

## Bertil Aldman Memorial Lecture

**International Research Contributions on the Biomechanics of Injury:  
Honoring the past and looking towards the future**

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It is, indeed, a privilege and great honor to deliver the distinguished Bertil Aldman Lecture. I am humbled and grateful. Dr. Aldman's doctoral thesis from the Karolinska Institute in 1962, *Biodynamic Studies on Impact Protection*, is still an excellent read that covers a wide range of research areas applicable even in today's environment. Dr. Aldman, also a medical doctor, in 1983 co-edited a book based on a course conducted in Amalfi, Italy, *Biomechanics of Impact Trauma*. It would not be a stretch for me to state that I had not known the depth of this interesting field at that time. My experience is, therefore, limited. Others, including some in this audience, are eloquent speakers, and some are more knowledgeable on the subject than I. It is with this acknowledgment and trepidation that I deliver this lecture.

I chose the title, International Research Contributions on the Biomechanics of Injury, to parallel the acronym of one of the most prestigious conferences in the world, and I will try to briefly recount its path and offer a peek into the future, based on my experiences as a perpetual knowledge seeker. Any and all mistakes are mine, and any credits belong to all from whom I continue to learn.

To the experienced folks in the audience, this text may be too simple. To the younger folks, I hope it serves as a history lesson and encourages future contributions to the betterment of our society to enhance safety and improve quality of life.

## **INTRODUCTION**

Injury Biomechanics is the Basic Science for injuries. It has wide-ranging applications in multiple areas of human life. Injury Biomechanics in Engineering has been applied to:

- replicate field-identified injuries,
- determine their mechanisms,
- quantify injury criteria,
- estimate human tolerances,
- evaluate injury mitigating systems,
- promulgate governmental and local standards, and
- advance human safety in various types of environments.

Injury Biomechanics in Medicine has been applied to:

- conceptualize, design, develop, and execute clinical trials for treatment,
- improve quality of life, and
- reduce the societal burden of injuries (and some diseases) around the world.

Biomechanics researchers from around the world have contributed to, influenced, and continue to play an important and active role in these areas. I will attempt to outline some of the important international research contributions and their worldwide applications. As the audience is mainly

from Engineering, my primary focus will be on this area. Quantifications of the effects of mechanical loading are one of the basic goals of biomechanics research.

## **A RECOUNT**

### HEAD

#### ***Head Injury Quantifications via Accelerations***

In 1880, Otto Messerer, a German surgeon, determined the breaking strength of human cadaver skulls and other bones. Different types of skull fractures were associated with different force magnitudes.

In the 1940s, Injury Biomechanics studies done in the US laid the foundation for the quantification of skull fractures and brain injuries. It was observed that patients sustaining skull fractures had associated brain injuries, especially loss of consciousness (80%). The hypothesis was, therefore, that preventing skull fractures mitigates brain injuries. A neurosurgeon and an engineer from Wayne State University (WSU), E. S. Gurdjian and Herbert Lissner, designed impact tests using different surrogates. They used embalmed human cadavers, now commonly termed post mortem human subjects/surrogates (PMHS). The origins of the PMHS acronym are from 1990s' European literature. Their experimental design was basic compared to current standards, which are more complex.

Forehead impact tests (initially sponsored by the Ford Motor Company) were conducted using 6 isolated embalmed PMHS heads and 2 whole-body PMHS. The specimens were dropped onto flat, rigid surfaces. Translation accelerations were recorded at the back of the head. Peak accelerations from 3 isolated and 2 intact whole-body PMHS and mean acceleration data from the other isolated head were used to develop the original head injury tolerance curve. Its maximum time duration was approximately 6 msec. The final response, known as the Wayne State University Tolerance Curve (WSTC) was obtained from animal, volunteer, PMHS, and medical data. While the tolerance curve is applicable to the adult, age-dependent tolerance curves were not derived.

In 1961, Charles Gadd from General Motors (GM) suggested a time-dependent approach to resolve the difficulty in applying the WSTC to complex wave forms. The resulting "weighted impulse criterion" was used in the early version of the US Federal Motor Vehicle Safety Standards (FMVSS) by limiting the Gadd Severity Index (GSI) to 1000 with no time limits. The index  $n$  was set at 2.5.

$$GSI = \int_0^t a^n(t) dt$$

In 1968, researchers from the Vienna Institute of Technology proposed the Vienna Institute index, JTI. It was based on a spring-dashpot model and WSTC.

In 1970, the Chrysler Corporation researchers proposed the effective displacement index, EDI. It was a better fit to WSTC than GSI. In 1971, W. F. S. Fan reported another model, termed the Revised Brain Model, which was also based on a single degree-of-freedom system, like the JTI and EDI, and based on WSTC. The Mean Strain Criterion (MSC) developed by Richard Stalnaker and James McElhaney was based on a two degrees-of-freedom model and not on WSTC.

Also in 1971, John Versace from the Ford Motor Company formulated the head injury criterion, HIC, using the acceleration-time curve. The HIC is perhaps the most frequently cited equation in biomechanics and safety engineering literatures. It is also cited in clinical papers.

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

Chosen to maximize HIC,  $t_2$  and  $t_1$  are arbitrary final and initial times in the acceleration pulse during which the HIC attains a peak value, and  $a(t)$  is the resultant acceleration at the center of gravity of the head (in g). The HIC represents the rate of change of energy stored in the system when modulated by the square root of average acceleration, and, thus, this formulation has a physical meaning.

In 1972, the US National Highway Traffic Safety Administration (NHTSA) chose a value of 1000 without duration limits as the threshold in FMVSS. In 1986, the interval over which HIC was computed was limited to 36 msec,  $HIC_{36}$ , with the same 1000 threshold.

In 1985, the GM and Ford researchers recommended to the International Standards Organization that they adopt the  $HIC_{15}$ , i.e., limiting the computations to within 15 ms. Although no human injury probability curves (HIPCs) were originally derived for the HIC, they developed HIPCs for skull fracture and brain injuries based on an analysis of PMHS and volunteer data. The HIC values of 1450 and 700 were associated with 50% and 5% risks of skull fracture.

Skull fracture studies at different anatomical sites using PMHS continued after the promulgation of HIC at WSU, the Medical College of Wisconsin (MCW), and other laboratories. In the 1970s and 1980s, researchers in France conducted drop tests using unembalmed PMHS following the French Standard (drop height of 2.5 m). Studies in Germany also ensued during this time frame to determine side impact tolerance using unembalmed PMHS. Again, acceleration was the primary biomechanical metric for head injury quantification. The current FMVSS 208 uses the  $HIC_{15}$  of 700, while the European regulations ECE R94 use the  $HIC_{36}$  of 1000 and limit the peak resultant acceleration to less than 80 g.

No presentation would be complete without discussions of other postulations, identifications, documentations, and quantifications of brain injuries attributed to other mechanisms and at local tissue levels by international researchers in medical and engineering disciplines.

In 1943, a physicist from England, A. H. S. Holbourn, hypothesized brain injuries to be a byproduct of angular motions. Between 1960 and 1980, researchers at the US National Institutes of Health advanced unique brain injury studies. Neurosurgeons from Pakistan and the US, Ayub Ommaya and Thomas Gennarelli, a bioengineer from the US, Lawrence Thibault, and neuropathologists from Scotland, Hume Adams and David Graham, are credited with having conducted conclusive first generation experiments, especially inertial angular accelerations. Their seminal research has fostered many candidate metrics for the quantifications of brain injuries.

Metrics based on the interpretations of linear and angular head motions include the Generalized Acceleration Model for Brain Injury Threshold (GAMBIT). This was developed in 1986 by the Canadian bioengineer James Newman, who delivered this lecture a few years ago. Since 2000, the Head Impact Power (HIP), Principal Component Score (PCS) or Head Impact Telemetry severity profile (HITsp), Skull

Fracture Criterion (SFC), Rotational Injury Criterion (RIC), and Brain rotational Injury Criterion (BrIC) have been proposed for different types of environments. The Rotational Velocity Change Index (RVCI) was proposed by a Japanese research team in 2015.

#### ***Head Injury Assessments with Helmet Use***

The British Standards Institute (BSI) set the early performance standards, BS 2826 1869-1952, for Racing Motorcyclists. In 1964 it included race car drivers. The tolerance was 5000 lbf. In 1959, George Snively, a medical doctor from Sacramento, CA, published the Snively Memorial Foundation-North American Standard: The limit was 400 g, equivalent to 4800 lbf. In 1970, Snell upgraded its standard to 300 g. In the 2015 Snell Standard, the limit is 243 to 275 g for motorcycle helmets, depending on the size of one of the 6 headforms.

In 1961, the American Standards Association set the limit to 400 g. It also had a 200 g 2 ms dwell time and a 150 g 4 ms dwell time specification. In 1969, the National Operating Committee on Standards for Athletic Equipment (NOCSAE) was formed for football helmets. The initial limit was a GSI of 1000. In 1982, it was revised to 1500. It was then lowered to 1200. In 1973, Harold Fenner and A. F. James, medical doctors, published a football helmet standard, at their personal expense. The first Canadian Standard for Ice Hockey was published in 1975.

In 1970, the Canadian Standard D230 used the 5000 lbf limit for vehicle helmets. In 1974, the US FMVSS 218 used the peak acceleration of 400 g, and set duration limits of 200 g for 2 ms, and 150 g for 4 ms. In 1979, the American National Standards Institute (ANSI) Z90.1 used 300 g as the limit for vehicle helmets; the earlier one was in 1971, corrected for an error in 1973. In 1980, the Snell standard adopted the 300 g peak limit. The 1997 Cost-327 report in Europe headed by Dietmar Otte from Germany contributed to the ECE Regulations 22-05 (originally established in 1972). In 2002, the revision 4 specified HIC of 2400 and peak acceleration of 275 g. In 2004, the World Motor Sport Council developed the FIA 8860-2004 standard. The updated standard has the limits of 275 g and 2400 as HIC for the chin guard and 300g and 3500 to 4000 HIC for the shell testing area.

## **CHEST**

### ***Frontal Impact***

The initial studies of Col. Stapp (1951-1970) and Eiband (1959), as well as industry researchers in the US, contributed to the chest injury criterion, which was specified as peak spinal acceleration not to exceed 60 g sustained for 3 ms or longer in the FMVSS 208. This was the only metric that could be measured in the Hybrid II dummy used for injury assessments. Researchers from the University of Heidelberg (UH), Laboratory of Biomechanics and Physiology, Peugeot-Renault in France, Laboratoire des Chocs et de Biomécanique Organisme National de Sécurité Routière (Onser) in France, and the US conducted approximately 100 PMHS tests. The Canadian and European road safety regulations adopted the deflection metric because the Hybrid III dummy better mimicked the human response than did the previous generation dummy.

In the 1990-1997 period, the MCW researchers (Frank Pintar and myself) and University of Virginia (UVA) researchers Walter Pilkey and Jeff Crandall (past President of IRCOBI) conducted whole body PMHS tests using the chestband, a new device for measuring external chest deflections, developed by Rolf Eppinger in 1989. Researchers at NHTSA further pooled and analyzed 63 tests from MCW, UVA, and UH to determine the role of deflections for chest injuries.

In 1998, the NHTSA researchers analyzed 71 PMHS tests from these 3 academic institutions and derived the chest trauma index, expressed as a linear combination of deflection and acceleration, which was proposed for the advanced airbag rule. In 1999, Prasad from Ford analyzed a reduced set of 68 tests and found that the deflection was a stronger predictor of chest injuries than acceleration. The FMVSS 208, applicable for vehicle model 2007+ years, used the 63 mm deflection and 60 g 3 msec clip acceleration limits, while the ECE R94 regulations used the V\*C of 1.0 and deflection limit of 50 mm. Since the promulgation of the current version of FMVSS 208, additional studies have been undertaken by the EU Thorax project. The NHTSA is sponsoring studies done by the MCW and UVA researchers to determine the thoracic injury biomechanics of occupants representing small size female occupants.

#### ***Nearside Pure Lateral Impact***

The NHTSA researchers analyzed a large test series in the 1980s to obtain an index representative of thoracic trauma from nearside impacts. Initial tests were performed at the UH, and the thoracic trauma index (TTI) was developed. The index was a side impact injury criterion in the FMVSS 214 standards promulgated in 1990, until the 2008 update.

Later tests involved whole body PMHS tests conducted under sponsorships from the industry and the US Centers for Disease Control and Prevention (CDC) to WSU and from NHTSA to MCW. In the 1990-1994 period, 17 WSU tests were used to obtain injury criteria from the Average Spine Acceleration (ASA), TTI, viscous criteria ( $VC_{max}$ ), and chest deflection. In a parallel venture, the NHTSA conducted tests at Ohio State University and MCW in the 1990s. Using 42 whole body PMHS tests conducted by MCW researchers, HIPCs for PMHS and injury assessment risk curves (IARCs) for the ES-2re dummy were developed. The metrics included the TTI, ASA, deflection, and others. The 2008 upgrade of FMVSS 214 is based on these analyses and IARCs for the ES2-re dummy. The FMVSS 214 specifies 44 mm for the peak deflection, while the ECE R95 specifies 42 mm.

#### ***Nearside Oblique Lateral Impact***

In the 2008-2013 period, the MCW researchers conducted sled tests to determine injury thresholds for oblique nearside impacts, which were based on US field data indicating that an angled loading vector is more susceptible to injury than the pure lateral vector. Considerations of different end conditions, load wall angulations representing different impact vectors, and velocities showed that deflection tolerances expressed in the form of HIPCs for oblique impacts are lower than those from pure lateral impacts.

In the 2015-2017 period, MCW researchers reported findings from whole body PMHS seated in full-scale vehicles equipped with side airbags and compared them to responses from the WorldSID device. The tests paralleled the US moving deformable barrier and pole tests with occupants seated in front and left rear positions. While the chest deflection contours were comparable between

surrogates, it was concluded that multi-point sensing may be required to better assess safety in nearside impacts.

### ***Far-side Impact***

Studies between 2000-2007 by a consortium from Australia, Sweden, and the US (MCW, George Washington University and Virginia Tech) have provided data on deflections and accelerations from PMHS tests and have assisted the industry in designing protections for the far-side occupant. The NHTSA is currently pursuing studies with MCW researchers using experiments and finite element modeling.

## **EXTREMITIES**

Studies conducted in the US during 1966-1976 showed that force and time could be used to protect the hip-femur complex. The FMVSS 208 uses the peak force limit of 10 kN, while the ECE R94 specifies both: 9.07 kN at zero time, dropping linearly to 7.58 kN for time duration up to 10 msec, and remaining at the same level thereafter. Studies conducted after 2000 by researchers from MCW and the University of Michigan Transportation Research Institute have proposed other limits for the knee-thigh-hip complex, applicable to restrained occupants. The tibia injury criterion is only adopted in ECE R94 and is based on GM research.

For the foot-ankle-lower tibial complex, efforts were advanced by MCW and UVA researchers under NHTSA and other sponsorships to determine tolerances in different loading modes and assist the design and evaluation of the THOR dummy lower leg. The HIPCs from axial loading published by MCW researchers in 1996 were used in the North Atlantic Treaty Organization (NATO) recommendations in 2007 for injury assessment. This criterion is currently used in the military Live Fire Testing and Evaluations in the US and elsewhere. As these tests were originally intended for automotive applications, under the DoD sponsorship, MCW researchers have produced newer HIPCs for underbody blast loadings and have also evaluated the military dummy leg, Mil-LX, for its injury predicting capability. Some of these results began to appear in literature in 2014.

## **SPINE**

### ***Early Studies***

In 1874, the President of the Obstetrical Society in Edinburgh, Matthews Duncan, reported the neck tensile strength of 4 fresh, newly born fetuses and a 2-week-old child cadaver. The failure force increased from the newborn to the 2-week-old specimen, representing the resistance change with increasing age.

In 1976, over a century later, 4 intact PMHS, 2.5 to 11 years of age, were tested in Germany, and this initial effort was followed by additional sled tests by increasing the sample size to 15. Studies in the US consisted of distracting functional units from an 8-hour old PMHS in 1993.

In 2005, Chinese researchers reported data from 10 PMHS tests, 2 to 12 years of age; the mean ultimate force for the 2- to 4-year-old specimens was lower than those for the 6- to 12-year-old specimens. In the 2008-2013 period, studies were done in the US. Tensile properties were determined from 18 specimens (20 weeks gestation to 14 years of age) using PMHS OC-C2 to C6-C7 segments; failure force increased with increasing age for all spinal levels.

### ***Unique Experiments from the US Naval Biodynamics Research Laboratory (NBDL)***

Researchers have resorted to animal models because of the sensitive nature of testing child PMHS. Tests were conducted using nonhuman primates (NHP) under the leadership of Captain Channing Ewing, an internist. Between 1972 and 1996, researchers logged 6600 total horizontal runs, including approximately 400 NHP tests. These types of tests may never be repeated, and recent attempts by the US Military to analyze and disseminate findings to the scientific community using modern analytical tools are worth noting. Results have been presented at the NATO-AGARD and other conferences. An engineer from the Netherlands, Jac Wismans, analyzed head-neck data from the US military in the 1980s. Researchers from the MCW and US Army Aeromedical Research Laboratory (USAARL) are currently analyzing information contained in the Biodynamics Data Resource (BDR).

### ***Neck Injury Criteria for Frontal Impact Based on Upper Neck Loads***

Two groups of animal studies conducted under sponsorships from GM and Ford have formed the basis for the present automotive neck injury criteria. The interaction of the force and moment was brought to the forefront, leading to the current standard for frontal tests by NHTSA between 1996 and 1998. The neck criterion introduced in FMVSS 208 in 2001,  $N_{ij}$ , is based on these tests ( $F_z$  is axial force and  $M_y$  is sagittal bending moment at the head-neck junction, and the criterion is the peak value). The criterion is scaled up and down from the 3-year-old to adult and younger populations for FMVSS specifications, and this includes scaling from male to female. Research conducted by the MCW and Duke University, under the sponsorships of the CDC, NHTSA, DoD, industry and others was used to determine critical thresholds. Some were presented at previous IRCOBI conferences.

$$N_{ij} = \left[ \frac{F_z(t)}{F_{intercept}} + \frac{M_y(t)}{M_{intercept}} \right]_{max}$$

### ***Neck Injury Criteria for Low Speed Rear Impact Based on Upper Neck Kinetics***

The original recognition of the injury as a Railway Spine in 1866 is attributed to a Danish-born Surgeon from London, John Erichsen. Harold Crowe first introduced the term “whiplash,” at a symposium of the Western Orthopaedic Association in San Francisco in 1928. In 1953, James Gay and Kenneth Abbot used this term when delivering a paper to the Section on Nervous and Mental Diseases of the American Medical Association, and now it is common, world-recognized terminology.

Since the 1990s, biomechanical researchers have played an active role in confirming/denying postulates of the mechanisms of injury, identifying injury metrics, supporting clinical observations regarding injury susceptibility of females, and designing injury mitigating systems. Early US studies were from the WSU, in the 1960s and 1970s. Researchers from MCW (including Brian Stemper who has presented at previous IRCOBI conferences) and Japan (Koshiro Ono) and Sweden (Per Lovsund, Mats Svensson, Lotta Jackobsson, Karen Brolin, Ola Bostrom, others) have contributed to the current understanding of head-neck dynamics, posture, injuries, and criteria through volunteer experiments with radiography and magnetic resonance imaging, modeling, and field studies. Theories of whiplash injuries include the hyperextension mechanism in 1965; spinal canal pressure (Bertil Aldman in 1986, Mats Svensson in 1993); facet joint shear (King Yang and Paul Begeman); facet joint shear coupled with ventral distraction, and dorsal compression mechanism contributing to joint-related symptomatology (MCW researchers). The mechanism of suboccipital headaches and neck pain were explained in early studies by MCW researchers, along with the female bias to injury. The



facet joint impingement mechanism was also confirmed by Ono and group. Swedish researchers proposed the Neck Injury Criterion, NIC, in 1996; the formulation is given:  $a_{rel}$  and  $v_{rel}$  represent the relative T1-C1 acceleration and velocity,  $L$  represents the length, 0.2 m. The NIC was set at  $15 \text{ m}^2/\text{s}^2$ .

$$NIC = a_{rel} * L + v_{rel}^2$$

In 2002, a team including a medical doctor and an engineer from Switzerland, Felix Walz and Kai-Uwe Schmitt, proposed a new injury criterion,  $N_{km}$ , which replaces the axial force in the  $N_{ij}$  with the shear force.

#### **Neck Injury Criteria for Rear Impact Based on Lower Neck Loads**

In 2016, MCW researchers derived a lower neck injury criterion, termed  $LN_{ic}$ , obtained by determining the shear force and bending moment from rear impact studies using whole body PMHS and isolated head-neck complex tests. The intercepts were determined using survival analysis-based HIPCs.

$$LN_{ic} = \left[ \frac{F_x(t)}{F_{intercept}} + \frac{M_y(t)}{M_{intercept}} \right]_{max}$$

#### **Thoracic-Lumbar Spine Injury Criteria via Acceleration**

In the 1960s, for aviation environments, as applied to ejection seats, the Dynamic Response Index (DRI) was developed using seat acceleration and assuming the human lumbar spine to be a lumped spring-mass model. The maximum DRI was correlated with experimental and operational injury data, and a value of 18 was the selected seat design goal. For energy-absorbing seats, the US Federal Aviation Administration (FAA) conducted tests with different pulses and seats using the Hybrid II dummy. An axial force of 1500 lb was associated with a DRI of 19. It was associated with a 9% risk of detectable injury. This compression only-based criterion is being revised through research at MCW from DoD and other sponsorships as lumbar spine injuries can be associated with bending moments. The US FAA is conducting studies with MCW researchers to obtain tolerance criteria for oblique facing seats following this team's success in developing neck and leg injury criteria for side-facing seats in this decade. This has led to some changes in FAA procedures.

#### **IMPORTANCE OF FIELD DATA**

The US National Automotive Sampling System (NASS) database initiated in 1977 is often used in epidemiological analysis and to identify priority areas for safety.

In 1967, focused data on medical and bioengineering aspects were initiated by Murray Mackay when he was in the Department of Transportation and Environmental Planning at the University of Birmingham, UK. He directed "a team comprised of a traffic engineer, a surgeon, a mechanical engineer, and a psychologist" and analyzed 425 cases (pedestrians, motorcyclists, front seat car occupants) in a 2-year period. In 1969, along with Anthony Ryan from the University of Adelaide, Australia, injury patterns were compared between urban and rural data from the US, the UK, and Australia. Field data were also gathered in Germany.

With limited medical information from the NASS database, MCW researchers proposed in 1990 to the CDC a more detailed method of data acquisition, analysis, computational modeling, and experiments for automotive occupant safety. While the CDC deemed the proposal to be ambitious,

in 1996 NHTSA initiated the Crash Injury Research and Engineering Network, CIREN, along similar lines. This program is ongoing with NHTSA sponsorship, with multiple Centers. The CIREN model is also used in other countries.

### **INJURY CODING**

In the 1940s, Hugh DeHaven coded injuries (DeHaven's scale) from aircraft crashes. In 1969, the origins of the Abbreviated Injury Scale (AIS) were drafted by a group of medical doctors, bioengineers, and public health personnel. In 1973, the Association for the Advancement of Automotive Medicine (AAAM) took the lead and published the AIS in manual format in 1976. Elaine Petrucelli (now Elaine Mackay) remained as the Executive Director of AAAM until early 2000. She single-handedly assisted in all early revisions 1980-1998. Together with Tom Gennarelli as Co-Chair and 7 additional members comprising the International Injury Scaling Committee, AIS 2005 Update 2008 was produced. The current version is from 2015. The AIS has been used for injury coding around the world, and it has helped meet research, education, clinical, and regulatory goals.

### **COMPUTATIONAL MODELING**

Models in the 1980s based on lumped parameter systems were applied for the head, helmet, and spine. Early finite element (FE) models of the spine were developed for the lumbar region. The MCW researchers initiated anatomically accurate cervical spine FE models in the early 1990s, while head injury models were already in progress. Because of computational efficiency, basic mechanics developments in the FE analysis and the availability of commercial software, complex models are being developed and used internationally in this field.

In 1997, Toyota initiated the development of a model termed the Total model for HUman Safety (THUMS), and it is being used by universities for automotive studies. In 2006 the Global Human Body Modeling Consortium (GHBMC) consisting of automakers and suppliers was formed. Its mission was to develop and maintain FE human body models for crash simulations. Recently, NHTSA has joined the consortium and Steve Ridella has played a role. Universities including Waterloo (Duane Cronin for neck) and WSU (King Yang for head), UVA (Matthew Panzer for thorax), and IFSTTAR (Philippe Beillas for abdomen) are academic partners for model development. The Elemance, headed by Joel Stitzel and Scott Gayzik, has issued free licenses to academic institutions in the US, Europe, Asia, and Australia. These researchers from Wake Forest University have also contributed to GHBMC model developments. It is currently used in automotive and military studies.

### **SOME FUTURE CONSIDERATIONS**

In drawing to my concluding remarks, it is worthwhile to acknowledge that research advancements have been valued and realized in the medical field for a long time. A medical doctor can influence the outcome of one patient at a time; however, improvements to human safety from injury mitigation have farther reaching consequences to the society as a whole.

The relatively new and applied field of Injury Biomechanics has already made significant progress in its short life span, an effective period of less than 75 years. Injuries to the major regions of the human body in different environments are identified by field data and replicated and quantified by acceleration, load deflection, or in combinations for establishing injury criteria in standards. This has served the society at large by significantly reducing the injury burden around the world. Certainly, not one country or group of researchers has solely carried the load.

Imagine a single acceleration metric (HIC) being responsible for mitigating head injuries in many environments - sports, aviation, military, automotive occupants and vulnerable road users, and the list goes on. Likewise, single chest acceleration or deflection metrics have greatly mitigated chest injuries in many scenarios. These simple metrics have also been responsible for bringing in new technologies, which are in the motor vehicles we use on a daily basis, for example. Mitigating injuries in the military environments are also by products of these results. Recognition of injury mechanisms and tolerances has played a role in sports (spearing in US football) and will continue to be a driver for safety hereafter. These achievements are unique, and we should all be proud.

Current practices have relied on standardizations from past experiments. This is true across continents and fields: aviation, sports, automotive, and military. Current standards for injury assessment use the binary approach. One has to ask, do these binary limits of head acceleration, chest deflection, femur and spine loads cover the whole injury spectrum (different probability levels, different severities, different consequences of injuries, and different levels of associations of one injury with another uninjured or injured component in the same body region or different body regions, and to all groups of the populations)? Do we need to stay confined to the strict significance level ( $p < 0.05$ ) in all of these pursuits?

The Injury Biomechanics field is undergoing a rapid transformation. We heard a lecture here about (cerebral) concussion and its perception/socialization from Tom Gennarelli not too long ago. It may be necessary to include the entirety or portions of the risk curve in safety assessments, as well as the effect of one risk curve on the other. For example, a 10% risk of an injury type/severity to one body region – how it affects, regulates, or modulates the other body region or component in the near and long term – may need to be assessed from statistical perspectives.

Incorporation of this coupling, risk levels, and injury severities in different situations in the analytical process will be of immense value as algorithms will be written of possible scenarios to direct or control the environment, e.g., autonomous vehicles. For this to be accurate, the time-based growth response of the biomechanical metric (e.g., force-time) to the insult may need to be considered in the analytical process. While complex computer models representing the mid-size male and small-size female, for example, are proven, models of varying local anthropometry and gender may be needed to assist data mining and improve the efficacy of algorithms. Regulators, Academia, and industry leaders have begun to adapt to this change because of advances in computational science, statistical modeling, human-machine interaction, and data mining. Interactions with other disciplines, especially biostatistics, will be more relevant, a fact realized by the medical field to cure diseases and develop drugs and devices for a long time.

From a medical perspective, patient/subject-specific models, also called personalized medicine, are gaining importance. Because of the unknown nature of the material properties, any experimental/FE model is only an approximation. It would be necessary to use experiences gained from clinical outcomes and combine them with statistical and analytical modeling to assist surgeons in choosing the best possible options for pre-surgical planning, surgical procedures, and postsurgical follow up. While not discussed here, the cyclic loading mode, not traditionally addressed in IRCOBI, is critical for the civilian. It is also true for the military, as training activities are used to determine and assign the appropriate individual to specific types of duty and also the Return-to-Duty (RTD) decisions. These considerations also require Injury Biomechanics researchers to pair with health care providers and others.

Injury Biomechanics researchers certainly have the background and expertise needed to accomplish further safety improvements and meet new challenges. President Ciaran Simms, Vice President Barclay Morrison, Treasurer Duane Cronin, and Secretary Kai-Uwe Schmitt, an experienced, diversified, multi-national team, are well poised within the IRCOBI structure. While an individual or a group's research output is always incremental, and while applications can take a long time—as we have seen with the federalization/regulation processes in the USA, for example—changing times require changing behaviors in order to remain relevant. The great object of Injury Biomechanics researchers is to continue to minimize/mitigate currently-discovered and future-expected injuries and improve quality of life of the injured via both Engineering and Medicine. I have no doubt that we are well poised to meet these new goals, and I conclude this lecture to call upon you to continue this noble work.

Good luck. Thank you all.

### **ACKNOWLEDGEMENTS**

The US Federal government has funded more than 95% of my research studies. I am thankful for the US Federal system that allows scientists like me to pursue research through funding opportunities. The support and interactions with the MCW faculty and VA Medical Center are acknowledged. I thank Alison Stadig for her suggestions and editing of this text. My research work could not have been done without the help of many staff members of the Neuroscience Laboratories. While it is too long to list all, special thanks are in order to Michael Schlick, John Humm, Jason Moore, and Christy Stadig for assisting all our faculty and staff over the years. They are dedicated, unselfish, and authentic researchers. All my colleagues have played a role in my career, and this list is also long. My accomplishments would not have come to fruition without help from my peers and reviewers, and to all of them, I owe a debt of gratitude. I continue to expect to learn from these uniquely qualified individuals. A considerable majority of my research has been a byproduct of my collaborations with Frank Pintar for more than three decades. He has shared his ideas, provided unselfish support, offered constructive criticisms, and continues to encourage and challenge me for further pursuits. To this true friend, a special place exists in my life.