

Pedestrian Lower Extremity Response and Injury: A Small Sedan vs. A Large Sport Utility Vehicle

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ABSTRACT

Vehicle front-end geometry and stiffness characteristics have been shown to influence pedestrian lower extremity response and injury patterns. The goal of this study is to compare the lower extremity response and injuries of post mortem human surrogates (PMHS) tested in full-scale vehicle-pedestrian impact experiments with a small sedan and a large sport utility vehicle (SUV). The pelvis and lower limbs of six PMHS were instrumented with six-degree-of-freedom instrumentation packages. The PMHS were then positioned laterally in mid-stance gait and subjected to vehicle impact at 40 km/h with either a small sedan (n=3) or a large SUV (n=3). Detailed descriptions of the pelvic and lower extremity injuries are presented in conjunction with global and local kinematics data and high speed video images. Injured PMHS knee joints reached peak lateral bending angles between 25 and 85 degrees (exceeding published injury criteria) at bending rates between 1.1 deg/ms and 3.7 deg/ms. In one case, the onset of tibia and fibula fracture may have prevented ipsilateral knee joint injury. Overall, the SUV appeared to be more aggressive than the sedan in causing injuries to the lower extremities and the pelvis of the tested PMHS.

INTRODUCTION

The World Bank estimates that between 41% and 75% of all road traffic fatalities worldwide are pedestrians (World Bank, 2006). While pedestrian fatalities are a significant problem, the number of pedestrians surviving vehicle impacts with injuries far exceeds the number of fatalities. In the United States (US), for instance, there are more than 13 injured pedestrians for every pedestrian fatality (NHTSA, 2006). Most pedestrian epidemiological data suggest that lower extremity injuries account for nearly a third of all reported injuries, resulting in the lower extremity being reported as either the most or second most frequently injured body region (Kong et al. 1996, Edwards and Green, 1999, Peng and Bongard, 1999, Chidester and Isenberg, 2001, Mizuno 2003, Toro et al. 2005). Among non life-threatening (AIS 2-4) injuries, lower extremity injuries account for

37% percent of all injuries (Chidester and Isenberg, 2001). Costs to society soar when pedestrians with lower extremity injuries require operative treatment and do not regain complete function of the injured limbs (Dischinger et al. 2004, Miller et al. 2004).

While mitigation of pedestrian injury is a complex problem given that many factors (pedestrian stature and stance, vehicle speed and geometry, etc.) affect the injury risk, automobile manufacturers and safety researchers have endeavored to design vehicle front-ends and front-end components aimed at reducing the frequency and severity of lower extremity injuries (e.g. Aldman et al. 1985, Ishikawa et al. 1994, Nagatomi et al. 1996, Detweiler and Miller 2001, Schuster 2004). Vehicle front-end geometry and stiffness in particular have been shown to have a large effect on both pedestrian response and injury (Bunketorp et al. 1983, Kallieris and Schmidt, 1988).

The goal of the current study is to compare pelvic and lower limb response and injury associated with differing production vehicle front-end geometries. Post mortem human surrogates (PMHS) were tested in 40 km/h vehicle-pedestrian impact tests with a small sedan (n=3) and a large SUV (n=3). Detailed descriptions of the injuries are provided in addition to kinematics data highlighting pelvic interaction with the vehicle and dynamic knee joint bending.

EXPERIMENTAL METHODS

Six full-scale vehicle pedestrian impact tests with PMHS were performed using the deceleration sled system at the University of Virginia (UVA) Center for Applied Biomechanics. Aside from using a different vehicle in the first three tests than in the second three tests, all test conditions remained identical to minimize variability in the tests and facilitate comparison of test data and results.

TEST SYSTEM – The test system used to perform the experiments has previously been described (Kam et al. 2005, Kerrigan et al. 2005a, Kerrigan et al. 2005b, Kerrigan et al. 2007, Untaroiu et al. 2007), but will be

briefly reviewed here. The test vehicles included a late-model small sedan (Figure 1) and a late-model large SUV (Figure 2) developed for the US Market. Production versions of both vehicles including all vehicle components were cut transversally, just rearward of the B-pillar, and attached to sled sub-frame assemblies, and then to a deceleration sled system. The vehicle wheels were removed and the front-suspensions were locked. The vehicle bucks were ballasted with additional mass until the sedan's weight matched the vehicle curb weight. The SUV buck weight could only be increased to 80% of the curb weight due to restrictions of the sled system. All damaged parts were inspected after each test and replaced prior to each subsequent test. A small light pedestrian sled was constructed to mimic the vehicle's ground-reference-level and to facilitate surrogate positioning prior to each test. Plywood, which has been shown to possess frictional characteristics similar to road surface (Kam et al., 2005), was used as the shoe-contact surface on the ground-reference-level. Since this study focused primarily on the vehicle-pedestrian impact and post-impact vehicle deceleration caused the surrogate to be thrown from the front of the vehicle, a catching mechanism was installed at the end of the sled tracks to catch the surrogate, prohibit ground contact, and prevent additional injuries (Kam et al., 2005).

PMHS PREPARATION – Four male and two female PMHS (Table 1) were selected for this study based on the absence of pre-existing fractures, lesions or other bone pathology as confirmed by computed tomography (CT) scans. The PMHS were obtained and treated in accordance with the ethical guidelines established by the National Highway Traffic Safety Administration, and all testing and handling procedures were reviewed and approved by an independent Oversight Committee at UVA. Upon arrival, the PMHS were refrigerated (2 degrees C) for 8-32 hours pending blood test results for infectious diseases. Provided the blood screening was negative, the PMHS were stored in a freezer (-15 degrees C) until they were removed and thawed at room temperature for 48-72 hours prior to the pre-test preparation.

Each specimen was instrumented with five six-degree-of-freedom (6DOF) cubes to facilitate kinematics measurement during the test (Figure 3). Each 6-DOF cube (or simply "cube") contained three accelerometers (model 7264B-2000, Endevco Corp., San Juan Capistrano, CA) mounted orthogonally and three magnetohydrodynamic (MHD) angular rate sensors (model ARS-06, Applied Technology Associates, Albuquerque, NM). The six sensors in each cube were arranged in a specially designed aluminum cube to allow measurement of both accelerations and velocities about the cube's three orthogonal axes. Each fully-instrumented cube measured 44 x 44 x 31 mm and weighed 180 g.

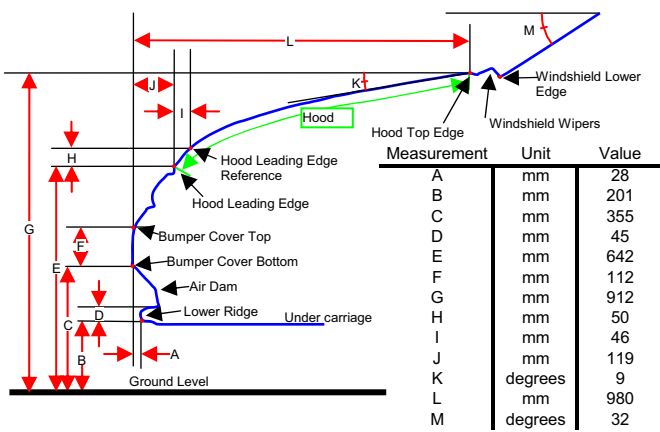


Figure 1. Schematic with dimensions of vehicle centerline cross section of sedan. Hood leading edge reference is based on EEVC specification (EEVC, 1998). Note that the scale of Figures 1 and 2 is not the same.

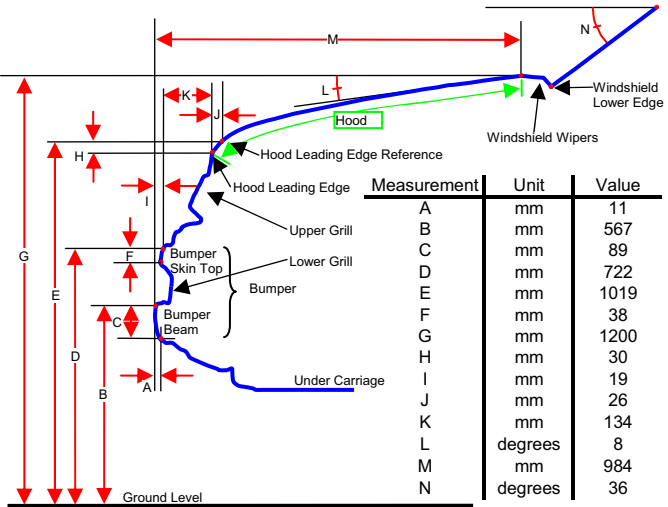


Figure 2. Schematic with dimensions of vehicle centerline cross section of SUV. Hood leading edge reference is based on EEVC specification (EEVC, 1998). Note that the scale of Figures 1 and 2 is not the same.

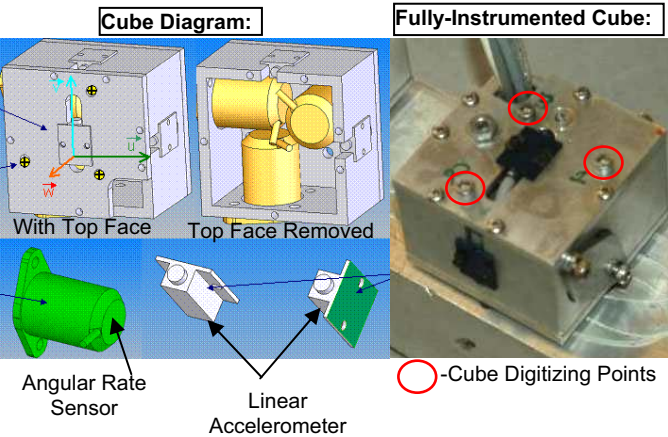


Figure 3. Six-degree-of-freedom (6DOF) cube with instrumentation: schematic (left), photograph (right).

Table 1. PMHS Information. Test numbers prefaced with “Sed” and “SUV” are for tests with the sedan and the SUV, respectively. BMI stands for Body Mass Index. Bone mineral density values are from standard femoral neck/trochanteric dual-energy X-ray absorptiometry readings (Model Delphi W, Hologic Inc., Bedford, MA, USA) where the total hip measurement is the average of the femoral neck, trochanteric and intertrochanteric scans. T-score is the number of standard deviations above/below the mean for a young adult of the same gender at peak bone mineral density. Z-score is the number of standard deviations above/below the mean for adults of the same age, race and gender.

Test #	Specimen Age/Gender	Pre-Test Stature (mm) /Mass (kg)/BMI	Cause of Death	Bone Mineral Density (g/cm ²)	T-Score	Z-Score
Sed1	61/F	1870/80.7/23.1	Ovarian Cancer	1.027	0.7	1.7
Sed2	70/M	1785/54.4/17.1	Cardiac Arrest	0.983	-0.3	0.3
Sed3	62/M	1859/81.6/23.6	Liver Cancer	1.062	0.2	0.7
SUV1	75/F	1765/46.7/15.0	Polycythemia Vera	0.655	-2.3	-0.5
SUV2	75/M	1782/56.7/17.9	Lung Cancer	0.938	-0.6	-0.3
SUV3	53/M	1760/104.2/33.6	Cardiac Arrest	1.051	0.1	0.5

Instrumentation mounting hardware was affixed to each PMHS to permit attachment of kinematics sensor cubes. An aluminum plate (35 x 90 x 3.2 mm) was attached to the sacrum, near the level of the pelvis center of gravity (CG), using two deep threaded screws affixed between the second and third sacral foramen. The skin and underlying tissue (approximately 35 x 90 mm) was removed prior to driving the screws into the sacrum. Aluminum plates (57 x 57 x 3.2 mm) were fixed bilaterally to the distal femora (junction of the distal and mid third portions) and the proximal tibiae (junction of the proximal and mid third portions). The plate mounting technique did not involve penetrating the bone cortex in an effort to minimize the possibility of causing stress concentrations that could contribute to fractures. Instead, steel cerclage wire was fed through holes in the plates and around the circumference of each bone at both the proximal and distal ends of the plates (Figure 4). A ball of fiberglass resin (approximately 20 mm diameter) was placed between the plate and the bone, and the cerclage wires were pulled tight squeezing the plate to the bone and causing the resin to spread around the interface of the bone and the plate (Figure 4). The wires were twisted and the resin dried creating a rigid fixation between the aluminum plates and the bones. To access the bones, the skin and underlying flesh was reflected at the mounting locations.

Each PMHS was also instrumented with ten uni-axial strain gages (CEA-06-062UW-350-P2, Vishay Micro-Measurements Division, Raleigh, NC) to facilitate identification of the time of fracture in the event of a long bone fracture (Figure 4). Four strain gages were affixed around a single cross section at the mid-diaphysis of the right tibia and right femur (struck-side) and a single gage was affixed to the mid shaft of the left femur and left tibia (contralateral side), laterally. The strain gages were mounted to the long bones via an established technique (Untaroiu et al. 2007). An incision was made in the right leg and right thigh at the midpoint of both the posterior

and anterior aspects, and in the left femur and left tibia posteriorly. The flesh was reflected to expose the long bone and the periosteum was removed at the gage mounting cross-section to permit drying of the underlying bone with ether. With their measurement axes aligned with the long axis of the bone, the gages were mounted to the bone surface using a cyanoacrylate adhesive (Cordey and Gautier, 1999).

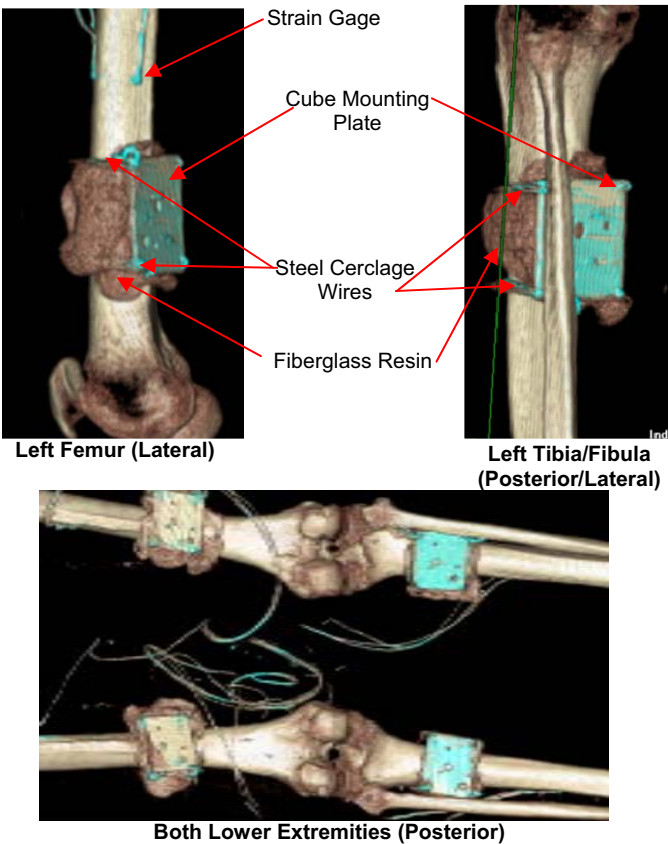


Figure 4. Images from 3-D reconstruction of pre-test CT scans showing cube mounting hardware affixed to the lower extremities.

Following preparation, each specimen underwent a computed tomography (CT) scan to document the orientations of the cube mounting hardware. All surrogate-mounted sensor data were sampled at 10 kHz from -1.5 s to 6 s from initial lower extremity-bumper contact via a wireless data acquisition system (TDAS G5, DTS, Seal Beach, CA) that was padded, water-sealed and attached to the PMHS posterolaterally in the lumbar area (Figure 5).

FINAL PREPARATION AND TEST EVENT – PMHS specimens were each outfitted with a cotton/lycra shirt and pants over a TYVEK body suit and a pair of athletic shoes. Holes were cut in the clothing to expose the cube mounting locations. The PMHS were also outfitted with a lift/support harness made from two sections of seat belt webbing. One longer piece (the shoulder strap) was directed under the arms of the PMHS anteriorly and across the posterior thorax. The second seatbelt strap was used to support the head. When the surrogate was hoisted into position over the pedestrian sled, support of the surrogate's weight was transferred to the clamp of a solenoid release mechanism. The release mechanism supported the weight of the surrogate until immediately prior to the impact.

Each PMHS was positioned with the right lateral side facing the vehicle along the vehicle centerline. The lower extremities were positioned to mimic a mid-stance position with the right (struck-side) lower extremity positioned behind the body and the left lower extremity positioned in front of the body (Figure 5). Fully extended knee orientations in both lower extremities were desired, although gravity drew the right knee forward, rendering the right thigh approximately vertical (Figure 5). The right thigh and leg were oriented at 84.9 (SD 2.3) and 64.2 (SD 3.6) degrees from the horizontal, respectively. Similarly the left thigh and leg were oriented at 82.8 (SD 1.7) and 82.9 (SD 4.8) degrees from the horizontal, respectively. The distance between the anterior tip of each shoe's sole measured 332 (SD 35) and 252 (SD 47) mm in the anterior-posterior and medial-lateral directions, respectively. The upper extremities of the surrogate were bound at the wrist, anterior to the body, with the left wrist closest to the abdomen, to ensure repeatable kinematics and the most severe impact (Kam et al. 2005). Once the final position of the surrogate had been set, a coordinate measurement machine (FARO Technologies, Lake Mary, FL) was used to digitize 88 points used to measure the exact position and orientation of the subject, and to measure the orientation of the 6DOF cubes with respect to the global coordinate system (Figure 5).

The test event was initiated by a pneumatic propulsion system that accelerated the vehicle sled, which was instrumented with a pair of damped linear accelerometers (Entran #EGCSD1S, Measurement Specialties, Inc., Hampton, VA) to 40 km/h. The vehicle sled passed an inductive sensor on the track that triggered the release of the surrogate approximately 20 ms before the initial bumper-lower extremity contact.

Additionally, the vehicle sled passed another inductance sensor used to measure the vehicle's velocity approximately 8 ms prior to initial bumper-lower extremity contact. Vehicle-PMHS interaction continued for 250 ms after bumper contact. The vehicle was then decelerated at approximately 6 g and the surrogate was thrown forward into the catching mechanism. Each test was captured from five different angles using high speed video cameras at 1000 frames per second.

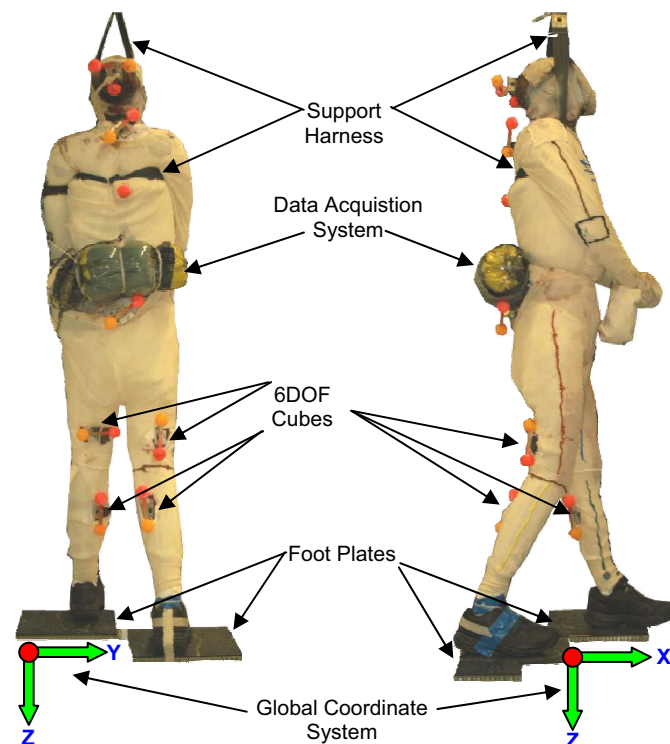


Figure 5. Typical pre-impact PMHS position: posterior view (left) and right side lateral view (right).

After each test, a second CT scan was performed to assist in documentation of the injuries. After the CT scan, each specimen underwent a detailed full body necropsy performed jointly by a pathologist's assistant and an orthopaedic surgeon. While a radiologist's report of the CT scan was used to assist exploratory efforts during the necropsy, all body regions, including those not indicated in the radiologist's report were explored and all injuries were identified and either photographed or hand-drawn. Finally, a standardized bone mineral density measurement (Table 1) was performed on the femoral head and neck regions of the left femora of each specimen by dual energy X-ray absorptometry (Model Delphi W, Hologic Inc., Bedford, MA, USA).

CUBE DATA PROCESSING

Using established cube data processing methods (Rudd et al. 2006) pelvic cube data were used to determine pelvic cube kinematics in a reference frame fixed with respect to the impact vehicle, and lower extremity cube data were used to determine varus/valgus knee bending angles. The orientation of each cube in the reference frame was determined prior to each test from

the coordinate measurement machine data. Data from the angular rate sensors in each cube were used to update the orientation of the cube at each time step of the impact interaction. By updating the orientation of the cube at each time step, the local sensor data could be expressed in the global reference frame. For the pelvis cube, the kinematics data were then expressed in the vehicle reference frame by accounting for the motion of the vehicle. For the lower extremity cubes, using CT image data cube kinematics in the global frame were transformed to leg and thigh segment kinematics by expressing the cubes' orientations with respect to the segment reference frames. Lastly, from the leg and thigh segment kinematics, knee joint kinematics were calculated.

INITIAL ORIENTATION OF THE CUBES – The global (inertial) coordinate system was defined by the direction of motion of the vehicle, which was specified by the orientation of the sled tracks. The orientation of the coordinate axes of the system was determined so that the goal in positioning the surrogate would be to align the surrogate such that it corresponded with the SAE convention (SAE J211/1, 1995) for a standing human: +X pointing anteriorly, +Y pointing to the right, and +Z pointing down (Figure 5). Using the pedestrian reference system as a basis, the vehicle's motion occurred in the -Y direction, with the +X axis pointing away from the vehicle to the vehicle right hand side, and +Z pointing in the direction of gravity.

To determine the cube's orientation, after setting the final pre-test position of the surrogate, the locations of three irregularly spaced points (screw heads) on the cube's top surface (Figure 3) were digitized using the coordinate measurement machine. From the locations of the three points, the locations of three of the cubes corners (on the top surface) were determined by minimizing the sum squared error of the distances between the digitizing points and the cube corners. Using the locations of the cube corners, three vectors defining orientation of the cubes local coordinate system in the global reference frame were calculated. From these vectors the 3 x 3 matrix $\mathbf{T}_g^l$ representing the vector transformation from the local (cube) reference frame l to the global frame g was determined where

$$\mathbf{r}_g = \mathbf{T}_g^l \mathbf{r}_l \quad (1)$$

transforms any vector $\mathbf{r}$ from its representation in the local frame ($\mathbf{r}_l$) to its representation in the global frame ($\mathbf{r}_g$).

UPDATING THE LOCAL TO GLOBAL TRANSFORMATION – Transformation of locally measured kinematics to the global reference frame relies on the calculation of the transformation matrix $\mathbf{T}_g^l$ at each time step. From the Equation 1 and the Coriolis equation in mechanics, the angular velocity tensor can be defined as (Shames, 1999):

$$[\omega_l \times] \equiv \begin{bmatrix} 0 & -\omega_{lz} & \omega_{ly} \\ \omega_{lz} & 0 & -\omega_{lx} \\ -\omega_{ly} & \omega_{lx} & 0 \end{bmatrix} = \mathbf{T}_g^l \dot{\mathbf{T}}_g^l \quad (2)$$

where $\dot{\mathbf{T}}_g^l$ is the time derivative of $\mathbf{T}_g^l$ and $\mathbf{T}_g^l$ is the inverse of the transformation defined in Equation 1 and ω_l is the local frame representation of the vector ω describing the angular velocity of the local frame. The elements of ω_l (ω_{lx} , ω_{ly} , ω_{lz}) are the angular velocities measured by the angular rate sensors on the cube. The definition of the angular velocity tensor can be rewritten as a set of nine ordinary differential equations

$$\dot{\mathbf{T}}_g^l = \mathbf{T}_g^l [\omega_l \times] \quad (3)$$

that can be used to determine the time history of the transformation from the body-fixed to the inertial reference frame. Bortz (1971) showed, using only geometry, that the vector transformation $\mathbf{T}_g^l$ could be written as a function of the orientation vector ϕ as

$$\mathbf{T}_g^l = \mathbf{I} + \frac{\sin \phi}{\phi} [\phi \times] + \frac{1 - \cos \phi}{\phi^2} [\phi \times]^2 \quad (4)$$

where $[\phi \times]$ is defined similarly to the angular velocity tensor (Equation 1) and the orientation vector ϕ defines the direction about which the global coordinate system g could be rotated an angle equal to the length of the orientation vector (ϕ) to align it with the local system l . Equation 4 assumes that both $\mathbf{T}_g^l$ and ϕ are functions of time and that at time $t=0$ the local frame is aligned with the global frame. It should be noted that ϕ does not have a subscript because it has the same representation in both the local and the global reference frames. Then, from Equation 4, Bortz (1971) showed that the "inertially measurable" angular velocity vector could be separated from the effects of the non-commutativity of finite rotations in a simple differential equation in ϕ :

$$\begin{aligned} \dot{\phi} &= \omega + \dot{\sigma} \\ \dot{\phi} &= \omega + \frac{1}{2} \phi \times \omega + \frac{1}{\phi^2} \left(1 - \frac{\phi \sin \phi}{2(1 - \cos \phi)} \right) \phi \times (\phi \times \omega) \end{aligned} \quad (5)$$

where $\dot{\sigma}$ is referred to by Bortz (1971) as the "non-commutativity rate vector" because its magnitude is indicative of the effects of the non-commutativity of finite rotations.

The nine ordinary differential equations defined in Equation 3 can be integrated to determine the time history of the transformation matrix $\mathbf{T}_g^l$ (see DiMasi, 1995). Alternatively, the three differential equations

defined in Equation 5 can be integrated and implemented in Equation 4 to determine the same solution (see Mital, 1978). In this study, a further simplification of the transformation calculation algorithm is implemented without sacrifice of the accuracy of the results. A step-wise version of Bortz' solution can be used to calculate the rotation of the body-fixed frame not over the entire time up to time t_i , but over a single time step from $t_{i-1}\Delta t$ to t_i . In other words, instead of calculating the orientation vector $\phi(t_i)$ required to align the global frame (or local frame at $t=0$) with the local frame at $t=t_i$ (as in Equation 4), an orientation vector ϕ can be defined at each time step to rotate the local frame at time $t_{i-1}\Delta t$ to align it with the local frame at t_i . Under this implementation, the effect of the non-commutativity of finite rotations ($\dot{\phi}$ in Equation 5) can be assumed to be negligible (over a single timestep) and the angular velocity vector ω can be integrated directly to determine the time history of the orientation vector ϕ (using Equation 5). Then using either Bortz's solution (Equation 4) or the Euler Parameter method (implemented by Rudd et al. 2006), an incremental rotation matrix can be determined at each time step. Then the matrix $\mathbf{T}_g^l$ can be determined at each timestep by multiplying the timestep's incremental rotation matrix by $\mathbf{T}_g^l$ from the previous step in a marching algorithm (Rudd et al. 2006).

PELVIS VELOCITY WITH RESPECT TO THE VEHICLE

– Since the linear accelerometers mounted within the 6DOF cube could not be all mounted at the exact same location, the acceleration signals included centripetal accelerations resulting from the angular motion of the cubes in addition to the accelerations resulting from the translation of the cubes. Thus, the centripetal accelerations were subtracted from the acceleration signals, which effectively moved the location of the acceleration measurement from the accelerometer to the center of the 6DOF cube, by applying the rigid body kinematics equation (Shames, 1999):

$$\mathbf{a}_p = \mathbf{a}_o + \omega \times (\omega \times \mathbf{p}_{op}) + \alpha \times \mathbf{p}_{op} \quad (6)$$

where, in this case, $\mathbf{a}_o$ is the acceleration of the center of the cube, $\mathbf{a}_p$ is the acceleration measured by the accelerometer (at a point p), $\mathbf{p}_{op}$ is the vector from the cube center to the accelerometer, and ω and α are the angular velocity and the angular acceleration of the cube, respectively. Since the position vector $\mathbf{p}_{op}$ has only one component, which is aligned with the sensitive axis of the accelerometer, only the local angular velocities, measured by the angular rate sensors, were required to separate the centripetal accelerations measured from the acceleration signals.

From the time history of $\mathbf{T}_g^l$, the cube's acceleration vector $\mathbf{a}$ was transformed from a local (cube) representation (a_l) to a global representation (a_g):

$$\begin{bmatrix} a_{gx} \\ a_{gy} \\ a_{gz} \end{bmatrix} = \mathbf{T}_g^l \begin{bmatrix} a_{lx} \\ a_{ly} \\ a_{lz} \end{bmatrix} \quad (7)$$

where the elements of a_l (a_{lx} , a_{ly} , a_{lz}) are the accelerations measured by accelerometers (after correcting for the centripetal accelerations measured by the sensors) mounted within the 6DOF cube fixed to the pelvis. The global accelerations (a_{gx} , a_{gy} , a_{gz}) were then integrated directly to determine the components of the cube's global velocity vector $\mathbf{v}$ represented in the global frame (v_{gx} , v_{gy} , v_{gz}). The vehicle's velocity time history was determined by integrating vehicle sled accelerations and using the track-mounted velocity sensor to seed the velocity calculation. The vehicle's velocity (approximately 40 km/h in the $-Y$ direction of the global frame) was added to the global Y-component of the pelvis velocity (v_{gy}) to determine the Y-component of the pelvis velocity with respect to the vehicle.

LEG AND THIGH SEGMENT REFERENCE FRAMES –

To determine leg and thigh body segment kinematics, cube (local) kinematics were expressed in coordinate systems defined by anatomic landmarks on the tibia (for the leg) and the femur (for the thigh). The leg and thigh body segment reference frames presented by Capozzo et al. (1995) were adapted for use in this study (Figure 6).

Although the specific location of the segment coordinate system origins is not important for this analysis, the origin of the thigh system was defined as the center of the line connecting the lateral and medial femoral epicondyles (Figure 6). The X-axis was defined to be perpendicular to the mid-coronal plane (defined by the two epicondyles and the center of the femoral head) with the positive direction pointing anteriorly. The Z-axis was defined as the vector connecting the origin to the center of the femoral head with the positive direction directed proximally. The Y-axis was defined to be perpendicular to both the X and Z axes, directed from the origin leftward.

The midpoint of a line connecting the most distal points of the malleoli was defined as the origin of the leg system (Figure 6). The X-axis was defined to be perpendicular to the mid-coronal plane (defined by the proximal protrusion of the fibular head and the most distal points of the malleoli), with the positive direction pointing anteriorly. The intersection of the mid-coronal and mid-sagittal planes (defined to be perpendicular to the mid-coronal plane and containing the center of the tibial tuberosity) established the Z-axis, which was directed proximally. The Y-axis was defined to be perpendicular to the X axis and the Z-axis, pointing leftward.