SIMULATION OF BIODYNAMIC RESPONSE TO UNDERWATER EXPLOSION

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BACKGROUND AND GOALS

NPS has been involved in research on "Underwater Explosion and Its Effects on Naval Ship" since 1983. The research on the effects of underwater explosion to surface ship and submarine has been active in U. S. since the World War II. The response of ship structural system, contained equipment and weapon system subjected to underwater explosion has been extensively investigated from the standpoint of susceptibility, vulnerability and survivability. However, in evaluating a ship's ability to remain a viable warfighting asset, crew survivability must also be addressed. For a ship to remain capable of fighting following damage resulting from enemy munitions such as mines or torpedoes, the ship's crew must remain sufficiently uninjured to be able to employ the weapons systems and fight the ship. This article is based on our recent research concentrated on investigating the effects of underwater explosions on shipboard crew vulnerability.

There are three basic goals for this research; (i) to develop a method for estimating crew survivability to underwater explosion, (ii) to use accelerometer data and video footage taken during live fire testing as a basis for the simulation, and (iii) to perform injury estimates for both male and female crew members.

The Articulated Total Body (ATB) Program [1] was used and it was primarily designed to simulate the three dimensional response of a system of rigid bodies subjected to dynamic applied and interactive contact forces. The ATB program was also developed to model the response of crash test dummies, but is used in many varied applications including human body motion, transient response of a MX missile in a wind tunnel, and pilot ejection from aircraft [1]. Use of the ATB program to model the response of a human in a shipboard environment is not quite different from using it to model the responses of a human or test dummy in an automobile or aircraft crash. Once a model is developed of the environment, the result of changes to the input excitation, such as improved shock isolation or varied charge size/location, can be estimated and potential injuries predicted.

EXPERIMENTAL SET-UP AND ATB MODEL

This study relies on data for vehicle excitation and dummy response provided by the Naval Surface Warfare Center Carderock Division Underwater Explosions Research Department. The data is from the Site Phase 3 (SSTV) Shock Test series, Shot 9991, conducted 17 June 1996. The anthropomorphic test device (ATD) modeled in this study was a Hybrid III 50th percentile male dummy [2] instrumented with triaxial linear accelerometers located at the centers of gravity of the head, thorax, and pelvis (Fig. 1). The ATD was seated, lap belt securely fastened, facing starboard in a standard operator's chair on the upper platform of a two platform test vessel that was subjected to an underwater explosion event. The test vessel was equipped with linear accelerometers. Accelerometers oriented in the vertical and athwartship directions were located at the base of the chair and the measured output was used as the basis of the excitation used in the ATB model.

A model of the operator's chair, based on measured physical dimensions, was constructed in the ATB program using plane contact surfaces for the seat pan, sides and back, and using contact segments for each arm rest. Force-deflection, energy absorption and damping properties for the seat surfaces were estimated. The physical dimensions, inertial properties and joint characteristics for the Hybrid III dummy were used directly as generated by the Generator of Body Data (GEBOD) program [3]. The model of the dummy was positioned in the simulation to match as closely as possible the position of the dummy as seen in the video of the test.

Figure 1. Seated Hybrid III ATD Dummy Setup and Sensor Locations [2]

HYBRID ATD DUMMY: MEASURED AND PREDICTED RESPONSE

Evaluation of the accuracy of the ATB model was made by comparing the accelerations of the head, thorax, and pelvis as predicted by the simulation to the measured values. In addition, the gross body motion as predicted by the simulation was compared to still frame images captured from standard video taken of the dummy during the underwater explosion event. Modifications were made to the initial position of the dummy and the characteristics of the chair in the simulation until reasonable agreement was obtained between accelerations and gross body motion. Figure 2 shows the sign convention used in reporting the accelerations of the head, thorax, and pelvis. As motion was predominantly in the sagittal plane and no lateral input acceleration (fore-and-aft) was used in the model excitation signal, accelerations in the Y direction were not compared.

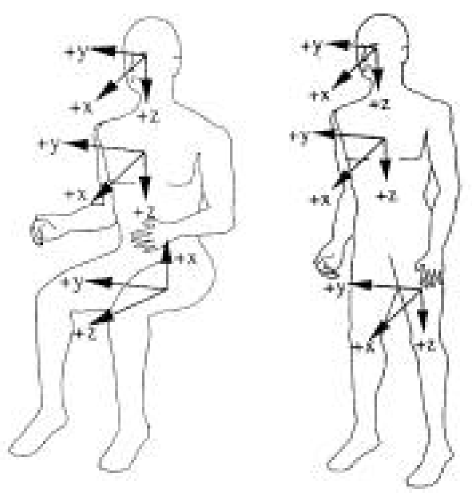
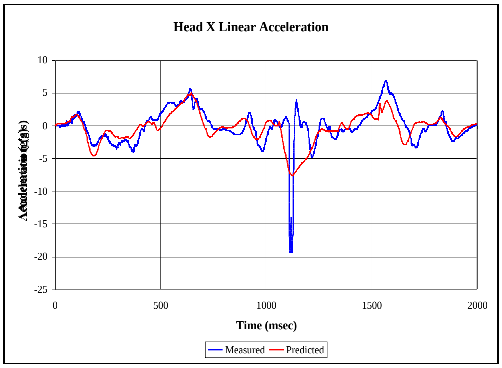


Figure 2. Dummy coordinate system [3]

Quite good overall agreement was seen in both the head X and Z directions (Fig. 3 and 4), with the phasing consistent and the many of the amplitudes closely matched. Agreement in the chest X direction (Fig. 5) is not as good, but the chest Z results (Fig. 6) show good phasing response even though the magnitudes in the peaks are for the most part under-estimated. Similarly, the pelvis X results (Fig. 7) are not as closely in agreement as the pelvis Z results (Fig. 8) which show good agreement both in phasing and amplitude.



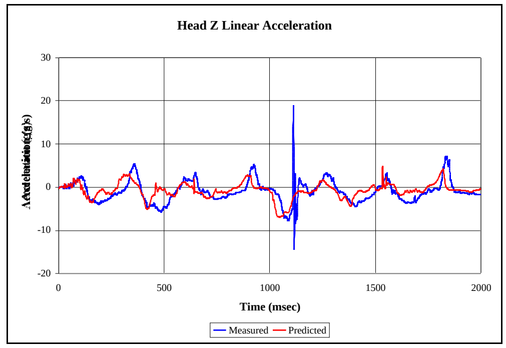
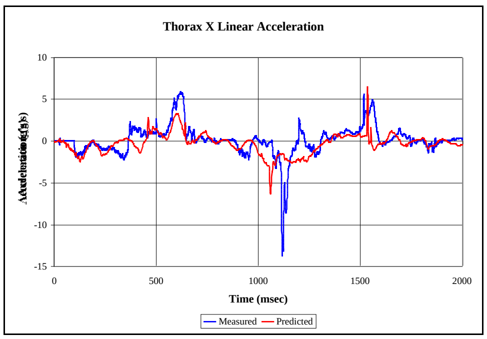


Figure 3. Comparison of head X accelerations

Figure 4. Comparison of head Z accelerations



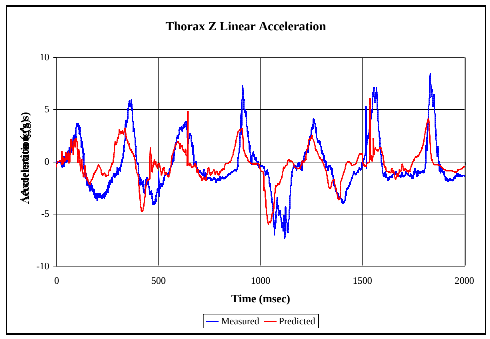
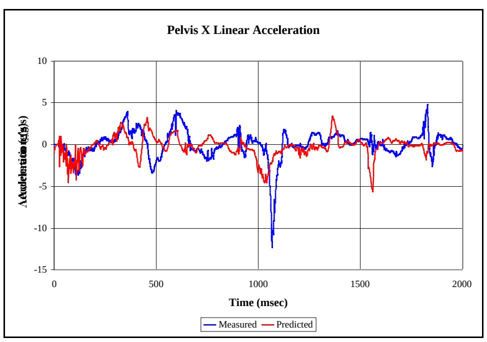


Figure 5. Comparison of chest X accelerations

Figure 6. Comparison of chest Z accelerations



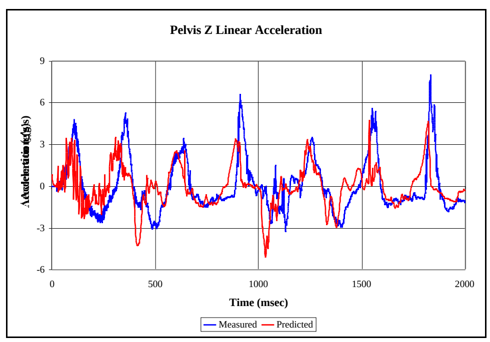


Figure 7. Comparison of pelvis X accelerations

Figure 8. Comparison of pelvis Z accelerations

The predicted gross bodily motion of the ATD dummy is also in reasonably good agreement with the images captured from the video of the test event. Figure 9 shows several frames comparing the test video with the predicted motion generated using the IMAGE program [4]. Basic phasing of the motion agrees well with the video although the arm motion is significantly different. One source of differing motion is the seat back. As can be seen the images from the test, the angle that the seat back makes with the seat pan in increased after the first recoiling of the dummy into the seat back. However, the seat back was not modeled as being able to rotate in the simulations.

Overall, the agreement between the predicted and recorded motions and accelerations was considered to be quite good and sufficient to demonstrate the validity of the model.

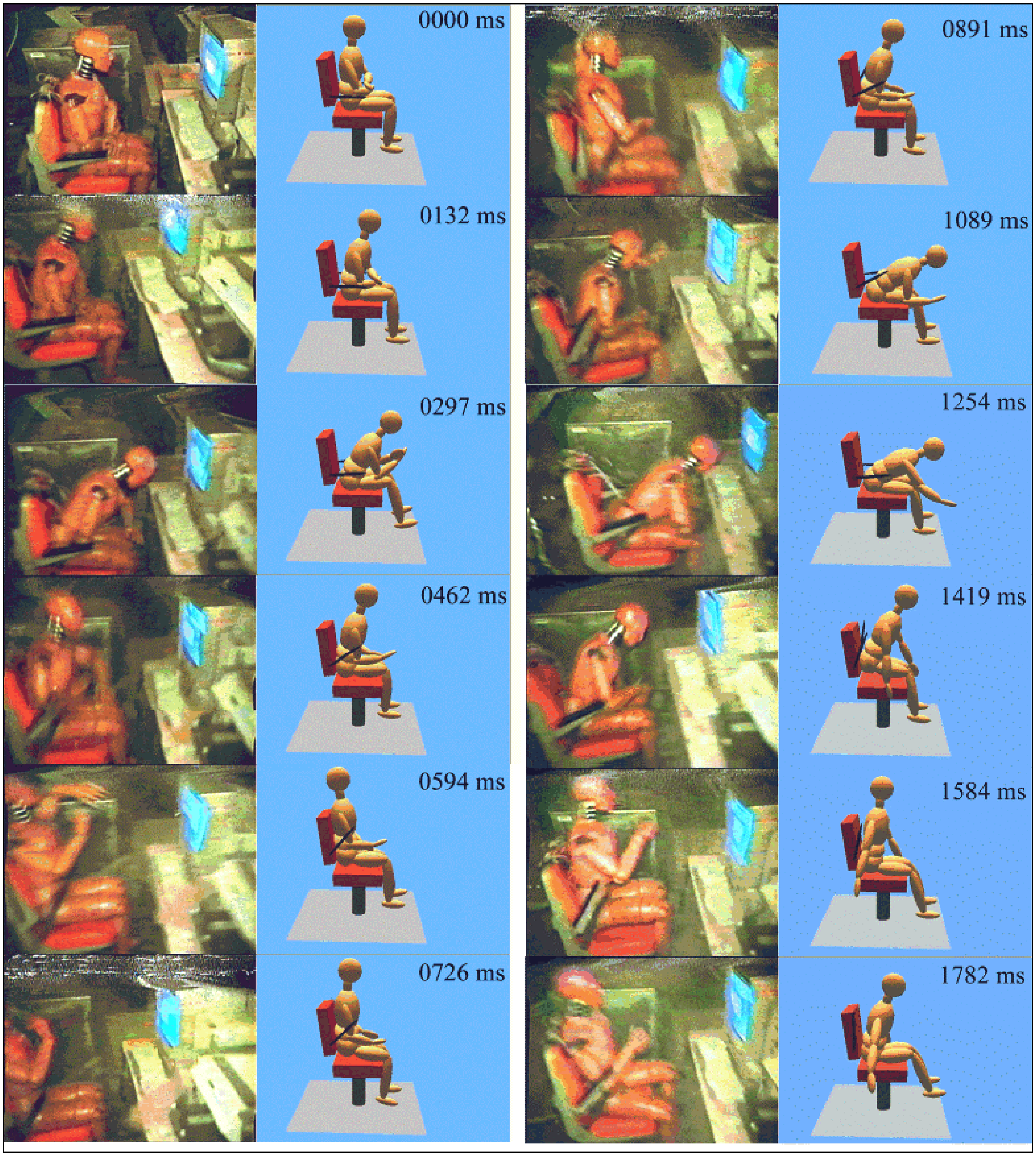


Figure 9. Motion Validation for Shot 9991

SIMULATION MODEL FOR HUMAN MALE AND FEMALE RESPONSES

The validated model of the chair and input excitation was used for predicting the response of a 50th percentile human male (5 ft 10 in tall, 173.5 lb) and of a 5th percentile human female (5 ft 0 in tall, 100 lb) in three separate situations. The first simulation was very similar to that of the Hybrid III dummy in the original test. The lap belt was still fastened, but the arms were positioned more naturally. The second simulation was with the lap belt removed and a desk surface placed in front of the chair. The position of the subjects were identical to that in the first simulation. The final simulation was the same as the second, but surfaces representing a computer keyboard and monitor were included. For each of these cases, the simulation was performed for both the male and the female human subjects. The initial position and simulation setup for each of the six cases may be seen in Figure 10 with the male subjects shown on the top row and the female subjects on the bottom row.

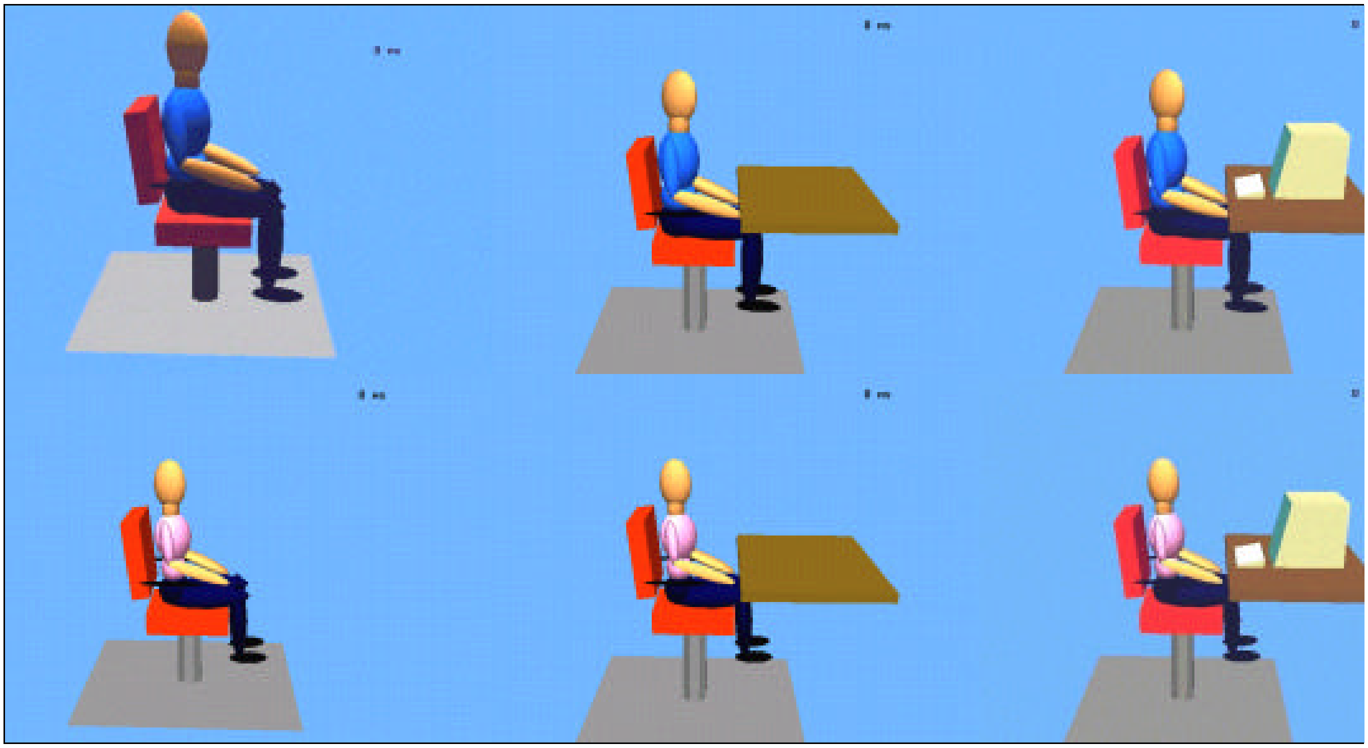


Figure 10. Simulation Setups for Male(top) and Female(bottom) Models

DESCRIPTION OF INJURIES AND INJURY CRITERIA

The only injuries for which estimates were performed for the seated subjects were those of the head and head-neck complex. Specifically, cerebral concussion, whiplash, fractures of bones of the face or skull, and injuries to the cervical spine due to axial loading. These specific injuries were considered to be not only the most likely to be sustained, but also those that would be most debilitating in the near or long term. Various parameters of the predicted response, as described below, were compared against their associated injury threshold values and estimates made of the likely injuries for each of the subjects.

Cerebral concussions, which, according to T*aber's Cyclopedic Medical Dictionary* [5] have associated symptoms of transient dizziness, paralyses, or unconsciousness; unequal pupils; shock; vomiting; rapid pulse; headache; and cerebral irritation, were evaluated by comparing the angular velocities and accelerations of the center of gravity of the head against the corresponding injury tolerances. According to Ommaya [6] the crucial injury mechanism leading to the onset of cerebral concussion is severe shear strain imposed by brain rotation and the proposed thresholds to predict a 50 percent probability of the onset of cerebral concussion in terms of angular velocity is 50 rad/sec, and in terms of angular acceleration is 1800 rad/sec^2. Thus, the resultant angular velocities and accelerations of the center of the head as predicted using the ATB program were plotted and compared to these threshold values to estimate the likelihood of the subject receiving a cerebral concussion.

Whiplash, which is a somewhat vague term referring to the broad collection of acceleration induced traumas to the cervical spine, is typically associated with rear end automobile collisions. In these instances, the body experiences a sudden forward acceleration, while the inertia of the head keeps it stationary. The force applied by the torso to the lower portion of the head causes a rotation of the head, resulting in an extension of the cervical spine. If the acceleration is sufficient, the inertial loading of the cervical spine can result in hyperextension and a whiplash injury. Kallieris [7] notes that restraining the torso during a deceleration event (such as a frontal collision) can lead to a hyperflexion of the cervical spine and an associated whiplash injury. Thus, the key components are excessive angle of the neck with respect to the torso, either in flexion or extension, combined with tensile loading. As reported in Panjabi and White [8], possible angular position thresholds are 80 degrees in extension and 58 degrees in flexion. Mertz and Patrick [9] claim that the torque developed at the occipital condyles (the point at which the skull articulates with the cervical spine) is a better predictor of whiplash injuries and that appropriate tolerance values are, for extension, 35 ft-lb (injury) and 42 ft-lb (ligamentous damage), and, for flexion, 44 ft-lb (pain), 65 ft-lb (injury), and 140 ft-lb (ligamentous or bone damage). Thus, for this research, the head angular position with respect to the torso predicted using the ATB program was considered to be a weak indicator for whiplash injury and the torque at the occipital condyles, also predicted using the ATB program, was considered to be a strong indicator for whiplash injury. In all cases, the axial force within the neck was check to verify that the loading was tensile.

Potential fractures of the bones of the face and skull were estimated by comparing the contact forces between the head segment and the desk or computer surfaces as computed using the ATB program against the respective fracture thresholds. The particular bone in question was estimated by close examination of the visualization of the motion generated using the IMAGE program. For each contact, the portion of the ellipsoid representing the head which was in contact with the desk or computer was associated as nearly as possible to the corresponding bone in the face or skull. The contact locations considered in this study and their associated threshold fracture forces, as summarized in Allsop [10] are: frontal, 900 lbf; temporoparietal, 450 lbf; zygomatic, 225 lbf; maxilla, 150 lbf; anterior-posterior mandible, 400 lbf; and lateral mandible, 200 lbf.

Significant injuries to the cervical spine due to axial loading, such as vertebral body fractures, facet dislocations, disk ruptures, and longitudinal ligament tears, were estimated by comparing the predicted axial load in the neck, along with the associated relative position (flexion, neutral, or extension), against the associated tolerance values. The neck axial load was computed from the forces in the joints between the upper torso and neck and between the neck and head. These joint forces were directly computed using the ATB program. Compression loading injury threshold forces as summarized in Sances, et. al. [11] are 6000 N for pure compression, 2000 N for compression-flexion, and 2200 N for compression-extension. These position specific tolerances were used in conjunction with duration of loading curves for pure compression provided in AGARD [12]. Tensile loading injury threshold forces as summarized in McElhaney and Meyers [13] as 1450 N for pure tension and 1160 N for tension-extension. As was the case for compression loading of the cervical spine, the previously listed threshold forces for tension loading were used in conjunction with the duration of loading curves for pure tension provided in AGARD [12]

Table 1 provides a summary of the injury criteria used in this research, along with the associated source for each.

PREDICTED RESPONSE AND ESTIMATED INJURY POTENTIALS

The predicted motions of the male and female subjects wearing the lap belt are shown in Figure 11 and Figure 12 respectively. Each subject went through multiple rebounds of the torso off the upper legs during the two seconds of the shock excitation. During the first rebound, occurring at approximately 380 msec, each of the subjects are likely to receive a whiplash injury and possibly a cerebral concussion. The male subject’s head reached a peak angle in flexion of 91.8 deg at 398 msec, with an associated torque at the occipital condyles of 44.1 ft-lb reached at 389 msec, and experienced a peak angular acceleration of 2242 rad/sec^2 at 388 msec. The corresponding injury threshold values are 58 deg [8], 44.1 ft-lb for pain [9], and 1800 rad/sec^2 [6]. The female subject’s head reached a peak angle in flexion of 87.8 deg at 386 msec, with an associated toque at the occipital condyles of 30.2 ft-lb reached at 389 msec, and experienced a peak angular acceleration of 1903 rad/sec^2. During subsequent rebounds of the torso off the upper legs, additional whiplash injuries are possible based solely upon the angle of the head. The male subject’s head reached peak flexion angles of 63.0 deg at 1020 msec and 79.1 deg at 1851 msec, during the second and fourth rebounds, respectively, while the female subject’s head reached a peak flexion angle of 74.8 deg at 999 msec during the second rebound.

The predicted motions of the male and female subjects not wearing the lap belt, but seated at a bare desk, are shown in Figure 13 and Figure 14, respectively. Since the desk is present, each subject experienced multiple impacts of the head against the desk rather than rebounds of the torso off the upper legs. During the first head to desk contact, each subject would possibly receive a cerebral concussion based on head angular accelerations in excess of the 1800 rad/sec^2 tolerance value [6]. The male subject experienced a peak acceleration of 2109 rad/sec^2 at 431 msec and the female subject experienced a peak of 074 rad/sec^2 at 346 msec. The female subject would possibly receive additional injuries during the first head to desk contact. The female subject experienced a peak axial load in the neck of 1614 N at 344 msec, which slightly exceeds the threshold for significant neck injury [13], and a contact force between the lateral mandible and the desk of 390 lbf, which exceeds the 200 lbf fracture threshold for that bone [10]. During the second head to desk contact, the male subject experienced a peak contact force between the maxilla and the desk of 465 lbf, which exceeds the 150 lbf fracture threshold for that bone [10] and would likely result in fracture. The female subject experienced a peak head angular acceleration of 1984 rad/sec^2, which slightly exceeds the 1800 rad/sec^2 threshold [6] and would possibly result in a cerebral concussion. The male subject experienced a third head strike with a peak contact force between the zygomatic bone and the desk of 309 lbf, which exceeds the 225 lbf fracture threshold for that bone [10] and would possibly result in fracture.

Table 1. Summary of Injury Criteria

Injury	Criteria	Source
Cerebral concussion	$\omega \geq 50$ rad/sec $\alpha \geq 1800$ rad/sec^2	Ommaya, et.al., [6]
Whiplash (extension)	$\theta \geq 80$ deg	Panjabi and White [8]
	35 ft-lb (injury) 42 ft-lb (ligamentous damage)	Mertz and Patrick [9]
Whiplash (flexion)	$\theta \geq 58$ deg	Panjabi and White [8]
	44 ft-lb (pain) 65 ft-lb (injury) 140 ft-lb (ligamentous or bone damage)	Mertz and Patrick [9]
Skull fracture	900 lbf (frontal) 450 lbf (temporoparietal)	Allsop [10]
Facial fracture	225 lbf (zygomatic) 150 lbf (maxilla) 400 lbf (anterior-posterior mandible) 200 lbf (lateral mandible)	Allsop [10]
Cervical spine injury due to axial loading (compression)	6000 N (pure) 2000 N (flexion) 2200 N (extension)	Sances, et. Al., [11]
	Duration of loading curve	AGARD [12]
Cervical spine injury due to axial loading (tension)	1450 N (pure) 1160 N (extension)	McElhaney and Meyers [13]
	Duration of loading curve	AGARD [12]

The predicted motions of the male and female subjects not wearing the lap belt, but seated at a desk with a computer, are shown in **Error! Reference source not found.** and **Error! Reference source not found.**, respectively. Similar to the unbelted case, each subject experienced multiple head strikes, but this time against the computer. Examining the predicted response parameters of the male subject revealed none in excess of the corresponding thresholds. Thus, the male subject is not likely to receive any of the injuries considered in this research. No consideration was made of possible lacerations resulting from breaking of the computer screen during impacts, so no prediction can be made of the likelihood of the male subject receiving this type of injury. The female subject experienced peak angular accelerations of 1880 rad/sec^2 during the first head strike against the computer, and 2427 rad/sec^2 during the second. The angular acceleration experienced during the first contact slightly exceeds the 1800 rad/sec^2 threshold value (Ommaya, et. al., 1993) and would possibly result in a cerebral concussion. The angular acceleration during the second peak is well in excess of 1800 rad/sec^2 and would thus be likely to result in a cerebral concussion. During the second head to computer contact, the female subject also experienced a peak contact force between the zygomatic bone and the computer of 360 lbf, which exceeds the 225 lbf fracture threshold for that bone (Allsop, 1993) and would possibly result in fracture. Additionally, during the second head to computer contact, the female subject experienced a peak axial loading of the neck of 2605 N (compression-extension) which exceeds the 2200 N threshold for significant neck injury (Sances, et. al., 1986).

A summary of the estimated injuries for the six subjects is provided in Table 2.

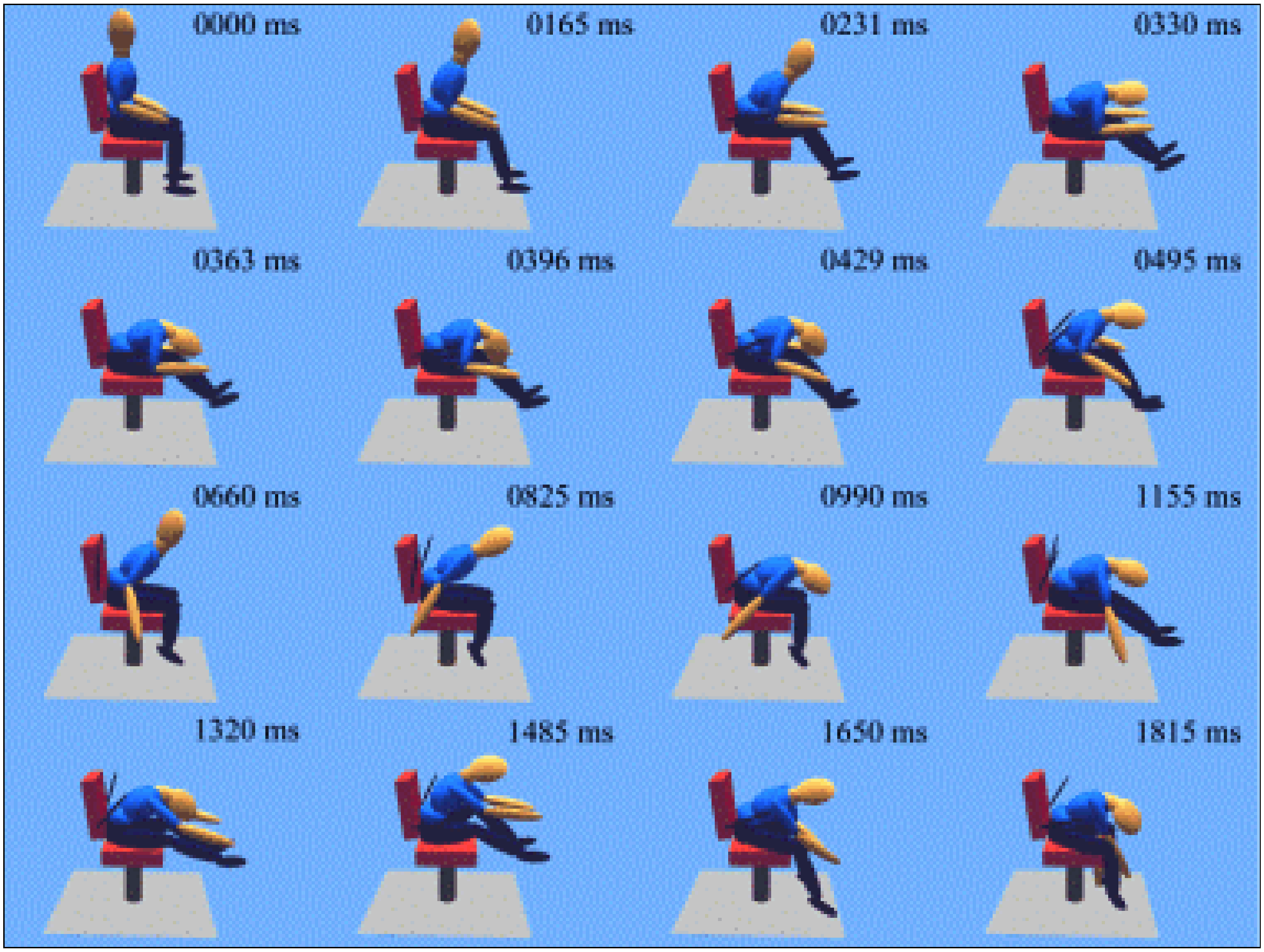


Figure 11. Predicted Motion of Male Subject Wearing Lap Belt

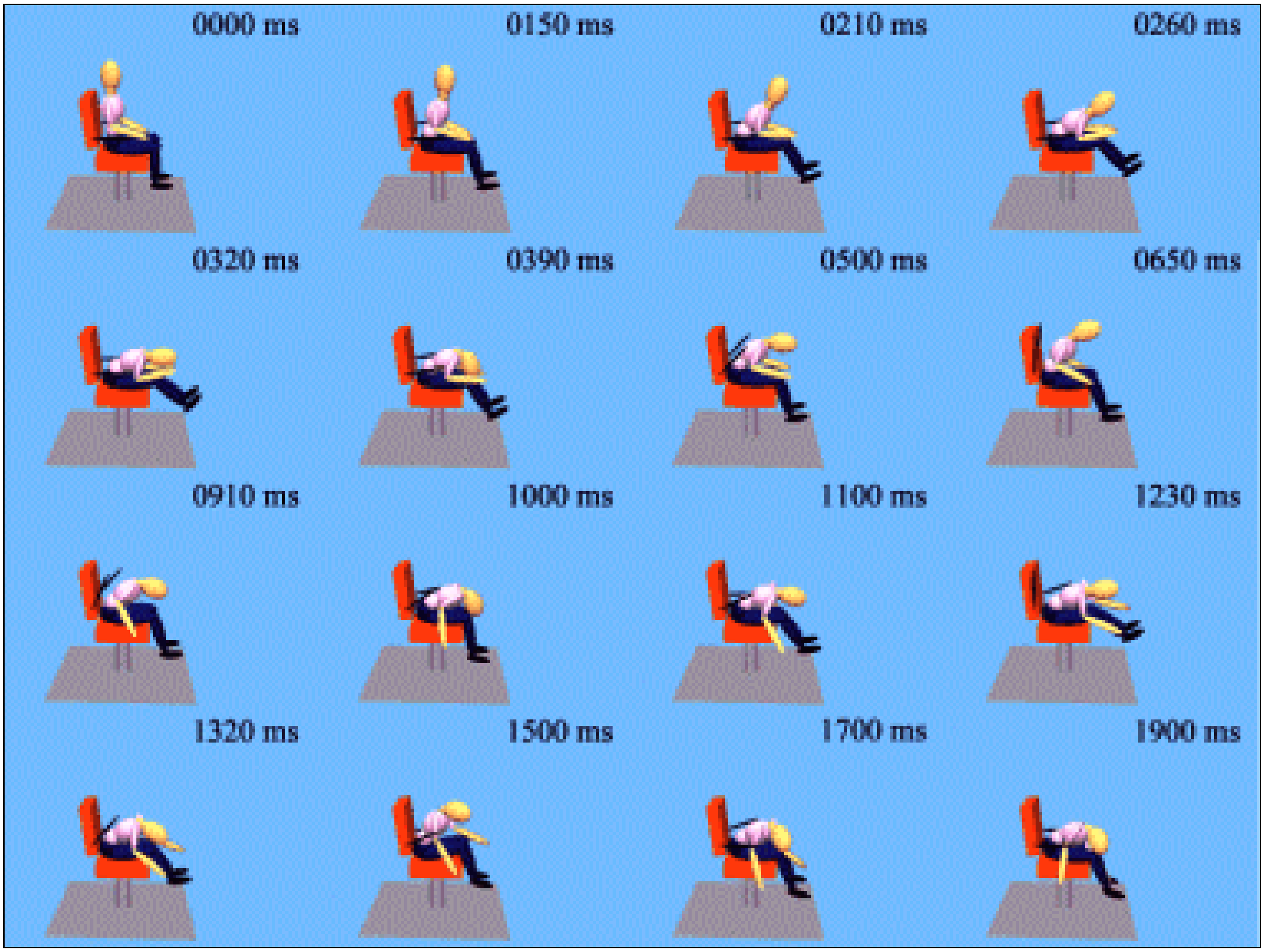


Figure 12. Predicted Motion of Female Subject Wearing a Lap Belt

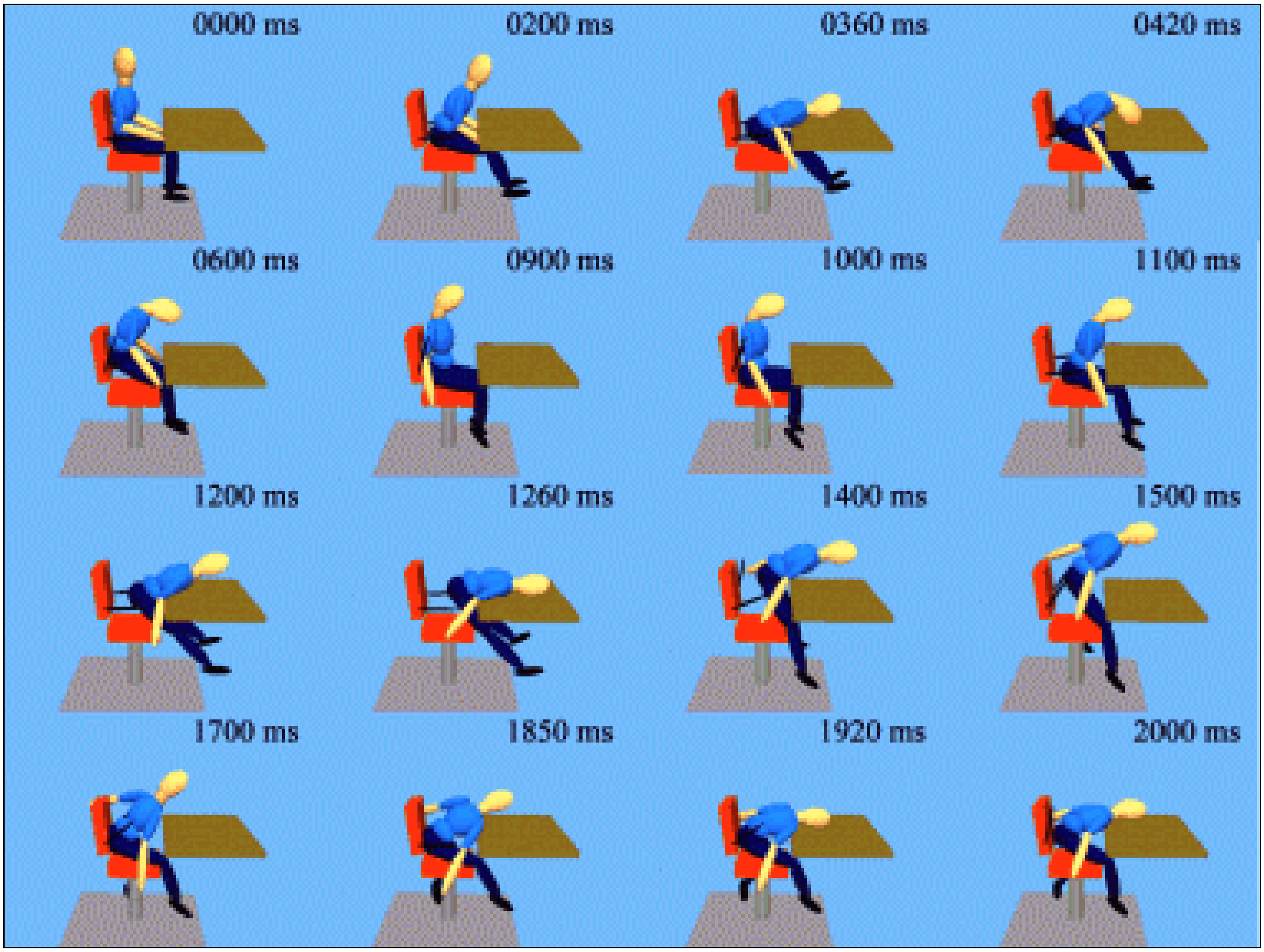


Figure 13. Predicted Motion of Unbelted Male Subject

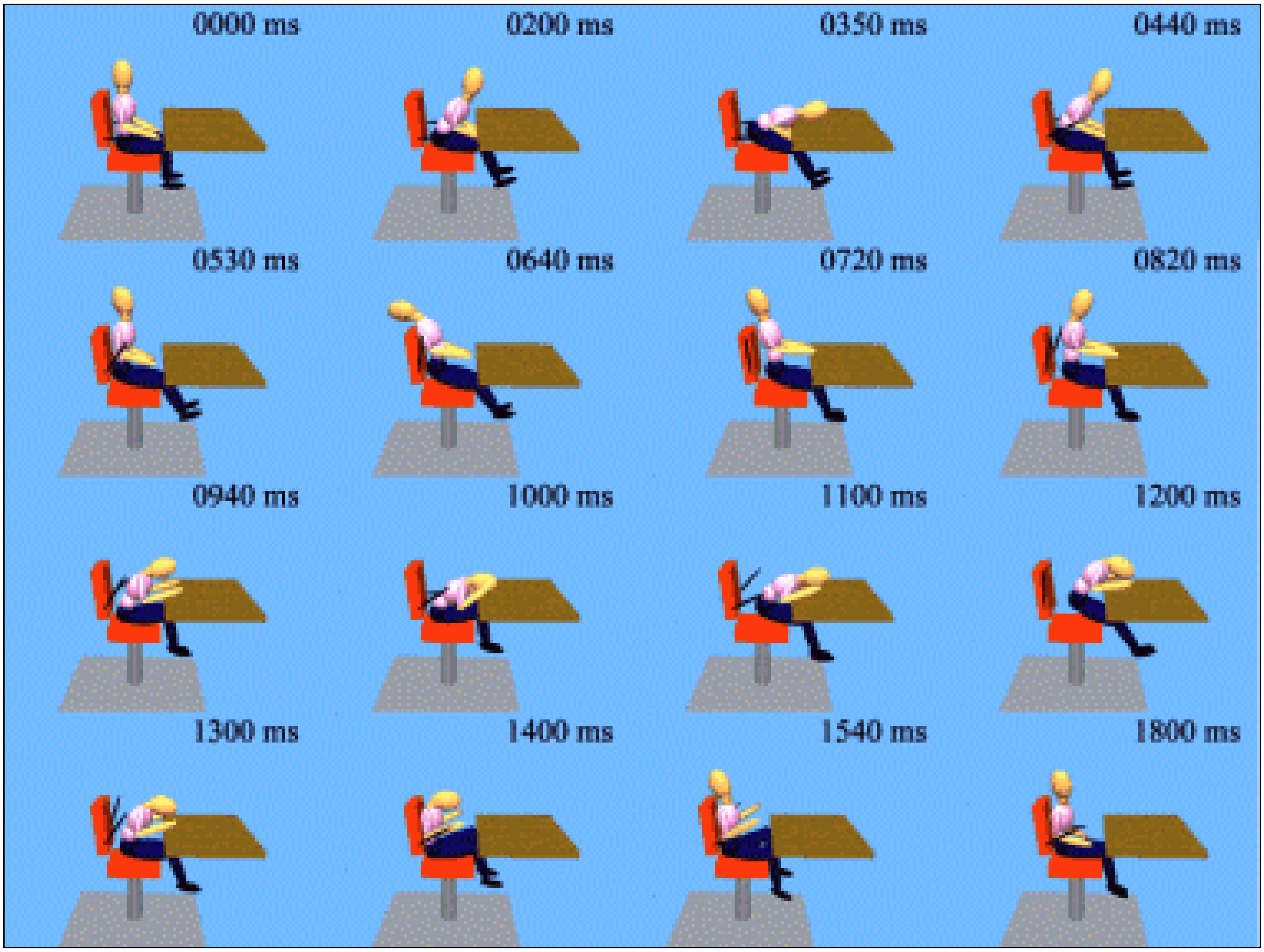


Figure 14. Predicted Motion of Unbelted Female Subject

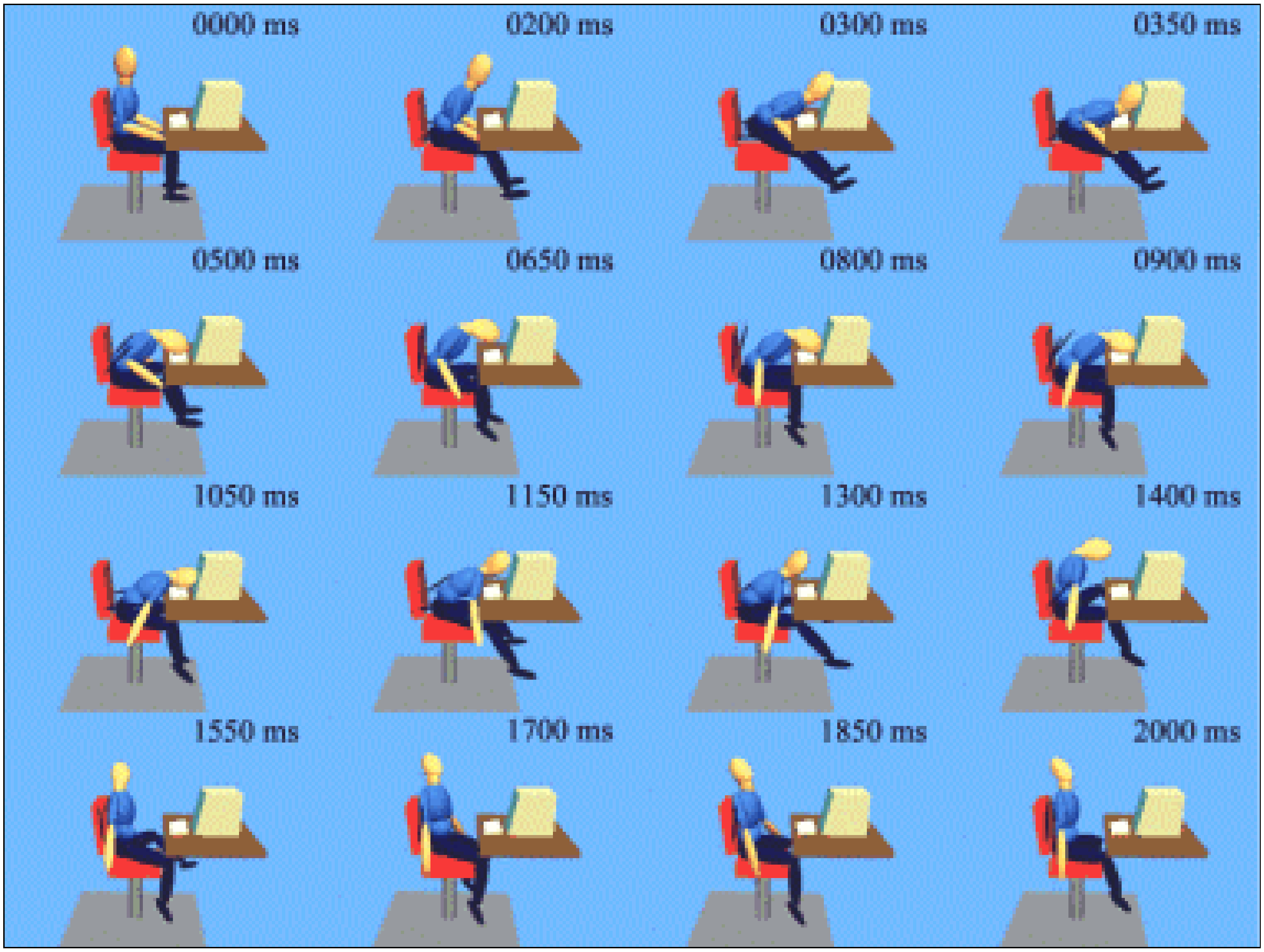


Figure 15. Predicted Motion of Male Subject at Computer

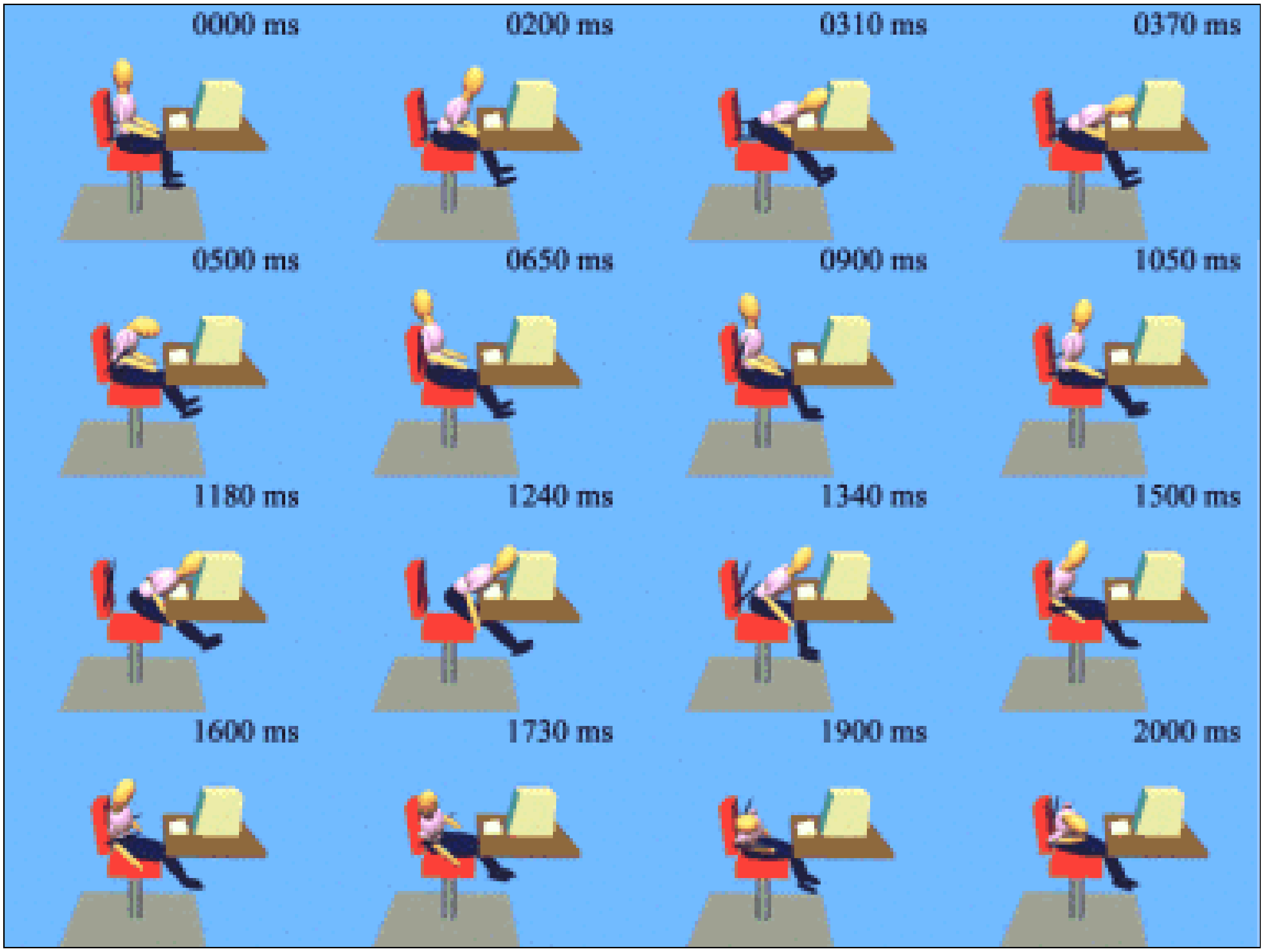


Figure 16. Predicted Motion of Female Subject at Computer

Table 2. Summary of Injury Estimates

		Time (msec)	Parameter	Value	Limit	Outcome
Belted Subjects	Male	388	Head α	2242 r/s^2	1800 r/s^2	Possible cerebral concussion
		398	Head θ	91.8 deg	58 deg	Probable whiplash injury
		389	Torque	44.1 ft-lb	44 ft-lb	
		1020	Head θ	63.0 deg	58 deg	Possible (not likely) whiplash injury
		1851	Head θ	79.1 deg	58 deg	Possible (not likely) whiplash injury
	Female	379	Head α	1903 r/s^2	1800 r/s^2	Possible cerebral concussion
		386	Head θ	87.8 deg	58 deg	Probable whiplash injury
		389	Torque	30.2 ft-lb	44 ft-lb	
		999	Head θ	74.8 deg	58 deg	Possible (not likely) whiplash injury
Unbelted Subjects	Male	431	Head α	2109 r/s^2	1800 r/s^2	Possible cerebral concussion
		1257	Head cont. force	465 lbf	150 lbf	Possible fracture of the maxilla
		1917	Head cont. force	309 lbf	225 lbf	Possible fracture of the zygomatic
	Female	346	Head α	2074 r/s^2	1800 r/s^2	Possible cerebral concussion
		983	Head α	1984 r/s^2	1800 r/s^2	Possible cerebral concussion
		346	Head cont. force	390 lbf	200 lbf	Possible fracture of the lateral mandible

		344	Neck axial force	1614 N	1450 N	Possible significant neck injury
Subjects at Computer	Male	No injury tolerances exceeded. Potential exists for lacerations resulting from possible breakage of computer screen during direct head impact.				
Subjects at Computer	Female	340	Head α	1880 r/s^2	1800 r/s^2	Possible cerebral concussion
Subjects at Computer	Female	1178	Head α	2427 r/s^2	1800 r/s^2	Possible cerebral concussion
Subjects at Computer	Female	1189	Head cont. force	360 lbf	225 lbf	Possible fracture of the zygomatic
Subjects at Computer	Female	1189	Neck axial force	2605 N	2200 N	Possible significant neck injury

CONCLUSIONS

On the basis of the presented results:

1. The Articulated Total Body model is a viable tool for simulating both male and female personnel in various positions in a shipboard environment during underwater explosion events.

2. Significant injuries can be expected for both male and female subjects in a shipboard environment subjected to a shock induced excitation.

3. The selection of application of injury criteria to predicted motion is extremely complicated.

ACKNOWLEDGEMENTS

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