

This bypasses erroneous attempts to solve the relative acceleration equation in differential form and eliminates the need for difficult accelerometer placements allowing for its use on human volunteers and PMHSs.

#### References

- [1] Alem N.M. The measurement of 3-D rigid Body motion. Proc. 2nd Ann. Int. Meeting of the Ad Hoc Committee on Human Subjects for Biomechanical Research, 1972.
- [2] Padgaonkar, A., et al. Measurement of angular acceleration of a rigid body using linear accelerometers. *Journal of Applied Mechanics* 1975; 42.
- [3] Viano D., et al. Measurement of head dynamics and facial contact forces in the Hybrid III dummy, SAE Paper No. 861891, 1986.

6840 We–Th, no. 15 (P60)

#### Modal analysis of the dry human skull

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Knowledge of the resonant frequencies and the mode shapes of the human skull can lead to a better understanding of the head injury mechanism upon impact on the one hand, and provide expertise in designing new protective headgear on the other hand.

We determined the resonant frequencies and associated mode shapes in the frequency band of 20 to 6000 Hz, for four defleshed skulls by means of experimental modal analysis. We used a hammer impact as excitation source and for each skull four different impact points were chosen, in those regions that are most frequently impacted in bicycle accidents. 60 evenly distributed points on the skull surface were used to measure the response of the vibrations. The geometrical model used to visualize the mode shapes was constructed by extracting the coordinates of the selected points from CT scans of the skulls.

We found 11 to 13 different resonant frequencies for these skulls. At some impact points not all the resonant frequencies were excited, and we found that impact in temporal areas will excite most of the resonant frequencies. In contrast with some other authors no frequency below 1000 Hz was found for the unconstrained defleshed skull. However when testing with a skull fixed by the foramen magnum two frequencies lower frequencies were found at 150 Hz and 850 Hz respectively. These preliminary results will be compared and validated against modal analysis of an intact human head in the near future.

6514 We–Th, no. 16 (P60)

#### A novel algorithm to measure linear and rotational head acceleration using single-axis accelerometers

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An algorithm was previously developed to measure linear head acceleration using orthogonal surface-mounted accelerometers [1]. A limitation of this algorithm is the estimation of rotational acceleration by assuming rotations about a fixed point in the neck. To address this limitation, we report on a new algorithm and demonstrate its accuracy. 6DOF directly measures rotational head accelerations, in addition to linear head accelerations, with six accelerometers placed tangential to the surface of the head. By assuming rigid body dynamics, the acceleration of any point ( $i$ ) on the head ( $\vec{a}_i$ ) undergoing linear and rotational acceleration is projected on the sensing axis of the accelerometer ( $\vec{r}_{a_i}$ ):

$$\|a_i\| = \vec{r}_{a_i} \cdot \vec{H} + \vec{r}_{a_i} \cdot (\vec{\alpha} \times \vec{r}_i),$$

where  $\vec{H}$  is the linear acceleration vector at the head center of gravity (CG),  $\vec{r}_i$  is the position vector of point  $i$  relative to head CG, and  $\vec{\alpha}$  is the rotational acceleration of the head. Centripetal acceleration is neglected based on the assumption that the accelerometer sensing axis is not radially oriented to the acceleration vector. Iterative optimization is used to solve for linear ( $\vec{H}$ ) and rotational ( $\vec{\alpha}$ ) accelerations that minimize the sum of square error between each accelerometer value and the expected acceleration:

$$\min \sum_{i=1}^n \left[ \|a_i\| - \vec{r}_{a_i} \cdot \vec{H}_E + \vec{r}_{a_i} \cdot (\vec{\alpha}_E \times \vec{r}_i) \right]^2,$$

where  $n$  is the number of accelerometers,  $\vec{H}_E$  is the estimated linear acceleration of the head CG, and  $\vec{\alpha}_E$  is the estimated rotational acceleration vector. Experimentally, peak linear and rotational acceleration and an industry standard time-weighted impact severity measure (HIC) were calculated by the

6DOF algorithm and compared to the results from the traditional 3-2-2-2 nine accelerometer configuration from a Hybrid III headform. Six accelerometers mounted on the surface of the Hybrid III were used as input to the 6DOF algorithm. A weighted pendulum impacted the headform over a variety of impact locations and speeds ranging from 3–7 m/s (86 impacts). Compared to the 3-2-2-2 measures, 6DOF had  $\pm 0.2\%$  ( $R^2=0.99$ ),  $\pm 2.5\%$  ( $R^2=0.99$ ), and  $\pm 0.6\%$  ( $R^2=0.99$ ) error for linear and rotational acceleration and HIC, respectively. Using only 6 accelerometers, 6DOF is capable of accurately measuring linear and rotational acceleration of the head.

#### References

- [1] Crisco JJ, Chu JJ, Greenwald RM. An algorithm and method for estimating the magnitude and direction of the linear acceleration of the head using multiple single-axis accelerometers. *Journal of Biomechanical Engineering* 2004; 126: 849–854.

## 5.6. Abdominal Injury Biomechanics

5422 We–Th, no. 17 (P60)

#### A pelvic fracture model for the assessment of treatment options in a laboratory environment

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**Background:** Optimal prehospital and clinical management of patients with severe pelvic trauma is discussed controversially. Prospective evaluations of different treatment strategies remain in many cases uncontrolled, and therefore, not evidence-based. The purpose of the present study was to develop a porcine model of reproducible severe pelvic trauma for subsequent controlled laboratory trials.

**Methods:** The study was performed on 13 juvenile porcine cadavers. Pelvic fractures were created by applying a pure anterior–posterior compression load to the pelvic ring using a servohydraulic material testing machine. Fracture patterns were classified according to the Young-Burgess classification and the Tile classification using postfracture CT scans including 3D-reconstructions.

**Results:** Disruptions of the posterior pelvic ring segment were in twelve cases unilateral and once bilateral transforaminal vertical sacrum fractures. Injuries of the anterior ring segment were obturator ring fractures bilateral, ipsilateral or contralateral to the injury of the posterior ring segment. According to the Tile classification this resulted in 12 Type C1 and one Type C3 fractures. In the Young classification all injuries were classified as Type APC III. In six cases transverse process fractures were found ipsilateral to the posterior ring disruption. Initial force drops indicating bony or ligamentous injuries occurred at mean forces of  $4030 \pm 269$  N (range, 3617 N to 4374 N).

**Conclusion:** The present model was able to create reproducible unstable pelvic fractures and can be used for controlled laboratory trials to study the management of patients with pelvic fractures.

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#### Stochastic process methods in the dynamics of deformation of tissues

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The stochastic process approach to the mechanics of tissues has been developed as the result of difficulties arising in the application of continuum mechanics methods for the description of mechanical behaviour of of tissue systems. The main idea of the stochastic approach is based on the similarities of equations of kinetics of deformed medium and equations encountered in the stochastic Markov process theory. The concept of the model is based on the simple idea according to which the deformation of the  $k$ -th surface in the tissues system implies the deformation of other  $l$ -th surface in the same system, i.e. there exist the transformation operator which expresses the relationship between the deformation vector  $W^k$  referring to  $k$ -th surface and deformation vector  $W^l$  referring to  $l$ -th surface. Assuming the integral representation and transitivity of the transformation operator the non-linear integral equation describing the deformation of the tissue system has been elicited. The deformation equation is very similar to the generalization of the Smoluchowski-Chapman-Kolmogorov equation well-known from the theory of stochastic Markov processes. In the stochastic process theory the kernel function involved in the Smoluchowski-Chapman-Kolmogorov equation represents the probability of state transition in the time interval  $[\tau, t]$ .

The non-linear integral equation applied for the description of deformations of tissues possesses the continuous and discontinuous solutions. The former correspond with continuous deformations of tissues whereas the latter with an injury of the tissue system. Discontinuous solutions may even be implied by continuous boundary conditions. Such a case enables the model to describe the situation when as the result of external force an internal trauma of the tissue system occurs without an explicit damage of its external surface.