INJURY BIOMECHANICS RESEARCH
Proceedings of the Twenty-fourth International Workshop

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PENDULUM IMPACT TEST SYSTEM TO STUDY WHIPLASH INJURY BIOMECHANICS

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ABSTRACT

Substantial experimental cervical spine biomechanics studies have been conducted using slowly applied forces and/or moments or, dynamically applied forces with contact. In contrast, few studies have been conducted to delineate the biomechanics of the structure under inertial "non-contact" type whiplash forces. This study was designed to develop an experimental methodology to induce inertial loads. A mini-sled pendulum system was designed to test specimens (such as intact human cadaver head-neck complexes) at sub-failure or failure levels under various loading modalities such as flexion, extension and lateral bending. The system allows acceleration/ deceleration input with varying wave form shapes. The system can dynamically record the input and output strength information such as forces, accelerations, moments, and angular velocities; it can also obtain the temporal overall and local kinematic data of the cervical spine segments at selected levels. This system permits a total biomechanical structural analysis of the cervical spine under whiplash or other modes. In this paper, the feasibility of the methodology is demonstrated by subjecting the human cadaver head-neck complex with musculature and skin to inertial flexion and extension loading. This method can be used to compare the human data and advanced dummy designs.

INTRODUCTION

Biomechanical evaluations of the human cervical spine are routinely conducted in a laboratory environment using mathematical and/or human cadaveric experimental models (McElhaney et al. 1983; Sances et al. 1984; Sances et al. 1986; Sherk et al. 1989; Yoganandan et al. 1989). These include an isolated component (e.g., ligament), intervertebral joint (e.g., functional spinal unit), intact ligamentous column (e.g., C2-T1 spine), and intact head-neck complexes with and without the accompanying passive musculature. Depending on the type of evaluation and the level of complexity, these models delineate the "passive" biomechanical response. Routinely, quasi-static load vectors (e.g., pure moment) have been used to induce the external loading (insult) at sub-failure levels to define parameters such as sagittal rotations and instability (Sherk et al. 1989). Recent studies have described the mechanisms of injury under traumatic forces induced by high-speed dynamic loading (Pintar et al. 1990; Pintar et al. 1995). These studies have replicated severe neck injuries such as wedge and burst compression fractures encountered in a clinical environment. These types of insult belong to the "contact type" force application.

It is well known that the human neck can be traumatized under dynamic forces applied to the structure in an inertial mode, i.e., impact forces are not directly applied via contact (Ewing et al. 1976; Aprill et al. 1990; Dwyer et al. 1990; Barnsley et al. 1994; Barnsley et al. 1995). To better understand the structural mechanics of the human head-neck complex, it is important to apply these non-contact type forces in a controlled environment.

The present study was undertaken to develop a methodology to induce inertial (dynamic) loading to an *in vitro* experimental model. In addition, the study was designed to allow for multi-axis (flexion, extension, lateral bending or any orientation in the sagittal or coronal plane) and multiple acceleration/deceleration testing at sub-failure and failure levels with varying wave forms (G-time histories). The methodology permits analysis of the dynamic temporal kinetics of the structure and the related pathoanatomical alterations.

MATERIALS AND METHODS

Testing Assembly: A mini-sled pendulum system was designed to house the specimen (e.g., human cadaver head-neck complex) and the loading assembly. The mini-sled consists of two 2.5 m long precision ground rails rigidly attached to a 1.5 m high steel frame. Four precision rolled ball bearings form the interface between the cart assembly and the rails. The cart assembly was designed to accept a six-axis load cell and allow for a vertical (z) axis of rotation of the specimen under test to permit different loading modes. A flat surface was attached to the cart assembly to accept the impact from the pendulum. The hollow pendulum impactor assembly was cylindrical in shape. It was designed to accept varying masses to its center. A load cell was attached at the leading face of the impactor. Varying energy absorbing materials surfaces were fixed to its face (Yoganandan and Pintar May 1997).

Instrumentation: An accelerometer (Entran, Model EGA-7231, Morgantown, CA) was rigidly attached to the rear face of the pendulum for measuring the input accelerations. A uniaxial load cell (Interface, Model 1210A0, Scottsdale, AZ) was attached to the leading face of the impactor to record the input longitudinal forces. A six-axis load cell (Denton, Inc., Rochester Hills, MI) was attached to the inferior fixation (T2 level) of the test specimen to record the generalized force (and moment) time histories. In addition, an accelerometer placed at this level recorded the T2 acceleration. At the superior end (head) a triaxial accelerometer and a triaxial angular velocity sensor were attached to measure the linear and angular components of the acceleration and velocity, respectively. All information were gathered using a digital data acquisition system (ODAS, DSP Technology, San Francisco, CA) according to the SAE J211b specifications at a sampling rate of 12,500 Hz. A high-speed video camera (Kodak, Model 4050, Rochester, NY) was used to capture the impact event.

Specimen Preparation and Mounting: The intact human cadaver head-neck complex was rigidly fixed at the T2-3 (inferior) end using polymethyl-methacrylate. The superior end (head) was appropriately positioned initially with break away tape to simulate an unrestrained system. The preparation was mounted with the posterior soft tissue including the ligamentum nuchae and the skin. The anterior region was devoid of spinal musculature and the skin to facilitate target placement for obtaining the localized kinematics. However, ligaments were not violated. Retroreflective targets were placed on the mastoid process of the skull, on the anterior regions of the exposed vertebral bodies, and on the lateral masses at each cervical spinal level (Pintar et al. 1990).

Loading: The specimen was inserted into the fixation on the mini-sled. The initial orientation was under the flexion (anterior region facing the impactor) or the extension (posterior region facing the impactor) mode. By rotating the specimen in the fixture housed in the mini-sled 180 degrees, the orientation was changed from flexion to extension or vice-versa. The specimen was subjected to non-contact type inertial loading by impacting at the inferior end (T2 level) with the pendulum impactor. All the biomechanical information were gathered under a velocity of 4.6 m/s.

RESULTS

The generalized force-time histories recorded by the six-axis load cell demonstrated bending moment (flexion/extension) and shear (anterior/posterior) forces to be predominant during the loading phase with minimal off-axis forces (lateral shear or axial load) or moments (torsion or lateral-flexion) suggesting that the head-neck complex of the specimen is primarily subjected to a planar-type dynamic force input. The acceleration-time histories delivered to the preparation were uni-modal.

The kinematic target analysis quantified the response of the cervical spine (Figure 1). The high-speed video photographic analysis demonstrated differing kinematics and head-cervical spinal column signatures under flexion and extension modes of loading at the same initial impact velocity in a temporal sequence. Under whiplash loading, the translation of T1 from the posterior to the anterior direction occurred immediately after the pendulum contacted the preparation. The motion was found to be similar to the human torso going forward secondary to a rear impact in real-world whiplash loading event (Geigl et al. 1995). The lower cervical spine went into an extension mode because of this translation. The inertia of the head with a concomitant T1 translation caused the head to lag producing an S-curve in the cervical spinal column. This S-curve was produced by flexion in the upper cervical complex (head-C2) and extension in the mid and lower cervical column (Figure 1). Motions of the various cervical spinal levels with respect to the base, i.e., head with respect to T1, C2 with respect to T1 are shown. The results quantify the head lag from the cervical spine motion on a temporal basis. The early local upper cervical flexion concomitant with the lower cervical extension was later followed by a uniform extension loading signature in the entire head-neck complex.

For loading from the anterior to the posterior direction to simulate a frontal impact, the upper cervical spine went into extension first, while the lower cervical spine went into local flexion, and later, the upper cervical spine also went into flexion resulting in a uni-modal flexion signature.

DISCUSSION

The objective of the present study was to develop a reproducible experimental method to induce inertial non-contact type forces to an *in vitro* cadaveric human head-neck complex model. Traditional biomechanical models using single-level or multi-level motion segments were not used in this study because of a lack of the connecting structures and its relevance to *in vivo* situations.

A mini-sled pendulum was developed in this study. This experimental set-up permitted a measurement of the biodynamic strength (e.g., forces, accelerations, moments) as well as the overall and local motion information (retroreflective target movements) using the principles of high-speed data acquisition and photography thus quantifying the kinetic and kinematic responses of the structure.

The mini-sled pendulum set-up has the ability to shape the acceleration pulse. The mass of the pendulum can be varied by adding/subtracting

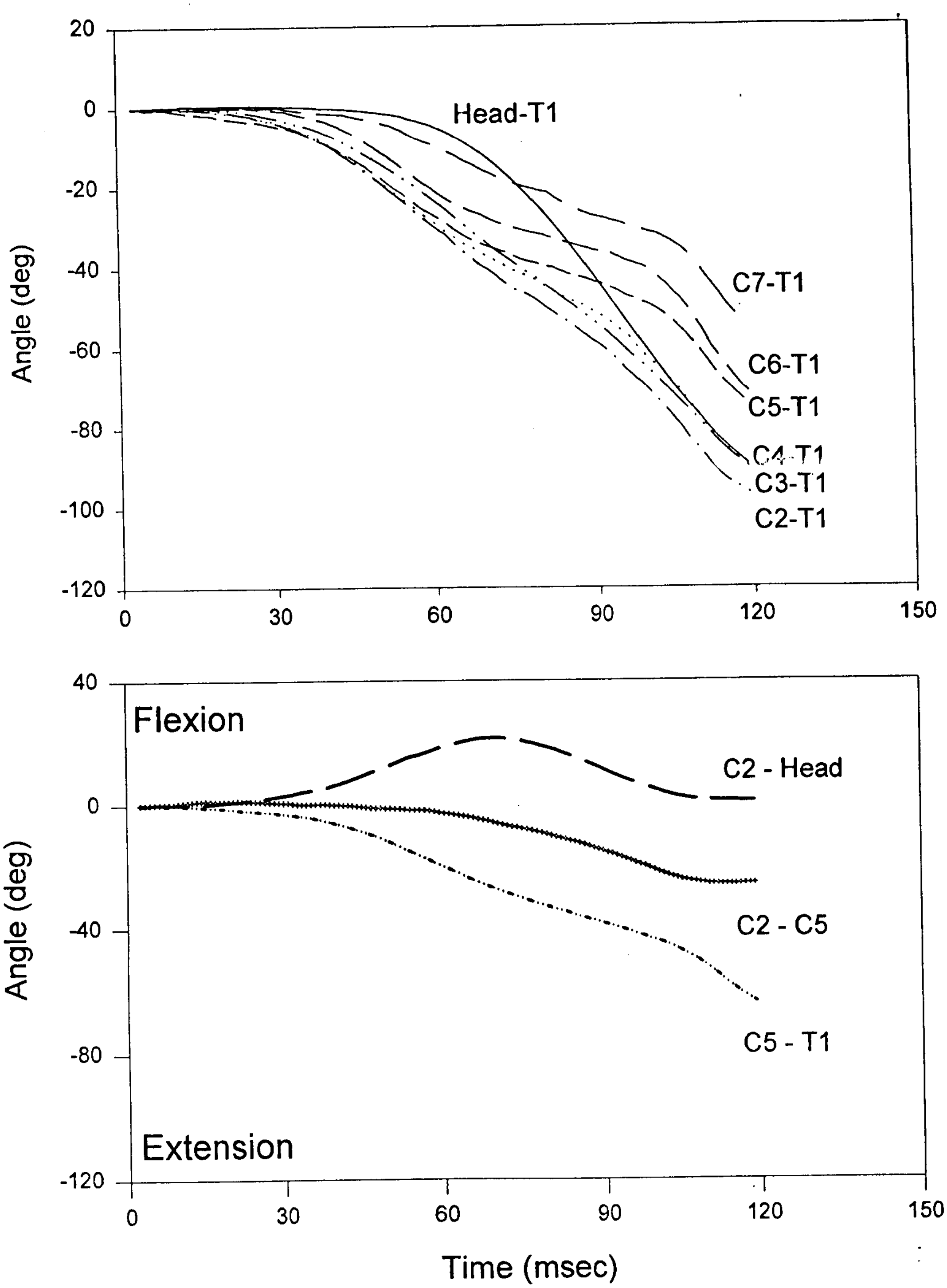

Figure 1: Kinematics of the head-neck complex due to whiplash loading

weights in the hollow cylinder, and the velocity of the impact can be varied. The acceleration level and the wave form shape can be tuned by a suitable combination of spring and dashpot (e.g., foam) materials placed at the front edge of the impacting striker or on the leading face of the cart assembly.

Instrumentation devices such as accelerometers at the rear face of the pendulum striker and the leading edge of the specimen together with the load cells on the pendulum and the specimen have the ability to provide a complete time history quantification of the biomechanical strength variables. The high-speed photographic equipment records the test event as well as the motion of the retroreflective targets. Because the optics and the strength information from all the channels are synchronized, a full analysis of the specimen response can be conducted. These data can be used for the development and validation of a detailed mathematical model (e.g., finite element model) of the human head-neck complex. The testing equipment and the associated high-speed instrumentation devices are an integral part of the test methodology. In addition, this experimental test set-up has the potential to use instrumentation devices such as a nine-axis accelerometer array on the head, or accelerometers on individual cervical vertebrae, if these data are of interest.

The optical tracking system is an integral part of the experimental design. This system consisting of the high-speed camera together with the retroreflective targets placed at every level of the head-neck complex will facilitate the analysis of localized temporal deformations of the spinal segments. For example, micro level motions representing facet joint capsule stretch and the associated anterior vertebral body and intervertebral disc compressions can be obtained. These data may have clinical relevance since the concept of facet motion has been implicated to cause pain in the human neck secondary to inertial non-contact loads (Bogduk and Aprill 1993). Research is underway to quantify the localized kinetics of the facet joints.

Another feature of the experimental test set-up includes testing of physical models such as anthropomorphic test devices (e.g., Hybrid III manikin head-neck structure) to investigate the dynamic mechanical response of the model in relation to the human. The system also allows data to be obtained from the upper six-axis load cell of the Hybrid III manikin structure by merely adding this instrumentation to the data collection apparatus. With the additional upper neck data, it will be possible to evaluate the load transmission from the occiput (upper neck) to the lower cervical region or vice-versa during inertial impact. Alternate designs of the physical model to closely match the human biomechanical response can be accomplished with the present methodology (Ono and Kanno 1996; White et al. 1996).

The posterior skin and the musculature included in the experimental model can be used to assess the contribution to the biodynamic response by conducting repeat experiments without these tissues. Likewise, the response of the human head-neck complex without the posterior musculature, i.e., the intact ligamentous cervical column, can be used to validate a detailed finite element model of the cervical spine (Kleinberger 1993; Yoganandan et al. 1996). Our methodology also allows testing of cervical columns with degenerative conditions such as stenotic canal, osteophytes, and spondylosis, and iatrogenically altered spines such as facetectomy and laminectomy to delineate their biomechanical characteristics under inertial impact.

In summary, in this study, we have developed a methodology to apply inertial non-contact flexion, extension, lateral bending, and oblique loading to an intact head-neck complex. Because of the nature of the mini-sled pendulum equipment design, it is possible to conduct dynamic studies simulating rear impact (whiplash type loading), frontal impact (forward flexion type loading), and oblique impact tests. The methodology included a dynamic recording of biomechanical data such as the generalized forces and moments, and accelerations, at high-sampling rates (Yoganandan et al. 1995). Furthermore, because the methodology used retroreflective targets, temporal local and overall kinematics of the segments of the head-neck complex can be delineated. A preliminary analysis of the segmental rotations are presented (Yoganandan et al. 1996). The hypothesis that differing local extension/flexion motions occur in the upper cervical region concomitant with local flexion/extension motions in the lower cervical spine under frontal/rear (whiplash) impact during the early stages of external loading, may elucidate the biomechanics of injury in real-world situations (Yoganandan et al. 1996).

ACKNOWLEDGMENT

This study was supported in part by DOT NHTSA Grant DTNH22-93-Y-17028, and the Department of Veterans Affairs Medical Research Service. We wish to acknowledge Narayan Yoganandan, Professor, Neurosurgery, Medical College of Wisconsin, and Frank Pintar, Director, Neuroscience Research Laboratories, VA Medical Center, and Associate Professor, Neurosurgery, Medical College of Wisconsin, for their assistance.

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DISCUSSION

PAPER: **Pendulum Impact Test System to Study Whiplash Injury Biomechanics**

PRESENTER: Tony Sances, Medical College of Wisconsin

QUESTION: Yih-Charnh Deng, General Motors Corporation
The mass of the head is about 3.5 kilograms.

ANSWER: Yes.

Q: Do you have trouble supporting the head with only a cervical spine?

A: Well, we use break away tape for the initiation of the experiment.

Q: OK. So, when impact occurs, that tape breaks.

A: Yes. That's true.

Q: OK. In some of the slides, you show some muscles attached in the back. Do you remove all the muscles?

A: Yes. I'm glad you asked that question. The muscles and the tissue are intact in the posterior region. However, in order to put our targets on, we had to remove the tissue in the front of the spine so we leave the posterior tissues intact.

Q: When you do the flexion?

A: Yes, both flexion and extension.

Q: So, only the back muscle is stiff.

A: Yes. That's right.

Q: OK. After the test, did you examine whether there was any soft tissue damage like disc or ligaments? Did you look at those?

A: Grossly, we didn't see anything but during some of the tests, we saw some tearing of the tissues near the fixation point as we went to the higher speeds. At about ten miles an hour we began to see some tissue disruption. But, at other places, from a gross inspection, no.

Q: Did you make any attempt to define some injury criterion for some soft tissue?

A: Not at these speeds. We haven't found it at this speed.

Q: OK. Thank you.

Q: Guy Nusholtz, Chrysler Corporation
Did you just do one test in a series or are those curves a cumulative of several?

A: It was one test in each series and so, we analyzed all of the curves from one test.

Q: OK. So, you haven't quite established a corridor on what the variation is for the different impacts?

A: Well, we have done about twelve to fifteen tests now and we're in the process of trying to get the corridors but as I've said, we really haven't seen any substantial injury at these levels so far. But, the important thing that we found is that the S curve develops and the upper part of the spine during the flexion mode goes into extension while the bottom part is in the opposite mode and vice versa.

Q: Have you given any thought to end conditions that you had and whether they might affect the response?

A: Well, if you look at the curves before we start having ripping of the tissue, we believe it's valid and then as it goes on, there is some tissue disruption at the fixation point.

Q: Have you thought about how that affects the response, not just the injury response but the physical response, the corridors in terms of displacement that you are generating?

A: Well, it's valid before we start seeing the large displacements between C7 and T1.

Q: Thank you.

Q: Warren Hardy, UMTRI
I have one quick question for you. I've done a number of rear end impact simulations in the past, a series of thirty-two. They were whole body tests. We were accelerating the torso, particularly T1 and T6 and the initial portion of the event, at least in a preliminary sense, can be broken into three stages and the first would be sort of a straightening of the spine making the cadaver sort of sit up and take notice which with the inertia of the head, tend to give a possible compressive effect on the cervical spine. Then after that, followed by a shear and then possibly tension, as the torso's trying to pull the head through. I was wondering what your thoughts were on the importance of that type of mechanism that you might get in a whole body cadaver test or any relevance that it might have on what you're doing.

A: Well, I think in some studies that were done on whole bodies, they have seen this. The people in Graz reported, for example, an extension. They see a slight bump or it begins to go into flection first and then you have the extension mode. But, of course, they couldn't do the segmental analysis that we were able to do by looking at the entire spine. In order to be able to compare the two, I think you would have to do an intact cadaver preparation. But, then you wouldn't have an opportunity to measure the moments and the forces at the bottom, as we've measured them. So, we were able to get the moments and the forces both in compression and the shear forces and so forth at C7-T1 and I think, as I recall, on one of the preparations, we had about a thousand inch lbs. of moment at C7-T1 and the forces were for shear in the 300 lb. range and compression in the several hundred lb. range.

Q: Larry Schneider, UMTRI

Could you tell us the initial conditions of your head and neck? How did you set those and are they representative of an automotive posture and if they are not, do you have plans to look at the effect of the initial orientation of the head relative to the neck with regard to this initial lag phenomenon?

A: Well, I think it would be interesting to do that, Larry, to place the head in a different position. We tried to put it in as normal position as we could for these first studies. If we wanted to have out-of-position studies, that would also be interesting.

Q: What do you mean by, "normal position as we could?"

A: The position that a person would have their head when they are sitting upright.

Q: And the neck as well?

A: Yes, the head/neck preparation.