Determination of a Whiplash Injury Severity Estimator (WISE Index) for Occupants in a Motor Vehicle Accident

by

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(ABSTRACT)

The diagnosis of a whiplash injury is a very subjective process. A claim of this type of injury is usually made on the basis of pain, which may or may not be accompanied by clinical signs of trauma. This study was aimed at providing a more objective, quantitative approach to identifying the potential for whiplash injury in a direct front-or-rear-end automobile collision.

The Whiplash Injury Severity Estimator (WISE Index) was created using data obtained from Dr. Schneck's personal library of case files, including the collision acceleration of the vehicle, and the height, weight, and sex of the occupant. Some extrapolated data was also used representing the low and high ranges of height, weight, and collision acceleration to increase the range of the WISE Index. Data was analyzed by the Dynaman computer program in conjunction with the Articulated Total Body Model, to calculate the response of the body to external forces and impacts. The dynamic response of the occupant, combined with preexisting medical statistics provided the information necessary to perform a regression analysis in MINITAB and thus construct the WISE Indices shown below.

Male WISE Index (R^2 = 0.993)

$$x = 0.2643 \pm 0.4071\left|\left(accel, g\right)\right| - 0.01428\left(PI\right)$$
$$1.1\text{ g} \le accel \le 5\ g; \quad 22.4 \le PI \le 25.0$$

Female WISE Index (R^2 = 0.978)

$$x = 0.6214 \pm 0.3429\left|\left(accel, g\right)\right| - 0.02929\left(PI\right)$$
$$0.8\text{ g} \le accel \le 5\ g; \quad 22.3 \le PI \le 31.0$$

Acceleration: Use the negative sign if it is a rear-end collision and the positive sign if it is a head-on collision.

ξ: A negative value means that potential injury results from backward head rotation, as in a rear-end collision. A positive value means that potential injury results from forward head rotation, as in a head-on collision.

$|\xi| < 1$ = "Safe"

$|\xi| > 1$ = "Dangerous"

The WISE Index allows one to predict the potential for a whiplash injury, as well as the intensity of the injury, based solely on collision acceleration, height, weight, and sex of the occupant. It is anticipated that this work and future efforts in this area will provide the information base necessary for anyone to effectively evaluate the validity of an alleged whiplash injury.

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1.0 Introduction and Problem Statement

According to the National Highway Traffic Safety Administration (NHTSA), in 1996 there were 6,842,000 reported automobile accidents resulting in 41,907 people being killed and 3,511,000 people being injured. Among the 3,511,000 injuries were many neck injuries, and 84% of the neck injuries were of the whiplash type.

Whiplash injuries affect the lives of over 1,000,000 people in the US every year, and claims of this injury have always been a subjective diagnosis at best. Whiplash injury is an imprecise term for damage to the ligaments, vertebrae, spinal cord, and nerve roots that is produced by a sudden jerking (in "whiplash" or whipping fashion) of the head with respect to the vertebral column. Specifically, the whiplash injury is caused by the inertial force of the head exerting a bending moment at the C7-T1 vertebra (neck pin), as shown in Figure 1. This figure shows the inertial force $\underline{F}$ acting at the center of gravity of the head (CG) a distance $\underline{r}$ away from the C7-T1 vertebra. This causes a bending moment ($\underline{M} = \underline{r} \times \underline{F}$) on the C7-T1 vertebra leading to whiplash. When a person makes the claim that he or she has suffered a possible whiplash injury, usually a physician's support is enough to substantiate the claim in a court of law. Since there is currently not an absolutely precise way to diagnose this injury, other than in the case of vertebral bone fracture, it is difficult to refute or substantiate the physician's testimony. If a Whiplash Injury Severity Estimator (WISE Index) could be developed, the potential for a whiplash injury could be more objectively quantified and used in conjunction with a physician's subjective testimony.

To quantify the potential for a whiplash injury, actual cases of occupants involved in motor vehicle accidents were used in conjunction with a computer model of human body dynamics. Height, weight, sex of the occupant, and kinematic information revealing the acceleration of the collision were obtained for each of these cases. Results of the dynamic analysis were then compared with known data regarding the ability of the human neck to tolerate loading without consequence, in order to develop the WISE Index. To increase the range of the WISE Index, some extrapolated cases were added to the data set, representing the low and high ranges of height, weight, and collision accelerations.

The work reported herein was undertaken as a modest first step towards the ultimate formulation of just such an objective Whiplash Injury Severity Estimator (WISE Index). That is to say, the objective of this research is to develop a procedure whereby one could combine the WISE Index with existing medical data to predict the potential and severity of a person's whiplash injury as a result of a motor vehicle accident. It is hoped that this research will begin to form the solid information base necessary for anyone to effectively evaluate the validity of an alleged whiplash injury based upon biomechanical data, not just subjective opinion.

2.0 Review of Literature

2.1 Anatomy of the Cervical Spine

The main focus of this research is on the articulation between the head and the neck, specifically the cervical spine. The cervical spine, shown in Figure 1, is formed by seven bones called vertebrae, which are designated C1 through C7 numbered from superior (top) to inferior (bottom). Of these vertebrae, C3-to-C7 are similar to each other in structure and function and can be described as "typical." The remaining two vertebrae, C1 and C2, are unique in size and shape because they have a special purpose. The functional unit of the spine consists of two adjacent vertebrae along with an intervertebral disc and adjoining ligaments, together referred to as the motion segment.

As shown in Figure 2, the typical vertebra is made up of a vertebral body, two pedicles, two laminae, and two spinous processes. The cervical vertebral body is elliptical in cross-section in a sagittal plane (cut into left and right sections) with a transverse "saddle-type" superior surface. On the sides of each body are raised lips which constitute the ulcinate processes. On the inferior (underneath) side of the body there is a protruding lip which articulates with the bevel or the upper surface of the underlying vertebra between the ulcinate processes. To each side of the vertebral body there is a doughnut-shaped transverse process which surrounds a transverse foramen, or passage, through which the vertebral artery travels up the cervical spine. The transverse process is grooved on its upper surface to allow for the passage of the spinal nerve roots.

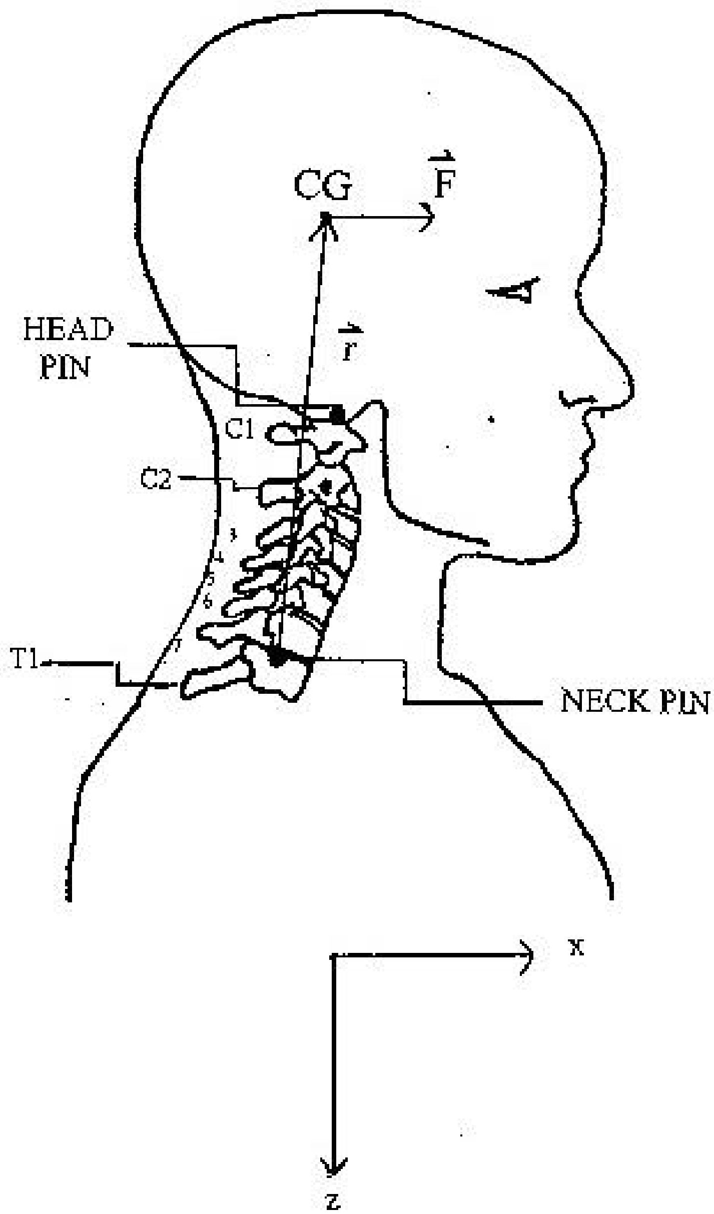

Figure 1 **Anatomical Representation of the Human Head / Neck System**

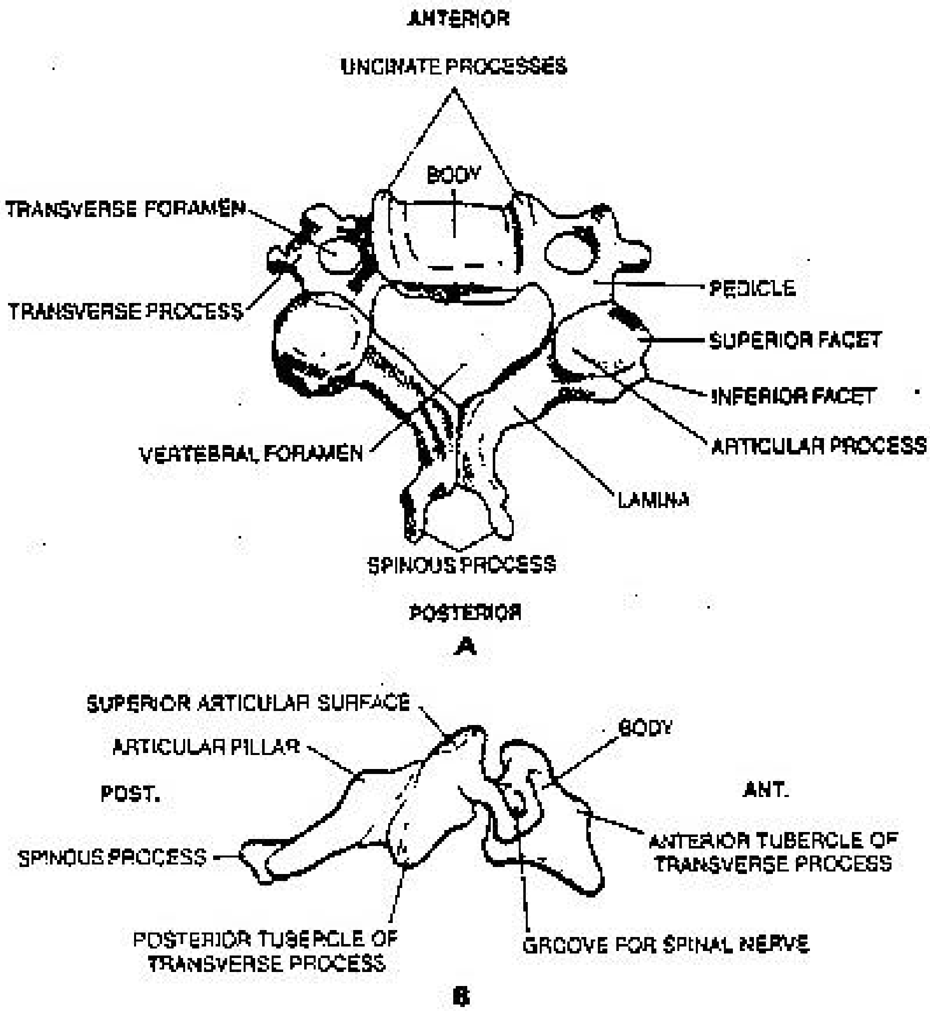

Figure 2 Schematic drawings of a typical cervical vertebra, representative of C3-C7. (From Nordin and Frankel, 1989, pg 211)
(A) Transverse View (B) Lateral (side) View

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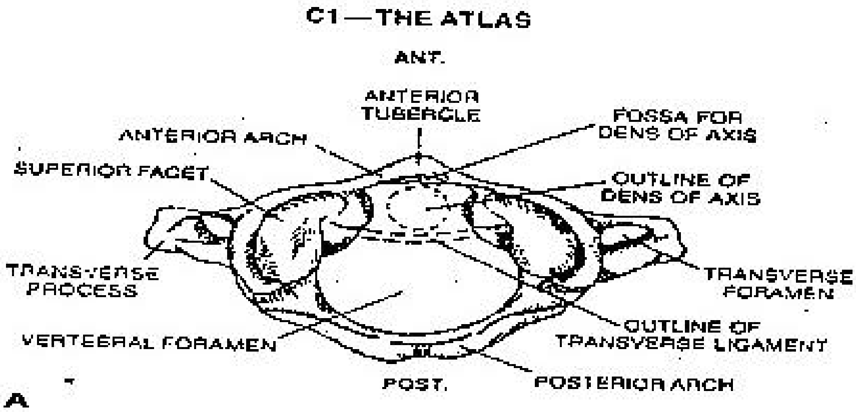
This grooving, combined with the presence of the foramen, weaken the process and predispose it to fracturing.

The facets of adjoining vertebrae form true synovial joints (fully movable joints in which a lubricant-filled cavity is present between the two articulating bones) with cartilage coating each articulating surface. The cervical facets are rotated with respect to the transverse plane so that the superior (upper) facets slant anterior (forward) 45 degrees and the inferior (lower) facets slant posterior (backwards) 45 degrees. These cervical facets guide the motion of the two vertebrae. The facets are located posterior to (behind) the transverse process. Posterior to the facets are the two lamina, each of which is angled toward the centerline from the lateral edge of the vertebra. These lamina unite to form the spinous process. The spinous process on each vertebra is bifid, or forked, in order to support the strong ligaments that attach from it to the head. Lying between the facets directly posterior to the vertebral body is an opening called the vertebral foramen. It is through this canal that the spinal cord passes. The foramen has a rounded triangular shape, and is about two thirds larger than the spinal cord to allow for motion without compressing the cord and other neurovascular structures.

The atypical vertebrae differ from C3-C7 in size and function, as shown in Figure 3. The second vertebra, C2, is named the axis because of the dens, a spinous process that protrudes superiorly from the vertebral body. It is around this structure that C1 rotates about the longitudinal axis of the spine. At the top of the dens lie points of attachment for the apical ligament and two alar ligaments, which extend to the sides of the skull and limit the rotation of the head. The facets and transverse process are modified to allow for rotational articulation with C1. The atlas, C1, has no vertebral body and is essentially a



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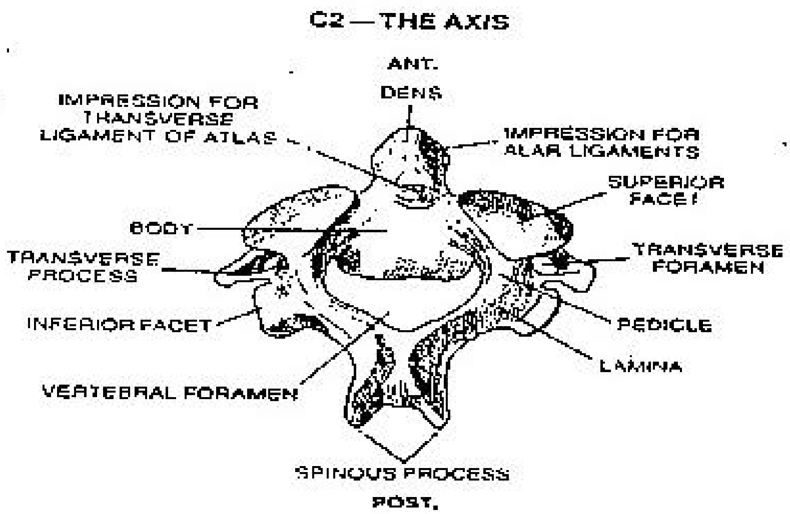

B

Figure 3 **The Atypical Cervical Vertebrae (Nordin and Frankel, 1989, pg 214)**
(A) Superior view of C1, the atlas
(B) Superior view of C2, the axis

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ring of bone (like a “washer” on a machine screw). The inferior facets of C1 correspond to the superior facets of C2, as was the case with C3-C7, but both are parallel to the transverse plane. This feature allows significant rotation around the longitudinal axis of the spine. The atlas derives its name from the fact that the weight of the skull is supported on its superior facets, which form the base of the atlanto-occipital joint, the head pin.

Essential to the stability of the spine are the supporting ligaments. The strong anterior longitudinal ligament runs along the anterior surface of the spine and attaches to the base of the skull, as shown in Figure 4. It is firmly attached to the vertebral bodies and more loosely attached to the intervertebral discs. The posterior longitudinal ligament spans the posterior surface of the vertebral bodies within the vertebral foramen. It is firmly attached to the discs and separated from the surface of the vertebrae by blood vessels that enter and leave the spinal canal. Bands of elastic fibers called the ligamentum flavum attach to each lamina and lie within the spinal canal on the posterior surface of the vertebral foramen (Figure 4). This ligament is considered the most elastic in the human body (Nordin and Frankel, 1989, pg 213). Other supporting ligaments lying on the posterior aspect of the vertebra are the capsular ligaments, the intertransverse ligaments, and the interspinous ligaments. As they run up the cervical spine, the interspinous ligaments blend into the supraspinous ligament, which is very dense and can be felt just beneath the skin on the back of the neck. The cruciform ligament is comprised of two main ligaments: a transverse ligament that attaches bilaterally to C1 and a longitudinal ligament that attaches to the body of C2 and connects to the rim of the

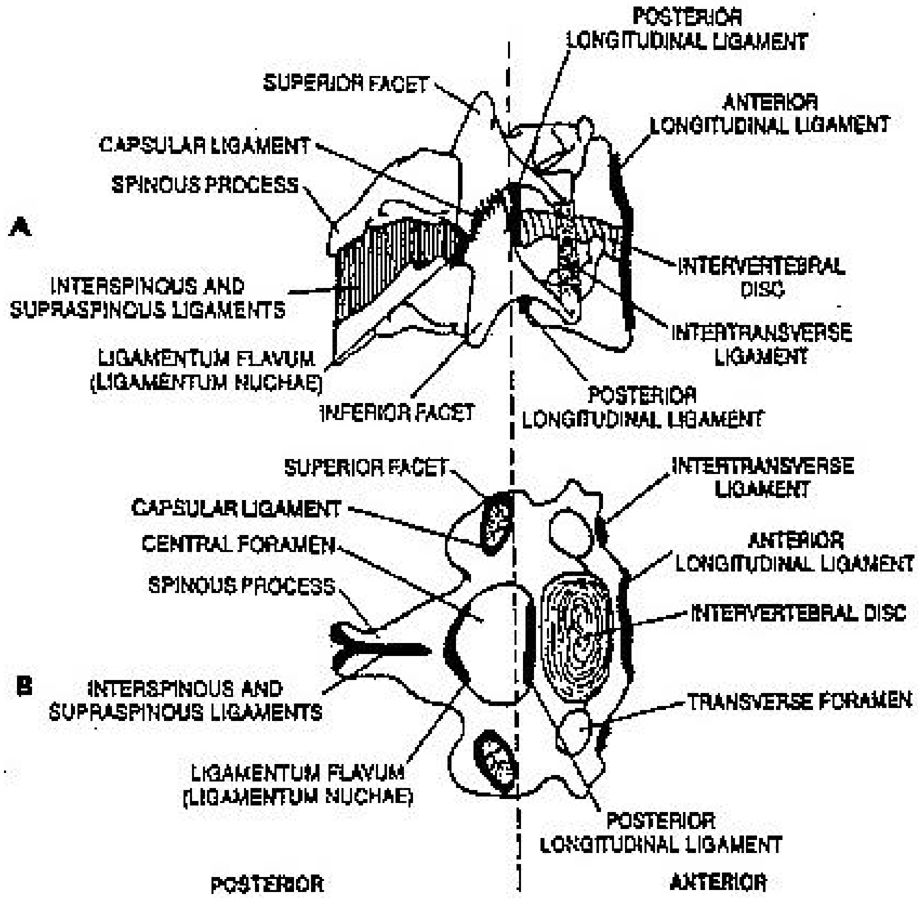

Figure 4 Schematic representations of a cervical motion segment composed of two typical vertebra, surrounding ligaments, and intervertebral disc. (Nordin and Frankel, 1989, pg 212)
(A) Lateral View (B) Superior View

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foramen magnum, the opening into the skull. This ligament prevents C1 and C2 from translating laterally relative to one another.

The intervertebral discs form specialized joints between the cartilage-coated surfaces of adjoining vertebral bodies. These discs are comprised of a central mass of gelatin like material, called the nucleus pulposus, surrounded by a tough outer covering, the annulus fibrosus. In addition to acting as a “spacer” between vertebra, giving the spinal column flexural maneuverability within restraining limits, the disc also serves a hydrostatic/viscoelastic function within the motion segment as it stores energy and helps to distribute loads.

The cervical spine is the most mobile region of the spine and allows the head a wide range of motion. The total range of motion includes about 145 degrees of flexion-extension (as if saying, “Yes”). Flexion refers to head motion around C1 that moves the chin towards the chest; extension is just the opposite. Flexion and extension of the neck are shown in Figure 5, which shows the various motions involved in head movement (Vulcan, King, and Nakamura, 1970). Axially, the head together with C1 can rotate around C2 approximately 180 degrees about the longitudinal axis of the spine, 90 degrees to both the left and the right (as if saying, “No”). The C1-C2 articulation provides roughly 50% of axial rotation with the other cervical vertebrae each contributing slightly to the remaining 50%. As a person ages, or experiences degenerating illnesses, this range of motion increases which can lead to injury.

Whiplash injury is a flexion-tension-extension injury and thus involves the ligaments of the cervical spine. An external impact can force the head into flexion beyond the normal physiologic range, damaging the ligaments in the form of a tensile

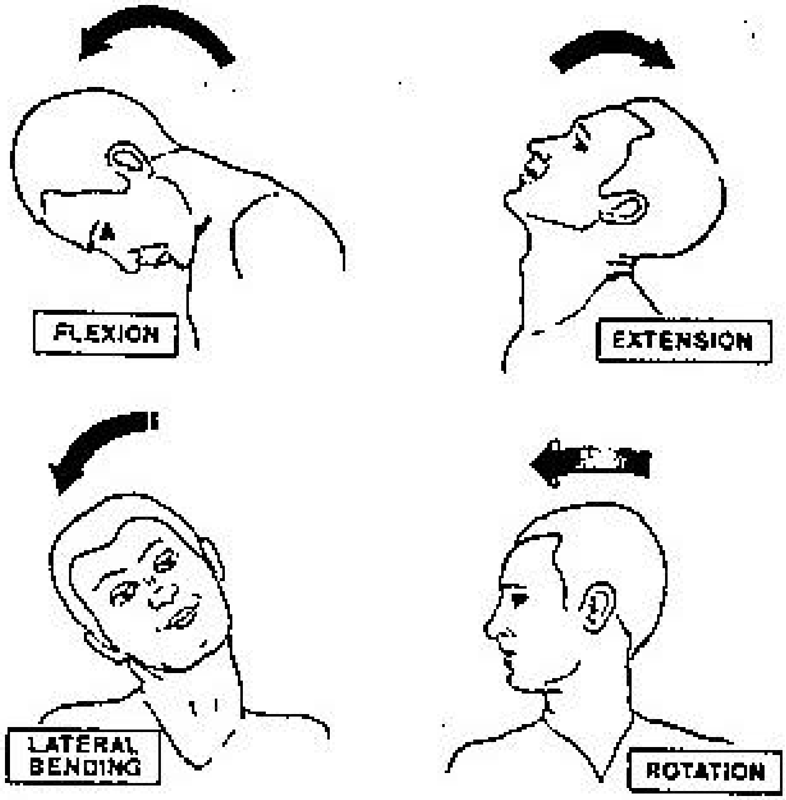

Figure 5 Motions Involved in Head Movement

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strain. With a higher force of impact, these ligaments may completely tear. A possible complication of this scenario is that the facets of the upper vertebra of the motion segment can override those of the lower segment and lock anteriorly when the spine recoils (Nordin and Frankel, 1989, pg 222). If the force is sufficiently severe, vertebral fracture may occur, and if more severe still, the intervertebral disc may rupture as well (commonly known as a herniated or “slipped” disc). Ultimately, if the whiplash is coupled with an impact such as the person’s head hitting the interior of the automobile, the range of motion caused may sever the spinal cord partially or completely leading to paralysis or death. When one considers that the head is basically a large mass (weighing some 10 pounds or more) on a small pivot, it is easy to imagine the potential for injury from a sudden change in motion.

2.2 Whiplash Injury Mechanisms

There are several different types of injury that are collectively termed "whiplash". The first of these is damage to the ligaments in the neck. A ligament is made up of dense connective tissue with fibers that are arranged parallel to each other. As in the case of other connective tissues in the human body, ligaments consist of relatively few cells suspended in an extracellular matrix material, most of which is water. The solid portion of the matrix is composed primarily of collagen and elastin fibers. The elastin content is especially important in contributing to the mechanical properties of the ligament (Nordin and Frankel, 1989, pg 62). Ligaments with high elastin content are more elastic (they can stretch when loaded and return to their original length when unloaded), which is the case for the ligaments present in the neck. Ligaments are also viscoelastic structures (their response depends both upon the amount of load to which they are subjected, and the rate at which they are loaded) which in general are pliant and flexible, allowing for the natural motion of the bones to which they attach. However, the collagen within the ligaments make them significantly strong and inextensible under tensile loading, which provides suitable resistance to applied forces (Nordin and Frankel, 1989, pg 63).

The upper limit for physiologic strain in ligaments has been found to be from 2 to 5% during normal activity such as running and jumping (Fung, 1981). These strains usually correspond to loadings less than 500 N or 112.4 lb. Some microfailure can occur during this range, but complete failure of the ligament does not usually occur until loading exceeds about 1000 N or 224.8 lb. (Noyes, 1977). When a ligament is subjected to a load that exceeds the normal range of about 400 N or 90 lb., microfailure occurs even before the yield point is reached. Once the yield point is exceeded, the ligament begins to undergo gross failure, which can cause the joint involved to become abnormally displaced. This can cause further complications involving the bones of the joint, surrounding muscles and ligaments, or local blood supply.

Injuries to the ligaments are categorized clinically in three ways depending on the level of severity. Injuries of the first category produce negligible clinical symptoms. Little pain is felt and there is no detectable joint instability. Second level injuries produce severe pain, and joint instability may be clinically detectable. In this case, the collagen fibers within the ligament matrix have failed progressively and partial ligamental rupture is possible. The strength and stiffness of the ligament may be reduced by as much as 50% because the amount of undamaged tissue is greatly reduced (Nordin and Frankel, 1989, pg 66). Third category injuries usually involve severe pain at the time of trauma and continued lesser pain thereafter. A majority of the collagen fibers are ruptured, which greatly reduces the load carrying capability of the ligament to almost zero. As a result, the joint is completely unstable, a fact that can be verified clinically.

Instability in the cervical spine due to ligament damage can have other complications as well. Weak ligaments can cause the vertebrae of the neck to articulate well beyond the normal range of motion. This can lead to fractures of the vertebral processes or even of the vertebral bodies, themselves. Also, this displacement can damage the nerves which lie in close proximity to the joint or, in worse case scenarios, damage the spinal cord itself.

Many mechanical forces are at work to cause a whiplash injury. Usually there is a sudden acceleration or deceleration of the involved person’s body. This commonly occurs when a car is struck suddenly and forcibly from behind. Due to the weight of the human head, a considerable amount of bending moment (flexure) can develop in the cervical spine. This is especially true for the atlanto-occipital joint between the head and neck, called the head pin in the ATB model, and the cervical joint between the neck and first thoracic vertebra, called the neck pin in the ATB model. In a direct rear-end injury as described above, the first thing that occurs is that the body is thrown forward with the seat but the head is thrown backwards, hyperextending the neck. Almost all whiplash injuries are flexion-extension injuries and tensile, meaning hyperextension, hyperflexion or both occur, pulling on the joints involved.

The hyperextension or hyperflexion can injure the neck in a number of other ways as well. Shear injuries can occur between vertebrae when the head is in the greatest extension and the torso is accelerating forward. Compression injuries can occur when the head is accelerated downward towards the spine, or when tissues are compressed during the extension phase of the collision. During this extension, which causes compression in the posterior area of the neck, tension can occur in the anterior of the neck, resulting in tearing of the musculature or anterior portion of the intervertebral disc. Tension injuries can also occur in a head on collision in which the neck is thrown forward. In this case, the flexion causes tension in the posterior area of the neck which could tear the posterior part of the intervertebral disc. The majority of all whiplash injuries, however, are flexure-related injuries. This bending is created at the neck pin by the inertial force exerted by rotational acceleration of the head around the y-axis (axis going from left to right through the neck) on the fulcrum of the top of the cervical spine.

It has usually been assumed that whiplash occurs at the maximum angle of flexion or extension during the accident. However, when the neck reaches its maximum extension/flexion it is found that it does not really exceed physiologic limits. Another theory, proposed by the American Society of Biomechanics (ASB), alleges that the injury occurs before maximum flexion/extension when the neck is just beginning to react (Panjabi, 1997). Using the example of a rear end collision, the injury occurs towards the beginning of the body's reaction, when the spine forms an S-shaped curve with flexion at the upper levels and hyper-extension at the lower levels. This antagonistic push/pull effect causes a tremendous amount of bending moment on the neck, and is hypothesized by ASB to be where the injury occurs (Panjabi, 1997).

All whiplash injuries are hard to predict because so many different variables are involved from accident to accident. The variables that are known to be most highly correlated with whiplash-related injuries are acceleration, and the individual's height, weight, and sex. Other variables that are important also but much harder to quantify are age, amount of warning or state of anticipation to the impact, and the health of the individual's bones, muscles, ligaments, and discs. When a physician testifies in court, he or she will base the whiplash diagnosis on the symptoms that the patient describes and the results of a clinical examination. As mentioned previously, clinical signs of joint instability are not always present, so a Whiplash Injury Severity Estimator (WISE Index) would help complement the clinical examination with some additional objective predictors of the probability of injury, which is what the present work addresses.

2.3 Other Injury Criteria

As mentioned before, the diagnosis of whiplash has always been a very subjective process at best. Since physical examinations and diagnostic imaging evaluations can be inconclusive even when a whiplash injury may have occurred, there are some standard guidelines that are generally taken into consideration by a physician or testifying engineer. First of all, experts have shown that 42 ft-lb of bending moment (flexure) is a critical value of bending moment that the neck can withstand before whiplash-type injury becomes highly probable (Mertz and Patrick, 1971). Usually, information involving the bending moment to which the subject's neck is subjected is unknown to anyone examining the patient, but information on acceleration of the collision can be readily determined by forensics experts. From such acceleration studies, "rule-of-thumb" guidelines have been proposed (personal communication, Forensic Technologies International, 1998), which are:

$a < 2g$ Virtually No Chance of any Injury
$2g \le a < 2.3g$ Gray Area, Need More Information
$2.3g \le a < 3.2g$ Injury Possible
$a > 3.2g$ Injury Highly Probable

From 2 to 2.3 g's the acceleration guidelines are listed as "Gray Area", meaning that injury is not ruled out, but is not likely unless there is an extenuating circumstance, such as old age or a pre-existing medical condition. From 2.3 to 3.2 g's the acceleration guidelines are listed as "Injury Possible", meaning that there is definitely a chance of ligamentous strain (whiplash), but symptoms should disappear in 7-14 days. For accelerations greater than 3.2 g's injury is very probable and can range from ligamentous strain as described above to much more severe and long-lasting injuries.

There have been some other injury criteria developed to try to quantify the risk or potential for injury in a motor vehicle accident. The two most prominent criteria are the Gadd Severity Index (GSI) and the Head Injury Criterion (HIC) (Kline, 1988, pg 529). Both of these criteria relate the possibility of life threatening injury to the acceleration of the collision only.

The Gadd Severity Index (GSI) was introduced by Charles W. Gadd (Gadd, 1961) after analyzing acceleration data obtained from automobile accidents. He found that when the whole body acceleration data was plotted on log-log coordinates, a straight line would fit the data. The equation of that line was $A_C T^{0.4} = 15.83$. By raising both sides of the equation to the 2.5 power, he obtained $A_C^{2.5} T = 1000$, where T is the duration in msec of constant acceleration, A_C, and 1000 is the tolerance level or critical value of the index. This equation later evolved into:

$$GSI = \int \left[A(t) \right]^{2.5} dt = 1000 \qquad (2.3.1)$$

In this equation, A(t) is the head acceleration in g's expressed as a function of time and the integration is over the impact duration in msec. According to Gadd, if GSI > 1000, the acceleration pulse can be considered dangerous. If GSI < 1000, the acceleration pulse can be considered not to be threatening. According to the GSI, for A(t) constant, a 2.5-g impact sustained for 100 msec would be dangerous, which is consistent with the FTI guidelines given above.

The Head Injury Criterion (HIC) was first proposed by J. Versace (Versace, 1971) and then modified by NHTSA (FMVSS, 1972). It was developed from the GSI and added a maximization procedure to help define the pulse duration, thus eliminating contributions from low-level long duration accelerations. The equation developed was:

$$HIC = \left[\frac{\int_{t_1}^{t_2} A(t)dt}{(t_2 - t_1)} \right]^{2.5} (t_2 - t_1) = 1000 \qquad (2.3.2)$$

Here t_1 and t_2 are varied to find the maximum HIC value. If HIC < 1000, the situation is not considered threatening while if HIC > 1000, the situation is considered to be dangerous.

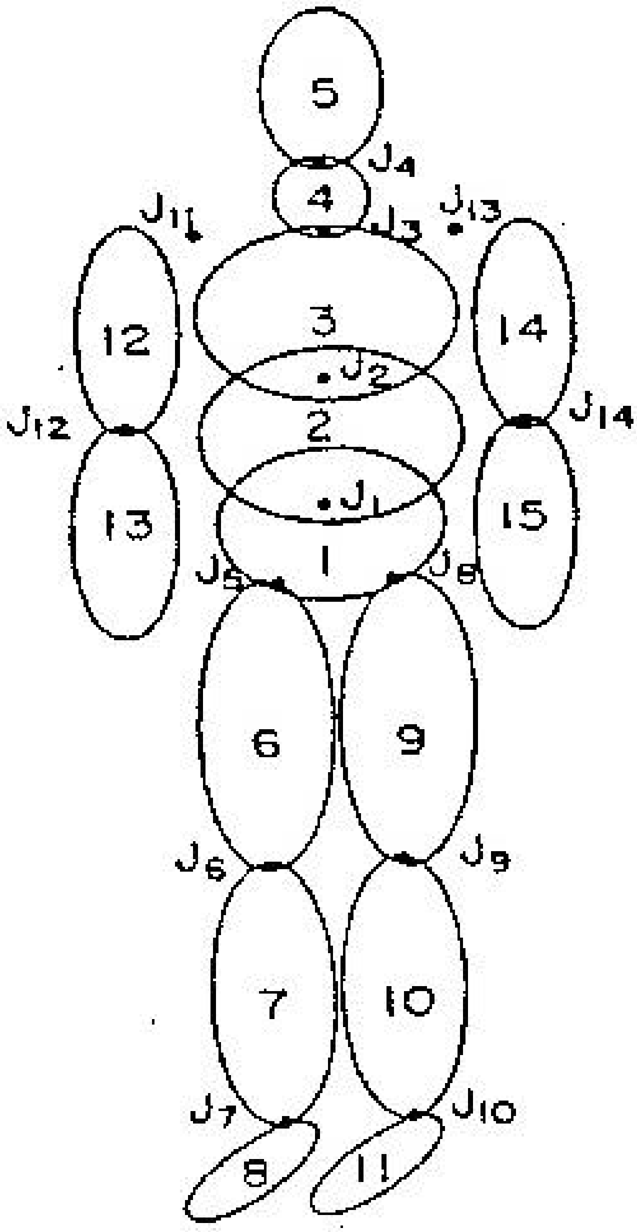
The HIC and GSI are very similar for clearly defined pulses and are equal for a square wave. In general, the HIC tends to be a little smaller than the GSI because the pulse duration is shorter. In other words, even though you are varying t_1 and t_2 to find the maximum HIC, the GSI still is evaluated over the entire time interval so it generally produces a greater value. Both the HIC and GSI base their analysis strictly on acceleration, which is a significant hindrance to their usefulness, because they do not take into account any features of the specific individual involved in an accident. The Whiplash Severity Estimator (WISE Index) developed in this study adds height, weight, and sex into the analysis also, which increases the accuracy and usefulness of the results.

3.0 Methods and Materials

3.1 ATB Model/Dynaman Generation of Data

The GEBOD III and Dynaman software programs were used to analyze the data set. The latter included 23 (5 male, 18 female) actual cases of occupants involved in motor vehicle accidents (obtained from Dr. Schneck's personal library of case files), and 36 (18 male, 18 female) cases extrapolated from the real data (user-defined height, weight, collision acceleration) to increase the range of the WISE Index. The data received from Dr. Schneck consisted of height, weight, and sex of the individual, along with the acceleration of the collision in g's. The acceleration of the collision was determined by a forensic analysis of the dynamics of the vehicle involved in the accident. Together, the two software programs constitute the personal computer version of the Articulated Total Body model which was designed by the US Air Force for the Armstrong Laboratories in the 1970's. The program has been used mainly to simulate aircraft ejections, but it is capable of simulating complex automobile accidents as well.

There are two phases of data acquisition using these programs. The first involves the GEBOD program. The user inputs information on height, weight, and sex of an individual to GEBOD which then outputs the physical parameters discussed below, for further computation by Dynaman. Dynaman uses a 15 segment, 14 joint model of the human body with each segment considered to be a three dimensional contact ellipsoid, as shown in Figure 6. The segments are assumed to articulate with one another as rigid



Joint j connects segment j with segment j+1

Figure 6 **Standard ATB 15 Segment / 14 Joint Model of the Human Body**[1]

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bodies that include the head, neck, upper torso, center torso, lower torso, two upper arms, two lower arms, two upper legs, two lower legs, and two feet. Hands are considered to be part of the lower arm. The segments are assigned identification numbers from 1 to NSEG, with NSEG in this case being 15. Joints connecting the segments are also assigned numbers from 1 to NJNT, NJNT equaling 14, as is illustrated in Figure 6. Although user-defined “sensitivity” coefficients may be assigned to each joint in order to account for their individualized contribution to the dynamic response of the system, in this study, all joints were assumed to be functioning normally (i.e., they had an unrestricted sensitivity of unity). Each of the ATB segments has a local coordinate system relative to a user-defined inertial coordinate system (Estep and Schneck, 1992). GEBOD provides all relevant physical data for each segment ellipsoid necessary to investigate the dynamics of the segment in terms of the local coordinate system. These parameters include the segment weight, center of mass, semiaxis lengths (geometry of segment ellipsoids), and principal mass moments of inertia for a “typical” individual (based on a regression analysis which is described in Baughman, July 1983). Also provided are values for yaw, pitch, and roll of the segment relative to its center of mass. Data for the joint locations are also provided by GEBOD, such as the location of the joint center with respect to both the adjoining segment’s reference frame and also the inertial reference frame. The output data also includes the viscoelastic, flexural, and torsional characteristics of the joints, which are essential for proper modeling of the body, especially the head-neck system. All of this data allows for the simulation of the effects of joint motion limitations, friction within the joints, and ligament forces within the head/neck system.

Six primary coordinate systems are used to describe the model data for use in Dynaman. They include: inertial, vehicular, principal, contact ellipsoid, joint, and local body segment coordinate systems. The six coordinate systems are necessary not only for the simulation itself, but to provide convenient reference frames for analyzing the large data sets generated. The three used in this study were the inertial, joint, and local body segment systems. The coordinate system used to describe the local body segments and joint systems are defined by Dynaman, while the inertial system is user defined. For the sake of uniformity and convenience, the inertial coordinate system was chosen with the same principal direction as the local and joint systems. The principal direction of the systems is defined as if one were to look in from the passenger side window in a car. Thus, the direction from the subject's back to chest and continuing forward is the positive x-direction, the lateral direction from the subject's left to right (toward the viewer) is the positive y-direction, and the direction from the top of the head down parallel to the Earth's gravity vector is the positive z-direction. The origin of the local body segment coordinate system is located at the segment center of mass, and joint coordinate systems are fixed at the location of the joint. These coordinate systems are shown in Figure 7.

For the purposes of this study, the segments of the body model that are associated with a "whiplash" injury were specifically examined. These segments were the head, neck, and upper torso along with the corresponding joints between them, the head pin and the neck pin, all modeled as three rigid segments connected by two ball and socket joints, shown in Figure 8. The selection of ball and socket joints allows only for relative rotational motion between the segments. The head pin ball and socket geometrically

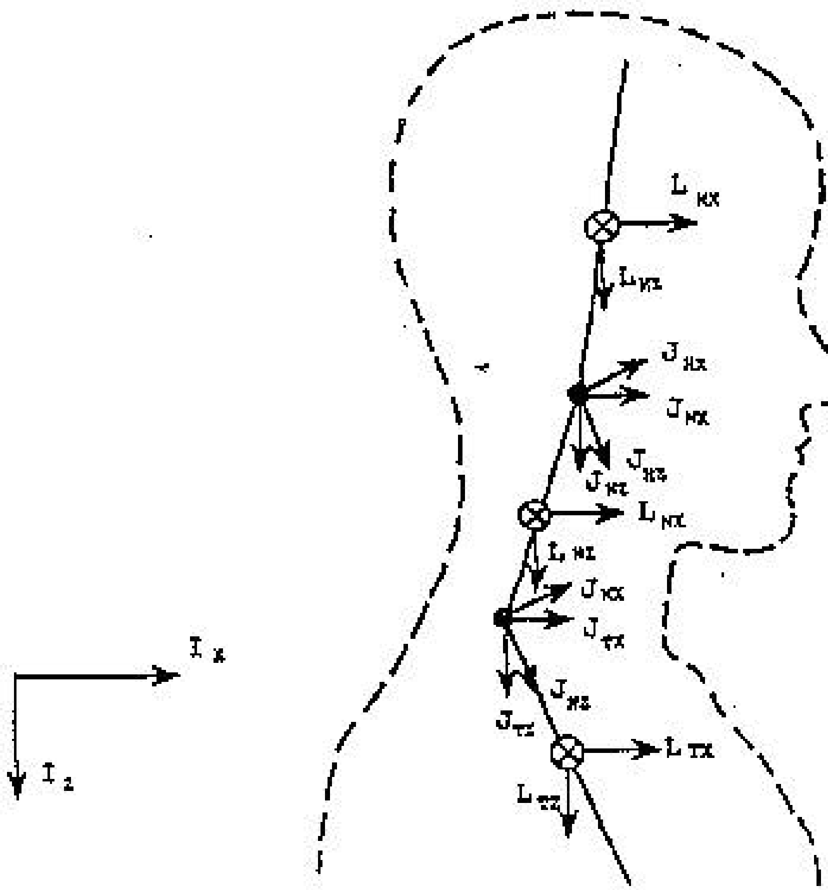

Figure 7 Inertial, Local, and Joint Coordinate Systems of the ATB Head/Neck Model

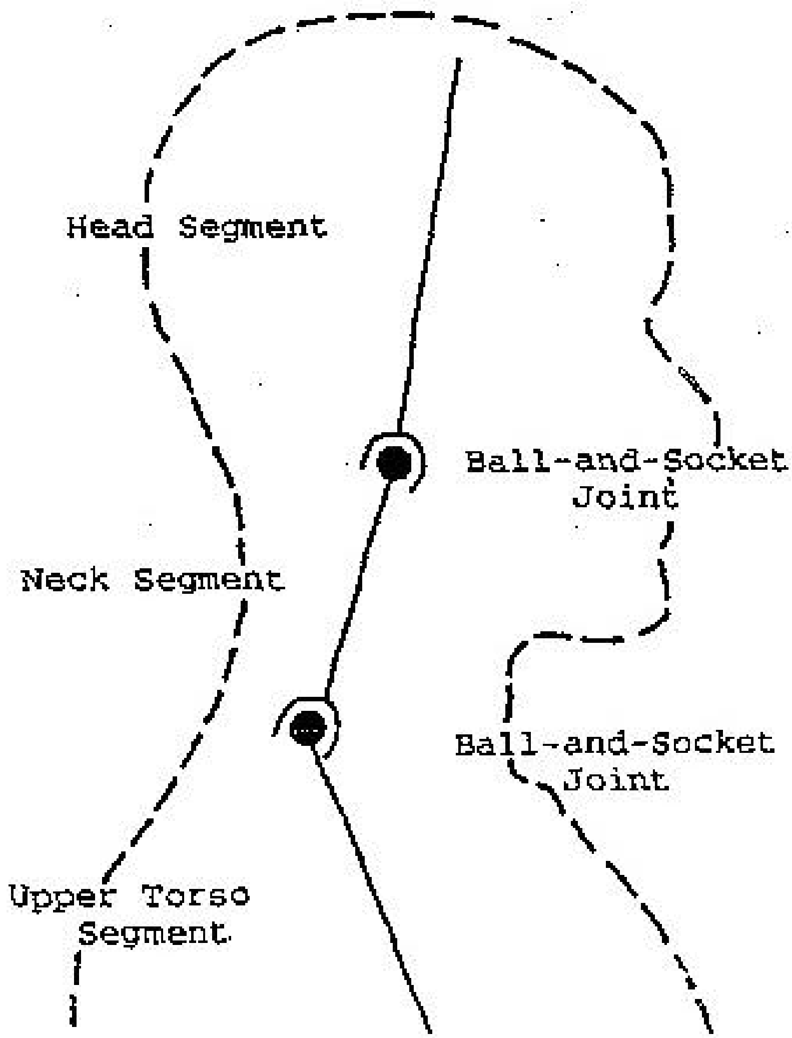

Figure 8 **Three Segment / Two Ball-and-Socket Joint Model of the Head/Neck System**

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corresponds to the anatomical location where the neck articulates with the head (atlanto-occipital joint) and simulates both the turning motion (as if saying “no”) of the head as well as the extension-flexion range of motion. The neck pin ball and socket simulates the cumulative rotational motion (extension-flexion) of the cervical vertebrae and correlates geometrically with the location between the seventh cervical vertebra and the first thoracic vertebra directly inferior to it. The use of ball and socket joints allows for total rotational freedom at the joint location. This is not the actual case *in vivo*. By using the joint data provided by GEBOD, it is possible to account for the resistance provided by the supporting structures (ligaments and muscles) of the cervical spine, and keep the motion of the model representative of the true motion of the head and neck.

Dynaman is the workhorse of the software package. Dynaman uses the GEBOD model data to predict the response of the human model to external environments. Depending on the demands of the user, one can simulate impact forces from any direction and account for variables such as seat-belt restraints or passive restraint systems such as airbags. In this study a typical shoulder belt and lap harness restraint were assumed. The program provides output defining the dynamics of the model and the internal forces in the joints during the time of the impact. These dynamics include linear and angular velocities of the segments, linear and angular acceleration of the segments, linear and angular displacements of both the segments and the joints, and joint forces and bending moments for all desired joints. Although it is possible to extract this data for all the segments and joints in the body model, the focus of this study concerned the head/neck system only.

3.2 Regression Analyses

The output dynamics of the head/neck system from Dynaman was analyzed in order to determine the WISE Index by relating it to the input information (acceleration of collision, height, weight, and sex of the occupant) and to the potential of obtaining a "whiplash" injury. Since it is known that 42 ft-lb of bending moment on the neck is a threshold for causing whiplash-like injuries, the ATB-calculated neck bending moment was considered to be the governing dependent variable involved. To develop the WISE Index, a regression analysis was then performed using MINITAB, with maximum neck bending moment as the dependent (response) variable, and acceleration, height, and weight as the independent (predictor) variables.

Since GEBOD requires sex to be specified along with height and weight, this is a good indication that there are significant physical differences between males and females when considering the dynamic response of the body. Therefore, it was decided to have two separate WISE Indices: one for males and one for females. Since it was not known a priori what form of the variables bending moment depended on most, all of the possibilities were tried and compared. The resulting equation that generated the highest R-Square statistic (where the latter is the square of the correlation coefficient) was the one finally accepted as giving the best regression analysis between the data and the functional form assumed to best describe it.

4.0 Data and Results from ATB/Dynaman

4.1 Male and Female Subjects

Twenty-three male subjects and thirty-six female subjects were evaluated using the aforementioned GEBOD and Dynaman software to allow for the construction of the WISE Index. Of the twenty-three male subjects, five were derived from actual case files of occupants involved in motor vehicle collisions (Dr. Schneck's personal library of case files), and of the thirty-six female subjects, eighteen were derived from actual case files received from Dr. Schneck also. The remaining male and female subjects had user-defined weights and heights, and were subjected to user-defined impact accelerations, as described below. Heights and weights for most of the actual cases were provided along with the acceleration of the collision determined from a forensic analysis of the dynamics of the vehicle. For some of the cases only a general body description was given such as "medium build", so heights and weights were assigned that reflected common average anthropometric values for someone with a "medium" build. These representative heights and weights were 5 foot 8 inches, 170 pounds for "typical" men and 5 foot 5 inches, 130 pounds for "typical" women. In a few cases there was no physical description provided, so for the purpose of this study these subjects were grouped in the medium build category to balance the samples with data specified, most of which were larger than average.

For both the male and female data sets, eighteen of the subjects were user-defined "average" value subjects used to increase the range, accuracy, and applicability of the

WISE Index. Representative small, medium, and large dimensions were chosen for both sexes to simulate what one would be most likely to find in a completely random sample of adults who were healthy, i.e., not grossly obese or grossly underweight. These values are shown in Table 1.

Table 1: "Average" Subjects Evaluated for the WISE Index

Sex	Small Build	Medium Build	Large Build
Male	5 ft 2 in, 120 lb	5 ft 8 in, 170 lb	6 ft 3 in, 185 lb
Female	4 ft 11 in, 95 lb	5 ft 5 in, 130 lb	6 ft 0 in, 150 lb

Each of the six "average" value models were evaluated by GEBOD and ATB at positive and negative accelerations of 1.5g, 3.0g, and 5.0g in the x-direction, representing acceleration vectors indicative of a low-moderate, moderate-high, and very high force rear end and head-on impact. This extrapolation provided eighteen different model runs for each sex, in addition to the actual cases received, and made the data consist of more uniform physical distributions and acceleration profiles, increasing the range and applicability of the WISE Index. A complete list of the data obtained for male and female subjects is shown in Table 2 and Table 3, respectively.

Table 2: Data for Male Subjects

Subject	Acceleration (negative: rear-end; positive: head-on)	Height	Weight
1	-1.5 g	5 ft 2 in	120 lb
2	-3.0 g	5 ft 2 in	120 lb
3	-5.0 g	5 ft 2 in	120 lb
4	-1.5 g	5 ft 8 in	160 lb
5	-3.0 g	5 ft 8 in	160 lb
6	-5.0 g	5 ft 8 in	160 lb
7	-1.5 g	6 ft 3 in	185 lb
8	-3.0 g	6 ft 3 in	185 lb
9	-5.0 g	6 ft 3 in	185 lb
10	1.5 g	5 ft 2 in	120 lb
11	3.0 g	5 ft 2 in	120 lb
12	5.0 g	5 ft 2 in	120 lb
13	1.5 g	5 ft 8 in	160 lb
14	3.0 g	5 ft 8 in	160 lb
15	5.0 g	5 ft 8 in	160 lb
16	1.5 g	6 ft 3 in	185 lb
17	3.0 g	6 ft 3 in	185 lb
18	5.0 g	6 ft 3 in	185 lb
19*	-2.1 g	6 ft 1 in	160 lb
20*	1.1 g	5 ft 10 in	175 lb
21*	-2.7 g	5 ft 10 in	175 lb
22*	-2.6 g	5 ft 11 in	200 lb
23*	-3.0 g	6 ft 5.5 in	255 lb

*** Actual case taken from Dr. Schneck's personal library of case files**

Table 3: Data for Female Subjects

Subject	Acceleration (negative: rear-end; positive: head-on)	Height	Weight
1	-1.5 g	4 ft 11 in	95 lb
2	-3.0 g	4 ft 11 in	95 lb
3	-5.0 g	4 ft 11 in	95 lb
4	-1.5 g	5 ft 5 in	130 lb
5	-3.0 g	5 ft 5 in	130 lb
6	-5.0 g	5 ft 5 in	130 lb
7	-1.5 g	6 ft 0 in	150 lb
8	-3.0 g	6 ft 0 in	150 lb
9	-5.0 g	6 ft 0 in	150 lb
10	1.5 g	4 ft 11 in	95 lb
11	3.0 g	4 ft 11 in	95 lb
12	5.0 g	4 ft 11 in	95 lb
13	1.5 g	5 ft 5 in	130 lb
14	3.0 g	5 ft 5 in	130 lb
15	5.0 g	5 ft 5 in	130 lb
16	1.5 g	6 ft 0 in	150 lb
17	3.0 g	6 ft 0 in	150 lb
18	5.0 g	6 ft 0 in	150 lb
19*	-1.8 g	5 ft 4 in	280 lb
20*	-2.6 g	5 ft 4 in	130 lb
21*	-2.5 g	5 ft 4 in	127 lb
22*	-2.2 g	5 ft 4 in	180 lb
23*	-2.2 g	5 ft 4 in	135 lb
24*	-2.0 g	5 ft 4 in	135 lb

Table 3: Data for Female Subject Continued

Subject	Acceleration (negative: rear-end; positive: head-on)	Height	Weight
25*	-3.0 g	5 ft 6 in	120 lb
26*	0.9 g	5 ft 4 in	135 lb
27*	-0.8 g	5 ft 4 in	128 lb
28*	-2.3 g	5 ft 1 in	111 lb
29*	-2.3 g	5 ft 4 in	135 lb
30*	-2.4 g	5 ft 8 in	170 lb
31*	-2.1 g	5 ft 4 in	130 lb
32*	-2.4 g	5 ft 6.5 in	185 lb
33*	-2.5 g	5 ft 7 in	200 lb
34*	-2.0 g	5 ft 1 in	117 lb
35*	1.0 g	5 ft 0 in	104.5 lb
36*	-4.527 g	5 ft 3 in	230 lb

*** Actual case taken from Dr. Schneck's personal library of case files**

4.2 Results Obtained From ATB Model/Dynaman

Shown in Tables 2 and 3 are the data that was input into GEBOD to determine the physical parameters necessary for Dynaman to calculate the dynamics of the respective subject response as a result of a motor vehicle accident. These physical parameters include the segment weight, center of mass, semiaxis lengths (geometry of the segment ellipsoids), and principal mass moments of inertia for the head, neck, and upper torso of the subject. This information, along with the acceleration profile of the vehicle as determined by forensics experts, enabled Dynaman to calculate linear and angular velocities, accelerations, and displacements of the head, neck, and upper torso, as well as the joint forces and bending moments experienced at the head pin and neck pin.

Although all of the dynamics of the subject response obtained from Dynaman are useful, the most pertinent variables in this study were the bending moment at the neck pin and the angular displacement of the head. The bending moment at the neck pin is important because bending moment is the governing variable in determining whiplash injuries. If, at any point, the bending moment at the neck pin exceeds 42 ft-lb (56.94 Newton-meters), the possibility of whiplash is highly probable. Dynaman provides bending moments at the head pin as well, but the neck pin bending moments are much higher because the moment arm r from there to the location (CG) of the head inertial force is larger. The angular displacement of the head is very important because it shows the magnitude of flexion/extension that the head undergoes as a result of the rotational acceleration caused by the inertial force of the head. Figures 9 and 10 respectively show the “typical“ graphical output obtained from Dynaman, of the bending moment (in foot-pounds) at the neck pin, and the angular displacement (degrees) of the head for male subject #8 (-3.0 g, 6 ft 3 in, 185 lb). This subject experienced a 3-g rear-end collision. Figures 11 and 12 respectively show the bending moment at the neck pin, and the angular displacement of the head for subject #17 which was a male of the same build, experiencing a 3-g head-on collision.

Also shown in Table 4 are the output obtained from Dynaman for the male subjects, including the subject number, acceleration of collision in g's, height in feet, weight in lbs, Ponderal index (to be defined later), maximum bending moment occurring during the collision, in ft-lb, time the maximum bending moment occurs, in ms following impact, and the maximum angular displacement in degrees. Shown in Table 5 are the same tabular output for the female subjects. Examination of these tables indicate that for a certain individual, a rear-end collision would produce a larger bending moment in the neck than a head-on collision of the same acceleration magnitude. Also, by looking at the time in ms at which the maximum bending moment in the neck pin occurs, it appears that time of maximum bending moment in the neck pin decreases with (is inversely proportional to) increasing magnitude of collision acceleration. The head angular displacement (degrees) at which the maximum bending in the neck pin occurs, however, increases with (is directly proportional to) increasing magnitude of collision acceleration.

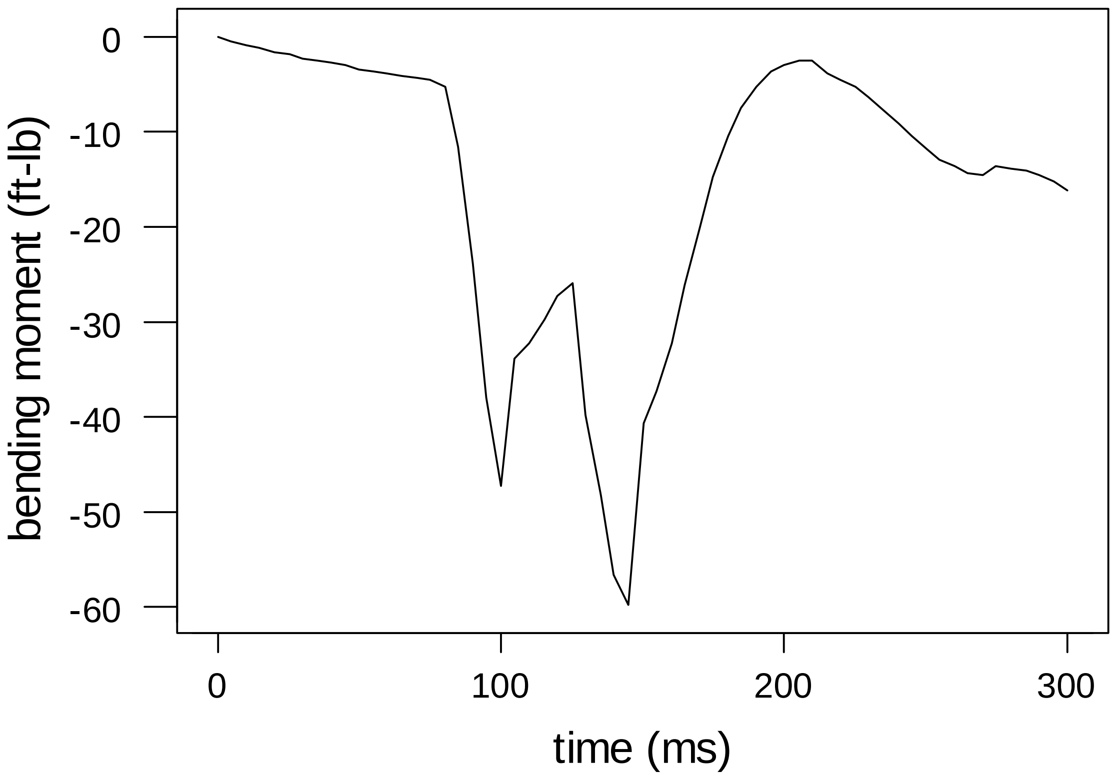

Figure 9 **Plot of Bending Moment at the neck pin vs time for Subject #8 (-3.0 g, 6 ft 3 in, 185 lb)**

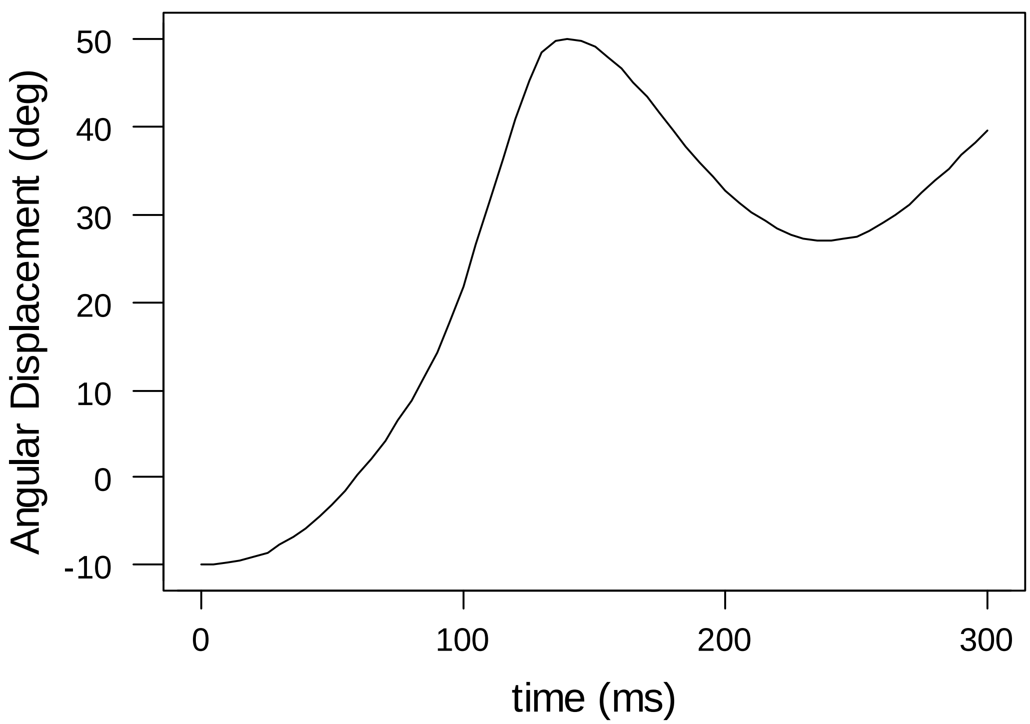

Figure 10 **Plot of Angular Displacement of the head vs time for Subject #8 (-3.0 g, 6 ft 3 in, 185 lb)**

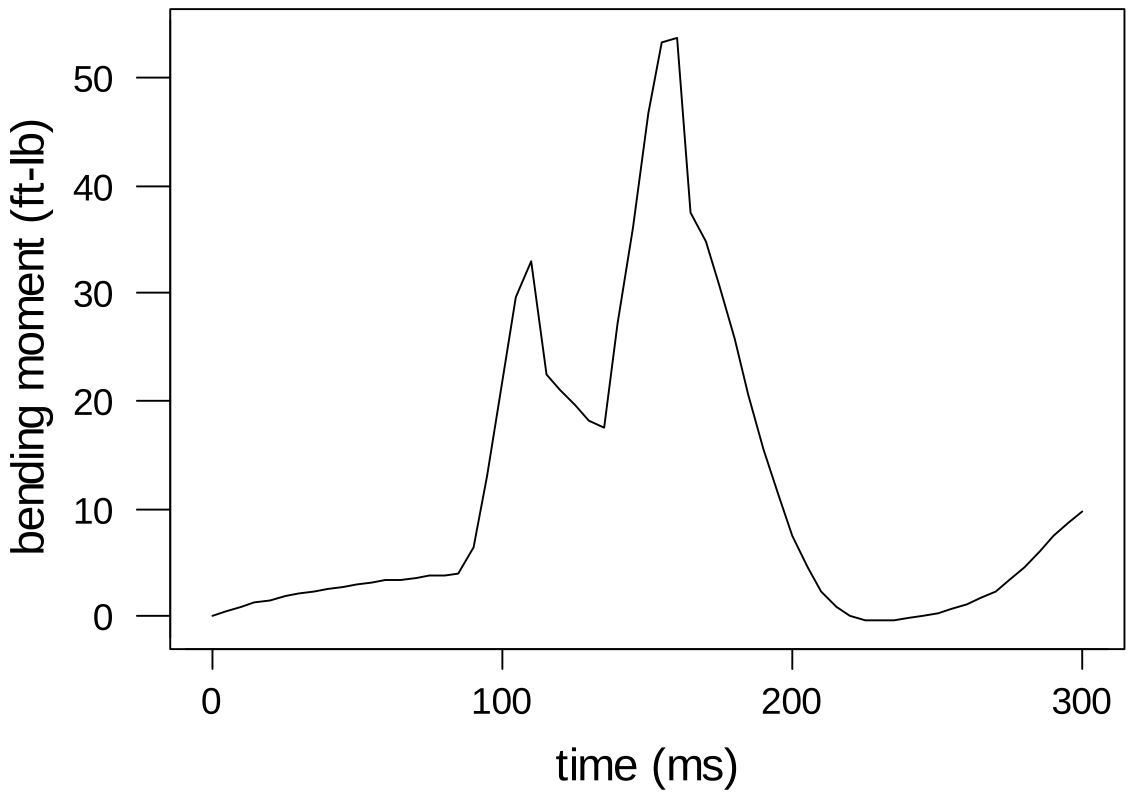

Figure 11 Plot of Bending Moment of the neck pin vs time for Subject #17 (3.0 g, 6 ft 3 in, 185 lb)

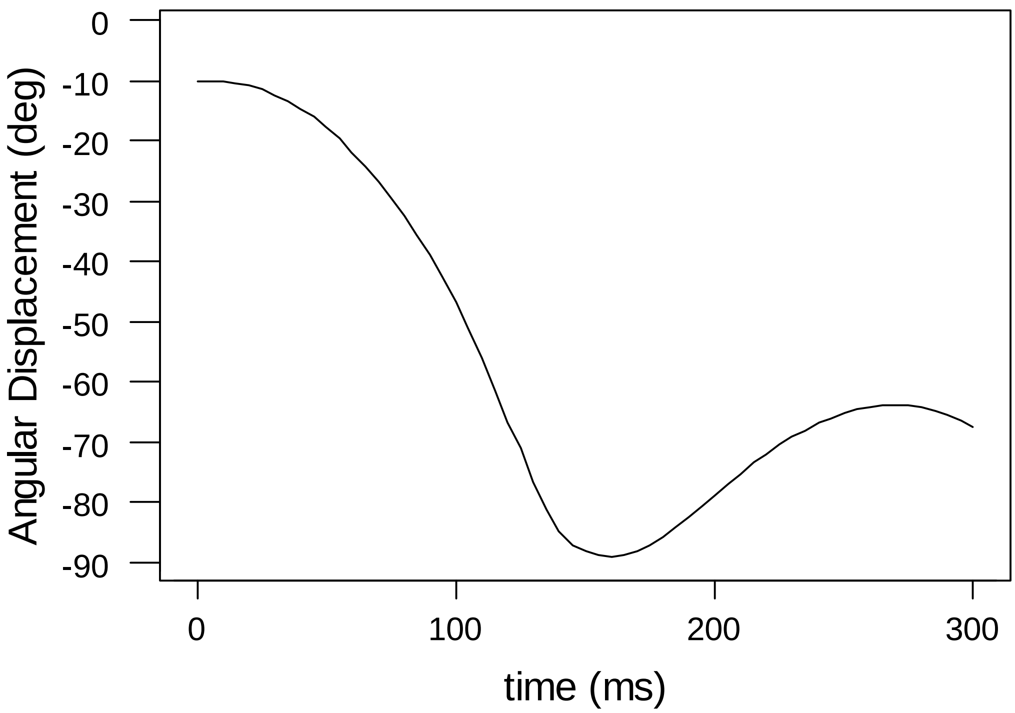

Figure 12 Plot of Angular Displacement of the head vs time for Subject #17 (3.0 g, 6 ft 3 in, 185 lb)

Table 4: Dynaman Output for Males

Subject	Acceleration	Height	Weight	PI	Neck Pin Maximum Bending Moment (ft-lb)	Time (ms) at which M_{max} occurs	Head Angular Displacement (deg) ++
1	-1.5 g	5 ft 2 in	120 lb	24.07	-25.0833	220	43.6
2	-3.0 g	5 ft 2 in	120 lb	24.07	-48.75	155	48.9
3	-5.0 g	5 ft 2 in	120 lb	24.07	-80.5	120	54.0
4	-1.5 g	5 ft 8 in	160 lb	24.15	-27.1667	220	43.8
5	-3.0 g	5 ft 8 in	160 lb	24.15	-53.6667	155	49.3
6	-5.0 g	5 ft 8 in	160 lb	24.15	-87.5	125	54.2
7	-1.5 g	6 ft 3 in	185 lb	22.98	-30.5	220	44.3
8	-3.0 g	6 ft 3 in	185 lb	22.98	-59.8333	160	49.8
9	-5.0 g	6 ft 3 in	185 lb	22.98	-94.1667	125	55.2
10	1.5 g	5 ft 2 in	120 lb	24.06	19.9167	220	-71.7
11	3.0 g	5 ft 2 in	120 lb	24.06	45.4167	155	-77.5
12	5.0 g	5 ft 2 in	120 lb	24.06	75.5	120	-82.3
13	1.5 g	5 ft 8 in	160 lb	24.15	21.5833	220	-72.1
14	3.0 g	5 ft 8 in	160 lb	24.15	48.25	155	-77.9
15	5.0 g	5 ft 8 in	160 lb	24.15	82.0833	125	-82.6
16	1.5 g	6 ft 3 in	185 lb	22.98	24.9167	220	-73.0
17	3.0 g	6 ft 3 in	185 lb	22.98	53.6667	160	-79.0
18	5.0 g	6 ft 3 in	185 lb	22.98	91.6667	125	-84.1
19*	-2.1 g	6 ft 1 in	160 lb	22.50	-42.1667	165	47.0
20*	1.1 g	5 ft 10 in	175 lb	24.17	11.1667	215	-66.1
21*	-2.7 g	5 ft 10 in	175 lb	24.17	-47.25	145	47.8
22*	-2.6 g	5 ft 11 in	200 lb	24.91	-47.75	110	24.2
23*	-3.0 g	6 ft 5.5 in	255 lb	24.71	-62.3333	105	23.8

Table 5: Dynaman Output for Females

Subject	Acceleration	Height	Weight	PI	Neck Pin Maximum Bending Moment (ft-lb)	Time (ms) at which M_{max} occurs	Head Angular Displacement (deg) ++
1	-1.5 g	4 ft 11 in	95 lb	23.39	-20.335	185	42.3
2	-3.0 g	4 ft 11 in	95 lb	23.39	-39.0833	135	47.2
3	-5.0 g	4 ft 11 in	95 lb	23.39	-63.5	105	51.5
4	-1.5 g	5 ft 5 in	130 lb	23.58	-23.9167	185	43.2
5	-3.0 g	5 ft 5 in	130 lb	23.58	-46.8333	135	48.3
6	-5.0 g	5 ft 5 in	130 lb	23.58	-70.6667	105	52.6
7	-1.5 g	6 ft 0 in	150 lb	22.32	-27.9167	185	44.1
8	-3.0 g	6 ft 0 in	150 lb	22.32	-53.6667	135	49.3
9	-5.0 g	6 ft 0 in	150 lb	22.32	-86.6667	110	54.3
10	1.5 g	4 ft 11 in	95 lb	23.39	14.5	220	-71.0
11	3.0 g	4 ft 11 in	95 lb	23.39	36.0833	155	-75.8
12	5.0 g	4 ft 11 in	95 lb	23.39	62.6667	120	-80.6
13	1.5 g	5 ft 5 in	130 lb	23.58	16.9167	220	-70.8
14	3.0 g	5 ft 5 in	130 lb	23.58	40.3333	155	-76.7
15	5.0 g	5 ft 5 in	130 lb	23.58	68.6667	120	-81.6
16	1.5 g	6 ft 0 in	150 lb	22.32	19.5833	220	-71.6
17	3.0 g	6 ft 0 in	150 lb	22.32	45.1667	155	-77.6
18	5.0 g	6 ft 0 in	150 lb	22.32	74.1667	120	-82.4
19*	-1.8 g	5 ft 4 in	280 lb	30.92	-27.4167	170	44.3
20*	-2.6 g	5 ft 4 in	130 lb	23.95	-38.25	140	46.7
21*	-2.5 g	5 ft 4 in	127 lb	23.76	-38.1667	145	46.6
22*	-2.2 g	5 ft 4 in	180 lb	26.69	-33.5	155	45.7
23*	-2.2 g	5 ft 4 in	135 lb	24.25	-33.9167	155	45.7

Table 5: Dynaman Output for Females Continued

Subject	Acceleration	Height	Weight	PI	Neck Pin Maximum Bending Moment (ft-lb)	Time (ms) at which M_{max} occurs	Head Angular Displacement (deg)++
24*	-2.0 g	5 ft 4 in	135 lb	24.25	-30.5833	160	44.9
25*	-3.0 g	5 ft 6 in	120 lb	22.61	-48	135	48.4
26*	0.9 g	5 ft 4 in	135 lb	24.25	9.1667	185	-47
27*	-0.8 g	5 ft 4 in	128 lb	23.82	-10.4167	235	29.2
28*	-2.3 g	5 ft 1 in	111 lb	23.83	-32.6667	150	45.5
29*	-2.3 g	5 ft 4 in	135 lb	24.25	-35.0833	150	45.9
30*	-2.4 g	5 ft 8 in	170 lb	24.64	-40.0833	150	47.0
31*	-2.1 g	5 ft 4 in	130 lb	23.95	-31.4167	155	45.2
32*	-2.4 g	5 ft 6.5 in	185 lb	25.92	-38.6667	150	46.8
33*	-2.5 g	5 ft 7 in	200 lb	26.41	-44.1667	145	47.0
34*	-2.0 g	5 ft 1 in	117 lb	24.26	-28.4167	160	44.4
35*	1.0 g	5 ft 0 in	104.5 lb	23.75	9.75	185	-51.2
36*	-4.527 g	5 ft 3 in	230 lb	29.42	-97.5	95	33.4

5.0 Results and Discussion

In order to develop the WISE Index, the results obtained from the ATB Model/Dynaman, shown in Tables 4 and 5, were used in conjunction with MINITAB to perform a regression analysis. Two separate WISE Indices were formulated (one for each sex), and ATB-calculated maximum bending moment at the neck pin was used as the dependent (response) variable, with FTI-calculated acceleration, the subject's height, and weight being used as the independent (predictor) variables.

What functional form the WISE Index would take was not known a priori, so many feasible possibilities were evaluated to identify which one produced the highest R-square statistic (square of the correlation coefficient). A regression equation for both sexes was first obtained with maximum bending moment at the neck pin depending linearly on acceleration only. The two resulting equations, and the definition of the variables included are shown below.

Linear male regression equation based on acceleration only (R^2 = 0.989)

$$M_{MAX} = -3.17778 \pm 17.141\left|(accel, g)\right| \qquad (5.1)$$
$$1.1\ g \le |accel| \le 5\ g$$

Linear female regression equation based on acceleration only (R^2 = 0.972)

$$M_{MAX} = -3.37191 \pm 14.6233\left|(accel, g)\right| \qquad (5.2)$$
$$0.8\ g \le |accel| \le 5\ g$$

Definition of Variables:

accel: Acceleration of the collision expressed in g's. For a rear end collision the negative sign should be used in front of the absolute value. For a head on collision the positive sign should be used in front of the absolute value.

M_{MAX}: Maximum bending moment, in ft-lbs, the subject experiences as a result of the given acceleration, height, and weight. If M_{MAX} is positive, then the potential injury would result from a forward head rotation such as in a front-end collision. If M_{MAX} is negative then the potential for injury is from rearward head rotation such as in a rear-end collision.

Next, a regression equation for both sexes was obtained with maximum bending moment at the neck pin depending linearly on each of the three separate terms, acceleration, height, and weight. The two resulting equations are shown below.

Linear male regression equation based on acceleration, height, weight (R^2 = 0.991)

$$M_{MAX} = -11.5 \pm 17.1\left|(accel, g)\right| - .0828(weight, lb) + 3.77(height,\ ft) \tag{5.3}$$
$$1.1\ g \leq |accel| \leq 5\ g; \quad 120\ lb \leq weight \leq 260\ lb; \quad 5.1\ ft \leq height \leq 6.5\ ft$$

Linear female regression equation based on acceleration, height and weight (R^2 = 0.974)

$$M_{MAX} = 2.5 \pm 14.4\left|(accel, g)\right| - .0696(weight, lb) + 0.67(height,\ ft) \tag{5.4}$$
$$0.8\ g \leq |accel| \leq 5\ g; \quad 90\ lb \leq weight \leq 280\ lb; \quad 4.9\ ft \leq height \leq 6.0\ ft$$

Since bending moment at the neck pin depended primarily on acceleration (as illustrated by the GSI and HIC), many different forms of acceleration were tried first in the regression analysis, while keeping height and weight as linear terms. For example:

the square root of acceleration, the cube root of acceleration, acceleration squared, acceleration cubed, the exponential of acceleration, and the natural log of acceleration, were all tried. All of these forms of acceleration, however, produced significantly lower R-Square statistics than did the equation using just a linear acceleration term.

Since a linear acceleration term produced the highest R-square statistic, the bending moment at the neck pin was presumed to depend linearly on acceleration, but the proper form of height and weight were still to be assessed. Thus, different forms each of height and weight were tried separately, keeping the remaining two variables, i.e., acceleration and either height, or weight, respectively, as linear terms. Height and weight were separately squared, cubed, taken to the square root, and taken to the cube root, but the R-square statistic in all cases was slightly lower than for the linear case. Next to be examined was the hypothesis that the maximum bending moment at the neck pin might be affected more by a measure of the height to weight ratio, rather than by the two variables separately. The weight was converted to mass, and using the height to mass ratio the R-square statistic improved slightly, yielding the new governing equations shown below.

Male regression equation using height to weight ratio (R^2 = 0.994)

$$M_{MAX} = -11.3 \pm 17.1\left|\left(accel, g\right)\right| + 6.92\left(\frac{height, ft}{mass, slugs}\right) \tag{5.5}$$

$$1.1\ g \le |accel| \le 5\ g; \quad 0.8 \le \frac{height}{mass} \le 1.4$$

Female regression equation using height to weight ratio (R^2 = 0.976)

$$M_{MAX} = -17.0 \pm 14.4\left|\left(accel, g\right)\right| + 10.2\left(\frac{height, ft}{mass, slugs}\right) \tag{5.6}$$

$$0.8\ g \le |accel| \le 5\ g; \quad 0.6 \le \frac{height}{mass} \le 1.7$$

Although the above regression equations are satisfactory to develop the WISE Index, further research revealed that a better and more “traditional” functional form that relates height and weight is the Ponderal Index (Kreighbaum and Barthels, 1996). The Ponderal Index is used as a measure of anthropometric build and is defined as shown below.

$$PI = 10^3 \times \frac{\sqrt[3]{Weight, kg}}{Height, cm}$$

For the purposes of this study, weight was converted to pounds and height was converted to feet, so the Ponderal Index then became:

$$PI = 25.2089 \times \frac{\sqrt[3]{Weight, lb}}{Height, ft}$$

Using the Ponderal Index instead of the height to mass ratio did not change the R-square statistic at all, but the Ponderal Index is more desirable because it is a pre-established and well-accepted measure of anthropometric build. The final regression equations used to determine the WISE Index are shown below. The regression plots for these equations are shown in Figures 13 and 14, for males and females, respectively. The regression plots show that the results from the regression analysis correspond with and confirm the “rule of thumb” guidelines proposed by FTI (See Section 2.3).

Male regression equation using the Ponderal Index (R^2 = 0.994)

$$M_{MAX} = 11.1 \pm 17.1\left|(accel, g)\right| - 0.6(PI) \tag{5.7}$$

$$1.1\, g \le a \le 5\, g; \quad 22.4 \le PI \le 25.0$$

Female regression equation using the Ponderal Index (R^2 = 0.976)

$$M_{MAX} = 26.4 \pm 14.4\left|(accel, g)\right| - 1.23(PI) \tag{5.8}$$

$$0.8\, g \le a \le 5\, g; \quad 22.3 \le PI \le 31.0$$

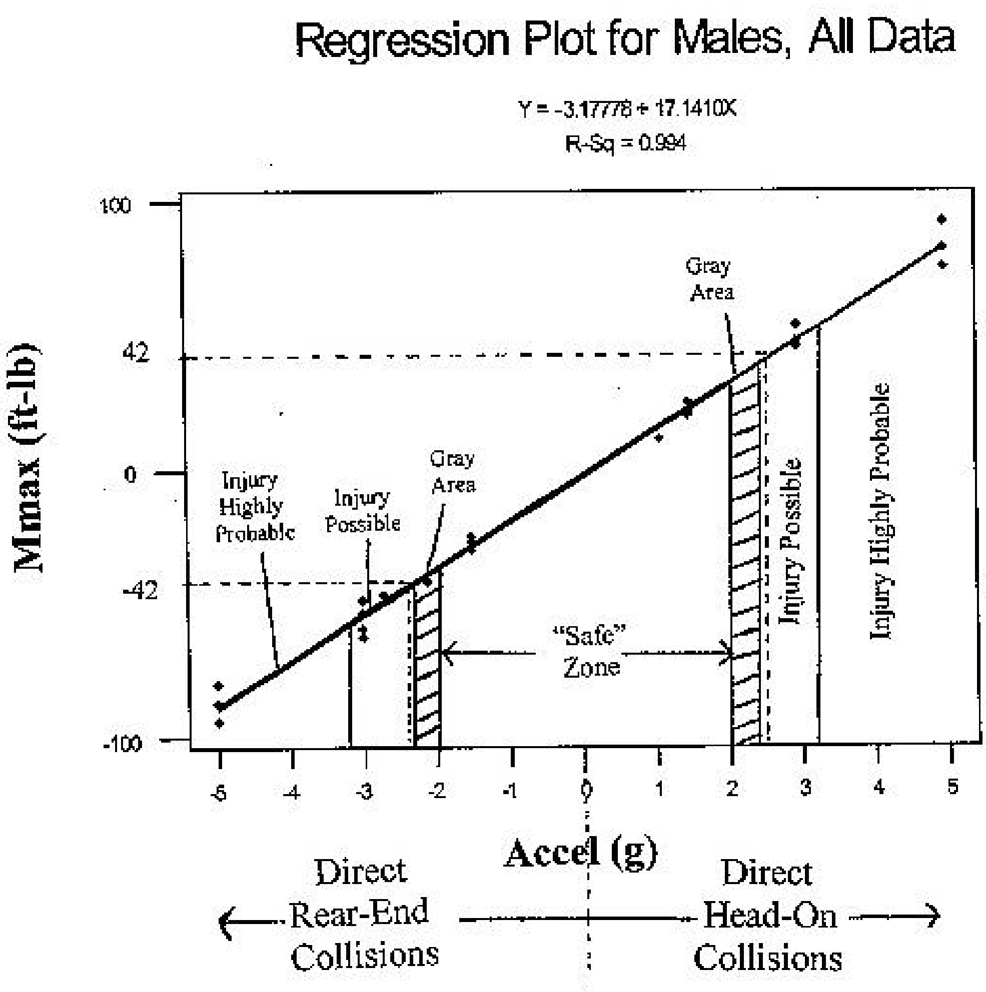

Figure 12 **Regression Plot for Males: Maximum Bending Moment at the neck pin vs. collision acceleration**

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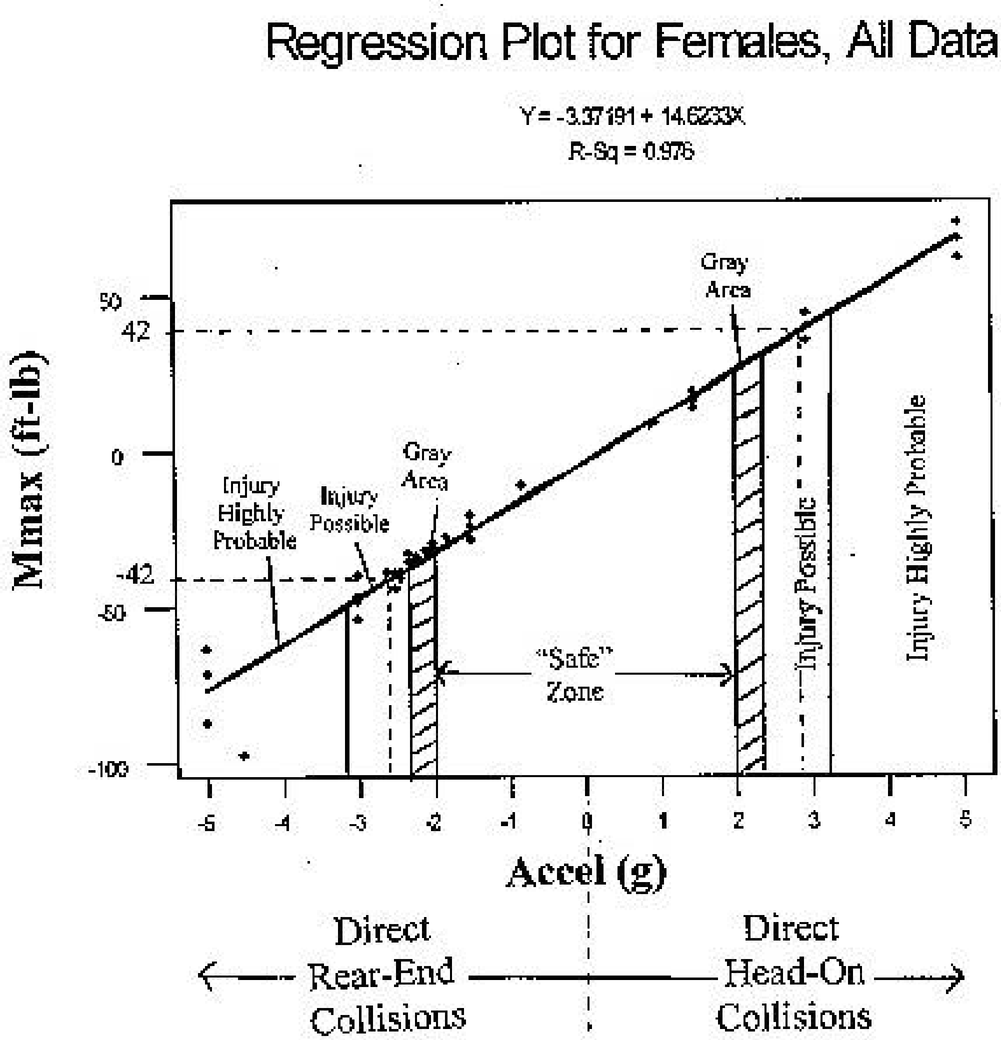

Figure 13 **Regression Plot for Females: Maximum Bending Moment at the neck pin vs. collision acceleration**

The previous equations are used by inputting the impact acceleration of the collision, and the Ponderal Index corresponding to the occupant(s) of the vehicle. The resulting absolute value of the maximum bending moment at the neck pin would then be either greater than or less then 42 ft-lb. If it is less than 42 ft-lb, it is considered to be below the threshold bending moment for whiplash injury and, therefore, relatively "safe". If it is greater than 42 ft-lb, it is considered to be potentially "dangerous". To make it easier to understand what the equations are telling the user, they can be nondimensionalized by dividing both of them by the threshold moment $M_{threshold}$, above which whiplash is highly probable. A new variable, ξ, was thus defined which is the ratio of $M_{max}/M_{threshold}$. If $|\xi| < 1$, the bending moment in the neck stays below the critical value and the possibility of injury will depend on how close to unity the ratio is. If $|\xi| > 1$, the bending moment is equal to, or exceeds the critical value so injury is highly probable, and a measure of intensity of injury is to note by how much the ratio exceeds unity. If ξ is positive, the potential injury would result from forward head rotation such as in a front-end collision. If ξ is negative, then the potential for injury is from rearward head rotation such as in a rear-end collision. The final WISE Indices are shown as follows.

Male WISE Index (R^2 = 0.993)

$$x = 0.2643 \pm 0.4071\left|\left(accel, g\right)\right| - 0.01428\left(PI\right) \qquad (5.9)$$

$$1.1\text{ g} \le accel \le 5\ g; \quad 22.4 \le PI \le 25.0$$

$$\text{Not for children; Most accurate for ages 18 - 55.}$$

Female WISE Index (R^2 = 0.978)

$$x = 0.6214 \pm 0.3429\left|\left(accel, g\right)\right| - 0.02929\left(PI\right) \qquad (5.10)$$

$$0.8\text{ g} \le accel \le 5\ g; \quad 22.3 \le PI \le 31.0$$

$$\text{Not for children; Most accurate for ages 18 - 55.}$$

Acceleration: Use the negative sign if it is a rear-end collision and the positive sign if it is a head-on collision.

ξ: A negative value means that potential injury results from backward head rotation, as in a rear-end collision. A positive value means that potential injury results from forward head rotation, as in a head-on collision.

$|\xi| < 1$ = "Safe"

$|\xi| > 1$ = "Dangerous"

Another good reason for choosing the Ponderal Index as a way to include the height and weight variables is that it is much easier to determine how the sex of the individual affects the likelihood of a whiplash injury. A male and a female are much more likely to have similar Ponderal Indices, than to have the exact same height and weight. Therefore, for any given (but the same) arbitrary acceleration and Ponderal Index, equations (5.9) and (5.10) yield a value of $|\xi|$ that is always greater for males than for females. This implies that a male of similar build, subjected to the same accident conditions as a female, is more likely to receive a whiplash injury in an automobile accident than is a female.

Since the Ponderal Index has a relatively narrow range for a correspondingly wide range of human heights and weights, it produces a nearly uniform term to account for height and weight of an individual, as opposed to the widely fluctuating term such as height or weight by itself. Even for the large range of heights and weights included in this study (4.9 ft to 6.5 ft for heights, and 95 lb to 280 lb for weights), the Ponderal Index only ranges from 22.4 to 25.0 for males, and 22.3 to 31.0 for females. Figures 15 and 16 show for males and females, respectively, the plot of the WISE Index regression equation for both the highest Ponderal Index and the lowest Ponderal Index for each sex. All of the data falls in the "envelope" of these two lines, so use of the Ponderal Index effectively collapses the data into a small region.

It appears from Tables 4 and 5 that for the same individual and a constant magnitude of collision acceleration, the maximum bending moment at the neck pin is apparently greater in rear-end collisions, than it is in head-on collisions. Thus, again from equations (5.9) and (5.10), one observes that for any arbitrary acceleration and

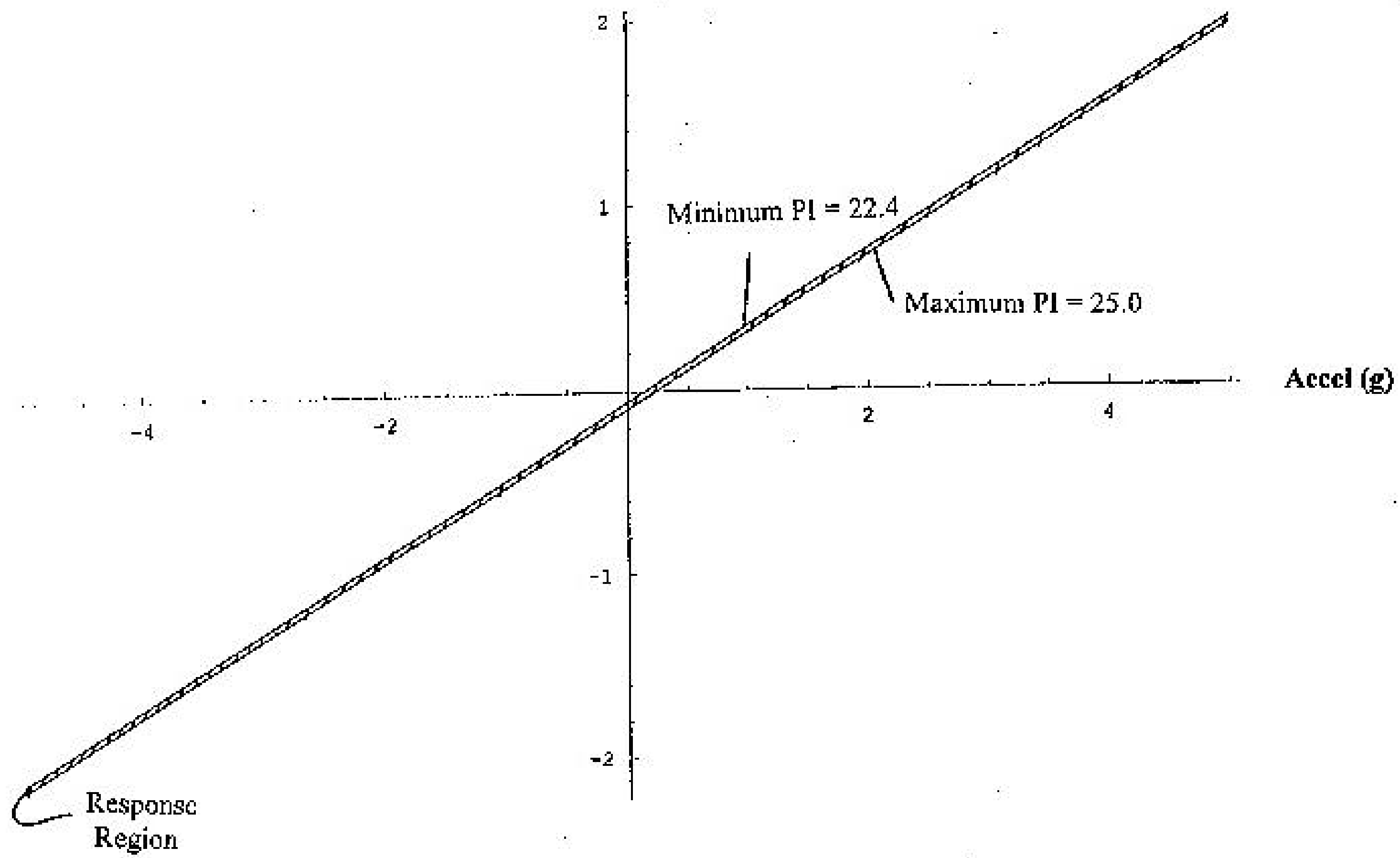

Figure 15 **Plot of ξ vs. accel for males using the largest and smallest Ponderal Index**

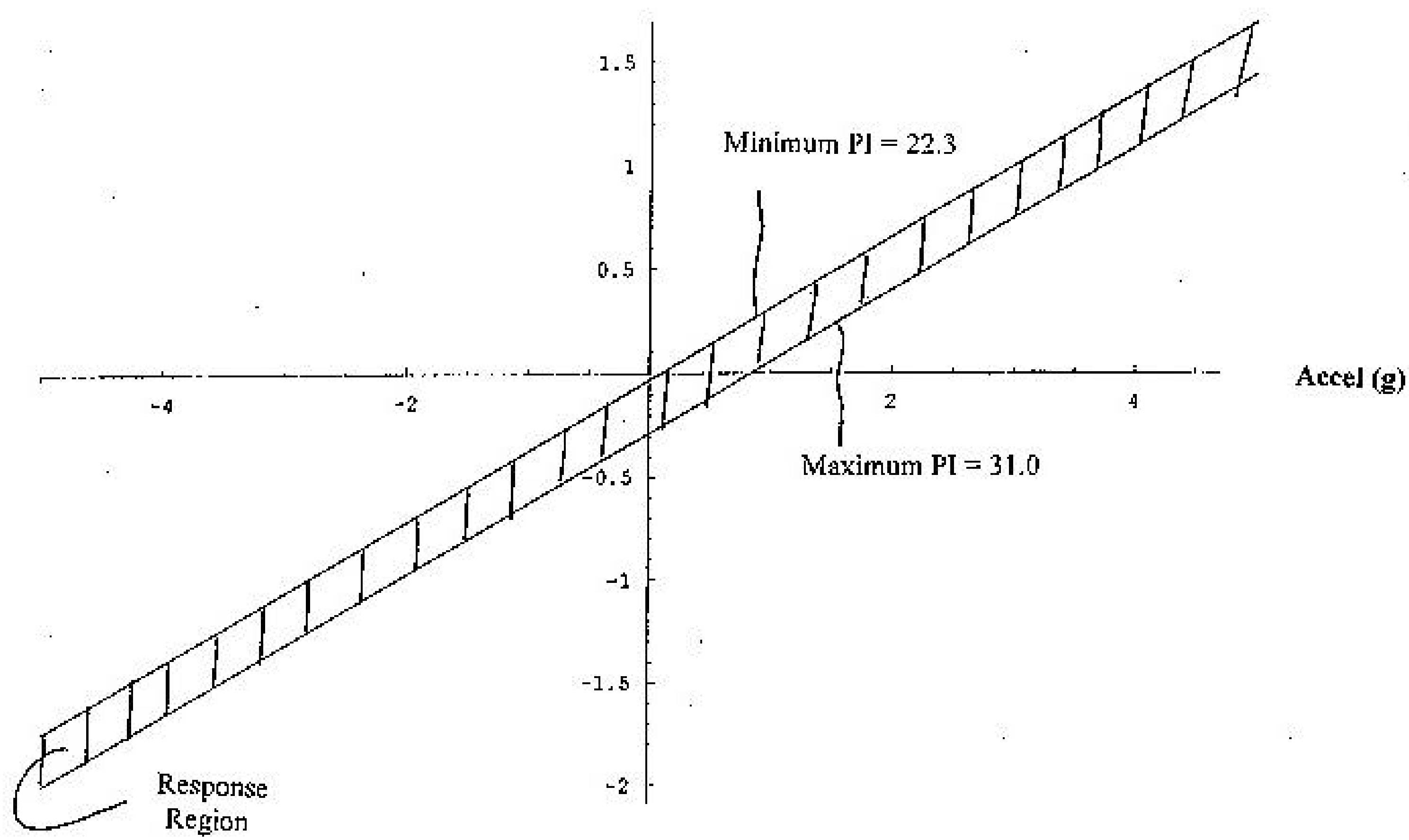

Figure 16 Plot of ξ vs. accel for females using the largest and smallest Ponderal Index

Ponderal Index, and for both sexes, the likelihood of getting whiplash is greatest in a rear-end collision. Moreover, inspection of Tables 4 and 5 further reveals that the time in ms after the collision at which the maximum bending moment occurs decreases with increasing magnitude of collision acceleration. A good illustration of this is shown in Figure 17. To quantify this fact, a regression analysis was carried out to see what kind of correlation the time of maximum bending moment has with the magnitude of collision acceleration. A linear analysis with the magnitude of acceleration yielded an R-square statistic of 0.683, showing that there is some direct correlation, but the dependency is not nearly as high as that between maximum bending moment and collision acceleration. When other functional forms of acceleration were used as before, it was found that the time in ms, depends best on the cubed root of the magnitude of collision acceleration. These two regression equations are shown below.

Male regression equation for time vs. acceleration (R^2 = 0.760)

$$time, ms = 390 - 162\sqrt[3]{\left|accel, g\right|} \tag{5.11}$$

Female regression equation for time vs. acceleration (R^2=0.798)

$$time, ms = 344 - 139\sqrt[3]{\left|accel, g\right|} \tag{5.12}$$

Inspection of Tables 4 and 5 also shows that the angular displacement of the head increases with increasing collision acceleration. This inherently makes sense because the faster the body is accelerated/decelerated, the more drastically the body will respond. To examine quantitatively how angular displacement of the head correlates with collision

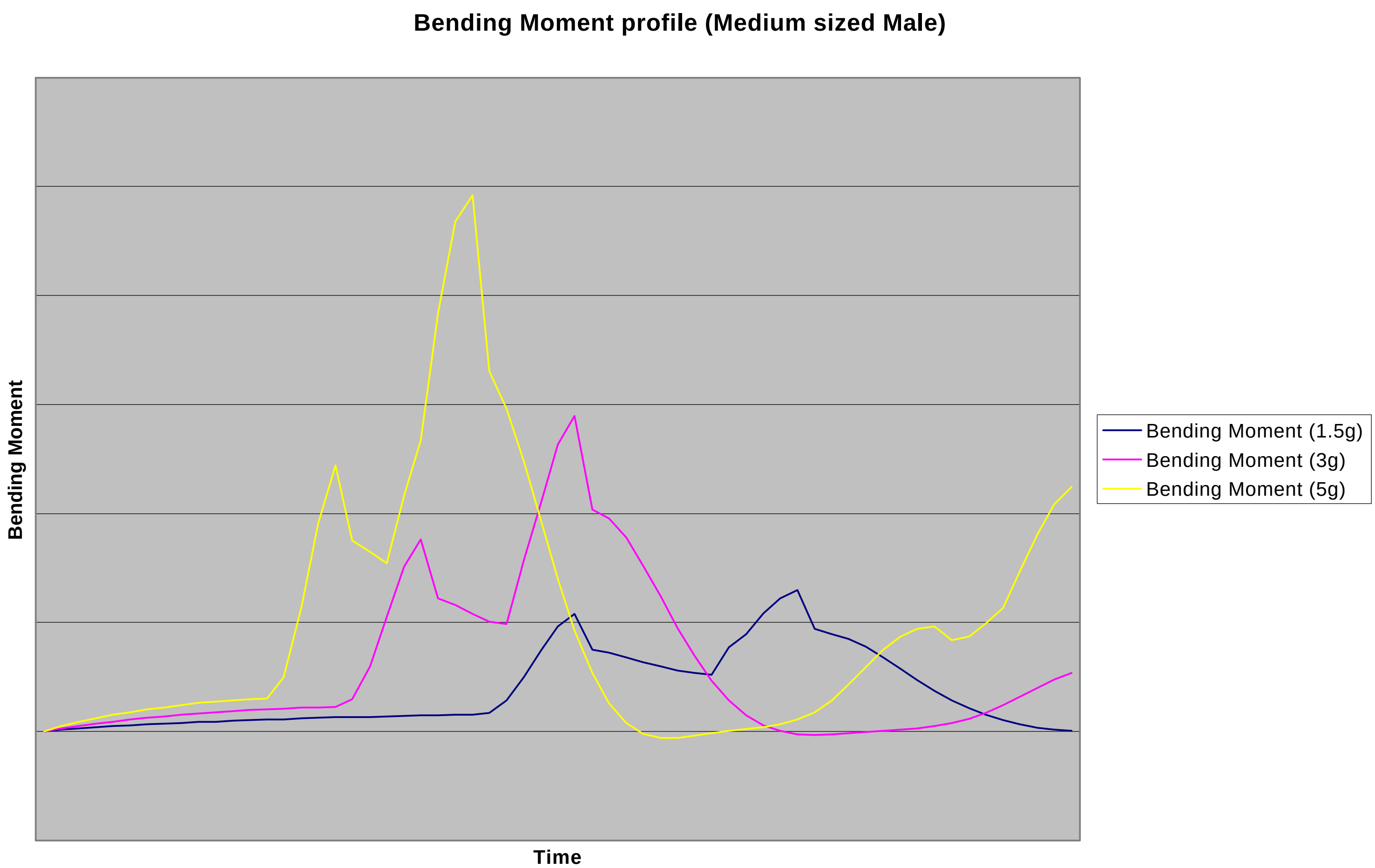

Figure 17 **Bending Moment vs time profile for an individual at 1.5 g, 3.0 g, and 5.0 g**

acceleration, a regression analysis was performed on these two variables. All the functional forms of acceleration were tried, with acceleration to the cube root giving the highest R-square statistic. In the regression equations shown below, a negative sign is used in front of the acceleration term if it is a rear-end collision, and a positive sign is used in front of the acceleration term if it is a head-on collision.

Male regression equation for angular displacement vs. acceleration (R^2 = 0.980)

$$J\,,\deg = -16.1 \pm 42.8\ \sqrt[3]{\left|accel, g\right|} \tag{5.13}$$

Female regression equation for angular displacement vs. acceleration (R^2 = 0.986)

$$J\,,\deg = -12.8 \pm 42.7\ \sqrt[3]{\left|accel, g\right|} \tag{5.14}$$

6.0 Conclusions and Future Work

Since whiplash has always been a very subjective diagnosis, a way to quantify the potential for whiplash injury in an automobile accident would help engineers, doctors, and lawyers, among others. To quantify whiplash injury the Whiplash Injury Severity Estimator (WISE Index) was developed to predict the likelihood of a healthy occupant in a motor vehicle accident receiving whiplash-type injuries as well as to measure the potential intensity of such injury.

To arrive at the WISE Index actual cases of occupants involved in motor vehicle accidents were used in conjunction with a computer model of human body dynamics. For each of these cases, height, weight, sex of the occupant, and kinematic information revealing the acceleration of the collision were obtained. Some extrapolated cases were also added to the data set representing the low and high ranges of height, weight, and collision accelerations, in order to increase the range of the WISE Index. The data given for each case was used in conjunction with GEBOD and Dynaman to obtain a dynamic analysis of the individual in the accident (specifically the head/neck region). Results of the dynamic analysis were then compared with known data regarding the ability of the human neck to tolerate loading without consequence.

Since it is known that 42 ft-lb of bending moment on the neck pin is the threshold moment above which injury is highly probable in a normal adult subject, neck pin bending moment is the governing variable in determining the probability of a whiplash-type injury. The variables that play the most important role in determining neck pin bending moment are acceleration of the collision, height, and weight of the individual. MINITAB was used to perform a regression analysis for each sex using neck pin bending moment as the dependent (response) variable, and collision acceleration, height, and weight as the independent (predictor) variables. Different forms of the predictor variables were tried and the corresponding forms with the highest R-square statistic were used in the WISE Index. It turned out that a linear term for acceleration and the Ponderal Index as a measure of anthropometric build were the best choices. The final nondimensionalized equations are as shown in equations (5.9) and (5.10).

Examination of these equations showed that normal males are more likely than females of a similar anthropometric build to receive a whiplash-type injury under the same accelerations. Also, a rear-end collision is more likely to cause whiplash than a head-on collision of the same acceleration implying that extension is more traumatic to the neck than flexion. The time following impact, at which the maximum bending moment in the neck pin is reached decreases with increasing collision acceleration, and the maximum angular displacement of the head increases with increasing collision acceleration. This makes sense conceptually because the faster your body is accelerated/decelerated, the faster the neck will respond and the more inertia the head will have to carry it a farther distance. Regression equations were developed to quantify both the time of maximum bending moment vs. acceleration and the angular displacement vs. acceleration (Equations (5.11), (5.12), (5.13), (5.14)).

This study was intended as a modest step into fully quantifying the potential for whiplash injury, and took the already existing injury severity indices a step further by accounting for the subject's anthropometric build. Although inclusion of anthropometric build led to a higher R-square statistic, the question may arise if the increase in the R-square statistic is statistically significant. The R-square statistic is a measure of how well a regression line can be fit to the data set that is used, not a measure of how exact the functional form of the governing equation is. Therefore, even if the R-square statistic did not improve at all when anthropometric build was added, the equation with anthropometric build would be the equation of choice because it accounts for more factors and still yields just as high an R-square statistic.

Since the data points lie in a very narrow band around the regression line, Figures 15 and 16 illustrate that the regression analysis gives the same slope of ξ for all values of Ponderal Index examined in this study. In reality, the slope of the regression line may, indeed, be a function, itself, of Ponderal Index (i.e., the WISE Index may actually be nonlinear in acceleration and Ponderal Index). This would mean that the apparent decrease in WISE Index with increasing Ponderal Index for a given acceleration may not be an entirely accurate trend. Certainly, no particular trend was apparent when the data was examined individually, plotting the WISE Index as a function of acceleration with Ponderal Index acting separately as a parameter. Thus, one would have to conclude at this point that a wider range of Ponderal Index needs to be examined to unmask the dependence of the WISE Index slope on Ponderal Index. That is, examination of a wider range of Ponderal Index would hopefully reveal any nonlinearities in the variation of the WISE Index with Ponderal Index, if there is any consistent such variation.

To improve on the WISE index even further, it would be desirable to make the index specific to the person involved in the accident rather than just a “typical” person of the same height and weight. This could be partially accomplished if one could measure the mechanical properties and viscoelastic behavior of the tissue involved. Currently, there is research into the use of ultrasound techniques to measure these mechanical properties non-invasively. When such measurements become possible, terms can be added to the WISE Index which could account for the specific physiologic condition of the occupant’s neck, thus taking into account such factors as age, previous medical history, and pregnancy.

Another way to improve this study prior to the regression analysis, would be to incorporate the Finite Element Method into the analysis of the human body response to impact, thus overcoming the limitations of the rigid body model. The present study accounted for direct rear-end or head-on collisions, but future work could account as well for lateral accelerations/decelerations. Other variables that can play a role in the body’s reaction to impact are dependent on the type of automobile involved. This would include the use of a supplemental restraint system such as an airbag. If no airbag were used and the occupant collided with the inside of the car, this factor, too, could be taken into account when measuring the subject impact response. Also, the properties of the occupant’s seat could be analyzed. Factors such as head rests, cushions, and seat position angle could all be involved in body response.

One of the most important elements that require consideration is the effect of active muscle response mechanisms such as reflexes. It has been shown that tensing the neck muscles prior to impact reduces the potential for injury (Mertz and Patrick, 1993).

Other active reactions such as preventing head motion by clasping the hands behind it and involuntary reflexes could have substantial effects on the motion of the head and thus the potential for injury. It is anticipated that some of these factors will be addressed in the future.

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Appendix

Sample ATB Model as Generated by GEBOD III for a Large Female

WEIGHT 150.0 LB.
STANDING HEIGHT 72.00 IN.
-COMPUTED BODY DIMENSIONS

0	WEIGHT	150.0 LB.
1	STANDING HEIGHT	72.00 IN.
2	SHOULDER HEIGHT	59.01 IN.
3	ARMPIT HEIGHT	54.11 IN.
4	WAIST HEIGHT	45.08 IN.
5	SEATED HEIGHT	37.11 IN.
6	HEAD LENGTH	7.518 IN.
7	HEAD BREADTH	5.807 IN.
8	HEAD TO CHIN HEIGHT	9.159 IN.
9	NECK CIRCUMFERENCE	13.81 IN.
10	SHOULDER BREADTH	15.02 IN.
11	CHEST DEPTH	9.592 IN.
12	CHEST BREADTH	11.47 IN.
13	WAIST DEPTH	6.862 IN.
14	WAIST BREADTH	10.07 IN.
15	BUTTOCK DEPTH	8.592 IN.
16	HIP BREADTH, STANDING	14.48 IN.
17	SHOULDER TO ELBOW LENGTH	13.79 IN.
18	FOREARM-HAND LENGTH	18.45 IN.
19	BICEPS CIRCUMFERENCE	10.09 IN.
20	ELBOW CIRCUMFERENCE	11.46 IN.
21	FOREARM CIRCUMFERENCE	9.592 IN.
22	WRIST CIRCUMFERENCE	6.256 IN.
23	KNEE HEIGHT, SEATED	21.49 IN.
24	THIGH CIRCUMFERENCE	6.256 IN.
25	UPPER LEG CIRCUMFERENCE	18.86 IN.
26	KNEE CIRCUMFERENCE	15.13 IN.
27	CALF CIRCUMFERENCE	14.00 IN.
28	ANKLE CIRCUMFERENCE	8.788 IN.
29	ANKLE HEIGHT, OUTSIDE	3.007 IN.
30	FOOT BREADTH	3.707 IN.
31	FOOT LENGTH	10.50 IN.

0WEIGHT CORRECTION FACTOR = 0.937

1largefemale
0CRASH VICTIM PARAMETERS (3-D)

CARDS B.2

SEGMENT			WEIGHT	PRINCIPAL MOMENT OF INERTIA (LB-SEC**2-IN)			SEGMENT CONTACT SEMIAXIS (IN)			ELLIPSOID CENTER(IN)			PRINCIPAL AXES (DEG)		
I	SYM	PLOT	(LB.)	X	Y	Z	X	Y	Z	X	Y	Z	YAW	PITCH	ROLL
1	LT	1	18.788	0.7222	0.3994	0.7931	4.296	7.241	4.399	0.286	0.000	1.408	0.00	0.00	0.00
2	CT	2	1.350	-0.0276	-0.0764	-0.0764	3.431	5.034	3.592	-0.229	0.000	0.248	0.00	0.00	0.00
3	UT	3	39.737	2.8318	2.2729	1.5809	4.796	5.736	6.271	1.199	0.000	0.000	0.00	0.00	0.00
4	N	4	1.991	0.0137	0.0168	0.0156	2.199	2.199	3.013	0.628	0.000	1.488	0.00	0.00	0.00
5	H	5	8.069	0.1523	0.1705	0.1153	3.759	2.904	5.679	0.752	0.000	0.000	0.00	0.00	0.00
6	RUL	6	24.591	0.1714	0.1690	0.0084	3.299	3.299	11.547	0.000	-0.330	0.899	0.00	0.00	0.00
7	RLL	7	7.258	0.4330	0.4334	0.0470	2.228	2.228	9.941	0.000	-1.432	0.488	0.00	0.00	0.00
8	RF	8	1.823	0.0318	0.0309	0.0067	1.504	1.853	5.248	-1.868	-0.751	1.050	0.00	0.00	0.00
9	LUL	9	24.591	0.1714	0.1690	0.0084	3.299	3.299	11.547	0.000	0.330	0.899	0.00	0.00	0.00
10	LLL	A	7.258	0.4330	0.4334	0.0470	2.228	2.228	9.941	0.000	1.432	0.488	0.00	0.00	0.00
11	LF	B	1.823	0.0318	0.0309	0.0067	1.504	1.853	5.248	-1.868	0.751	1.050	0.00	0.00	0.00
12	RUA	C	3.524	0.1070	0.1087	0.0166	1.605	1.605	6.894	0.000	-0.033	0.000	0.00	0.00	0.00
13	RLA	D	2.836	0.1714	0.1690	0.0084	1.527	1.527	9.224	0.436	0.524	0.000	0.00	0.00	0.00
14	LUA	E	3.524	0.1070	0.1087	0.0166	1.605	1.605	6.894	0.000	0.033	0.000	0.00	0.00	0.00
15	LLA	F	2.836	0.1714	0.1690	0.0084	1.527	1.527	9.224	0.436	-0.524	0.000	0.00	0.00	0.00

0

JOINT					LOCATION(IN)-SEG(JNT)			LOCATION(IN)-SEG(J+1)			JOINT AXIS(DEG)-SEG(JNT)			JOINT AXIS(DEG)-SEG(J+1)		
J	SYM	PLOT	JNT	PIN	X	Y	Z	X	Y	Z	YAW	PITCH	ROLL	YAW	PITCH	ROLL
1	P	M	1	0	-1.60	0.00	-0.97	-2.07	0.00	2.69	0.00	0.00	0.00	0.00	5.00	0.00
2	W	N	2	0	-1.18	0.00	-0.99	-0.05	0.00	7.59	0.00	0.00	0.00	0.00	5.00	0.00
3	NP	O	3	0	0.03	0.00	-7.85	-0.45	0.00	1.79	0.00	0.00	0.00	0.00	10.00	0.00
4	HP	P	4	0	0.34	0.00	-2.57	0.22	0.00	2.65	0.00	0.00	0.00	0.00	10.00	0.00
5	RH	Q	1	0	1.25	2.11	2.55	0.99	-1.98	-7.19	0.00	0.00	0.00	0.00	-45.00	0.00
6	RK	R	6	1	0.41	0.80	11.18	0.44	-0.57	-7.21	0.00	0.00	0.00	0.00	60.00	0.00
7	RA	S	7	0	0.48	-0.77	10.38	-1.87	-0.25	-1.71	0.00	90.00	0.00	0.00	10.00	0.00
8	LH	T	1	0	1.25	-2.11	2.55	0.99	1.98	-7.19	0.00	0.00	0.00	0.00	-45.00	0.00
9	LK	U	9	1	0.41	-0.80	11.18	0.44	0.57	-7.21	0.00	0.00	0.00	0.00	60.00	0.00
10	LA	V	10	0	0.48	0.77	10.38	-1.87	0.25	-1.71	0.00	90.00	0.00	0.00	10.00	0.00
11	RS	W	3	0	-1.39	5.72	-4.33	-1.03	-0.03	-5.87	0.00	0.00	0.00	0.00	-4.10	0.00
12	RE	X	12	1	0.13	-0.04	5.92	0.52	0.26	-6.97	0.00	0.00	0.00	0.00	-70.00	0.00
13	LS	Y	3	0	-1.39	-5.72	-4.33	-1.03	0.03	-5.87	0.00	0.00	0.00	0.00	-4.10	0.00
14	LE	Z	14	1	0.13	0.04	5.92	0.52	-0.26	-6.97	0.00	0.00	0.00	0.00	-70.00	0.00

1 JOINT TORQUE CHARACTERISTICS

CARDS B.4

FLEXURAL SPRING CHARACTERISTICS

TORSIONAL SPRING CHARACTERISTICS

	Flexural: SPRING COEF. (IN.LB./DEG**J)					Torsional: SPRING COEF. (IN.LB./DEG**J)				
				ENERGY	JOINT				ENERGY	JOINT
JOINT	LINEAR (J=1)	QUADRATIC (J=2)	CUBIC (J=3)	DISSIPATION COEF.	STOP (DEG)	LINEAR (J=1)	QUADRATIC (J=2)	CUBIC (J=3)	DISSIPATION COEF.	STOP (DEG)
1 P	0.000	10.000	0.000	0.700	20.000	0.000	10.000	0.000	0.700	5.000
2 W	0.000	10.000	0.000	0.700	20.000	0.000	10.000	0.000	0.700	35.000
3 NP	0.000	5.000	0.000	0.700	25.000	0.000	10.000	0.000	0.700	35.000
4 HP	0.000	5.000	0.000	0.700	25.000	0.000	10.000	0.000	0.700	35.000
5 RH	0.000	10.000	0.000	0.700	70.000	0.000	0.800	0.000	0.700	40.000
6 RK	0.000	1.800	0.000	0.700	60.000	0.000	0.000	0.000	0.000	0.000
7 RA	0.000	7.000	0.000	0.700	35.000	0.000	10.000	0.000	0.700	26.000
8 LH	0.000	10.000	0.000	0.700	70.000	0.000	0.800	0.000	0.700	40.000
9 LK	0.000	1.800	0.000	0.700	60.000	0.000	0.000	0.000	0.000	0.000
10 LA	0.000	7.000	0.000	0.700	35.000	0.000	10.00	0.000	0.700	26.000
11 RS	0.000	10.000	0.000	0.700	122.500	0.000	10.000	0.000	0.700	65.000
12 RE	0.000	1.800	0.000	0.700	70.000	0.000	0.000	0.000	0.000	0.000
13 LS	0.000	10.000	0.000	0.700	122.500	0.000	10.000	0.000	0.700	65.000
14 LE	0.000	1.800	0.000	0.700	70.000	0.000	0.000	0.000	0.0	0.000

CARDS B.5
JOINT VISCOUS CHARACTERISTICS AND LOCK-UNLOCK CONDITIONS

JOINT	VISCOUS COEFFICIENT (IN.LB.SEC./DEG)	COULOMB FRICTION COEF. (IN.LB.)	FULL FRICTION ANGULAR VELOCITY (DEG/SEC.)	MAX TORQUE FOR A LOCKED JOINT (IN.LB.)	MIN TORQUE FOR UNLOCKED JOINT (IN.LB)	MIN. ANG. VELOCITY FOR UNLOCKED JOINT (RAD/SEC.)	IMPULSE RESTITUTION COEFFICIENT
1 P	0.100	0.00	30.00	0.00	0.00	0.00	0.00
2 W	0.100	0.00	30.00	0.00	0.00	0.00	0.00
3 NP	0.100	0.00	30.00	0.00	0.00	0.00	0.00
4 HP	0.100	0.00	30.00	0.00	0.00	0.00	0.00
5 RH	0.100	0.00	30.00	0.00	0.00	0.00	0.00
6 RK	0.100	0.00	30.00	0.00	0.00	0.00	0.00
7 RA	0.100	0.00	30.00	0.00	0.00	0.00	0.00
8 LH	0.100	0.00	30.00	0.00	0.00	0.00	0.00
9 LK	0.100	0.00	30.00	0.00	0.00	0.00	0.00
10 LA	0.100	0.00	30.00	0.00	0.00	0.00	0.00
11 RS	0.100	0.00	30.00	0.00	0.00	0.00	0.00
12 RE	0.100	0.00	30.00	0.00	0.00	0.00	0.00
13 LS	0.100	0.00	30.00	0.00	0.00	0.00	0.00
14 LE	0.100	0.00	30.00	0.00	0.00	0.00	0.00

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SEGMENT INTEGRATION CONVERGENCE TEST INPUT
CARDS B.6

SEGMENT NO.	SYM	ANGULAR VELOCITIES (RAD/SEC.) MAG. TEST	ABS. ERROR	REL. ERROR	LINEAR VELOCITIES (IN./SEC.) MAG. TEST	ABS. ERROR	REL. ERROR	ANGULAR ACCELERATIONS (RAD/SEC.**2) MAG. TEST	ABS. ERROR	REL. ERROR	LINEAR ACCELERATIONS (IN./SEC.**2) MAG. TEST	ABS. ERROR	REL. ERROR
1	LT	0.010	0.010	0.0100	0.010	0.010	0.0100	0.100	0.100	0.1000	0.100	0.100	0.0100
2	CT	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
3	UT	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
4	N	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
5	H	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
6	RUL	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
7	RLL	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
8	RF	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
9	LUL	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
10	LLL	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
11	LF	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
12	RUA	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
13	RLA	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
14	LUA	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000
15	LLA	0.010	0.010	0.0100	0.000	0.000	0.0000	0.100	0.100	0.1000	0.000	0.000	0.0000

VITA

Kevin M. Moorhouse

EDUCATION

Masters of Science in Engineering Mechanics. Expected graduation May 1998.
VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY, Blacksburg, Virginia.
August 1996 to present.

- Concentration in Biomechanics.
- Thesis – Determination of a Whiplash Injury Severity Estimator (WISE Index) for Occupant involved in a Motor Vehicle Accident
- GPA of 3.72 on a 4.0 scale.

B.S. in Engineering Science & Mechanics. Graduated With Honors, May 1996.
VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY, Blacksburg, Virginia.
August 1992 to May 1996.

- Concentration in Biomechanics.
- Senior Project -- Determination of an Accident Severity Index for Occupants involved in a Motor Vehicle Accident.
- GPA of 3.4 on a 4.0 scale.

MCDONOUGH HIGH SCHOOL, Pomfret, Maryland.
Graduated June 1992 with High Honors and a GPA of 4.1 on a 4.0 scale.

HONORS

- Full Tuition Scholarship, Assistantship Award, and Presidential Fellowship Award, Virginia Tech Department of ESM, 1998-completion of PhD Degree
- Full Tuition Scholarship and Assistantship Award, Virginia Tech Department of ESM, 1996-8
- Passed Engineer in Training (EIT) Test in the Top Tenth Percentile, Spring 1996
- Engineering Scholarship, Virginia Tech Department of ESM, 1993-4
- Member of the University Honors Program
- University Dean's List, 1992-1996
- National Merit Commended Student
- Principal Trumpet, Maryland All-State Band

EXPERIENCE

Teaching Assistant, VIRGINIA TECH DEPARTMENT OF ESM, Blacksburg, Virginia. 08/96-05/98.

- Graded papers and held office hours for Computational Methods (a numerical methods course), And Mechanics of Deformable Bodies.

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Clinical Internship in Biomedical Engineering, ROANOKE MEMORIAL HOSPITAL, 05/97-08/97.

- Rotated through all hospital departments and activities observing the main types of surgery, and learning how the various equipment worked.

Legal Assistant, PATTON BOGGS, L.L.P, Washington D.C. Summer 1995-1996.

- Preparing documents for trial, quality controlling documents, researching for attorneys.
- Wrote *Engineering Analysis of Resilient Pipe Hangers*, to be consulted in trial.

ESM Lab Assistant, VIRGINIA TECH DEPARTMENT OF ESM, Blacksburg Virginia. Spring ’95.

- Researched and simulated the possibility of whiplash in an automobile accident using Dynaman.