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Dynamic Bending Mechanics of the Pediatric Cervical Spine

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ABSTRACT

While pediatric cervical spine injuries are not extremely common, when they occur, the life-long debilitating consequences for the child, their family, and society are catastrophic. In order to mitigate these injuries to children, we must understand the mechanics of the child neck and the inputs which create these deleterious injuries. Automotive crashes represent a significant percentage of these injuries and often involve high rates of speed and inertial loading of the head and neck complex. Thus, in an effort to prevent child neck injuries, we set out to investigate the bending mechanics of the maturing cervical spine as a result of dynamic loading. Sixteen baboon cadaver specimens were utilized spanning the pediatric populace from 2 to 23-human equivalent years. The C5-C6 functional spinal units of these specimen were dissected free and rigidly fixed to a custom dynamic bending apparatus. This device applied dynamic angular displacements(mean of 27.6-rad/sec) to the superior vertebrae while minimizing the shear forces traveling through the specimen. The specimens were divided into a flexion and extension group for testing to failure and the loads, displacements, and accelerations of the event were recorded. The flexion and extension stiffness increased with maturation as did the failure moments for both flexion and extension. Further, these data were utilized to generate scaling from the child to adult for spinal mechanics. The raw data and these scaling values provide data for computational models and anthropomorphic test devices which may lead to a meaningful neck injury prevention scheme for children.

INTRODUCTION

Although cervical spine injuries in children account for less than 10% of all cervical injuries, the fatalities resulting from cervical trauma to children are four times the rate of adult fatalities (Myers and Winkelstein, 1995). A large percentage of these injuries are a result of motor vehicle accidents which induce dynamic bending of the cervical spine. Unfortunately, the mechanical response of the pediatric cervical spine to dynamic bending inputs is not well understood, making prevention of these injuries unfeasible.

Previous research has been performed to understand the response of adult cervical spine tissues to pure bending moments both quasi-statically and dynamically. A number of studies have investigated the quasi-static bending range of motion of the adult cervical spine (Dvorak et al., 1992; Lind et al., 1989; White and Panjabi, 1990; Panjabi et al., 2001); however, only a few studies have investigated the stiffness and failure characteristics (Nightingale et al., 2002; Voo et al, 1998). Nightingale et al. (2002) applied pure bending moments quasi-statically to both the upper and lower cervical spine segments. They found the failure moment to be significantly greater for extension compared with flexion of the upper cervical spine. Further, when comparing the upper and lower cervical spine, they measured significantly larger extension failure moments in the upper cervical spine. This result which is not consistent with epidemiological data, they explained, was likely due to active musculature load sharing. In a study by Voo et al. (1998), dynamic bending moments (18 to 35-rad/sec) were applied to cervical spine tissues and their stiffness values were compared with quasi-static tests performed on the same tissues. They discovered that the dynamic stiffness was statistically greater than the quasi-static stiffness. Therefore, data exists for the adult populace in quasi-static and dynamic loading rates for range of motion and failure experiments.

An examination of the pediatric literature reveals but one study examining the bending mechanics of the cervical spine (Pintar et al., 2000). Pintar et al. (2000) utilized a goat model to examine the effects of maturation on the stiffness of isolated functional spinal units in flexion and extension. These quasi-static experiments provided the first data to estimate the mechanics of the immature spine in bending. In an effort to supplement this data set for child injury prevention, this research project was initiated to investigate the dynamic stiffness and failure mechanics of maturing tissues.

Therefore, the objective of this study is to examine the pediatric cervical spine dynamic bending mechanics. These data will enable accurate child neck injury prediction in computational models and anthropomorphic test devices (ATD). To our knowledge, this is the first study examining the dynamic bending mechanics in the pediatric population.

METHODS

This research effort aimed to design and build an accurate dynamic bending apparatus and then utilize it in assessing the mechanics of maturing tissues. We developed an apparatus based upon the work of Crawford et al, (1995) and augmented the device to minimize shear forces and perform at dynamic rates. This device converts the axial motion of a hydraulic Material Testing System (MTS) to a pure bending moment. Also, the design allows translation on the horizontal axis in line with the coupled moment, thus minimizing shearing forces (Figure 1).

Specimen Preparation

Due to the limited availability of human pediatric tissues, we employed a cadaveric baboon model for this investigation. Sixteen male baboon specimens were obtained from the University of Washington Regional Primate Research Center. Each specimen was radiographically assessed for its skeletal maturity to ascribe a human equivalent age to the specimen (Ching, 2000). The specimens obtained for this experiment ranged in age from 2 to 23-human equivalent years. Each baboon specimen was prepared by removing all musculature and maintaining the osteoligamentous structures for biomechanical testing. Following this gross dissection, each specimen had the C5-C6 functional spinal unit dissected free for the bending experiments. These functional spinal units were fixed using wire through the vertebrae and embedded in polymethylmethacrylate on the superior and inferior vertebrae to ensure rigid purchase between the specimen and testing apparatus.

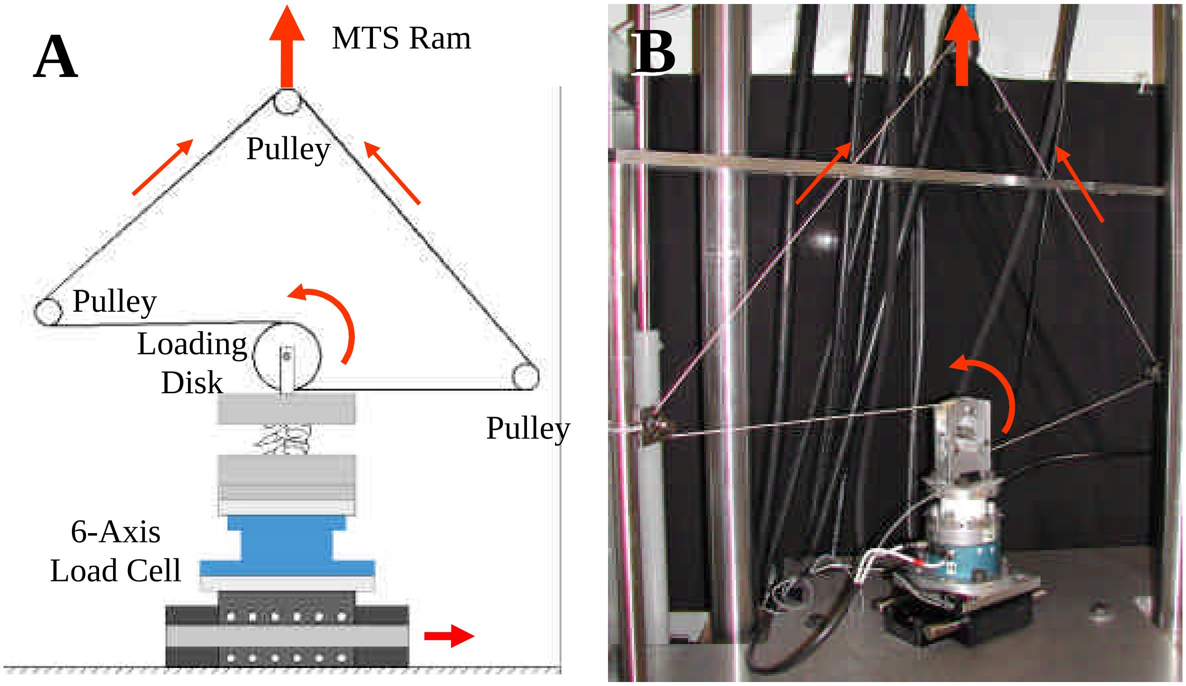

Figure 1: Apparatus designed to apply dynamic pure bending moments to the cervical spine by translating the MTS vertical displacement into a pure torque about the specimen. **A**. is a schematic demonstrating the operation and **B**. is a photograph of the actual apparatus shown with a 'dummy' specimen (white) between the load cell and the upper loading disk.

Experimental Protocol

All of the specimens were placed in age order and then every other specimen was assigned to a group. This enabled the broadest distribution of ages for both the flexion and extension groups. Once assigned to a loading group, each specimen was placed into the dynamic bending apparatus with a six axis load cell mounted inferiorly. High speed video (Fastcam, Photron Inc., San Diego, CA) images were obtained at 2,000-frames/sec for displacement measurement purposes. A linear accelerometer was placed on the carriage to measure the inferior accelerations and a rotational accelerometer was fixed to the superior loading disk to obtain the angular acceleration of the superior end of the specimen.

A review of the literature provided a range for the dynamic bending moment loading rate. A study by Siegmund et al. (2000) reported a head-to-T1 extension bending rate of 6.7-rad/sec for a 3Gx pulse in human subjects experiments. Another study reporting full cervical spine angular velocity as a function of crash impulse, determined a 21-rad/sec flexion rate as a result of a -15Gx impulse (Margulies, 1998). While these studies evaluated total cervical angular velocity, one study measured segmental angular velocities on human subjects. Ono et al. (1997) used cineradiography techniques to reveal a 10 to 15-rad/sec segmental angular velocity for C5-C6 extension when exposed to a 5Gx input pulse.

Based upon these studies, our experimental protocol targeted the application of dynamic rotational displacements at 25-rad/sec in both flexion and extension which was slightly higher than rates observed during non-injurious human subject tests. Thus, each specimen was exposed to these displacements until tissue failure and the loads, displacements, and accelerations were recorded.

Data Analysis

Validation of the loading apparatus involved an evaluation of the forces and moments traveling through the specimen. Dynamic analysis of the experimental setup also revealed the inertial components of the shear forces and moments as well as those actually present within the tissues. Further, the actual loading rate of each specimen was computed using video analysis (WinAnalyze, Mikromak, Erlangen, Germany) to ensure repeatability. Once this investigation demonstrated the strengths and weaknesses of our apparatus, analysis of the failures in maturing tissues was initiated. This involved measuring the bending stiffness of each specimen using a 10% to 90% method and accepting $r^2 > 0.80$. The ultimate failure moment of each specimen was noted as the maximal measured moment observed during the test. Due to the small sample size, only descriptive statistics were performed on the results.

RESULTS

Our initial goal was to evaluate our dynamic bending apparatus using a known material as our test specimen. Numerous tests revealed that at 25-rad/sec the inertial component of our carriage carried a shear load of not more than 100-N while the moments reached 25-Nm. These validation experiments demonstrated that we had a working range for our apparatus which was

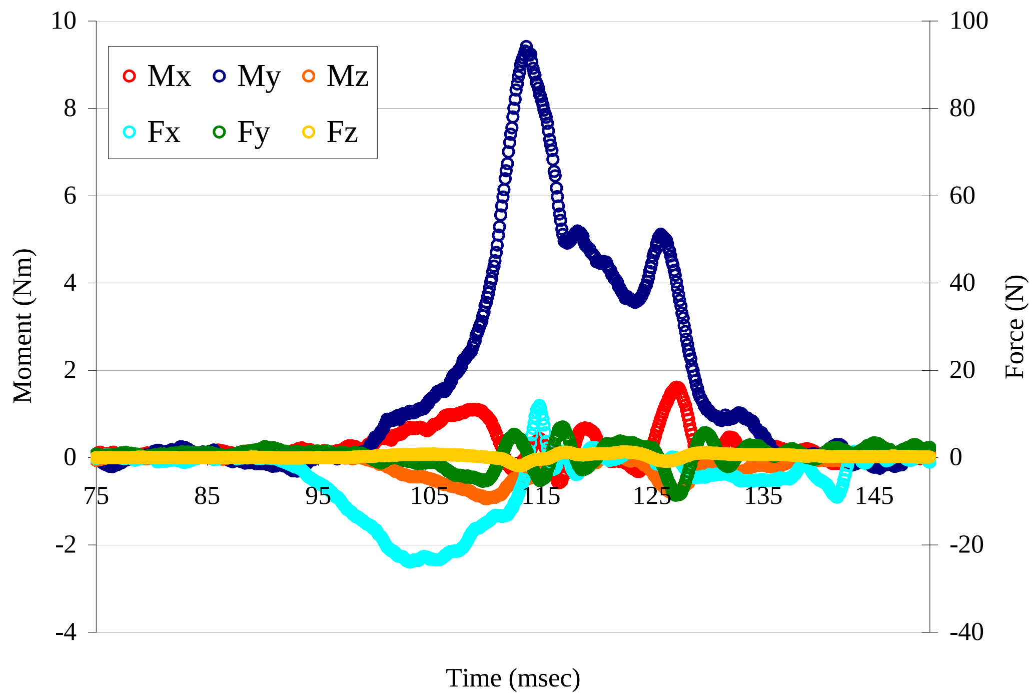

Figure 2: Time history of one experiment demonstrating both the force and moment data to demonstrate that the apparatus supplies a moment input while minimizing all other forces. The peak shear force in this case was -23-N and the peak moment was 9.5-Nm.

dynamic, but also enabled bending moments up to specimen failure without excessive shear loads. Next, we evaluated actual specimen loading curves to evaluate whether or not our prediction was accurate. Figure 2. demonstrates a load time history of one specimen where the primary moment rose to failure, while all other forces and moments were minimized. In all cases, the shear force in the direction of the applied angular displacement was less then 80-N. This has been shown to be less than half of the load required to cause failure in these tissues (Medley et al., 2001). Finally, a mean dynamic bending rate of 27.6 ±5.1-rad/sec was observed for both flexion and extension of all specimens tested. Thus, our apparatus created repeatable dynamic bending inputs which enabled a comparison of the stiffness and failure moment across our test sample of maturing spinal tissues.

Given a repeatable and accurate input and test environment, the effects of maturation were then evaluated. The bending stiffness increased with spinal development for both the flexion and extension cases (Figure 3). Further, the ultimate failure moment in both flexion and extension increased with maturation (Figure 4). Larger ultimate failure moments were recorded for extension than flexion when comparing similar age tissues. Finally, both the flexion and extension stiffness and the ultimate failure moment were found to be correlated with developmental age.

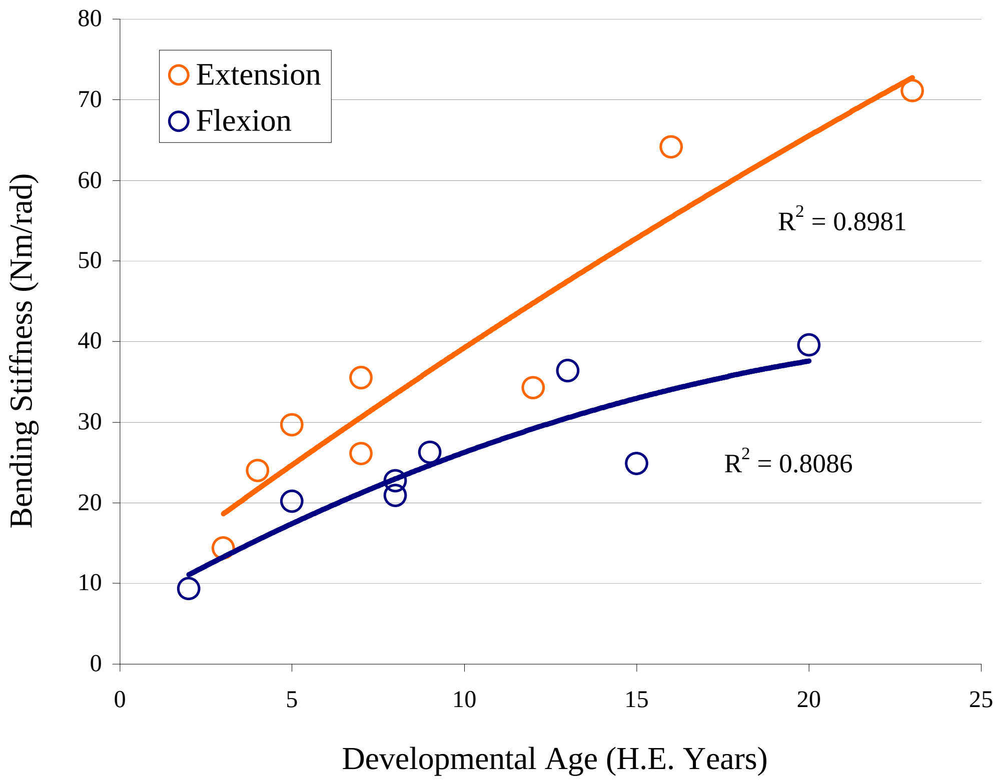

Figure 3: Bending stiffness of the C5-C6 functional spinal unit as a function of maturation. Both the flexion and extension data fit second order polynomial curves to represent their change in mechanics with age.

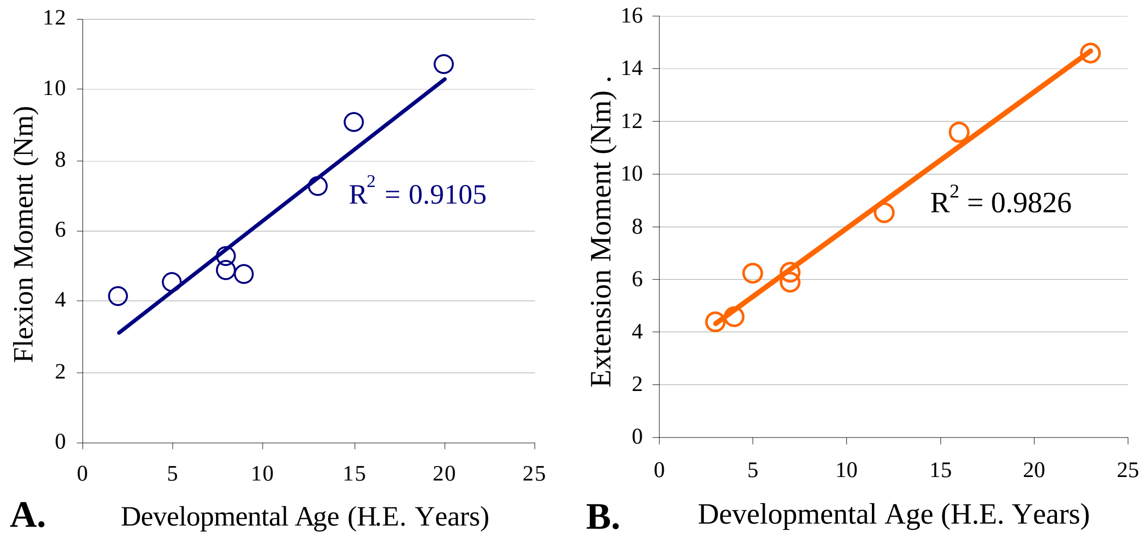

Figure 4: Ultimate failure moment as a function of developmental age. **A**. demonstrates the flexion failure moment with a linear fit with age and **B**. depicts the extension failure moment as a linear relationship with maturation.

The tissue failures observed in these experiments were consistent and unique for the extension failures and bifurcated for the flexion injuries. All of the extension injuries were inferior C5 physis or endplate failures while half of the flexion injuries were of this type and half were superior C6 physis or endplate injuries (Figure 5). These injuries are similar to those exhibited in the clinic, with the addition that bilateral facet dislocation would result from many of these in a closed injury scenario.

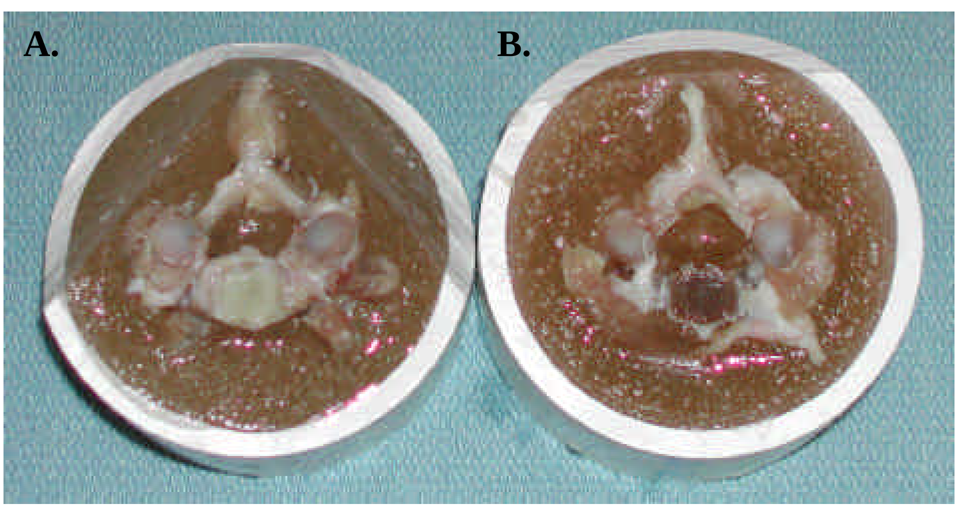

Figure 5: Extension failure of a 7-H.E. year old specimen. The failures shown are **A**. superior surface of C6 with the intervertebral disc attached and **B**. inferior surface of C5 at the zone of the failure. Note the facet capsules are ruptured as well, indicating a bilateral facet dislocation mechanism is possible, which is consistent with child injury epidemiology.

DISCUSSION

Our experimental apparatus was designed to impart dynamic bending displacements to isolated spinal segments in an effort to measure the resulting moments and loads up to and through failure. Dynamic bending was imparted on C5-C6 spinal segments at 27.6 ±5.1-rad/sec in both flexion and extension. These experiments on maturing spinal tissues resulted in tissue failures (no potting failures) that were indicative of real-life injuries, thus providing credence to the moments measured at failure.

The relationships discovered herein for the maturation effects on bending mechanics are similar to those discovered in tension, compression, and shear (Ching et al., 2001; Nuckley, 2002; Medley et al., 2001). Further, we have identified a scaling algorithm for the maturing cervical spine to predict bending stiffness (Figure 6). The flexion stiffness appears to be greater than the extension stiffness throughout development, but in general these data follow a similar trend to the Federal Motor Vehicle Safety Standard 208 recommendations. Comparisons between the data collected herein and those by Pintar et al. (2000) reveal that dynamic stiffness may be larger than quasi-static stiffness in children as well as adults. Further, these data together empower the scaling of child neck bending mechanics for modeling and anthropomorphic test device biofidelity experiments.

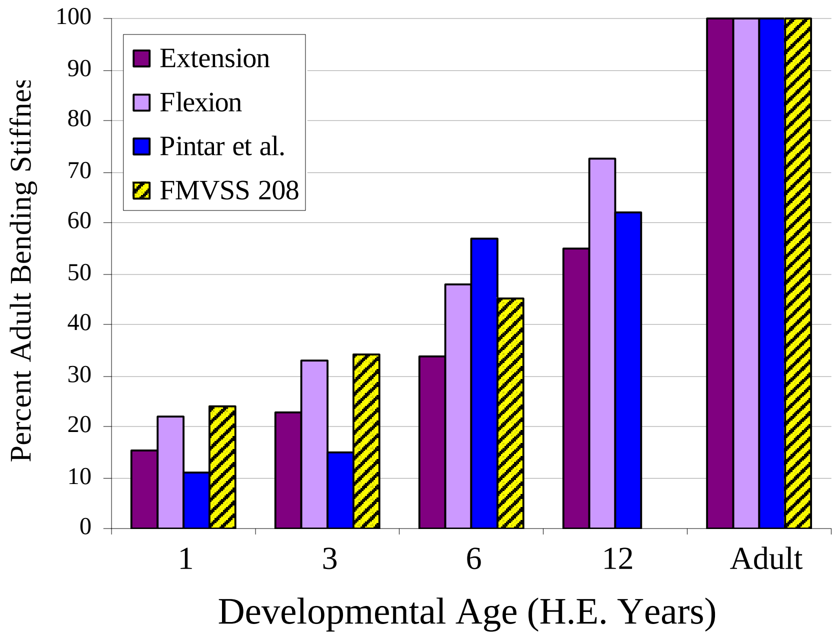

Figure 6: Scaling of pediatric bending stiffness to the adult. The flexion and extension data collected herein are scaled along side the quasi-static data of Pintar et al. (2000) and the Federal Motor Vehicle Safety Standard 208.

CONCLUSIONS

Maturation affects the bending mechanics of the cervical spine through an increased stiffness and failure moment with age. These results were determined in dynamic experimental tests which resulted in injuries consistent with epidemiological data. The relationships discovered herein provide insight into the developing cervical spine and through age based scaling may be applied to modeling efforts aimed at mitigating child spinal injuries. By enabling these modeling efforts, this dynamic bending mechanics data can foster the creation of more biofidelic anthropomorphic test devices and injury prevention strategies.

ACKNOWLEDGEMENT

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DISCUSSION

PAPER: **Dynamic Bending Mechanics of the Pediatric Cervical Spine**

PRESENTER: ***Dr. David Nuckley, Applied Biomechanics Lab, University of Washington***

QUESTION: *Guy Nusholtz, DaimerChrysler*
What type of scaling rules are you using to map between the different surrogates? You mentioned scaling to compare. At some point, you want to take the baboon data and map it into actual human pediatrics, and you mentioned something about scaling to do the comparison.

ANSWER**:** Correct.

Q: But, there's a lot of different scaling or size scaling, mass scaling, physiological scaling, geometric issues, material scaling. How did you do that?

A: We're doing all–Our goal is to do material property scaling or structural properties scaling. In this case here, what I showed were structural properties scaling of stiffness. So, we took the adult stiffness value for the baboon–that became 1–and we scaled all the pediatric data down from that. And if we were then to compare this and apply this to the human, we would take whatever value of adult human stiffness we chose, that would be times the 100% or 1 and then would scale that value down using our pediatric baboon scaling relationship. Scaling using material properties would be ideal. But in this case, I chose stiffness.

Q: So, you're taking–But in order to do that, your baboon represents pediatric. You then have to have the pediatric equivalent of stiffness.

A: Correct. Yes. That would be the material property, the modulus, etc. We would go to, probably, elastic modulus for that, or some other property.

Q: So, you just take–Okay. How good of a scale–That seems to be very rough and crude and it ignores a lot of the geometric changes that occur. How good do you think that type of scaling would be?

A: I actually think that's going to be a little bit more accurate if we sort of tease out all of the individual parts of the scaling, I think our end product might be more accurate if we have the material properties correct and we scale those material properties. We also include in that equation a scale by size and a scale by gender, as well, if you will. So, I think that breaking it down to the material properties is more important, or will be more accurate in the long-term.

Q: Thank you.

Q: *Frank Pintar, Medical College of Wisconsin*
Very nice presentation. I'm trying to get a feel for what kind of a relationship there is between static and dynamic, and you kind of know that for adults. But, do you have an indication of the same kind of static to dynamic ratios that hold for the adult will hold for the pediatric population?

A: In the limited data we've collected, I don't believe so. The very young pediatric spine seems to be more rate-dependent than the older tissues that we've examined, and that's–

Q And, you would expect that as a function of ossification?

A: I think so. Ossification.

Q: Okay. It would have been–It's logical because of the increased soft tissue.

A: Yeah. And, that's not–I don't have any data that would specifically support that. That's based on some tensile data that we collected, as well as this bending data. So, we didn't specifically investigate that. In fact, we are currently doing a compression study that will specifically investigate the interaction term between age and loading rate.

Q: And, it's probably non-linear as opposed to–

A: I would think so.

Q: Alright. Thanks.

Q: *Barry Myers, Duke University*
Dynamic bending is hard to do, so good for you for taking a poke at it. I was curious. Did you measure the cart, your load cell cart's acceleration?

A: Yes. We did. That's how we acceleration compensated our moments because, because the cart, basically, was accelerating. We have the mass in the cart, and we made sure that our F_Z–our shear force–I'm sorry, F_X. Basically we broke out our F_X so that we knew how much was a frictional component, how much was the inertial component to validate that that was, in fact, going through the specimen and that was due to the inertia so that when that force dropped down, when we saw the acceleration pulse drop down, we knew the load through the specimen at all times. The reason we didn't run these much faster is that the faster you run them, the higher the acceleration in the cart and the more the inertial force F_X there is.

Q: Yeah. I wasn't so worried about the moment. I was just curious. Did you do an MA calculation to figure out what the shear load was on the specimen because there's going to be an MA difference between–not just inertially compensating the load...MA?

A: Sure. And, that was–It ended up–time-dependent, but it was based on our maximum rate. I think it was a maximum 45 newtons, or 40 newtons of shear force. I have the data. I'd be happy to–

Q: It's surprising that you need to let that cart run.

A: Yes. When we fixed it, we were getting shear forces that were much, much higher, and–

Q: That's what I expect is that what you're doing or that what's happening is when that cart accelerates laterally, you get a big MA term that hides the, protects the load cell. The load cell doesn't see that. So, the load cell measures less shear, but it's motion segment may have pretty similar shear.

A: Yes. We thought about that. That's why we instrumented our cart set-up. The shear force at the MA term is actually less than the F_X shear force term. The rate that we chose was to minimize that shear component.

Q: Thank you.