Explicit Finite Element Method Applied to Impact Biomechanics Problems

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Abstract

Biofidelic numerical models require accurate geometry, material properties, representative loading conditions, and proper verification and validation based on relevant experimental studies. Imaging techniques such as CT or MRI are typically used to generate anatomically correct model geometry. Accurate biological material properties and constitutive models present a challenge since tissues are typically anisotropic with nonlinear viscoelastic behavior and must be represented using a continuum approach. Verification and validation is a crucial development step to demonstrate the biofidelity of the model against experimental studies and is differentiated from model calibration, which involves adjusting FE modeling parameters to improve the agreement between the model and experimental data. Validation of the model should be performed in a continuous stepwise process, with increasing levels of complexity. Evaluating a numerical model against a variety of different loading conditions is the best means to obtain a fully validated model and this should be considered as a continuous process throughout the useful life of the model.

Keywords explicit finite element analysis, impact biomechanics, verification and validation.

I. INTRODUCTION

Impact Biomechanics is the study of human body response to impact loading, and injury resulting from mechanical interaction, where events are typically 100-200ms in duration corresponding to acute exposure [1]. Biomechanics is "the science that examines forces acting upon and within a biological structure and the effects produced by such forces"[2] whereas the term impact refers to physical impact or collision between two or more objects. Treatment of this impact may be rather straightforward, where conservation of momentum and energy may be sufficient to describe rigid or elastic body impact, while the concepts of wave mechanics and rate dependant material properties are required for interacting deformable bodies [3] when the impact transients are on the order of the natural frequency of the body. This is generally the case for the human body comprising many low impedance materials with low natural frequency relative to typical engineering structures. The concept of stress waves resulting from collision between two deformable bodies, and the transmission of these waves through a structure is a fundamental concept that must be considered in all impact biomechanics problems.

An important outcome from impact biomechanics research is protection of the human body from 'serious' injury [4], commonly encountered in automotive crash scenarios, accidental injury, assault, sports injuries, and the area of personal protection. A good example is the development of vehicle restraint systems including seatbelts which are credited with saving 13,000 lives per year [5]. Impact Biomechanics can be characterized by four main areas of study [4]: injury mechanisms, mechanical response to impact, human tolerance to injury, and simulation of impact through the use of surrogates or human analogs. The subject of impact biomechanics is often concerned with specific injuries, and their consequence in terms of threat to life. The Abbreviated Injury Scale (AIS) [6], originally introduced in 1971 and most recently updated in 2008, is an anatomically-based severity scoring system that classifies injuries according to body region on a six level ordinal scale providing a common language for the many groups involved in solving biomechanics challenges including medical professionals, epidemiologists, engineers and biomechanists. This scale is only concerned with ranking the severity of the injury but not the consequences [1]. For injury prediction, this is often implemented through an injury-risk curve, expressing the injury probability for a specific severity based on mechanical input or response.

A primary challenge in Impact Biomechanics is the development and design of improved protection, which often requires the 'test that cannot be done…', in that accurate assessment of the injury requires tests on living

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humans at injurious levels of loading. To address these needs, several predictive experimental techniques have been pursued including live human subjects with sub-injurious exposure levels, and a variety of physical human surrogates or simplified representations of the human body [1, 7]. Physical surrogates are used to evaluate impact response and predict trauma, with the primary requirements being biofidelity (a surrogate that closely resembles the response of the human body) and frangibility (a surrogate that sustains damage similar to human tissues). Physical surrogates include: Post Mortem Human Subjects (PMHS), animal models, and Anthropometric Test Devices (ATDs). We are typically concerned with evaluating a physical insult to the human body, which is done by predicting a response and correlating this to the corresponding biological injury, but must also consider longer term sequelae such as contusion.

Interest in impact biomechanics and protection from insult has existed for centuries [4], with some of the first directed experimental testing occurring in the 1920's and 1940's [4]. In the 1950's Col. John Stapp pioneered human tolerance research using rocket sled tests where he survived up to 40G decelerations to understand aircraft ejection, and began to investigate human tolerance to acceleration in auto crash events. Anthropomorphic Test Dummies (ATD) development began in the 1950's and led to the development of the Hybrid III in the 1970's, the basis for many crashworthiness and safety developments in vehicles to date. Several mathematical models have been proposed, for example the Lobdell lumped-mass thorax model [7, 8]; however these approaches are often limited in terms of the type, direction and severity of impact. Although physical surrogate testing continues to provide important guidance and safety improvements, there are limitations including a lack of physiological response at injurious levels of loading, ethical issues for PMHS and animal testing, surrogate response variability, and challenges with detailed experimental measurements for correlation to injury mechanisms at the tissue level.

Recently, numerical surrogates have become more common with increased speed and decreased cost of computing [1, 4, 7, 9] coupled with improved numerical techniques, material models [10, 11], and material data [11]. The primary benefit of numerical models is the ability to provide detailed insight into impact response at the tissue level, depending on the level of detail in the model, and to identify the importance of different tissues in terms of response and the potential for injury. In general, numerical surrogates include any numerical approach or algorithm to solve the multiple nonlinear equations resulting from many biomechanics problems. The methods include specific computer codes written to solve a series of equations, and more generalized methods including dynamics equations as applied to multi-body models and the finite element method.

The finite element method is a numerical approach used to solve partial differential equations, but is an approximation to the original problem, where a volume is discretized into smaller sub-volumes (elements) with a prescribed mass distribution and material behaviour. The finite element formulation allows us to consider complex geometry, non linear materials, collision and contact, material damage, and failure; however it must be noted that numerical models are idealizations, and contain many assumptions. It is the job of the modeler to ensure that the assumptions are appropriate, and to recognize the 'implied consent' associated with the use of a commercial finite element package and/or model to analyze an impact problem. In particular, a series of verification cases based on relevant experimental tests is essential for any code or model to ensure that the embodied assumptions are reasonable for impact scenario considered.

The explicit finite element method is used to simulate transient problems where the excitation frequency applied to the structure is approximately one third of the first or lowest natural frequency of a structure. This can also be expressed in terms of stress waves, when the transients in a loading pulse occur over time periods that are on the order of the time it takes for a stress wave to cross the structure (length divided by stress wave velocity) the problem is considered transient. In this case, material inertia and damping are significant and must be considered. The explicit formulation typically requires many computation cycles over very small time increments on the order of nanoseconds to microseconds, and thus benefits from a faster processor, or the use of multiple processors with smaller memory requirements than for the implicit method. In this formulation, the equations are solved at discrete time increments using a central difference integration scheme, where the selection of the time increment is important and must be sufficiently small to avoid divergence of the solution. The base time step is typically determined from the Courant condition [12, 13], the time required for a stress wave to cross an element in the model. In a simple form, this can be expressed as the length of an element divided by the acoustic wave speed (Appendix A). This method can present challenges for relatively long simulation times, such as simulating stress relaxation problems, where cumulative round-off error may require

the use of increased accuracy (double precision, for example). Explicit codes also allow for deformable bodies to come into contact during an impact scenario.

The basic elements of a finite element model include geometry (anatomy, anthropometrics), material properties (constitutive equations and material data) and boundary conditions (input or loading) (Fig. 1). It is essential that the model requirements be defined prior to model development to ensure that the appropriate level of detail be included in terms of specific tissues, connectivity and type of impact. As with any surrogate, the kinematic or kinetic response must be correlated to trauma (Fig. 1) and the predictive capabilities generally improve with smaller mesh scales provided the geometric and material information is available. It is important to ensure the model design is reasonable and balanced, keeping in mind computational costs, the prediction goals, available material properties and relevant validation data.

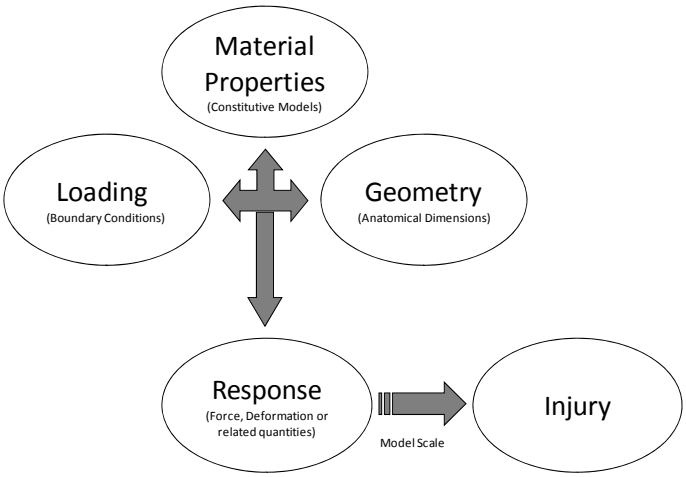


Fig. 1: Model input requirements and relation to predicted values

The primary goal of explicit finite element models is to predict human body response to physical insult and relate this to injury (Fig. 1). A primary challenge in this process is predicting or understanding biological injury and the importance of longer term sequelae such as contusion. Trauma can be predicted at multiple levels based on many different measurable responses at different scales, depending on the model requirements and available supporting information (Fig. 2). Early injury evaluations were based on global criteria such as gross vehicle acceleration or loads on occupants [1, 7] applicable to specific types and directions of loading. More recent focus has been on biologically-based injury criteria utilizing kinematic or kinetic response of the body, which are often still directional in nature. Ultimately, the goal of impact biomechanics is to predict response and injury at the tissue level, which requires detailed geometric and constitutive models of tissues, and an understanding of injury mechanisms and thresholds. At the organ/tissue level, trauma is associated with stress, deformation, strain rate, and wave phenomena. The actual mechanical parameters depend on the material properties and injury mechanism. This is still a developing area requiring information from the medical, experimental, numerical and epidemiological communities.

In practical terms, the two primary challenges are to ensure the finite element model is sufficiently refined to approximate the deformation gradients and forces in a body, and that the material models represent the material behavior over the range of strains and strain rates encountered in the impact scenario. The model or mesh may be refined in selected areas of importance, but care must be taken to avoid drastic changes in element size since the concentrated nodal mass will result in non-physical internal wave reflections within the model. This paper describes application of the explicit finite element approach to impact biomechanics problems, using simple impact between two cylinders (Appendix A) and a detailed thorax model (Fig. 3) as examples. The impact in Appendix A represents a classical impact problem for which mathematical solutions and a large body of experimental data exists [3, 12], while Fig. 3 demonstrates application of a three dimension impact biomechanics model [14, 15].

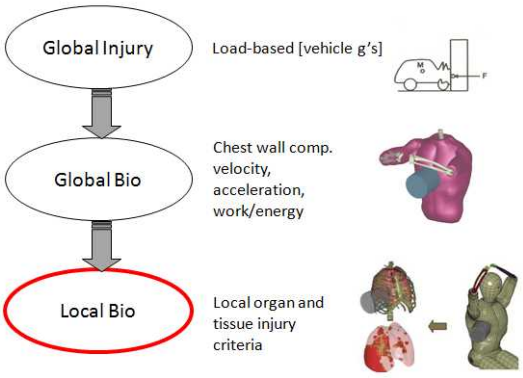


Fig. 2: Trauma prediction at multiple levels

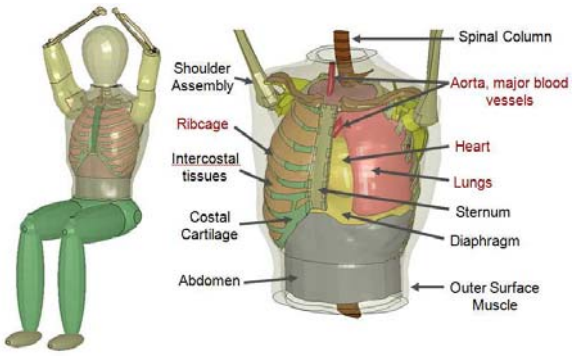


Fig. 3: Detailed finite element thorax model [15]

II. FINITE ELEMENT MODEL REQUIREMENTS

The first discretized thorax numerical models began to appear in the literature in the 1970's [16], incorporating rigid elements connected by springs and joints [17], and followed by more complex models with beams, plates and solid elements [18]. These multi-body approaches have continued to the present and have the benefit of capturing the general anatomy and mass distribution of the human body while being computationally efficient so that a large number of scenarios or scenarios of very long duration can be investigated. However, one of the limitations of this approach is the inability to predict local tissue response, which could be used to infer or understand injury mechanisms. To address this, detailed human body models have been pursued. Plank and Eppinger [19] developed a model with the individual ribs modeled and the internal organs considered as a lumped viscoelastic solid. This model was evaluated against PMHS tests and used for restraint system investigations. Huang [20] developed a thoracic model with representations of the internal organs, validated with PMHS tests. This was further developed by Shah [21] to evaluate aortic injury in crash scenarios. Although these early developments showed great promise, the level of detail in the models required significant assumptions so their applicability was limited to specific loading cases.

The level of detail in thorax and human body models has increased dramatically over the last decade due to the availability of detailed anatomical information and increased computing power [16]. In parallel with increased computing power allowing for significant increases in the number of elements and the level of detail in a model, significant advances have also been made in terms of material constitutive models, their implementation in finite element codes, and the required physical material test data at appropriate strain rates. At present, there are several three dimensional human body and detailed thorax models in existence including: the Total Human Model for Safety (THUMS) [22, 23], the Wayne State University torso model [24], the Johns Hopkins University model [25], L3 Communciations thorax model [26], and the Human Model for Safety (HUMOS) [27]. Many of the current models contain approximately 100,000 to 500,000 elements, with corresponding element sizes on the order of 2 to 10 mm. A detailed description of recent human body model developments is provided in [9]. For the purposes of this study, a detailed thorax model [14, 15, 28] with a simplified body was used as an example. The model (Fig. 3) includes 93,986 solid elements, 20,903 shell elements and 1,129 beam elements with 256 material definitions. Typical run times for a single analysis ranged from 1.5 to 2 days (4 cores, Intel i7 2.67 GHz 64 bit processor, LS-Dyna Version 971 R4.2.1, single precision). The primary requirements for the model included: (Phase 1) Numerical stability for common impact scenarios, the ability to predict gross kinematics of the thorax with a focus on side impact, validation using pendulum impact and side sled scenarios, incorporation of simplified internal organs, (Phase 2) development and integration of internal organs, and (Phase 3) integration with a full vehicle model to investigate crash scenarios [14, 15, 28].

Model Geometry

Model geometry or anatomical dimensions are available from a wide variety of sources including online databases such the Visible Human Project [29] (Fig. 4) or can be developed directly from CT, micro CT and MRI scans. In general, these methods provide data that must be represented using surfaces and subsequently converted to volumes. Subject-specific data should always be compared against the accepted anthropometric

dimensions and posture for the gender and body size considered.

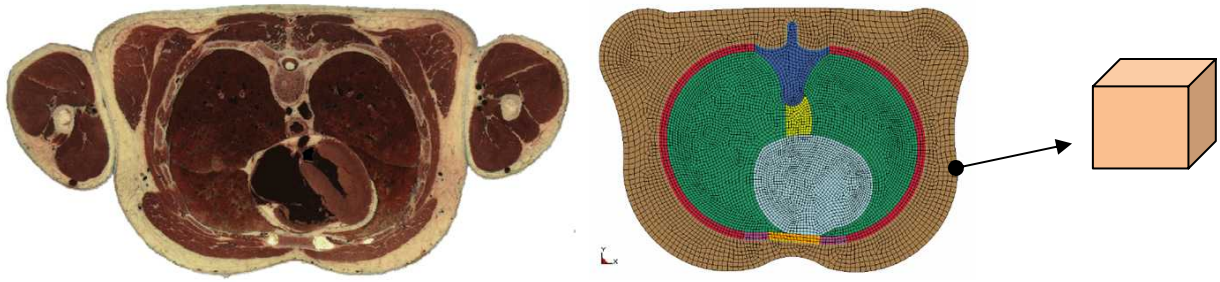


Fig. 4: 50th percentile male torso, transverse section from Visible Human Project [L] and mid-sternum cross-section [R]

The finite element method requires that a surface or volume be discretized into elements of a specified size and shape, with a prescribed response based on mass distribution and the element formulation, where elements are actually defined using a series of points in space (nodes) and the element mass (density times volume) is evenly distributed between the element nodes. Thus, the size of the elements and resulting mass distribution can play an important role in the response of a tissue particularly for localized impacts on large elements where non-physical distribution of mass can lead to erroneous results.

Tissues can be discretized using one dimensional beam, truss or tension-only elements. For example tension-only elements are often used to simulate a ligament with the assumption that the force line of action is directed between the element nodes and the ligament attachment occurs at a specific location on the corresponding hard tissue. Two-dimensional shell elements are often used for structures that are very thin relative to their planar dimensions and provide computational efficiency compared to three-dimensional elements. The cortical bone shell occurring on hard tissues, such as the ribs, is on the order of 1mm in thickness and is often modeled using shell elements. Solid or hexahedral elements are required to model larger volumes, for example cancellous bone or lung tissue. The basic assumption in the discretization process is that a continuum approach is appropriate. It is often not feasible to model at the structural length scale, for example the alveolar level (approx 300 µm diameter) and we assume a continuum approach with appropriate material properties to model the tissue as a continuum. In general, it is recommended to use the simplest element formulation available that will meet the model requirements, in order to reduce computational costs.

An important aspect of meshing a volume is to ensure that the geometry, mass distribution and total mass are represented accurately. When discretizing curved surfaces with linear (straight side) elements, the faceted surface approximation results in missing material (mass or volume) as shown in the end view for a ¼ cylinder (Fig. 5) as a function of the element size normalized by the radius of the surface. The error in volume/mass can become significant for elements which are large relative to the radius of curvature. The individual components and complete model should be reviewed for conformance to the original CAD surfaces and expected volume.

It should be noted that triangular shell and tetrahedral solid elements are commonly available (full and reduced integration) with the benefit that these elements are suitable for meshing complex geometry with automated meshing algorithms; however, the linear or single integration point triangle or tetrahedral elements are often subject to locking for incompressible materials and should be used with caution. Although fully integrated hexahedral (brick) elements are available in most explicit finite element codes, single integration point or reduced integration elements are commonly used due to computational efficiency and numerical stability. The primary challenges with single integration point elements include the resulting constant stress, requiring more elements to model a stress or deformation gradient compared to fully integrated elements, and the potential for 'hourglass' deformation referring to modes of deformation which result in nonphysical oscillations with no stress in the element. This numerical artifact is controlled through the addition of artificial viscous damping or stiffness to control these modes of deformation and the model should be monitored to ensure the hourglass energy remains small (<~10%) relative to the system energy. However, significant hourglassing of elements is best controlled through finite element mesh refinement.

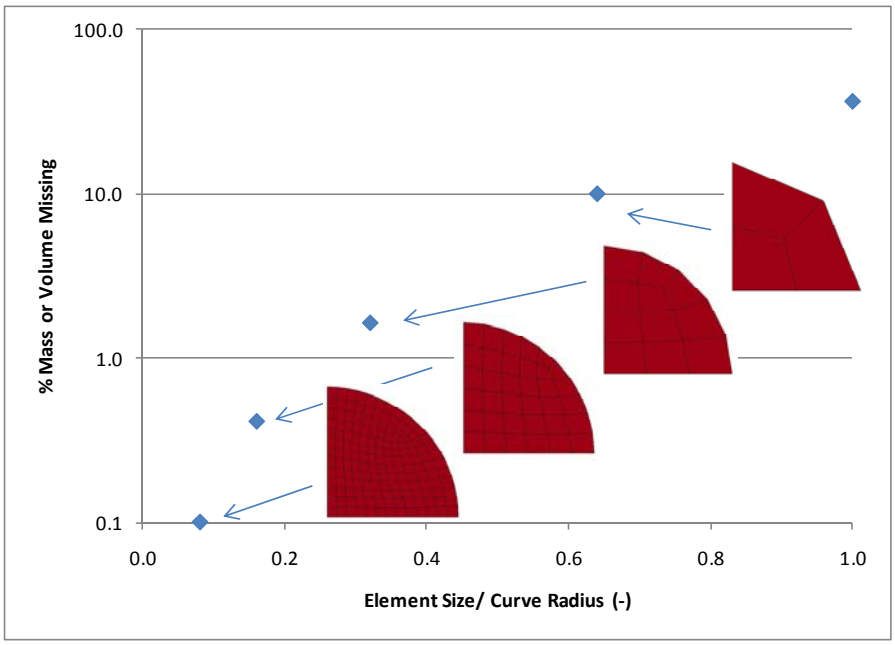
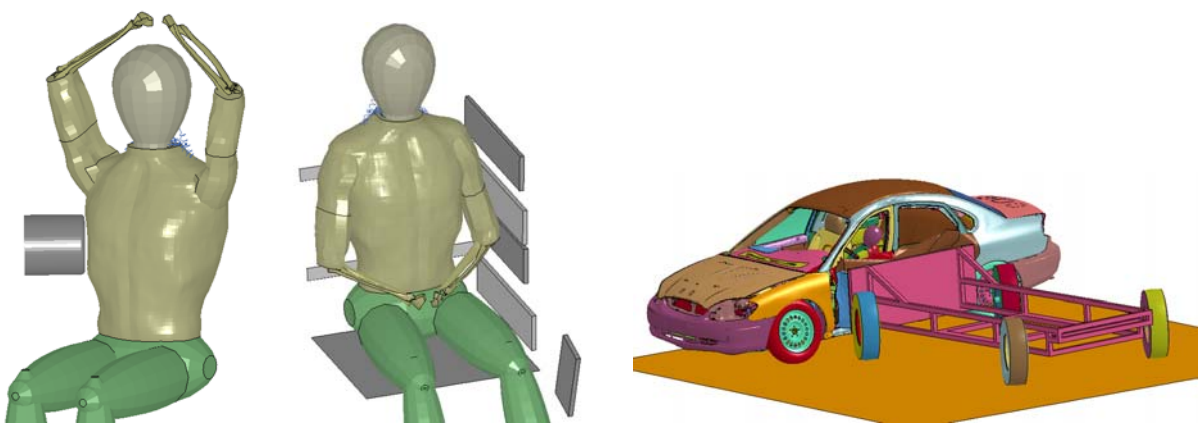


Fig. 5: Discretization of curved surfaces (1/4 cylinder) and corresponding missing volume

Loading (Boundary Conditions)

Boundary conditions represent the loading and applied displacements experienced during an impact scenario. This can range from a simple fixed or applied velocity condition to modeling a human body within a full vehicle model. For the simple example of impact between two cylindrical bars (Fig. A2), the full impact can be modeled (Bars A and B), or Bar A can be omitted and the interface velocity can be applied directly to Bar B. This approach can be successful and significantly reduce computation time if the physics of the impact are completely understood. Unfortunately, for most problems of interest this is not the case and a model of the actual impact scenario is required. A model should be constructed in a logical stepwise fashion, initially using the most simple impact scenario available to ensure that these initial models can be used to validate the human model. For more complex scenarios, the impact scenario should be validated and verified separately from the human model as much as possible to ensure the predicted response is physically correct. A primary challenge in this area is coupled loading, where the body response is affected by the impacting structure, and the structural response is similarly affected by the presence of the body. A coupled problem, such as occupant interaction with a vehicle door in side impact [30] presents unique challenges in terms of model validation.

In general, we seek a representative and repeatable loading condition to validate the model with subsequent cases progressing in complexity. An example is the thorax model, subjected to simple free-flight pendulum (frontal, oblique) and limited stroke (lateral) impacts (Fig. 6). This was followed by side sled tests, simplified side impact with deformable door, and finally integration in a full vehicle model.



Oblique Pendulum [15] WSU Side Sled [15] Full Vehicle with Torso Model [30]

Fig. 6: Torso model impact cases, progressing from simple to complex loading scenarios.

Material Properties

The measurement of physical material properties of tissues, description using constitutive relationships, and implementation in numerical codes represents one of the largest areas for development in the area of impact biomechanics. Mechanical properties refer to the response of tissues to a disturbance, often a prescribed displacement at a prescribed displacement rate, and the corresponding force. These values are then typically presented in terms of stress and strain to remove dimensional dependence of the properties. A constitutive model is simply an equation relating stress and strain, with the simplest version being Hooke's law for an isotropic linear elastic material. In some cases, such as modeling ligaments with one-dimensional tension-only elements, the measured force and displacement may be used directly by the element.

Although biological materials are well-known to exhibit hyperelastic (non-linear elastic) behavior with viscoelastic (strain rate) effects [11](Fig. 7) it is often recommended that model development begin with simple material models to allow for initial development and scoping studies. Complex material models should be developed in a stepwise fashion and tested independent of the larger model using single element and small multi-element test cases to ensure the material model provides the expected results. It must be emphasized that the applicability of material properties to a detailed model is a question of scale, and implementation of a specific set of material properties depends on the sample size used in the physical testing to measure the material properties.

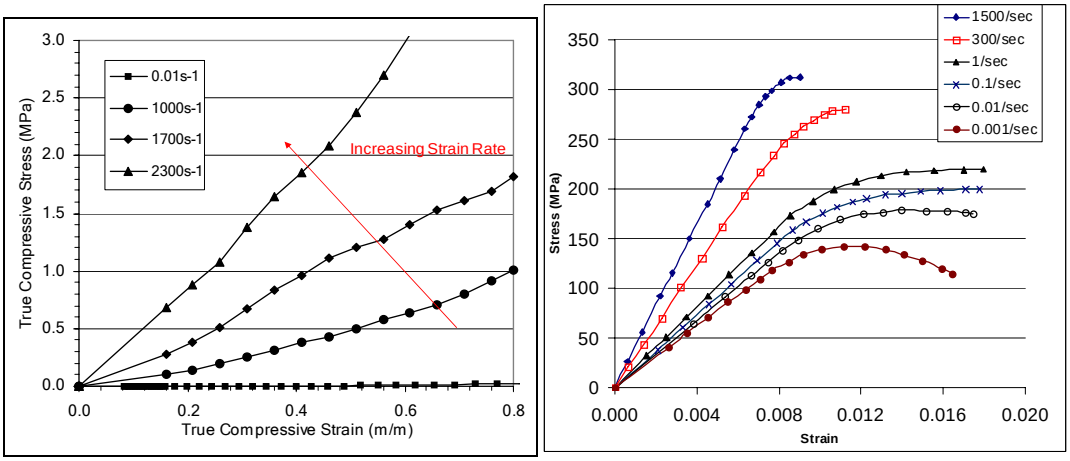


Fig. 7: Typical mechanical properties of soft tissue (muscle, left [31]) and hard tissue (cortical bone, right [11])

Mechanical properties are typically separated in terms of deviatoric or deformational components (Eqn. 1) and bulk or volumetric components (Eqn. 2) for implementation in constitutive models. The deviatoric components refer to strain, and the corresponding stresses, that result in distortion of the material, while the bulk or hydrostatic components result in dilation of the element. Importantly, many constitutive models assume the material is nearly incompressible with a relatively high Poisson's ratio and corresponding bulk modulus. Although this assumption is valid in cases where the hydrostatic pressure is not significant, these assumptions do have an important effect on the calculated material wave speed and resulting time step. Importantly, incorrect assumptions can lead to spurious high pressures and changes in deviatoric stresses. Both hard and soft tissues are known to exhibit a dependence on strain rate, and the need for rate dependency in a model should be determined by first simulating the impact scenario using quasi-static properties to provide a conservative estimate of the expected strain rates. If the predicted rates warrant inclusion of rate dependant properties, this should be included. If there are no mechanical properties available to judge this aspect, then it should be listed as a limitation of the model. A complete treatment of continuum mechanics and constitutive models is not possible in this study, and the reader is referred to constitutive modeling and biomechanics references [10, 11, 12].

$$\sigma = f(\varepsilon', \dot{\varepsilon}), \qquad (1)$$

where σ is deviatoric stress (Pa), ε' is deviatoric strain (-) and $\dot{\varepsilon}$ is strain rate (1/s).

$$p = f(\varepsilon_V), \qquad (2)$$

where p is pressure or volumetric stress (Pa), ε_V is volumetric strain (-).

The relationship between pressure and volume can include nonlinearities which is necessary for relatively high pressures, and the dependence of pressure on internal energy is required for a complete equation of state.

III. MODEL EVALUATION

Numerical models require appropriate verification and validation to be considered predictive within the assumptions and limitations of the model. The ASME Guide for Verification and Validation in Computational Solid Mechanics [32] provides some basic definitions and guidance for numerical model development.

A model should be constructed using a bottom-up approach through identification of the relevant model sub-structures and development of the model at this level. For example, the primary structural components in the thorax include the ribs, spine and sternum with the associated connectivity. Secondary structures include the musculature, ligaments and finally the internal organs. Once identified, each of these structures should be included with appropriate connectivity to adjacent structures. The level of detail required will depend on the intended uses for the model.

Key to model evaluation is the identification of response metrics, commonly based on measured quantities such as displacement, acceleration, force or other derived quantities. Identification of these parameters is important since they should be numerically robust, supported by experimental evidence, and be meaningful in terms of the intended model purpose. A numerical model is inherently precise, since we predict the same answer for the same inputs, and we are primarily focused on accuracy. However, both precision and accuracy need to be considered for experimental results. The selection of relevant metrics is often driven by the available experimental results, and we are somewhat limited in terms of what and how much we can measure experimentally, but should still be evaluated using a Phenomena Identification and Ranking Table (PIRT) [33], where metrics or response phenomena are ranked according to the importance of the response (low, medium, high) and the level of confidence in the model to predict these quantities (low, medium, high). It is also important to consider a measure that is sensitive to relevant changes in inputs.

The comparison of numerical and experimental results can be undertaken at several levels of complexity, with the relevance of a given metric and sensitivity to input changes being the most important criteria. One of the biggest challenges is that the responses are transient for impact phenomena, making meaningful direct comparison challenging. An incremental approach to comparison is important, beginning with simple approaches to ensure the model is within the correct order of magnitude, followed by more complex processes to evaluate specific aspects of the model.

Simple approaches include the comparison of measured and predicted peak values and the corresponding timing of the event, and this should always be considered as a starting point to evaluate a model. Transient experimental data is often provided as response corridors, with an 'average' response quoted. Care must be taken in interpreting this data since generating corridors (statistical bounds based on the average or an envelope to capture the maximum/minimum response) can remove some aspects of the physical response including local peaks that may occur at different times for different tests. ISO [14] provides qualitative descriptions for this process: good (falling within the experimental corridor), reasonable (falling outside the corridor, but within one corridor width) and poor (falling outside the corridor by more than one corridor width). To address the subjective nature and limitations of simple approaches, several methods have been proposed to facilitate comparison using cross-correlation methods [34, 35]. Cross-correlation methods, involving techniques such as Fourier transforms, are used to provide a measure of how well a transient or time-varying signal correlates with another signal as measured by a dimensionless correlation coefficient ranging from -1 to +1. Comparison of transients necessitates phase shift and this is often undertaken to maximize the correlation coefficient.

Jacob et al. [33] have proposed four approaches to model evaluation:

- Global evaluation method – comparison is made to discrete values from a parameterized response
- Threshold evaluation method – the predicted response is compared to relevant threshold values for the signal
- Criterion evaluation method – this method assumes that a relevant assessment criterion, typically a mathematical formula such as the Head Injury Criterion or specified signal processing approach,

which is compared to a critical value or injury-risk curve

- Limit evaluation method – a method for comparing a transient signal to response corridors, where the proportion of the signal within the corridors determines the correlation between the measured and predicted response.

In all cases, there are often multiple measures of performance and the results are combined using a weighting scheme to produce an overall score or single value. These approaches have also been captured in software such as CORA (Partnership for Dummy Technology and Biomechanics, Ingolstadt, Germany). It should be noted that this method requires meaningful average experimental response to be defined, and the definition of several fit and weighting parameters to calculate a correlation coefficient. The most important aspect of any detailed comparison method is transparency in terms of the flexibility in the fit and evaluation parameters so that the resulting correlation value can be interpreted correctly.

An important aspect for all transient signals is filtering. This is undertaken to remove unwanted portions of the signal or artifacts, such as high frequency resonance, in experimental data. An ideal low-pass filter allows only frequencies below the cutoff point to pass, and rejects all frequencies above this point. The approach and requirements are specified in SAE J211, providing a common basis for comparison of test data from different sources. Further, it is important that the sampling frequency (experimental and numerical) be sufficient to avoid aliasing, an effect that causes different continuous signals to become indistinguishable (or aliases of one another) when sampled. In general, the sampling frequency must be greater than the difference between the maximum and minimum frequencies of the signal sinusoidal components (bandwidth).

Following initial model development and the identification of simple test cases, model verification should be initiated. Model verification is the process of ensuring that the computational model accurately represents the intended mathematical model and the corresponding solution, as implemented in the solver. Importantly, the assumptions in the model and corresponding intended uses must be documented through a configuration management process. Initial model developments should include modeling of simple cases with known solutions to ensure the model and code or solver is working as expected. A series of baseline verification cases with known solutions should be developed, and must be evaluated when moving to a new version of the solver, or modifying the model. The verification process should include an investigation of finite element mesh refinement to ensure the mesh is sufficient to model the process. A refinement study begins with an initial model, meshed to sufficiently represent the geometry of the tissue or structure. An evaluation parameter, typically selected from the available data in the verification cases, is predicted in an impact scenario and the mesh is then refined. This is typically accomplished through element splitting since most preprocessors can do this in a straightforward manner. Care must be taken to avoid generating a mesh that is not computationally feasible, and may require selective refinement. The results from three different mesh sizes can be used with Richardson extrapolation [36] to estimate the 0 mm element size solution and with the Grid Convergence Index (GCI) can provide guidance on the appropriate mesh size. The GCI provides an error bound estimate for a given element size, but still requires judgment in terms of defining an acceptable mesh size. However, this process provides associated performance metrics to support the mesh size decision.

Model validation incorporates the assembly of data to support the model implementation and predictions, and is defined as “the process of determining the degree to which a model is an accurate representation of the real world (based on the intended uses for the model)” [32]. This process should include the collection and evaluation of a wide range of data including different modes and rates of loading to provide a comprehensive evaluation of the model. Validation of a complex model or system should be undertaken in a hierarchical fashion beginning with simple impact scenarios followed by more complex scenarios. It should be noted that, for any model, the validation is not a final statement regarding the model, but a continuous process of evaluation which will develop with the model and as available validation data develops. It should also be noted that validation is specific to a particular computational model for a particular intended use.

It is important to distinguish between verification and validation and model calibration. Calibration refers to the process of adjusting model parameters, in many cases to nonphysical levels, so that the model predicts the intended result. This can involve adjustment of the level of geometric detail, finite element mesh design, but most often is undertaken with material properties. In some cases, this may be undertaken with parameter identification, which can produce valid material properties if the constitutive model is appropriate, the properties are within physically expected bounds, and the model can be independently validated. Model

calibration is typically a response to incorrect representations or insufficient detail in the model. This can be considered a valid process, as long as the assumptions and calibrated parameters are clearly stated, and the intended use of the model is within the calibrated limits, but can also be useful in highlighting areas for improved development. However, this inherently limits the predictive capabilities of the model, one of the primary motivations for the development of detailed human body models. Concurrent with this process, the sensitivity of a model to the input parameters should be investigated to better understand the impact scenario and response of the model. This is of particular importance with respect to the loading conditions and material properties.

Given our goal to develop detailed and predictive human body models to understand response and injury, the process of model development, verification and validation is a significant undertaking and should be considered a continuous process throughout the life of the model.

IV. CONCLUSIONS

The explicit finite element approach provides a means for detailed simulation of nonlinear impact biomechanics problems. Numerical models, in conjunction with physical surrogates, impact experiments and tissue mechanical properties can provide important insights into impact phenomena that are not easily measured experimentally and are too complex for simple mathematical calculations. The goal of impact biomechanics research is to protect the human body from 'serious' injury, with the primary challenges being lack of physiological response in models and the need for tissue-level injury criteria.

A model should be designed to answer specific and directed questions, and therefore must be designed with intent and clear performance specifications. A model is a surrogate, not a substitute, for tissues and organs and only represents specific aspects of the tissue depending on the level of detail in the model. The primary inputs include geometry, loading or boundary conditions, and mechanical properties at a length scale relevant to the scale of the model. The human body consists of many hierarchical structures and can be described in terms of scale as atoms, molecules, cells, tissues, organs and finally the organism. We generally work at the organ or tissue level, so that a continuum approach is acceptable. Mechanical properties capturing the strain and strain rate ranges experienced during an impact and associated tissue damage criteria represent the most significant areas for development at present.

Numerical models require appropriate verification and validation to be considered predictive within the assumptions and limitations of the model. Key to model evaluation is the identification of relevant response metrics and comparison of numerical and experimental results, which can be undertaken at several levels of complexity. The development of experimental tests and accuracy requirements should be based on the intended use of the model with the relevance of a given metric in terms of injury prediction, and sensitivity to input changes, being most important. Model verification, an assessment of the implementation and associated assumptions, must precede validation. Validation, or assessment of the extent to which a model represents the phenomena investigated, should be pursued in a hierarchical fashion (simple to complex) and is specific to a particular computational model for a particular intended use.

The ultimate goal of developing detailed and predictive human body models is to understand response and predict injury. To meet these requirements the process of model development, verification and validation is a significant undertaking and should be considered a continuous process throughout the life of the model.

V. ACKNOWLEDGEMENT

The author would like to thanks the many graduate students who have contributed to the development of the models discussed, and to Jason Fice and Jeff Moulton for reviewing the manuscript.

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VII. APPENDIX A - STEREOMECHANICAL IMPACT OF CYLINDRICAL BARS

Collinear impact of two linear elastic cylindrical bars is a simple impact scenario to provide a basis for more complex impacts between dissimilar materials. This approach is widely used in the high strain rate testing of materials in the form of a Kolsky or Hopkinson bar impact apparatus.

Consider two cylindrical bars (**A** and **B**) shown in Fig. A.1. Bar **A** is of finite length **L**, with a diameter **d** much smaller than the bar length, and is moving at velocity **V** to the right. Bar **B** is much longer than bar **A** and is initially stationary. At time t=0 seconds the bars make initial contact.

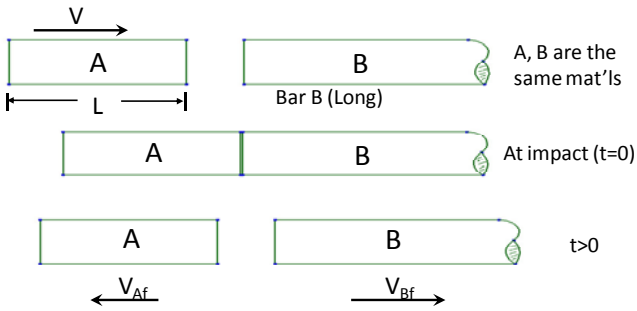

Fig. A.1: Simple impact between cylindrical bars

Straightforward application of conservation of mass, momentum and energy provides us with a simple rigid body approximation of the final conditions. In this case, some of the energy of bar **A** is transferred to bar **B**, and at some point later in time bar **A** has velocity V_{Af} to the left and bar **B** is moving to the right with a velocity V_{Bf} where $m_A<m_B$. The resulting velocity depends on the relative masses of bars **A** and **B** and assumes the bodies are rigid with an effective mass occurring at the centroid of the body.

$$V_{Af} = \frac{(m_A - m_B)}{(m_A + m_B)}V \qquad V_{Bf} = \frac{2m_A}{(m_A + m_B)}V \qquad \text{(A.1)}$$

A numerical simulation of two bars (Table A.1) impacting was conducted using a commercial explicit finite element program (LS-Dyna, LSTC). A one-quarter model with symmetry conditions was used for computational efficiency (Fig. A.2) with a finite element mesh size of approximately 2 mm. In the first case, the bars were treated as rigid in the numerical model, and the resulting velocities were in agreement with equations A.1. For V=10m/s, V_{Af}=-4.12 and V_{Bf}=5.88 m/s

Table A.1: Bar properties

	Bar A	Bar B
Initial Velocity (m/s)	10	0
Density (kg/m^3)	1850	1850
Modulus (GPa)	18.8	18.8
Poisson's Ratio '(-)	0.312	0.312
Diameter (mm)	25	25
Length (mm)	250	600
Mass (kg)	0.23	0.54

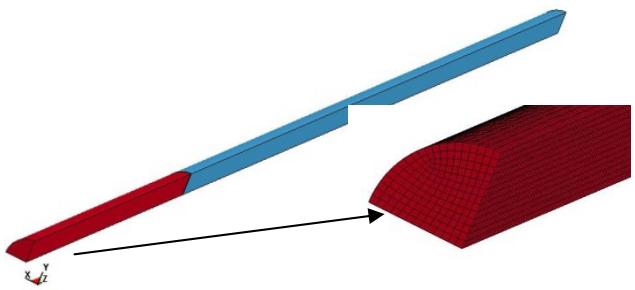

Fig. A.2: One quarter model of bar impact problem

Although valuable for simple analysis of impact problems, the previous analysis neglected the internal energy in the bodies due to wave transmission, and the resulting vibration. If we consider the impact for the case of elastic (deformable) bars, we need to consider that an elastic wave will be generated by the impact and will travel at a finite speed through the body (Fig. A.3). The wave speed, also known as the acoustic or sonic speed is a characteristic of the material, which depends on the material stiffness and density. The response of the bars at the interface can be described in terms of the particle velocity (Up) and the wave velocity (C). At t=0, bar **A** contacts bar **B**. This contact occurs over a relatively short but finite time, the time it takes for the wave to travel twice the length of bar **A**, and results in a particle velocity (Up) at the interface between the bars. The free surface of bar **B** moves with particle velocity Up to the right, and the free surface of bar **A** moves with the relative particle velocity Up to the left. For the case where both materials are the same, and the bars are the same diameter, Up is equal to one half of the impact velocity **V**. The particle velocity results in a compressive stress wave, which propagates through each bar at the wave speed C. The physical length of the wave in bar **B** is twice the length of bar **A**.

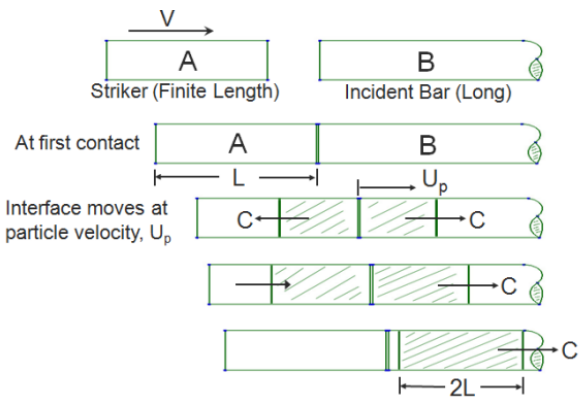

Fig. A.3: Simple impact between cylindrical bars

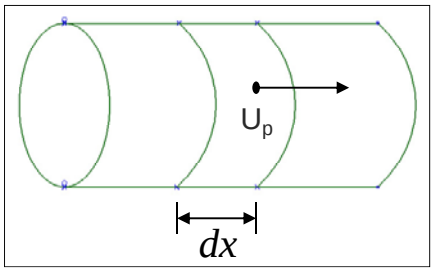

Fig. A.4: Conservation of momentum

Application of the conservation equations

Applying the conservation equations, we can determine the magnitude of the stress wave in the bars, and the velocity of the wave.

Consider a small length of the bar (of length dx) (Fig. A.4), where the applied impulse (F.dt) is equal to the change in momentum. The mass of the small length is m with cross sectional area A, and this section moves with the particle velocity Up.

$$F\,dt = d(mU_p)$$

$$\sigma A dt = \rho A dx U_P$$

$$or$$

$$\sigma = \rho \frac{dx}{dt} U_P \qquad \text{(A.2)}$$

$$\sigma = \rho C U_P$$

$$where$$

$$U_P = \frac{1}{2} V$$

$$\sigma = \frac{1}{2} \rho C V$$

Equations A.2 provide us with an estimate of the stress wave magnitude σ resulting from the impact.

Applying Newton's second law for the small section (Fig. A.4):

$$F = m\,a$$

$$-\left[A\cdot\sigma - A\left(\sigma + \frac{\partial\sigma}{\partial x}\cdot\delta x\right)\right] = A\cdot\rho\cdot\delta x\cdot\frac{\partial^2 u}{\partial t^2} \qquad \text{(A.3)}$$

$$\frac{\partial\sigma}{\partial x} = \rho\frac{\partial^2 u}{\partial t^2}$$

And including the stress and strain relationships, we can arrive at the one-dimensional wave equation:

$$E = \frac{\sigma}{\varepsilon},\ \varepsilon = \frac{\partial u}{\partial x} \qquad \frac{\partial^2 u}{\partial t^2} = \frac{E}{\rho}\frac{\partial^2 u}{\partial x^2} \qquad \text{(A.4)}$$

We can solve this equation using separation of variables to give:

$$u(x,t) = F(x - C\cdot t) + G(x + C\cdot t) \qquad \text{(A.5)}$$

F and G are functions describing the wave in the positive and negative directions, respectively. Noting that C is the elastic wave velocity:

$$C = \sqrt{\frac{E}{\rho}} \qquad \text{(A.6)}$$

It should be noted that this result is a consequence of the assumptions in the elementary theory approach. In particular, that the bar diameter is small relative to the length of the wave resulting from the impact. In general, the speed of wave propagation in an unbounded medium is:

$$C = \sqrt{\frac{K + 2/3G}{\rho}} \qquad \text{(A.7)}$$

where K is the bulk modulus and G is the shear modulus.

For the properties in Table 1, the wave speed is 6163 m/s and the magnitude of the stress wave for a 10m/s impact is 29.25 MPa. The physical length of the wave in bar **B** is 0.5m (twice the length of bar **A**), and this corresponds to a time duration of 379 µs. On impact the energy of bar **A** is transferred to bar **B** and following the impact bar **A** is stationary. Advanced analysis of this scenario shows that the situation is somewhat more complex, and the response includes a higher frequency component resulting from free surface effects and depending on the ratio of bar diameter to wave length, known as Pochhammer-Chree oscillations.

Vibration Analysis

Vibration analysis of a cylindrical bar provides the following relationship:

$$f = C/(2l) \qquad \text{(A.8)}$$

Where f is the natural frequency or 1st mode of vibration in hz (cycles per second), C is the wave speed and l is the length of the bar. For the properties in Table 1, the frequency is 2635 hz for bar **B**.

Numerical Impact Model

The one-quarter impact model (Fig A.2) was analyzed using different finite element mesh sizes to evaluate the impact. The results (Fig A.5) demonstrate that, as the mesh size decreases, the predicted results are closer to the plateau predicted by elementary theory. Also, for larger element sizes the initial peak is higher. All methods predict the Pochhammer-Chree oscillations, with smaller element sizes. Additional analyses (not shown) demonstrated nonphysical responses when considering tetrahedral elements and non-uniform finite element meshes. All element sizes were found to provide a good estimate of the natural frequency of the bar (Fig. A.6). This result is expected since this depends directly on the defined material properties and bar length (Eqn. A.8), and fundamental to the explicit formulation and element definition. However, this also highlights the need for the selection of appropriate response metrics since this particular quantity is relatively insensitive to the finite

element mesh size. One aspect of this model that requires further analysis is the dispersive effects introduced in the explicit finite element formulation.

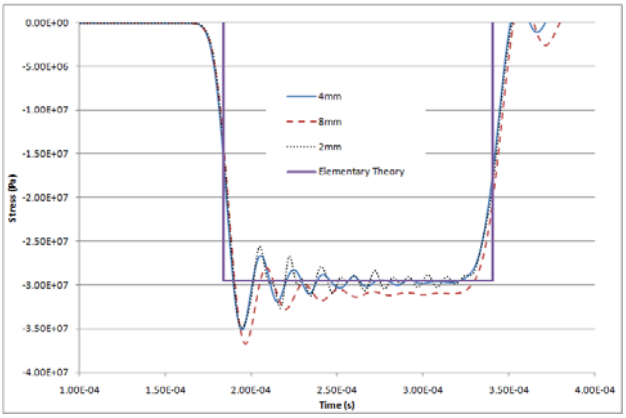

Fig. A.5: Predicted impact response (bar **B**) compared to elementary theory.

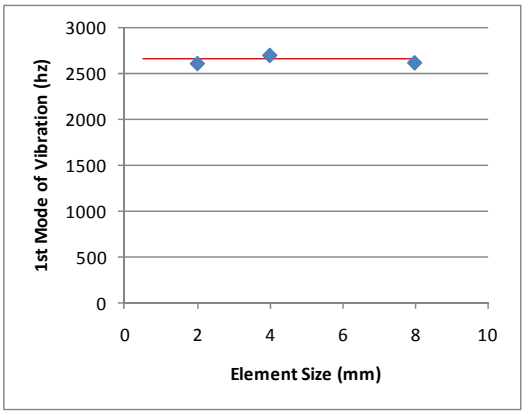

Fig. A.6: Predicted natural frequency