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ABSTRACT

A lower extremity model has been developed to study occupant injury mechanisms of the major bones and ligamentous soft tissues resulting from vehicle collisions. The model is based on anatomically correct digitized bone surfaces of the pelvis, femur, patella and the tibia. Many muscles, tendons and ligaments were incrementally added to the basic bone model. We have simulated two types of occupant loading that occur in a crash environment using a non-linear large deformation finite element code.

INTRODUCTION

The Federal Government and the automobile industry are both concerned with vehicle occupant safety. The automobile industry has increased its research in assessing the response of vehicles and passengers in accident scenarios. Passive passenger restraints, supplemental restraint systems, energy absorption areas, and side impact stiffeners all work to minimize the effect of vehicle collisions on the occupants. Due in part to a lack of restraint, the lower extremities remain very susceptible to severe injury during collision.

A finite element model has been developed to allow the study of injuries to the major bones and ligamentous soft tissues as a result of automobile collisions, but may also be used for studying sports injuries. The nonlinear finite element code DYNA3D (Whirley and Englemann, 1993) was used to determine the potential for injury as a result of various loading scenarios. The kinematic/kinetic behavior of the lower extremity is determined by bones and ligaments, tendons, the interaction between these components, and the applied external loading.

The finite element method is an obvious choice for modeling joint mechanics because it provides a consistent way to discretize structures. It also yields an accurate description of the complex geometry, material inhomogeneities and contact. The method can accommodate large deformations in conjunction with anisotropic, viscoelastic, or nearly incompressible material behavior, all of which are characteristic of biological tissues (Wismans et al, 1980), (Puso et al, 1995), (Weiss et al, 1994).

METHODS

A commercial mesh generator was used to construct the finite element model (XYZ Scientific Applications, Livermore, CA). The starting point for the development of the model was a detailed, 3-D surface description of all the major bones and soft tissue in the lower extremity (Viewpoint DataLabs Int'l, Orem, Utah). Then, 3-D blocks of solid elements were mapped to the surface descriptions of the bones and soft tissues. This resulted in an accurate solid hexahedral element model of all essential bones of the lower extremity. This same approach was used for some of the soft tissues. The bones were modeled as isotropic, elastic-perfectly plastic materials of initially homogeneous density (E=2.0E+06 psi $\nu = 0.3$, ρ = 1.404E-06 lb sec^2/in^4). The modulus lies within accepted tensile and compressive values (Yamada, 1970). For stress levels in excess of 12 ksi (approximately the ultimate strength of compact bone), a bilinear constitutive model was used. The long bones were modeled as solid when, in fact, the shafts of long bones are filled with marrow. The lateral and medial menisci were explicitly represented with solid elements using isotropic material properties based on a published report, while the articular cartilage was neglected. Future model improvements will include the cartilage layers, hopefully providing for more accurate prediction of the load transfer into the distal femur and the proximal tibia due to loading across the joint during impact.

Because of the high-rate dynamic nature of the simulations of automobile collisions, inclusion of appropriate inertial properties in the model is essential. We chose to lump the muscle mass at nodes of underlying bone elements. This approach underestimates the effective inertia of the leg about its long axis, but is accurate for calculating the inertia corresponding to flexion and extension about the joints.

The major ligaments and tendons of the lower extremity were approximated either by 1-D discrete elements or by 3-D continuum elements. Estimates for the stiffness of the discrete elements were taken from the literature, while insertion and origin points for all soft tissue were obtained from Delp, 1990.

RESULTS

As an approximation of a crash event, occupant deceleration and loading during a sled test has been simulated. The pelvis was fixed in a manner that represents a belted occupant. The foot was fixed to the toepan of the vehicle and the actual toepan displacement (sled test data) was used as a boundary condition. The soft tissues (ligaments, tendons and muscles) were pre-loaded to represent an occupant that has his foot on the floor and relaxed at the time of impact. The simulation was compared against results of existing sled deceleration test data incorporating the Hybrid-3 crash dummy or instrumented cadavers. The computer simulation indicated qualitative agreement with the sled test observations. Thirty milliseconds after a deceleration pulse to the foot, the foot is pushed back and the knee forced upward. This initial simulation produced high levels of von Mises stress in the patella, tibial shaft and the femoral head region.

The second crash simulation is that of the knee impacting a generic knee bolster. For this type of loading we have calculated excessive stress levels in the knee and pelvic regions that correspond to observed fracture patterns. The results of this simulation still need to be compared with existing test data.

CONCLUSIONS

A modeling approach was adopted that accurately captures the mechanics of lower extremity motion. This approach assumed that the leg was passive during its response to the excitation, that is, no active muscular contraction and therefore no active change in limb stiffness. The approach called for detailed geometric modeling of the bones to allow accurate definition of muscle origin and insertion points and to determine the spatial variation of response quantities. The approach recognized that the most important contributions of the muscles to lower extremity response are their ability to define and modify the impedance of the limb.

The biological materials that comprise the lower extremity are complex both in composition and response. This model has taken a simple approach to produce computational results. When nonlinear material behavior in a component of the leg model was deemed important to response, a nonlinear constitutive model was incorporated. The accuracy of these assumptions can be verified only through a review of analysis results and careful comparisons with test data.

As currently defined, the model meets the objective for which it was created. Much work remains to be done, both from modeling and analysis perspectives, before the model can be considered complete. The model implements a modeling philosophy that can accurately capture both kinematic and kinetic response of the lower limb. We have demonstrated that the lower extremity model is a valuable tool for understanding the injury processes and mechanisms. We are now in a position to extend the computer simulation to investigate the clinical fracture patterns observed in actual crashes. Additional experience with this model will enable us to make a statement on what measures are needed to significantly reduce lower extremity injuries in vehicle crashes.

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