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Development of a computational framework to adjust the pre-impact spine posture of a whole-body model based on cadaver tests data

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ABSTRACT

A method was developed to adjust the posture of a human numerical model to match the pre-impact posture of a human subject. The method involves pulling cables to prescribe the position and orientation of the head, spine and pelvis during a simulation. Six postured models matching the pre-impact posture measured on subjects tested in previous studies were created from a human numerical model. Posture scalars were measured on pre- and after applying the method to evaluate its efficiency. The lateral leaning angle θ_L defined between T1 and the pelvis in the coronal plane was found to be significantly improved after application with an average difference of $0.1 \pm 0.1°$ with the PMHS ($4.6 \pm 2.7°$ before application). This method will be applied in further studies to analyze independently the contribution of pre-impact posture on impact response using human numerical models.

1. Introduction

Post-Mortem Human Subjects (PMHS) are the primary surrogates used in injury biomechanics to study injury mechanisms and develop injury risk functions that can be used for the design of restraint systems to improve road safety. In recent studies, chestbands, high-speed optical motion-capture systems, and digital and laser scanning devices have characterized the impact environment including thorax deflections, segmental kinematics, and boundary conditions (Lessley et al. 2010, Shaw et al. 2014). These methodological advances permit greater description of boundary conditions, allowing studies to focus on particular parameters in side impact as the actual subjects' posture.

In the published literature, the term 'posture' is used with vastly different meaning: while some researchers refer to posture as any variations between positions that can be obtained by a rigid body translation and rotation from a reference position (Park et al., 2013), others used it in the form 'out-of-position' to describe inadequate position that could be detrimental to the effectiveness of a restraint systems (Kemper et al., 2008), or to actually define the shape of human body. In the present study, the posture is defined as the positions and orientations of limbs and body regions relative to each other independently of the impact environment.

The question of 'posture', regardless of its definition, has gained considerable of interest as it can now be better controlled, and therefore can be an input for a PMHS test, or at least accurately measured. The computational models of the human body that are now available allow for a fine control of the initial posture of the body by using the 3D kinematics data collected during the experiment and input the actual posture rather than the nominal posture (Pipkorn et al., 2014).

It is common in impact tests to report large variations in injury outcomes between PMHS subjected to the same loading, and it is hypothesized that the pre-impact posture of the subjects played an important role in the reported variability by modifying the load path to the spine, shielding and protecting the ribcage (Lessley et al., 2010, Donlon et al., 2014). While some methods exist to modify the angle of the joints that connect long bones such as the knee and the shoulder, they do not apply to complex structures such as the spine as they require definition of discrete joints and their associated kinematics. Therefore, the objective of the present study was to develop and evaluate a method to reproduce the pre-impact posture of a PMHS with a human numerical model.

2. Methodology

Two recent studies on side impacts where three PMHS were impacted by a rigid wall (without side airbag in one case - Lessley et al., 2010, termed SideRigid; with side airbag in the other case - Shaw et al., 2014, termed SideAB) were used in the current study, as they provide a set of six PMHS tested based on the same 'posturing' protocol.

2.1. Experimental data

In Lessley et al. (2010), three approximately 50th percentile adult male PMHS were subjected to right side lateral impacts at 4.3 ± 0.1 m/s

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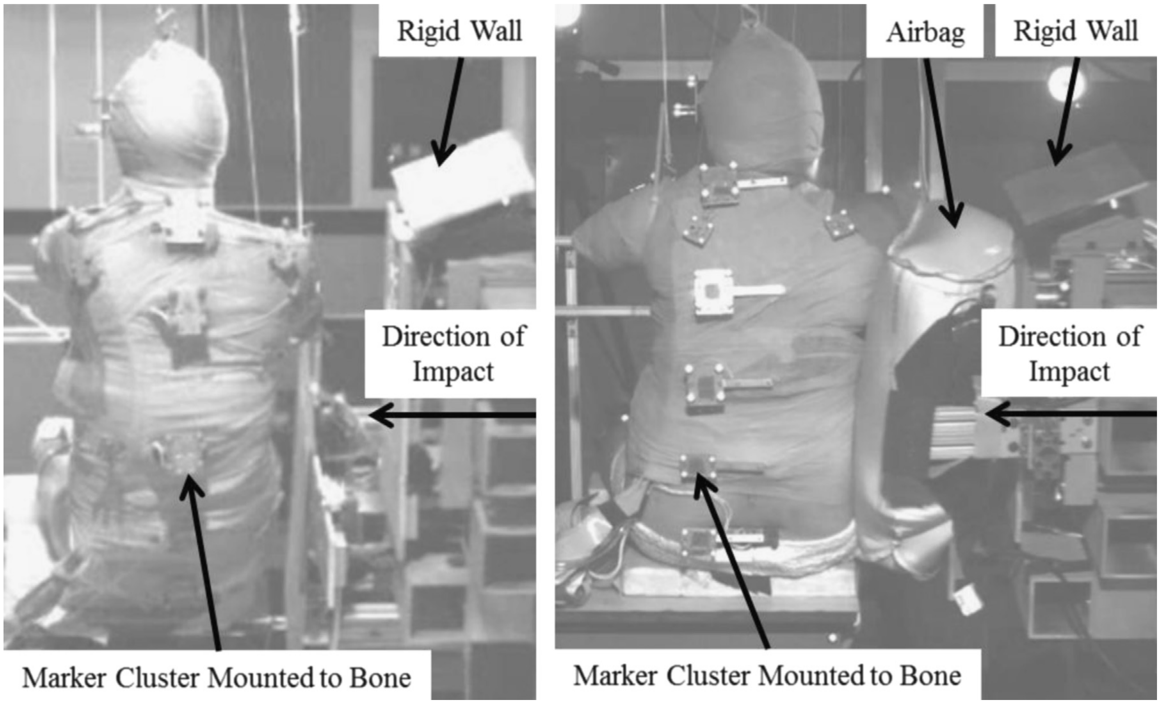

Fig. 1. Setup for experiments (Lessley et al. 2010, Shaw et al. 2014).

using a rigid wall mounted to a rail-mounted sled (Fig. 1). In Shaw et al. (2014), the same methodology was applied on three other PMHS but a side-impact airbag was deployed during the event (Fig. 1). In both studies an optoelectronic stereophotogrammetric system tracked the position of retro-reflective four marker clusters during the impact event (Fig. 1). Prior to each test, the clusters were surgically secured to selected anatomical locations on each subject including the head, the scapulae, the spine (T1, T6, T11, and L3) and the pelvis. The collected cluster data allowed the position and orientation of the corresponding underlying bone to be determined at each time step during the impact event using a rigid body motion analysis that provided 6DOF motion data for each of the selected bone segments. Anatomical coordinate systems (ACS) were defined for each bone based on a set of bony landmarks (Fig. 2). Bones for which position data was recorded are listed in Table 1.

The pre-impact posture in each test was defined as the position of the subject at time 0: the time of first contact between the wall or airbag and the subject. In the SideRigid impacts, the first contact occurred when the wall contacted the greater trochanter; while in the SideAB tests, the first contact occurred when the airbag contacted the subject.

Fig. 3 shows the initial positions of all the PMHS from the posterior view. While none of the PMHS was subjected to spinal pathology, significant variations in spinal posture were observed between PMHS due to gravity (Fig. 3). A global quantification of the spine curvature consisting of a lateral lean angle θL was defined as the angle between the vertical axis and a line connecting the midpoint of the posterior superior iliac spines to the center the T1 vertebra (Donlon et al., 2014). Three subjects were found to lean away from the wall (subject 1413, $\theta_L = 6.4°$; subject 1415, $\theta_L = 7.7°$; subject 1569, $\theta_L = 6.5°$), two subjects leaning closer to the wall (subject 1570, $\theta_L = -3.7°$; subject 1571, $\theta_L = -3.1°$) and only one has a straight spine (subject 1414, $\theta_L = 0.4°$).

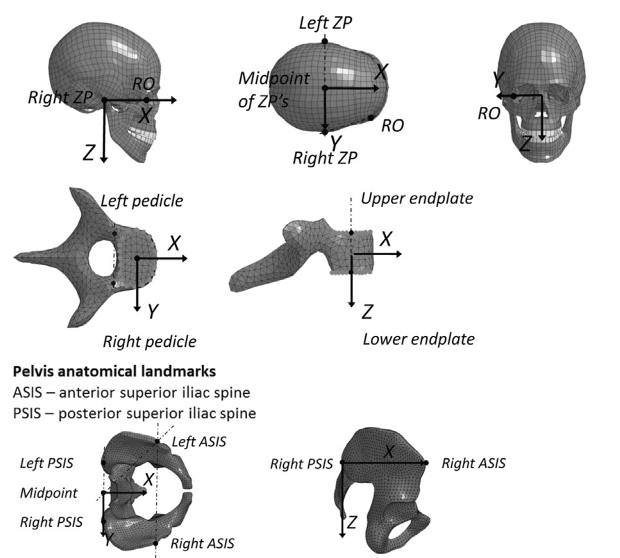

Fig. 2. Anatomical coordinate system definition showed using THUMS geometry.

coordinate systems defined for each bone in the experiments were defined on THUMS by locating the specific anatomical landmarks using the 3D geometry (Fig. 1). The solver used was LS-DYNA (mpp971sR6.1.1 Rev. 78769, SVN. 80485, LSTC, Livermore, CA, USA). Simulations were performed on a 48 nodes cluster (Dual Opteron 6238, 24 cores/node, 64 GB/node). The pre and post processing work was carried out with LS-PREPOST (v4.1, LSTC, Livermore, CA, USA) and scripts written in Matlab (R2012a, The MathWorks Inc., Natick, MA, USA).

2.2. Human numerical model

The Total Human Model for Safety (THUMS, version 4.0) was used for the computational work. The bilateral arms were cut through the proximal third of the humerus as in the experiments. The anatomical

2.3. Applying pre-impact posture

The positioning simulations utilized the pulling cables technique described below. It was applied to prescribe the position and orientation of the head, the T1, T6, T11, L3 vertebral bodies, and the pelvis.