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Development of a computational framework to adjust the pre-impact 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. The method was found to be effective since the average error, from the head to pelvis, between the posture of the model and the PMHS posture was 7.67 ± 1.49 mm for position and 4.7 ± 0.8 ° significantly lower ($p<0.001$) than it was prior to the method (40.0 ± 12.0 mm, 8.5 ± 1.0 °). This method will be applied in further studies to analyze independently the contribution of pre-impact posture on impact response using human numerical models

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. The recent development of techniques and tools to measure the 3D kinematics of PMHS during impact tests opened ways to better describe the boundary conditions specific to an impact, such 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 a lot of interest as it can now be better controlled, and therefore can be an input for a PMHS test, or at least accurately measured. Simultaneously, 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 like 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.

METHODS

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; with side airbag in the other case - Shaw et al., 2014) were used in the current study, as they provide a set of six PMHS tested based on the same ‘posturing’ protocol.

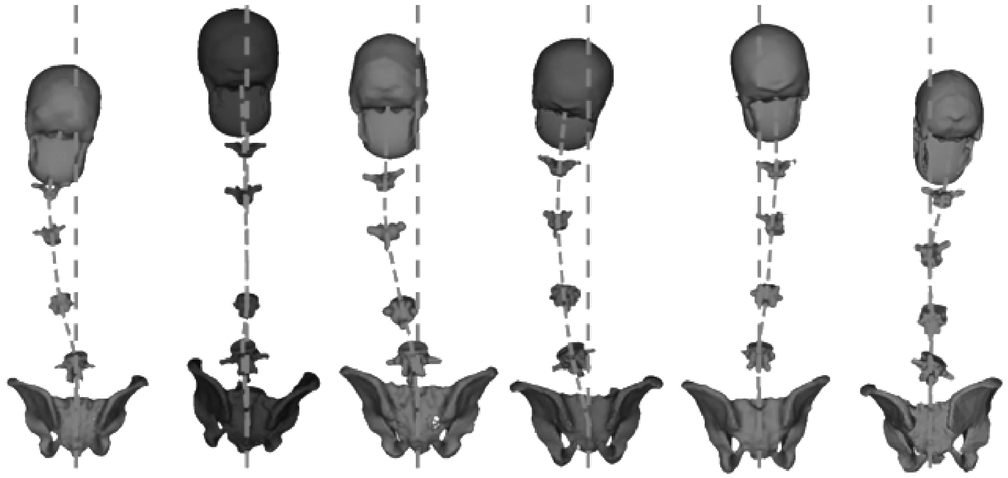
Experimental data

In Lessley *et al* (2010) and Shaw *et al* (2014), the three-dimensional (3D) kinematics data (3 translations and 3 rotations) of certain bones on the PMHS were recorded for every millisecond of the tests according to the method reported by Lessley *et al* (2010). Anatomical coordinate systems (ACS) were defined for each bone based on a set of bony landmarks (Figure 2). Bones for which position data was recorded are listed in Table 1.

Table 1: Vicon-tracked bones for each test

	Rigid wall, no airbag (SideRigid)			**Rigid wall, with airbag (SideAB)**		
Bone target	Test 1413	Test 1414	Test 1415	Test 1569	Test 1570	Test 1571
Skull (Head)	Yes	Yes	Yes	Yes	Yes	Yes
T1	Yes	Yes	Yes	Yes	Yes	Yes
T6	Yes	T5	Yes	Yes	Yes	Yes
T11	Yes	T12	Yes	Yes	Yes	Yes
L3	Yes	Yes	Yes	Yes	Yes	Yes
Pelvis	Yes	Yes	Yes	Yes	Yes	Yes

The pre-impact posture in each test was the position of the subject at time 0, defined as 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. Figure 1 shows the initial positions of all the PMHS from the posterior view.



Test 1413:
16 Rib Fractures

Test 1414:
Uninjured

Test 1415:
Uninjured

Test 1569:
Uninjured

Test 1570:
2 Rib Fractures

Test 1571:
9 Rib Fractures

Figure 1: Posture view of the pre-impact (t = 0 ms) spine posture in the rigid-wall and airbag tests with corresponding injury outcomes obtained in Lessley et al 2010 and Shaw et al., 2014 (Donlon *et al*, 2014).

Human numerical model

The Total Human Model for Safety (THUMS, version 4.01) was used for the computational work. The bilateral arms were cut through the proximal third of the humerus as in the experiments. The anatomical coordinate systems defined for each bone in the experiments were defined on THUMS by locating the specific anatomical landmarks using the 3D geometry (Figure 2). 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).

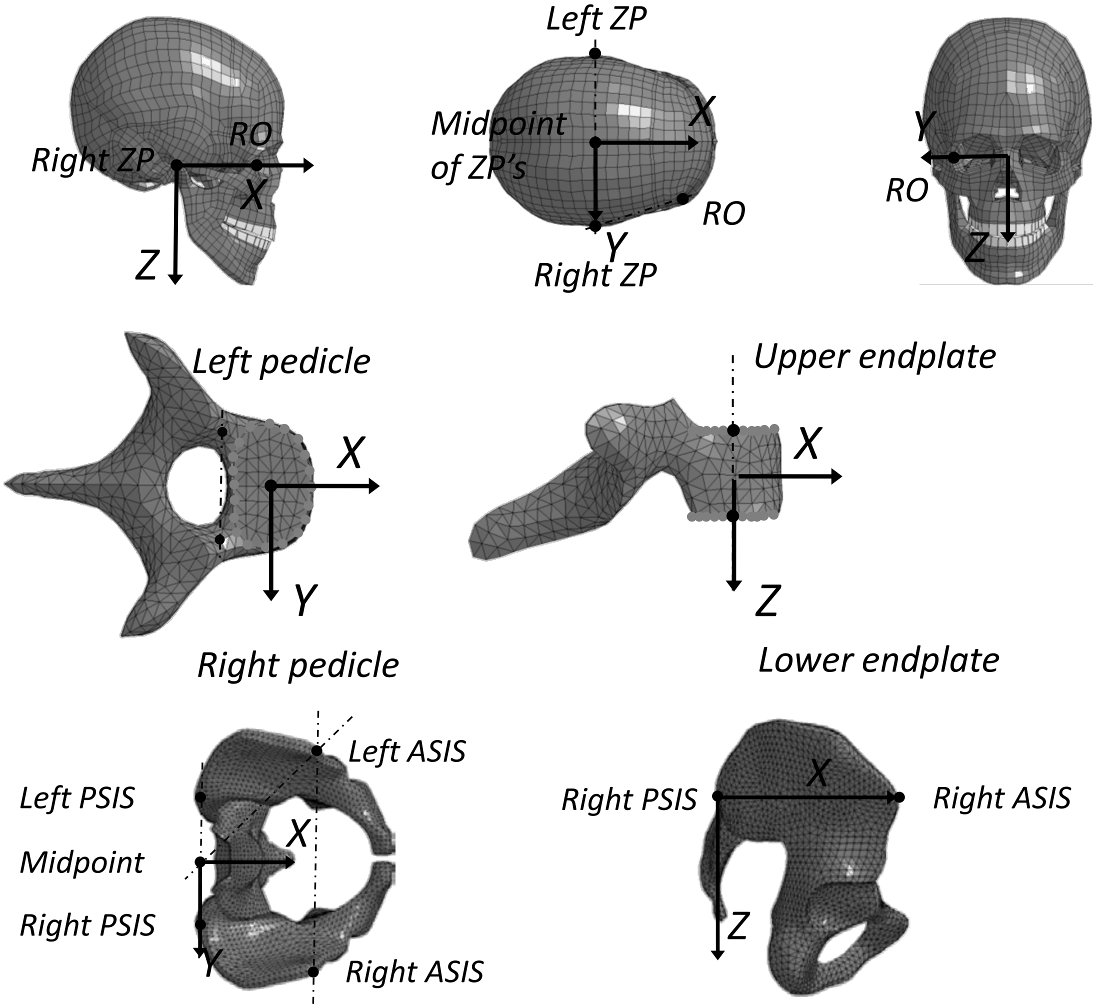


Figure 2: Anatomical coordinate system definition showed using THUMS geometry.

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.

The purpose of this technique is to move smoothly a part from a start location to a target location while maintaining the physiological constraints on the joints. This technique consists of a cable (*SEATBELT in Ls-Dyna), two rigid plates in aluminum (30x30x1 mm), a spring and a slipring (Figure 3). One plate is linked by the spring to the part to be moved (plate 1) and the other plate is fixed to the target location (plate 2). The two plates have the same orientation. One extremity of the cable is fixed to plate 1, and the other extremity is driven by a prescribed acceleration. A slipring anchored to plate 1 prevents plate 2 from moving past plate 1.

During the simulation, the cable pulls the part up to the target location during the first 150 ms. Next, the model is let stabilize during 150 ms. When the model stopped oscillating, the nodes coordinates of the model (excluding the posture apparatus) are exported to be used as an input for a postured model (Figure 3). By combining three pulling systems, both the final position and the orientation of a body segment can be controlled (Figure 3). For each bony segment, three targets define its position and orientation : three for the local vectors extremities of the local coordinate system (Figure 3). The positions of these landmarks are the targets used to position THUMS using the pulling cable technique.

However, if the positions of the all the PMHS landmarks were used directly as targets for the corresponding landmarks on THUMS, the spine would extend unrealistically because of the differences between THUMS and the PMHS. To overcome this issue, the PMHS posture was normalized in order to deal with the difference in spine length and anatomy (Figure 4). First, the PMHS spine was normalized in the x, y and z directions to match that of THUMS, and the curvilinear abscissa of each anatomical landmark along the spine was calculated for both the PMHS and THUMS. Second, landmarks that correspond to the curvilinear abscissae of the THUMS landmarks were created on the PMHS spine and used as targets when modifying THUMS posture. The orientation of the body segments was applied as is (e.g. the orientation of T1 measured for the PMHS was used for THUMS T1).

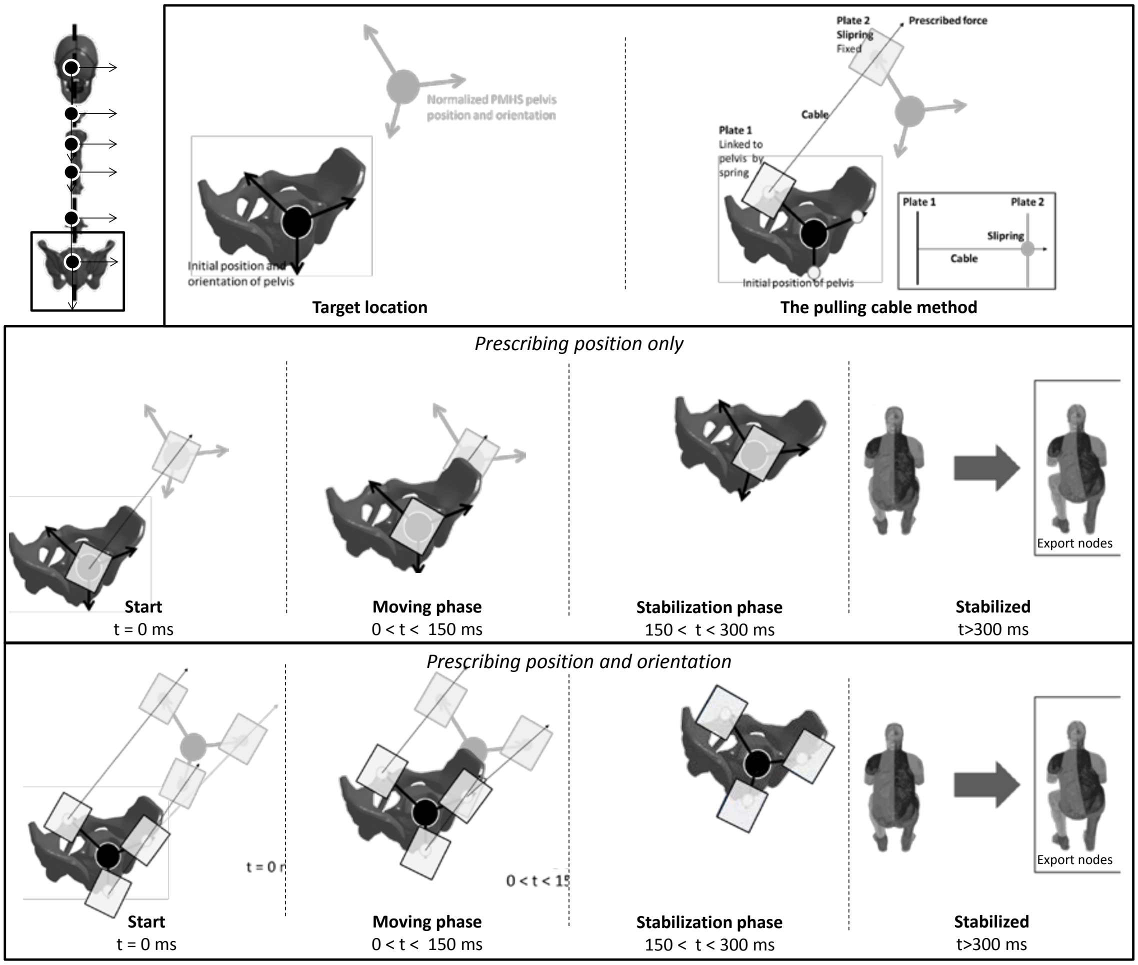


Figure 3: Pulling cable time sequence (pelvis case).

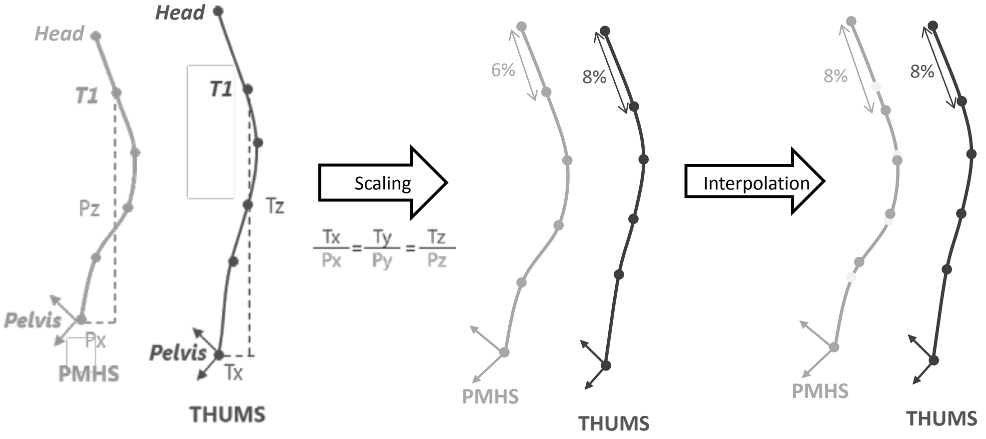


Figure 4: Determination of virtual landmarks on the PMHS spine based on the landmark position in the THUMS spine.

Three methods were evaluated to modify THUMS posture to match that measured in the PMHS tests (Table 2):

- M1: the position and orientation of the pelvis and the position of T1 were constrained, while the head, T6 and T11 were free to translate and rotate.
- M2: the position of all segments was constrained, but the orientation was left unconstrained,
- M3: the position and orientation was constrained for each segment
- In all cases, the arms were constrained (but not the scapula) and the legs were free to move without constraints. Bone deformations were disabled by making the bones rigid.

The three methods were tested with each of the six subjects. A total of eighteen simulations were run. The postured THUMS obtained as a result of these simulations are referred to as THUMS_[method]_[subject id], where [method] is M1, M2 or M3, and [subject id] indicates the target posture (1413, 1414, 1415, 1569, 1570, and 1571). The initial THUMS model (with no modification of the posture) is referred to as THUMS_Initial.

Table 2: Summary of the constraints for the three methods evaluated. 'Constrained' means that the measurement from the PMHS (position or orientation) was used as a target.

	Head	**T1**	**T6**	**T11**	**L3**	**Pelvis**
M1	Free	Constrained	Free	Free	Free	Constrained
	Constrained	Free	Free	Free	Free	Constrained
M2	Constrained	Constrained	Constrained	Constrained	Constrained	Constrained
	Free	Free	Free	Free	Free	Free
M3	Constrained	Constrained	Constrained	Constrained	Constrained	Constrained
	Constrained	Constrained	Constrained	Constrained	Constrained	Constrained

Evaluation of the differences in posture

The posture of the model and the exact pre-impact posture of the subjects were compared first qualitatively by visually comparing the postured THUMS to the corresponding PMHS, using the 3D kinematic models developed by Donlon et al. (2014). For each test, a three-dimensional model of the PMHS skeleton was constructed in OpenSim 3.0 (Delp *et al*, 2007) to represent the experiments: OpenSim is an open-source software that can combine STL files for the bone geometries and the three-dimensional time-history positions of reference points on these geometries to create three dimensional animated models of the experiment (Figure 5).

Second, a quantitative assessment was performed using two independent scalars:

- Position: the distance between the origin of the bony segment (head, T1, T6, T11, L3, pelvis) in THUMS posture and PMHS normalized posture.
- Orientation: the difference in angle between the bony segment orientation between THUMS posture and PMHS normalized posture. The orientation is decomposed in three rotations according to the global coordinate system (extrinsic rotations).

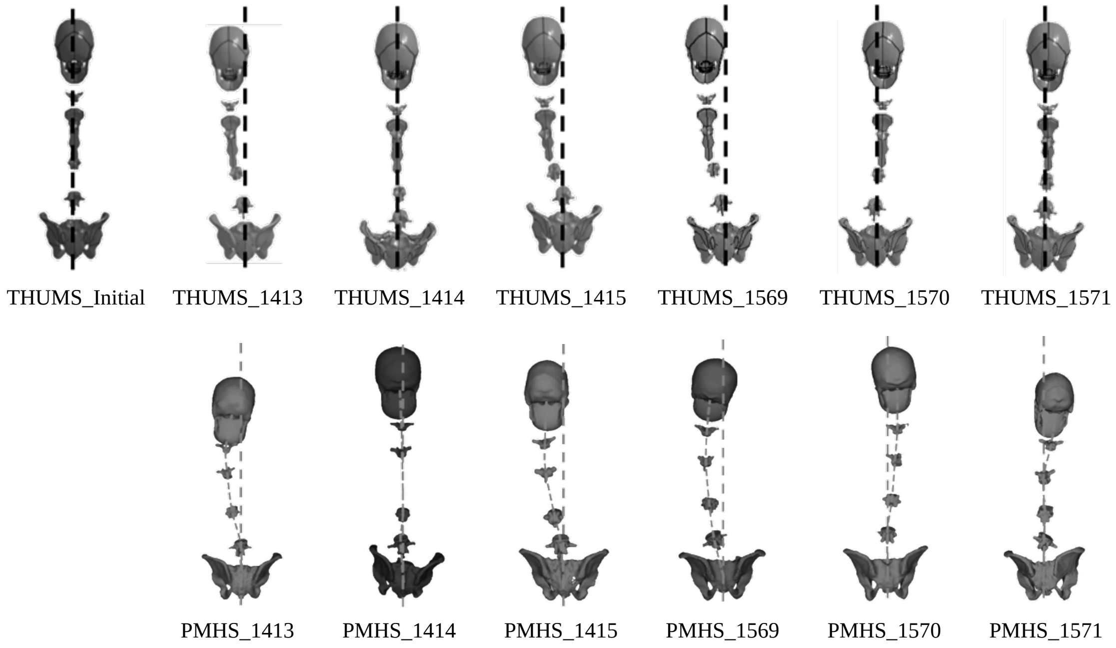
These scalars were computed before and after applying the posturing methods to evaluate their performances and paired t-tests were used to determine if the improvements were significant ($p<0.05$).

RESULTS and DISCUSSION

From the eighteen postured models created using the three methods (M1, M2, M3), only the models created using M1 (precise match of the position and orientation of the pelvis, and of the position of T1) were found to be stable. These models are shown in Figure 5.

Quantification of differences in posture between THUMS and subjects is shown in Table 3 for each body segment. The average error, all body segments combined, between the posture of the model after applying the method and the PMHS pre-impact posture was 7.67 ± 1.49 mm for position and 4.7 ± 0.8 °, significantly lower ($p<0.001$) than it was prior to the application of the method (40.0 ± 12.0 mm, 8.5 ± 1.0 °).

In all cases, the position error of the pelvis was below 7 mm after application of the method. Similar results were obtained for the orientation except for one case (1414), where the pelvis was tilted 11 degrees off.


THUMS_Initial THUMS_1413 THUMS_1414 THUMS_1415 THUMS_1569 THUMS_1570 THUMS_1571

PMHS_1413 PMHS_1414 PMHS_1415 PMHS_1569 PMHS_1570 PMHS_1571

Figure 5: Qualitative comparison of pre-impact posture between PMHS and THUMS

Table 3: Quantification of difference in posture

	PMHS 1413 vs.							
	THUMS_Initial				**THUMS_1413_M1**			
	Position	**Orientation (angle)**			**Position**	**Orientation (angle)**		
	Distance	**X (°)**	**Y (°)**	**Z (°)**	**Distance**	**X (°)**	**Y (°)**	**Z (°)**
Head	26.8	-2.6	-4.3	-7.6	29.1	-3.1	-3.5	-8.5
T1	0.0	-6.6	11.2	-6.8	0.0	-4.1	8.2	-6.2
T6	7.9	-7.8	5.0	13.4	6.6	-2.5	3.6	6.2
T11	21.7	-9.4	-19.0	14.1	5.3	-2.8	-5.1	2.8
L3	56.4	-12.6	-12.6	6.8	5.7	-9.9	8.3	0.6
Pelvis	78.2	1.7	-13.0	2.6	2.0	0.6	-2.6	-0.3
Average	*31.8*	*6.8*	*10.9*	*8.5*	*8.1*	*3.8*	*5.2*	*4.1*
		8.7				4.4		
	PMHS 1414 vs.							
	THUMS_Initial				**THUMS_1414_M1**			
	Position	**Orientation (angle)**			**Position**	**Orientation (angle)**		
	Distance	**X (°)**	**Y (°)**	**Z (°)**	**Distance**	**X (°)**	**Y (°)**	**Z (°)**
Head	17.4	0.3	-24.4	2.1	18.9	1.0	-11.7	2.0
T1	0.0	-1.8	-8.2	-0.9	0.0	-1.8	-3.3	-0.6
T5	25.0	-5.6	1.6	7.0	6.4	-5.6	0.5	4.6
T12	42.9	-1.4	-21.5	-1.8	8.7	-2.6	-4.7	-2.6
L3	27.1	-2.6	-32.3	11.3	12.1	2.2	-8.9	12.7
Pelvis	107.7	1.7	-25.2	-5.7	3.8	-1.6	-11.1	-4.3
Average	*36.7*	*2.3*	*18.9*	*4.8*	*8.3*	*2.5*	*6.7*	*4.5*
		8.6				4.6		
	PMHS 1415 vs.							
	THUMS_Initial				**THUMS_1415_M1**			
	Position	**Orientation (angle)**			**Position**	**Orientation (angle)**		
	Distance	**X (°)**	**Y (°)**	**Z (°)**	**Distance**	**X (°)**	**Y (°)**	**Z (°)**
Head	9.9	-4.0	-11.2	-0.6	7.0	-5.5	-7.0	-1.4
T1	0.0	-2.2	1.0	6.7	0.0	-0.3	0.4	6.2
T6	23.9	-4.4	-10.2	6.4	6.1	1.9	-1.7	-3.8
T11	46.7	-5.4	-32.2	4.7	4.3	0.7	-8.7	-10.8
L3	86.2	-8.3	-14.5	7.5	9.5	-6.1	9.2	1.6
Pelvis	128.2	-2.2	-19.4	2.6	3.1	-1.9	-3.6	1.0
Average	*49.1*	*4.4*	*14.8*	*4.8*	*5.0*	*2.7*	*5.1*	*4.1*
		8.0				4.0		
	PMHS 1569 vs.							
	THUMS_Initial				**THUMS_1569_M1**			
	Position	**Orientation (angle)**			**Position**	**Orientation (angle)**		
	Distance	**X (°)**	**Y (°)**	**Z (°)**	**Distance**	**X (°)**	**Y (°)**	**Z (°)**
Head	18.4	0.2	-37.1	-9.9	26.3	-2.8	-13.7	-8.9
T1	0.0	0.7	-8.4	-11.0	0.0	0.8	-2.4	-3.8
T6	15.5	-2.7	-0.3	2.3	5.4	1.1	1.1	-5.4
T11	22.5	4.7	-8.4	-3.9	11.5	5.7	-1.7	-5.3
L3	38.4	12.1	-12.2	22.1	10.3	9.2	1.6	8.7
Pelvis	73.9	4.0	-5.7	1.3	3.6	0.8	-3.7	0.9
Average	*28.1*	*4.1*	*12.0*	*8.4*	*9.5*	*3.4*	*4.0*	*5.5*
		8.2				4.3		
	PMHS 1570 vs.							
	THUMS_Initial				**THUMS_1570_M1**			
	Position	**Orientation (angle)**			**Position**	**Orientation (angle)**		
	Distance	**X (°)**	**Y (°)**	**Z (°)**	**Distance**	**X (°)**	**Y (°)**	**Z (°)**
Head	21.9	3.8	-19.5	7.6	15.6	5.4	-16.8	7.5
T1	0.0	1.6	-13.9	4.9	0.0	1.2	-13.2	4.4
T6	20.9	0.3	12.8	10.7	8.0	-0.6	11.8	11.0
T11	30.5	-4.4	-26.2	-1.9	8.6	-5.3	-16.7	1.7
L3	60.2	1.2	-6.3	-6.1	4.5	1.4	4.4	-2.3
Pelvis	74.3	4.8	-2.2	1.7	7.0	2.5	1.5	-1.2
Average	*34.6*	*2.7*	*13.5*	*5.5*	*7.3*	*2.7*	*10.7*	*4.7*
		7.2				6.0		
	PMHS 1571 vs.							
	THUMS_Initial				**THUMS_1571_M1**			
	Position	**Orientation (angle)**			**Position**	**Orientation (angle)**		
	Distance	**X (°)**	**Y (°)**	**Z (°)**	**Distance**	**X (°)**	**Y (°)**	**Z (°)**
Head	27.0	15.4	-17.0	-3.1	17.8	14.8	-12.1	-2.2
T1	0.0	-0.7	-5.5	-12.0	0.0	-0.1	-1.2	-4.7
T6	42.8	8.2	14.0	14.4	6.9	10.9	11.9	6.7
T11	53.1	-7.7	-36.9	2.0	7.1	-4.6	-6.0	3.3
L3	110.3	4.4	-11.0	-25.6	9.2	-1.9	5.8	-6.3
Pelvis	126.4	-1.0	-0.6	2.6	5.8	0.5	0.1	1.8
Average	*59.9*	*6.2*	*14.2*	*10.0*	*7.8*	*5.5*	*6.2*	*4.1*
		10.1				5.3		

CONCLUSIONS

The computational framework presented in this paper was successfully used to alter the posture of a whole body computational model: by running a simulation where the kinematic of the T1 vertebra and of the pelvis were prescribed, the overall curvature of the spine was altered to better match that of the PMHS. While the modified spine posture does not match exactly that of the PMHS because of differences in morphology and physiological degrees of freedom, the changes in spine posture were sufficiently significant to create different postured THUMS model based on the PMHS. This paper provides a tool that will be used to further evaluate how the differences on posture alter the impact response of a PMHS and the injuries they sustained.

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