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Normalized frontal impact biofidelity kinematic corridors using post mortem human surrogates

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A B S T R A C T

Due to reducing cost and powerful computing resources and the ability of finite element human body models (FEHBM) to predict human body response more realistically, they are gaining acceptance to be a substitute for mechanical surrogates. Unlike mechanical surrogates, FEHBM can realistically simulate human kinematics and kinetics. Moreover, an array of quantities can be directly measured from FEHBMs. However, similar to Anthropomorphic Test Devices (ATDs), in order to evaluate the biofidelity, these models must be validated using PMHS response corridors. Therefore, availability of such PMHS corridors that can be used to validate both ATD and FEHBM kinematics is of primary importance. The current study presents normalized biofidelity corridors of head CG, T1, T12, and sacrum accelerations using PMHS frontal sled tests that were previously conducted. In addition, rotational accelerations and displacements of the head are also presented. The experimental data were collected using four specimens. Each specimens were tested with non-injurious pulses using two different velocities (low: 3.6 m/s and medium: 6.9 m/s). These data were normalized using mass-based technique to represent mid-sized United States population. Using the normalized data, average and plus/minus one standard deviation response corridors were generated that can be used to evaluate the biofidelity of ATDs and FEHBMs.

1. Introduction

Preventing and mitigating injuries in automotive crashes require studies to improve understanding of occupant kinematics. Generally, these studies are performed under controlled laboratory conditions, using an array of testing devices, and test subjects. Commonly used testing devices are drop tower (Yoganandan et al., 2013a, 2013b, 2013c, 2013d, 2014a, 2014b), hydraulic test devices (Pintar et al., 2005), sled systems (Yoganandan et al., 2013a, 2013b, 2013c, 2013d), etc., whereas, animals (Kent et al., 2006), human volunteers (Patrick et al., 1965), isolated body regions (Yoganandan et al., 2013a, 2013b, 2013c, 2013d), whole body Post mortem human surrogates (PMHS) (Humm et al., 2012; Yoganandan et al., 2013a, 2013b, 2013c, 2013d), Anthropometric testing devices (ATD) (Yoganandan et al., 2011a, 2011b, 2011c), and mathematical models (Yang et al., 2006) are commonly used as human surrogates. Using animals in impact biomechanical testing is generally not preferred, but used to obtain certain data that are impossible to collect otherwise, such as pediatric response to severe impacts. Unlike human volunteers, isolated test specimens and PMHS can be exposed to loadings that help researchers to delineate various injury mechanisms, and obtain injury tolerances for different body regions. Even though PMHS lack in muscle activity, they are generally preferred in biomechanical testing due to their similarity in responses to living humans.

However, PMHS testing has several disadvantages, such as anthropometric variations in an ensemble of specimens (Yoganandan et al., 2014a, 2014b), inability to measure in-situ loads in certain body regions (like spine, head CG, femur force etc.) (Pintar et al., 2010), and ethical requirements. It is well established that injury tolerance, and mechanical response varies with anthropometry (Bose et al., 2011). When experiments are performed using PMHS, all specimens in the ensemble may not represent a specific population. In order to minimize variations in PMHS responses due to differences in anthropometries, several normalization techniques are proposed in literature (Eppinger, 1976; Mertz, 1984; Moorhouse, 2013). In addition, it is difficult to measure certain parameters – such as spine, head CG, and femur loads – without compromising the mechanical integrity of a PMHS. Frequently, in PMHS testing, these loads are derived from externally attached sensors, using assumptions based on principles of rigid bodies and joints; these assumptions add inaccuracies to derived loads (Pintar et al., 2010). Further, some countries restrict performing PMHS experiments due to ethical reasons. In order to circumvent these disadvantages, researchers have developed various ATDs and computational models to mimic human response.

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ATDs are mechanical surrogates that are designed to respond like live human beings that represent a specific population. The popular frontal impact ATDs are Hybrid III, and THOR (Yoganandan et al., 2011a, 2011b, 2011c). For other modes of impacts – such as rear, and side impacts – different ATD's are used (Yoganandan et al., 2011a, 2011b, 2011c; Humm et al., 2012). In recent years as the data collection and instrumentation technologies improved, several sensors are incorporated in ATDs to measure different quantities. Nevertheless, these quantities must be validated with normalized PMHS or human volunteer experiments, performed under similar input conditions. Traditionally, ATD responses are validated using biofidelity corridors obtained from PMHS tests (de Lange et al., 2005; Ridella and Parent, 2011; Parent et al., 2013). ATD responses are typically deemed acceptable if the responses lie between the corridors. One of the major disadvantages of using ATDs is the high stiffness of these devices that result in poor kinematics prediction – especially involving flexible body region like spine (Demetropoulos et al., 1998, 1999). In addition, high device and instrumentation costs, limitations in measuring capabilities – for example, ATDs cannot measure stresses and strains – prompted researchers to find an alternate surrogate that is more biofidelic, and more accessible to researchers. One such surrogate that has recently received wide attention is finite element human body models (FEHBM) (Kitagawa et al., 2006, Gayzik et al., 2011, Gayzik et al., 2012, Schinkel-Ivy et al., 2014). These models use finite element principles to mimic human body responses under impact. Due to reducing cost and powerful computing technologies, and the ability of FEHBM to predict human body response more realistically, they are considered to be a substitute for mechanical surrogates. Unlike these surrogates, FE models can realistically simulate human kinematics and kinetics. Moreover, an array of quantities can be measured from FEHBMs. However, similar to ATDs, in order to evaluate the biofidelity, these models must be validated using PMHS response corridors. Therefore, availability of such PMHS corridors that can be used to validate both ATD and FEHBM kinematics is of primary importance.

The objective of the current study is to derive normalized biofidelity corridors of head CG, T1, T12, and sacrum accelerations, using PMHS tests under frontal impact. The raw acceleration data were taken from previously performed experiments. Head CG accelerations were measured using PNAP data collection system, and accelerations from other regions were collected using tri-axial accelerometer packages.

2. Methods

All the raw acceleration data were obtained from a previous study performed by (Pintar et al., 2010). For brevity, a brief description of the experimental setup is provided and a detailed description can be found in the original study. Four unembalmed (age: 60 ± 23 years, weight: 59 ± 21 kg, height: 170 ± 7 cm) prescreened PMHS were included for testing. The restrained specimens were tested on a custom rigid seat and input pulses were applied using a hydraulically driven or a bungee cord driven sled system (Table 1).

The setup included a three-point belt system, a knee bolster, and a head restraint. The head restraint was adjustable fore-aft, and up-down

Table 1
Details of the sled tests.

Test#	Speed	Sled Type
FC133	Low	Bungee
FC134	Medium	Bungee
FC197	Low	Bungee
FC198	Medium	Bungee
FC220	Low	Hydraulic
FC221	Medium	Hydraulic
FC237	Low	Hydraulic
FC238	Medium	Hydraulic

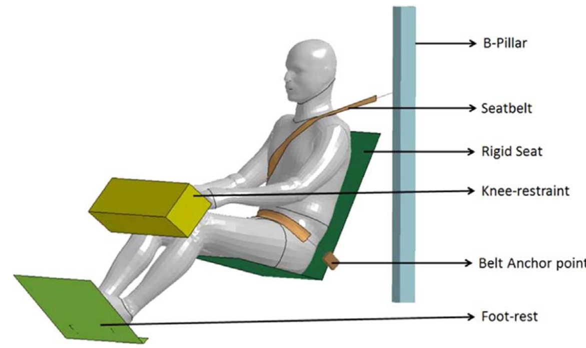

Fig. 1. Pre-test PMHS sled setup.

with respect to the seat. The knee bolster assembly – intended to mimic instrument panel - included a honeycomb padding and supporting fixtures (Fig. 1). A three point seatbelt system was used to restrain the specimens. The seatbelt material property corresponded to 5% elongation at 11 kN. The D-ring was fixed in the fore-aft direction and was adjusted up and down to the specimen anthropometry such that it was level with the auditory meatus. The lower anchor positions were typical of a mid-size sedan in the US vehicle fleet. Belt positioning followed FMVSS-208 specifications. To apply a generic belt pre-tensioner scenario, the belt was pulled 10 cm at the D-ring after the FMVSS-208 belt positioning procedure. The entire preparation was mounted on a sled and tested at two different speeds (low: 3.3 m/s and medium: 6.7 m/s), using two input pulses (Fig. 2). A pyramid-shaped nine accelerometer package (PNAP) was mounted on the head to collect accelerations at the center of gravity (CG). In addition, custom mounts were used to attach tri-axial accelerometers at T1, T12, and sacrum.

All acceleration signals were filtered, normalized, and average plus/minus one standard deviation corridors were generated. Filtering was performed using SAE (Society of Automotive Engineers) class-60 filter to remove high frequency vibrations from the collected signals. Anthropometric variations are unavoidable in biomechanical experiments. It is well documented that these variations influence experimental responses. Hence to minimize these variations, individual responses were normalized to a predefined reference population of a 50th percentile American male. The normalization procedure performed in this study used a basic approach proposed by Eppinger (1976) that assumes linear relation between length, mass, and time units. Furthermore, the procedure assumes identical tissue density and elastic modulus between individual PMHS and reference PMHS. The following equations were used to normalize the collected acceleration data to a 50th percentile American male for the acceleration (a) and time (t) data.

$$a_n = \lambda^{-\frac{1}{3}} \times a_i \tag{1}$$

$$t_n = \lambda^{\frac{1}{3}} \times t_i \tag{2}$$

Where, the subscripts n and i denote normalized and individual signals. Lambda is the normalizing factor, which is given by the ratio between the reference mass (76 kg) and individual PMHS mass. Following the normalization procedure, average and plus/minus one standard deviation biofidelity corridors are generated.

3. Results

Component wise biofidelity corridors for head CG accelerations are given in Figs. 3 and 4. The average peak x-direction accelerations were approximately −4 and −13 g for the low and medium speeds. Whereas, in the y-direction the peak was approximately −1 and −4 g; and in the z-direction the peak was 3.5 and 17, for low and medium speeds.

Component wise biofidelity corridors for T1 accelerations are given