a = \frac{d^2 x_{\text{pivot}}}{dt^2}\) we get:

\[-a \sin \phi = -x_{\text{pivot}} \sin \phi \left(\frac{d\phi}{dt}\right)^2 - x_{\text{pivot}} \cos \phi \frac{d^2 \phi}{dt^2} + \frac{2l}{3} \frac{d^2 \phi}{dt^2}\]

Since body B does not penetrate the obstacle to a great length we can assume that \(x_{\text{pivot}}<<1\) which yields:

\[
\frac{2l}{3} \frac{d^2 \phi}{dt^2} \approx -a \sin \phi \Leftrightarrow \frac{2l}{3} \frac{d^2 \phi}{dt^2} \approx -a \frac{d\phi}{dt} \Leftrightarrow \\
\Leftrightarrow \frac{2l}{3} \frac{d}{dt} \left( \frac{1}{2} \frac{d\phi}{dt} \right) \approx \frac{a}{dt} \frac{d}{dt} (\cos \phi) \Leftrightarrow \frac{d}{dt} \left[ \frac{l}{3} \frac{d\phi}{dt} - a \cos \phi \right] = 0 \quad (1)
\]

or equivalently:

\[
\frac{l}{3} \left( \frac{d\phi}{dt} \right)^2 - a \cos \phi = C \quad (2)
\]

The initial conditions for (1) are the following:

\[
t=0, \; \phi=90\text{deg}, \; x_{\text{pivot}}=0 \text{ and } dx_{\text{pivot}}/dt=0
\]

so:

\[
L = \frac{ml}{2} \left( \frac{2l}{3} \frac{d\phi}{dt} \right) = L_{\text{before crash}} \Leftrightarrow \frac{m}{l} \int \left( \vec{r} \times \vec{v} \right) dr = \frac{ml}{2} v_o \Leftrightarrow \\
\frac{d\phi}{dt} \bigg|_{t=0} = \frac{3 v_o}{2 l}
\]

from (2) for \(t=0\):

\[
\left( \frac{d\phi}{dt} \right)^2 \bigg|_{t=0} - a0 = C \Leftrightarrow C = \left( \frac{3 v_o}{2 l} \right)^2 = \frac{3v_o^2}{4l}
\]

Then (2) becomes:

\[
\left( \frac{d\phi}{dt} \right)^2 \bigg|_{t} = \left( \frac{3 v_o}{2 l} \right)^2 + \frac{3a}{l} \cos \phi(t) \quad (3)
\]

At \(t=t_{\text{impact}}\), \(\phi=\phi_{\text{imp}}=\phi^*_{\text{imp}}\) we can solve (3):

\[
\left. \frac{d\phi}{dt} \right|_{t_{\text{imp}}} = \sqrt{\frac{9v_o^2}{4l^2} + \frac{3a}{l} \cos \phi^*_{\text{imp}}}
\]

But since \(v_{\text{head}} = v_c = \frac{d\phi}{dt} l\),

\[
v_{\text{head}} = \sqrt{\frac{9v_o^2}{4} + 3al \cos \phi^*_{\text{imp}}} \quad (4)
\]
Equation (4) suggests that the velocity of the head increases as the deceleration of the pelvis (a) increases. Equivalently, faster pelvis stops result in higher HIC's.

The vehicle model was modified to include a roof and in order to examine its effects to the overall injury sustained by the driver. The dummy used in the simulations was the 95th percentile male since he is the driver most likely to experience such injury. The results indicate that roof contributions to the overall head injury are secondary. The head impact with the steering wheel or the windshield was an order of magnitude larger than the impact with the roof. Emphasis however should be placed on the fact that the simulations ran with MADYMO in this study assume a strictly linear motion of the car along the x-axis. The only acceleration field acting on the z-axis is gravity and there is no acceleration along the y-axis (rotational). Depending on a multitude of factors, including the suspension and the structure of the vehicle, many crashes introduce a z-axis acceleration component. Figure 5.2.4 illustrates a crash sequence where rotation along the y-axis occurs, which induces acceleration along the z-axis.

![Vertical acceleration due to a frontal car crash](image.png)

Even though there is contact of the driver with the roof, the x-axis acceleration is still an order of magnitude higher and it is responsible for the majority of the head injury. While the roof is a secondary source of injury, it can help reduce the overall upper body injury because it "guides" the driver toward the windshield. As we will see, the windshield acts as a restraint and absorbs some of the impact energy.

As hinted earlier, an interesting finding has to do with the contribution of the windshield to the injury sustained by the driver. From an injury point of view, contact with the windshield is desirable.
The results of the simulation indicate that windshield impact is "milder" than impact with the instrument panel and the steering wheel. To further investigate this non-intuitive result, a 95th percentile dummy (where contact with the windshield is most likely to occur) was placed at different driving positions in such a way that contact with the windshield would occur. Curves A, B, and C in Figure 5.2.5 illustrate the correlation between HIC velocity and windshield contact. The 95th percentile is leaning forward so that the windshield absorbs a significant amount of the impact. The more energy the windshield absorbs, the lower the HIC number.

![Diagram of simulated driver positions](a)

![Graph of HIC vs. velocity](b)

Figure 5.2.5.
(a) Driver positions to investigate windshield effects
(b) HIC vs. velocity for different windshield impacts

The "windshield effect" is responsible for what seems like a discrepancy in the simulation results. For example, the HIC for a 95th percentile for a linear velocity of 6m/sec is 200, whereas for v=8m/sec it is only 100. At a first glance, this contradicts our finding of direct correlation between injury and linear velocity. The observed values however hold since faster speeds result in greater knee impact onto the knee bolster. This causes the pelvis to decelerate sharply and the occupant to lean forward. Consequently, his head hits first the windshield and then it hits the instrument panel and steering wheel. In the 6 m/sec case, there is little or no energy absorbed by windshield due to the limited contact with the head. The deceleration of the head
(and equivalently the head injury) results solely from the impact of the chest on to the steering wheel.

The restraining behavior of the windshield stems from its structural properties and its positioning relative to the motion of the occupant. Figure 5.2.6 shows the loading-unloading characteristics of the windshield and the steering wheel surfaces. Structurally, it provides a better "cushion" than the other interior surfaces of the car.

Figure 5.2.6. Loading-unloading characteristics of (a) the windshield and (b) the steering wheel

Figure 5.2.7 shows the force deflection due to the orientation of the windshield. We used MADYMO to track the path of the head during a crash involving windshield and I/P impact. Effectively, the windshield pushes the occupant down into the seat deflecting some of the impact energy. Assuming that the windshield is tilted 45 degrees, the force exerted by the windshield is half than that of a perpendicular steering wheel.
A dramatic indication of the windshield effect occurs during the course of a high severity crash. Surprisingly, when the linear impact velocity was set to 16 m/sec for a 95th percentile dummy, the resulted HIC is only 500. The high speed impact with the toe pan and the knee bolster make the upper body rotate forward just enough so that the right amount of energy is divided between the head and the windshield first and the chest and the steering wheel subsequently. This creates a fairly mild head-steering wheel impact which is the primary source of injury. The combination of the crash variables (initial position, linear velocity, car interior etc.) yields an acceptable HIC for a frontal vehicle collision with a rigid body at about 55 mph. The simulation however did not account for any structural deformations that are bound to happen in such severe accidents.

The windshield effect is more significant as the size of the dummy gets larger. For smaller dummies (i.e. 5th percentile female) the steering wheel will almost always prevent the dummy from hitting the windshield. Simulations indicate a strong correlation between the severity of the injury and the angle of the steering wheel. It turns out that the steeper the angle of the steering wheel the higher the injury sustained. The effect of the steering wheel is more pronounced in the case of a smaller driver. A 5th percentile female driver dissipates a large part of her impact energy on the steering wheel. As the steering wheel becomes leaner, the chest comes in contact before the head and absorbs some of the impact energy. The coupling through the neck lowers the amount of deceleration due to that first impact. As the steering wheel
gets steeper, the impact energy dissipated at the head increases which results in higher head acceleration and consequently higher HIC numbers. The steering wheel opposes the motion of the driver during the crash, rather than redirecting it as was the case with the windshield.

The driver affected most by the "steering wheel" effect is the 5th percentile female. Simulations show that as the size of the dummy increases, the interactions with the toe pan and the knee bolster become more significant and absorb more of the overall impact energy. In addition, larger occupants are positioned so that their heads are higher than the steering wheel. Consequently, there is limited contact between the steering wheel and the head independently from the angle of the steering column. Figure 5.2.9 summarizes the observations and substantiates the intuitive analysis above.
While on the subject of analyzing parameters that affect occupant injury, it should be emphasized that the results of MADYMO should always be analyzed and justified by simple models or intuition. Some peaks contained in the data can be attributed to the limitations of the simulation package. In the case of the 95th percentile driver leaning forward, the HIC at $v=14\text{m/sec}$ is 2995 whereas for $v=12$ is 602 and $v=16$ is 446. A closer look at the MADYMO output reveals that the HIC in the case of $v=12\text{ m/sec}$ was not measured at the time of the impact rather some time later when the head came in contact with the boundary of the steering wheel. MADYMO's depiction of elastic forces as surface penetration further complicates further the analysis. Therefore much attention is needed when using MADYMO as an analysis tool.

### 5.3 Experiment 3

The results of Experiment 3 (Figure 4.3.1) suggest a direct correlation between the distance of the driver from the steering wheel and upper body injury. One should have expected this sort of correlation since from basic physics:

$$a = \frac{2S}{t^2}$$

However, much attention is required in order to identify the critical distances past which the injury becomes fatal.

As driver distance we defined the combined distance of the center of the surface of the sternum and the face from the center of the steering
wheel (Figure 3.7.2). We chose such a definition because upper body injury depends both on the distance of the head and the chest. This is due the upper body impact energy which is transferred into both the chest and head. As MADYMO suggests, different injury values correspond to the same head distance but different chest distances and vice versa.

Findings of this experiment confirm many of the occupant behavior observations of Experiment 2. The results validate the “steering wheel” effect which suggests that the steeper the steering wheel angle, the higher the overall head injury. When the steering wheel angle was defined as the free variable, the angle-injury correlation became immediately apparent. The plots indicate that for $\theta=0$ degrees, the injury versus distance curve has a higher slope than in the case of $\theta=27$ degrees. In a sense, the angle of the steering wheel regulates the distribution of impact between the head and the chest --and therefore the injury-- as shown in the analysis of Experiment 2.

In the case of the 50th percentile and larger drivers, analysis reveals a strong dependence between the vertical position of the seat and the sustained injury. When the seat position along the z-axis was defined as a free variable while the rest of the variables of the experiment are held constant --including the distance of the driver-- the simulation yielded an inverse correlation between the height of the seat and the amount of injury. It is safe to argue that higher seat positions result in lower HIC’s. The explanation lies primarily in the “windshield effect” presented earlier. The height of the seat effectively regulates the amount of windshield impact. As was discussed in the analysis of Experiment 2, the structure and position of the windshield are responsible for the reduction of the injury. In addition to regulating windshield impact, the seat height regulates the distribution of impact among the head and the chest. Higher seat positions might mean little or no primary head contact with the steering wheel while lower seat positions might mean the exact opposite.

The effects of the toe pan and the knee bolster as investigated in Experiment 2 appear --as expected-- in the series of simulations of Experiment 3. To maintain simplicity and focus they were treated as fixed variables.
As discussed in the introduction, one of the goals of this experiment was to establish an occupant distance past which injury is unacceptable (Figure 5.3.2). As our intuition might have hinted, the occupant size is a crucial factor in the determination such distance thresholds.

The larger the occupant size, the more relaxed the distance thresholds. The size of the occupant dictates his seating position and consequently the degree of windshield impact and energy absorption by the chest. As simulations indicate, larger occupants usually experience more windshield contact, and their chests absorb a larger amount of impact energy. Consequently their heads undergo a “milder” deceleration and sustain lighter injury.

It was attempted to link all the injury contributing variables quantitatively to form an equation of the type:
\[ \text{HIC} = c_1 \times (\text{head distance})^{\alpha_1} + c_2 \times (\text{chest distance})^{\alpha_2} + c_3 \times (\text{seat height})^{\alpha_3} + c_4 \times (\text{velocity})^{\alpha_4} + K \]

where \(K\) is a constant incorporating the design of the car. The result was not reliable due to the fact that there was a large number of variables with non linear correlations to their injury contributions. Real world car crashes are very complex processes with results that hardly ever repeat themselves. The set of variables involved greatly expands if one takes into account the structural properties of the car and how they affect the motion of the occupant.

However, one can come up with a first order qualitative approximation to estimate the HIC. As our analysis indicated, the following table summarizes some key dependencies between the HIC and various accident parameters:

\begin{align*}
\text{HIC} &\propto D_{\text{occupant}} & (1) \\
\text{HIC} &\propto \frac{1}{D_{\text{seat}}} & (3) \\
\text{HIC} &\propto \frac{1}{S_{\text{occupant}}} & (5) \\
\text{HIC} &\propto V_{\text{car}} & (2) \\
\text{HIC} &\propto \frac{1}{H_{\text{seat}}} & (4) \\
\text{HIC} &\propto \frac{1}{\theta_{\text{steering wheel}}} & (6)
\end{align*}

\(D_{\text{occupant}}\): distance of the occupant from the steering wheel  \\
\(S_{\text{occupant}}\): size of the occupant  \\
\(H_{\text{seat}}\): height of seat  \\
\(V_{\text{car}}\): velocity of car  \\
\(\theta_{\text{steering wheel}}\): angle of the steering wheel  \\
\(D_{\text{seat}}\): distance of seat from steering wheel

Contrary to the head injury, the chest injury remains within the acceptable limits for all driver positions considered in this simulation. For example, at \(d_{\text{seat}}=14\) inches the chest injury is still sustainable, while the HIC for the same distance is 350% over the fatality limit (HIC=1000). Such behavior is supported by the simple two dimensional model presented in the analysis of Experiment 2 and by the physical properties of the sternum itself. The main difference between the chest and the head dynamics is due to their different distances from the back joint and --in the case of the chest-- the lack of the neck joint.
5.4 Experiment 4

A glance at the results of Experiment 4 does not immediately provide any insight about the effects of varying the airbag deployment time. As we will see below, a definite correlation between injury and deployment time is very difficult to establish. Such an experiment introduces an enormous amount of complexity. Part of the complexity comes from accurately modeling the airbag, especially the size, mounting, position, materials, gases and timing characteristics. In addition, modern airbags utilize straps (also called tethers) to connect the back and the front of the airbag. This postpones the instant the occupant comes into touch with the airbag. Since the injury depends significantly on the initial interaction between the airbag and the driver, the straps play a crucial role. Further difficulty is introduced by the fact that the straps exhibit a nonlinear behavior. While extreme care was taken in the setup of the experiment, the achieved accuracy is still subject to interpretation.

An intuitive approach about the head injury vs. time would suggest the following:

![Figure 5.4.1. An intuitive speculation of head injury vs. airbag deployment time](image)

If the airbag is deployed much earlier than the crash (anticipatory sensing) it will inflate fully and it will start deflating at the time of contact with the driver. The head injury is thus expected to be higher since the restraining capability of the airbag is reduced. In the limiting case ($\Delta t=-\infty$), the airbag is fully deflated and the HIC reaches its maximum value. In other words for sufficiently early deployment times the airbag has no restraining capabilities. On the
other hand, if the airbag is deployed much later than \( t_o \) (the normal deployment time) the driver will be very close to the airbag at the time of contact. Since the airbag inflation is an uncontrolled explosion pushing the airbag membrane at 160 mph, the resulting HIC is expected to be high. Thus the later the airbag deployment time, the closer to the steering wheel the driver is at the time of deployment and consequently the higher the HIC.

Although Figure 4.4.1 might not show a definite trend for the correlation of injury versus deployment time, a case by case analysis of each simulation revealed many interesting aspects of variable deployment time.

The distance of the driver from the steering wheel and the size of the airbag are very crucial to the overall injury. A late deployment might just touch the head of the driver when it is fully inflated. This will change the direction of the head (force it to lean backward) and consequently affect the HIC number. At the final stage of inflation an airbag might overinflate -- extend-- toward a particular direction due to the straps. It will eventually retract to assume its steady state where the gas is uniformly distributed throughout its volume. If the deployment time is sufficiently late or if the driver is placed sufficiently close, this final inflation stage extension will result to a “slap” at the driver’s face. While this primary contact is not the result of significant injury there has been cases where it tilted the driver’s head so that during the main airbag-driver impact and the subsequent steering wheel-driver impact the HIC figures were much higher. Figure 5.4.2 illustrates graphically the concept of airbag overextension.
The so called "submarining" effect introduces a source of "noise" in our efforts to correlate airbag deployment time and injury. The phenomenon is encountered usually in simulations with smaller occupants where the dummy tends to "submarine" and move under the steering wheel. While the effect is observed throughout Experiments 2 and 3 it is more emphasized with the presence of the airbag. Such a behavior (graphically illustrated in Figure 5.4.3) can be readily explained if we consider the interactions of the dummy with the airbag, seat cushion and knee bolster.
Figure 5.4.3 A 5th percentile female driver “submarining”

A simple physical model can easily explain the event. Figure 5.4.4 illustrates the forces acting on the dummy. Assume that the dummy is moving in the x direction with a velocity $u$ and that her knees hit the kneebolster. The orientation of the kneebolster is such that it redirects the motion of the dummy pushing her “down”

![Diagram of forces acting on a dummy](image)

Figure 5.4.4 Analysis of the “submarining” effect

In addition, the inflated airbag tends to push the smaller driver back and down amplifying the effect. The seat cushioning with its soft loading and unloading characteristics “gives in” allowing the driver to get under the steering column. A solution to this problem is the introduction of a steel plate under the seat cushion which would prevent the occupant from going under the steering column.

A series of simulations indicate the submarining effect is closely related to the airbag deployment time. Earlier deployment times are most likely to result in little or no submarining, while later deployment times tend to amplify the effect. This assertion is contingent upon the distance and the size of the driver. In the case of earlier trigger, the airbag inflates while the momentum of the dummy relative to the car is small. If the dummy is positioned sufficiently close, one can think of the airbag hitting a standing object while
trying to expand. In the case of later deployment times, the momentum of the dummy is considerable. This introduces a complex interaction between the dummy and the expanding airbag which results to the airbag being pushed “upward” and the dummy being pushed downwards. This does not only limit the restraining capabilities of the airbag, but it also causes the “submarining effect.”

From the analysis above, it becomes apparent that the correct modeling of the airbag mounting is very crucial. This in turn introduces an extra degree of complexity and rigorous model validation with high speed film is necessary --something which was beyond the scope of this study. Although it is hard to produce a quantitative correlation between variable deployment time and injury, qualitatively --for a properly seated, unbelted driver-- earlier deployment seems to be beneficial. In many simulation sequences, earlier airbag deployment results in better airbag position during the driver-airbag impact.

Depending on how the airbag is mounted on the steering wheel, if the airbag deploys late(r) the driver-airbag impact occurs when the bag is partially inflated. The driver then pushes the airbag upwards (between the windshield and the steering wheel) so that the sternum comes in direct contact with the steering wheel. In the case of earlier deployment, the airbag inflates fully before the driver-airbag impact and gets “caught” in the right position (between the driver and the steering wheel).

In conclusion, the correlation between injury and airbag deployment time is not straightforward. The interaction between driver and airbag is very complex and there are several limitations introduced by the simulation process.
Some of the key concepts revealed by the four experiments can be used in the design process of advanced restraint systems. The following proposed applications are a direct result of this study on occupant dynamics.

6.1 An occupant sensing system

As it was mentioned in the introduction, unbelted drivers, drivers out of position (leaning forward etc.), and small drivers (females under 5’1”) can find themselves dangerously close to the airbag at the time
of deployment so that they can't withstand the enormous forces associated with the airbag inflation. There have been occasions where the inflation induced injuries were responsible for the death of a driver in an otherwise non-fatal accident. Drivers, usually because they hit another object or they slam the brakes, can find themselves in an “out of position situation.”

The proposed system utilizes an array of sensors to predict the injury to be sustained by the driver. If it is determined that the injury is sustainable (injury criteria values are below a threshold) it inhibits airbag deployment. The system is an add-on to the existing hardware used to determine airbag deployment also known as Sensing and Diagnostic Module (SDM). The output of the SDM is a single bit set to 1 (high) if there is airbag deployment and to 0 if there is no deployment. The output of the occupant sensing system (OSS) is also a single bit set to 0 if the predicted injury does not justify airbag deployment and to 1 if airbag deployment is allowed. The product of the outputs of the two systems is taken and it is passed to the airbag trigger mechanism:

![Diagram of SDM and OSS systems](image)

Figure 6.1.1. The OSS and SDM systems

The Occupant Sensing System receives its inputs from the following sensors:

- a multi beam infrared LED array mounted at the rearview mirror which can measure the distance of a set of points lying on the same line
- an array of weight sensors located into the cushion of the seat
- sensors that relay information about the position of the seat in the X and Y axes
• a sensor that measures the angle of the back of the seat
• a sensor to measure the angle of the steering wheel (assuming an adjustable steering column)
• a connection to the Engine Control Module that relays information about the speed of the vehicle
• a connection to the SDM that relays information about the severity of the crash (filtered accelerometer output)
• a seat belt sensor that senses if the seat belt is used

The arrangement of the sensors is shown in Figure 6.1.2

The weight system along with the seat positioning sensors are used to determine the size of the occupant. While it is easy to conclude that a 100 lb. person is a 5th percentile female, more information is necessary when the driver’s weight can correspond to a wide range of heights. Therefore, the information relayed by the weight sensor alone is insufficient. Assuming that the seat is set to a comfortable driving position, information about the position of the seat --given by the
appropriate sensors-- can be combined with the output of the weight sensor to obtain an estimate of the size of the occupant.

The array of the IR sensors is used to determine the distance of the occupant from the steering wheel. Since it's a multi beam system, it has the ability to "scan" a series of point along the same line. That distance information in conjunction with the occupant size can be used to determine the positions of the driver's chest and head.

The steering wheel sensor relays information about the angle of the steering wheel. Information about the knee bolster, the toe pan, the windshield and the layout of the car are constant and can be incorporated into the system's controller.

The Engine Control Module (ECM) provides information about the speed of the vehicle. All sensors are connected to a micro-controller which assesses the level of the driver injury for non deployment. The estimation of the injury, can done with an algorithm based on the analysis of Experiment 3. If the predicted injury is below a preset threshold, the system assumes that the injury is sustainable without the use of the airbag and it sets its output to zero. The controller monitors the driver at all times so that its output is always set to either one or zero.

In order to optimize such a system, more research is necessary to correlate driver position with injury for airbag deployment scenarios. If that data is available the system can estimate the two injury cases (deployment or non deployment injury) and make the decision that minimizes injury.

Further research is also necessary to accurately set the thresholds for sustainable injury and optimize the algorithm for injury prediction. While a more elaborate algorithm would produce better results an algorithm based on the findings of this study would also be very beneficial.

In a hypothetical situation, the system identifies an unbelted 130 pound driver who has positioned the seat very close to the steering wheel. The vehicle is moving at 50 mph when severe braking occurs. The braking results in driver displacement so that her sternum and her head are at 3'' and 4'' respectively from the steering wheel --which is adjusted to
25 degrees. Immediately after the braking, there is a severe impact with a 20 g filtered peak longitudinal acceleration, which normally requires airbag deployment. Based on the findings of this study, the injury sustained by a 5th percentile female or a 50th percentile female is about 5 times below the federal threshold. Therefore the output of the OSS is set to zero while the output of the SDM set to one due to the severity of the impact. Consequently, the output of the AND gate is zero and there is no airbag deployment. This seems to be the optimal decision --despite the severity of the crash-- since according to the research of Kress, Porta and Duma [8] the inflation induced injury for a driver located 4 inches away from the airbag is fatal.

6.2. Folding Toe Pan

In Experiment 2 we identified the toe pan as a major contributor to the amount of injury sustained by the driver. We saw that leaner toe pans result in slower pelvis stops and thus less upper body injury. A mechanical system can be devised which in case of a crash --involving an unbelted driver-- it uses a combination of pulleys to “pull” the toe pan and make it leaner. The system is mechanical in nature, and utilizes the extreme forces associated with a crash to deform the toe pan. Figure 6.2.1 illustrates graphically the system concept. In addition, soft deformable material can be placed on the toe pan so that it absorbs some of the energy of the lower body. To further reduce injury the same mechanical system that folds the toe pan can also retract the pedals and push the seat backwards.

![Figure 6.2.1. The “folding” toe pan](image-url)
6.3 Self adjusting steering wheel

In both Experiments 2 and 3 it was shown that the injury increases as the angle of the steering wheel --as this is defined in Figure 3.7.3-- decreases. Therefore, in vehicle models with an adjustable steering wheel a mechanical system can be devised which with a combination of linkages and pulleys can make the steering wheel leaner in the case of an accident. Once again the extreme forces associated with a car crash can be utilized in order to achieve the tilting of the steering wheel.

![Figure 6.3.1. The self adjusting steering wheel](image)

6.4 Adjustable pedals

Experiment 4 indicated clearly that the airbag needs room to inflate. Many drivers however position themselves dangerously close to the airbag in order to be able to reach car pedals. Adjustable car pedals would enable shorter drivers to position themselves further and higher. In an airbag deployment scenario, this would leave enough room for the airbag to inflate fully. In a non deployment scenario, a higher seating position would introduce windshield impact and as shown in Experiments 2 and 3 it would reduce the overall injury.
6.5 Anti submarining plate

The submarining effect explained in the analysis of Experiment 4 is more pronounced with smaller drivers. The combined interaction of the driver with the airbag and the kneebolster forces the driver under the steering column. The effect can be reduced by inserting a steel plate into the cushion of the seat. The hard surface will prevent the driver from being pushed down into the seat.

![Steel plate mounted on the seat structure](image)

Figure 6.5.1. The antisubmarining plate

6.6 Self retracting steering column.

As shown in Experiment 4, in order to maximize the restraining capabilities of the airbag and to minimize inflation induced injuries, the airbag needs room to expand. As it was discussed, it is often the case that drivers seated too close prevent the airbag from inflating properly. A possible solution to this problem is a mechanical system that retracts the steering column during the course of the accident creating thus more room for the airbag to inflate properly. Additionally, the retracted steering column will allow for more windshield contact.
Figure 6.6.1 The self retraction steering column
Chapter 7

Conclusions

7.1 Computer Simulation of Car Crashes with MADYMO

Computer simulation of car crashes is by far the most efficient and cost effective method to study occupant dynamics. The alternative is the careful study of high speed videos of actual car or sled crashes. A computer simulation offers the flexibility of selecting free variables and running series of simulations trying to understand how the free variable affects occupant dynamics. For example, one can run a series of simulations varying the angle of the toe pan or the coefficient of friction of the seat, in order to isolate and study the specific effect of the selected free variable to the occupant behavior.
On the other hand, a car crash is an enormously complex event and an accurate simulation of all aspects can be very hard to model and execute. In this study, structural deformations of the vehicle were not taken into account. Though MADYMO is not a reliable tool to predict the absolute value of the HIC for a certain crash accident scenario, it can be very helpful in providing insight about the different parameters that affect occupant injury and how they affect it.

7.2 Airbag deployment based on occupant displacement

One of the motives for this study was to examine the feasibility of an airbag deployment system based solely on the motion of the occupant. From the analysis of Experiment 1 it becomes apparent that this concept does not constitute a viable solution. As it was shown, the displacement of the occupant during the first stages of the crash is very small in order to accurately describe the crash event and determine airbag deployment.

7.3 Evaluation of the freebody algorithm

The freebody algorithm is a very efficient and effective way to calculate a “worst case scenario” occupant displacement during a frontal crash. Though it is a crude estimate, it predicts accurately the motion of the occupant at least during the first stages of the crash. As it was shown in the analysis of Experiment 1, the freebody algorithm compares favorably with the MADYMO prediction. The same analysis also suggests that much attention should be paid to the acceleration profile used in conjunction with the freebody algorithm.

7.4 Variable deployment time

It is very hard to quantify the effects of varying the airbag deployment time. As shown in the analysis of Experiment 4 the interaction between the driver and the airbag is a very complex process which is difficult to model and predict. It was shown that there is a strong injury
dependence on the first stages of the airbag-driver interaction. However for a certain set of conditions presented in the analysis of Experiment 4, it appears that earlier deployment times are more beneficial in terms of the injury sustained by the driver.

### 7.5 Occupant dynamics

The series of experiments conducted revealed some key relationships between various parameters that affect the behavior of the occupant during a car crash. It was shown that:

- fast pelvis stops result in high head injury
- the windshield often behaves as a restraining device reducing head injury
- steeper steering wheel angles (defined in the setup of Experiments 2 and 3) result in higher combined head and chest injury
- in frontal crashes, the injury contribution of the roof is secondary compared to the primary source of injury (usually the steering wheel or instrument panel)
- steeper toe pan angles result in higher upper body injury
- the knee bolster contributes to the “submarining effect”
- the seat cushioning also plays a key role in the “submarining effect”
- during the initial stages of a car crash the head lags the chest
- the injury is proportional to the distance from the steering wheel (as this was defined in Experiment 3)
- the injury is also proportional to the linear velocity of the driver at the time of impact (as this was defined in Experiment 2)
- airbag deployment contributes to the submarining effect (Experiment 4)
- injury depends on the initial airbag-driver interactions which are very complex
Chapter 8

Summary

A set of four experiments were designed in order to answer specific questions associated with occupant behavior during a car crash. The four core experiments were used as the vehicle to identify other parameters that contribute to the driver injury and affect his or her overall behavior. A multi-body model was set up to simulate the interior surfaces of a mid-size passenger car. In addition, a finite element model of a driver airbag was devised in order to simulate airbag deployment and airbag-driver interactions. Standard models for various dummies were used from the TNO dummy library in order to cover a range of occupants. MADYMO’s multi-body and finite element modules were used to conduct all simulations and interpret the results.
The injury contribution of various parameters associated with vehicle design was identified and quantified to a first degree. A comparative analysis between the multibody simulation and the freebody occupant displacement algorithm verified the algorithm and presented arguments about its validity. A correlation between the occupant's linear velocity at the time of impact and the driver's injury was shown as well as between the driver's distance from the steering wheel and his or her injury. An attempt was made to quantify the effects of variable airbag deployment time which did not yield a definite correlation, but exposed the complexity of driver-airbag interactions and their contribution to injury.

Finally, the findings of the study were used to present an occupant detection system that promises to reduce inflation induced injuries. The analysis of the study was also used to present a series of additional restraining systems which promise to reduce injury associated with driver-vehicle interaction. Such systems include a self adjusting steering wheel, a folding deformable toe-pan, a steel plate into the cushion of the seat and a retractable steering column.

The use of computer simulation for the study of occupant dynamics was assessed, and some of the limitations of MADYMO in simulating contact interactions and occupant airbag interactions were exposed. Where applicable classic mechanics were used to justify the results.

The Head Injury Criterion and the G 3ms injury criteria were used along with the Federal Regulations to assess the injury sustained by the driver and characterize its severity.

Future research should investigate occupant dynamics for belted drivers and expand the current work for passengers.
Appendix I

Injury Criteria

A.1 Injury Parameters

The field of injury biomechanics deals with the effect on the human body of mechanical loads, in particular impact loads. Due to the mechanical load, a body region will experience mechanical and physiological changes, the so called biomechanical response. Injury will take place if the biomechanical response is of such nature that the biological system deforms beyond a recoverable limit. The mechanism involved is called injury mechanism. The severity of the resulting injury is indicated by the expression injury severity. An injury criterion is a physical parameter or a function of several physical parameters which correlates well with the injury severity of the body region under consideration.
Many injury criteria are based on accelerations, forces, displacement and velocities. These quantities can be obtained with the standard features offered by MADYMO.

In order to assess the upper body injury, this study utilized the acceleration based HIC and 3ms criteria. Both criteria are carried out on the linear acceleration signals of the head and the sternum respectively.

A.2 Head Injury Criterion (HIC)

The Head Injury Criterion (HIC) is defined as follows:

$$HIC = \max_{T_0 \leq t_1 \leq T_E} \left[ \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} R(t) \, dt \right]^{2.5}$$

where $T_0$ is the starting time of the simulation, $T_E$ is the end time of the simulation, $R(t)$ is the resultant head acceleration in g’s (measured at the head’s center of gravity) over the time interval $T_0 < t < T_E$, $t_1$ and $t_2$ are the initial and final times (in s) of the interval during which the HIC attains a maximum value.

A value of 1000 is specified for the HIC as concussion tolerance level in frontal (contact) impact. For practical reasons, the maximum time interval $(t_2 - t_1)$ which is considered to give appropriate HIC values was set to 36 ms.

Some limitations of the HIC include:
- HIC only considers linear acceleration, while biomechanical response of the head also includes angular motion which is believed to cause head injury,
- HIC is only valid if hard contact occur, thus the time duration of the impact is limited
- The value of HIC is greatly affected by the time interval used for its calculation.

Despite some of the drawbacks, HIC is the most commonly used criterion for head injury in automotive research and is believed to be an...
appropriate discriminator between contact and non-contact impact response.

A.3 3ms criterion (3MS)

The thorax contains, after the head, the next most critical organs to protect from injuries. The bony cage structure of the thorax consists of twelve thoracic vertebrae, the sternum and twelve pairs of ribs which form a relatively rigid though movable shell.

A commonly stated human tolerance level for severe chest injury (AIS>4) is maximum linear acceleration in the center of gravity of the upper thorax of 60 g, sustained for 3 ms or longer. Thus, the criterion is not based on a single maximum value but on a sustainable level of linear acceleration.
Appendix II

A MADYMO input file

B.1 A representative MADYMO input file

The following code is a representative input file for a MADYMO simulation.

!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!
!EXPERIMENT TO DETERMINE INJURY VS DISTANCE
!FILE: ../AIRBAG/5TH/DATA9.DAT
!MID SEVERITY FRONTAL CRASH 20 G MAX
!MID SIZE CAR
!IOANNIS HARRIPOULOS
!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!!
ONE UNRESTRAINED DRIVER
57TH FEMALE DRIVER
OCTOBER 1996 MIT 6THG

Simulation Duration
0 0.24
RUKU4 0.0001 0.001 0
0 0.5 0.01 0.1
INERTIAL SPACE
COMPACT VEHICLE ELLIPSOIDS
0 0 0.020000 0.200000 0.200000 -2.650000 0.500000 0.820000 2.000000 1 0 0
STEERING WHEEL
FUNCTIONS
2
0.000000 0.000000
0.500000 50000.000000
END FUNCTIONS
ORIENTATIONS
1
0 1 2 0.610900
END ORIENTATIONS
PLANES
0 -3.231739 0.849999 0.941782 -3.231739 -0.850001 0.941782 -3.021739 -0.850001 0.431782 1
CAR SEAT BACK
0 -2.200000 0.850000 0.200000 -2.200000 -0.850000 0.200000 -1.600000 -0.850000 0.200000 3
CAR SEAT CUSHION
0 -2.560000 -0.850000 0.800000 -2.560000 -0.850000 0.800000 -2.560000 -0.850000 0.800000 5
FLOOR
0 -3.200000 0.850000 0.200000 -3.200000 -0.850000 0.200000 -2.200000 -0.850000 0.200000 3
CAR SEAT CUSHION FRONT
0 -2.600000 -0.850000 0.800000 -2.600000 -0.850000 0.800000 -2.600000 -0.850000 0.800000 5
TOE PAN
0 -2.400000 -0.850000 0.550000 -2.600000 -0.850000 0.740000 -2.600000 0.850000 0.740000 4
BOLSTER
0 -0.090000 -0.850000 0.900000 -2.766000 -0.850000 1.296000 -2.766000 0.850000 1.296000 5
WINDSHIELD
0 -2.000000 -0.850000 0.500000 -2.000000 -0.850000 0.500000 -2.000000 -0.850000 0.500000 3
MIDDLE END PLANES
FUNCTIONS
* Loading Functions
* SEAT BACK
4
0.000000 0.000000
0.050000 100.000000
0.050000 100.000000
0.050000 100.000000
0.050000 100.000000
* SEAT CUSHION BACK
13
0.000000 0.000000
0.020000 197.199997
0.047100 394.399994
0.064000 592.000000
0.072000 789.000000
0.087500 986.400024
0.095300 1183.599976
0.101500 1381.000000
0.106500 1578.199951
0.111000 1775.400024
0.115400 1972.699951
0.127700 2367.199951
0.150000 1972.000000
* SEAT CUSHION FRONT
* FLOOR, TOE PAN,
2
0.000000 0.000000
0.050000 100.000000
* KNEE BOLSTER
6
0.000000 0.000000
0.024000 6409.000000
0.073000 788.000000
0.096000 788.000000
0.120000 1183.000000
0.160000 1183.000000
* W/S, I/P
2
0.000000 0.000000
0.050000 100.000000
END FUNCTIONS
END INERTIAL SPACE
SYSTEM
FIFTH HybridII
---TNO Hybrid II dummy--
---confidential data omitted---
INITIAL CONDITIONS
-2.018789 0.500000 0.527703 0.000000 0.000000 0.000000 1 -1 0
ORIENTATIONS
! reference position
! lower torso straight
-1 1 2 0.314160
! spine backward 5 deg.
0 0 1 2 -0.087270
! upper torso forward 18.7 deg.
3 0 1 2 0.326380
! neck bracket in zero offset position (frame forward 5.05 deg.)
4 0 1 2 0.088140
! head straight on neck
5 0 1 2 0.000000
! left clavicle backward 8.7 deg., dropped 9.5 deg.,
<table>
<thead>
<tr>
<th>6 0 1 2</th>
<th>-0.151840 1 -0.165810</th>
</tr>
</thead>
<tbody>
<tr>
<td>! right clavicle backward 8.7 deg., dropped 9.5 deg.,</td>
<td></td>
</tr>
<tr>
<td>7 0 1 2</td>
<td>-0.151840 1 0.165810</td>
</tr>
<tr>
<td>! left upperarm straight downwards</td>
<td></td>
</tr>
<tr>
<td>8 0 1 1</td>
<td>0.165810 2 -0.087270</td>
</tr>
<tr>
<td>! right upperarm straight downwards</td>
<td></td>
</tr>
<tr>
<td>9 0 1 1</td>
<td>-0.165810 2 -0.087270</td>
</tr>
<tr>
<td>! left lowerarm straight downwards</td>
<td></td>
</tr>
<tr>
<td>10 0 1 2</td>
<td>-1.200000</td>
</tr>
<tr>
<td>! right lowerarm straight downwards</td>
<td></td>
</tr>
<tr>
<td>11 0 1 2</td>
<td>-1.200000</td>
</tr>
<tr>
<td>! left hand straight downwards</td>
<td></td>
</tr>
<tr>
<td>12 0 1 2</td>
<td>0.000000</td>
</tr>
<tr>
<td>! right hand straight downwards</td>
<td></td>
</tr>
<tr>
<td>13 0 1 2</td>
<td>0.000000</td>
</tr>
<tr>
<td>! left upper leg straight forwards</td>
<td></td>
</tr>
<tr>
<td>14 0 1 2</td>
<td>0.139630</td>
</tr>
<tr>
<td>! right upper leg straight forwards</td>
<td></td>
</tr>
<tr>
<td>15 0 1 2</td>
<td>0.139630</td>
</tr>
<tr>
<td>! left lower leg straight downwards</td>
<td></td>
</tr>
<tr>
<td>16 0 1 2</td>
<td>-0.139630</td>
</tr>
<tr>
<td>! right lower leg straight downwards</td>
<td></td>
</tr>
<tr>
<td>17 0 1 2</td>
<td>-0.139630</td>
</tr>
<tr>
<td>! left foot dropped 17 deg.</td>
<td></td>
</tr>
<tr>
<td>18 0 1 2</td>
<td>-0.296710</td>
</tr>
<tr>
<td>! right foot dropped 17 deg.</td>
<td></td>
</tr>
<tr>
<td>19 0 1 2</td>
<td>-0.296710</td>
</tr>
<tr>
<td>! sternum straight (frame backward 13.7 deg.)</td>
<td></td>
</tr>
<tr>
<td>20 0 1 2</td>
<td>-0.239110</td>
</tr>
</tbody>
</table>

END ORIENTATIONS

END SYSTEM

FORCE MODELS

ACCELERATION FIELDS

FUNCTIONS

END ACCELERATION FIELDS

CONTACT

PLANE-ELLIPSOID

END PLANE-ELLIPSOID

ELLIPSOID-ELLIPSOID
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