SPINAL LOADING ON WHEELCHAIR OCCUPANTS WITH POSTURAL DEFORMITIES IN A REAR IMPACT DURING SURFACE TRANSPORT

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ABSTRACT

The rise in surface transport by wheelchair users raises significant safety considerations. In particular rear impact injuries have not been adequately addressed. In this paper, rear impact crash simulations of a wheelchair occupant with scoliosis have been used to quantify the influence of scoliosis on spinal loading. The results show an increase in effective sagittal plane stiffness introduced by the curvature of the scoliosis, which results in increased spinal loading. Furthermore, the complexity of the deformed spinal curves means that the commonly used Cobb angle classification is not an appropriate means of assessing wheelchair user spinal loading during impact.

Keywords: DISABLED (persons), KINEMATICS, REAR IMPACTS, SPINE, WHEELCHAIRS

IT IS ESTIMATED that in Ireland 700 wheelchair users make 500,000 road journeys annually, remaining in their wheelchairs during transit. This growing usage of surface transport by wheelchair occupants highlights the need for research to be conducted to provide these occupants with the same level of crash protection as an automotive seat occupant. To date most wheelchair crash protection research has focused on the performance of the wheelchair and the wheelchair tie down and occupant restraint system (WTORS) in an impact. The WTORS consists of a 4-point tie down and occupant restraint connected to specially adapted vehicles, figure 1, the size of which varies from small privately owned vehicles (M1 vehicles) to medium-sized transit buses (M2 vehicles) used for transport of school children. Legislation such as ISO 10542 regulates the requirements for a WTORS and their connection to the wheelchair and vehicle [ISO 10542, 2001]. However, to date there has been no evaluation of injury criteria, and this is in contrast to crash safety standards for passenger vehicles (FMVSS 202, 208 etc.), which require that injury thresholds assessed using dummies representing the general population are not exceeded. In contrast, the primary goal of wheelchair crash safety standards has been wheelchair and occupant retention, and hence prevention of serious subsequent impacts with the vehicle interior.

Figure 1: Wheelchair Tie Down and Occupant Restraint System (WTORS)

The ISO standard 10542 requires a WTORS to retain the occupant and secure the wheelchair in a 48kph/20g frontal impact [ISO 10542, 2001], with no reference to rear

impact. However automotive research has shown that although rear impact accounts for only 5% of fatalities [Viano, 2002], this crash mode results in 30% of automotive related trauma in the general population, and for more long term injury from low severity impacts than any other crash mode [Viano, 2002]. The lack of rear impact investigation for wheelchair crash safety will be addressed in this work.

The second shortcoming of research to date is the use of standardised crash dummies which are generally a poor representation of many wheelchair occupants, since both their shape and mechanical response may not correspond well to given wheelchair user. Scoliosis which is commonly found among wheelchair users is a 3D deformation of the spinal column. It has been found that gender, age, height and weight all influence the risk of injury for the general population [Temming, 1998]. It therefore seems likely that the presence of a postural deformity will also significantly influence injury likelihood. In 2000, Bertocci et al investigated existing injury criteria and noted that thresholds are typically based on values at which 25% of the general population would experience serious injury, but stated that "levels of risk associated with the use of able-body population based injury criteria may actually result in higher risk in the disabled population" [Bertocci, 2000].

Therefore the overall goal of this work is to assess the hypothesis that wheelchair occupant injury risk is influenced by postural deformity, and to then quantify this risk. The hypothesis was tested in the case of rear impact, using a combined physical and numerical approach. This method is broadly used in the automotive industry and has been successfully adapted for wheelchair crash safety research to determine the optimal parameters for safety such as WTORS position [Kang, 1998; Bertocci, 2000] and the effect of seatback stiffness/angle on seat loading [Bertocci, 2000]. Quasi-static tests of the seat, seatback and headrest have also been conducted to determine the optimal parameters for improved safety in a frontal impact [Karg1996, van Roosmalen2000, Ha 2000, 2002]

In order to provide baseline validation data, crash tests using a BioRID II occupant in a surrogate wheelchair were conducted. These tests formed an integral part of the goals of the research team in Trinity College as the data from the physical tests has also been used to investigate the provision of a detachable head restraint for wheelchair occupants and the performance of seating attachment hardware.

A computational model of the baseline wheelchair crash test was developed using MADYMO (version 6.4.1, 2007). Scoliosis based on x-ray data, was implemented into this model and these scoliotic dummy models were then subjected to simulated crash tests. The validation of the baseline model and the development of the scoliosis models are reported in detail in Walsh et al [2008a; 2008b], and an overview of this process is given in the methodology section of this paper.The key findings of the paper detailing the scoliosis models were that the vertebral joint constraint forces in the direction of the crash (X component) in the thoracic region of the spine increased substantially (approximately 200%) when scoliosis was introduced [Walsh et al 2008b]. The maximum increase when scoliosis is applied was at the apex of the scoliotic curve, figure 2. Figure 2 shows the graph of the peak forces found in the 'moderate' scoliosis case (Cobb angle = 42°) and it is seen to directly correspond to the apex of the deformed spine shown in the x-ray. This pattern was also found for the 'mild' degree of scoliosis (Cobb angle = 18°). The forces on the spine are found to increase due to the altered shape of the surface of the back which leads to a point loading effect on the spine when it contacts the seatback. The forces in the lumbar region of the spine were seen to decrease in the scoliosis models when compared to the baseline case.

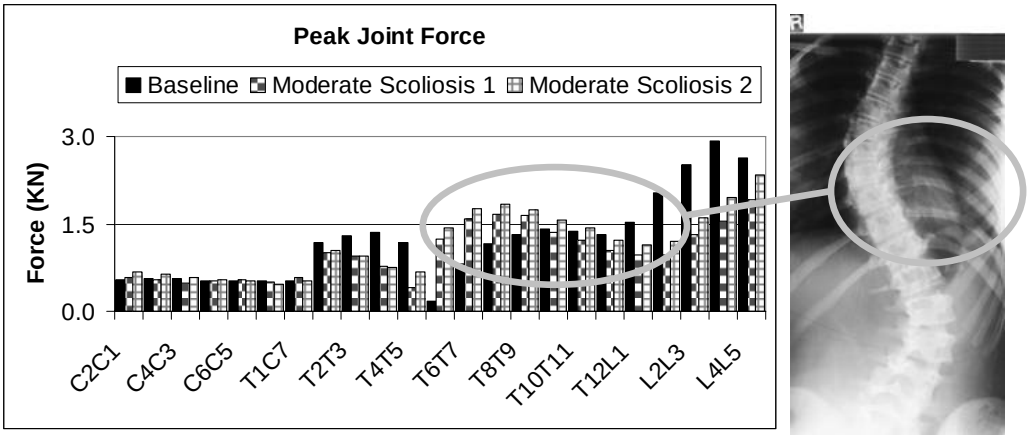

Figure 2: Joint Constraint Forces on the Spine in the X Direction for Moderate Scoliosis Case 1 and 2 (Adapted from Walsh et al 2008b)

It was found that when scoliosis was introduced to the model it was necessary to adjust the position of the occupant in the wheelchair as the deformed spine increased the sitting length of the occupant and to accommodate this new posture the dummy was moved. It was also found that adjusting the position (case 2) of the scoliotic dummy within the wheelchair produced higher loading than adjusting its orientation (case 1). The effect this has on the results is shown as the difference between Moderate Scoliosis 1 and Moderate Scoliosis 2 in Figure 2 This process was also necessary for the mild case of scoliosis and the adjusted position resulted in an average increase in force of 160N. The increase in force was attributed to the larger distance between the torso and the seatback which allows the rearward momentum of the dummy to increase and so the contact force is greater.

The models have also been assessed using injury criteria such as the *NIC* and the N_{km}. The *NIC* value for the baseline case was found to be $31m^2/s^2$ which is above the threshold value of $15m^2/s^2$. This result was not surprising as the test was carried out in a rigid wheelchair without a head restraint in place as it is not mandatory for wheelchair occupants to use a head restraint in transit. It was found that when scoliosis was introduced an increased level of injury, up to $49m^2/s^2$ was predicted. The N_{km} was also found to increase from 1.21 to 2.04 and 2.06 for the mild and moderate case respectively. The moment about the global Y axis was found to remain the dominant component of the N_{km} equation with values of up to 1.4 for the moderate case of scoliosis. These increases were found to be due to the increased acceleration that is passed to the dummy due to the point loading effect of the altered shape of the back.

The results presented in the paper detailing the scoliosis models provides the first evaluation of the influence of scoliosis in a rear impact, however further understanding of the mechanics underlying this is required. Previous investigations have shown that these occupants are more vulnerable in a crash and so knowledge is key to providing solutions [Walsh et al 2008b]. The level of deformity of each case of scoliosis is unique however to understand the general influence of scoliosis the specific behaviour of the models used in this study must first be determined, and this more in depth analysis is presented here. Improvements to crash safety for these more vulnerable wheelchair users can then be determined based on this full evaluation.

METHODOLOGY

A sled test of a wheelchair and occupant without a head restraint secured using a WTORS, figure 3(a), was conducted. It was subjected to a rear impact using the crash pulse specified by the International Insurance Whiplash Prevention Group of 16kmph/10g, figure 3(b) [IIWPG, 2005]. A BioRID dummy was used to simulate the movement of the occupant when subjected to a rear impact. The wheelchair used in the test was a surrogate wheelchair constructed approximately according to ISO 10542 [2001] which was designed to represent a powered wheelchair. This reinforced wheelchair was developed for repeated testing to

determine the requirements of the WTORS in a frontal impact. As this test had not been conducted previously a new protocol was developed which combined wheelchair frontal impact with automotive rear impact procedures.

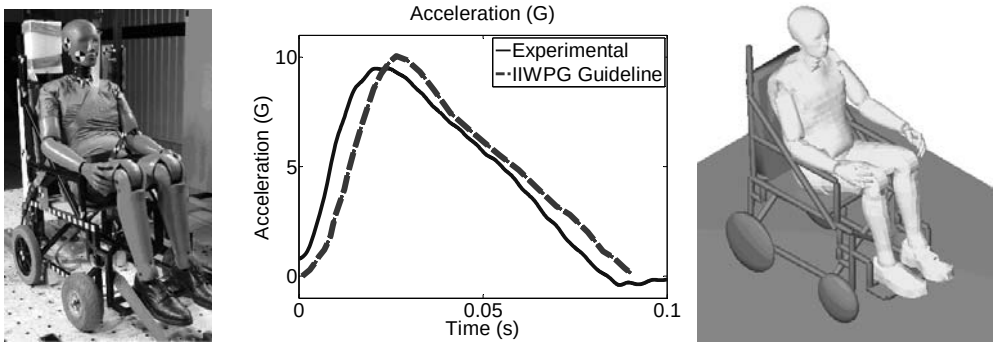

Figure 3(a): Test Set-up for Sled Test (b): IIWPG rear impact pulse (c): Computational Model of BioRID II in a Surrogate Wheelchair

The outputs from this fully instrumented test provided the data required for validation of a numerical model of the crash. This computational model was created using crash simulation software MADYMO (version 6.4.1 2007) which uses multi-body modelling techniques to create a model of the crash as shown in figure 3(c). Contact characteristics of the wheelchair, which define the force penetration response of the wheelchair, were required. The wheelchair was therefore tested and the penetration level found in response to a known force was recorded. The model was then validated kinematically and dynamically against the physical sled tests.

To validate the model comparisons were made between the crash video and the animation output and a very good kinematic correlation was found. Key events such as time of contact between the dummy and the wheelchair as well as hyperextension of the neck were seen to correspond. Snapshots of the models at 60ms time steps, figure 4, show good correlation between the physical and numerical outputs. This figure also shows the coordinate system used for the models. Time history plots of the outputs were also analysed. The acceleration of sections of the dummy were compared and the average difference between the peak outputs in the x direction was found to be 13.3%. The X direction is the principle axis of investigation as it is the direction of the crash pulse. The time history graphs of the head and first thoracic vertebra (T1) in this principle direction are shown in figure 5, and these show good correlation. The force (Fx) and moment (My) exerted on the upper neck are shown to correspond well for the physical and numerical model as shown in figure 5. These outputs are key elements for analysing the outcome of a rear impact as they are the components of injury criteria. From these results and further analysis in Walsh et al, 2008a it was concluded that the MADYMO model is a good representation of the physical test, and is therefore a suitable baseline model.

This baseline was used to quantify and qualify the injury level predicted for a wheelchair occupant subjected to a rear impact in the surrogate wheelchair, which had not been determined previously.

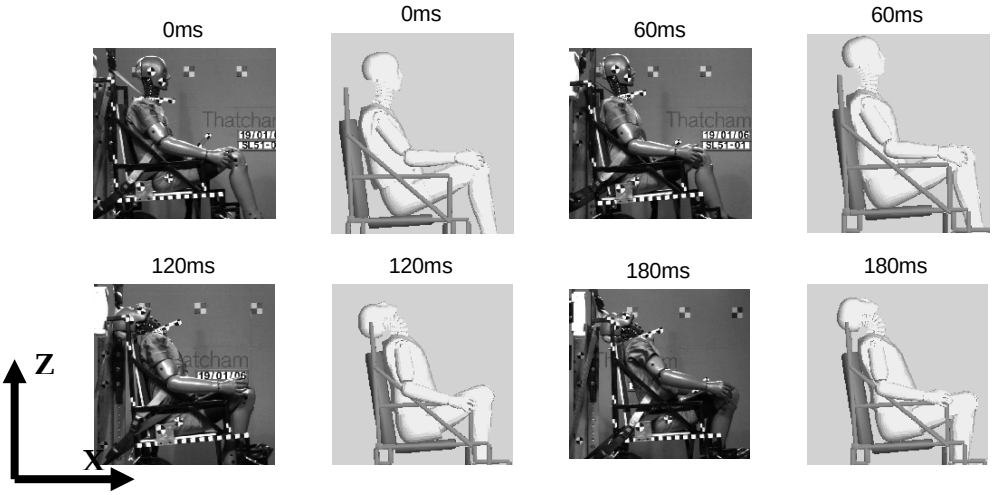

Figure 4: Still Shots of Video and Animation Output at 60ms Intervals

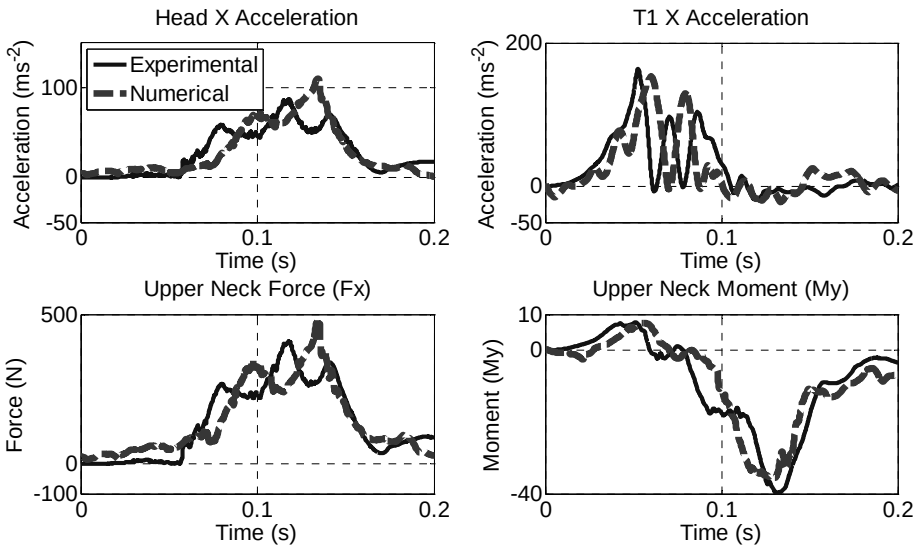

Figure 5: Head and T1 X Component of Acceleration Force (Fx) and Moment (My) Exerted on the Upper Neck, (Adapted from Walsh et al 2008a)

The validated model was then used as a baseline that was adapted to better reflect the altered posture of a scoliotic occupant. The three dimensional orientation of the vertebrae was determined from x-rays and applied to the validated model to create a crash test dummy with scoliosis. The x-rays were supplied by the scoliosis clinic in Crumlin Hospital, Dublin, Ireland. The scoliosis applied in each case is therefore based on real life data from x-rays of scoliotic spines. Each vertebra is orientated in 3D space so although the geometry of the vertebrae may differ between the x-ray and the BioRID the curvature of the spine is accurate. A mild and moderate degree of scoliosis was applied to the model, figure 6. The level of scoliosis is quantified by the Cobb Angle [Morrisy, 1990] which is based on the curvature of the spine as seen on an A-P x-ray of the patient. These two models were then crash tested and the outputs analysed using injury criteria and the forces throughout the spinal column were also assessed. The predicted spinal loading from crash simulations using the two cases of scoliosis was compared to the baseline case in order to assess the influence of the postural deformity.

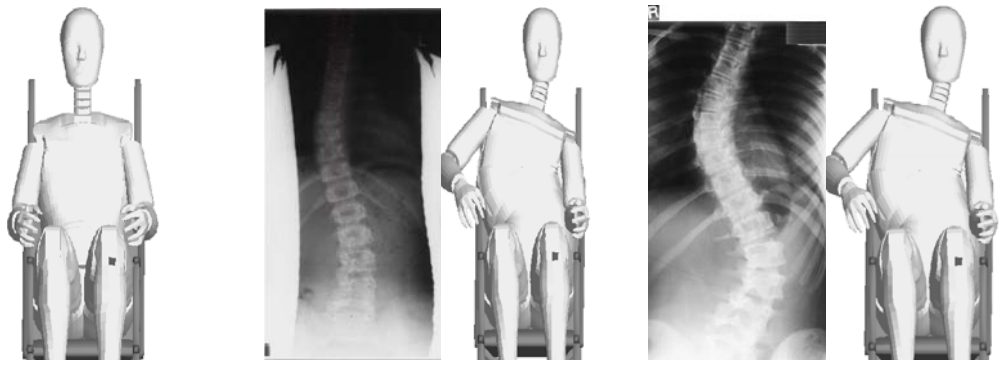

Figure 6: Frontal View of Baseline, Mild and Moderate Scoliosis with Corresponding X-rays

When scoliosis was introduced the models were found to have an increased seating length as the distance from the pelvis to the tip of the kyphotic curve in the thoracic region is increased. Therefore the altered seating posture when placed at the baseline position and orientation results in the upper torso penetrating through the seatback. To counteract this, the dummy's position needed to be altered without changing the shape of the scoliotic spine. Instead the dummy's position was changed by rotating (case 1) about the H point or translating (case 2) the dummy about the H-point as shown in figure 7. This was carried out for both the mild and moderate cases. From the animation it can be seen that the influence of scoliosis is to introduce asymmetry to the dummy motion which alters the kinematic response.

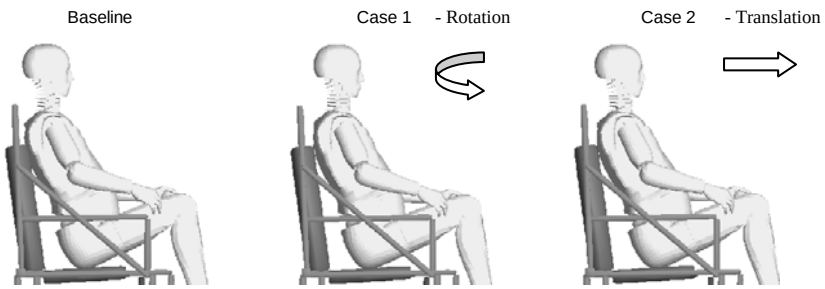

Figure 7: Lateral View of Scoliosis Model at Baseline Position and Orientation, Case 1 with Altered Rotation and Case 2 with Altered Position [adapted from Walsh et al 2008b]

RESULTS

The models previously used to evaluate the effect of a postural deformity in a rear impact previously were further analysed and the results are presented here. The degree of curvature of the spine was related to the altered loading pattern found in the dummy. The time of occurrence of the peak force values was also investigated to characterise the altered behaviour of the model with scoliosis.

ALTERED KINEMATICS: In all cases the scoliotic dummies were seen to pass through a different range of motion in the animation files. The loading pattern in the force outputs was also found to be different, and the underlying mechanisms were investigated. To do this the orientation in space of each of the vertebrae of the scoliotic and baseline spines when seated in the wheelchair, figure 8, was determined. This figure also shows the coordinate system of the model. These lateral views show the lordosis and kyphosis associated with the mild and moderate cases of scoliosis, the frontal views of which are shown in figure 6. A comparison of the orientation about the global Y axis, which is the axis of sagittal rotation of the spine, shows that the vertebrae with the greatest deviation from the baseline orientation were at the

apex of the scoliosis curve, figure 9. This deviation was seen to exaggerate the kyphosis and lordosis normally found in the spine. The largest orientation (10.5^o) in the thoracic region of the moderate case of scoliosis is at the apex of the curve. The joints in this area were previously found to be subjected to a large increase in force when compared to the baseline [Walsh, 2008b]. In the lumbar region the largest forces were also found at the joints between the vertebrae with the largest deviation from the baseline spine. In the mild case of scoliosis the greatest deviation in the lumbar region was 20°.

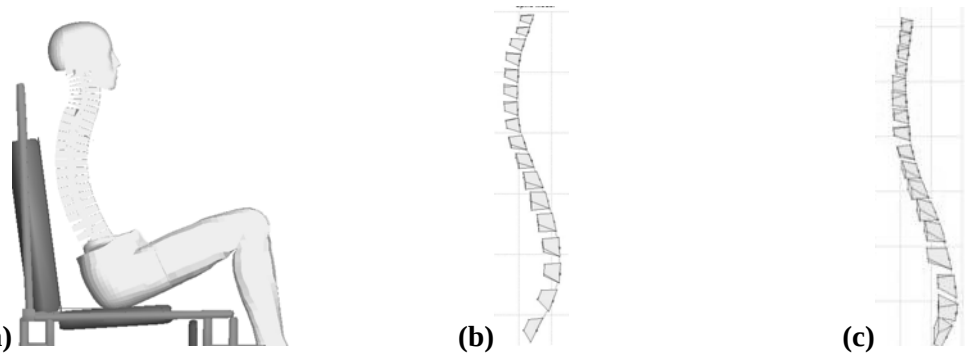

Figure 8: (a) Lateral View of Curvature of the Spine when Seated (b) Lateral View of Mild Scoliosis (c) Lateral View of Moderate Scoliosis

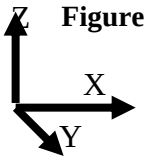

Figure 9 also shows that the orientation of the baseline case is quite uniform whereas in the scoliosis models it is less so. The general shape of lordosis and kyphosis is maintained by a combination of the larger and smaller angles that were found. The greatest deviation from the baseline case in the thoracic region is clearly shown to correspond to the apex of the curve at the joint between the 8th and 9th thoracic vertebra.

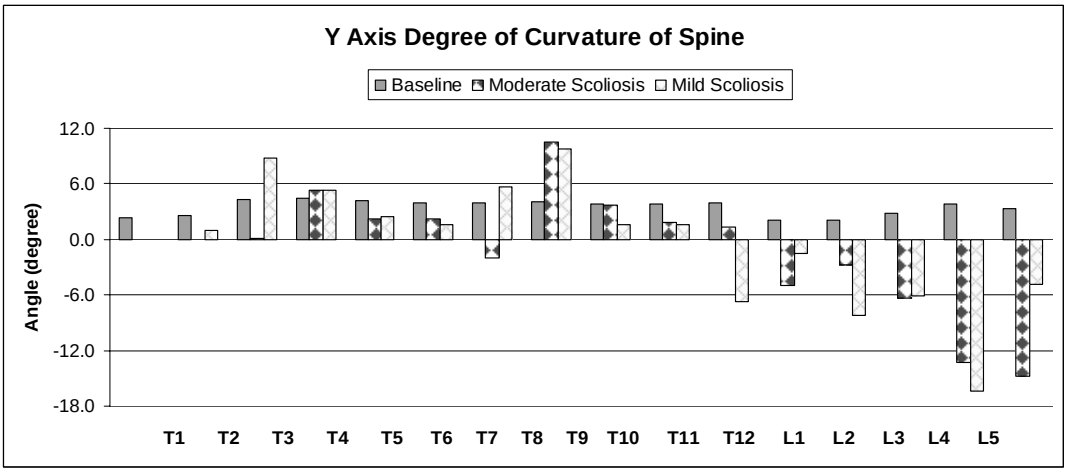

Figure 9: The Degree of Curvature about the Global Y Axis of each Vertebrae in the Thoracic and Lumbar Region for the Baseline, Mild and Moderate Scoliosis

COBB ANGLE: The level of scoliosis is quantified by the Cobb Angle [Morrisy, 1990] which is based on the curvature of the spine as seen on an A-P x-ray of the patient. The Cobb angle for the mild case of scoliosis is 18° and 42° for the moderate case of scoliosis. The orientation of the vertebrae about the global X axis as shown in figure 10 again shows the compensatory nature of a scoliosis curve with both positive and negative values.

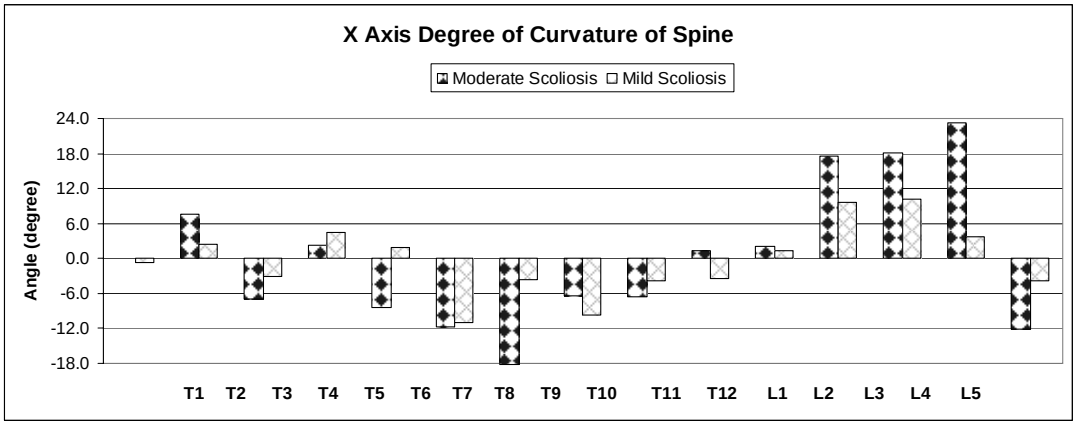

Figure 10: The Degree of Curvature about the Global X Axis of each Vertebrae in the Thoracic and Lumbar Region for the Mild and Moderate Case of Scoliosis

TIMING OF PEAK FORCES: The timing of the peak forces on the spine shows a distinct pattern for the baseline case as the spine straightens which allows full contact with the seatback. This pattern is displayed in figure 11 which shows the time of peak force in the global Y direction on the spine for the cervical, thoracic and lumbar regions. It can be seen that for this case the mid thoracic region contacts the seatback initially causing straightening of the kyphotic curve, and then the lower thoracic region contacts the seatback. The lumbar region then follows and contacts the seatback and finally the neck reaches peak force when it is fully extended. In figure 11 the positive force values due to the torso being pushed forward by the seatback are shown in black. The negative force values are shown in grey and represent the joints of the spine which are above the seatback and so the force is created by the backwards motion of the spine. The joint where these two areas meet is found to reach its peak force value at a later stage due to the shift in the movement of the spine.

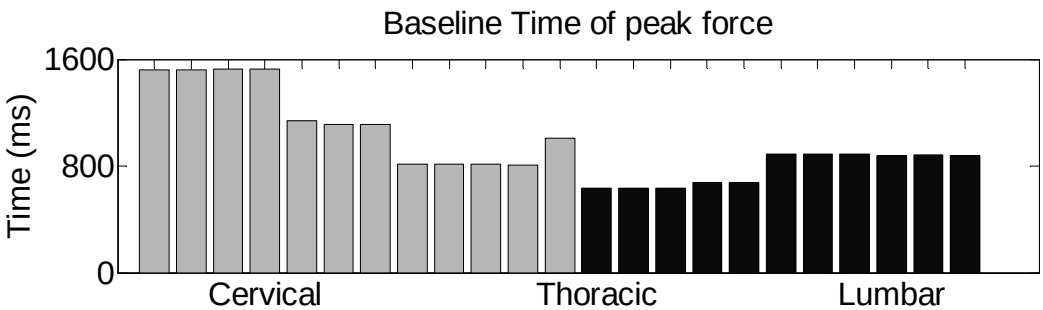

Figure 11: Time of Peak Forces in the Baseline Case, Black Denotes Positive Forces and Grey Denotes Negative Forces

A similar pattern was found in the scoliosis cases as shown figures 12 and 13, with peak forces in the thoracic region followed by the lumbar region and then the cervical area. However, the cervical region of each scoliosis model was seen to reach its maximum values earlier in the scoliosis simulation than in the baseline case. The other marked difference is in the lumbar region of the mild scoliosis case which is seen to reach its maximum at a much later time, this is due to the altered interaction of the deformed spine with the seatback. The effect of the sitting height of the dummy is also seen in these graphs, as when a moderate case of scoliosis is introduced the higher level of deformity causes a lower sitting height, and so

more of the spine contacts the seat back and is thus loaded with positive forces via contact with the seatback. In the figures shown the altered position (case 2) is presented, but it was found that the altered orientation (case 1) of each type of scoliosis behaved in the same manner. For this analysis it was the degree of scoliosis which influenced the results rather than the positioning.

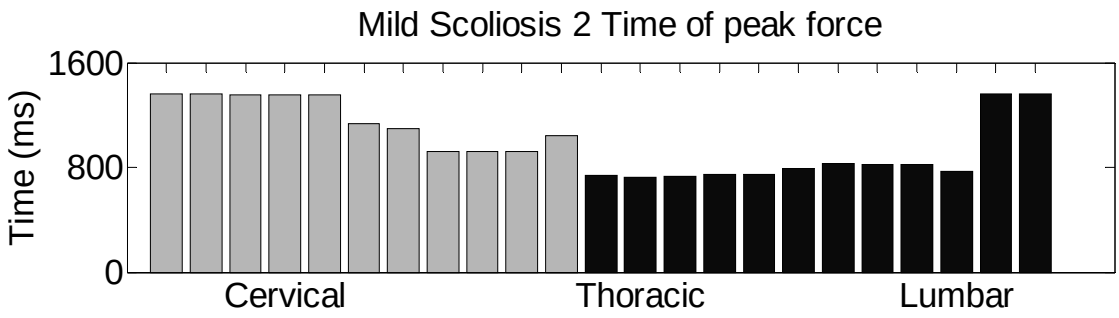

Figure 12: Time of Peak Forces in Mild Scoliosis 2. Black Denotes Positive Forces and Grey Denotes Negative Forces

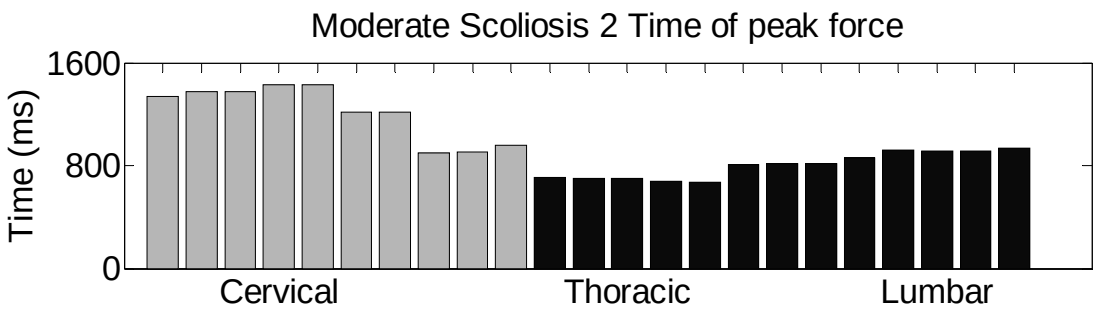

Figure 13: Time of Peak Forces in Moderate Scoliosis 2. Black Denotes Positive Forces and Grey Denotes Negative Forces

DISCUSSION

Previous evaluation of the forces on the thoracic and lumbar joints has found that at the apex of the deformed spine the forces increased by about 200% [Walsh, 2008b]. In the work detailed here analysis of the movement patterns of the baseline simulation compared to the scoliotic cases shows that although the conventional hyperextension of the neck occurs, there is a significant degree of asymmetry in the scoliotic models. Evaluation of the degree of curvature introduced in the spine due to scoliosis shows that as the joint rotation axis is shifted away from the global Y direction the effective stiffness of the spine in a sagittal (X-Z) plane is increased. The degree of deviation from the normal curvature of the seated spine was seen to have a direct effect on the resulting forces on the spine. The peak force in the thoracic region of the scoliotic spine was seen to correspond to greatest deviation from the global Y-axis. This restricts the movement of the dummy and prevents it from fully straightening and flattening against the seatback, as shown in figure 14 of the baseline and the mild case of scoliosis.

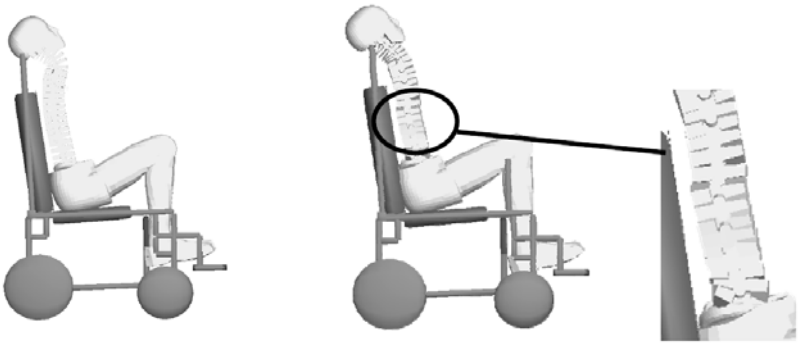

Figure 14(a) Non Scoliotic Dummy (b) Scoliosis Dummy (c) Close up View of Scoliotic Spine

It was found that although the Cobb angle of the moderate case of scoliosis is over twice that of the mild case this is not reflected in a two fold increase in the joint forces. The forces on the joints are due to the interaction of the level of deformity, the consequent lordosis and kyphosis levels introduced which affects the seating position of the dummy, and finally the shape of the surface of the back due to the orientation of the vertebrae. This complex interaction means that Cobb Angle which quantifies the level of scoliosis in the frontal plane may not be an accurate predictor of increased risk of injury as it measures the curvature in a single plane and does not account for the deformity about the Y axis. The increased acceleration passed to the dummy due to the point loading effect as found previously was found to have an influence on the timing of the peak forces in the spine. In the scoliotic models the peak forces in the neck occur at an earlier time due to the greater acceleration pulse being passed up through the spine which causes the neck to reach its full extension at an earlier stage. Investigating the time of peak forces revealed that the peak force in the lumbar region of the mild scoliosis case is not upon contact with the seatback but occurs during the rebound phase. This is due to the reduced gap between the torso and seatback which does not gain as much rearward momentum and therefore has a reduced contacting force.

These graphs (figures 11, 12, 13) also show the influence of sitting height on the direction of the forces exerted on the spine. The greater degree of scoliosis was found to reduce the sitting height of the dummy. The result of this was that less of the dummy's upper thoracic region was pulled backwards. Therefore the reduced height may provide some shielding against extension and possible damage of the joint and associated tendons and ligaments. This may help to reduce injury associated with rear impact. However the increased acceleration that the head and neck are exposed to due to the point loading effect may over-ride this and cause increased injury.

The predicted behaviour reflects the movement of the vertebrae in real life cases which may have an altered geometry. The interaction of the facet joints, which control the range of motion, is therefore changed [Veldhuizen, 1987]. This prevents the straightening of the spine as found in the baseline model which spreads the force throughout the spine, figure 14(a). In the scoliosis models, however the force is concentrated at the contact region in the spine as shown in figure 14(c). Scoliosis due to muscle weakness is found in 90% of all patients with Duchenne Muscular Dystrophy [Stern 1988; Huynh, 2007] and lower bone porosity is also commonly associated with these patients [Huynh, 2007] which may make them more vulnerable in a crash when subjected to such high loading.

LIMITATIONS

A limiting factor of this work is that the BioRID II is an approximated model of the human form and so some differences are found between the two which may be further highlighted by adapting the model. The use of revolute joints in the vertebral pairs is a simplification of the more complex real life case, which is further complicated by adding scoliosis. However the BioRID II dummy was developed to simulation the articulation of the spine in low speed rear

impacts and to investigate injury in this region [Davidsson, 1998]. The analysis tool developed here can provide guidelines for developments in crash safety for this area.

CONCLUSION

This work has required the rear impact sled testing of a crash test dummy in a wheelchair and a new procedure was created for this. The output data was then used to create and validate a computational model which provided a baseline against which comparisons can be made. This analysis tool allows the user to determine the effect that a postural deformity such as scoliosis has on the outcome. From the work presented here it can be seen that the baseline model shows that the acceleration pulse causes motion of the spine in the direction of the impact. The altered orientation of the scoliotic spines however does not allow the acceleration pulse to cause rotation of the spine about the same axis, which introduces an effective sagittal stiffness of the spinal column. The combined point loading and altered orientation of the vertebrae results in increased forces on the spine. This reflects real life cases which have a reduced range of motion due to the altered orientation of the facet joints in a scoliotic spine. However the Cobb angle which defines the degree of deformation of the spine was not found to be an accurate predictor of the level of risk in a rear impact as it focuses on the frontal plane only. This research can be used to inform future wheelchair occupant safety guidelines.

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