Novel experimental techniques using PMHS and CT scans to assess performance of wearable sensors for mild Traumatic Brain Injury applications for different populations

John Humm, Klaus Driesslein, Alok Shah, Mark Begonia, Brian Stemper,
Frank Pintar, Narayan Yoganandan

Abstract Increased interest in the short and long-term health risks of mild traumatic brain injury (mTBI) has led to the development of wearable sensors designed to monitor real-time head kinematics. To be effective, sensors should be accurate, unobtrusive, low-maintenance, user-friendly, and scalable to a large population. They should also be evaluated with appropriate Post-Mortem Human Subject (PMHS) studies, as real-world impact tests are necessary. Therefore, the objectives of the present study were to develop a methodology to instrument the PMHS head with sensors mounted within the cranium and close to the cranium's center-of-gravity (cg) as determined by CT images.

The cg kinematics from the CT-placed internal sensor matched well with the computed cg kinematics of the externally placed sensors standardly used in impact biomechanics tests, though these tests did not include Personal Protective Equipment (PPE). To demonstrate the feasibility of the methodology and data acquired from the internal sensors, controlled tests were conducted on PMHS-helmet systems. This process can be effectively used to conduct efficacy evaluations with different types of helmets and other head-mounted wearable sensors, something urgently needed in the sports and military communities. Additionally, this methodology may be useful in the design of experiments used for validating human body models in highly automated vehicle environments.

Keywords Head injury, wearable sensors, CT imaging, mTBI

I. INTRODUCTION

The Center for Disease Control (CDC) estimated 2.5 million traumatic brain injury (TBI) emergency department (ED) visits in 2013 [1]. During the same year, the Department of Defense (DoD) reported over 27,000 TBIs in soldiers from the four branches of the armed services [2]. Increased interest within the medical community to the short and long-term risks of TBI has led to development of wearable sensor technologies to quantify exposure. To be effective, these devices should be unobtrusive to the user, require little maintenance, have easy data management, be scalable to a large population, and accurately measure head metrics. While Anthropomorphic Test Device (ATD) experiments are a good preliminary assessment of sensor accuracy, they fail to account for important biofidelity characteristics such as skin, fat, mouth, and teeth. Post-Mortem Human Subjects (PMHS) provide more realistic test conditions for the evaluation and comparison of wearable head kinematic sensors. To accurately assess the efficacy of these sensors, it is necessary to have a reference sensor that can measure accelerations in proximity to the PMHS head center-of-gravity (cg). Because traditional methods of acquiring these accelerations cannot be used in helmeted PMHS tests, an alternative method must be used to position the sensor near the head cg. The objectives of the present study are, therefore, to evaluate the performance of wearable sensors designed to measure head kinematics using a unique/feasible instrumented PMHS model.

II. METHODS

Head cg linear accelerations and angular velocities in PMHS tests are typically determined from instrumentation mounted to the skull [3] or palate [4]; however, these methods are precluded, as one of the design criteria of the study is that such instrumentation should not be placed in a location that may interfere with wearable sensors,

J. Humm (tel: +1 414 3842000 x43512; e-mail: jhumm@mcw.edu), K. Driesslein, and A. Shah are Research Engineers, M. Begonia is a Postdoctoral Fellow, and N. Yoganandan is a Professor in the Department of Neurosurgery at the Medical College of Wisconsin. B. Stemper is an Associate Professor and F. A. Pintar is a Professor in the MCW-Marquette University Department of Biomedical Engineering.

including sites such as such as the mouth (mouth guard sensors) [5], ears (earpiece sensors) [6], and external cranium (headband and helmet mounted sensors) [7]. Therefore, the instrumentation for calculating head linear cg accelerations and head angular velocities must necessarily be placed at a location that does not interfere with wearable sensors, preferably close to the head center of gravity. The current study uses a novel technique whereby a six-degree-of-freedom package (6DOF), (DTS, Seal Beach, CA) consisting of three linear accelerometers and three angular-rate sensors is inserted into the cranium cavity in proximity to the head cg. This method allows for measurement of the PMHS head kinematics using the 6DOF reference sensor with little post-processing.

Two series of experiments were performed for this study. The first used a whole-body non-helmeted PMHS to test the feasibility and accuracy of an internally mounted sensor. The second used an isolated head-neck helmeted PMHS with an internal sensor.

Specimen Preparation

Whole Body Tests

An unembalmed whole-body PMHS was procured and screened against hepatitis B, hepatitis C, and HIV. The approximate cg location of the head, estimated as 12 mm anterior to the auditory meatus, was marked bilaterally [8]. The specimen was thawed to the point where the neck was semi-mobile, but the brain remained mostly frozen and solid. Next, a 38 mm hole was drilled along the medial-lateral (M/L) axis of the head at the previously marked fiducials. A 38 mm O.D cylinder aluminum pipe was inserted through the head and secured bilaterally to the skull using an aluminum plate and bone screws. A six-DOF sensor was mounted in the center of pipe at the mid-point of its length. A tetrahedron-style nine accelerometer package with three angular rate sensors (NAP+3ARS) was mounted externally to the skull. The internal sensor was aligned to the Frankfurt plane using measurements of the sensor and anatomic fiducials. The external sensor was similarly aligned, and the head accelerations at the location of the internal sensor were computed using equations of rigid-body motion.

Isolated Tests

Using the results from the whole-body experiments, the instrumentation and techniques were refined. An unembalmed PMHS, isolated from the head to T2, was procured and screened against hepatitis B, hepatitis C, and HIV. Pretest computed tomography (CT) scans and radiographs were taken to rule out pre-existing trauma, assess specimen quality, and identify anatomic landmarks. Prior to instrumentation, an alignment fixture was attached to the superior cranium. The fixture consisted of an aluminum cylinder, terminated on one end by a 51 mm sphere and welded on the other end perpendicular to a flat plate. The underside of the plate was fixed to the skull via an aluminum stand-off block. Pre-instrumentation CT scans of the isolated specimen with fixture were obtained. Three-dimensional reconstruction of the scan was used to measure the orientation of the Frankfurt plane and M/L axis of the head relative to the alignment fixture. The spherical end of the alignment fixture was clamped to a rigid platform mount at the base of a single-axis drill. The platform mount had a socket-type joint that allowed rotation of the sphere of the rotation fixture to align the M/L axis of the head with the axis of the drill. A small pilot hole was made on the left side of the head at the approximated location of the projected head cg, and the tissue surrounding this point was removed. The M/L axis of the head was aligned with the drill axis using measurements calculated from CT scans. The drill was then used to bore a 35 mm diameter hole through the head from the left to right side. Brain tissue was evacuated, and a 35 mm outer diameter plastic sleeve was inserted along the mediolateral axis through the head and sealed at the edges using putty. Another hole, approximately 19 mm in diameter, was made at the external occipital protuberance; the specimen was inverted, and the cranium was filled with ballistic gel. The gel was then allowed to set. Next, a custom-made 25 mm diameter x 136 mm aluminum plug, which had a machined recess at the center for mounting the 6 DOF sensor, was inserted through the sleeve, and the ends were secured to the head to prevent rotation of the plug. A schematic of the instrumentation and picture of the instrumented plug are shown in Figure 1. The isolated specimen mass (head with brain and neck) prior to instrumentation was 5.05 kg, and the instrumented mass (head with gel, sleeve, plug, reference sensor, and end caps) was 5.22 kg. The inferior end of isolated subject was potted in polymethylmethacrylate (PMMA) such that the C7-T1 joint was free. After potting, the alignment fixture was removed from the head, and another set of CTs were taken to measure the position and orientation of the sensor relative to the Frankfurt plane.

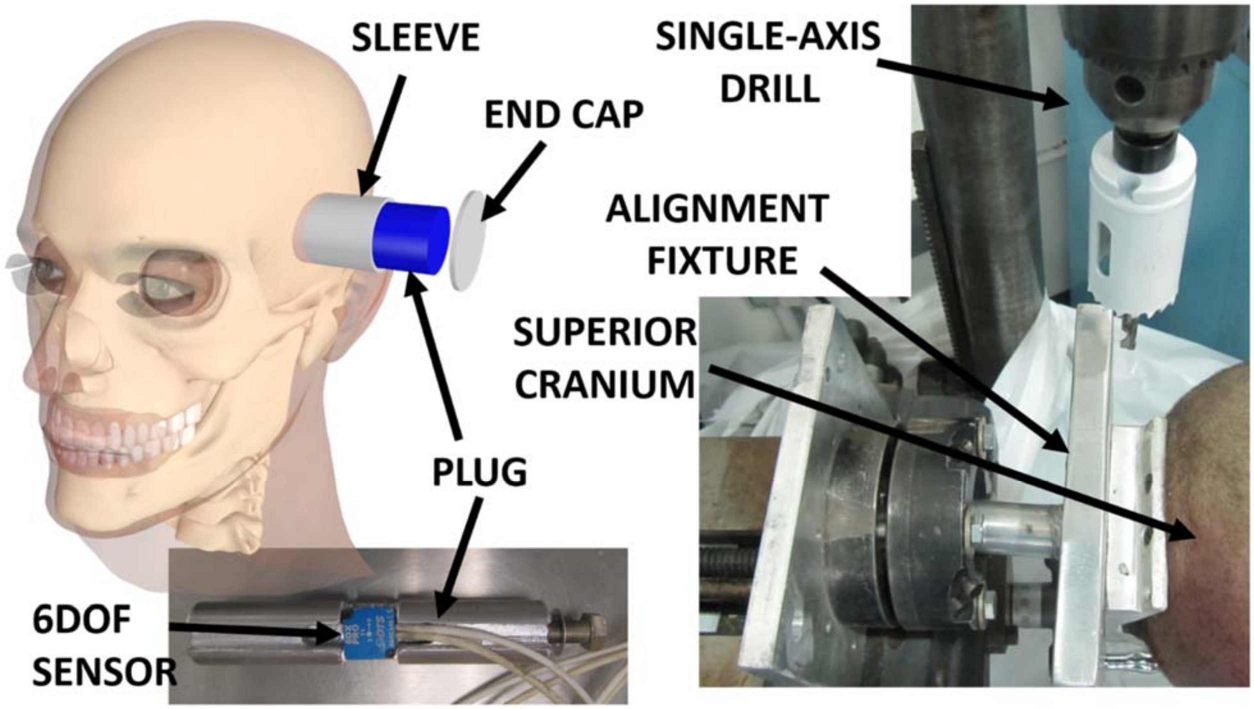

Figure 1: Left is schematic of specimen instrumentation showing the location of the sleeve, plug, and endcap. Bottom left is picture of plug with 6 DOF reference sensor. Right is picture of specimen attached to alignment plate with single-axis drill.

Experimental Testing

Whole-body Tests

Efficacy of the internal sensor was first verified via a series of non-helmeted, whole-body PMHS tests impacted with a 16 kg pendulum from 1 to 4 m/s. The striking end of the pendulum was a 108 mm polyurethane hemisphere. A 25 mm Ethafoam pad was placed between the impacting surface of the pendulum and the head. The specimen was placed on a rigid seat with the Frankfurt plane approximately horizontal. The head was held in position using masking tape affixed bilaterally from the top of the head to the shoulders. The center of the pendulum contacted the forehead at the midsagittal plane superior to the glabella. Head motion was limited to approximately 45 degrees of extension using an angled metal plate with padding to prevent premature failure of the specimen. Table 1 lists the conditions of the Whole-body test series. A schematic of the test setup is shown in Figure 2.

Table 1: Whole-body PMHS test parameters

Test ID	Impact Location	Impact Speed (m/s)
WB101	Front	1
WB102	Front	2
WB103	Front	3
WB104	Front	4

All reference data was sampled at 20 kHz according to SAE-J211 protocols. Head linear accelerations were digitally filtered at CFC1000, and head angular velocities were filtered at CFC180. A high-speed video camera set perpendicular to the sagittal plane of the specimen recorded the impact at 1 kHz.

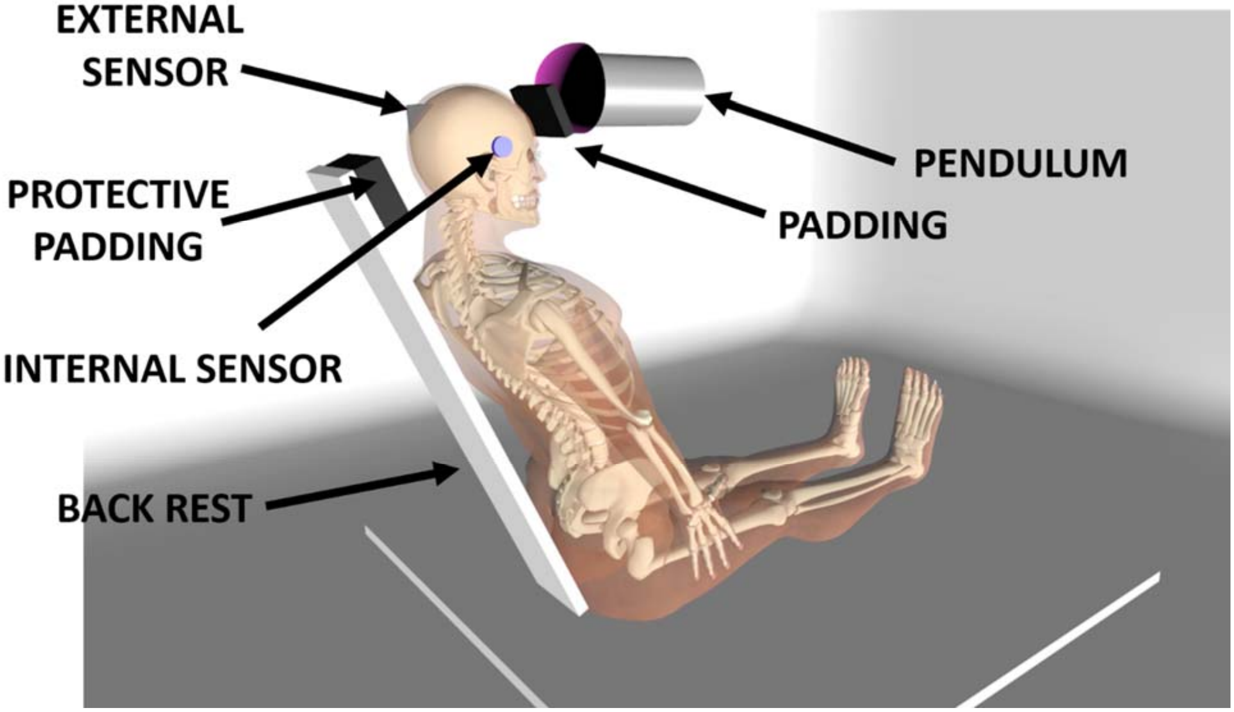

Figure 2: Schematic of whole-body test setup showing PMHS with external and internal sensor, back rest, and pendulum with padding

Isolated Tests

The inferior end of the PMMA block was attached to a six-axis load cell fixed to the top of the mini-sled cart. The cart was attached to precision roller bearings, which could slide freely on two precision rails that were approximately 3.5 m long. The entire mini-sled sat atop a hydraulic lift table to control the pendulum impact location. Additionally, between the six-axis load cell and cart sat a rotational plate that could adjust the axial orientation of the specimen relative to the pendulum. A foam 'catch-device' was placed on the non-struck side to limit head excursion and prevent premature lower cervical spine failure. The ACH was mounted on the specimen using standard donning procedures adopted for PMHS testing. Each impact was delivered via a rigid-arm mini-sled pendulum device. The striking-end of the pendulum consisted of a 2-kg impactor that conformed to National Operating Committee Athletic Equipment (NOCSAE) standards for helmet testing (Standard No. ND081). The total mass of the rigid arm was 19 kg. The head was supported using custom-designed pneumatic cylinders that released prior to impact to allow free motion of the head. The test setup is shown in Figure 3. The specimen was impacted on the helmet at 3 m/s twice in the anteroposterior direction and once in the lateral direction (Table 2). A uniaxial accelerometer was fixed to the back of the rigid arm pendulum at the level of the impactor. All reference data was sampled at 20 kHz according to SAE-J211 protocols. Impact magnitude was measured using a light-velocity trap. Two high-speed video cameras, one perpendicular to plane of impact and one approximately 45 degrees oblique to the plane, recorded the impact at 1 kHz. Pre-test radiographs were taken using a digital x-ray to ensure consistency of the specimen's intra-test initial position.

Table 2: Isolated head-neck PMHS test parameters

Test ID	Impact Location	Impact Speed (m/s)
ISO100	Front	3
ISO101	Front	3
ISO102	Left	3

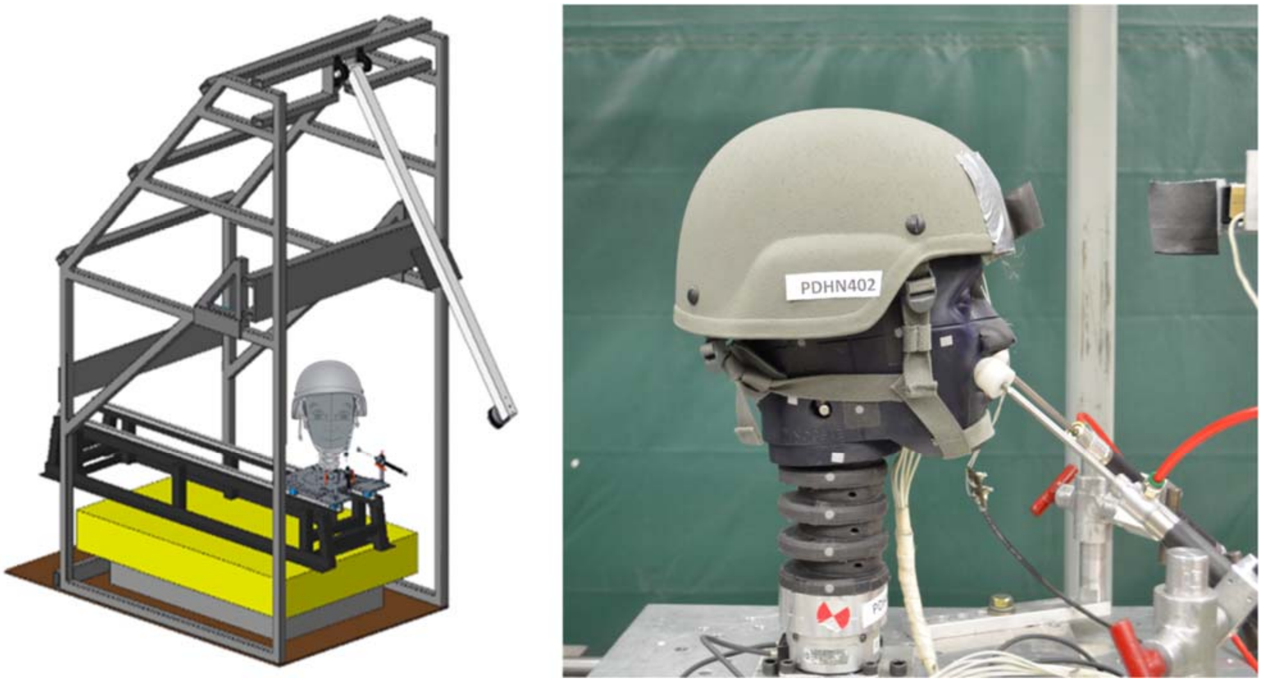

Figure 3: Left is schematic of rigid-arm pendulum with mini-sled. Right is picture of test with helmeted surrogate on mini-sled with hydraulic support arms.

III. RESULTS

A comparison of internal and external sensor time-history in the three axes for linear accelerations and angular velocities from the 4 m/s whole-body pendulum test is shown in Figures 4 and 5.

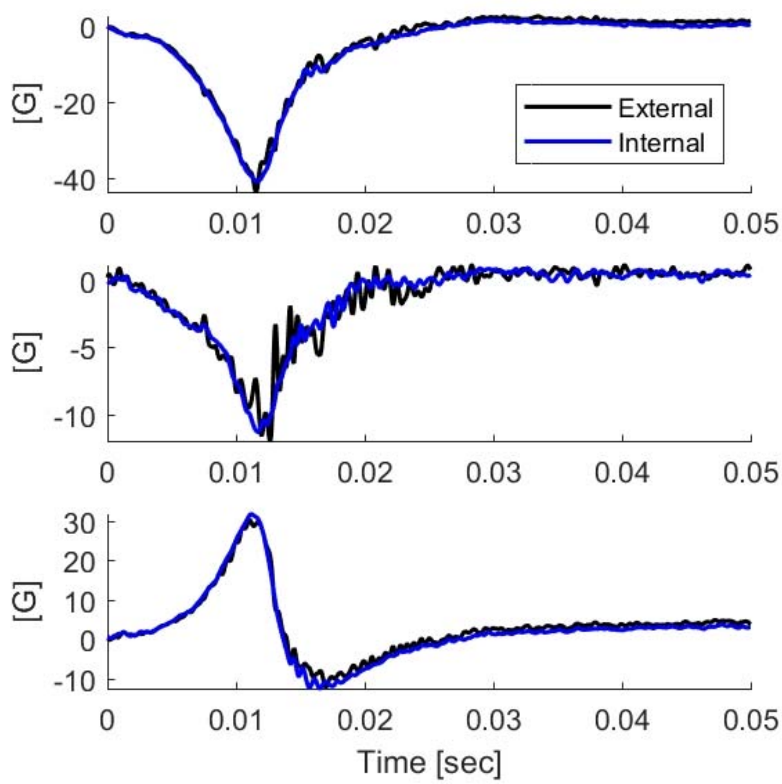

Figure 4: Plots of head linear accelerations in the posteroanterior (upper), lateral (middle), and superoinferior (lower) directions for the internal (blue) and external (black) sensors. The external sensor data were aligned to the Frankfurt plane and calculated at the internal sensor location using equations of rigid-body motion.

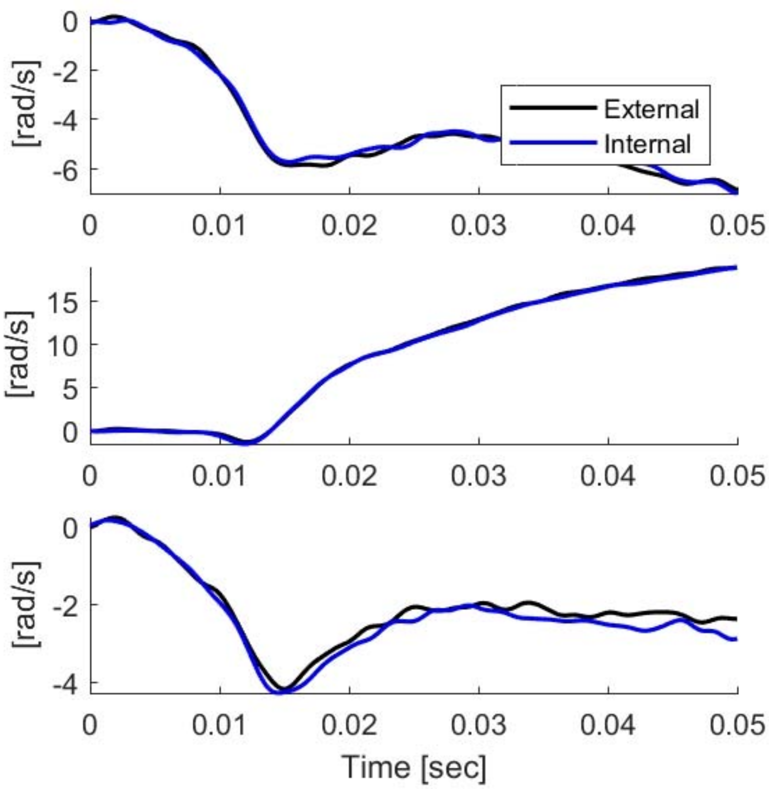

Figure 5: Plots of head angular velocities about the posteroanterior (upper), lateral (middle), and superoinferior (lower) axes for the internal (blue) and external (black) sensors. The external sensor data were aligned to the Frankfurt plane and calculated at the internal sensor location using equations of rigid-body motion.

Normalized root mean-square error between the internal and external sensor are compared in Table 3. Maximum and minimum values and times of attainment for the head kinematics in all three axes are similarly compared for the four whole-body tests in Table 4.

Table 3 Normalized root mean square error for Whole-body PMHS tests

	Linear Acceleration			*Angular Velocity*		
	X (%)	Y (%)	Z (%)	X (%)	Y (%)	Z (%)
1 m/s	6.5	27.4	21.7	23.3	4.5	13.3
2 m/s	1.7	13.0	7.9	12.3	1.8	6.7
3 m/s	1.6	7.0	4.2	6.2	1.1	5.3
4 m/s	2.8	6.7	2.8	2.3	0.9	5.0

Table 4: Whole-body maximum and minimum head kinematics for internal and external sensor

		Param	Units	1 m/s Int	1 m/s Ext	2 m/s Int	2 m/s Ext	3 m/s Int	3 m/s Ext	4 m/s Int	4 m/s Ext
Linear Acceleration	X	Max	g	0.3	0.0	1.5	1.4	3.4	3.4	2.5	1.3
		Time	sec	0.050	0.047	0.038	0.040	0.049	0.050	0.033	0.031
		Min	g	-3.9	-4.0	-15.6	-15.2	-37.9	-36.5	-43.8	-40.9
		Time	sec	0.024	0.023	0.016	0.016	0.015	0.014	0.012	0.012
	Y	Max	g	0.6	0.5	1.3	1.3	4.1	2.0	1.1	0.9
		Time	sec	0.047	0.044	0.015	0.014	0.020	0.024	0.028	0.033
		Min	g	-0.7	-0.6	-1.4	-1.3	-9.2	-8.6	-12.0	-11.3
		Time	sec	0.012	0.027	0.027	0.026	0.015	0.014	0.013	0.012
	Z	Max	g	1.4	0.8	2.8	2.6	15.4	15.1	30.3	31.7
		Time	sec	0.028	0.023	0.013	0.014	0.013	0.014	0.011	0.011
		Min	g	-0.9	-1.2	-1.8	-1.9	-9.2	-7.7	-12.2	-12.6
		Time	sec	0.045	0.048	0.026	0.026	0.020	0.020	0.017	0.017
Angular Velocity	X	Max	rad/s	0.1	0.2	0.7	0.7	0.0	0.0	0.1	0.0
		Time	sec	0.019	0.002	0.017	0.020	0.000	0.004	0.002	0.003
		Min	rad/s	-0.4	-0.4	-0.7	-0.6	-3.3	-3.1	-6.9	-7.0
		Time	sec	0.047	0.045	0.048	0.050	0.018	0.019	0.050	0.050
	Y	Max	rad/s	2.6	2.6	11.2	10.8	18.1	17.9	18.8	18.8
		Time	sec	0.050	0.049	0.050	0.050	0.050	0.050	0.050	0.050
		Min	rad/s	-0.1	-0.1	0.0	0.0	0.0	0.0	-1.4	-1.6
		Time	sec	0.002	0.003	0.001	0.002	0.000	0.001	0.012	0.012
	Z	Max	rad/s	0.1	0.1	0.3	0.2	0.4	0.3	0.2	0.1
		Time	sec	0.002	0.003	0.000	0.003	0.032	0.029	0.002	0.001
		Min	rad/s	-0.7	-0.8	-2.2	-2.4	-2.8	-3.0	-4.2	-4.3
		Time	sec	0.024	0.050	0.050	0.048	0.016	0.017	0.015	0.015

IV. Discussion

Head cg linear accelerations are an important metric used in estimating risk of head fracture and brain injury. Current Federal Motor Vehicle Safety Standards (FMVSS) developed by NHTSA use the resultant head cg linear acceleration to estimate head injury risk. In ATDs, triaxial accelerometers are placed within the skull at the anatomically equivalent head cg location and therefore can directly measure head cg linear accelerations. However, accurate measurement of these accelerations in PMHS is significantly more challenging. Typically, sensor package(s) are mounted externally to the skull, measurements are made relating the position of the sensor to anatomic fiducials, and equations of rigid-body motion are used to calculated head cg accelerations. One study used tetrahedral-style nine accelerometer and three angular rate sensors package to measure head cg accelerations in PMHS [9]. This instrumentation set was later refined with the development of a 6aw package by Kang et. al. and consisted of six linear accelerometers and three angular velocity transducers [10]. While both packages have been shown to accurately calculate head cg linear acceleration, externally mounted transducers are not practical when conducting helmeted PMHS tests. Mounting these sensor packages inside the head would not be feasible due to their size, and their installation would require the removal of large sections of the cranium, thereby compromising the strength and biofidelity of the skull. A more ideal method for measuring head cg linear accelerations for helmeted PMHS tests is to honor the concept of ATD head instrumentation but place the accelerometers at/near the head cg. This is the primary motivation/rationale for the development of the current experimental methodology.

The goal of the whole-body PMHS test series was to confirm the accuracy of the internal sensor using the externally mounted nine-accelerometer package. Curve morphologies (Figures 4 and 5) of the internal sensor essentially overlapped in all three axes for the linear accelerations and angular velocities with the calculated traces from the external sensor. Peak values and times of attainment demonstrated good agreement at tests above 1 m/s. Above 2 m/s the normalized root mean square error was below 5% for the sagittal plane kinematics (x and linear accelerations and y angular velocity). Less agreement was observed in the off-axis directions. The head linear accelerations were computed from the external nine-accelerometer package using techniques from Padgaonkar et al [11]. The differences in the signals between computed and directly measured head cg accelerations found in the current study have also been observed in others [12]. This study also highlighted the difficulties in placing the sensor near the head cg, as well as the need for more refined techniques and instrumentation.

Several factors make mounting a sensor inside the head near the cg difficult. First and foremost, there is no flat boney surface within the head at the location of the cg; thus, the sensor must be rigidly suspended above the cranial vault. This was accomplished in the isolated tests by making an aluminum plug with a recessed space in the center where the 6 DOF sensor was attached. The outer ends of the plug were tapped such that the sensor position could be adjusted to align with the center of the head using 1/2”. Two aluminum plates were attached to the skull—one on either side of the head—using bone screws, and the plug was rigidly fixed to the skull. The position of the plug was carefully selected to coincide with the mediolateral axis of the head near the head cg as measured from the auditory meatus. Pre-instrumentation CT scans were used to select a place within the head that was near the head cg but did not interfere with the dorsum sellae of the sphenoid bone. A 35 mm diameter cylindrical model was created and placed within a 3-D model of the head using 3-D Slicer. The cylinder was placed as close as possible to the approximate head cg location, and the center of the diameter was marked on the left and right side of the skull on the model. Measurements were taken of the alignment plate so that these points could be transcribed to the head.

Another issue when placing a sensor in the head is replacement of the internal contents of the cranium cavity. In the isolated tests, once the hole was drilled through the M/L axis of the skull and prior to insertion of the aluminum plug, the contents of the skull were emptied, as the post-mortem consistency of the brain does not resemble in situ characteristics. Ballistic gel was used as a surrogate for the brain. To fill the cranial space, the specimen was inverted (head down, neck up), and a small hole was drilled at the base of the skull. A 35 mm diameter plastic sleeve was inserted through the M/L axis of the head to keep the gel from flowing out of the holes on the lateral sides of the head. The gel was then poured and allowed to set overnight. This technique minimized air pockets and maintained an open space for placement of the aluminum plug and sensor.

The use of the gel paralleled previous head injury studies in automotive and military literatures [13], as replacing the intracranial contents with this simulant removes air from the cranium—as air is normally present in PMHS specimens—rendering improved handling of the specimen while maintaining the skull’s structural integrity. These processes render the specimen preparation more in line with the normal brain-skull medium. Because of the lack of air gaps within the cranium, any brain motion that might occur due to external loads—inertial or contact—is eliminated, rendering the experimental preparation a better-controlled test condition.

Accurate placement of the plastic sleeve and aluminum plug insert along a M/L axis was an important instrumentation goal. This was controlled by carefully measuring the pre-instrumentation CTs and transcribing these orientations to the fixture at the base of the single axis drill, thereby aligning it with the M/L axis of the head. The technique was successful as the anteroposterior and superoinferior deviations of the sensor axes were less than 2 degrees. Significant deviations of the plug from the M/L axis would make it harder to control the placement of the sensor to its target position. The final position of the head sensor was within 1 mm of the center of the head and within 3 mm of the targeted sagittal plane position. Additionally, misalignment of the plug with the M/L axis may cause protrusion of the aluminum end plates that fix the plug to the skull. In a severe case, this could also lead to asymmetry in the sagittal plane and affect the response of the head.

The largest difference in alignment occurred about the M/L axis. It should be noted that this was not due to the error in aligning the drill but rather a rotation of the plug inside the plastic sleeve about the M/L axis prior to fixing the plug to the skull with the aluminum end plates. Also, it should be noted that the sensor data is aligned mathematically to the anatomic reference frame using the rotation matrix calculated from the instrumentation CT scan. Deviation of the sensor coordinate system from the anatomic system can be corrected regardless of the offset; however, from a signal perspective, the rotation misalignment about the M/L axis does not affect the quality of the measured head kinematics. Thus, this difference in the M/L axis should be expected and, more importantly, is not an issue if the above points are addressed during the analysis, which was conducted in the postprocessing phase of the experimentally-gathered data. There are other points to consider before fully evaluating helmet and head-based wearable sensors. One issue is the resonant frequency of the head, head and helmet, and reference sensor. While the resonant frequency of the head/skull has been reported in literature (approximately 900 Hz) [14] additional characterizations are necessary. This is a limitation of the current study and a topic of future research.

It should be noted that the cg location of a PMHS head is not known a priori, as the experimental model included an intact head-neck complex. Determination of the head mass properties requires disarticulation at the level of the occipital condyles, which compromises the integrity of the specimen. The estimation of the cg was made with respect to the auditory meatus. It should be noted that variations exist in the contours of the periphery of the head and the internal structures. Additional tests are needed to refine estimation technique.

The CT-based techniques described in this paper provide a detailed methodology for researchers to adopt regarding inserting a 6 DOF sensor in an intracranial cavity. This type of instrumentation is needed to evaluate wearable sensors in the military and athletic environments, as well as in the automotive field, where the impact location is uncertain (i.e., rollovers, pedestrian impact). Additionally, non-standard positions in future automotive seating environments may need this type of experimental approach to properly validate human body models for evaluating crashworthiness and advancing safety in highly automated vehicles.

V. CONCLUSIONS

Recognizing the need to accurately measure head kinematics to predict brain injuries in sports and military situations wherein PPE is used, this study developed a methodology to instrument the PMHS head with the sensors within the cranium. The CT-based technique demonstrated herein is valuable to future experimentalists, as PMHS tests are critical in evaluating the efficacy of any helmet- or head-mounted wearable sensor. The cg kinematics from the internal sensor was found to match well with the computed cg kinematics from externally placed sensors, used commonly as the standard in impact biomechanics tests; these tests did not include the PPE. To demonstrate the feasibility of the methodology and the gathering of data from the internal sensors, controlled tests with PMHS-helmet system were conducted. This process can be effectively used to conduct tests with different types of helmets and other head-face wearable sensors for their efficacy evaluations, an urgent need in the sports and military community.

VI. ACKNOWLEDGEMENT

This study was supported by Defense Health funding from the Office of the Assistant Secretary of Defense for Health Affairs (DHA) through the U.S. Army Medical Research and Materiel Command (USAMRMC) Broad Agency Announcement under the Joint Program Committees (JPC-5 and JPC-6) and Award No. W81XWH-16-1-0010. This study was also supported with resources and use of facilities at the Zablocki VA Medical Center, Milwaukee, Wisconsin.

VII. REFERENCES

1. Taylor CA, B.J., Breiding MJ, Xu L, *Traumatic brain injury-related emergency department visits, hospitalizations, and deaths - United States, 2007 and 2013*, in *Morbidity and mortality weekly report*. 2017, **CDC**. p. 1-16.
2. DVBIC, *DoD Numbers for Traumatic Brain Injury Worldwide - Totals*. 2017, Defense and Veterans Brain Injury Center.
3. Yoganandan, N., et al., *Lightweight low-profile nine-accelerometer package to obtain head angular accelerations in short-duration impacts.* J Biomech, 2006. **39**(7): p. 1347-54.

4. Ewing, C.L. and D.J. Thomas, *Torque Versus Angular Displacement Response of Human Head to-GDx Impact Acceleration.* STAPP Car Crash Journal, 1973.
5. Funk, J., et al., *Validation and Application of a Methodology to Calculate Head Accelerations and Neck Loading in Soccer Ball Impacts*. 2009.
6. Panzer, M., et al. *Evaluation of ear-mounted sensors for determining impact head acceleration*. in *IRCOBI*. 2009. York (UK).
7. Duma, S.M., et al., *Analysis of real-time head accelerations in collegiate football players.* Clin J Sport Med, 2005. **15**(1): p. 3-8.
8. Yoganandan, N., et al., *Physical properties of the human head: mass, center of gravity and moment of inertia.* J Biomech, 2009. **42**(9): p. 1177-92.
9. Yoganandan, N., et al., *Head motions using nine accelerometer package and angular rate sensors.* Biomed Sci Instrum, 2008. **44**: p. 256-61.
10. Kang, Y.S., K. Moorhouse, and J.H. Bolte, *Measurement of six degrees of freedom head kinematics in impact conditions employing six accelerometers and three angular rate sensors (6aomega configuration).* J Biomech Eng, 2011. **133**(11): p. 111007.
11. Padgaonkar, A.J., K.W. Krieger, and A.I. King, *Measurement of Angular Acceleration of a Rigid Body Using Linear Accelerometers.* Journal of Applied Mechanics, 1975. **42**(3): p. 552-556.
12. Pintar, F.A., N. Yoganandan, and J. Baisden, *Characterizing occipital condyle loads under high-speed head rotation.* Stapp Car Crash J, 2005. **49**: p. 33-47.
13. Shah, A.S., et al., *Methodology to Study Attenuation of a Blast Wave Through the Cranium.* 2011(54884): p. 17-24.
14. Yoganandan, N. and A. Banerjee, *Survival Analysis-Based Human Head Injury Risk Curves: Focus on Skull Fracture.* J Neurotrauma, 2018. **35**(11): p. 1272-1279.