DESIGN OF A BIOFIDELIC INSTRUMENTED 3.4 KG INFANT DUMMY

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ABSTRACT

A new infant dummy has been designed, manufactured, and tested, representing an average newborn of mass 3.4 kg. This dummy is a successor of the 2.5 kg newborn dummy developed earlier which had represent at 10th percentile Japanese newborn. Gross data such as total weight, length, and head circumference were taken from several sources including the Centers for Disease Control [CDC, 2000]. More detailed measurements were obtained from newborns in two Japanese clinics. Dynamic response data for head, neck, thorax, and abdomen were defined by scaling adult data. The dummy has 11 segments (head, neck, torso, upper arm, lower arm and hand, upper leg, lower leg and foot). The torso is further divided into shoulder, chest, abdomen, and pelvis, all connected to a flexible spine. Segments are connected by joints which provide human like range of motion. The dummy is instrumented with 26 sensors, including triaxial accelerometers at the head CG, upper and lower neck, thorax CG and pelvis CG; 3-axis angular velocity sensor in the head; uniaxial load cells in the neck and lumbar spine; string potentiometer to measure chest deflection; and five force sensors on the abdomen. This paper describes the methodology used to develop the design and the results from biofidelity testing.

INTRODUCTION

In late 2000, Aprica Childcare Institute funded GESAC, Inc to design and develop a new, small, infant dummy, which is now known as the Aprica 2.5 kg infant dummy. The development of the 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 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 Japanese 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 was not instrumented.

An overview of the 2.5 kg infant dummy is given by Rangarajan [2002]. Following the successful testing of the 2.5 kg infant dummy, Aprica funded GESAC to modify the design of 2.5 kg infant dummy to represent a 50th percentile newborn with a mass of about 3.4 kg. The objective of the new dummy, apart from representing a wider range of infants, was to improve its biofidelity and also to enhance its instrumentation.

ANTHROPOMETRIC DATA

The anthropometric data of an average infant were collected from several sources including the CDC for the overall height, weight and head size, the Hirokawa and Nishikawa Clinics for measurements on actual subjects, and a 1975 report on anthropometry of US infants and children [UMTRI, 1975]. Scaling from data of the next closest age group was also employed to determine values of certain dimensions when measured data were not available from existing sources. Some major target dimensions of the 3.4 kg infant dummy are listed in Table 1. Mass distribution of the dummy is listed in Table 2. The actual values are also listed in the two tables for comparison.

Table 1.
Key target and actual dimensions of the 3.4 kg infant dummy.

Parameters	Target (mm)	Actual (mm)
Overall height	504	530
Upper arm length (shoulder to	77	94
Lower arm length (elbow to wrist)	74	70
Hand length	62	38*
Upper leg length (hip to knee)	89	115
Lower leg length (knee to ankle)	86	95
Leg length (knee to heel)	120	123
Foot length	76	67
Foot breadth	32	38
Mid-thigh circumference	160	170
Leg at knee circumference	135	130
Calf circumference	125	120
Ankle circumference	87	90
Head circumference	348	352
Head depth	117	115
Head breadth	94	98
Head height (head top to chin)	130	108
Neck length (OC to C7/T1)	60	66
Mid-neck circumference	200	195
Shoulder breadth	158	160
Shoulder to crotch length	218	225
Shoulder circumference	360	310
Chest breadth	109	106
Chest depth	84	74
Chest circumference	340	310
Waist breadth	105	106
Waist depth	90	91
Waist circumference	335	334
Hip breadth	113	115
Hip circumference	338	295
Spine length (T1 to L5)	179	180

* dummy's hands folded

Table 2.
Target and actual mass distribution of the 3.4 kg infant dummy.

Segment mass	Target (g)	Actual (g)
Head	1067	999
Neck	168	79
Torso (chest, abdomen, pelvis)	1229	1204
Upper arm	77	83
Lower arm and hand	59	92
Upper leg	206	192
Lower leg and foot	126	152
Total mass	3400	3320

INFANT DUMMY DESIGN

Figure 1 shows a picture of the 3.4 kg infant dummy in its sitting posture with instrumentation wires at the side. The dummy consists of 11 segments, i.e., the head, neck, torso (which includes chest, abdomen and pelvis), upper arms, lower arms with hands, upper legs, and lower legs with feet.

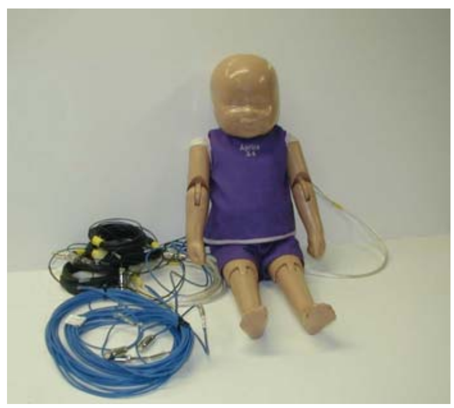

Figure 1. The 3.4 kg infant dummy with instrumentation wires.

The head has an inner aluminum housing which provides room for instrumentation of the head. The metal housing is then covered by a specially formulated urethane skin. Figure 2 shows the metal housing and the cap of the head.

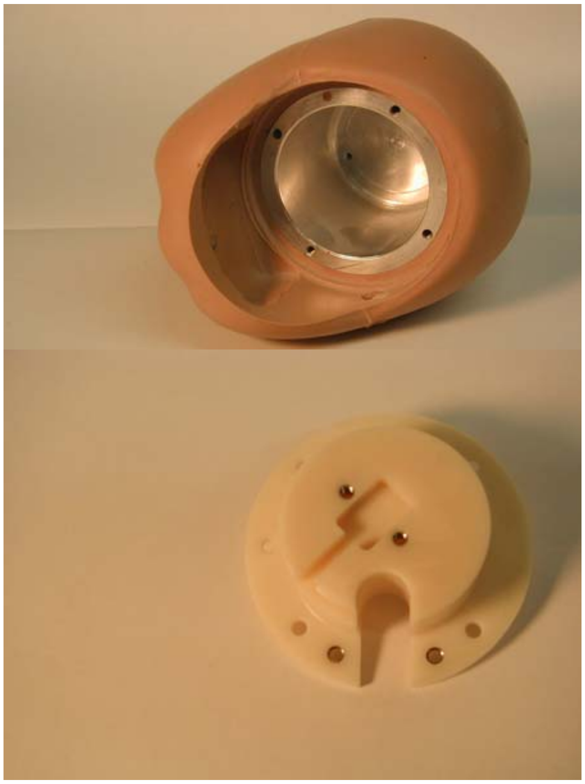

Figure 2. Metal housing and cap of the infant head.

The spine structure consists of the neck, thoracic spine, and lumbar spine. The shoulder and pelvis assemblies are integrated within the spine structure, along with instrumentation, as shown in Figure 3. The neck is connected to the head at the Occipital Condyle (OC) with a pin joint. The neck is molded as a urethane column with reinforcement at the center. Its lower end is attached to the shoulder block through a uniaxial load cell.

Figure 3. Neck-spine assembly with shoulder, chest block and pelvis. Wires are of two uniaxial load cells and a string potentiometer.

The shoulder block is machined of Delrin plastic. Two spherical ends made of aluminum serve as the shoulder joints and are attached to it. The shoulder block is fixed to the chest block which houses a string potentiometer.

Between the chest block and the pelvis there is a flexible lumbar joint. The lumbar joint is molded as a urethane column with reinforcement at the center. A uniaxial load cell is also attached to the lower end of the lumbar joint.

The pelvis is also made of Delrin and sits under the lumbar load cell. The hip joints are attached under the pelvis block. Like the shoulder joints, these consists of spherical ends made of aluminum.

The upper leg is connected to the pelvis by the ball and socket hip joint. The friction of the joint can be adjusted by tightening or loosening a screw attached to the joint for this purpose. The friction can be set to the standard 1G level by this method. The lower leg is connected to the upper leg by a pin joint. Its friction can also be adjusted. The femur and tibia bones are made of ABS plastic. A specially formulated layer of Urethane is molded outside the bones to simulate the flesh and skin.

The upper arm is connected to the shoulder by a ball and socket joint. The friction of the joint can be adjusted by tightening or loosening a screw attached to the joint for this purpose. The lower arm is connected to the upper arm by a pin joint. Its friction can also be adjusted. The humerus and forearm bones are made of ABS plastic. A specially formulated layer of Urethane is molded outside the bones to simulate the flesh and skin.

Figure 4. A prototype of the rib cage.

The rib cage is made of a polycarbonate outer shell lined with strips of damping material. An aluminum mass is attached to the front of the cage to simulate the sternum. Figure 4 shows a prototype of the rib cage (the final design is dimensionally different from the one shown here, but structurally the same). The actual rib cage can be seen in Figure 5.

The torso flesh/skin is molded as one piece of specially formulated Urethane. A slit runs along the center of the back of the flesh/skin to provide access to internal parts. A zipper is sown to the flesh/skin at the slit. Figure 5 shows the torso and its inside structure.

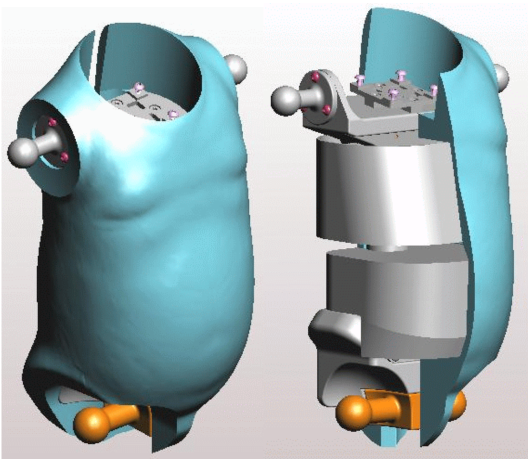

Figure 5. The torso of the dummy.

The abdomen of the dummy is filled with a piece of foam with a stiffness that is tuned to meet the abdomen biofidelity requirements.

INSTRUMENTATION OF THE INFANT DUMMY

The goal of the instrumentation with this dummy is to provide as many channels as possible for various testing conditions. This dummy is expected to be used in the evaluations of child restraint system, comfort of strollers, as well as in the study of shaken baby syndrome. A deliberate design effort was made to meet this goal. The 3.4 kg infant dummy is able to measure a maximum 26 channels of data. A full list of the instrumentation is shown in Table 3.

Table 3.
Instrumentation of the 3.4 kg infant dummy.

Location	**Sensor type**	**Channels**
Head CG	Angular velocity	3
Head CG	Accelerometer	3
Upper neck	Accelerometer	3
Lower neck	Accelerometer	3
Lower neck	Load cell	1
Thorax CG	Accelerometer	3
Thorax	String potentiometer	1
Lower lumbar end	Load cell	1
Pelvis CG	Accelerometer	3
Abdomen	Flexible Force	5

Angular acceleration is assumed to be an important dynamic variable which correlates with possible injury to the head [Duhaime, et al, 1987], and may be more important than simple linear acceleration. Thus the capability of measuring angular accelerations was considered to be an important instrumentation requirement. The angular velocity sensor model ARS-06S by ATA Sensors was used. The workable frequency range of these sensors is 0.4 Hz ~ 1.0 kHz, which is deemed suitable for most impact and shaken testing. Figure 6 shows the triaxial angular sensor mounted inside the head housing.

Figure 6. Triaxial angular velocity sensor housed inside the head.

The accelerometer model ASM-200BA by Kyowa Electronic Instruments was used for all acceleration measurements. This model is compact, has a capacity of 200 g, and offers high resolution. It is suitable for all testing conditions currently planned for the infant dummy.

The uniaxial load cell model 6398 by Robert A. Denton, Inc. was used for the axial load measurements at the lower neck and the lower lumbar spine. This load cell has a capacity of 200 lbf (890N) and a compact size (35.0 x 35.0 x 7.6 mm). It was introduced to provide additional information on tension and compression forces at the top and bottom of the spine.

The displacement of the chest when chest compression occurs is also of interest in evaluating thoracic injury. A string potentiometer is installed at the position of about the 6th/7th thoracic spine to provide an estimate of chest compression. One end of the string is attached to the spine and the other end is connected to the sternum mass. The compact string potentiometer (model 174-0321TR) made by SpaceAge Control was used. It has a maximum travel of 1.5" (38 mm), and is small enough to be housed within the thoracic spine.

In testing the child seat, it is also desirable to know the forces applied to the abdomen area of the dummy due to belt or other interactions. Since the abdomen is very flexible and has very limited space for instrumentation, a structurally flexible measuring mechanism is desirable. To meet this requirement, FlexiForce sensor model A201 by Tekscan was used. These sensors are paper-thin, flexible, and have a capacity of 1~1000 lbf (4.4~4450 N). Five of these sensors of 100 lbf (445 N) capacity were affixed to a piece of cloth with an area corresponding to the face of the abdomen. This set of sensors is put between the molded flesh and the abdomen foam when it is necessary to measure the force sustained by the abdomen. The set of sensors can be taken out of the dummy when the measurement is not required. Figure 7 shows the set of FlexiForce sensors.

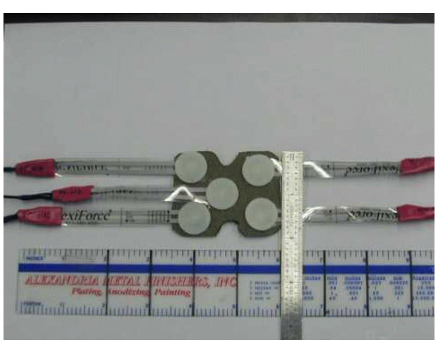

Figure 7. A set of 5 FlexiForce sensors used for abdomen force measurement.

BIOMECHANICAL REQUIREMENTS

Scaling

Since there are only very limited test data on newborn infants, scaling was employed to define the biomechanical requirements for the newborn. The biomechanical requirements for the 50th percentile male dummy were scaled using the following scaling factors to obtain the requirements for the newborn.

Three basic scaling factors:

$$\lambda_m = m_s / m_p \quad (1).$$

$$\lambda_\rho = \rho_s / \rho_p \quad (2).$$

$$\lambda_E = E_s / E_p \quad (3).$$

where, λ is the scaling constant, m the mass, ρ the density, and E the modulus of elasticity, respectively. The subscript s refers to scaled data, and p to prototype or standard data. In this study, the mass density is assumed to be the same for adults and newborns, therefore, λ_ρ=1.0.

Other scaling factors can be obtained by using their mathematical relationships as follows.

Scaling factor for velocity:

$$\lambda_V = \sqrt{\lambda_E} \quad (4).$$

Scaling factor for time:

$$\lambda_T = \lambda_L / \sqrt{\lambda_E} \quad (5).$$

Scaling factor for acceleration:

$$\lambda_a = \lambda_E / \lambda_L = \lambda_E / \sqrt[3]{\lambda_m} \quad (6).$$

Scaling factor for force:

$$\lambda_F = \lambda_L^2 \lambda_E = \lambda_m^{2/3} \lambda_E \quad (7).$$

where λ_L is the length ratio between the scaled and prototype objects.

For the 3.4 kg infant dummy, the following biofidelity tests were defined: the head impact, the head drop test, thorax impact test, the abdomen impact test, and the neck pendulum test. The response requirements for these tests were scaled from the corresponding responses in the adult 50th perecentile dummy.

Head Impact Requirement

Head impact requirement was scaled from the 50th percentile male dummy requirement [Hodgson, et al., 1975]. Based on the scaling procedure, the impact at the forehead of the 3.4 kg dummy should be performed at a speed of 0.8 m/s using a rigid impactor of 6.04 kg. The impactor head has a shape of an oval of 50 mm x 75 mm. The scaled corridor for the force-time curve is shown in Figure 8.

In obtaining the scaling factor, λ_E, of the human skull, data from McPherson, et al., quoted by Melvin [1995] and Thibault et al. [1999] were averaged. Both studies had limitations in regard of sample sizes.

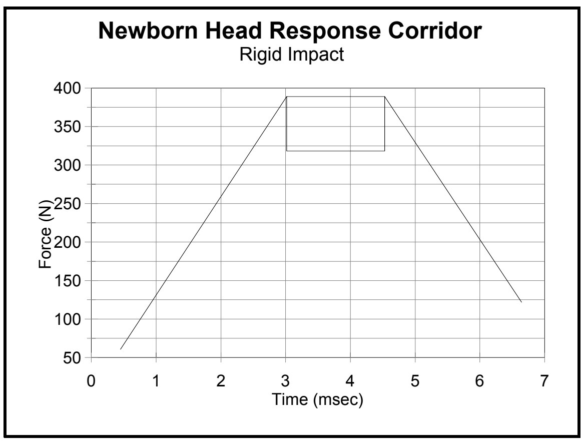

Figure 8. Biomechanical corridor for infant head impact.

Head Drop Test Requirement

Head drop requirement for the newborn was scaled from the 50th percentile male corridor [Hubbard, et al, 1974]. Using the scaling factors mentioned above, the head drop test of the 3.4 kg newborn should be performed at a height of 376 mm and should produce a peak resultant acceleration of 132~162 g.

A new study based on a small sample size by Prange et al. [2004] at Duke University has indicated lower peak acceleration for this test. These data can be incorporated into developing a new corridor in the future when more data are available. For the design of the current 3.4 infant dummy, the scaled corridor was used as a design target.

Thorax Impact Requirement

The thorax requirement for the 3.4 kg infant dummy was scaled from that for the 50th percentile male dummy [Neathery, 1974; Ratingen, et al, 1997]. Based on the aforementioned scaling method, the impact test for the 3.4 kg dummy thorax should be performed at a velocity of 3.3 m/s using a rigid impactor of diameter 50 mm with a mass of 1.1 kg. The force-deflection curve of the dummy at these conditions should be within the corridor shown in Figure 9.

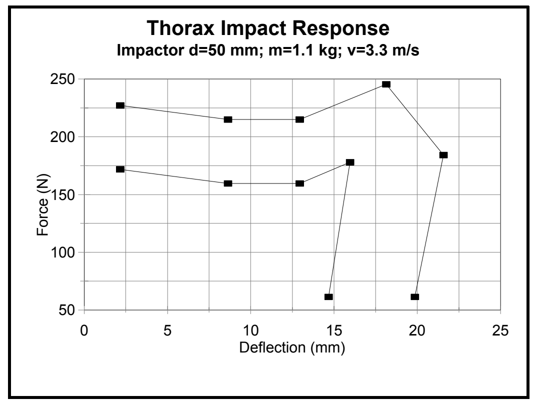

Figure 9. Biomechanical corridor for infant thorax impact.

Abdomen Impact Requirement

The biomechanical corridor for the abdomen impact of the 3.4 kg dummy was scaled from that of the 50th percentile male dummy [Cavanaugh, et al, 1986; Hardy, et al, 2001]. Based on the scaling method, the certification for the 3.4 kg dummy should be performed at a velocity of 4.7 m/s using a rigid rod impactor of diameter 10 mm with a mass of 1.4 kg. The force-deflection curve under these conditions should be within the corridor shown in Figure 10.

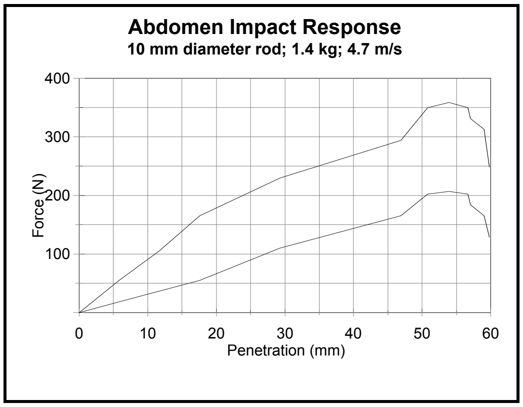

Figure 10. Biomechanical corridor for the infant abdomen impact.

Neck Pendulum Test Requirement

There are two kinds of requirements for the neck pendulum test, that is, the dynamic and kinematic requirements [Mertz, et al, 1971, Patrick, et al, 1976]. The dynamic requirement is characterized by a curve of the moment about the OC joint vs. the head rotation angle, and the kinematic requirement is a time history of head angle. Since the 3.4 kg infant dummy is not sufficiently instrumented to measure the OC moment, the current dummy design effort only focused on meeting the kinematic requirement.

The scaled kinematic requirement for the neck under the test pulses (see Figures 21 & 22) are shown in Table 4.

Table 4.
Kinematic requirement for neck pendulum test under the specified pulses.

Test	Peak angle (deg)	Peak time (ms)
Frontal	65 ~ 79	68 ~ 83
Lateral	36 ~ 46	63 ~ 77
Extension	-73 ~ -89	77 ~ 94

DISCUSSIONS

Anthropometric Resemblance

Comparing the target anthropometric data with the actual data of the dummy listed in Tables 1 and 2, it is clear that the design basically achieves its goal to produce an anthropometrically human-like dummy. It provides a successful prototype for future improvement. Some dimensions and mass distributions can be further fine-tuned in later modifications. Since some target values are either estimates or scaled values from other age groups, the small difference between the target and actual values is deemed insignificant.

Biofidelity Certification Tests

The biofidelity tests described above, were performed to examine the performance of the designed infant dummy. Before the tests, the dummy was soaked in a temperature controlled room for 24 hours. The temperature was controlled between 69°F and 72°F.

The tests were carried out using a linear impactor shown in Figure 11 driven by a pendulum from one end. This impactor was designed to have the flexibility of attaching different moving masses and different impactor heads that are required by each test. Different speeds can be achieved by changing the drop height and the mass of the pendulum. The linear impactor was instrumented with an LVDT, a uniaxial load cell, and an accelerometer. The speed of the impactor was monitored by a velocity gate which is not shown in the picture. The speed gate can be seen in Figures 12, 16 and 18. During the tests, the impactor was fixed to a frame which also supported the driving pendulum and a height-adjustable platform for sitting the dummy.

The setups and results of each test are discussed below.

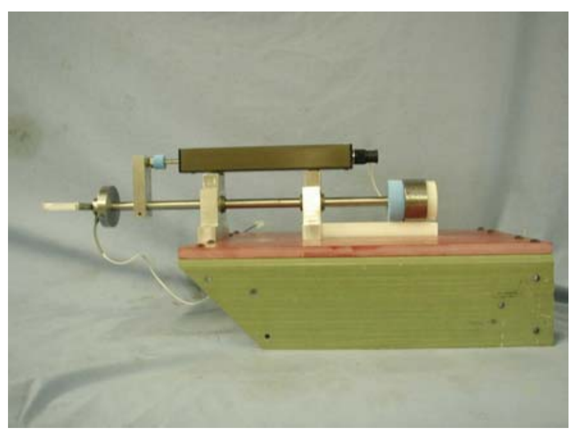

Figure 11. The linear impactor used for testing the infant dummy (with the impactor head for abdomen test).

Head Impact Test

Figure 12 shows the setup of the head impact test. The dummy was sitting upright on a hard plastic surface without any other support. The impactor was aimed at the center of the forehead of the dummy. The impact speed was 0.8 m/s.

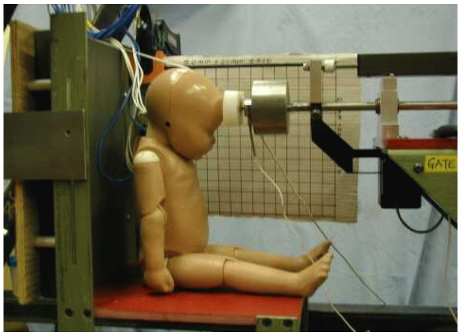

Figure 12. Setup of the head impact test.

Figure 13 shows the result of the head impact compared to its biomechanical corridor. It can be seen that the repeatability of the two tests were very good. The average peak force of the tests was within the scaled corridor, while the timing of the peak is about 1ms behind the scaled corridor. The performance of the head in the impact test was found to be acceptable.

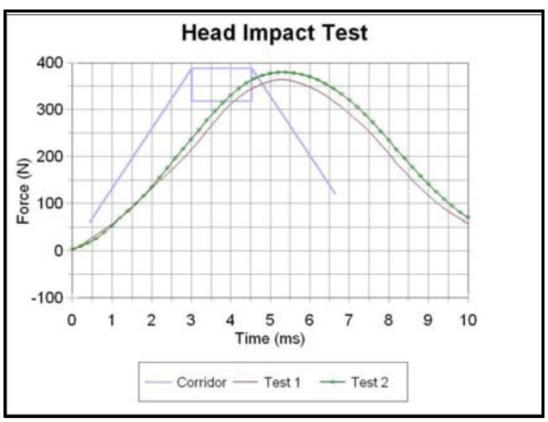

Figure 13. Head impact result of the infant dummy.

Head Drop Certification Test

The setup of the head drop test is shown in Figure 14. The setup is the same as used for testing the HIII dummy head. The head was dropped from a height of 376mm to a rigid surface (a thick steel plate). Accelerations on three directions were measured to calculate the resultant value.

Figure 15 shows the drop test result. The tests showed good repeatability, and the average peak value of the three tests was within the scaled corridor.

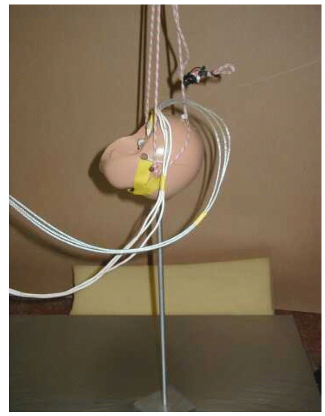

Figure 14. Setup for head drop test.

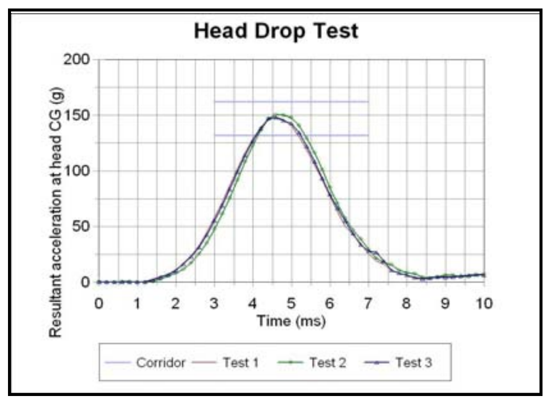

Figure 15. Head drop test result of the infant dummy.

Thorax Impact Certification Test

Figure 16 shows the setup of the thorax impact test. The dummy was seated so that the anterior chest is approximately parallel to the impactor head. The center of the impactor is aimed at the center of the chest, but avoiding impact with the shoulder block inside the dummy. The head of the dummy drooped naturally. The forearms of the dummy were raised to allow better view for the high speed camera.

Figure 17 shows the result of the thorax impact tests. It can be seen that the two tests repeated well. Though the force is close to the upper limit of the scaled corridor, the maximum deflection of the chest is within the corridor. The design successfully achieved the relatively flat part of the force-deflection as in the scaled corridor, as well as, the hysteresis. The performance of the chest was considered to be acceptable.

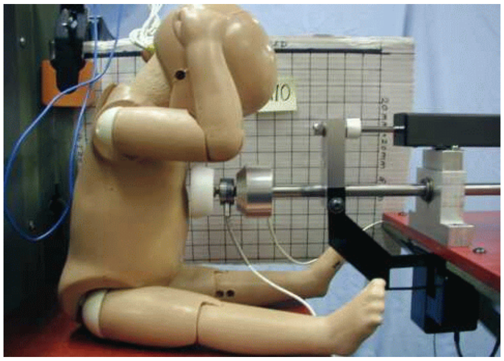

Figure 16. Setup of the thorax impact test (the forearms were raised to allow better high speed camera view).

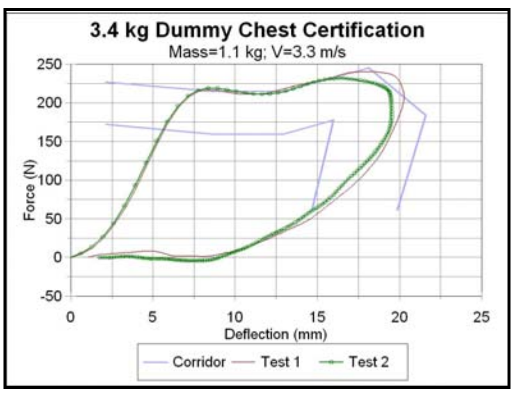

Figure 17. Test result of the thorax impact compared with scaled corridor.

Abdomen Impact Certification Test

Figure 18 shows the setup of the abdomen impact test. The dummy was seated with its torso upright. The impactor head was aimed at the center line of the abdomen. The arms were placed out of the way for better viewing by the high speed camera.

Figure 19 shows the result of the abdomen impact tests. The repeatability of the two tests was good. The abdomen was compressed by about 41mm, and the force during the compression fell into the scaled corridor, which demonstrated that the current abdomen design was able to meet the target requirements.

Neck Pendulum Certification Test

The neck pendulum certification was performed in all three modes - frontal flexion, extension and lateral flexion. Currently the response of the neck is focused on the kinematic requirement. The angular motion of the head was calculated from the angular rate sensors located at the head CG. The time history of this curve was plotted against its requirement to see how the neck performs under different mode of dynamic loading. Neck axial load from the lower neck load cell and accelerations at head CG were also measured, but were not examined for response purposes.

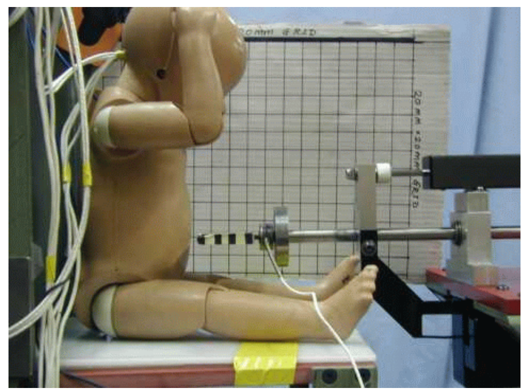

Figure 18. Setup of the abdomen impact test.

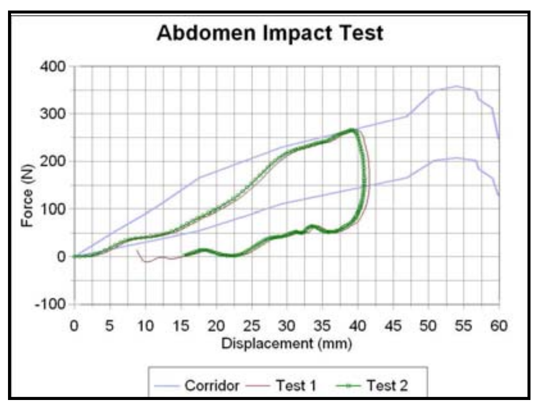

Figure 19. Test result of the abdomen impact compared with scaled corridor.

Figure 20 shows the setup of the neck pendulum test. The head-neck assembly was attached to the pendulum arm through an adapter. The drop angle of the pendulum was monitored by an angular potentiometer at the rotation axle. The pendulum was stopped by a piece of foam attached to a rigid frame. The foam was carefully selected to produce the desired pulses for the testing.

Figures 21 is the pulse produced for frontal flexion and extension tests, and Figure 22 the pulse for the lateral test. For frontal flexion and extension, the pendulum was dropped from the same height, creating a pulse of about 25 g with a duration of about 20 ms. For lateral flexion, the pendulum was dropped from a lower height, creating a pulse of 12.6 g with a duration of 30 ms.

The results of the tests are shown in Figures 23, 24 and 25 for frontal flexion, extension and lateral flexion respectively. For the flexion test, it can be seen that the timing of the peak is close to the target kinematic requirement though the maximum head angle is slight higher. In the extension test, the maximum angle is also slightly higher than the required value, but the time of the peak is within the requirement. In the lateral test, the maximum angle is higher than the required value, but the time of peak is within the requirement.

Overall, the time of peak angles were within the kinematic requirement. The maximum angles were found to be slightly higher than the requirements. Since the requirements were based on scaling the adult 50th percentile neck responses on the pendulum, it is felt that additional biofidelity data should be developed before proceeding with any modifications to better tune the necks.

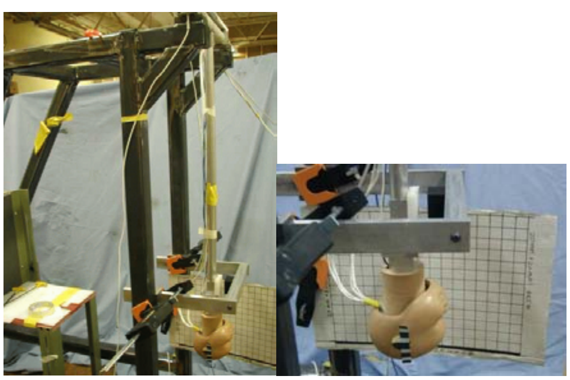

Figure 20. Setup of the neck pendulum test for frontal flexion (left: overall view, right: local view).

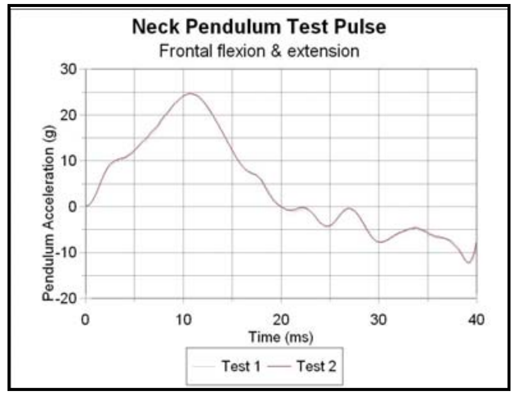

Figure 21. Acceleration pulse produced by the pendulum for frontal flexion and extension tests.

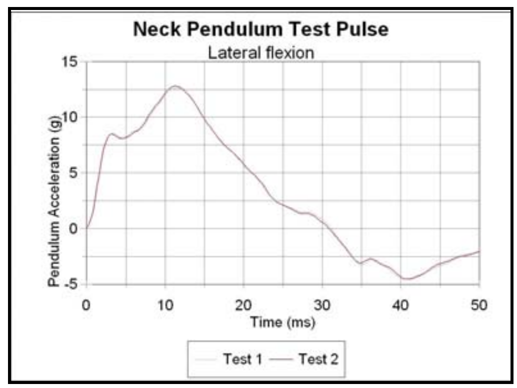

Figure 22. Acceleration pulse produced by the pendulum for lateral flexion test.

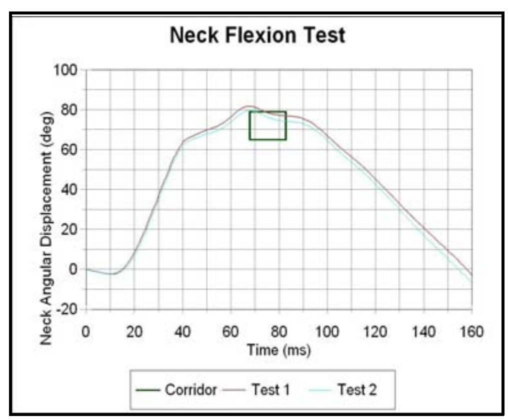

Figure 23. Time history of head angle during neck frontal flexion.

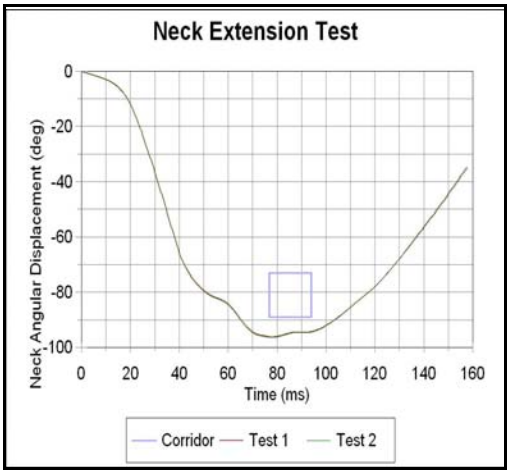

Figure 24. Time history of head angle during neck extension.

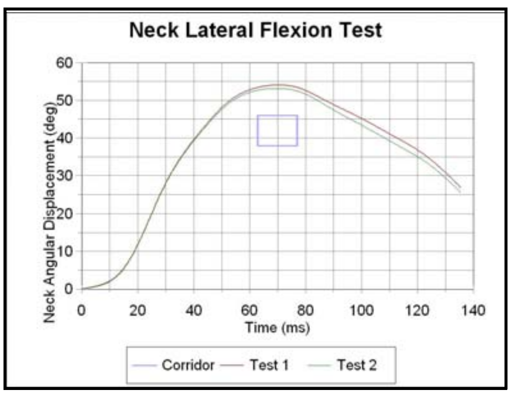

Figure 25. Time history of head angle during neck lateral flexion.

CONCLUSION

A 3.4 kg infant dummy representing an average of newborn baby was developed. The dummy was designed to be used for various testing setups including child restraint system evaluation, stroller comfort evaluation, shaken baby syndrome studies, and others. The dummy includes 26 channels of data to provide sufficient data for the different tests planned using it. Biofidelity requirements for the dummy were obtained by scaling the requirements for the 50th percentile male dummy. The biofidelity tests have shown that the performance of the dummy either meet or are close to the target requirements.

As new data from cadaver testing on newborns come out in the future, some of the biofidelity requirement may need to be revised. However, the success with this dummy demonstrated that the design approach used this study can be employed to meet those possible revisions in the future.

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