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Development of a Biofidelic 2.5 kg Infant Dummy and Its Application to Assessing Infant Head Trauma During Violent Shaking

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This paper has not been screened for accuracy nor refereed by any body of scientific peers and should not be referenced in the open literature.

ABSTRACT

A small infant dummy has been developed, known as the Aprica 2.5 kg infant dummy. The basic dimensions of the dummy were obtained from direct measurements of newborns. The dummy is segmented according to conventional cut planes at the principal joints. The neck of the dummy has been designed to model the approximate stiffness of a newborn. Selected biomechanical testing has been conducted on the dummy to evaluate its response relative to scaled response corridors. The head response requires some stiffening, the neck response is within the scaled corridors, as is the abdomen impact response. The chest response requires a greater amount of damping.

The new, infant dummy has been used in a number of selected drop tests and shaking tests to evaluate head accelerations under inertial loading and impact loading. Violent shaking of the Aprica dummy generated much high angular acceleration than has been previously reported.

INTRODUCTION

The development of the Aprica 2.5 kg infant dummy was motivated by the need to have a more biofidelic and instrumented dummy which would represent a small infant for evaluating restraint systems. Aprica Childcare Institute had used the TNO P0 dummy for such evaluation, but it was felt that the dummy was unsuitable because:

1. Its weight was over the Aprica target of 2.5 kg, the weight of a 5% percentile Japenses newborn.
2. The neck was too stiff. It is known that the neck of the newborn is unstable and generally unable to support the head. The P0 dummy appears to model the instability of the neck by integrating a nearly unconstrained A-O joint in the dummy. The head pitches around the A-O joint easily but without any involvement of the neck.
3. The dummy is not instrumented.

Rangarajan [2002] has provided an overview of the development of the Aprica infant dummy.

DESIGN METHODOLOGY

At the beginning of the project, it was decided by Aprica and GESAC that the infant dummy structure should consist of the following segments:

1. Head
2. Neck
3. Thorax including spine and shoulder structure
4. Pelvis
5. Lower extremities
6. Upper extremities

This structure was chosen as it would make it easier to instrument and measure the response of various segments. In addition, this segmentation plan would make it possible to modify design of various parts of the dummy as appropriate and to replace parts that might be damaged during testing. Most important of all, this segmentation plan was chosen as it would lead to better biofidelity in the response to impact since the segmentation corresponded with the normal joint locations of the human infant.

Each of these segments consists of skeleton and flesh. The skeletal part of the dummy gives it structural strength and is used to mount sensors. In general, most skeletal parts of the dummy were machined from Delrin whose density is similar to the overall density of the human bone. Dummy flesh was mostly molded from Urethane. The Urethane flesh in the thorax was backed up by open cell foam to provide more biofidelic response.

Anthropometry

GESAC conducted a literature search to identify anthropometric data for a 2.5 kg infant. Preliminary anthropometric data were obtained from anthropometric specifications for a premature infant in CMVSS (Canadian Motor Vehicle Safety Standards) 213.5 test procedures [Transport Canada, 2000].

At the same time, pediatricians from Aprica measured the dimensions of several body segments of an infant weighing approximately 2.5 kg. These dimensions were compared to the segment dimensions published by the Japanese Ministry of Transport for a 10% infant of mass 2.5 kg. It was found that the measurements from the infants compared favorably with those of the 10% infant.

Mass scaling procedures commonly used in anthropometry [Mertz, 1989, Melvin, 1995] were used to adjust the dimensions of the infant data in CMVSS 213.5 test procedures and compared with the data developed by Aprica. It was found that the CMVSS 213.5 data compared well with data reported by Aprica. It was therefore decided to accept data provided by Aprica as design guidelines.

Data from four infants were averaged and it was found that the average was quite close to the data obtained from the first infant. It was felt that the sample size was too small to be statistically relevant. In addition, it was felt that it was better to make the dummy representative of at least one infant as the averaged data were close to that of the first infant measured.

These data are presented in Table 1. The first column lists the name of the body segment. Measurement obtained from the first infant is listed in column 3 (column heading “Original Data”) and column 4 (column heading "Final Data") lists the average measurements obtained from the next three infants measured. Design goal for each segment is provided in column 5 (column heading "Design Goal") and the final measurements obtained from the infant dummy delivered to Aprica are listed in column 6 (column heading "Measured Value"). The design goals were typically obtained by rounding the original data. In some instances when original data were not available, they were established from estimating from photographs of the infants.

Table 1. Aprica Infant Dummy Anthropometry

Parameter	**Units**	**Original Data**	**Final Data**	**Design Goal**	**Measured**
Mass	gm	2572	2603	2600	2600
Height	mm	450	440	450	450
Head circumference	mm	310	340	349	340
Head length	mm	118	118	118	120
Head width[1]	mm	88	88	95	94
Head depth[1]	mm	140	140	112	110
Neck circumference	mm	180	187	172	161
Neck length	mm	50 [2]		54	53
Shoulder circumference	mm	300	322	305	340
Width @ shoulders (center to center)	mm	135		115	115
Shoulder width	mm	120	128	140	140
Depth @ shoulders (sternum to spine)	mm			76	76
Chest circumference	mm	290	315	297	298
Width @ chest	mm			99	100
Depth @ chest	mm			81	74
Waist circumference	mm	310	323		318
Width @ waist	mm			95	114
Depth @ waist	mm			86	79
Hip circumference	mm	280	285	286	285
Hip breadth (distance between ball socket centers)	mm			56	56
Upper arm circumference	mm	80	93	80	80
Elbow circumference	mm	70	83	70	70
Wrist circumference	mm	50	63	50	50
Leg at hip circumference	mm	135	155	130	125
Circumference @ thigh	mm	130		130	115
Leg circumference at knee	mm	100	107	100	110
Leg at circumference at ankle	mm	60	75	60	60
Arm length (arm to tip of hand)	mm	180	183	180	180

Parameter	Units	Original Data	Final Data	Design Goal	Measured
Upper arm length	mm			69	69
Lower arm length	mm			66	66
Hand length	mm			45	38
Leg length (crotch to heel)	mm	150	152	150	133
Top of head to shoulder	mm	110	108	110	110
Leg length (knee to heel)	mm	70	92	90	85
Upper leg length	mm			79	79
Lower leg length	mm			76	67
Foot height (ankle to bottom of foot)	mm			14	15
Under foot length	mm	60	63	60	63
Head weight [3]	gm			800	772
Neck weight	gm			126	62
Torso (shoulder, thorax, pelvis)	gm			1273	1244
Upper arm weight	gm			29	39
Lower arm weight	gm			22	32
Upper leg weight	gm			82	79
Lower leg (w/ foot) weight	gm			48	73

Notes:

1. The final dimensions were adjusted based on measurements made on an infant CPR manikin
2. The design goal for neck length was suggested by the pediatric consultant at Aprica.
3. Design goal for segment weights were obtained by calculating the volume of the segments and multiplying the volume by a density of 1 gm/cc. Estimated segment weights were compared to generally accepted values in the literature. For instance, it is often estimated that in an infant, the weight of the head is about a third of the body weight. So, the estimate of 0.8 kg for a 2.5 kg dummy was accepted as being reasonable.

Biomechanical Response Requirements

There are very limited data available for impact response of the newborn under normal loading conditions. There are scattered data e.g. for the static load limits for neck fracture in tension [Melvin, 1995], and the mechanical properties of fetal cranial bone [McPherson and Kriewall, 1980]. Response requirements were developed based on a simple, theoretical scaling of the response expected for a 50th percentile adult male as described by Melvin. The scaling accounts for the change in the material strengths of tissue between the newborn and adult using two scales, one for calcaneal tendon tissue and one for skull bone. The requirements, specially for the neck, have to be treated with some degree caution, since they imply behavior of structures based on just these two types of materials. But we believe the scaled responses can be a useful guide to estimating response for a number of different impact conditions.

The basic scaling parameters are:

$S_E = E_{infant} / E_{50M}$
$S_l = l_{infant} / l_{50M}$
$S_m = m_{infant} / m_{50M}$

where: S_E = elastic modulus scale factor
E_{infant} = elastic modulus for infant tissue
E_{50M} = elastic modulus for 50th male tissue
S_l = length scale factor
l_{infant} = typical length for infant (e.g. head length, chest circumference, etc.)
l_{50M} = corresponding length for 50th male
S_m = mass scale factor
m_{infant} = typical mass for infant component (e.g. head mass)
m_{50M} = mass of corresponding component for 50th male

All variables of interest, e.g. displacement, force, velocity, etc. can be scaled using some combination of the above scale factors. If the density of the tissue is assumed the same in the infant and adult male, then the length and mass scale factors are usually related by the relation:
$S_m = S_l^{\,3}$

We opted to keep the mass and length scale factors independent in deriving the performance requirements for the dummy.

Instrumentation

The Aprica 2.5 kg infant dummy is instrumented with triaxial accelerometers at the head C.G., upper neck, lower neck, chest C.G., and pelvis. The accelerometers can be used to compute injury criteria such as the head injury criteria (HIC). During testing, additional accelerometers can be added to the arms and legs.

RESULTS

The assembled Aprica 2.5 kg infant dummy is pictured in Figure 1.

Figure 1: Aprica 2.5 Kg Infant Test Dummy With Instrumentation.

Biomechanical response requirements and test procedures to evaluate them were developed for the head, neck, thorax, and abdomen. The test procedures were based on scaling the requirements defined for the 50th percentile Thor dummy, developed for the National Highway Traffic Safety Administration (NHTSA) and described in the document: "Thor Certification Manual (Revision 2001.02)" [GESAC, 2001]. This document is available from the NHTSA website. In general, the impactor size, mass, and velocity used in the adult male dummy certification tests were scaled down appropriately. In order to simplify the test procedures, the tests for all the four components mentioned above were carried out using a pendulum device with a rigid arm and the various impactors attached to the base of the pendulum arm. In the case of the neck tests, the pendulum tests for the Thor neck were scaled down for the infant dummy. The response from preliminary tests conducted on the new infant dummy are described in the subsections below.

Head Impact Response

Two separate tests were performed to evaluate the biomechanical response of the head to impact. The first is a scaled version of the whole-body, head impact test that is used for the 50th percentile male Thor dummy. The second is a scaled version of the head drop test that is commonly used for dummies in the Hybrid III family. The responses from both test configurations are shown in Figures 2 and 3.

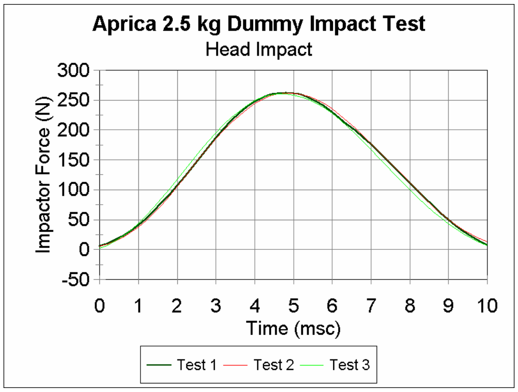

Figure 2: Response Of Head To Impact In Whole Body Testing

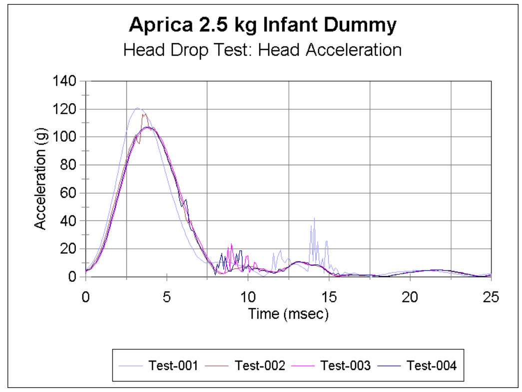

Figure 3: Response Of Head To Drop Testing

At the present time, there is uncertainty in defining the appropriate response one should expect from a head impact to the infant. The elastic modulus of the bones in the infant skull are significantly lower than an adult. In addition the bones are connected by soft cartilage, so that it is relatively easy to deform the skull under quasi-static loading. The problem arises in trying to quantify the effective stiffness of the skull during dynamic impact. The stiffness of the bones may not be very velocity sensitive, but the brain material, because it is viscoelastic, will stiffen significantly under rapid rates of loading. But the material has not been properly characterized, and Irwin and Mertz [1997] have argued that the model used by Melvin [1995] should be used. Duhaime, et al [1987] conducted a series of experiments with a soft doll head which was tightly filled with water-absorbed cotton. They impacted the head against a rigid surface, both with and without a stiff shell covering the head. They found that there was no significant difference in the head accelerations for the two configurations. A possible explanation of this would be if the water-soaked cotton acted as an incompressible body, leading to a high stiffness comparable to the stiffness of the stiff shell.

If we assume that the adult response should be scaled using the scaling factors for the geometry and the elastic modulus, then for the head impact response (shown in Figure 2) we would expect the peak force to be in the range 450-525 N. Our peak of 250 N would indicate that the skull/flesh stiffness is too low at the present time. Similarly for the head drop test, the expected range is 212-260 g, while our peak of 105 g is also too low. Both tests indicate a similar ratio between expected stiffness and observed stiffness. The results do show that the responses in both configurations are very repeatable.

Neck Dynamic Response

Some preliminary data have been obtained on the neck response of the new infant dummy by attaching the head and neck to the end of a pendulum and allowing the pendulum to impact a deformable stop. This is similar in concept to the standard head/neck pendulum used for calibrating the adult male necks [GESAC, 2001]. The response is measured in terms of moment acting on the head as a function of the angle of the head relative to the base of the neck. Currently there is no load cell at the head/neck joint (occipital condyle), and the moment is measured indirectly from the acceleration of the head. The response of the Aprica infant dummy neck in flexion is shown in Figure 4. The angle was measured by digitizing the high speed video of the event.

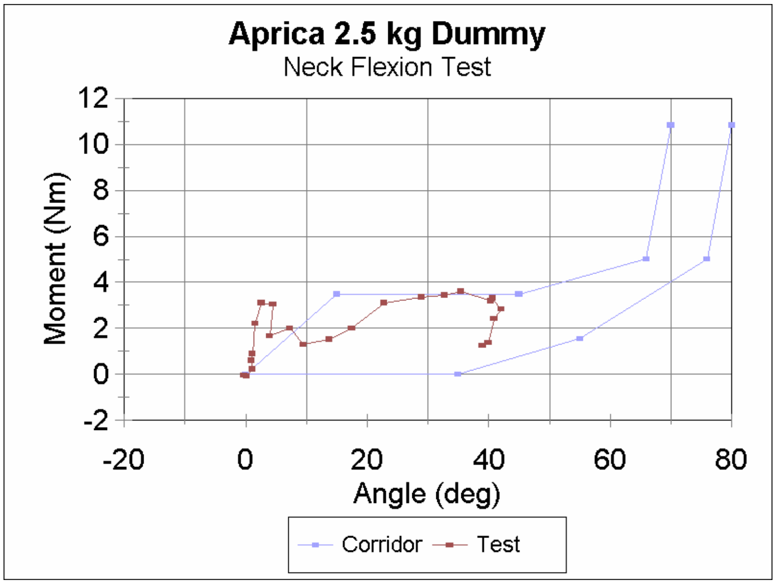

Figure 4: Moment Vs. Angle Response Of Infant Dummy Neck In Frontal Flexion

Thorax and Abdomen Impact Responses

A scaled version of the standard Kroell test was used to design an impactor to evaluate the chest impact response of the infant dummy. The scaling was similar to the procedure described by Ratingen [1997]. The scaled impactor had a diameter of 50.3 mm, a mass of 1.2 kg, and the impact speed was 3.3 m/s. The scaling used a scale factor of .6 for the elastic modulus which was derived from the scaling suggested by Melvin for the calcaneal tendon [1995].

In a similar manner, the abdomen impact procedure was scaled from the one developed by Cavanaugh for adults [1986]. A rigid rod is used as an impactor, with a diameter of 9.4 mm, a mass of 1.5 kg, and an impact speed of 4.7 m/s. The length of the rod was made so as to engage the width of the dummy abdomen.

In both these cases, the response is usually measured as a force-deflection function. Currently there is no deflection measurement instrument for the infant dummy and deflections were measured by digitizing the high speed video. The time histories of the impact force for the chest and abdomen impacts are shown in Figures 5 and 6.

From scaling the Kroell response, the expected peak force should in the range of 170-210 N, and the expected peak deflection should be in the range 14-17 mm. The current test produced a peak force of 350 N, and a deflection of 21 mm. The response indicated that there was not enough damping in the chest system. The initial impact energy was not used up quickly enough leading to the higher, final deflection and corresponding force. For the abdomen, the expected force at 28 mm of deflection is in the range: 187-230N, while the test result produced 190N. For the abdomen, the force response appeared to be reasonably biofidelic.

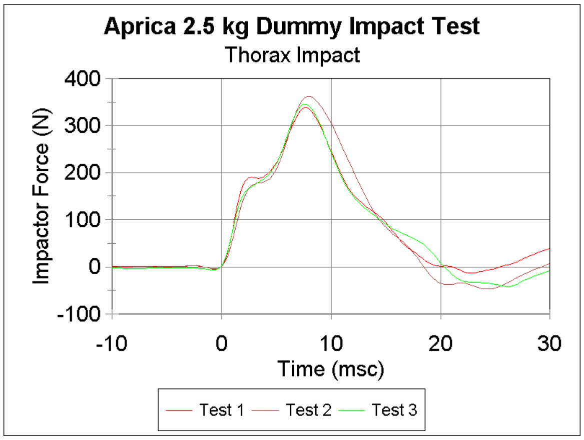

Figure 5: Force Vs Time Response Of Impactor During Chest Impact

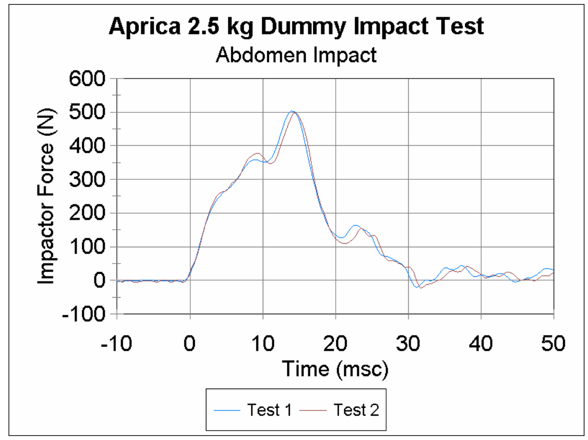

Figure 6: Force Vs. Time Response Of Impactor For Rod Impact To Abdomen

SHAKING AND IMPACT EXPERIMENTS

The Aprica 2.5 dummy was used to perform a preliminary study of estimating biomechanical parameters in events which may result in abusive and accidental infant injuries, including violent shaking, violent shaking followed by forceful slamming of the infant to various surfaces, and falls from various heights.

A 50^{th} percentile Japanese adult male was recruited to shake and slam the dummy. Tangential acceleration was measured using internal accelerometers at the head center of gravity and at the top of the head. Angular acceleration and velocity were measured using high-speed video. Each time, the dummy was shaken for 4 seconds. Each scenario (shaking, shaking plus slamming, falling) was repeated five times.

Table 2 shows the maximum and mean peak resultant linear accelerations found in various events. The maximum peak acceleration is the maximum value of the peak accelerations obtained from all the tests of the same type, and the mean peak acceleration is the average of the peak accelerations obtained from each set of tests. Figure 7 shows the linear acceleration at impact of head of dummy from various heights to a concrete surface.

Table 2. Linear Acceleration Recorded During Events

	Maximum linear acceleration		**Mean linear acceleration**	
	Head center of gravity	***Top of head***	***Head center of gravity***	***Top of head***
Violent shaking alone	27.7 g	67.8 g	26.2 g	64.8
Violent shaking followed by slamming to thin carpet over wood floor	368.7 g		268.3 g	
Violent shaking followed by slamming to sofa	136.3 g		103.4 g	
No shaking, slamming to a tatami mat	510.3 g		433.0 g	
Dropped from chest level when carrier stumbles when walking	333.0 g		281.1 g	
Rolls off sofa	95.5 g		90.9 g	

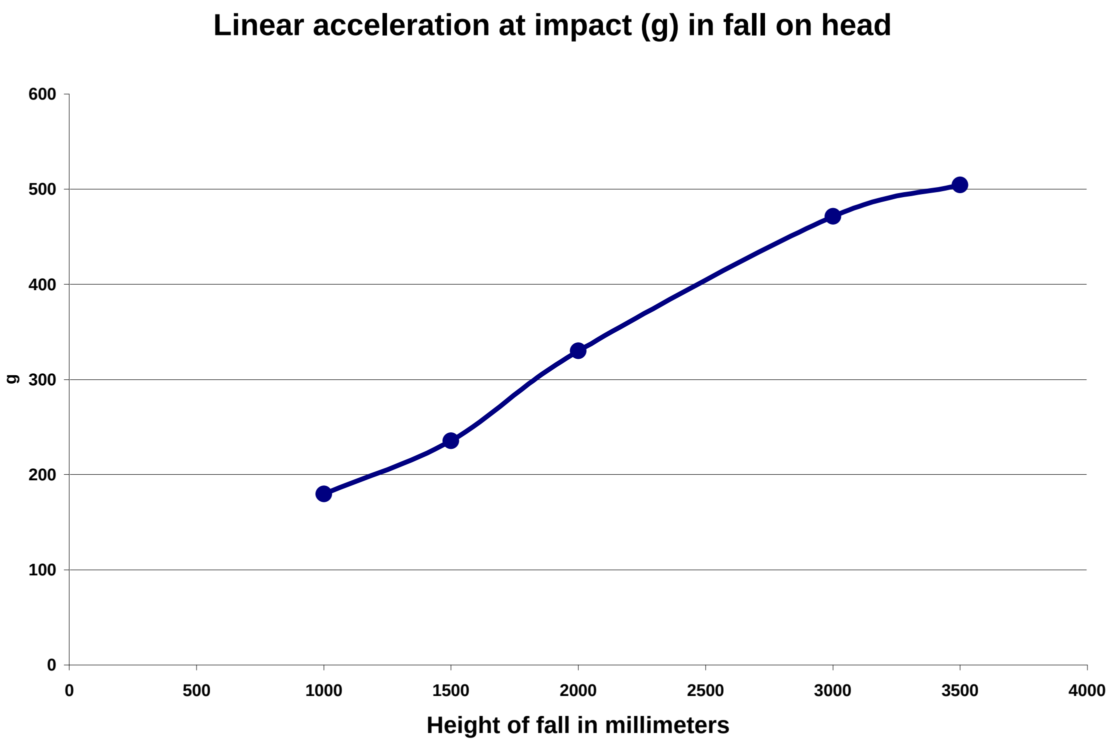

Figure 7: Linear Acceleration At Impact (G) Falling From Various Heights.

During shaking events, the dummy was shaken four to five times per second. The average angular excursion of the head was 88° forward and 133° to the back (relative to the base of the neck). The maximum total angular excursion of the head relative to the neck experienced during the five shaking episodes was 248°.

The angular motion of the head of the infant dummy was obtained by digitizing selected markers on the dummy head from a high speed video of the event. A typical frame at the time of maximum extension of the head is shown in Figure 8. The digitized was averaged using a moving average of five points and filtered at CFC 60. The resulting data was differentiated filtered again at

CFC 60 to estimate the angular velocity. Finally, the angular velocity was differentiated once more and filtered to estimate the angular acceleration.

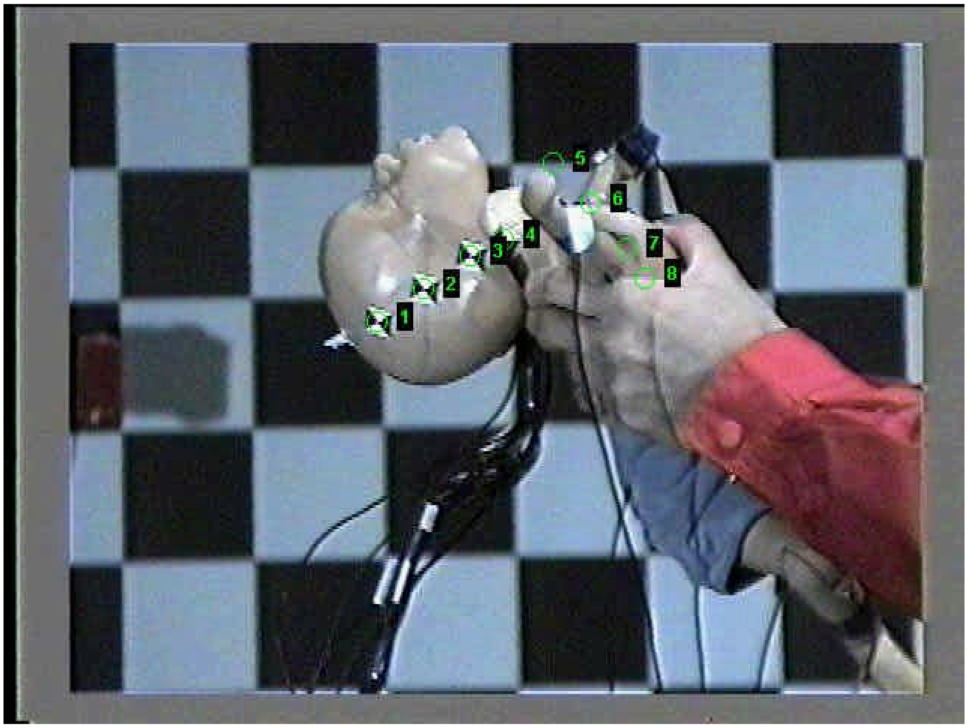

Figure 8: Typical Frame Used In Estimating Angular Motion

The maximum peak-to-peak angular velocity and acceleration obtained from digitizing the angular motion data during the violent shaking tests are shown in Figure 9. The highest angular acceleration obtained was 13,252 radians/sec^2 at an angular velocity of 153 radians/sec.

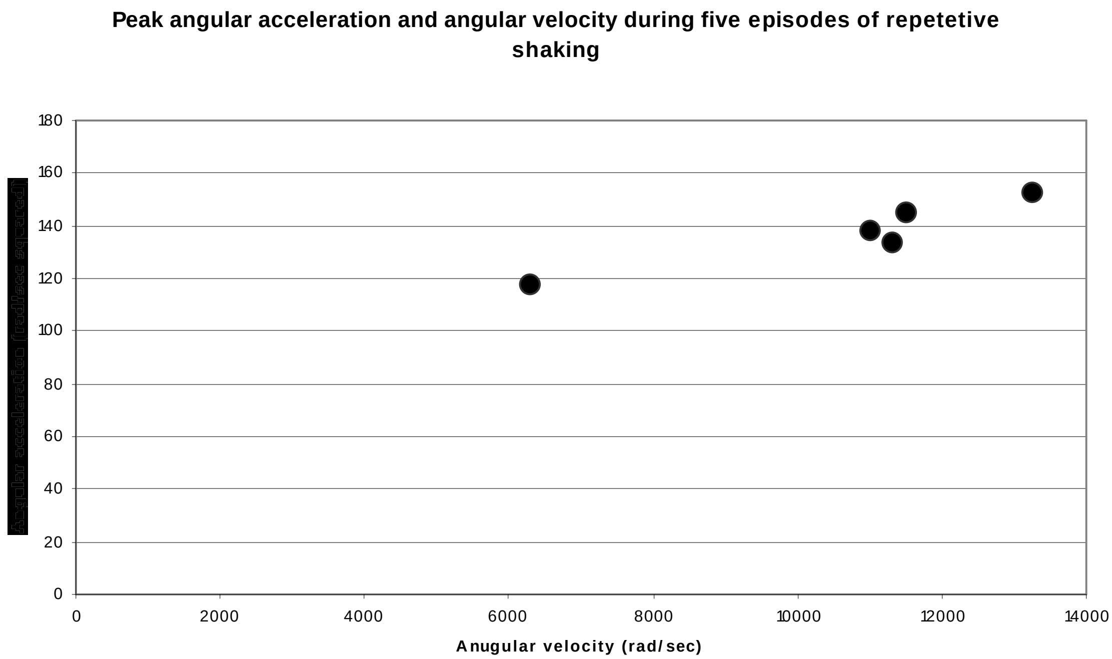

Figure 9: Maximum Angular Acceleration As A Function Of Maximum Angular Velocity

The linear acceleration data from the accelerometers mounted at the CG of the head, top of the head, and just below the O.C. are being currently analyzed to estimate the angular motion from the accelerometer data. These data will provide an independent estimate of the angular motion and wll be compared to the data obtained from digitizing to establish consistency of the digitizing method.

CONCLUSIONS

A new infant dummy has been designed and developed. The dummy anthropometry corresponds to a 10th percentile Japanese infant with a mass of 2.5 kg. The dummy is segmented in a human-like manner with realistic ranges of motion at the joints and with head, neck, thorax, and abdomen stiffnesses designed to correspond to values scaled from a 50th percentile male dummy. The procedure for scaling includes both the scaling for geometric size and the scaling for material properties. Test procedures have been defined to evaluate the biofidelity of the dummy. These tests have been scaled from the corresponding tests conducted on the 50th percentile male Thor dummy. The preliminary results from these tests indicate that the stiffness of the head may be too low, though further research is needed to determine the appropriate stiffness range of the human infant. The neck response indicates it is in the lower range of the scaled Mertz corridor. The chest impact response indicates that greater damping is required, while the abdomen response appears to be within the biofidelity requirements.

Since the dummy neck and spine stiffness have been made to correspond approximately to the expected stiffness of an infant, the dummy can be used in crash testing of infant restraint systems.

In a special study, the dummy was used to determine the peak angular accelerations of the head experienced during violent shaking. The angular acceleration obtained from digitizing the high speed video was substantially greater than that previously reported by Duhaime and colleagues [Duhaime, et al., 1987].

ACKNOWLEDGEMENTS

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DISCUSSION

PAPER: **A Biomechanical Model of Abusive Infant Head Trauma**

PRESENTER: ***Tariq Shams, GESAC***

Carole Jenny, Brown University

QUESTION: *Guy Nushultz, Daimler/Chrysler*
Just a quick question. When you scale, you don't just scale for size which I think that's what you're alluding to. You have to scale for physiological and, as well as material properties.

ANSWER: Right.

Q: And if you did that and you're still coming up with higher angular acceleration?

A: Yes. The–As far as the neck properties, which, because the person was holding the thorax rigidly so it's basically what the neck stiffness is about. That was scaled by scaling both the material elastic modulus, assuming that the ligament strength could be scaled.

Q: Aren't we talking head here? Aren't we talking about the head?

A: No. Okay. The rate at which the head is moving depends on the strength because the neck was the moving part, because the chest–the thorax was held rigid.

Q: But the injury parameters were related–

A: Yes.

Q: So it's gonna be scaled by three factors, not just one.

A: Right. And, the head characteristics–because this was non-impact-was that the mass properties, the dimensions of the head, the C9 location and the effective moments of inertia were scaled. There it's basically–because–

Q: So, you just did geometric scaling?

A: The head–Stiffness was scaled using scaling of the elastic modules, as well. So if you had impact of the head with the hard surface, we expect the accelerations that are generated will–

Q: So, I understand you did two scalings instead of three.

A: Yes.

Q: Okay. There's a physiological scaling which you have–which she was alluding to before which has to do with the way the physiology of the brain is set up.

A: Right, because this is a rigid structure so–

Q: But the head is not a rigid structure.

A: The head is–The mechanical head is a rigid structure.

Q: Okay.

A: So, we do not have any internal moving elements within the head.

Q: Okay. I realize that, but I–I'll talk to you later.

A: But that's the stiffness of the head scalp, skull is the scale that's–

Q: Your scaling the dummy, but we're talking about the–

A: Scaling of the elastic modulars as the–

Q: We're talking about the injury criteria.

A: Right.

Q: So, you've got all three scalings in there?

A: Yes.

Q: Physiological?

A: Yes.

Q: Material?

A: Yes.

Q: And geometric, and you still come up with a higher injury criteria.

A: Right.

Q: For the head.

A: Yes. So, our angular accelerations are very high.

Q: *John Melvin, Tandelta*
It's well known that the adult brain is well-protected from translational acceleration by the incompressibility of the brain and the fact that the skull is rigid; and therefore, the CSF can't go anywhere during an impact. Now, I've always felt that one of the problems in the Shaken Baby Syndrome is that with the fontanelles and such, you don't have a rigid container. And therefore, translational accelerations can produce very large brain motions that we don't see in adults, and that really has to be comprehended in understanding this problem. Basically it makes it easier to cause these large motions, and I suspect that's what's contributing to some of the eye injuries. It's just stretching the optic nerve because the brain is really moving around in there.

A: Actually, one of the theories about the eye injury is that infant vitreous is very adherent to the retina whereas in the adult it's not, and that there may be more traction on the retina than you would get in an adult because of that adherence. Nobody's proven that. That was just a theory.

Q: But, it doesn't take much in the way of volume change to really release the brain into violent motion. We've seen that in finite element models when you don't use a very high ratio. So, that really needs to be explored and really, I suppose a finite element model is really the only way to study this in the long run. But, it's probably a big effect.