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Construction and evaluation of thoracic injury risk curves for a finite element human body model in frontal car crashes

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ABSTRACT

There is a need to improve the protection to the thorax of occupants in frontal car crashes. Finite element human body models are a more detailed representation of humans than anthropomorphic test devices (ATDs). On the other hand, there is no clear consensus on the injury criteria and the thresholds to use with finite element human body models to predict rib fractures. The objective of this study was to establish a set of injury risk curves to predict rib fractures using a modified Total HUman Model for Safety (THUMS). Injury criteria at the global, structural and material levels were computed with a modified THUMS in matched Post Mortem Human Subjects (PMHSs) tests. Finally, the quality of each injury risk curve was determined. For the included PMHS tests and the modified THUMS, DcTHOR and shear stress were the criteria at the global and material levels that reached an acceptable quality. The injury risk curves at the structural level did not reach an acceptable quality.

1. Introduction

Frontal crashes account for 10% of all fatalities and 21% of all MAIS3+ of belted occupants in the US (Eigen and Martin, 2005). It has also been shown that thoracic injuries are frequent and severe in this type of crashes, Cuerden et al. (2007) found that as many as 84% of all drivers killed in frontal crashes sustained AIS3+ thoracic injuries. The most common thoracic injuries are rib fractures as described by Carroll et al. (2010) and Eigen et al. (2007). Crandall et al. (2000) found that approximately 61% of all AIS2+ thoracic injuries were rib fractures and that the maximum thoracic AIS was defined by rib fractures for approximately 72% of the occupants sustaining a maximum thoracic AIS2+. Furthermore, Wanek and Mayberry (2004) found that the number of rib fractures is a good indicator of other thoracic and abdominal injuries. A review of the literature illustrated that age was the most important occupant characteristic that influences the injury risk (Mendoza-Vazquez, 2014). Tools that assess the risk of rib fractures and are sensitive to restraint design changes are important for the improvement of restraint systems in cars, which help to reduce the occurrence of rib fractures in frontal car crashes.

Anthropomorphic test devices (ATDs) and finite element human body models (FE-HBMs) are tools used to develop and evaluate restraint systems. Typically, ATDs for frontal crashes are instrumented to measure chest compression and spine acceleration, often referred to as global criteria, to study the thoracic injury risk. The maximum chest compression (Cmax) (Kroell et al., 1974), defined as the mid-sternal chest deflection divided by the original chest depth, is a thoracic injury criterion used in regulations and consumer tests to assess the risk of thoracic injuries in frontal crashes. The maximum viscous criterion (VCmax) (Lau and Viano, 1986) was developed in order to predict soft tissue injuries in the thorax by taking the product of the chest deflection and the deflection rate. The maximum chest deflection (Dmax) (Kleinberger et al., 1989) was proposed to capture localised deflections in the chest, as produced by a belt, by measuring the maximum deflection of five different points on the chest. The combined deflection criterion (DC) (Song et al., 2011) and differential deflection criterion (DcTHOR) (Davidsson et al., 2014) were proposed to account for asymmetry in the compression of the thorax. FE-HBMs represent humans in greater detail than ATDs, which allow FE-HBMs to measure these global criteria as well as criteria at the structural and material levels. Development of global criteria for ATDs is ongoing (Song et al., 2011; Trosseille et al., 2013; Davidsson et al., 2014) and FE-HBMs can provide valuable fundamental knowledge on the differences between and potential strength of these global criteria (Brolin et al., 2012).

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With criteria at the material level the FE-HBMs can potentially analyse injury mechanisms at a tissue level (Wismans et al., 2005) and predict injuries to specific organs and tissues. Shen et al. (2005) found that stress obtained from subject specific models best correlated with rib fractures induced by high speed impacts in animal tests. Song et al. (2011) and Forman et al. (2012) have assessed the rib fracture risk using strain-based criteria with FE-HBMs. For criteria at the structural level (Charpail et al., 2005) found that rib deflection at fracture was about 21% in tests for isolated ribs under anteroposterior loading.

Several FE-HBMs have been developed in recent years (Iwamoto et al., 2002; Vezin and Verriest, 2005; Vavalle et al., 2013). One frequently used model is the Total HUman Model for Safety (THUMS) (Iwamoto et al., 2002; Shigeta et al., 2009). The rib cortical bone material in THUMS, and other state of the art FE-HBMs, is modelled with an isotropic elasto-plastic material that eliminates elements that reach a certain strain level, indicating the occurrence of a rib fracture. This approach is deterministic in the sense that an exact number of fractures can be predicted given a single set of occupant characteristics, as in Kitagawa and Yasuki (2013). Deterministic approaches are limited for prediction of injury occurrence in a population with varying physical characteristics (Forman et al., 2012), who compared the greatest strain for every rib with a distribution of ultimate strain values obtained from tensile tests and calculated the risk of rib fractures. Hence, material criteria that are not coupled to element elimination can be used to create injury risk curves that capture variability in occupant characteristics. Furthermore, the use of an isotropic model to represent the anisotropic material of rib cortical bone (Viano, 1986), may limit the use of injury criteria at the material level. Therefore, it is not evident that material criteria are best suited to predict thoracic injuries with the FE-HBMs available today, hence structural and global level criteria should also be considered for FE-HBMs.

The objective of this study was to recommend a set of injury risk curves for an FE-HBM that predict rib fractures in frontal car crashes. To achieve this, injury risk curves were constructed by relating PMHS injury outcome with computed injury criteria at the global, structural and material levels extracted from a modified THUMS v3 in matched simulations. In addition, age was included as a covariate to construct age adjusted injury risk curves. Finally the injury risk curves for each criterion were compared using the Akaike Information Criterion (AIC) and quality index. The recommended injury risk curves have the potential to contribute to a reduction of thoracic injuries by increasing the use of FE-HBM during the development of restraint systems.

2. Method

The FE-HBM used in this study was a modified THUMS v3 (Mendoza-Vazquez et al., 2013), hereafter referred to as modified THUMS. This model represents the trabecular rib bone with hexahedral elements and the cortical bone with shell elements with an elastic modulus of 13 GPa, and a yield stress of 93.5 MPa. For the modified THUMS, the element elimination controlled by plastic strain was deactivated and a refined mesh in the intercostal muscles, bones and flesh of the ribcage was introduced as described by Mroz et al. (2010) and Pipkorn and Kent (2011). This finer mesh, illustrated in Fig. 1, resulted in a mean element length of 3.5 mm for the rib cortical bone elements, compared to 8.4 mm in THUMS v3. The solid elements representing the thoracic flesh in the modified THUMS have a bulk modulus of 1.33 MPa compared to the original 2.29 MPa. The strain–stress curve for the solid elements representing the thoracic internal organs in the modified THUMS was scaled down to provide a better correlation to experimental data (Mendoza-Vazquez et al., 2013). The biofidelity of the modified THUMS has been verified against impactor, table top and sled tests by Mendoza-Vazquez et al. (2013). The pre- and post-processor used were LS-PREPOST (v2.4, LSTC, Livermore, CA, USA), Primer (v10.0, Oasys Ltd., UK), respectively. The finite element solver was LS-DYNA (version 971 R4.2.1, LSTC, Livermore, CA, USA). Data was analysed using in-house code developed in MATLAB (R2007b, The Math Works Inc., Natick, MA, USA).

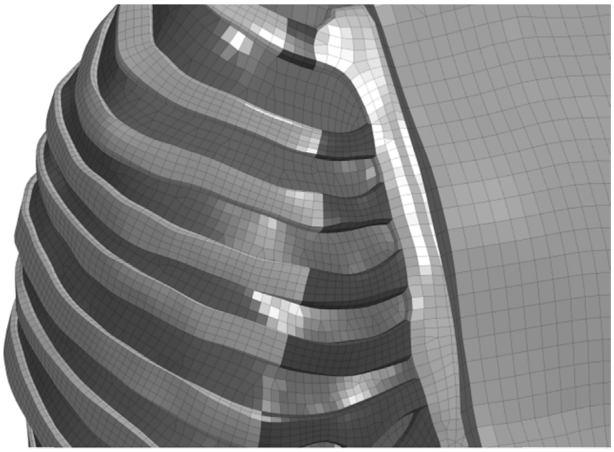

Fig. 1. Detail of the finer mesh, showing the ribs, intercostal muscles, rib cartilage, sternum and thoracic flesh.

A total of twenty-three PMHS tests with an average stature of 1.77 m, average weight of 70.2 kg and average age of 61 years at time of death were reproduced with the modified THUMS, as shown in Table 1. These tests include impactor, table top and sled tests illustrated in Fig. 2. The impactor tests were reported by Nahum et al. (1970), Kroell et al. (1974) and Bouquet et al. (1994); table top tests by Kent et al. (2004) and sled tests by Shaw et al. (2009). In all impactor tests, the PMHSs were impacted at the middle of the thorax at the height of the 4th intercostal space with a cylindrical pendulum of 152 mm in diameter. In the table top tests, PMHSs were laying freely on a rigid bench while loads were applied at a displacement rate of 1 m/s. The load was either of a hub, belt, double diagonal belt or band type. In the sled tests, PMHSs were seated on a rigid plate, restrained with a three point belt and a knee bolster and subjected to an acceleration pulse. Only tests with rib fracture data and performed on fresh PMHSs, weighing between 54 and 88 kg at a stature between 1.57 and 1.92 m were included. If a PMHS was subjected to multiple tests, as in the table top test case and MRS04 impactor test, only one test per PMHS was considered and only if rib fractures were not detected after that test.

A scaling law was applied to the mass of the impactor hitting the modified THUMS, without making any changes to the initial velocity or the modified THUMS. The scaling was based on the method proposed by Mertz (1984), who advised basing the scaling on the effective mass and chest depth, a brief description of this method is included in Appendix A. Injury criteria were extracted from simulations at the time when the chest compression in the modified THUMS reached the same level as the matched PMHS in the table top test. Sled tests were not scaled. The results for the scaling of the impactor tests are shown in Table 2.

The descriptions of the models used in the simulations are available in Mendoza-Vazquez et al. (2013). For each simulation, injury criteria were calculated according to the definitions in Appendix B. The global level criteria were maximum chest compression (Cmax) (Kroell et al., 1974), maximum viscous criterion (VCmax) (Lau and Viano, 1986), maximum deflection (Dmax) (Kleinberger et al., 1989), combined deflection (DC) (Song et al., 2011), combined deflection for THOR (DcTHOR) (Davidsson et al., 2014), and total internal energy in the rib cortical bones (TIE). The structural level