SIMULATING THE ROAD FORWARD: THE ROLE OF COMPUTATIONAL MODELING IN REALIZING FUTURE OPPORTUNITIES IN TRAFFIC SAFETY

Jeff Crandall
University of Virginia

ABSTRACT

Current vehicle and restraint designs consider a relatively small percentage of the variation that exists within the population of humans and the distribution of crashes. The breadth of these variations, coupled with the added complexity of ever-changing societal trends, will not permit designs developed using conventional techniques to realize established goals for injury and fatality reduction. Within the realm of passive and integrated safety, widespread implementation of modeling into the design and evaluation process provides the only viable means of achieving significant reductions in injury and fatality. Ultimately, the versatility and efficiency of computational modeling will establish a new paradigm in which injury is described by causal mechanisms including statistical representations of population variations. While this transformational change will be guided and directed by simulation technology, it will require commensurate evolution of techniques within the fields of experimental biomechanics, crash reconstruction, and epidemiology.

Keywords: Models, Biomechanics, Procedures, Sensitivity Analysis

The World Health Organization estimates that 1.2 million people die each year in road crashes worldwide and as many as 50 million are injured or disabled (WHO, 2004). To confront this global epidemic, governments have put forth programs with identified targets and timeframes to reduce these fatalities. The European Commission created a target of a 50% reduction in fatalities within seven years for their EC Road Transport Action Program (2003). Recently, Japan set a goal to halve the number of fatalities within the next decade (Shima, 2009). Countless other nations have implemented regional and national road traffic casualty reduction programs with the objective of drastic reductions in fatalities in the foreseeable future. With perhaps the most ambitious plan, the Swedish Parliament adopted a progressive plan invoking the principle of Vision Zero with the goal of eliminating deaths and serious injuries from motor vehicle crashes by 2020 (Tingvall and Haworth, 1999).

While the results to date from the fatality reduction programs have been both impressive and commendable, additional gains will grow increasingly difficult to achieve as opportunities, potentially characterized as lower-hanging fruit, are realized. As the marginal costs of reducing injuries and fatalities grow, new paradigms of injury prevention must be implemented. This paper contends that computational modeling of human response and injury will drive the paradigm shift in passive and integrated safety. While the utility of computational modeling is envisioned well beyond its current manifestations, it has yet to be fully implemented in injury reduction strategies. Growth in computational modeling does not imply that complementary areas of research, such as experimental biomechanics, crash data analysis, and epidemiology, will stagnate or shrink. Conversely, it is anticipated that these disciplines will experience evolution in both the quantity and quality of their contributions as they play supporting roles guided and directed by the needs of simulation technologies. Similarly, it is envisioned that computational modeling will assist educational efforts, infrastructure development, and regulatory programs in injury prevention by providing estimates of future trends with proposed priorities determined from simulation data.

The complexity of the impact environment cannot be recreated in laboratory investigations nor fully described in retrospective analysis of crashes and field data. Modeling techniques provide a viable method for addressing the requisite range of variation inherent in the human, vehicle and crash environments that are otherwise intractable using conventional approaches. The versatility of simulation will allow engineers to systematically analyze the complex array of intrinsic and extrinsic factors experienced in the field that are too large, too complex or too expensive to tackle with

conventional approaches. Despite the enormity of the problem, computational modeling has the potential to render meaningless the term "accident" as a causal mechanism of road traffic injuries and instead to define them in the context of a fully understandable, tractable, and preventable problem.

To achieve its potential, human body modeling must evolve commensurate with other developments in the field of traffic safety. On the immediate horizon, additional opportunities in injury reduction will be garnered when the anticipated benefits of current active safety systems are fully realized. Even with the presence of active safety systems, however, it must be expected and accepted that crashes cannot be totally avoided and that continued refinements in passive safety must also occur. Unfortunately, characterization of passive safety as a mature field in relation to the emerging science of active safety may mistakenly lead some to believe that additional gains in passive safety are unattainable. This pessimistic opinion may result from safety professionals assessing future opportunities in injury reduction using current capabilities and practices. While computational modeling comprises a regular component of evaluation and design programs, the transformational change that will occur when injuries are characterized and prevented in a simulated reality is not fully appreciated in most predictions of future opportunities.

Current evaluations of vehicle and restraint design through experimental testing cover a relatively small percentage of the range of crash conditions and occupants observed in the field. While manufacturers, governments, and other institutions work towards other crash conditions, regulatory and compliance tests provide fixed targets for design and evaluation that remain a primary focus. While regulatory and compliance testing have provided significant contributions to restraint and vehicle design, Kent et al. (2005a) showed that crash testing programs such as the US NCAP and FMSS 208 cover roughly 20% to 40% of the driver fatalities in frontal crashes. Moreover, the authors found that nearly 85% of the driver fatalities in an elderly population occur at frontal impact speeds less than the current US NCAP conditions. While these test programs serve a useful purpose as stand alone evaluations at specific conditions, the validity of extrapolating their benefits to the extremes of the crash and occupant populations is questionable. The opportunities for continued refinements in performance and evaluation testing procedures are highlighted in the recent paper by Segui-Gomez et al. (2007). After controlling for crash direction and impact speed, the authors were unable to detect a statistically significant relationship between EuroNCAP safety ratings and severe injuries suffered by drivers and front passengers in real world crashes with the same vehicle. Given the number of controlling variables, a logical conclusion is that the tests are incapable of representing the diversity and complexity of the occupant and crash environment even under seemingly comparable conditions.

Past and current efforts by manufactures, suppliers, and others have provided tremendous improvements in vehicle and restraint design that have transformed the chances of survival in a collision. Despite these advances, however, the most effective safety systems have not approached the levels required for a Vision Zero approach to injury prevention and control. Unquestionably, the seat belt remains one of the greatest passive safety success stories in terms of injuries prevented and lives safety. The belt restraint system is approximately 42% to 45% effective at preventing fatal injuries in occupants (e.g., Evans, 1990). As a supplemental restraint, the air bag system provides a 12% increase in restraint effectiveness (Viano, 1995). While a 60% effectiveness rate for the belt and air bag system is impressive, the potential, or rather need, exists for the remaining 40% to be protected. Furthermore, any estimate of effectiveness provides only a static snapshot of what is an ever changing mix of vehicles, occupants, collision types, and injury. Thus, evaluations of effectiveness must be continuously reassessed and reevaluated in a dynamic process. The ability of modeling to conduct predictive, as opposed to retrospective, evaluations of countermeasure effectiveness including varied occupant, vehicle, and crash populations has the potential to revolutionize the manner in which safety opportunities are identified and priorities are established.

While models of the human body have played a supporting role in understanding response and injury since their introduction in the 1960's (Prasad and Chou, 2002), increases in the commercial availability of software coupled with growth in the efficiency, power, memory, and storage capacity of computer hardware have resulted in relatively sophisticated models of the human body in relation to the two-dimensional constructs of forty years ago. These advances in computationally modeling have created the potential for simulations to address a wide spectrum of problems within injury biomechanics. Unfortunately, this potential is likely unattainable using current human models because they do not adequately describe the real or underlying behavior of a human during an impact.

While models, by definition, represent an approximation of physical phenomena, they must provide reasonably accurate estimates in order to ensure confidence in their results. As broader segments of the crash and human population are simulated, the task of a robust model validated for a diverse range of conditions becomes more challenging. Thus, the goal of future developments in human models will be to enhance the model's predictive capabilities by expanding the data available for development and validation and minimizing controllable sources of error.

This paper highlights the potential of human models to handle, in a quantifiable fashion, the multi-dimensional complexity of the crash environment. Furthermore, it demonstrates how models will permit more accurate characterization of injury as an event, not simply as mechanical failure at a prescribed tolerance level. By embracing the inherent diversity among individuals and enhancing knowledge regarding the phenomena that underlie the injury process, advances in the predictive ability of human models are viable at a variety of scales (time and length). To support the need for this continued development and refinement of human models, this paper provides a review of the population factors that influence injury in traffic incidents and estimates the challenges for continued injury reduction presented by these population factors. The ability of modeling to address these challenges is reviewed relative to current and future modeling approaches. Concurrent with the advances in modeling capabilities, this paper highlights the need for evolution within the fields of experimental biomechanics, crash reconstruction, and epidemiology to be both guided and directed by the human modeling activities.

HUMAN FACTORS

Human response and injury during impact loading result from the interplay of a vast array of intrinsic and extrinsic factors. The list of human parameters alone is extensive and includes, but is not limited to, variations in size, obesity, gender, age, biomechanical tolerance, pre-existing medical conditions, sitting position, muscle tensing, and posture (Mackay, 2007). Expanding the complexity of the problem, this diversity may vary temporally within individuals (e.g., changes in muscle tensing during a crash event)) and across populations (e.g., societal increases in elderly and obese populations). Combined with the range of variation presented by collision and restraint parameters, the realm of possible paired occupant states and crash scenarios form an n-dimensional space whose characterization and analysis are inconceivable using conventional techniques. While the utility and practicality of modeling to incorporate this breadth are demonstrated later in this paper, several of the key human factors are examined in detail to justify the need for including them explicitly in future injury prevention strategies.

ANTHROPOMETRY: One of the easiest variables to characterize and the most extensively studied is the size of the occupant. Anthropometric data exists for external measures of body dimensions (e.g., Robinette et al., 2002; Gordon et al., 1988) with relatively less data for mass distributions and moments of inertia of specific body segments (e.g., Schneider et al., 1983). Historically, the combination of anthropometric data sets has enabled manufacturers to develop a range of dummies at target sizes for adults, namely the 5th percentile female, 50th percentile male, 95th percentile male, as well as an assortment of child dummies. The 50th percentile male or mid-size male dummy has served as the standard for normalization of response and evaluation of performance for many of the current injury countermeasures.

To the extent that occupant sizes have been considered in evaluations of safety devices, it is clear that characterization by mass and stature neither captures a large percentage of the underlying population (Figure 1) nor describes adequately the complexity of body shape and composition inherent within the human body. Customary attempts by health professionals to characterize body shape use the Body Mass Index (BMI) as a measure of patient's body fat relative to an "ideal" weight (Figure 2). Favored for its ease of calculation, BMI is defined as the subject's mass in kilograms divided by the square of the subject's length in meters. While BMI makes no distinction between body weight from muscle and body weight from fat, it provides a convenient variable for evaluating societal trends. Within the context of injury biomechanics, it provides a convenient variable with which to assess restraint performance for obese individuals.

The World Health Organization (WHO) defines "overweight" as a BMI≥25, and "obesity" as a BMI≥30. Estimations by the WHO indicate that globally 1.6 billion adults (age 15+) are overweight and at least 400 million adults are obese (WHO, 2006). WHO projects that by 2015, approximately 2.3 billion adults will be overweight and more than 700 million will be obese (WHO, 2006). More than half of the adult population is now defined as being either overweight or obese in a majority of countries represented by the Organisation for Economic Co-operation and Development (OECD) (OECD, 2007). An interesting study by Wang et al. (2008) projects that all American adults will be overweight or obese by the year 2048. While a public health problem in and of itself, obesity must be considered in traffic injury prevention strategies of the future.

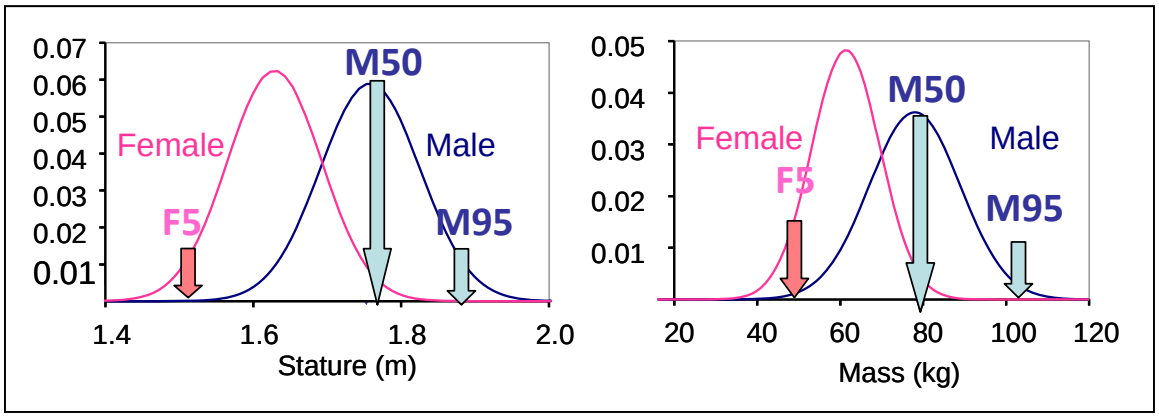

Figure 1- Stature and mass distributions by Gender based on the data of Gordon (1988) with inserted markers for standard dummy sizes (5th female (F5), 50th male (M50), 95th male (M95)).

To date, studies on the effects of obesity on crash-induced injuries have primarily focused on epidemiologic evaluations of field data. Mock et al. (2002) observed increased risk of mortality and severe injury in occupants with increased body weight or BMI. Zhu et al. (2006) indentified increased risk of driver death for males in crashes at both high and low extremes of BMI but did not find the same trends for women. Other studies have shown that the influence of increased BMI on injury risk is not evenly distributed across all body regions (Cormier, 2008) and may even be protective for some areas of the body (Wang et al., 2003). While the complex effects of obesity on injury distribution are not understood, the overall increase in injury risk is well established. This injury risk for obese occupants can be attributed to a combination of more mass (energy) to manage during the crash, a poor restraint fit due to the additional subcutaneous tissue at points where the belt loads the body, and the presence of comorbidities (e.g., heart disease, high blood pressure, diabetes, various types of cancer) associated with obesity. While not explicitly intended to be investigations of obesity, variation in restraint performance has been demonstrated both experimentally and computationally for occupants of varying mass and stature (Miller and Maripudi, 1996; Adomeit et al, 1997; Bose et al., 2008). Projected changes in the occupant demographics related to weight and BMI will obviously lead to additional challenges for future vehicle and restraint designers. Given the capability to morph and to scale the human geometry, modeling provides one of the best opportunities to evaluate population parameters using anthropometrically specific models. While relatively few simulation studies have been done that include even BMI as an explicit factor, several investigations have been able to show the overall sensitivity of response to obesity measures but have not been able to reproduce the region specific injury risk patterns seen in the evaluations of field data (e.g., Bose et al., 2008).

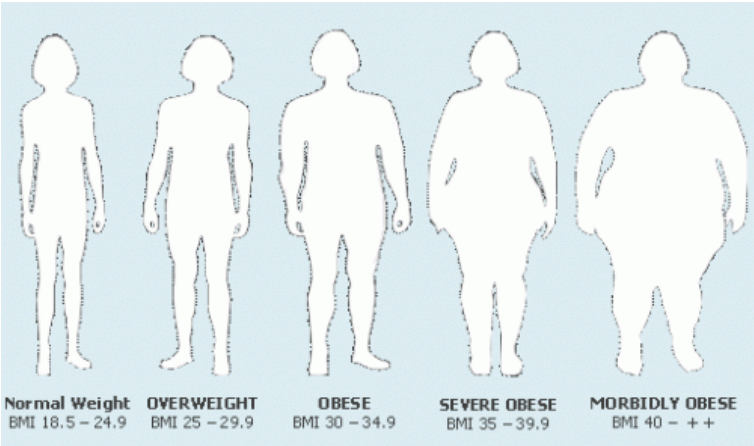

Figure2 - Descriptive terms associated with Body Mass Index ranges (WHO, 2006).

These simulations studies of obesity have used multi-body models morphed using direction-specific scale factors. In the field of impact biomechanics, this technique of direction-specific scaling has been widely used in the area of pediatric model and dummy specifications (e.g., Irwin and Mertz, 1997). While capable of reproducing gross changes in body anthropometry, scaling of body dimensions by external markers without regard to changes of tissue composition and internal organ position likely overlooks essential details that are necessary to accurately predict the injuries. Given the natural variation among individuals of nominally the same body type and asymmetric changes in internal and external mass distributions caused by factors such as obesity, a framework for handling and characterizing the shape of body's internal and external structures must be utilized. Fortunately, imaging modalities combined with subsequent finite element (FE) modeling and analysis have the potential to provide much richer information about the body shape than traditional anthropometric measurements that measure circumferences, breadths, and linear distances between anatomical landmarks. The digital expression of the three-dimensional geometry and topology of tissue structures from conventional imaging modalities (e.g., computed tomography (CT) and magnetic resonance imaging (MRI)) and manual sectioning (e.g., Visible Human) is used extensively in current FE human modeling. What is underdeveloped in modeling efforts is a framework of analysis for characterizing individual anatomical variations relative to distributions within the population. While the utilization of models derived from single individuals of varying sizes (e.g., Combest, 2009) will constitute a step forward relative to a single anthropometry (i.e., mid-size male), expression of human variation at a few discrete geometries will be a relative simple approximation to the potential array of models with differing morphology at the tissue, subsystem, and whole body levels. Comparison of simulated tissue responses and injury relative to laboratory tests or field data will likely require statistical analysis of body and organ shapes and locations as well as their interactions. Fortunately, these procedures are currently utilized in medical imaging processing to develop patient specific models and in studies of anthropometric variation outside the realm of injury biomechanics.

For organs and other body structures, characteristic shapes can be defined relative to a standardized reference coordinate system for the body. Alignment and characterization of mean dimensions for individual organs and structures can be performed using techniques such as Procrustes Analysis (e.g., Takhounts et al., 2008). Whether evaluating the shape of the body or individual anatomical structures, it is essential to retain the entire shape and not to reduce the structure to a measures of individual dimensions. Once reference shapes are established, correlations between the shapes of co-varying anatomical structures can be performed by decomposing sets of shapes using techniques such as principal component analysis (PCA). PCA summarizes patterns of variation in a set of variables so that they measure different aspects of shape variation (Figure 3). This method takes into account not only size variance but proportional variability as well. Thus, it can account for not only individuals who are uniformly large or small, but those whose measurements may combine small torsos with long limbs (Azouz et al., 2006). Since PCA relies on the correlation structure of the variables, it can facilitate understanding and quantification of co-varying body dimensions. In order to develop computational models that include quantification of shape variation as well as corresponding modes

of variation, techniques such as Procrustes Analysis and PCA must be used in the analysis of anthropometric data. Given the positional dependence (i.e., occupant posture) of internal and external relationships among shapes and the alignment and position of structures as well as variations with age (e.g., elderly occupants) and gender, considerable effort will be required to characterize and model the breadth of the population and conditions observed in the field.

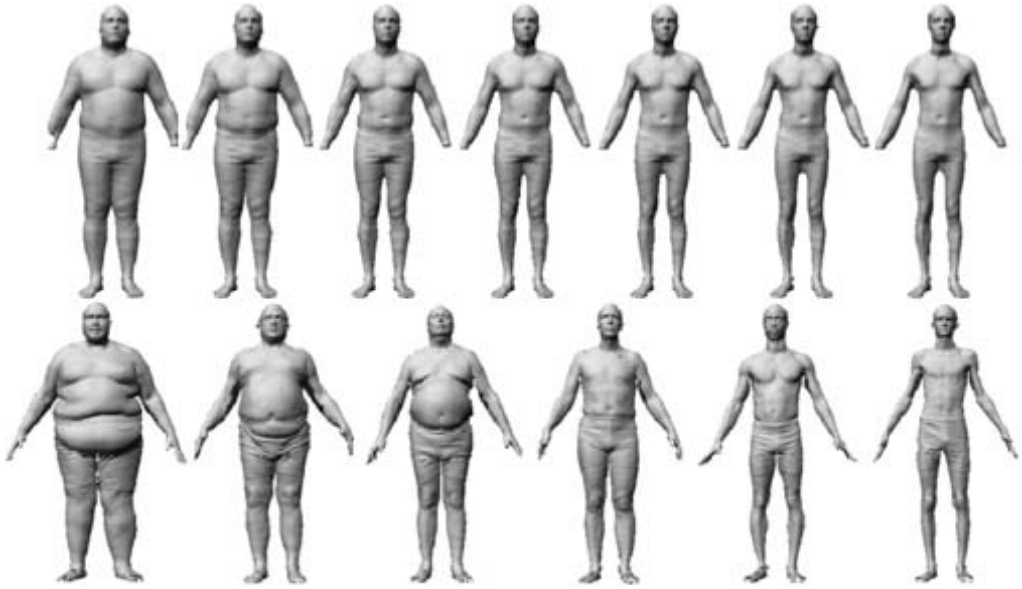

Figure 3 - Variation mass (weight) given a normalized height (adapted from Azouz et al. (2006)).

POSITION AND POSTURE: Analysis of crash data in the field provides an important counterbalance to the relatively controlled and contrived conditions of the laboratory environment. For occupants, literature on seating position has primarily focused on drivers (Schneider et al, 1983; Manary et al., 1998; Flannagan et al., (1998)) with more limited information on front passengers (Zhang et al., 2004) and rear seat occupants. The nominal driving posture utilized in most experimental investigations is based on geometric dimensions and joint angles measured for preferred seating posture of adult drivers (Schneider et al., 1983). Accommodations for occupant size and vehicle parameters have been addressed parametrically in terms of seating position relative to interior vehicle components and joint angles (Reed et al., 2001). While a particular strength of these investigations is the number of vehicles and volunteers included in the experimental design, their goal has been to develop a nominal driving position. The static measurements of drivers in a preferred driving position were not intended to reflect the entire range of occupant positions that occur during driving activities. A much larger degree of variation has been observed in field investigations (Bingley et al., 2005; Mackay et al., 1998). To classify seating posture and position, Zhang et al. (2004) used video-photography of passengers at tollbooths in combination with a survey to solicit occupant's estimations of their own occupant posture. Although the video-photography and survey provided limited quantitative information about the overall occupant posture, the qualitative categories did permit Bose et al. (2008a) to develop eight non-nominal occupant postures for use in a computational study. The authors used the eight postures to prescribe positioning of the lower extremities, proximity of the upper-body to the steering column, and sideways lateral bending (in the coronal plane) of the occupant (Figure 4). The study demonstrated both the importance of postural orientation in predicting collision-induced injuries and the ability of modeling to implement occupant states as variables. While the modeling by Bose et al. provides a subjective interpretation of the body positions, quantitative information is required for accurate elucidation of response sensitivity as a function of occupant posture whether it is achieved voluntary or involuntarily (e.g., pre-impact braking). In addition, there exists a need to expand the information beyond the static measurements of the position of the body regions and to include information on the muscle tensing required to

achieve these positions, the changes in location of internal organs with changes in occupant position, and the changes in the residual stress state of the musculoskeletal components in these systems. Given the sensitivity of occupant response to position, expanded information on occupant orientation and position will be crucial to the optimization of passive and integrated safety systems.

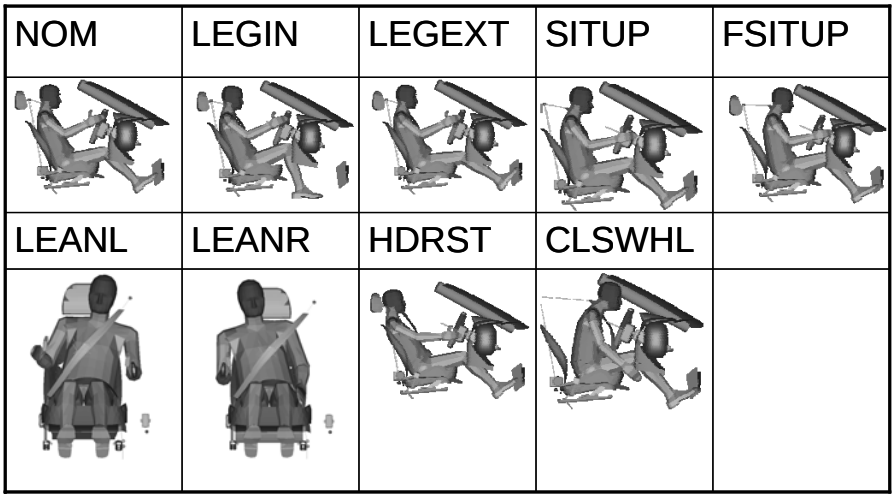

Figure 4 - The occupant model oriented in the occupant postures evaluated by Bose et al. (2008a)

While the positional variation of the occupant can vary significantly, the confines of the vehicle interior place at least some bounds on the initial position of the occupant. In comparison, pedestrians can be in any number of positions across the vehicle (e.g., various regions of the vehicle front) and in any number of postures (running, standing, lying down) when struck by vehicles. While contact on the front corners of the vehicle is most likely, experimental studies have primarily focused on the pedestrian striking the mid-line of the vehicle. Similarly, few pedestrian studies have specifically investigated the sensitivity of the kinematic response and injuries to changes in the position and orientation of the pedestrian at impact. Bhalla et al. (2002) showed that pedestrian throw distance is affected by pedestrian orientation (i.e., facing toward, away, or lateral to the vehicle). Beyond simply the orientation of the pedestrian's body, the orientation of the limbs is important for predicting the distribution of injuries on the pedestrian's body. For instance, it is common for crash investigators to decide the stance phase of the pedestrian based on whether the anterior or the posterior aspect of the head was injured during the crash (e.g., van Rooij et al., 2003). Meissner et al. (2004) used a computational model to demonstrate that stance has a substantial effect on the subsequent impacts of the head and thorax with the vehicle. The variation in stance changes the severity of an injury incurred during an impact by altering the region impacted and the subsequent upper body kinematics. In addition to variations of stance and the overall orientation of the body, differences in the ground reaction forces and muscle tensing state play a role in the subsequent interactions of the pedestrian with the vehicle. To account for differences associated with gait of the pedestrian, Untaroiu et al. (2008) developed a continuous sequence of walking as a design parameter with prescribed relationship between the ankle, knee, hip, shoulder and elbow joint angles (Figure 5). While there are obviously variations in these relationships for different gaits (e.g., running is not simply walking done more quickly), parameterization of position in this manner should facilitate design of countermeasures for a broader range of conditions and should produce improved correlation when actual vehicle-pedestrian crashes are reconstructed and simulated.

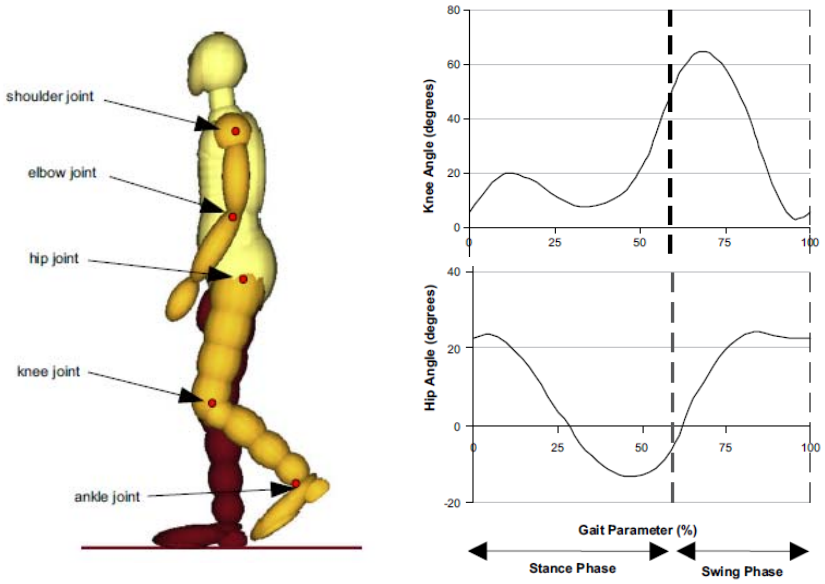

Figure 5 - Pedestrian model with joint angles as a function of gait phase (Untaroiu et al., 2008)

Meissner et al. (1994) evaluated the in-position designation for restrained child occupants and found that most children, due to a propensity to explore their surroundings and a dislike of being restrained, turn in the child restraint, play with their harness, and even remove themselves completely from their restraints. By observing behavioral patterns of children using a video acquisition system, the authors summarized the children's behavior as "squirmy" and "squiggly". While not quantitative assessments, these alterations from a nominal in-position orientation obviously have the potential to lead to compromises in restraint performance. Furthermore, optimization of child restraint features without regard to the rather common positions that differ drastically from those utilized in regulatory or standard laboratory testing will likely produce suboptimal performance in the field. Marshall et al., (2005) used a MADYMO model of the 12 month CRABI dummy in a rearward facing child restraint to investigate child restraint performance for what were deemed typical positions for children: nominal position, head and neck bent forward, body bent forward at upper thoracic spine, body bent forward at the hips (Figure 6). In the frontal crash simulations, head acceleration, head contact force, neck tension, and head injury criterion (HIC) showed increases up to 600% with increasing distance from the nominal position.

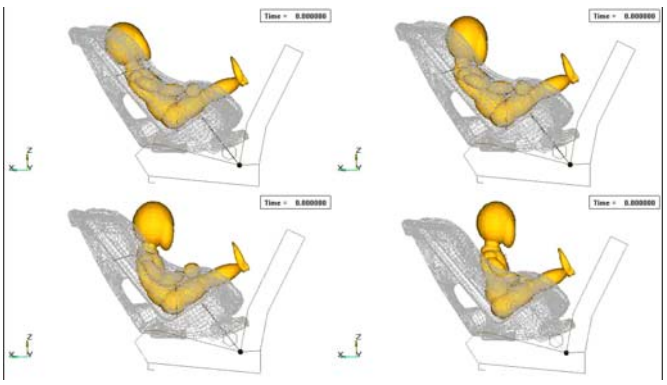

Figure 6 - CRABI 12 month positions clockwise from upper left: nominal position, neck leaning forward, upper torso leaning forward, leaning forward at hip.

AGE: The number of persons over the age of 60 has tripled over the last 50 years and will more than triple again over the next 50 years (UN, 2002). In the U.S., life expectancy has doubled since the beginning of the 20th century (Oskvig, 1999) and the percent of the population over the age of 65 is

forecast to exceed 25% by the year 2030 (OECD 2001). Due to longer life expectancy and decreasing birth rates, the population growth rate of older Americans is expected to be 3.5 times that of the total population by 2010 (Kent et al., 2005). Aging is not just a U.S. domestic public health issue, however, as more than two in every five persons projected will soon be at least 60 years of age in some developed countries (UN, 2002). Perhaps most alarming is the projection that China will have 285 million people over the age of 60 by 2025.

It is well documented that older people are more susceptible to injury than younger people, and exhibit higher morbidity, mortality, and treatment costs for a given injury. Kent (AAAM, 2009) noted that older occupants are more fragile than younger occupants with a tendency to sustain a greater level of injury for a given magnitude of loading and they are also frailer than younger occupants with worse outcomes for the same injury. For these reasons, Mackay (2007) concluded that the elderly population may have the most to gain from improvements in occupant protection.

While diminished structural response and injury tolerance of the elderly partially result from declines in the hard and soft tissue structures following middle age. At the structural level, degradations in much of the musculoskeletal system have been quantified at least quasi-statically (e.g., Yamada, 1970) but relatively few injury risk functions for dynamic loading have incorporated age as a covariate. Notable exceptions at the structural level (i.e., body region) include criteria for lateral loading of the chest (Cavanaugh et al., 1993; Eppinger et al., 1984; Kuppa et al., 2003); frontal loading of the thorax (Kent et al., 2003), and axial compression of the hindfoot (Funk et al., 2002). Some researchers (e.g., Hardy et al., 2005) have proposed a simple shift in the injury tolerance curves to reflect changes in injury tolerance as a function of age. Due to differences in types, combinations and severities of injury that exist between elderly and young adults, simple scaling of criteria with age is likely misleading. Kent et al. (2005b) noted that complexity is added to the issue of characterizing the tolerance of the elderly since some older people are in good condition and some are not. This results in a senescent population that is unique in the degree of its heterogeneity and suggests that while the use of age as an indicator of injury tolerance is convenient and necessary, it is insufficient in and of itself.

Structural tolerances and response result from a combination of material and geometric properties. Unfortunately, a limited number of studies have explicitly addressed shape and postural changes in the elderly. Kent et al. (2005) identified changes in the rib angle as a function of age and used a parametric FE study to demonstrate the significance of these changes on rib fracture and rib cage response. Using three-dimensional landmarks, Gayzik et al. (2008) quantified the overall age-related shape changes of the human rib cage and spine. More recently, Choi et al. (2009) used three-dimensional whole body laser scans to obtain external geometry of an elderly occupant in the driving position and combined this data with registered CT, X-ray, and ultrasonic data to obtain the internal skeletal and soft tissue structures (Figure 7). While this process has been completed for only a single adult male (size and shape of 50th percentile Korean elderly male), it provides a first step at developing the overall morphology of the elderly.

Due primarily to the interest in osteoporosis, researchers have extensively characterized age-related changes in the quality, architecture, and material properties of bone. It is well established that bone strength and quality diminishes with advancing age (Figure 8). Classic measures of bone quality such as bone mineral content (e.g., Hannson et al. 1980) and apparent density of bone (e.g., Carter and Hayes, 1977) have been widely used as age-dependent predictors of bone strength. Since these measures fail to capture the underlying inhomegeneities, architecture and structure of the bone, investigations of cortical bone strength and bone mineral density frequently show low correlation coefficients (Snyder and Schneider, 1991; Squillante and Williams, 1993; Rho et al., 1995) but possess some predictive value. Within the automobile safety framework, Hardy et al. (2005) developed a prototype ultrasonic system for measuring bone mineral content with the intent of incorporating this information into restraint systems that could be customized for a given bone condition.

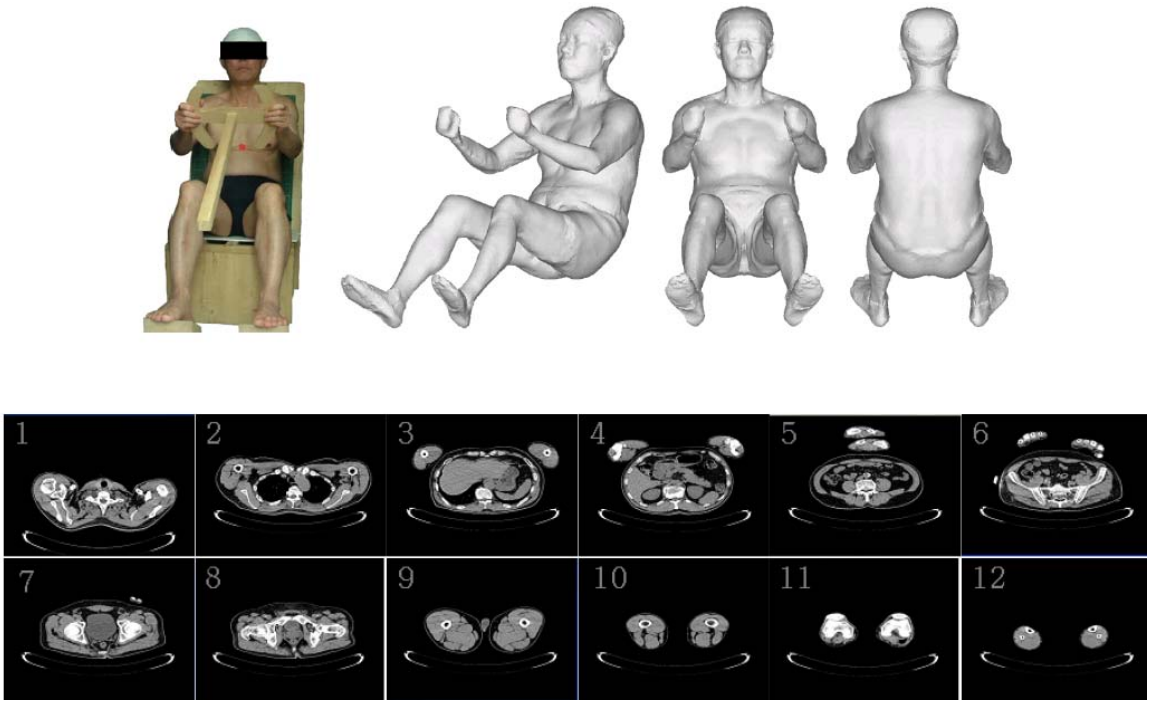

Figure 7 - Body scan and CT slices of elderly volunteer in a driving position (Choi et al., 2009)

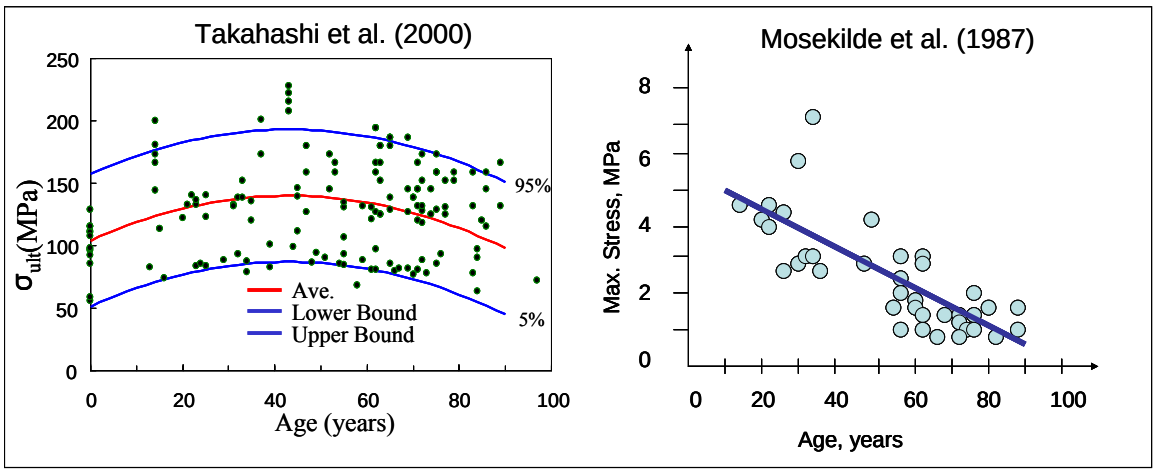

Figure 8 - Corridors of Ultimate Stress as a function of Age for cortical bone (femur, Takahashi et al., 2000) and trabeculare bone (vertebral body, Moskilde et al., 1987).

With regard to age-related effects, characterization of soft-tissue changes has lagged that of the hard tissue structures. It is well known that muscle atrophies with age but significant degradation in mechanical performance also occurs in the ligaments, tendons, articular cartilage, intervertebral discs, internal organs, and skin (Figuers, 2005). These changes in the response and failure of connective tissue result from complex alterations of the collagen type, collagen cross-bridging (e.g., Seto and Brewster, 1991), diameter of elastin and collagen fibers (e.g., Dressler et al., 2000) and water content (Menard and Stanish, 1989). Beyond connective tissue, changes in the function and properties of various organs and systems occur throughout the body. For example, decreases in brain volume and corresponding increases in cerebrospinal fluid volume have been noted (e.g., Gur et al., 1991) and shown to be significant for brain response during impact loading (e.g., Kleiven and Holst, 2001).

Despite the compromised injury tolerances of the elderly, the same basic design maxims apply for restraints design: maximize time of force application, maximize distance of force application, minimize body articulations, and distribute forces over strongest anatomical structures (Eppinger, 2002). Given the complex structural and material changes that occur with aging, adjustments in

design may need to reflect regional differences in tolerance. It is well documented that chest injuries, even a single rib fracture, are problematic for the elderly (Elmistekaway and Hammad, 2007). In order to understand prevention of these rib fractures, biomechanical engineers must have an improved understanding of the aging body to withstand different loading patterns and magnitudes. Zhou et al. (2006) indicated that differing degrees of change in soft and hard tissue properties as a function of age resulted in different amounts of tolerance reduction for distributed (e.g., air bag) versus concentrated (i.e., belt loading).

It is clear that the geometric and material changes of the elderly result in a significantly different occupant to be protected relative to young adults. This paper has not expounded on the developmental changes in children but these present an equally, if not more, demanding challenge for the biomechanical engineer. While basic response changes such as regional stiffness may be incorporated into a physical dummy, the complexities associated with variations in response and tolerance at the tissue level (i.e., differing changes in soft and hard tissue) along with interactions between these hard and soft tissue structures will require more advanced representations of the human. Fortunately, incorporation of this complexity into a FE model is already underway within the impact biomechanics community (e.g., Ito et al., 2009, Choi et al., 2009) and, when combined with other covariates such as size and gender, should permit tailored performance of future restraints and countermeasures for the occupants of varying ages.

MUSCULATURE EFFECTS: Bracing or muscle tensing strongly influences the occupant dynamics during a collision by modifying the kinematics of the occupant, changing the external forces at the interfaces with the vehicle, increasing the effective mass of body regions, and altering the stress distributions and injury patterns of internal tissues. Studies involving low-speed volunteer sled tests and driving simulator tests have demonstrated these effects subsequent to muscle forces generated in the pre-collision environment (Begeman et al., 1980; Choi et al., 2005; Ejima et al., 2005; Manning et al., 1997). While investigations have indicated that isometric or voluntary muscle action prior to a collision alters the occupant dynamics, the effect of these muscular actions combined with reflexive contractions during the impact on injury risk remains largely unreported. This likely results from the fact that human volunteers are tested at low severity levels and human cadavers and anthropometric test devices provide only limited insight to the influence of muscle tension. In particular, dummies (e.g., Hybrid III) include musculature in the neck where the effects are significant but also include musculature in areas such as the chest where the influence on response is not significant at injurious levels of loading (Shaw et al., 2005). In the areas where muscle is implemented, the dummies exhibit fixed levels of tensing that cannot be adjusted either before or during an impact event. Additionally, musculature effects are not implemented in either the abdomen or extremities despite the fact that active muscle effects for these areas are substantial. Muscle tensing in these regions has either not been sufficiently characterized or is difficult to implement without establishing a musculature system that ensures whole-body equilibrium prior to the crash event.

As noted above, the imminent occurrence of a collision frequently generates conscious protective behaviors (i.e., avoidance maneuvers, braking and bracing) provided sufficient time is available prior to the collision (Klopp et al., 1995). These muscle activation levels may vary during the impact event with additional transient increases provided by reflexive responses. Begeman et al. (1980) identified lengthy delays in reaction of more than 200 ms in frontal crashes with volunteers although onset latencies and peak muscle activity have been show to be closer to 100 ms in rear impact tests with volunteers (Brault et al., 2000; Siegmund et al., 2004). While these times indicate that pre-impact bracing will influence occupant kinematics only in higher speed frontal (e.g., Klopp et al., 1995; Pithioux et al., 2005) and in near-side impacts, reflexive contributions during the impact event could occur in certain far side scenarios (e.g., Meijer et al., 2008), rear impacts (e.g., Siegmund et al., 2004) and certainly in rollover events (e.g., Hu et al., 2008).

The consequences of not incorporating muscular effects into design and evaluation of vehicles and countermeasures are potentially significant. At least half of all occupants engage in emergency bracing prior to a frontal collision (Ore, 1992). While muscle activation can potentially provide an alternate means of restraint and energy transfer (i.e., propriotonic restraint – Armstrong et al., 1968) that results in significant reduction in injuries, simulation studies have shown that these effects

unequally influence the distribution of the crash forces within various body regions (Bose and Crandall, 2008). This implies that assumptions considering muscle tensing forces to be always protective are not valid and only through a characterization and understanding of their influence can the potential benefits of these actions be realized. Furthermore, this finding suggests that advanced and adaptive restraint systems will ultimately adjust the restraint characteristics to account for musculature effects as one of the control parameters (Bose and Crandall, 2008).

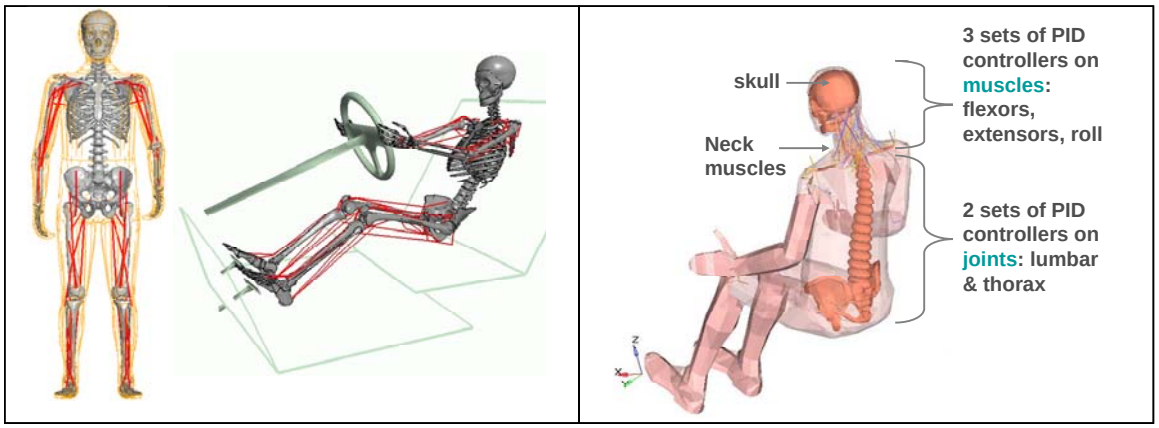

Figure 9 - Human model with skeleton and muscles (Choi et al., 2005), Multi-body model with neck muscle control and stabilizing spine (Fraga et al., 2009)

Given the significance on injury response and the dynamic nature of the effects, computational models of the human body must ultimately involve neuromuscular representations with detailed muscle structures (Figure 9) as well as a behavioral law that describes the individual muscular contributions during the pre- and post-impact events. Recent advances in modeling muscle with the finite element framework have shown the ability to simultaneously address the kinematic, geometric (e.g., wrapping), and material (i.e., nonlinear anisotropic) complexity of muscle (Blemker and Delp, 2005; Blemker et al., 2005). The traditional optimization techniques used for gait and sports analyses (i.e., equilibrium, minimization of total muscle forces, minimization of fatigue factors, minimization of joint torque, and minimization of energy) and associated constraints (Buchanan and Shreeve, 1996) are unlikely to be capable of fully explaining the distribution of forces experienced in crash events. While most studies evaluate the pattern of muscle activities by EMG, EMG signals represent the tendency and regulation of muscle activities in a relative manner rather than as true muscle forces. The fact that many muscles cross a joint often leads to an under-determined problem with an infinite number of possible force patterns that can produce the required forces and moments Herzog and Leonard, 1991 W. Herzog and T. Leonard, Validation of optimization models that estimate the forces exerted by synergistic muscles, *Journal of Biomechanics* **24** (1991), pp. 31–39. **Article** | PDF (968 K) | View Record in Scopus | Cited By in Scopus (63)(Herzog and Leonard, 1991). Solving this problem may require future non-invasive methods of directly measuring muscles forces exerted by all individual muscles comprising a synergistic group.

In order to evaluate human-machine interactions associated with crash avoidance and pre-impact occupant behavior, feedback control models will be required. Recent innovations in modeling have allowed position-control strategies to be implemented in evaluations of motorcycle rider behavior (Fraga et al., 2009). Coupling neuromuscular control laws with realistic architectural models of musculature should provide internal and external biofidelity of models necessary to represent the effects experienced by occupants in real world collisions.

PRE-EXISTING MEDICAL CONDITIONS: The presence of preexisting medical conditions in occupants has a demonstrated association with increased mortality and mortality independent from age (Wutzler er al, 2009). Examples of pre-existing medical conditions shown to be risk factors include arterial occlusive disease, heart disease, hepatitis/liver cirrhosis, carcinoma/malignant disease, coagulation disorder, obesity, cardiopulmonary disease, and diabetes (Wutzler et al. 2009; Broos et al.,

1993). Mackay (2007) notes that even daily variations in the state of organs (e.g., stomach contents, bladder volume) or presence of alcohol, drugs, and pharmacologic agents can influence injury potential. Once these effects have been characterized in terms of biomechanical and biochemical parameters, there is theoretically no impediment to the inclusion of these or other conditions in computational models of injury.

MODELING OF RESPONSE AND FAILURE OF TISSUE

Current concepts in injury assessment using FE models focus on the mechanical failure of tissues according to ultimate strain or stress criteria, or combinations thereof (e.g., Von Mises criteria). State-of-the-art failure models in bone typically involve a relatively simple elastic-plastic response curve with rate sensitivity occasionally implemented. Damage is normally represented by element elimination rather than a formal crack propagation approach due to problems inherent with mesh-dependent damage growth and the requisite time steps. Meanwhile, failure of bone under experimental impact loading is now characterized using instrumentation capable of discerning the initiation of microfracture as well as the cumulative effects of microdamage prior to a global peak load (e.g., Salzar et al., 2006; Salzar et al., 2008).

Relatively speaking, failure representation in hard tissue is well characterized in comparison to that used for soft tissues. Most soft tissue representations in computational models use an elastic or viscoelastic constitutive model that remains invariant during the impact event; that is they do not provide for alterations in response or structure at either the micro- or macro-level. The importance of simulating nonrecoverable changes in soft tissues at finite strains has been documented in experimental studies, using the term strain conditioning. Strain conditioning alters the viscoelasticity of the tissue and involves decreases in the magnitude of the linear complex modulus and corresponding changes in phase values. For this reason, strain conditioning is sometimes referred to as strain softening. These strain conditioning effects have been shown both at the material property level as well as the cellular level in brain testing of an animal model (Darvish and Crandall, 2002). Poroelastic and poroviscoelastic models have been used extensively for modeling of tissues such as cartilage (e.g., Mow et al., 1980), and when coupled with an appropriate damage term (e.g., a viscous flow – see van Dommelen et al., 2006) may be capable of characterizing the nonrecoverable behavior of soft tissues. In order to reflect accurate stresses and strains within soft tissues during an impact event, FE models of soft tissue must move beyond the current practice of constant material properties and must implement these damage and softening effects.

Models of poroelasticity highlight the importance of body fluids in distributing forces within and around tissues. For head injuries, the importance of modeling the behavior of the cerebrospinal fluid as a liquid and its effect on strain distributions in the brain has been demonstrated computationally (e.g, Darvish and Crandall, 2002). Lee (2008) demonstrated the influence of not only fluid pressure but also shear stresses at the vascular wall interface in predicting traumatic aortic rupture. Given the proportion of tissues that are fluid, modeling of fluid-solid interaction provides an open area of study that may lend insight into soft tissue response and failure. In fact, modeling at this level will likely elucidate complex injury mechanisms that cannot be replicated, viewed, or measured experimentally.

While the ultimate application of more advanced constitutive models will be in FE applications, modeling techniques may in themselves provide a vehicle for determining more advanced material models. Many tissues involve complex shapes with residual stresse states that do not make them amenable to in vitro testing using specimens machined as standard engineering coupons. Meanwhile, testing in vivo or in vitro with intact organs presents complex geometries and boundary conditions that are not easily evaluated analytically. One approach to determine material parameters in these cases is to establish an inverse-problem of parameter identification. FE calculations of the intact organ experiments are varied to match the measured boundary conditions describing the global structural behavior along with any available direct measurements of tissue strain. An optimization tool is used to adjust the material parameters in an effort to minimize the error between the simulated and measured responses (e.g., Mukherjee et al., 2007). Through use of models of the tissue tests, the inverse-FE methodology holds an exciting potential for obtaining more accurate material properties of biological models needed for development of advanced FE human models.

While increasing complexity can be introduced into the constitutive laws, biomechanists must not lose sight of the fact that tissues and cells are living structures with inherent biophysical and biochemical processes. One of the underexplored areas of modeling involves simulating the process of failure, damage, and death of tissue as sequences of coupled mechanical and chemical events. While the process of incorporating physiologic functions into mechanical models is undoubtedly challenging, researchers in injury prevention can learn from work that has been done in other areas of biomechanics. In particular, the processes of bone remodeling (e.g., Carter and Beaupre, 2001) as well as pathological manifestations of tissue such as the growth of aneurysms in vessels (e.g., Ryan and Humphrey, 1999) have been studied with FE modeling and can provide insight into the potential for modeling the physiologic progression of injury. Since deformation can activate pathophysiologic cascades below failure tolerances, this modeling will inherently involve modeling over scales of several orders of magnitude to include responses such as hemorrhage, inflammation and cell death and dysfunction with coupling of simulations at the tissue, cellular, and even molecular level. Biomedical researchers (e.g., Cater et al., 2006) are already developing mathematical relationships between engineering parameters and cellular death as a function of time. These functions can be incorporated into finite element models to add another dimension beyond a single failure threshold to predict the severity of injury. While this will necessitate increased collaborations between clinicians and biomechanical engineers, the recent findings that injury outcome and tolerance are dependent on genetic type (Friedman et al., 1999; Gennarelli et al., 2003) suggests expansion of current collaborations to include basic scientists in the areas such as systems and molecular biology, biophysics, biochemistry, and molecular immunology. Once the complexities of the injury process are characterized, the potential exists not only to identify the causal mechanisms and processes of injury but to evaluate appropriate interventions and therapies through simulation.

MODELING SCALE

Biological materials are hierarchical in structure with a temporal component to their development, proliferation, damage, and death. Thus, injury biomechanics must potentially bridge length and time scales of several orders of magnitude in order to characterize the response and failure of tissues. Whereas the upper limit of scale for human models is readily defined at the level of the whole body, the lower of the scale of problems needs to be carefully considered by the modeler depending on the question at hand. Within each level of hierarchy, the scales lead to different modeling challenges and techniques with differing associated computational costs and benefits (Fung, 2003). Multi-body modeling is the least computationally intensive approach and may be sufficient if whole body kinematics is the primary interest. FE modeling is likely required if tissue-level injury prediction is desired. For modeling of processes at sub-mm or smaller, micro-mechanics or even nano-mechanics must be considered. Similar to the anatomical hierarchy, the timeframe of interest can range from microseconds or milliseconds for blast and impact events to hours, days, or even weeks if the progression of injury cascades and outcome is to be simulated. While there may be a tendency to relate decreasing length and times scales to increased utility, the question of cost-benefit must be considered especially in light of the fact that the quality and availability of data diminish at lower length scales and both extremes of the time scales. These limitations combined with computational expense advocate for an evaluation on the part of the modeler as to the choice the appropriate model for addressing the specific problem of interest. Longer term, injury modeling will likely entail coupling and interactions of models of varying time and length scales.

Multi-body models have been integral to the study of injury biomechanics due to their efficient balance of accuracy and computational cost. Using a combination of inertial and geometric representations of the body, structural response characteristics, and nonlinear contact algorithms, these models are readily applicable to variational studies of the occupant population that involve scaling and morphing of anatomical structures (e.g., Bose et al., 2008). Similarly, these models allow efficient implementation of simple line models of muscle tensing and the associated passive-active behavior. Their efficiency and relative ease-of-use will ensure they remain an essential component of current restraint and vehicle design efforts, particularly in cases where optimization and parameter sensitivity analysis are involved.

The lack of failure descriptions at the tissue level, however, limits the utility of the multi-body models since they have to rely on implementation of structural injury criteria, many of which have been tailored to dummies rather than computational models. Furthermore, using of these criteria precludes the use of multi-body models in developing improvements and extrapolations of already established criteria. As noted previously, finite element models represent tissue level response using governing equations based on the concepts of continuum mechanics. In addition, they facilitate digital expression of the three-dimensional geometry and topology of tissue structures with an adjustable level of accuracy. This combination of anatomical and material details allows the complex topology and boundary conditions to be modeled; the inhomogeneous, anisotropic and non-linear materials to be simulated, and the stress distributions in the various tissue structures to be delineated.

Sorely lacking from the current biomechanics database for FE model development and validation is a statistical framework for handling variability within the populations of human data. This shortcoming exists both within the structural properties used for validation of global response of the body and components as well as within the material properties. Validation and development of future models will require that the variance inherent in experimental data is characterized and embraced with statistical quantification of not only the structural and material parameters but relevant covariates of age, size, gender, and other factors. Specifically, utilization of mean values for any material and structural parameters or even time-history responses is woefully inadequate and improved characterization of response corridors, including statistically valid confidence intervals derived from both normalization and signal analyses, will be necessary for accurate presentation of experimental results.

The arbitrary use of any set of mean material data belies the fact that the response of the computational models themselves are dependent on mesh, geometry and numerical factors. For a given model and mesh, the choice of material parameters must consider this inherent variability and allow for variation within a range of the experimental data (e.g., confidence intervals). In addition to developing statistically-based corridors from normalized sets of experimental data, systems of evaluation (i.e., rating) systems must develop methodologies that are not based on subjective weightings of model response parameters relative to experimental curves, peak values, or times (e.g., Jacob et al., 2000), but rather assess the model's ability to reflect the underlying response phenomena observed in the experiments using assessments of causal variables with prioritized weightings determined from sensitivity analyses.

Given the lack of experimental constitutive data and three-dimensional material models, finite element model formulations frequently assume the tissue stress–strain response is linear, despite the fact that many tissues undergo finite deformations with nonlinear responses during typical impact events. Even outside the impact environment, the validity of this assumption is questionable as virtually all tissues exhibit a nonlinear constitutive response (Fung, 1993). The nonlinear effects in the constitutive relation of soft tissue are of particular importance since they are more accurate for finite deformations (Darvish and Crandall, 2001). In addition, waves in response to a step-wise input are diffusive and broaden in materials with memory that behave according to linear theory. Conversely, materials that behave according to nonlinear theory can develop self-preserving waveforms as well as discontinuities in the form of shock or acceleration waves in response to a step-wise input (Spence, 1973). Clearly, temporal and spatial nonlinear models must be experimentally determined and implemented into FE codes if accurate injury prediction is to be achieved.

Since response of the model represents an interplay between geometry and material properties, standard finite element schemes are mesh dependent with stress levels dependent on the mesh density. FE models that lack detailed anatomical structures show that the tissue level stresses and strains that are predicted from these models do not represent the actual tissue stresses and strains (e.g., Cloots et al., 2007). Representation of tissue behavior without knowledge of the underlying structure and function has the potential to result in phenomenological constitutive models that are over-parameterized (Fernandez et al., 2006) and do not adequately represent the tissue response over a robust range of conditions. As an added complication, modelers face a range of challenges when trying to capture the complexities that arise in tissues due to anisotropy and inhomogeneity. The anisotropies primarily arise from fibers (e.g., collagen, elastin, axons) forming a highly organized microstructure or the presence of larger structures (e.g., vasculature) within the surrounding tissue. Representations of the anisotropy (Untaroiu et al., 2005) and the heterogeneity influence the levels of

stress predicted and the injury tolerances and mechanisms. Near term progress in improving the accuracy and distribution of stresses within the tissues will likely involve multi-scale coupling of models at the tissue and micro-structural levels. The invention of micro-computed tomography (μCT) will enable more detailed visualization and measurement of cortical and trabecular structures which in turn can be implemented into micro-FE models. From the standpoint of population variations, statistical variation in material geometry and orientation has received considerably more attention at the micro-structural level (e.g., collagen fiber orientation in soft tissues as an explicit factor in material models) than in phenomenological models at either the structural or material levels. In particular, models have been developed that include volume fractions (e.g., Medvedev et al., 2006), fiber orientations (e.g., Thomopoulus et al., 2005), geometry (Lucas et al., 2009), and degrees of undulation (e.g., Karami et al., 2009). At even lower length scales, research in cellular mechanotransduction may ultimately provide insight into the cellular response of stress and strain that initiates the injury process.

While it is established that forces at higher scales influence behavior at lower scales and that lower-scale properties influence higher-scale response, these relationships and interactions are rarely included explicitly in computational models (Tawhai et al., 2009). Furthermore, researchers have shown that structural responses of tissues can be equally matched by a variety of constitutive models but this global validation can result in wide variations in performance at the material level (Untaroiu et al., 2006). In injury biomechanics, the coupling of multi-scale models is relatively novel (Cloots et al., 2008). In order to establish hierarchical links between and across the spatial scales, multi-scale modeling will require more intricate representation of material response within scales and interactions among scales. This will require the development of scale models based on different physical behaviors involving different frameworks of mathematical approaches. While representation of microstructural response in FE is done for small pieces of isolated tissue, increased computational power and custom (i.e., non-commercial) codes must be developed if whole-body or multi-organ loading is to be evaluated. If simulation studies within injury biomechanics will entail simultaneous interactions across spatial scales including cellular, tissue, organ, and multi-organ systems, a computational framework in which to link and to interpret these multi-scale results must be generated.

MODEL COMPELXITY

INTERIM APPROACHES: Given the current levels of uncertainty in prescribing material processes, an interim process of injury evaluation is in use which augments experimentally measured results through a finite element interpretation of the data. Eppinger (2000) refers to this class of modeling as Computer Aided Trauma Evaluation (CATEv), of which the simulated injury monitor (SIMon) finite element head model is probably the best known example (Takhounts et al., 2008). For the CATEv approach, a given geometry and mesh are utilized for the model based on either a given individual or, preferably, a representative topology based on shape analysis. Given this geometry and mesh, constitutive properties and provisional injury criteria are developed by calibrating the models with existing experimental observations. Uncertainties in material properties, geometry, and numerical parameters are handled through the calibration effort with values prescribed or bounded by constraints of the experimental data. Similar to other types of phenomenological descriptors, the viability of interpolation and/or extrapolation of these results within and beyond the realms of validation data are highly dependent on the quality and breadth of the experimental data. This approach provides for significant advancements over conventional structural descriptors of injury used in dummies and permits and permits tissue level evaluations of injury without having to model the complexity of interactions involved between occupant and vehicle environment since these are measured directly. As such, it may provide a reasonable balance of efficiency and accuracy for shorter-term applications of modeling. By reflecting contributions of each model parameter to the injury prediction, it lends itself to identifying the variables which need to be more thoroughly characterized (e.g., human material parameters) or controlled (e.g., vehicle and design variables) relative to those which may prove insensitive.

Surprisingly, the use of FE models to examine the quality of the experimental data used in its development and validation has received relatively little attention. Murakami et al. (2006) preformed an interesting parametric analysis of how the extrinsic experimental factors, including boundary

conditions, influenced the resulting response curves in thoracic testing. Soni et al. (2008) performed a similar study using a FE model of a pedestrian to evaluate the effects of varying boundary conditions in dynamic tests. Assuming reasonable biofidelity for these models, analyses of this type would ideally be performed prospectively rather than retrospectively so that experimental testing could refine the measurement and precision of relevant extrinsic factors. Philosophically, both the CATEv and parametric analysis technique address the problem that despite the level of sophistication involved in experimental procedures, none of the experimental or biological parameters can be known exactly and must be represented by a distribution. This distribution can arise from natural variation among individuals or within individuals (e.g., preconditioning of tissues or spatial variation of properties within the same tissue) or from random and systematic errors introduced through the modeling process itself. Despite the significant uncertainties that exist in material properties, geometric configuration and loading conditions, finite element models of the human body frequently assign unique material, geometric, and numerical parameters without recognition of the consequences. In addition to the underlying uncertainty of the experimental input data, the response of biomechanical models constructed to represent these systems will also contain uncertainty. This uncertainty can be attributed to inherent randomness in physical parameters and to both random and systematic errors introduced through the modeling process. Introducing specific values without regard to an understanding of the variability inherent in the underlying data, essentially accurate implementation of inherently inaccurate data, can lead to erroneous conclusions being drawn from the model output. Similar to design of experiment approaches used in laboratory investigations, a more effective approach would utilize computational investigations whereby parameters of interest and levels of interaction are evaluated.

STOCHASTIC MODELING: Probabilistic analysis provides a methodology allowing computer models to capture the uncertainties as well as the natural variation present in biological data (e.g., Thacker et al., 1997; Thacker et al., 1998). The principle underlying stochastic methods is that the input parameters of the model are defined, not by a single value, but by a statistical distribution. Due to unavoidable inaccuracies of loading, boundary conditions and measurement devices, experimental data used for model development and validation includes errors.

The values of each input parameter are sampled randomly from the appropriate distribution and used in the model. The model is solved many times to build up a distribution of the output of interest. One advantage of obtaining a distribution over the single value obtained from a deterministic model is that confidence limits, giving an indication of the spread of the response, can be estimated. The sensitivity and the confidence of the calculated parameters represent important information for those using the model output to evaluate and design countermeasures. In addition to providing a more accurate representation of the model output, this information should permit more accurate comparisons and validation using experimental and field data since statistical measures of equivalence can be evoked. An interesting consequence of the probabilistic approach is that the most likely response can be determined and this response will not necessarily be the same as the mean value (Marczyk, 1999). Clearly, progress in probabilistic models depends critically upon the continued development and refinement of validated deterministic models and the collection and synthesis of data as probabilistic inputs. The calculation of probability of injury requires that all potential injury mechanisms and their potential interactions be included in the probabilistic model. For a complex system, these would require the simulation of hundreds of injury mechanisms and thousands of interactions. While probabilistic analyses potentially present several challenges, including significant computational time due to the many trials required, they may represent the best option for capturing and quantifying the variation in response and injury present in the human population.

APPLICATION: While the creation of models with phenonmena across multiple scales and with populational variations will be significant technical achievements, the metric of success for modleing must be ultimately measured in terms of injuries prevented and lives saved. Specifically, the computational studies must translate to innovative information, tools, and countermeasures that permit tackling the problems in a wholistic fashion reflective of the actual traffic safety environment. Philosohically, the process must employ a “workstation to roadside” approach with biomechanical engineers and designers providing new injury countermeasures created through simulations for implementation in the field. In turn, feedback must occur in the reverse direction (i.e., “roadside to workstation) with epidemiologists and crash reconstructionists making novel observations about the

effectiveness of these countermeasures in a contemporary collision environment and providing this information to the modelers and designers for continued model refinements.

Given the abilitiy of the models to provide the requisite scales and estimates of population variation, the opportunity exists to develop interventions at the level of both a population and an individual. While the metrics may be different at the two levels (i.e., minimization of injury, cost, or disability at the individual vs. societal level), customization of protection for the individual implemented in a sufficiently widespread manner will ultimately lead to overall reductions in the road user population. Rather than differences in approach, the difference between occupant and population protection will likely exist in establishing the priorities for implementation of countermeasures since the progress will be incremental and cost-benefit analyses must be employed.

Given a set of parameters that describe the vehicle, occupant, and collision, restraint optimization for the individual can be readily performed using optimization techniques that minimize both regional and whole body metrics of injury. This analysis can be done for passive systems that simply manage the energy of the occupant and crash or active systems that introduce energy into the system. The effectiveness of these countermeasures will be dependent on the ability of sensor information with in the vehicle to assess the state of the occupant (e.g., position, biomechanical tolerance, etc.) and the crash event (e.g., Bose et al., 2008). In addition to variability of the underlying crash and occupant populations, enhanced sensing will introduce additional uncertainties in the system. In addition to quantifying these uncertainties, the optimization process must involve a strategy to predict future actions from current information (e.g., a very basic example of this concept is the fire - no fire decision algorithms for air bags). These predictions will allow tailoring of adaptive restraint systems that can synthesize past, current, and future, albeit estimated, data. Finally, the uncertainties and predictions inherent in these systems will introduce errors and biases that, in the case of traffic safety, may directly relate to unintentional injuries sustained. Thus, a quantitiatve methodology for evaluating and weighting tradeoffs at both the individual and population levels needs to be developed.

COMPLEXITY: In modeling, one of the most difficult questions arises out of a need to balance answering the questions at hand relative to the required complexity of model. Obviously the balance between the predictive and descriptive capabilities of the model relative to and the computational cost requires interpretative knowledge of modeling tools as well as judgment and experience. While considerable effort in this paper has highlighted the ability of models to expand in both breadth and depth the knowledge base in injury biomechanics, the objective is not to develop more complex models or to explore new length and time scales simply as an academic exercise. In reality, parametric studies that incorporate variables provide the only method of accurately assessing whether or not they are significant and can be excluded from future studies. In matching utility with complexity, it is important to remember that all modeling is an approximate solution technique used to solve a mathematical model that is itself an idealized representation of the actual physical problem being studied (Wayne, 2008). At best, our finite element and multi-body models approximate the response of the actual human based on a set of simplifying assumptions. At worst, they provide accurate representation of inaccurate data. Above all, models must allow flexibility depending on the application and problems of interest and the judgment of the modeler. The statistician George Box concluded 'all models are wrong, some are useful' (Box, 1979) and this sentiment can be rightly applied to simulations of the human body with the realization that their utility may provide the most viable path forward for addressing the breadth and depth of the traffic injury problem.

CONCLUDING REMARKS

The compelling case for modeling has been developed in light of the complexity of the injury process and the variation among individuals and within populations of individuals. While current models lack the detail and biofidelity necessary to achieve their envisioned potential, the path forward can be reasonably envisioned. Empirically-derived injury criteria that lack a solid scientific foundation must be systematically replaced with injury estimates derived from simulations based on causal mechanisms and the principles of mechanics. Similarly, researchers must no longer be content to model an "average" or "typical" system but must embrace population variation as a necessary component. This combination of mechanical principles and statistical descriptions will create

opportunities for hypothesis-driven research in a field where it has been previously nonexistent. To develop and test these hypotheses, the research must include multi-disciplinary research collaborations using "team science" (Viceconti, 2008). Thus, future advances in modeling are inextricably bound to the continued evolution of experimental biomechanics, crash data analysis, epidemiology, and clinical studies. Fortunately, the field of injury prevention has been multi-disciplinary since its inception and is well positioned to accept computational modeling as a new player. The broader challenge will be for researchers in these complementary disciplines to come to the realization that their own research activities in traffic safety should be guided and directed by the needs of and the output from the computational models. In the area of experimental injury biomechanics, simulation technology will generate opportunities for extrapolation of test data and enhancement of laboratory techniques. It is clear that the level of detail measured in experiments must increase in order to provide the requisite response and boundary conditions necessary for model development and validation. However, this increased detail will undoubtedly lead to increased cost and time of experiments which in turn may have the paradoxical consequence of reducing the number of tests conducted unless the overall objectives of both increased detail and statistical representation are emphasized. Even with increasingly sophisticated experimental methodologies, limitations exist on being able to characterize physiologic processes (e.g., muscle tensing, blood flow, and injury cascades) and to measure stress and strain patterns in vivo for complex anatomical structures. Many of the most exciting challenges for improved injury prediction through modeling lay at the boundaries between length scales (e.g., micorstructure-tissue interactions), between sub-systems (e.g., vascular and nervous), and between time scales (e.g., injury cascades). To examine these problems, computational modeling will have to grow as a field with advances in the design of algorithms and code that improve the performance of simulations including the time and accuracy of simulations. Similarly, universally accepted criteria for model validation and acceptance will need to be developed.

The physicist Heinrich Hertz (1894) stated that "it is the first, and in a way the most important task of science to enable us to predict future experience, so that we may direct our present activities accordingly." Even to those of us who are not modelers, the utility of computational models to advance the field of injury prevention should be unquestionable. Beyond simple acceptance of the benefits of computational modeling, safety professionals must accept that piecemeal incorporation of modeling into future injury reduction strategies will not be sufficient to realize targets as ambitious as Vision Zero given the complexity of the injury problem. In reality, the ultimate realization of modeling's potential will require a fundamental transformation in the field of injury prevention by which simulation methodologies are essential components of planning, design, and evaluation and influence the activities of the entire traffic safety community. In summary Vision Zero must be virtually simulated before it is physically realized if every individual on the road is to be protected.

ACKNOWLEDGMENT

The author would like to thank the staff and researchers at the University of Virginia Center for Applied Biomechanics for their contributions to the author's work cited in this paper. Additionally, the author would like to acknowledge Costin Untaroiu, Richard Kent, Maria Segui-Gomez, Tim Gillispie, and Jason Kerrigan for their thorough review and insightful comments.

REFERENCES

Abbreviated Injury Scale Update (1998) The Abbreviated Injury Scale- 1990 Revision (Update 98). Des Plaines, IL, AAAM.

Adomeit, HD,Wils, O,Heym, A. (1997) Adaptive Airbag-Belt-Restraints - An Analysis of Biomechanical Benefits. Paper 970776, Society of Automotive Engineers.

Armstrong, RW, Stapp, JP, Waters, HP. (1968) Human Muscular Restraint During Sled Deceleration. Stapp Car Crash Conference Proceedings, Paper 680793.

Azouz, ZB, Rioux, M, Shu, C, Lepage, R. (2006) Characterizing human shape variation using 3D anthropometric data. Visual Comput 22, 302–314.

Bhalla, KS, Montazemi, P, Crandall, JR, Yang, J, Liu, X, Dokko, Y, Takahashi, Y, Kikuchi, Y, Longhitano, D. (2002) Vehicle impact velocity prediction from pedestrian throw distance, Trade-offs between throw formulae, crash simulators, and detailed multi-body modeling. IRCOBI Conference on the Biomechanics of Impact.

Bingley, L, Morris, R, Cross, G. (2005) Determination of real world occupant postures by photo studies to aid smart restraint development. International Technical Conference on the Enhanced Safety of Vehicles (ESV), 19.

Blemker, SS, Delp, SL. (2005) Three-Dimensional Representation of Complex Muscle Architectures and Geometries. Annals of Biomedical Engineering, 33(5): 661-673.

Blemker, SS, Pinsky, PM, Delp, SL. (2005) A 3D Model of Muscle Reveals the Causes of Nonuniform Strains in the Biceps Brachii. J Biomech, 38(4): 657-665.

Bose, D, Crandall, JR, Untaroiu, CD, Maslen, E. (2008a) Influence of pre-collision occupant properties on the injury response during frontal collision. IRCOBI Conference on the Biomechanics of Impact.

Bose, D, Crandall, JR. (2008) Influence of Active Muscle Contribution on the Injury Response of Restrained Car Occupants. Annals of Advances in Automotive Medicine, 52, 61-72.

Box, GEP (1979) Robustness in the strategy of scientific model building. in Robustness in Statistics. eds Launer, RL, Wilkinson, GN. New York: Academic Press.

Broos, PL, D'Hoore, A, Vanderschot, P, Rommens, PM, Stappaerts, KH. (1993) Multiple trauma in patients of 65 and over. Injury patterns. Factors influencing outcome. The importance of an aggressive care. Injury, 24(6): 365-8.

Buchanan, TS, Shreeve, DA. (1996) An Evaluation of Optimization Techniques for the Prediction of Muscle Activation Patterns During Isometric Tasks. J Biomech Eng, 118(4): 565-574.

Carter, DR, Beaupre, GS. (2001) Skeletal Function and Form, Mechanobiology of Skeletal Development, Aging, and Regeneration. New York, Cambridge UP.

Carter, DR, Hayes, WC. (1977) The compressive behavior of bone as a two-phase porous structure. J Bone Joint Surg Am, 59-A: 954-962.

Cavanaugh JM, Zhu Y, Huang Y, King AI. (1993) Injury and Response of the Thorax in Side Impact Cadaveric Tests. Stapp Car Crash Conference Proceedings, Paper 933127.

Choi, HY, Sah, SJ, Lee, B, Kang, S, Mun, MS, Lee, I, Lee, J. (2005) Experimental and numerical studies of muscular activations of bracing occupant. Paper 05-0139, International Technical Conference on the Enhanced Safety of Vehicles (ESV), Proc. 19th ESV.

Choi HY, Han J, Park Y, Kyung HY, (2009) Digital Elderly Human Body Modeling, SAE, Paper 2009-01-2262, Society of Automotive Engineers.

Cloots, RJH, Gervaise, HMT, van Dommelen, JAW, Geers, MGD. (2008) Biomechanics of Traumatic Brain Injury, Influences of the Morphologic Heterogeneities of the Cerebral Cortex. Annals of Biomedical Engineering, 36(7): 1203–1215.

Combest, J. (2009) Status of the GHBMC –Developments with the Six Centers of Expertise, SAE Government Industry Meeting. (http,//www.sae.org/events/gim/presentations/2009/johncombest.pdf)

Cormier JM. (2008) The Influence of Body Mass Index on Thoracic Injuries in Frontal Impacts. Accid Anal Prev, 40(2): 610-615.

Darvish, KK, Crandall, JR. (2001) Nonlinear Viscoelastic Effects in Oscillatory Shear Deformation of Brain Tissue. Medical Engineering and Physics, 23: 633-645.

Darvish, KK, Crandall, JR. (2002) Influence of Brain Material Properties and Boundary Conditions on Brain Response during Dynamic Loading. IRCOBI Conference on the Biomechanics of Impact.

Darvish, KK, Crandall, JR. (2002a) Influence of Brain Material Properties and Boundary Conditions on Brain Response during Dynamic Loading. IRCOBI Conference on the Biomechanics of Impact.

Darvish, KK, Crandall, JR. (2002b) The Effect of Axonal Injury on the Material Properties of Brain Tissue. Proceedings of the 4th World Congress of Biomechanics, Calgary, Canada.

Darvish, KK, Stone, J. (2004) Changes in Viscoelastic Properties of Brain Tissue Due to Traumatic Injury. Paper IMECE2004-60849, ASME International Mechanical Engineering Congress and RD&D Congress.

Dressler, M, Butler, D, Wenstrup, R, Awad, H, Smith, F, Bovin, G. (2000) A Mechanism for Age-Related Decline in Tendon Biomechanics. 46th Annual Meeting, Orthopaedic Research Society, March 12-15, Orlando, Florida

Ejima, S, Ono, K, Holcombe, S, Kaneoka, K, Fukushima, M. (2007) A Study on Occupant Kinematics Behaviour and Muscle Activities during Pre-Impact Braking Based on Volunteer tests. IRCOBI Conference on the Biomechanics of Impact.

Elmistekawy, E, M, Hammad, A. A. M. (2007) Isolated rib fractures in geriatric patients Annals of Thoracic Medicine, 2(4): 166-168.

Eppinger, R, Marcus, J, Morgan, R. (1984) Development of Dummy and Injury Index for NHTSAs Thoracic Side Impact Protection Research Program. Paper 840885, Society of Automotive Engineers.

Eppinger, R. (2000) A Biomechanical P.E.E.E.P. Show. in Human Biomechanics and Injury Prevention, editors, Kajzer, J, Tanaka, E, Yamada, H. Tokyo, Springer.

Eppinger, R. H. (2002) Occupant Restraint Systems in Accdidental Injury, Biomechanics and Prevention edited by A. M. Nahum and J. W. Melvin, 2nd Edition, Springer.

Evans L (1990) Restraint effectiveness, occupant ejection from cars and fatality reductions, Accident Analysis and Prevention, 22: 167-175.

Fernandez, JW, Scmid, H, Hunter, PJ. (2006) A Framework for Soft Tissue and Musculoskeletal Modeling, Clinical Uses and Future Challenges. in Mechanics of Biological Tissue. eds, Holtzapfel, GA, Ogden, RW, New York, Springer.

Figuers, CC. (2005) Injury Prevention and Rehabilitation for Active Older Adults edited by K. P. Speer, Human Kinetics Publishers.

Flannagan, CAC, Manary, MA. (1998) An Improved Seating Accommodation Model with Application to Different User Populations. SAE International Congress & Exposition. Paper 980651, Society of Automotive Engineers.

Fraga, F, van Rooij, L., Happee, R, Wismans, J, Symeonidis, I, Pledschus, S. (2009) Development of a Motorcycle Rider Model with Focus on Head and Neck Biofidelity, Recurring to Line Element Muscle Models and Feedback Control. Paper 09-0244, International Technical Conference on the Enhanced Safety of Vehicles (ESV), 21.

Friedman, G, Froom, P, Sazbon, L, Grinblatt, I, Shochina, M, Tsenter, J, Babaey, S, Yehuda, B. Groswasser, Z. (1999) Apolipoprotein E-4 genotype predicts a poor outcome in survivors of traumatic brain injury. Neurology 52(2): 244-8.

Fung YC. (1993) Biomechanics , mechanical properties of living tissues. New York, Springer-Verlag.

Fung, Y.C. (2003) Keynote Lecture to the WCCBME Biomechanical Engineering's Many Foundations. in Frontiers in Biomedical Engineering, eds. Hwang, NHC, Woo, S. New York, Kluwer.

Funk, JR, Crandall, JR, Tourret, L, MacMahon, C, Bass, CR, Khaewpong, K, Eppinger, R. (2002) The Axial Injury Tolerance of the Human Foot/Ankle Complex and the Effect of Achilles Tension. J Biomech Eng, 124, 750-757.

Gayzik, FS, Yu, MM, Danelson, KA, Slice, DE, Stitzel, JD. (2008) Quantification of Age-Related Shape Change of the Human Rib Cage Through Geometric Morphometrics. J Biomech, 41(7): 1545-1554.

Gennarelli, TA, Pintar FA, Yoganandan, N. (2003) Biomechanical Tolerances for Diffuse Injury and a Hypothesis for genetic type variability in response to trauma. Annual Proceedings/Association for the Advancement of Automotive Medicine (AAAM).

Gordon, C, Churchill, T, Clauser, C, Bradtmillter, B, McConville, J, Tebbetts, I, Walker, R. (1988) Anthropometric Survey of U.S. Army Personnel, Summary Statistics Interim Report, NATICK/TR-89/027

Gur, RC, Mozley, PD, Resnick, SM, Gottlieb, GL, Kohn, M, Zimmerman, R, Herman, G, Atlas, S, Grossman, R, Berretta, D. (1991) Gender differences in age effect on brain atrophy measured by magnetic resonance imaging. PNAS 88(7): 2845-2849.

Hansson, T, Roos, R, Nachmson, A. (1980) The Bone Mineral Content and Compressive Strength of Lumbar Vertebrae. Spine, 5(1),46-55.

Hardy, R, Watson, J, Frampton, R, Page, M, Zioupos, P, Cook, R, Kennedy, A, Sproston, P, Forrester, B, Peach, S. (2005) BOSCOS - Development and Benefits of a Bone Scanning System. Proceedings of the International Research Council on the Biomechanics of Impact (IRCOBI).
Herzog, W, Leonard, TR. (1991) Validation of Optimization Models That Estimate the Forces Exerted By Synergistic Muscles. J Biomech, 24(Suppl 1): 31-39.
Hu, J, Yang, KH, Chou, CC, King, AI. (2008) A Numerical Investigation of Factors Affecting Cervical Spine Injuries During Rollover Crashes. Spine, 33(23),2529-35.
Irwin, AL, Mertz, HJ. (1997) Biomechanical Bases for the Crabi and Hybrid III Child Dummies. Stapp Car Crash Conference Proceedings, Paper 973317.
Ito, O, Dokko, Y, Ohashi, K. (2009) Development of Adult and Elderly FE Thorax Skeletal Models. SAE World Congress, 2009-01-0381.
Jacob C, Charras F, Trosseille X, Hamon J, Pajon. M, Lecoz JY. (2000) Mathematical models integral rating. International Journal of Crashworthiness, 5(4): 417-431.
Karami, G,Grundmana, N, Abolfathia, N, Naika, A, Ziejewskia, M. (2009) Micromechanical hyperelastic modeling of brain white matter under large deformation. J of the Mech Behav of Biomed Mat. 2(3),243-254.
Kent, RW, Henary, BY, Matsuoka, F. (2005a) On the Fatal Crash Experience of Older Drivers. Annual Proceedings/Association for the Advancement of Automotive Medicine, 49, 371-391.
Kent, RW, Lee, S, Darvish, KK, Wang, S, Poster, C. Lange, A, Brede, C, Lange, D. (2005) Structural and Material Changes in the Aging Thorax and Their Role in Crash Protection for Older Occupants. Stapp Car Crash J, 49, 231-249.
Kent, RW, Lee, S, Darvish, KK, Wang, S, Poster, C. Lange, A, Brede, C, Lange, D. (2005b) Structural and Material Changes in the Aging Thorax and Their Role in Crash Protection for Older Occupants. Stapp Car Crash Journal, 49: 231-249.
Kent, RW, Lopez-Valdes, F, Trowbridge, M, Ordoyo, R, Segui-Gomez, M. (2009) How Many People Are Injured and Killed as a Result of Aging? Frailty, Fragility, and the Elderly Risk-Exposure Tradeoff Assessed via a Risk Saturation Model. Annual Proceedings/Association for the Advancement of Automotive Medicine. (in press).
Kent, RW, Patrie, J, Poteau, F, Matsuoka, F, Mullen, C. (2003) Development of an age-dependent thoracic injury criterion for frontal impact restraint loading. Paper 72, International Technical Conference on the Enhanced Safety of Vehicles (ESV), 18.
Kleiven, S, von Holst, H. (2001) Consequences of Brain Size Following Impact in Prediction of Subdural Hematoma Evaluated with Numerical Techniques. Proceedings of the International Research Council on the Biomechanics of Impact (IRCOBI).
Klopp, GS, Crandall, JR, Sieveka, EM, Pilkey, WD. (1995) Simulation of Muscle Tensing in Pre-Impact Bracing. IRCOBI Conference on the Biomechanics of Impact.
Kuppa, S, Eppinger, RH, McKoy, F, Nguyen, T, Pintar, FA, Yoganandan, N. (2003) Development of Side Impact Thoracic Injury Criteria and Their Application to the Modified Es-2 Dummy with Rib Extensions (Es-2Re). Stapp Car Crash J, 47: 189-210.
Lee, SH. (2008) A Three-Dimensional Computational Analysis of Blood Flow and Fluid-Structure Interactions in the Human Aorta During Traumatic Rupture Conditions. Ph.D. Dissertation, University of Virginia.
Liu Z, Bilston, L. (2002) Large deformation shear properties of liver tissue. Biorheology, 39,735–742.
Lucas, SR, Bass, CR, Crandall, JR, Kent, RW, Shen, FH, Salzar, RS. (2009): Viscoelastic and failureproperties of spine ligament collagen fascicles. Biomechanics and Modeling in Mechanobiology (in press).
Mackay M, (2007) The increasing importance of the biomechanics of impact trauma, Sadhana, 32(4): 397-408.
Mackay, M, Hassan, A, Hill, J. (1998) Observational Studies of Car Occupants Positions. Paper 98-S6-W-42, International Technical Conference on the Enhanced Safety of Vehicles (ESV), 16.
Manary, M, Reed, M, Flannagan, C, Schneider, L. (1998) Atd Positioning Based On Driver Posture and Position. Stapp Car Crash Conference Proceedings, Paper 983163.
Marshall, R, Sherwood, C, Crandall, JR. (2005) Estimating Head Impact Velocities in Rear Facing Child Restraint Systems, SAE Govt. Industry Meeting.

Medvedev, A, Samsonov, V, Fomin, V. (2006): Rational Structure of Blood Vessels, J. Applied Mechanics and Technical Physics, 47(3),324-329.
Medvedev, A, Samsonov, V, Fomin, V. (2006): Rational Structure of Blood Vessels, J. Applied Mechanics and Technical Physics, 47(3),324-329.
Meijer, R, Rodarius, C, Adamec, J, van Nunen, E, van Rooij, L. (2008) A first step in computer modelling of the active human response in a far-side impact. Int J of Crash, 13(6): 643-652.
Meissner, MU, van Rooij, L, Bhalla, KS, Crandall, JR, Longhitano, D, Takahashi, Y, Dokko, Y, Kikuchi, Y. (2004) A Multi-Body Computational Study of the Kinematic and Injury Response of a Pedestrian with Variable Stance upon Impact with a Vehicle. Paper 2004-01-1607, Society of Automotive Engineers.
Meissner, U, Stephens, G, Al, fredson, L. (1994) Children in Restraints. Proceedings of the Association For the Advancement of Automotive Medicine (AAAM): 38.
Menard, D, Stanish, WD. (1989) The aging athlete. Am J Sports Med. 17(2):187–196.
Miller, HJ, Maripudi, V. (1996) Restraint Force Optimization for A Smart Restraint System. Paper 960662, Society of Automotive Engineers.
Mock, CN, Grossman, DC, Kaufman, RP, Mack, CD, Rivara, FP. (2002) The Relationship Between Body Weight and Risk of Death and Serious Injury in Motor Vehicle Crashes. Accid Anal Prev, 34(2): 221-228.
Montalvo, F, Bryant, RW, Mertz, HJ. (1982) Possible positions and postures of unrestrained front seat children at instant of collision, International Technical Conference on Experimental Safety of Vehicles (ESV), 9.
Mosekilde, L, Mosekilde, L, Danielsen CC. (1987) Biomechanical Competence of Vertebral Trabecular Bone in Relation to Ash Density and Age in Normal Individuals. Bone, 8(2): 79-85.
Mow, VC, Kuei, SC, Lai, WM, Armstrong, CG. (1980) Biphasic Creep and Stress Relaxation of Articular Cartilage in Compression? Theory and Experiments. J of Biomech Eng, 102(1): 73-84.
Mukherjee, S, Chawla, A, Karthinkeyan, B, Soni, A. (2007) Finite element crash simulations of the human body, Passive and active muscle modeling. Sadhana Vol. 32, Part 4, August 2007, pp. 409–426.
Murakami, D, Kobayashi, S, Torigaki, T, Kent, RW. (2006) Finite Element Analysis of Hard and Soft Tissue Contributions to Thoracic Response, Sensitivity Analysis of Fluctuations in Boundary Conditions. Stapp Car Crash J, 50, 169-189.
OECD (2007) OECD Health Data 2007, OECD, Paris.
Ore, LS. (1992). Design requirements and specifications, dummy lower extremity development task. Event Report, NHTSA.
Oskvig RM, (1999) Special Problems in the Elderly, Chest 115:158S-164S.
Pithioux, M, Chavet, P, St Onge, N, Nicol, C. (2005) Influence of Muscle Preactivation of the Lower Limb on Impact Dynamics in the Case of Frontal Collision. Int J of Crash, 10(6): 557-565.
Prasad, P, Chou, C. (2002) A Review of Mathematical Occupant Simulation Models. in Accidental Injury, Biomechanics and Prevention, eds Nahum, AM, Melvin, JW. 2nd Edition, NY, Springer.
Reed, M, Manary, M, Flannagan, C, Schneider, L, Arbalaez, R. (2001) Improved Atd Positioning Procedures. Paper 2001-01-0117, Society of Automotive Engineers (SAE).
Rho, JY, Hobatho, MC, Ashman, RB. (1995) Relations of mechanical properties to density and CT numbers in human bone. Med Eng Phys, 17(5): 347-55.
Robinette, KM, Blackwell, S, Daanen, H, Boehmer, M, Fleming, S. (2002) CAESAR, Summary Statistics for the Adult Population (Ages 18-65) of the United States of America, Final Report, AFRL-HE-WP-TR-2002-0170.
Ryan, JM, Humphrey, JD. (1999) Finite Element Based Predictions of Preferred Material Symmetries in Saccular Aneurysms. Ann of Biomed Eng, 27(5): 1573-1586.
Salzar, RS, Bass, CR, Kent, RW, Millington, SA, Davis, M, Lucas, SR, Rudd, RW, Folk, B, Donnellan, L. (2006) Development of Injury Criteria for Pelvic Fracture in Frontal Crashes. Traffic Injury Prevention, 7,299-305.
Salzar, RS, Genovese, D, Bass, CR, Bolton, JR, Guillemot, H, Crandall, JR. (2008) Load path distribution within the pelvic structure under lateral loading. Proc. ICRASH Conference.
Schneider, LW, Robbins, DH, Pflug, MA, Snyder, RG. (1983) Anthropometry of Motor Vehicle Occupants, Vol 1, Procedures, Summary Findings and Appendices Final Report, UMTRI-83-53-1.

Segui-Gomez, M, Lopez-Valdes, F, Frampton, R. (2007) An evaluation of the EuroNCAP crash safety rating in the real world. Annual Proceedings/Association for the Advancement of Automotive Medicine, 51:282-298.
Seto, JL, Brewster, CE. (1991) Musculoskeletal conditioning of the older athlete. Clin Sports Med. 10(2): 401-29.
Shaw, CG, Lessley, DJ, Crandall, JR, Kent, RW. (2005) Elimination of Thoracic Muscle Tensing Effects for Frontal Crash Dummies. SAE Transactions, Journal of Passenger Cars-Mechanical Systems, 114(6): 205-219.
Sheldon, W, Stevens, S, Tucker, W. (1940) The Varieties of Human Physique. New York, Harper and Brothers.
Shima, M. (2009) Government Status Report of Japan. International Technical Conference on the Enhanced Safety of Vehicles (ESV), 21.
Siegmund, GP, Sanderson, DJ, Inglis, JT. (2004) Gradation of Neck Muscle Responses and Head/Neck Kinematics to Acceleration and Speed Change in Rear-End Collisions. Stapp Car Crash J, 48: 419-430.
Snyder, SM, Schneider, E. (1991) Estimation of mechanical properties of cortical bone by computed tomography. J Orthop Res, 9(3): 422-31.
Soni, A, Mukherjee, S, Chawla, A. (2008) Effects of boundary conditions in dynamic tests for pedestrian safety, a finite element simulation study. Int J of Veh Safety, 3(3): 295-306.
Spence, DA. (1973) Nonlinear wave propagation in viscoelastic materials. In Nonlinear elasticity, ed Dickey RW. New York, Academic Press.
Squillante, RG, Williams, JL. (1993) Videodensitometry of osteons in females with femoral neck fractures. Calcif Tissue Int. 52(4): 273–277.
Takahashi, Y, Kikuchi, Y, Konosu, A, Ishikawa, H. (2000) Development and Validation of the Finite Element Model for the Human Lower Limb of Pedestrians. Stapp Car Crash J, 44: 335-355.
Takhounts, EG, Ridella, SA, Hasija, V, Tannous, RE, Campbell, JQ, Malone, D, Danelson, K, Stitzel, J, Rowson, S, Duma, S. (2008) Investigation of Traumatic Brain Injuries Using the Next Generation of Simulated Injury Monitor (Simon) Finite Element Head Model. Stapp Car Crash J, 52: 1-31.
Tawhai, M, Bischoff, J, Einstein, D, Erdemir, A, Guess, T, Reinbolt, J. (2009) Multiscale Modeling in Computational Biomechanics, Determining Computational Priorities and Addressing Current Challenges. IEEE Engineering in Medicine and Biology.
Thacker, BH, Wu, YT, Nicolella, DP, Anderson, RC. (1997) Probabilistic Injury Analysis of the Cervical Spine. Proceedings of the AIAA/ASME/ASCE/AHS/ASC 38th Structures, Structural Dynamics, and Materials (SDM) Conference, Kissimmee, Florida, April 7-10.
Thacker, BH, Wu, YT, Nicolella, DP. (1998) Probabilistic Model of Neck Injury. in Frontiers in Head and Neck Trauma, Clinical and Biomechanical, Probabilistic Model of Neck Injury. eds Yoganandan, N, Pintar, FA, Larson, SJ, Sances Jr, A. Amsterdam: IOS Press.
Thomopoulos, S, Marquez, JP, Weinberger, B, Birman, V, Genin, GM. (2006) Collagen Fiber Orientation at the Tendon to Bone Insertion and Its Influence on Stress Concentrations. J Biomech, 39(10): 1842-1851.
Tingvall, C, Haworth, N. (1999) Vision Zero - An ethical approach to safety and mobility. 6th ITE International Conference Road Safety & Traffic Enforcement, Beyond 2000, Melbourne, 6-7 September.
United Nations (UN) (2002) World Population Ageing 1950-2050, Population Division, DESA, United Nations.
Untaroiu, CD, Darvish, KK, Crandall, JR, Deng, B, Wang, JT. (2005) Identification of Viscoelastic Properties of Human Medial Collateral Ligament using Finite Element Optimization. 20th Congress of the International Society of Biomechanics.
Untaroiu, CD, Kerrigan, JR, Crandall, JR. (2006) Material Identification using Successive Response Surface Methodology, with Application to a Human Femur subjected to Three-Point Bending Loading. SAE Transactions, Journal of Passenger Cars-Mechanical System, 115(6): 26-35.
Untaroiu CD, Meissner MU, Crandall JR, Takahashi Y, Okamoto M, Ito O, (2008) Crash reconstruction of pedestrian accidents using optimization techniques, Intl. J. Impact Eng. 36(2): 210-219.

van Dommelen, J, Minary, M, Ivarsson, BJ, Millington, S, Raut, M, Kerrigan, JR, Crandall, JR, Diduch, D. (2006) Viscoelastic behavior of human knee ligaments, nonlinearity of the inelastic response. Ann of Biomed Eng, 34(6): 1008-1018.

van Rooij, L, Bhalla, KS, Meissner, MU, Ivarsson, BJ, Crandall, JR, Longhitano, D, Takahashi, Y, Kikuchi, Y. (2003) Pedestrian Crash Reconstruction using Multi-body Modeling with Geometrically Detailed Validated, Vehicle Models and Advanced Pedestrian Injury Criteria. Paper 468, International Technical Conference on the Enhanced Safety of Vehicles (ESV), 18.

Viano, D. (1995) Restraint Effectiveness, Availability and Use in Fatal Crashes, Implications to Injury Contol. J of Trauma, 38(4): 538-546.

Viceconti, M. (2008) Multiscale modelling and team science, the future of orthopaedic biomechanics. J Foot Ankle Res. 1(Suppl 1): 1-2.

Wang, SC, Bednarski, B, Patel, S, Yan, A, Kohoyda-Inglis, C, Kennedy, T, Link, E, Rowe, S, Sochor, M, Arbabi, S. (2003) Increased Depth of Subcutaneous Fat Is Protective Against Abdominal Injuries in Motor Vehicle Collisions. Annual Proceedings/Association for the Advancement of Automotive Medicine (AAAM), 47: 545-559.

Wang, Y, Beydoun, M, Liang, L, Caballero, B, Kumanyika, S. (2008) Will All Americans Become Overweight or Obese? Estimating the Progression and Cost of the US Obesity Epidemic. Obesity, 16 (10): 2323-2330.

Wayne, JS. (2008) Finite Element Analysis. in Encyclopedia of Biomaterials and Biomedical Engineering, Vol. 2, eds. Wenk, GE, Bolwin, GL. New York, InformaHealthcare USA.

World Health Organization (WHO) (2004) World Report on Road Traffic Injury Prevention. April 7.

World Health Organization (WHO) (2006) Fact sheet N°311, Obesity and Overweight, Sept. 2006.

Wutzler, S, Maegele, M, Marzi, I, Spanholtz, T, Wafaisade, A, Lefering, R. (2009) Association of Preexisting Medical Conditions with In-Hospital Mortality in Multiple-Trauma Patients. J Am Coll Surg, 209(1): 75-81.

Yamada, H, (1970) Strength of Biological Materials. Eds Evans, FG. Baltimore: Williams & Wilkins.

Zhang L, Chen L, Vertiz AM. (2004) Survey of Front Passenger Posture Usage in Passenger Vehicles. SAE World Congress & Exhibition. Paper 2004-01-0845, Society of Automotive Engineers.

Zhou, Q, Rouhana, SW, Melvin, JW. (1996) Age Effects On Thoracic Injury Tolerance. Stapp Car Crash Conference Proceedings, Paper 962421.

Zhu, S, Layde, PM, Guse, CE, Laud, PW, Pintar, F, Nirula, R, Hargarten, S. (2006) Obesity and Risk for Death Due to Motor Vehicle Crashes. Am J Public Health, 96(4): 734-739.