

Dynamic Impact Response and Predicted Brain Tissue Deformation Comparisons for an Impacted Hybrid III Headform With and Without a Neckform and Torso Masses

Evan S. Walsh, Marshall Kendall, T. Blaine Hoshizaki, and Michael D. Gilchrist

Abstract Helmet assessment protocols differ with respect to neckform and torso presence. To determine the relative effects of neckform and torso masses on predicted three-dimensional human impact response, a suspended Hybrid III headform was impacted with a pendulum striker. Trials were completed with and without a Hybrid neckform attached to the base of the headform. Surrogate torso masses were increased in 11.3 kilogram increments to the base of the neckform simulating body masses up to 95 kilograms. These trials were then compared to impacts on a linearly translating table.

The dynamic impact responses from the trials were recorded and used as loading curves for finite element analysis using the University College Dublin Brain Trauma Model to calculate brain tissue deformations within the Cerebrum.

Statistical significance between impact conditions was assessed with ANOVA and Tukey's post hoc tests at an alpha value of 0.05. Differences were identified between the suspended conditions and the table conditions for linear acceleration; the no neck and the other suspended conditions for angular acceleration; and the no neck condition and all but the suspended no torso impact condition for maximum principal strain. This shows that torso mass and the neckform that mechanically links the mass to the headform should be considered when measuring headform impacts.

Keywords Anthropometric Test Device, Helmet Assessment, Neurotrauma Biomechanics, Torso Mass.

I. INTRODUCTION

Helmets and their associated assessment protocols and standards were primarily introduced into sport to prevent catastrophic head injury, the risk of which has effectively been reduced to near zero because of the success of head protection development [18][21]. Recent attention towards concussion in sports has led to questions surrounding protective headgear and how it is tested [30]. Currently there are differing methodologies that exist for helmet assessment protocols and head impact reconstructions based on multiple physical characteristics, including the presence or absence of surrogate neckforms [26][38]. The European impact standards for ice hockey [17], alpine [15] and cycling [16] helmet assessment all use an untethered headform drop, whereas the current North American standards all have a solid metal neckform for guided headform drops [1][6][7][9][32]. Although both the untethered and rigidly attached methodologies effectively deliver consistent impact measures that are reproducible between laboratories, they likely do not represent a truly biofidelic response to impacts. If the goal is to produce a standard that considers the biofidelic risks of both catastrophic and mild traumatic brain injuries to mechanical loading [2], the connection between the headform and a torso must be addressed. Collision protocols proposed for American football and ice hockey employ a compliant Hybrid neckform. It is not clear which of these methods best represent impact dynamics. The biofidelity of modern mechanical neckforms has been critiqued for not ideally representing the dynamic compliance and resistance of human necks under inertial or impact loading [12], while the lack of any basal tethering does not mimic a live human condition. A study that varied the striking mass reported that the relative mass of the two objects significantly alters both the dynamic impact response and the predicted tissue deformation [27]. This is an important consideration when reconstructing injury events involving sporting events, where collisions between two players are common occurrences. How the mass of the impacted player

E. S. Walsh and M. Kendall are PhD candidates in Neurotrauma Biomechanics at the University of Ottawa in Canada (613-562-5800 x7212, ewals056@uottawa.ca). T. B. Hoshizaki is a Professor of Biomechanics in the department of Human Kinetic at the University of Ottawa. M. D. Gilchrist is a Professor in the department of Mechanical Engineering at University College Dublin in Ireland and is cross-appointed to the University of Ottawa.

affects brain tissue response has yet to be defined. Thus, the purpose of this study was to compare the dynamic responses of the headform and calculated brain tissue deformations while varying torso mass and impact parameters.

II. METHODS

A fiftieth percentile Hybrid III headform instrumented with a 3-2-2-2 linear accelerometer array [33] was centrally impacted to the forehead at a velocity of 4.4 metres per second using a 3.6 kilogram pendulum striker with a vinyl nitrile 602 foam impact surface. The nine mounted single-axis Endevco 7264C-2KTZ-2-300 accelerometers were sampled at 20 kilohertz by a Diversified Technical Systems (TDAS) Pro Lab system and filtered using the Society of Automotive Engineers (SAE) J211 class 1000 low pass Butterworth protocol before being collected by TDAS software. Data collection was triggered when any of the accelerometers reached a 29.4 metres per second squared, or 3 g, threshold and terminated after 15 milliseconds. These data collection techniques and parameters are common to collision impact testing [23][37][38][41][42]. The three orthogonal positive linear axes were defined from the headform's centre of gravity as projecting forward, to the left and cranially for the x-, y- and z-axes respectively.

Trials were divided into three categories by headform condition. For categories A and B, the headform was suspended from the rear aspect of its crown, creating a slight positive elevation of the surrogate. Impact condition one had no neckform connected to the Hybrid III headform (A1) and therefore was free to move in all directions upon impact. This lack of basal tethering was similar to European helmet assessment drop protocols. The second category of impacts had a Hybrid neckform attached, with steel nodding joints [41], to the base of the Hybrid III headform, B1. The steel nodding joints replaced the standard rubber nodding joints and reduced the high magnitudes of mechanical noise reported by Foreman by forming a more rigid head-neck connection without significantly affecting the desired signals [19]. Masses were subsequently added to the base of the neckform in four 11.3 kilogram increments, from 11.3 to 45.5 kilograms, to simulate the effect of torso mass (B2, B3, B4, and B5). Neither the base of the neckform for condition B1 nor the bottom of the surrogate torso masses from B2, B3, B4, and B5 was attached to any additional restrictive or resistive boundary condition.

Anthropometric data from Clauser [8], Dempster [11], and de Leva [10] were used to equate the simulated torso masses to adult male total body masses. Table 1 illustrates how the B4 condition approximates a seventy kilogram male and the B5 condition similarly represents a ninety-five kilogram individual, or close to the total masses for the fiftieth and ninety-fifth percentile Hybrid III anthropometric test devices of 77.7 and 101.2 kilograms, respectively. This added torso mass was predicted to affect the headform's response due to the altered moment of inertia of the system. According to anthropometric data, the desired torso mass for average test devices was 40.3 kilograms; the fiftieth percentile Hybrid III has a combined torso mass of 40.2 ± 0.28 kilograms [19].

TABLE I
SIMULATED BODY MASS IN KILOGRAMS FROM SURROGATE TORSO MASS PROPORTIONS

Subset	Torso	Clauser	Dempster	de Leva	Mean
Mass Proportion		0.5070	0.4970	0.4346	0.4795
B1	0	0	0	0	0
B2	11.36	22.4	22.9	26.1	24
B3	22.7	44.8	45.7	52.3	48
B4	34.1	67.2	68.6	78.4	71
B5	45.5	89.7	91.5	104.6	95

The third category had the base of the Hybrid neckform being affixed to a 12.78 kilogram translating table on linear rails that was first allowed to move freely in the sagittal plane (C1) and then immobilized (C2). The headform was kept at the same positive elevation as the suspended conditions for the group C impacts.

Angular accelerations were calculated by comparing the six peripheral linear accelerometer signals to the three orthogonal centre of gravity sensors of the 3-2-2-2 array using the principles of rigid body mechanics. This created three-dimensional linear and angular dynamic response loading curves that were applied to the University College Dublin Brain Trauma Model (UCDBTM) in order to calculate the brain tissue deformation metric of maximum principal strain (MPS) within the Cerebrum.

Consistent with previous Hybrid III impact testing, each of the impact conditions was tested three times [27][28][37][38][41]. Statistical significance between each of the category subsets was assessed with ANOVA and Tukey's post hoc tests at an alpha value of 0.05, with particular focus on comparing the dynamic impact responses and brain tissue deformations of the neck less surrogate, group A, to each of the other suspended subsets of categories group B and the two translating table subsets of the third category, group C. Impact responses by neck and torso condition that are not significantly different from the neck less surrogate.

Computational Modeling

Finite element analysis of all headform impacts was performed using a refined version of the University College Dublin Brain Trauma Model. The original finite element model defined by Horgan and Gilchrist [23] with 26 000 elements defining the tissues of the Cerebrum and Cerebellum was refined to 32 999 elements by [13] to reduce the risk of discretization error. The computational surrogate is defined as a linearly viscoelastic material finite element model that uses a large deformation theory and has been validated against cadaveric testing. [13][19]

The behaviour of this tissue was characterized as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus. The hyperelastic law in Pascal was given by:

$$C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15}$$

where C_{10} and C_{01} are the temperature-dependent material parameters and t is time in seconds. The compressive behaviour of the brain was considered elastic. Material properties of the ten distinct biological tissues modelled in the UCDBTM [22][23] are presented in Table 2, below.

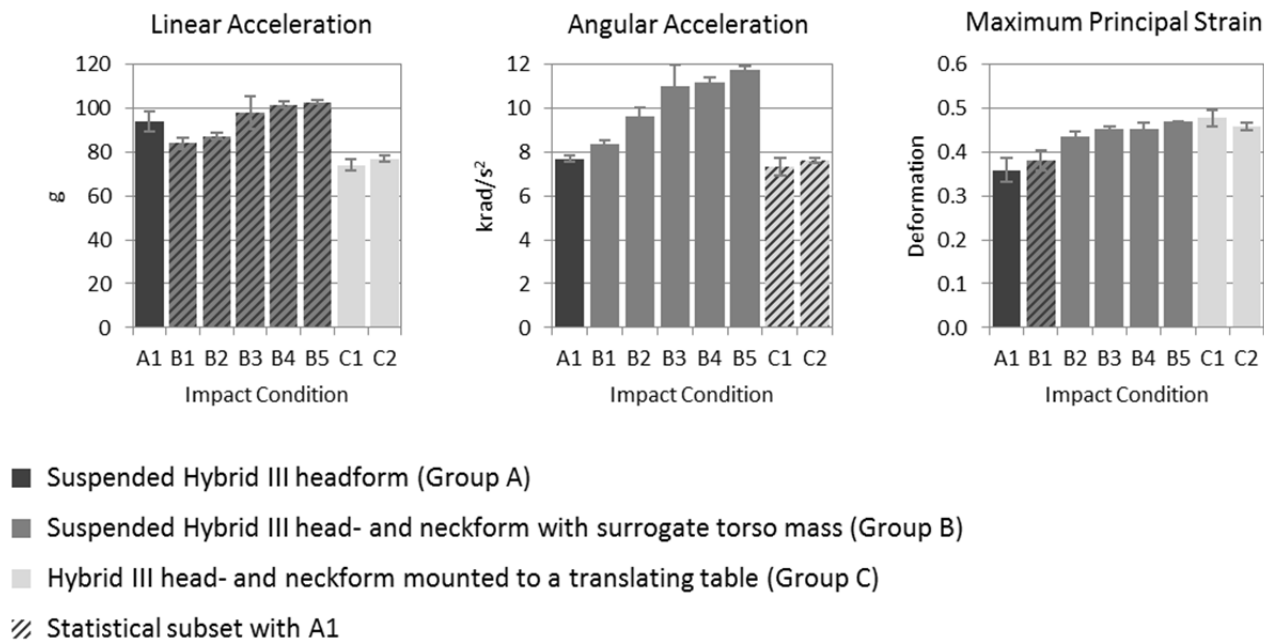
TABLE 2
UCDBTM MATERIAL PROPERTIES

Material	Young's Modulus (MPa)	Poisson's Ratio	Density (kg/m ³)
Scalp	16.7	0.42	1 000
Cortical Bone	15 000	0.22	2 000
Trabecular Bone	1 000	0.24	1 300
Dura Mater	31.5	0.45	1 130
Pia Mater	11.5	0.45	1 130
Falx, & Tentorium	31.5	0.45	1 140
Cerebrospinal Fluid	15 000	0.50	1 000
Grey & White Mater	Hyperelastic	0.49	1 040

III. RESULTS

Figure 1 illustrates the peaks and standard deviations for each of the dependent variables for each impact condition. Statistical subsets with, or values that were not considered to be significantly different from, the no neck condition (A1) are indicated with hashed bars. Significant differences in terms of peak linear acceleration were found between the suspended conditions, groups A and B, and the table conditions, group C. Further, significant differences in peak angular acceleration were found between the no neck condition, A1, and the other suspended conditions, group B. Finally, significant differences in maximum principal strain were found between the no neck condition (A1) and all but the suspended no torso impact condition (B1).

Figure 1: Peak and standard deviation resultant linear accelerations, angular accelerations, and predicted maximum principal strains.



IV. DISCUSSION

With the increasing occurrence of concussions in sports, there is a need for more accurate and realistic impact reconstruction protocols [3]. Representative boundary conditions applied to the head-neck system drastically change the kinematic loading of the head and therefore the potential for brain injury [5]. The current European helmet testing standard uses an untethered headform drop on to an anvil [15][16][17]. Conversely, the North American helmet test standard uses a headform attached to a neck form which is then attached to a monorail [1][6][7][9]. These helmet assessment testing protocols are developed to ensure that helmets meet a minimum standard for protection. The question is whether or not the impact condition using an untethered or tethered headform is a realistic reconstruction of an impact event. Recent research investigating concussive injury reconstructions have been using a linear impactor system to impact a headform attached to a neck and 20kg sliding table. This 20kg sliding table is said to mimic the torso mass of an impacted football player [35]. How this torso mass affects the dynamic response of the head and the resulting brain tissue response needs to be determined and compared to these current impact set-up conditions.

Helmet testing standards aim to protect against traumatic brain injuries and therefore use the correlated metrics of peak linear acceleration or a linear tolerance function, such as the Gadd Severity Index or Head Injury Criterion, to determine the efficacy of the protective equipment. While peak linear acceleration has been associated with catastrophic head and brain injuries [40], angular acceleration has been shown to be linked to more diffuse type injuries such as concussions. With the present push for more comprehensive traumatic brain injury protection, particularly against the prevalent risk of concussion [4][14][30][35], alternative tolerance criteria and impact reconstruction methodologies are being considered. While previous research has focused on the effects of impacting mass [27] and the Hybrid III neckform's bias [2], this study focused on the effects of different torso mass conditions operating through the fiftieth percentile neckform. The results of this study showed that peak linear acceleration, the typically measured indicator for risk of traumatic brain injury, was not sensitive enough to differentiate between any of the suspended headform conditions regardless of torso masses – as indicated by the lack of significant difference between the conditions. Impacts predicting the risk of purely catastrophic traumatic brain injuries provide reasonable measures with or without a torso mass.

Peak linear and angular acceleration have been associated with varying levels of concussive injury risk [29][43]. The results of this study show that the presence and quantity of torso mass may have significant effect in the prediction of concussive injury. This is apparent with the results of the peak angular and resulting MPS values within the brain for the conditions where torso mass was increased (B1 to B5). This supports previous injury reconstruction research [27][33], stressing the importance of torso mass of the impacting player. The

UCDBTM human head and brain finite element model predicted brain tissue deformations caused by each impact where only the response of the B1 condition (attached Hybrid III neckform but no torso mass) was not significantly different from A1 (the free headform). Injury reconstruction research has shown that the peak value of the dynamic response alone may not be the best indicator of brain tissue deformation. The sensitivity of the brain tissue deformation metric shows differences between impacting conditions which may be masked by peak linear and angular accelerations [28][37], among other curve characteristics.

In terms of prediction of risk of injury, the results of this study show that the headform condition with neck unattached may underestimate maximum principal strain for frontal impacts when compared to the other conditions. The unattached condition also had the highest proportional standard deviation for the maximum principal strain measures. Though peak kinematic response may be similar for pass/fail tests, it does not account for differences in the characteristics of the dynamic response loading curves, which have significant effects on brain tissue response. This may help explain the lower peak linear and angular acceleration response in the C1 & C2 conditions, which resulted in the highest MPS values of all impact conditions [28][37].

Limitations

The physical and computational modelling methods of head impacts, although some of the most advanced techniques currently available to assess risk of traumatic brain injury, are not without limitations. The Hybrid III anthropometric test device was originally developed for the measurement of loads imparted during automotive collisions [25][30]. However, it has been recently used for direct head impact reconstructions and is a good surrogate for human response. The UCDBTM finite element model has likewise been used to predict brain tissue deformations resulting from head impacts and has been accepted as a reliable method for determining risk of traumatic brain injury, but any results must be interpreted with the limitations associated with the dynamic tissue response to loading common to all biological finite element models.

V. CONCLUSION

The purpose of this study was to determine the effects of different torso mass of the impacted surrogate on the peak dynamic headform response and MPS of the brain tissue. The results of this study show that though peak linear acceleration was unable to differentiate between different torso masses, brain tissue response (MPS) could. This study confirms that peak acceleration may be underestimating actual brain tissue deformation values and thus may not be sensitive enough of a metric to predict MPS of brain tissue associated with risk for concussive injuries. While mass of the impactor has previously been confirmed to have significant effects on dynamic response and brain tissue response, this study shows that torso mass should also be considered for injury reconstruction methodologies.

VI. REFERENCES

- [1] American Standards of Testing and Materials. F1045-07: Standard Performance Specification for Ice Hockey Helmets. 2007, 10 pages.
- [2] Bartsch A, Benzel E, Miele V, Morr D, and Prakash V. Hybrid III anthropometric test device (ATD) response to head impacts and potential implications for athletic headgear testing. *Accidental Analysis and Prevention*, 2012, 48: 285-291.
- [3] Bishop PJ. Head protection in sport with particular application to ice hockey. *Ergonomics*, 1976, 19(4):451-464.
- [4] Boden BP, Tacchetti RL, Cantu RC, Knowles SB, and Mueller FO. Catastrophic head injuries in high school and college football players. *The American Journal of Sports Medicine*, 2007, 35(7):1075-1081.
- [5] Bourdet N and Willinger R. Coupled head-neck-torso and seat model for car seat optimization under rear-end impact. *Journal of Sound and Vibration*, 2008, 313:891-907.
- [6] Canadian Standards Association. Z262.1-09: Ice Hockey Helmets. 2009, 29 pages.
- [7] Canadian Standards Association. Z263.1-08: Recreational Alpine Skiing and Snowboarding Helmets. 2008, 28 pages.
- [8] Clauser CE, McConville JT, and Young JW. Weight, Volume and Center of Mass of Segments of the Human Body. *AMRL Technical Report*, 1969.
- [9] Consumer Product Safety Commission. 16 CFR Part 1203: Safety Standard for Bicycle Helmets; Final Rule. *Federal Register*, 1998, 63(46):11711-11747.

- [10]de Leva, P. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics*, 1996, 29(9):1223-1230.
- [11]Dempster WT, et al. The anthropometry of the manual work space for the seated subject. *American Journal of Physiology and Anthropology*, 1959, 17:289-317.
- [12]Deng Y. Anthropomorphic dummy neck modeling and injury considerations. *Accident Analysis & Prevention*, 1989, 21(1):85-100.
- [13]Doorly MC and Gilchrist MD. The use of accident reconstruction for the analysis of traumatic brain injury due to head impacts arising from falls. *Computer Methods in Biomechanics and Biomedical Engineering*, 2006, 9(6):371-377.
- [14]Duma SM, et al. Analysis of Real-time Head Accelerations in Collegiate Football Players. *Clinical Journal of Sports Medicine*, 2005, 15(1):3-8.
- [15]European Committee for Standardization. EN 1077: Helmets for Alpine Skiers and Snowboarders. 2007, 19 pages.
- [16]European Committee for Standardization. EN 1078: Helmets for Pedal Cyclists and for Users of Skateboards and Roller Skates. 2012.
- [17]European Committee for Standardization. Working Draft EN 10256: Head and Face Protection for Use in Ice Hockey. 2006, 65 pages.
- [18]Flik K, Lyman S, and Marx RG. American collegiate men's ice hockey: an analysis of injuries. *The American Journal of Sports Medicine*, 2005, 33(2):183-187.
- [19]Foreman SG. The dynamic impact response of a hybrid III head- and neckform under four neck orientations and three impact locations. *MSc dissertation, University of Ottawa, Canada*. 2011.
- [20]Foster J, Kortge J, and Wolanin M. Hybrid III – A biomechanically-based crash test dummy. *Society of Automotive Engineers*, 1994, Technical Paper 70938.
- [21]Goodman D, Gaetz M, and Meichenbaum D. Concussion in hockey: there is cause for concern. *Medicine & Science in Sports & Exercise*, 2001, 33(12):2004-2009.
- [22]Horgan T. A finite element model of the human head for use in the study of pedestrian accidents [Doctoral dissertation]. 2005. University College Dublin.
- [23]Horgan TJ & Gilchrist MD. The creation of three-dimensional finite element models for simulating head impact biomechanics. *International Journal of Crashworthiness*, 2003, 8(4):353-366.
- [24]Hoshizaki TB, Walsh ES, Post A and Gilchrist MD. The application of brain deformation values in assessing the safety performance of ice hockey helmets. *Proceedings of the Institute of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology*, 2012, 226: 11 pages.
- [25]Hubbard R & McLeod G. Definition and development of a crash test dummy head. *Proceedings of the 18th Stapp Car Crash Conference*. 1974: SAE Paper 741193.
- [26]International Standard Organization. ISO TC 83/SC 5N Working draft ISO/EN 10256 Head and face protection for use in ice hockey. 2006, 65 pages.
- [27]Karton C, Hoshizaki TB, & Gilchrist MD. The influence of inbound mass on the dynamic impact response of the Hybrid III headform and tissue deformation response characteristics. *Journal of ASTM international*. In Press.
- [28]Kendall M, Post A, Rousseau P, Oeur A, Gilchrist MD, and Hoshizaki TB. A comparison of dynamic impact response and brain deformation metrics of head impact reconstructions for three mechanisms of head injury in ice hockey. *Proceedings of IRCOBI Conference*, 2012, Dublin, IRL.
- [29]Kleiven S. Predictors for traumatic brain injuries evaluated through accident reconstruction. *Stapp Car Crash Journal*. 2007, 51:81-114.
- [30]McIntosh AS, et al. Sports helmets now and in the future. *British Journal of Sports Medicine*, 2011, 45:158-1265.
- [31]Mertz HJ. Biofidelity of the Hybrid III Head. 1985: SAE Paper 851245.
- [32]National Operating Committee on Standards for Athletic Equipment. Standard Performance Specification for Newly Manufactured Hockey Helmets: NOCSAE DOC 030-11m12. 2012, 12 pages.
- [33]Newman J, Barr C, Beusenbergh M and Whitnall C. A new biomechanical assessment of mild traumatic brain injury - Part 2 - results and conclusions. *Proceedings of IRCOBI Conference*, 2000, Montpellier, France.
- [34]Padagaonkar AJ, Krieger KW, and King AI. Measurement of angular acceleration of a rigid body using linear accelerometers. *Journal of Applied Mechanics*, 1975, 42:552-556.
- [35]Pellman EJ, Viano DC, Tucker AM, Casson IR, and Waeckerle JF. Concussion in professional football: reconstruction of game impacts and injuries. *Neurosurgery*, 2003, 53(4):799-814.

- [36] Pellman EJ, Viano DC, Withnall C, Shewchenko N, Bir CA, and Halstead D. Concussion in professional football: helmet testing to assess impact performance – part 11. *Neurosurgery*, 2006, 58(1):78-96.
- [37] Post A, Hoshizaki TB, and Gilchrist MD. Finite element analysis of the effect of loading curve shape on brain injury predictors. *Journal of Biomechanics*, 2012, 45:679-683.
- [38] Post A, Walsh ES, Hoshizaki, TB, and Gilchrist MD. Analysis of loading curve characteristics on the production of brain deformation metrics. *Proceedings of the Institute of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology*, 2012, 226: 8 pages.
- [39] Towner E, Dowswell T, Burkes M, Dickinson H, and Hayes M. Bicycle helmets: review of effectiveness. *UK Department for Transport*, 2002.
- [40] Viano D, King A, Melvin J, and Weber K. Injury biomechanics research: An essential element in prevention of trauma. *Journal of Biomechanics*, 1989, 22(5):403-417.
- [41] Walsh ES, et al. Dynamic impact response characteristics of a helmeted Hybrid III headform using a centric and non-centric impact protocol. *Proceedings of the Institute of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology*, 2012, 226:220-225.
- [42] Walsh ES, Rousseau P, and Hoshizaki TB. The influence of impact location and angle on the dynamic impact response of a hybrid III headform. *Sports Engineering*, 2011, 13(3):135-143.
- [43] Zhang L, Yang KH and King AI. A proposed injury threshold for mild traumatic brain injury. *Journal of Biomechanical Engineering*, 2004, 126:226-236.