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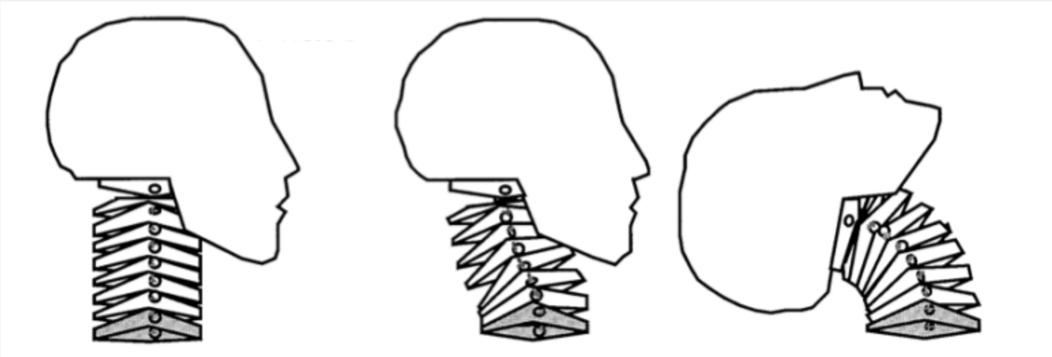

Assessment of a hydrodynamic model of Whiplash induced transient pressure in the spinal canal

Using in-vivo Whiplash simulation data

Master's thesis in Biomedical Engineering

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MASTER’S THESIS IN BIOMEDICAL ENGINEERING

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Abstract

During rear-end collisions, the occupant's body is accelerated and pushed forward by the seat-back while the unsupported head lags behind, resulting in the Whiplash motion to the cervical spine. Whiplash injuries are one of the most common injuries in rear impacts, that often cause pain and can cause symptoms such as dizziness, headaches, vision disorder, and neurological and upper extremities. During the rapid Whiplash motion in the neck, blood redistribution in the internal vertebral venous plexus will compensate for the rapid volume change inside the spinal canal, and as a result impulsive pressure transients are induced. It has been hypothesized that the transient pressure changes can result in injurious loading on the cervical dorsal root ganglia, This may explain the mechanism of injury and the source of the symptoms that whiplash injury victims suffer from. This project aimed to analyze high speed X-ray movies of porcine subjects in in-vivo whiplash experiments of varying severity, to digitize vertebral motion of the neck during the tests, and to evaluate the accuracy of a previously developed MATLAB-Simulink hydrodynamic model of transient pressure changes in Whiplash motion using the data from the mentioned experiments. Overall, the simulated results are in the same order of magnitude compared to the experimental readings, but the timing seems to have a delay. The raw X-ray movies had a limited noise-signal ratio due to the relatively thick soft-tissue layer in the porcine neck that resulted in a limited contrast in the video frames. There is potential for more accurate tracking of the vertebral angular motion with modified signal filtering or possibly by applying machine learning techniques for improved pattern recognition in the video frames.

Keywords: Whiplash motion, Neck injury, Pressure transients, Internal vertebral venous plexuses, Dorsal root ganglion, Hydrodynamic system of tubes

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1 Introduction

This chapter presents an overview of the background and key concepts related to whiplash injuries in motor vehicle accidents and outlines the objectives of this thesis project.

1.1 Background

Whiplash associated disorder is the most common injury in motor vehicle accidents yet poorly understood [1]. In Sweden, Whiplash associated disorder accounts for approximately 50% of traffic injuries with long-term symptoms. Frontal impacts account for 30% of long-term WAD cases, rear-end impacts for 50%, and side impacts for 10%. [2]. The majority of whiplash injuries tend to occur in urban areas. Thus, with the increase in urban traffic, the incident rate of whiplash is also increasing [3, 4]. The annual incident rate of whiplash injury in north America and west-European countries was reported to be greater than 300 per 1000000 people [5]. The neck injuries sustained in rear-end crashes are mostly classified as minor injuries (AIS 1) on the abbreviated injury scale (AIS). However, it leads to permanent disability in approximately 10% of the cases, affecting the life quality of victims and having high social costs [6]. The annual cost of WAD was reported to be approximately £3.1 billion in United Kingdom and €10 billion in Europe, including costs such as medical care, sick leave, and lost work productivity [7].

The development of Whiplash prevention seats has shown variable effects on lowering the risk of whiplash injury [3, 8]. A better understanding of the mechanism and the source of injury is crucial to improve the preventive, diagnostic and medical intervention methods.

1.2 Whiplash motion kinematics

Head and neck kinematics during whiplash motion in rear impact consists of three main phases. The first phase (retraction) begins with the transmission of the impact force from the seatback to the occupant's torso. Consequently, the torso is pushed forward, but the head while not in contact with the head-restraint tends to remain in its initial state of motion due to inertia and lags behind the torso. Hence, the upper cervical spine is forced into flexion, and the lower cervical spine is forced into extension, resulting in an S-shape curvature (Fig. 1), that is regarded as a most critical and potentially injury producing time point in the Whiplash motion sequence. This phase typically occurs within the first 100ms of the whiplash motion. At the end of this phase the lower cervical spine reaches its natural maximum extension limit, and the upper cervical spine reaches the maximum flexion limit.

In the second phase, the head begins to rotate rearward, and the upper cervical spine goes into extension while the lower cervical spine tends to go into a less extended posture. This phase corresponds to the next 50 ms of whiplash motion, from 100-150 ms.

During the last phase the whole cervical motion goes into full extension. At the end of this phase the head rotation stops, and the rebound phase begins as the motion is reversed [9].

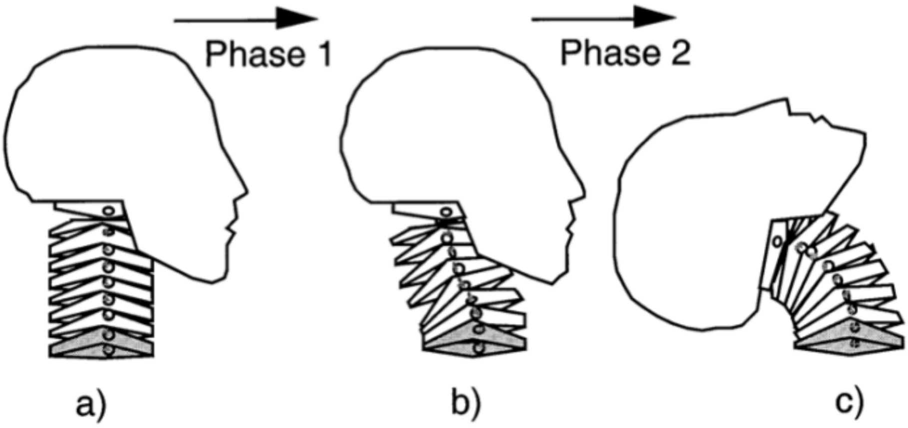

Figure 1. Schematic drawing of the retraction and extension phases during Whiplash motion in rear-end collisions. [10]

1.3 Potential anatomical injury sites:

Six anatomical sites including vertebral arteries, facet joints, intervertebral discs, neck muscles, ligaments, and dorsal root ganglia (DRG) have been proposed to be the potential site of injury in Whiplash motion [1]. During neck extension the vertebral canal is shortened and the ligamentum flavum protrudes into the spinal canal, and thus the cross-sectional area and the inner volume of the cervical canal decreases (illustrated in Figure 2). During neck flexion, the opposite takes place. As the tissue and fluids inside the spinal canal are almost incompressible, the epidural veinous-plexus compensates for the change in the spinal canal inner volume by allowing the vein blood volume inside the canal to vary during flexion and extension, respectively.

During rapid flexion-extension movements such as whiplash motion, due to fluid acceleration and blood flow resistance, the mentioned mechanism will compensate for the volume change but due to the rapid dynamics this leads to a transient pressure gradient in the spinal canal and along the intervertebral foramina [11].

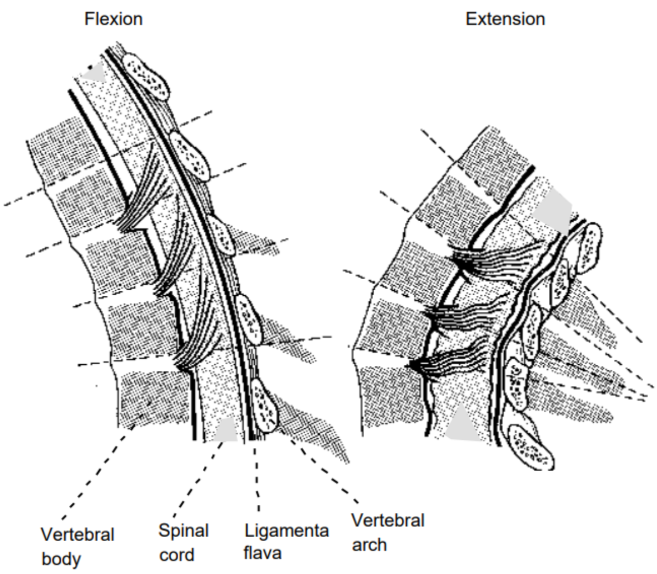

Figure 2. Volume change inside the cervical vertebral canal during flexion and extension visualized in a sagittal cross section.[12], as cited in [13]

Aldman hypothesized that the mentioned pressure gradients may cause stress and strain and consequent nerve tissue injury [11] in the cervical nerve root region. Studies on porcine subjects verified the hypothesis and revealed pressure pulses during the whiplash motion that were caused by the mentioned hydrodynamic mechanism. Signs of nerve cell body membrane dysfunction and leakage were evident in the DRG [14]. Cairns et al. also suggested neuroinflammation in the DRG as the potential mechanism for acute to chronic pain transition in whiplash injury victims [15]. Additionally, Panjabi showed that cervical extension leads to foraminal width narrowing that may cause ganglion compression, especially in subjects with foraminal spondylosis with smaller bony dimension of foramina [16].

1.3.1 in-vivo porcine test

While previous porcine whiplash studies used a pre-tensed strap to transmit force to the head and generate whiplash exposures on porcine model [14], in a more recent study servomotors were used to program specific motion profiles to simulate more controlled and repeatable whiplash exposures [17]. The two experimental set-ups are visualized in Figure 3. In the later study, a biteplate was used as the device-animal interface to actuate the animal's head via four attachment points to the motors. Due to the anatomical and scale similarities to the human neck, Yorkshire pigs were selected for this study. To enable live spinal CSF pressure measurement, each pig underwent surgery to implant 3 fiber optic pressure transducers in the spinal cord intrathecal space with their sensing tips at C2, C5 and C7 vertebral level. Each animal was exposed to several whiplash exposures with different severities and motion profiles. These were generated using 4 parameters including: 1. Retraction distance, 2. Retraction duration, 3. Extension angle, 4. Extension duration. To increase severity during tests, extension angle and retraction distance were kept unchanged while duration of the retraction and extension phases were decreased to increase test speed and severity.

The vertebral motion was recorded during each test using high-speed X-ray movie instruments with a 1000 fps frequency [17]. The output from this study will be analyzed in the current project as described in the next sections.

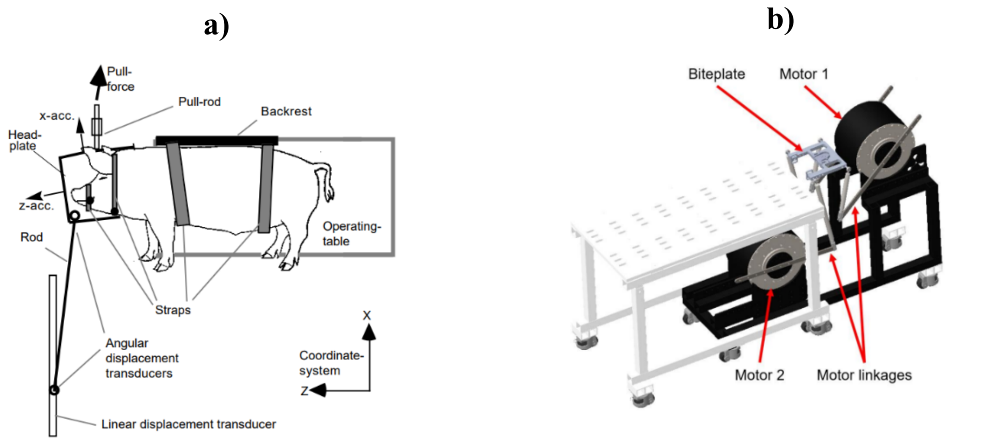

Figure 3. Test setup of two in-vivo procine experimentss, simulating Whiplash motion in rear-end collisions. a) using a pull-rod to pull the head-plate. b) using servomotor to program the motion profile.

1.3.2 Computer simulation

Yao et al. [19] modelled the transient pressure changes in the spinal canal during whiplash motion and was able to quantify the relation between pressure and neck motion. MATLAB-Simulink software was used to develop a model that simulated the venous blood flow and the corresponding pressure- time history in the spinal canal during whiplash motion.
As previous studies had indicated that the flow resistance of the CSF is relatively high, this fluid was omitted in the modelling. Thus, only blood transportation in the Internal Vertebral Venous Plexus (IVVP) was included [18]. In Simulink a virtual hydrodynamic system of cylindrical tubes was representing the IVVP as well as the lateral intervertebral vein connections. With this Simulink model it was possible to predict pressure changes inside the vertebral canal during whiplash motion, only using the angular motion data of each neck vertebra as the input [19].

While the Simulink model only calculated the generated transient pressure changes but was unable to model local mechanical soft-tissue loading and deformation, a more recent three-dimensional fluid-structure interaction model was developed to study the pulling and pressing processes on the DRG caused by the transient pressures. It was observed that the DRG is stretched due to the negative pressure phase, and prominent Von Mises stress took place near the fixed ends of the DRG. Additionally, these simulations indicated that the the negative pressure phase leads to a pulling process on the DRG, which is more efficient in deforming the DRG [20].

1.4 Aims

The main aim of this project is to validate the hydrodynamic whiplash simulation model developed by Yao et al [19] and to evaluate its accuracy in predicting transient pressure changes during whiplash motion. More specific objectives of the project were:

1. Digitize vertebral motion from high-speed x-ray movies of previously performed in vivo porcine experiments.
2. Run the model for the derived motion data and compare the simulated pressure pulses with the measured pressure values from transducers during the experiments to evaluate the model's validity.

The results of this project are intended to enable additional validation of the hydrodynamic model to explore the transient pressure hypothesis and potentially adding guidance to further refinements of the model. Additionally, the results may become helpful in future implementation of the hydrodynamic model developed by Yao et al. as a post-processor software to Human body models.

2 Methodology

2.1 Movie preprocessing

Four high-speed X-ray movies of in vivo porcine tests simulating Whiplash motion from the described experiment at university of British Columbia were selected for analysis. The chosen tests were the simulations with the highest severity (lowest duration) and an initial retraction distance in 3 different pigs. A less severe test from the first pig was also chosen to evaluate pressure calculations for different severities in one test subject. As the movies' quality were limited, each movie was down-sampled by averaging each of the three neighbor frames in order to have a more constant contrast and brightness throughout the whole movie. This process was done using MATLAB R2019b. Various image enhancement techniques were tested to reach a quality high enough to digitize vertebral motion. Finally, manual annotation of all frames before analysis was chosen as it led to more stable outputs compared to the automated annotation option. For this purpose, all frames of each movie were extracted using a MATLAB script, and a surface pen was used to draw borderlines of cervical vertebrae in all frames manually in Microsoft Photos as shown in Figure 4. Consequently, a MATLAB script was used to turn the annotated frames into a movie again.

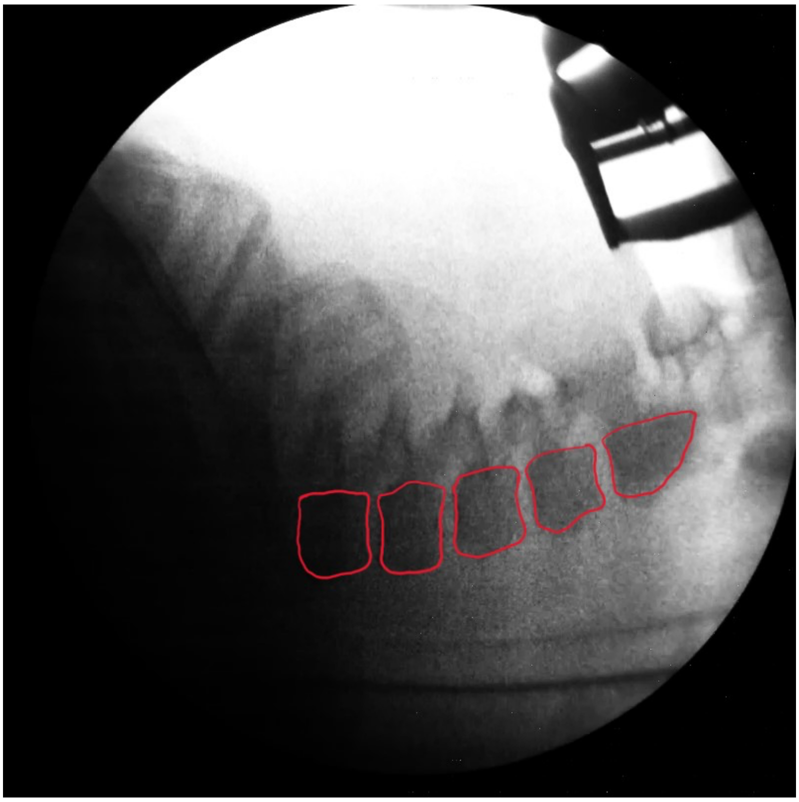

Figure 4. An annotated frame of the X-ray movie from test #1.

2.2 Motion digitization

TEMA Automotive software, version 3.8 was used to calculate angulation of each vertebra throughout the tests. Processed movies were fed to the software, and two points were marked on the posterior edges of each vertebra as shown in figure 5. The line between the two points was defined as the coordinate at time zero, and angulations were calculated as the angle between the line at each time point and the line at time zero. Due to the low movie quality, the automatic tracking feature of the software could not be used, and markers were set manually at each frame. Angulation of each vertebra over time was calculated automatically in the software.

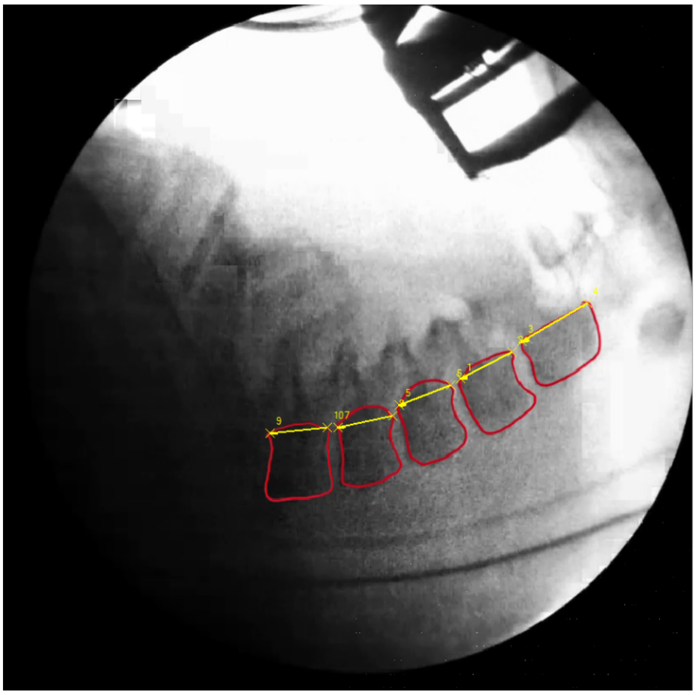

Figure 5. Marked points on the posterior part of the cervical vertebrae to track angular motion in a frame from test#1.

2.3 Data processing and filtering

Despite the annotation, it was not possible to derive smooth motion plots. Due to the sensitivity of the Simulink model to artefacts such as small fluctuations in the motion, it was crucial to filter the motion data before using it as the input to the Simulink model. Figure 6 and 7 display the raw angular motion plots and the processed data after filtering for exposure 1 and 2 (Table 1) respectively. To process the raw data, a 147 msec zero-padding was added to the beginning of the data to prevent sharp slopes at the initial phase of motion after polynomial curve fitting. Consequently, a MATLAB script was used to fit polynomial curves to the raw data and a moving average filter was used to further smooth the initial phase in the fitted curves. As in most frames C1, C6 and C7 were not completely visible for tracking, a simplified assumption was introduced assuming that C2 and C1 move together and thus have the same angulation data, and motion data of C6 and C7 were considered to be 2/3 and 1/3 of angulation of C5 at each time point respectively. T1 was defined as stationary in all experiments.

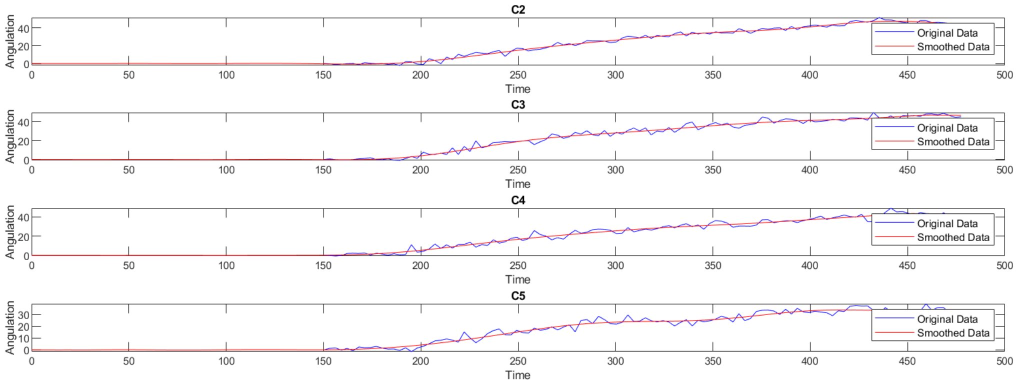

Figure 6. The angular motion of a) C2, b) C3, c) C4, d) C5, during exposure #1, before and after data processing. A 147 msec zero-padding was added to the raw data.

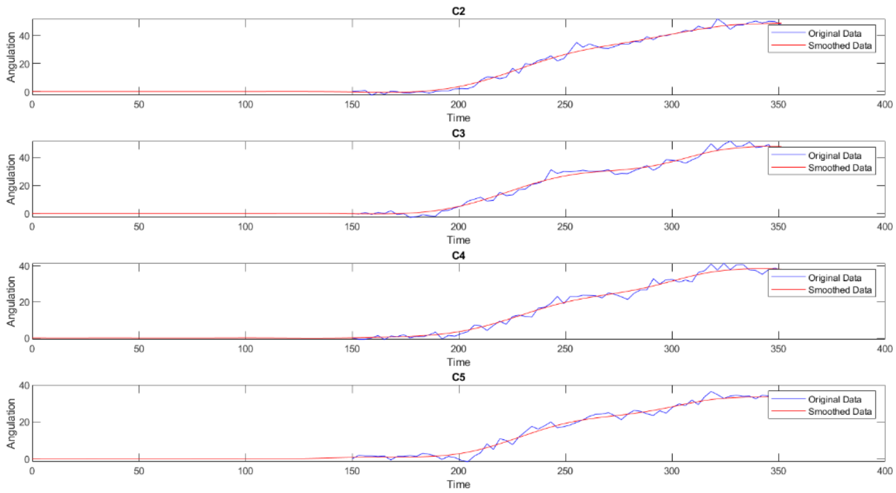

Figure 7. The angular motion of a) C2, b) C3, c) C4, d) C5, during exposure #2, before and data processing. A 147 msec zero-padding was added to the raw data.

2.4 Simulation

After data processing, digitized motion data of all cervical vertebrae and stationery T1 for each experiment were fed to the model, and the pressure pulses were calculated by the model at various cervical levels. As the pressure data from transducers were at the vertebral level of C2, C5 and C7, the same measurements from the simulation were also selected and plotted to compare to the experimental data.

3 Results

In the following chapter, the calculated vertebral motion and the corresponding simulated pressures are presented and compared to the sensor readings. Out of the four pigs that underwent Whiplash tests, the available data for the first two pigs included whiplash simulation with an initial retraction, whereas the available data for third and the fourth pig was from pure extension tests with no initial retraction and were less priority for this project, thus not analyzed. Additionally, the quality of the X-ray data from the second subject was insufficient and the corresponding analysis was omitted in the present study. Table 1 shows the details of the analyzed experiments. In these experiments, the duration of the extension and the retraction phases were used as the altering exposure parameters to create different severities in Whiplash motion exposures.

Table 1. The test settings for the two analyzed experiments.

#	Test subject	Initial retraction [m]	Maximum head extension [deg]	Extension duration [s]	Retraction duration [s]
Exposure 1	Pig 1(test #04)	0.05	80	0.20	0.060
Exposure 2	Pig 1(test#09)	0.05	80	0.11	0.033

Figures 8 and 9 display the relative angular motion and pressure plots for both tests. The start of the angular motion is delayed about 60 and 33ms for exposures #1 and #2 respectively that indicates that the neck is mainly moving translationally during the initial retraction.

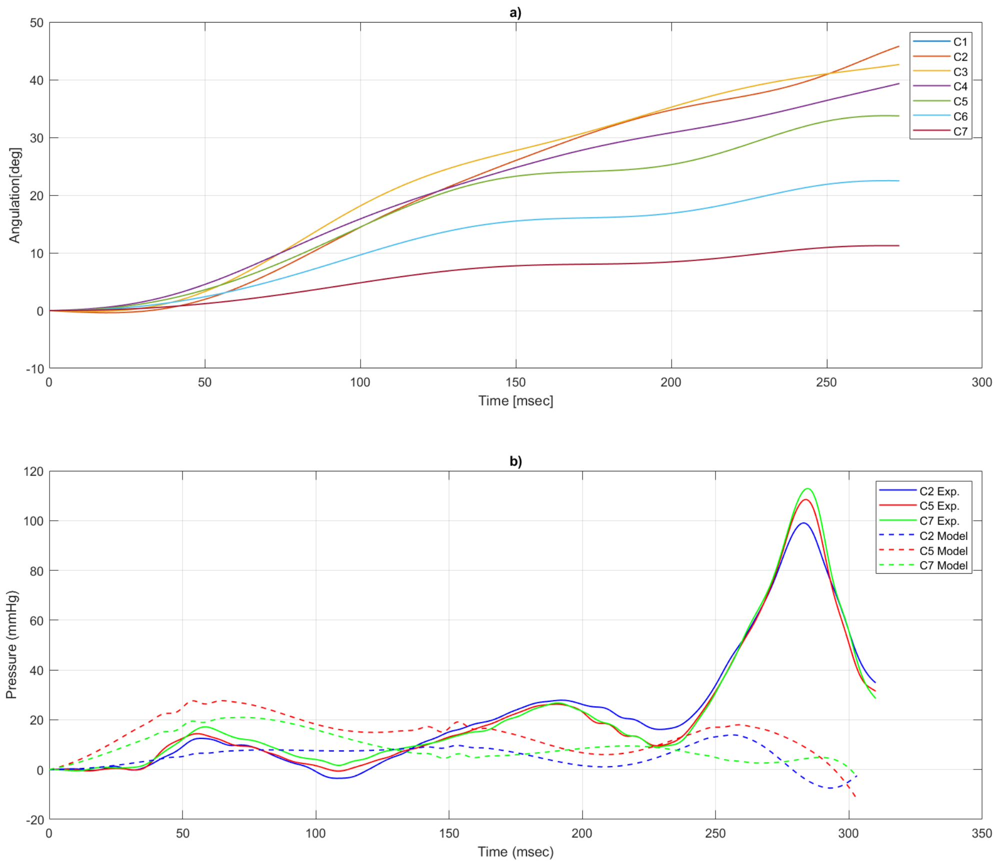

Figure 8. The derived a) angular motion b) experimental and simulated pressure pulse plots for exposure#1.

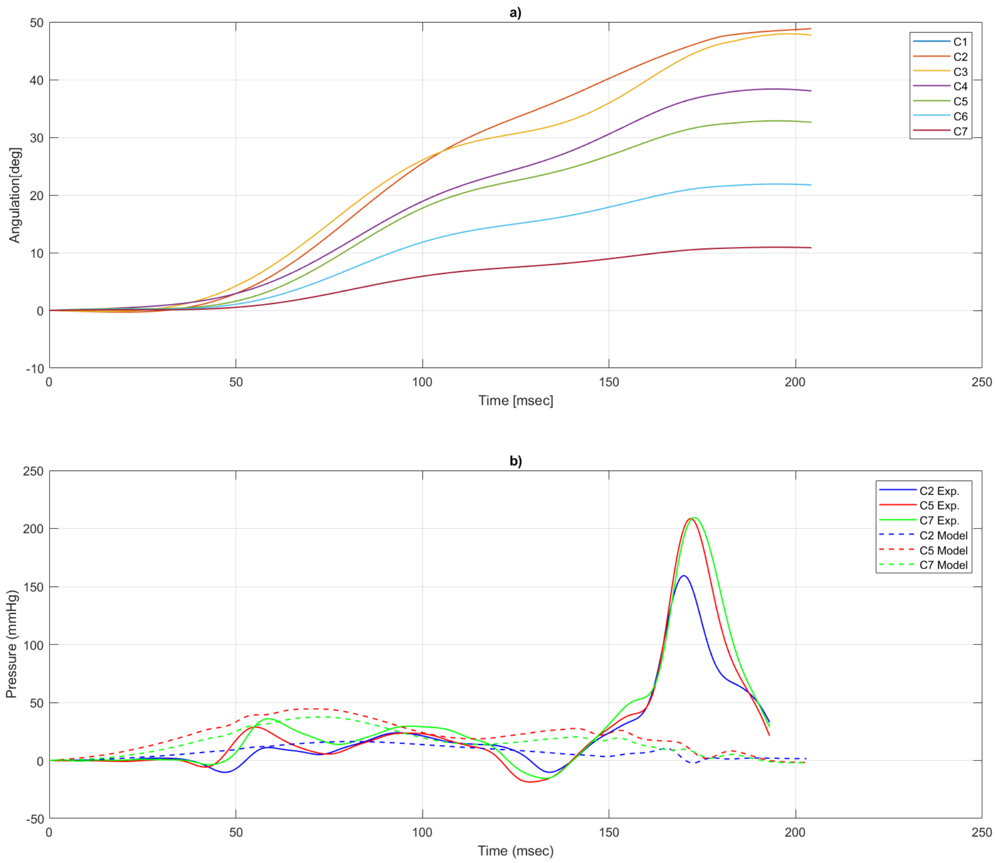

Figure 9. The derived a) angular motion b) experimental and simulated pressure pulse plots for exposure#2.

The simulation results for both exposures show two pressure maxima, the first one at the beginning of the motion and the second one about 50 ms before the maximum head extension (around 240 ms and 140ms for exposure #1 and #2 respectively) followed by a small dip at the maximum extension. There is a distinct time delay in the simulated pressure pulses compared to the sensor readings.

4 Discussion

In this project, X-ray movies from two experimental in-vivo whiplash exposures in an anaesthetized piglet, with low (1#) and high (2#) severity, were analyzed and used as input to the hydrodynamic model of Yao et al. [19] to predict pressure transients. The general trend of the simulated pressure magnitudes was consistent with the trend in the sensor measurements, comparing between exposures 1# and 2#. However, the similarity in timing between experiment and simulation was limited. In the low severity case, the first maximum pressures are slightly higher than the measurements and the second maximum pressures are slightly lower. The high severity case displays a better resemblance in predicting the pressure magnitudes. In both tests, experimental data shows that during the retraction phase the pressure values at the level of C7 are slightly higher than C5, while in the simulations this is vice versa.

There are dips in the experimental pressure signals at about 100 ms and 70 ms respectively. These dips occur at the end of the retraction phase in each of the exposures. In the earlier work of Svensson et al. [13] these dips occurred as very distinct negative pressure peaks. These negative peaks were hypothesized to cause nerve cell membrane dysfunction, or leakage, in the posterior nerve root ganglion cell bodies. This hypothesis was supported by the findings of Yao et al. [20].

In both exposures the transducers measured a high peak at maximum extension (around 280 ms and 170 ms respectively) which was not predicted by the model. These peaks are most probably artifacts caused by excessive loading on the tip of the catheter. This is most likely the effect of squeezing and bending of the transducer as the soft tissue in the spinal canal presses against the transducer while the spinal canal reaches its maximal shortening and maximum rearward bending. Thise peaks should thus be disregarded in the analysis.

The main limitation in this study was the limited resolution and contrast of the X-ray movies that lowered the accuracy in the tracking and digitizing of vertebral angular motions. Consequently, the vertebral angular motion raw data plots were very noisy. The soft tissues of the biological body do not allow such high-frequency fluctuations as displayed in figure 6, so this is an artifact that needs to be filtered out before the simulation, to prevent artificial pressure transients in the output. Thus, fitting the motion curves to a polynomial function followed by applying a moving average filter to the created curves were used to smooth the angular motion data. Due to high level non-periodic noise in the data, it was impossible to precisely reproduce the actual motion and exact time derivatives using filtering techniques. As the model predictions are mainly based on time derivatives, this may explain the trend and magnitude differences between the measurements and the simulated results.

One opportunity for further refinement of the method would be to refine the angular motion output filtering. In fact, the raw signals in Figure 6 indicate that the C3 angulation starts about 40 ms followed by C2 and C4 at 50ms and last C5 at 55ms. This corresponds to an s-shape. In the filtered curves however, these time differences are not clearly visible, so highly essential information has been filtered out.

Additionally, as in most frames C1, C6 and C7 were invisible or too blurry to track. It was decided to simplify the calculations by assuming that C1 and C2 move together and have the same motion plot, in addition the angular displacements of C7 and C6 were assumed to be 1 third and 2 thirds respectively of the C5 angular displacement. These simplified assumptions would potentially cause deviations in the simulation results compared to the sensor readings.

Another possible explanation for these differences is that the model has been designed based on the anatomical dimensions and mechanical parameters of the human vertebral canal whereas the data used in this study came from porcine experiments. Even though there is a high similarity between the cervical spines of the human and the pig, they are not identical, and the differences can affect the model output.

While the results indicate overall magnitude and trend similarity between the simulation output and experimental results, further work is required before additional validation of the pressure simulation model is feasible. Acquiring X-ray movies of better contrast and resolution would be highly beneficial in digitizing motion more accurately and decreasing artificial pressure artefacts caused by noisy input. In theory, validating the model using data from human subjects would be more beneficial as the pressure simulation model parameters are defined based on human cervical spine dimensions and dynamic properties.

In this project, an alternative movie tracking method was initially tested with the second pig subject. The approach included a physical template made of a transparent sheet that was used in the annotation of the borderlines of the vertebrae in each frame. This was done in an attempt to increase the accuracy in the tracking and to lower the output noise caused by the limited repeatability in the identification of distinct locations on the vertebral body images in consecutive video frames. While this approach appeared promising it did in the end not improve the tracking accuracy. It did not result in high enough accuracy to derive meaningful pressure predictions from the data of the second test subject.

Due to lack of time and training data, it was out of the scope of this project to use machine learning models for this purpose, but it is recommended to explore the use of such models for object detection in future studies. That may allow annotation of the vertebral positions more accurately before digitizing the angular motion.

5 Conclusion

The aim of this study was to assess the accuracy of the MATLAB-Simulink model developed by Yao et al. in predicting whiplash motion induced pressure transients in the cervical spinal canal using experimental data from an in-vivo whiplash injury porcine model at UBC, Vancouver. For this purpose, first the X-ray movies from the experiments were analyzed to digitize the angular motion of the vertebrae during the whiplash simulation. Consequently, the output motion data was filtered and conditioned and used as the input to the hydrodynamic model.

The results indicate meaningful similarities between the computationally predicted pressure pulses and the transducer outputs in both trend and the magnitudes for both low and high severity tests. However, the pressure time history is not well reproduced. The main explanation for these differences is inaccuracy of the input data due to the limited quality of the x-ray movies and the artifacts in the digitized motion plots and their time derivatives as a consequence of the manual motion tracking as well as the filtering process and simplifications in calculating motion values for C1, C6 and C7. Another, lesser limitation is the use of experimental data from porcine subjects while the pressure simulation model has been developed using the properties of the human cervical spine and neck tissue. Even though the cervical spinal anatomy of a pig and a human are similar, they are not identical, and the differences can affect the model's calculations and the final pressure pulses predicted by the model.

Overall, further assessment and validation of the pressure simulation model using results from in-vivo porcine experiments will be helped by X-ray movies of higher quality. Another potential improvement could be to adopt machine learning techniques to overcome the quality limitations of the current x-ray movies. Using data from experiments with various severities and settings would be beneficial to evaluate the model's robustness and may provide improved possibility to relate the pressure pulses with the potential dorsal root ganglion injury in test subjects.

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