Modeling of Occupant Biomechanics with Emphasis on the Analysis of Lower Extremity Injuries

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Abstract -

This paper addresses the use of finite element method to the simulation of humans in an impact environment. An extensive literature survey reveals several rigid body dynamics models of the human and sub-components, however, relatively little work has been performed using finite elements. Difficulties remain in characterizing the material properties of the associated biological materials because of non-linear, inhomogeneous, anisotropic, and rate dependent behavior. Physical testing using cadavers remains the primary means for determining human impact response and for validating the models created. A method for creating human finite element models is presented which requires the model to be developed in a ground-up approach and in a componentwise manner. Each component has five levels of detail with level one being a collection of rigid segments connected by simple joints and level five consisting of a highly detailed model with proper material properties and injury mechanisms included.

This method has been employed to create a level two human lower extremity model. The model consists of accurate geometric segments of the individual bones in the leg. Each segment is connected using joint definitions in LS-DYNA3D that contain the non-linear stiffness characteristics of the hip, knee, and ankle. The model was used to simulate the loading conditions of a 50% overlap frontal collision of two mid-sized cars with a

closing speed of 112 km/hr (70 mph). A parametric study to determine the effects of muscle tensioning was performed for twenty-seven different joint loading cases. Results indicated that muscle tensioning greatly affected the kinematics of the leg during high speed impact events. Greater stiffness in the hip and knee directly resulted in a higher potential for injury in the ankle. In addition, higher levels of muscle activation in the ankle reduced injuries from the deceleration pulse of the impact, however, toepan intrusion still presented potential harm to the ankle.

Introduction

The field of biomechanics has been rapidly advancing for the past several years. With the large number of transportation devices in use worldwide, one area of biomechanics that receives considerable attention is the development of dynamic simulation models of vehicle occupants. Automobile accidents alone caused 40,676 fatalities and over 3.2 million injuries in the United States in 1994 [1]. Medical costs relating to the direct care and rehabilitation of these fatalities and injuries exceeds $120 billion dollars per year. With automobile accidents being the leading cause of death for those between the ages of 1 and 34, the costs associated with production losses due to mortality is estimated to exceed $50 billion per year [2, 3].

These statistics alone show the necessity to improve the safety aspects associated with automobiles. A first step, in making these improvements, is understanding the nature of the injuries sustained in collisions. This involves detailed knowledge of the composition, structural integrity, and impact response of the human body. Since determining these characteristics directly, through some form of impact testing, using living human subjects is neither reasonable nor ethical, it is necessary to develop some type of human surrogate. Because the human body is arguably one of the most complex structures in existence, it is in turn one of the most difficult to simulate.

With the increasingly rapid development of higher powered computing techniques, computer modeling is becoming a more realistic and credible form of crashworthiness evaluation. The Finite Element Method (FEM) is currently the most advanced tool for simulating an impact event. It's advantages over other modeling techniques include accurate geometric representation, advanced contact algorithms, and material models for representation of the large deformations experienced in high speed impacts. Additionally, FEM allows for the collection of more data than any other type of modeling. This includes quantities such as stress, strain, displacement, velocity, acceleration, energy, etc. at virtually any location in the model.

Literature Review

Computer modeling

Mathematical modeling of vehicle occupants has progressed rapidly since the first model was proposed by McHenry in 1963, at the Cornell Aeronautical Laboratory (currently named Calspan Corporation) [4]. Models are generally divided into two categories based on the analytical technique used to formulate the solution. These are either a rigid body dynamics formulation or a finite element approach. Rigid body dynamical models of human occupants are more commonly referred to as gross motion simulators. These range in complexity from simple one-dimensional mass-spring component models to three-dimensional whole body simulators. In general these models assume the occupant to be a set of rigid bodies linked by various types of joints in an open loop system. This formulation is referred to as a tree structure. Planes, ellipses, ellipsoids, and hyperellipsoids are used as visual representation of the various body segments, and provide surfaces for contact of the segments in the crash environment. The governing equations of motion are derived using either a Lagrangian technique (relating the system's potential, kinetic, and dissipated energies to the generalized forces and coordinates) or a Newtonian approach (conservation of linear and angular momentum). The equations are solved using various integration schemes depending on the program used. The two most advanced and commonly used are Articulated Total Body (ATB) developed at Armstrong Aerospace Medical Research Laboratory (AAMRL) [5] and MAthematical DYnamical MOdels (MADYMO) from the TNO Crash Safety Research Center in the Netherlands [6].

Because of the limitations previously mentioned in these gross motion simulators, FEM models are becoming more widely used. While most effective, it is also the most difficult and computer intensive. A much higher level of detail and knowledge is needed of the physical specimen being modeled. Because of this, relatively few whole body human FEM models exist. More commonly developed are models of the various dummies and individual components of the human body (i.e. thorax, head, joints, etc.). One FEM model that is close to a complete body is the Wayne State University human cadaver side impact model developed by Huang *et al*. in 1994 [7]. The intention of the model is to simulate the gross motion of a human cadaver in side impact. The notion of specifying a cadaver rather than a living human was based on the availability of validation data.

Head injury constitutes approximately 50 percent of all injuries sustained in transportation accidents, and is a common injury in sports and other human activities [8]. While head injury is one of the most common types of injury, the mechanisms of head and brain trauma still have many unanswered questions. Because of the necessity to eliminate head injury, the scientific studies focused on this body component have been the most extensive. Mathematical modeling of head response to impact can be a useful tool in establishing brain injury mechanisms, particularly when used in conjunction with appropriate experimental studies. King and Chou reviewed the many models that were developed between 1966-1975 [9]. Over 25 models were covered, most of which modeled the head as a fluid filled spherical or ellipsoid shell. The effects of translational acceleration, rotational acceleration, and flexion/extension of the upper cervical were studied to assess injury potential. One conclusion made by King and Chou was that because of the complex geometry and material properties involved, the use of the finite element approach was the best suited for modeling the head. This became apparent as the FEM models quickly outperformed other types of models when their development began in the mid 1970's. Currently, one of the most advanced FEM head model was developed by Ruan *et al.* at Wayne State University, now of the Ford Motor Company [10,11,12].

Along with head injury, thoracic trauma can be very serious and many times life threatening. The thorax, in simple terms, is a structural body consisting of the rib cage which encloses the heart and lung organs. Its primary function is to resist atmospheric pressure and develop a negative pressure differential thus permitting air inflow to the lungs. In addition, it protects the internal organs from injury. In the automobile crash environment, increased safety features have reduced chest injury; however, the nature of safety restraints, such as airbags and seat belts, still require the majority of the impact to be absorbed by the thorax. For this reason, detailed thorax models are beneficial for the assessment of the various restraint systems. An advanced finite element model of the human thorax was developed in 1991 by Plank and Eppinger of the U.S. Department of Transportation [13].

For a more comprehensive review of human modeling for both rigid body and FEM models the reader is referred to reference numbers [14, 15, 16, and 17]. In each of these sources several models in each class are described in detail.

Biological material characterization

Probably the most difficult and unascertained area of human modeling is the representation of material characteristics and mechanical behavior. Things that are taken for granted in conventional engineering do not apply to a biological system. The structural components of the body are living organisms and respond to stress and strain biologically not just mechanically. External forces acting on a body are coupled with a biological response that applies internal forces as well. In all areas of the body, especially soft tissues, the materials can behave in a non-linear, viscoelastic, inhomogeneous, and highly anisotropic manner. Adding to this already difficult problem is the great variability between individuals and the inability to collect the necessary information by standard methods. For ethical reasons the engineer can not perform accepted material characterization tests on a living subject. Nonetheless, methods are being developed that allow for a good understanding of the issues at hand.

Hard tissue

Hard tissue, primarily bone, is one of the easier biological tissues to study. In a mechanical sense, bone is a composite material with several distinct solid and fluid phases. There are two major forms of bone tissue: cortical or trabecular. Cortical bone is very compact and dense providing much of the structural support of bones throughout the body. Trabecular bone, often called spongy, comprises much of the inside area of long bones, short bones, and irregular bones such as the vertebrae. Cortical bone density is in the range of 1.85 to 2.00 g/cm3, whereas, trabecular bone can range from 0.15 to 1.0 g/cm3 [18].

The stress-strain relationship of cortical bone was found to be strain rate dependent [19]. Figure 1 displays the stress-strain relation at various strain rates. As the strain rate increases, the bone stiffens. This was found to be true throughout the bones in the body. For strain rates comparable to those seen in automobile impact situations, the elastic modulus is in the range of 5 to 20 GPa. In addition the plastic region is eliminated at these high strain rates and bone may be modeled as linearly elastic.

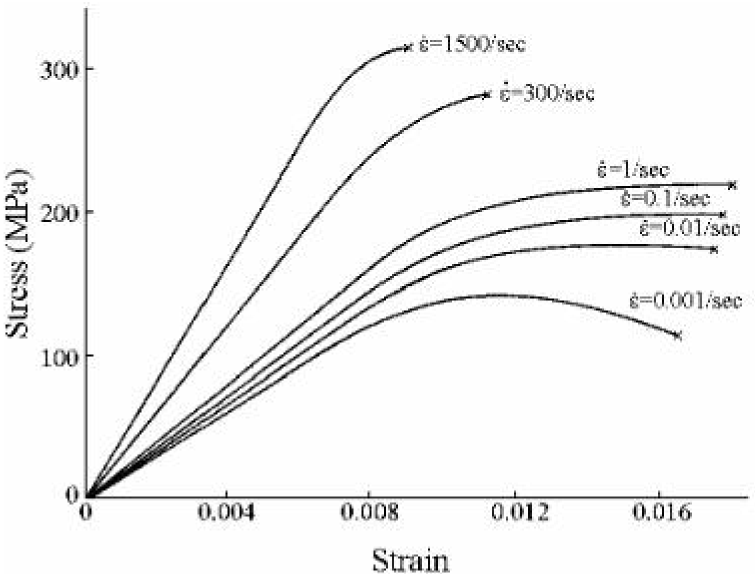

Figure 1: Strain rate dependence of the stress-strain relationship in cortical bone

Soft tissue

Soft tissue mechanics is a more difficult subject to address. There is a wide range of soft tissues in the body that includes skin, muscles, ligaments, tendons, blood vessels, all of the major organs, and more. Not only is there variability between various tissues but also within the same tissue type. For example, depending on the alignment in the lungs, similar blood vessels can be either rigid or compliant. The mechanical properties of soft tissues are related directly to its structure, geometry, function, and relationship to its neighboring organs. Soft tissues are pseudoelastic; meaning they are not elastic, but under periodic loading each tissue will have a steady state stress-strain relationship which is not strain rate dependent. The loading and unloading phases of the tissue are not coincident, thereby exhibiting certain levels of hysterisis. The existence of this loop formed in the loading and unloading phases indicates that the tissues are also viscoelastic. Tissues, when stretched suddenly to a new length then held constant, will exhibit stress relaxation. As time increases the amount of stress in the tissue will decrease. The degree of stress reduction depends upon the tissue [18].

Muscle mechanics is an important aspect of modeling the human. Muscle activation is the primary means for articulating body segments and plays an important role in defining the kinematics of the human in any activity or event. There are two components of a muscle that contribute to its influence on the system: the passive component and the active component. The passive component acts as a tension element and has a direct non-linear relationship between force and muscle length. Because a muscle cannot passively shorten, a passive tensile force only exists when the muscle is at its resting length or being stretched. The active component is slightly different. As a muscle contracts, it can shorten its length and, at the same time, produce a tension force. Actively, the greatest force can be produced when the muscle is at a length equivalent to its resting length. In addition little to no force can be produced when a muscle shortens to one-half its original length, or when it is stretched to 1.5 times its original length [20]. The active state is also controlled by the level of activation, often described by the function, $a(t)$. This function varies between 0 and 1 depending on the level of muscle fiber recruitment for contraction. A fully activated or "flexed" muscle has an activation level of 1. All of these characteristics are displayed in Figures 2 and 3. To mathematically describe these characteristics, several models have been presented. They are all models of viscoelasticity consisting of combinations of springs and dampers. These models include the Maxwell model (spring in series with a damper), the Voight model (spring in parallel with a damper), and the Kelvin model (spring in parallel with a Maxwell model) [21].

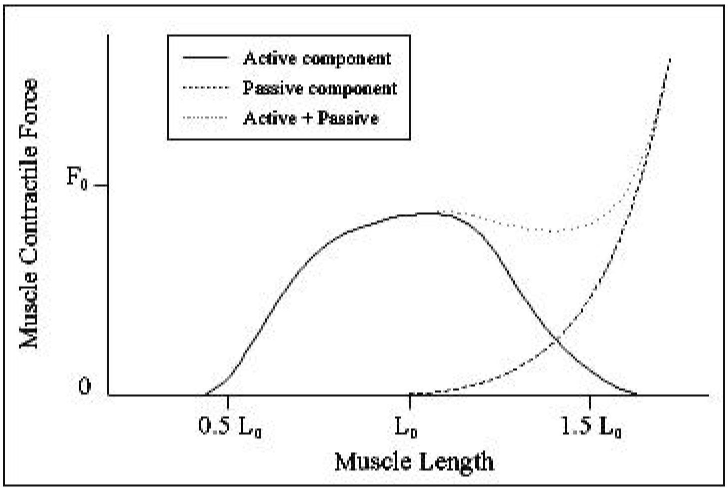

Figure 2: Force-length characteristics of skeletal muscle

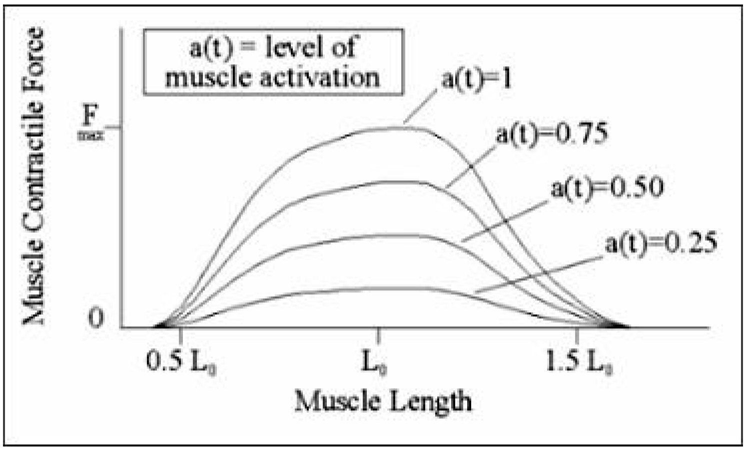

Figure 3: Muscle activation level effects on force-length properties

Joint mechanics

In all instances, there are several different soft tissues that span each joint. These soft tissues are the primary means for articulating the joint as well as providing its stability. Moments produced by the soft tissues result in body segment motion. Like muscles, there are two components to these moments: the passive and the active. The passive moment is due to the influence of the viscoelastic structures such as passive muscle components, ligaments, cartilage, etc. situated across the joint. Audu and Davy characterized the passive moment with the following equation [22]:

$$M_{pass}(\theta,\dot{\theta}) = k_1 e^{-k_2(\theta-\theta_2)} - k_3 e^{-k_4(\theta_1-\theta)} - c\dot{\theta} \quad [1]$$

The *k* values are stiffness constants and depend upon the joint being modeled, *c* is the passive damping constant, and the values q1 and q2 are angles approximating the joint's limit in each direction. The composition of this equation is such that, as q reaches the joint's limit, the joint becomes extremely stiff. For joints with more than one degree-of-freedom, constants for the equation in each direction are needed. Therefore, joints such as the hip, have there passive moment described by three equations (flexion/extension, abduction/adduction, and internal/external rotation).

The active moment is more difficult to characterize because it is caused by the active muscles that span the joint. Because several muscles often span a joint, there is dynamic load sharing between the muscles and it is difficult to determine the contribution of each. When muscles are activated to tighten a joint, there is a positive and negative moment applied simultaneously to the joint. These will act to stiffen the joint over a certain range-of-motion; then, as the joint is forced in a certain direction toward its limit, the passive component will control the joint. Limited data has been published which characterizes the specific torques that are developed at each joint. Figure 4 displays a) the general characteristics of the passive behavior and b) that of an active locked joint.

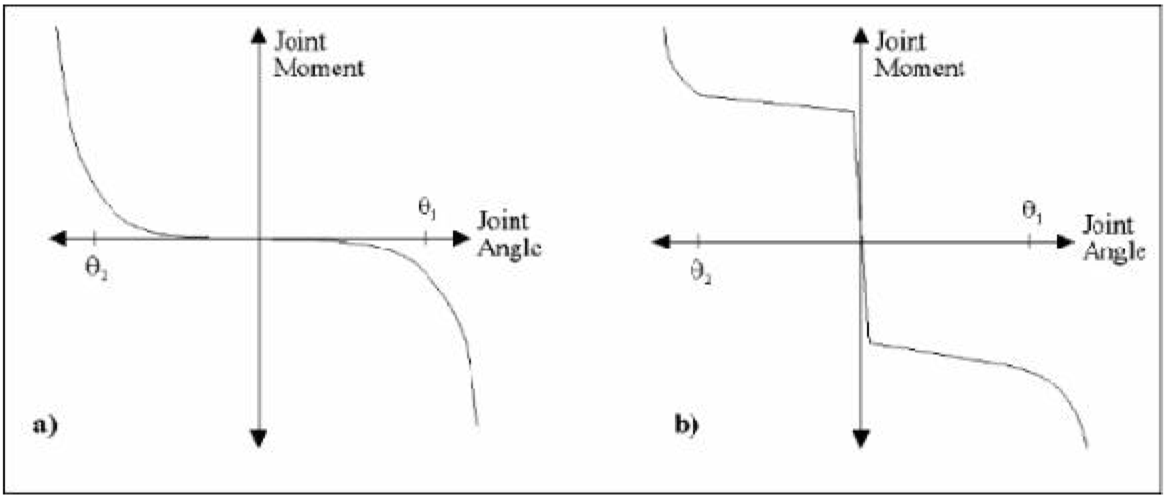

Figure 4: Joint moment vs. joint angle characteristics. a) passive joint b) fully activated joint

Physical testing

Physical testing is an important component of biomechanics studies, especially in automotive safety applications. In the automotive field, physical testing (for biomechanics purposes) is primarily done with the use of cadavers, however, some animal and living human subject testing is also performed. These tests provide the researcher with key information to be used in the following:

1. biological material characterization
2. development of injury assessment criteria

3. development of anthropomorphic test devices
4. assessment of crashworthiness based on injuries sustained
5. development of industry standards and regulations for crashworthiness
6. development of loading corridors for computer model validation purposes

Physical testing can range in scale from simple impact tests of a body component to full scale car to car collisions with several occupants in each.

Extensive work in this area began in the 1960's for the purpose of developing a crash test dummy. At that time, data was used to develop certain loading corridors and impact response curves that related such quantities as joint moment to joint angle, force to deflection, acceleration to impact velocity, etc. Since then, work has been done to modify these initial curves and fine tune them for a more accurate representation. In addition several quantities or formulas were developed that related the test data to injury potential. Injury parameters for the head, chest, and femur are a few that have been developed.

Table 1: Summary of major works in human impact response and injury tolerance.

Body Region	Author(s)	Year
Head	Lissner, Lebow, and Evans [23]	1960
Head	Ommaya and Hirsch [24]	1971
Head	Tzeng, Tseng, and Lee [25]	1993
Face	Nyquist, Cavanaugh, Goldberg, and King [26]	1986
Neck	Mertz and Patrick [27]	1971
Neck	Patrick and Chou [28]	1976
Thorax	Kroell, Schneider, and Nahum [29,30]	1971, 1974
Thorax	Neathery and Lobdell [31]	1973
Thorax	Viano, Lau, Ashbury, King, and Begeman [32]	1989
Abdomen	Nusholtz, Kaiker, Huelke, and Suggitt [33]	1985
Abdomen	Cavanaugh, Nyquist, Goldberg, and King [34]	1986
Lower	Torso Nyquist and Murton [35]	1975
Spine	Begeman, King, and Prasad [36]	1973
Pelvis	Nusholtz, Alem, and Melvin [37]	1982
Pelvis	Viano, Lau, Ashbury, King, and Begeman [32]	1989

Femur	Horsch and Patrick [38]	1976
Knee	Viano, Culver, Haut, Melvin, Bender, Culver, and Levine [39]	1978
Tibia	Nyquist, Cheng, El-Bohy, and King [40]	1985
Ankle	Begeman and Prasad [41]	1990

For the general assessment of injury to any region of the body emergency physicians use an Abbreviated Injury Scale (AIS), developed by the Association for the Advancement of Automotive Medicine [42]. It ranks the level of injury on a scale from AIS 0 to AIS 6, with the following correlations:

AIS Code	Injury Level	Fatality Range
0	no injury	0.0 %
1	minor	0.0 %
2	moderate	0.1 - 0.4 %
3	serious	0.8 - 2.1 %
4	severe	7.9 - 10.6 %
5	critical	53.1 - 58.4 %
6	maximum	virtually unsurvivable

Injuries around AIS 3 are considered tolerable, and it is injuries of AIS 4 and higher that safety regulations try to eliminate under reasonable impact scenarios. Although there is some controversies over the level of severity of certain injuries, AIS coding has been developed for all regions of the body. For a more complete survey of human impact response and tolerance the reader is referred to the sources listed in Table 1.

METHODOLOGY FOR HUMAN FEM MODEL DEVELOPMENT

From the previous sections it can be concluded that the finite element method is the best suited tool to mathematically model the human for reasons of geometrical representation, material characterization, and environment interaction. The modeler is now faced with an extremely difficult task. Like with the development of any new concept, it must start from the simple and gradually build to the complex in a systematic ground up approach. There are two goals that provide a basis for constructing this model:

7. At each stage of development, no matter how simple or complex, the model must be able to be integrated into a full scale crash simulation. If the model can not do this then it loses value as an engineering tool.
8. The model must be constructed in a way such that more detailed component models can be easily transitioned into a base model.

With the goals in mind, the development of the occupant model and its components can be divided into five stages, level 1 to 5, based on the level of detail in the model. Issues addressed at each level of the model concern geometric and physiological representation. Geometric representation includes the shape of the mesh, the mesh density, and to what level of detail will objects be included in the model. The physiological representation includes material properties, joint modeling, failure and injury modes, and environment interaction.

It is not necessary for the whole body model to be constructed of components from the same level of development. A model may contain a level 1 head and neck, with a level 4 thorax, and level 3 legs. This discretion will be left to the modeler and is dependent upon certain issues. These include, but are not limited to:

9. Certain aspects or safety devices being studied will not require a detailed complete body and may only need a detailed component. For example, the effects of a new airbag and seat belt system will need a more detailed upper body model rather than the lower portion. It is in the interest of computational time and cost to limit a model to the necessities for the particular study.
10. It may not be possible to have models developed for all components that are of the same level. Limitations on the scientific data available will preclude the modeler from developing higher level models of certain segments. As an example, it is much more difficult to model major organs than it is the knee joint; therefore, a level 5 knee is likely to be placed in the model before a level 5 heart and lungs is available.

In all cases, however, geometric compatibility must be preserved to ensure a good interchangeability between various segments.

Level 1

Following the previously stated outline for levels of progression, the entry level model, although simple in definition, is a very important part of the overall problem. It will form the foundation for higher level models, therefore, it is important to construct it with care and always plan past the present model to prevent any downfalls in the future. The first step in constructing the model is creating the geometry. To address the issue of great variability among humans, some assumptions must be made. The first of which will be the selection of a single size and mass for the model. (Once completed, the model can be scaled to represent a wide range of individuals.) The most logical choice is that of a 50th percentile male for two reasons: 1) this will represent the average human, and 2) it is nearly a scientific convention in biomechanics to refer to this size person, with a majority of the published data reflecting results scaled for a 50th percentile male.

Accurate representation of the body's anthropometry is important. To maintain the proper kinematics, each segment must be the correct size and have the correct mass distribution. Centers of mass, moments of inertia, and the locations of joint's axis of rotation are included in the anthropometry. An excellent report for obtaining the anthropomorphic data needed for developing the geometry was released in 1983 from the Transportation Research Institute by Robbins [43]. This report provides segment dimensions and masses, as well as providing joint locations and joint angles for a 50th percentile male seated in the average automobile driving position.

With geometric data now available, construction of the model can begin. The level 1 occupant will serve as a tool similar to the gross motion simulators of MADYMO and ATB. For simplicity, the body segments can be constructed of basic geometric shapes such as ellipsoids and spheres. These shapes, of course, should approximate the size of the component they represent. Each segment will be connected by a simple joint definition in LS-DYNA3D. A level 1 model can be seen in Figure 5. The mesh for the model pictured was developed at Livermore Software Technology Corporation (LSTC), however joint properties were modified by this author at the National Crash Analysis Center (NCAC). Joint types are either spherical or hinge and are given constant stiffness values which approximate a human. These values were taken from literature related to the development of the Hybrid III dummy [44]. Each segment on this model is constructed of rigid shell elements of uniform thickness. Densities are adjusted so that segments will have the proper segment masses.

As a gross motion simulator, segment accelerations will closely resemble those seen by more detailed models. Assessment of an automobile's crashworthiness is possible because quantities such as the Head Injury Criteria, Thoracic Trauma Index, and 3ms Chest Acceleration Criteria can still be calculated. However, a good injury assessment is not plausible because loads developed at the joints will be higher than those expected from an actual human. This is because the model lacks compliance in the segments and all forces are transmitted directly to the joints. The model has also satisfied the prescribed goals of this modeling approach. It is capable of being integrated into a full scale crash simulation and it is constructed of well defined segments that can easily be interchanged with more advanced ones.

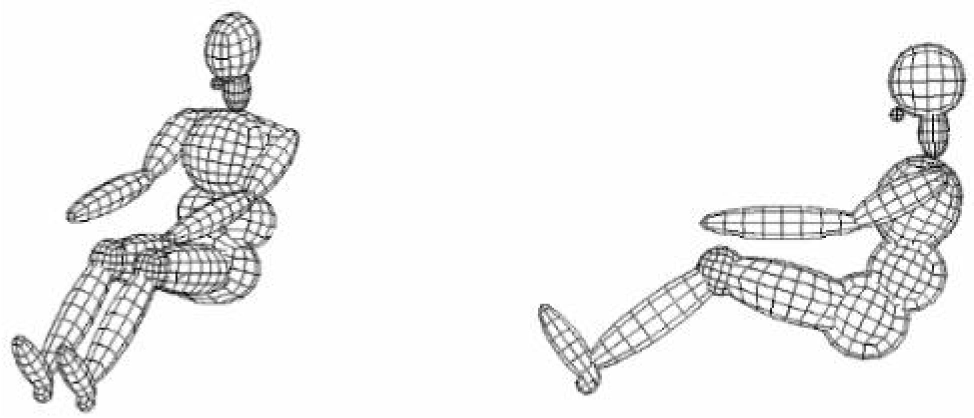

Figure 5: Level 1 occupant model (developed by LSTC and modified by NCAC)

Level 2

The next level of development is similar to level 1 with a few exceptions. This level takes advantage of the use of finite elements for geometric representation and the nonlinear features of LS-DYNA3D for human-like joint modeling. While maintaining rigid segments, geometry is greatly improved upon. Since today's computers can handle large models with ease, the level of detail in each segment should not be sacrificed. The model will take on the true form of a human. Because the geometry of the human is very complex and difficult to recreate in a computer environment, certain methods have been developed to enter geometry to the finite element pre-processor. Digitization of a human, using a robot arm to enter data from three-dimensional space, or the use of MRI and CAT scan images are two methods that are commonly used. The former method is used at the NCAC [45]. Complex geometric models can also be purchased from companies who perform the digitization. One such company, with a large library of data-sets, is Viewpoint Datalabs located in Orem, Utah [46].

There are three major advantages to using accurate geometry. First, each segment will now have accurate moments of inertia and centers of gravity. This will improve the kinematics of the model and thus the gross motion behavior. Second, contact interaction with the vehicles environment is greatly improved. This will allow for a better evaluation of the interior components of the vehicle for improved occupant interaction. Thirdly, parts will fit together more accurately especially when higher level components are interchanged.

The other major difference in this level of model is the use of more accurate joint characteristics. Joints are still modeled using the pre-defined LS-DYNA3D joint models, however, the stiffness characteristics are modeled differently. Non-linear torsional springs and dampers located at the joint centers of rotation are used to define the behavior of the joint. This level of model is also the first that allows different level components to be integrated together. If lower limb reactions are desired, it is not necessary to incorporate the detailed joints in the upper body. Figure 6 displays a level 2 occupant model that was developed at the NCAC. As with level 1, this model can be integrated into the crash environment and components are easily interchangeable.

Figure 6: Level 2 occupant model developed by the NCAC

Level 3

This stage is where FEM models begin to separate themselves from other modeling techniques. This most significant change is the introduction of the deformable material models available in LS-DYNA3D. At his point each component will begin to take on its unique form in the body, as opposed to previous levels where all components were rigid shells. The major issues that will arise from the introduction of deformable materials are:

11. New levels of geometry will be necessary in each component to represent the external and internal structures.
12. The connectivity and sliding interface definitions between these new levels of geometry will be important.
13. The introduction of material properties and choice of material models for each component.

A majority of the advanced finite element models of the human components that have been developed today can be classified as a level 3 model.

At this level most materials can be modeled as linear elastic or viscoelastic. The internal and external components should be introduced as separate materials. For structural integrity and kinematic response, the skeletal bones should be accurately represented. For occupant compartment interaction and some injury evaluation, the outer surface of the body must be represented. At this level the response of specific internal organs is not yet an issue, therefore, they can be modeled as a continuum of solid elements. Joints throughout the body can still be defined using LS-DYNA3D joints; however, active joint properties should now be represented by translational Kelvin elements for bending modes and torsional Kelvin elements for the axial rotation models.

Some concerns with the level 3 models are to maintain the proper segment masses, centers of gravity, and other anthropomorphic data. Assigning material densities must be done with care so that they are accurate and preserve the proper performance of the model. Overall, the level 3 component models have made many improvements over previous levels. The skeletal structure of the human is complete, estimated material properties have been applied to all regions of the body, and joints are more accurately defined. These models are now useful for assessing injury potential in each segment of the body. Although failure mechanisms have not been introduced, quantities measured such as stresses, pressures, joint forces, etc. can be compared to experimental values at harmful levels.

Level 4

This level of model will improve upon level 3 in a few distinct areas. Material properties can be improved with each component represented with a material model that more accurately depicts it's behavior. In addition, some failure mechanisms such as bone fracture can be incorporated at this level. Second, any structure being modeled will be represented by finite elements rather than an LS-DYNA3D feature. This means that the LS-DYNA3D joints will be replaced with advanced joint models comprised of finite elements representing the necessary structures comprising a human joint. Additionally, muscle representation would be done with continuum models of the individual muscles rather than by Kelvin spring-damper elements. Finally, the major organs will be defined and individually represented. In general, the level 4 model will be representative of the most advanced model possible with current knowledge. Because the level 5 model is defined as the highest possible level, more knowledge of the human body will be needed from a scientific standpoint before FEM models can be taken to that level.

With respect to current capabilities, the complete level 4 human would be the most advanced tool for simulating vehicle occupants in an impact event. Each component is not beyond the realm of today's scientific capabilities; however, the major issue that currently limits its development would be the material representations and subsequently its validation. Although its complete validation would be difficult, it would still be very useful in crashworthiness evaluations. It is important to note, that at many levels, it will be more accurate at representing a human than any of the advanced "crash dummies" in use today.

Level 5

The level 5 model is the complete human model that can accurately predict the injuries of an occupant in any impact scenario. The feasibility of this model is currently low, especially for components of the body that are difficult to define in an absolute scientific manner. Certain characteristics of a level 5 model such as advanced material representation of all biological tissues will greatly depend upon the continued development of the FEM codes and the scientific characterization of the biological system.

For continued advancement in vehicle occupant models this staged approach must be accompanied by increased efforts to characterize the human. Validation of each component model at each level will be an iterative process with comparisons being made to experimental results, published data on human and cadaveric response, and in depth real world case study investigations. Because these are human models, the traditional means of validating against sound experimental data is not possible. It is therefore the modelers responsibility to use good modeling techniques made possible through a firm understanding of the analysis tools and the subject at hand.

APPLICATION TOWARD LOWER EXTREMITY MODELING

Problem definition and lower extremity characterization

For years, the largest safety concern for vehicle manufacturers has focused on designing cars that protect the occupant from life threatening injuries, particularly, those to the chest, head, and neck. With the use of airbags and improved seat belt protection, occupants are surviving what were once unsurvivable accidents. While the more serious issues have been addressed, what has received far less focus is the protection of the lower extremities. The lower extremities provide the primary means of locomotion for the human body. Injuries to this region are rarely fatal; however, they often require significant periods of hospitalization and subsequent rehabilitation. In addition, severe injuries to this region can result in degenerative arthritis, particularly if a joint is involved [47]. For these reasons, it is necessary to properly address the issues of the lower extremities in crash testing and simulation.

Preliminary assessment of the issues surrounding lower extremity injury can be done through low level finite element modeling. A level 2 lower limb model can be very useful for addressing joint behavior, muscle tensioning, and injury potential in a high speed impact event. Additionally, it can provide a foundation for more complex models and dictate the direction of their development. This study focused on three topics:

14. The development of a level 2, lower extremity, finite element model with accurate joint characterization and muscle tensioning behavior
15. Performing a parametric study to determine the influence of muscle tensioning in the ankle, knee, and hip on the injury potential in the ankle.
16. Provide a basis for the development of a more detailed level 3 model.

Table 2: Passive joint moment coefficients for flexion and extension of the lower extremity.

Coefficients for Passive Joint Moment		
Ankle	K1 = 2.0 K2 = 5.0 K3 = 9.0 K4 = 5.0	C1 = .943 theta1 = 0.349 theta2 = -0.524
Knee	K1 = 3.10 K2 = 5.90 K3 = 10.5 K4 = 11.8	C1 = 3.17 theta1 = 0.00 theta2 = -1.92
Hip	K1 = 2.6 K2 = 5.8 K3 = 8.7 K4 = 1.3	C1 = 1.09 theta1 = 1.92 theta2 = -0.1744

Table 3: Active joint properties for the lower extremity in the sagittal plane.

Joint	Maximum Active Torque [N-m]	
	Extension	Flexion

Hip	-215	200
Knee	-160	80
Ankle	-125	250

The lower extremity consists of three segments: the thigh, leg (often referred to as the shank or lower leg), and foot. Though the leg is anatomically the portion from the knee to the ankle, the entire lower extremity is often referred to as the leg. For clarification and ease of use, the entire lower extremity will be termed the leg unless otherwise noted. The segments are spatially linked via several soft tissue components, most noticeably approximately 29 muscles. Most of the muscles in the leg span only one joint however several span two joints. Each of the three major joints (hip, knee, and ankle) is unique; however, all are similar in that they are each weight bearing, synovial joints surrounded by a ligamentous joint capsule [47]. Yamaguchi determined the constants for flexion and extension in the sagittal plane for each of the joints in the leg to satisfy Equation 1 [20]. These constants can be found in Table 2. Instead of modeling each active muscle individually, their overall contribution on a joint can be lumped into an active torque applied at the joint's center of rotation. Gordon *et al*. characterized this active torque in the flexion-extension modes of the leg joints [48]. The maximum active torques at each joint in the leg are found in Table 3.

Various studies have been performed to examine the impact response of the ankle and other leg joints. Quantities such as joint moment and joint angle can be measured and compared with injuries observed. Comparisons to the experimental data can be done with a low level FEM model, and estimates of joint failure and injury potential can be made in the post-processing analysis. For ankle injury, Begeman and Prasad determined the impact response in dorsiflexion through pendulum tests of nine sets of ankles [41]. The study indicated that injury to the ankle joint could only be directly correlated with the maximum angle of dorsiflexion. Injury could not be correlated with joint moment, pre-impact range-of-motion, or peak force values. It was determined that if the ankle exceeded 45° in dorsiflexion, injury was very likely.

Model description and simulations performed

Table 4: Mass properties of leg segments.

Segment	Surface Area [mm2]	Density [Mg/mm2]	Mass [Mg]

Pelvis	6.475E+04	8.814E-08	5.707E-03
Upper Leg	5.965E+04	1.444E-07	8.614E-03
Lower Leg	6.569E+04	5.460E-08	3.587E-03
Foot	6.923E+04	1.417E-08	0.981E-03

The level 2 leg model created has accurate three dimensional geometry; however, the joints are restricted to motion in the sagittal plane. The focus of the model being developed is on joint characteristics. The geometry for the finite element mesh consists of the pelvis, femur, patella, tibia, fibula, and the 26 bones of the foot. The data for the geometry is of a 50th percentile male, digitized by Viewpoint Datalabs. A finite element mesh was applied to the bones and consisted of a total of 5477 thin shell elements. The mesh was then divided into four groups: the pelvis (pelvic bone), upper leg (femur), lower leg (patella, tibia, and fibula), and the foot (calcaneus, cuboid, 3 cuneiforms, 5 metatarsals, navicular, talus, and 14 phalanges). Because this is still a level 2 model, each of the groups was applied a rigid material type. The thin shell elements were given a uniform unity thickness and the density of the elements were adjusted so that the mass of a segment would match that of a 50th percentile. The segment masses were taken from the anthropomorphic study by Robbins [43] and can be found in Table 4. Additionally a 40 kg point pass was positioned at the top of the pelvis to simulate the upper portion of the body in hopes of maintaining some of the inertial influences of the body. With the material properties and mesh applied, the leg was positioned into an average vehicle occupant's sitting position. The specifications for this average seated position were also given by Robbins [43]. Figures 7 displays the finite element mesh of the leg in the seated position and Figure 8 displays a rendered isometric view of the leg with the joints' axis of rotation shown.

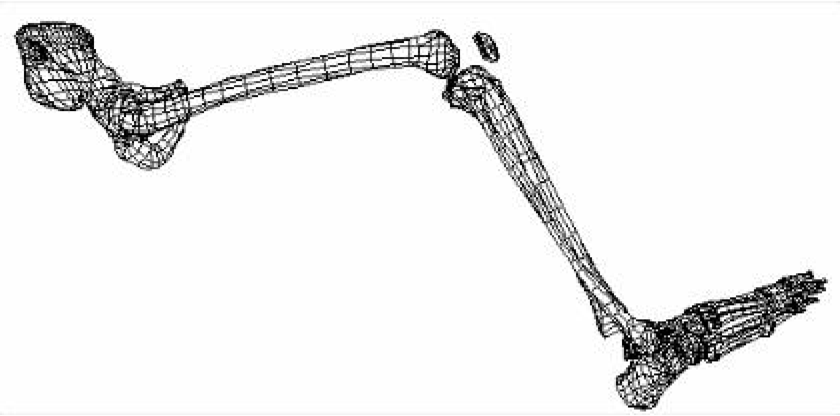

Figure 7: Finite element model of level 2 leg model in vehicle occupant seated position

The model was translated into an input deck for LS-DYNA3D version 930 [49]. The joints were defined with a non-linear load curve for the passive stiffness and damping being applied to the three respective joints. The load curve is in the form of joint angle vs. stiffness moment and follows the equation described by Audu and Davy [22] for passive stiffness with the coefficients defined by Yamaguchi [20] applied.

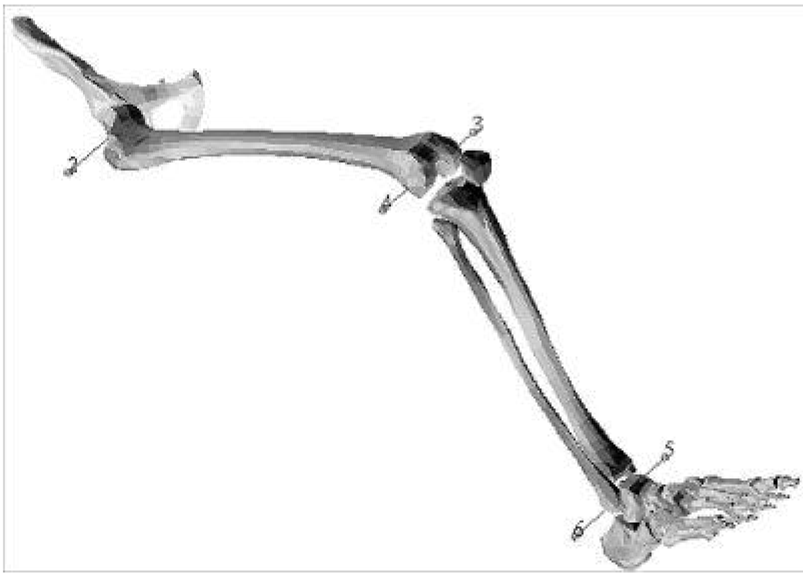

Figure 8: Isometric view of the seated leg with axis of rotation for each joint displayed

This study was to determine the effects of muscle tensioning on the joints in the leg during a high deceleration event and investigate the injury potential to the ankle in such a case. With this in mind it was necessary to apply the external loading conditions in order to simulate a high speed impact event. A seat belt was added in the form of a parallel spring and damper connecting the pelvis and a newly defined belt anchor. The element constants for the seat belt were K=6.0104 N/m for the spring and C=1000 N·s/m for the damper. The simulated crash pulse was of the NCAC's simulation of NHTSA Test #1400, a Taurus to Taurus head on collision, at 35 mph each, with a 50% overlap [50]. The seat belt, belt anchor, and the leg were given an initial velocity of 56 km/hr (35 mph). For the impact event, at time t=0, the belt anchor was applied the velocity profile of the car's center of gravity while the velocity profile of the toepan was input to the foot. The total simulation time is 150 ms. For test run #1400 there is a maximum toepan intrusion of 40 mm horizontally. All data taken from the Taurus-Taurus run was filtered using a standard SAE 60 Hz filter.

To study muscle tensioning effects, a joint could be in one of three states: 100%

passive, 100% passive and 50% active, or 100% passive and 100% active. Three joints, with three possible joint conditions each, results in a possibility of 27 combinations in which the joints could be loaded from internal torques. The values for 100% activation were taken from Gordon *et al.* [48] and the values for 50% activation were simply scaled from the 100% values. The investigation was to compare results such as joint position, joint moments, and other reactive forces to determine which combinations of muscle activity were potentially harmful.

Simulation results

For this study, comparisons between joint angle, joint moment, and muscle activation level were made. It was previously stated that joint angle is the best indicator of injury for loading the ankle in dorsiflexion, therefore, injury predictions could be made by comparing joint angles to the failure levels set forth by Begeman and Prasad [41]. For each test run, the maximum angle of the ankle and the corresponding moment at this angle are found in Table 5. The test number contains three numbers. The first represents the hip, the second the knee, and the third the ankle. Since a joint could have one of three levels of activation a number 1 corresponds to passive only, a 2 corresponds to passive and 50% active muscles, and a 3 corresponds to passive and 100% active muscles. For example, test run number 1-2-3 indicates that the hip is passive only, the knee is passive and 50% active, and the ankle is passive and 100% active.

Table 5: Maximum dorsiflexion angle and corresponding ankle joint moment for each test case.

Test #	Max. Angle [degrees]	Moment [N-m]
111	55.2	40.6
112	12.1	27.1
113	18.3	81.6
121	59.7	60.4
122	12.2	27.4
123	16.6	73.9
131	61.7	77.0
132	13.3	30.0
133	15.1	60.8
211	61.8	77.6
212	25.2	58.0
213	18.7	83.3
221	63.4	102.0
222	26.7	61.5
223	16.9	75.2

231	62.8	92.4
232	28.6	66.4
233	15.2	67.3
311	62.1	81.9
312	44.7	113.0
313	17.7	78.6
321	63.5	106.0
322	45.0	115.0
323	15.3	68.1
331	64.2	120.0
332	45.1	115.0
333	13.0	57.5

There was a large spread on the results and some noticeable trends formed. The angle in dorsiflexion ranged from 12.1° to 64.2°. Of the 27 test runs, 11 of them exceeded the 45° injury threshold set forth by Begeman and Prasad. The highest values for joint angle came when the hip was 100% activated and there was some form of tension in the knee. A completely tense leg had a dorsiflexion angle of 13.0° while a completely relaxed leg had a maximum angle of 55.2°. Some trends that were noticed involving joint angle include:

17. Increasing the activation level in the ankle decreased the maximum angle of dorsiflexion in the ankle.
18. Increasing the activation level in the hip increased the maximum angle of dorsiflexion in the ankle.
19. The effect of tensing the knee, while the ankle and hip remained constant, had no distinct pattern of results in ankle joint angle.
20. When the ankle was only passively activated, the peak dorsiflexion angle occurred early in the impact event, between 45 and 70 ms.
21. When the ankle was only passively activated, a second sharp increase in dorsiflexion was noted around 125 ms.
22. When the ankle was 50% or 100% activated, the peak dorsiflexion angle occurred later in the impact event, between 110 and 120 ms.
23. The point of maximum dorsiflexion was successively followed by the peak value in plantarflexion as the ankle swiftly rotated back

Ankle joint moment ranged from 27.1 to 120.0 N·m. Ankle joint moment did not have a direct correlation with joint angle but it did correlate to the tensing patterns of the joints. As the level of activation remained constant in the ankle and the other joints were gradually stiffened, the moment in the ankle joint rose sharply. A six image sequence of the 1-1-2 (lowest peak dorsiflexion) and the 3-3-1 (highest peak dorsiflexion) simulations can be found in the Appendix accompanied with the respective data plots. It should be noted that all of the pre- and post-processing was performed on a Silicon Graphics Indigo2 Extreme Workstation and all of the simulations were performed on a single processor of a Silicon Graphics Power Challenge XL.

Discussion and future development

This study intended to focus on the developmental process of a finite element model of the lower extremity. Although the model presented here is rather elementary, it did demonstrate the feasibility of analyzing muscle tensioning effects and injury potential in low level models. It is probably best to first discuss the limitations of this model and how these limitations will effect the results. These limitations include:

- In simulating the external loading from the crash pulse on the leg, there was no interaction will the vehicles interior. Loading on the ankle will be different if the knee impacts and crushes into a knee bolster. This effect will be to essentially lock the knee and hip in place, subsequently loading the ankle more aggressively. Other concerns are secondary impacts with the foot space and interactions with the seat cushions.
- There is no failure associated with this level of model; therefore, when the joint reaches injury levels it will always "spring" back. When the stiffness in the joint causes this spring back, the foot is likely to move too extremely in the opposite direction. This response is likely the cause of the large values of plantarflexion following each peak in dorsiflexion.
- Joints were only permitted to move in one plane, therefore, all forces and loading were not distributed through a joints entire range-of-motion. Motions such as inversion-eversion in the ankle may account for some energy redistribution thus lowering injury potential.

Although these downfalls are inherent in this model, they do not detract from the purpose of its creation and its use as a preliminary analysis tool. The model demonstrated, that even at low levels of detail, muscle tensioning greatly effects the kinematics of the lower extremities under impact loading. By producing such a wide range of angles and moments, it is most definitely an issue that must be addressed when discussing safety and design. The model also demonstrated the ease to which muscle tensioning can be incorporated into finite element models. Other human surrogates have a much more difficult time to do this. Dummies need adjustable joints and cadavers require some form of joint locking device.

Concerning the results, some conclusions can be made from this parametric study. These include:

- The ankle showed the highest level of injury potential when it was only passively tensed. Additionally, the peak values occurred early in the crash event (~ 50 ms). This indicates that the ankle joint is susceptible to injuries caused by rapid deceleration when the occupant is not aware of an oncoming collision. These peak levels can not be associated with toepan intrusion because that occurs after 100 ms. It must be noted that it is highly unlikely for the ankle to be only passively activated. The simple nature of operating an automobile requires some form of muscle tension through pedal or toepan interactions.
- Higher levels of ankle loading were seen when the other joints were tensed. The leg can be viewed as a jointed beam. It makes sense that the weakest joint will have to absorb most of the loading applied, because, this load will be transmitted across stiff joints. Stiffening of the hip joint had the greatest effects on the ankle. For cases when an oncoming collision is apparent to the driver, braking is applied most readily through a rapid extension of the leg at the hip and knee. This action leaves the ankle more susceptible to increased loading.
- For cases in which the ankle was tensed, the peak loading occurred around 110 ms. This is the same time at which toepan intrusion occurs in this simulation. In these runs (50% or 100% ankle tension), the intrusion showed to be the greatest contributor to ankle loading. Additionally for cases with no ankle tensing, a second peak loading of lower magnitude occurred at about 110 ms indicating intrusion influences.

The model has demonstrated it's potential as a safety research tool; therefore, it is necessary to develop this model to the next level. The most immediate changes will involve the incorporation of three dimensional joints for complete joint articulation. By incorporating these joints, additional data is needed for passive and active stiffness characteristics for motion out of the sagittal plane. Also, active damping properties are needed for all ranges of motion. The model would then be a complete level 2 model. As discussed in Chapter 4, development to a level 3 will involve the incorporation of bone properties and the viscoelastic properties of the outer surface of the leg for contact and vehicle compartment interaction.

CONCLUSIONS

This research accomplished three major tasks: 1) an extensive literature review of computer modeling of humans, 2) the development of a methodology for creating human finite element models, and 3) the application of this method toward the creation of a finite element model of the human lower extremities. This research has shown that finite element modeling is a viable tool for the future of automobile safety and crashworthiness studies. The complexity and accuracy of these models is no longer limited by the FEM codes or computational power. It is crucial to the development of human models that more efforts are put forth to characterize the biological system. Advanced techniques that allow for non-invasive and safe measuring of the living system are the ideal goal. These efforts must include a high level of communication between researchers. Additionally, the finite element modeler must be incorporated as much as possible to express the true needs and desired outcome from the physical experimentation. From the medical and trauma teams through to the engineers and model developers, in a collective effort, the problem can be solved in an effective and timely manner.

The future of this research will be to further advance the field of lower extremity safety through experimental tests and computer model development. Experimental needs include:

24. Investigation into the active properties of the joints and muscles in three dimensions, and the muscle force sharing across the joints during activation
25. The design of test fixtures to simulate vehicle occupant loading conditions
26. Material characterization of biological tissues at high strain rates
27. Collecting experimental data in a form best suited to validating FEM models

From a simulation standpoint, the future of the models will be to incorporate the following:

28. Full degree-of-freedom joints which accurately depict the motions of the human leg
29. The addition of discrete and finite elements to represent the active joint properties
30. The addition of viscoelastic contact surfaces around the skeletal structures
31. The application of material properties to all finite elements
32. A simultaneous effort to create highly detailed sub-component models (i.e. bone fracture, tendon rupture, etc.)

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