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STUDY OF HEAD INJURIES IN CHILDREN

Author(s)

K.A. McCaul, A.J. McLean, C.N. Kloeden and R.W. Hinrichs

Performing Organisation (Name and Address)
NH&MRC Road Accident Research Unit
University of Adelaide
GPO Box 498, ADELAIDE SA 5001

Sponsor (Name and Address)
Federal Office of Road Safety
GPO Box 594
CANBERRA ACT 2601

Available from (Name and Address)
Federal Office of Road Safety
GPO Box 594
CANBERRA ACT 2601

Abstract

The literature describing the biomechanical response of children to impact is reviewed. The results of a computer simulation of a real child pedestrian impact using the computer program MADYMO are presented. The results of this simulation show good agreement with currently accepted tolerance levels for head impact in children.

Keywords children, head injury, accident, pedestrian, impact tolerance, computer simulation, biomechanics, anthropometry.

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STUDY OF HEAD INJURIES

IN CHILDREN

K.A. McCAUL

A.J. McLEAN

C.N. KLOEDEN

R.W. HINRICHS

NH&MRC Road Accident Research Unit,
University of Adelaide,
ADELAIDE, S.A. 5001.

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EXECUTIVE SUMMARY

The aims of this report were: 1) to review the literature describing the anthropometry of children and their biodynamical response to impact; 2) to simulate a real child pedestrian impact using the computer program MADYMO; and 3) to relate the results of this simulation to the injuries sustained by the child, with special reference to any head injuries sustained.

The computer program, MADYMO, is a versatile tool for examining the biomechanics of a pedestrian struck by a motor vehicle. It uses detailed data describing the anthropometry of the pedestrian and the force-deflections characteristics of individual body segments. Given this data, and additional information describing the shape and velocity of the striking vehicle, MADYMO will provide a graphical representation of the impact event, together with velocities and accelerations of, and forces acting on, chosen body segments.

While information describing the anthropometry of children is readily available, the literature review found very few sources which described the impact response of children. In addition, the biomechanical parameters obtained from these sources were not directly applicable to the MADYMO program. Consequently we were forced to use [redacted]

The child pedestrian suffered a skull fracture and brain injuries consistent with exposure to high levels of rotational acceleration. The simulation of this impact showed good agreement with these injuries. In particular, the forces acting on the head were sufficient to have produced a skull fracture and the rotational acceleration of the head was above currently accepted tolerance limits.

INTRODUCTION

The NH&MRC Road Accident Research Unit is conducting a long-term study of the mechanisms of brain injury in pedestrian crashes (Gibson et al., 1985). Data for this study, which commenced in 1983, are obtained from fatal pedestrian crashes occurring in the Adelaide metropolitan area. A significant proportion of the cases studied so far have involved children. This is not surprising since they are a high risk group for this type of accident. The small stature of a child pedestrian results in an impact event which is kinematically quite different from that which occurs when an adult pedestrian is struck by a passenger vehicle. Consequently, there are differences in the patterns of injury. The child is more likely to sustain thoracic or abdominal injuries, since those body regions are directly impacted, and is also more likely to be thrown in front of the vehicle, thereby increasing the risk of head injury from striking the road surface and also increasing the risk of being subsequently run over by the vehicle.

The purpose of this project is to review the literature on child impact tolerances by body region, with particular reference to the head, and to prepare a data set for the MADYMO computer simulation program which will adequately describe a collision between a child pedestrian and a car.

THE MADYMO PROGRAM

MADYMO is a crash victim simulation computer program which has been developed by the Research Institute for Road Vehicles of TNO, in The Netherlands (Maltha and Wismans, 1980). This program is a versatile tool for the investigation of the biomechanics of impact, for either vehicle occupants or pedestrians. The program requires input data that describe the relevant characteristics of the vehicle structure and the impact victim. The NH&MRC Road Accident Research Unit took delivery of the two-dimensional (2D) version of the program in 1983. Following some modifications to the dataset, the program has been used to simulate adult pedestrian accidents. The input data required to describe the characteristics of the pedestrian are of two types: anthropometric (e.g., overall height and lengths of body segments, and weight ; these dimensions are usually obtained at autopsy) and biomechanical (e.g., estimates of the stiffness of joints and body segments obtained from a review of the relevant literature).

This two dimensional model had been shown previously to provide an adequate simulation of the pedestrian/vehicle collision when a 7 segment model of the pedestrian was used (van Wijk et al., 1983). The NH&MRC Road Accident Research Unit has developed a 9 segment model of the pedestrian with stiffness characteristics closer to those of an adult pedestrian. This model has been validated by Gibson et al. (1986) using results from cadaver tests reported by Backaitis et al. (1983) and Cesari et al. (1985).

Initially we thought that a simple collision between a car and a pedestrian would be able to be modelled adequately using a 2D simulation. However, it was apparent in the cadaver tests conducted by Cesari et al. (1985) that when a pedestrian is struck on the side by the front of the car there is considerable rotation of the upper torso about its longitudinal axis before the head impacts the car. Furthermore, we became increasingly interested in being able to measure the rotational acceleration imparted to the head on impact. For these reasons we obtained a copy of a three-dimensional (3D) version of MADYMO which was released in 1985/86 (Wismans et al., 1985). While the 3D version does enable us to carry out a more realistic simulation of the pedestrian/car collision it also places greater demands on the user to provide more detailed input data.

The human body as represented by MADYMO comprises nine segments: the head, neck, thorax, abdomen, pelvis, thighs and lower legs (see Appendix). For each of these segments the following parameters are needed: length, mass, the location of the centre of mass and the moments of inertia about each of the major axes. The response to impact for each segment is defined by the user in terms of force-deflection relationships. The user also defines joint stiffness characteristics in force-angular deflection terms.

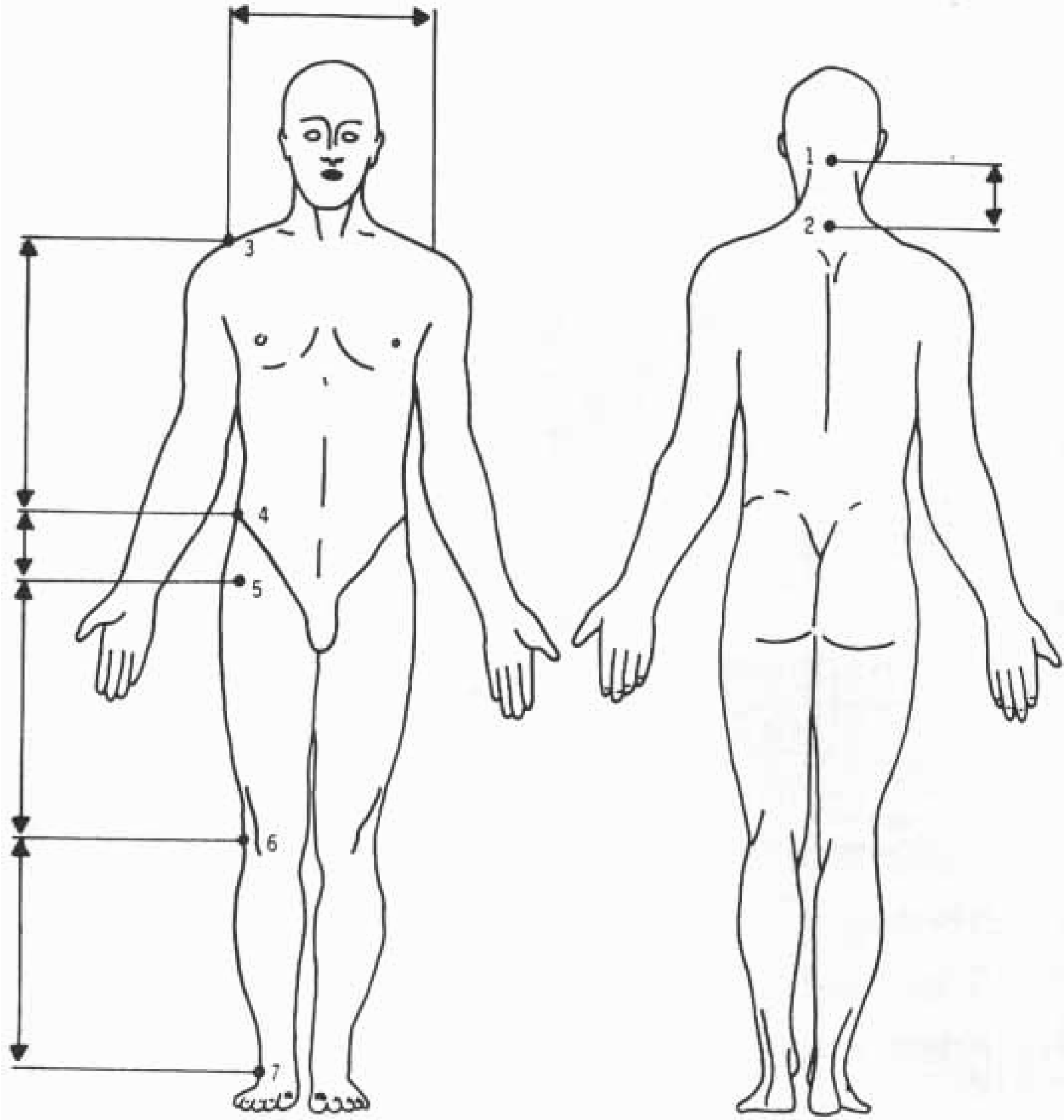
The length of each body segment can usually be obtained by direct measurement. When this is not possible estimates can be obtained from published data from anthropometric surveys. However, in a review of sources of child anthropometric data, Snyder et al. (1974) noted that there was a complete lack of data for a number of functional measures which would be required by industrial designers. Subsequently, Snyder et al. (1977) conducted a survey which provided much of the anthropometric data required by the MADYMO computer program (Figure 1).

The mass of each body segment can be estimated using regression equations derived by Jensen (1986). This paper also provides regression equations for the radius of gyration of various body segments from which estimates of moments of inertia about the three principal axes can be derived.

BIOMECHANICAL DATA RELEVANT TO CHILDREN

While the anthropometric requirements of the MADYMO program can be met reasonably well by direct measurement, supplemented by the current literature, the data required to describe the impact response of each body segment is much more difficult to obtain.

The response of living humans to impact can only be investigated experimentally at non-injurious levels. The available biomechanical data on the response of the human body to injurious levels of impact has been primarily obtained from tests conducted on cadavers. However, the data obtained from such tests do not reflect the true biodynamic response of a living human because of factors such as the absence of any muscle activity in cadavers. The data obtained from cadaver tests is also not



REFERENCE POINTS:

1. Occipital Condyle.
2. Seventh Cervical Spine.
3. Acromioclavicular Joint.
4. Iliac Crest.
5. Hip Joint (Greater Trochanter)
6. Femoral-Tibial Joint.
7. Heel.

Figure 1: Anthropometric measurements

representative of the general population since most cadavers used in these tests are those of older adults. There has been comparatively little work done using child cadavers and consequently the literature from which child biomechanical data can be obtained is quite sparse.

Sturtz (1980) carried out an extensive review of the medical and technical literature relating to the biomechanical response of children to impact. However, very few of the force-deflection characteristics required by the MADYMO program are available in this review. There is information on the results of impact tests on the 6th and 7th ribs of children under the age of 14 years which indicated an average breaking load of 234N at a deflection of 15 to 20 mm. Sturtz also reported the results of post-mortem seat-belt tests conducted on child cadavers which indicated that forces of 4000 to 5400 N were generated without any injury being observed. No estimates of the deflection caused by these impacts was given however.

Curry and Butler (1975) conducted a series of tests on the femurs of 18 subjects ranging in age from two to 48. They concluded that although the femurs of children had a lower modulus of elasticity and lower bending strength than adults, they did tend to deflect more before breaking. They also absorbed more energy after fracture than did adult femurs.

Martin and Atkinson (1977) carried out tests on four cross-sectional specimens from the left femurs of 37 cadavers. These cadavers ranged in age from two and a half to 82 years but only three were children (15 years or younger).

The biomechanics of head impacts suffered by children in falls was examined by Mohan et al. (1979). Thirty cases involving children 10 years of age or younger who had fallen head first onto a flat surface were examined, with 6 of these cases being simulated using the MVMA Crash Victim Simulation computer program (Bowman et al., 1979). It was assumed that the stiffness of child skulls was less than the stiffness of adult skulls because calcification of the human skull continued until adulthood. The authors estimated child skull dynamic stiffness by assuming the following: that the static loading curve for a child's head would be of a similar shape to that of an adult; the relationship between static loading and dynamic loading would be the same for children as that already determined for adults; the deflection at the end of the first static loading segment (the elastic range) would be proportional to the length of the head; and that the ratio of child static skull stiffness to adult static skull stiffness could be estimated as a function of age. Mohan et al. presented a curve of age-dependent static head stiffness which indicated that by the age of 7 years a child would have a static skull stiffness which would be 70 per cent of that of an adult. By the age of 13 years this would have increased to 90 per cent. Head accelerations at impact were also calculated and 200-250g was suggested as a tolerance range for AIS = 2 injuries.

Sturtz (1978) investigated 10 child pedestrian crashes and simulated them using an Alderson VIP 6 dummy. From the results of these tests he suggested a head impact acceleration of 50 g for 15-20 msec as a tolerance level associated with unconsciousness for a duration of less then 15 minutes.

SIMULATION OF A CHILD PEDESTRIAN CRASH

A child pedestrian crash has been chosen for simulation from the files of the NH&MRC Road Accident Research Unit Head Injury Study. The vehicle involved was a Datsun traytop utility. The child, an 8 year old male, ran across the road and into the path of the oncoming vehicle which was estimated to be travelling at 50km/h at impact.

The child was struck on the left side and sustained a skull fracture in the left anterior cranial fossa; a closed fracture of the left radius and ulna; a lacerated left lower lobe of the lung; extensive lacerations of the left kidney; some bruising on the outside of the right kidney; a left peritoneal haematoma and a ruptured spleen. Externally there were abrasions distributed over most parts of the body, mostly caused by contact with the road. A neuropathological examination of the brain was conducted and revealed, macroscopically, a laceration of the left inferior frontal lobe; scattered patchy cortical contusions, the largest involving the superior surface of the left frontal lobe; haemorrhages in the left frontal lobe, corpus callosum and the rostral pons; and small bilateral gliding contusions (bridging vein ruptures). A microscopic examination of the brain did not reveal any additional significant injuries. The methodology of the neuropathological examination is described in Gibson et al. (1985).

All of the major injuries were probably caused by the initial impact with the vehicle rather than by contact with the road surface.

Damage to the vehicle was confined to the front driver's side corner. There was a large dent, 180 x 240 mm, in the bonnet of the vehicle centred approximately 200 mm from the right front fender and 120 mm from the leading edge of the bonnet. There was no displacement of the leading edge of the bonnet itself. The bumper bar was dented, the point of maximum deflection being 120 mm from the right hand end of the bumper bar. The height of the lower edge of the bumper bar was 535 mm, the upper edge 635 mm, and the height of the bonnet leading edge was 865 mm.

The child weighed 21 kg and was 1190 mm tall. The body segment lengths, mass distribution and moments of inertia used in the MADYMO simulation are listed in Table 1.

As already stated in the preceding section, the available literature does not provide adequate data on either the force-deflection or the joint-stiffness characteristics required by MADYMO. In order to proceed with a simulation of a real child pedestrian accident the joint and body segment stiffness characteristics of a child anthropometric dummy have been used.

TABLE 1: Length, Mass and Moments of Inertia for each Body Segment.

Body segment	Length (mm)	Mass (kg)	I_X (Kgm^2)	I_Y (Kgm^2)	I_Z (Kgm^2)
Head	139	2.16	.0041	.0041	.0041
Neck	94	0.75	.0008	.0008	.0008
Thorax	168	6.05	.0075	.0075	.0023
Abdomen	127	2.64	.0034	.0034	.0010
Pelvis	63	2.40	.0059	.0059	.0018
Thigh	286	2.06	.0129	.0129	.0032
Lower leg	312	1.44	.0158	.0158	.0040

SIMULATION RESULTS

The graphical output from the MADYMO simulation of this case is presented in the appendix to this report. The kinematic representation is for the first 200 msec of the impact event calculated at 10 msec intervals.

Graphs of the resultant linear acceleration and the resultant angular acceleration of the head are also presented, together with the resultant force on the head due to impact with the bonnet, the resultant torque at the top of the neck (occipital condyles), and the resultant velocity of the head during the impact event. A summary of the maximum values for each of these parameters is given in Table 2.

The values in Table 2 are consistent with the head injuries observed at autopsy. The 214 g maximum head acceleration is within the 50-300 g tolerance range for adult brains (Interdisciplinary Working Group for Accident Mechanics, 1986). Sturtz (1980) used scaling techniques to estimate tolerance limits for 6 year old children. Using tolerance criteria derived from the Wayne State Curve, which stipulates that head acceleration should not exceed 80 g for more than 3 msec, Sturtz suggested a tolerance limit for 6 year olds of 82.1 g. The Wayne State Curve is, however, based on data derived from tests on animals (dogs), cadavers, and human volunteers (Newman, 1980) and therefore any criteria derived from it should be treated with some caution.

TABLE 2: Maximum Values of Kinematic Variables

Kinematic variable	Time	Maximum value
Linear acceleration of head	46.25 msec	2096 m/sec/sec (214 g)
Angular acceleration of head	46 msec	24973 rad/sec/sec
Force on head during impact with bonnet	46.5 msec	3451 N
Torque at occipital condyles	36 msec	81 Nm/rad
Velocity of head	97.25 msec	19.8 m/sec

The maximum rotational acceleration of the head occurs when the head strikes the bonnet of the vehicle. The computed value of 24973 rad/sec/sec far exceeds the cerebral concussion tolerance limit of 1800 rad/sec/sec suggested by Ommaya and Hirsch (1971) for adults. Sturtz (1980) has derived, via scaling techniques, a value of 1823 rad/sec^2 as a tolerance limit for a 6 year old child. If the value computed by MADYMO is correct, the child should have sustained severe brain injuries of the type normally associated with rotational acceleration: diffuse axonal injury and micro-haemorrhages. However, diffuse axonal injury develops with time and is, therefore, more evident in head injury victims who survive for much longer periods than the one hour survival time in this case. The macroscopically evident petechial haemorrhages in the corpus callosum, and the bilateral gliding contusions are, however, indicative of rotationally induced injury (Lowenhielm, 1973).

The MADYMO output includes an estimated maximum torque of 81 Nm/rad at the occipital condyles. Mertz and Patrick (1971) computed whiplash tolerance limits of 57 Nm for extension and 190 Nm for flexion from experiments conducted on adult volunteers and cadavers. Based on these tolerance limits the computed torque at the occipital condyles would be unlikely to have produced significant injury.

The force on the head at impact, as calculated by MADYMO, was 3451 N which exceeds the minimal force for fracture tolerance of 2000 N for the temporo-parietal region of adult skulls derived experimentally by Schneider and Nahum (1972). Consequently, the force predicted by MADYMO was sufficient to produce a skull fracture in the left anterior fossa. The laceration of the left inferior frontal lobe of the brain is associated with this fracture.

CONCLUSION

A review of the literature related to child impact biomechanics has failed to produce meaningful data from which body region and joint stiffnesses can be determined. The body region and joint stiffness data from a child anthropometric test device has therefore been used to provide the necessary data for the MADYMO Crash Victim Simulation computer program. A real child pedestrian crash has been simulated using MADYMO. The results of this simulation are consistent with the injuries observed at the autopsy. The results of the detailed neuropathologic examination of the child's brain have revealed significant injuries which have been associated with exposure to high levels of rotational acceleration.

Further work on the simulation of child pedestrian crashes as a means of investigating the mechanisms of brain injury to child pedestrians, will involve comparison of body region and joint stiffness data based on similar data from adult cadavers with the stiffness data derived from anthropometric dummies.

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APPENDIX

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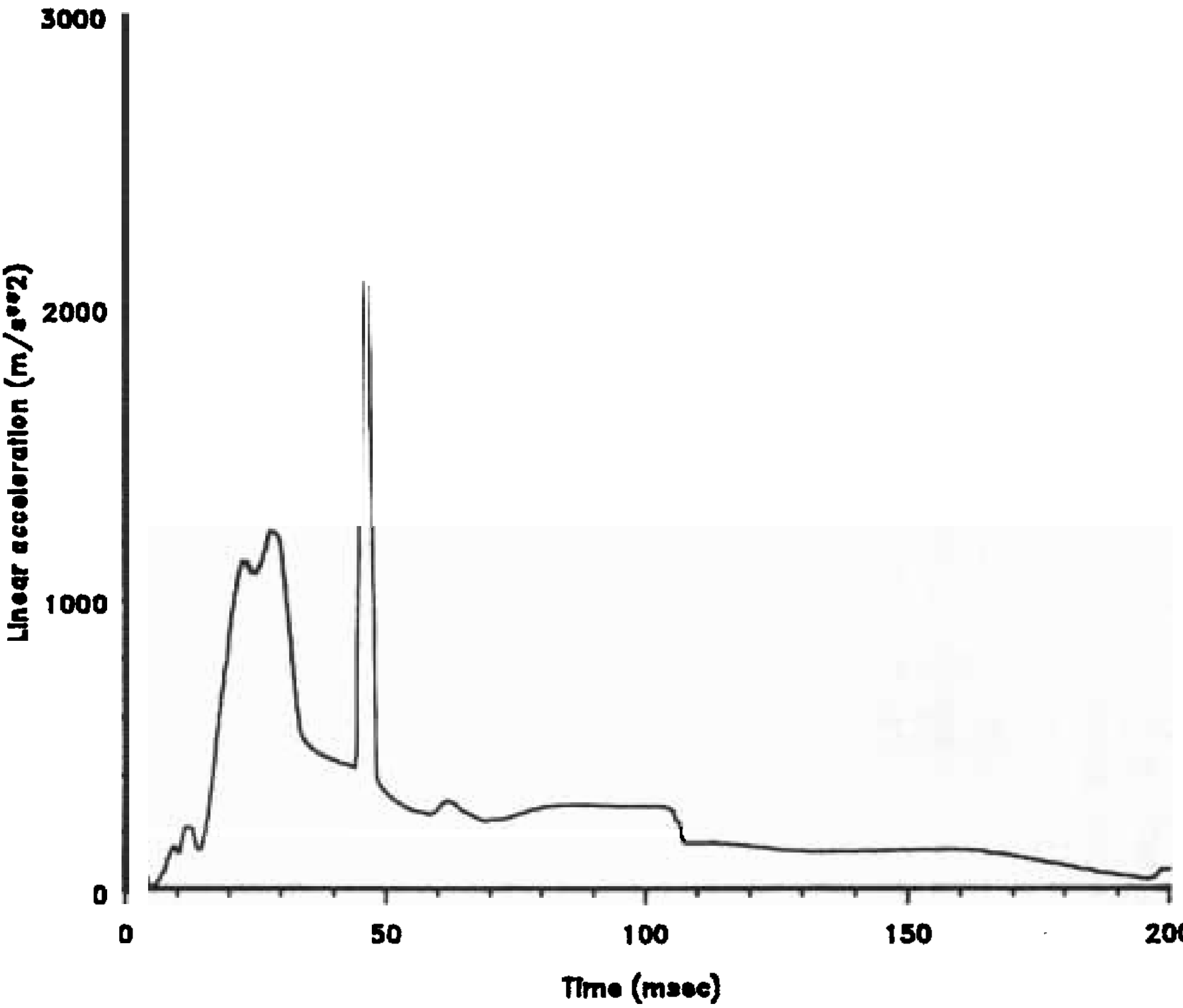
RESULTANT LINEAR ACCELERATION OF HEAD



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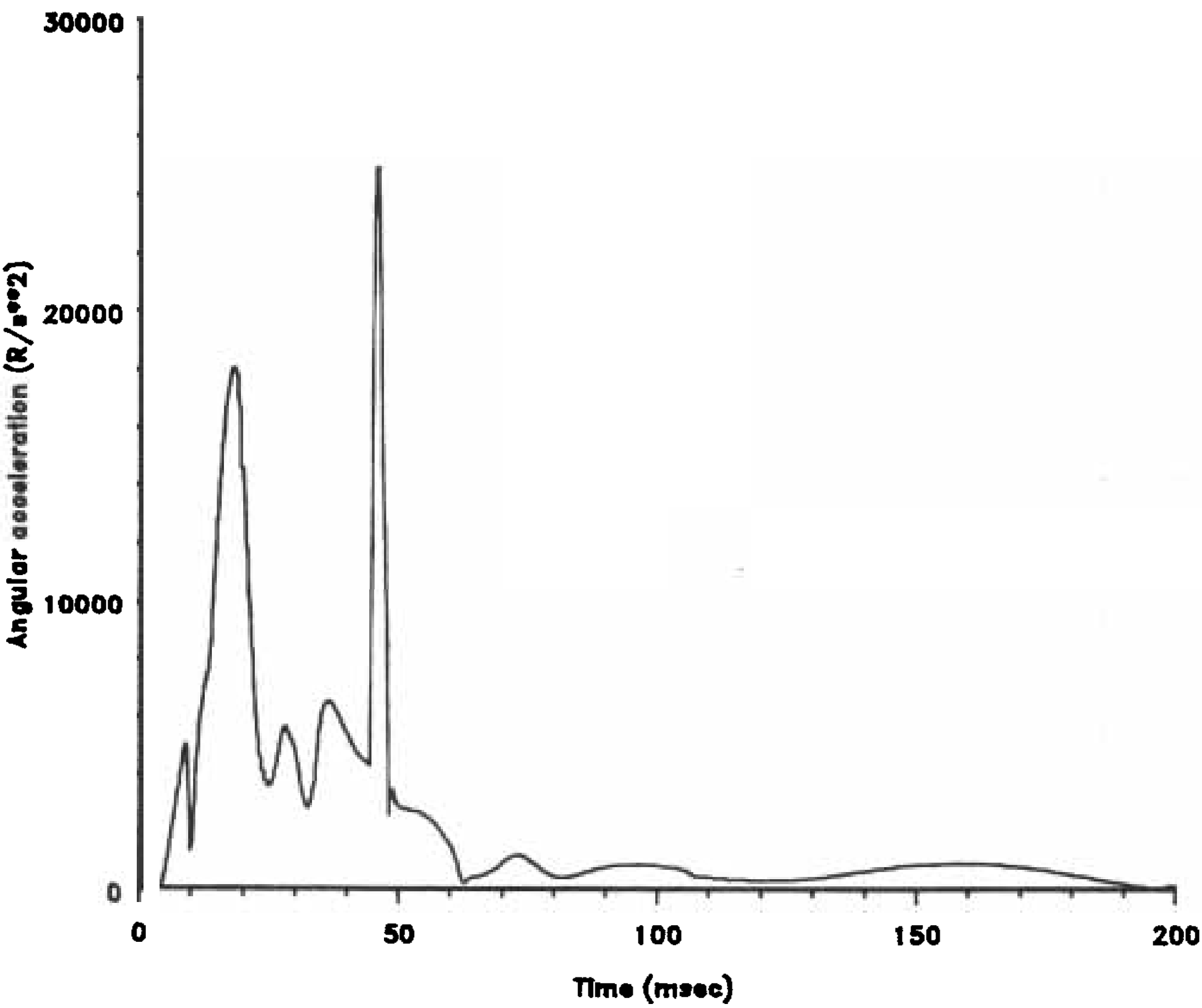
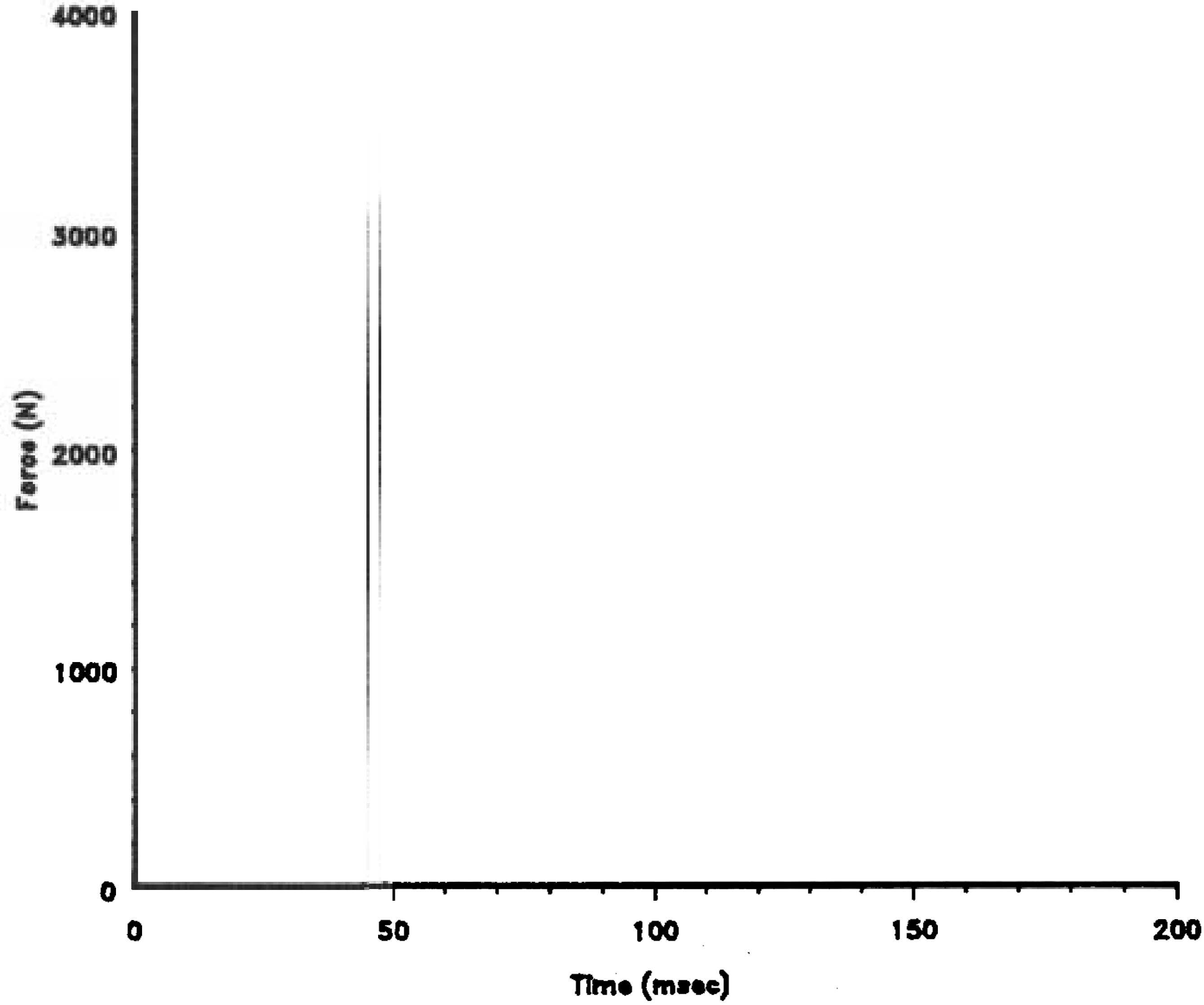
RESULTANT ANGULAR ACCELERATION OF HEAD



H024-87
RESULTANT FORCE OF HEAD ON HOOD



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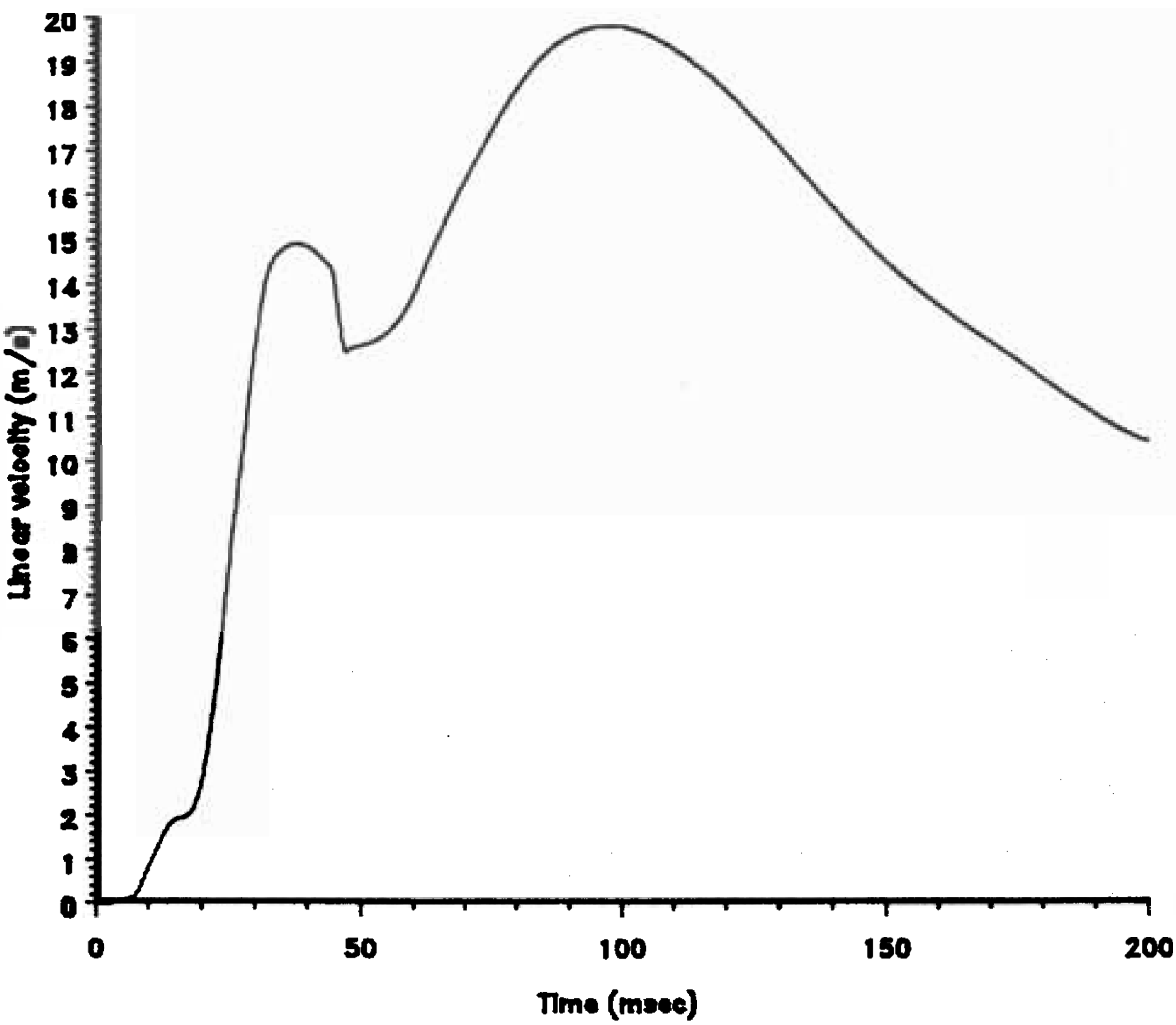
RESULTANT LINEAR VELOCITY OF HEAD



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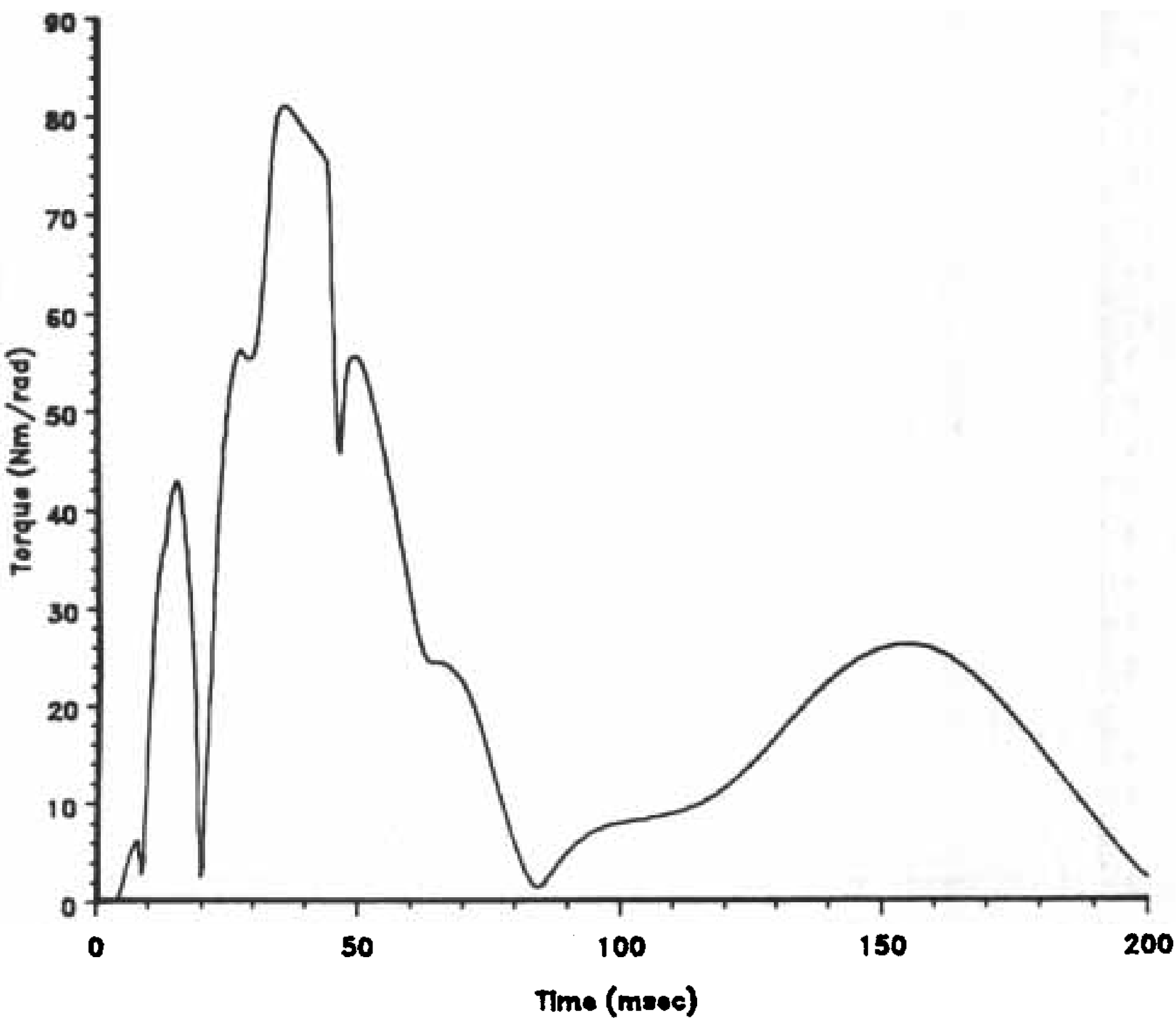
RESULTANT TORQUE AT TOP OF NECK



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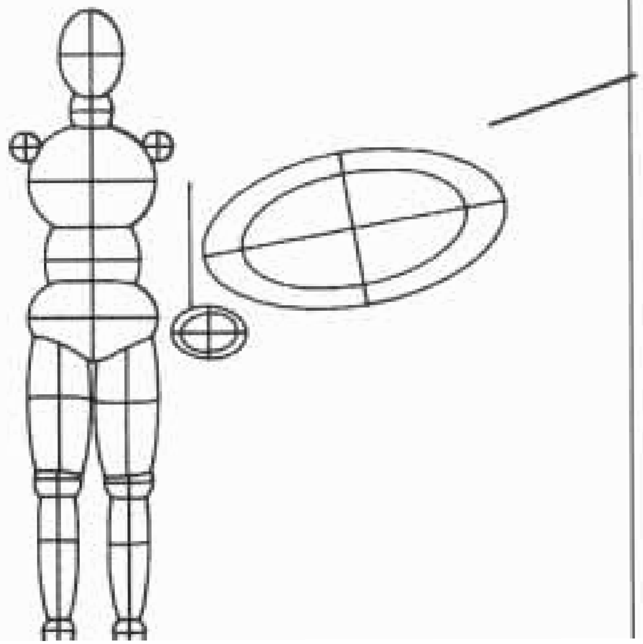

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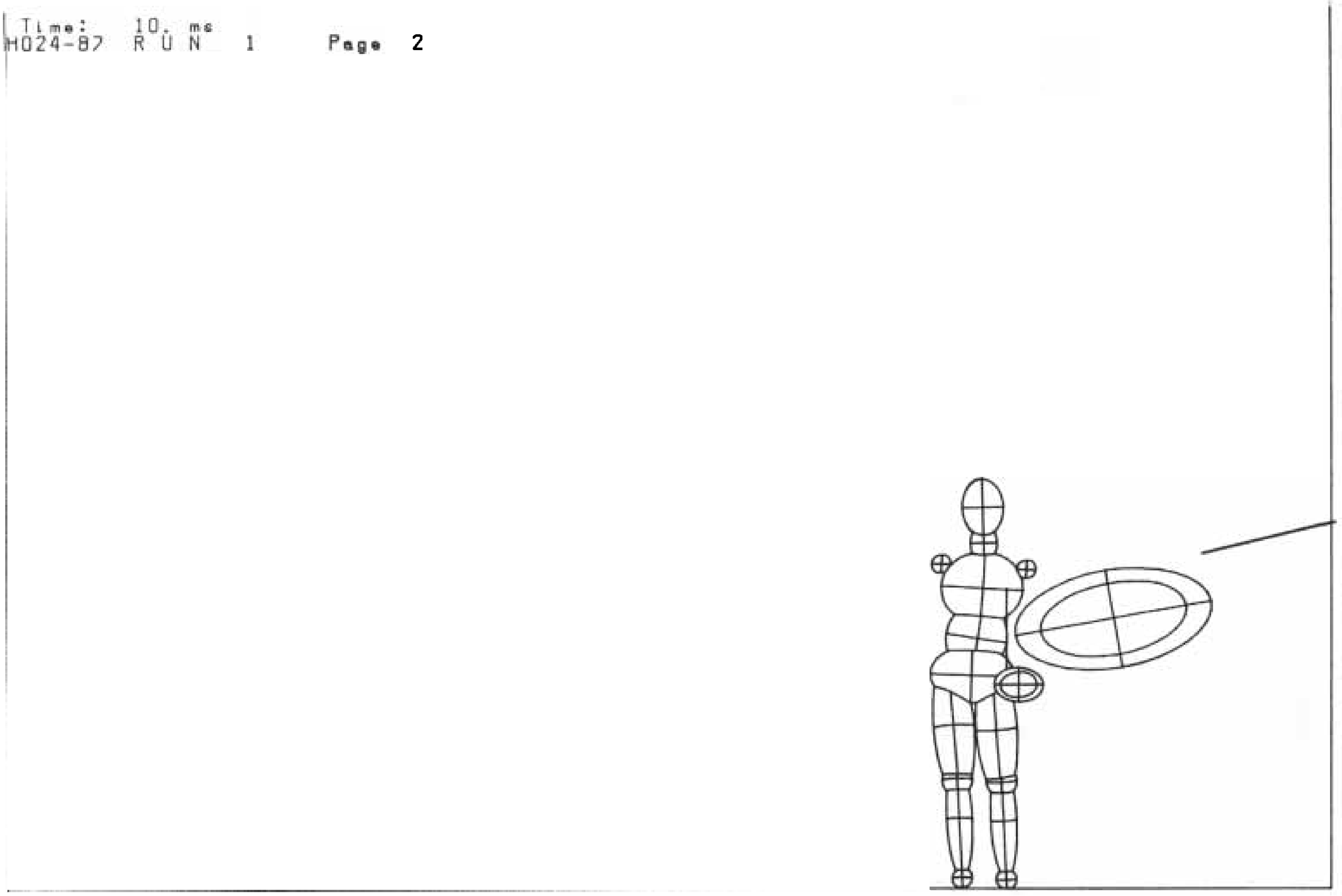

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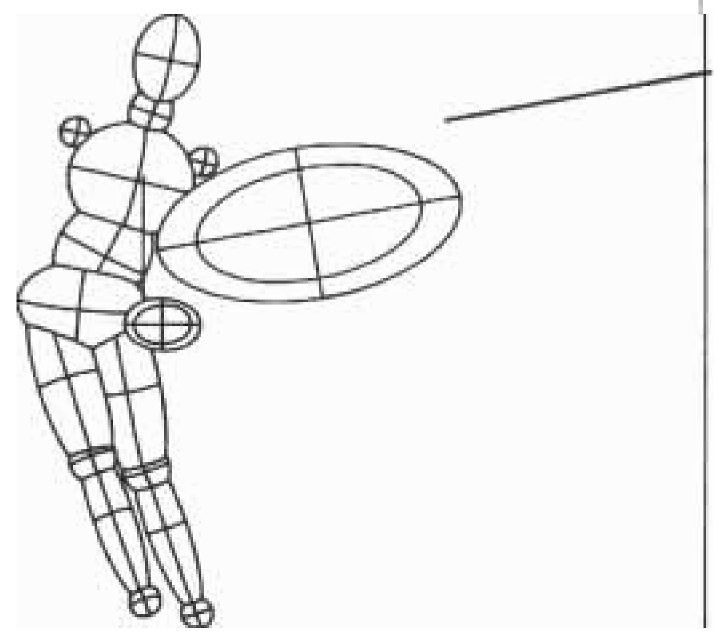

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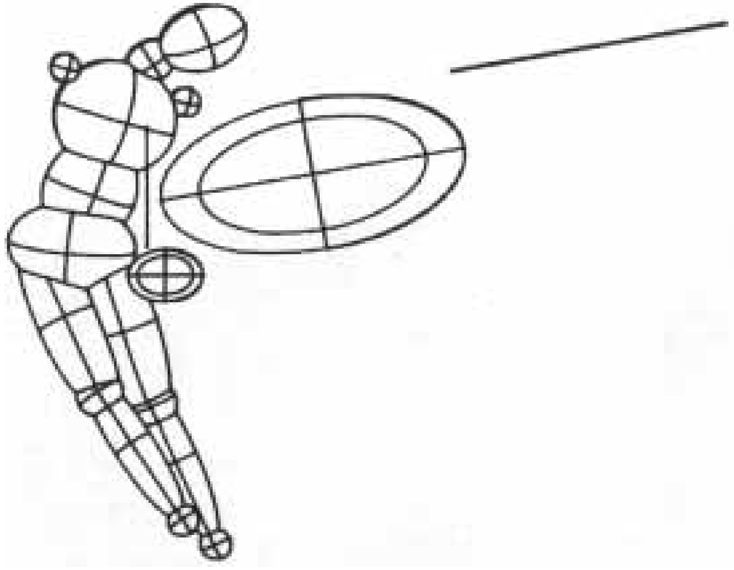

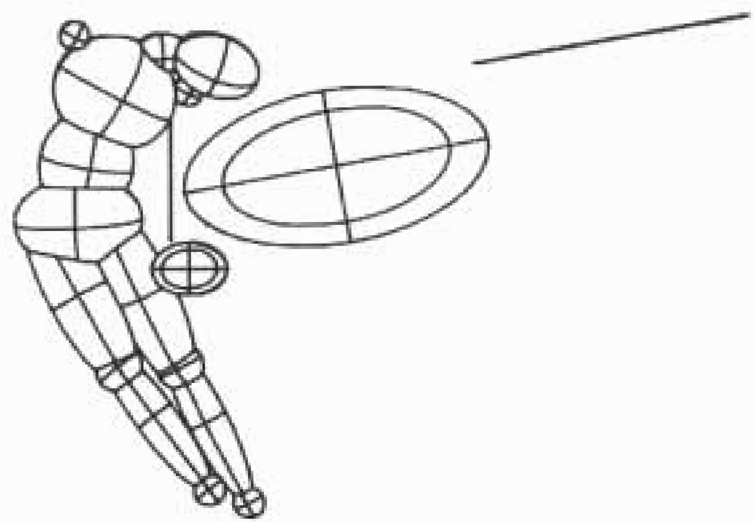

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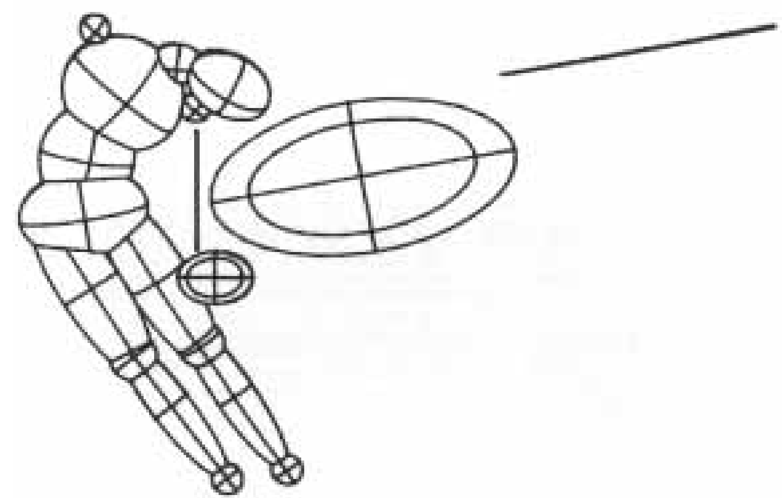

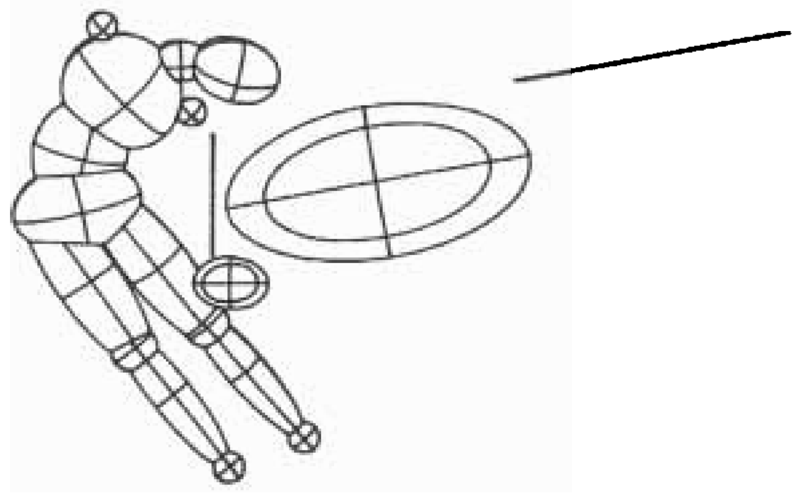

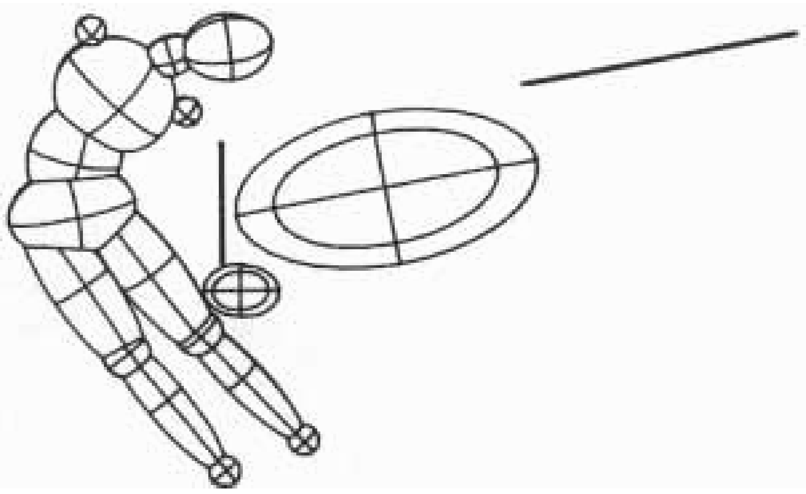

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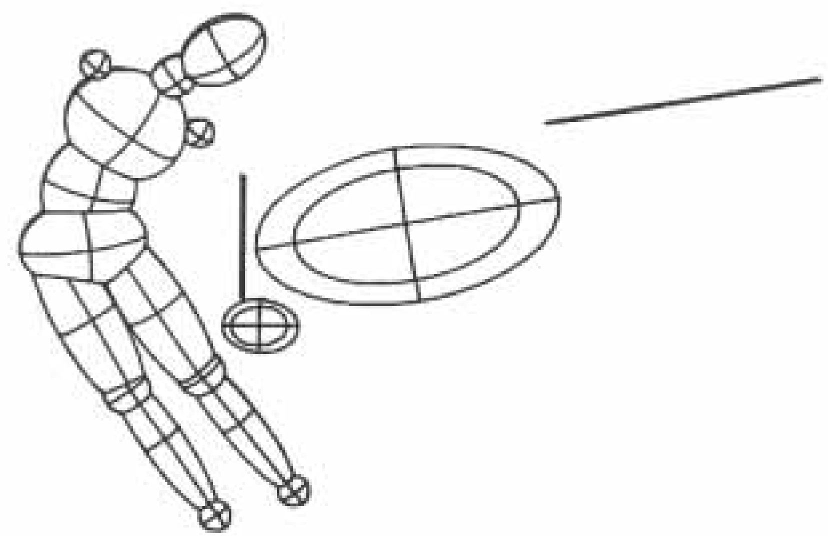

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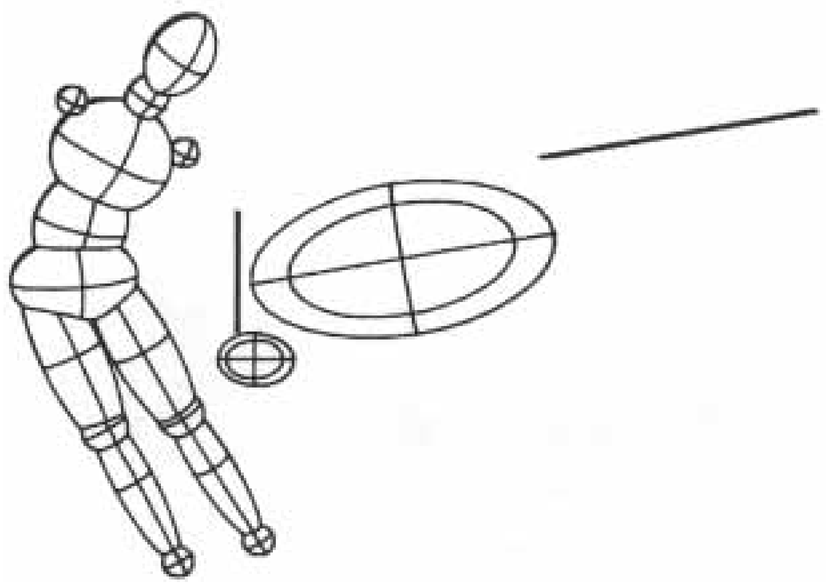

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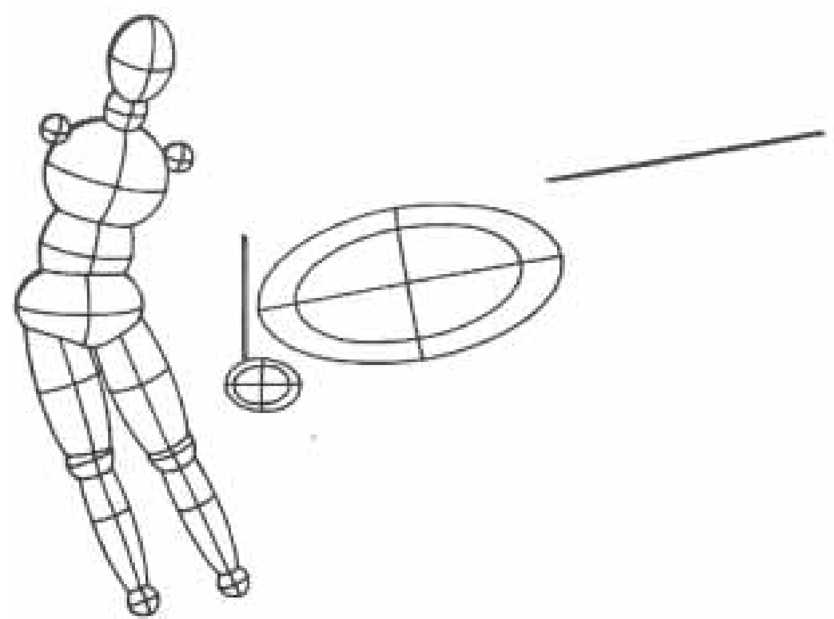

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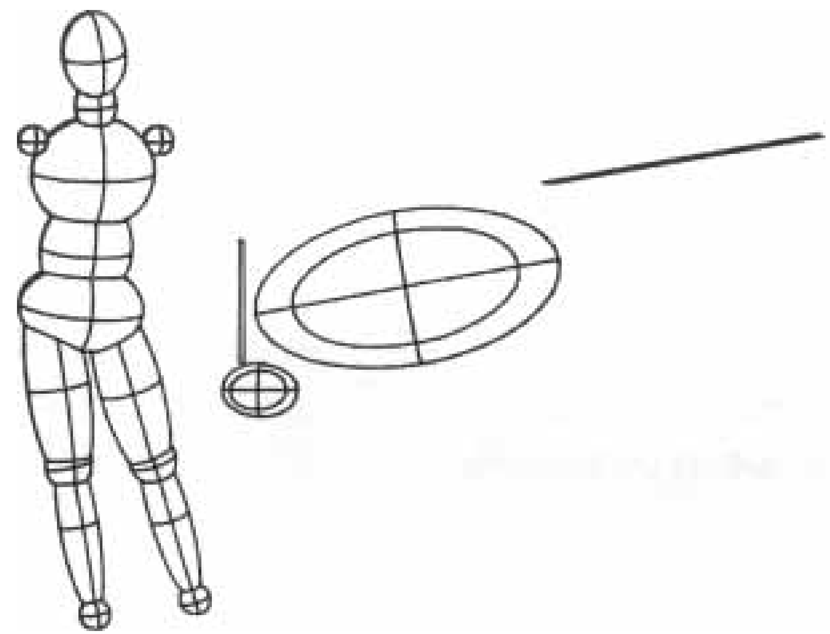

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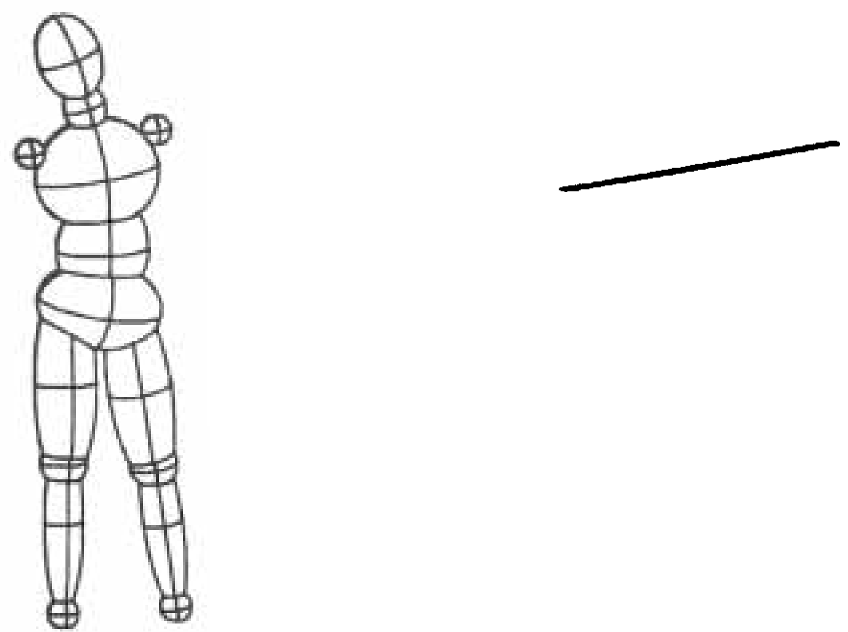

Time: 130. ms
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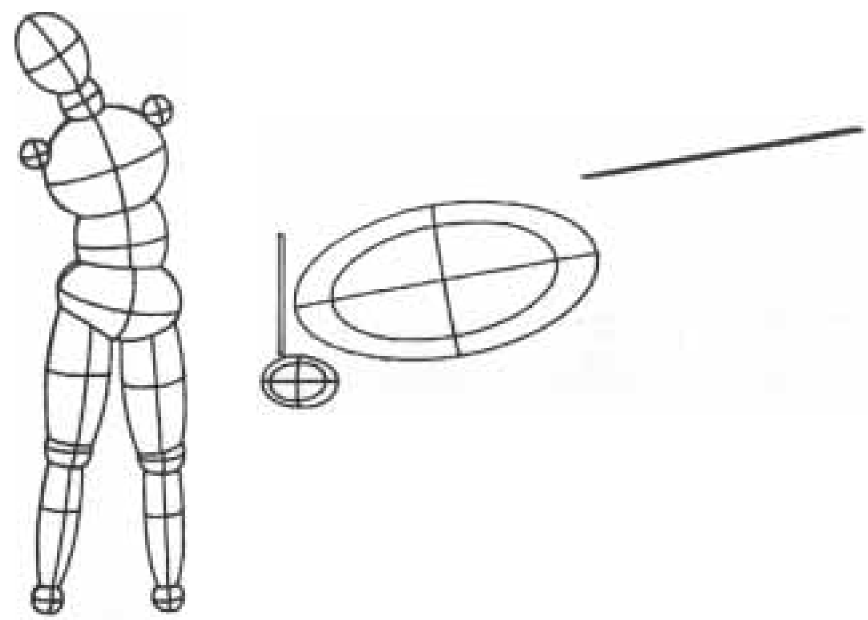

Time: 140. ms
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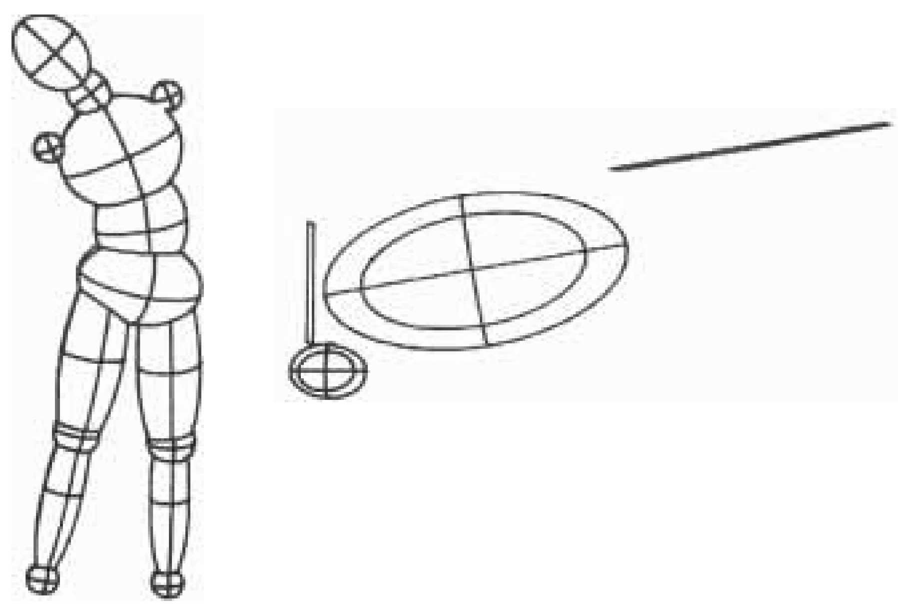

Time: 150. ms
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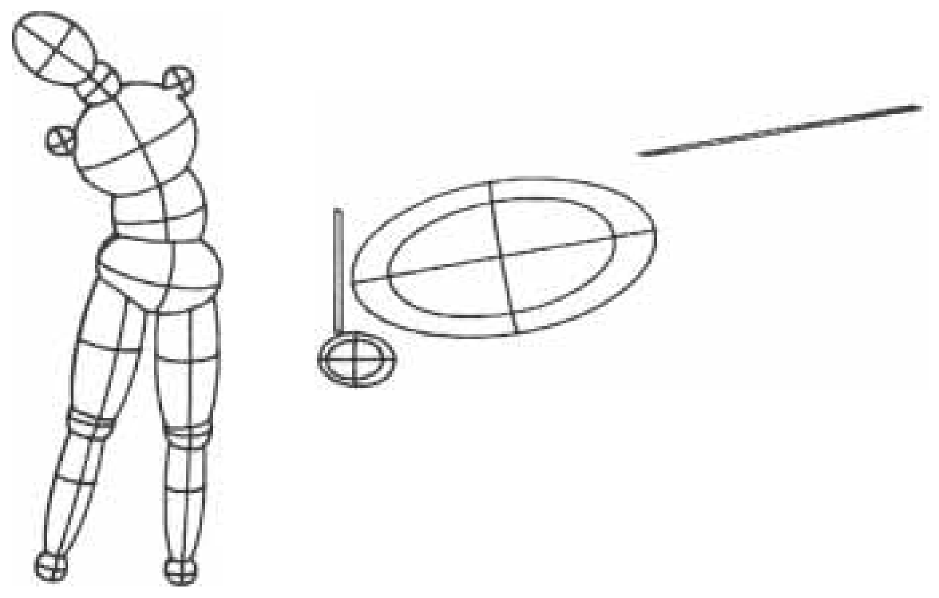

Time: 160. ms
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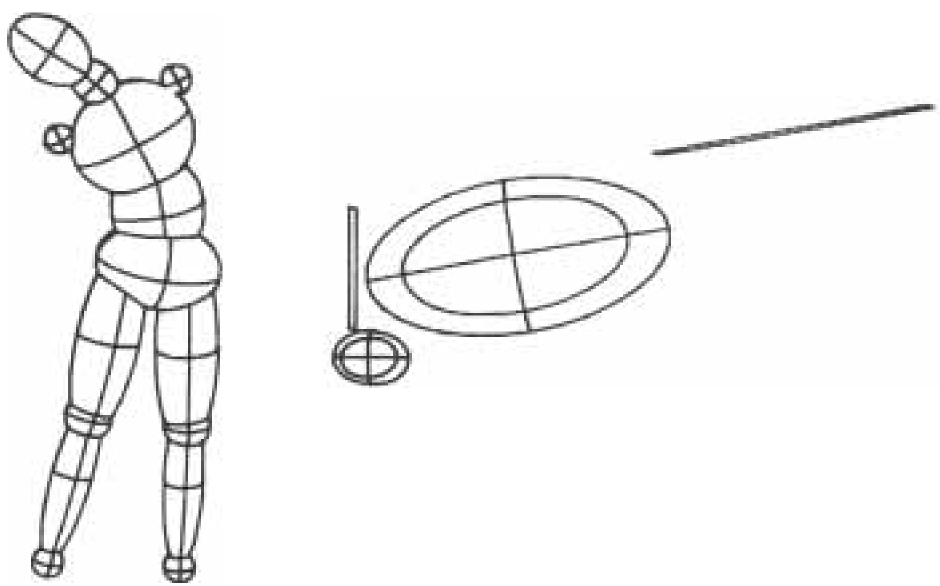

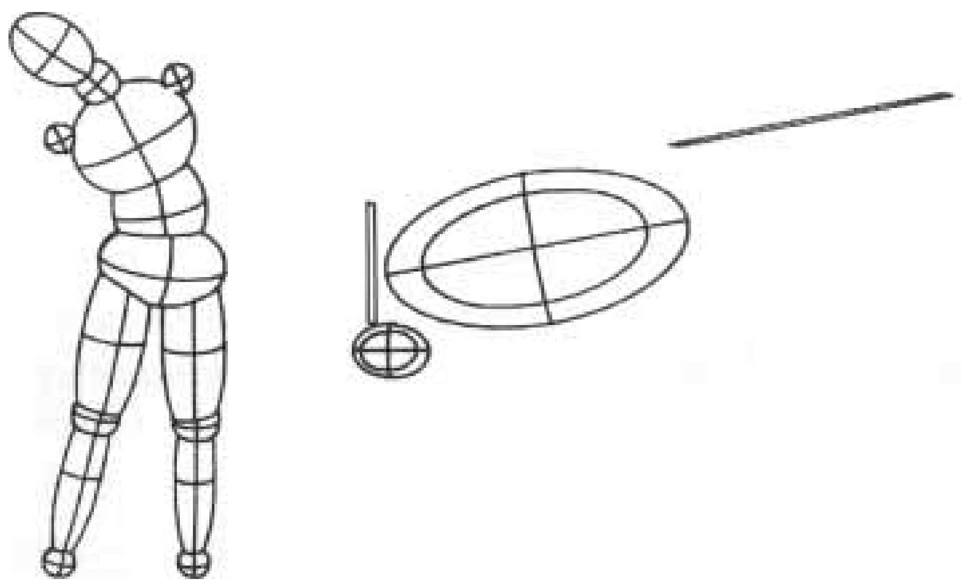

Time: 180. ms
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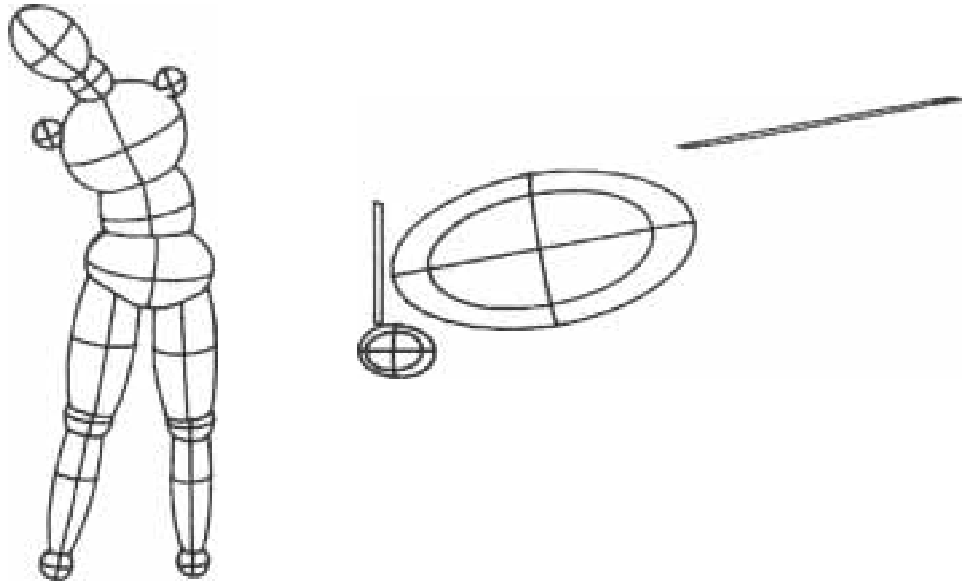

Time: 190. ms
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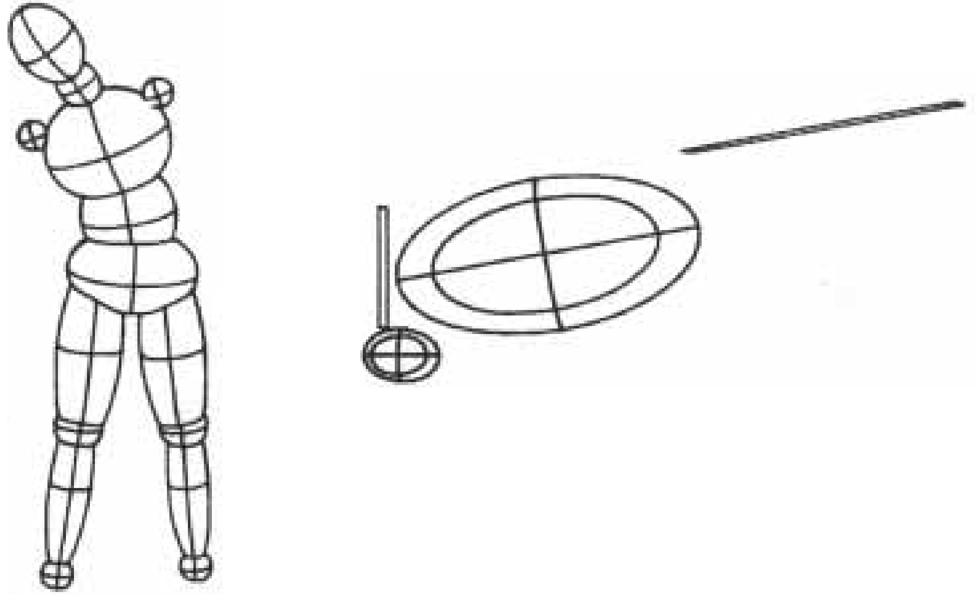

Time: 200. ms
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