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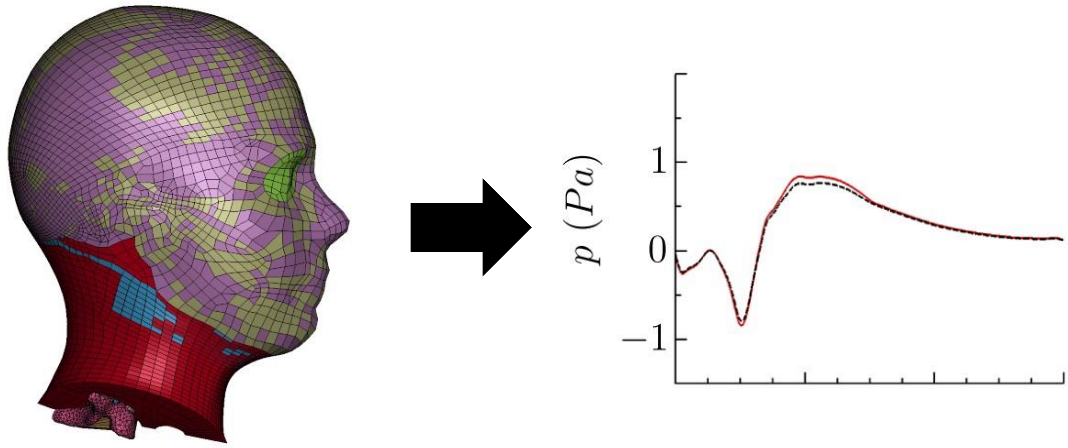

Development of general hydrodynamic modelling method for whiplash nerve injury

Using high-fidelity data from the ViVA+ human body model

Master’s thesis in Automotive Engineering

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CHALMERS UNIVERSITY OF TECHNOLOGY
Gothenburg, Sweden 2022
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Abstract

Vehicle collisions are an issue in the automotive industry, and one of the most common injury in vehicle collisions are whiplash injuries. The reason for the rise of such injury is caused by the occupant's torso being accelerated along the collision direction while the unsupported head lags. Some usual sections in the human body which can be affected by a whiplash motion are spinal ligaments, dorsal root ganglion, and invertebral discs in the neck. In addition, there are studies that has recorded pressure transients in the spinal canal when necks are exposed of whiplash motions. These pressure transients explain some symptoms that are associated with whiplash injuries.

This thesis aimed to develop an existing Matlab-Simulink program that computes pressure transients in the human spinal canal for all directions of neck motions. The current program was only customized for rear-end collisions, taking sagittal neck motion into account. Furthermore, the input in the program was customized to calculate volume changes in the spinal canal modelled with vertebral angular displacement, whereas the modifications done for this study was based on volume changes from human body model simulations. To obtain the volume changes for each vertebra, a human body model was used called ViVA+. With this approach, the purpose was to get similar results of volume change as the old program did with angular displacement for the sagittal direction.

To perform modifications in the program and obtain the desired results, it was divided into different steps. Firstly, modeling in ViVA+ was completed, which was also based on computational settings with crash pulses. The modeling was created in different segments for the vertebras and in sections between each vertebra. Once the modeling was concluded, the desired volume could be achieved, which was put into the MATLAB program. Further, the MATLAB program had to be modified in such a way that it was possible to compute with different types of collisions and directions.

For rear-end collisions, something that was noticed was that the airbags had an effect on the motion of the neck. With simulations completed, with and without airbags, it could be concluded that the one with airbags did not reach full extension. However, the amplitude of the pressure obtained was similar to the program's old version. With the airbags capturing the volume of the segments in a three-dimensional way, it was possible to do the same for other directions, such as lateral motion. One issue regarding the rear-end collisions for not reaching full extension was that the implementation of airbags or its properties had an impact on the stiffness of the HBM. As the aim was to develop the program so that it was possible to calculate the pressure build-up for different directions, it could be concluded that the new modifications were successful, as it was possible to implement on side-collisions.

Keywords: vehicle collision, whiplash motion, pressure transients, spinal canal, angular displacement, ViVA+, airbags, human body model

Contents

Preface

This master's thesis has been performed during the spring of 2022 at the Department of Mechanics and Maritime Sciences within the Division of Vehicle Safety in Chalmers University of Thechnology, Sweden. The thesis was the last degree project in the masters programme Automotive Engineering. In this study, a hydrodynamic model that calculates pressure transients in the human spinal canal has been developed. The study is a part of several research studies and projects concerning assessments in vehicle safety with usage of Human Body Models.

This thesis would not have been feasible without the support and assistance of numerous individuals who have provided essential guidance and preparations in various ways in order to finalize the thesis.

First and foremost, we would like to express our special thanks to our examiner, Professor Mats Svensson for granting us the opportunity to embrace such an interesting thesis. Thank you for providing your advision, feedback and guidance throughout the project. Your comprehensive knowledge regarding the treated topics has been of very much value for the thesis.

We are highly grateful to our supervisor, Assoc. Professor Huadong Yao for being supportive during the development of the hydrodynamic model. Thank you for giving us guidance and generating valuable ideas for problem solving when we worked with your excellent model.

We would also give a special thanks to Asst. Professor David Abellan Lopez from Miguel Hernández University of Elche, who has provided an incredible effort in teaching and advising us on how to model in LS-dyna. Thank you for the support and guidance during your time in Chalmers.

Thanks to Dr. Johan Iraeus and Dr. I Putu Alit Putra for your support and help regarding several issues during the project. Thank you for providing necessary and useful tools and data in order to proceed with the project.

Gothenburg, June 2022-06-01

HUSSEIN AL-DEBIS & ARAN MAHMOUD

Notations and abbrevations

Notations and abbrevations used throughout the report are presented below:

3D	Three-dimensions
AIS	Abbreviated Injury Scale
CNS	Central Nervous System
CSF	Cerebospinal Fluid
C1 - Atlas	First cervical verterba
C2 - Axis	Second cervical verterba
C3	Third cervical verterbra
C4	Fourth cervical vertebra
C5	Fifth cervical vertebra
C6	Sixth cervical verterba
C7	Seventh cervical verterbra
DRG	Dorsal Root Ganglion
HBM	Human Body Model
IV	Intervertebral Vein
IVVP	Internal Vertebral Venous Plexus
L1-L5	First to fifth lumbar vertebra
T1-T12	First to twelfth thoracic vertebra

V_{seg}	Volume of each segment
V_{spinal}	Total volume of all segments
$\dot{V}_{interverebral}$	Volume change of segments in the intervertebral foramen
$\dot{V}_{segment}$	Volume change of each segment

1 Introduction

The following paper aims to present a degree project within the field of road vehicle safety. This project is the basis for developing of a hydrodynamic programming model that calculates the pressure build-up inside the human spinal canal. This section will intriduce, the theoretical and technical background that underlies the project. Accordingly, the aim of the project, several limitations, and the specifactions of issue will be reviewed.

1.1 Background

In vehicle collisions, the most common injury for an occupant is the exposure to a whiplash motion on the neck (Siegmund et al., 2009). The mechanism of the whiplash motion is caused by the occupant torso being accelerated along the collision direction while the unsupported head lags behind. Consequently, this rapid motion in the neck results in a neck injury and includes multiple potential anatomical injury locations in the neck. These locations are the spinal ligaments, facet joints, intervertebral discs, dorsal root ganglion (DRG), vertebral arteries, and neck muscles. In addition, according to experiments conducted on a porcine model, pressure transients in the spinal canal were recorded during exposure of a whiplash motion (Svensson et al., 1998; Örtengren et al., 1996). Accordingly, these records supported the Aldman-hypothesis about pressure transient causing DRG dysfunction, which explains several symptoms related to a whiplash injury (Aldman, 1986).

Based on these and other findings, Yao et al. (2016) developed a hydrodynamic model that simulates the pressure inside the human spinal canal in a Matlab-Simulink program. The program simulates the volume compensation inside the spinal canal by simplifying the internal vertebral venous plexus (IVVP) and the canal into cylindrical tubes. Furthermore, geometric relations between the vertebras in the cervical spine were defined by using angular displacement of the vertebras as inputs. These inputs were extracted from different crash test results and human body model (HBM) simulations. However, this program is intended for tests and simulations of rear-end vehicle collisions, therefore only considering the sagittal motion of the neck. Accordingly, the results from the program simulations presented similar pressure behavior as the experiments on the porcine model conducted by Svensson et al. (1998).

1.2 Aim

The project aims to develop the Matlab-Simulink program developed by (Yao et al., 2016), to calculate the pressure for all directions of neck motions. The program will be modified by changing the input from vertebral angular displacement to direct volume input. The volume changes in the spinal canal of each vertebra will be obtained from a specific HBM to be implemented in the given program. With this method, it will be possible to extract the exact volume used to compute the pressure. The purpose is to attain the same results as previous computations when the angular displacements were used as input in the sagittal direction. If possible, due to the time frame, the program will also be developed so it can be adjusted for other types of collisions such as lateral and oblique for future use, but also to investigate how for example the position of the seat and head restraint influences the pressure build-up.

1.3 Limitations

During the project, there will be some limitations. Only collisions of vehicles will be taken into regard. This will also mean that the program will be modified in purpose to gain knowledge regarding how the pressure differs in sagittal and lateral motion, and with the input modification, the pressure build-up should give similar results. When a collision occurs, it is often dependent on many reasons such as road conditions and driver behavior. However for this paper these reasons will not be investigated. Therefore, the anatomy of the human body will only be related to developing the Matlab-Simulink program for vehicle collisions, specifically rear-end collisions but also side collisions.

To perform the program the input will be given from a specific HBM called ViVA+, a model developed in Chalmers University of Technology. The previous calculations used inputs from the same program, and therefore, the comparison will be consistent and clear in which areas they differ. Another limitation that needs to be considered is that only injuries related to whiplash motion will be analyzed. All work will be made in cooperation with Chalmers, and crash test data and simulations will be acquired from Chalmers as inputs for the program.

1.4 Specifications of the issue under investigation

Based on the described background and the objectives of the project, the following research questions were specified and formulated:

RQ1: How shall the MATLAB program of (Yao et al., 2016) be modified to use direct volume to replace the current vertebral angular displacement input?

RQ2: Can the new program be used also for vehicle side collisions? How do the pressure transients compare to those of rear-end collisions?

2 Theory

In order to understand the essential biomechanics in the human cervical spine, a literature study was conducted regarding its anatomical structures and motions. Furthermore, neck injuries and whiplash motion were addressed with connections to vehicle collisions. Earlier studies related to the topic of whiplash motions and injuries were reviewed to gain knowledge of previous findings.

2.1 Structure of the cervical spine

The spinal column in the human body consists of 33 individual vertebras and is divided into four spine segments, the cervical, thoracic, lumbar, and sacrococcygeal (Kapandji, 2008). The latter segment is formed by two composite bones, the Sacrum and the Coccyx, with five and four fused vertebras, respectively, see Figure 1. The location of the cervical spine in the spinal column lies between the head base and the thoracic level and is formed by seven vertebras named C1-C7. Additionally, the thoracic and lumbar segments are formed with twelve and five vertabras named T1-T12 and L1-L5 respectively. The main objective of the spinal column is to support the trunk and allow the body's bending, rotation, and upright standing (Kapandji, 2008; Sabia & Mathur, 2018). Moreover, another objective is to protect the neuraxis that provides an efficient and pliable casing to the spinal cord along the vertebral canal.

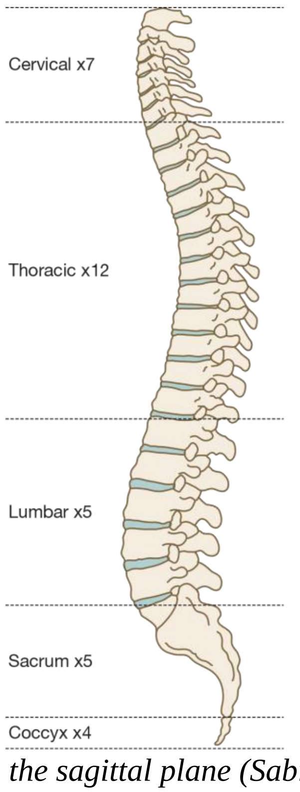

Figure 1 Spinal column in the sagittal plane (Sabia & Mathur, 2018).

The cervical spine is located along the human neck and functions as a supporter to the head, allowing head movement in different directions (Kapandji, 2008). It is divided into two distinct segments, the upper and lower, having different functionalities and anatomical structures. The upper segment contains the first and second vertebra, atlas, (C1) and axis (C2), which are connected to the head base and offer three axes and three degrees of freedom. Additionally, the shapes of C1 and C2 are different from each other and to the remaining vertebras in the cervical spine. The remaining five vertebras form the lower segment, C3 to C7, were all of them has similar shape. This cervical spine segment has two movements, flexion-extension and combined lateral flexion-rotation. Between two adjacent vertebras, intervertebral discs allow movements relatively to each other (Lundon & Bolton, 2001). These discs have also the function of transmitting loads along the vertebral column.

2.1.1 Spinal cord

Through the vertebral canal runs the spinal cord, a part of the body's central nervous system (CNS) (Nógrádi & Vrbová, 2006). It extends from the foramen magnum in the head base down to the upper section of the lumbar spine. The spinal cord has a tubular shape and is covered by three membranes, the dura mater, arachnoid, and pia mater (Sabia & Mathur, 2018). Dura mater is the outer layer of strong connective gray tissue and that covers the nerve roots and spinal cord. The archanoid membrane contains veins and arteries and is connected directly to the dura mater. Pia mater is the inner membrane that allows blood to travel to the nerve roots and spinal cord. The pia mater and the arachnoid membrane are separated with a clear body fluid, also known as cerebrospinal fluid (CSF). From the dorsal horn in the center, dorsal roots enter the DRG into the intervertebral foramen. Figure 2 illustrates a transverse cross-section of the mid-cervical spine.

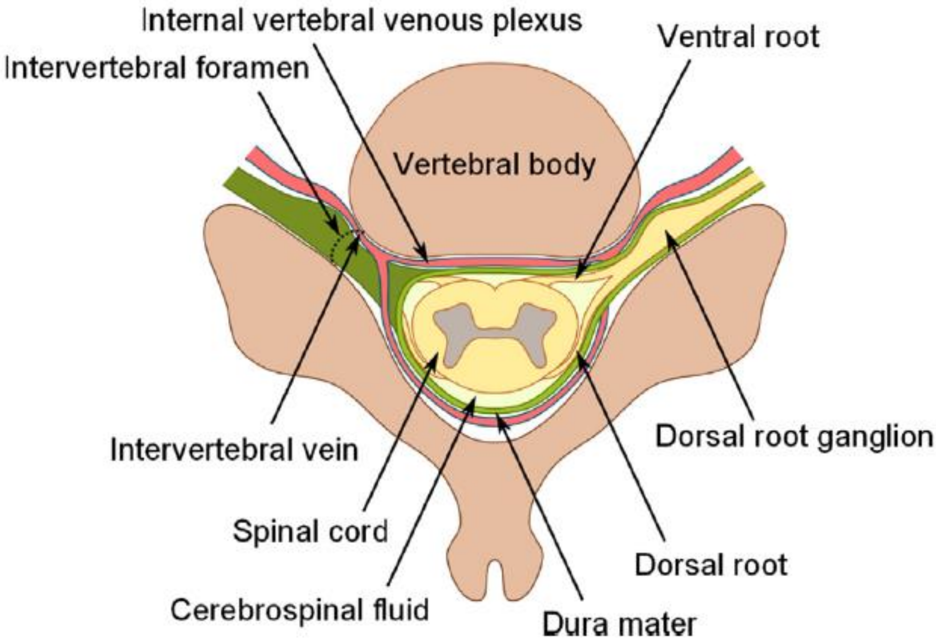

Figure 2 Transverse cross-section of a verterbra in the mid cervical spine (Yao et al., 2016).

2.1.2 Biomechanics and motions of the cervical spine

The joint between C1 and the head is called the atlanto-occipital joint and covers only a flexion-extension movement (Bogduk & Mercer, 2000). According to Kapandji (2008), this joint's pure axial rotation is impossible since it actively stretches out certain ligaments. A rotation of this joint would include translations and lateral flexion of the occiput, therefore seen as an impure axial rotation. Conversely, flexion-extension is established through the rolling and sliding of the occipital condyles along the anterior walls. Bogduk and Mercer (2000) presented studies on cadavers, and showing mean ranges of flexion-extension movements of the atlanto-occipital joint reaching 18.6°. For lateral flexion, the mean range was 3.9°. Furthermore, Bogduk and Mercer (2000) describe the second joint between C2 and C1, the atlanto-axial joint, to permit a large range of axial rotation in the cervical spine. The motion is described as accommodated at the median of the joint, where the anterior arch of C1 pivots around the odontoid process. Cadavers experiments show that this joint reaches about 47° of axial rotation, 10° of flexion and extension, and approximately 5° of lateral flexion.

In the lower segment of the cervical spine, the vertebras are connected with their oblique facet joints (Kapandji, 2008). The facet joints are slightly slanted, both posteriorly and inferiorly. During extension, the verterbral body tilts posteriorly, making the intervertebral space narrower. According to Jonsson (2008), the length of the cervical spine increases during flexion and decreases during extension. Therefore, the spinal canal alters volume during a flexion and extension motion inside the cervical vertebral column. Kapanji has presented ranges of motions of flexion and extension with a reference plane at skull level, showing a complete range in flexion-extension of 110° for the lower cervical spine.

2.2 Neck injuries

A common injury during rear-end collisions is neck injury (Norris & Watt, 1983). Once a crash of this type happens, there will be flexion and extension motion in the neck, specifically of the cervical spine. These movements lead to whiplash injuries and are dependent on the sudden acceleration, which impacts the severity of the injury. As mentioned, flexion and extension motion will cause damage to the neck, with the main action being hyperextension, causing the severity of the injury. These injuries are a kind of whiplash injury, which occurs during whiplash motion, and can be ranked in Abbreviated Injury Scale (AIS), to know the severity of the injury. The most common severity in these types of collision and injuries is identified as minor injury (Örtengren et al., 1996). Örtengren et al. mention that neck injuries are very common during rear-end collisions, giving a quantity of 80% when there are injuries in these situtations. From these injuries, symptoms will occur for the driver or passenger, and they are the same for different types of collisions, such as frontal and side collisions but also rear-end collision (Svensson et al., 2000). However, according to Svensson et al., the motion is different between these types of collisions. For the rear-end collision the neck will have a retraction movement to then be developed to extension motion. While for a frontal collision, the neck will start to have a protraction motion, eventually leading to a flexion motion. These movements will be described further in the sub-sections below, along with the symptoms. The neck injuries during rear-end collisions are directly related to whiplash motion which will cause some symptoms for the specific passenger or driver.

2.2.1 Whiplash motion

During a car collision, such as rear-end or frontal collision, the movement that leads to injuries in the neck, as mentioned in Section 2.2, is whiplash motion. The explanation behind the action is due to a lateral force being applied to the driver on the torso, while the head will remain in the same position (Liu & Yang, 2008). The force comes from acceleration due to the collision, which leads to the whiplash motion and can be described in different phases. Once the force is applied, the first phase that will be developed will be the retraction motion of the neck as described in Section 2.2, with the second phase being extension motion (Svensson et al., 2000).

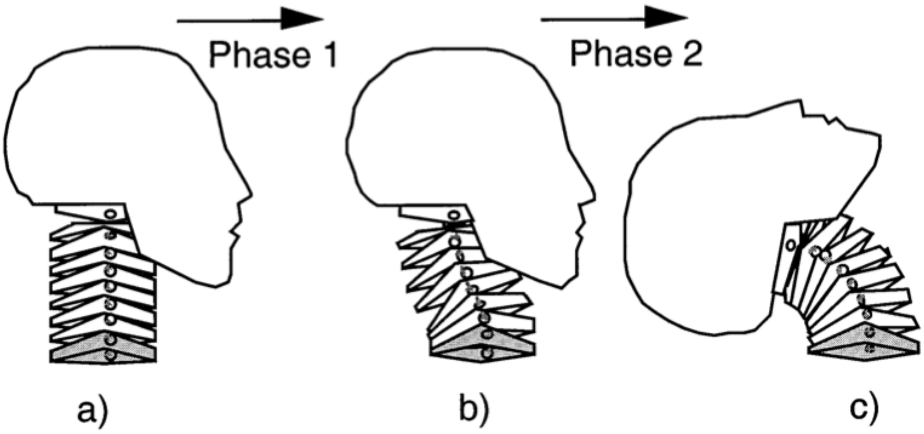

Figure 3 Movement of the neck during a rear-end collision (Svensson et al., 2000).

As seen in Figure 3, the cervical spine will begin to look like an S-shape between the two phases. In step “c” the head has been moving backward and is not in the S-shape form any longer. This occurs once the flexion and extension have attained the limit for the cervical spine segments, both the lower and the upper (Liu & Yang, 2008). The lower is related to the flexion during the motion, while the upper parts will reach maximum extension, eventually leading to the head having a movement, as seen in “c” in Figure 3. During the collision, a rapid change of extension to flexion and vice versa will happen for the upper and lower segments, respectively, due to the acceleration.

2.2.2 Symptoms

When a collision has occurred, some symptoms will be shown in the driver regarding the injuries. A common symptom that can be identified is neck pain, but besides this others, symptoms that the driver can be exposed to are neurological symptoms in the upper torso (Liu & Yang, 2008; Örtengren et al., 1996). Örtengren et al. continue to mention that injuries related to whiplash motion can, in some cases, lead to disability (10% of the cases). Neck trauma is also a symptom the driver can suffer from and is related to the sensory nervous system. Pain can be one of the aspects taken into account (another can be weakness) and are connected with the central nervous system through the roots of cervical nerves. It is also common to suffer from the symptoms considered visual such as dizziness, headache, and unconsciousness. According to Örtengren et al., during low-speed rear-end collisions, it is uncommon that any significant injury will occur and, therefore, there are rarely any deficits regarding the functional and distinct structural areas.

2.3 Pre-studies

This section introduces previous studies regarding the topic that will be investigated and presented. These studies that are connected to this are a hydrodynamic model along with the HBM ViVA+. The purpose of the hydrodynamic model was to see if it would support the hypothesis of Aldman (1986). The theory was based on pressure gradients in the canal that would be the outcome of sagittal neck motion once there are volume changes in the spinal canal. Aldman mentions that once the whiplash motion is in act, the spinal canal volume would variate, which would lead to the incompressible contents in the canal, CSF and IVVP, in need of redistribution. For the HBM ViVA+, the purpose was to see how the human body, specifically the neck is in motion during a rear-end collision and where the body is most impacted.

2.3.1 The hydrodynamic model

As mentioned in Section 1.1, a hydrodynamic model that simulates the pressure transients inside the vertebral canal was developed by Yao et al. (2016). The model's main goal was to investigate analytically and theoretically the hypothesis that pressure transients in the canal causes DRG dysfunctions. The model is made up of two major subsystems that were implemented in Matlab-Simulink to create a program. Firstly, the IVVP and the canal at each vertebra were simplified into cylindrical tubes after defining a set of geometrical relationships, see Figure 4. This was a part of the first subsystems where the main objective was to calculate the altering volumes between two adjacent vertebras during motion. Typically, the rotation center of a vertebra in the cervical spine is located in the body of the inferior vertebra. However, Yao et al. assumed an instant center of rotation within the vertebral body. This geometric model uses vertebral angular displacement as input to calculate the relative motions between vertebra pairs. The derivations for the volume changes are not included for brevity.

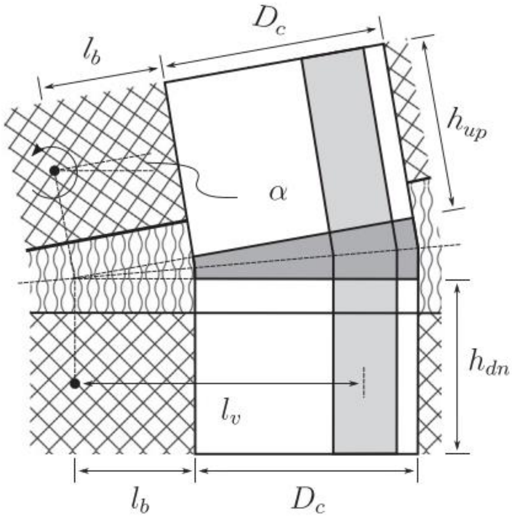

Figure 4 Illustration of simplified tubes of two adjacent verterbras with geometric relations (Yao et al., 2016).

The second subsystem covers modeling of the vein network by decomposing the network into hydrodynamic components. Following the decomposition, the components are constructed concerning their function and location, see Figure 5. The local vein network between two adjacent vertebras is illustrated in the figure and describes the connection between the components. The *Canal* components include the IVVP inside the vertebral canal. Furthermore, the component *Canal joint* consists of the IVVP located in the space between the vertebras. Additionally, the intervertebral vein (IV) is located inside the *Branch* component. This part of the subsystem uses inputs of volume change calculated from the first subsystem for each vertebra joint. Each element aims to calculate the flow in the IV and IVVP tubes by using a toolbox in Matlab-Simulink called SimHydraulics.

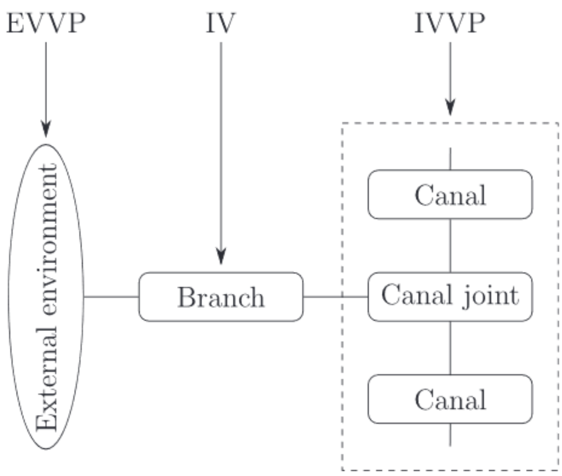

Figure 5 *Schematic relations of the vein network between a pair of vertebras (Yao et al., 2016).*

2.3.2 The HBM ViVA+

Human body models are a digital representation of the human anatomy in software and are primarily used for road safety developments. HBMs address different types of injury risks directly from a software environment. These models have been developed with a high degree of biofidelity and detail by analogy to a human body. Chalmers University of Technology in Gothenburg, Sweden, has developed an open-source HBM that represents the average female called ViVA OpenHBM (Östh et al., 2017). The model's dimensions are based on anthropemtric measurements within 5% deviation from measurements of the 50th percentile female. Accordingly, the HBM is a 31-year-old female with a weight of 60.8 kg and a height of 161.6 cm. The reason behind developing a female model beacuse that females have a higher injury risk than males in comparable vehicle collisions. Therefore, a detailed and simplified version of the model was created where the major difference between them was the detailing of the neck. Hence, the clear version includes more soft tissues and connections to model the neck's kinematics. On the other hand, compliant kinematic joints are instead used between the vertebras in the simplified version. Figure 6 illustrates an overview of the ViVA OpenHBM.

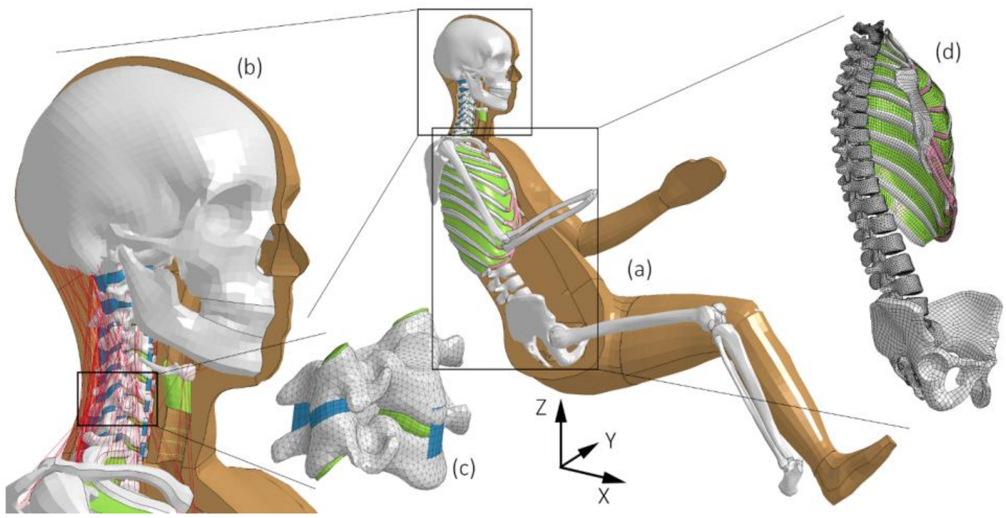

Figure 6 *a) overview of the ViVA OpenHBM. b): cervical spine. c): a verterbral segment. d): the axial skeleton, pelvis, an thoracic cage. (Östh et al., 2017).*

In recent years, enhancement of the ViVA model has been implemented and includes detailing of the legs, pelvis rib cage, and cervical spine. The newer model is hereafter called ViVA+. In addition, the model has been developed to be prepared for other types of postures, such as standing. Other anthropometries have also been used to create a male version of the ViVA+ model. However, for this project, the seated female model will be used. See Figure 7 for an overview of the seated female ViVA+ 50F model.

Figure 7 An isometric view of the ViVA+ 50F model.

3 Methodology

To accomplish the desired results and specifications of issue, several steps had to be gone through. These steps were modeling in the ViVA+, which consisted of how the modeling was done, computational settings and prescribed crash pulses. Beside this, MATLAB programming was performed to get results so that it was possible to analyze what was accomplished to see if the specifications of issue were fulfilled.

3.1 Modifications on ViVA+

The desired aim was to investigate the possibility of acquiring the pressure build-up in the spinal canal for all directions of neck motion. Hence, it will be possible to investigate lateral and oblique vehicle collisions. To achieve the described aim, the issue under investigation needed to be approached in a three-dimensional way. Compared to the previous method of investigation, the hydrodynamic model of Yao et al. (2016), only takes sagittal motion of the neck into account.

The current version of the ViVA+ model does not contain any CNS and spinal cord modeling. Hence, the vertebral column in the HBM is hollowed and is only formed through the stacked vertebras. Figure 8 illustrates an overview of the cervical spine and a sagittal cross-section, where the spinal canal can be viewed. To acquire the volume changes inside the spinal canal, a volume representation that encloses the domain of the vertebral column was needed to be modeled.

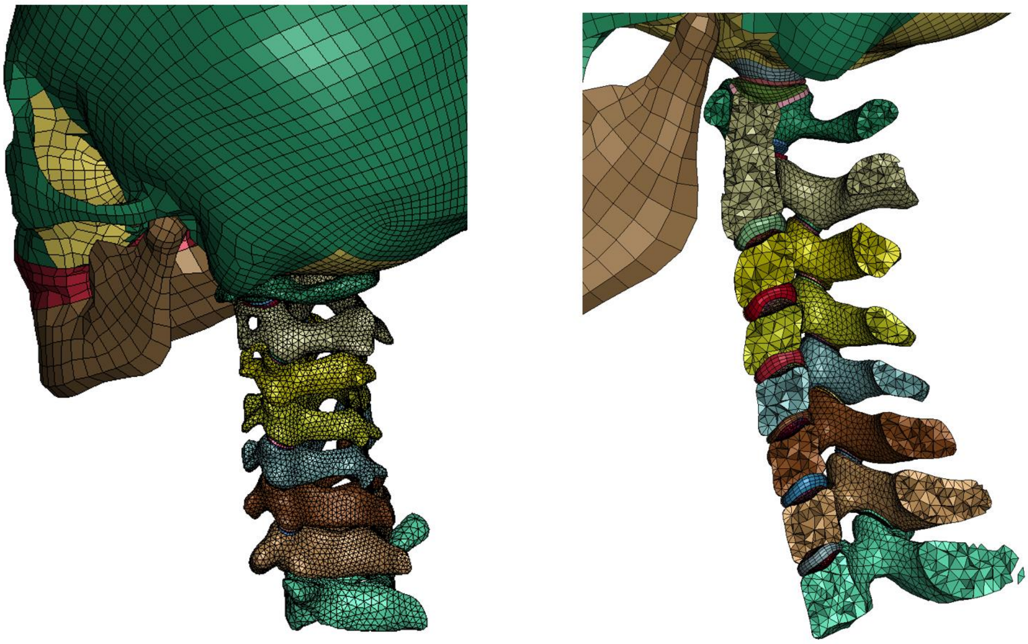

Figure 8 Overview of the ViVA+ cervical spine.

The reason to model in ViVA+ was due to different causes related to the specifications of the issue. However, the method used in this project allowed movements in all directions possible since the program, LS-Dyna, allows modeling in 3D. Besides this, an advantage was that the direct volume could be acquired, which compared to the model of Yao et al. (2016) had an approximative volume using other prerequisites.

3.1.1 Modelling segments

As Jonsson (2008) mentioned, the length of the cervical spine elongates during flexion and shortens during extension. Therefore volume change occurs in the intervertebral space between the vertebras. Thus, capturing this volume alteration from the HBM during a vehicle collision simulation involving a whiplash motion was essential. This was done by modeling new parts in the ViVA+ from existing nodes that enclose the cervical spine's spinal canal. Existing nodes were exploited to attach the new modeled parts with the current model without any free nodes. This was done to allow the added parts to follow the kinematics of the cervical spine during simulations. Furthermore, this was done in segments, meaning that a part was modeled for each vertebra and each pair of vertebras. However, the modeling is limited to the vertebras in the cervical spine and the first thoracic vertebra T1. Accordingly, the segments are presented in Table 1 and numbered in order according with their location and beginning from the first vertebra C1.

Table 1 The segments created in ViVA+

$V_{seg,i}$ - Segment							
No. 1	*No. 2*	*No. 3*	*No. 4*	*No. 5*	*No.6*	*No. 7*	*No. 8*
C1	C1-C2	C2	C2-C3	C3	C3-C4	C4	C4-C5
No. 9	*No. 10*	*No. 11*	*No. 12*	*No. 13*	*No. 14*	*No. 15*	
C5	C5-C6	C6	C6-C7	C7	C7-T1	T1	

These segments were created by generating new shell elements that utilize the existing mesh around the vertebral foramen of each vertebra. This method made it possible to create parts that enclose the vertebral foramen. An arbitrary cylindrical form was therefore created and thereafter closed on the top and bottom. These parts correspond to the odd segment numbers in Table 1, segments No. 1,3,5,…,15. After that, new parts were created between the superior and inferior vertebras to enclose the intervertebral space. An example of a segment can be seen in Figure 9, and complete visualization of all segments in Appendix A.1. The shell elements were defined with an element formulation according to Belytschko-Tsay, due to its robustness and fast computations (Haufe et al., 2013). Furthermore, the thickness of the shells was defined to be 2 millimeters.

The ViVA+ model has been validated in terms of material and element qualities, stiffnesses and kinematics in its current versions. Therefore, it is crucial not to add any stiffness to the entire model when creating the airbags. Hence, this was controlled by defining the part segments material as Nulls. Defining parts as Null materials allows the state equations to be considered and eliminates the need to compute deviatoric stresses. The density was set to 1e-7 kg/mm^3, Poisson's ratio to 0.4, and Young's modulus to 0.01 GPa. This methodology allowed the parts to be connected to the nodal points without affecting the overall stiffness of the model.

The final step was to define the created segments as airbags since LS-Dyna directly calculates the volumes inside airbags during simulations. Here, the option *AIRBAG_LINEAR_FLUID was used with the same density as for the Null materials and low values of bulk modulus with approximately 0.0002 GPa.

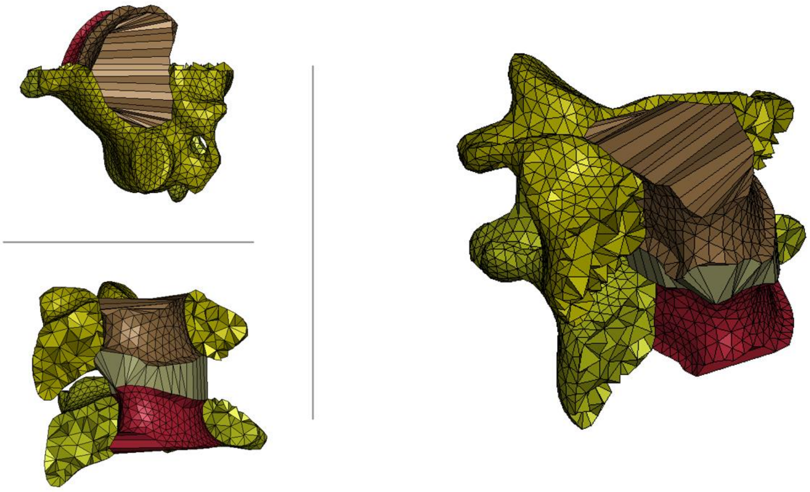

Figure 9 A top, side and isometric view of segment No. 5,6 and 7 together with the third and fourth cervical vertebra.

3.1.2 Crash simulations

The second step was to initialize crash simulations with the added segments on the HBM in a simulation program called LS-run. To minimize the calculation time of the simulations, only the head and neck parts of the ViVA+ were used, see Figure 10. This was done since the interest was only in the neck's motion and the airbag's behavior. Therefore, a setup including a head-neck model of the ViVA+ model with a prescribed acceleration pulse was used to obtain new data from the created segments. The aim was to perform two simulations with prescribed acceleration pulses applied to the T1 vertebra from different directions. The two simulations were to represent impacts of type rear and side. These types of effects will generate new data that will be used to get a broader understanding of the performance of the hydrodynamic model at a later stage. The prescribed acceleration pulse was based on a study conducted by (Sato et al., 2014), where the aim was to clarify the charasterisitcs of dynamic cervical vertebral motion for rear-impacts. In the study by Sato et al., two test series for rear-impact were tested, one with an inclined sled and one with a mini-sled. The latter test series was used for the simulations of this study. The peak acceleration of the test was 42 m/s^2 and with a ΔV of 5.8 km/h. The time history of the acceleration pulse of the sled is seen in Appendix A.2.

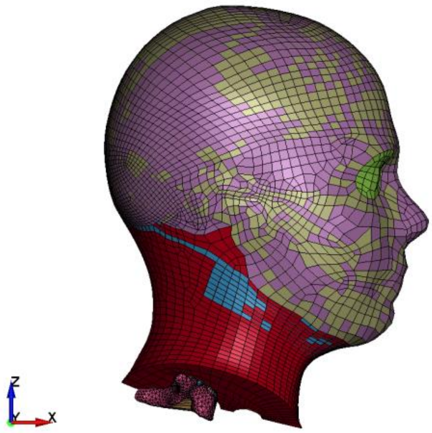

Figure 10 The head-neck model of the ViVA+.

Nonetheless, the results of the kinametics on T1 from that study was prescribed on the setup that includes the head-neck in LS-dyna, see Figure 11. The acceleration pulse was prescribed on the neck in positive x-direction for the rear-impact scenario and in positive y-direction for the side-impact method. Simulating with the described crash pulse allowed the model to generate a whiplash motion on the ViVA+ HBM, capturing the volumes in the created airbag models.

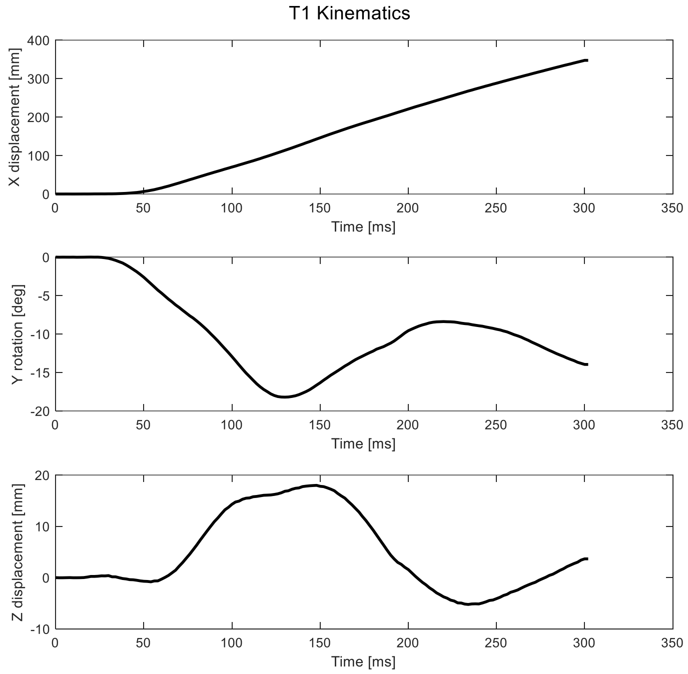

Figure 11 The kinematics of T1 from the tests conducted by Sato et al. (2014)

3.2 MATLAB programming

After performing the desired simulations, it was necessary to process the data to be properly read into the hydrodynamic model. As mentioned in Section 3.1.1, airbag segments were created to calculate the volume alterations in the spinal canal during a crash simulation. The data is dependent on time and captures, therefore, the volume inside each segment at each time step. It is crucial to correctly input these data points into the hydrodynamic model for it to be read rightly. There is no limit for the length of the data. However, it should contain 16 columns totally, where the first column corresponds to the simulation time and the rest to the volumes. The volumes correspond to the segments in Table 1 and should be arranged with the odd segment numbers first and thereafter the even segment numbers. An example of how the data columns should be placed is listed in Table 2, where * corresponds to data points. Accordingly, the data file should be tab-delimited and of the type .dat file.

Table 2. An example of the data file arrangement that concerns the volumes obtained from ViVA+ simulations.

	Segment No.														
Time	*1*	*3*	*5*	*7*	*9*	*11*	*13*	*15*	*2*	*4*	*6*	*8*	*10*	*12*	*14*
*	*	*	*	*	*	*	*	*	*	*	*	*	*	*	*
*	*	*	*	*	*	*	*	*	*	*	*	*	*	*	*
*	*	*	*	*	*	*	*	*	*	*	*	*	*	*	*
…															

The program used to acquire the pressure build-up in the spinal canal was MATLAB, i.e., the hydrodynamic model, with input from the ViVA+ simulations. Since the output from the HBM simulations was volume for the different segments, the simulation's results were used as inputs into the hydrodynamic model. As mentioned in section 2.3.1, the hydrodynamic model was divided into subsystems, where the latter accept volume changes in each vertebra segment. Therefore, the input volumes were differentiated with respect to time to acquire volume changes at each time step. However, the vertebra segments are defined differently in the hydrodynamic model compared to the executed modeling in LS-Dyna. The actual difference lies in the definition of the vertebral heights, where vertebral height is defined as the length of the vertebral canal including the length to the centre of the disc in the neutral-position, see h_{up} and h_{dn} in Figure 4. Therefore, eight input positions for volume change are placed for each verterbra segment in the hydrodynamic system. Conversely, the volume change occurs in the intervertebral space, meaning seven positions. Thus, it was necessary to divide the data to make it compatible with the hydrodynamic model.

This resulted in additional steps that included partitioning of all the modeled segments to apply the same definition of vertebral height. The partitioning process could be simplified into three steps:

1. Calculation of the total volume in the spinal canal in every time step – V_{spinal}.
2. Calculation of the sum of volume changes in all intervertebral foramen in every time step – $\dot{V}_{intvertebral}$.
3. Calculation of the actual volume change for each segment – $\dot{V}_{segment}$.

The first step was performed by summing the volumes of all segments from the input data in every time step. Here n denotes the segment numbers, and the calculation is given by,

$$V_{spinal} = V_{seg,1} + V_{seg,2} + \ ... + V_{seg,n}, \qquad \text{where } n = 1,2, \ldots ,15. \tag{1}$$

In the second step, the volumes in the intervertebral foramen were differentiated and then summed up. For this step, only the data corresponding to the junctions between each pair of vertebras are included, therefore denoted with j for even segment numbers. The volume changes are calculated by,

$$\dot{V}_{intervertebral} = \frac{d}{dt} V_{seg,2} + \frac{d}{dt} V_{seg,4} + \ ... + \frac{d}{dt} V_{seg,j}, \qquad \text{where } j = 2,4, \ldots ,14. \tag{2}$$

Lastly, the volume changes that cover the vertebral heights for each canal, $\dot{V}_{segment}$, are calculated for each segment of the vertebral foramens. Hence, acquiring eight volume changes that are compatible with the hydrodynamic model. The calculation is given by,

$$\dot{V}_{segment} = \frac{\dot{V}_{intervertebral}}{V_{spinal}} * V_{seg,i}, \qquad \text{where } i = 1,3 \ldots 15. \tag{3}$$

As mentioned in Section 2.3.1, within the hydrodynamic model, the spinal canal is divided into cylindrical tubes. Thus, the tube dimensions should represent the vertebral heights and dimensions of the body on which the crash tests are based on.

Yao et al. (2016) implemented the vertebral dimensions of asymptomatic Chinese women reported by Lim and Wong (2004). In addition, the vertebral heights were written by Vasavada et al. (2008). To increase the accuracy of the pressure build-up simulations, the vertebral heights and dimensions of the ViVA+ HBM should be implemented into the hydrodynamic model. However, such data was not available. Therefore the same values used by Yao et al., were implemented, with the different dimensions for the vertebras presented in Table 3.

Table 3. Paramters regarding the different vertebras.

Verterbra	Half of the antroposterior diameter [mm]	Canal diameter [mm]	Canal height [mm]	Segment height [mm]
C1	16.70	18.50	16.78	6.78
C2	16.70	18.50	10.78	14.29
C3	16.40	16.10	13.50	19.71
C4	16.10	15.70	12.60	17.82
C5	15.80	16.00	12.20	17.16
C6	16.40	16.10	12.00	16.84
C7	16.90	16.00	13.00	17.55
T1	16.90	16.00	13.00	15.15

In the MATLAB program, new function files were written and implemented in a new path, meaning that the previous version, which accepts angular displacements as inputs, is preserved. This implementation allows the program user to choose between the old and new versions when simulating the pressure build-up in the spinal canal. The function files are written mainly to read the input data properly and execute the partitioning calculations of the segments. Moreover, this implies that the first subsystem of the hydrodynamic model is different from the older version, while the second subsystem is preserved.

3.2.1 Pressure build-up simulations

Modifying the MATLAB program made it possible to run the pressure build-up simulations with input from the LS-Dyna HBM simulations. Here, the inputs of volumes from the different crash-impact simulations were implemented into the hydrodynamic model for further analysis. The older version of the hydrodynamic model has been validated in terms of performance regarding the calculations of the pressure transients in the spinal canal. Therefore, the older version will be used in the validation process for the new model. In the validation process, the old version of the program is used to validate the performance of the latest version. This was done by comparing the pressure transients from the conducted simulations. Moreover, input data for angular displacement and direct volume was extracted from the same HBM simulation for the rear-impact scenario to make the two versions comparable.

4 Results

In the following chapter, the simulation results from the hydrodynamic model are presented. Broadly, the pressure build-up in the spinal canal will be given for the rear-impact and the side-impact crash simulations. The present computations utilized inputs from ViVA+ simulations which in turn uses prescribed vertebral motions from Sato et al. (2014). The results show the pressure transients in pascal and during the time interval between 0.15-0.45 seconds since the crash pulse was applied 0.15 seconds later during the simulations.

Additionally, different phase demonstrations of the rear and side-impact simulation on the ViVA+ model are visualized further. Also, a rear-impact simulation without the airbags implementation was performed to be visualized. These visualisations were included to highlight how the created airbags affect the neck motion and the possibility of affecting the stiffness of the ViVA+ model.

4.1 Rear-impact

As described, two rear-impact scenarios were simulated, one simulation with the implemented airbags and one simulation without the airbags. Figure 12 (a) and 12 (b) show the animation phases of the two simulations. Although the same crash pulse was applied to the two simulations, a significant difference in neck movement was observed. In the case without airbags in Figure 12 (b), the neck movement is similar to the neck movement described for rear-end collisions by Svensson et al. (2000) in Figure 3.

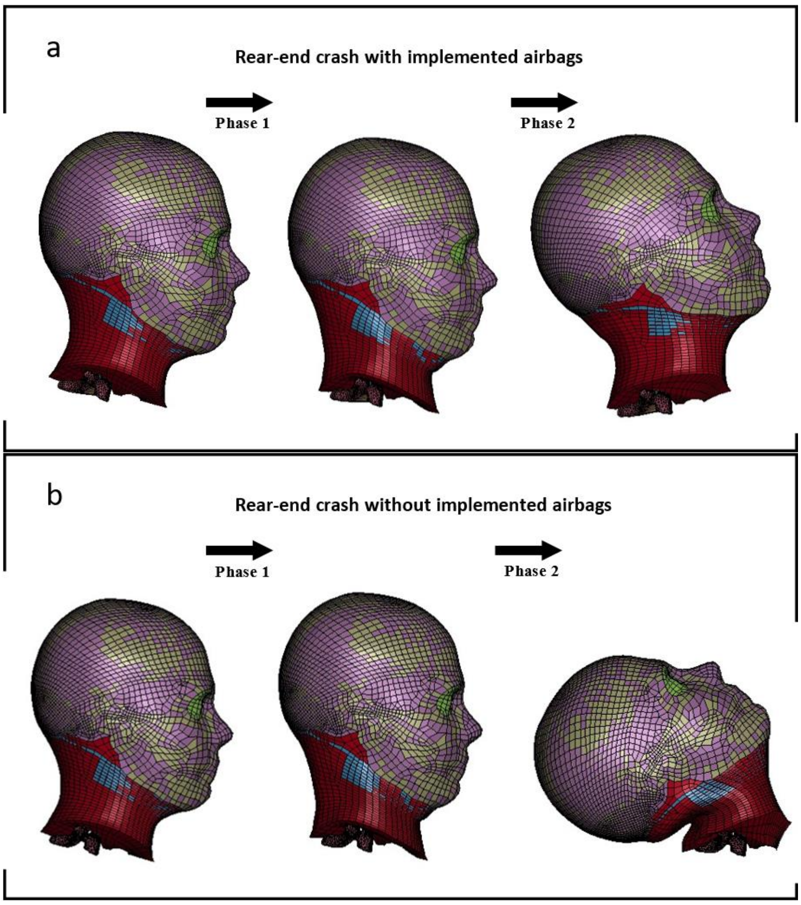

Figure 12 Demonstration of the different phases during rear-end crash simulations with included and excluded airbag implementation.

On the contrary, the case that includes the airbag implementation demonstrates a different neck motion. Primarily, during *Phase 2* in Figure 12 (a), the neck did not complete full extension. Instead, the neck motion reached a stopping point of the extension where after that, the motion changed direction towards full flexion at the end of the simulation, but the full flexion is not shown. However, the neck movement between the two phases in both cases shows the retraction of the neck that forms an S-shape, which is consistent with the literature findings presented in Section 2.2.1.

Figure 13 presents the pressures for each vertebra segment from the rear-impact simulation. The amplitude of the pressures is similar to the results presented by Yao et al. (2016). Theoretically, the retraction of the neck in which the S-shape is formed causes negative pressure, which can be seen around second 0.26. This time step corresponds to the time in which the neck formed the S-shape in the LS-Dyna simulation.

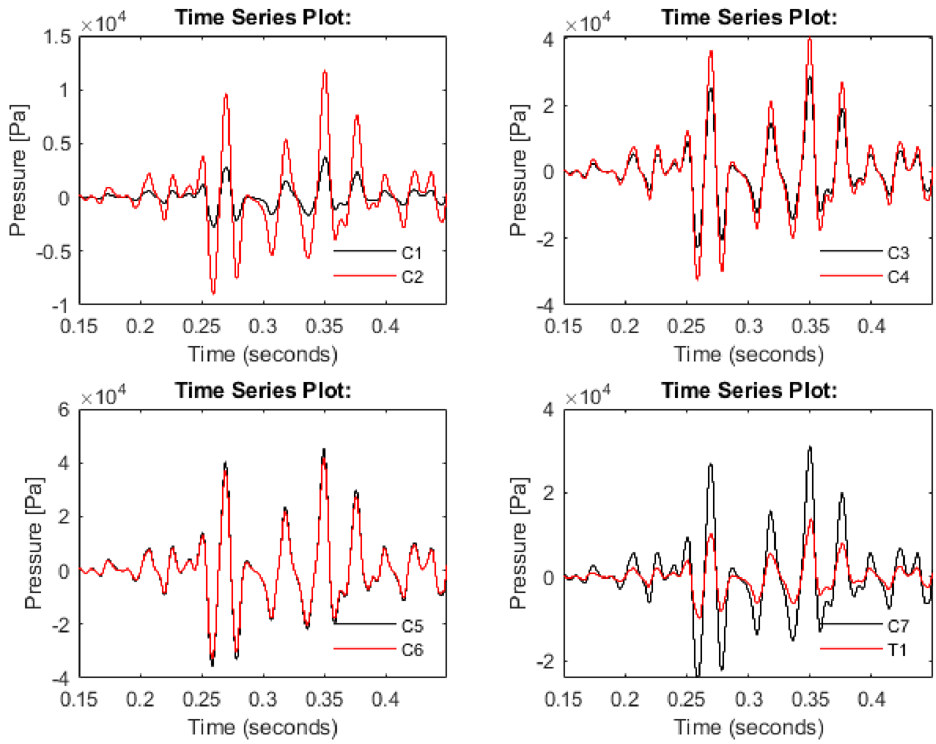

Figure 13 Pressure results of the rear-impact simulation for each verterba segment with the new version. Direct volume input.

Figure 14 presents the pressure transients from the same rear-impact scenario but with angular displacement data and a simulation conducted with the old version of the hydrodynamic model. Further, a clear difference in the behaviors of the pressures can be seen between the simulations of the new version and the old version. These differences are based on different reasons concerning the cylindrical tubes' calculations. However, the magnitudes of the pressures are on the same level between the different versions.

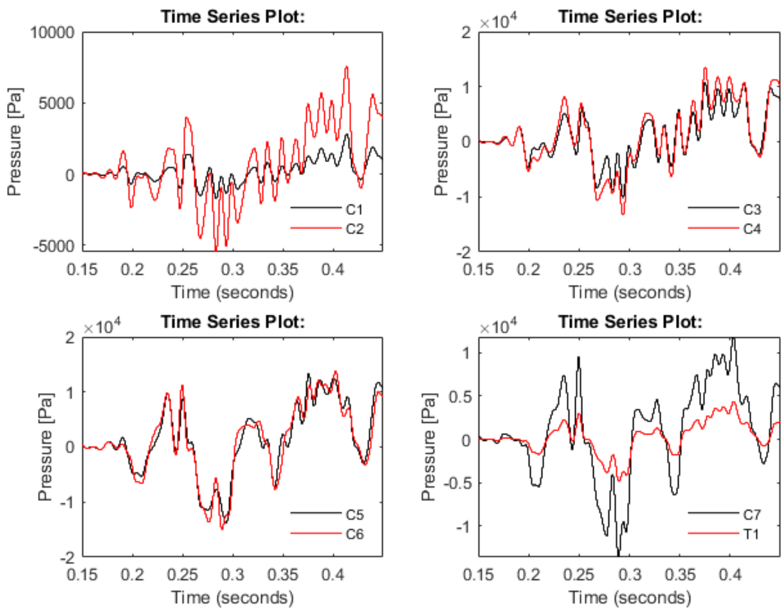

Figure 14 Pressure results of the rear-impact simulation for each verterba segment with the old version. Angular displacement input.

4.2 Side-impact

Since the airbags capture the segments' volumes in a three-dimensional approach, the modeled representation of the spinal canal in the ViVA+ model could capture the volumes for different directions. In Figure 15, the pressures for the side-impact scenario are presented. The previous hydrodynamic model took only sagittal neck motions into account. Nevertheless, the new model can provide an analysis of the pressure transient in the spinal canal for other neck movements.

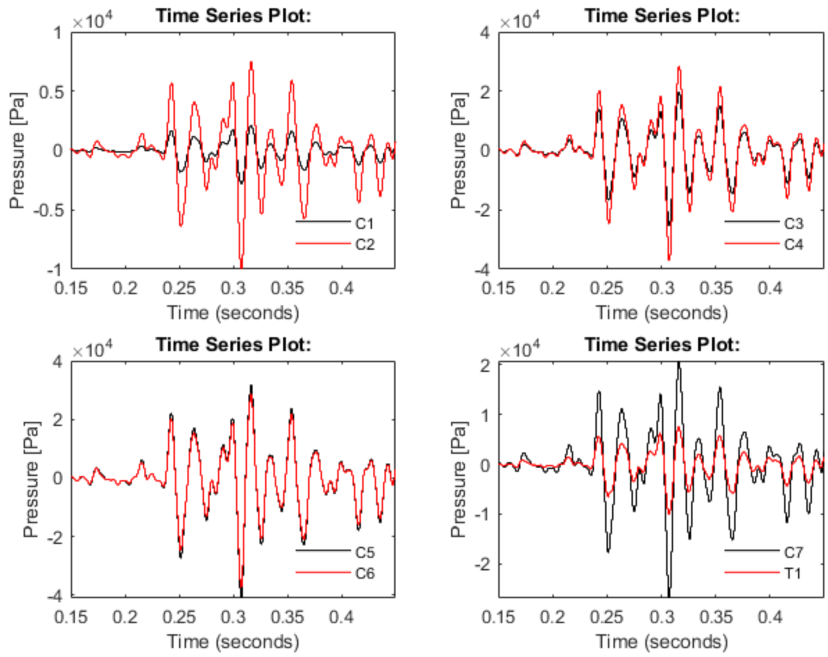

Figure 15 Pressure results of side-impact simulation for each verterbra segment.

In Figure 16, a demonstration of the neck motion for the side-impact simulation is visualized and shows that a lateral movement is performed. It has been clarified that the implemented airbags affect the ViVA+ model. Therefore, there is uncertainty about the lateral neck motion for this scenario. This is the case since a simulation for side-impact without the implemented airbags was not performed.

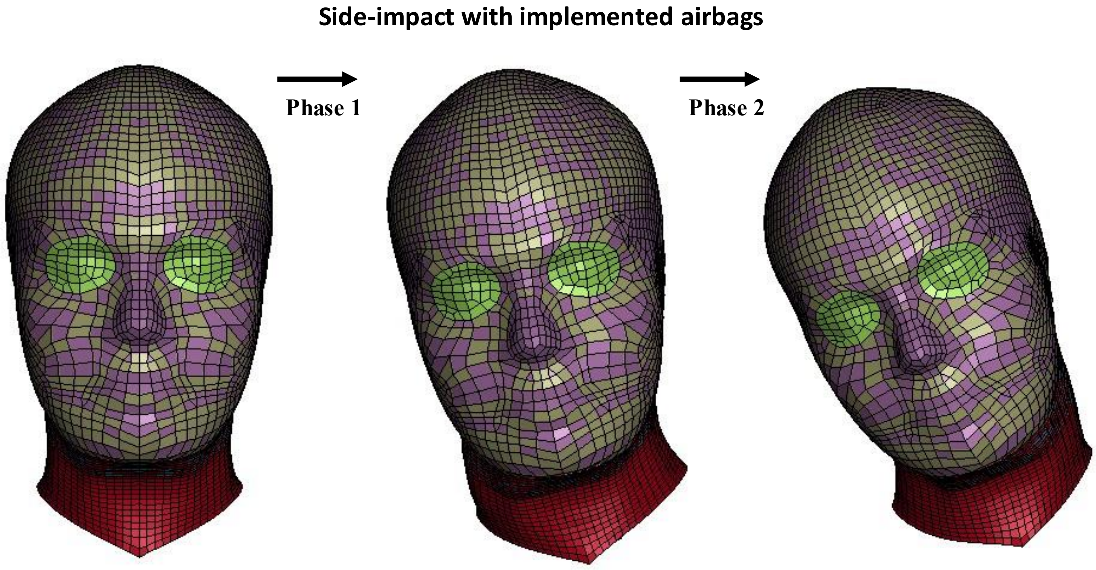

Figure 16 Animations of the different phases during side-impact simulations with airbag implementation.

5 Discussion

Based on the results presented in Chapter 4, this chapter provides a discussion and analysis regarding the outcomes of the results. Furthermore, the methodology used throughout the work of this paper will be discussed to highlight possible factors that may have influenced the results.

This study presented a development of an existing hydrodynamic system in which the blood flow in the spinal canal can be simulated. Moreover, the development concerns modifying the model to allow a different type of input data to be utilized. Primarily, the input data shall be based on volume output from modeled airbags in the HBM ViVA+.

It was noticed that either the properties of the airbags or the implementation in its entirety did affect the stiffness of the HBM. This impacted the actual whiplash motion of the neck during the simulation. In other words, the traditional whiplash motion was not performed due to the implementation of the airbags. Therefore, the properties and the actual modeling of the airbags should be optimized in order to avoid uncertainties in the extracted volume data. Figure 12 in 4.1 shows the difference between the simulation with and without airbags, where the full extension was not completed with airbags. Due to the timeframe of the study, there was no time available to develop the modeling of the airbags. Optimizating the airbags would lead to more stable and consistent simulations of the ViVA+. Consequently, this would provide more reliable data on the volumes that should represent the spinal canal volume.

It was, however, possible to demonstrate volume variations with the ViVA+ from the implemented airbags along the cervical spine. The implementation of the airbags allowed extraction of the volume data directly from the airbags without excessive complications to be used as input into the hydrodynamic model. A drawback of this implementation is that it only adapts to a specific HBM. New modeling of the airbags is therefore required if a similar methodology is intended to be used for another HBM.

The modification in the MATLAB program allowed the data from the HBM to be used as input as long as the input file holds the correct amount of columns and arrangement. This implies that volume alterations in the spinal canal can be calculated directly utilizing the more accurate representation of the canal compared to the previous model by Yao et al. (2016). The previous model uses angular rotations of the vertebras to estimate the volume alterations in the canal. However, the subsystem in which the pressure transients are calculated still uses cylindrical tubes as a representation, therefore requiring vertebras' dimensions. In this part, the dimensions of the ViVA+ model were not implemented, which can generate lower accuracy in the pressure estimations. Huadong et al. (2016) have concluded that different vertebral dimensions only produce minor differences in pressure magnitudes. Therefore, implementing the vertebral dimension of the ViVA+ would not have affected the results to a greater extent.

6 Conclusion

The aim of the study was to develop the Matlab-Simulink hydrodynamic program developed by Yao et al. (2016) to calculate pressure transients for all directions of neck motion. Mainly, the central focus was to allow the program to accept a different type of input parameter sets. The previous program utilized angular displacement of the vertebras in the cervical spine as inputs. Thus, the main aim was to change the program to allow direct volume input from ViVA+ simulations. In this chapter, conclusions will be drawn as to whether the aim was reached by answering the specifications of issue under investigation. In addition, an evaluation of the study is carried out to provide recommendations for future research regarding the topic.

6.1 Answers to research questions

The conclusions regarding the research questions are presented further:

- *RQ1: How shall the MATLAB program of (Yao et al., 2016) be modified to use direct volume to replace the current vertebral angular displacement input?*

Firstly, the foremost step required to implement this modification is approaching the issue from an three-dimenisonal perspective. That is, capturing the volume alterations in the cervical spine directly from a three-dimensional model. The reason for executing such a step is to actually gain knowledge on how such data is formed. Based on that information, the program can be modified to satisfy the received data. As explained in the methodology of the study, the subsystem in the program that calculates the pressure transients accepts volume changes for each segment in the cervical spine. Therefore, the implemented airbags in the ViVA+ model needed to be divided in such segments to account for the different volume changes respectively. After that, it was necessary to differentiate the data to consider the volume change between each time step of the simulations.

- *RQ2: Can the new program be used also for vehicle side collisions? How do the pressure transients compare to those of rear-end collisions?*

Since the input data is based on volumes from a three-dimensional model representation, the new program can be used for any vehicle collision. Nevertheless, the data should be based on simulations performed on HBMs. Therefore, the reason to conclude that the new program can be used for side collision is dependent on how the crash pulse was prescribed on the HBM. One simulation that uses lateral prescribed crash pulse was performed in this study, and data information on volumes from the airbags was acquired. Thus, the new program can be used for vehicle side collisons and all direction types of vehicle collisions. There was no significant difference in the pressure transient between rear-end impact and side-impact. It is possible to argue that this question is still open for future developments since it is concluded that the airbags' implementation affected the neck motion during the simulations.

6.2 Recommendations for future developments

With the program having new modifications dependent on the volume in the human spinal canal, it would not be desired to use this kind of method in all cases. One recommendation would be to apply an interface in the MATLAB program which allows the user to choose which version of the program to utilize. If, for instance, the user prefers to make calculations with input based on volume change, the new program would be suitable. The interface can be developed even further in such a way as to choose between different data files, such as rear-end collision or side collision. This would imply a user-friendly interface and be used for fast computations during vehicle safety assessments. Furthermore, the modeling part of the airbags should be investigated further to obtain a better representation of the spinal canal.

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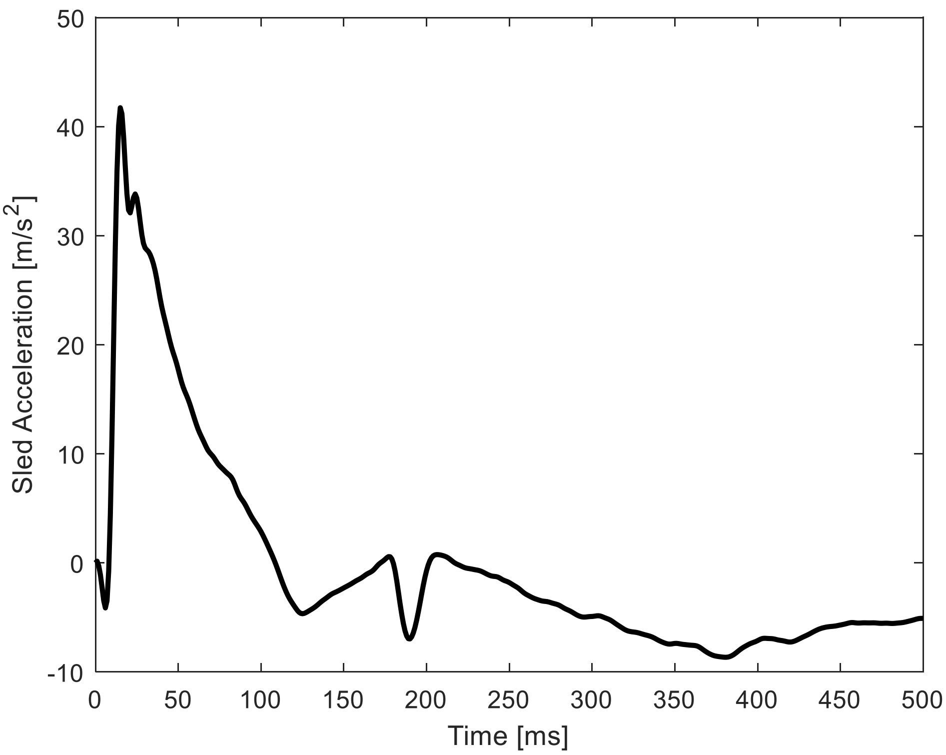
Appendix A

A.1

An overview of the modelled airbags in the ViVA+ model.

A.2



The time history of the acceleration pulse from the test conducted by Sato et al. (2014).

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