

Development of an In Vitro Model of Head-first Impact with a Hybrid III Head, Surrogate Spinal Cord, and Simulated Neck Muscles

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This paper has not been screened for accuracy nor refereed by any body of scientific peers and should not be referenced in the open literature.

ABSTRACT

Although the role of muscles was once thought to be minimal in axial compression, it is now thought that their stabilizing effect may influence the buckling behavior of the inherently unstable articulated column that is the cervical spine. This has been previously investigated experimentally using a very simplified model of neck musculature. The objective of this study was to create and establish the merit of a model of head-first impact with an advanced muscle force replication (AMFR) system using a Hybrid III anthropometric test device head, cadaveric cervical spine, and surrogate spinal cord.

An osteoligamentous cervical spine (occiput-T2, age 60 yrs) was set in dental stone at T1/2 and attached to a Hybrid III head at the occiput using custom adapter plates. The T1/2 potting cup was fixed to a six axis lower neck load cell and mounted to the carriage of a drop tower. The AMFR system modeled four bilateral muscles and three follower loads using fishing line tied to the vertebrae or to the Hybrid III head mounting plate. A radiopaque biofidelic surrogate spinal cord was inserted into the spinal canal. The specimen was dropped from a height of 60 cm onto a padded impact platform overtop a uni-axial load cell. The impact was captured at 1000 frames per second with two high speed video cameras and a high speed x-ray system. Injuries were diagnosed by a fellowship-trained spine surgeon (JS) using x-rays, CT scans, and through dissection.

Lordosis was removed from the specimen and it was aligned with zero eccentricity (anteroposterior distance between the occipital condyles and the T1 vertebral body) using the AMFR system. The peak impact platform and lower neck axial loads were 7930 N and 3830 N, respectively. Injuries included a burst fracture of C1 and a disc and bilateral facet capsule rupture at C5/6. The C1 fracture produced a concurrent peak spinal cord compression of 20%.

This pilot study showed that compression injuries in head-first impacts can be reproduced with a cadaver spine, Hybrid III head, and simulated muscle forces. This model advances in vitro testing as it permits measurement of spinal cord compression and simulation of neck muscle forces with tied attachments on the vertebrae allowing this to be achieved without damaging the structural integrity of the spinal column.

INTRODUCTION

Spinal cord injury (SCI) can have devastating quality of life consequences for victims, their families and society, and the financial costs are substantial (DeVivo et al., 1992; Priebe et al., 2007). SCIs are a significant health concern with an annual incidence of 13,000 in North America (Dryden et al., 2003; NSCISC, 2006). Axial loading of the cervical spine leading to SCI may result from any head-first impact, such as those in diving, hockey, snow skiing, rugby, cheerleading and baseball (Torg et al., 2002), as well as those resulting from motor vehicle accidents (Clayton et al., 2004; Reed et al., 2006) and rollover motor vehicle accidents (Moffatt et al., 2003). In some of these cases, the head impacts a surface, such as the ground, and the inertia of the incoming torso results in compression of the cervical spine. Axial loading has been documented as the primary cause of catastrophic cervical spine injuries in football (Torg et al., 2002) and has been implicated as the prime mechanism leading to SCI in motor vehicle accidents (Yoganandan et al., 1989).

To control head and neck posture for head-first impact simulations, cadaveric cervical spine specimens have previously been positioned with sutures, string, cables, springs, and/or fishing line applied to the ear, nose and/or forehead (Yoganandan et al., 1986; Pintar et al., 1990; Pintar et al., 1995; Nightingale et al., 1996b; Nightingale et al., 1997; Saari et al., 2006). These applied loads act along non-physiologic lines of action. Alignment of the head is critical in experimentally creating cervical spine injuries as a posture with “zero eccentricity” (placing the occipital condyles superior to the T1 vertebral body without any anterior or posterior offset) is considered to be a requirement for the creation of cervical spine burst fractures (Maiman et al., 2002). Physiologically, this is achieved through contraction of postural and phasic neck muscles which act on the temporal, occipital, and hyoid bones and the cervical vertebrae, (Mayoux-Benhamou et al., 1997; Brelin-Fornari, 1998) resulting in segmental compression forces in the cervical spine (Hattori et al., 1981; Moroney et al., 1988; Pospiech et al., 1999). Although the role of muscles was once thought to be minimal in axial compression (Nightingale et al., 1996b), their stabilizing effect may influence the flexural rigidity of the osteoligamentous spine, which is an inherently unstable segmented column (Panjabi et al., 1998), and alter its dynamic deformations (Nightingale et al., 2000; McElhaney et al., 2002). This has only been previously investigated experimentally using a very simplified model of neck musculature (Saari et al., 2006).

Experimental models of the cervical spine investigating the role of neck musculature have used cables and springs to simulate muscle forces on osteoligamentous specimens; however, the techniques used to attach muscle forces altered the structural integrity of the spinal column, as rods, screws, and/or eyelets were inserted into the vertebrae (Nolan and Sherk, 1988; Bernhardt et al., 1999; Panjabi et al., 2001; Kettler et al., 2002). Since axial compression typically results in vertebral fractures, any stress risers in the bone, due to screws or rods, would be undesirable as they could cause non-physiologic fracture initiation. To our knowledge, the role of deep and surface muscle activation on cervical spine kinetics and kinematics as a result of head-first impact has not been examined in an experimental model. Such a model would be useful for incorporating the activation of neck muscles from in vivo studies (Yamaguchi et al., 2005) and to validate computational models (Newberry et al., 2005).

The objective of this study was to create and evaluate the merit of an in vitro model of whole cervical spine compression due to head-first impact using an advanced muscle force replication (AMFR) system. Other advanced aspects of the model include the use of a Hybrid III anthropometric test device (ATD) head and a radiopaque surrogate spinal cord in the spinal canal but these will not be discussed in detail here.

METHODS

A fresh-frozen human cervical spine (occiput-T2) was harvested, screened radiographically for bony abnormalities, loss of disc height or other signs or abnormal degeneration, and frozen until use. The donor was male and 60 yrs of age. The specimen was tested within 72 hours after defrosting and hydration was maintained throughout preparation and testing by spraying with saline solution and wrapping the specimen in saline soaked gauze.

The experimental apparatus was designed to simulate a head-first impact. The specimen was mounted in an inverted posture on the carriage of a four-rail drop tower (Saari 2006) and dropped from a height of 60 cm to achieve an impact velocity of approximately 3 m/s, which is the estimated threshold for

injury to the cervical spine (Nightingale et al. 1996b). The impact platform was padded with two layers of a yoga mat and a 0.8 mm sheet of neoprene rubber to create a high friction constraint surface. A uni-axial load cell was placed underneath the impact platform (Omega LC 402-5K). The mass of the carriage assembly (17 kg) approximates the effective mass of the following torso in a diving injury (Nightingale et al. 1996b).

Surrogate Spinal Cord

T2 and the inferior half of T1 were potted in a casting cup with dental stone such that the anterior margin of the T1 vertebral body was vertically oriented in the casting cup. An elastomeric surrogate of the spinal cord was used (Kroeker et al. 2009), lateral diameter 11.5 mm, anterior-posterior diameter 6 mm. The surrogate cord was manufactured from QMSkin 30 (Quantum Silicones, Richmond, VA) and dosed with barium sulfate (BaSO_4) to make it radiopaque.

Hybrid III Head Attachment

A portion of the occiput was attached to a 50th percentile Hybrid III ATD head using two custom plates sandwiching a Delrin sealing cylinder (Figure 1). The center of rotation of the specimen's C0/1 joint (Chancey et al. 2007) was placed in the same medial-lateral and anterior-posterior position as the center of the Hybrid III head/neck joint (determined using machine drawings); however, it was placed slightly inferior to that of the Hybrid III head/neck joint, since accurate inferior-superior placement of this joint would have required redesign of the Hybrid III head. The occiput was mounted to the Hybrid III head using screws, wire, and epoxy putty to distribute compressive load. The Hybrid III head was wrapped with cellophane to protect the instrumentation from moisture and this was covered with two layers of nylon stocking. The Hybrid III head was equipped with three uni-axial accelerometers (Endevco 7264C) at its center of mass. Head Injury Criteria with a time duration of 15 ms (HIC15) was calculated as follows:

$$HIC = (t_2 - t_1) \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} (a(t)) dt \right]^{2.5}$$

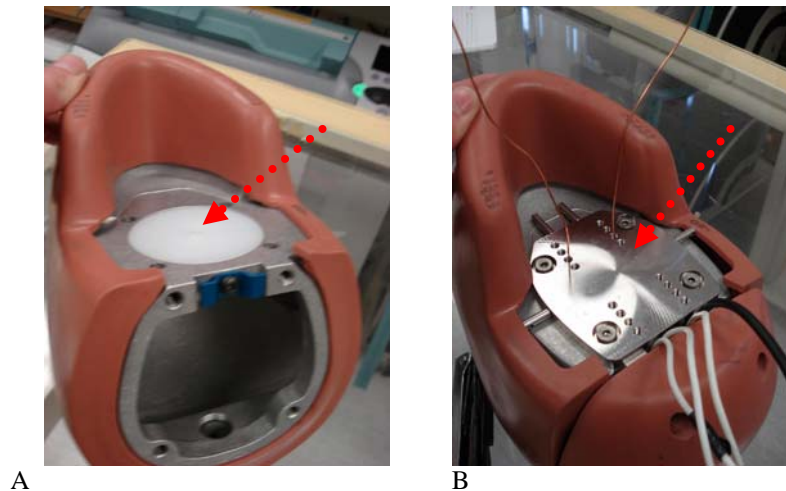


Figure 1: (A) Sealing cylinder and (B) external adapter plate in place on the Hybrid III ATD head (shown with dotted arrows).

Advanced Muscle Force Replication (AMFR) System

The AMFR system was developed to simulate the forces of four bilateral muscles or groups of muscles: semispinalis capitis, sternocleidomastoid, hyoids (omohyoid, sternohyoid, and sternothyroid), and

trapezius. These muscles were selected due to their dominant role in generating moments in flexion (hyoid, sternocleidomastoid), extension (semispinalis capitis, trapezius), axial rotation (trapezius), and lateral bending (sternocleidomastoid) (Vasavada et al., 1998; Oi et al., 2004). The stabilizing mechanical effect of deep anterior neck musculature (longus colli, longus capitis, and longissimus cervicis) was simulated with an anterior follower load and two lateral follower loads (Patwardhan et al., 2000; Panjabi et al., 2001), which were guided at each vertebral level (C1-C7) to provide compression along the length of the spine. The sternocleidomastoid and hyoid muscle cables and the follower load cables were fixed superiorly on clevis pins on the external adapter plate superior to the occiput, while the trapezius muscle was fixed superiorly onto each vertebra C0-C7. These cables were also connected at the lower (T2) level. The semispinalis capitis muscle was attached bilaterally to four vertebrae (C4-C7) and to the occiput.

To achieve fixation of the muscle force cables on each vertebra while avoiding stress concentrations resulting from screw fixation, braided fishing line (Berkely Gorilla Tough, 50 lb test, Spirit Lake IA) was passed through the right and left transverse foramina on each vertebra (C1-C7) and tied across the anterior vertebral bodies (Figure 2). Hinged clamps (callotte clamshell, BeaZu Beadworks, Vancouver BC, Canada) and split rings (round split ring, 20 lb rating, Mustad, Auburn NY) were fixed to this string bilaterally and on the centerline at each level for guiding the bilateral follower load and anterior follower load cables. The lateral follower cable fixation points approximated the centers of rotation of each segment (Dvorak et al., 1991). Bilateral posterior attachment points were provided by passing fishing line around each lamina and tying these tightly against the lamina (Figure 2). Hinged clamps and split rings were fixed to this string to provide attachment points for the trapezius and semispinalis capitis muscle force cables.

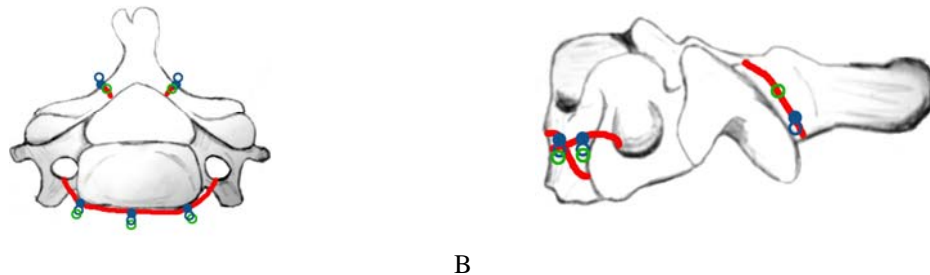


Figure 2: Superior (A) and anterior-lateral (B) views of a typical cervical vertebra showing the path of the braided fishing line (red) tied tightly against the vertebra and used along with hinged clamps (blue / open circles attached to smaller filled circles) and split rings (green / open circles only) to provide non-destructive attachment points for muscle force cables in the AMFR system.

An aluminum plate (thickness - 1.3 cm, width - 17.8 cm, length - 27.9 cm) was attached to the T1/2 casting cup and to a six-axis lower neck load cell (Denton 4366J), which was mounted to the carriage of the drop tower. This plate located the inferior end of all muscle forces (except semispinalis capitis) and follower load cables as the cables passed through holes in the plate. The location of these holes were based on published quantitative anatomical dissections (Chancey et al., 2003; Oi et al., 2004). Each cable that was passed through this plate was tied to a screw, the end of which was passed through a compression spring (11.43 cm long, 3.15 N/mm, McMaster-Carr Supply Company, Atlanta GA) and a washer, such that turning a nut at the end of this screw applied compression to the spring and tension to the cable. Trapezius muscle force cables from adjacent segments were attached to a force limiter (Figure 3A), so that the forces in both segments were equal and were controlled by one spring; four springs on each side of the specimen controlled the force in the trapezius cables for the C0/1, C2/3, C4/5, and C6/7 segments. Eight extension springs (2.37 cm long, 1.19 N/mm, McMaster-Carr Supply Company, Atlanta GA) were attached bilaterally between each of C4-C7 and the occiput to simulate the semispinalis capitis muscle. These were attached at the vertebrae with split rings and at the occiput with four screws. Extension in the springs was controlled through the use of circlips and nuts on the screws. The spring stiffnesses were selected to represent passive stiffness of neck muscles as they stretch from their length in neutral posture (Vasavada et al., 2007). Bilateral flexion limiters were tied between a screw in the occiput and a hinged clamp on the C1 lamina to prevent hyperflexion of the

specimen, which would anatomically be prevented by contact between the chin and chest (Panjabi et al., 2001). In total, the AMFR model included 33 cables and 23 springs (15 compression and eight extension springs).

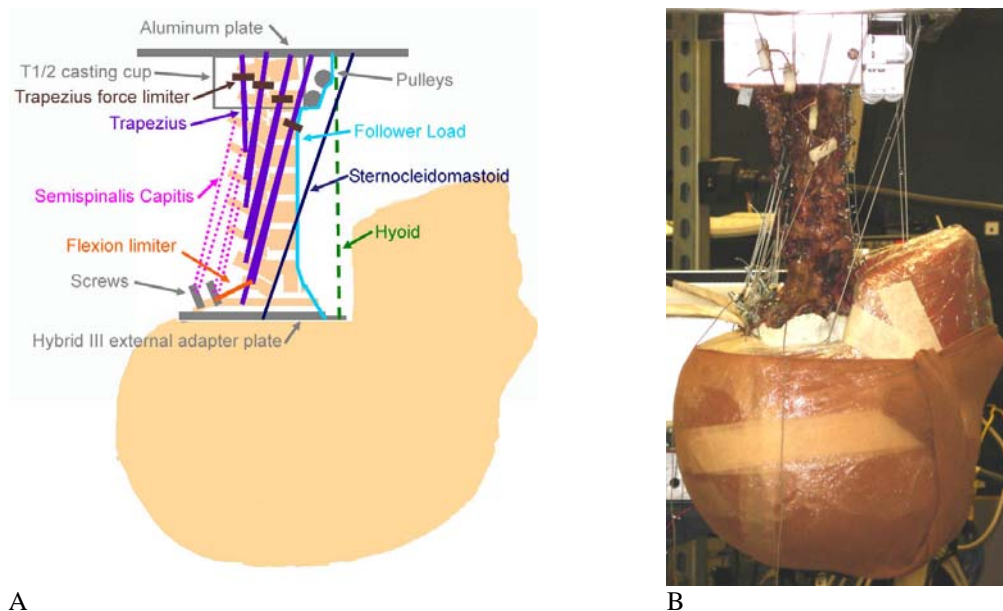


Figure 3: (A) Schematic of the AMFR system (lateral view) with muscle cables and key components identified. (B): Photograph of the AMFR system (lateral view) with the Hybrid III head.

The muscle forces were adjusted (by turning nuts on the compression and extension spring assemblies) as necessary to align the impact point on the head, the occipital condyles, and the center of the T1 vertebral body to create a lordosis-removed and stable posture which is thought to be required for incurring compressive type fractures (Maiman et al., 2002). Pre-drop muscle forces were calculated from the spring stiffnesses and the spring length (measured with a vernier caliper) following specimen alignment. Segmental compression forces were calculated as the sum of the muscle forces acting through each segment of the specimen, without taking into consideration the angles of the lines of action of the muscles or the posture of each segment.

High Speed Video & X-Ray

Each impact was captured with two high speed video cameras (Phantom V9, Vision Research, Wayne NJ) at a resolution of 1632 x 1200 pixels and at 1000 frames per second. Load cell and accelerometer data channels were image-synchronized, sampled at 78 kHz and filtered with pre A/D anti aliasing filters to satisfy SAE J211b. All data from the instrumentation were processed with a custom program (Matlab 7.0, Mathworks).

A high speed x-ray system (generator/source: Phillips MCN 160/167, 22.9 cm image intensifier: Precise Optics, camera: Kodak SR 1000) captured a lateral view of the spine at 1000 frames per second at a resolution of 240 x 256 pixels and a shutter speed of 1/2000 seconds. Contours of the surrogate spinal cord were manually segmented (Analyze 8.1, Mayo Clinic, USA) for each x-ray image and these two-dimensional data points were exported for further analysis in a custom program (Matlab 7.0, Mathworks). Cord diameters were computed using an iterative closest point technique at each axial location and for each time point after head impact. Peak spinal cord compressions were computed as the difference between the cord diameter at a time after impact and that at the same axial location just before impact. Cord compressions were expressed as a percentage of the cord diameter just before impact.

Injury diagnoses

The specimen was imaged with x-ray prior to the test and with High Resolution Peripheral CT (XTreme CT, Scanco Medical AG, Bassersdorf, Switzerland) after the test. Injuries were diagnosed by a spine surgeon (JS) using the pre- and post-test x-ray, post-test CT, and post-test dissection.

RESULTS

By increasing the spring compression/extension in the muscle force and follower load cables, the cervical lordosis was removed from the specimen and it was aligned with zero eccentricity. With the AMFR system, the specimen felt qualitatively much more stiff and stable than the osteoligamentous spine without the AMFR system. Moving the AMFR head felt qualitatively like trying to manipulate the head of a person who consciously resists head motion through tensioning their neck muscles. The loads in each of the cables are summarized in Table 1. The total force in all cables was 176.8 N. Segmental compression force (sum of the muscle forces acting through each segment, without taking into consideration the angles of the lines of action of the muscles or segmental posture) at C0/1 was 56 N and this increased inferiorly to a maximum of 170 N at C7/T1 (Figure 4). At C4/5 the segmental compression force was 132 N.

Table 1. Force (N) in each of the AMFR system cables.

Muscle Cable		Left	Right	Anterior
Follower Load		0.3	2.5	7.3
Sternocleidomastoid		4.4	3.8	n/a
Hyoid		12.9	12.6	
Semispinalis Capitis	C4	0.2	0	
	C5	0	0.4	
	C6	0.2	1.2	
	C7	0.2	0.6	
Trapezius	C0/1	7.9	9.1	
	C2/3	17.0	24.6	
	C4/5	28.4	24.6	
	C6/7	9.5	8.5	

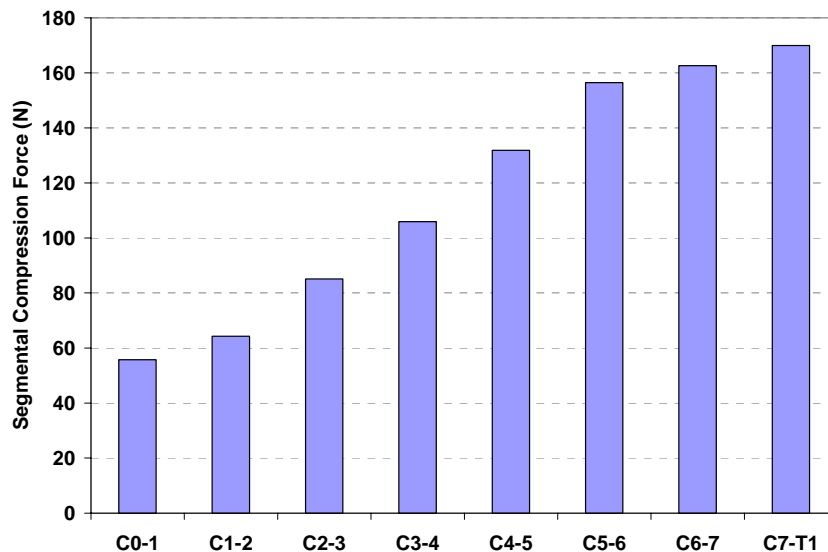


Figure 4: Segmental compression forces in the AMFR model.

During impact, the muscle force cables were observed to become slack and the flexion limiter (between the occiput and C1) was not taut at any point during the test. The injuries produced included a three-part burst fracture of C1 (posterior and anterior ring fracture with transverse ligament rupture), a left occipital condyle fracture, and a disc rupture with associated bilateral incomplete ligamentum flavum and facet capsule injuries at C5/6. The C1 burst fracture was observed on the high-speed video and x-ray images (Figure 5) which corresponded approximately to a local peak in lower neck axial load (Figure 6). The C1 fracture produced a peak spinal cord compression of 20% adjacent to this level, which occurred 2 ms after the peak head acceleration.

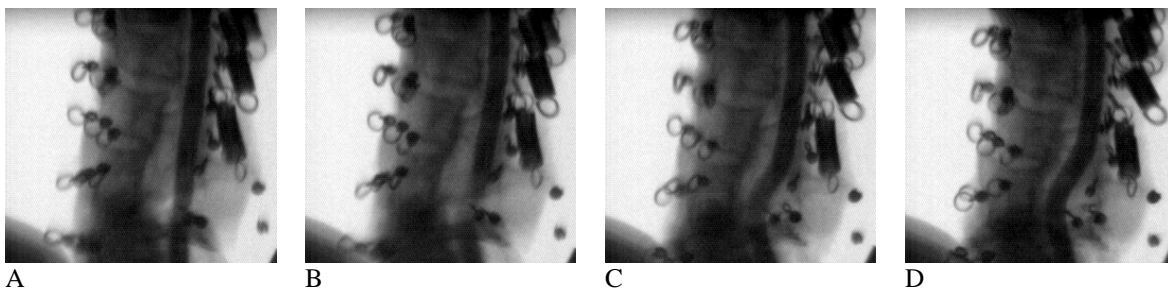


Figure 5: High speed x-ray images in sequence (each is 1 ms after the previous image) of the specimen at the time of the C1 fracture (bottom of the image). The chin of the Hybrid III head can be seen to the left of the specimen in each image. Split rings, clamshell rings, and extension springs of the AMFR system can also be seen.

The peak impact platform load was 7930 N (not shown). The peak lower neck axial load and flexion moment were 3830 N and 112 Nm, respectively (Figures 6 & 7). The peak HIC15 was 85. The local peak in lower neck axial load that corresponded approximately to the C1 fracture was 1993 N and the flexion moment at this time was 75 Nm.

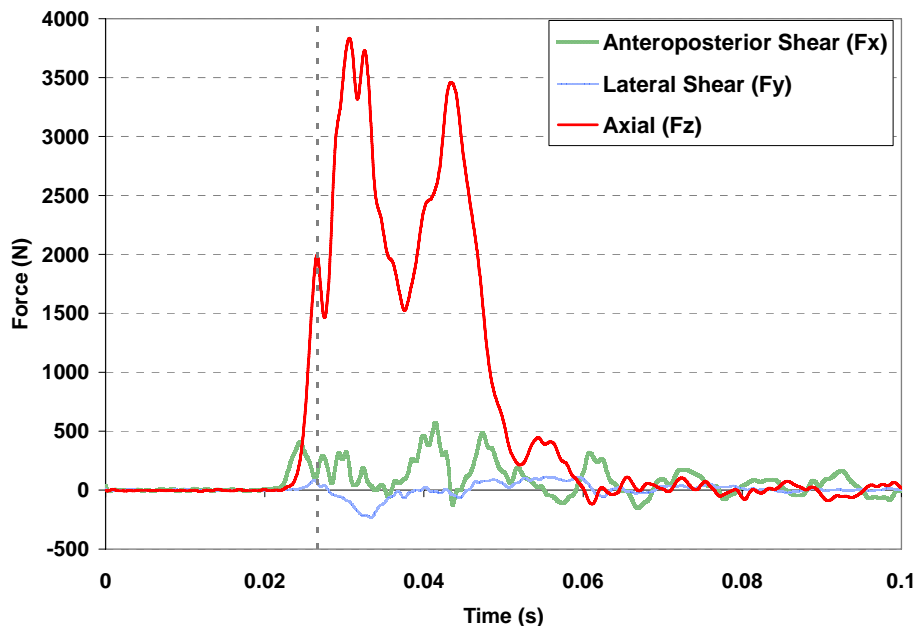


Figure 6: Lower neck forces. The time of the local peak in axial force corresponding approximately to the C1 fracture is shown with a dotted vertical line.

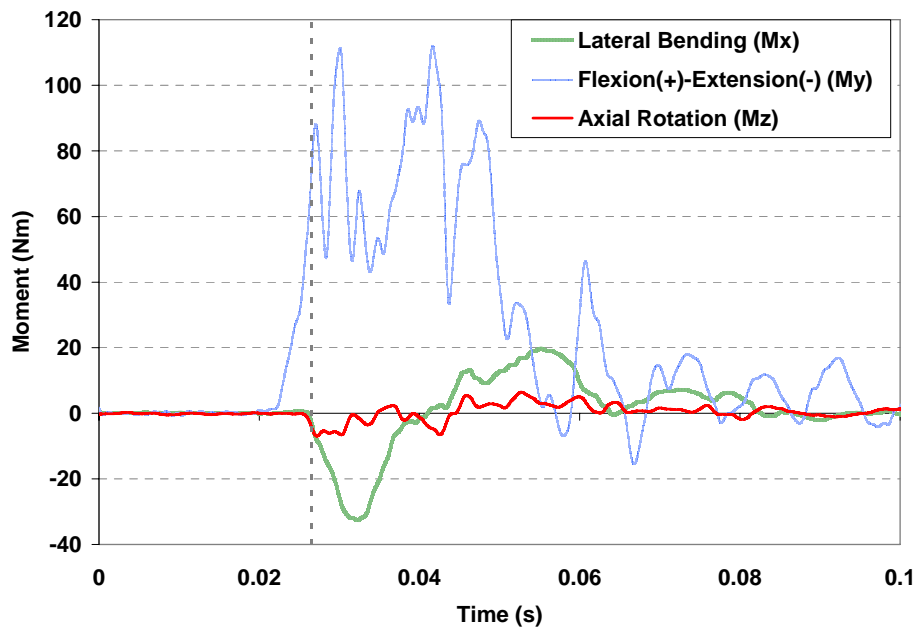


Figure 7: Lower neck moments. The time of the local peak in axial force corresponding approximately to the C1 fracture is shown with the dotted vertical line.

DISCUSSION

Cervical spine kinetics are central to understanding injury mechanisms resulting from head-first impact and may play a key role in developing injury prevention devices and diagnosis, first responder, and treatment strategies for these injuries. Neck muscle activation stiffens and stabilizes the head and neck complex (Stemper et al., 2006) and it has been shown that it can preferentially protect the upper cervical spine in flexion (Brolin et al., 2005); however, the effect of deep and surface neck muscle activation has not

been examined experimentally for head-first impacts. An experimental model with simulation of neck muscle activation without altering the structural integrity of the spinal column is of interest as this would allow for incorporation of neck muscle activation from *in vivo* studies (Yamaguchi et al., 2005) and could be used to validate computational models (Newberry et al., 2005).

The use of the AMFR system in the present experiments was thought to have three main effects. Firstly, it allowed control of posture at the vertebral level, rather than indirect posture control through positioning of the head (Yoganandan et al., 1986; Pintar et al., 1990; Pintar et al., 1995; Nightingale et al., 1996b; Nightingale et al., 1997; Saari et al., 2006). Secondly, it allowed control of pre-impact segmental loading. Although the muscle cables were observed to become slack during impact, they acted to stabilize the spine during the initial increase in neck loading. The axial loads and flexion-extension moments at the lower neck, impact force, and HIC15 observed during impact were comparable to those previously reported using cadaver models without simulation of neck muscles (Pintar et al., 1989; Pintar et al., 1990; Nightingale, 1993; Nightingale et al., 1997). The stabilizing effect of neck musculature may become more apparent in comparing the vertebral kinematics between these studies; this analysis is not presented here. Thirdly, the AMFR system cables may also apply forces to the head and vertebrae during impact, particularly during impacts with an eccentricity as the head may flex or extend. However, since spine injuries in head-first impacts occur before large head motions are produced (Nightingale et al., 1996a), this is expected to have the least significant effect on the injury-producing portion of the impact.

In the present experimental model, head and cervical spine specimens were held in an inverted posture and the application of simulated muscle forces generated a segmental compression force of 132 at C4-5. *In vivo* compression forces in the cervical spine have been examined using electromyography-driven biomechanical models and using needle-type pressure transducers inserted into the disc (Hattori et al., 1981; Moroney et al., 1988). Using a biomechanical model, mean calculated compression reaction forces at the C4 level of 122 and 558 N were determined for relaxed and maximal flexion postures, respectively (Moroney et al., 1988). Measurements of nucleus pressure have indicated that segmental compressive forces in the cervical spine (calculated using an average area of the nucleus (Pooni et al., 1986)) are approximately 53 N when laying supine, 75 N in the neutral posture, and 100 N in flexion (Hattori et al., 1981). The segmental compression forces determined in the present study are within the range of those previously reported; however, no previous study has examined segmental compression forces in the cervical spine in an inverted posture, or in a startled or frightened individual as may be the case prior to a head impact, with or without muscle activation. Furthermore, the segmental compression forces calculated in the present study overestimate the actual compression forces since the angles of the muscle lines of action and posture of the segments were not taken into account, thus assuming that the muscles applied only compression to the spine. In each segment, compression, anteroposterior shear, and medial-lateral shear forces were induced.

The use of the Hybrid III head was advantageous in this model as it allowed direct measurement of kinematics at the center of mass of the head, it was easier to access than a cadaver head, and it had improved biofidelity compared to previous surrogate head models (Saari et al., 2006). The surrogate cord allowed direct observation of spinal cord deformation during the impact. The AMFR system allows simulation of neck muscle forces without compromising the structural integrity of the spine, which is essential for investigating the biomechanics of the spine when fractures are expected to occur. With use of the AMFR system, there was qualitatively more resistance to movement when attempting to manually move the Hybrid III head on the cadaver spine; it felt similar to trying to move the head of a person who resists head motion through tensioning their neck muscles. The AMFR system allowed for improved posture control with positioning loads that are physiologic in magnitude and act along anatomic lines of action. A limitation of the AMFR system is that it required a considerable amount of time to attach the muscle cables in a nondestructive manner. In addition, the muscles were modeled as linear spring elements with point attachments while muscles are known to be viscoelastic and to have distributed attachments at tendon insertions. However, the line of action approximations were based on published studies of muscle morphometric measurements (Vasavada et al., 1998; Chancey et al., 2003; Oi et al., 2004; Vasavada et al., 2007). Furthermore, prior to impact, the AMFR system modeled muscle forces representative of pre-impact muscle activation, which is consistent with bracing observed in subjects aware of an impending impact (Siegmond et al., 2003; Yamaguchi et al., 2005). During impact, the AMFR system modeled the passive stiffness of musculature; simulated muscles were not capable of activation. This is appropriate for axial

impacts as the primary loading phase has a very short duration (4 ms) and negligible active muscle forces could be produced during this time (Valkeinen et al., 2002).

This pilot study showed that compression injuries in head-first impacts can be reproduced with a cadaver spine, Hybrid III head, and simulated muscle forces through the use of the AMFR system. This model advances *in vitro* testing as it permits measurement of spinal cord compressions and simulation of neck muscle forces with attachments tied to the vertebrae without damaging the structural integrity of the spinal column.

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DISCUSSION

PAPER: **Development of an In Vitro Model of Head-first Impact with a Hybrid III Head, Surrogate Spinal Cord, and Simulated Neck Muscles**

PRESENTER: *Carolyn Greaves, University of British Columbia*

QUESTION: *Guy Nushultz, Daimler Chrysler*

I have two questions: One, how did you validate your compression algorithm? And two, how did you match the impedances at the endpoints where you're attaching to the dummy head and where you're attaching the drop-tower structure? You're looking at the compression of the spine. You said you use some algorithm, but how did you know that that's actually telling you what compression is? And when you say compression, what do you mean? Are you looking at it as a maximum strain across that? Are you looking at, "I have two points and I'm going to talk about how those two points compress?"

ANSWER: Yes. The compressions that are presented, the segmental compressions—It's an overall compression force at the segment. Actually, we haven't done the kinematic analysis of these specimens yet so it's not taking into account the different orientations of the specimens. And so, it's more like a general, overall force in the segment that may not be completely a compression. There could be some sheer components that cancel out on either side, but our goal in applying the compression forces was just straight to the spine. We weren't trying to achieve something that was physiologic. But in the end, we compared to physiologic compression forces just to see if the model ended up with some physiologic compression forces.

Q: So just so I'm clear: You've got two points and you're talking about a percentage of how those two points move? Is that right? I mean it's not clear what you mean by spinal cord compression.

A: Squeezing.

Q: This way?

A: Yes. Squeezing this way.

Q: It's a maximum--

A: We gave segmental compressed forces in the axial direction. So we had a bunch of strings that were traveling in different directions and attached to different vertebrae. We measured how much the springs at the end of the model were compressed or extended in terms of the extension springs and then got— We calculated compression, but not exactly compression because it's sort of overall compression without taking into account the orientation of the vertebrae.

Q: Okay. If you matched the impedance, how did you do it? If you didn't, why do you think it might not be important or important?

A: I'm not sure I know what you mean.

Q: Well, I'm going to attach your spine segment to the skull, right?

A: Yes.

Q: To the Hybrid III dummy head.

A: Yes.

Q: Well, a skull will not respond exactly the way the Hybrid III head did. So there's an impedance mismatch between attaching to the head and attaching to the spine.

A: Right.

Q: That could increase or decrease fractures in either the occipital or in terms inside of the spine. So the question is: If you don't understand the question, you probably didn't do it. But then, do you think it's important?

A: Yes. That's a good question. It could be important in our model, like taking into account that we're attaching to the rigid plate at the head rather than attaching to a skull that deforms, if that's what you mean.

Q: As well as where it attaches to the spine because in a person, you have a lot of give here.

A: Yes.

Q: Where it can move laterally in your simulation.

A: Yes. We didn't take that into account. It could be something that's important, but there are a lot of approximations with attaching muscle forces at one place in the model as well. So I don't know if that's the biggest approximation that will have the most affect on our results, but that's a good question and we didn't look at that.

Q: Okay. Thank you. Now, I'll turn you over to John.

Q: *John Melvin, Tandelta Inc.*

This is amazing work. I really like the detail you're doing there.

A: Thanks.

Q: Why didn't you use the Hybrid III neck load cell in this set-up because it would allow you to make some measurements very near the spine that you're not measuring right now?

A: We used the lower-neck load cell from the Hybrid III. You mean the upper one? As I understand, the upper-neck load cell will actually replace a few of the vertebrae of our model because we wanted to include the upper cervical spine as well. As it was, our occipital condyles were probably a bit too low compared to the dummy; and if we added a load cell, they would be even lower. So our spine would be, I think, too far from the center of gravity of the head. So that might have changed some of the interactions between the head and the spine that would have shifted everything in the model.

Q: I understand. Given all the work you're doing to prepare this spine for testing, it would seem to me that you could modify the Hybrid III head and stick that load cell up a little farther in it once. And then, you've got it up there.

A: Yes.

Q: Its position doesn't matter. It can be moved up into the head. It's just a very good measuring system of all the forces that are being transmitted to the spine.

A: Yes. I agree. That would be great, and this would be a great way to measure upper-neck loads in a cadaveric specimen as well. That's a good point. Thanks.

Q: *Jingwen Hu, UMTRI*

Very nice study.

A: Thank you.

Q: I think it's a very complicated task to set up. I am just curious. Comparing with previous cadaver tests, adding the muscle force, in your opinion, is increasing the neck injury risk or decreasing it? I thought it's a very interesting question there.

A: Yes, that is an interesting question. I guess there are a number of different factors that will affect that, like the muscles, I think, will stabilize the spine so it might change some of the buckling characteristics that you see if you don't have the muscle forces. It might keep the forces more axially than letting the vertebrae move around, so that could be one change.

Q: So in your opinion, it's hard to say if it actually increases or decreases the injury risk.

A: Yes, I think it's hard to say. I think it will change some of the characteristics of the impact in the way that the vertebrae move. I'm not sure if it will necessarily increase or decrease the fracture risk. It might change the kind of fracture that you get.

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Q: With more tests, you probably can compare with the series.

A: Yes.

Q: I'm looking forward to your results.