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UNIVERSIDAD PONTIFICIA
ICAI

Master in Mobility and Security

Master Thesis Project

Prediction of Injury in the Spinal Columna due to Severe Axial Loading

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I declare, under my responsibility, that the Project presented with the title

PREDICTION OF INJURY IN THE SPINAL COLUMNA DUE TO SEVERE AXIAL LOADING

at the ETS de Ingeniería - ICAI de la Universidad Pontificia Comillas in the

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Predicción de lesiones en columna espinal debido a una carga axial grave

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RESUMEN DEL PROYECTO

Las lesiones causadas por explosiones originadas desde debajo del vehículo, conocidas como explosiones bajo el cuerpo (UBB, por sus siglas en inglés), aunque no son comunes, representan un riesgo para la seguridad tanto de los conductores como de los pasajeros, especialmente en vehículos militares. Hasta la fecha, se han llevado a cabo numerosos estudios sobre las lesiones en diversas partes del cuerpo resultantes de estas explosiones, sin embargo, varios aspectos aún deben ser investigados. Un criterio de lesión relacionado con la probabilidad de lesiones en la columna vertebral es el Índice de Respuesta Dinámica (DRI), pero aún carece de estudios más profundos. El objetivo de este proyecto es aclarar si el DRI es una métrica plausible para la probabilidad de lesiones en la columna vertebral basada en una carga axial. Esto se logra mediante la simulación de un modelo rígido de la columna vertebral al que se le aplica una carga axial en el sacro.

Introducción

A lo largo de los años, se han llevado a cabo numerosos experimentos para mejorar la seguridad automotriz y la prevención de lesiones. Estos han contribuido al desarrollo de muchas leyes de seguridad en todo el mundo, como limitaciones de velocidad en tráfico urbano, así como a crear conciencia sobre la importancia de un uso correcto del vehículo al conducir.

Sin embargo, existen ciertos incidentes que no pueden ser mejorados mediante legislación o concientización pública. Estos incidentes son deliberadamente causados por otras personas con el objetivo de dañar a los pasajeros del vehículo a través de medios como colisiones múltiples, lanzamiento de proyectiles hacia un vehículo o detonación explosiva desde debajo de un vehículo. Aunque estos incidentes no son de los más comunes, aún representan un gran riesgo para los pasajeros del vehículo, especialmente en zonas de guerra.

En el caso de los dispositivos explosivos desde abajo, ejercen una fuerza axial que resulta en lesiones en todo el cuerpo. Una de las áreas más afectadas debido a esta fuerza es la columna vertebral, la cual requiere un mayor entendimiento de sus riesgos de lesiones para poder aplicar las medidas de protección necesarias para mitigarlos.

El objetivo de este proyecto es desarrollar nuevas curvas de lesiones para la columna vertebral bajo una variedad de fuerzas axiales, como las causadas por una explosión. Esta investigación utiliza un modelo computacional rígido de la columna lumbar desarrollado en MSC Adams (MSC.Software) para simular la carga de un UBB y obtener las fuerzas y desplazamientos de cada parte de la columna lumbar. Luego, estos datos se utilizan para obtener el Índice de Respuesta Dinámica (DRI) de la columna vertebral y la probabilidad de lesión de una cierta parte de la columna vertebral basada en un artículo previo. Se comparará el DRI con cada riesgo de lesión para obtener una relación entre los valores del DRI y el riesgo de lesión en ciertas partes de la columna vertebral.

Metodología

El primer paso del proyecto es examinar proyectos anteriores en los cuales se haya estudiado la lesión de cierta parte de la columna vertebral cuando se aplica una explosión bajo el cuerpo (UBB, por sus siglas en inglés). Los siguientes proyectos han sido elegidos para ser examinados en este proyecto:

1. Yoganandan 2013 [1]: Este estudio revisa la tolerancia humana para UBB en el campo militar. En él se revisa un estudio de 12 PMHS de los cuales 7 sufren fractura cervical debido a la carga axial. Esto permite obtener una curva de riesgo de lesión cervical basada en el impacto de la fuerza pico.

2. Yoganandan 2018 [2]: Este estudio buscaba obtener una relación entre la edad y el riesgo de lesiones debido a un UBB. En este proyecto se establece un grupo de 20 PMHS de diferentes edades en posición vertical e intactos, posteriormente se les aplica una fuerza axial en T1 para obtener curvas de riesgo de lesión en la zona cervical según su edad y la fuerza aplicada. 13 de estos PMHS experimentaron fracturas en al menos una vértebra cervical.

3. Ortiz-Paparoni 2021 [3]: Este estudio se enfoca en el riesgo de lesión en la zona lumbar debido a un UBB. El estudio experimenta en 75 PMHS a los que se les aplica una carga de compresión desde abajo en diferentes posturas para determinar su respuesta a la lesión. De esos 75, 64 resultaron en al menos una fractura vertebral.

4. Stemper 2017 [4]: En este estudio se realizan un total de 38 pruebas en 23 especímenes. Las pruebas consisten en la aplicación de una carga axial en la columna lumbar de un PMHS, causando 26 fracturas y permitiendo el desarrollo de curvas de riesgo de lesión en la zona lumbar. Estas probabilidades de lesión dependen de la fuerza y la aceleración pico aplicada.

5. Bailey 2015 [5]: Este estudio buscó replicar las condiciones de carga de un UBB con fines de investigación. Para el experimento, se completaron 10 pruebas en PMHS donde los sujetos fueron sometidos a una carga axial desde debajo de la pelvis, lo que resultó en la fractura de 3 de ellos. Esto permite desarrollar curvas de riesgo de lesión para la zona de la pelvis basadas en la aceleración pico y la velocidad pico.

6. Rooks 2019 [6]: Este estudio analiza trabajos anteriores que involucran la probabilidad de lesiones en varias partes de la columna vertebral. La suma de todos estos trabajos anteriores tiene un total de 24 pruebas en PMHS, de las cuales 15 sufren lesiones por fractura y 9 no. Esto ayuda a desarrollar una curva de riesgo de lesión según la aceleración pico del sacro.

Se realiza un examen de las ecuaciones DRI y BRIC en cuanto a la obtención de ambos valores basados en la compresión de la columna vertebral o los momentos de la cabeza, respectivamente. Estas ecuaciones, así como las ecuaciones relacionadas con las curvas de riesgo de lesión de los documentos anteriores, se codifican en Matlab, donde se obtienen los resultados en función de ciertas entradas.

Estas entradas se obtienen de un modelo rígido sólido desarrollado por Lucas Low, que simula las propiedades de rigidez y amortiguamiento de una columna vertebral, como se observa en la Figure 1.

Figure 1: Modelo rígido desarrollado para la simulación

El modelo de la Figure 1 se somete a un movimiento de pulso triangular en el cual se aplica una carga axial mediante el aumento de la velocidad en el sacro con una aceleración constante hasta alcanzar un punto de velocidad pico, donde comienza a disminuir a la misma deceleración hasta llegar al final del pulso.

Se realizan un total de 58 simulaciones en las cuales se aplica un valor de velocidad pico y una duración total del pulso para cada caso, siendo el rango de velocidad pico entre 0.1 y 35 m/s, mientras que el rango de duración del pulso está entre 1 y 10 ms.

Cada simulación proporciona un valor entre 0 y 1 que se refiere a la probabilidad de lesión (siendo 0 ninguna y 1 probabilidad total) de ciertas partes del cuerpo. Los valores obtenidos en cada simulación son 11, de los cuales uno corresponde al DRI, otro al BRIC, otro a la lesión espinal desarrollada por Lucas y los 8 restantes a los trabajos mencionados previamente, de los cuales dos de ellos [4] [5] tienen dos valores cada uno según diferentes parámetros.

Resultados

Los resultados obtenidos se muestran en 10 gráficos, en los cuales el eje x corresponde a la probabilidad de lesión DRI y el eje y a los demás valores obtenidos.

Cada gráfico muestra una línea discontinua que sigue un camino lineal. Esta línea está destinada a separar los resultados que pueden considerarse correctos de los que no pueden considerarse correctos. Esto se debe a que los puntos ubicados sobre la línea de puntos significan que en esas simulaciones el riesgo de lesión del área específica que se analiza es superior al riesgo de lesión del DRI. Esto no puede ser correcto porque el riesgo de lesión del DRI debería ser superior a todos los riesgos de lesión anteriores, ya que todas las secciones analizadas en los trabajos anteriores son partes de la columna vertebral y, por lo tanto, para que tengan un riesgo de lesión, al menos debería ser el mismo que el de la columna vertebral.

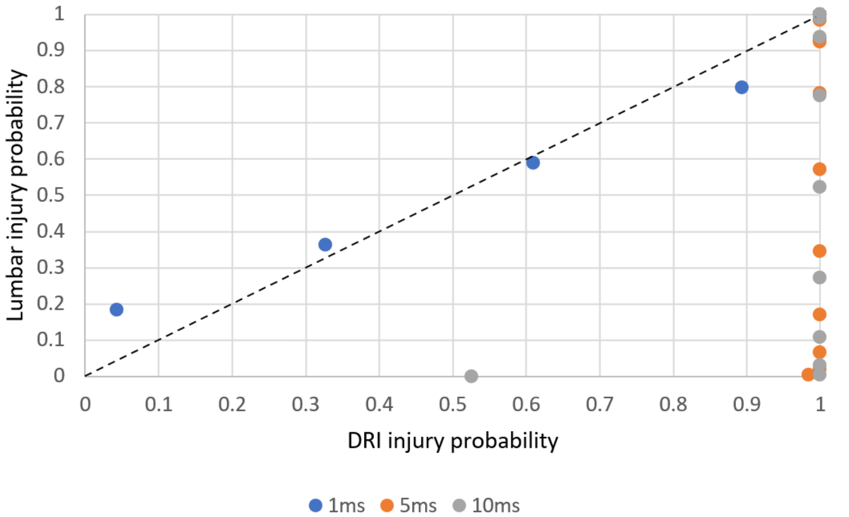

Figure 2: Gráfico de correlación entre la probabilidad de lesión de la zona lumbar y la probabilidad de lesión de la columna vertebral

En casos como el de Figure 2, se puede observar una correlación lineal entre el riesgo de lesión lumbar y el riesgo de lesión DRI cuando la duración total del pulso es de 1 ms, pero lo mismo no se aplica para 5 y 10 ms.

Conclusiones

Los valores de entrada aplicados para las simulaciones parecen ser demasiado excesivos para los resultados requeridos, ya que la mayoría de los valores de DRI obtenidos son de 1, lo que dificulta una posible correlación con los riesgos de lesión de los trabajos anteriores. Para una correlación más precisa, se deberían realizar más proyectos con un rango de velocidad pico entre 0.2 y 2 m/s o experimentar con un rango más amplio de duración total del pulso.

En cuanto a la correlación entre BRIC y DRI, podría existir alguna aunque la cabeza no forme parte de la columna vertebral, pero debido a su conexión con la columna cervical parece haber alguna correlación que se puede lograr a valores más bajos de velocidad pico cuando se aplica un pulso de duración de 10 ms, como se puede observar en Figure 3.

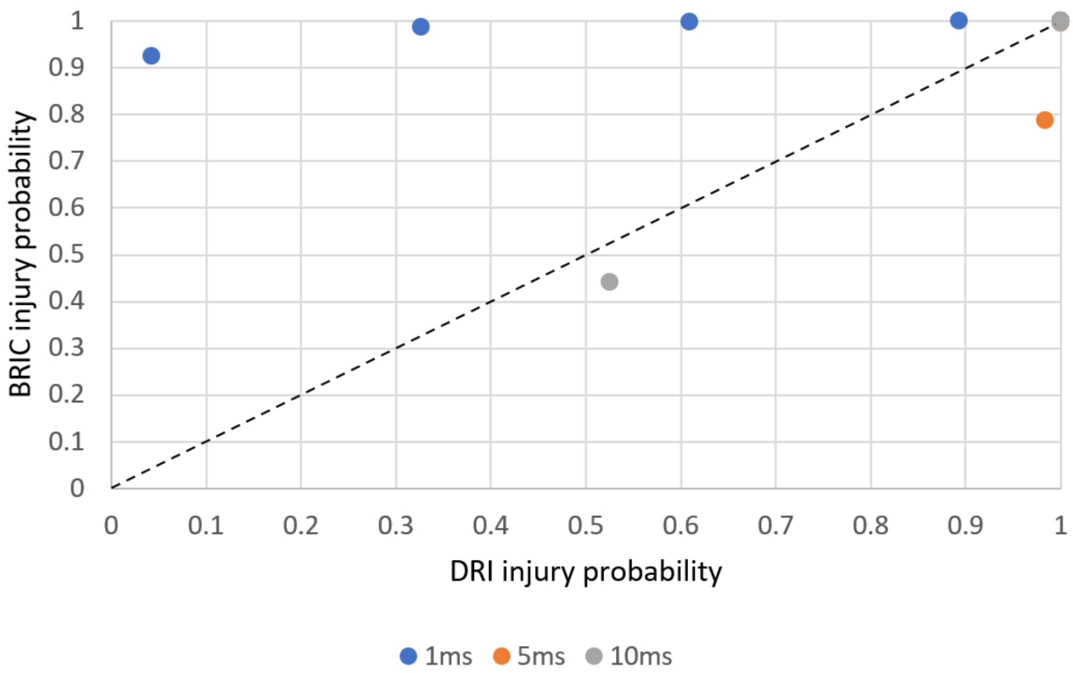

Figure 3: Gráfico de correlación entre la probabilidad de lesión de cabeza y la probabilidad de lesión de la columna vertebral

Prediction of injury in the spinal columna due to severe axial loading

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Abstract

Injuries caused by explosions originating from under body blasts (UBB) on a vehicle, although not common, pose a risk to the safety of both drivers and passengers, particularly in military vehicles. Numerous studies have been conducted to date on injuries to various body parts resulting from these blasts, yet several aspects remain to be investigated. An injury criteria that is related to spine injury probability is the Dynamic Response Index (DRI), but still lack of further study. The aim of this project is clarify if DRI is a plausible metric for spinal injury probability based on an axial loading. This is performed through the simulation of a rigid spinal model which has an axial loading applied on the sacrum.

Introduction

Throughout the years, many experiments have been conducted to improve automotive safety and injury prevention. They have helped developed many safety laws throughout the world such as speed limitation or urban traffic, as well as raising awareness of the importance of a correct use of the vehicle while driving.

However, there are certain incidents that cannot be improved through legislation or public awareness. These incidents are deliberately caused by other people with the objective to cause harm to vehicle passengers through means such as multiple vehicle collisions, projectile launching towards a vehicle or an explosive detonation from beneath a vehicle. Although these incidents are not among the most common, they still pose a great risk to vehicle passengers, especially in warfare areas.

In the case of explosive devices from beneath, they exert an axial force which results in injuries across the body. One of the most damaged areas due to this force is the spinal column, which requires further understanding in its injury risks in order to be able to apply the necessary protective measures to mitigate it.

The objective of this project is to develop new injury curves to the spinal column under a range of axial forces, such as those caused by an explosion. This investigation uses a rigid computational model of the lumbar spine developed in MSC Adams (MSC.Software) to simulate the load of a UBB and obtain the forces and displacements of each part of the lumbar spine. This data is then used to obtain the DRI of the spinal column and the injury probability of a certain part of the spinal column based on a previous paper. The DRI will be compared to each injury risk to obtain a relation between DRI values and the injury risk in certain parts of the spinal column.

Methodology

The first step of the project is to examine previous projects in which the injury of a certain part of the spinal column was studied when an UBB is applied. The following projects are chosen to be examined in this project:

1. Yoganandan 2013 [1]: This study reviews the human tolerance for UBB in the military field. In which it reviews a study of 12 PMHS of which 7 suffer cervical fracture due to the axial loading. This allows to obtain a cervical injury risk curve based on peak force impact.

2. Yoganandan 2018 [2]: This study sought to obtain a relation between age and injury risk due to an UBB. In this project a group of 20 PMHS of different ages are set upright and intact, afterwards an axial force is applied into them on T1 to obtain an injury risk curves on the cervical area depending on their ages and on the force applied. 13 of those PMHS experienced fractures in at least one cervical spine.

3. Ortiz-Paparoni 2021 [3]: This study focuses on injury risk in the lumbar area due to an UBB. The study experiments on 75 PMHS which where applied a loading compression from beneath at different postures to determine its injury response. From those 75, 64 resulted in at least one vertebral fracture.

4. Stemper 2017 [4]: In this study a total of 38 tests are performed on 23 specimens. The tests consist on the application of an axial loading onto the lumbar spine of a PMHS, causing 26 fractures and allowing for the development of injury risk curves in the lumbar area. These injury probabilities depend on the force and peak acceleration applied.

5. Bailey 2015 [5]: This study sought to replicate the loading conditions of an UBB for investigating purposes. For the experiment, 10 PMHS test are completed were the subjects are subjected to an axial loading from beneath the pelvis resulting in the fracture of 3 of them. This allows to develop injury risk curves for the pelvis area based on peak acceleration and peak velocity.

6. Rooks 2019 [6]: This study analyses previous works involving injury probability in several parts of the spinal column. The sum of all those previous works has a total of 24 PMHS tests of which 15 sustain fracture injuries and 9 don't. This helps develop an injury risk curve depending on the sacrum's peak acceleration.

An examination of DRI and BRIC equations is performed regarding the obtaining of both values based on the spinal column's compression or the head's moments respectively. These equations, as well as the equations regarding the injury risk curves from the previous papers is coded in Matlab were it gives the results based on certain inputs.

These inputs are obtained from a solid rigid model developed by Lucas Low, which simulates the stiffness and damping properties of a spinal column as observed in Figure 4.

Figure 4: Rigid model developed for the simulation

The model from Figure 4 is applied a triangular pulse motion in which a motion located on the sacrum creates an axial loading by increasing its velocity at a constant acceleration until it reaches a point of peak velocity where it starts to decrease at the same deceleration until it reaches the pulse's end.

A total of 58 simulations are performed in which a certain peak velocity and total duration pulse value are applied for each case, being the peak velocity's range between 0.1 and 35m/s, whereas the duration pulse ranges between 1 and 10ms.

Each simulation gives a value between 0 and 1 that refers to the probability of injury (being 0 none and 1 total probability) of certain parts of the body. The obtained values in each simulation are 11, belonging one of them to the DRI, another to the BRIC, another to the spinal injury developed by Lucas and the remaining 8 to the mentioned works previously done, of which two of them [4] [5] have two values each according to different parameters.

Results

The results obtained are shown in 10 graphs in which the x axis belongs to the DRI injury probability and the y axis to the rest of values obtained.

Each graph shows a dashed line that follows a linear path. This line is meant to separate the results that can be considered correct from those that cannot be correct.

This is due to the fact that the points located over the dash line mean that in those simulations the injury risk of the certain area that is analysed is superior to the injury risk of the DRI, this cannot be because the DRI injury risk should be superior to all the previous injury risks since all the sections analysed from previous papers are parts of the spinal column and therefore for them to have an injury risk it should at least be the same for the spinal column.

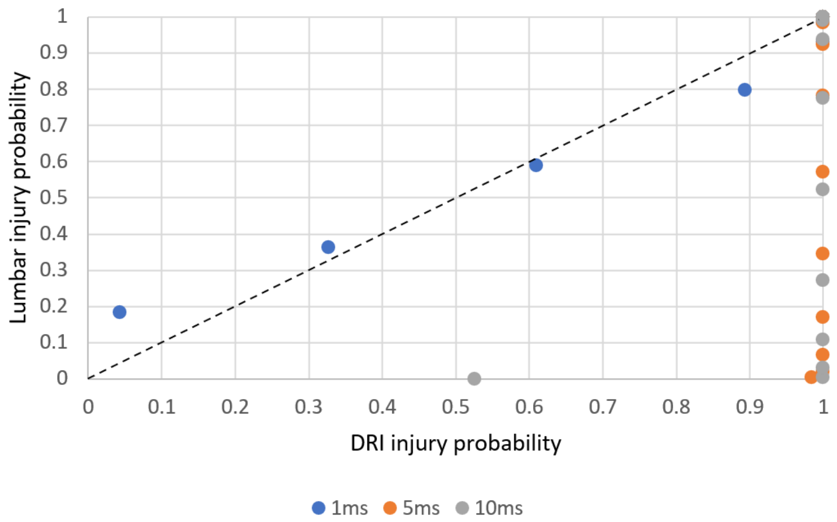

Figure 5: Graph of a correlation between DRI and lumbar injury

In cases such as Figure 5 a linear correlation can be observed between lumbar injury risk and DRI injury risk when the total duration pulse is of 1ms, but the same does not apply for 5 and 10ms.

Conclusions

The inputs applied for the simulations seem to be too excessive for the required results, since most of the DRI values obtained are of 1, complicating a possible correlation with the injury risks from previous papers. For a more precise correlation, further projects should be performed with a peak velocity range between 0.2 and 2 m/s or experience with a wider range of total duration pulse.

Regarding the correlation between BRIC and DRI, there could exist some even though the head is not part of the spinal column, but due to its connection to the cervical column there seems to be some correlation that can be achieved at lower peak velocity values when applied a 10ms duration pulse as it can be observed in Figure 6.

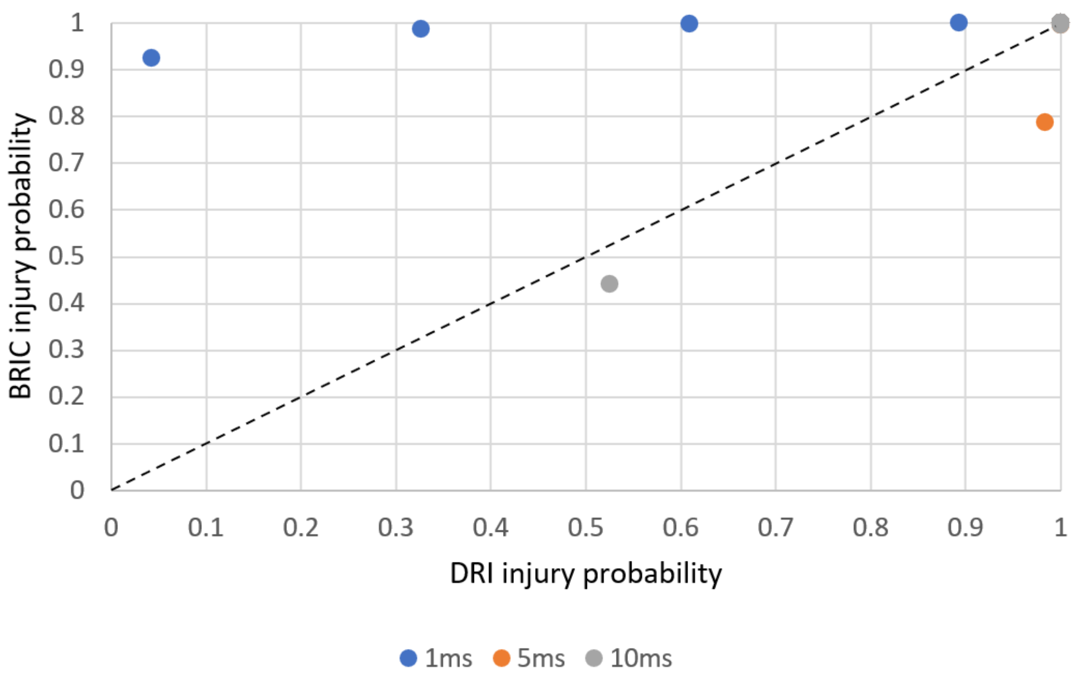

Figure 6: Correlation between DRI and BRIC

Gratitude

To my parents, for their unwavering support throughout my academic journey and for teaching me to persevere in pursuit of my goals.

To the professors who have made a significant impact on my academic path, not only in terms of education but also in my overall development.

To my siblings and closest friends, for being there in both the highs and lows of my academic journey and for their unconditional support.

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Chapter 1

Introduction

Ever since the creation of combustion engines, the incidents related to vehicle use has been a subject of concern to both developers and the public eye. Since then, motor vehicle collisions have become common, resulting in injuries for both the occupants and the VRUs.

Throughout the years, many experiments have been conducted to improve automotive safety and injury prevention. They have helped developed many safety laws throughout the world such as speed limitation or urban traffic, as well as raising awareness of the importance of a correct use of the vehicle while driving.

However, there are certain incidents that cannot be improved through legislation or public awareness. These incidents are deliberately caused by other people with the objective to cause harm to vehicle passengers through means such as multiple vehicle collisions, projectile launching towards a vehicle or an explosive detonation from beneath a vehicle. Although these incidents are not among the most common, they still pose a great risk to vehicle passengers, especially in warfare areas.

In the case of explosive devices from beneath, they exert an axial force which results in injuries across the body. One of the most damaged areas due to this force is the spinal column, which requires further understanding in its injury risks in order to be able to apply the necessary protective measures to mitigate it.

The objective of this project is to develop new injury curves to the spinal column under a range of axial forces, such as those caused by an explosion. This investigation uses a rigid computational model of the lumbar spine developed in MSC Adams (MSC.Software) to simulate the load of a UBB and obtain the forces and displacements of each part of the lumbar spine. This data is then used to obtain the DRI of the spinal column and the injury probability of a certain part of the spinal column based on a previous paper. The DRI will be compared to each injury risk to obtain a possible relation between DRI values and the injury risk in certain parts of the spinal column, which will help determine if DRI is a plausible option for measuring the probability of spinal injury due to UBB.

1.1 Motivation

The driving force of this project is to obtain enough data to perform a breakthrough in injury analysis related to the spinal column. Some other crucial factors for the motivation of this project are:

Human welfare: Severe axial loading incidents are a potential cause for life threatening spinal injuries, as well as physical, emotional and socioeconomically aspects if the injury acquires a certain severity and is not properly treated afterwards. The study of velocity, forces and accelerations of those incidents can contribute to the development of a more equipped vehicle or safety gear.

Injury avoidance: The knowledge obtained from this study can also help develop safety protocols to minimize the damage caused by an axial impact on the lumbar area.

Technological advancements: The project also contributes to the development of simulation techniques. The use of a simulation software such as MSC Adams has not been fully applied for experiments such as this and it can be a breaking point in the development of software predictions through this type of rigid models.

In summary, the motivation driving this project emanates from a steadfast commitment to preserving human well-being, enhancing safety measures, harnessing technological advancements, and yielding positive societal and economic impacts. By attaining a comprehensive understanding of the forces, accelerations, and injuries associated with severe axial loading on the spinal column, the project aims to contribute to injury prevention, improve safety standards, and ultimately effect meaningful change in the lives of individuals vulnerable to spinal injuries.

1.2 Memoir

The project has the following structure, which is the one specified for Universidad Pontificia de Comillas (ICAI).

The Introduction presents the main idea of the project and gives a brief summary of what it is going to be about. It also includes a Motivation section where it specifies the driving factors behind the involvement in this project, as well as the memoir which acts as the index making a summary of the whole project.

Afterwards, comes the State of the Art where all the previously known knowledge which has helped act as base of this project is commented. This is divided in three sections. The first one is regarding the structure of the vertebral column which summarises the different parts which will later be analysed. The second one explains the studies about injuries which are related to the topic of the subject, as well as, simulation studies and the injury risk studies in which this project will investigate. The third one is about the study of DRI and BRIC.

Later, the Project definition shows the project's objectives and the methods, applications and data that are used to develop the project and obtain the results.

In Model development it is explain the methodology of how the project is made, starting from the review of past projects, to the simulation and obtaining of results.

In Result analysis, the results obtained from the simulation are shown and briefly explained, dividing them into each of the areas that each of them correspond.

Future projects explains the possible projects that could be developed based on this project and the results that are obtained in it.

Finally, in Conclusions a more detailed explaining of the results obtained is done, focusing on the results of each part of the spinal column.

Chapter 2

State of the art

This section explains the studies previously made, as well as the projects from which the injury risk data of the spinal column sections is obtained.

2.1 Vertebral column

The vertebral column supports the body's physical structure and nervous system, enabling movement and sensation. Pathology of the spine can lead to debilitating outcomes on quality of life. The vertebral column (spine) defines the animal subphylum Vertebra, or vertebrates, of the phylum Chordata. In humans, it is composed of 33 vertebrae that include 7 cervical, 12 thoracic, 5 lumbar, 5 sacral, and 4 coccygeal. Along with the skull, ribs, and sternum, these vertebrae make up the axial skeletal system [7].

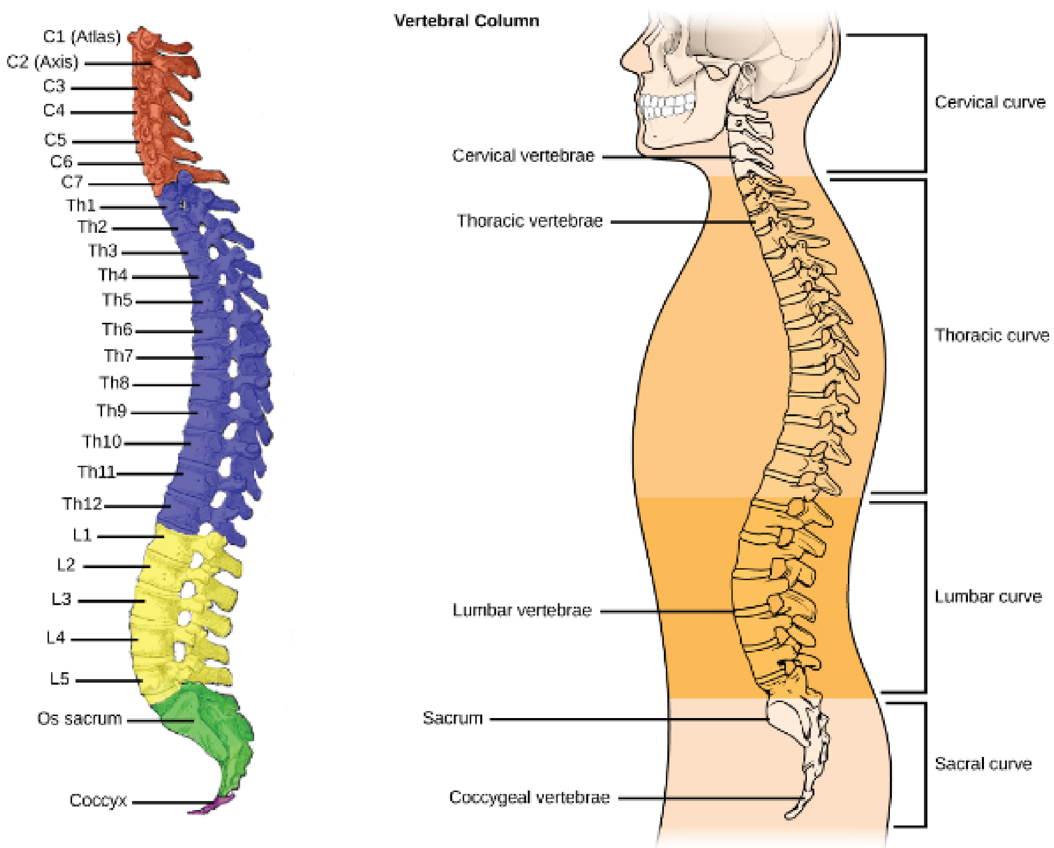


Figure 2.1: Vertebral column [8]

The spinal column forms the central axis of weight-bearing and supports the head as well as transfers the weight of the trunk and abdomen to the legs.

Its unique jointed structure provides bending and rotation, whereas the spine nerves provide attachments to the muscles and the thoracic region attachment sites to the ribs [7].

The spinal column is composed of three sections: lumbar, thoracic and cervical spine. Each of these sections' vertebrae is joined by inter-vertebral discs, which are cartilaginous structures composed of annulus fibrosus and nucleus pulposus. They make up 25% of the vertebral column and provide support to the anterior and posterior longitudinal ligaments [7].

2.1.1 Lumbar spine

The lumbar spine consists of the five bones (vertebra) located in the lower back. The lumbar vertebrae, known as L1 to L5, are the largest of the entire spine. They are located below the 12 chest (thoracic) vertebra and above the five fused bones that make up the triangular-shaped sacrum bone [9].

Compared with other spine vertebrae, the lumbar vertebrae are larger, thicker and more block-like bones. They provide stability for the back and spinal column and allow for a point of attachment for many muscles and ligaments. They also support most of the body's weight and are also the center of the body's balance. The lumbar spine and the muscle and ligaments that attach to them allow the human body to walk, run, sit, lift and move the body in all directions [9].

The spinal cord is a bundle of nerve tissue that extends from the lower part of the brain to about the L1 vertebra. It carries messages between the brain and muscles. The remaining nerve roots, called the cauda equina, descend down the rest of the spinal canal [9].

The lumbar spine is connected to the lumbar muscles which, along with the abdominal muscles, work to move the trunk and lower back. Muscles and ligaments provide strength and stability to the lower back and allow to bend forward, backward and rotate. The muscles that attach to the lumbar spine include: [9]

Latissimus dorsi: This is the large, flat, wide triangular-shaped muscle. It starts at the bottom of the sixth thoracic vertebrae and the last three or four ribs and covers the width of the middle and lower back. A portion of the latissimus attaches to the upper arms and the "lats" help you use the arms to pull up the body weight, help breathing by lifting the rib cage and help you bend to the side.

Iliopsoas: This three-muscle group moves the hip joint. The iliopsoas, one on each side of the body, flexes and stabilizes the hip and lower back as the body walks, runs and gets out of a chair.

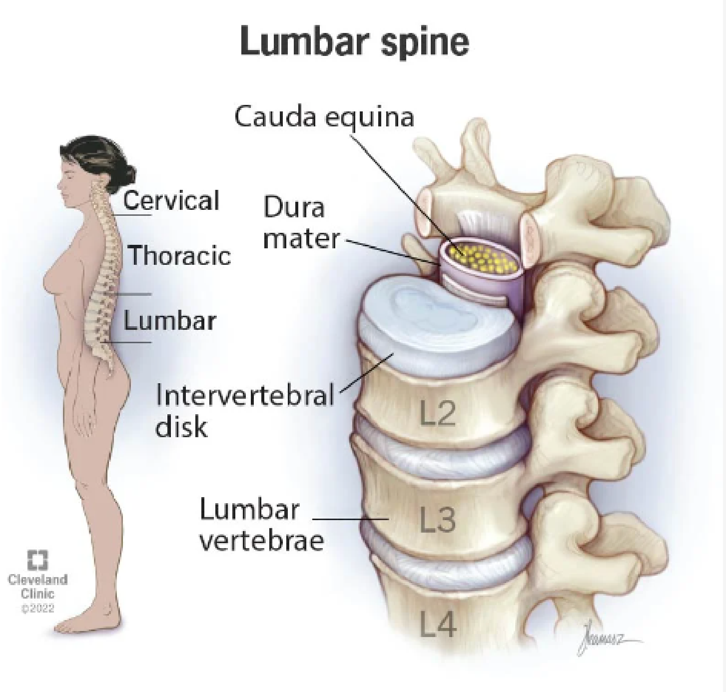


Figure 2.2: Lumbar spine [9].

Paraspinals: This group of three muscles is located along the length of the spine. These muscles help the body extend, side bend and rotate. They also help keep the body upright body posture.

2.1.2 Thoracic spine

The thoracic spine is the middle section of the spine. It starts at the base of the neck and ends at the bottom of the ribs. It's the longest section of the spine and it consists of 12 vertebrae, labeled T1 through T12 [10].

Vertebrae are the 33 individual, interlocking bones that form the spinal column. These bones help protect the spinal cord from injury while allowing the body to twist and turn. Between the vertebral bones are disks that provide cushioning for the vertebrae and flexibility for the body muscles [10].

The thoracic spine is also surrounded by muscles, nerves, tendons and ligaments that help with movement and flexibility. The spinal cord runs through the center of the entire spine. It sends and receives messages from the brain, which controls all aspects of the body's functions [10].

The thoracic spine has several important functions, including: [10]

Protecting the spinal cord and branching spinal nerves: The nerves of the spinal cord pass through a large hole (called the vertebral foramen) in the center of all of the vertebrae in the spine. Taken together, all the stacked vertebrae of the spine form a protective central canal that protects the spinal

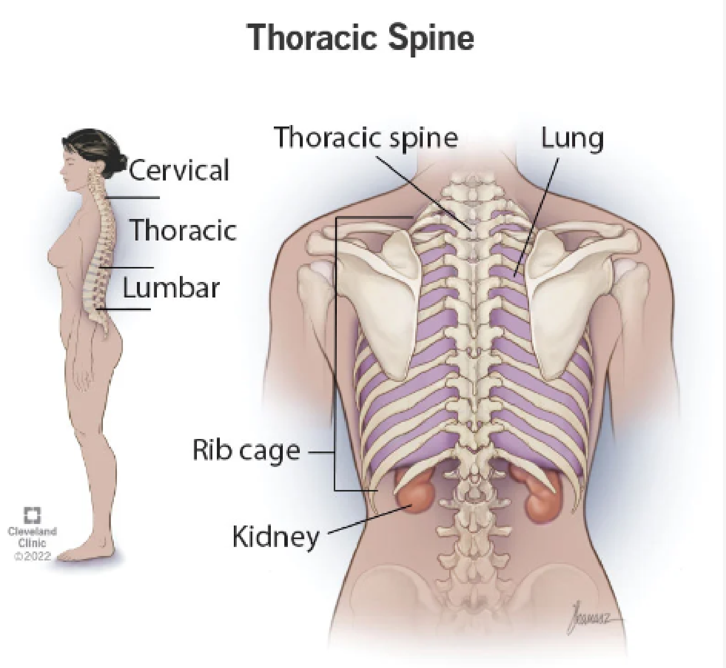


Figure 2.3: Thoracic spine [10].

cord.

Providing attachments for the ribs: Thoracic vertebrae are unique in that they have the role of providing attachments for the ribs, except for the two at the bottom of the ribcage.

Supporting the chest and abdomen: The thoracic spine helps stabilize the rib cage, and the rib cage, in turn, helps stabilize the thoracic spine. Together, the thoracic spine and ribcage protect the heart and lungs. The joints in the thoracic spine are tight enough to protect these vital organs, but loose enough to allow for the movements of breathing — inhaling and exhaling.

Allowing movement of the body: The soft inter-vertebral disks between the vertebrae make it possible for the body to twist and bend without sacrificing the supportive strength of the vertebral column. The joints in the thoracic spine allow the body to have the greatest range of rotation of the entire spine. However, the thoracic region has the least flexion, or extension, of the entire spine.

2.1.3 Cervical spine

The cervical spine — the neck area of the spine — consists of seven stacked bones called vertebrae. The first two vertebrae of the cervical spine are unique in shape and function. the first vertebra (C1), also called the atlas, is a ring-shaped bone that begins at the base of the skull. It's named after Atlas, of Greek mythology, who held the world on his shoulders. The atlas holds the head upright. the second vertebra (C2), also called the axis, allows the atlas to pivot against it for the side-to-side “no” rotation of the head [11].

The seven cervical vertebrae (C1 to C7) are connected at the back of the bone by a type of joint (called facet joints), which allow for the forward, backward and twisting motions of the neck [11].

The cervical spine is also surrounded by muscles, nerves, tendons and ligaments. “Shock-absorbing” disks, called intervertebral disks, are positioned between each vertebra. the spinal cord runs through the center of the entire spine. the spinal cord sends and receives messages from the brain, which controls all aspects of the body’s functions [11].

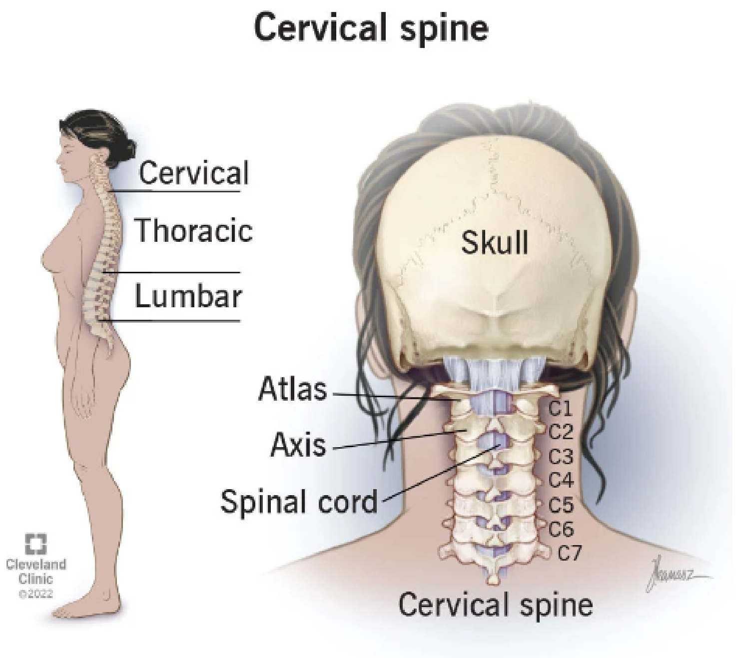


Figure 2.4: Cervical spine [11].

The cervical spine has several functions, including: [11]

Protecting the spinal cord: The nerves of the spinal cord pass through a large hole (called the vertebral foramen) that passes through the center of all of the vertebrae — from the base of the skull through the cervical vertebrae, the thoracic (middle back) vertebrae and ending between the first and second lumbar (lower back) vertebrae. Taken together, all the stacked vertebrae of the spine form a protective central canal that protects the spinal cord.

Supporting the head and allowing movement: The cervical spine supports the weight of the head (average weight of 10 to 13 pounds). It also allows the head and neck to tilt forward (flexion), backward (extension), turn from side to side (rotation) or bend to one side (ear-to-shoulder; lateral flexion).

Providing a safe passageway for vertebral arteries: Small holes in cervical spine vertebrae C1 to C6 provide a protective pathway for vertebral arteries to carry blood to the brain. This is the only section of vertebrae in the entire spine that contains holes in the bone to allow arteries to pass through.

The cervical spine is also connected to a series of muscles and ligaments. Among the muscles connected there are: [11]

Sternocleidomastoid: This muscle, one on each side of the neck, runs from behind the ear to the front of the neck. It attaches to the breast bone (sternum) and collarbone. This muscle allows the body to rotate the head side-to-side and tilt the chin upward.

Trapezius: This pair of triangular muscles extend from the base of the skull down the cervical and thoracic spine and out to the shoulder blade. They help tilt the head upward/move the neck backward, rotate the head right or left or lift the shoulder blade.

Levator scapulae: This muscle attaches to the first four cervical vertebrae and the top of the shoulder blade (scapula). It helps lift the shoulder blade, bend the head to the side and rotate the head.

Erector spinae: Several muscles make up this muscle group. In the cervical spine area, these muscles help with posture, neck rotation and backward neck extension.

Deep cervical flexors: These muscles run down the front of the cervical spine. They allow the body to flex the neck forward neck and help keep the cervical spine stable.

Suboccipital muscles: These four pairs of muscles connect the top of the cervical spine with the base of the skull. They allow the body to extend and rotate the head.

Whereas among the ligaments in the cervical spine which connect bone to bone to help to keep the cervical spine stable. There are three major cervical spine ligaments which are: [11]

Anterior longitudinal ligament: This ligament extends from the base of the skull, down the front of the cervical vertebra. It stretches to resist backward neck motion.

Posterior longitudinal ligament: This ligament starts at C2 and extends down the back of the cervical vertebrae. It stretches to resist forward neck motion.

Ligamentum flava: These ligaments line the backside of the inside opening of each vertebra where the spinal cord passes. These ligaments cover and protect the spinal cord from behind.

2.2 Injury studies

There are several paths to understanding the injuries of the spinal column and its consequences. They can be epidemiological, which result in the study of the patterns developed in society or at war that result in spinal injuries. They can also be done by conducting tests on human surrogates such as dummies, and PMHS to replicate as precise as possible the injuries of a live person. Finally, in the most recent years with the growth of technology and computational programs, computational surrogates are developed to simulate with various levels of precision the injuries.

2.2.1 Epidemiology studies

In 2015, Wojcik and her team arranged a study concerning the injuries in the spinal cord or vertebral column from military soldiers due to both the Afghanistan and Iraq wars. The data was acquired through the different hospitals which attended the injured soldiers [12]

Although the study mainly compares the data between the injuries caused in one war to the other, Wojcik deduced that spinal hospitalizations represented 8.2% of total injury admissions and that the amount of column injuries had severely increased in comparison to previous wars. It also noted the increase of paralysis due to those injuries being between 16 and 13% depending on the war battled [12].

That same year, Spurrier and his team developed a study to clarify if DRI is an appropriate variable to predict injury in the spine. They acquired data from previous papers that showed a total of 258 fractures in 189 patients due to blast injuries. The data was acquired with the authorization of the United Kingdom Ministry of Defence, as it was classified information related to conflict areas [13].

The model developed on this project showed two different patterns for injuries in blasts and aircrafts, proving that DRI was not valid for that project and that a new model had to be developed to be able to predict spinal injury due to UBB [13].

2.2.2 Experiments

The following studies regard experiments undergone to study injury risk curves on several body parts related to the spinal column. The studies are:

1. Yoganandan 2013 [1]: This study reviews the human tolerance for UBB in the military field. Based on previous studies where 13 PMHS were tested, this study performs a series of analysis centered around the injury risk of the cervical spine due to an explosive device from underneath. The project manages to obtain a risk injury curve of the cervical when a certain axial force is applied from beneath T1 as observed in Figure 2.5.

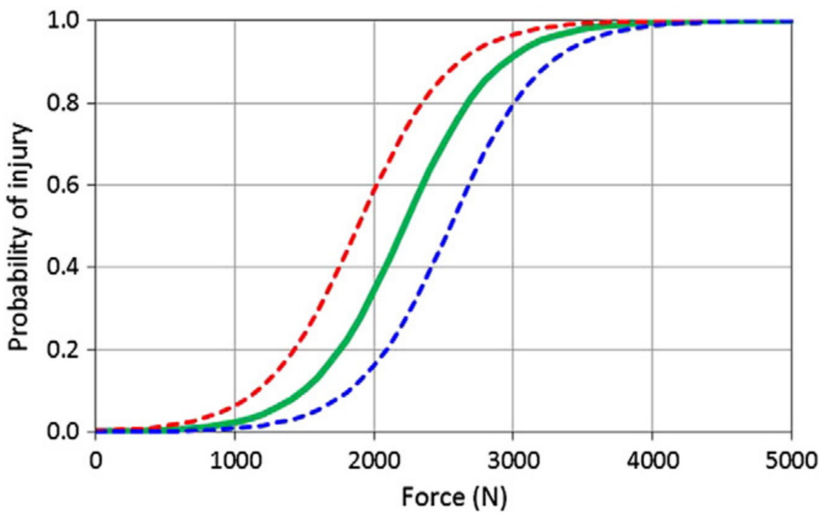


Figure 2.5: Injury risk curve on T1 based on peak force impact [1].

The injury risk probability follow a log-logistics distribution, an equation applied for survival analysis which follows a cumulative injury risk that increases the risk from 0 until it reaches a full injury risk after a certain value.

2. Yoganandan 2018 [2]: This study sought to obtain a relation between age and injury risk due to an UBB. In this project a group of PMHS of different ages are set upright and intact, afterwards an axial force is applied into them on T1 to obtain the following injury risk curves on the cervical area depending on their ages. This project aspires for a further study of axial loading on T1 from the data obtained in the previous study on Figure 2.5.

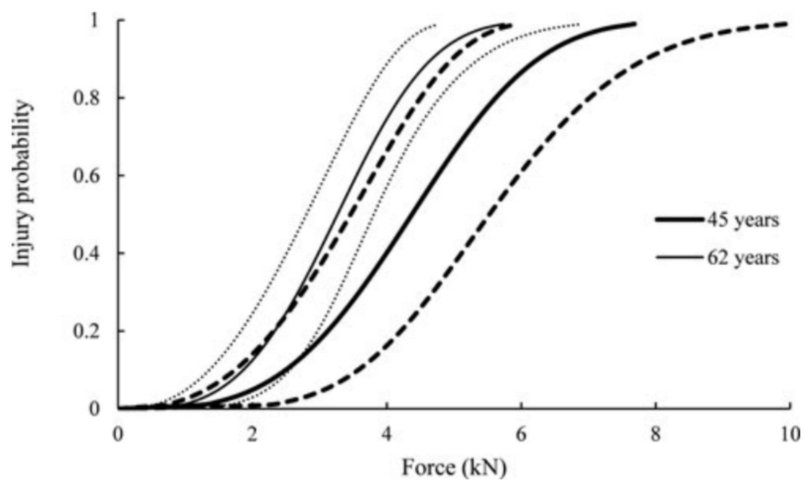


Figure 2.6: Injury risk curve on T1 based on peak force impact [2].

The curves from Figure 2.6, which are also determined by the log-logistic distribution such as in Figure 2.5, show a considerable increase in injury risk when age is applied, as it takes less axial force to cause the same injuries of 62 year old's than on 45 year old's with a higher axial UBB.

3. Ortiz-Paparoni 2021: This study focuses on injury risk in the lumbar area due to an UBB. The study experiments on 75 PMHS which where applied a loading compression from beneath at different postures to determine its injury response. From those 75, 64 resulted in at least one vertebral fracture. The injury risk is applied through a Weibull survival function which develops the injury risk curves of Figure 2.7.

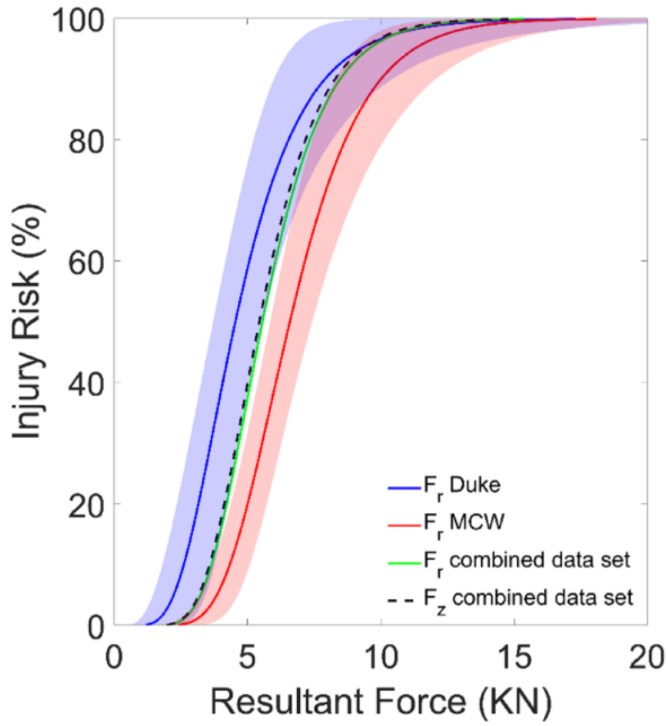


Figure 2.7: Lumbar injury risk caused by a UBB in the T12-L1 joint [3].

4. Stemper 2017 [4]: In this study a total of 38 tests are performed on 23 specimens. The tests consist on the application of an axial loading onto the lumbar spine of a PMHS, causing an axial loading which generated lumbar injuries, as it's shown in Figure 2.8.

   The experiment resulted in the development of injury risk curves related to axial force and peak acceleration, helping in the further understanding of biomechanical tolerances of the lumbar spine. L1 was the most affected section of the lumbar spine and the experiment provided no significant difference between male and female subjects [4].

5. Bailey 2015 [5]: This study sought to replicate the loading conditions of an UBB for investigating purposes. Ten PMHS test are completed were the subjects are subjected to an axial loading from beneath the pelvis. Afterwards, a series of injury curves are developed based on sacrum acceleration Figure 2.10a and sacrum velocity Figure 2.10b.

   Pelvis injury curves display the probability of suffering an injury on the pelvis according to the maximum acceleration or velocity value that is applied on the sacrum due to an UBB loading.

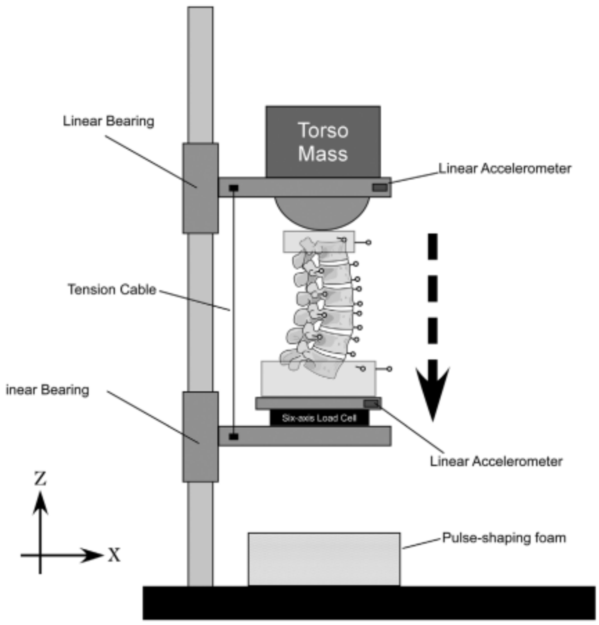


Figure 2.8: Vertical accelerator test setup to simulate high rate loading of the specimen. [4].

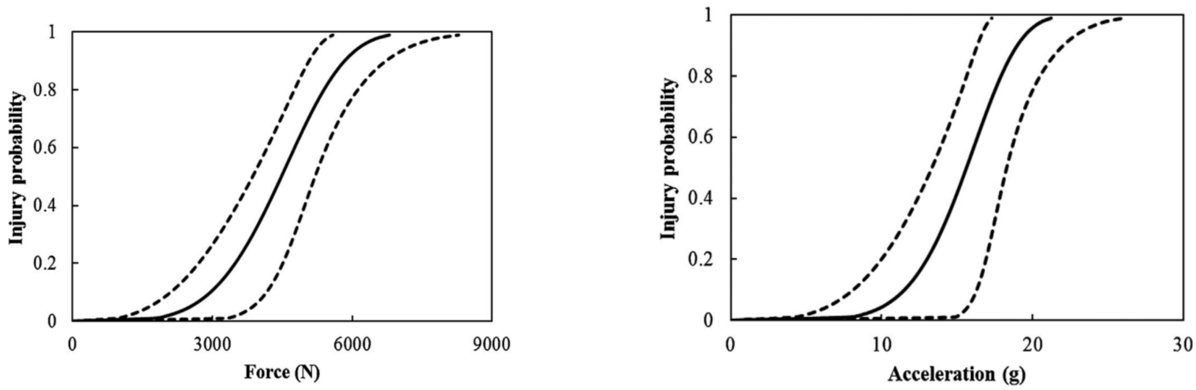
6. Rooks 2019 [6]: This study analyses previous works involving injury probability in several parts of the spinal column. The sum of all those previous works has a total of 24 PMHS tests of which 15 sustain injuries and 9 don't. This helps develops an pelvis injury probability based on the acceleration applied into the pelvis due to an UBB.

   Data from Figure 2.11 is divided into censored and uncensored injury probabilities, being the uncensored data only dependant on the 15 PMHS test subjects that were injured, while the uncensored includes the 9 subjects that were not injured.

2.2.3 Simulations

In 2016, Coogan et al. developed a finite-element model of lumbar intervertebral discs in LS-Dyna. The aim was to study the effects of axial forces upon L3 and L4 to design an optimal nucleus replacement device [14].

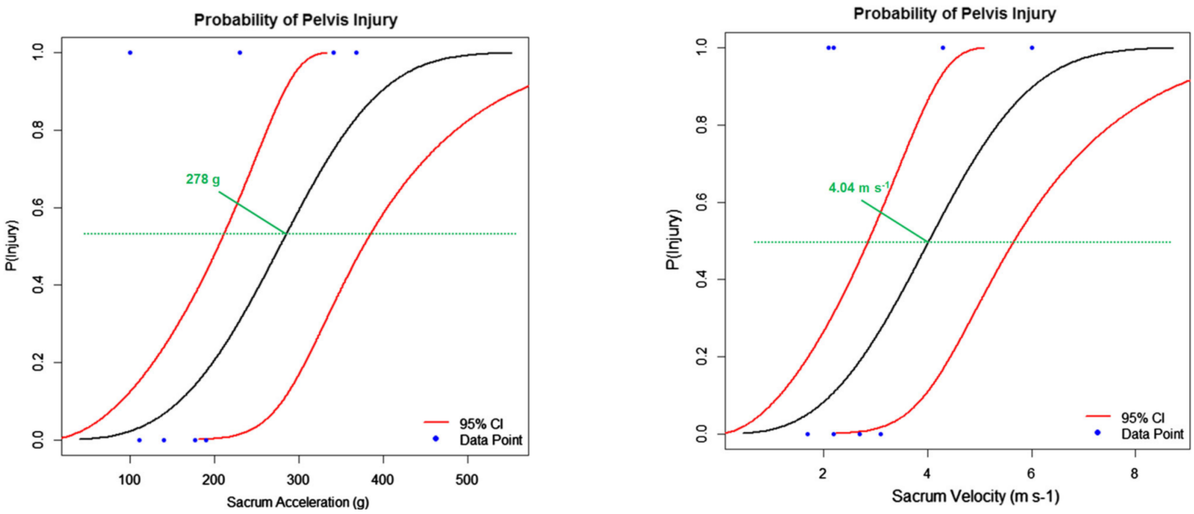
In 2023, Rebelo et al. developed a finite-element model to study the effect of stature and of mitigation systems on injury risk to the leg. This project developed a FE model of the lower limb based on a cadaveric lower limb anthropometry close to that of the 50th percentile American male, which was tested on two sets of UBB simulations. The first set incorporated the IMPAXX, a commercial foam for vehicle floors, foam placed between the loading plate that represented the vehicle's floor and the sole of the combat boot.The second set used two sets of statues, being one shorter and the other larger than the original, to study the effect of stature in lower limb injury as it can be observed in Figure 2.12.



(a) Force/injury curve

(b) Peak acceleration/injury curve

Figure 2.9: Injury risk curves regarding force and peak acceleration when axial loading is applied in the spinal column [4]



(a) Pelvis injury probability based on sacrum acceleration

(b) Pelvis injury probability based on sacrum velocity

Figure 2.10: Pelvis injury probability curves [5]

2.3 Injury criteria

Injury criteria are metrics used to establish the likelihood of a certain bone of the human body to fracture due to a variable such as a force, displacement, peak acceleration, etc. In this project, the two main injury criteria that are studied are Dynamic Response Index (DRI) and Brain Injury Criteria (BRIC).

2.3.1 Dynamic Response Index (DRI)

The Dynamic Response Index (DRI) is an injury prediction model which was first developed over 60 years ago to study the maximum dynamic compression of the spinal column when a vertical force is applied. In the recent years, the DRI has been studied for mili-

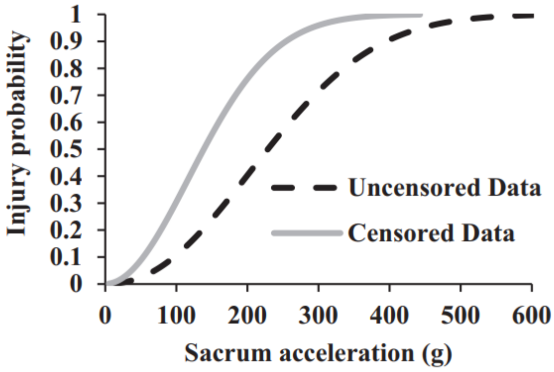


Figure 2.11: Injury risk curves for the censored and uncensored data analysis [6].

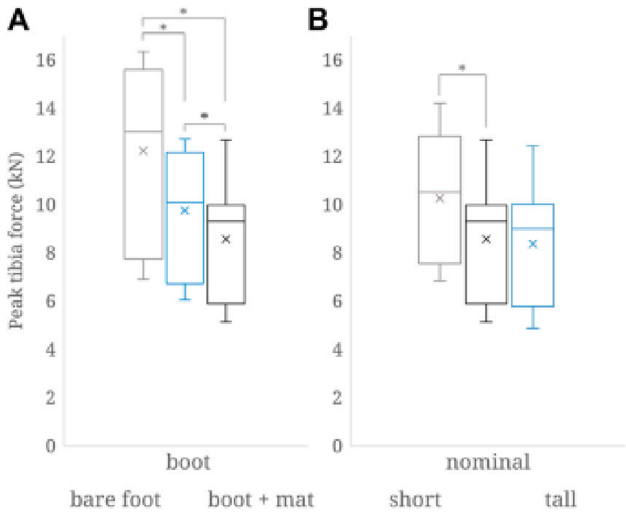


Figure 2.12: Comparison of peak tibia force distributions between (A) original baseline model and mitigation represented by the combat boot [15] and the IMPAXX floor mat; (B) short, tall, and the nominal stature all with a combat boot and the floor mat. Statistical significance ($p < .05$) using the Student's t-test between the distributions is noted with the asterisk [16]

tary ground vehicles to predict the injury level of the spinal column due to the effect of explosive devices.

To develop the DRI equation, a spinal model is created which consists of a single degree of freedom mass-spring-damper as seen in Figure 2.13. This spring's mass has a stiffness represented by the constant k, while the damping behavior of the body is captured by the damping coefficient c [17]. The equation obtained is the following:

$$\ddot{\delta} + 2\zeta\omega_n\dot{\delta} + \omega_n^2 + \delta = a_z \tag{2.1}$$

where:

$$
\begin{aligned}
a_z &\text{ is the acceleration-time input to the model} \\
\delta &\text{ is the relative displacement of the model - compression of the lumbar spine} \\
\omega_n &\text{ is the natural frequency of the system with } \omega_n = \sqrt{\frac{k}{m}} \\
\zeta &\text{ is the damping ratio with } \zeta = \frac{c}{2\sqrt{mk}}
\end{aligned}
\tag{2.2}
$$

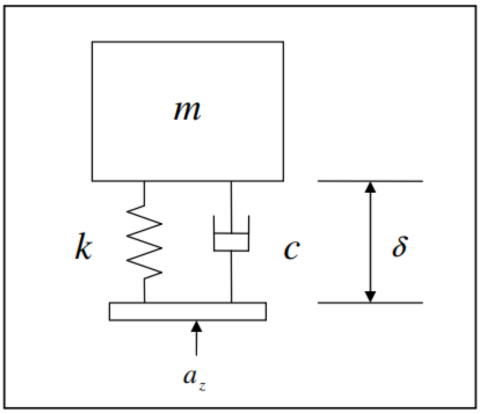


Figure 2.13: Single degree of freedom mass-spring-damper model of the human spine [17].

This motion equation was later developed in a study [17] by solving the differential equation through the application of a second order towards the relative displacement. This allows for the obtaining of a DRI equation based on maximum compression, undamped frequency and acceleration due to gravity, as seen in Equation 2.3.

$$
DRI = \frac{\omega_n^2 \delta_{max}}{g} \tag{2.3}
$$

It has also been developed a correlation between the DRI and the injury risk of a possible injury in any part of the spinal column, although this only works if the DRI values range between the values shown in Figure 2.14.

2.3.2 Brain Injury Criterion (BRIC

Since the 1940s, the rotational motion of the head as a brain injury has been proposed and studied. At first, experiments were made using animals as subjects, but later, through the development of dummies such as the THOR models, more breakthroughs were developed related to the brain injury [18].

The most recent research [18] proposed a model to obtain a brain injury criteria based on the angular velocity on all three axis and the critical value of max resultant angular velocity, which is the maximum angular velocity obtained in each test, as observed in Equation 2.4.

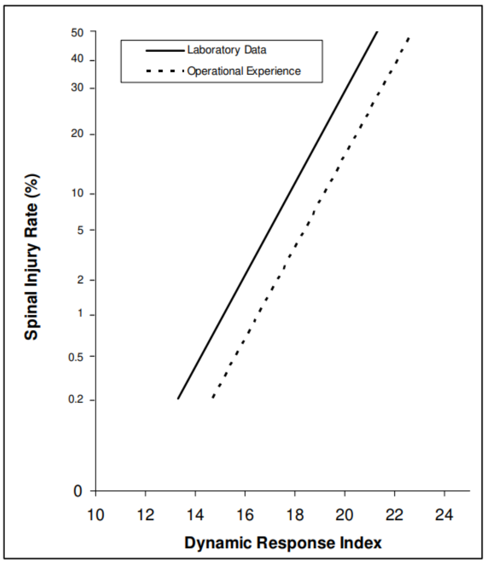


Figure 2.14: Probability of spinal injury estimated from laboratory data compared to operational experience [17].

$$BRIC = \sqrt{(\frac{\omega_x}{\omega_{xC}})^2 + (\frac{\omega_y}{\omega_{yC}})^2 + (\frac{\omega_z}{\omega_{zC}})^2} \tag{2.4}$$

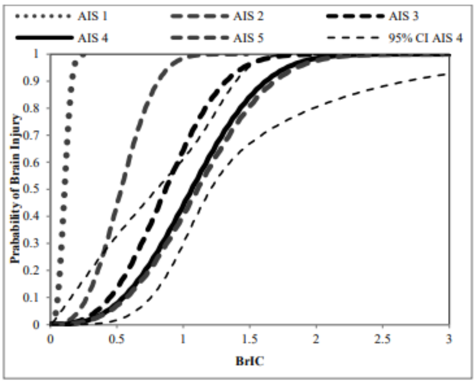


Figure 2.15: Injury risk curve based on Equation 2.4 [18].

Chapter 3

Project definition

3.1 Objectives

The main aim of this research project is to develop an extensive computational surrogate with the capacity to accurately predict spinal column injuries in the face of severe axial loading. Once this computational surrogate is developed, the objective is to use it to prove a correlation between DRI and other spinal injuries due to UBB by applying this model.

3.2 Method application

The data is obtained through the simulation of a solid rigid body in MSC Adams. The body is composed of a lumbar spine, which has the appropriate stiffness and dumping properties of human body. Afterwards, a motion is applied into the lower parts of the body to simulate an axial force. This motion follows a frequency pulse, accelerating constantly until it reaches a maximum velocity to then decrease back to rest.

3.3 Software

The development of this project requires the use of two softwares to obtain the required appropriate results, being these MSC Adams and Matlab.

3.3.1 MSC Adams

MSC Adams is the multibody dynamic simulation software used to perform the respective tests for the project. It can generate rigid bodies or export them from other similar multibodies. Those rigid bodies can experience several types of inputs such as forces or motions which modifies there trajectory and meshing displacement.

MSC Adams also allows to visualise the simulations and obtain various types of graphs such as acceleration, velocity or displacement.

3.3.2 Matlab

Matlab is a programming language capable of developing numerical computations and data analysis. It can also perform several functions such as matrix operations, data visualization, integration or simulations.

It can also import data from other computational softwares such as MSC Adams and perform operations and data analysis with it, which is the reason why it's used for this project, as it allows to obtain data in a much more optimized way than through MSC Adams.

3.4 Data

There are two types of previously developed data for this project: the rigid body model of a spinal column and the previous injury probability works.

The rigid body's geometry and its properties regarding stiffness and damping are developed by PHd student Lucas Low. It's composed of a full spinal column with its cervical, thoracic and lumbar spines, the skull on top and the sacrum beneath the inferior lumbar.

The data to which the DRI and BRIC are compared is obtained from papers previously published which study the injury probability of certain parts of the spinal column due to an UBB.

3.5 Planning

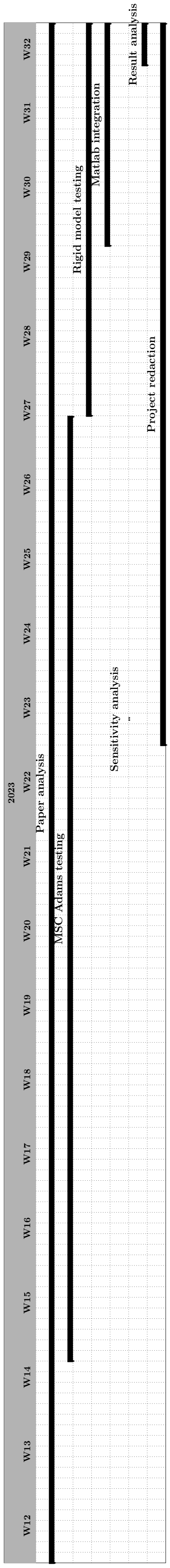
The project planning consists of several stages. The fist one is a paper analysis. In this stage projects and experiments related to injury analysis are searched and studied. At first the studies are related to any type of biomedical subject that relates to injuries due to vehicle crashing, but as it advances, it begins to specify in finding papers related to spinal column injuries due to an axial loading or vehicle crash.

The second stage begins when access to MSC Adams is granted. In this stage the objective is to practice with the Adams software and learn how to use its multiple applications as well as the creation and development of rigid bodies, even though they will not be used for this project.

The third stage starts once the rigid body is fully developed and given. A study in MSC Adams begins, as well as a study in the integration of Matlab with MSC Adams for simulation. This fourth stage also requires the development of the code needed to acquire the results from simulation.

After the code has been verified and tested, several simulation runs are performed under various inputs to obtain the desired outputs of the project. Once all the outputs have been obtained, they are analysed to obtain the desired hypothesis results.

When all the results have been properly documented and graphed, the project redaction can end.



(a) Project timeline

Chapter 4

Model development

This chapter explains the steps needed to fulfill the development of this project. It includes the process undergone from when the rigid model is obtain until the results are fully developed.

4.1 Past projects

Before the project development it is required to study past project which are related to injury probability in the spinal column due to an UBB. Each project performs a study either through research from previous projects or through the experimentation on PMHS. The studies observed in Table 4.1 show each paper's applied survival function as well as the parameter by which the injury curve depends. Even though all tests are due to UBB caused on the sacrum, the analysed section of the spinal column varies in each case.

Paper	Survival function	Spinal column section	Parameter
Yoganandan 2013	Log-logistic	Cervical	Resultant Force
Yoganandan 2018	Log-logistic	Cervical	Axial Force
Ortiz-Paparoni 2021	Weibull	Cervical	Axial Force
Stemper 2018	Weibull	Lumbar	Axial Force
Stemper 2018	Weibull	Lumbar	Acceleration
Bailey 2015	Weibull	Pelvis	Acceleration
Bailey 2015	Weibull	Pelvis	Velocity
Rooks 2019	Weibull	Pelvis	Acceleration

Table 4.1: Paper characteristics

4.2 MSC Adams

The rigid model previously mentioned is prepared to have a certain force applied in MSC Adams. For each simulation different input values will be applied so that it can obtain the desired outputs regarding its different body parts.

For every simulation of the process, a velocity motion is applied to the rigid model developed. To simulate the effect of an explosion located beneath the body, the motion is applied to the pelvis, following an axial direction in the Z coordinate upwards as seen in Figure 4.1.

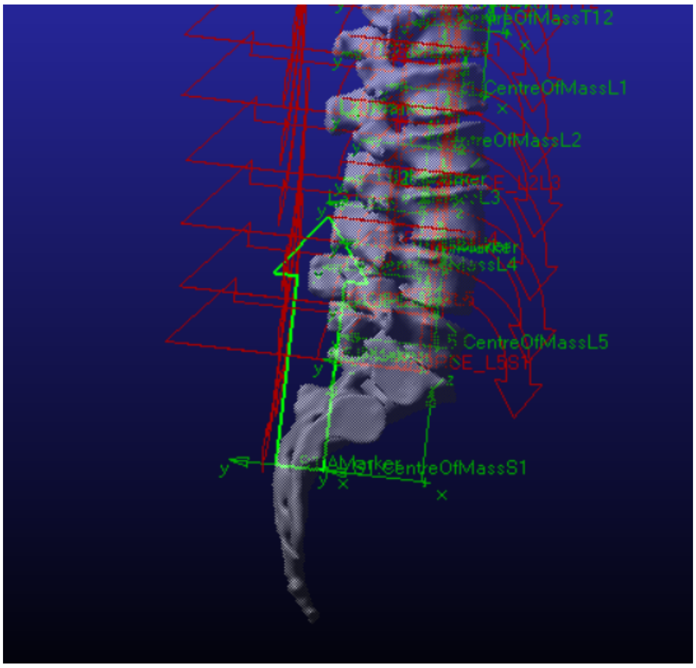


Figure 4.1: Velocity motion applied under pelvis

The acceleration force is simulated through a triangular pulse in velocity. A triangular pulse consists of a motion which increases at a constant trajectory until it reaches its maximum point. Once it does that, the motion decreases following the inverted path of the one followed before as shown in Figure 4.2.

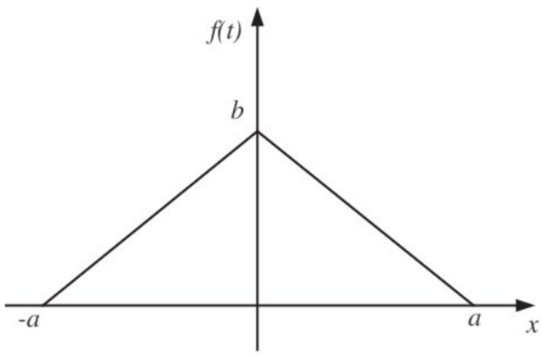


Figure 4.2: Triangular pulse function [19]

The triangular pulse seeks to simulate an acceleration on the Z axis by applying an increasing velocity on the designated place. This velocity motion forces the rigid body upwards, the displacement of the body parts vary depending on the maximum velocity value and the time it takes for that velocity value to be obtained from a resting position.

Although the rigid body commences on a static position, it has gravity forces applied to each of its parts. Must be also noted that during the simulation the body can translate and rotate in all rotations as it has no boundary conditions other than the positive force applied in Figure 4.1.

Once the data of for a simulation is established, it's exported into Matlab to undergo the simulation and data obtaining.

4.3 Matlab

Through Matlab, a code is run to acquire the desired values for each case. The values obtained in each case are:

4.3.1 DRI

A DRI value that represents spinal injury is obtained through a Matlab code by applying Equation 2.3. In this equation the natural frequency applied is a constant value of 52.9 rad/s, which is the same value that was applied in the DRI project [17], whereas the maximum compression value represents the minimum distance between the sacrum and the head through the simulation, since its the instant were peak compression is applied into the spine.

4.3.2 BRIC [18]

A BRIC value is obtained through a Matlab code by applying Equation 2.4. In this equation, the value ranges between 0 and 1, referencing the probability of cervical injury. The angular velocities of the equation are the angular velocities of the head in each axis, whereas the maximum critical angular velocities are values that remain constant being the followed in Table 4.2.

The critical angular velocities applied in Table 4.2 represent the injury probability of AIS 4 cervical injury, a logistic survival function as observed in Figure 4.3.

Critical Max. Angular Velocity	Rad/s
ω_x	66.25
ω_y	56.45
ω_z	42.87

Table 4.2: Critical max. angular velocity values

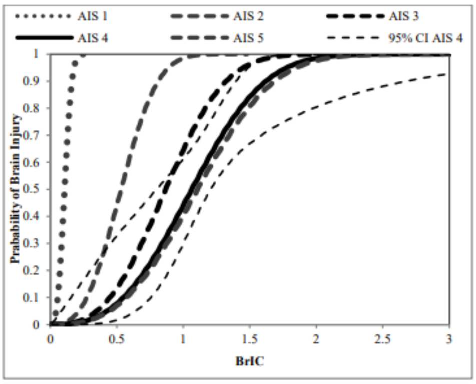


Figure 4.3: Injury risk curve based on Equation 2.4 [18].

4.3.3 Cervical injury risk due to axial force

The Matlab code gives a value obtained from an injury risk curve [1] that follows a log-logistic survival function. The experiment that obtained this values was performed with 13 PMHS which had an axial force applied at T1 to obtain injury risk values that extend from 0 to 1 when applied to axial forces in between 0.4 to 4.3 kN.

4.3.4 Cervical injury risk regarding age due to axial force

The Matlab code gives a value obtained from an injury risk curve [2] that follows a log-logistic survival function, which was obtained from PMHS tests of ages around 45 years old with an axial force applied at T1. The injury risk curves extend form 0 to 1 when applied forces in between 2 to 4.4 kN.

4.3.5 Cervical injury risk due to T12-L1 axial force

The Matlab code gives a value obtained from an injury risk curve [3] that follows a log-logistic survival function, which was obtained using 75 PMHS with an axial force applied between T12 and L1. The injury risk curves extend from 0 to 1 when applied to forces between 2 to 15 kN.

4.3.6 Lumbar injury risk curve due to peak force

The Matlab code gives a value obtained from an injury risk curve [4] that follows a Weibull survival function, which was obtained by applying an axial load on L5 from a lumbar spine of 23 PMHS. The injury risk curves extend from 0 to 1 when applied to forces between 2.1 to 7.3 kN.

4.3.7 Lumbar injury risk curve due to peak acceleration

The Matlab code gives a value obtained from an injury risk curve [4] that follows a Weibull survival function, which was obtained by applying an axial load on L5 from a lumbar spine of 23 PMHS. The injury risk curves extend from 0 to 1 when applied to peak accelerations between 8 to 23g.

4.3.8 Pelvis injury risk due to peak acceleration

The Matlab code gives a value obtained from an injury risk curve [5] that follows a Weibull survival function, which was obtained by applying an axial load on the sacrum of 10 PMHS. The injury risk curves extend from 0 to 1 when applied to peak accelerations from 20 to 600g.

4.3.9 Pelvis injury risk curve due to peak velocity

The Matlab code gives a value obtained from an injury risk curve [5] that follows a Weibull survival function, which was obtained by applying an axial load on the sacrum of 10 PMHS. The injury risk curves extend from 0 to 1 when applied to peak velocities from 0.3 to 9m/s.

4.3.10 Pelvis injury risk due to peak acceleration

The Matlab code gives a value obtained from an injury risk curve [6] that follows a Weibull survival function, which was obtained by applying an axial load on the sacrum of 24 PMHS. The injury risk curves extend from 0 to 1 when applied to peak accelerations from 0 to 500g.

4.3.11 Spine injury risk due to peak acceleration

The Matlab code gives an injury risk value between o and 1. It follows a Weibull survival value of the spinal column, which predicts the probability of a possible injury in any place of the spinal column due to the value of the peak acceleration of the sacrum because of UBB. This equation is developed by Lucas Low.

4.4 Sensitivity analysis

After all parameters are established the sensibility analysis commences. This analysis is composed of 58 simulation results. In each simulation, two parameters are modified to obtain the desired results:

1. Velocity (mm/s): Maximum velocity that the triangular pulse reaches during the simulation. Its range of values are from 100 to 30000mm/s.
2. Total pulse duration (s): Corresponds to the time at which the triangular pulse ends, which is when the velocity falls to 0. Its range of values are from 0.001 to 0.01s.

Even though all triangular pulses end at between 1 and 10ms depending on the time variables, the simulation ends at 20ms to allow for further displacement of the spinal column after the velocity ends.

In Figure 4.4 the inputs of all 58 simulations can be observed. For lower time values more peak velocity is needed to obtain an injury probability risk of 100% in each case, whereas for load pulses that require more time it is less needed to apply such high values, but instead many lower values are applied.

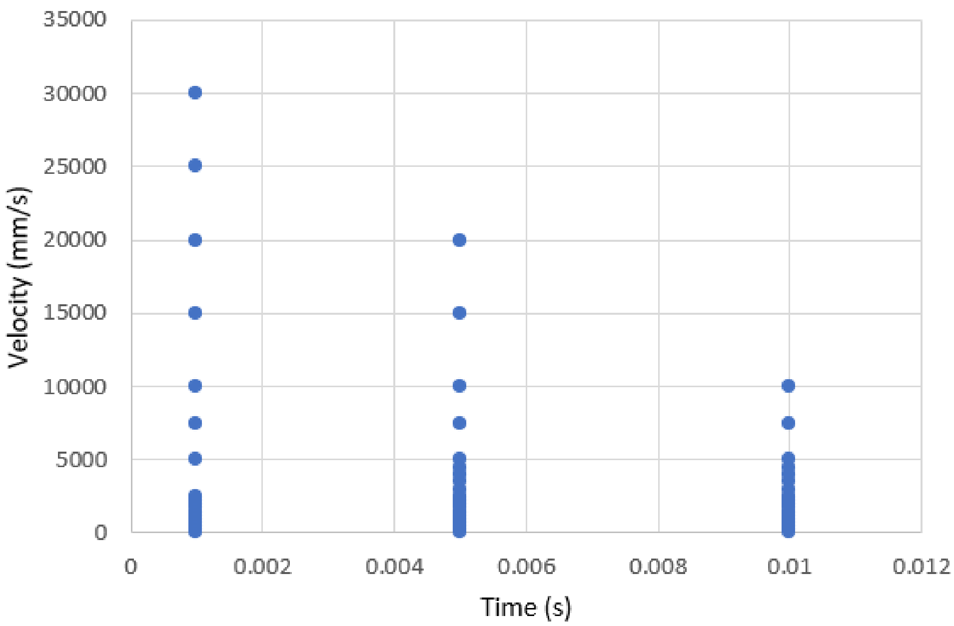


Figure 4.4: Velocity and time data of each simulation performed.

Chapter 5

Result analysis

This chapter displays the data obtained through the methodology previously mentioned. The data is represented in a series of graphs in which the x value is DRI which represents the probability of having a spinal injury and the y value is the obtained value of a certain injury risk criteria that represents the probability of a fracture in that certain area. Each point of the graph is the result of a simulation in which a certain peak velocity was a applied as well as a certain total pulse duration.

Each case displays the increase of the peak velocity when applying a certain pulse duration. Those pulse duration values are 1ms, 5ms and 10ms. The reason for distributing them this way is due to the fact that and increase of velocity at a constant pulse causes a direct increase in acceleration, which is a variable in which most of the original graphs are based on. However, since the total duration of the simulation is constantly of 20ms, the lower the pulse duration is the fewer the displacement will be compared to larger pulses.

The injury risk value of each graph is obtained from a different equation based on a previous projects and they are distributed according to the section of the spinal column to which they are related which is one of the following:

1. Head
2. Cervical
3. Lumbar
4. Pelvis
5. Spine

These graphs seek to obtain a relation between DRI and a probability of injury in each of the sections mentioned. The since the DRI shows the probability of injury in any place of the spinal column, the results should show a higher risk of injury in the DRI than in the rest of values, since the DRI involves all sections of the spinal column and the other parameter just one that can be the cervical, lumbar or pelvis.

5.1 Head area

Even though the head doesn't belong to the spinal column, its still attached to the cervical spine so a correlation between the BRIC [18] is established through Figure 5.1 to search for a possible injury correlation between DRI and BRIC.

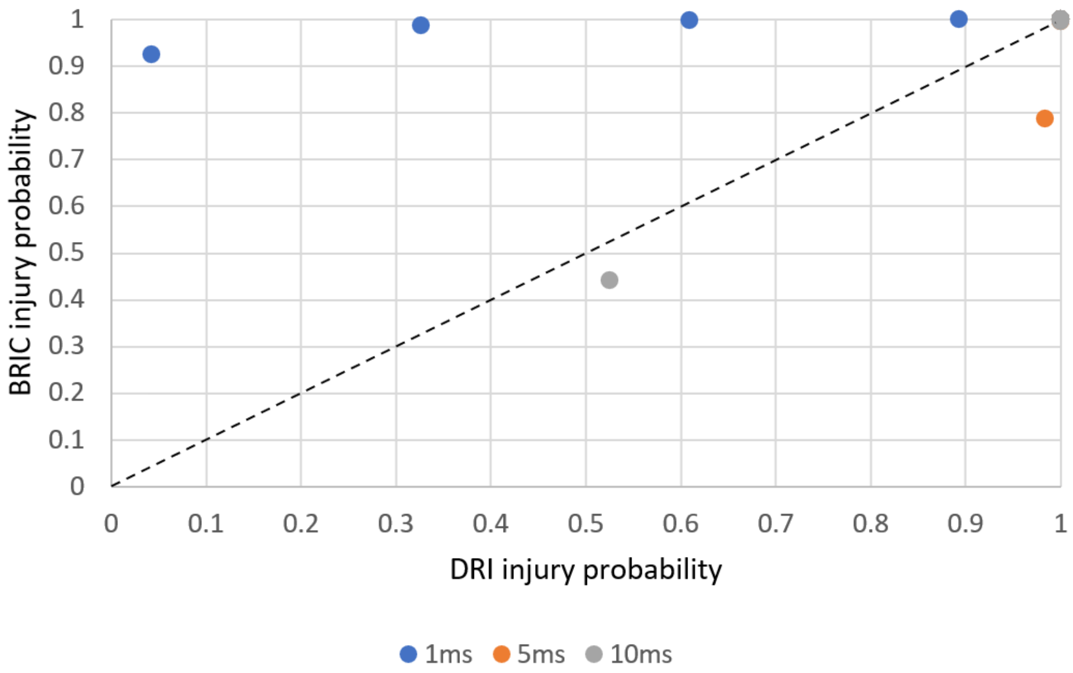


Figure 5.1: Correlation between DRI and BRIC

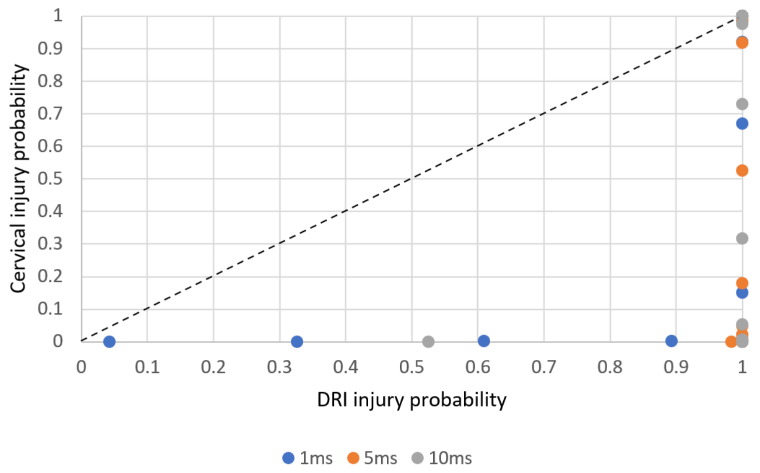
It can be observed that most of the values have acquired a total injury risk value of 1 for both BRIC and DRI. However, a value of 5ms and 10ms are located beneath the dashed line, meaning that those values indicate that there is a higher chance of having a spinal injury than a head injury. For values of 1ms the probability of getting a head injury seems to be always superior or equal to that of DRI, specially since in that pulse duration the BRIC values always give a higher chance of injury than 90%.

5.2 Cervical area

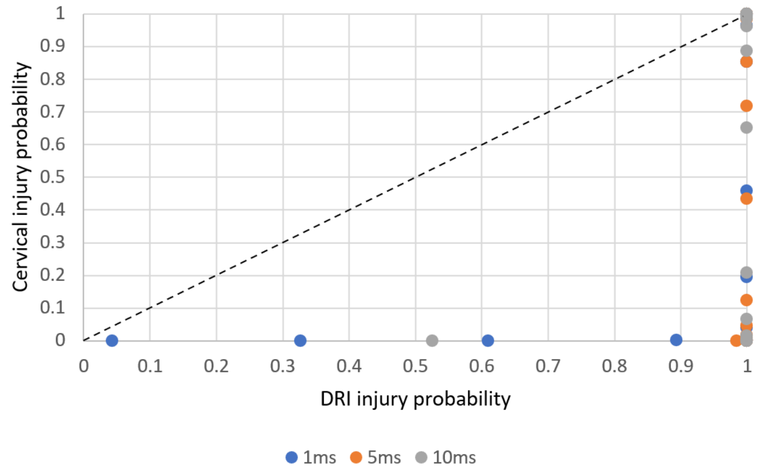
The cervical area is studied through the comparison between probability of spinal injury due to DRI values and the cervical injury due to three previously performed test runs [1] [2] [3].

For both graphs of Figure 5.2 the results are always below the dashed line meaning that the cervical injury probability is always lower or equal to the spinal injury probability. Most of the graphs' results present a total spinal injury probability although its cervical injury values vary slightly between Figure 5.2a and Figure 5.2b.

As for Figure 5.3 values seem quite similar to the previous graphs, although it can be noted that for 1ms as spinal injury increases, so does the cervical injury, but only to lower values and not reaching a 10% injury risk until it reaches a total spinal injury risk.



(a) Correlation between DRI and cervical injury



(b) Correlation between DRI and cervical injury

Figure 5.2: Injury risk curves for cervical area regarding axial force in T1

5.3 Lumbar area

The data regarding the lumbar area was obtained from an experiment [4] were 38 tests were performed and 26 resulted in fractures in the lumbar area.

Figure 5.4 regards lumbar injury risk based on the force of the UBB and it shows an linear increase of injury risk between the lumbar and spinal injury, although the distribution begins with a higher chance of lumbar injury that is later on decreased in relation to the DRI as the injury risk grows. For the other two duration pulses the spinal injury probability is almost always total, meaning that for values superior to 1ms lower peak velocity values should be applied to understand if the relation between both injury risks is always linear.

On Figure 5.5 which regards acceleration instead of force as in Figure 5.4 there seems to be no logical correlation between spinal and lumbar injury as it shows very high lumbar injury values for 1ms and very low values for 10ms. Although for 10ms there could be a possible linear correlation between both injury risks.

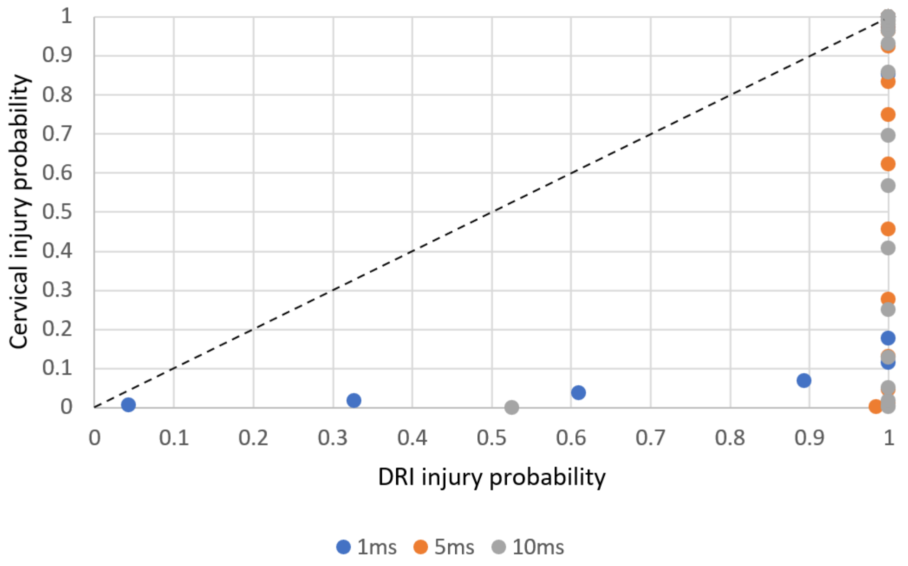


Figure 5.3: Correlation between DRI and cervical injury

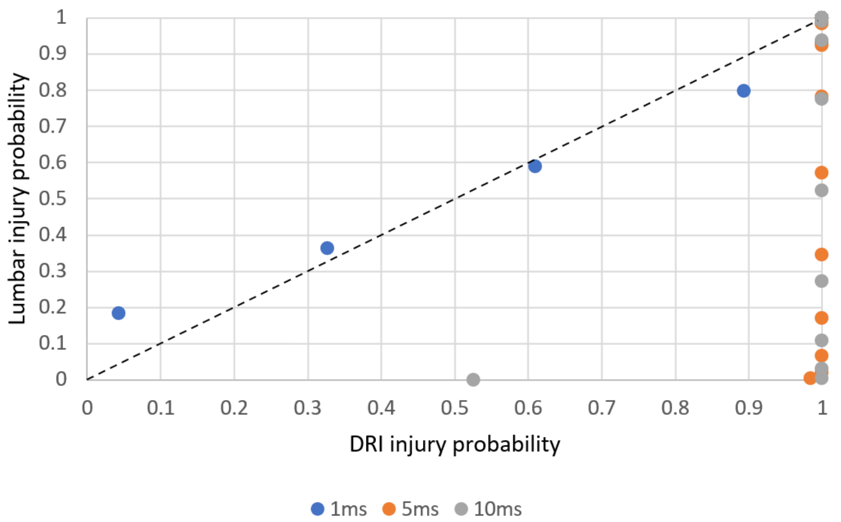


Figure 5.4: Correlation between DRI and lumbar injury due to force

5.4 Pelvis area

The pelvis area compares the injury risk of the pelvis with the spinal injury values obtained by the DRI. This analysis is based on three injury risk curves acquired from two different studies, where the first one [5] experimented with 10 PMHS of which 3 resulted with pelvic fractures, and the second one [6] with 24 PMHS of which 15 resulted with pelvic fractures.

In Figure 5.6 there is a considerable difference between Figure 5.6a and Figure 5.6b regarding the values of 1ms duration pulse. While in Figure 5.6a the values are above the dashed line, meaning that there is a higher risk of pelvic injury than of spinal injury which is wrong, in Figure 5.6b they are below the dashed line and slightly increase linearly, reaching a pelvic injury risk of approximately 10% when the spinal injury is of 100%. It is also noted that for Figure 5.6a there is a considerable gap at total spinal injury risk

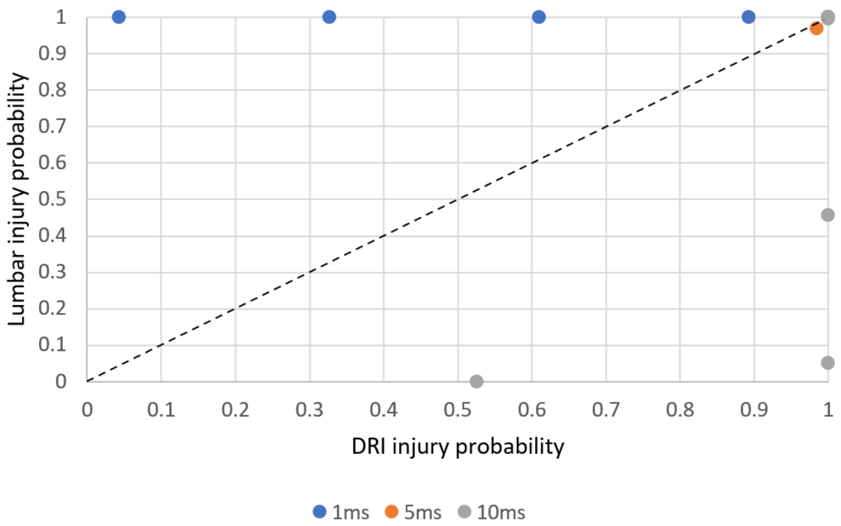


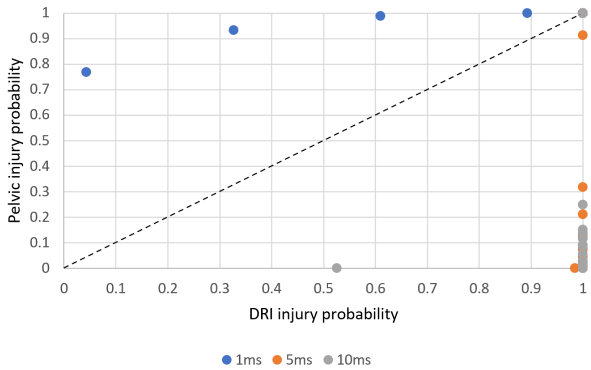
Figure 5.5: Correlation between DRI and lumbar injury due to acceleration

between 40 and a 90% pelvic injury risk, unlike in Figure 5.6b where it is more distributed.

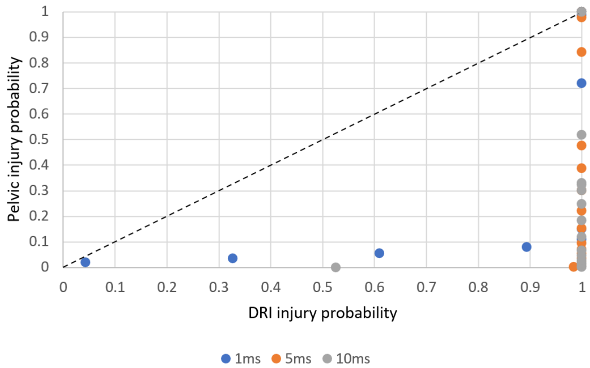
In Figure 5.7 the results are similar to the previous graphs where the injury risk is dependant on acceleration, since the pelvic injury risk is located at around 100% independently of the DRI values. however, there is a clear distribution across the 5 and 10ms values that are located around the total spinal injury risk.

5.5 Spinal area

The results of Figure 5.8 are not as linear as expected and are more related to Figure 5.6a since there is a gap at total DRI injury between 50 and 90% injury risk for spinal and the values of 1ms are located ta higher values than 90% spinal injury risk.



(a) Correlation between DRI and pelvis injury due to acceleration



(b) Correlation between DRI and pelvis injury due to velocity

Figure 5.6: Injury risk curves for pelvis area regarding axial force in sacrum

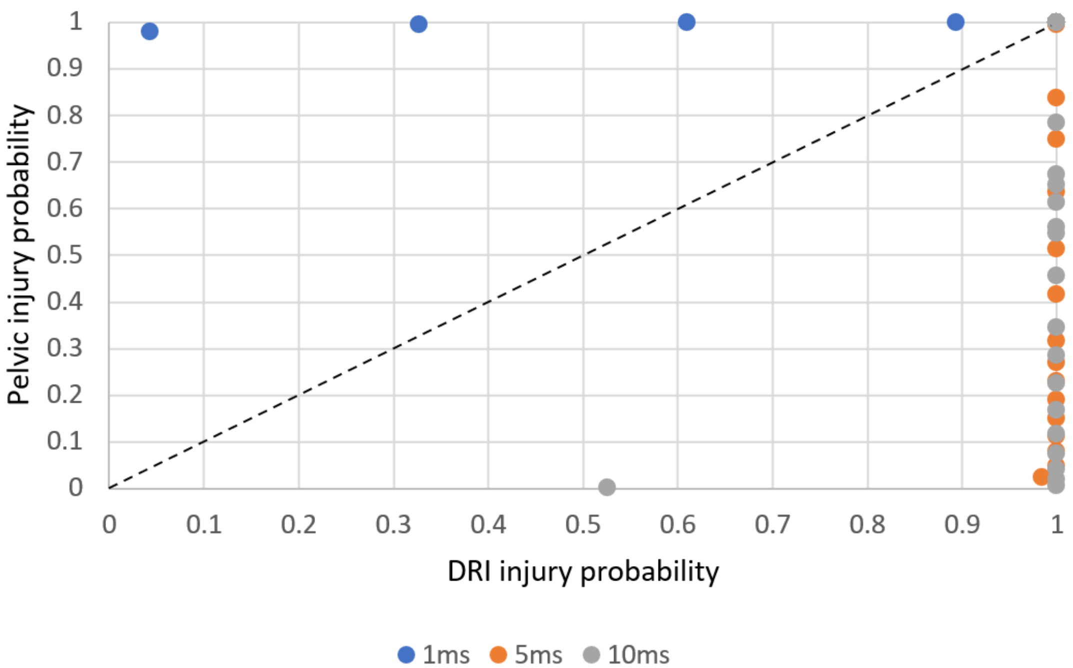


Figure 5.7: Correlation between DRI and pelvis injury due to acceleration

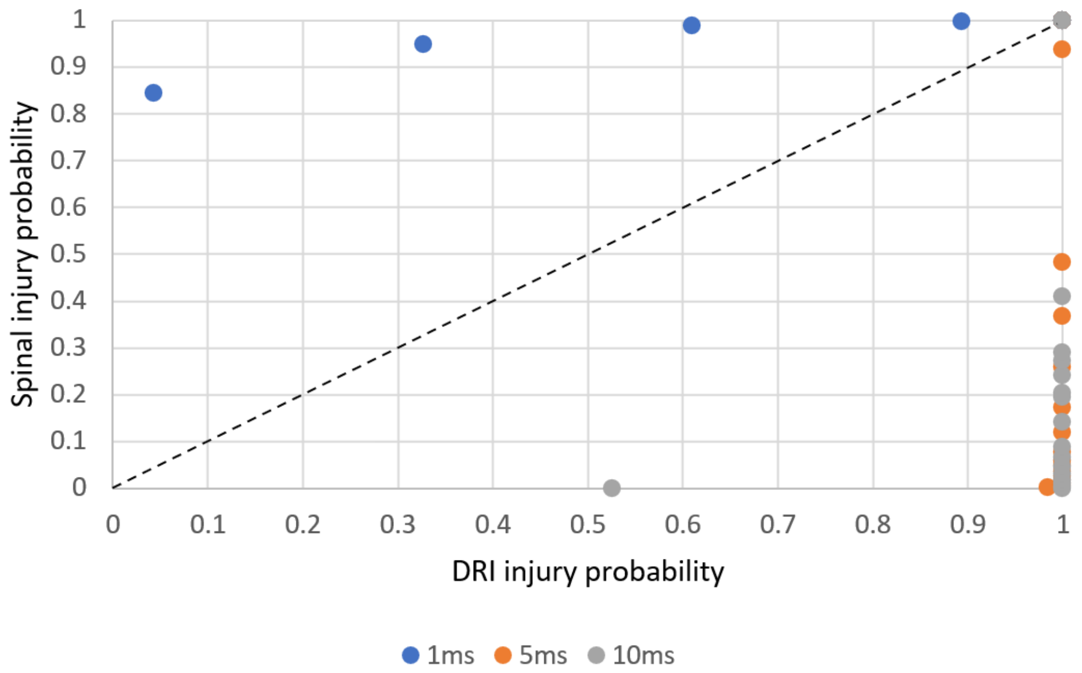


Figure 5.8: Correlation between DRI and spinal column injury

Chapter 6

Future projects

Several studies can be performed in the near future based on the surrogate program developed as well as on the data recollected.

6.1 Extensive sensitivity analysis

The current sensitivity analysis is composed of 62 simulations with values ranging between 1 to 10ms for time and 0.1 to 10m/s for peak velocity. In some cases it can be observed that further velocity values such as 20m/s are needed in order to obtain the desired injury risk curve.

Therefore, a future research could be performed applying further values of both time and peak velocity to get more precise injury curves.

6.2 Further parameters

DRI has been compared to several parts of the spinal column, but there are still several more parameters from the spinal column that can be compared to the DRI such as, for example, the Neck Injury Criterion (Nij).

6.2.1 Neck Injury Criterion (Nij)

Neck injuries are originated around the cervical column, most of them usually consist of fractures at C2 (32%) and C7 ((20.9%), from which 15% result fatal and are most commonly caused by motor vehicle crashes [20].

The calculation of the Nij is based on a combination of axial force (Fz) and sagittal moment (My) normalized by an axial force critical intercept (Fzc) and sagittal moment intercept (Myc) as seen in Equation 6.1.

$$Nij = \frac{F_z}{F_{zC}} + \frac{M_y}{M_{yC}} \tag{6.1}$$

6.3 Further application

This type of injury prediction through the use of MSC Adams combined with Matlab allows for further use of the surrogate programming, such an extension of it towards other body parts due to UBB or the application of that criteria due to a different type of load.

6.3.1 Lower extremities for UBB

The current program applies for parts of the spinal column due to the application of the DRI. However, it could be developed to compare previous papers based on the effects of UBB on lower extremities such as injury in the lower leg due to an axial loading [21].

6.3.2 Changing load

A different approach to this program could be the application of a different type of loading on the rigid model such as, for example, the use of a frontal load due to a car crash that would allow to determine the DRI of the spinal column based on previous works that perform studies on this type of incidents.

Chapter 7

Conclusions

The following conclusions are regarding the efficiency of the developed program as well as the results obtained due to that same program an its comparisons.

7.1 Project compilation

Based on the results obtained, the surrogate program allows to produce injury risk curves that follow either a log-logistic or a Weibull survival function. Also can be noted that this values can be compared to those of the DRI values in certain cases, but it still requires for further compilations in which the peak velocity values must be lower to obtain a more precise comparison between DRI and the injury probability of each spinal area. Acceptable values could be of a maximum peak velocity of 2m/s and a wider range of pulse duration values ranged between 1 and 10ms.

7.2 DRI and BRIC correlation

Although further test should have to be performed, there seems to be a possible correlation between BRIC and DRI injury risk values, but it requires of more data, specially for values at a range between 0.2 and 0.5m/s when applied a 10ms duration pulse.

7.3 DRI comparison

Based on the obtained results, only a linear correlation can be stated between DRI spinal injury and lumbar injury due to force as stated in Figure 5.4. Although, other tests do also share a linear similarity with the spinal column, but not so much since they only reach the 10% injury risk when the DRI has already reached the 100% such as the case of Figure 5.5 or Figure 5.6b.

To obtain a clearer correlation between DRI injury risk and cervical, lumbar or pelvic risk more simulations need to be made since the only possible conclusion yet is that injury risks that are obtained from projects in which the value is obtained due to acceleration do not correlate well with DRI as observed in Figure 5.5 or Figure 5.6a were the injury risk values of the Y axis when applied a total duration of 1ms resulted always superior to the injury risk values of DRI.

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