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HELMET IMPACT TEST SYSTEM DEVELOPMENT

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FINAL REPORT

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THE UNIVERSITY OF MICHIGAN
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HELMET IMPACT TEST SYSTEM DEVELOPMENT

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16. Abstract

Protective head gear is currently evaluated using several ANSI guidelines, such as ANSI 789.1. The rigid headform specified in this standard does not realistically represent the head-neck-torso complex of the potential helmet wearer; further, the impact conditions specified therein do not simulate the variety of impact conditions encountered in actual industrial accident.

The objectives of this project were to: 1) examine the literature to define the required impact characteristics, 2) identify the unavailable data then conduct tests to obtain such data, 3) develop a helmet impact test system based on acquired response data, and finally, 4) construct such a device and deliver it along with its user's manual.

These objectives were met, and the work leading to the final goal is described in this final report. The device itself is delivered seperately along with all engineering drawings and the assembly and operation manual.

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PREFACE

Protective head gear is essential in preventing head injuries in a hazardous environment where the head may be subjected to various mechanical and/or electrical shocks. The worthiness of safety helmets is currently evaluated using several American National Standards Institute (ANSI) guidelines.

The procedure for testing industrial safety helmets, as specified in ANSI Z89.1, has several shortcomings. First of all, the rigid headform specified in this standard does not realistically represent the head-neck-torso complex of the potential helmet wearer. Second, by limiting the impact testing to the vertex of the head in the superior-inferior direction, this procedure disregards other hazardous situations which may be just as serious. Finally, even if more realistic headforms and impact modes are introduced, the injury mechanisms of the head or cervical and thoracic spines are ignored by using the peak transmitted force as the only tolerance criterion.

A test system which simulates the response of a fiftieth-percentile adult male to impacts at any location above a plane 2.5 cm above the basic head anatomical (Frankfort) plane is therefore needed. Based upon this need, NIOSH contracted the Highway Safety Research Institute (HSRI) of the University of Michigan to develop a helmet impact test system.

The contract specifies that the work shall be done in 5 phases: (I) Examine the literature to define the required impact characteristics;(II) identify the unavailable but needed data, then conduct tests to obtain such data; (III) propose three levels of impact test system sophistication both in software and hardware;(IV) construct and validate the most feasible of the proposed three systems; and finally, (V) deliver the system and its documentation.

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1. PHASE I - LITERATURE SURVEY

This is the final report on Phase I of the Helmet Impact Test System Development project. The objectives of this phase are outlined in section 1.1. The mechanisms of head, neck, and upper-torso injuries are briefly discussed in the next section. In section 1.3, the requirements of the desired system are discussed, followed by a discussion of the injury tolerance criteria and the unavailable data. The findings of this phase are summarized in section 1.6 and a bibliography list is given in the last section.

1.1. OBJECTIVES

A test system that simulates the response of a fiftieth-percentile adult male to impacts at any location 2.5 cm above the basic plane is needed to realistically evaluate the impact resistance of industrial helmets.

The first phase in developing such a system is to examine the available research literature with two objectives in mind. The first objective is to define the response requirements of the desired system. This involves an understanding of the dynamics of typical impacts to the helmeted head, the biomechanics of the resulting head-neck-torso injuries, and the correlation between impact descriptors and injury patterns.

The second objective is to determine which of these response requirements have already been developed, through previous research efforts, and which responses are needed to complete the data base necessary for designing the desired test system. This would provide guidelines for designing, conducting, and analyzing the experiments in subsequent phases of this project.

1.2. HEAD-NECK-TORSO INJURY MECHANISMS

The objectives of Phase I of this research program are best served by understanding the individual injury mechanisms of the head, neck, and upper torso. In the following discussion, the types of injuries which are most likely to occur as a result of head impacts are emphasized. The intent of the discussion is to introduce the reader to some aspects of the problem rather than present an in-depth analysis.

1.2.1. Head Injuries

These may be soft-tissue (scalp) injuries, skull fractures, and brain and brainstem injuries. In an industrial environment where a worker is wearing a protective helmet, superior-inferior (S-I) impacts occur most frequently when the helmet is struck by a falling object. Since helmets act as load distributors. remote linear skull fractures are the most likely types of head injuries to occur.

Remote linear skull fractures occur when the applied forces of impact are well distributed, causing cranial shell bending and creating tensile stresses away from the point of impact. These can be so excessive that a crack in the skull is initiated and propagated.

In many cases, skull fractures are not considered serious injuries by themselves. Nontheless, they serve as indicators of the severity of injury, since they are often associated with brain injuries.

Three types of brain injuries are common: laceration, contusion and concussion. Cerebral laceration may be caused by direct invasion of the cranial cavity by foreign objects or by violent motions of the brain relative to the skull. Contusion is characterized by ruptures of small blood vessels. Cerebral concussion, which is usually associated with unconsciousness, is the least severe because it is often reversible.

The exact mechanisms of brain injuries today remain unknown in most of the cases. However, researchers are able to relate, with some success, a given type of injury to certain loading modes of the head. This is useful since it allows the safety engineer to design protective devices for specific types of impact hazards.

1.2.2. Cervical Injuries

The neck is perhaps the weakest link in the head-neck-torso complex structure. The mechanisms of acute cervical spine injuries may be classified in four categories.

1) FLEXION of several types, which may produce subluxation and bilateral interfacetal dislocation, which are principally soft tissue injuries; simple wedge fracture (anterior compression of the vertebral body); Clay-shoveler's fracture of the spinous process (usually C6 or C7); and the most serious type of tear-drop fracture, where a triangularly-shaped, separate fragment is displaced and may impinge upon the surface of the cervical spinal cord.

2) FLEXION-ROTATION, which may produce "locked" or "perched" vertebra. This type of injury refers to the anterior dislocation of the inferior facet of the involved vertebra with respect to the superior facet of the one below.

3) VERTICAL COMPRESSION, which produces bursting fractures. Least common of these is the "Jefferson" anterior and posterior fracture of the ring of C1, and the bilateral displacement of its lateral masses. The most common vertical compression occurs in the mid or lower cervical segments and is caused by intervertebral disc material being impelled through an end-plate into the vertebral body, causing it to burst. The posterior fragment is displaced and may impinge upon the spinal cord.

4) EXTENSION, which may produce as simple an injury as a fracture of the posterior neural arch resulting from compression during maximum extension, or an injury as serious as the tear-drop fracture of an upper segment, usually C2, in which the triangularly-shaped fragment is pulled away from the main vertebral body.

In most cases, cervical spinal injuries can be determined with radiographic examination of the neck; however, it is possible for a spinal cord damage to be present in the absence of radiographic evidence of vertebral fracture or dislocation. The use of x-ray diagnosis simplifies the experimental studies of injuries using cadavers and permits the correlation of the exerted forces and resulting injuries, leading to the establishment of tolerance levels and injury criteria.

1.2.3. Upper Thoracic Spine and Torso

It is conceivable that a heavy load imposed on the top of the head would push the head-neck structure inferiorly into the upper torso, causing damage to the upper thoracic spinal column and possibly fracturing the clavicles. For this to occur, the loads must be so large that head and neck injuries would have occurred in the process. Therefore, a conservative tolerance limit should be based on injuries to the head and neck and not on those to the upper torso.

The importance of the upper torso and thoracic spine in the current project stems from the role they play in the production of neck injuries. During impact to the helmeted head, the forces applied through the occipital condyles must be countered by reaction forces applied by the upper thoracic vertebrae to the neck. Therefore, the compliance of the upper thoracic spine plays an important role in absorbing the energies transmitted through the neck.

1.3. PARAMETERS OF THE TEST SYSTEM

The response of the human head, neck, and upper torso to impacts at any location 2.5 cm above the basic plane may be characterized by the following dynamic variables: (1) the contact forces, controlled by the impactor weight and velocity and dynamically measured using force transducers,(2) the location, direction and distribution of the forces transmitted to the head through the protective helmet,(3) the gross kinematics of the head, described by angular and translational accelerations, velocities, and displacements,(4) the shear and axial forces and the moments applied by the head to the neck at the occipital condyles, and(5) the spinal axial and bending deformations characterized by the relative motions between the head, neck, and torso.

Most of these dynamic variables are measurable quantities, either directly using electronic transducers or indirectly using direct measurements in conjunction with physical laws. The results are expressed quantitatively as time-histories, averages, and/or peak values.

In addition to these dynamic variables, the impact response is characterized by the pattern and severity of injury sustained by the involved body segments, usually described in qualitative medical terms. Although several quantitative scales have been devised to assess the severity of injury, this assessment remains primarily descriptive. Nontheless, the design and evaluation of protective devices must incorporate some means of injury assessment.

A test system which evaluates the performance characteristics of protective helmets must therefore take into account the biomechanical parameters described above. However, an effective system should be designed to simulate

the most frequent impact situations.

In an industrial environment where a worker is required to wear a protective helmet, a falling object is the most common hazard encountered. The configuration of the worker at the time of impact determines the location and direction of the applied impact forces; however, unless the worker's head is tilted, or the object is not falling along the "vertical," the impact forces will most likely be in the general superior-inferior (S-I) direction, and may be normal or oblique with respect to the helmet surface.

Throughout this report, the expressions "S-I impacts" and "top-of-the-head impacts" are therefore loosely used to indicate impacts at locations above the basic anatomical plane and in the general superior-inferior direction.

1.4. INJURY TOLERANCE CRITERIA

The central concern of this research program is to minimize trauma to the head, neck,and upper torso, caused by falling objects onto the helmeted head of a worker. The possible mechanisms of injuries were briefly discussed in section 1.2. In this section, the tolerance criteria currently used for injury assessment and prediction are discussed. This discussion is based on an extensive review of the available research literature, and most of the statements made are digested from the references given at the end of this report. The interested reader may further consult these references for more detailed presentations.

1.4.1. Head Injury Tolerance

The first serious attempt at establishing a human tolerance level was the Wayne State Tolerance Curve derived from rigid impacts to cadaver heads in the anterior-posterior (A-P) direction. In each of these tests, the uniaxial acceleration of the head was measured and skull fracture was used to indicate an injurious impact. Since cerebral concussion is often a reversible injury, it may be associated with a conservative estimate of head injury tolerance. Furthermore, since skull fractures are associated with cerebral concussion, it is reasonable to use linear skull fractures as indicators of overall head injury tolerance. That is the reason for the popularity of the WSU Tolerance Curve as an overall head injury criterion. This curve became the basis for a host of methods for determining the severity of a head impact and later evolved to the Head Injury Criterion (HIC) currently used as a government standard in automotive crash testing.

Today, the HIC is the most widely used criterion for assessment of overall head injury, although there is room for debate on the critical value that should be used. It should be noted that the WSU tolerance data, from which the HIC evolved, includes only A-P accelerations resulting from A-P impacts. Furthermore, the structural assymmetries of the brain and head suggest that the impact response may be dependent on the direction and location of impact. Therefore, there is little experimental biomechanical justification for using the resultant head acceleration for head injury assessment as required by the HIC.

It should finally be pointed out that the HIC applies to the translational acceleration of the head. Recent investigations have shown that rotational accelerations (with little or no translational accelerations) produce cerebral concussions. However, no rotational motion tolerance limits have been established.

1.4.1. Spinal Injury Tolerance

Most published data on the tolerance limits of the cervical and thoracic spines falls into two categories. The first involves force levels which can be tolerated by individual vertebral elements, usually in compression modes, while the second deals with forces and moments which can be tolerated by multi-vertebrate sections of the spine.

The average ultimate static compressive strength of a typical cervical vertebra is reported to be somewhere between 1.75 and 2.0 kN. In general, however, dynamic strength (and tolerance) of these elements may be twice as much. Furthermore, the interaction between adjacent vertebrae, their initial configuration as well as the mode and rate of loading may influence to a great extent the level of dynamic forces which can be tolerated before any of the spinal injuries described earlier may occur.

Dynamic loading of the cervical spine have extensively been reported in the Proceedings of the Stapp Car Crash Conferences. These publications are concerned with impulsive loadings of the head due to violent motions of the torso. Such work does not specifically deal with S-I impacts to the top of the head, but the reported data may be used to derive estimates of tolerance levels which are not otherwise available and which are necessary for the design and development of the desired helmet test system.

Data for S-I impacts to eleven cadavers, obtained in a recent study at HSRI, indicate that cervical spine fractures occur for peak forces of 5.7 kN with an energy of 380 J transfered early in the impact from a 10-kg mass moving at 7.5 m/s. The same study also found that these values are greatly influenced by the physical condition of the cadaver's cervical spine, its initial orientation and the mode of loading. Finally, these eleven tests failed to produce basal skull fractures, a serious mode of injury thought to occur in S-I impacts. Most of the damage occured in the lower cervical and upper thoracic vertebrae, suggesting the importance of including these elements in the test system being developed.

1.5. UNAVAILABLE DATA

The current research literature lacks conclusive data on the human response and tolerance to impacts to the top of the head in the general S-I direction. Most of the available documentation relates to skull and brain injury mechanics and impact tolerance. Head injury criteria are based on A-P impacts but have been applied with some degree of success to other directions impacts. Neck injuries and tolerance data are available for loads which are typified by "whiplash" motion. Direct impacts to the top of the head are not well documented in terms of spinal injuries.

The review of literature, therefore, suggests that the biomechanics of cervical and upper thoracic spinal injuries should further be investigated. In particular, the parameters of location, direction and distribution of impact forces, and the initial configuration of the head and spine should be evaluated as they effect the injury patterns and severities during direct impacts in the S-I direction.

A comprehensive tolerance study is not possible in a single study because of time and funding limitations. Therefore, an in-depth investigation of one of the above parameters may prove to be most productive. It appears, from the HSRI pilot study, that emphasis should be placed on padded impacts to the top of the head when the head is flexed (forward) about 20 degrees. With this configuration, the cervical spine is nearly "straight," a worst-case situation where most of the impact energy would be absorbed by the vertebral column. The goal of this study is to generate tolerance and kinematic data and observe injury mechanisms, which would be the basis for both further testing and development of helmet impact test system.

1.6. SUMMARY

A review of the research literature on human response and tolerance has been conducted. The most significant publications which are pertinent to the head, neck and upper torso are listed in the bibliography, section 1.7.

Most of the available data deals with the automotive crash environment with little or no emphasis on S-I impacts. Head and brain injury tolerance for this type of impacts is incomplete but adequate for use in developing the desired helmet impact test system. Documentation of neck and spinal injuries resulting from S-I impacts are virtually non-existent. Such data must be generated, even on a limited basis, before a realistic helmet test system is developed.

1.7. BIBLIOGRAPHY

The following is a list of selected references dealing with topics relevant to the understanding of the biomechanics of head, neck and upper torso injuries. Most of these publications were generated around the automotive crash environment; however, they provide the reader with insight into the injury mechanisms, tolerance and protections of the above body regions. The citations are listed in chronological order and are grouped by body region.

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Appendix A

REVIEW OF PUBLISHED DATA

In this Appendix, the data available in the research literature concerning the biomechanics of head and neck injuries is reviewed. Although detailed, this presentation is by no means exhaustive since the volume of publications dealing directly or indirectly with this subject is massive. Furthermore, papers offering quantitative data on head and neck impact tolerance are oriented to automotive crash environment, with little or no emphasis on S-I impacts as may be encountered in industrial hazardous conditions. Therefore, the review presented here is intended to familiarize the reader with the concepts of defining human tolerances to impacts, to present typical examples to illustrate the type of data that is usually or should be monitored, and to give typical results that have consistently been obtained by various researchers.

It should be pointed out that much of the data currently available does not deal directly with humantolerances to impacts in the general S-I direction. Extrapolation of these results to impa cts to the top of the human head is, at best, speculative. Nonetheless, the experience gained from previous research is extremely valuable in gaining insight into the problem at hand, in pointing out possible directions for new research, and in alerting researchers to the pitfalls and difficulties of defining human tolerance to impact.

Of the publications examined, those which contain quantitative assessment of impact-injury offer such a wide range of test conditions, methodologies, findings, and interpretations that it would not be feasible to summarize them in a simple easy-to-read table. Instead, the most relevant information from individual publications will be highlighted with emphasis on:

a) quantification of the impact conditions, such as force level and duration, impactor velocity and mass, loading rate, location and direction, of applied forces as well as shape of impactor and force distribution, and

b) quantification of response parameters such as measured deformation/displacement, acceleration level and duration, pressure/stress time histories as well as type and degree of specimen failure or injury.

A.1. PROPERTIES OF HUMAN SKULL

Although skull fracture has been studied for over one hundred years, it has been only recently that experiments to determine the mechanical properties of skull bone as a material have been performed. The first study was that of Evans and Lissner [1] in 1957. The average ultimate strength of specimens of parietal compact bone was found to be 70 MPa in tension, 152 MPa in compression when

loaded in the same tensile load direction, and 167 MPa in compression when loaded perpendicular to the tensile load direction. The average compressive strength of specimens of the cancellous diploe layer bone was found to be 25 MPa.

More recently, Wood [2] has reported on a study of the mechanical properties of unembalmed human cranial bone in tension based on tests of over 120 specimens from thirty subjects. The specimens were taken from the compact layers of parietal, temporal and frontal bone and tested at strain rates ranging from static to 150 sec^{-1}. The modulus of elasticity, the breaking stress and strain were found to be strain-rate sensitive, while the energy absorbed to failure were not. At low strain rates, the average tensile strength found to be 69 MPa agrees closely with the value obtained by Evans and Lissner. The average modulus of elasticity ranged from 12 to 20 GPa, and the average value of energy absorbed to failure was 347 kJ/m^3.

The fracture of the skull as a whole has been under investigation for over a century. Thus, quantitative data on the magnitude of the force required for fracture has appeared in the literature as early as 1859 when Weber [3] found that the skull of a small boned tuberculous girl 27 years of age required only 4.95 kN, while the skull of a robust 37-year old woman did not fracture under 6.12 kN.

In the classical study in 1880 by Messerer [4], the skulls from 25 men and women with ages ranging from 18 to 82 years were loaded either in the transverse direction or in the longitudinal one. Judging from the forces required for fracture, Messerer found that skulls of women were stronger in the transverse direction than those of men, while the skulls of men were stronger in the longitudinal direction than those of women. In both sexes, the longitudinal strength was higher than the transverse strength. When all tests were combined, the average load was 5.08 kN (3.4 - 7.8 kN) for transverse loading and 6.36 kN (3.9 - 11.8 kN) for longitudinal loading.

In addition to his tests described above, Messerer investigated compression in a direction perpendicular to base of the skull, or superior-inferior direction. This was done on 8 skulls with 3 or 4 attached cervical vertebrae. In this series of tests, the base of the skull was destroyed before the compression had much effect on the entire skull. In many cases, the first or second vertebra fractured before the skull did. The average breaking load was found to be 2.64 kN (2.2 - 2.9 kN).

The amount of energy and the time for its absorption required to fracture 55 intact human cadaver heads was investigated in 1949 by Gurdjian, Webster and Lissner [5]. Data obtained in these tests showed that energy varying from 45-100 J was required to produce a single linear fracture, with insignificant correlation between location of impact and the amount of energy.

Evans, Lissner and Lebow [6] studied in 1958 the relation of energy, velocity and acceleration to skull deformation and fracture in intact human heads taken from embalmed adult cadavers, by dropping the head on a 1954 model automobile instrument panel, and producing blows to the forehead. fractures were produced with peak impact accelerations of 337, 344, 555 and 724 g having a total time duration of 11.25, 4.88, 9.03 and 3.38 ms, respectively. In some cases the head tolerated, without fracture, peak acceleration as high as 686 g and available kinetic energy as great as 782 J.

In 1968 Nahum et al. [7] conducted a series of experiments on 10 human skulls using a drop tower to apply imapcts to the frontal and tempero-parietal junction with a 1-square inch impactor area. The data obtained from their experiments led the authors to suggest the following critical values when the contact area is effectively one square inch: 4.9 kN for frontal area, 2.5 kN for parietal area, and 1.0 kN for zygomatic area. In this study, Nahum et al. also concluded that the thickness of the soft tissue plays an important role in increasing the tolerable impact forces.

A.2. BRAIN INJURY TOLERANCE

Many researchers have concluded that, as far as injury to the brain or brain stem injury is concerned, the ultimate physical cause is shear stress. Quantification of this shear is so difficult that other physical, more measurable factors, were related to head injuries by investigators. Thus, head accelerations (translational and rotational) as a peak value, average value or mean value associated with a pulse duration, impulse, energy and velocity were used to quantify tolerance of the brain to head impacts in the experimental animal research. Others focused on the pressure gradient that is produced in the cranium in the hope of establishing a correlation between the level of impact and the resulting brain injury.

In a series of experiments reported by Ommaya in chapter 23 of [8], 80 Rhesus were used in head impact tests under varying conditions. Cerebral concussion was defined as the loss of voluntary movement and aversive response to ear pinch when these were present immediately before impact. Severity of impact and response was measured by piston (impactor) velocity, head tangential velocity, head linear acceleration, impact force and intracranial pressure, or by calculated values of kinetic energy, and impulse. Ommaya concluded in his analysis that the impulse of impact was a reliable and statistically significant index that may be used to relate the input and the dissipation of energy of occipital blow to experimental concussion. Thus, the probability of concussion is 10% for an impulse between 0.20 to 0.33 N.s, 50% for an impulse between 1.85 to 1.94 N.s, and 90% for an impulse between 13.32 and 13.46 N.s.

Another index that was found to be reliable in predicting concussion is the head acceleration with levels of 9.9 - 13.7 g's, 100.1 - 102.5 g's and 865.3 - 869.1 g's associated with 10%, 50% and 90% concussion probabilities, respectively. Surprisingly, no statistical correlation between intracranial pressure and concussion was found to exist in this series of measurements.

In another series of experiments, Ommaya, Faas, and Yarnell [9] studied the effects of whiplash on the production of cerebral concussion, using 50 Rhesus monkeys. The angular acceleration of the head was measured from 1000 frames/sec highspeed movies. The results indicate that, as the duration of the angular acceleration increases from 3, 4, 5, 6, 7 to 10 ms, the concussion threshold of its peak magnitude decreases from 500, 150, 90, 70, 60 to 40 krad/s^2.

The dynamic structural characteristics of monkey skull and brain were determined over a wide frequency range by Stalnaker and McElhaney in a 1972 study and reported in [10]. The measured property was the driving point impedance

which allowed the conceptual characterization of the head as two masses coupled by a spring and a dashpot. It was determined that the shape of the impedance curve was similar for several species of subhuman primates and the fresh human cadaver.

In order to validate their model, which produces a mean strain as output when the input is a measured acceleration, the authors tested 30 Rhesus monkeys by impacting the head at increasing levels in various directions (front, side, back, top and mid-front). In this study, McElhaney, Stalnaker and Roberts found that for front, side, top and rear impacts, the critical accelerations were 1300, 1500, 980 and 1000 g with durations of 3.6, 2.8, 7.0 and 3.4 ms, respectively.

While Ommaya and his colleagues emphasized the importance of head deformation/brain rotation as the cause of shear stress formation in the brain, others (led by Gurdjian) suggested that pressure gradients/cavitation are the cause for these shear stresses. Measurements of intracranial pressures during head impact has been attempted and results used in head modeling [11, 12, 13].

The most recent study is that of Nahum, Smith and Ward [14] where two series of cadaver head impact experiments were conducted. Measured intracranial pressures at various sites were correlated with other impact parameters. The authors found strong correlation between the head acceleration and individual pressures, with correlation coefficient (r^2) varying between 0.89 to 0.95. Since cadavers were used, no injury tolerance criterion was established for the living human.

In order to estimate tolerances for the living humans, scaling has been suggested by Ommaya and Hirsh [15] and by McElhaney et al. [10]. Scaling Rhesus monkey tolerance to rotational acceleration to that of man, Ommaya proposed the use of the mass of brain and obtained a tolerance threshold of about 1200 rad/s^2 of angular acceleration of the human head. McElhaney used a dimensionless parameter using average skull dimensions and weight, along with impact dynamics to extrapolate data from 3 sub-human primates species to man. He concluded that the human head would tolerate up to 2.24 m/s impact by a flat rigid striker, that the peak tolerable acceleration for the side of the human head would be 56 g, and that a peak force of 800 lb. would be tolerated by the human head.

A.3. SPINAL INJURY LITERATURE

The simplest structural element of the spine is the vertebra. Strengths of the vertebra (primarily in compression) and of spinal sections consisting of several vertebrae have been determined by various investigators. Most extensive reporting was done by Evans and his colleagues in the 1950's and 1960's.

Evans reported [16] in 1962 that the end plates of 28 fresh vertebrae failed with an average load of 3 kN (1.9 - 4.0 kN) statically applied. In a previous study [17], Evans and Lissner studied the response of sections of the spine (deflection, energy absorption, moment) to compression and bending. They found that lower spinal section (T12 - L5) deflects on the average 3.5 cm and absorbs 66 J of energy when subjected to an average 680 lb of compressive loading.

Hodgson, Lissner and Patrick [18] studied the effects of jerk on the human spine. They defined a dynamic load factor to be used to estimate the dynamic tolerance of the spine from static data. They concluded that a load factor of 2.2 and 2.4

should be applied when the rate of onset of the applied acceleration is between 800 and 2000 G/sec.

In their classical study in 1967, Mertz and Patrick [19] summarized the voluntary static human tolerance levels based on reactions acting at the occipital condyles as follows: for the normal head position, 178 N (P-A) and 356 N (A-P) shear force 1100 N (I-S) axial force; 25.8 J extension and 14.2 J extension torque. These values change significantly for extended or flexed head. Furthermore, when dynamic whiplash tests were performed on volunteers, and the results compared to similar tests on cadavers, the maximum dynamic head response indicated that the calculated torque, axial and shear forces in cadavers were 2.5, 0.6 and 0.5 times those in volunteers, respectively.

In a subsequent study in 1971, Mertz and Patrick [20] proposed response envelopes of the human neck hyperextension and extension. These envelopes of the tolerable moments at the occipital condyles as functions of the head angle relative to the torso remain today the design basis for anthropomorphic dummy necks.

Most other data on neck injuries are qualitative in nature, with fragmented pieces of numerical results, obtained primarily from estimates based on impact reconstruction. These publications are valuable for the understanding o- the mechanisms of neck injuries.

A.4. CONCLUDING REMARKS

In reviewing the publications that deal with injuries to the head, neck and upper torso, two primary conclusions were reached. First, the bulk of the data is qualitative in nature, giving limited definite numerical answers to the "how much" question. Second, the majority of numerical data was generated in the last three decades with emphasis on the automotive crash environment.

Since this project is geared toward helmet development and testing for protective worthiness in industrial environment, the search of literature was concentrated on superior-inferior modes of impact, or any impact situation that might result in an axial loading of the head-spinal column structure.

The human tolerance data which was presented in this appendix may be used as a basis to estimate tolerance to axial loading and response, but such estimates will only be speculative. The direct approach of obtaining this tolerance data under the desired impact conditions remains, therefore, the best method for defining that human response and tolerance.

A.5. PUBLICATIONS CITED

The following is a list of references which have been cited in this Appendix. Some have already been included in the general bibliography list given in the last section of the Phase I Final Report. The remaining were not included in the original list.

[1] F.G. Evans and H.R. Lissner, "Tensile and Compressive Strength of Human Parietal Bone." J. Appl. Physiol., 10:493-497, 1957.

[2] J.L. Wood, "Dynamic Response of Human Cranial Bone." J. Biomechanics, 4:1-12, 1971.

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[4] O. Messerer, "Über Elastiticät und Festigkeit der Menschlichen Knochen." J.G. Cotta, Stuttgart, 1880.

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[7] A.M. Nahum, J.D. Gatts, C.W. Gadd and J. Danforth, "Impact Tolerance of the Skull and Face." Proc. 12th Stapp Car Crash Conf., pp. 302-316, SAE, 1968.

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[9] A.K. Ommaya, F. Faas and P. Yarnell, "Whiplash Injury and Brain Damage, an Experimental Study." J. Am. Med. Assoc., 204(4):285-289, 1968.

[10] J.H. McElhaney, R.L. Stalnaker and V.L. Roberts, "Biomechanical Aspects of Head Injury." Proc. Symposium on Human Impact Response (Oct. 1972), Plenum Press, New York, 1973.

[11] E.S. Gurdjian, V.L. Roberts and L.M. Thomas, "Tolerance Curves of Acceleration and Intracranial Pressure and Protective Index in Experimental Head Injury." J. Trauma, 6(5):600-604, 1966.

[12] V.L. Roberts, V.R. Hodgson and L.M. Thomas, "Fluid Pressure Gradients Caused by Impact to the Human Skull." ASME Publication No. 66-HUF-1, 1966.

[13] J.V. Benedict, "An Analysis of an Impact-Loaded, Fluid-Filled, Thin Spherical Shell as a Mathematical Model for an Investigation of the Cavitation Theory of Brain Damage." Dissertation Prospectus, M.E. Department, Tulane University, 1968.

[14] A.M. Nahum, R. Smith and C.C. Ward, "Intracranial Pressure Dynamics During Head Impact." Proc. 21st Stapp Car Crash Conference, SAE paper 770922, SAE, 1977.

[15] A.K. Ommaya and A.E. Hirsch, "Protection of the Brain from Injury during Impact: Experimental Studies in the Biomechanics of Head Injury." AGARD Conf. Proc. No. 88 on Linear Acceleration of Impact Type, Oporto, Portugal, 1971.

[16] F.G. Evans, H.R. Lissner and L.M. Patrick, "Acceleration-Induced Strains in the Intact Vertebral Column." J. Appl. Physiol., 17(3): 405-409, 1962.

[17] F.G. Evans and H.R. Lissner, "Biomechanical Studies on the Lumbar Spine and Pelvis." J. Bone & Joint Surg., 41-A(2):278-290, 1959.

[18] V.R. Hodgson, H.R. Lissner and L.M. Patrick, "The Effect of Jerk on the Human Spine." ASME publication 63-WA-316, 1963.

[19] H.J. Mertz and L.M. Patrick, "Investigation of the Kinematics and Kinetics of Whiplash." SAE Publication No. 670919, Proc. 11th Stapp Car Crash Conf., 1967.

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2. PHASE II - CADAVER IMPACT TESTS

The second phase of the project is designed to generate human kinematic response which remains unavailable in the research litterature, and which could feasibly be obtained with cadaver testing. It was determined that very little data is available on the human response to impacts in the S-I direction. Therefore, all the tests performed in this phase were conducted by delivering the impact to the crown of the head in the spinal S-I direction.

Five fully instrumented tests were performed. The data generated is massive and, therefore, will not be included here in detail. Instead, processed time-histories of the response of various parameters are presented, along with supported documents and brief description of the experimental and analytical methods employed in the testing and data analysis.

It was felt that the sample is too small to draw general conclusions; therefore, none was drawn. Instead, specific observations were made about consistent response trends which are then used to provide guidelines for the design of the helmet test device itself.

The test subjects were unembalmed cadavers obtained through the Anatomy Department at the University's Medical School. The protocol for the use of cadavers in this study was reviewed by the Committee to Review Grants for Clinical Research and Investigation Involving Human Beings of the University of Michigan Medical Center and follows guidelines established by the U.S. Public Health Service and recommended by the National Academy of Science/ National Research Council.

2.1. OBTAINABLE TEST DATA

There are two categories of responses that could be determined from fully instrumented cadaver testing. The first is tolerance response which is based on post-test autopsy examination of the physical damage caused by the impact. The second category is kinematic and dynamic responses of various body segments, either directly measured with transducers such as accelerometers and load cells, or indirectly obtained by mathematical manipulation of direct measurements.

It is this kinematic and dynamic response which was sought in the cadaver testing phase of the project. Furthermore, emphasis was placed on the accuracy and completeness of measurement. New instrumentation techniques allow the measurement of the full three-dimensional motion of the head as a rigid body and the monitoring of the motion of vertebral bodies during impacts to the head in the S-I direction. Therefore, the following measurements were made:

a) the velocity and energy of impact;

b) the direction and location of impact force with respect to the head

coordinate system and the orientation of the cervical spine relative to the line of impact;

c) the 3-D rigid body motion of the head, including displacement, velocity and acceleration of the six degrees of freedom;

d) the resultant acceleration at the head center of mass and the HIC (head injury criterion);

e) the forces and moments (in 3 directions) at the occipital condyles;

f) the components of the applied force, resolved the head center of mass.

Most of the above parameters were measured and/or computed as functions of time for periods of 75 ms, from the initiation of impact. In many instances, supporting measurements had to be made in order to apply laws of dynamics for computation of various forces from kinematic measurements. In addition, sophisticated experimental methods were employed to express measured quantities in standard anatomical reference frames.

Finally, the tests were filmed at 1000 frames/second using 2 high-speed motion picture cameras aimed in orthogonal directions. The intent was to measure the 3-D motions of the 1st and 12th vertebrae. After carrying out the data reduction for the first test and considering the availability of the triaxial acceleration measurement at T1 and T12, it was decided to continue the film coverage but not to carry out the data reduction for the 3-D motion of T1 and T12.

The pertinent results of 5 tests (79H201 through 79H205) are reported here. The methods used in obtaining the results are described in detail only when such methods have not been reported previously elsewhere.

2.2. INPUT TO ANALYSIS PROGRAMS

Pre-analysis processing of x-ray motion picture film as well as transducer signals was applied in order to prepare for analysis of test data.

Figures 1 through 5 show the input used for the data analysis programs which compute the 3-D motion of the head and those which determine the initial orientations of the head, the location of the occipital condyles in the head anatomical reference frame and the orientations of the T1 and T12 vertebral mounts in the laboratory reference frame.

Figures 6 though 10 contain the 9 acceleration components used to determine the 6 degrees-of-freedom motion of the head. Each triax components are resolved about the standard head anatomical reference frame (PA, LR, and SI). The actual nine accelerometer readings were taken in another instrumentation frame whose orientation and origin is known (through x-ray 3-D film analysis) with respect to the standard head anatomical reference frame.

The accelerations were filtered at 300 Hz and sampled at 1600 Hz. Filtering was done by performing a FFT (Fast Fourier Transform) on each signal, then throwing out all components whose frequency is above 300 Hz.

FIGURE 1. TEST 79H201

PROGRAM	INPUT*		
3DNINE			
Instrumentation Frame wrt Anatomical Frame			
3 Translations *(cm)*	- 9.59	0.42	4.53
3 Rotations *(deg)*	100.29	-79.61	-102.53
Anatomical Frame wrt Lab Frame			
Anatomical Center Positions *(cm)*	74.23	0.66	- 3.82
Euler Angles *(deg)*	-174.97	82.30	10.95
3DCOND			
Condyle Center wrt Anatomical Center			
Condyle (midpoint, *cm*)	- 2.46	- 0.34	0.26
3DFILM			
Positions of Head wrt Cross *(mm)*	51.3	-11.6	-111.2
Rotations of Head wrt Cross *(deg)*	- 59.0	83.7	130.0
Initial Position of Head wrt Lab *(mm)*	742.3	6.6	- 38.3
Initial Angle of Head wrt Lab *(deg)*	-175.0	82.3	10.9
Positions of T1 wrt Cross *(mm)*	- 2.0	0.0	-115.0
Rotations of T1 wrt Cross *(deg)*	180.0	118.0	0.0
Initial Position of T1 wrt Lab *(mm)*	849.0	- 2.8	- 2.9
Initial Angle of T1 wrt Lab *(deg)*	1.5	54.6	-165.3
Positions of T12 wrt Cross *(mm)*	- 14.0	0.0	-105.0
Rotations of T12 wrt Cross *(deg)*	180.0	83.0	0.0
Initial Position of T12 wrt Lab *(mm)*	859.1	13.4	64.3
Initial Angle of T12 wrt Lab *(deg)*	137.3	69.5	- 46.8

*Order of INPUT is x,y,z for translations; and roll, pitch, yaw for rotations.

FIGURE 2. TEST 79H202

PROGRAM	INPUT*		
3DNINE			
Instrumentation Frame wrt Anatomical Frame			
3 Translations *(cm)*	- 9.38	0.14	7.23
3 Rotations *(deg)*	1.48	- 61.43	0.05
Anatomical Frame wrt Lab Frame			
Anatomical Center Positions *(cm)*	51.51	2.00	- 0.60
Euler Angles *(deg)*	-167.74	79.03	11.57
3DCOND			
Condyle Center wrt Anatomical Center			
Condyle (midpoint, *cm*)	- 1.57	0.23	- 2.09
3DFILM			
Positions of Head wrt Cross *(mm)*	14.2	1.5	-136.9
Rotations of Head wrt Cross *(deg)*	177.2	60.8	- 1.3
Initial Position of Head wrt Lab *(mm)*	515.1	20.0	- 6.1
Initial Angle of Head wrt Lab *(deg)*	-167.7	79.0	11.6
Positions of T1 wrt Cross *(mm)*	0.0	0.0	-119.0
Rotations of T1 wrt Cross *(deg)*	180.0	115.0	0.0
Initial Position of T1 wrt Lab *(mm)*	662.6	22.3	18.5
Initial Angle of T1 wrt Lab *(deg)*	12.0	62.5	-169.9
Positions of T12 wrt Cross *(mm)*	- 8.0	0.0	- 94.0
Rotations of T12 wrt Cross *(deg)*	180.0	90.0	0.0
Initial Position of T12 wrt Lab *(mm)*	910.1	23.1	87.8
Initial Angle of T12 wrt Lab *(deg)*	164.1	84.9	- 20.2

*Order of INPUT is x,y,z for translations; and roll, pitch, yaw for rotations.

FIGURE 3. TEST 79H203

PROGRAM	INPUT*		
3DNINE			
Instrumentation Frame wrt Anatomical Frame			
3 Translations *(cm)*	- 9.13	- 0.86	0.07
3 Rotations *(deg)*	185.06	- 70.83	166.55
Anatomical Frame wrt Lab Frame			
Anatomical Center Positions, *est.* *(cm)*	62.87	1.33	- 2.21
Euler Angles, *est.* *(deg)*	-171.35	80.66	11.26
3DCOND			
Condyle Center wrt Anatomical Center, *est.*			
Condyle (midpoint, *cm*)	- 2.8	0.0	- 2.8
3DFILM			
Positions of Head wrt Cross, *est.* *(mm)*	0.0	0.0	0.0
Rotations of Head wrt Cross, *est.* *(deg)*	0.0	0.0	0.0
Initial Position of Head wrt Lab, *est.* *(mm)*	628.7	13.3	- 22.2
Initial Angle of Head wrt Lab, *est.* *(deg)*	-171.3	80.7	11.3
Positions of T1 wrt Cross *(mm)*	0.0	0.0	- 94.0
Rotations of T1 wrt Cross *(deg)*	180.0	122.0	0.0
Initial Position of T1 wrt Lab, *est.* *(mm)*	0.0	0.0	0.0
Initial Angle of T1 wrt Lab, *est.* *(deg)*	12.0	62.5	-169.9
Positions of T12 wrt Cross *(mm)*	0.0	0.0	-115.0
Rotations of T12 wrt Cross *(deg)*	180.0	75.0	0.0
Initial Position of T12 wrt Lab, *est.* *(mm)*	0.0	0.0	0.0
Initial Angle of T12 wrt Lab, *est.* *(deg)*	164.1	84.9	- 20.2

*Order of INPUT is x,y,z for translations; and roll, pitch, yaw for rotations.

FIGURE 4. TEST 79H204

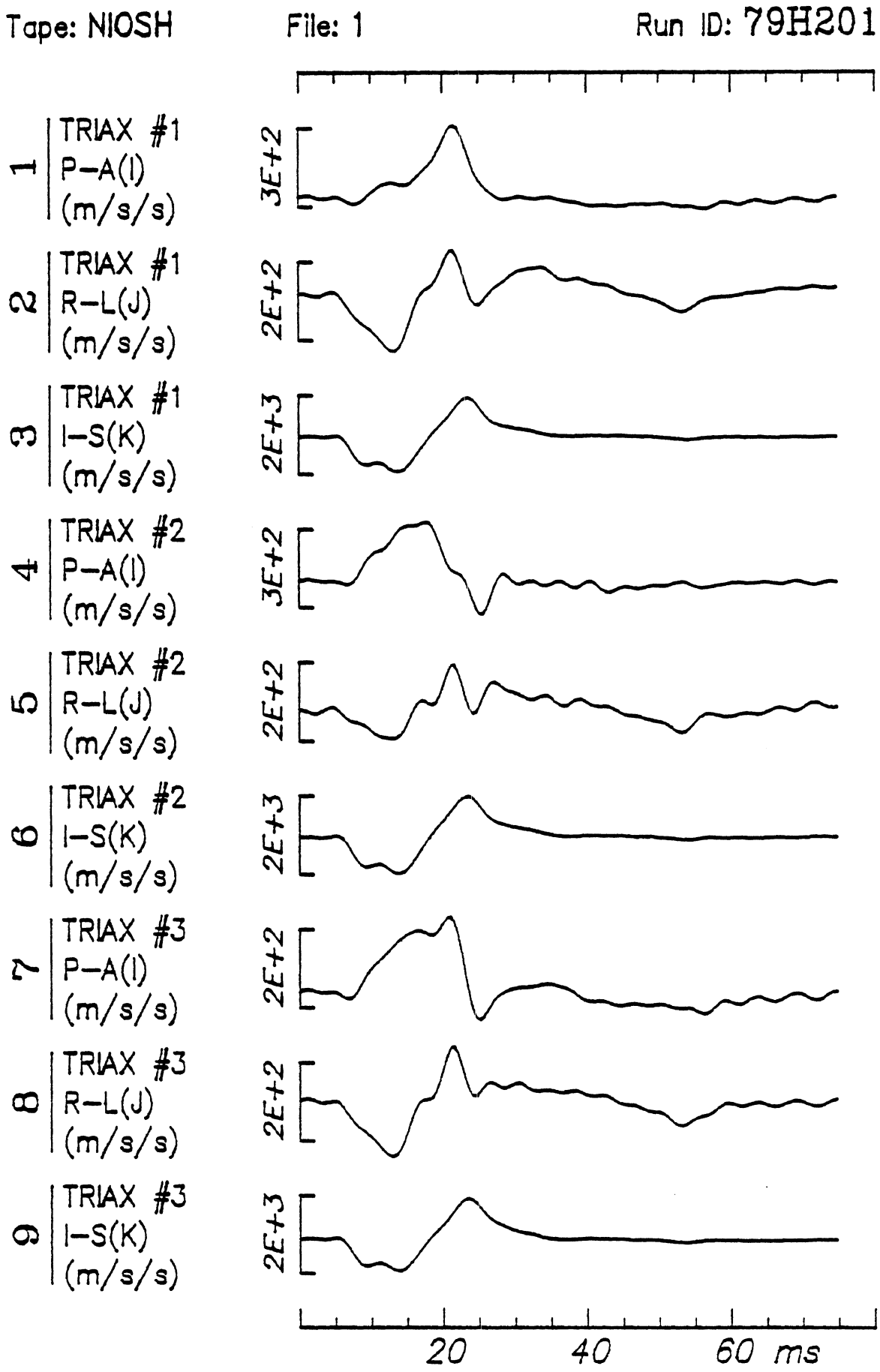
PROGRAM	INPUT*		
3DNINE			
Instrumentation Frame wrt Anatomical Frame			
3 Translations *(cm)*	- 8.93	2.90	3.03
3 Rotations *(deg)*	- 24.09	83.99	170.28
Anatomical Frame wrt Lab Frame			
Anatomical Center Positions *(cm)*	51.51	2.17	- 0.43
Euler Angles *(deg)*	171.14	76.12	- 4.18
3DCOND			
Condyle Center wrt Anatomical Center			
Condyle (midpoint, *cm*)	- 0.86	- 0.27	- 1.16
3DFILM			
Positions of T1 wrt Cross *(mm)*	- 13.0	0.0	-112.0
Rotations of T1 wrt Cross *(deg)*	180.0	105.0	0.0
Initial Position of T1 wrt Lab *(mm)*	646.2	18.9	14.8
Initial Angle of T1 wrt Lab *(deg)*	7.2	64.3	-165.6
Positions of T12 wrt Cross *(mm)*	- 13.0	0.0	-105.0
Rotations of T12 wrt Cross *(deg)*	180.0	90.0	0.0
Initial Position of T12 wrt Lab *(mm)*	928.7	22.2	52.8
Initial Angle of T12 wrt Lab *(deg)*	165.1	71.0	- 10.1

*Order of INPUT is x,y,z for translations; and roll, pitch, yaw for rotations.

FIGURE 5. TEST 79H205

PROGRAM	INPUT*		
3DNINE			
Instrumentation Frame wrt Anatomical Frame			
3 Translations *(cm)*	- 9.87	1.84	4.55
3 Rotations *(deg)*	63.77	80.14	-110.23
Anatomical Frame wrt Lab Frame			
Anatomical Center Positions *(cm)*	52.27	2.93	- 2.79
Euler Angles *(deg)*	-166.42	73.29	12.71
3DCOND			
Condyle Center wrt Anatomical Center			
Condyle (midpoint, *cm*)	- 1.32	0.07	- 2.34
3DFILM			
Positions of T1 wrt Cross *(mm)*	- 14.0	0.0	- 99.0
Rotations of T1 wrt Cross *(deg)*	180.0	80.0	0.0
Initial Position of T1 wrt Lab *(mm)*	634.1	32.1	- 26.0
Initial Angle of T1 wrt Lab *(deg)*	- 40.8	87.0	139.8
Positions of T12 wrt Cross *(mm)*	1.0	0.0	-137.0
Rotations of T12 wrt Cross *(deg)*	180.0	60.0	0.0
Initial Position of T12 wrt Lab *(mm)*	1009.4	46.4	7.6
Initial Angle of T12 wrt Lab *(deg)*	167.3	67.1	- 4.3

*Order of INPUT is x,y,z for translations; and roll, pitch, yaw for rotations.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2
TRIAX #1
R-L(J)
(m/s/s)
2E+2
3
TRIAX #1
I-S(K)
(m/s/s)
2E+3
4
TRIAX #2
P-A(I)
(m/s/s)
3E+2
5
TRIAX #2
R-L(J)
(m/s/s)
2E+2
6
TRIAX #2
I-S(K)
(m/s/s)
2E+3
7
TRIAX #3
P-A(I)
(m/s/s)
2E+2
8
TRIAX #3
R-L(J)
(m/s/s)
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9
TRIAX #3
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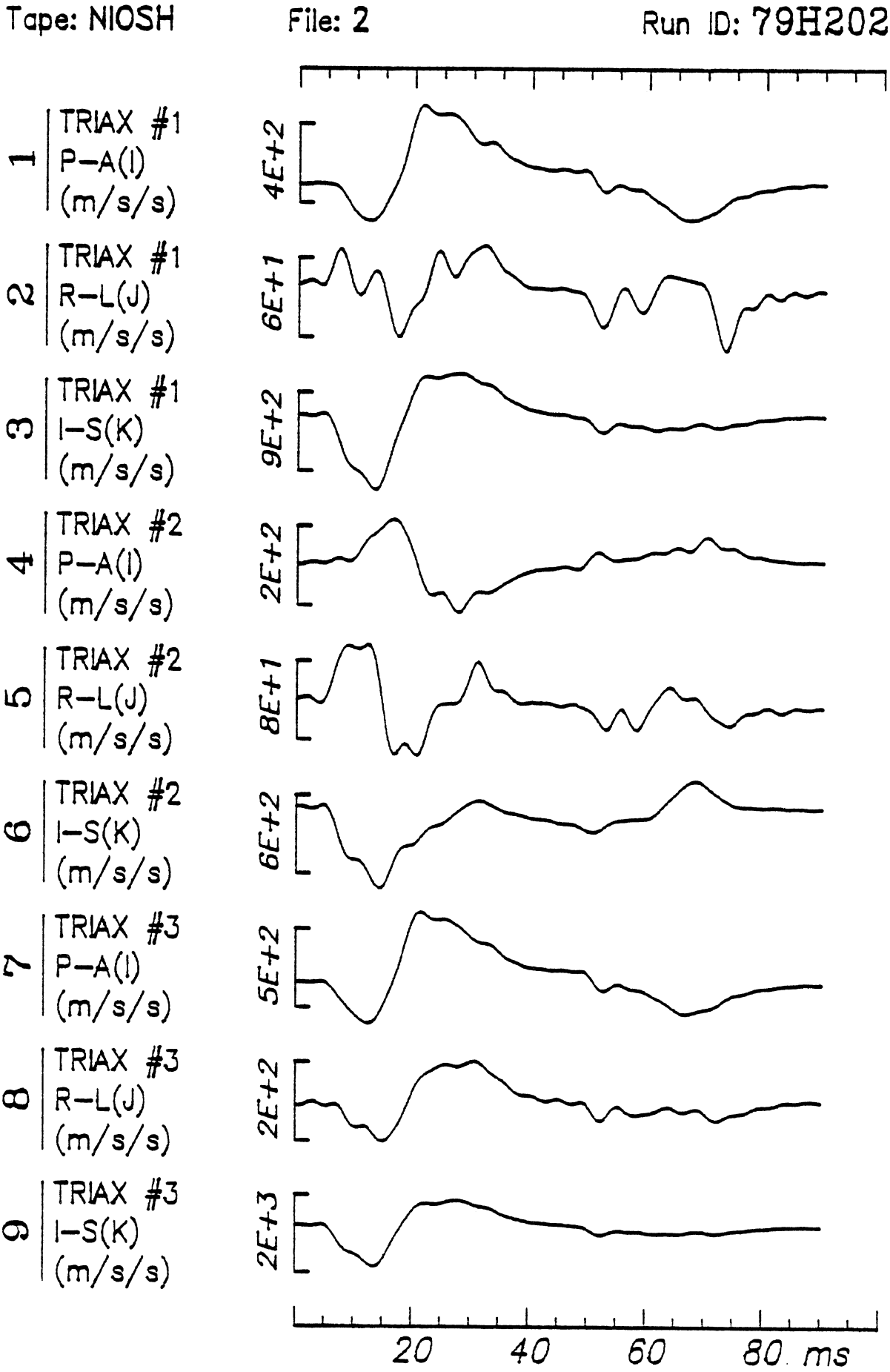
(m/s/s)
2E+3
20
40
60 ms

Figure 6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(m/s/s)
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3
TRIAX #1
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(m/s/s)
9E+2
4
TRIAX #2
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(m/s/s)
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(m/s/s)
6E+2
7
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5E+2
8
TRIAX #3
R-L(J)
(m/s/s)
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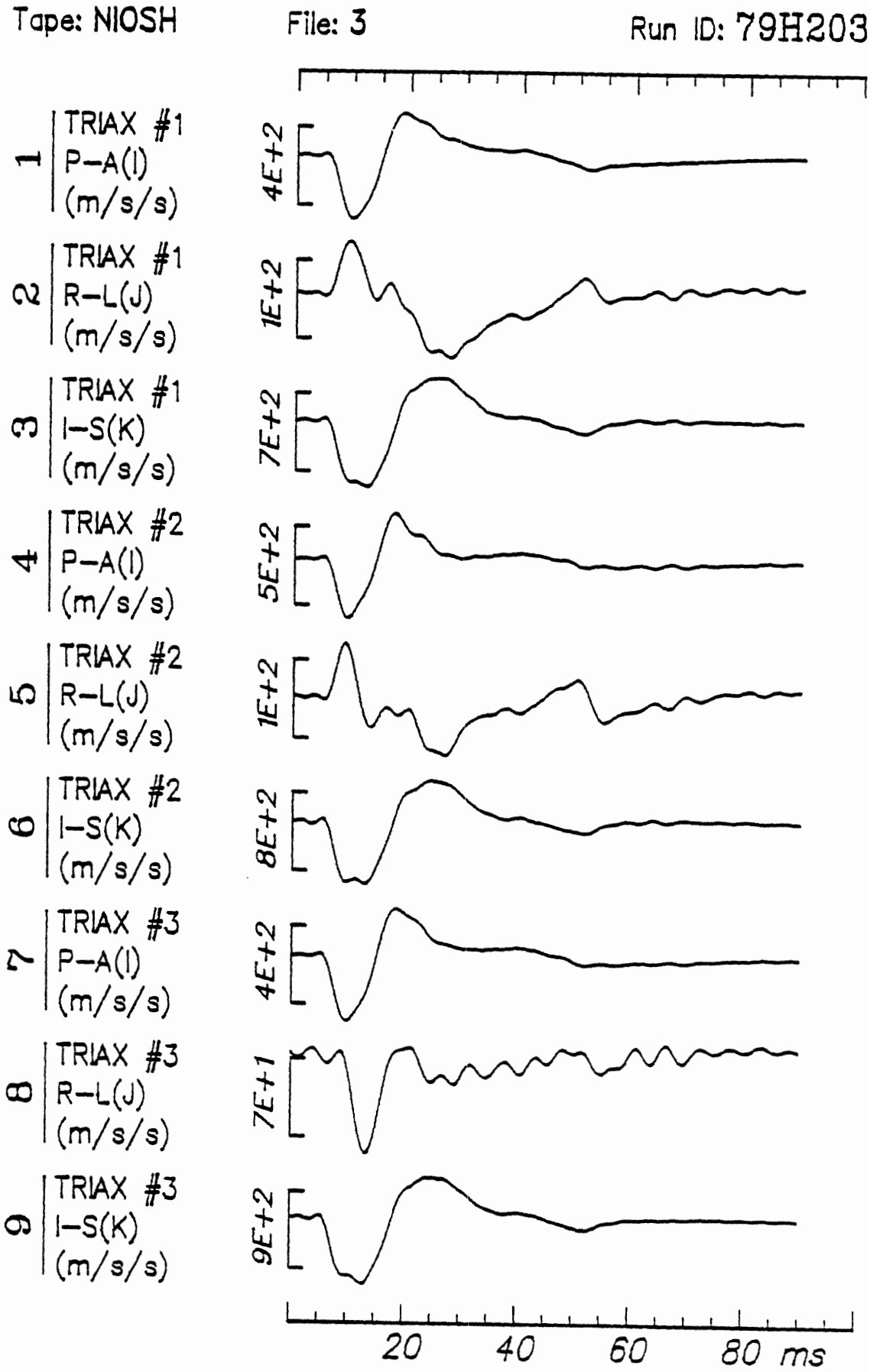
TRIAX #3
I-S(K)
(m/s/s)
2E+3
20
40
60
80 ms

Figure 7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(m/s/s)
1E+2
3
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I-S(K)
(m/s/s)
7E+2
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TRIAX #2
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(m/s/s)
1E+2
6
TRIAX #2
I-S(K)
(m/s/s)
8E+2
7
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(m/s/s)
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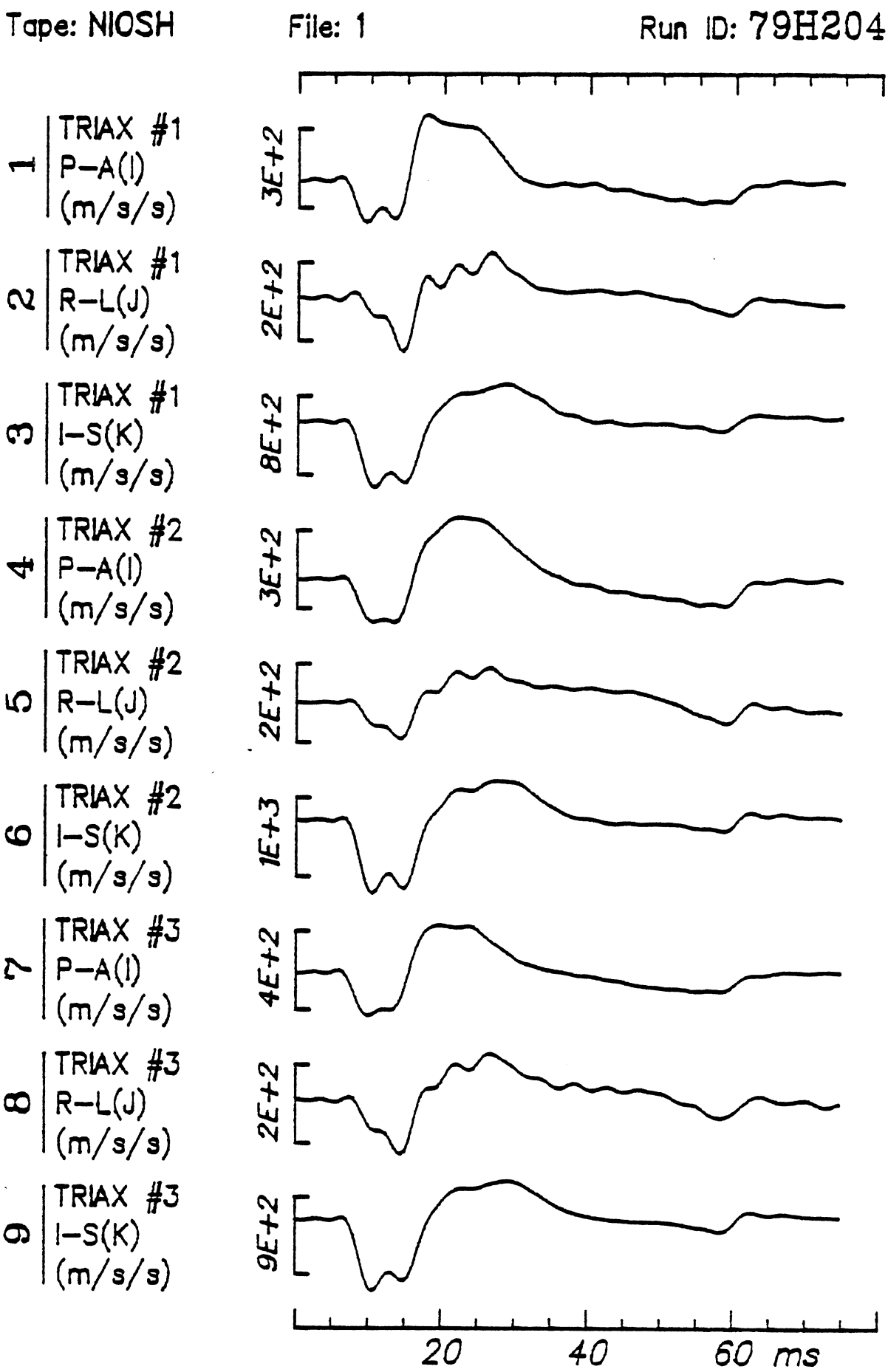
9E+2
20
40
60
80 ms

Figure 8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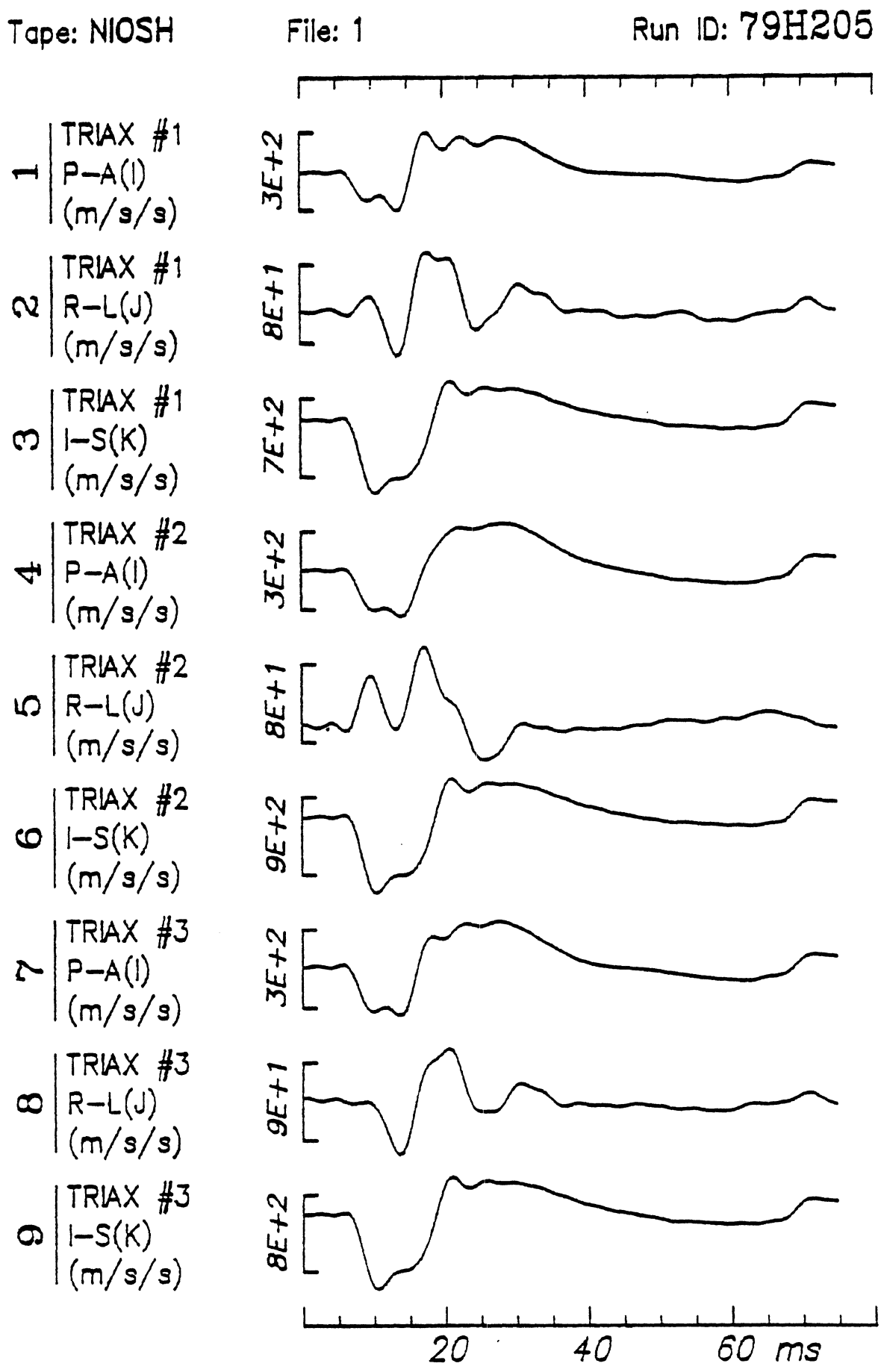
(m/s/s)
2E+2
3
TRIAX #1
I-S(K)
(m/s/s)
8E+2
4
TRIAX #2
P-A(I)
(m/s/s)
3E+2
5
TRIAX #2
R-L(J)
(m/s/s)
2E+2
6
TRIAX #2
I-S(K)
(m/s/s)
1E+3
7
TRIAX #3
P-A(I)
(m/s/s)
4E+2
8
TRIAX #3
R-L(J)
(m/s/s)
2E+2
9
TRIAX #3
I-S(K)
(m/s/s)
9E+2
20
40
60 ms

Figure 9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3E+2
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8E+1
3 TRIAX #1 I-S(K) (m/s/s)
7E+2
4 TRIAX #2 P-A(I) (m/s/s)
3E+2
5 TRIAX #2 R-L(J) (m/s/s)
8E+1
6 TRIAX #2 I-S(K) (m/s/s)
9E+2
7 TRIAX #3 P-A(I) (m/s/s)
3E+2
8 TRIAX #3 R-L(J) (m/s/s)
9E+1
9 TRIAX #3 I-S(K) (m/s/s)
8E+2
20
40
60 ms

Figure 10

2.3. HEAD KINEMATIC RESPONSE

Processing of the 9 acceleration readings of the head produced as many as 60 variables which can be used to characterize the response. Not all these variables are included here, since many of them are significant only in special rigid body motion.

The head linear (i.e., translational) and angular velocities and accelerations were chosen to be reported here, since they have a direct bearing on the design of the helmet test system, and since most injury tolerance criteria are defined in terms of these variables.

The angular motion of the head for the 5 tests is shown in Figures 11-15. The angular acceleration and velocity vectors are shown both as components in the head anatomical directions (which are moving in the laboratory frame during the impact) and as resultants. It is interesting to note that most of tne tests (except H203) indicate that primary motion is about the L-R axis (i.e., flexion-extension). The linear motion of the head, i.e., translation of the head anatomical center, is given in Figures 16-20. Here also, the linear acceleration and velocity are given both as resultants and as components in the head anatomical reference frame. As expected, the translational motion is most severe in the S-I direction; however, motion in the L-R and A-P direction is generated to a lesser extent.

The HIC for the resultant acceleration has been computed. It is interesting to note that none of the tests resulted in HIC higher than 325, even though the impact forces and energies were near what is thought to be the fracture tolerance limits of the neck. This point will be discussed in detail in chapter 4.

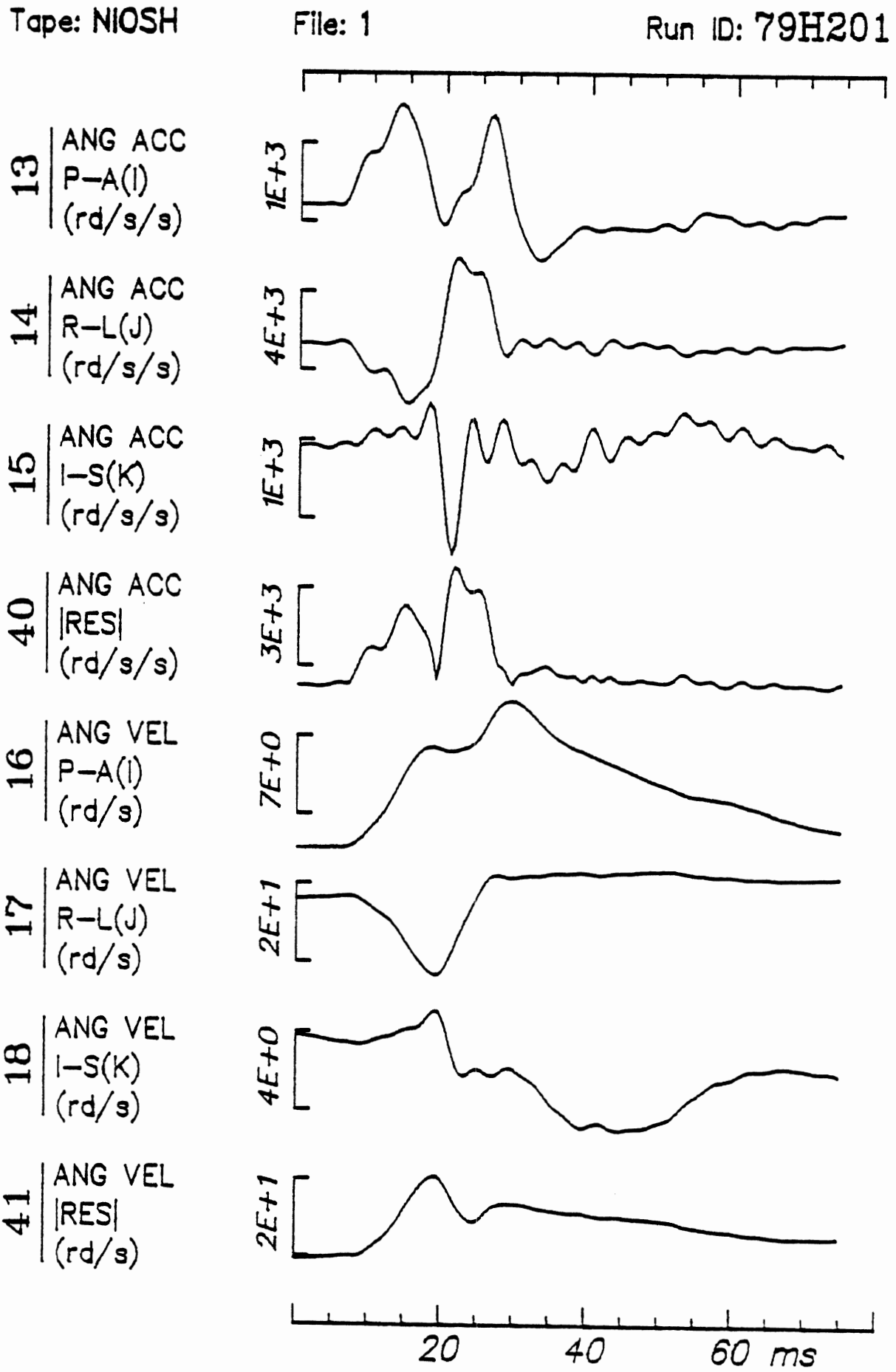

Tape: NIOSH
File: 1
Run ID: 79H201
13
ANG ACC
P–A(I)
(rd/s/s)
1E+3
14
ANG ACC
R–L(J)
(rd/s/s)
4E+3
15
ANG ACC
I–S(K)
(rd/s/s)
1E+3
40
ANG ACC
|RES|
(rd/s/s)
3E+3
16
ANG VEL
P–A(I)
(rd/s)
7E+0
17
ANG VEL
R–L(J)
(rd/s)
2E+1
18
ANG VEL
I–S(K)
(rd/s)
4E+0
41
ANG VEL
|RES|
(rd/s)
2E+1
20
40
60 ms

Figure 11

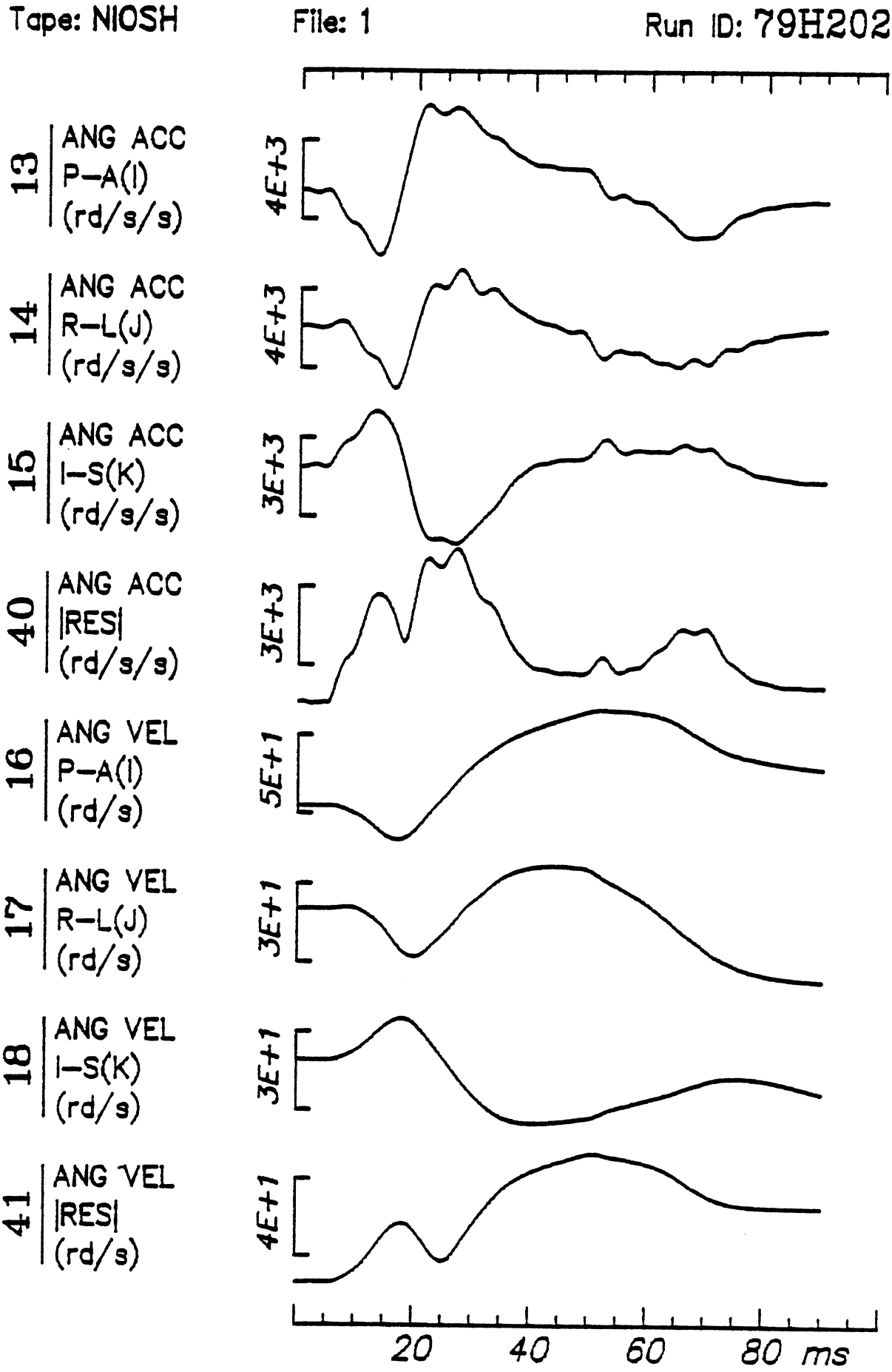

Tape: NIOSH
File: 1
Run ID: 79H202
13
ANG ACC
P–A(I)
(rd/s/s)
4E+3
14
ANG ACC
R–L(J)
(rd/s/s)
4E+3
15
ANG ACC
I–S(K)
(rd/s/s)
3E+3
40
ANG ACC
|RES|
(rd/s/s)
3E+3
16
ANG VEL
P–A(I)
(rd/s)
5E+1
17
ANG VEL
R–L(J)
(rd/s)
3E+1
18
ANG VEL
I–S(K)
(rd/s)
3E+1
41
ANG VEL
|RES|
(rd/s)
4E+1
20
40
60
80 ms

Figure 12

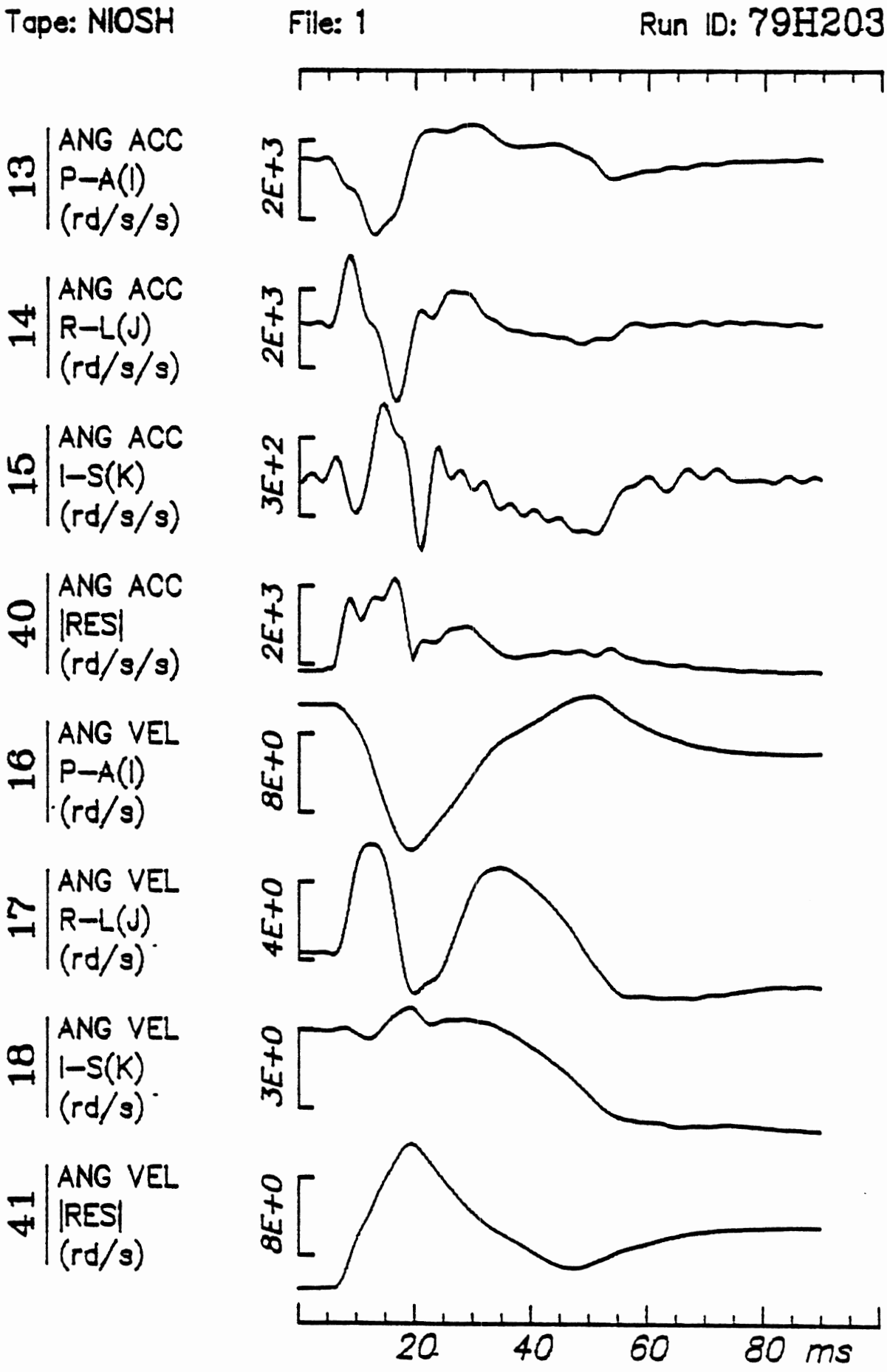

Tape: NIOSH
File: 1
Run ID: 79H203
13
ANG ACC
P–A(I)
(rd/s/s)
2E+3
14
ANG ACC
R–L(J)
(rd/s/s)
2E+3
15
ANG ACC
I–S(K)
(rd/s/s)
3E+2
40
ANG ACC
|RES|
(rd/s/s)
2E+3
16
ANG VEL
P–A(I)
(rd/s)
8E+0
17
ANG VEL
R–L(J)
(rd/s)
4E+0
18
ANG VEL
I–S(K)
(rd/s)
3E+0
41
ANG VEL
|RES|
(rd/s)
8E+0
20
40
60
80 ms

Figure 13

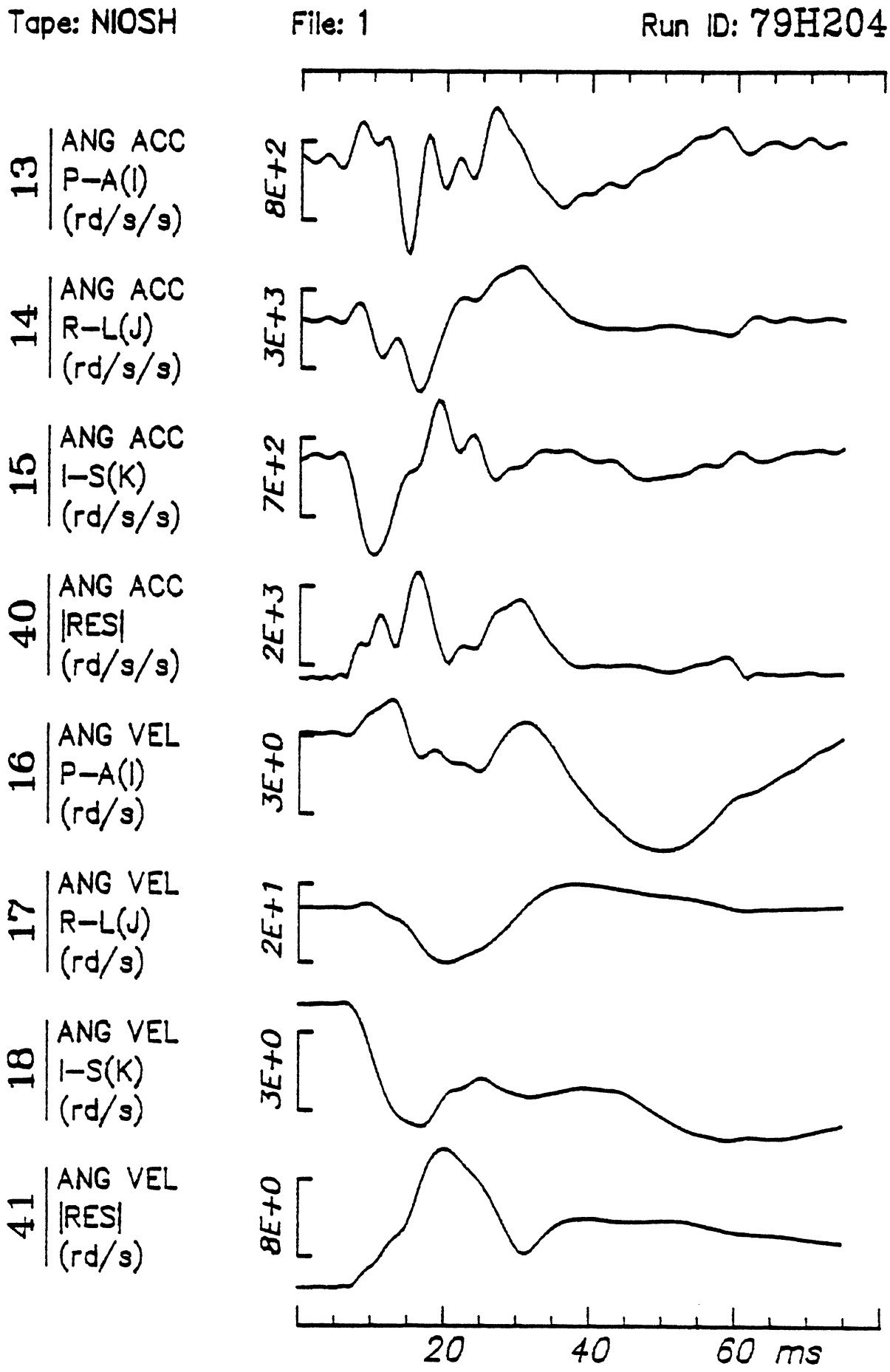

Tape: NIOSH
File: 1
Run ID: 79H204
13
ANG ACC
P–A(I)
(rd/s/s)
8E+2
14
ANG ACC
R–L(J)
(rd/s/s)
3E+3
15
ANG ACC
I–S(K)
(rd/s/s)
7E+2
40
ANG ACC
|RES|
(rd/s/s)
2E+3
16
ANG VEL
P–A(I)
(rd/s)
3E+0
17
ANG VEL
R–L(J)
(rd/s)
2E+1
18
ANG VEL
I–S(K)
(rd/s)
3E+0
41
ANG VEL
|RES|
(rd/s)
8E+0
20
40
60 ms

Figure 14

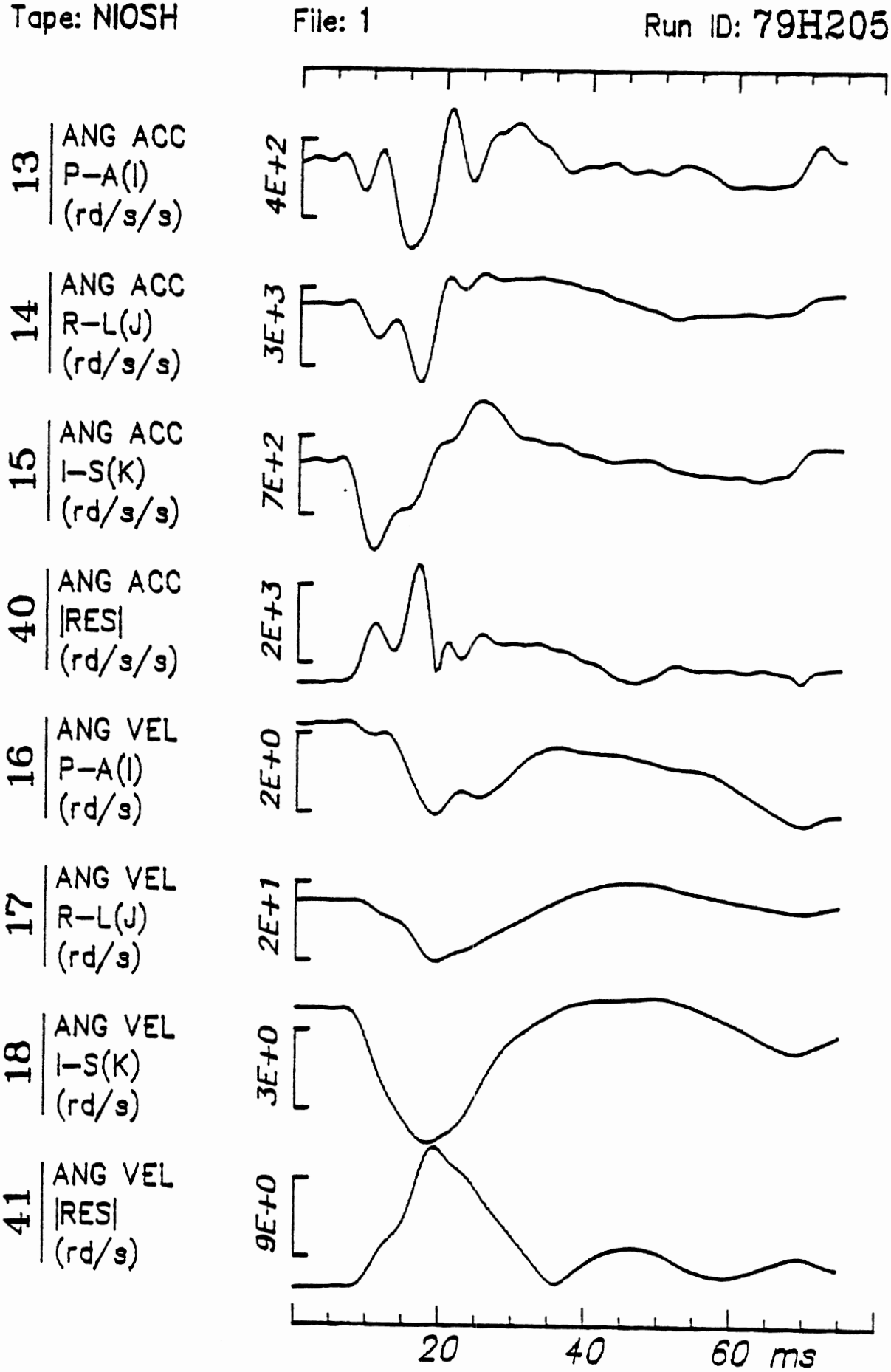

Tape: NIOSH
File: 1
Run ID: 79H205
13
ANG ACC
P-A(I)
(rd/s/s)
4E+2
14
ANG ACC
R-L(J)
(rd/s/s)
3E+3
15
ANG ACC
I-S(K)
(rd/s/s)
7E+2
40
ANG ACC
|RES|
(rd/s/s)
2E+3
16
ANG VEL
P-A(I)
(rd/s)
2E+0
17
ANG VEL
R-L(J)
(rd/s)
2E+1
18
ANG VEL
I-S(K)
(rd/s)
3E+0
41
ANG VEL
|RES|
(rd/s)
9E+0
20
40
60 ms

Figure 15

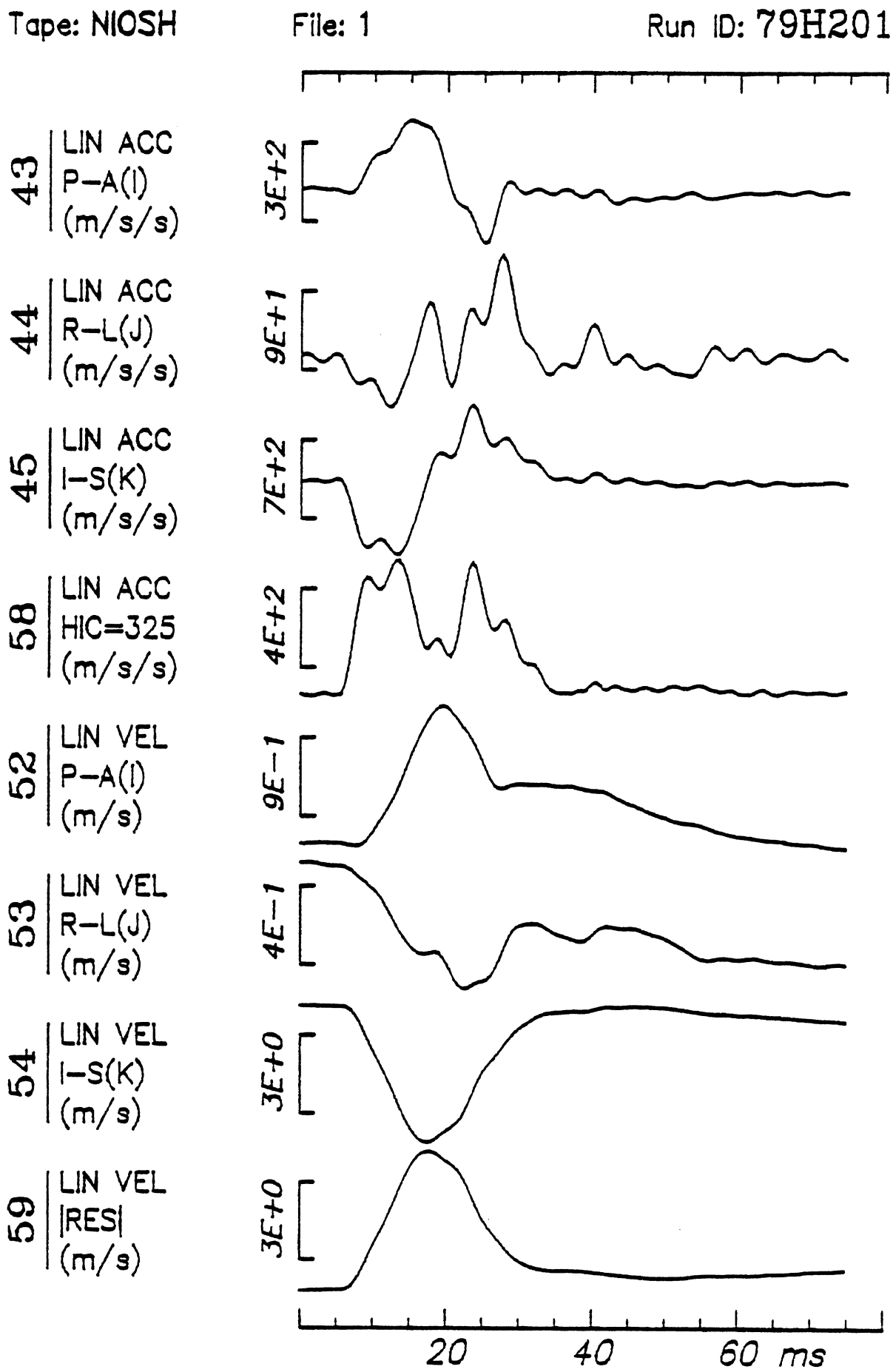

Tape: NIOSH
File: 1
Run ID: 79H201
43
LIN ACC
P-A(I)
(m/s/s)
3E+2
44
LIN ACC
R-L(J)
(m/s/s)
9E+1
45
LIN ACC
I-S(K)
(m/s/s)
7E+2
58
LIN ACC
HIC=325
(m/s/s)
4E+2
52
LIN VEL
P-A(I)
(m/s)
9E-1
53
LIN VEL
R-L(J)
(m/s)
4E-1
54
LIN VEL
I-S(K)
(m/s)
3E+0
59
LIN VEL
|RES|
(m/s)
3E+0
20
40
60 ms

Figure 16

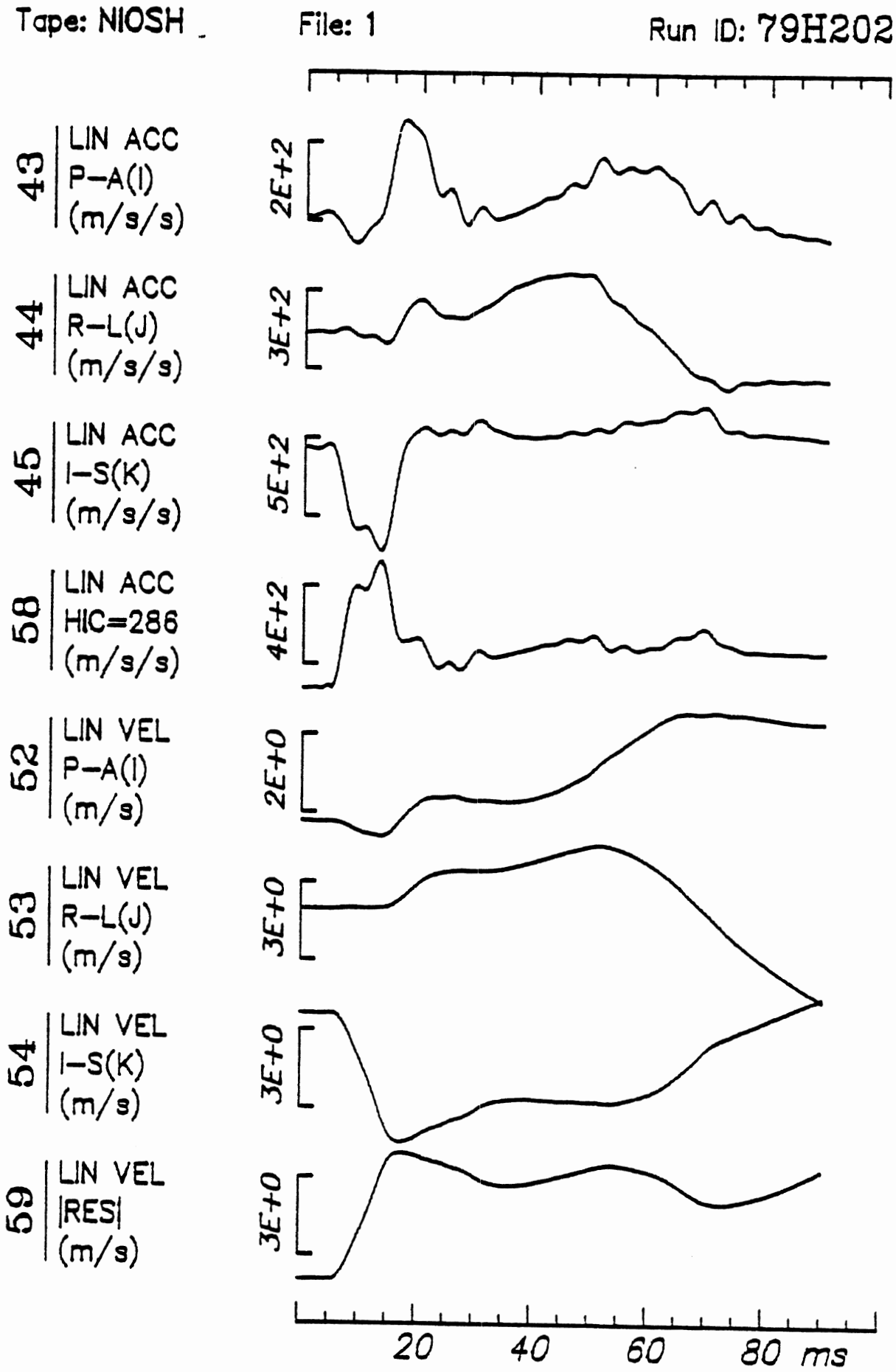

Tape: NIOSH
File: 1
Run ID: 79H202
43
LIN ACC
P-A(I)
(m/s/s)
2E+2
44
LIN ACC
R-L(J)
(m/s/s)
3E+2
45
LIN ACC
I-S(K)
(m/s/s)
5E+2
58
LIN ACC
HIC=286
(m/s/s)
4E+2
52
LIN VEL
P-A(I)
(m/s)
2E+0
53
LIN VEL
R-L(J)
(m/s)
3E+0
54
LIN VEL
I-S(K)
(m/s)
3E+0
59
LIN VEL
|RES|
(m/s)
3E+0
20
40
60
80
ms

Figure 17

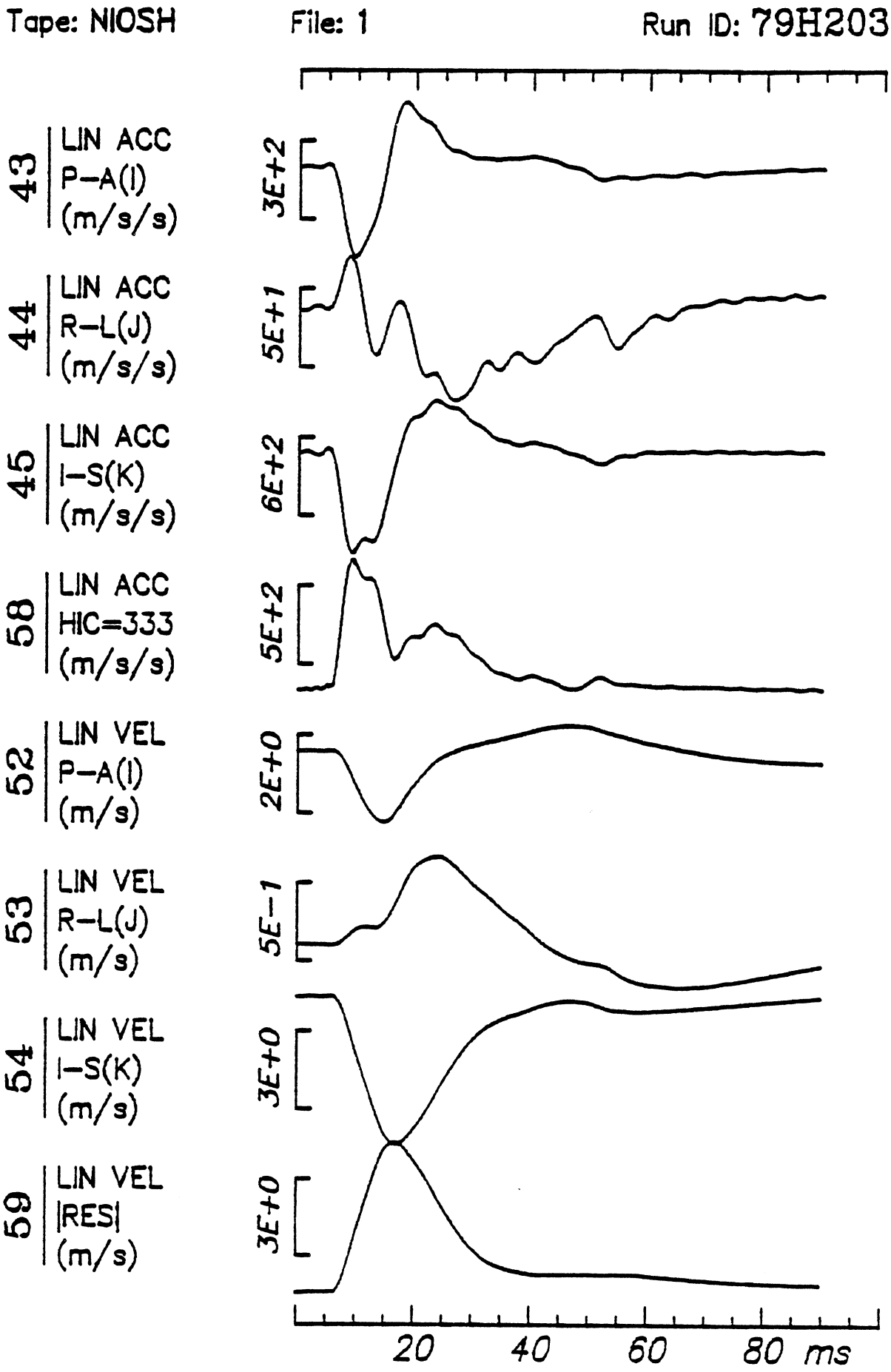

Tape: NIOSH
File: 1
Run ID: 79H203
43
LIN ACC
P-A(I)
(m/s/s)
3E+2
44
LIN ACC
R-L(J)
(m/s/s)
5E+1
45
LIN ACC
I-S(K)
(m/s/s)
6E+2
58
LIN ACC
HIC=333
(m/s/s)
5E+2
52
LIN VEL
P-A(I)
(m/s)
2E+0
53
LIN VEL
R-L(J)
(m/s)
5E-1
54
LIN VEL
I-S(K)
(m/s)
3E+0
59
LIN VEL
|RES|
(m/s)
3E+0
20
40
60
80
ms

Figure 18

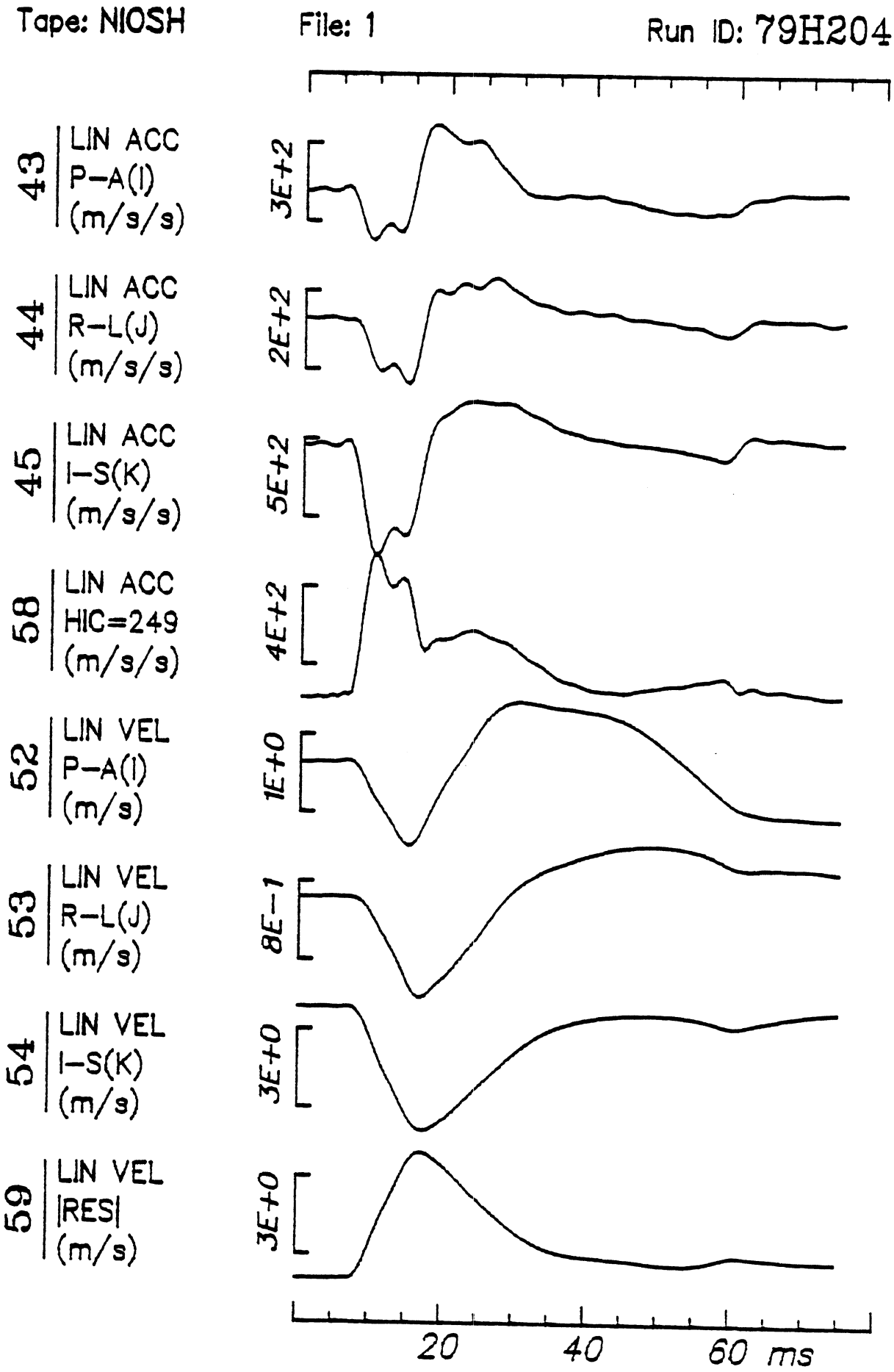

Tape: NIOSH
File: 1
Run ID: 79H204
43
LIN ACC
P-A(I)
(m/s/s)
3E+2
44
LIN ACC
R-L(J)
(m/s/s)
2E+2
45
LIN ACC
I-S(K)
(m/s/s)
5E+2
58
LIN ACC
HIC=249
(m/s/s)
4E+2
52
LIN VEL
P-A(I)
(m/s)
1E+0
53
LIN VEL
R-L(J)
(m/s)
8E-1
54
LIN VEL
I-S(K)
(m/s)
3E+0
59
LIN VEL
|RES|
(m/s)
3E+0
20
40
60 ms

Figure 19

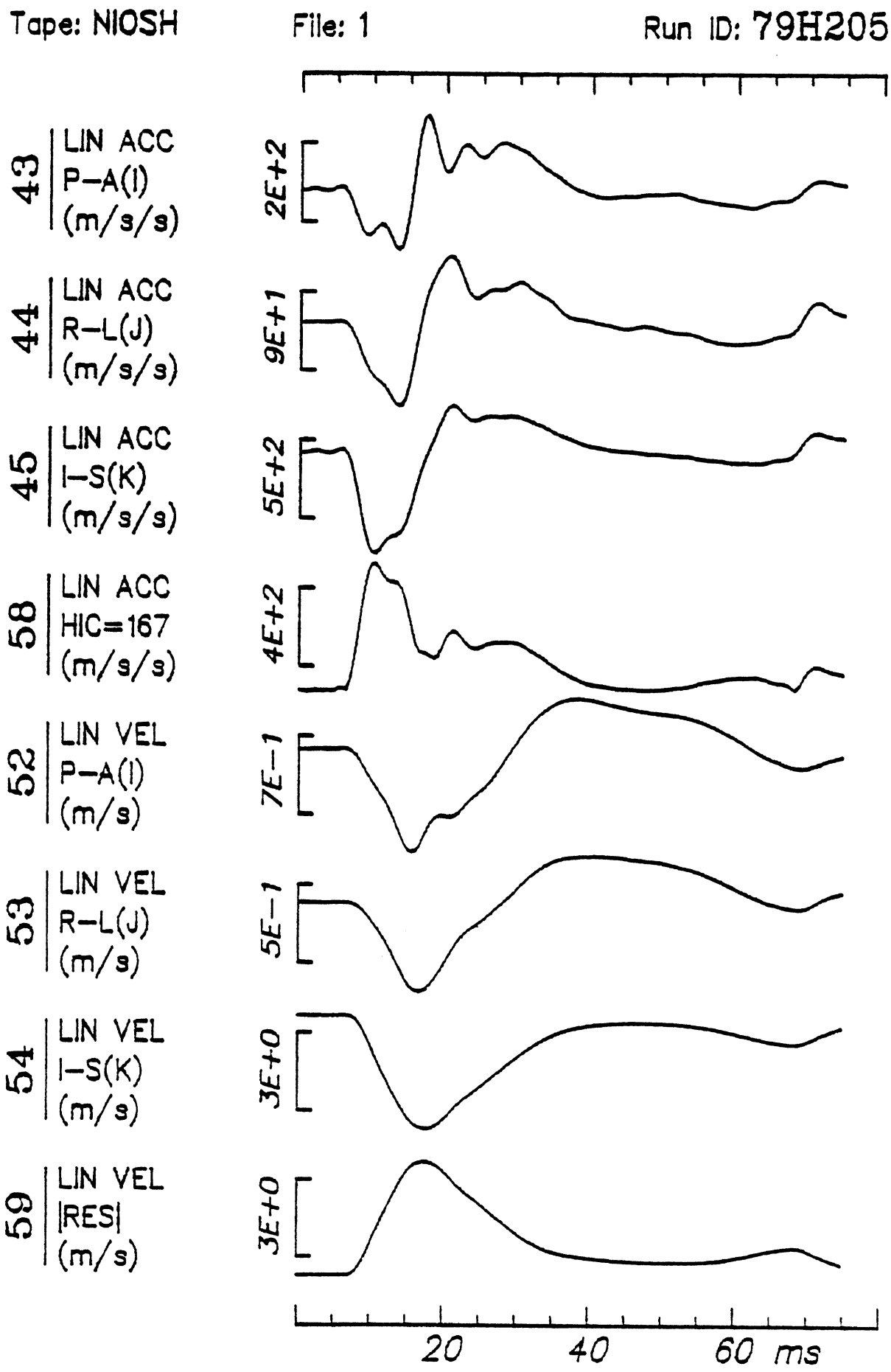

Tape: NIOSH
File: 1
Run ID: 79H205
43
LIN ACC
P-A(I)
(m/s/s)
2E+2
44
LIN ACC
R-L(J)
(m/s/s)
9E+1
45
LIN ACC
I-S(K)
(m/s/s)
5E+2
58
LIN ACC
HIC=167
(m/s/s)
4E+2
52
LIN VEL
P-A(I)
(m/s)
7E-1
53
LIN VEL
R-L(J)
(m/s)
5E-1
54
LIN VEL
I-S(K)
(m/s)
3E+0
59
LIN VEL
|RES|
(m/s)
3E+0
20
40
60 ms

Figure 20

2.4. RESPONSE OF T1 AND T12 VERTEBRAE

Figures 21-25 show the accelerations of T1 in the P-A, R-L and I-S directions and the resultant of these components.

Figures 26-30 show the accelerations of T12 broken in the same directions as T1.

Figures 31-40 show the "velocities" of these accelerations, obtained by simple integration of the components, then finding the resultant of the integrated velocities as the square root of sum of the squared components.

As expected, the highest accelerations and velocity changes are in the S-I direction, i.e., along the impact axis. Note that, in general, the response of T12 is lower than that of T1, indicating that, as the point of observation is moved away from the point of impact, the motion is dissipated. This observation is used as a general guideline for the design of a realistic helmet impact device.

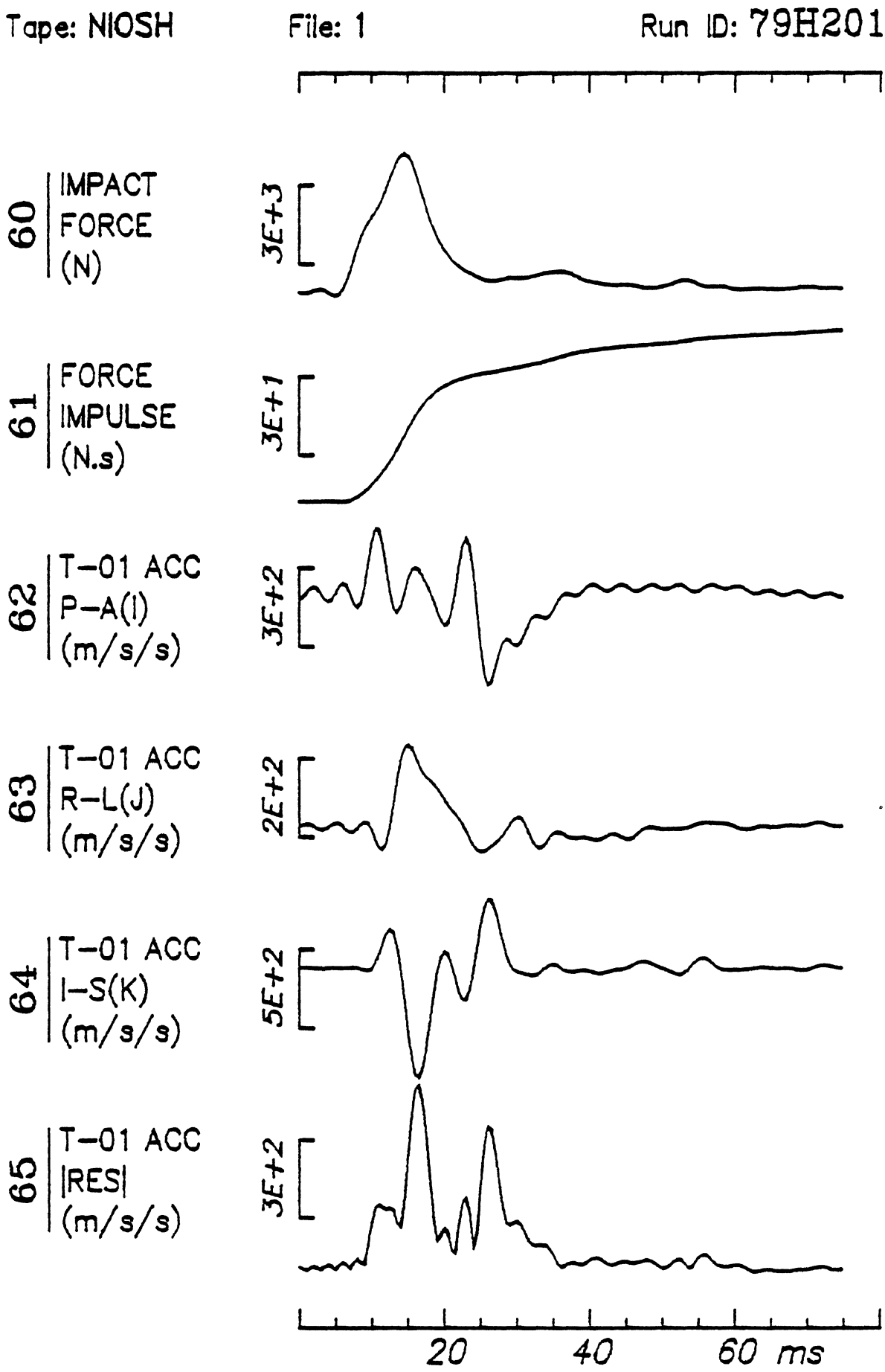

Tape: NIOSH
File: 1
Run ID: 79H201
60
IMPACT
FORCE
(N)
3E+3
61
FORCE
IMPULSE
(N.s)
3E+1
62
T-01 ACC
P-A(I)
(m/s/s)
3E+2
63
T-01 ACC
R-L(J)
(m/s/s)
2E+2
64
T-01 ACC
I-S(K)
(m/s/s)
5E+2
65
T-01 ACC
|RES|
(m/s/s)
3E+2
20
40
60 ms

Figure 21

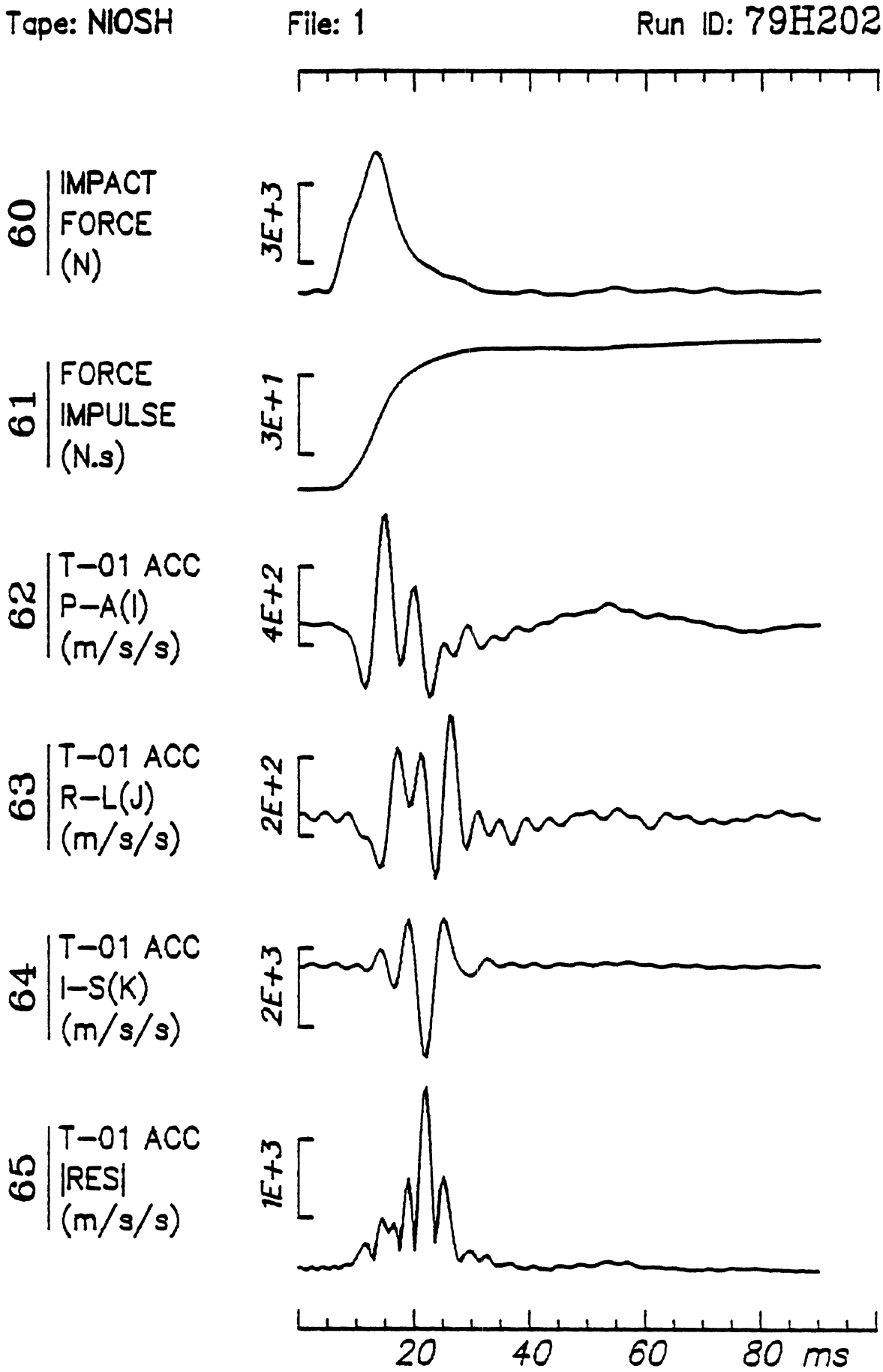

Tape: NIOSH
File: 1
Run ID: 79H202
60
IMPACT
FORCE
(N)
3E+3
61
FORCE
IMPULSE
(N.s)
3E+1
62
T-01 ACC
P-A(I)
(m/s/s)
4E+2
63
T-01 ACC
R-L(J)
(m/s/s)
2E+2
64
T-01 ACC
I-S(K)
(m/s/s)
2E+3
65
T-01 ACC
|RES|
(m/s/s)
1E+3
20
40
60
80
ms

Figure 22

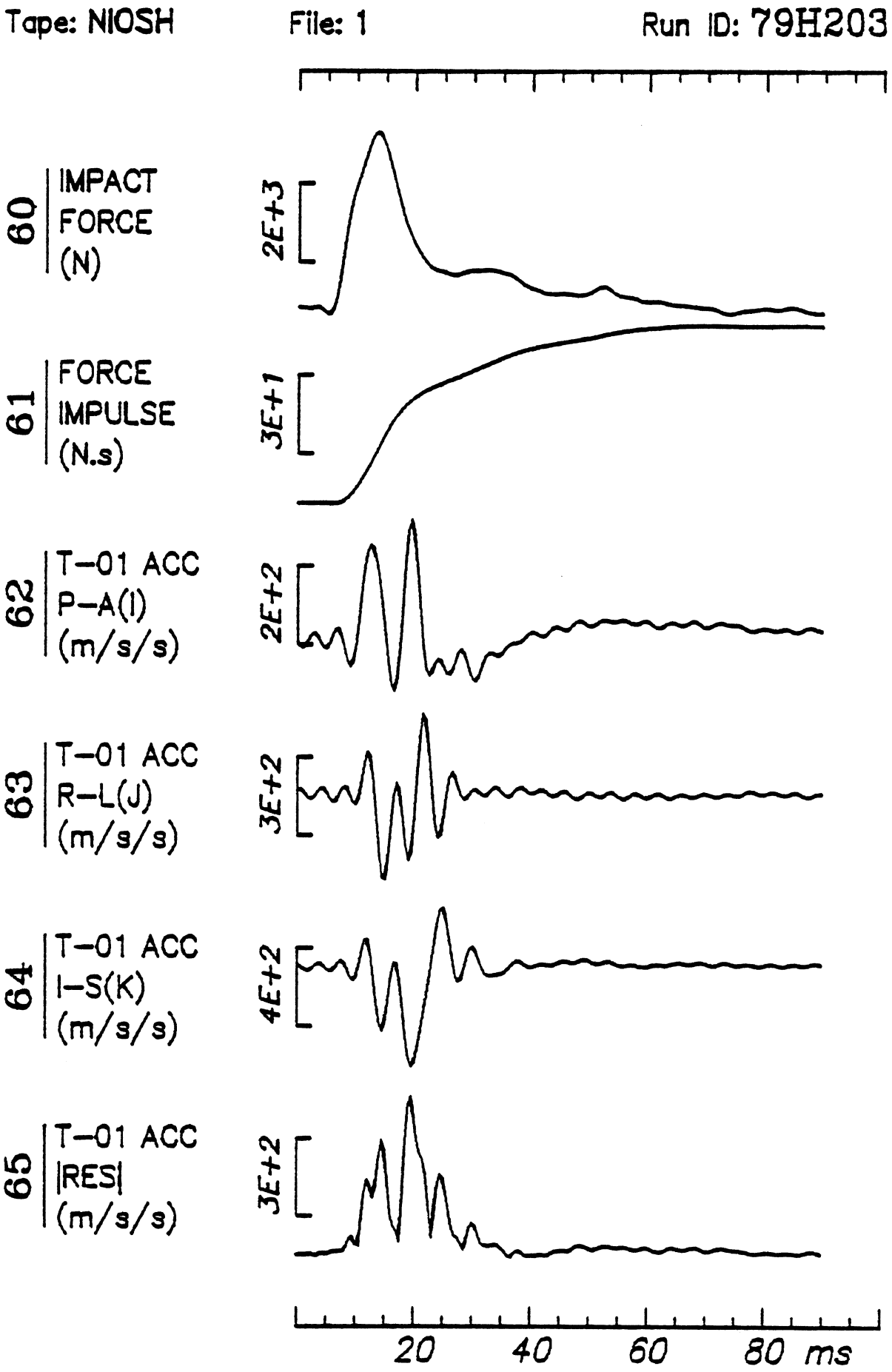

Tape: NIOSH
File: 1
Run ID: 79H203
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
3E+1
62
T-01 ACC
P-A(I)
(m/s/s)
2E+2
63
T-01 ACC
R-L(J)
(m/s/s)
3E+2
64
T-01 ACC
I-S(K)
(m/s/s)
4E+2
65
T-01 ACC
|RES|
(m/s/s)
3E+2
20
40
60
80
ms

Figure 23

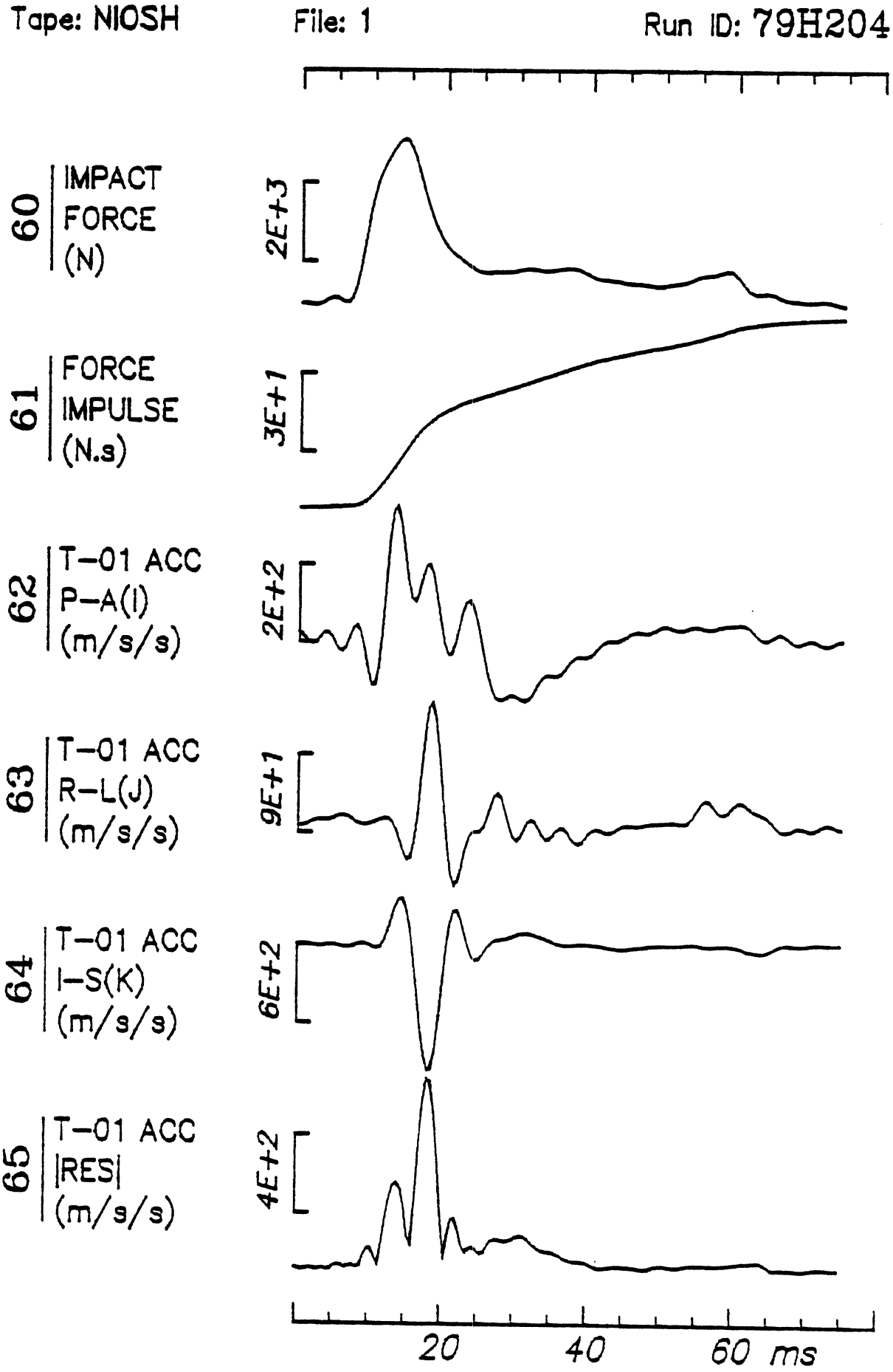

Tape: NIOSH
File: 1
Run ID: 79H204
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
3E+1
62
T-01 ACC
P-A(I)
(m/s/s)
2E+2
63
T-01 ACC
R-L(J)
(m/s/s)
9E+1
64
T-01 ACC
I-S(K)
(m/s/s)
6E+2
65
T-01 ACC
|RES|
(m/s/s)
4E+2
20
40
60 ms

Figure 24

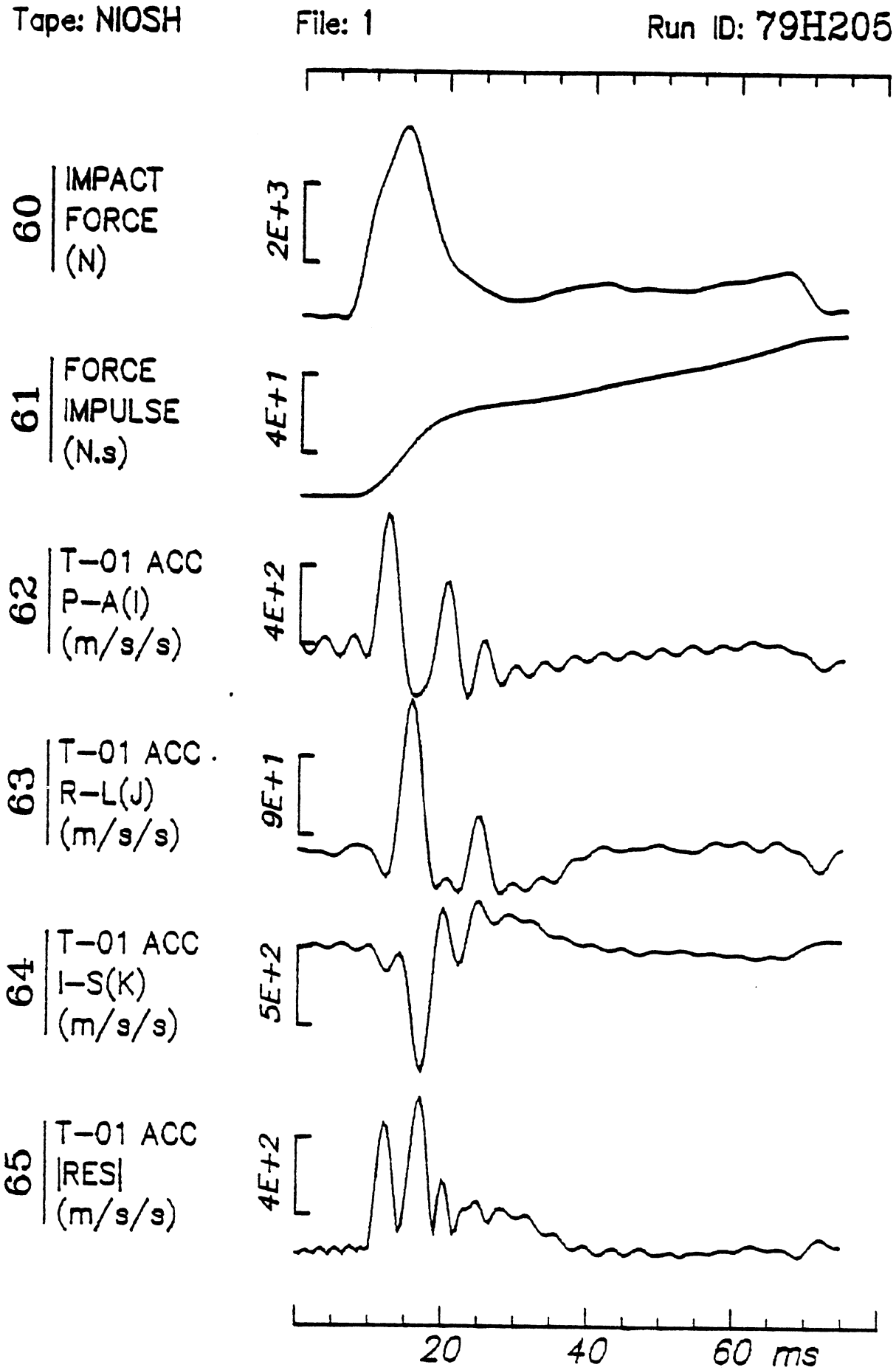

Tape: NIOSH
File: 1
Run ID: 79H205
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
4E+1
62
T-01 ACC
P-A(I)
(m/s/s)
4E+2
63
T-01 ACC
R-L(J)
(m/s/s)
9E+1
64
T-01 ACC
I-S(K)
(m/s/s)
5E+2
65
T-01 ACC
|RES|
(m/s/s)
4E+2
20
40
60 ms

Figure 25

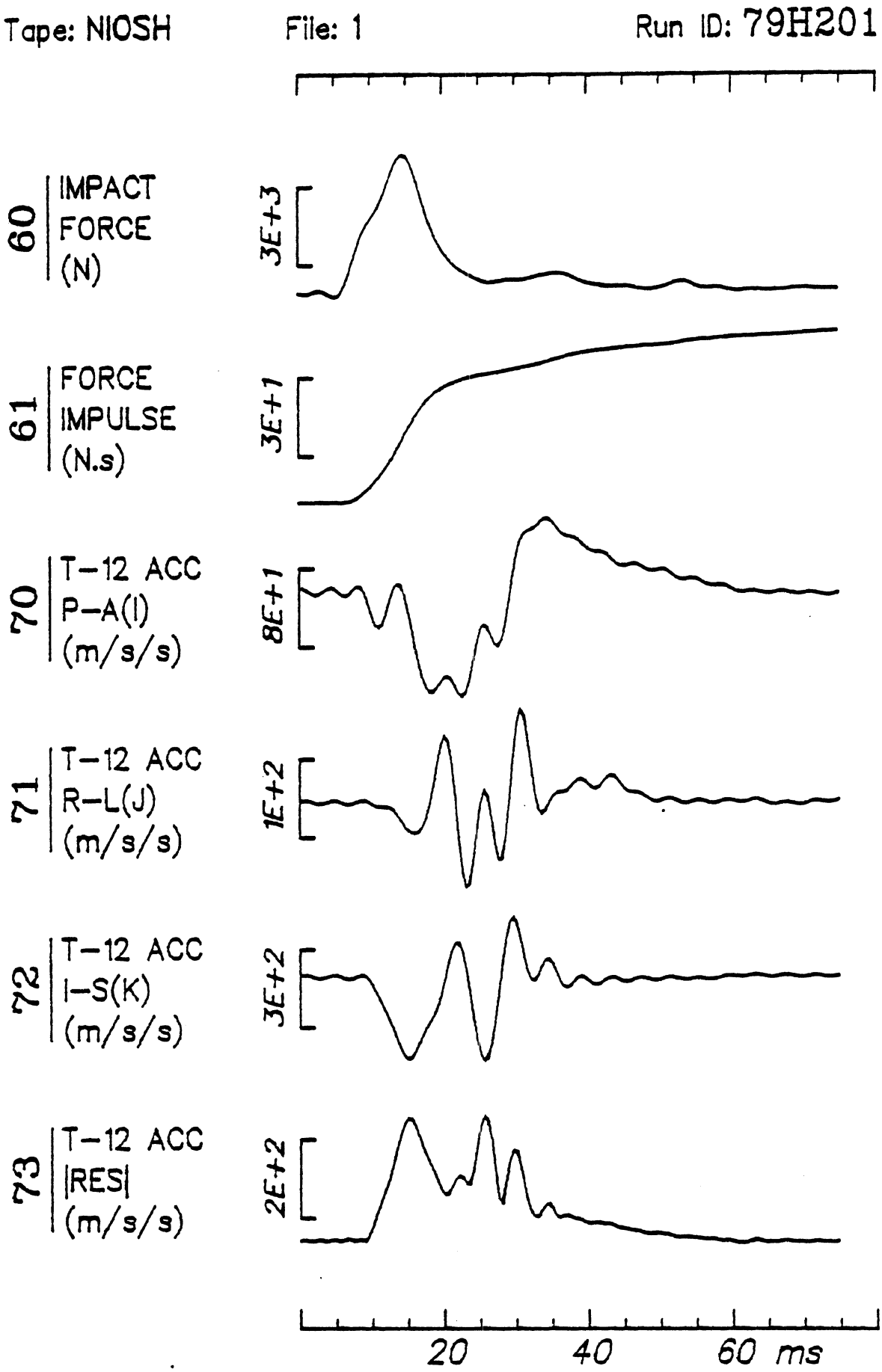

Tape: NIOSH
File: 1
Run ID: 79H201
60
IMPACT
FORCE
(N)
3E+3
61
FORCE
IMPULSE
(N.s)
3E+1
70
T-12 ACC
P-A(I)
(m/s/s)
8E+1
71
T-12 ACC
R-L(J)
(m/s/s)
1E+2
72
T-12 ACC
I-S(K)
(m/s/s)
3E+2
73
T-12 ACC
|RES|
(m/s/s)
2E+2
20
40
60 ms

Figure 26

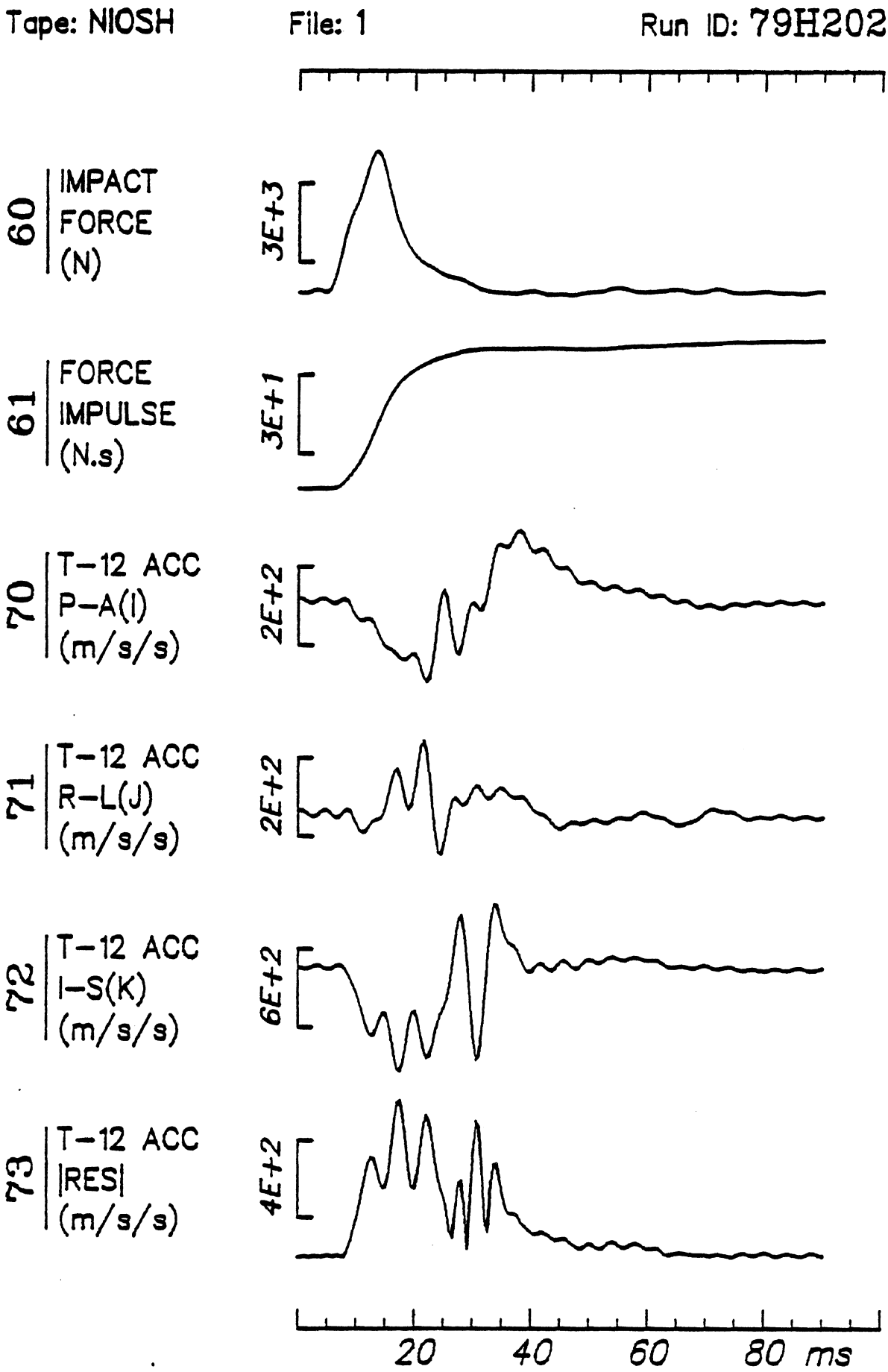

Tape: NIOSH
File: 1
Run ID: 79H202
60
IMPACT
FORCE
(N)
3E+3
61
FORCE
IMPULSE
(N.s)
3E+1
70
T-12 ACC
P-A(I)
(m/s/s)
2E+2
71
T-12 ACC
R-L(J)
(m/s/s)
2E+2
72
T-12 ACC
I-S(K)
(m/s/s)
6E+2
73
T-12 ACC
|RES|
(m/s/s)
4E+2
20
40
60
80
ms

Figure 27

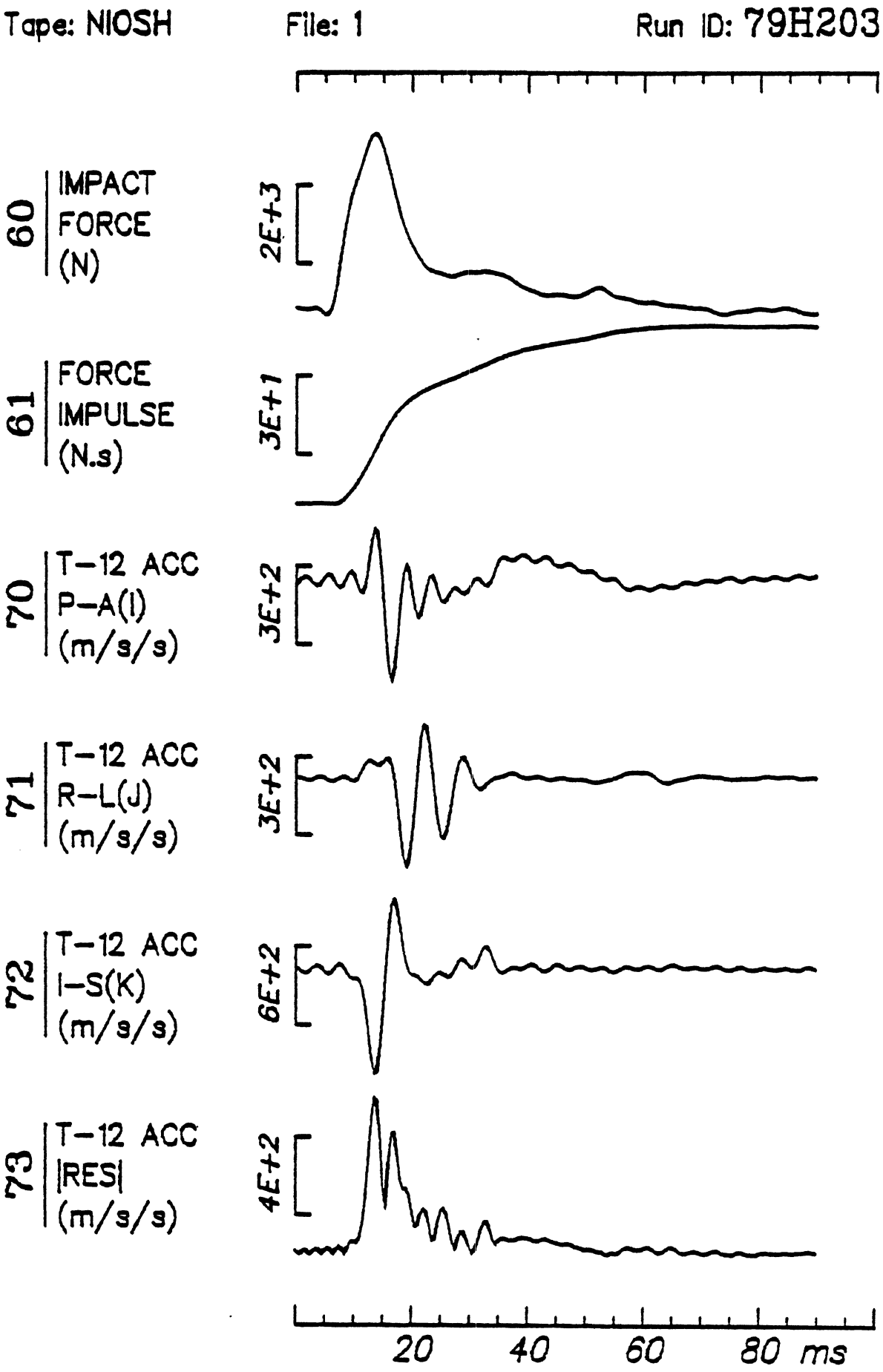

Tape: NIOSH
File: 1
Run ID: 79H203
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
3E+1
70
T-12 ACC
P-A(I)
(m/s/s)
3E+2
71
T-12 ACC
R-L(J)
(m/s/s)
3E+2
72
T-12 ACC
I-S(K)
(m/s/s)
6E+2
73
T-12 ACC
|RES|
(m/s/s)
4E+2
20
40
60
80
ms

Figure 28

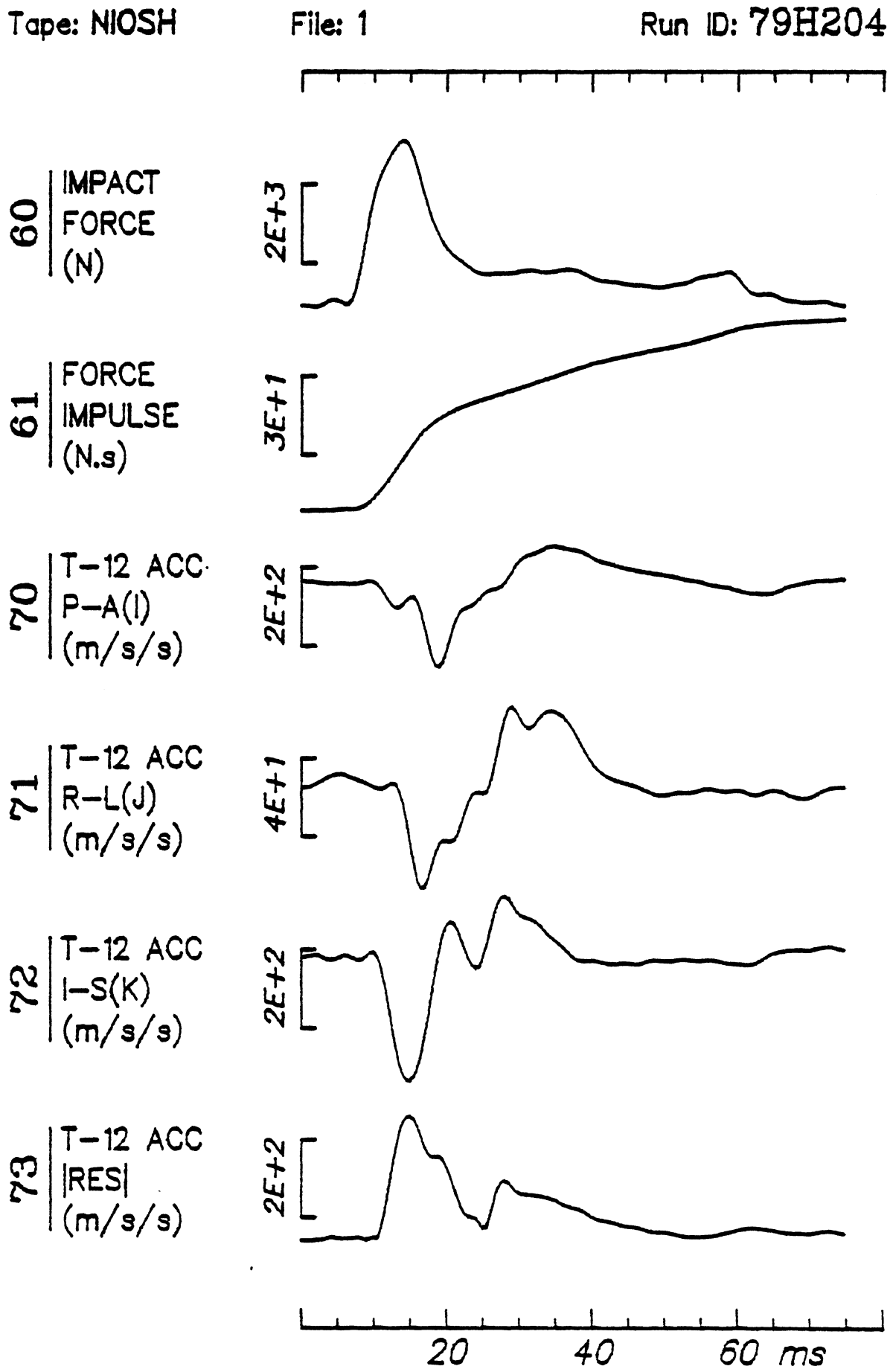

Tape: NIOSH
File: 1
Run ID: 79H204
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
3E+1
70
T-12 ACC
P-A(I)
(m/s/s)
2E+2
71
T-12 ACC
R-L(J)
(m/s/s)
4E+1
72
T-12 ACC
I-S(K)
(m/s/s)
2E+2
73
T-12 ACC
|RES|
(m/s/s)
2E+2
20
40
60 ms

Figure 29

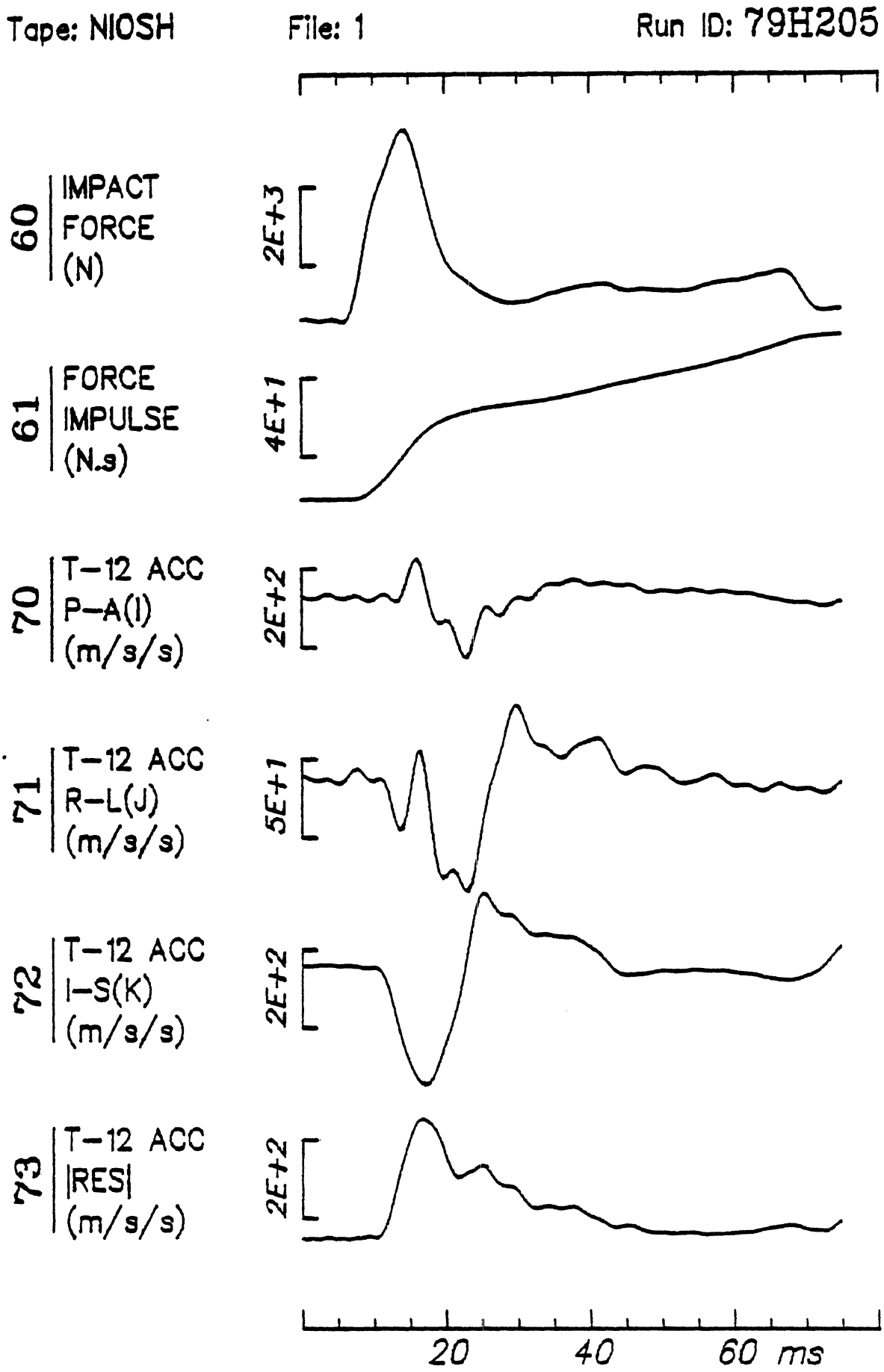

Tape: NIOSH
File: 1
Run ID: 79H205
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
4E+1
70
T-12 ACC
P-A(I)
(m/s/s)
2E+2
71
T-12 ACC
R-L(J)
(m/s/s)
5E+1
72
T-12 ACC
I-S(K)
(m/s/s)
2E+2
73
T-12 ACC
|RES|
(m/s/s)
2E+2
20
40
60 ms

Figure 30

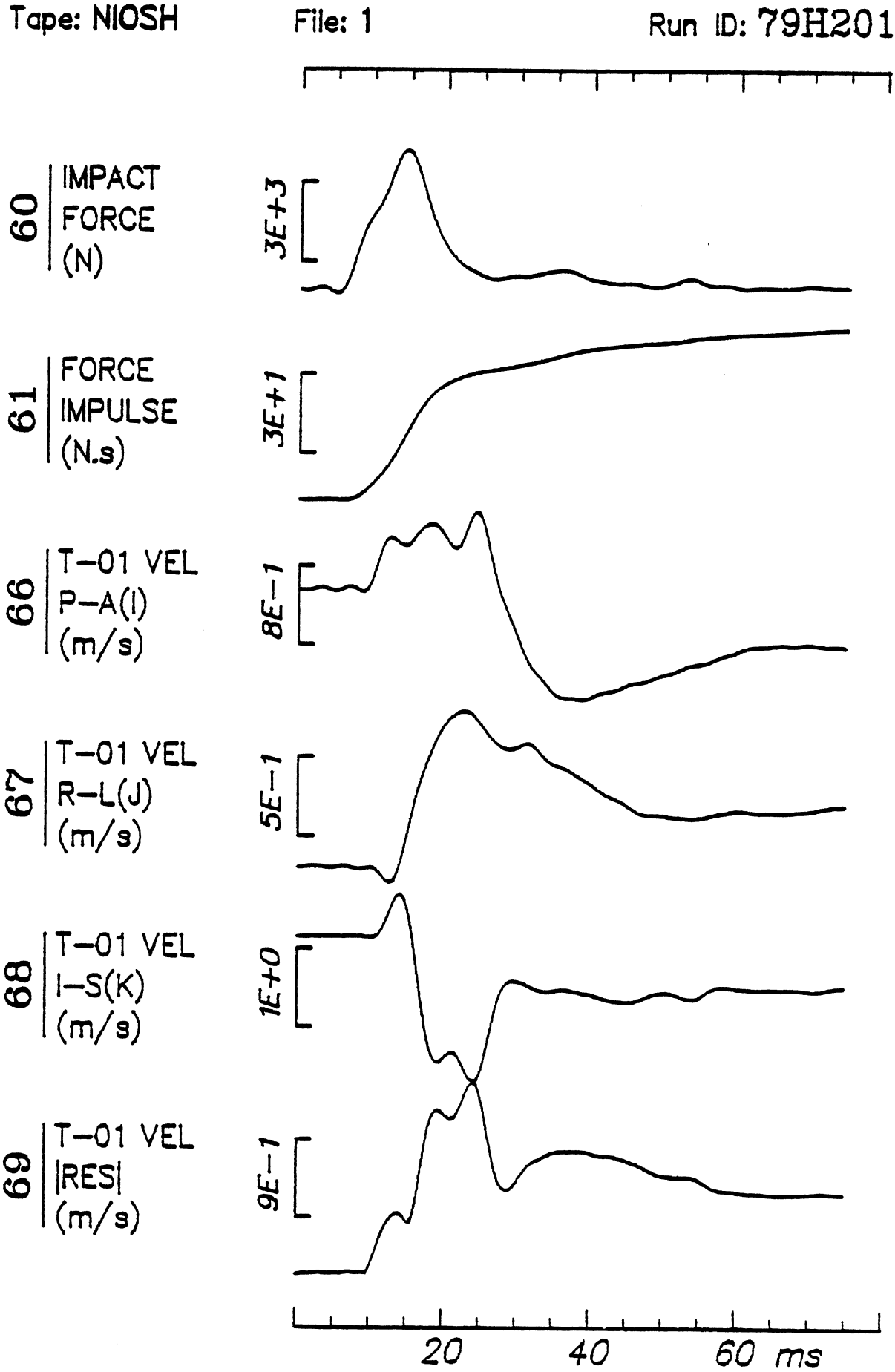

Tape: NIOSH
File: 1
Run ID: 79H201
60
IMPACT
FORCE
(N)
3E+3
61
FORCE
IMPULSE
(N.s)
3E+1
66
T-01 VEL
P-A(I)
(m/s)
8E-1
67
T-01 VEL
R-L(J)
(m/s)
5E-1
68
T-01 VEL
I-S(K)
(m/s)
1E+0
69
T-01 VEL
|RES|
(m/s)
9E-1
20
40
60 ms

Figure 31

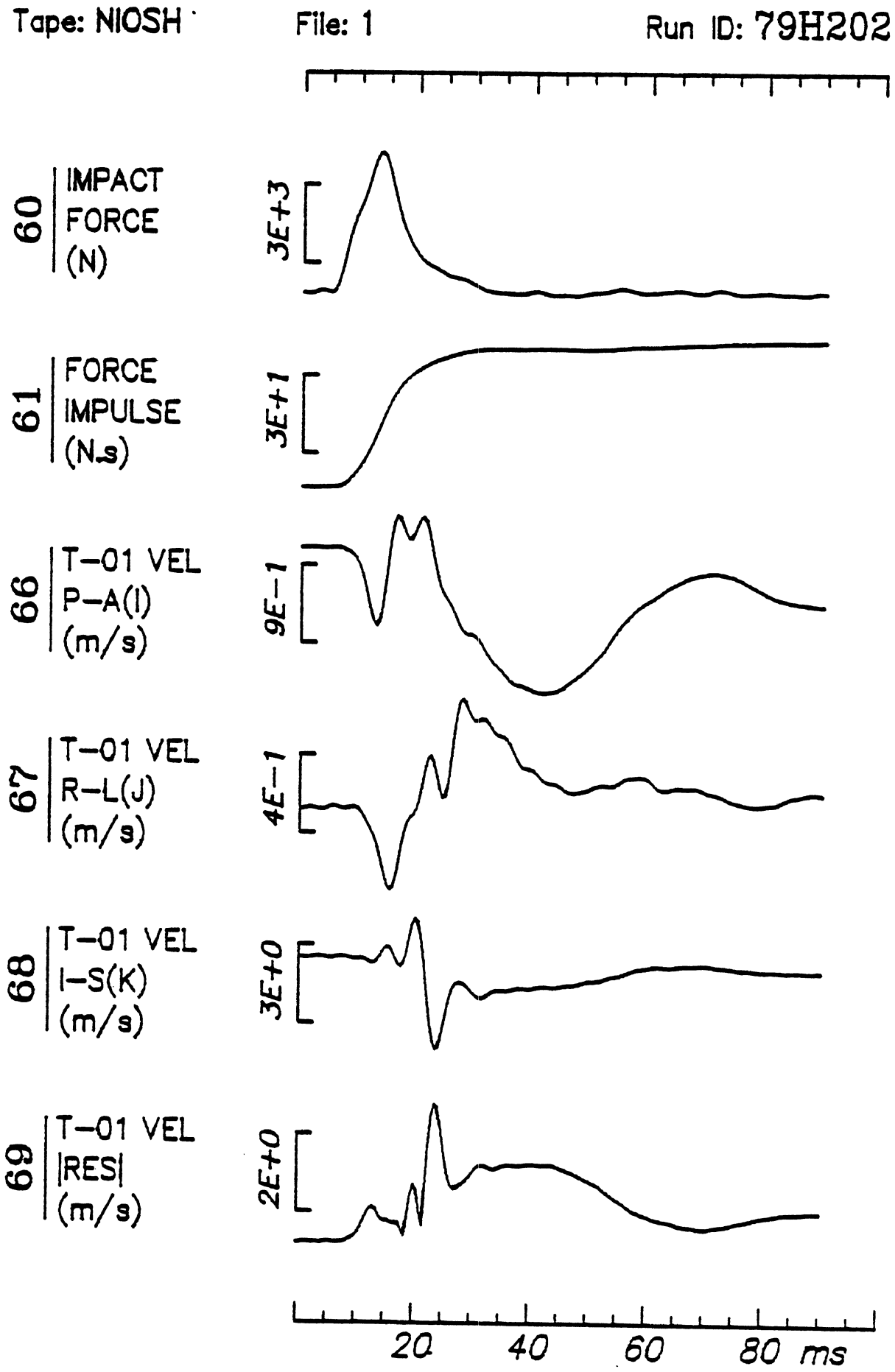

Tape: NIOSH
File: 1
Run ID: 79H202
60
IMPACT
FORCE
(N)
3E+3
61
FORCE
IMPULSE
(N.s)
3E+1
66
T-01 VEL
P-A(I)
(m/s)
9E-1
67
T-01 VEL
R-L(J)
(m/s)
4E-1
68
T-01 VEL
I-S(K)
(m/s)
3E+0
69
T-01 VEL
|RES|
(m/s)
2E+0
20
40
60
80
ms

Figure 32

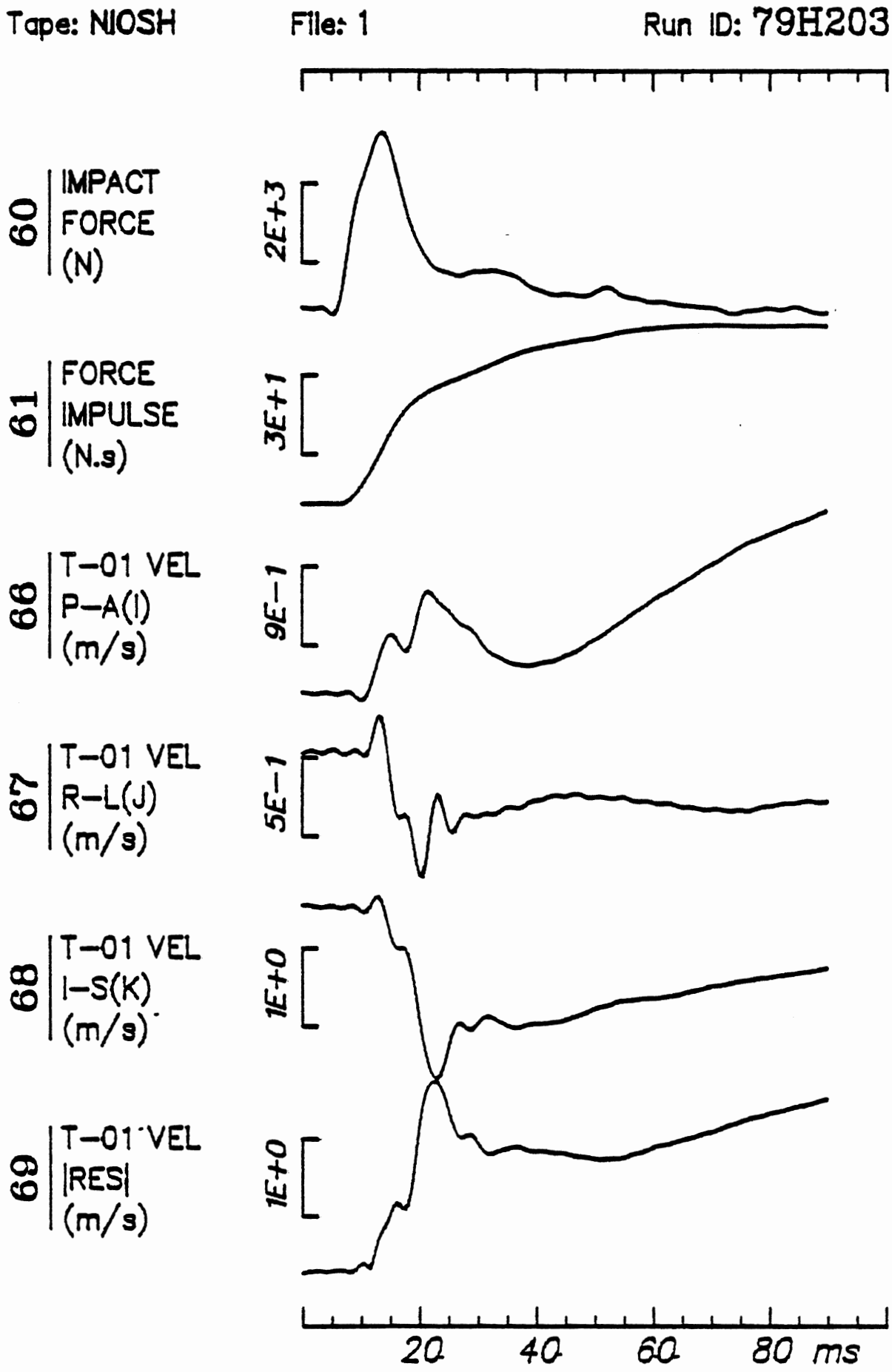

Tape: NIOSH
File: 1
Run ID: 79H203
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
3E+1
66
T-01 VEL
P-A(I)
(m/s)
9E-1
67
T-01 VEL
R-L(J)
(m/s)
5E-1
68
T-01 VEL
I-S(K)
(m/s)
1E+0
69
T-01 VEL
|RES|
(m/s)
1E+0
20
40
60
80 ms

Figure 33

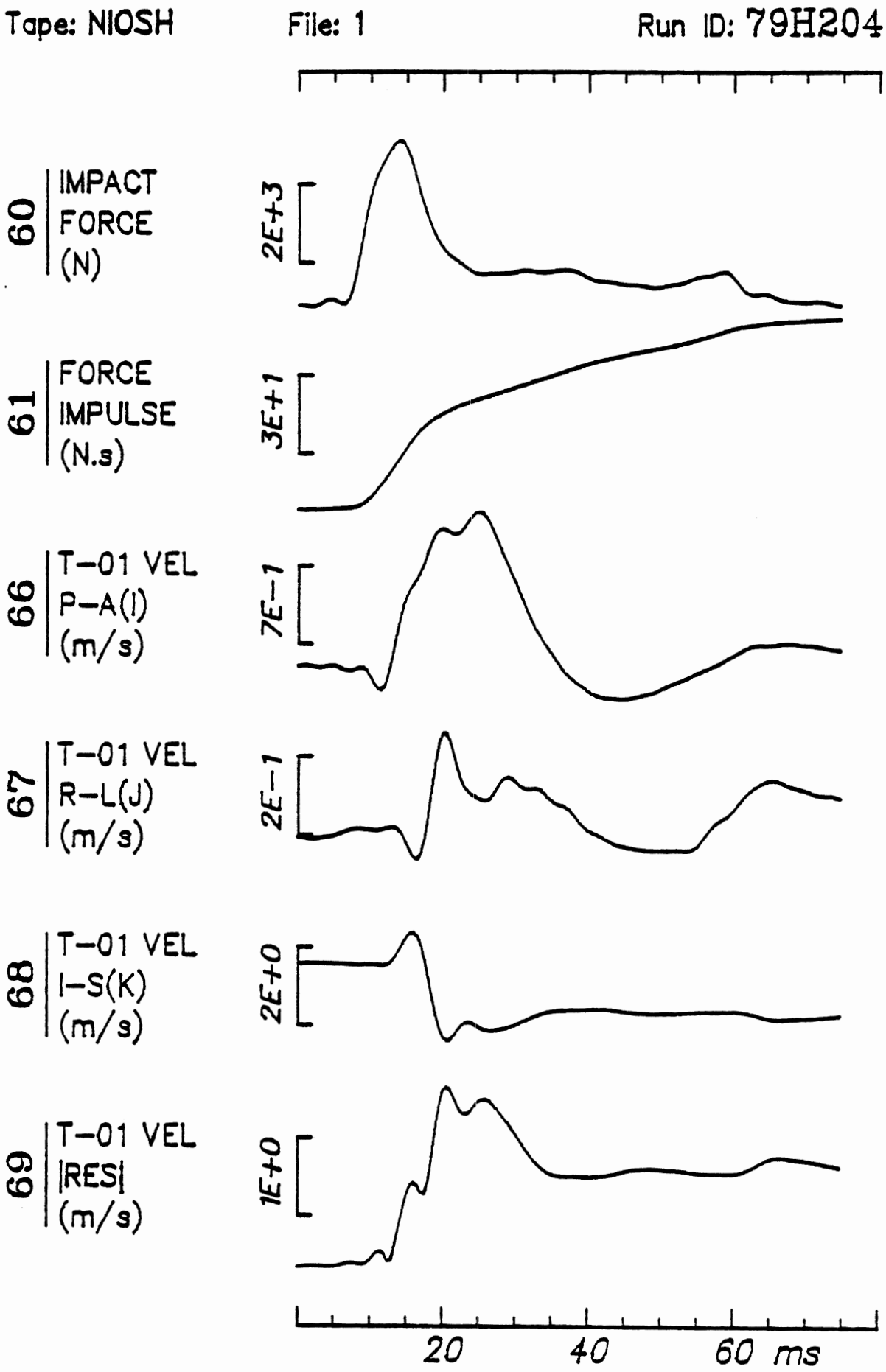

Tape: NIOSH
File: 1
Run ID: 79H204
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
3E+1
66
T-01 VEL
P-A(I)
(m/s)
7E-1
67
T-01 VEL
R-L(J)
(m/s)
2E-1
68
T-01 VEL
I-S(K)
(m/s)
2E+0
69
T-01 VEL
|RES|
(m/s)
1E+0
20
40
60 ms

Figure 34

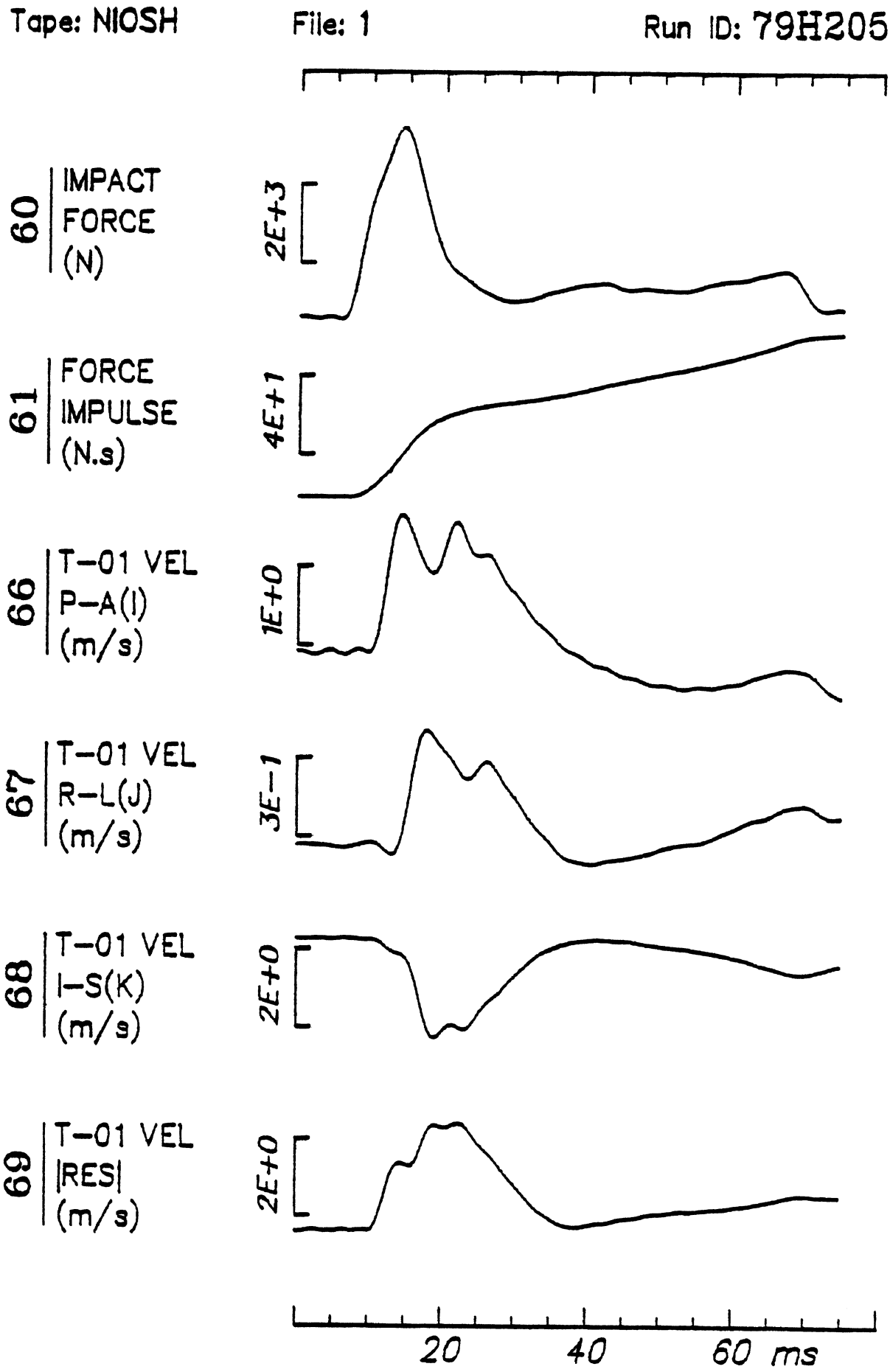

Tape: NIOSH
File: 1
Run ID: 79H205
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
4E+1
66
T-01 VEL
P-A(I)
(m/s)
1E+0
67
T-01 VEL
R-L(J)
(m/s)
3E-1
68
T-01 VEL
I-S(K)
(m/s)
2E+0
69
T-01 VEL
|RES|
(m/s)
2E+0
20
40
60 ms

Figure 35

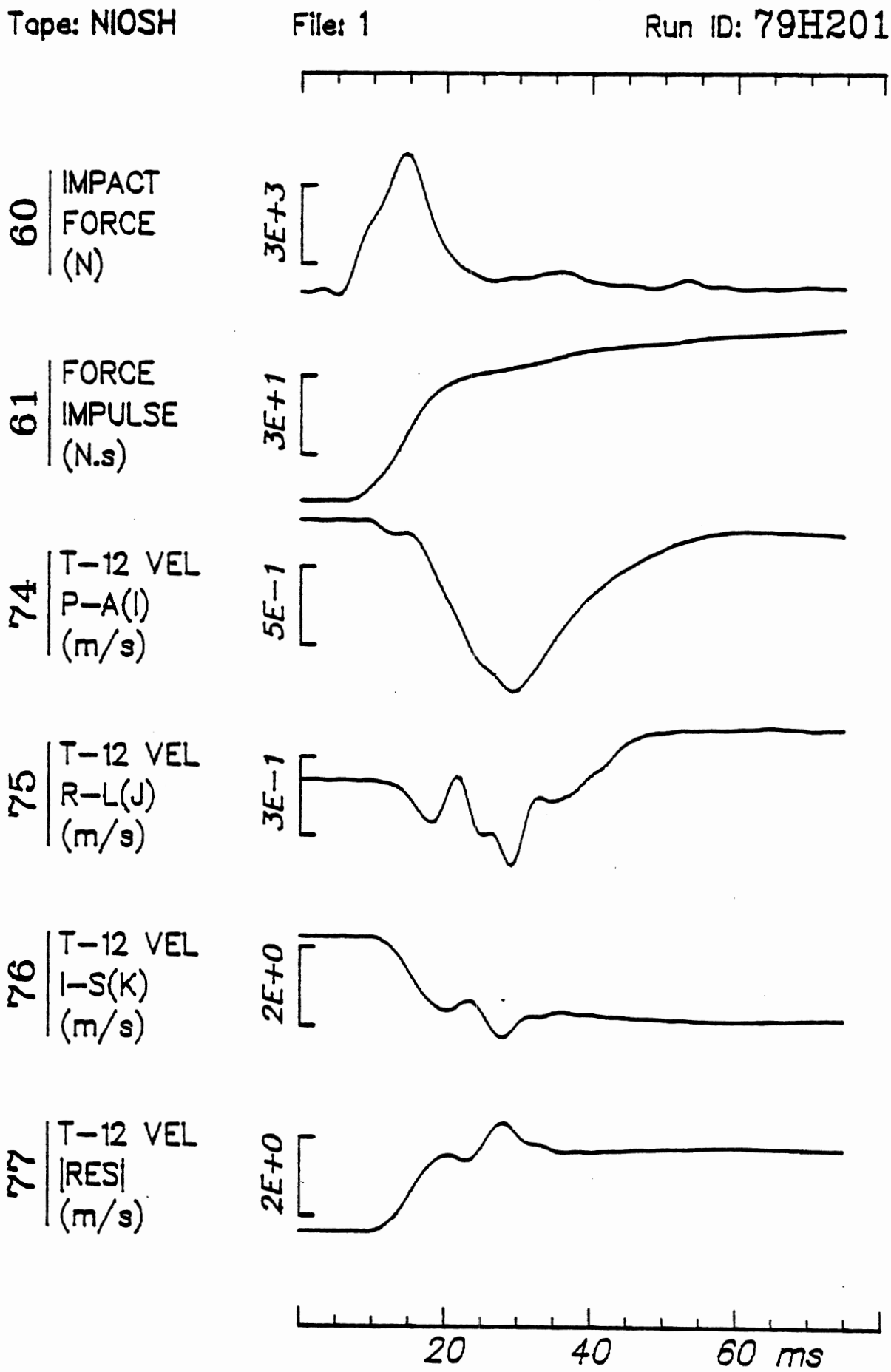

Tape: NIOSH
File: 1
Run ID: 79H201
60
IMPACT
FORCE
(N)
3E+3
61
FORCE
IMPULSE
(N.s)
3E+1
74
T-12 VEL
P-A(I)
(m/s)
5E-1
75
T-12 VEL
R-L(J)
(m/s)
3E-1
76
T-12 VEL
I-S(K)
(m/s)
2E+0
77
T-12 VEL
|RES|
(m/s)
2E+0
20
40
60 ms

Figure 36

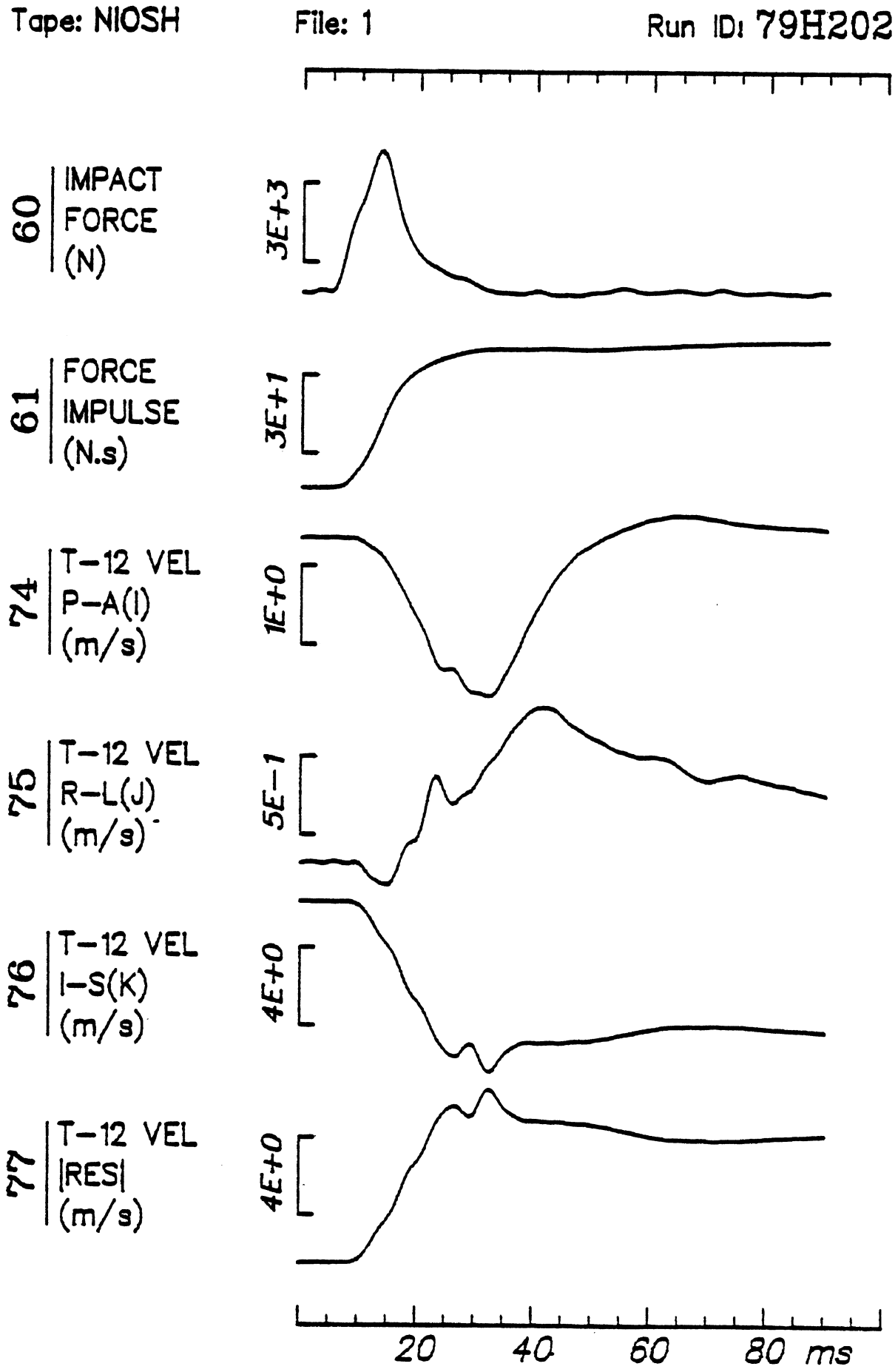

Tape: NIOSH
File: 1
Run ID: 79H202
60
IMPACT
FORCE
(N)
3E+3
61
FORCE
IMPULSE
(N.s)
3E+1
74
T-12 VEL
P-A(I)
(m/s)
1E+0
75
T-12 VEL
R-L(J)
(m/s)
5E-1
76
T-12 VEL
I-S(K)
(m/s)
4E+0
77
T-12 VEL
|RES|
(m/s)
4E+0
20
40
60
80
ms

Figure 37

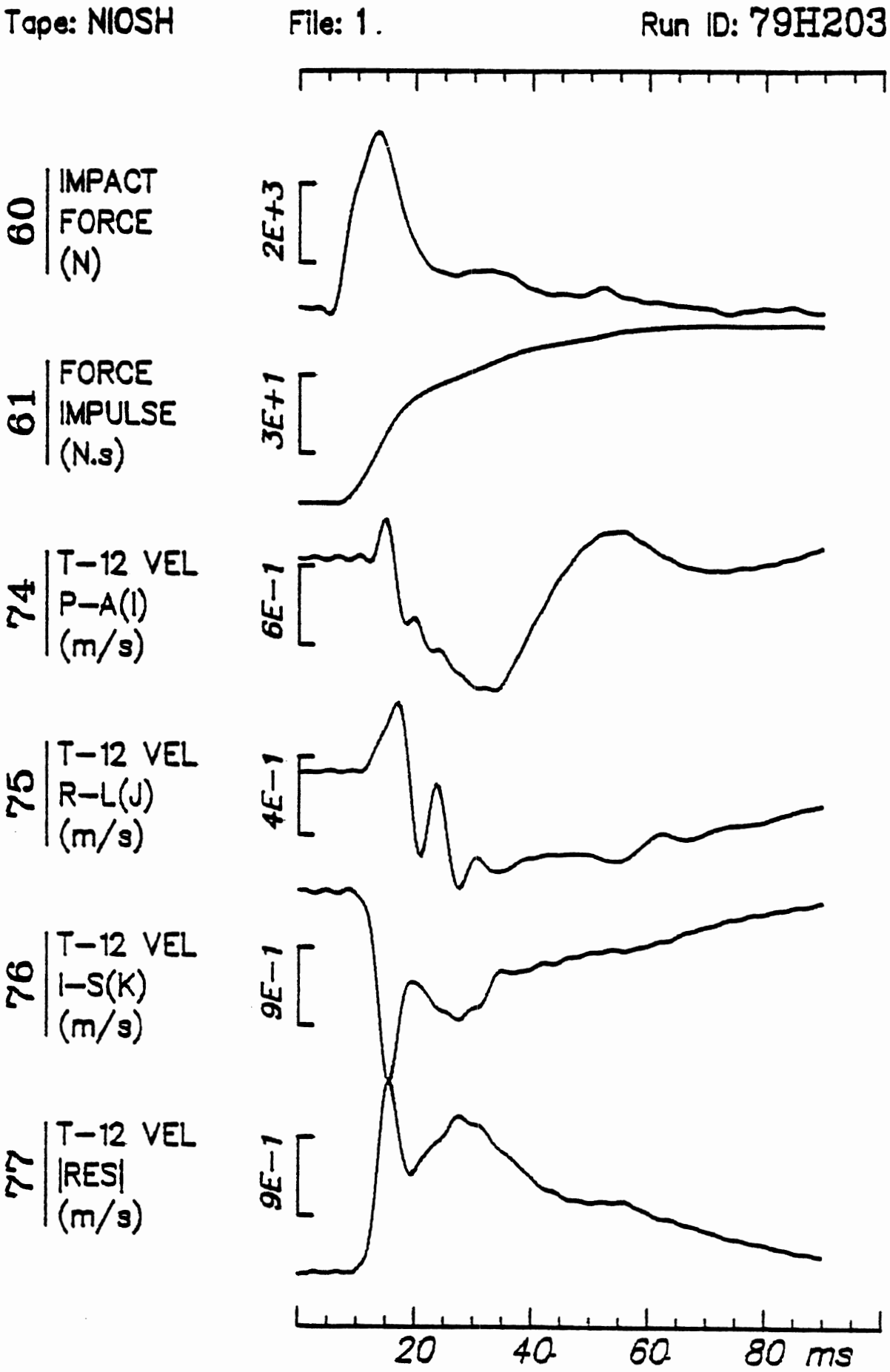

Tape: NIOSH
File: 1.
Run ID: 79H203
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
3E+1
74
T-12 VEL
P-A(I)
(m/s)
6E-1
75
T-12 VEL
R-L(J)
(m/s)
4E-1
76
T-12 VEL
I-S(K)
(m/s)
9E-1
77
T-12 VEL
|RES|
(m/s)
9E-1
20
40
60
80 ms

Figure 38

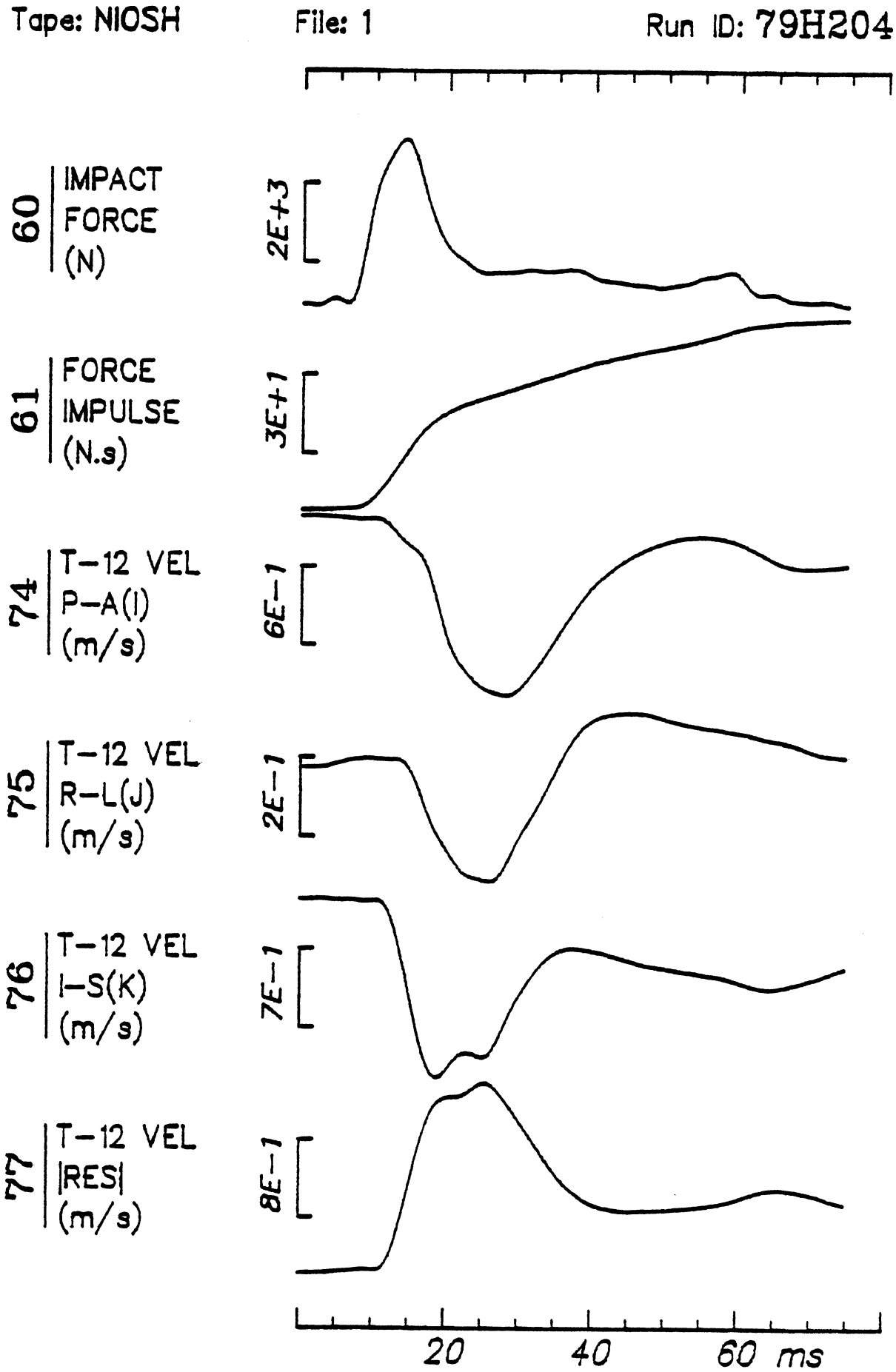

Tape: NIOSH
File: 1
Run ID: 79H204
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
3E+1
74
T-12 VEL
P-A(I)
(m/s)
6E-1
75
T-12 VEL
R-L(J)
(m/s)
2E-1
76
T-12 VEL
I-S(K)
(m/s)
7E-1
77
T-12 VEL
|RES|
(m/s)
8E-1
20
40
60 ms

Figure 39

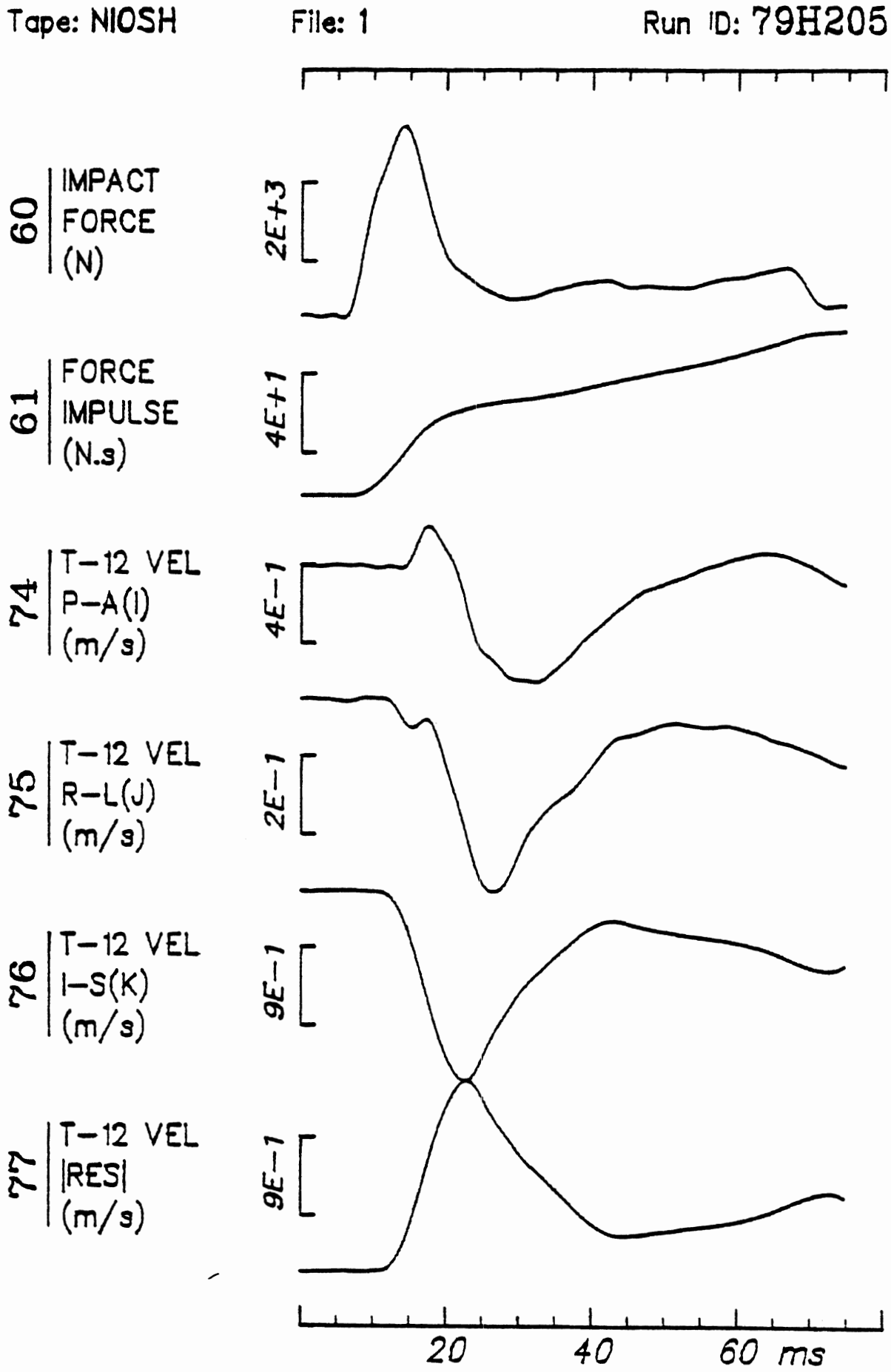

Tape: NIOSH
File: 1
Run ID: 79H205
60
IMPACT
FORCE
(N)
2E+3
61
FORCE
IMPULSE
(N.s)
4E+1
74
T-12 VEL
P-A(I)
(m/s)
4E-1
75
T-12 VEL
R-L(J)
(m/s)
2E-1
76
T-12 VEL
I-S(K)
(m/s)
9E-1
77
T-12 VEL
|RES|
(m/s)
9E-1
20
40
60 ms

Figure 40

2.5. DYNAMICS OF HEAD IMPACT

The head data presented earlier may be described as kinematic data which apply to any rigid body regardless of its inertial properties. In order to describe the dynamics of impact, the mass of the rigid body (i.e., the head) and its moment of inertia matrix must be known. The dynamic analysis of impact should then yield the reactions of the head, assumed to be a rigid body, at the neck joint.

Although the head motion with respect to the neck is a sliding-rolling motion about the occipital condyles, a reasonable model for this joint is to assume a simple "joint" at which the head exerts on the neck a single force vector and a single moment vector which may be resolved along 3 orthogonal directions.

Therefore, the following quantities, needed for the dynamic analysis, were defined:

a) the location of the head-neck connection point (assumed to be mid-way between the two condyles) relative to the head reference frame;

b) the location of center of mass in the head reference frame, which is near but not the anatomical center;

c) the mass of the head;

d) the moment of inertia matrix of the head about the head standard reference frame. Alternately, the principal axes of inertia must be defined along with 3 principal moments of inertia.

The above quantities were determined as follows:

a) Condyle locations were determined by x-ray analysis of corresponding lead targets. These coordinates are included in the earlier Figures 1-5.

b) Head center of mass was always assumed to be in the mid-sagittal plane, with coordinates (1-3, 0-0, 2,1) cm along the head (P-A, R-L, I-S) axes, respectively. These coordinates are justified in a study by E. B. Becker entitled "Measurement of Mass Distribution Parameters of Anatomical Segments".

c) Head mass was computed by a regression model developed using R. F. Chandler's, "Investigation of Inertial Properties of the Human Body", a study where exhaustive measurements were made on 6 cadaver heads.

d) Head principal moments of inertia were computed using a regression model developed by D. G. Lett in a report entitled, "Estimating Moments of Inertia of the Head From Standard Anthropometric Data".

Once all these quantities are determined, the forces and moments at the condyles may be calculated. This method is the only way available for this determination since it is hardly feasible to implant let alone develop a six-channel transducer to measure the 6 reactions at the condyles.

In the regression models, inertial properties of the head were predicted using 4 anthropometric measurements which are:

LEN . . . . A-P length;
BRT . . . . L-R breadth;
HGT . . . . S-I height;
CIR . . . . Circumference of the head.

The regression models are presented in Figures 41 and 42, the dependent variables are the mass of the head (MAS), and the 3 principal moments of inertia (IUU, IVV, IWW). The correlation coefficients range from 0.86 to 0.95, an extremely high correlation for biological materials and systems where varation is wide.

Using these models, and anthropometric measurements of the 5 tested cadavers, shown in Figure 43, and the kinematic results of the 3-D rigid body motion described earlier, it was possible to calculate the time histories described below.

The impact force was assumed to act in a fixed laboratory direction while the head (and its reference axes) were moving. The impact force was therefore resolved in the 3 moving axes of the head as shown in Figures 44-48. Also shown in these figures is the linear acceleration vector (3 components) of the head calculated at the head CG (center of gravity.)

The next set of figures, 49 through 53, shown the condyles reactions which consist of 3 force and 3 moment components.

Note that the I-S force component and the moment about the R-L axis are the highest, indicating that it may be reasonable in the design of a realistic helmet device, it may be sufficient to monitor the flexion - extension torque at the bottom of the dummy head along with the load at that same point in the S-I direction.

Dependent Variable: MAS

Independent Variable: X = CIR

Model: MAS = [-0.128114E+02] + [0.294774E+02]*X

K	Dependent Variable (MAS) Act	Est	Act-Est	Independent Variable (X)	Subj
1	0.402500E+01	0.396121E+01	0.637875E-01	0.569000E+00	1
2	0.415200E+01	0.434442E+01	-0.192422E+00	0.582000E+00	2
3	0.482100E+01	0.460971E+01	0.211288E+00	0.591000E+00	3
4	0.335800E+01	0.331271E+01	0.452852E-01	0.547000E+00	4
5	0.410500E+01	0.422650E+01	-0.121502E+00	0.578000E+00	5

Correlation Coefficient: r = 0.950450E+00
Standard Error: s = 0.184888E+00

Dependent Variable: IUU

Independent Variable: X = (MS/5)*(BR**2 + CR**2)

Where: MS = [-0.128114E+02] + [0.294774E+02]*CIR
BR = B
CR = 1/SQRT(SIN(E)**2/A**2 + COS(E)**2/C**2)
A = LEN/2, B = BRT/2, C = HGT/2, E = -49.58 deg

Model: IUU = [-0.723156E-02] + [0.188201E+01]*X

K	Dependent Variable (IUU) Act	Est	Act-Est	Independent Variable (X)	Subj
1	0.181000E-01	0.181583E-01	-0.583343E-04	0.134909E-01	1
2	0.207000E-01	0.217112E-01	-0.101121E-02	0.153787E-01	2
3	0.251000E-01	0.230940E-01	0.200601E-02	0.161134E-01	3
4	0.133000E-01	0.128644E-01	0.435594E-03	0.106779E-01	4
5	0.197000E-01	0.210742E-01	-0.137422E-02	0.150402E-01	5

Correlation Coefficient: r = 0.950072E+00
Standard Error: s = 0.154145E-02

Figure 41

Dependent Variable: IVV

Independent Variable: X = (MS/5)*(AR**2 + CR**2)

Where: MS = [-0.128114E+02] + [0.294774E+02]*CIR
AR = 1/SQRT(COS(E)**2/A**2 + SIN(E)**2/C**2)
CR = 1/SQRT(SIN(E)**2/A**2 + COS(E)**2/C**2)
A = LEN/2, B = BRT/2, C = HGT/2, E = -49.58 deg

Model: IVV = [-0.495799E-02] + [0.139723E+01]*X

K	Dependent Variable (IVV) Act	Est	Act-Est	Independent Variable (X)	Subj
1	0.207000E-01	0.203624E-01	0.337619E-03	0.181218E-01	1
2	0.232000E-01	0.251008E-01	-0.190086E-02	0.215131E-01	2
3	0.277000E-01	0.251267E-01	0.257325E-02	0.215317E-01	3
4	0.146000E-01	0.146509E-01	-0.509508E-04	0.140341E-01	4
5	0.231000E-01	0.240581E-01	-0.958137E-03	0.207668E-01	5

Correlation Coefficient: r = 0.936374E+00
Standard Error: s = 0.193817E-02

Dependent Variable: IWW

Independent Variable: X = (MS/5)*(AR**2 + BR**2)

Where: MS = [-0.128114E+02] + [0.294774E+02]*CIR
AR = 1/SQRT(COS(E)**2/A**2 + SIN(E)**2/C**2)
BR = B
A = LEN/2, B = BRT/2, C = HGT/2, E = -49.58 deg

Model: IWW = [-0.524091E-03] + [0.103282E+01]*X

K	Dependent Variable (IWW) Act	Est	Act-Est	Independent Variable (X)	Subj
1	0.144000E-01	0.138360E-01	0.564013E-03	0.139037E-01	1
2	0.141000E-01	0.159075E-01	-0.180748E-02	0.159094E-01	2
3	0.182000E-01	0.163632E-01	0.183678E-02	0.163506E-01	3
4	0.108000E-01	0.108471E-01	-0.471547E-04	0.110099E-01	4
5	0.152000E-01	0.157431E-01	-0.543080E-03	0.157502E-01	5

Correlation Coefficient: r = 0.860866E+00
Standard Error: s = 0.155521E-02

Figure 42

HEAD ANTHROPOMETRY

Test No.	LEN (cm)	BRT (cm)	HGT (cm)	CIR (cm)
79H201	19.8	17.7	24.2	57.8
79H202	19.7	14.7	22.6	56.5
79H203	18.5	15.0	23.3	54.7
79H204	19.5	15.4	22.8	55.8
79H205	21.2	15.4	23.7	59.2

where: LEN = A-P length
BRT = L-R Breadth
HGT = S-I Height
CIR = Circumference.

Figure 43

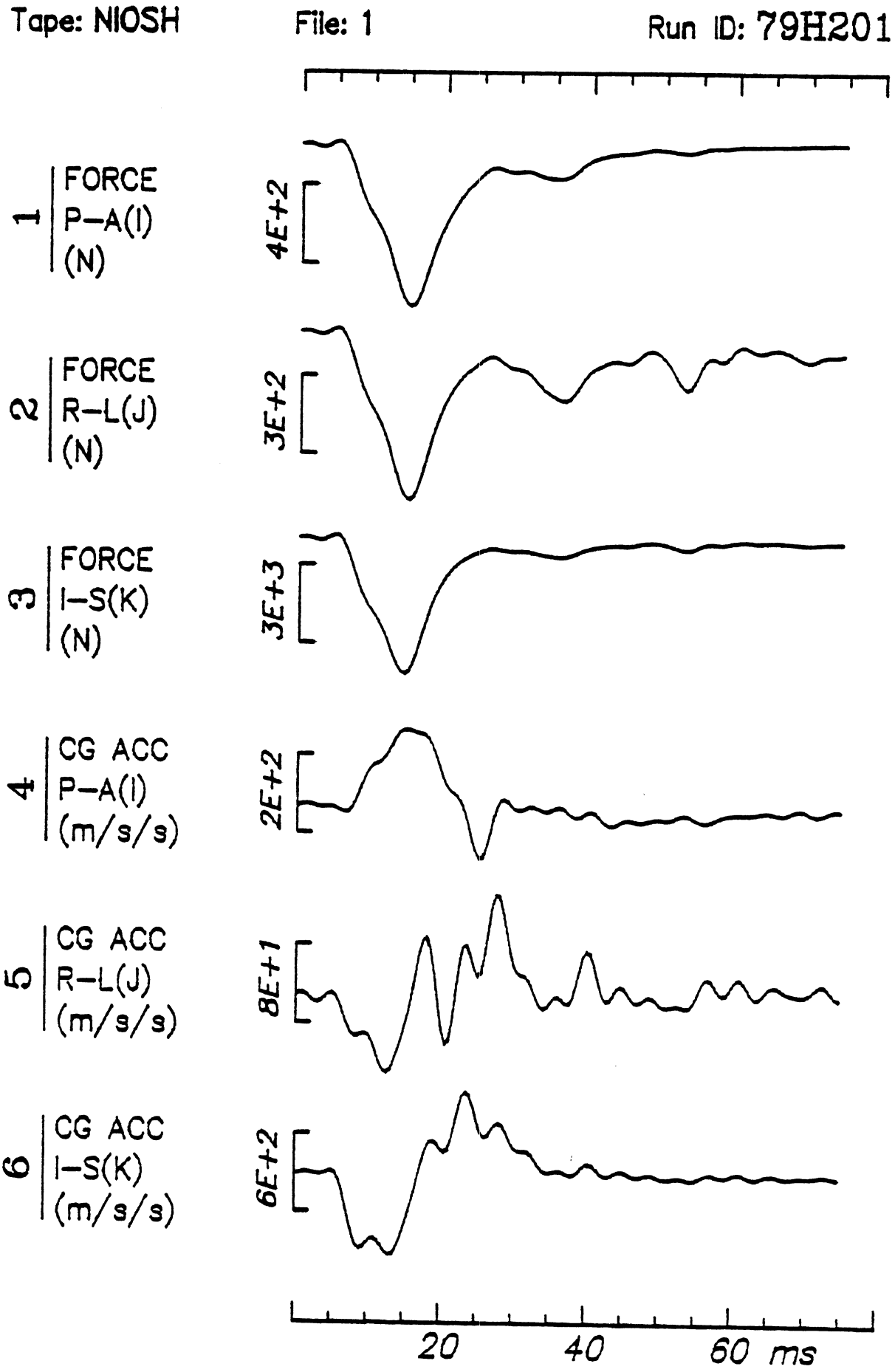

Tape: NIOSH
File: 1
Run ID: 79H201
1
FORCE
P–A(I)
(N)
4E+2
2
FORCE
R–L(J)
(N)
3E+2
3
FORCE
I–S(K)
(N)
3E+3
4
CG ACC
P–A(I)
(m/s/s)
2E+2
5
CG ACC
R–L(J)
(m/s/s)
8E+1
6
CG ACC
I–S(K)
(m/s/s)
6E+2
20
40
60 ms

Figure 44

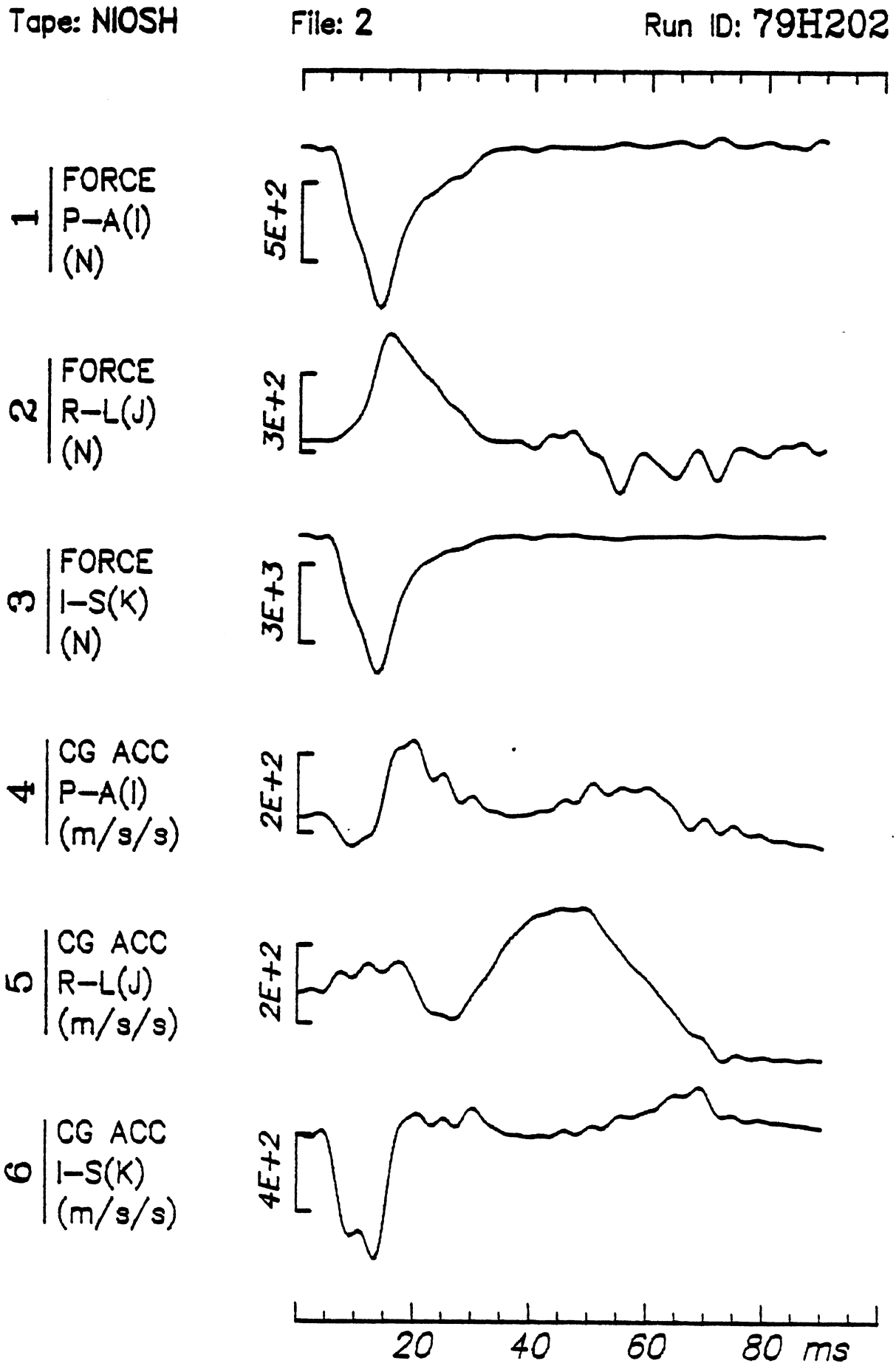

Tape: NIOSH
File: 2
Run ID: 79H202
1
FORCE
P–A(I)
(N)
5E+2
2
FORCE
R–L(J)
(N)
3E+2
3
FORCE
I–S(K)
(N)
3E+3
4
CG ACC
P–A(I)
(m/s/s)
2E+2
5
CG ACC
R–L(J)
(m/s/s)
2E+2
6
CG ACC
I–S(K)
(m/s/s)
4E+2
20
40
60
80 ms

Figure 45

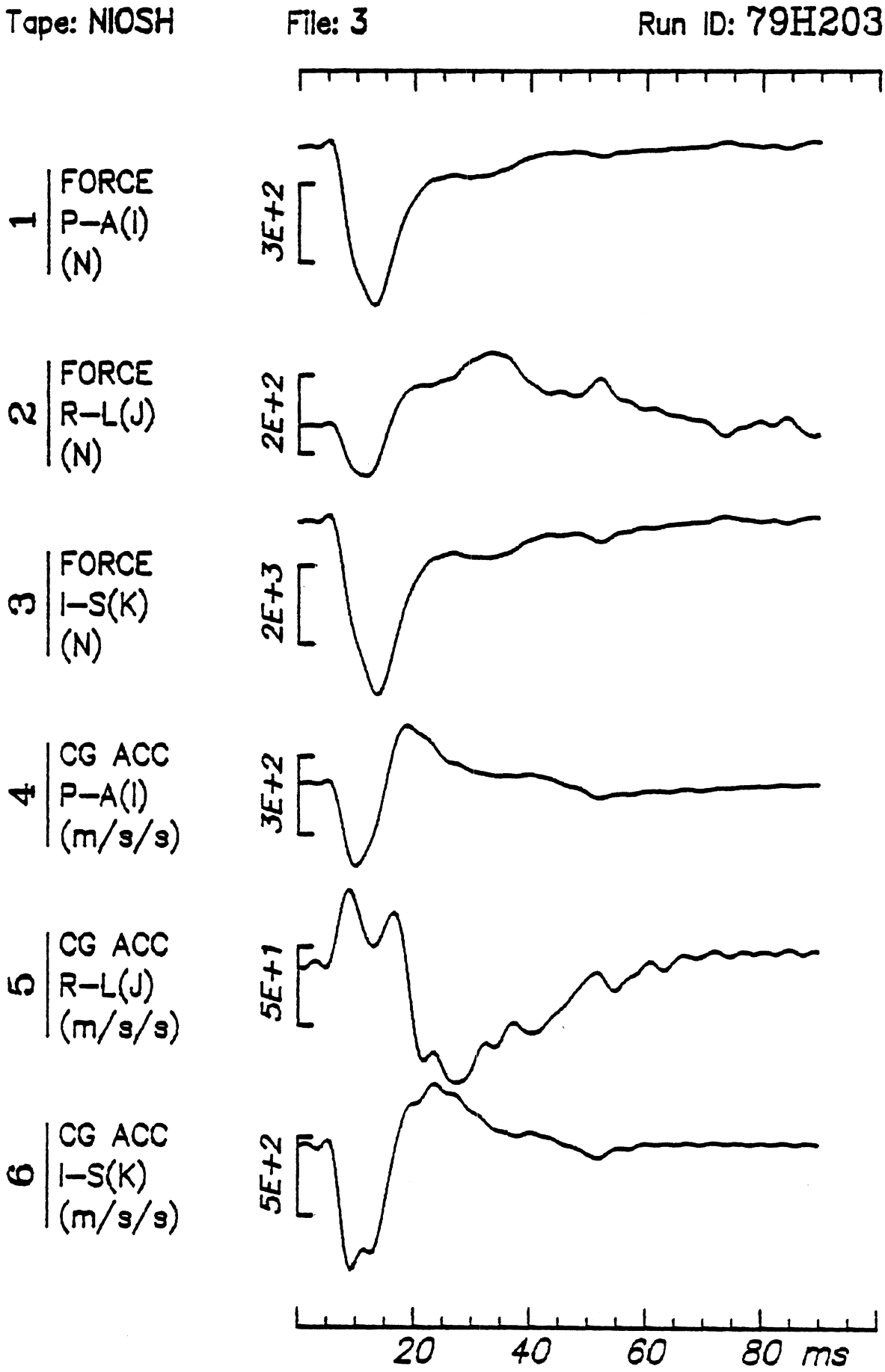

Tape: NIOSH
File: 3
Run ID: 79H203
1
FORCE
P–A(I)
(N)
3E+2
2
FORCE
R–L(J)
(N)
2E+2
3
FORCE
I–S(K)
(N)
2E+3
4
CG ACC
P–A(I)
(m/s/s)
3E+2
5
CG ACC
R–L(J)
(m/s/s)
5E+1
6
CG ACC
I–S(K)
(m/s/s)
5E+2
20
40
60
80 ms

Figure 46

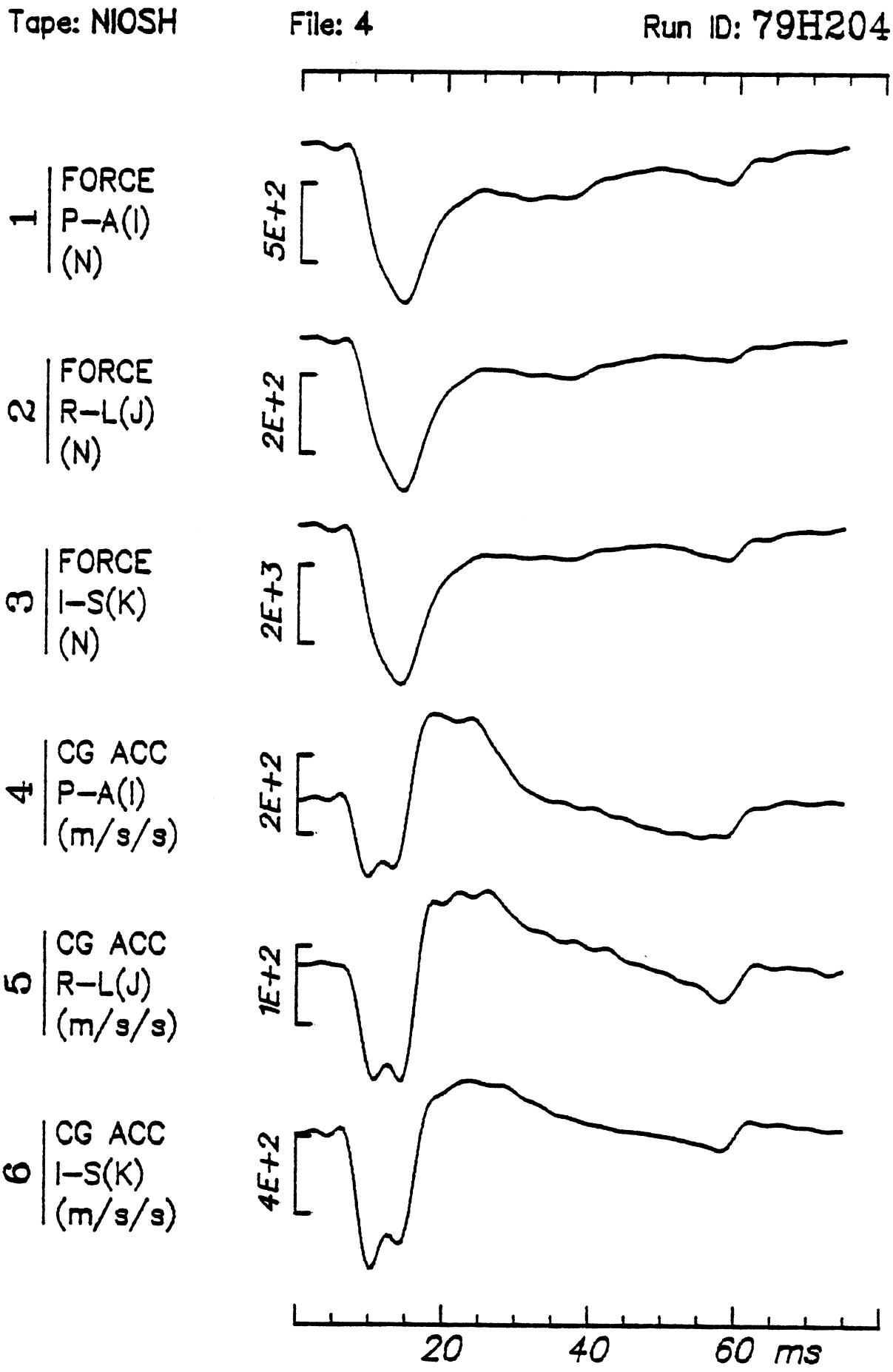

Tape: NIOSH
File: 4
Run ID: 79H204
1
FORCE
P-A(I)
(N)
5E+2
2
FORCE
R-L(J)
(N)
2E+2
3
FORCE
I-S(K)
(N)
2E+3
4
CG ACC
P-A(I)
(m/s/s)
2E+2
5
CG ACC
R-L(J)
(m/s/s)
1E+2
6
CG ACC
I-S(K)
(m/s/s)
4E+2
20
40
60 ms

Figure 47

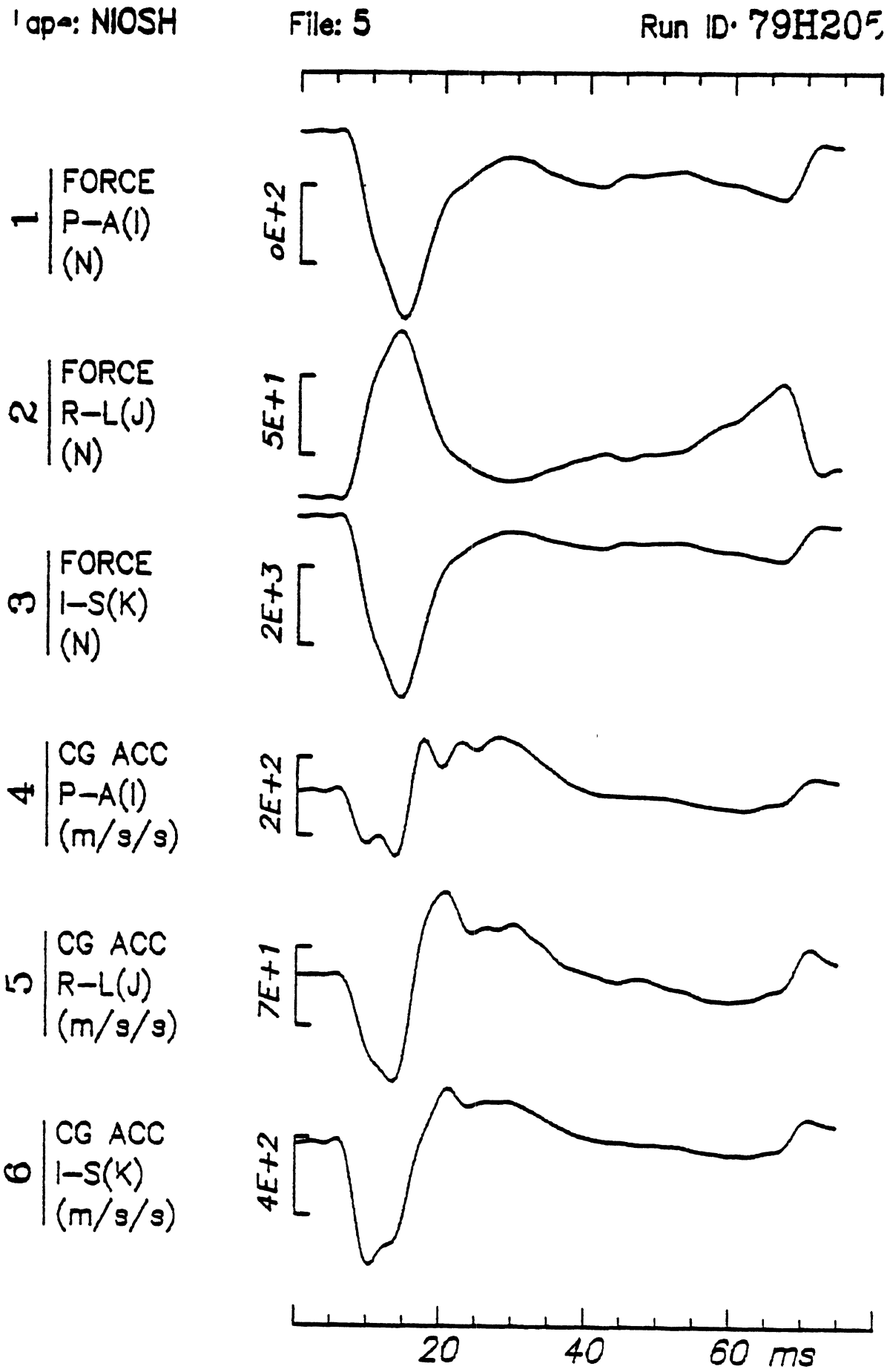

Tape: NIOSH
File: 5
Run ID: 79H205
1 FORCE P-A(I) (N)
oE+2
2 FORCE R-L(J) (N)
5E+1
3 FORCE I-S(K) (N)
2E+3
4 CG ACC P-A(I) (m/s/s)
2E+2
5 CG ACC R-L(J) (m/s/s)
7E+1
6 CG ACC I-S(K) (m/s/s)
4E+2
20
40
60 ms

Figure 48

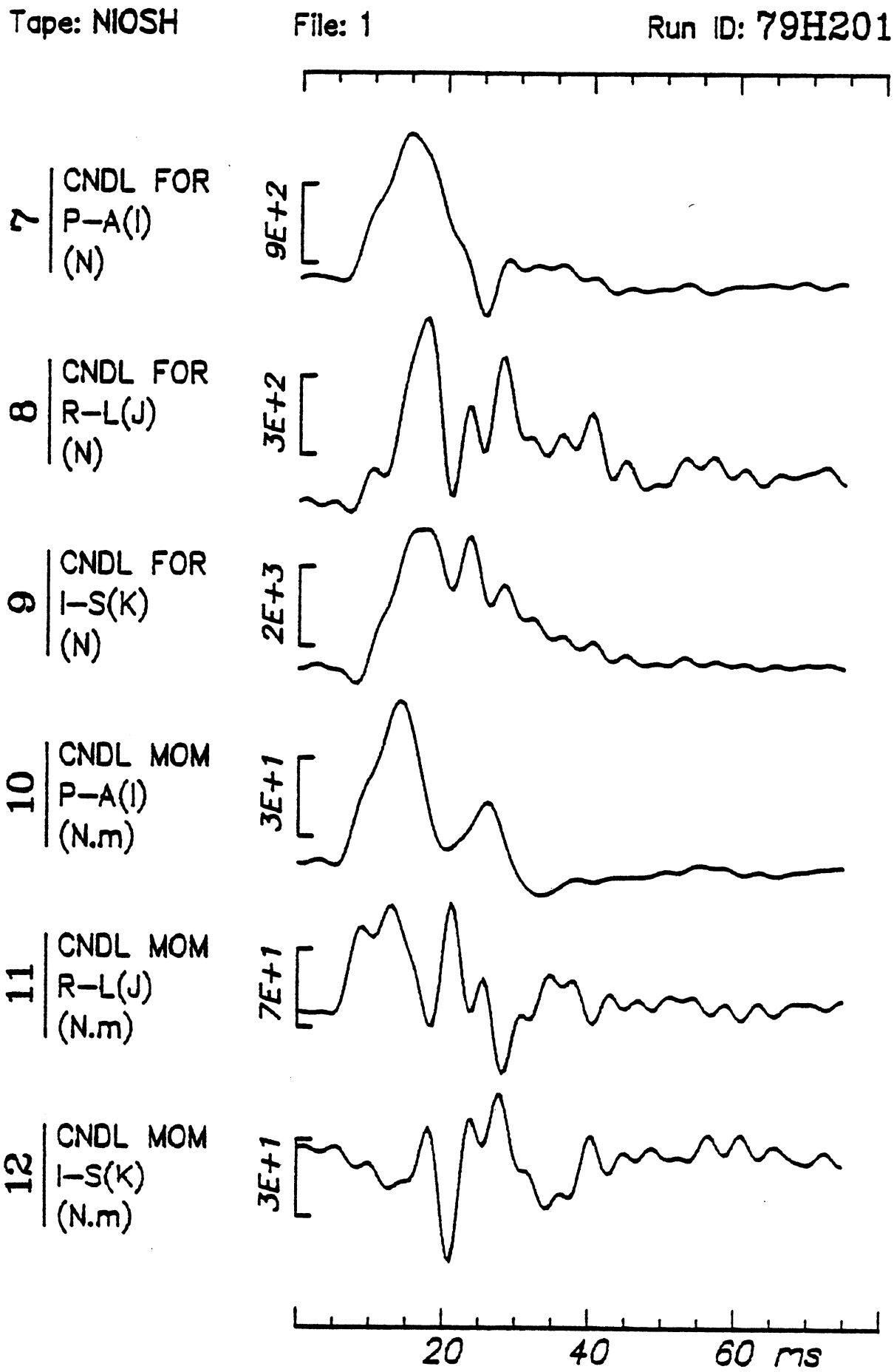

Tape: NIOSH
File: 1
Run ID: 79H201
7
CNDL FOR
P–A(I)
(N)
9E+2
8
CNDL FOR
R–L(J)
(N)
3E+2
9
CNDL FOR
I–S(K)
(N)
2E+3
10
CNDL MOM
P–A(I)
(N.m)
3E+1
11
CNDL MOM
R–L(J)
(N.m)
7E+1
12
CNDL MOM
I–S(K)
(N.m)
3E+1
20
40
60 ms

Figure 49

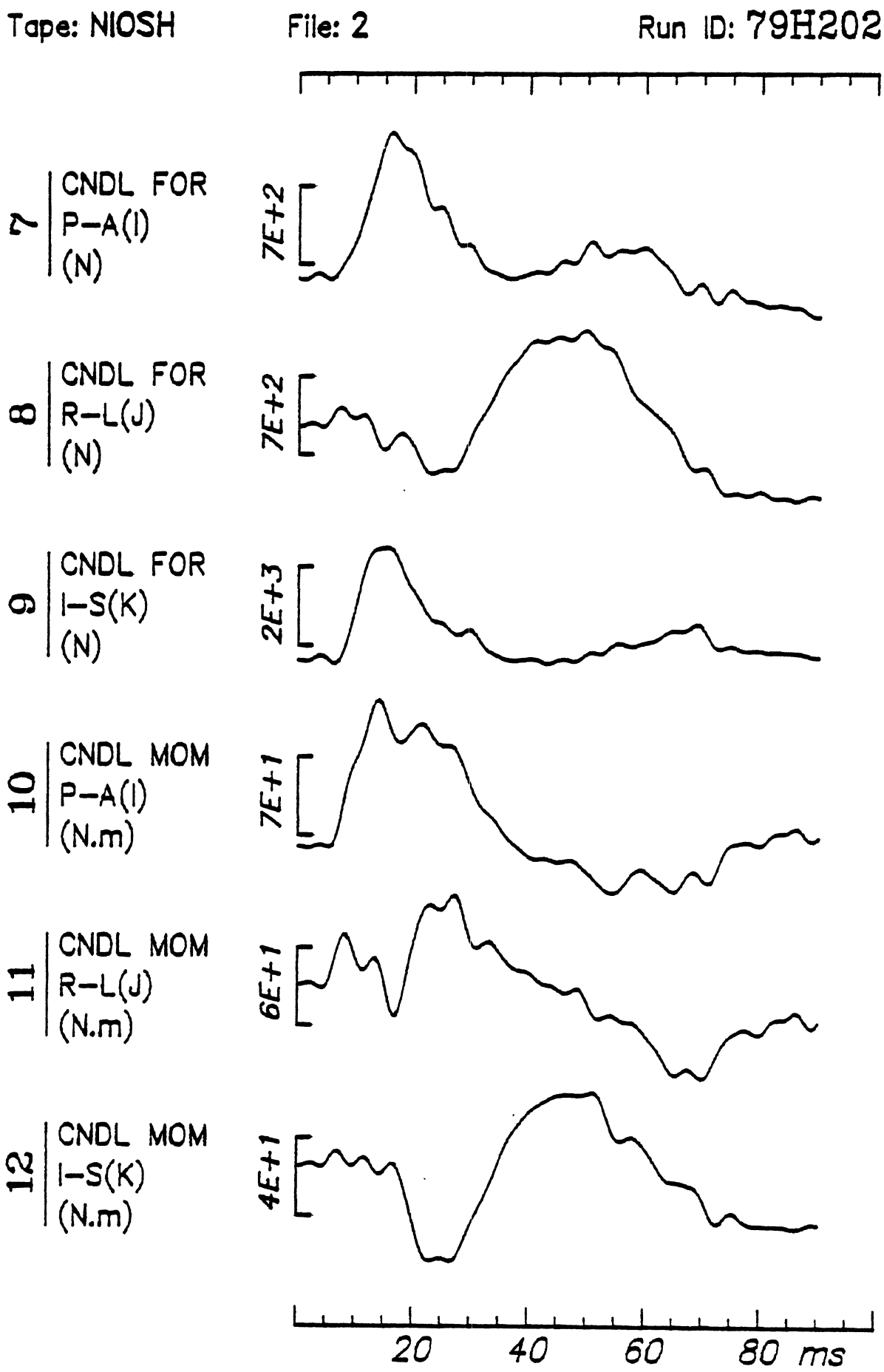
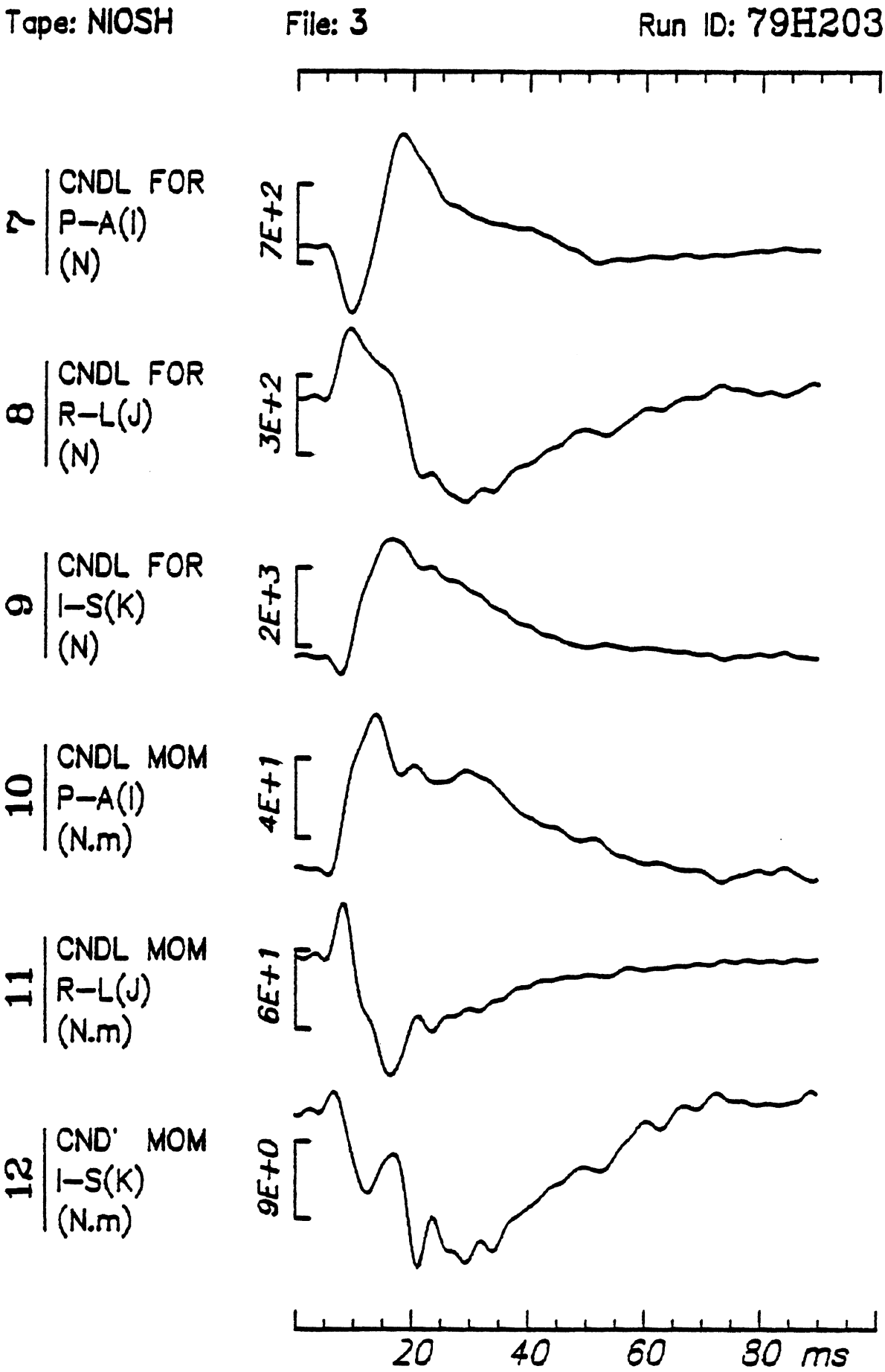

Tape: NIOSH
File: 2
Run ID: 79H202
7
CNDL FOR
P-A(I)
(N)
7E+2
8
CNDL FOR
R-L(J)
(N)
7E+2
9
CNDL FOR
I-S(K)
(N)
2E+3
10
CNDL MOM
P-A(I)
(N.m)
7E+1
11
CNDL MOM
R-L(J)
(N.m)
6E+1
12
CNDL MOM
I-S(K)
(N.m)
4E+1
20
40
60
80 ms

Figure 50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Run ID: 79H203
7
CNDL FOR
P-A(I)
(N)
7E+2
8
CNDL FOR
R-L(J)
(N)
3E+2
9
CNDL FOR
I-S(K)
(N)
2E+3
10
CNDL MOM
P-A(I)
(N.m)
4E+1
11
CNDL MOM
R-L(J)
(N.m)
6E+1
12
CND' MOM
I-S(K)
(N.m)
9E+0
20
40
60
80
ms

Figure 51

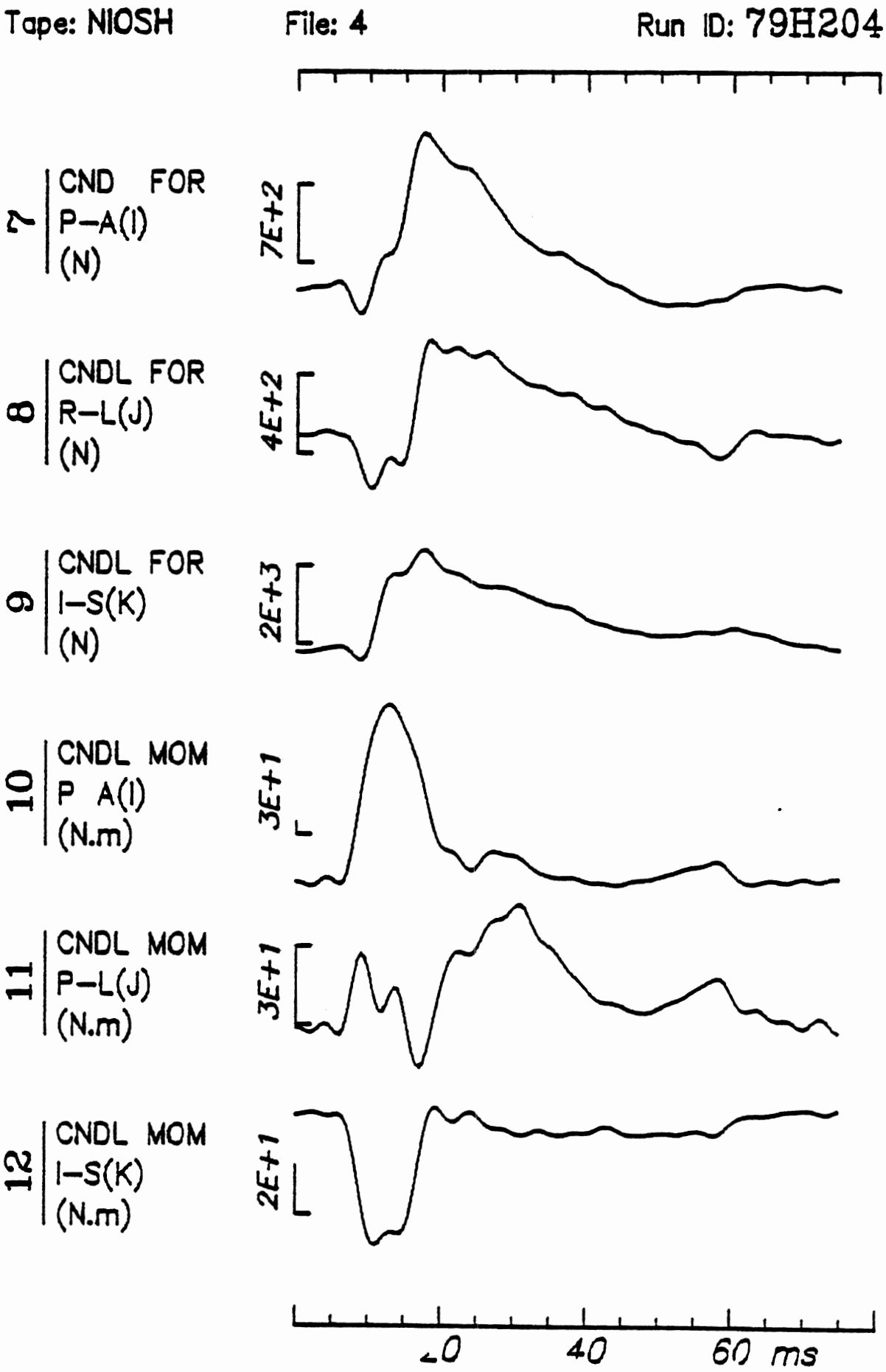

Tape: NIOSH
File: 4
Run ID: 79H204
7
CND FOR
P–A(I)
(N)
7E+2
8
CNDL FOR
R–L(J)
(N)
4E+2
9
CNDL FOR
I–S(K)
(N)
2E+3
10
CNDL MOM
P A(I)
(N.m)
3E+1
11
CNDL MOM
P–L(J)
(N.m)
3E+1
12
CNDL MOM
I–S(K)
(N.m)
2E+1
0
40
60 ms

Figure 52

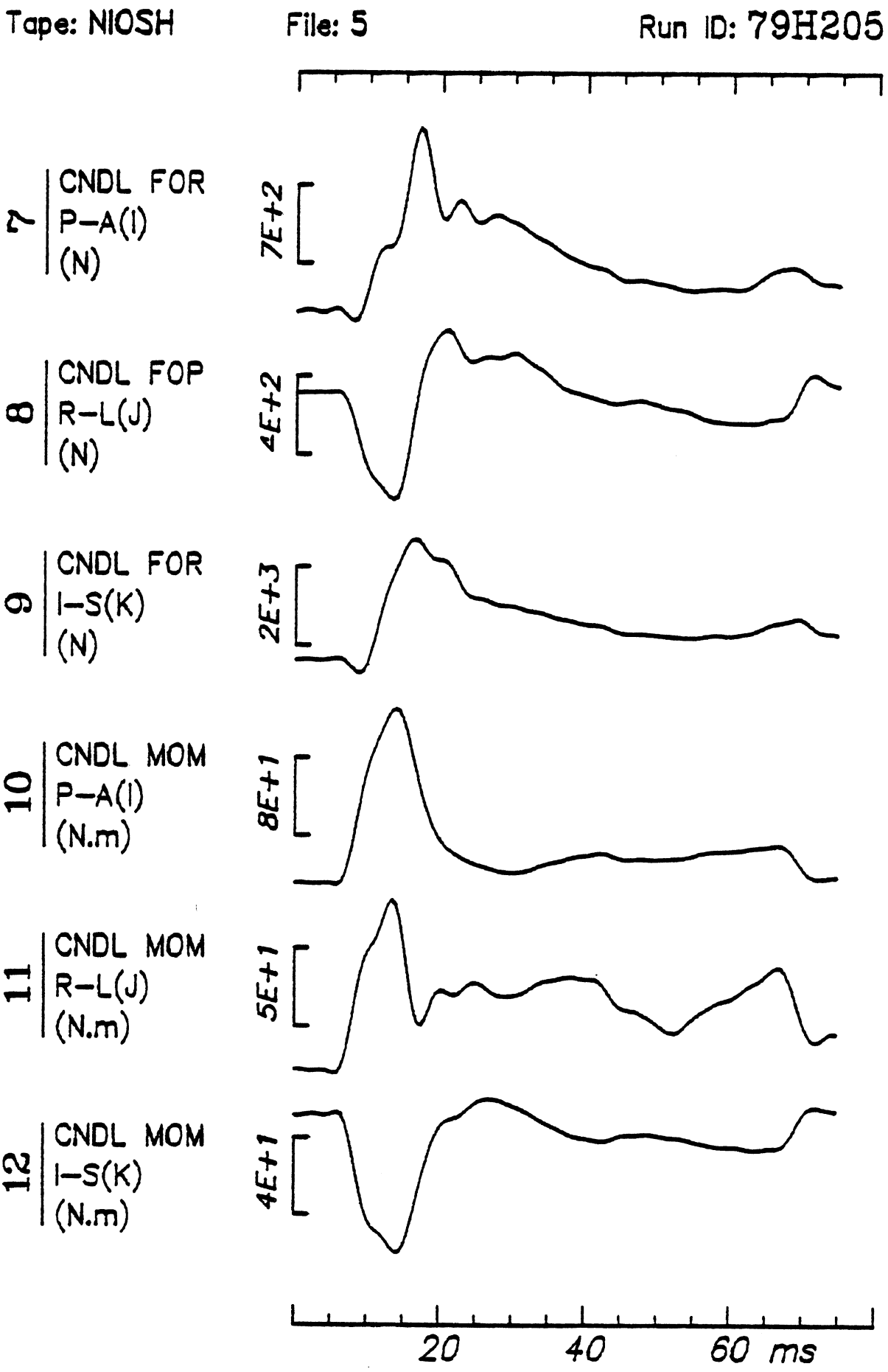

Tape: NIOSH
File: 5
Run ID: 79H205
7
CNDL FOR
P-A(I)
(N)
7E+2
8
CNDL FOP
R-L(J)
(N)
4E+2
9
CNDL FOR
I-S(K)
(N)
2E+3
10
CNDL MOM
P-A(I)
(N.m)
8E+1
11
CNDL MOM
R-L(J)
(N.m)
5E+1
12
CNDL MOM
I-S(K)
(N.m)
4E+1
20
40
60 ms

Figure 53

2.6. CHARACTERIZATION OF THE HUMAN BODY

The portion of the human body effected by S-I impacts to the top of the head include the head itself, the neck and spine and the upper thorax. This portion may be considered as a physical system consisting of many inter-acting elements. Analysis of the inter-actions may render practically impossible the understanding of the biomechanics of this portion of the human physical system.

It is a usual practice in complex system analyses to consider input-output relationships as characterization of such a system. This relationship is called the transfer function of the system and may or may not be independant of time. This transfer function is a process which transform the given input into an output. It is assumed here that this process is stationary or time-invariant.

There are a number of input and output parameters which have been measured during the 5 cadaver tests. Thus, the measured impact force is an input quantity, while acceleration and velocity responses at the head anatomical center, at T1 and at T12 are all output quantities. It is therefore legitimite to characterize the upper portion of the body by transfer functions or processes which transform the impact force into any one of the resulting responses. The usefulness of such characterization is the development of a "black box" model which, given the impact force, would predict the human response to impact,

One such transfer functions is the mechanical impedance, defined as the ratio of "force" over "velocity". Here, "force" and "velocity" are assumed to be the magnitudes of these quantities when the system has reached a steady state under sinusoidal excitation. Mechanical impedance (with a magnitude and phase angle) is usually generated by exciting a given system with a given frequency, then sweeping the frequency over a desired range. At each frequency the magnitude of the steady-state velocity (also sinusoidal) resulting in an impedance which is function of the frequency.

Unorthodox techniques are used in this project to obtain the mechanical impedance of the system as function of frequencies. The method makes the following assumptions:

a) the system is time-invariant;

b) the system is linear, therefore the principle of superposition may be applied

c) the initial conditions of the system are all zero, allowing to assume that the magnitude of response at any given frequency is the result of an excitation of the same frequency.

Armed with these reasonable assumptions, and with the understanding that any irregular function of time (e.g., impact force, acceleration response) may be considered as one period of a periodic function, each of the input and output quantities were transformed to the frequency domain, resulting in a frequency spectrum at discrete frequencies ranging from the fundamental to the Nyquist rate. The fundamental is equal to the inverse of the signal duration, while the Nyquist rate is equal to half of the Sampling rate. However, because of rounding errors of the Fast Fourier Transform (FFT) and since magnitudes of components in the upper frequency range (higher than 100 Hz) are

small approaching the rounding error, output/input ratio are noisy and should not be considered highly reliable.

Now that all signals of interest have been transformed via FFT to the frequency domain, it is possible to characterize the system at each discrete frequency, resulting in an overall impedance curve which is function of frequency. Finally, note that the input to the mechanical system may be at any location and in any direction, and the output also in any different (or same) direction and location.

The following four sections contain transfer impedances between the impact force as input, and the velocity as output measured at three different "locations" (Head, T1 and T12) in 4 different "directions" (resultant, P-A, R-L, I-S), as tabulated below:

Sec. 2.6.1. Resultant impedance curves at the head CG for 5 tests, at T1 for 5 tests and at T12 for 5 tests.

Sec. 2.6.2. P-A impedance curves also at the head CG, at T1 and at T12 each for the 5 tests;

Sec. 2.6.3. R-L impedance curves, similar to the above,

Sec. 2.6.4. I-S impedance curve, also similar to the above.

2.6.1. Resultant's Transfer Impedance Curves

Figures 54-58 give the impedance of the head when the "output" is the "integrated resultant velocity" which differs from the resultant acceleration by a factor equal to the frequency at which the impedance is calculated. Note that, generally, the low-frequency behavior of the head is mass-like. Note also that there is an anti-resonance at 30-40 Hz and a resonance (natural frequency of the head) around 60-80 Hz.

Figures 59-63 give the transfer impedance between the impact point and T1 in the resultant "direction". Note the mass-like behavior at low frequencies, which indicates, as might be expected, a higher mass than the equivalent mass of the head.

Figures 64-68 give the transfer impedance between the impact point and T12, also in the "direction" of the resultant. This time, the low-frequency is consistently mass-like, with equivalent masses definitely higher than both head or T1 equivalent masses.

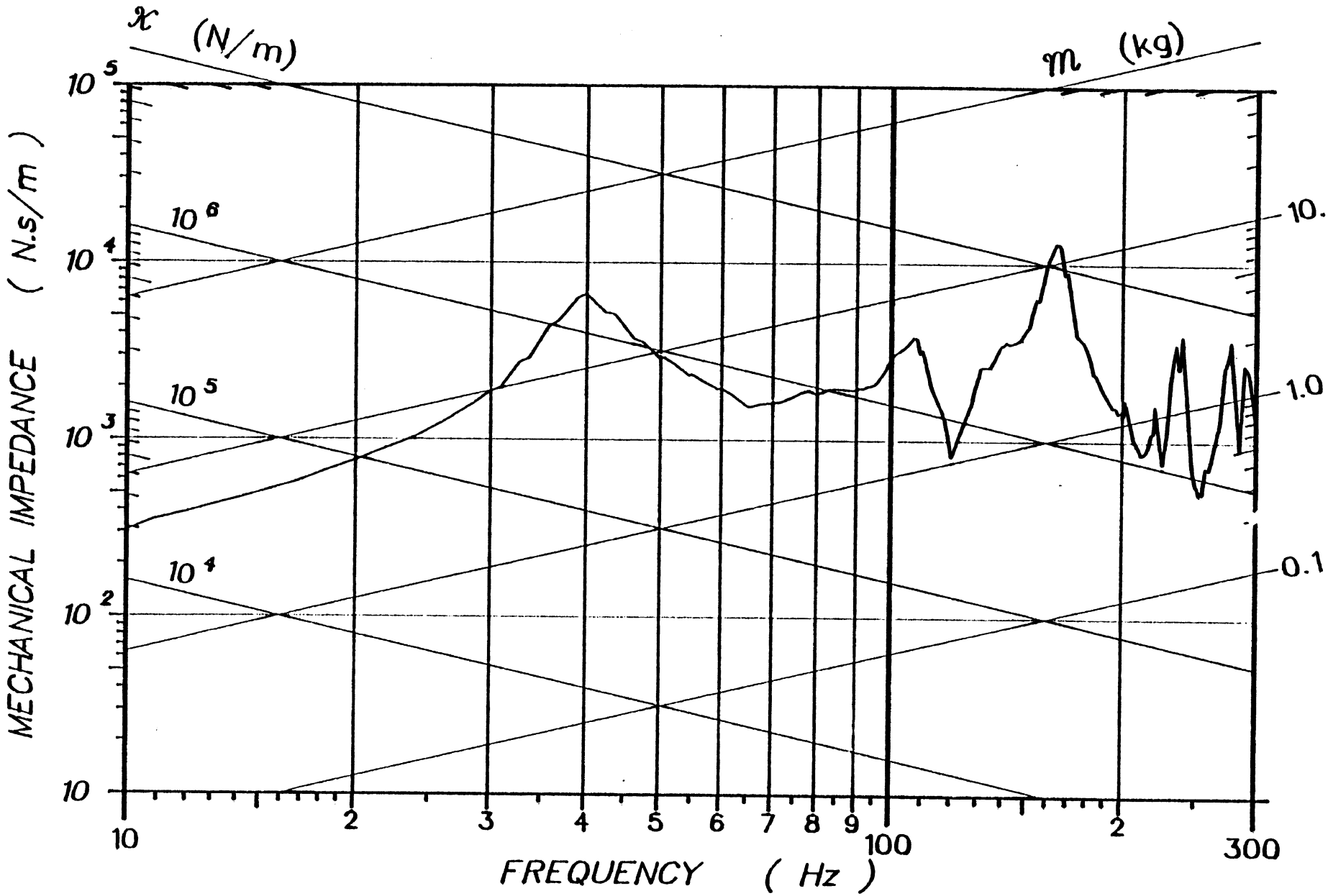


Figure 54: HEAD (RESULTANT)

79H201

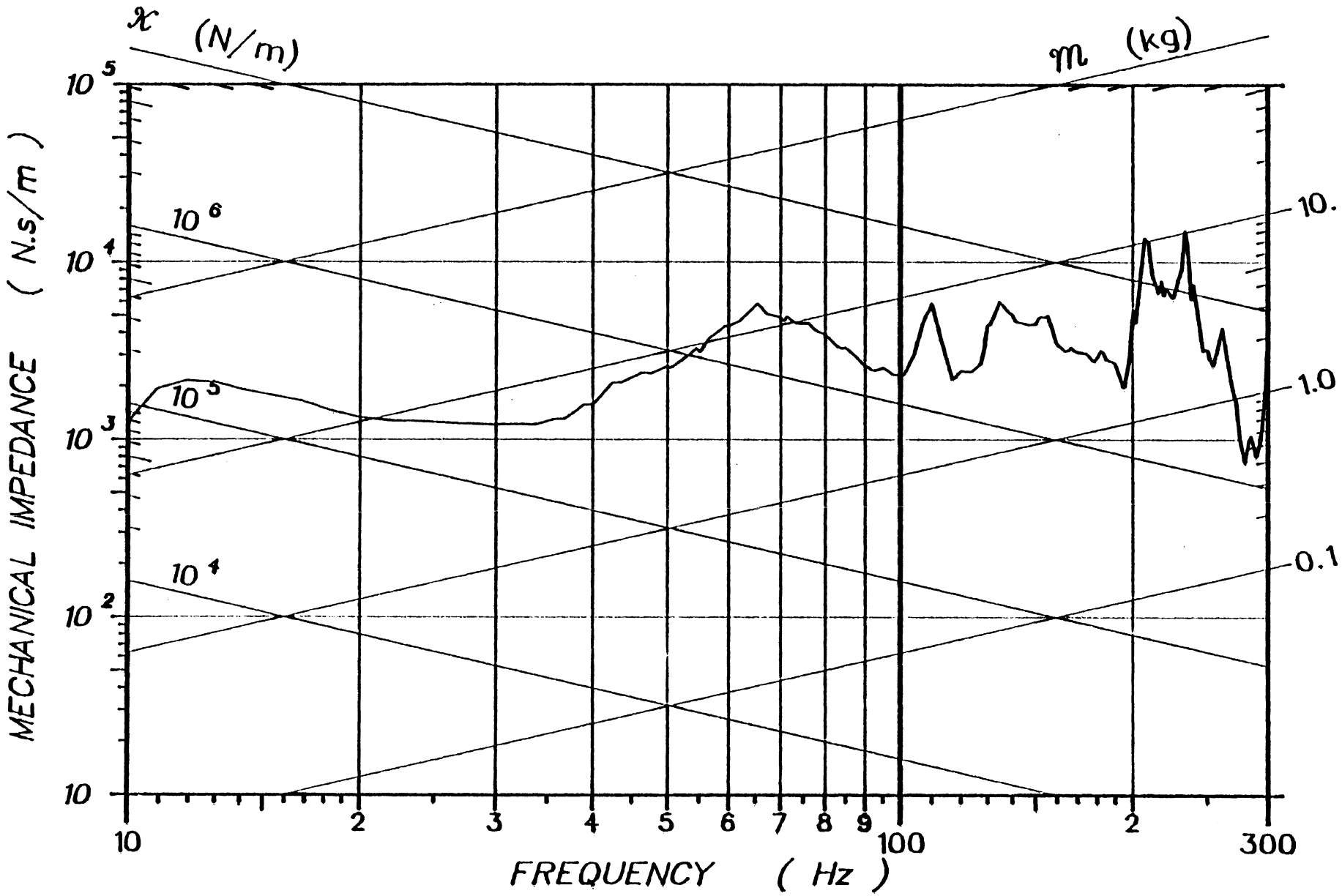


Figure 55: HEAD (RESULTANT)

79H202

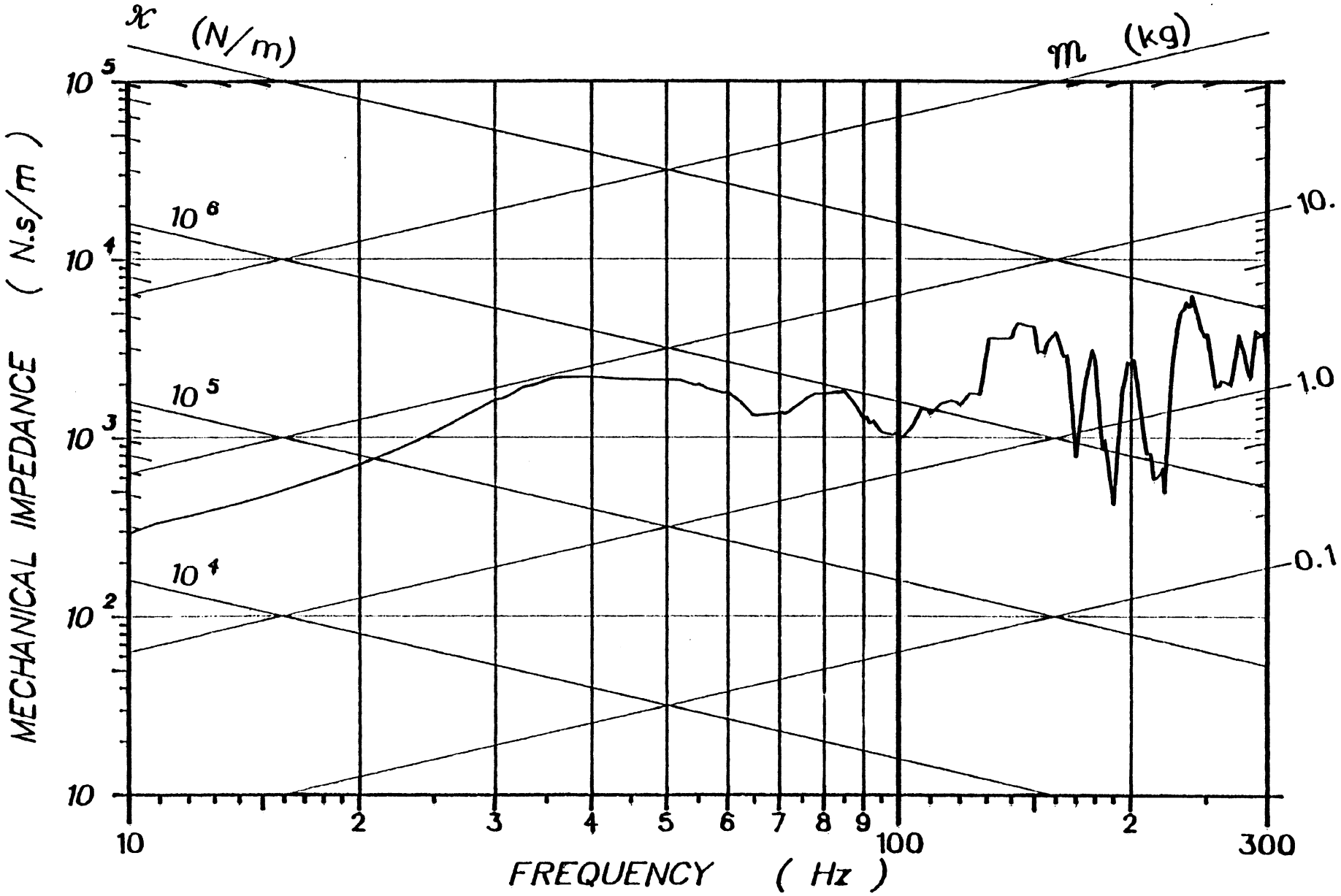


Figure 56: HEAD (RESULTANT)

79H203

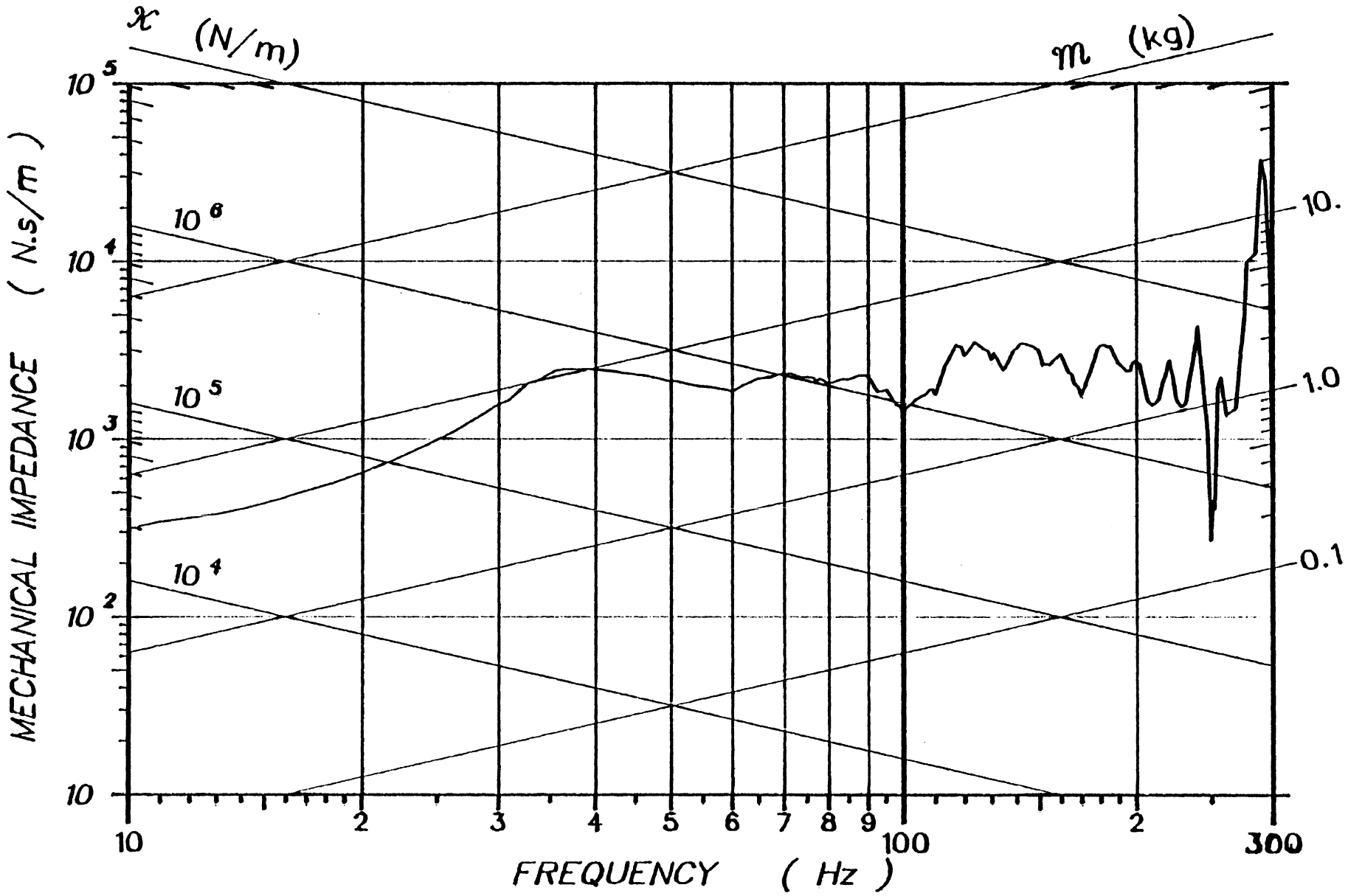


Figure 57: HEAD (RESULTANT)

79H204

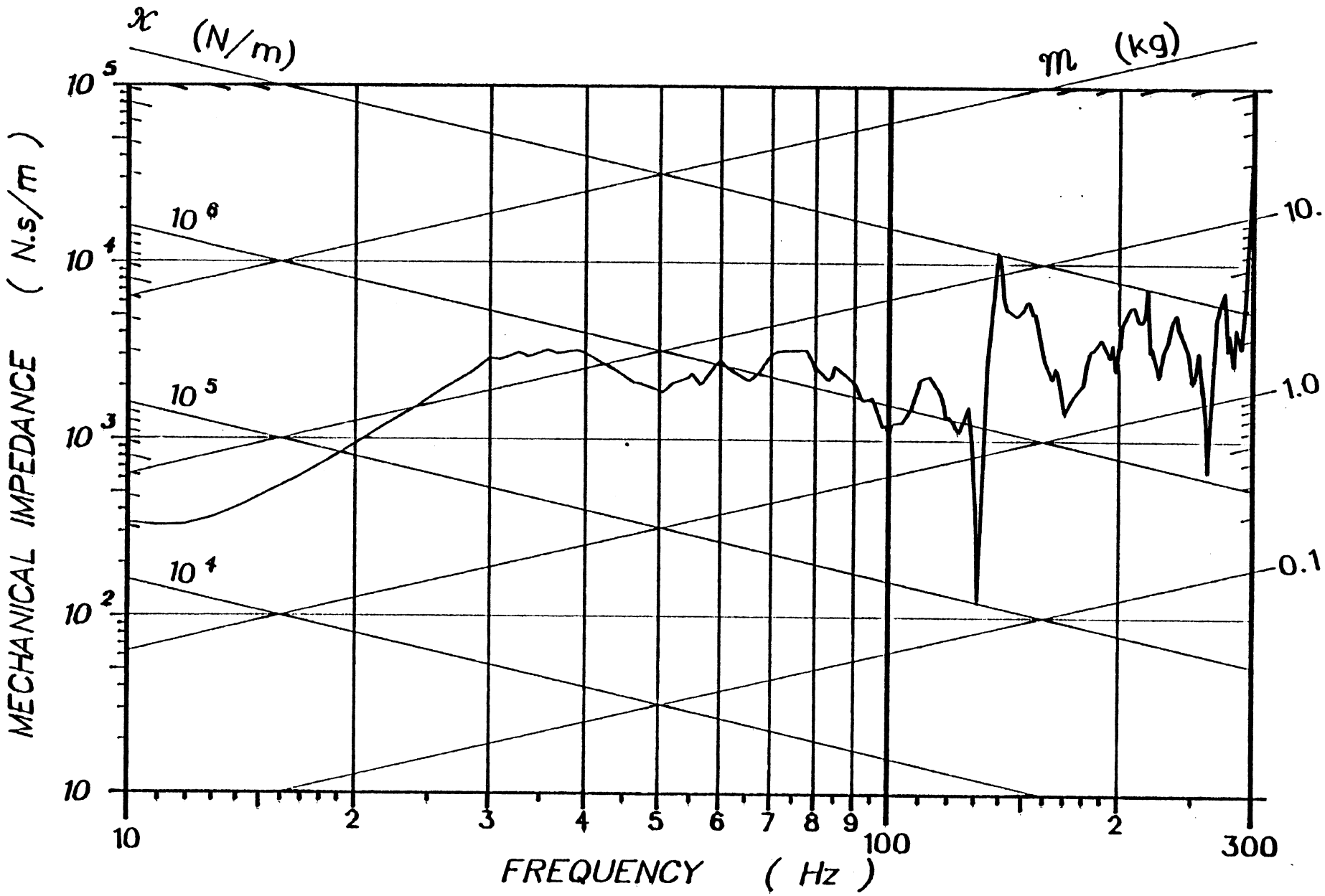


Figure 58: HEAD (RESULTANT)

79H205

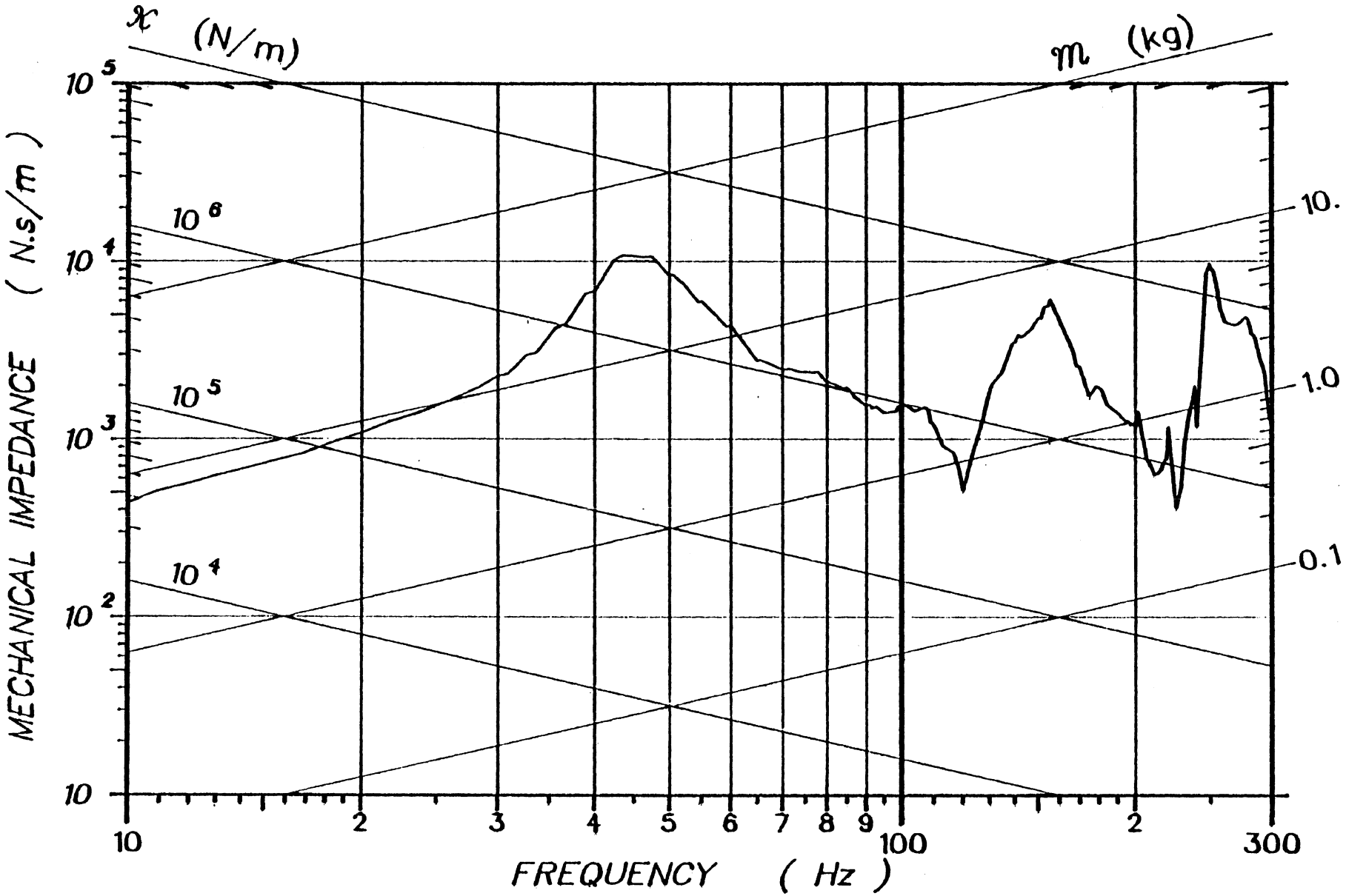


Figure 59: T1 (RESULTANT)

79H201

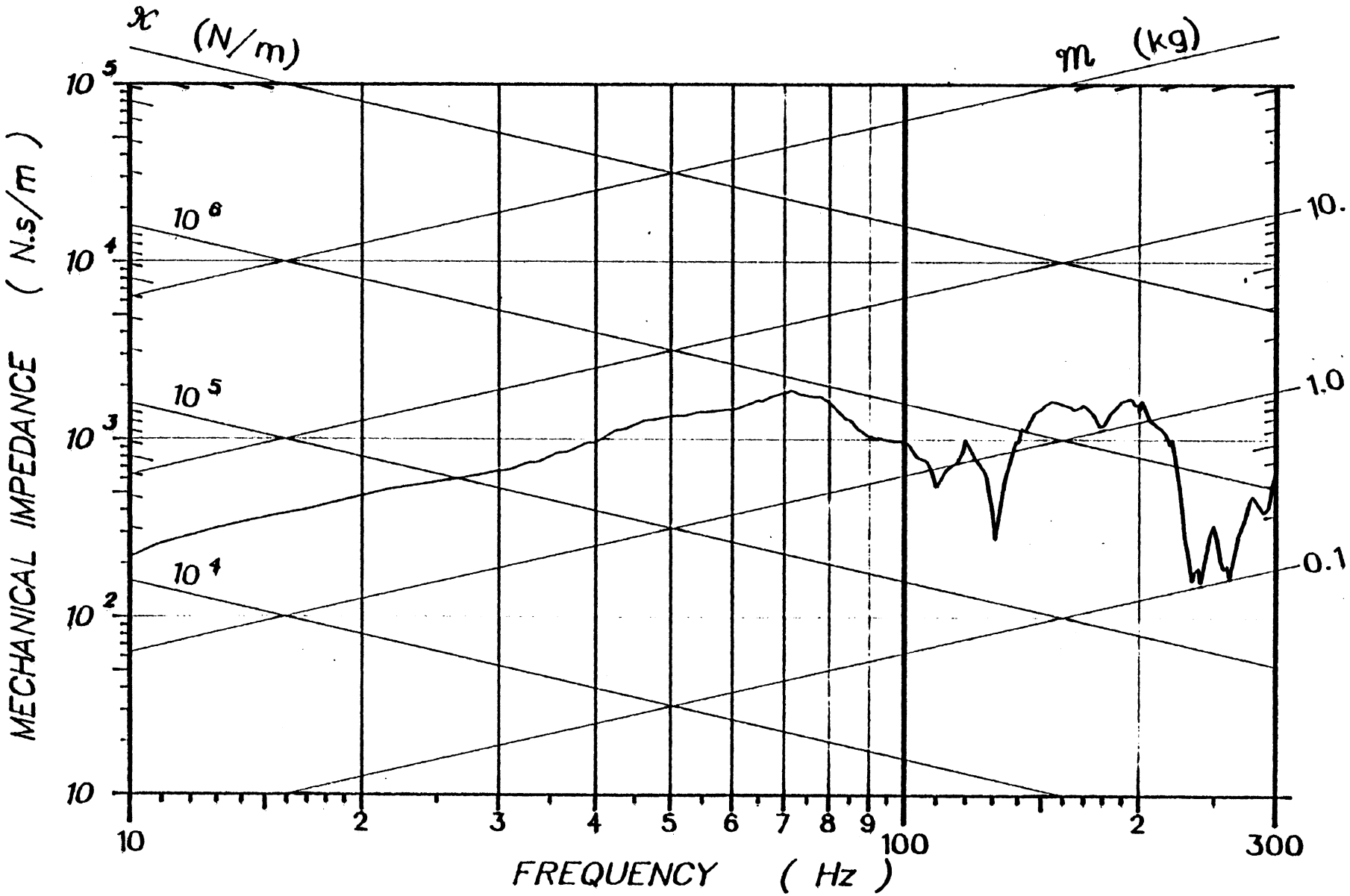


Figure 60: T1 (RESULTANT)

79H202

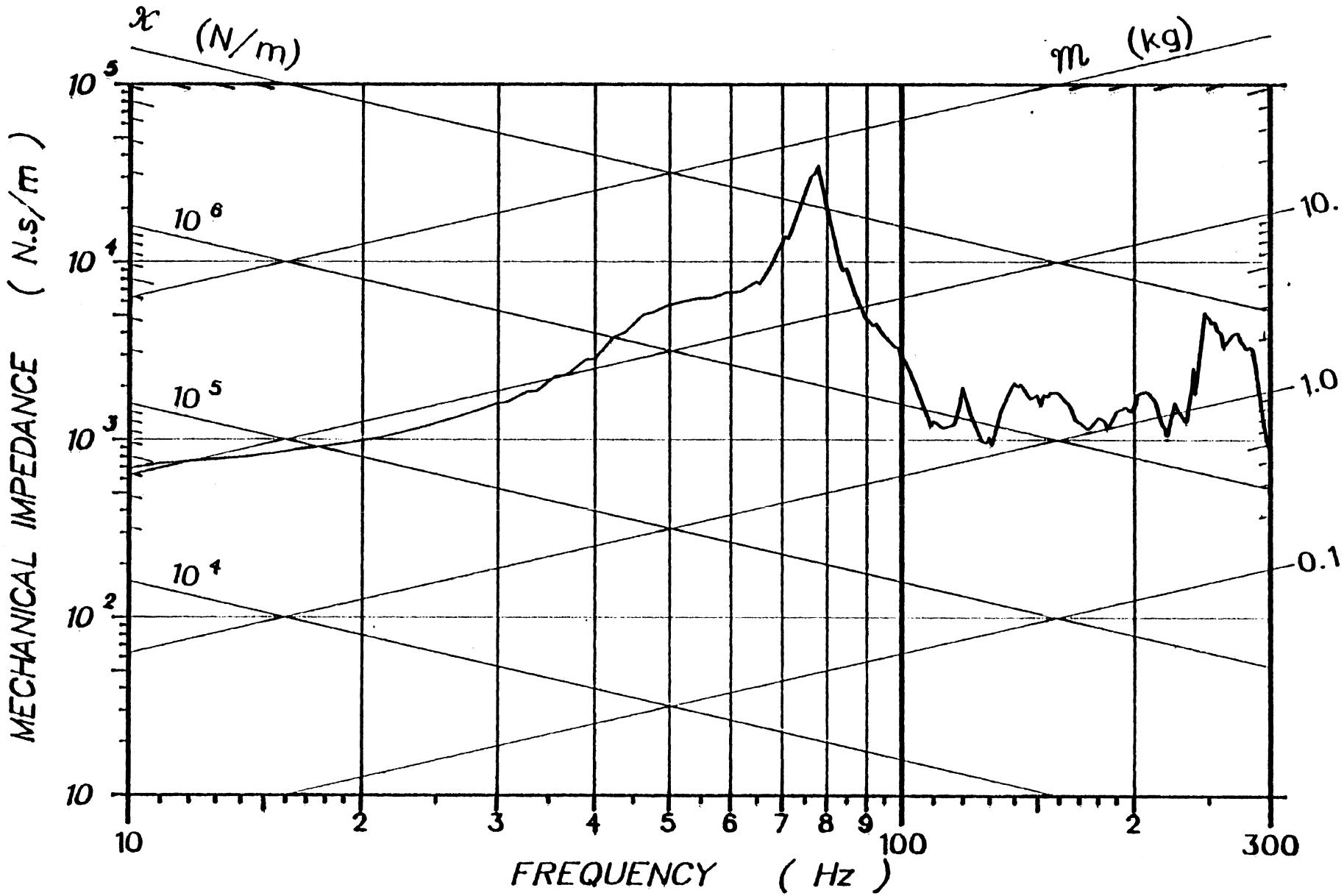


Figure 61: T1 (RESULTANT)

79H203

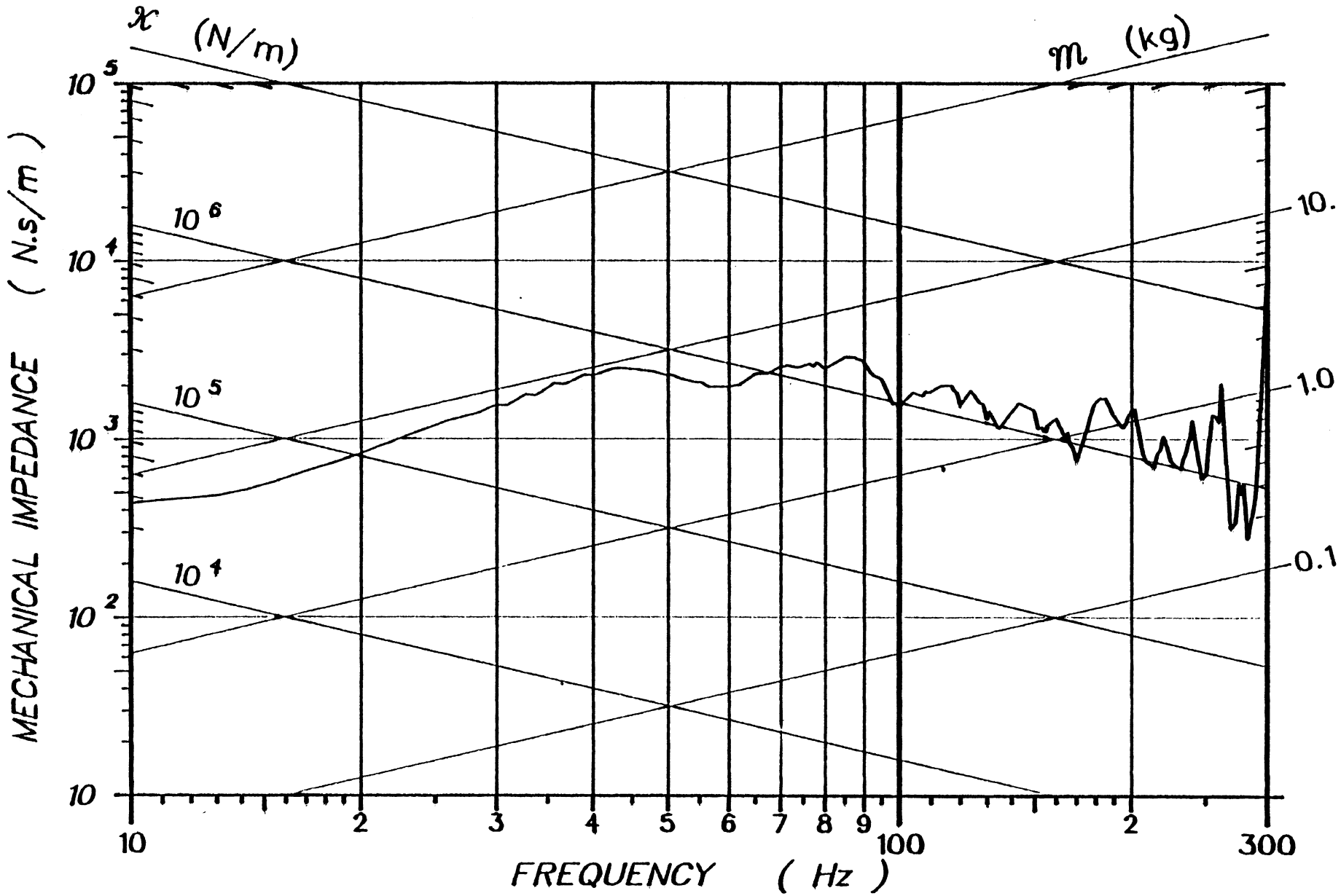


Figure 62: T1 (RESULTANT)

79H204

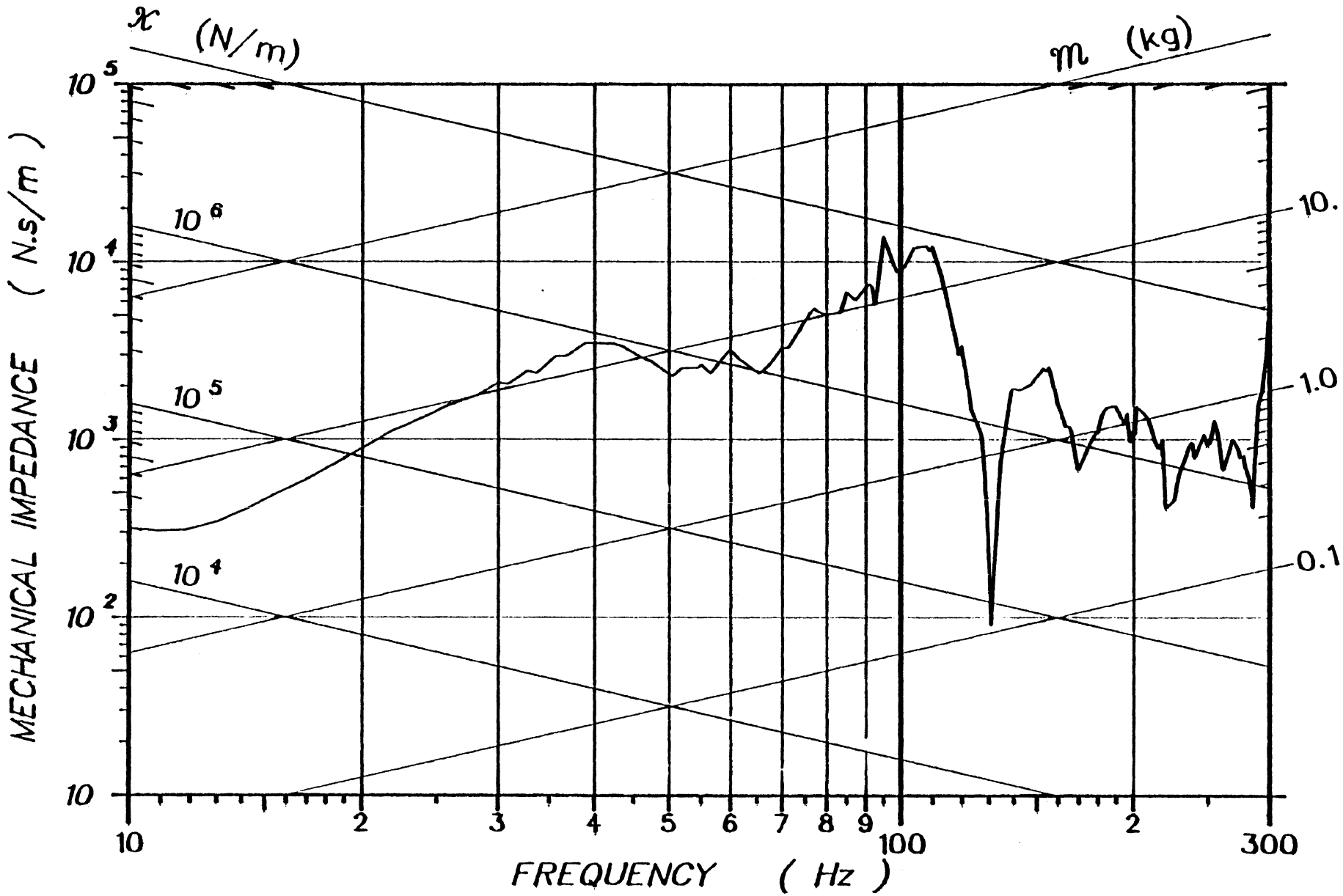


Figure 63: T1 (RESULTANT)

79H205

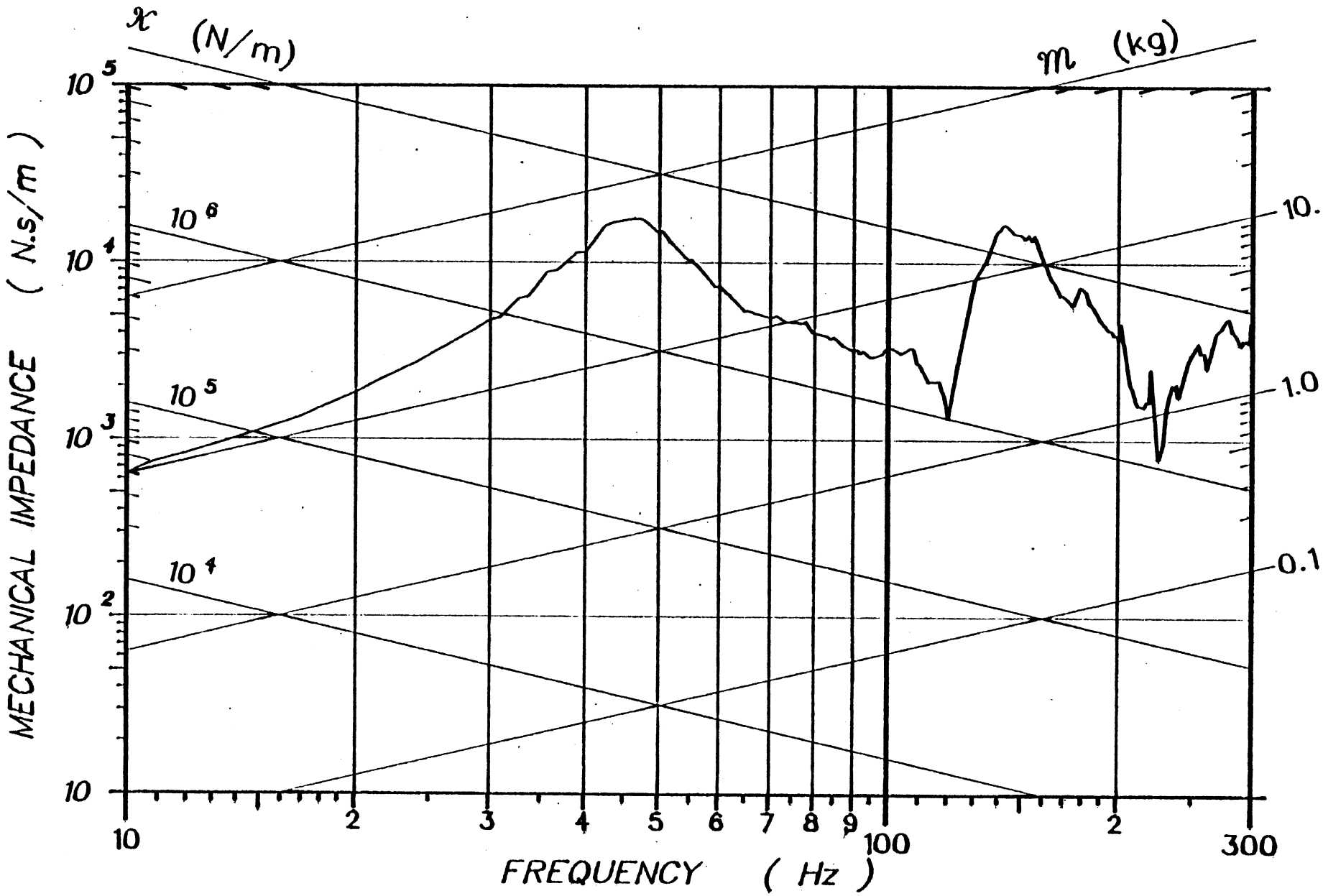


Figure 64: T12 (RESULTANT)

79H201

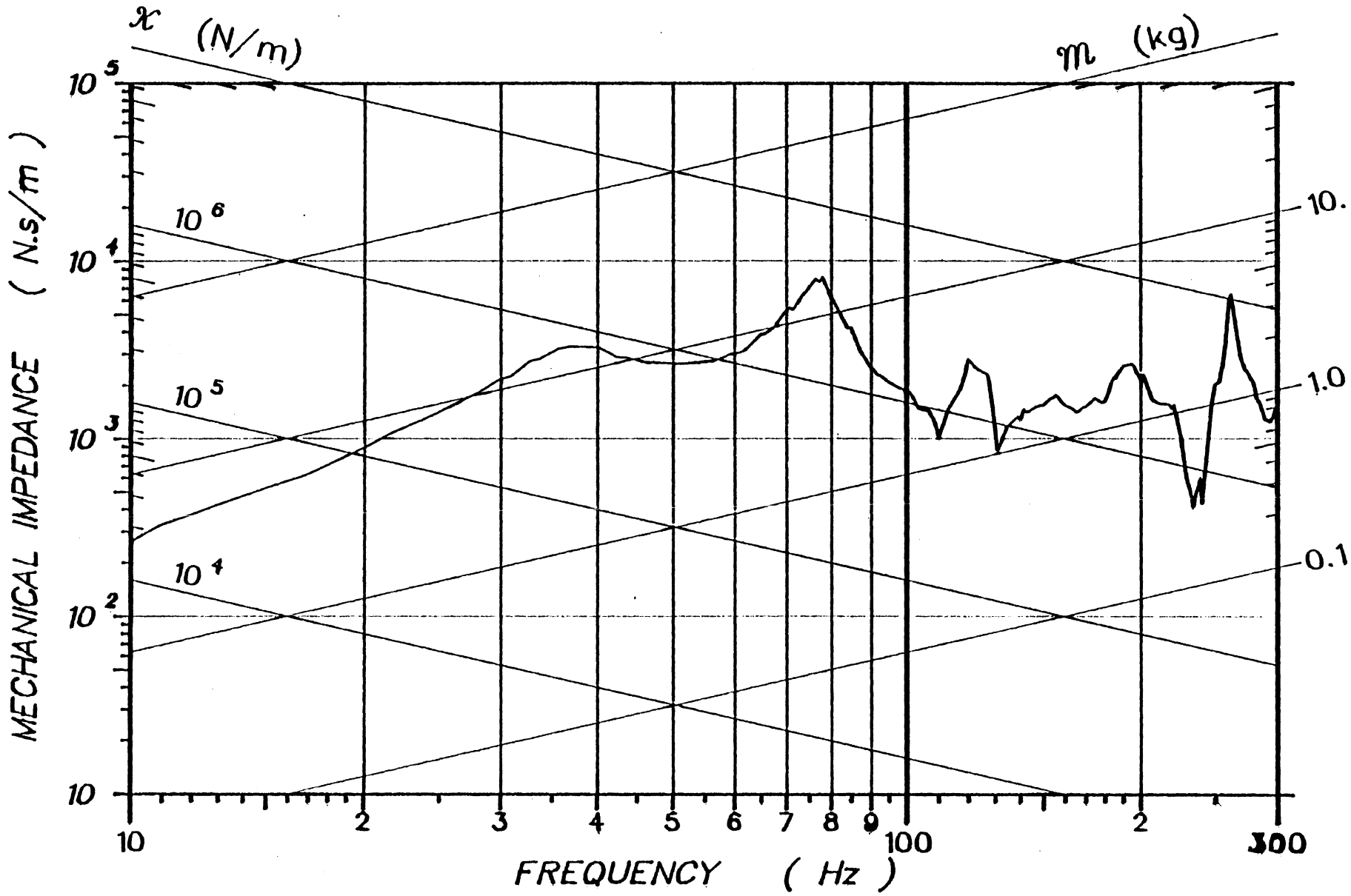


Figure 65: T12 (RESULTANT)

79H202

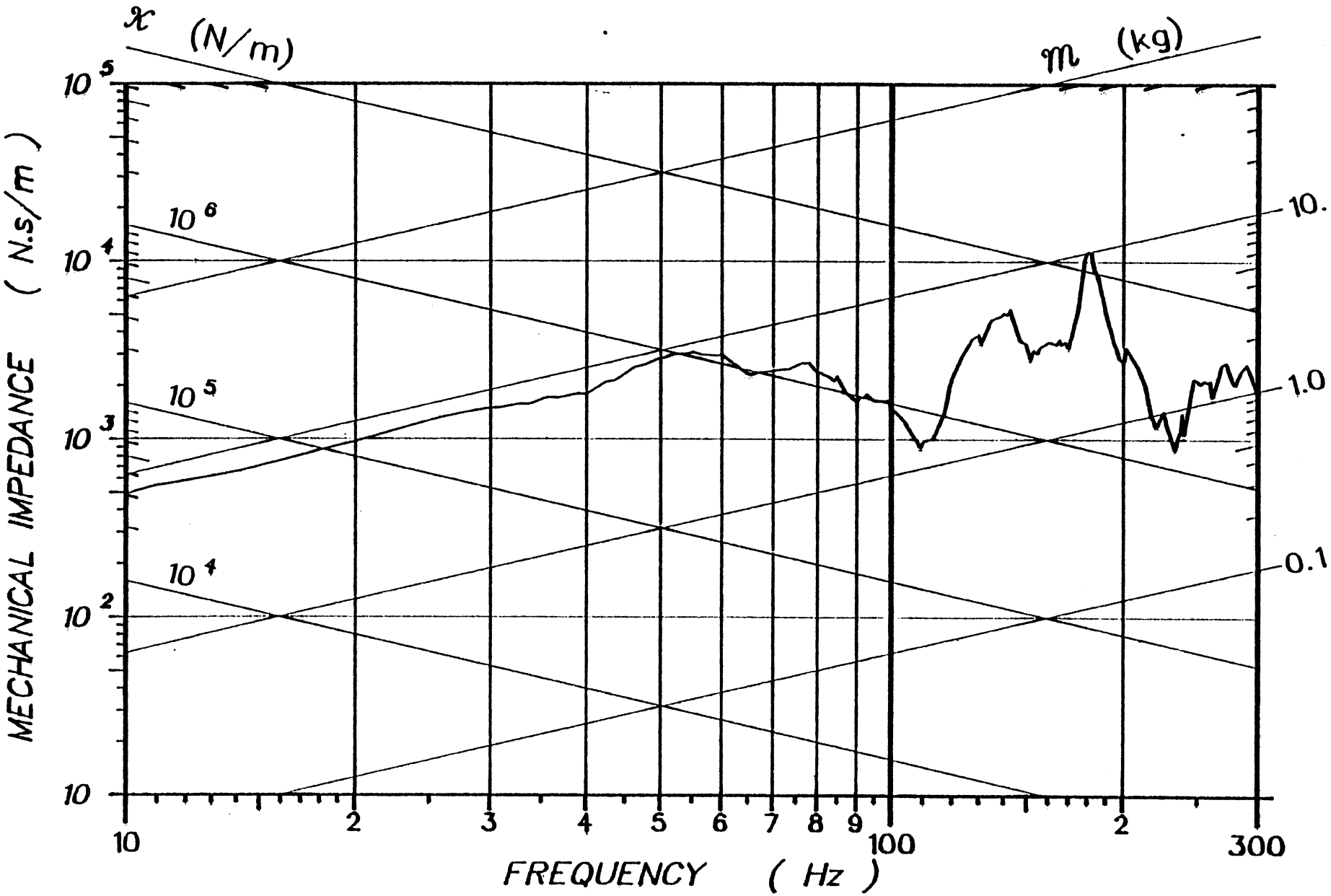


Figure 66: T12 (RESULTANT)

79H203

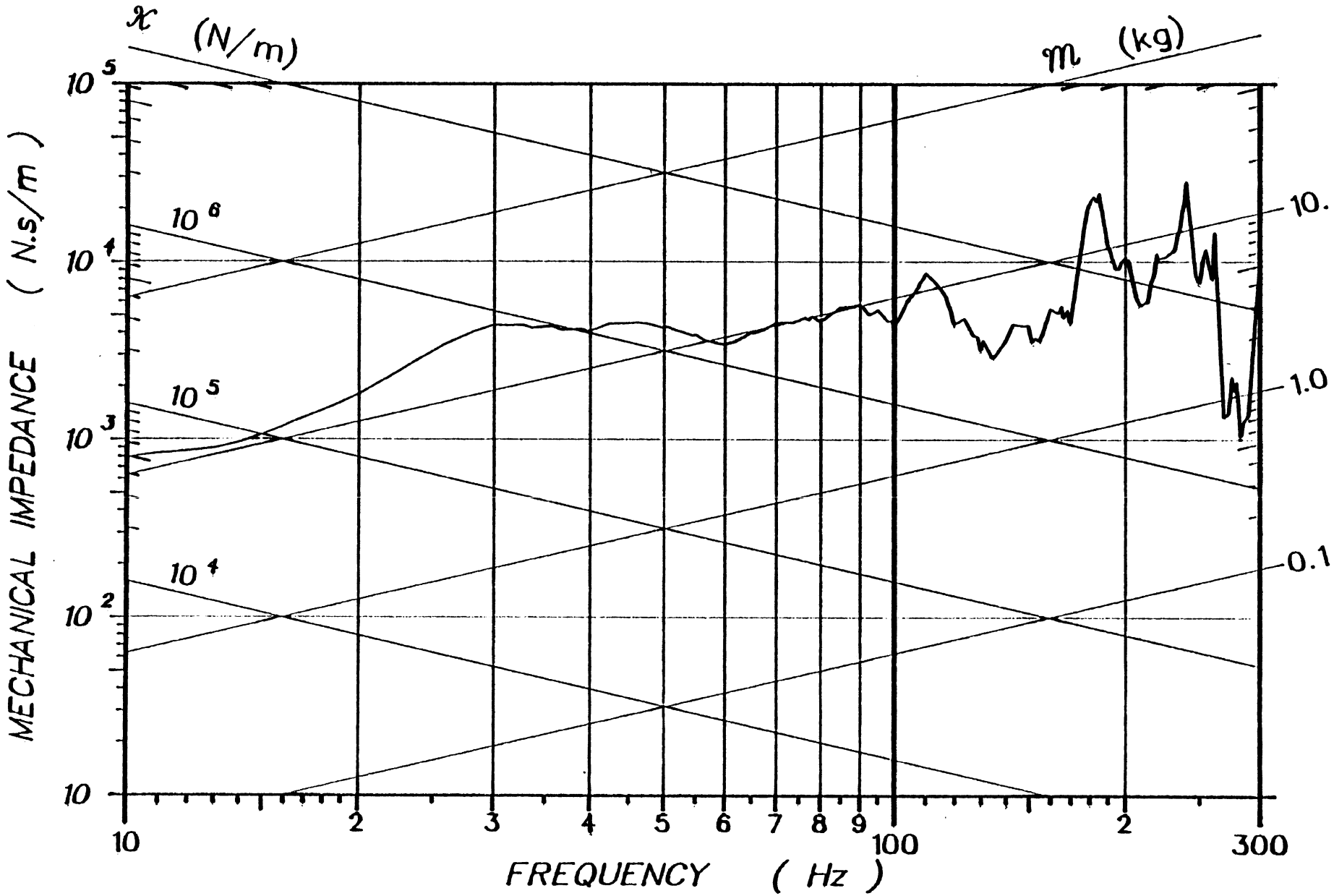


Figure 67: T 12 (RESULTANT)

79H204

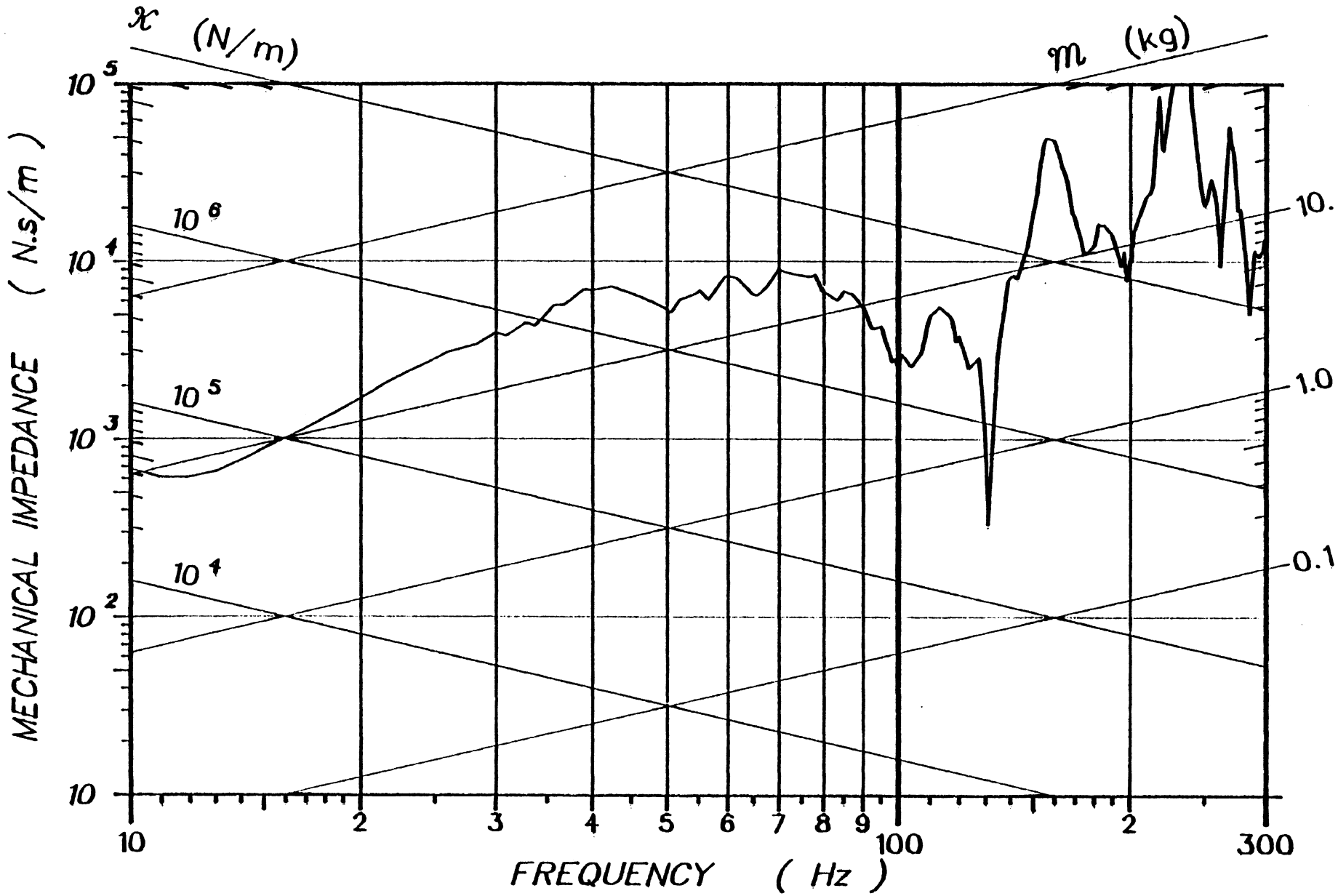


Figure 68: T12 (RESULTANT)

79H205

2.6.2. P-A Transfer Impedance Curves

Figures 69-73 are five transfer impedance curves between the impact point and the head CG in the P-A direction. Note that the impedances are generally higher than those for the resultants, indicating the head "refuses" to move in the P-A direction as much as it moves in the resultant direction. Simply stated, the impedance curves confirm the fact that resultant accelerations are higher than the component in the P-A direction.

Figures 74-78 give the transfer impedances at T1, while figures 79-83 give impedancy at T12. The general trend is that T12 impedances are higher than those at T1, indicating a dissipation of energy as one moves away from the point of impact.

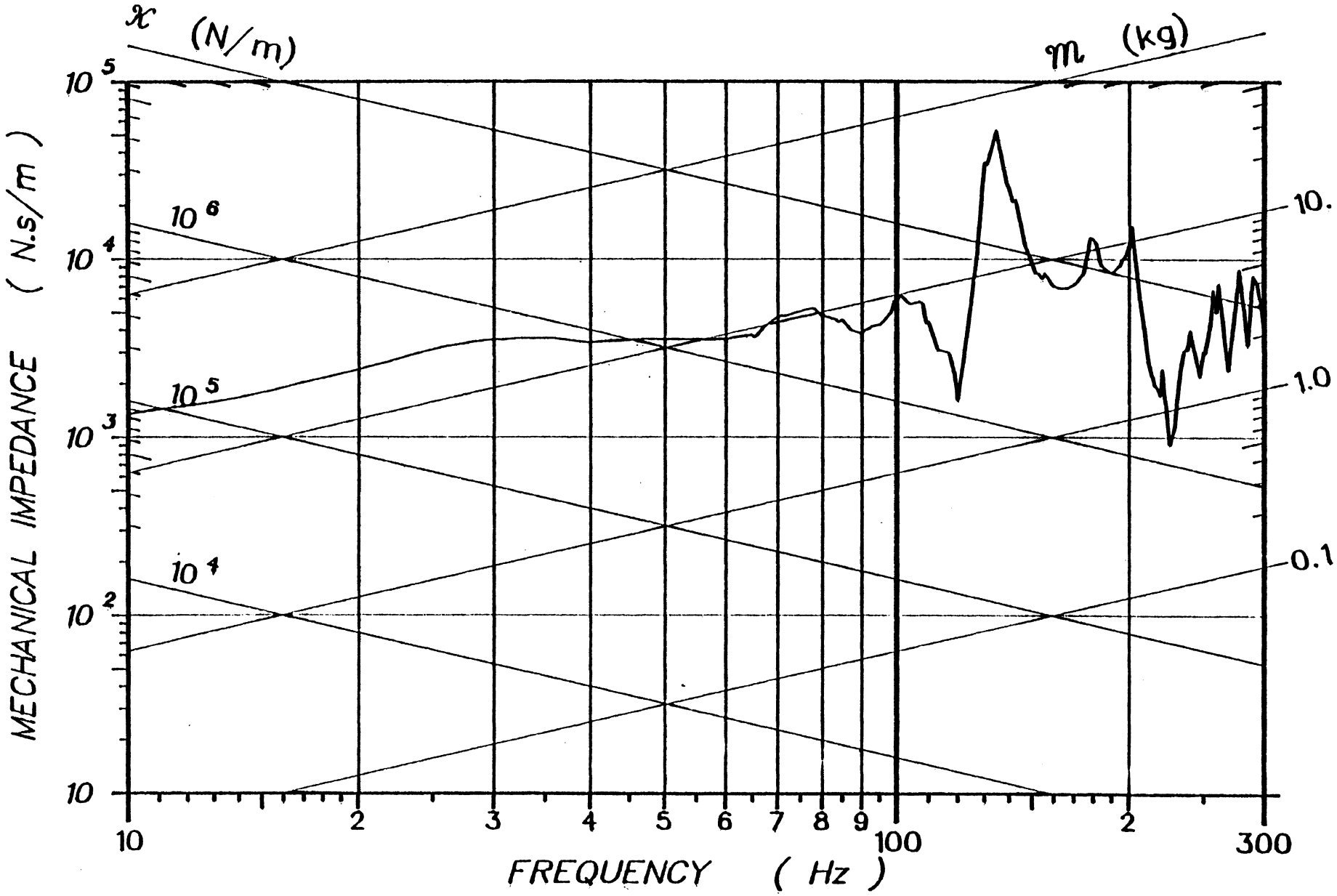


Figure 69: HEAD (P—A)

79H201

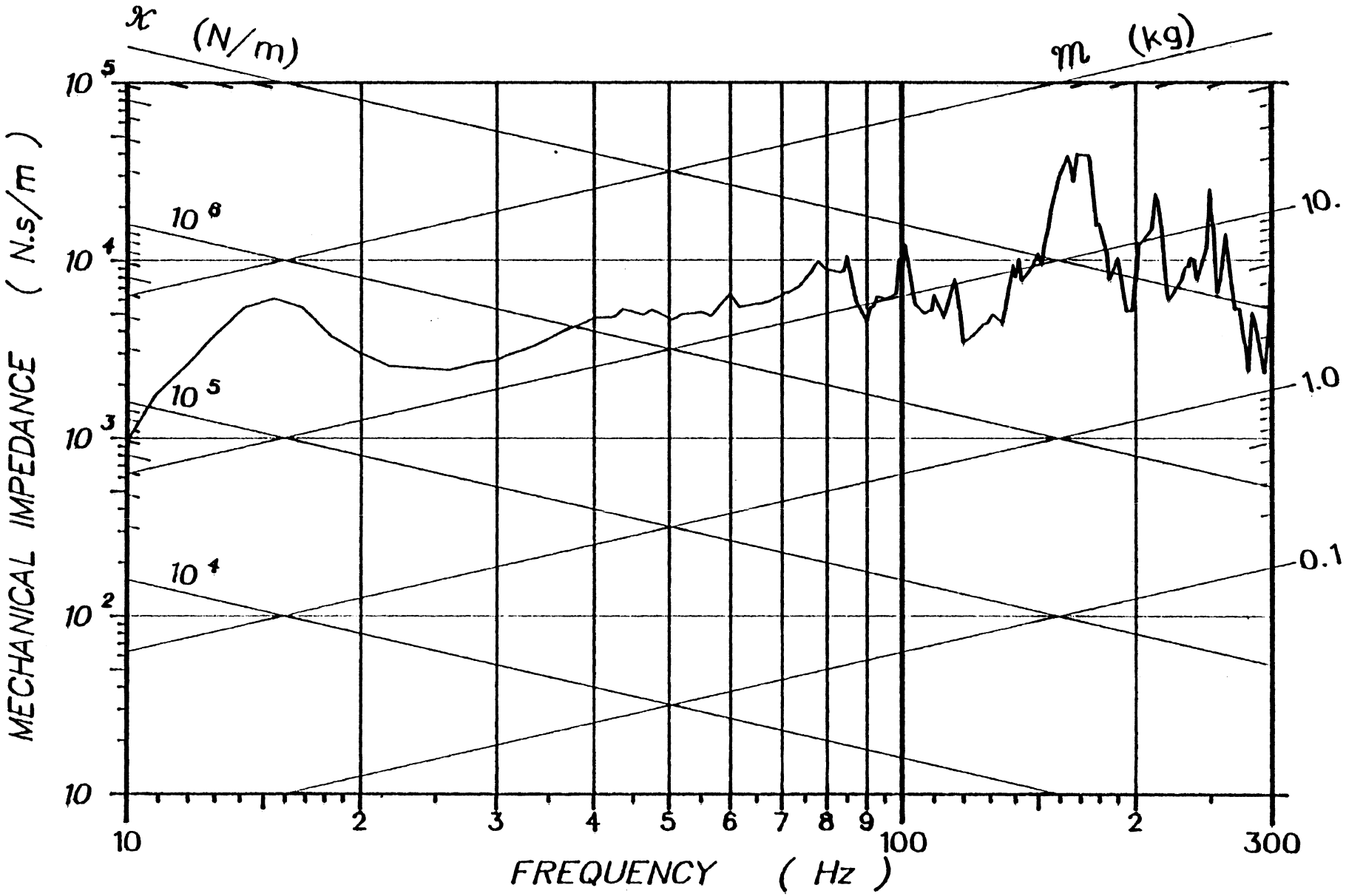


Figure 70: HEAD (P–A)

79H202

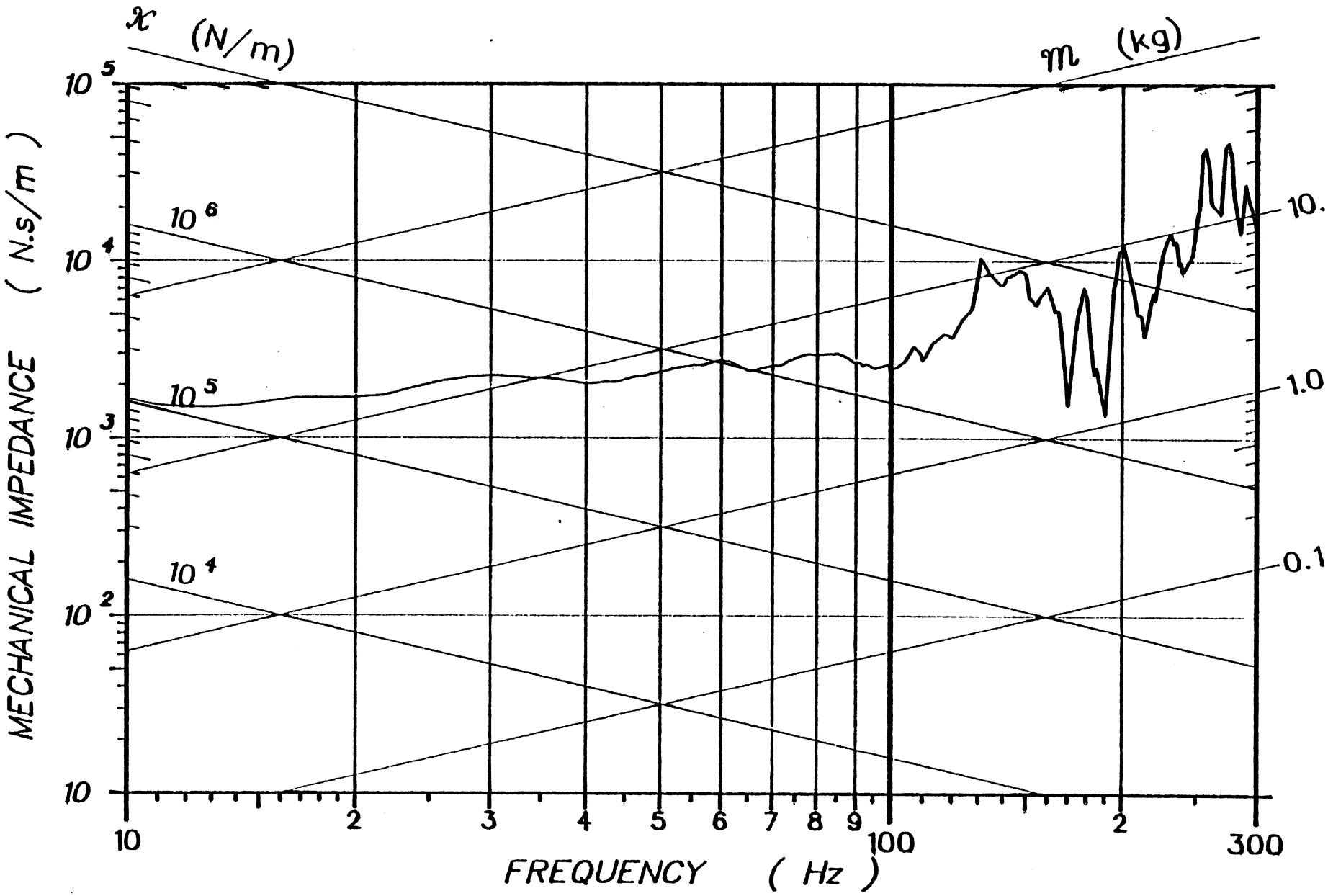


Figure 71: HEAD (P—A)

79H203

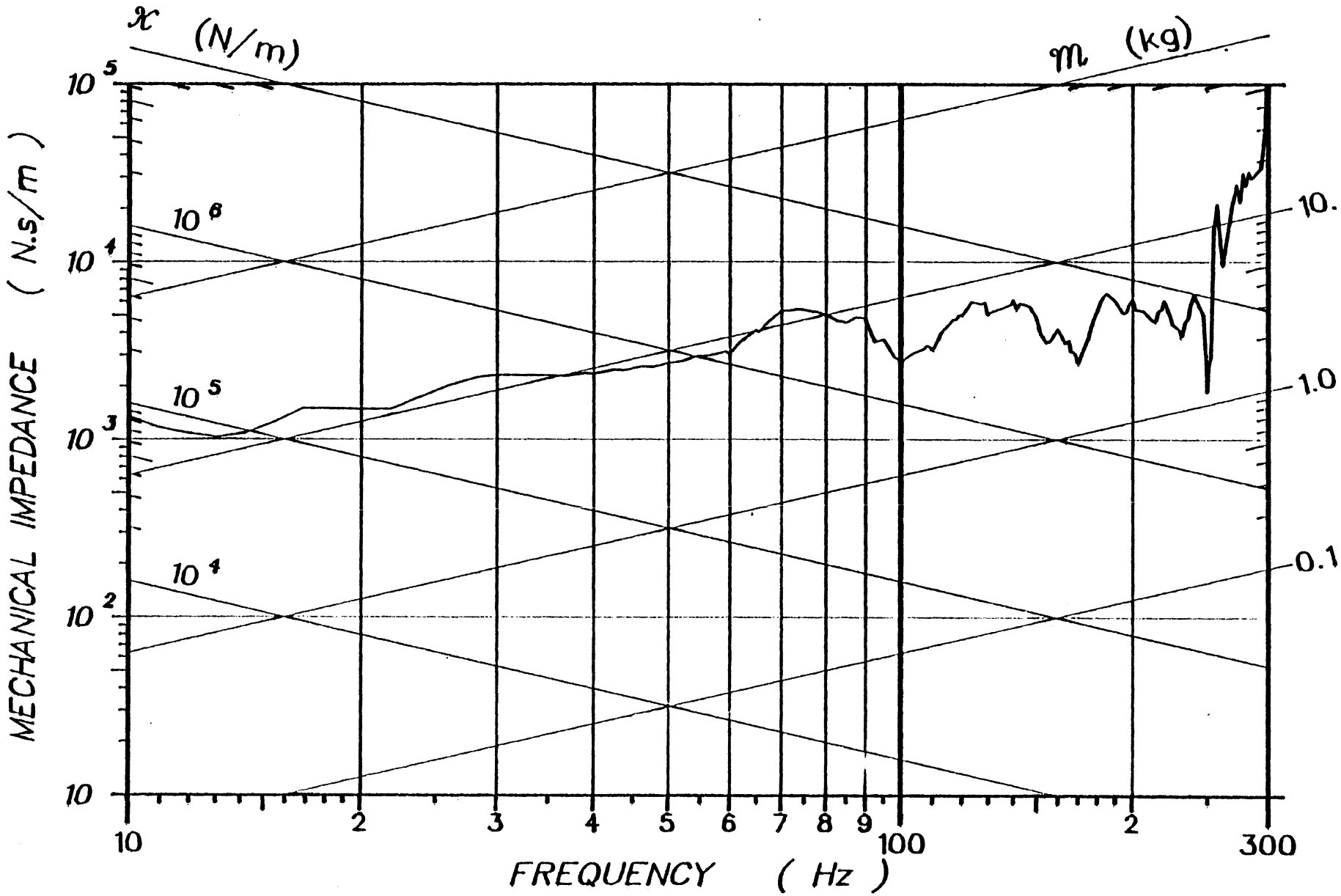


Figure 72: HEAD (P—A)

79H204

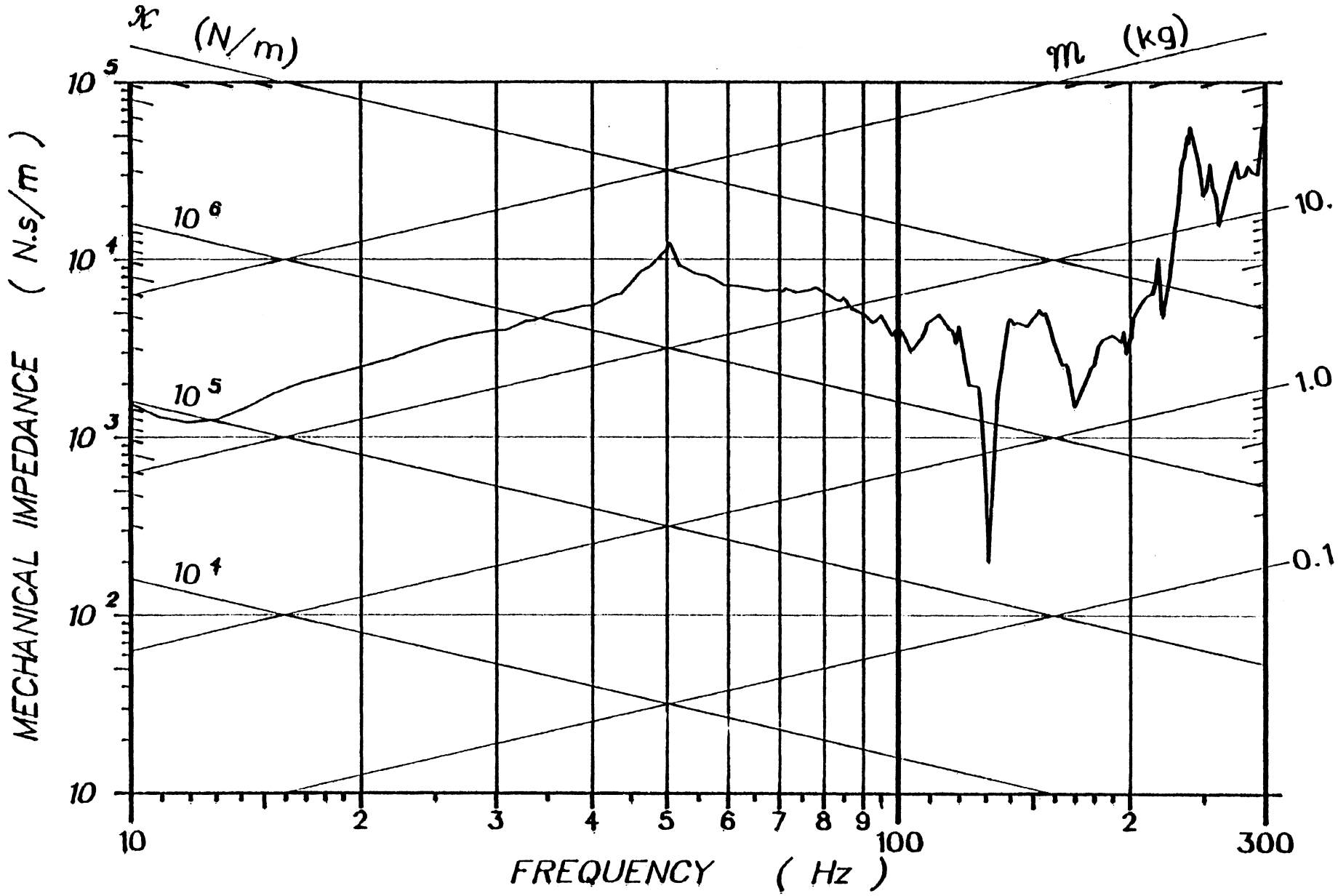


Figure 73: HEAD (P–A)

79H205

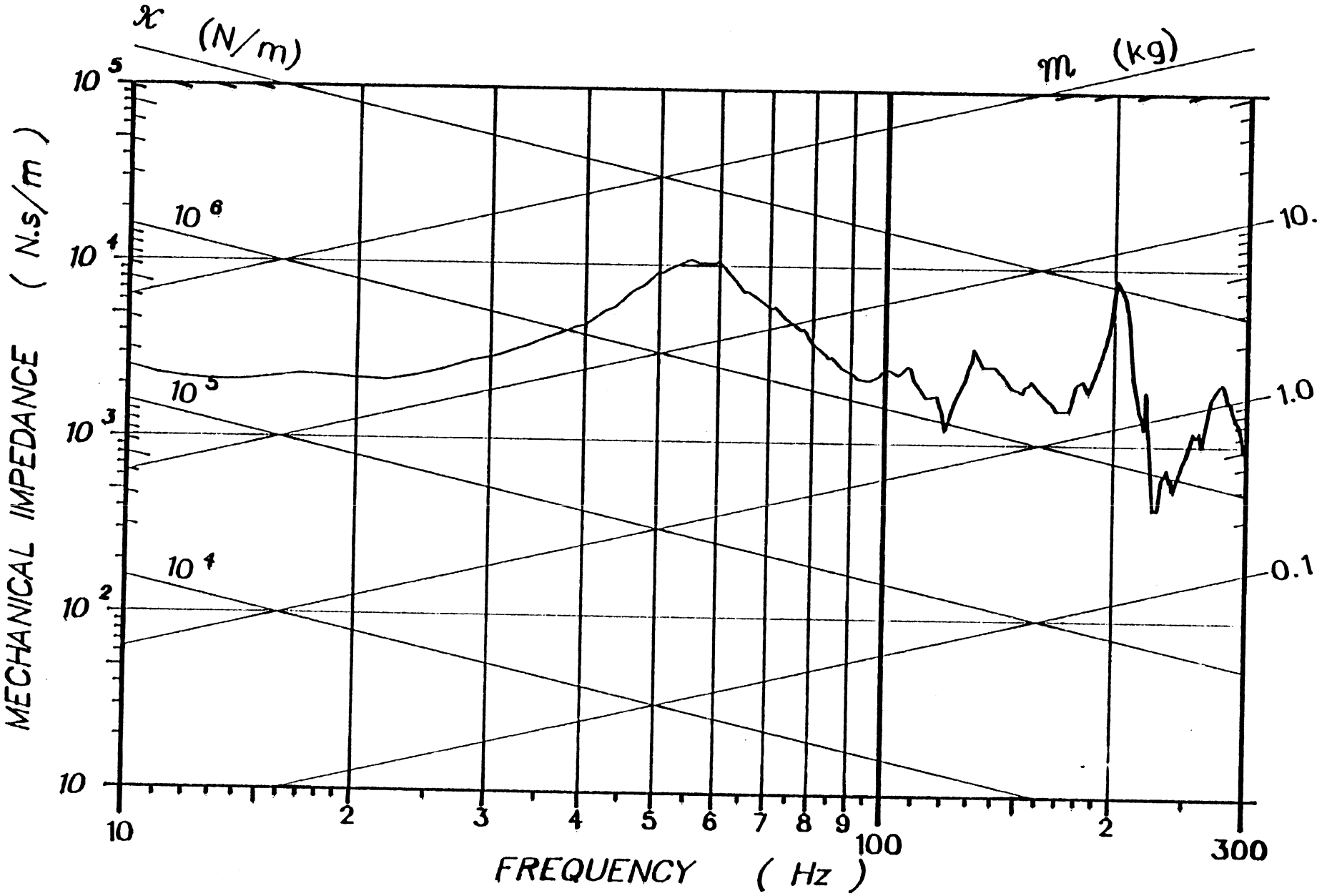


Figure 74: T1 (P—A)

79H201

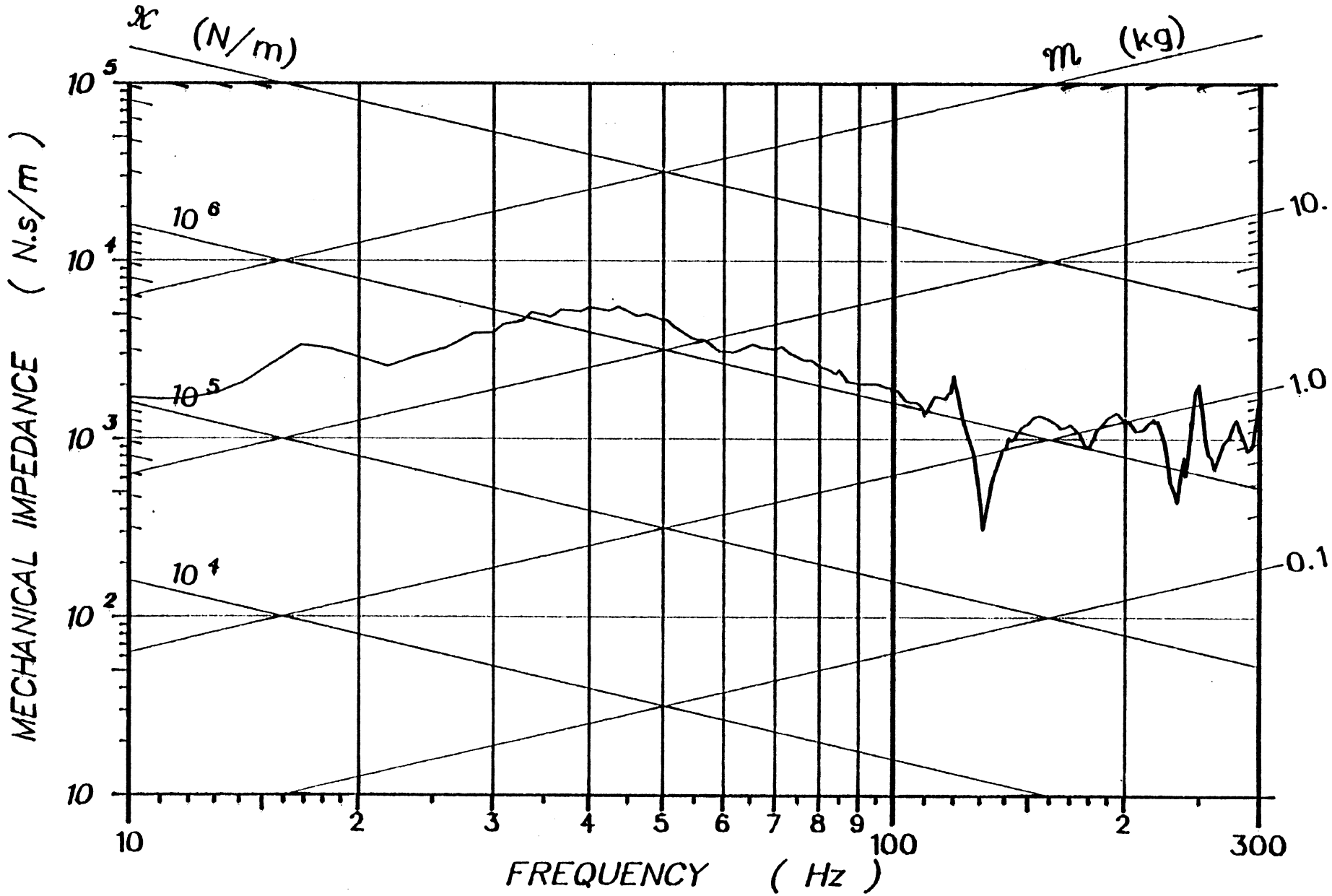


Figure 75: T1 (P—A)

79H202

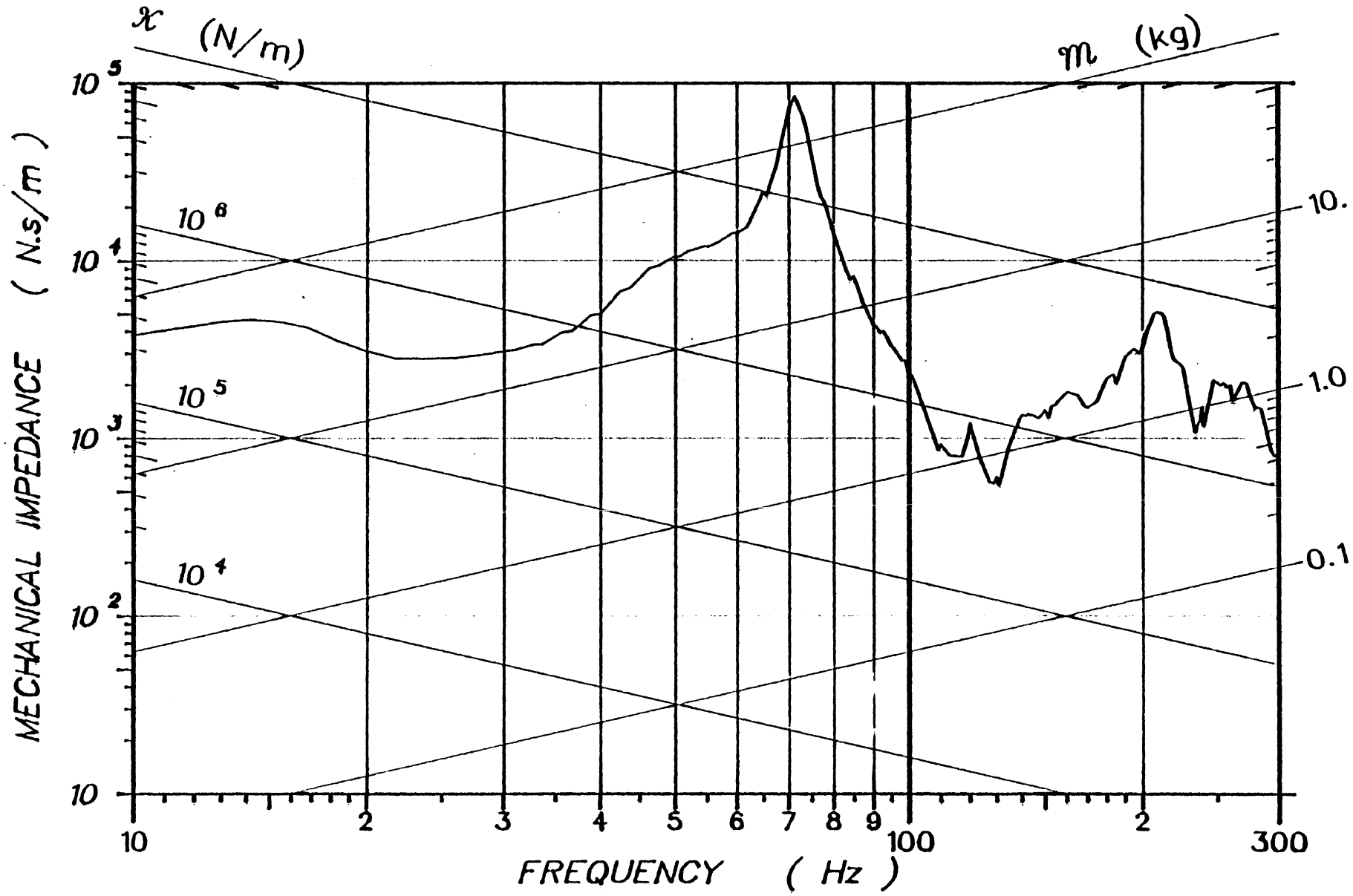


Figure 76: T1 (P—A)

79H203

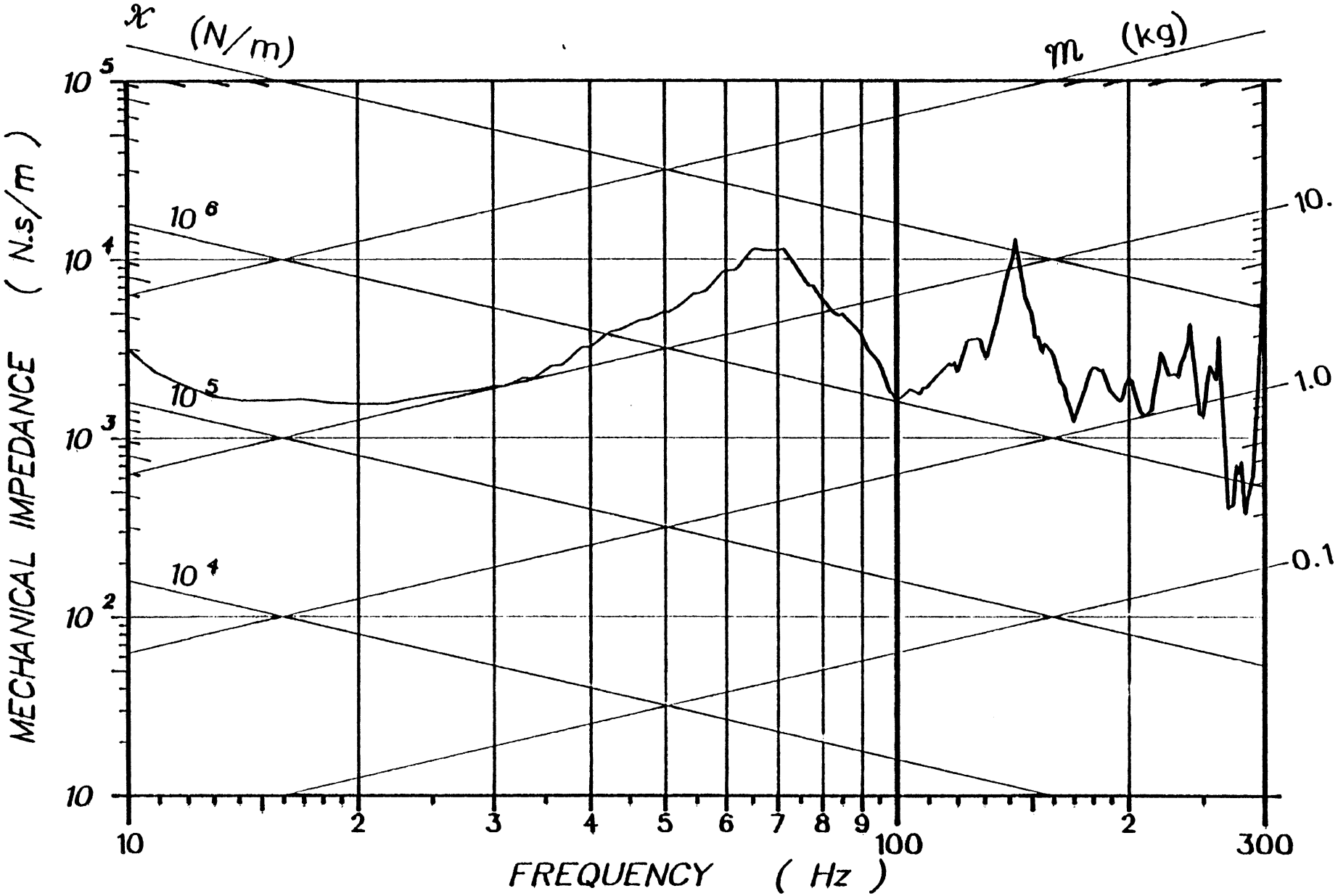


Figure 77: T1 (P—A)

79H204

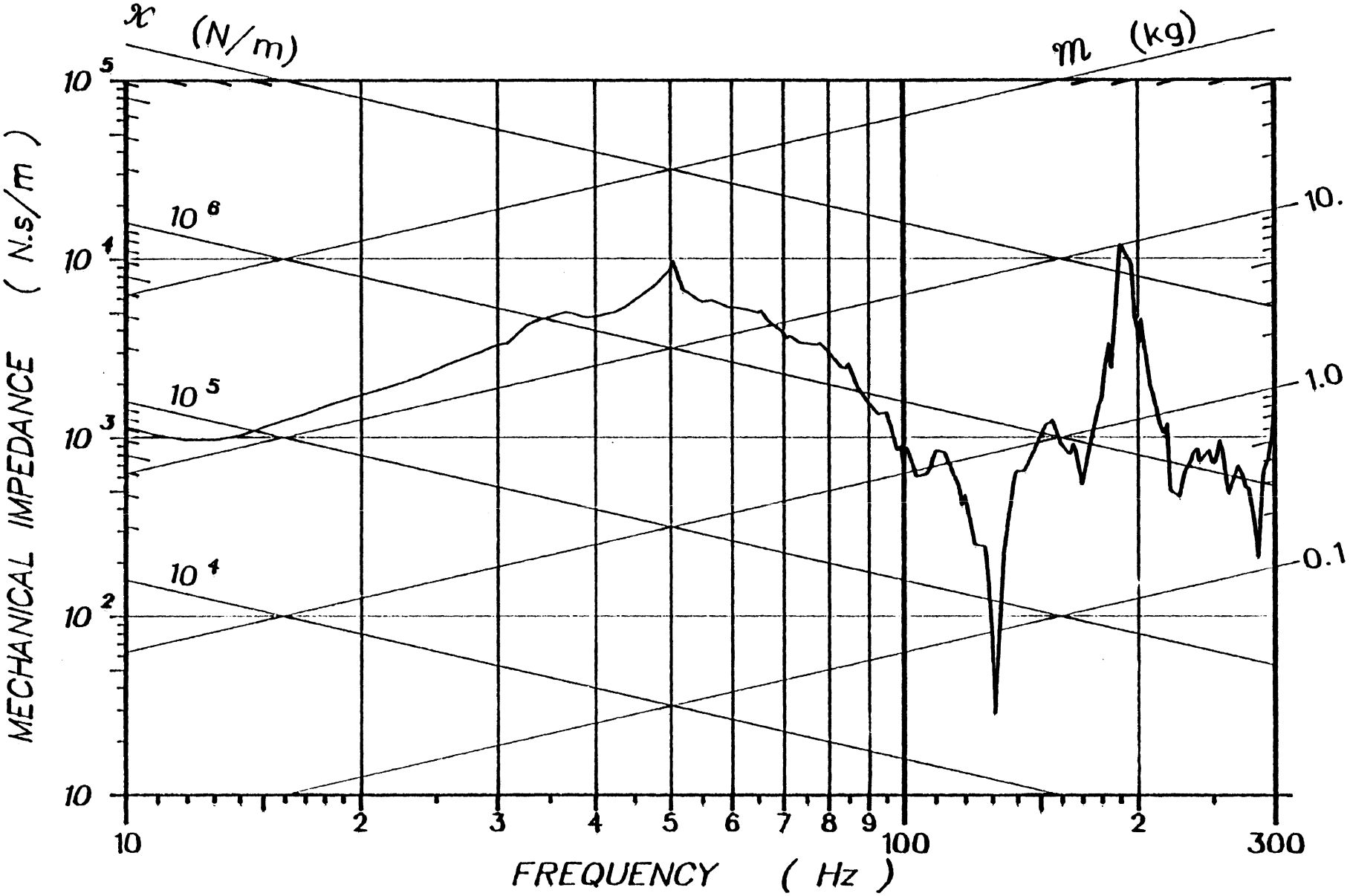


Figure 78: T1 (P—A)

79H205

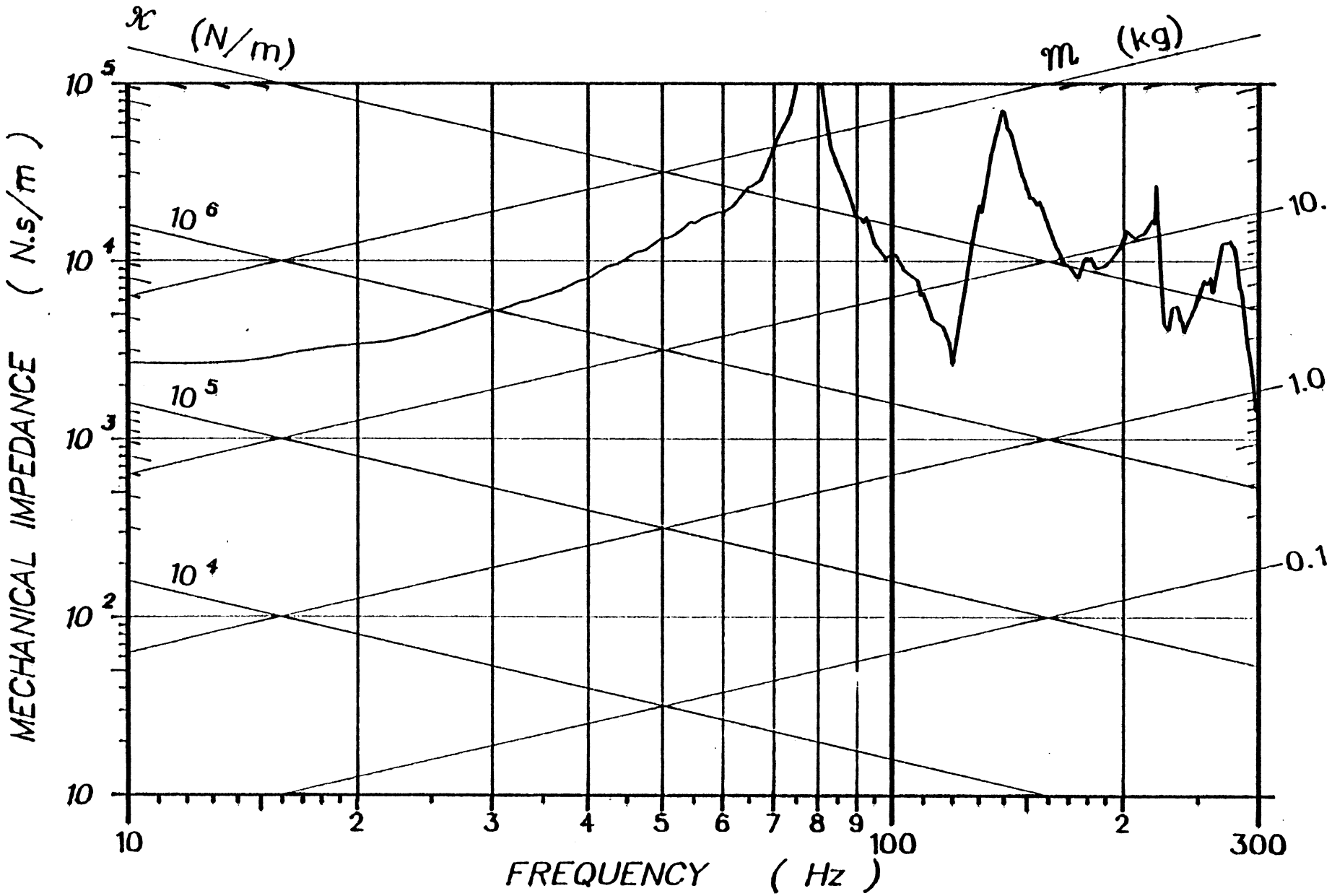


Figure 79: T12 (P—A)

79H201

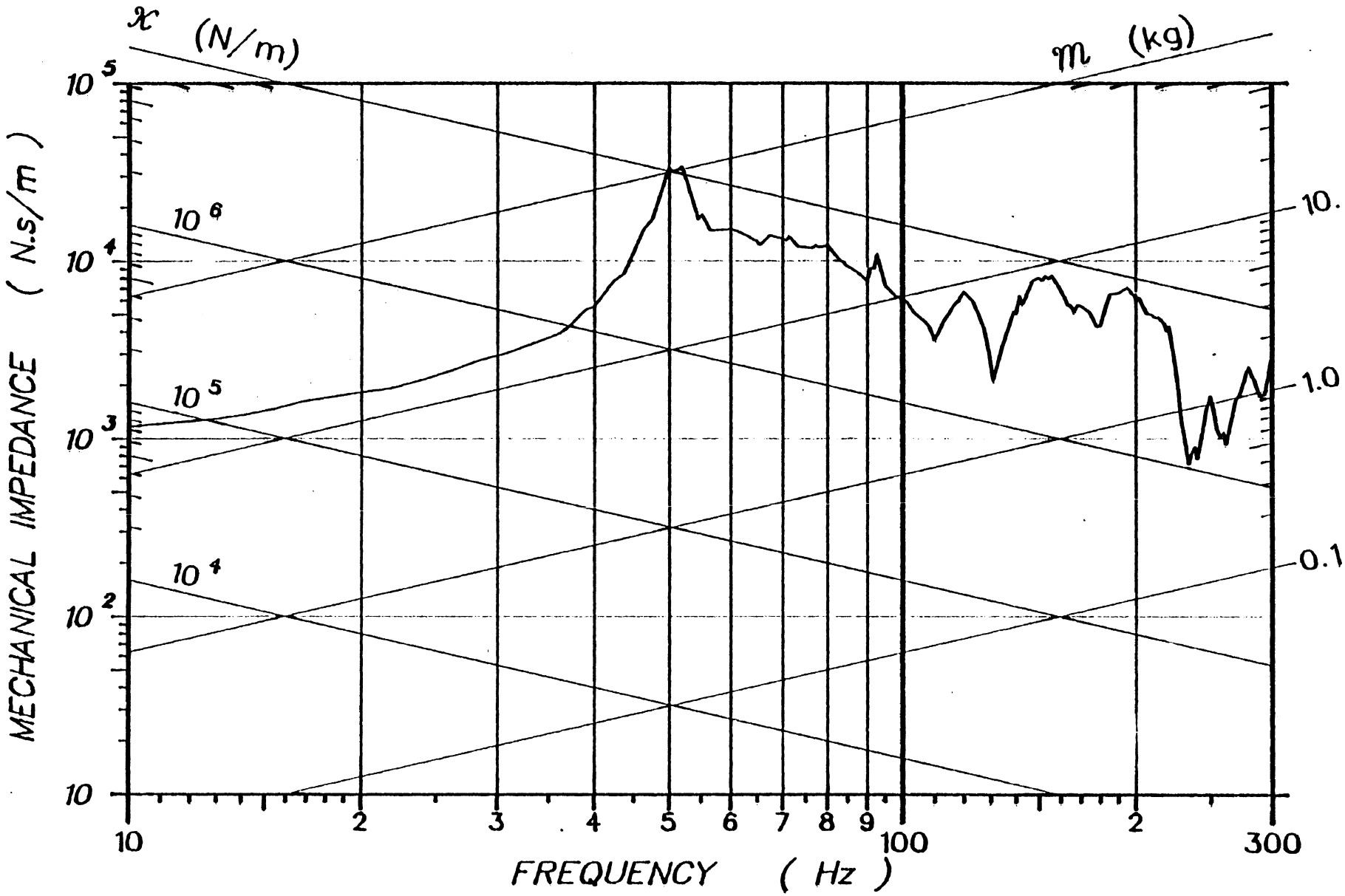


Figure 80: T12 (P–A)

79H202

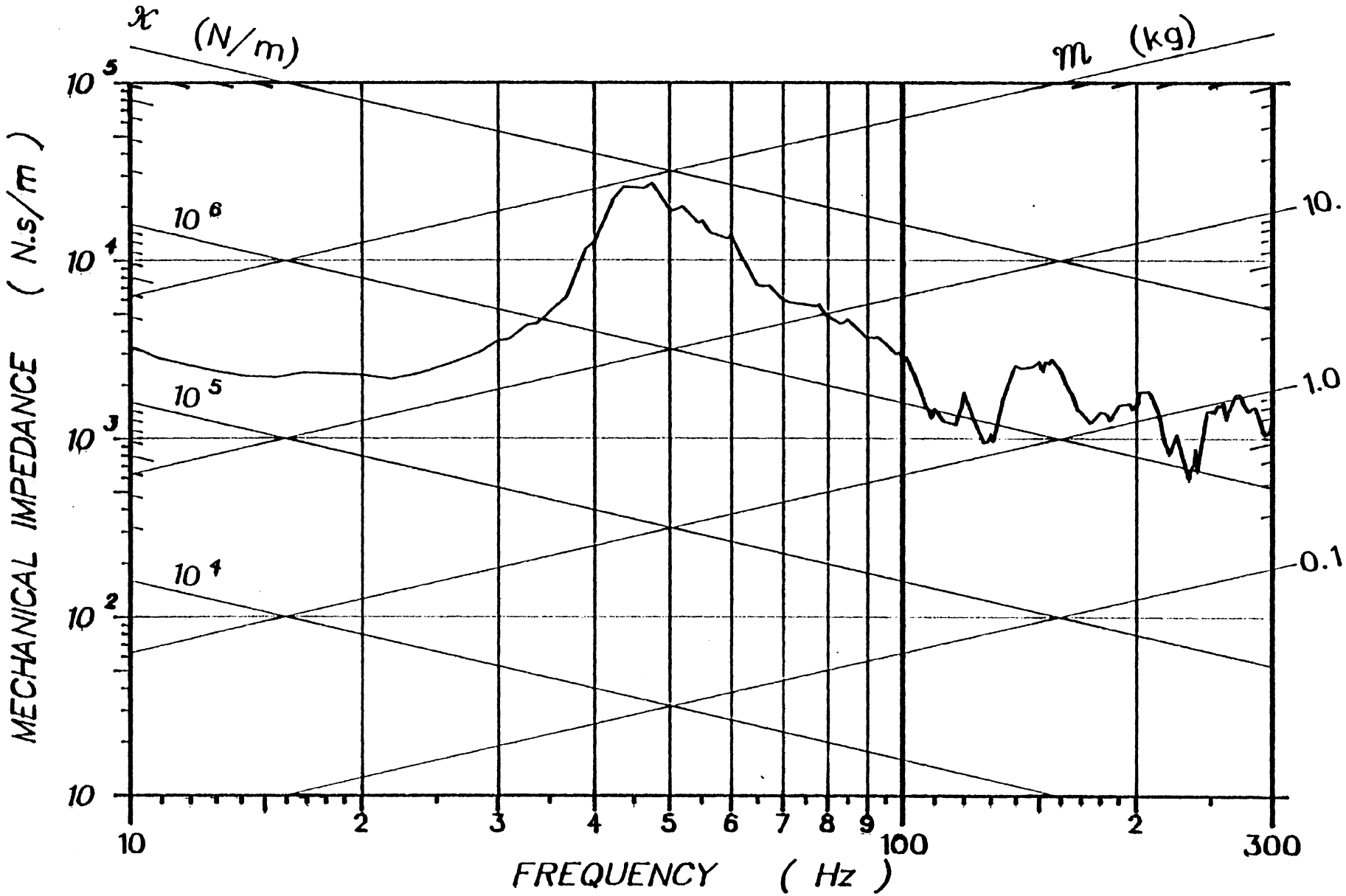


Figure 81: T12 (P—A)

79H203

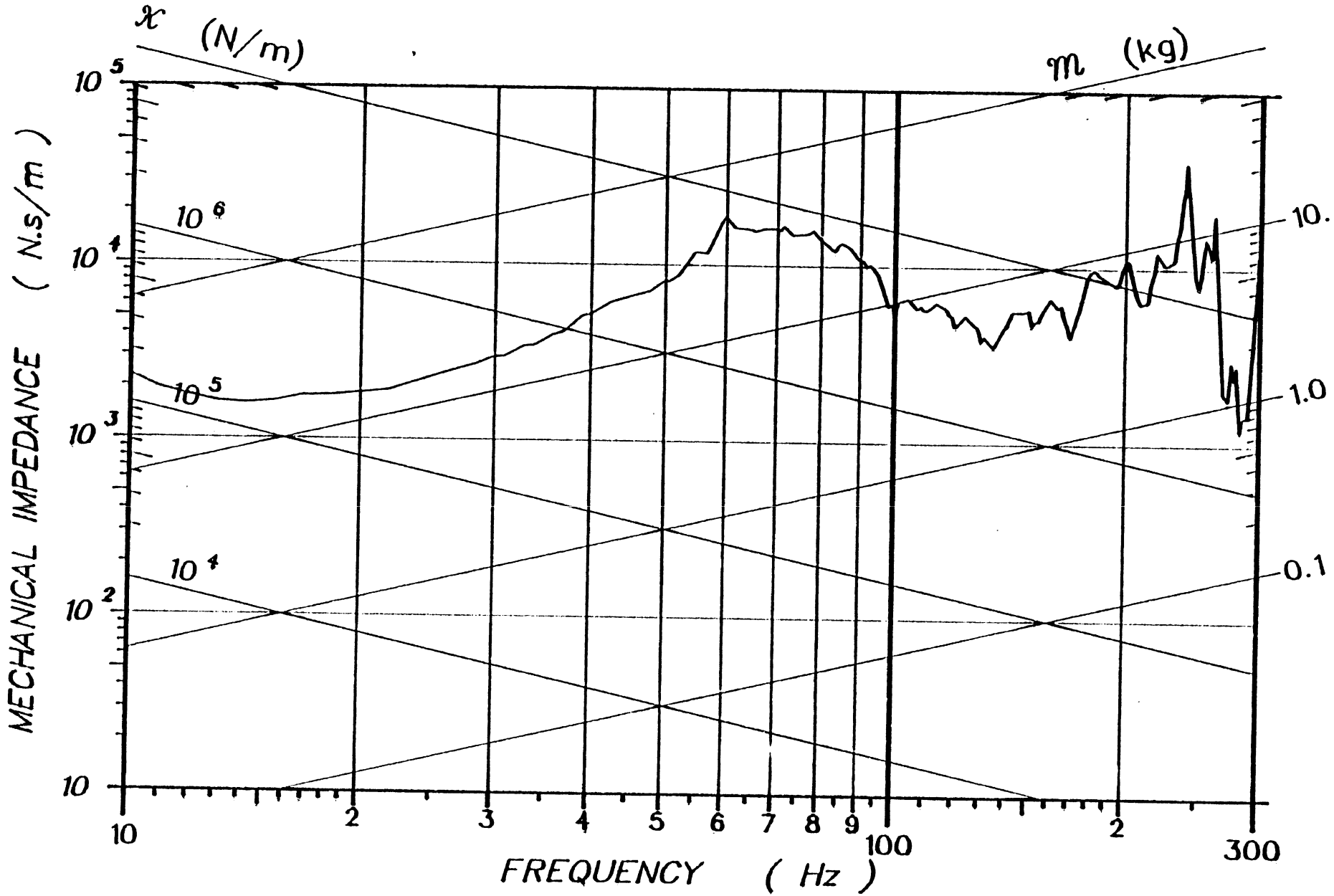


Figure 82: T12 (P–A)

79H204

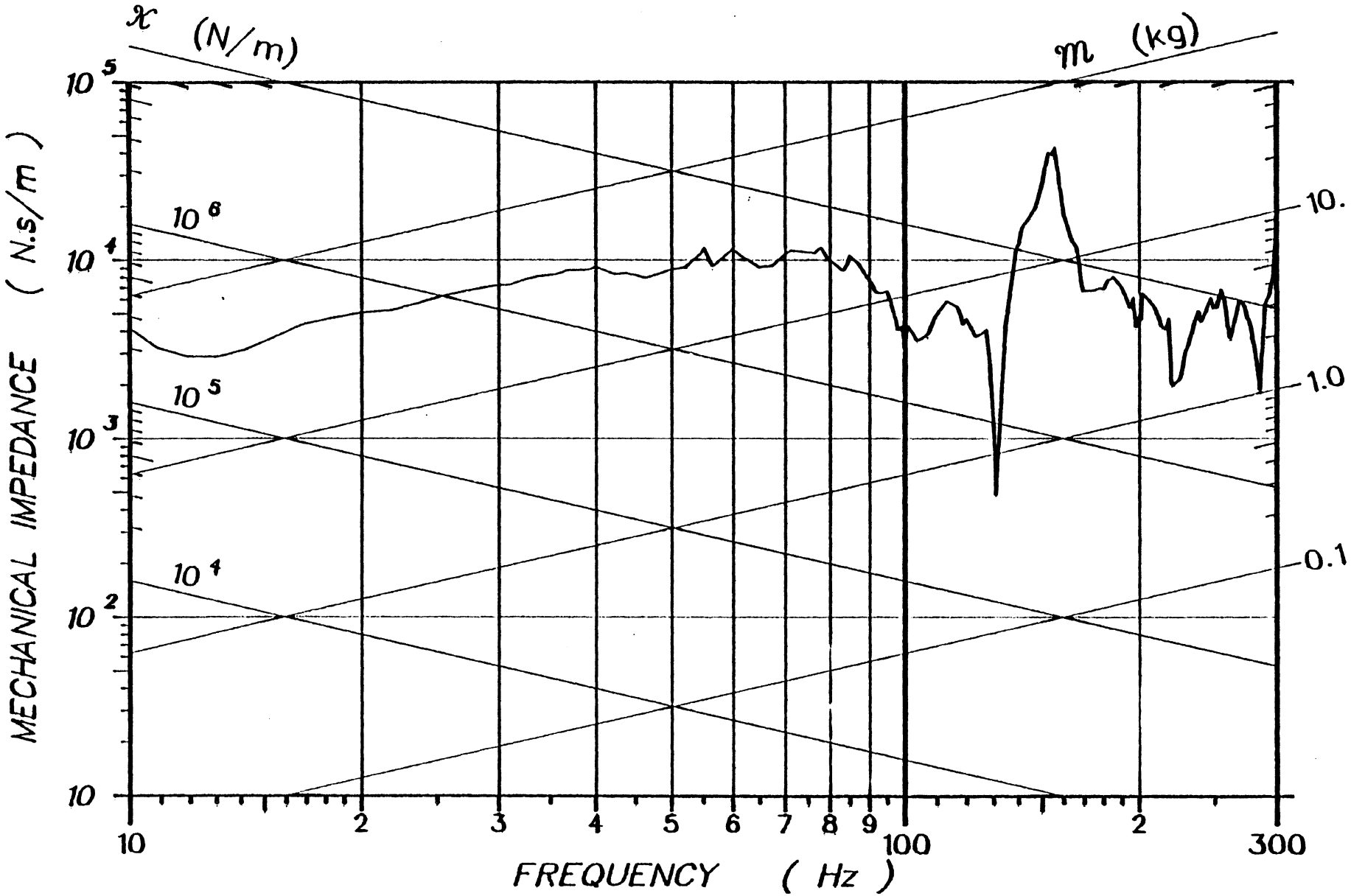


Figure 83: T12 (P—A)

79H205

2.6.3. R-L Transfer Impedance Curves

The transfer impedances between the point of impact and various observation points in the R-L direction are given in figures 84-88 for the head CG, in figures 89-93 for T1 and in figures 94-98 for T12.

As with the P-A impedances, these R-L impedances are higher than the resultant impedances and even higher than P-A impedances themselves. This indicates that the system response to S-I impact is less sensitive in the R-L direction than other direction responses.

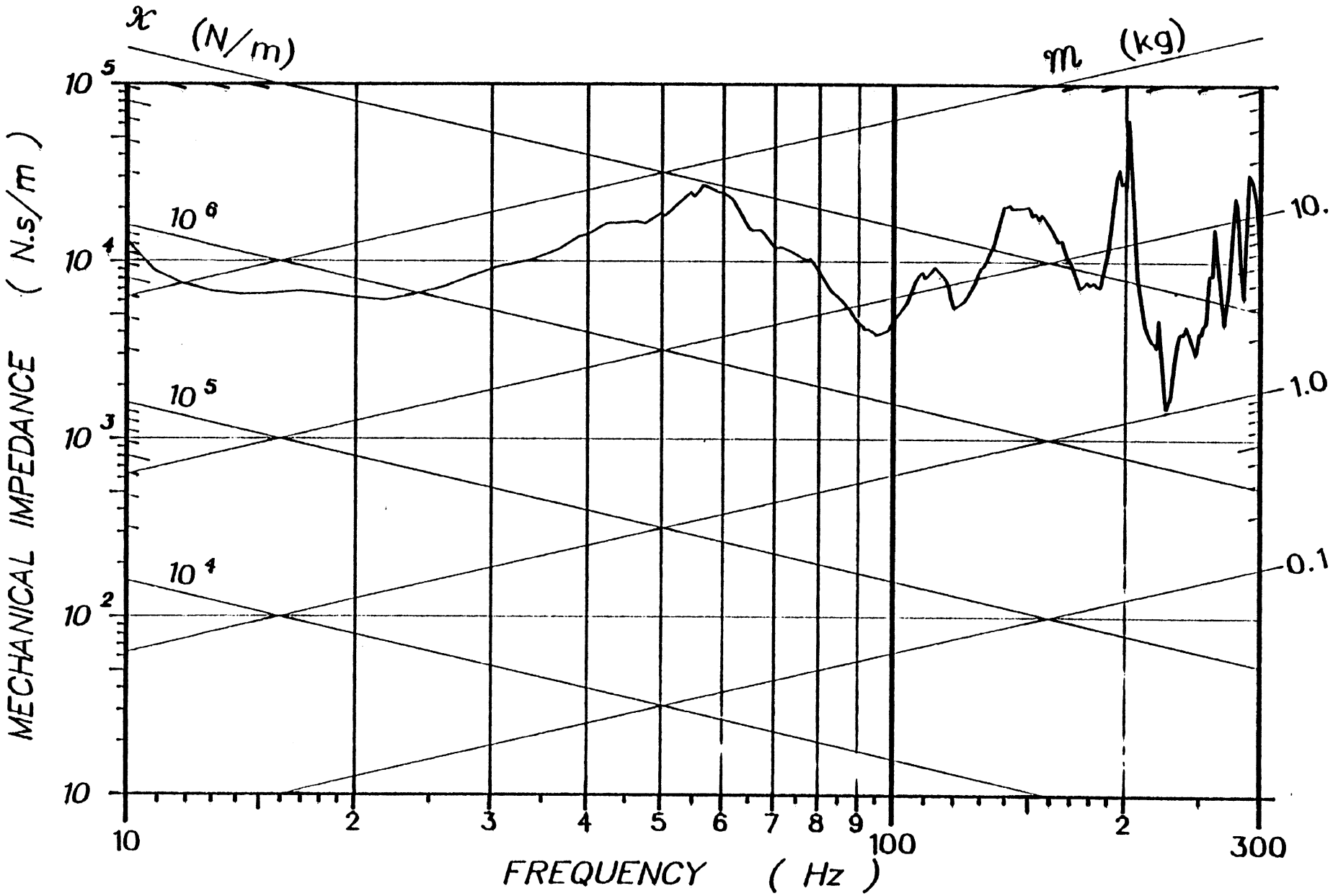


Figure 84: HEAD (R–L)

79H201

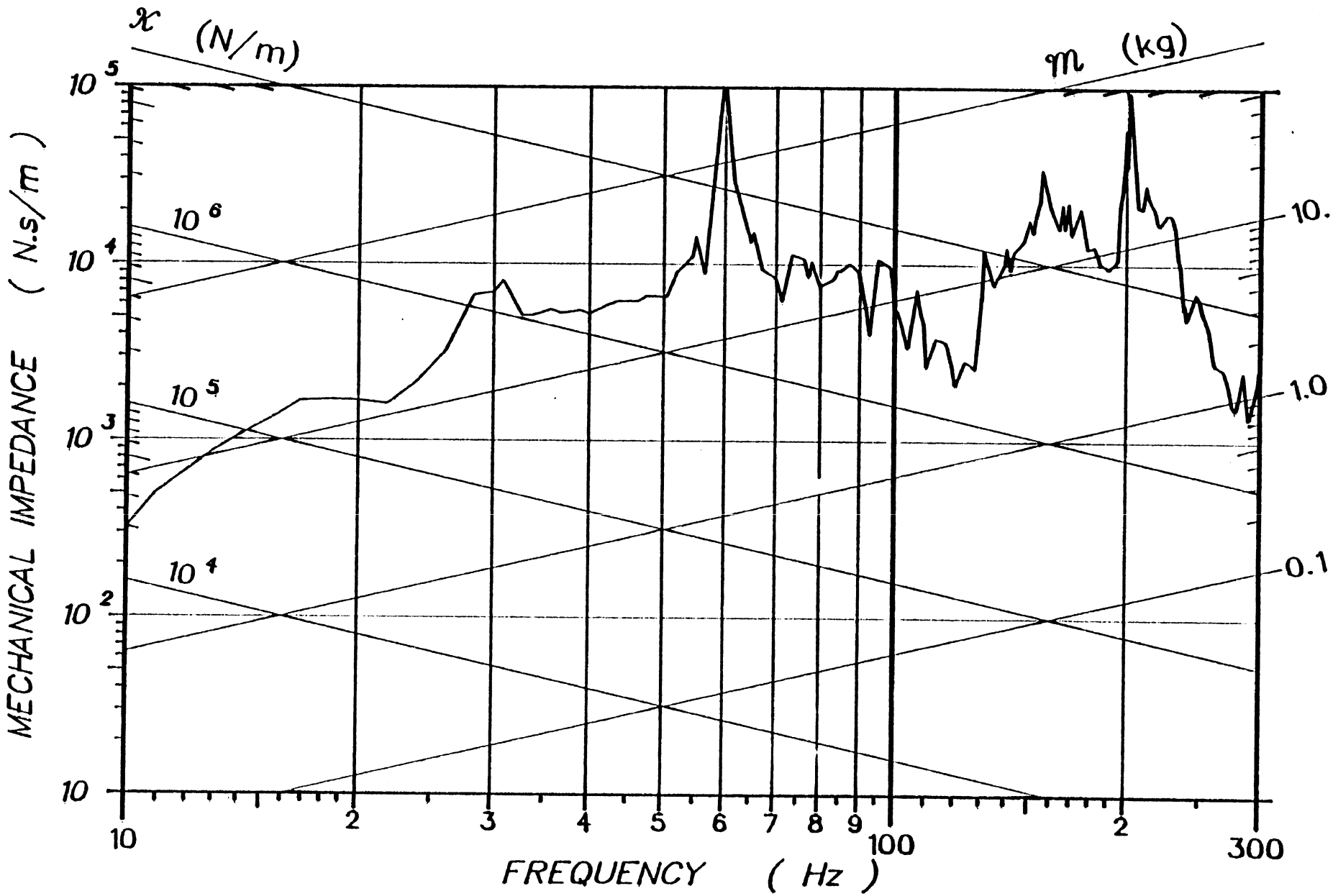


Figure 85: HEAD (R–L)

79H202

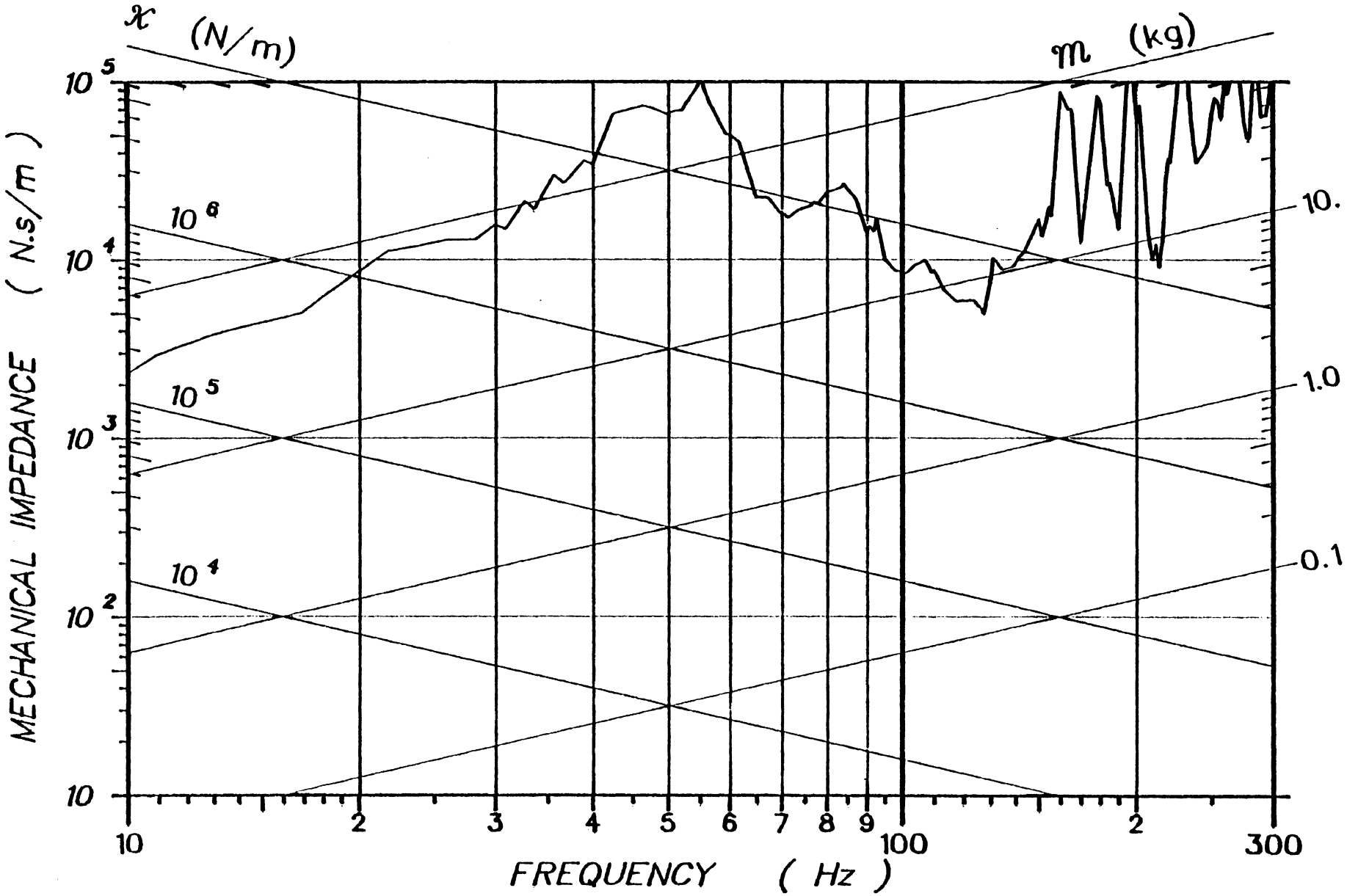


Figure 86: HEAD (R–L)

79H203

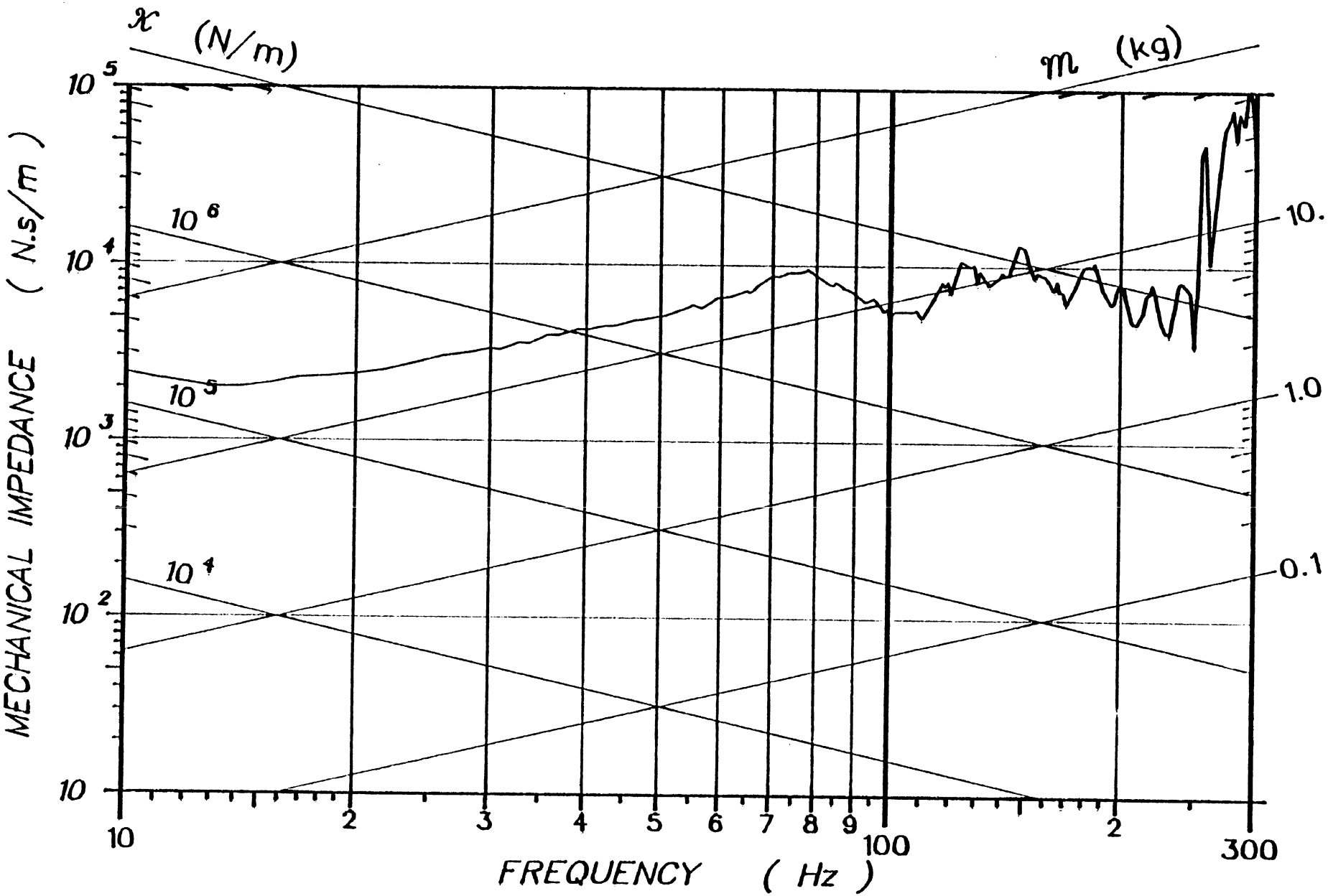


Figure 87: HEAD (R–L)

79H204

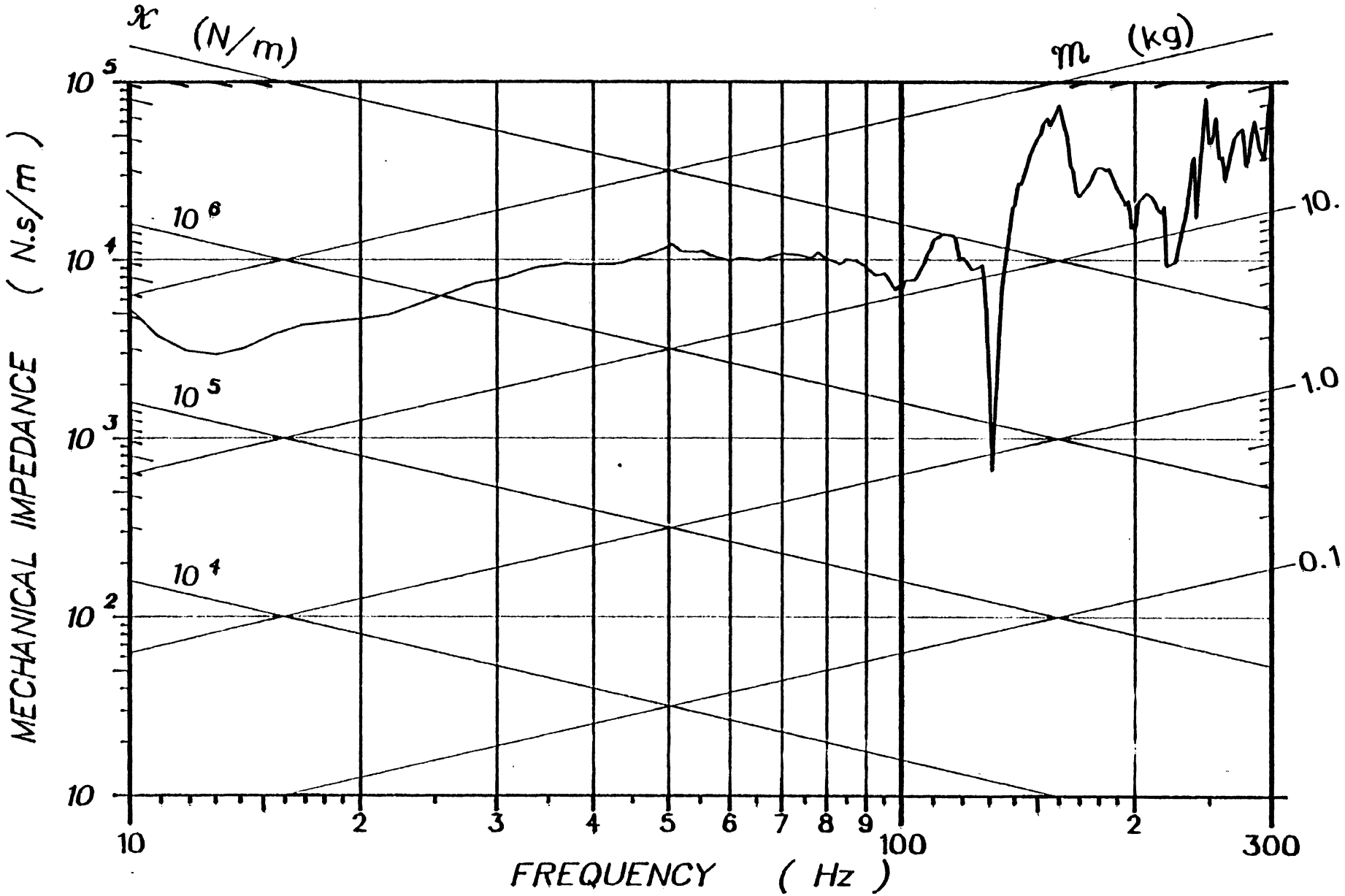


Figure 88: HEAD (R–L)

79H205

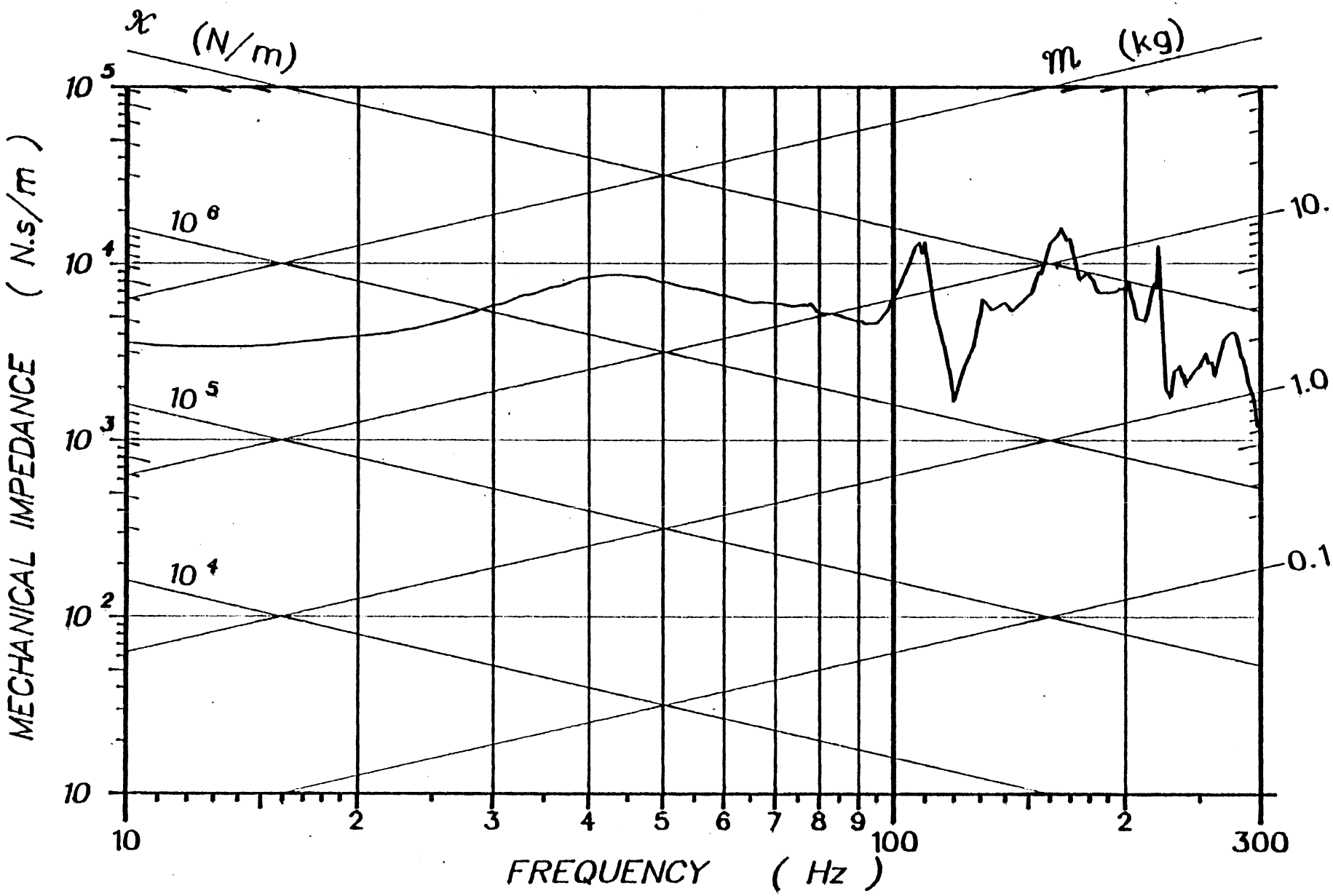


Figure 89: T1 (R–L)

79H201

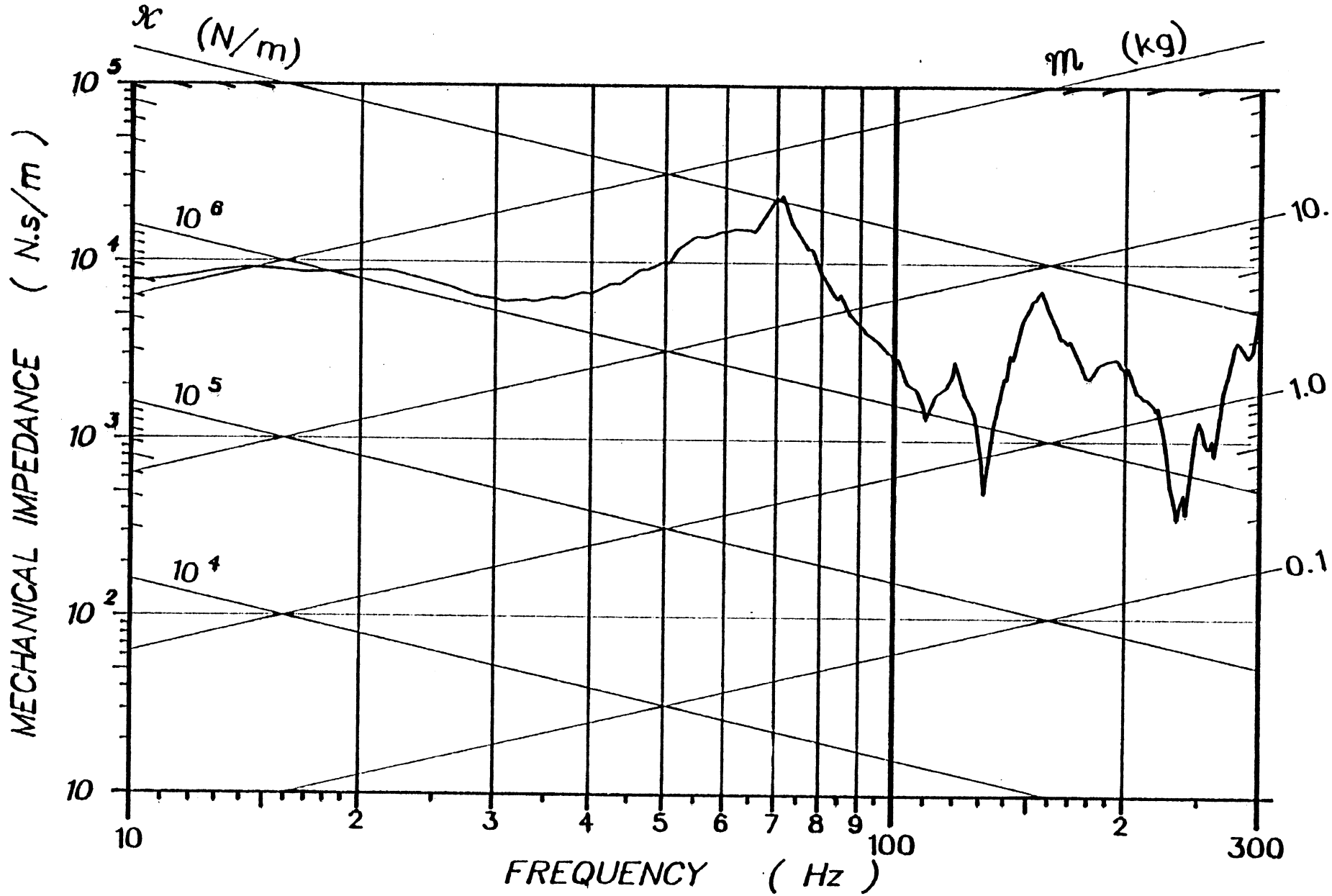


Figure 90: T1 (R–L)

79H202

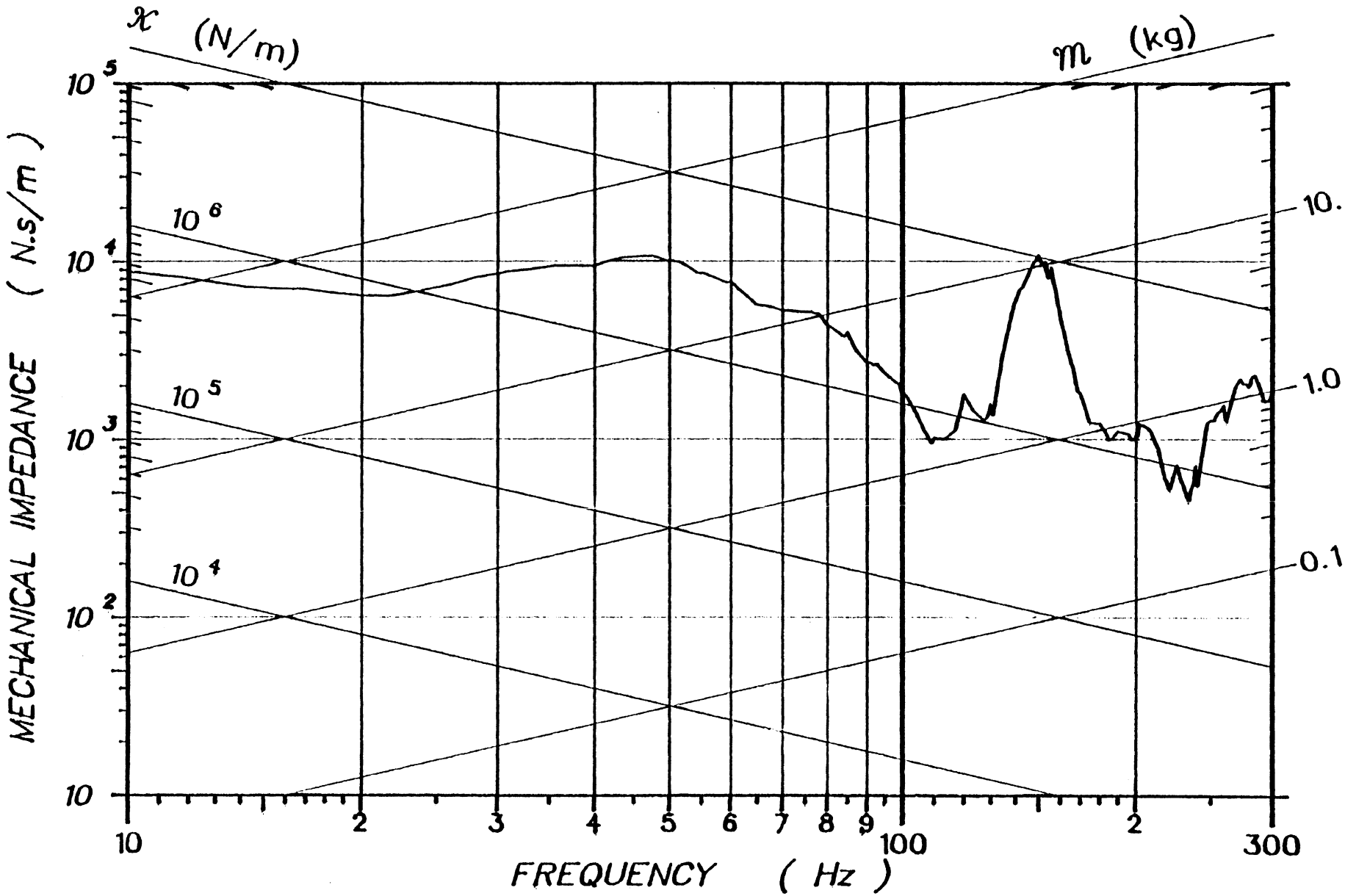


Figure 91: T1 (R–L)

79H203

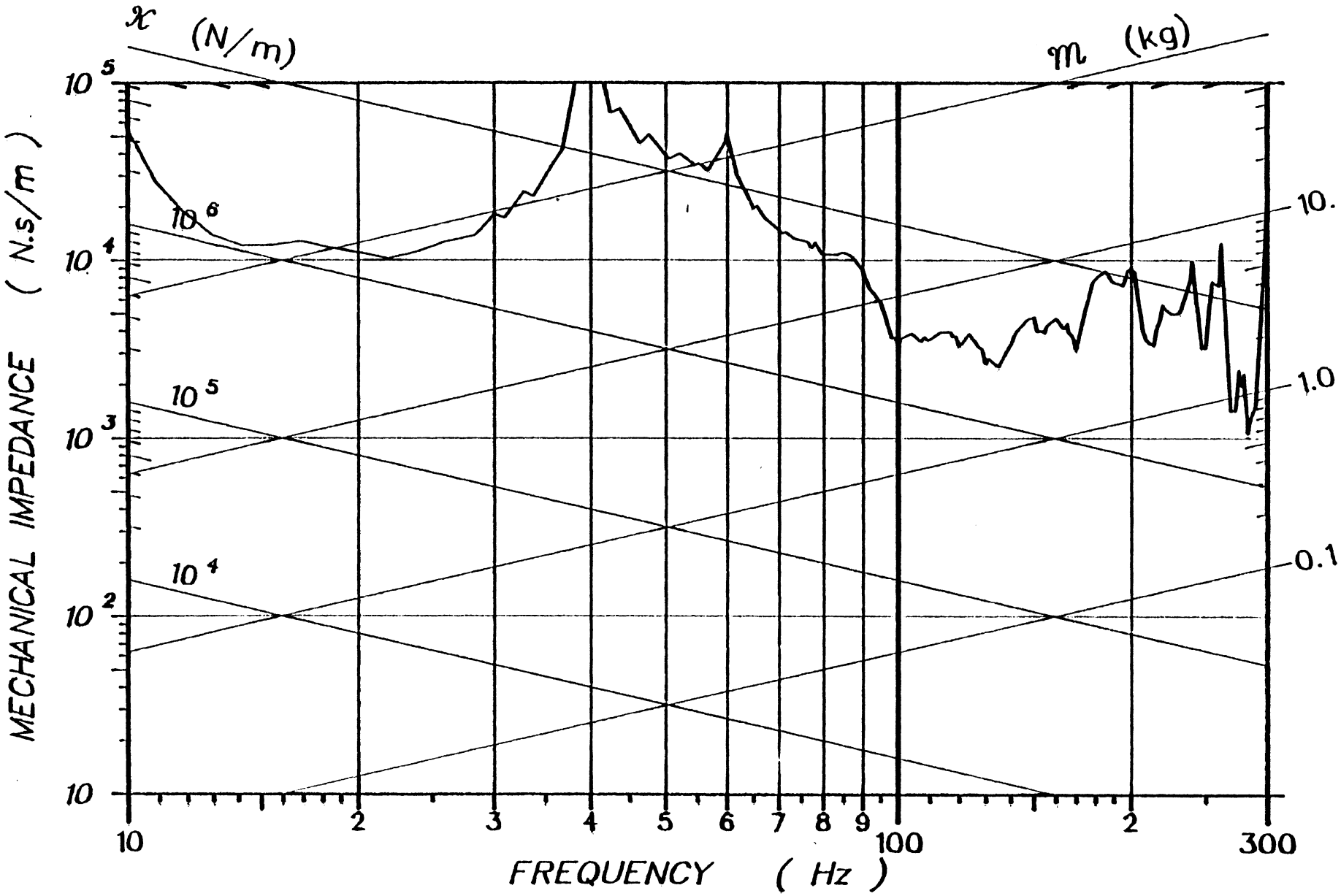


Figure 92: T1 (R–L)

79H204

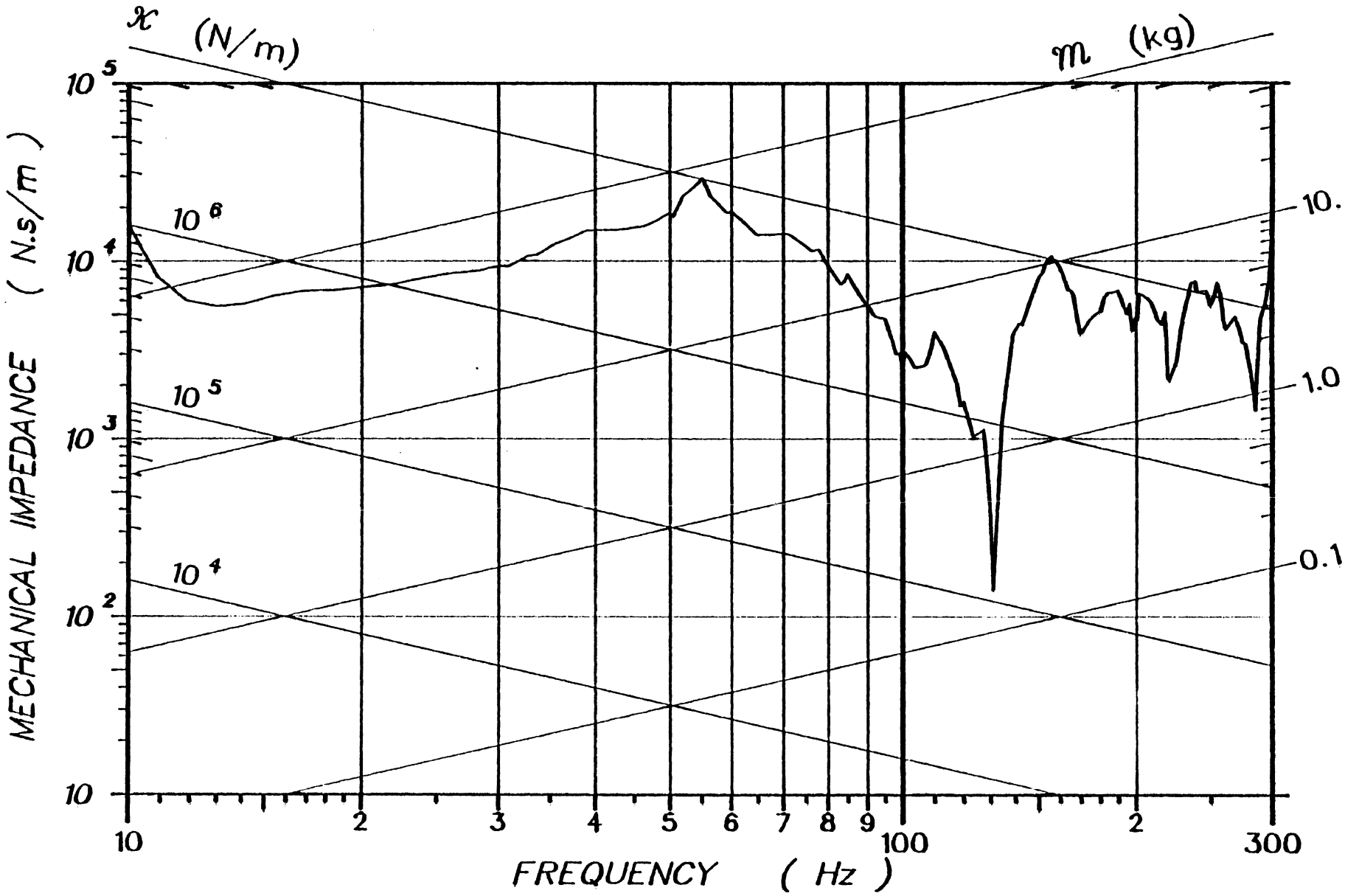


Figure 93: T1 (R–L)

79H205

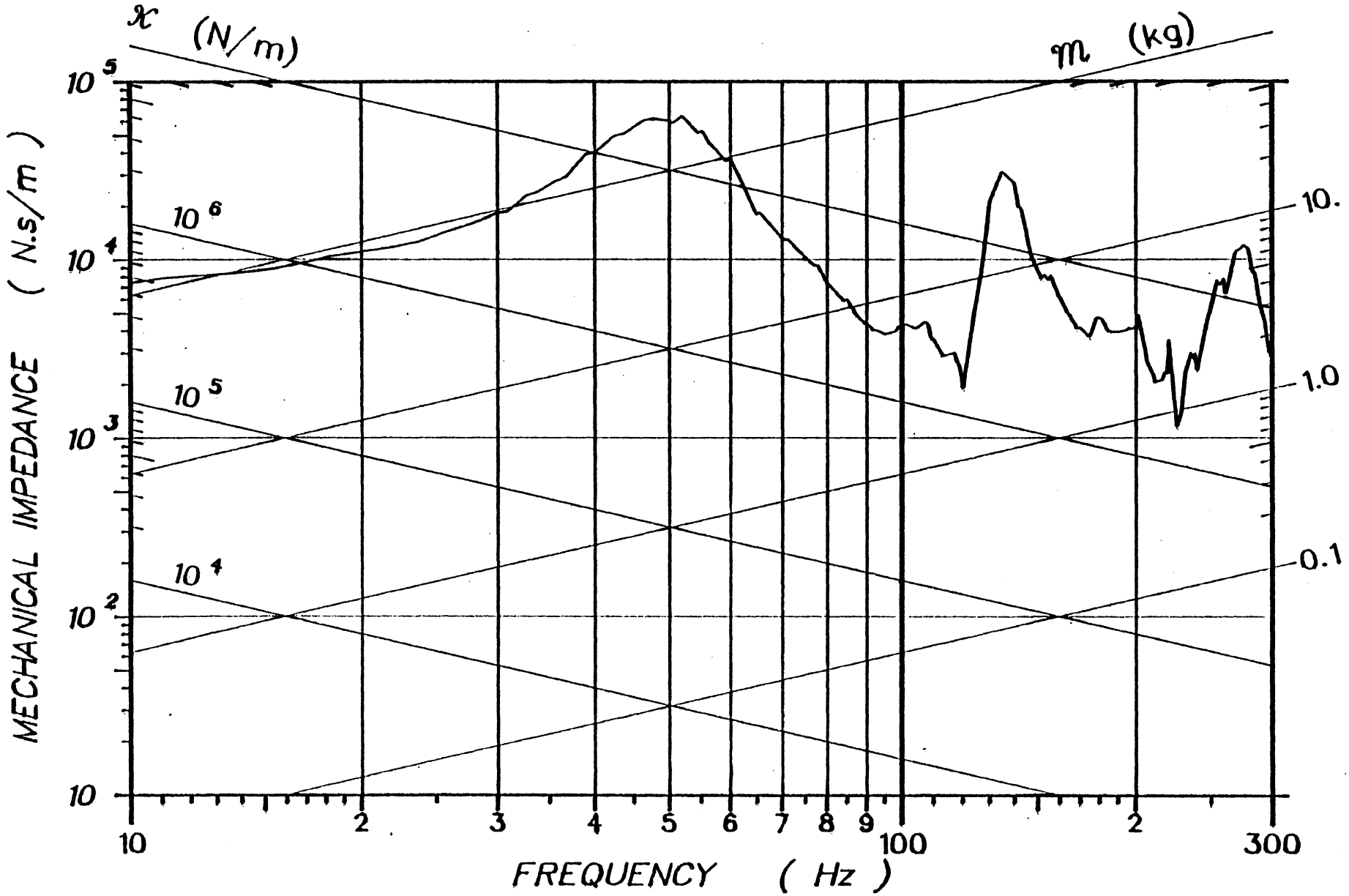


Figure 94: T12 (R–L)

79H201

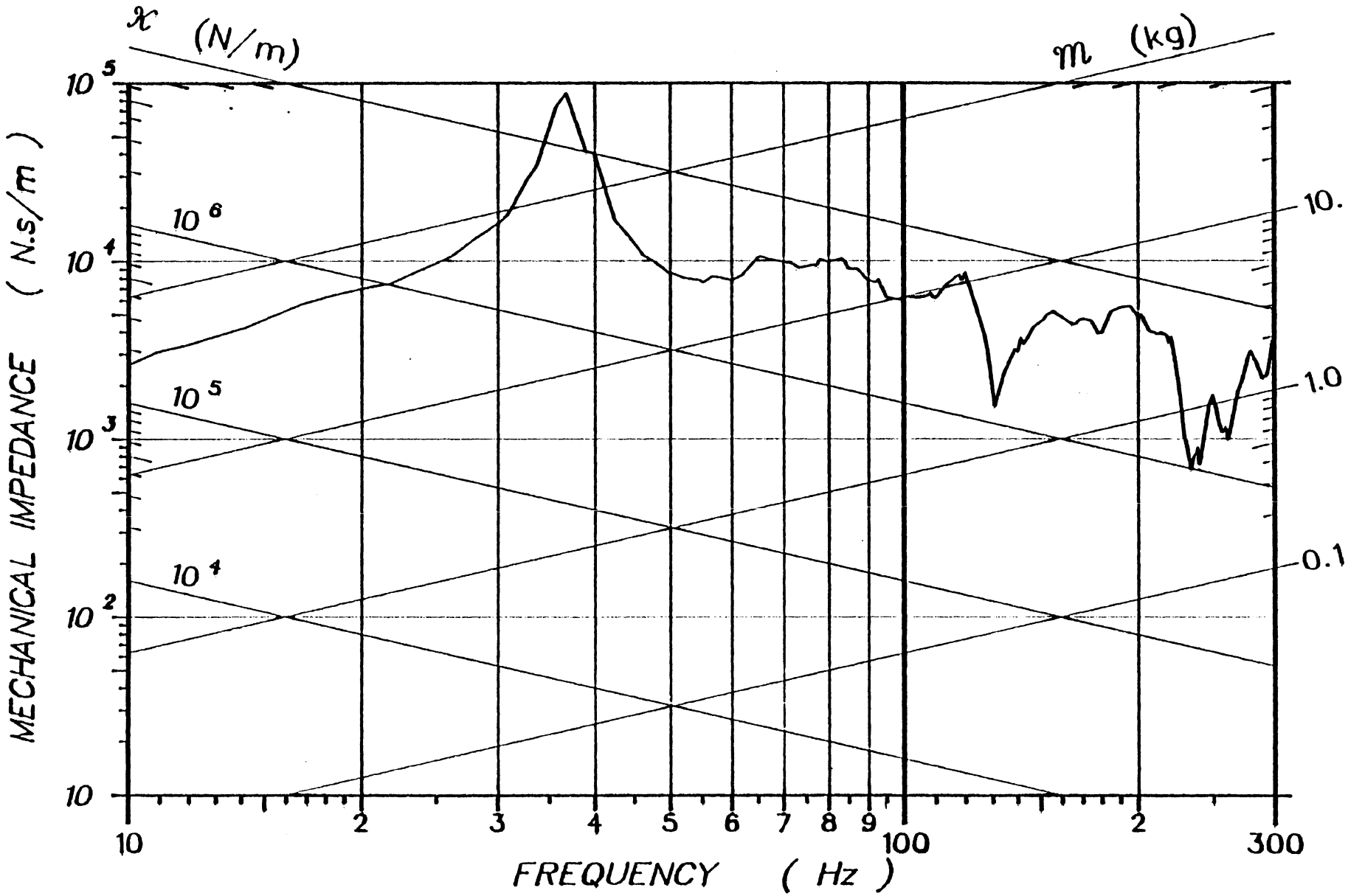


Figure 95: T12 (R–L)

79H202

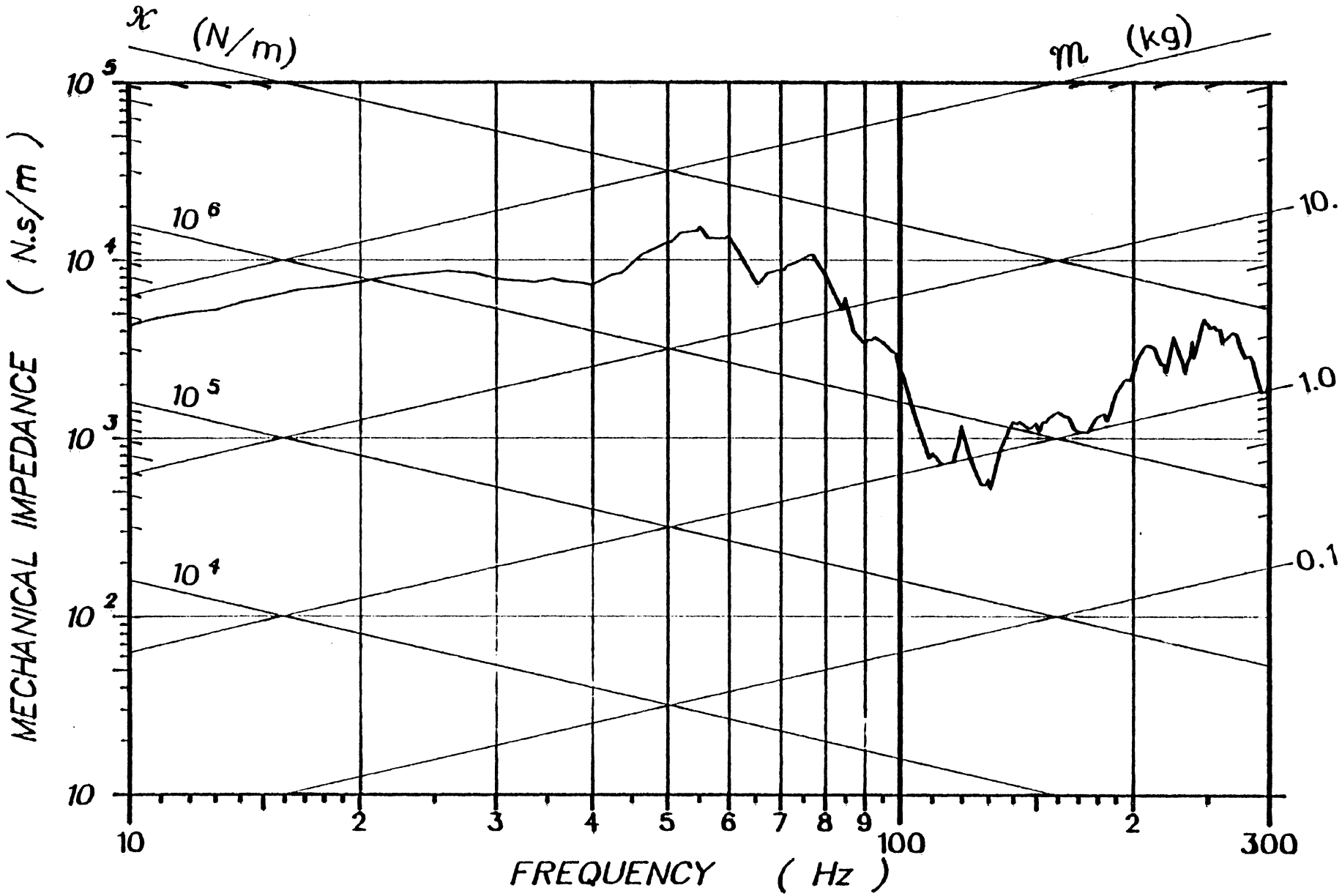


Figure 96: T12 (R–L)

79H203

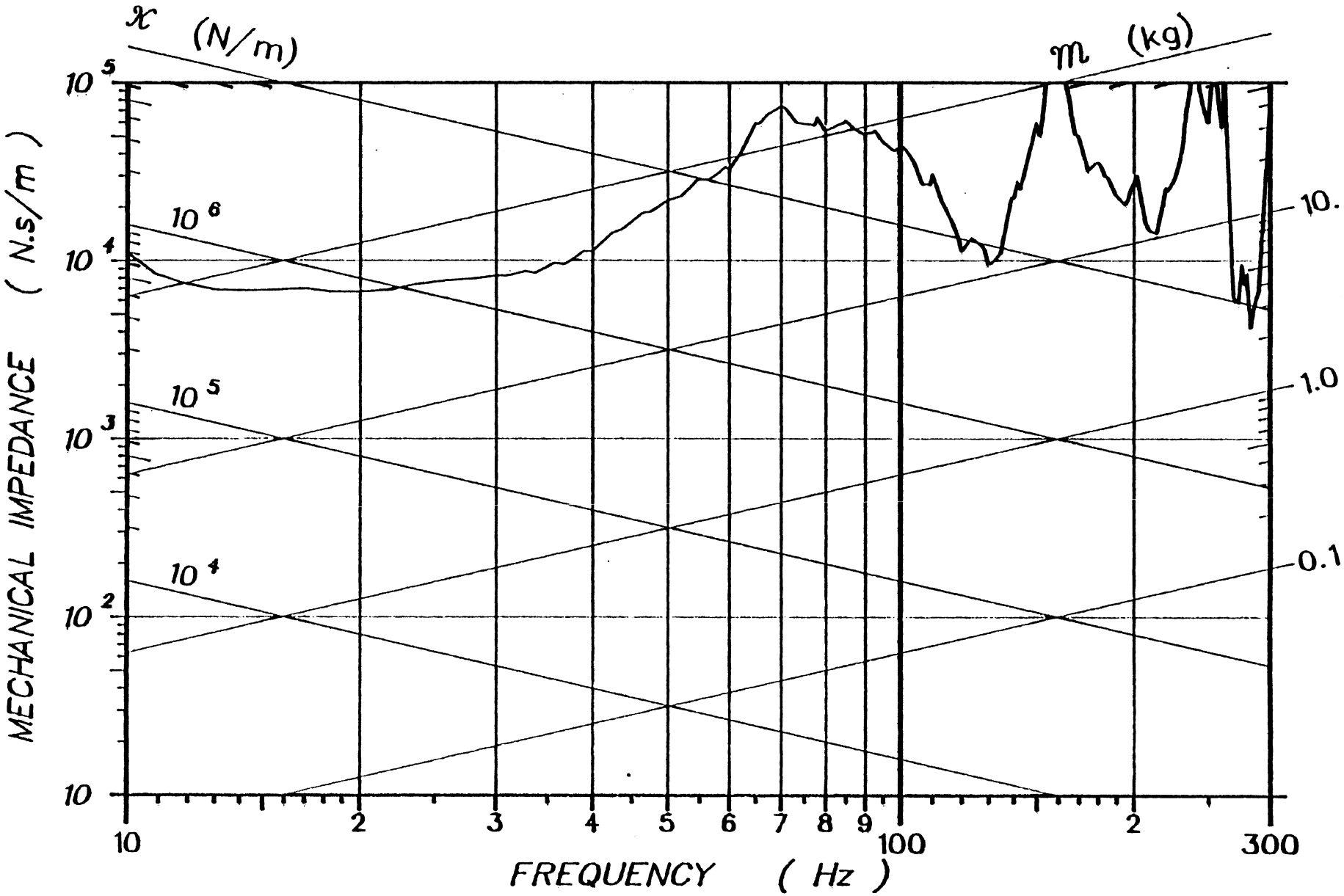


Figure 97: T12 (R–L)

79H204

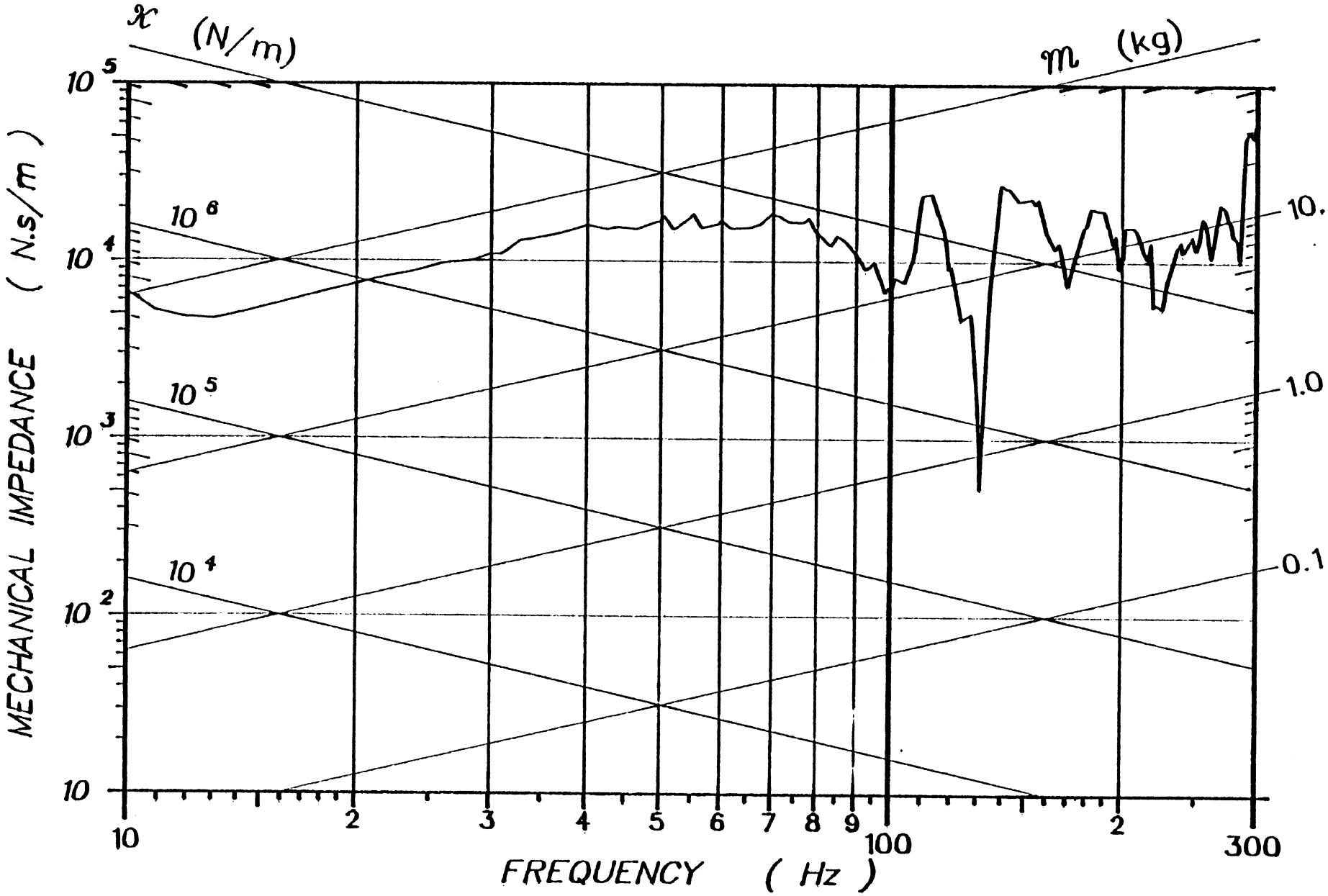


Figure 98: T12 (R–L)

79H205

2.6.4. I-S Transfer Impedance Curves

The final group of transfer impedance curves are those in the I-S direction; thus, figures 99-103 are those for the head, figures 104-108 are for T1 and figures 109-113 are those for T12.

These impedances are the most important group since the output (acceleration) is in the same direction as the input (impact force). Several observations may be made about these curves.

First of all, the impedances are lower for the head, and higher for T12, while T1 impedances are somewhere in between. This supports earlier observations that the impact energies are dissipated by the system, so that it effects mostly the head and to lesser extent the lower thoracic vertebrae.

The second observation is that, at low frequencies below 30 Hz, the system acts like viscous damping (dash pot) with constant impedance which is independant of the frequency. This observation is important since it suggest that any realistic test device must have elements which dissipates energy without returning this energy back to the system, as is the action of a spring.

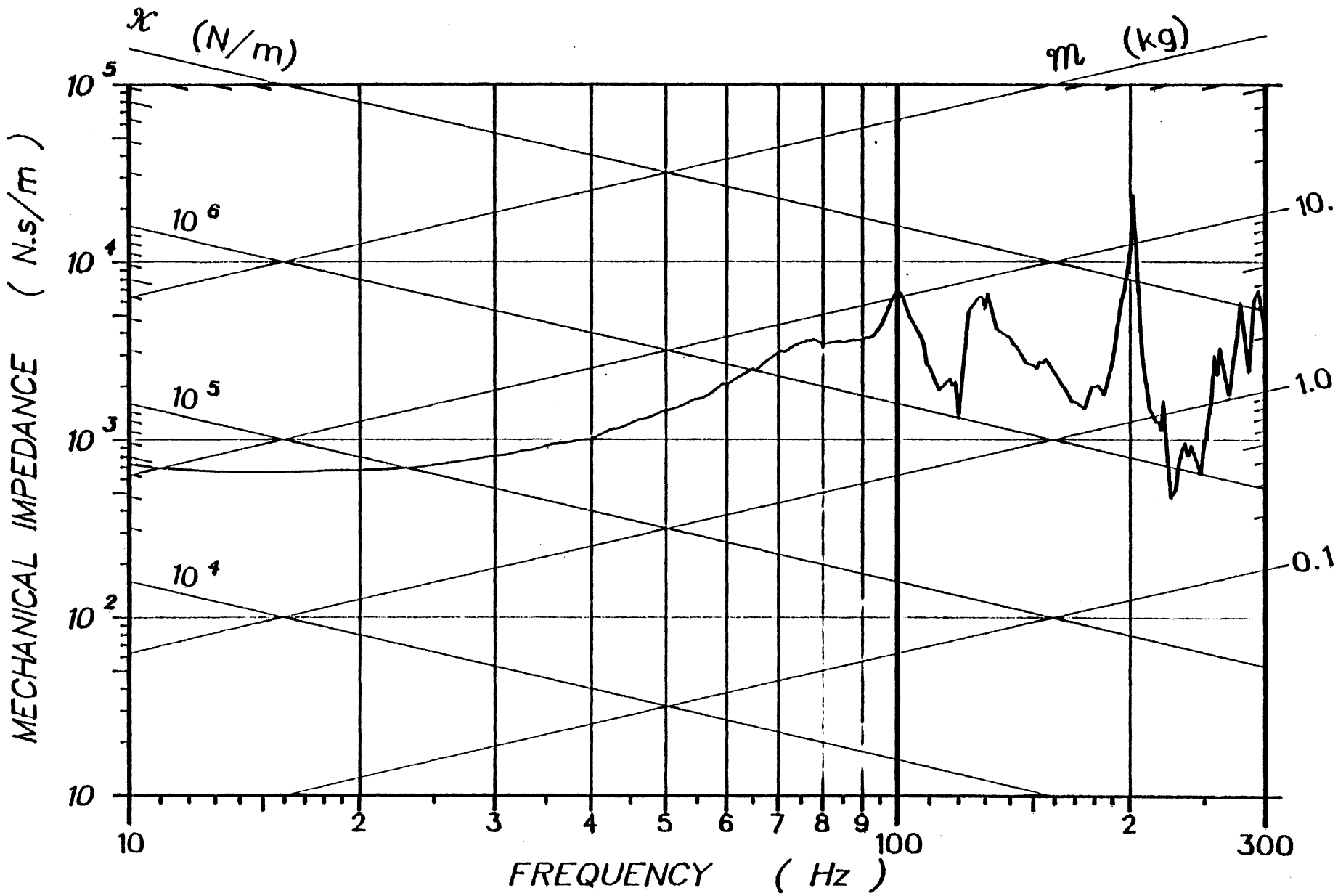


Figure 99: HEAD (I–S)

79H201

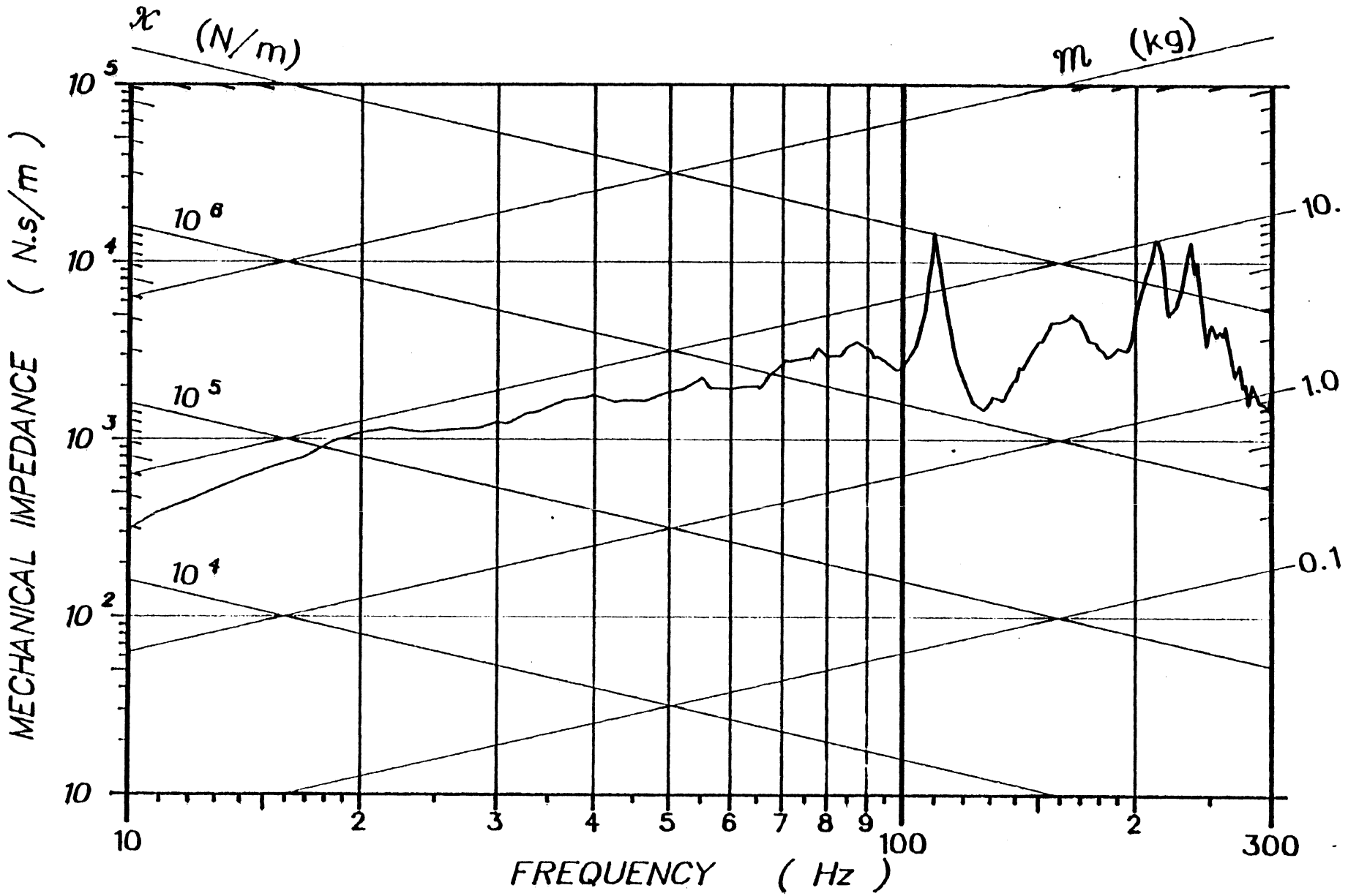


Figure 100: HEAD (I–S)

79H202

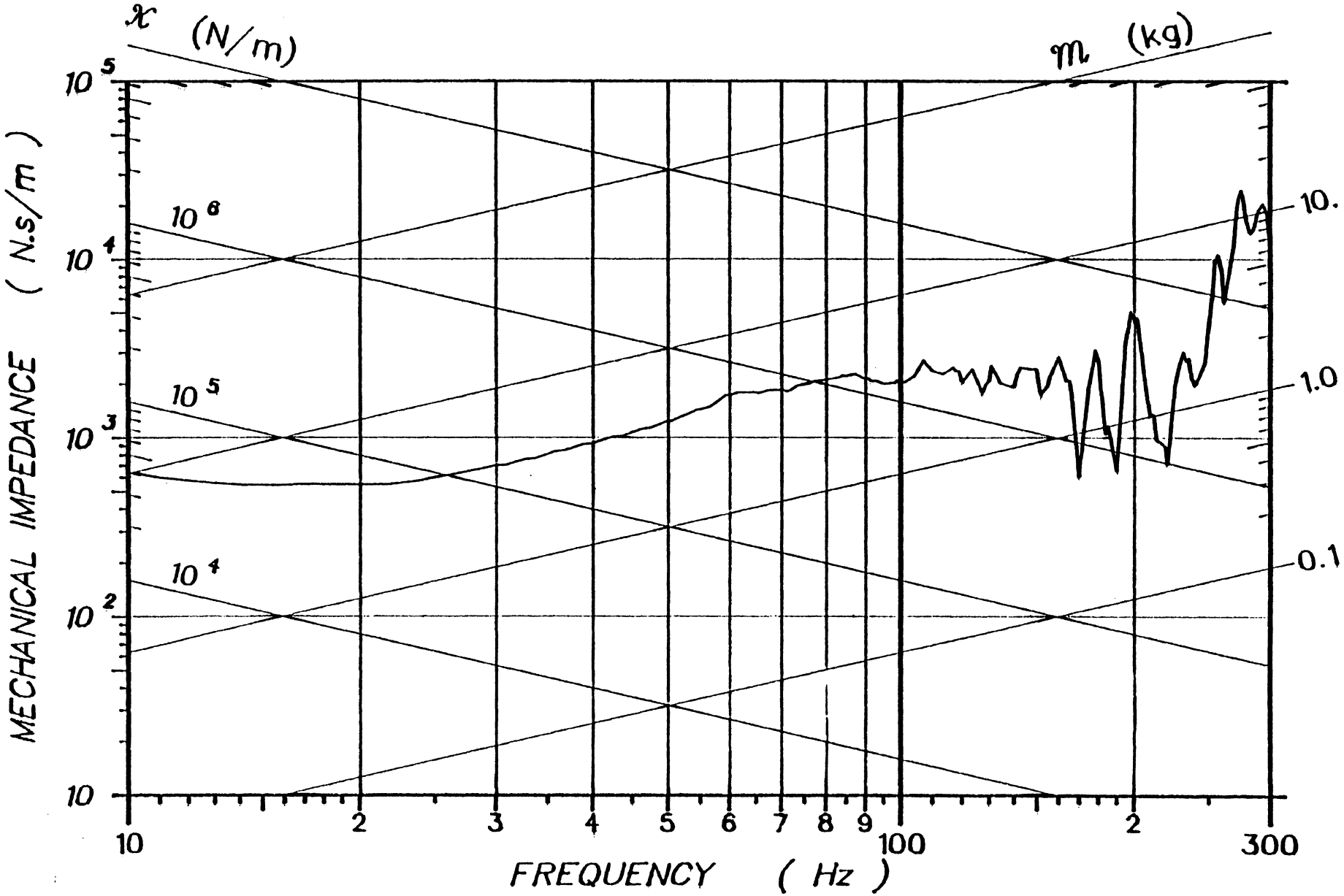


Figure 101: HEAD (I–S)

79H203

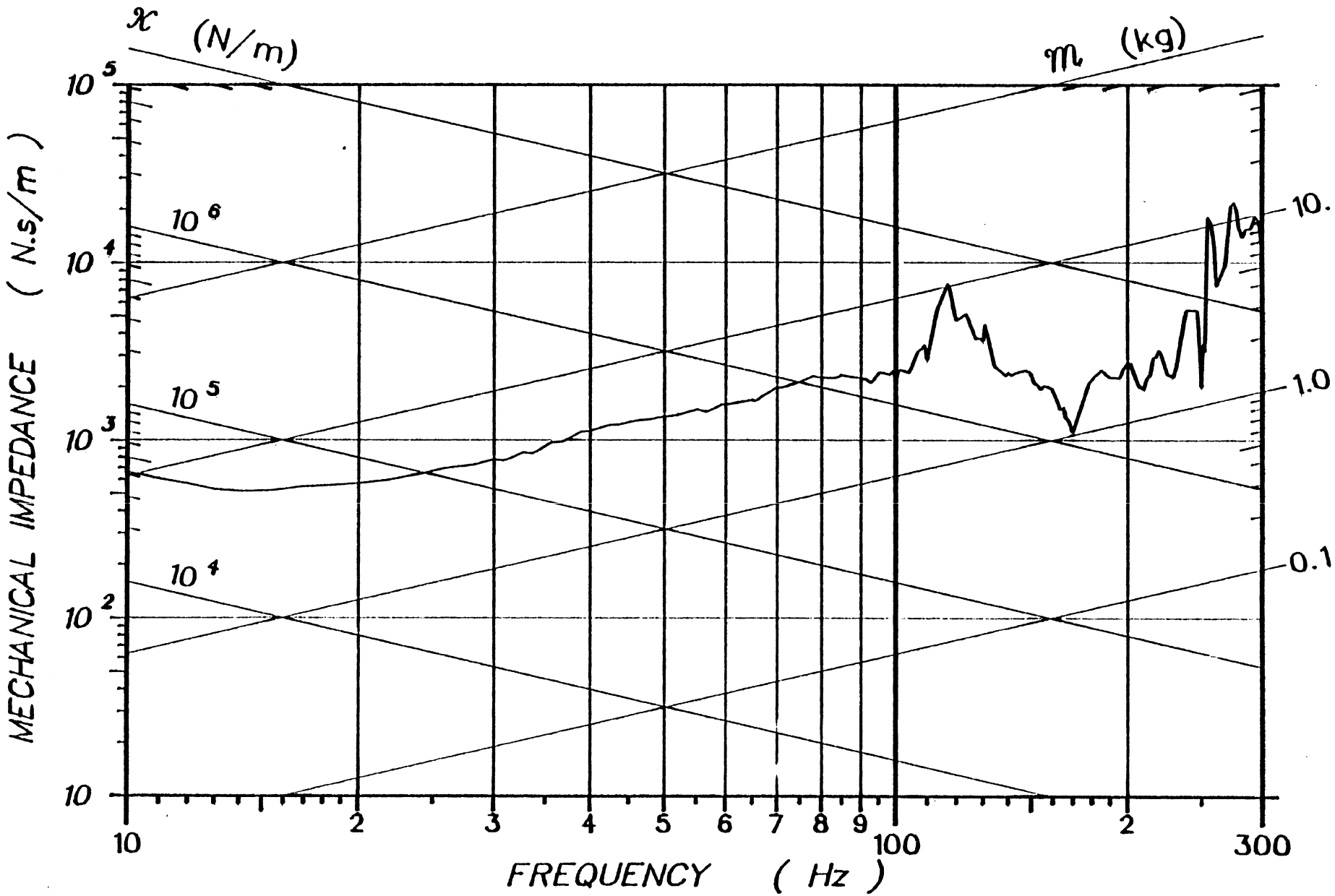


Figure 102: HEAD (I–S)

79H204

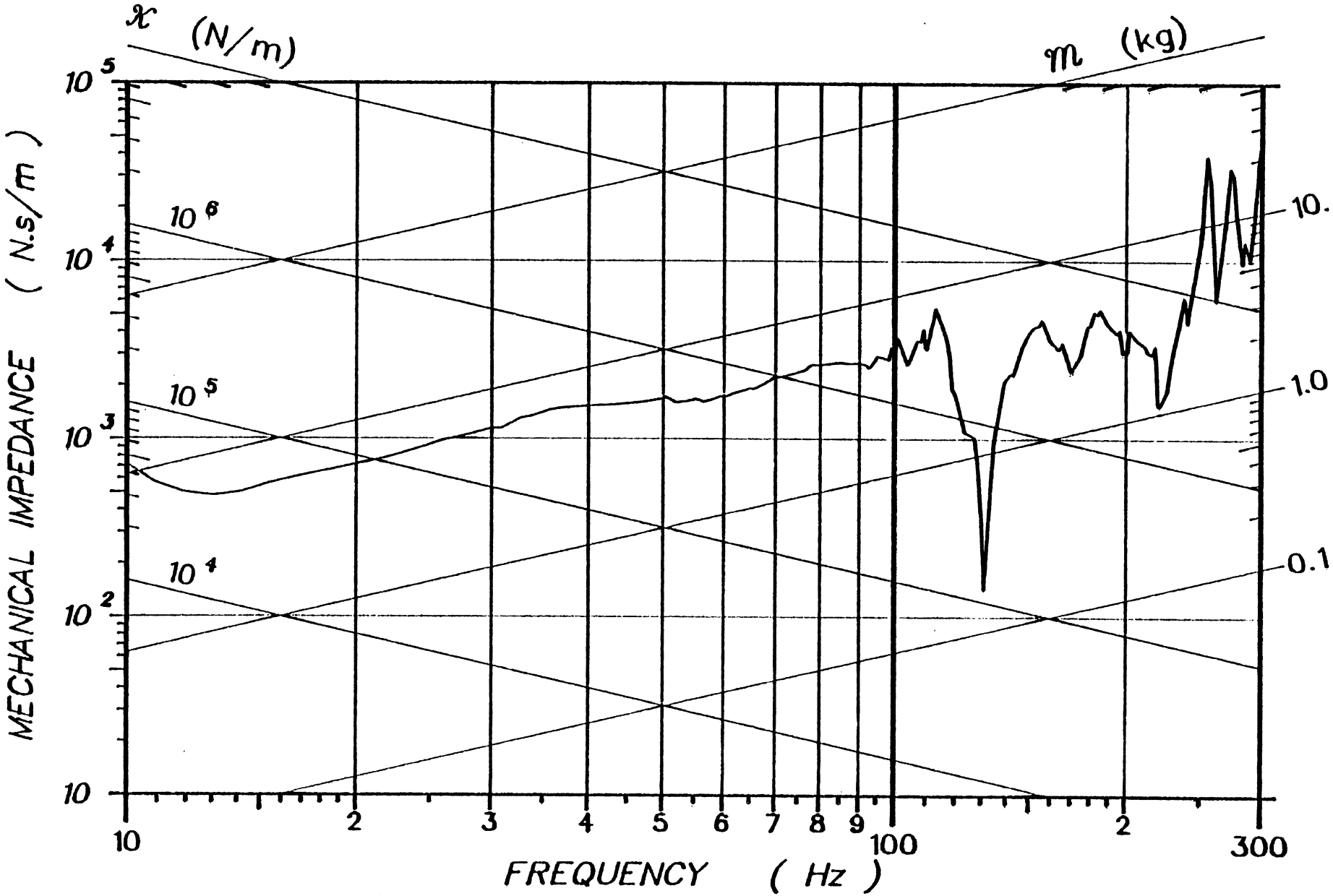


Figure 103: HEAD (I—S)

79H205

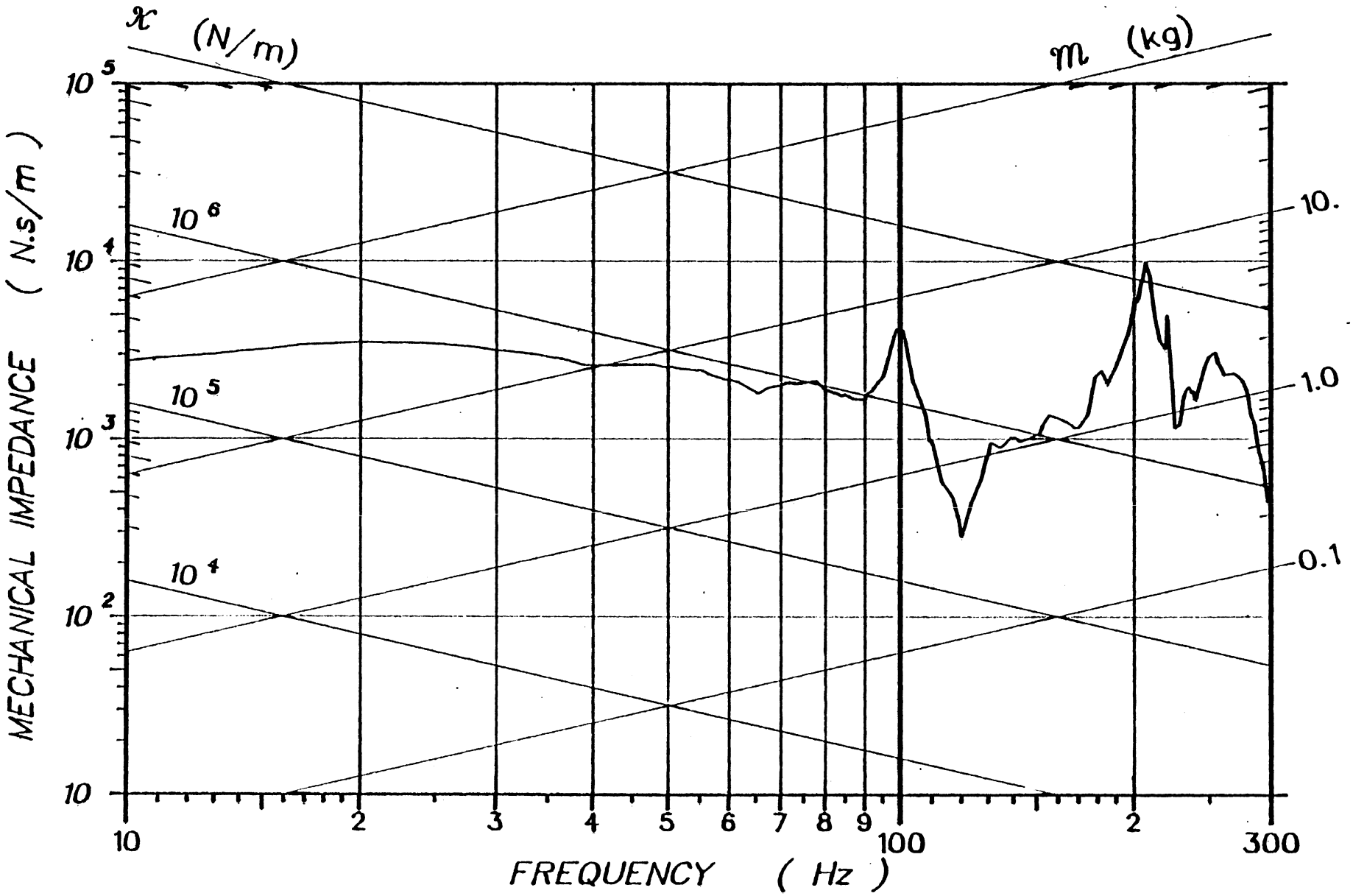


Figure 104: T1 (I–S)

79H201

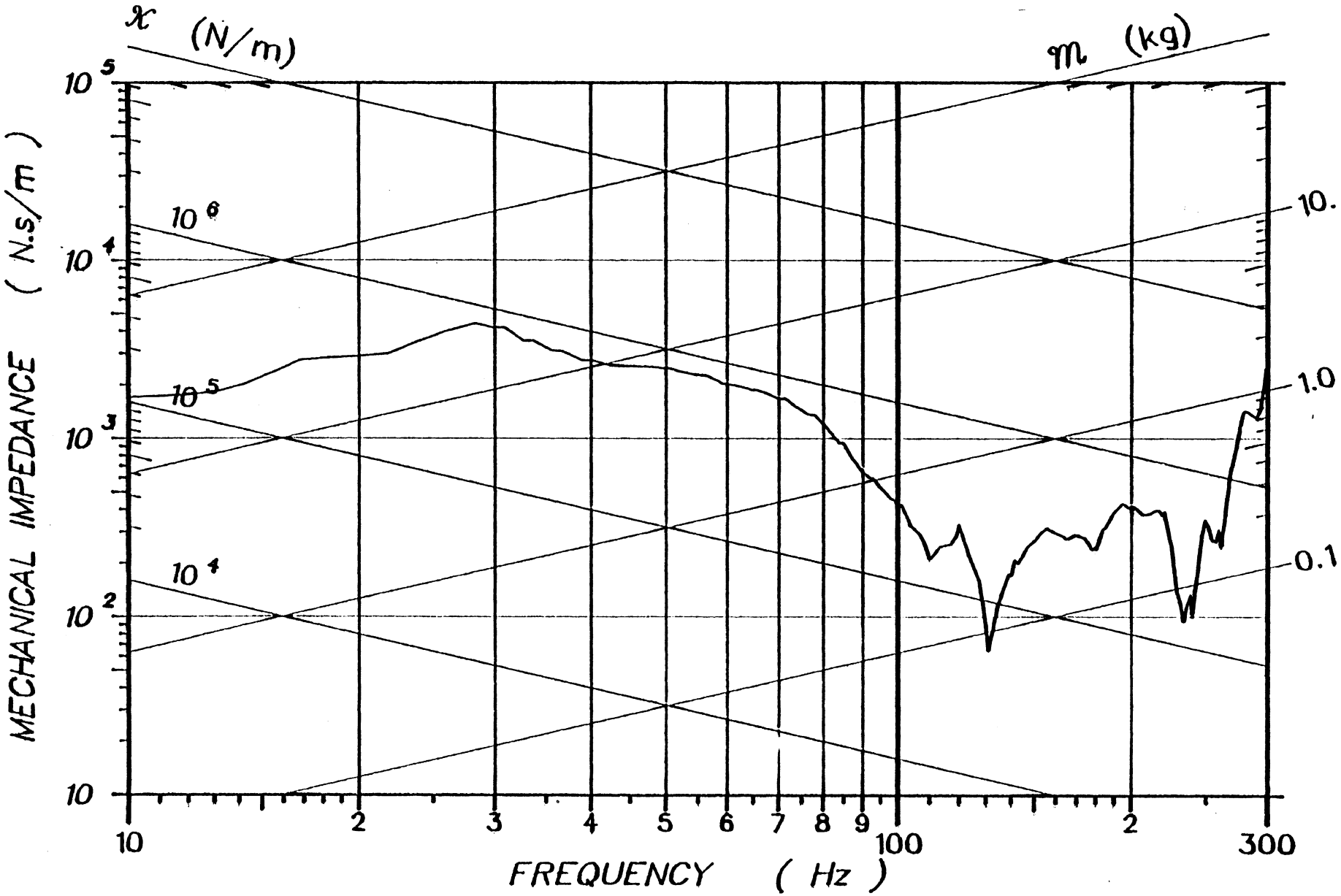


Figure 105: T1 (I–S)

79H202

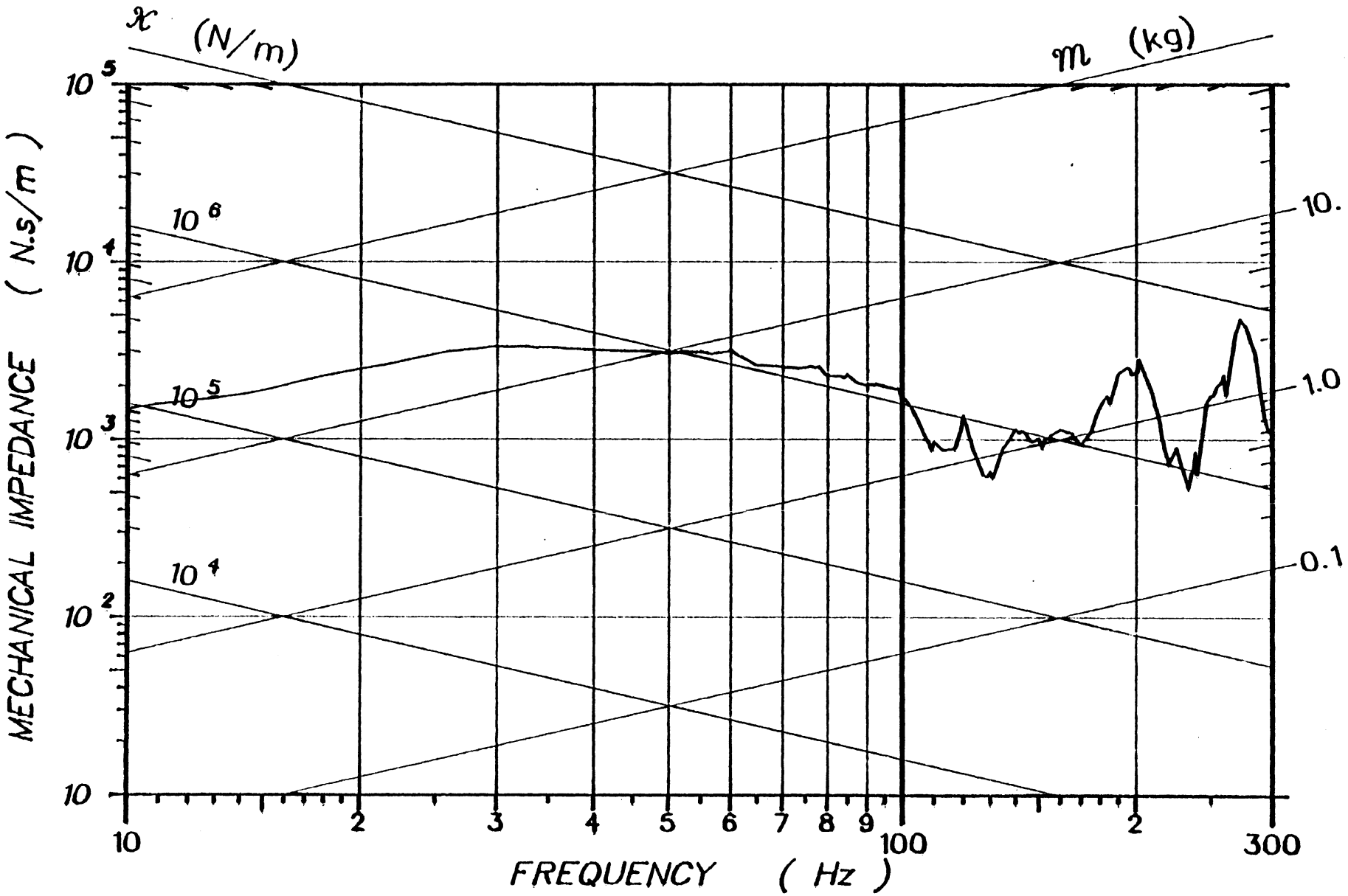


Figure 106: T1 (I–S)

79H203

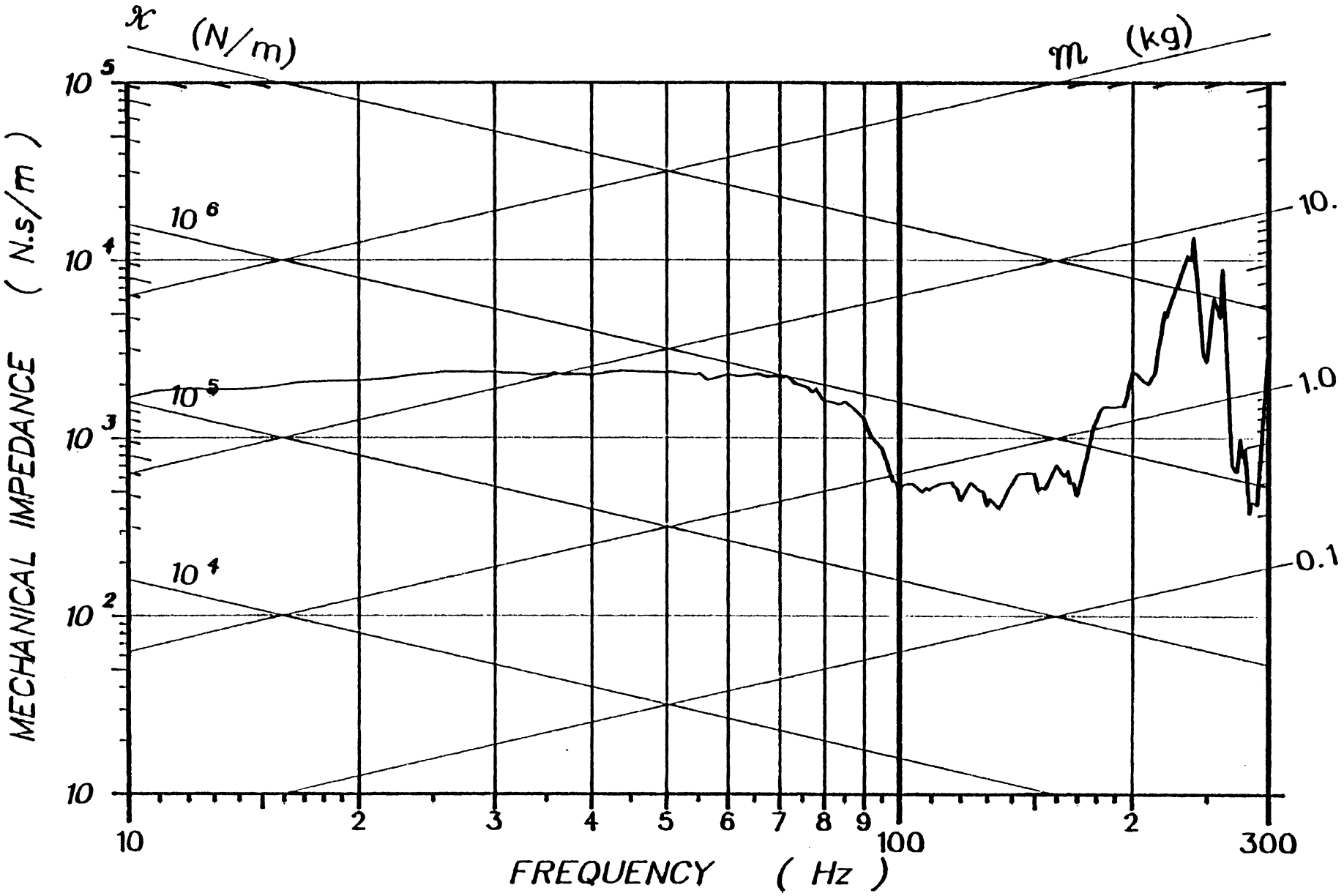


Figure 107: T1 (I–S)

79H204

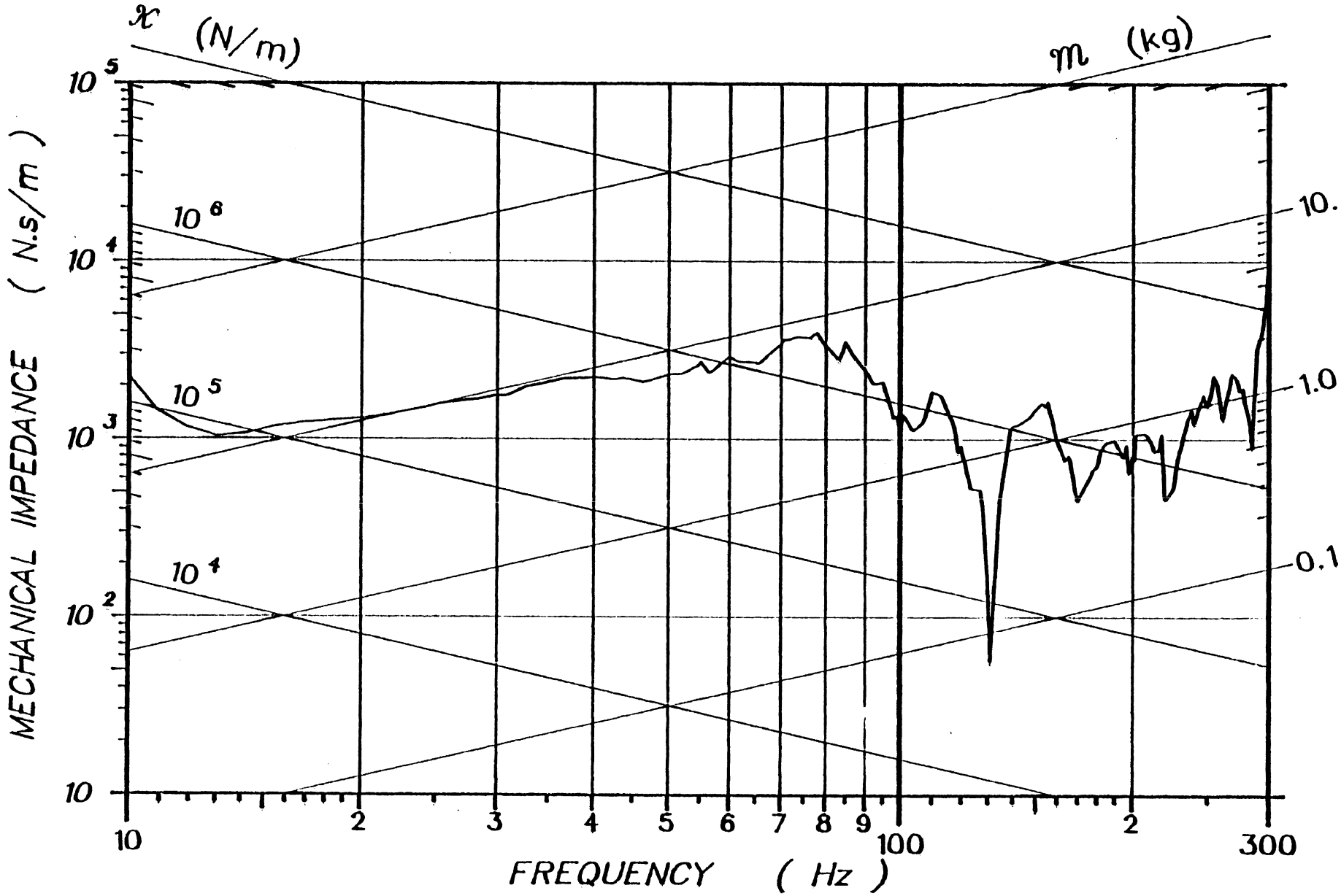


Figure 108: T1 (I–S)

79H205

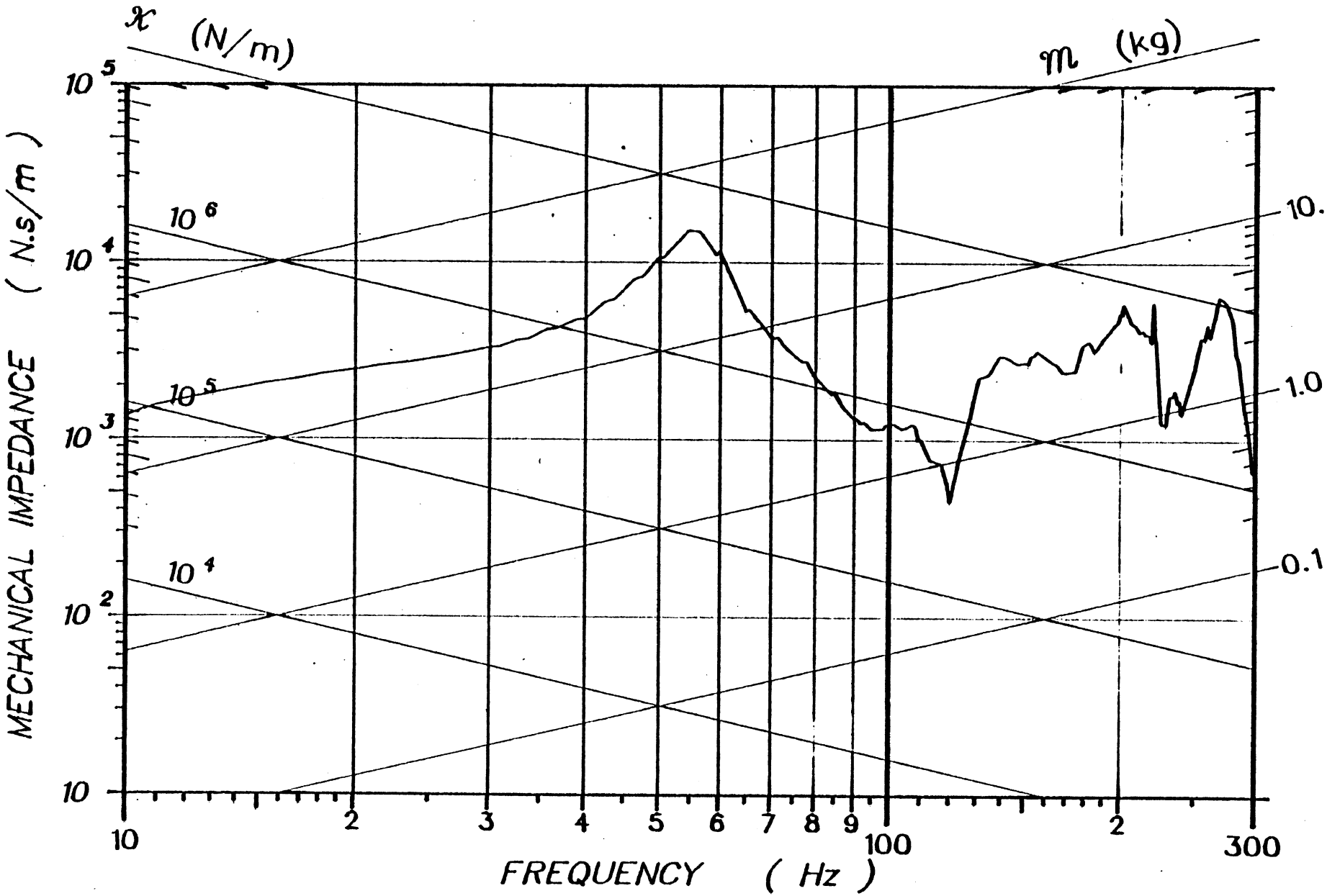


Figure 109: T12 (I–S)

79H201

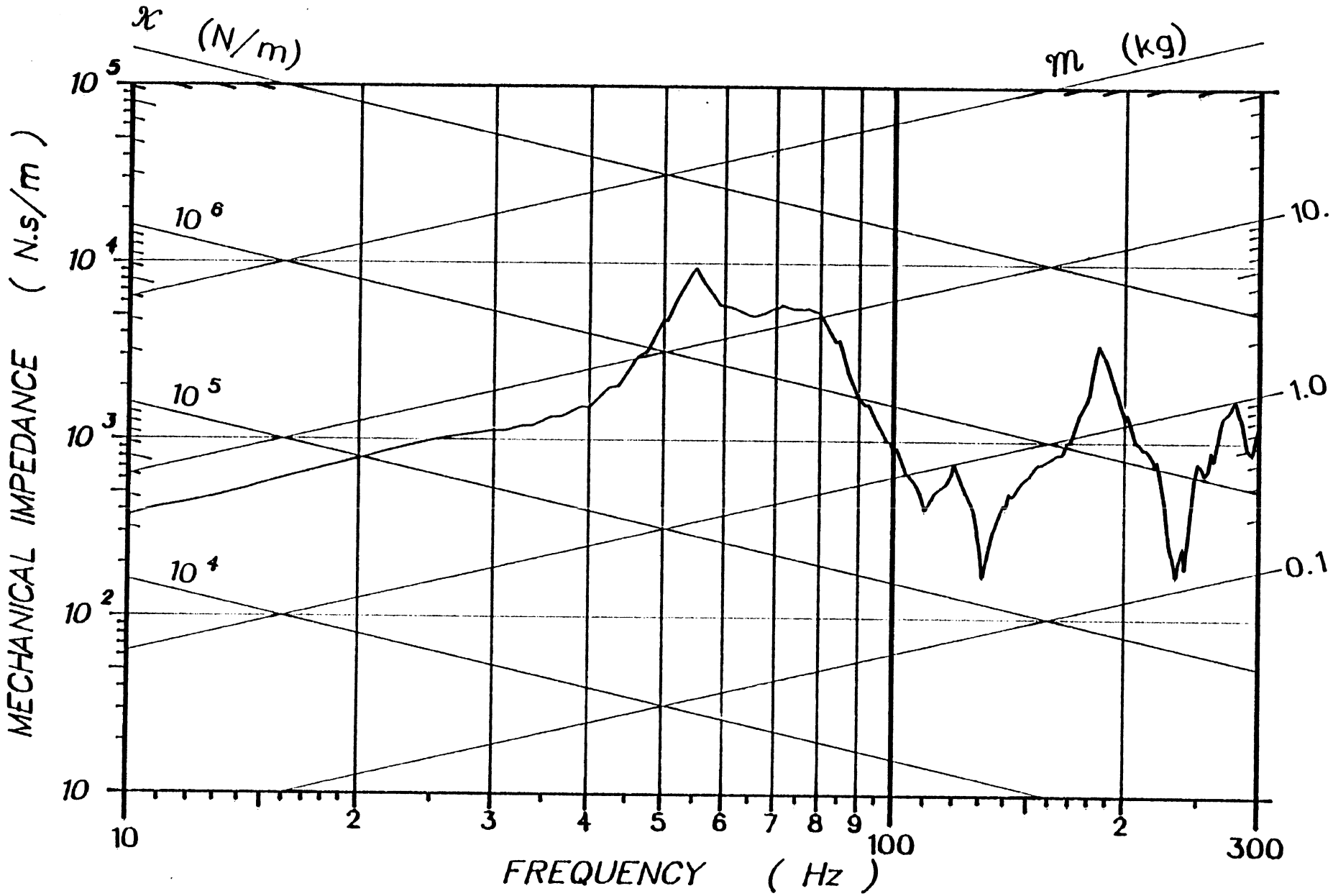


Figure 110: T12 (I–S)

79H202

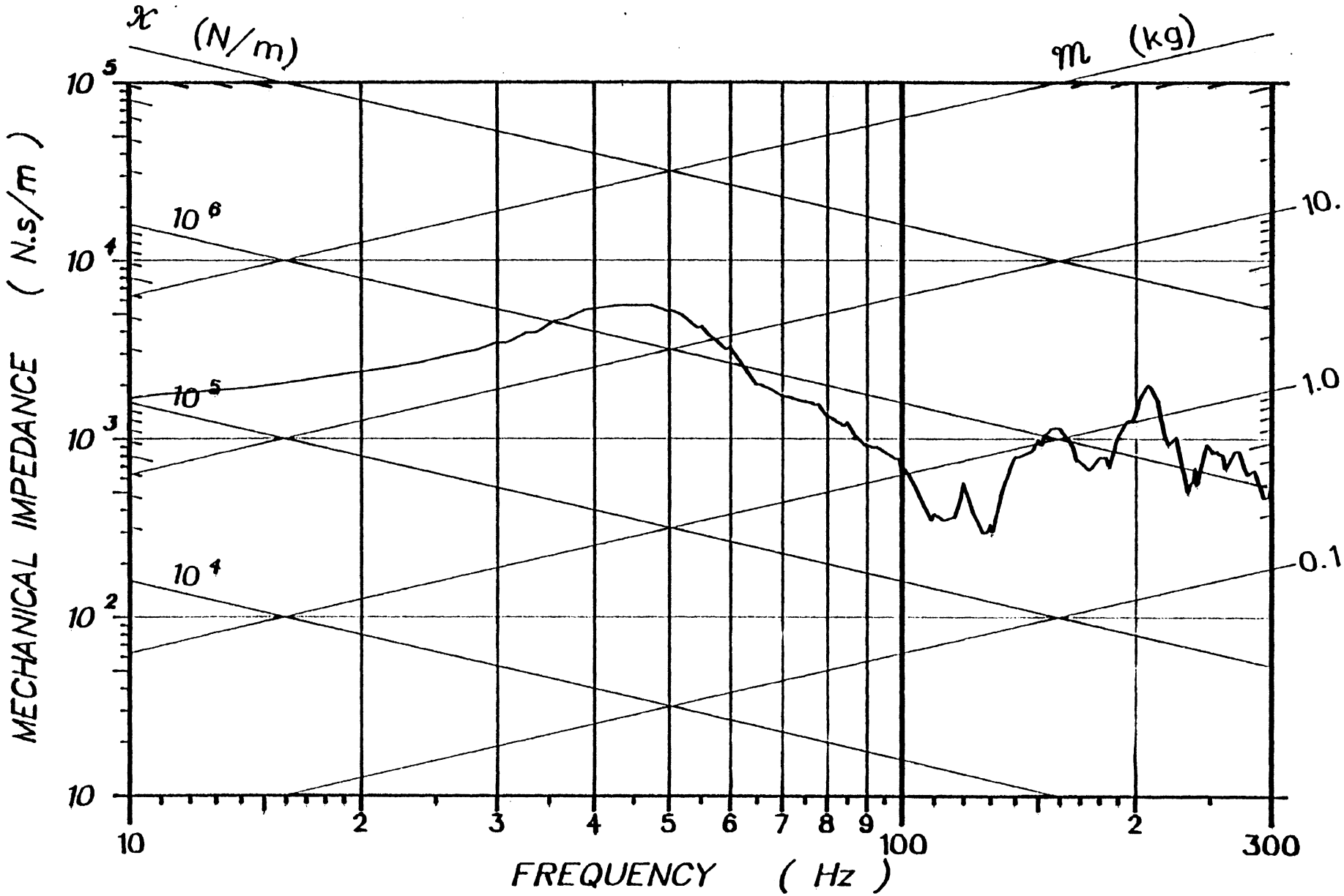


Figure 111: T12 (I–S)

79H203

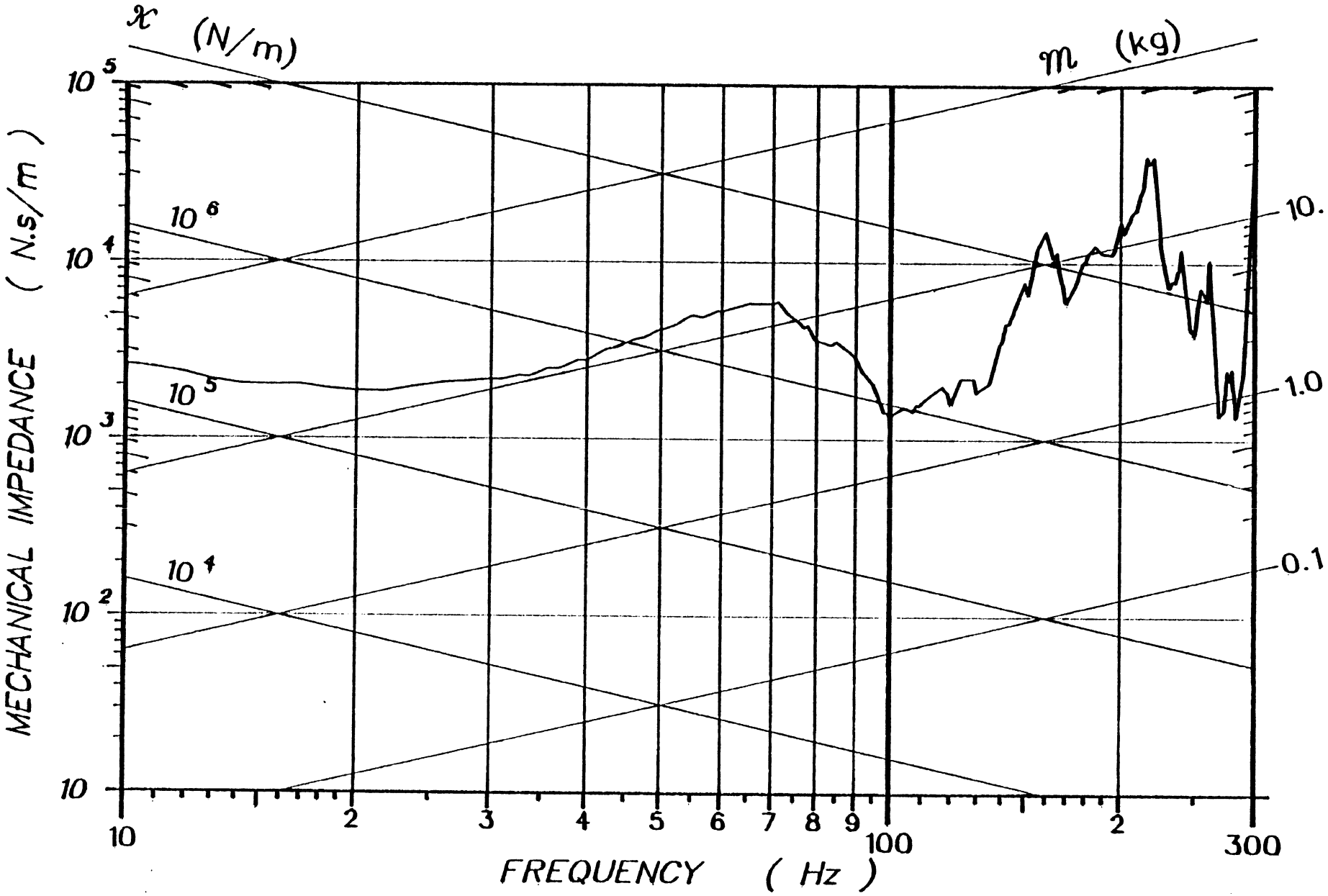


Figure 112: T12 (I–S)

79H204

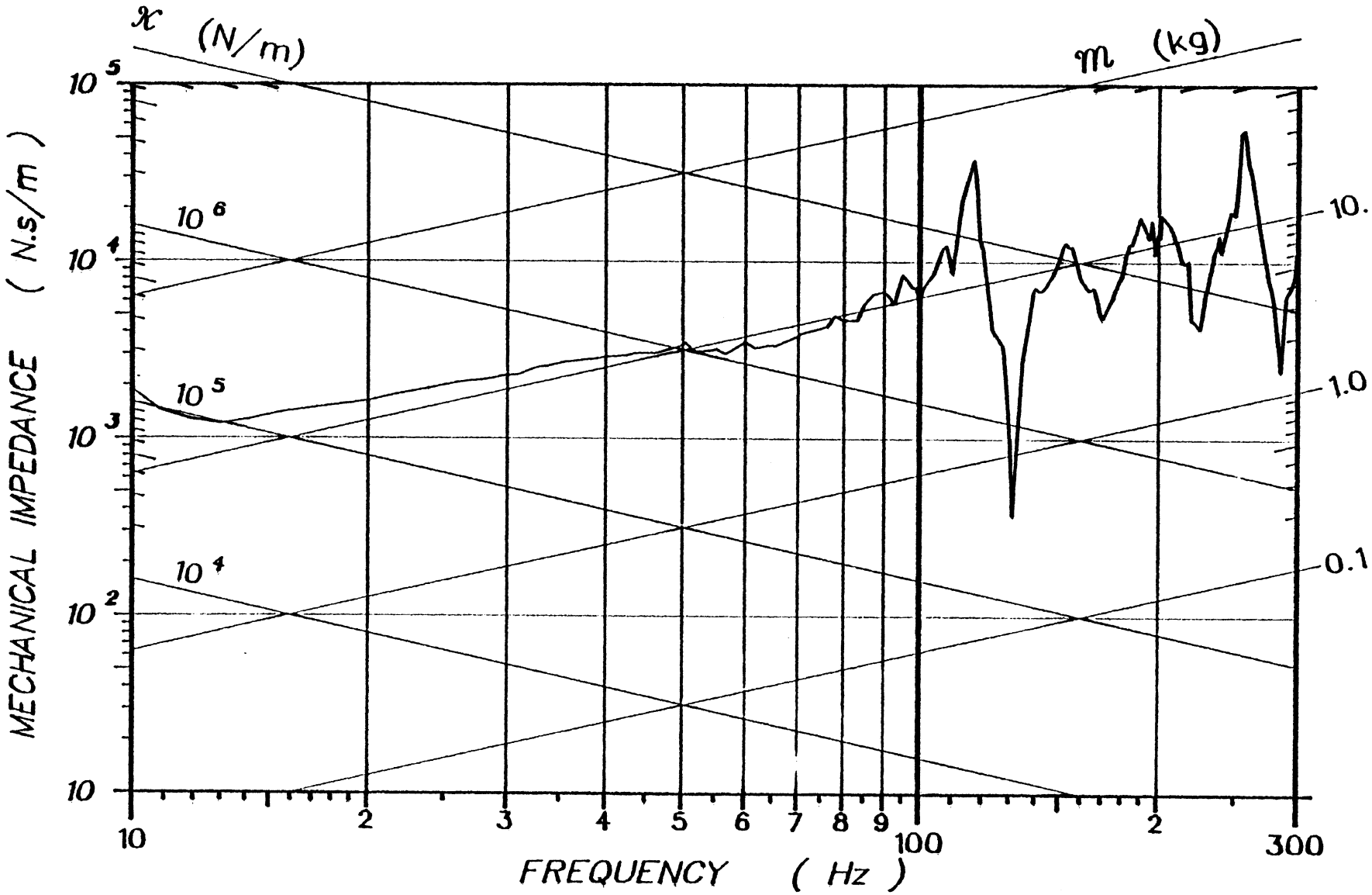


Figure 113: T12 (I–S)

79H205

2.7. REQUIREMENTS FOR HELMET TEST DEVICE

The test results presented in the previous sections are those from 5 cadaver tests, a sample too small to draw definite general conclusions about human response to S-I impact. There were, however, some observations which were consistently made regardless of the test being considered or the parameter used in making the observation.

The cadaver tests have consistently suggested that any realistic helmet impact test system must respond to S-I impacts as follows:

a) the motion of the head should be primarily in the S-I direction;

b) the head rotation relative to the neck should be primarily about an L-R axis

c) the motion of the head should be absorbed by non-conservative elements which dissipate the energy.

3. PHASE III - PROPOSAL OF IMPACT TEST SYSTEMS

3.1. BACKGROUND

The requirements of Phase III of this project may be summarized as follows: 1) propose three levels of impact test systems sophistication, 2) develop an analytical model for each proposed system, and 3) submit sketches and narrative description of each system.

The design of these systems should incorporate results from cadaver testing in Phase II. Although this testing phase is not completed, there is enough data to suggest trends in response and to allow the formulation of a preliminary design criteria of the desired test system which is human-like in response.

The overlap between phases II and III is necessary to compensate for the unavoidable earlier delays. As testing in Phase II continues, the design requirements of Phase III will be modified to accommodate additional results from cadaver tests. With this in mind, three systems are proposed based on our preliminary findings.

3.2. PRELIMINARY TRENDS IN RESPONSE

Results from 3 tests (79H201, 202 and 203) indicate the following trends in the measured cadaver response to S-I impacts:

1) The motion (acceleration, velocity and displacement) is primarily but not exclusively in the direction of impact. Thus, while accelerations of the head, T1 and T12 are highest in the S-I direction, accelerations in the A-P and L-R directions are generated as well.

2) As the point of observation moves away from the point of impact (from the head, to T1 to T12,) the magnitudes of responses are reduced indicating some dissipation of energy through the head, neck and spine.

3) From impedance analysis of one test (79H201) in the frequency domain, the S-I and resultant responses at low frequencies is consistant with the masses of the head, head-neck and head-neck-upper spinal segments of the body. The A-P and L-R responses exhibit a dashpot-type behavior characteristic of a constant force-velocity relationship, independant of frequency. Results from tests 79H202 and 79H203 generally support this characterization; however, no statistically valid conclusion can be made with a sample of only 3 tests.

No attempt will be made at this stage to interpret or explain these results except to include them as Appendix 3.A.

3.3 DESIGN REQUIREMENTS

Any realistic impact test device should provide some kind of damping since preliminary results indicate that energy is being dissipated by the head-neck-spine system.

The test system should include, as a minimum, a headform where a helmet could be installed and a lower linkage representing the neck-torso. Kinematic response of the head and of the neck should be monitored along with the interaction forces at the head-neck junction.

Several levels of sophistications could be achieved by the selection of the headform, the design of the neck-torso structure and by the requirements imposed on the number and accuracy of measurements being made of the response. In all cases, the device must be capable of being positioned to receive blows at any location above a reference plane which is 2.5 above the head basic plane.

Most importantly, the measured response of the device to impact should match, as closely as possible, that of the living human under similar conditions. The emphasis placed on this human-like behavior underscores the difficulties in achieving such a device. Considerable effort will therefore be spent in tuning the constructed device to produce the desired response.

3.4. THREE PROPOSED SYSTEMS

In the following sections, three helmet impact test systems are proposed. These systems differ by the complexity of their construction and by the number of transducers used to monitor their impact response; however, they all have the same basic configuration: a headform, a "neck" structure and a mounting base. The use of commerically available components would reduce the cost of constructing the device, but would result in limited tuning flexibility. On the other hand, an extremely flexible design dictates that all components be designed, from the ground up, to produce the desired human-like response and to meet the design requirements spelled out earlier. The compromise system would incorporate some available components that would be moderately modified, with improved linkages between the various parts.

3.5 PROPOSED MINIMUM COST SYSTEM

In order to keep the cost at a minimum, existing dummy head and neck would be used. The mounting base would be adapted from a ball-and-socket vise that is commercially available. This system is sketched in Figure 114.

3.5.1. The Head

Part 572 ATD head would be used in this design. The vinyl skin would also be included to bring the head to the appropriate circumference and weight and to absorb some of the high frequency resonances that might be generated.

Instrumentation of the head would be limited to a triaxial (or equivalent) accelerometer mounted at the head center of gravity.

SIMPLE STRUCTURE, SIMPLE INSTRUMENTATION

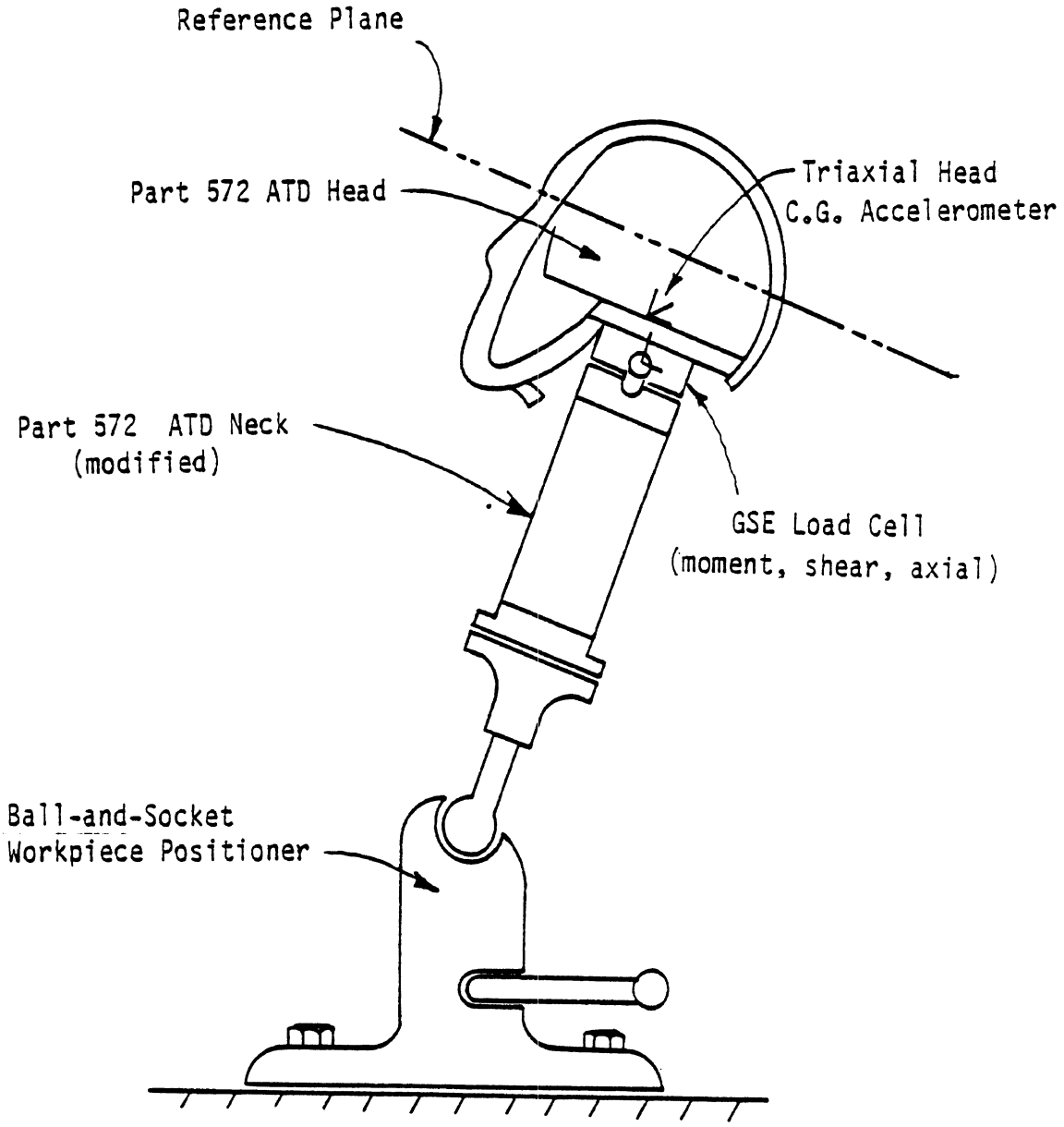


Figure 114 - "Minimum-Cost" Helmet Impact Test System

3.5.2. The Neck

The neck in this low-cost design would also be that of Part 572 ATD, because of its availability and simplicity. This neck a monolithic rubber casting, cylindrical in shape, with a flexible steel cable along its longitudinal axis, connected to two washers at the ends of the cylinder.

Connection between the neck and head would be designed around a special GSE load cell to measure the moment and shear and axial forces at the condyles. This interface would be similar to that used in the construction of the GM Hybrid III dummy.

3.5.3. The Base

The neck-head assembly would be mounted to the laboratory floor (or work table) through a heavy-duty ball-and-socket work positioner, such as Wilton's Pow-Rarm Work Positioner No. 302. This support would provide up to 1000-lb loads at 12 inches, and has a wide range of angular adjustments.

3.5.4. Advantages

This "minimum cost" system can be constructed without major effort in designing its components. The use of a standard anthropomorphic head ensures a human-like response of the head, at least under impact conditions similar to most of those encountered in frontal automobile crashes.

A standard dummy head has also the advantages of provisions for mounting a triaxial accelerometor at its center of gravity, a minimum instrumentation requirement.

The advantages of using an existing neck is the commercial availability of such a component, and its acceptance as an anthropomorphic surrogate of the human neck. The choice of this neck (as opposed to other available necks) is the simplicity of its construction, its ruggedness, and the possibility of specifying a longer neck if it became necessary.

The mounting base is a fairly common machine shop equipment used to position a work piece at various angles for machining. The ball joint vise, which has a quick release/clamp mechanism, allows the head-neck assembly (work piece) to be oriented in any position within a wide but limited range.

The proposed instrumentation package allows the monitoring of the head c.g. accelerations and of the reaction forces at the condyles. These same response measures are obtained in cadaver testing, so that tuning of the device be limited to matching head accelerations and condyles reactions.

3.5.5. Disadvantages

While this design is appealing because of its simplicity and low cost, it presents three potential sources of problems: 1) the mounting base, 2) the neck response, and 3) the monitoring of response.

The problem with a ball-and-socket vise is that it cannot be clamped down tightly, to eliminate possible slippage during impact, unless its size is increased beyound practical limits. Another problem is the limit of ranges of adjustment because of the design of the opening in the socket. A third problem is that, once the vise is released from a given position, it would be very difficult to repeat the same position for another test. Finally, the location

and direction of impact would be very difficult to document by the angular position of the top plate, a desirable feature for conducting parametric studies.

The second potential source of problems is the neck itself, primarily because of the absence of damping. While hysteresis exists in the Part 572 dummy neck, it is not sufficiently high to account for the documented energy absorption characteristics of the head-neck structure. In addition, the effects of the upper thorax on the head response is not accounted for in this design, unless those effects are incorporated in the neck by designing an unrealistically long neck. Tuning the device to produce human-like response would be limited to adjustments of the neck length, which may not critically effect the response.

The third potential area of trouble has to do with the monitoring of response. While head accelerations at the c.g. have been associated with injury, other paramenters such as angular accelerations of the head or the kinematics and reactions at the C7-T1 connection may be just as important, and may even be the critical factors in determining the injury potential in S-I impacts. This design does not provide for these measurements to be used if and when future testing so indicates.

3.6. PROPOSED BEST POSSIBLE SYSTEM

The "best" system that could possibly be designed is one that eliminates the disadvantages of the low-cost system while retaining most of its desirable features. Improvements in the design would therefore: 1) provide for a rigid and repeatable mounting device, 2) increase the neck damping and incorporate the effects of the non-rigid thoracic sub-structure, and 3) monitor as many response variables as possible. The proposed "best" system is illustrated in the sketches of Figure 115 and is described below.

3.6.1. The Head

The Hybrid III dummy head is selected for this design because of its improved response to rigid impacts. A complete 3-D motion measurement package, such as the HSRI 3-3-3 or the WSU 3-2-2-2 nine-accelerometer arrangement would provide the 3-D kinematics of the head.

3.6.2. The Neck

The link between the head and the adjustable rigid mounting base would consist of two elements which simulate the cervical and the upper thoracic portions of the spine.

To simulate the cervical spine, the GM Hybrid III neck would be used. This neck is a one-piece, flexible component with biomechanical bending and damping responses in both flexion and extension. Three rigid aluminum vertebral elements are molded in butyl-elastomer to form the neck structure.

The purpose of the second element is to simulate the energy absorbing characteristics of the upper thoracic structure and to provide additional control over the axial and bending response of the neck. This element would amount to

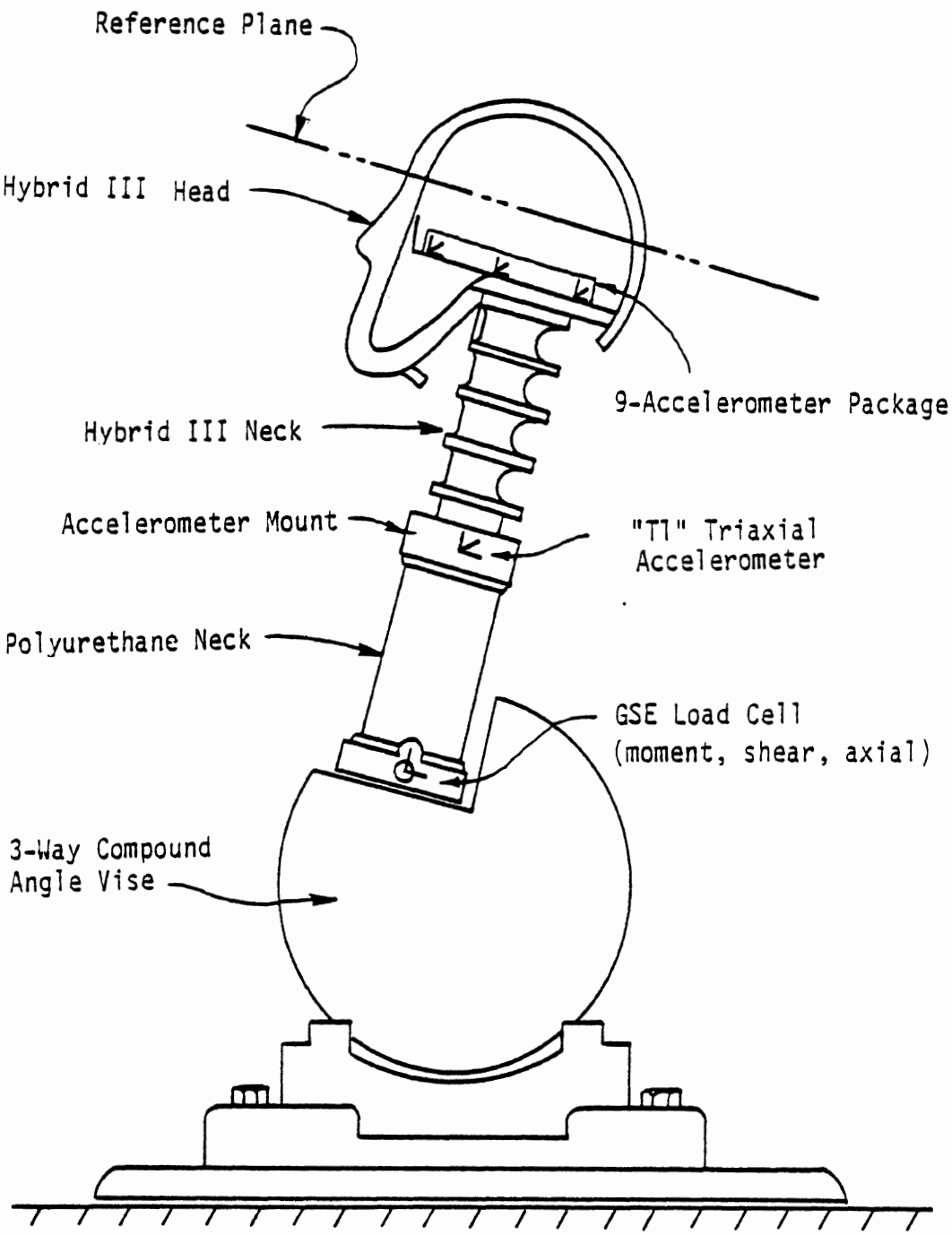


Figure 115 - "Best Possible" Helmet Impact Test System

a cylindrical casting from polyurethane material. The stiffness (or softness) of this "sub-neck" would be controlled specifying the dimensions and the chemical composition. Two end plates would be used to provide attachment surfaces for the neck and the mounting base.

The interface between the upper and lower necks would be instrumented with a triaxial accelerometer to monitor the kinematics of a point simulating the C7 or T1 vertebra. The GSE load cell would be utilized to monitor the moment and axial and shear forces transmitted to the bottom of the head-neck assembly, since reactions at the condyles (all 6 of them) would be calculated from the 3-D rigid body dynamics of the head.

3.6.3. The Base

The "best" improvement that could be made over the low-cost design is to use a more sophisticated work piece positioner. Therefore, the mounting base in this design is proposed to be a heavy duty universal angle vise, such as Wesson's No. 1 VR. This vise maintains set up accuracy and swings through 360 degrees in the horizontal plane and 90 degrees in the vertical plane. The body and cradle can be locked to take up to 2000 lb loading.

3.6.4. Advantages

This system eliminates most of the problems associated with the low-cost design proposed in section 3.5, while retaining those desirable features. Thus, the use of existing components is maintained wherever possible to keep the cost of the system at a reasonable level. The performance of the device is enhanced by incorporating additional elements that have to be custom-made.

By selecting a more rugged, easily adjustable and very flexible mounting base, the problems of ball-joint support is eliminated. It would be possible to design and construct a new mounting base to increase the range of positioning, however, the additional cost of this effort cannot be justified when compared to the additional benefits.

The same argument holds true for using an existing dummy head which is human-like in its biomechanical response and its anthropometry. Any attempt to improve the response is a major effort which involves additional data that is not currently available. The addition of a 3-D nine-accelerometer package to the standard dummy head enhances the response monitoring capability.

The major advantage of this design is, however, in the concept of simulating both neck and thoracic structures and in monitoring the motion of a point equivalent to first thoracic vertebra, T1. The specified upper neck (Hybrid III neck) has flexion and extension responses that were validated against those of human volunteers and cadavers. In the construction of this neck, the butyl elastomer was chosen for its high damping characteristic in order to approximate the biomechanical hysteresis requirements.

Since this neck was designed primarily for sagittal bending, lateral bending as may occur in L-R impacts was not taken into account. Furthermore, this neck was not intended to be mounted on a rigid platform, but rather on the thorax of a dummy. Finally, the axial deformation (e.g. stretch) of this neck is not allowed to a reasonable extent. Because of all these reasons, a substructure to which the head-neck is mounted would be included in this design.

The use of a polyurethane casting allows flexibility in tuning this device to produce the closest match between its response and that of the available S-I, A-P and L-R human impact response.

3.6.5. Disadvantages

One of the problems in the proposed "best" possible helmet impact test device is the practicality of such a design. While the proposed system can be constructed, it is very difficult to achieve an exact match between human and device response. This is true especially for the neck axial displacement which approaches 10 mm when the thorax is included in the impact. Such displacement is necessary to produce the appropriate accelerations observed at the head level.

The proposed device will have some compliance in the axial direction, but because of the limit on the amount of displacement which would occur, it is expected that the tuning process will be time-consuming, a major disadvantage.

The primary disadvantage is, however, in the amount of measured data to be processed. By using the 9-accelerometer package, a data analysis program would be required to extract the angular and translational acceleration components in the A-P, L-R and S-I direction. This would be a burden that slows down the process of evaluating protective helmets, and may require the availability of a large digital computer.

3.7. PROPOSED COMPROMISE SYSTEM

In proposing the "minimum cost" system in section 3.5, the goal was to provide improvements over the current headform specified in the ANSI standard, while keeping the cost to a minimum. For the "best possible" system proposed in section 3.6., the cost was not a major concern. The underlying philosophy in designing that system was the achievement of the "best device response" possible without resorting to outlandish schemes and mechanisms. While either of the proposed systems can be constructed, a more reasonable compromise between them has been conceived, and its selection is recommended.

The proposed compromise system, shown in the sketch of Figure 116 is mechanically more sophisticated than the "low-cost" version, so that a more human-like response can be obtained. However, this system does not include all of the instrumentation package specifiec in the "best" system, a reasonable sacrifice. This system is described below.

3.7.1. The Head

The head used in the compromise system is a direct carryover from the two other proposed systems. This head was retained because it represents the state-of-the-art knowledge of human head anthrompometry and biomechanical response. The precision-cast aluminum head and vinyl skin would be purchased from one of the dummy manufacturers.

Instrumentation inside the head would be limited to the standard triaxial accelerometers in the A-P, L-R and S-I directions at the head center of gravity.

SOPHISTICATED STRUCTURE, SIMPLE INSTRUMENTATION

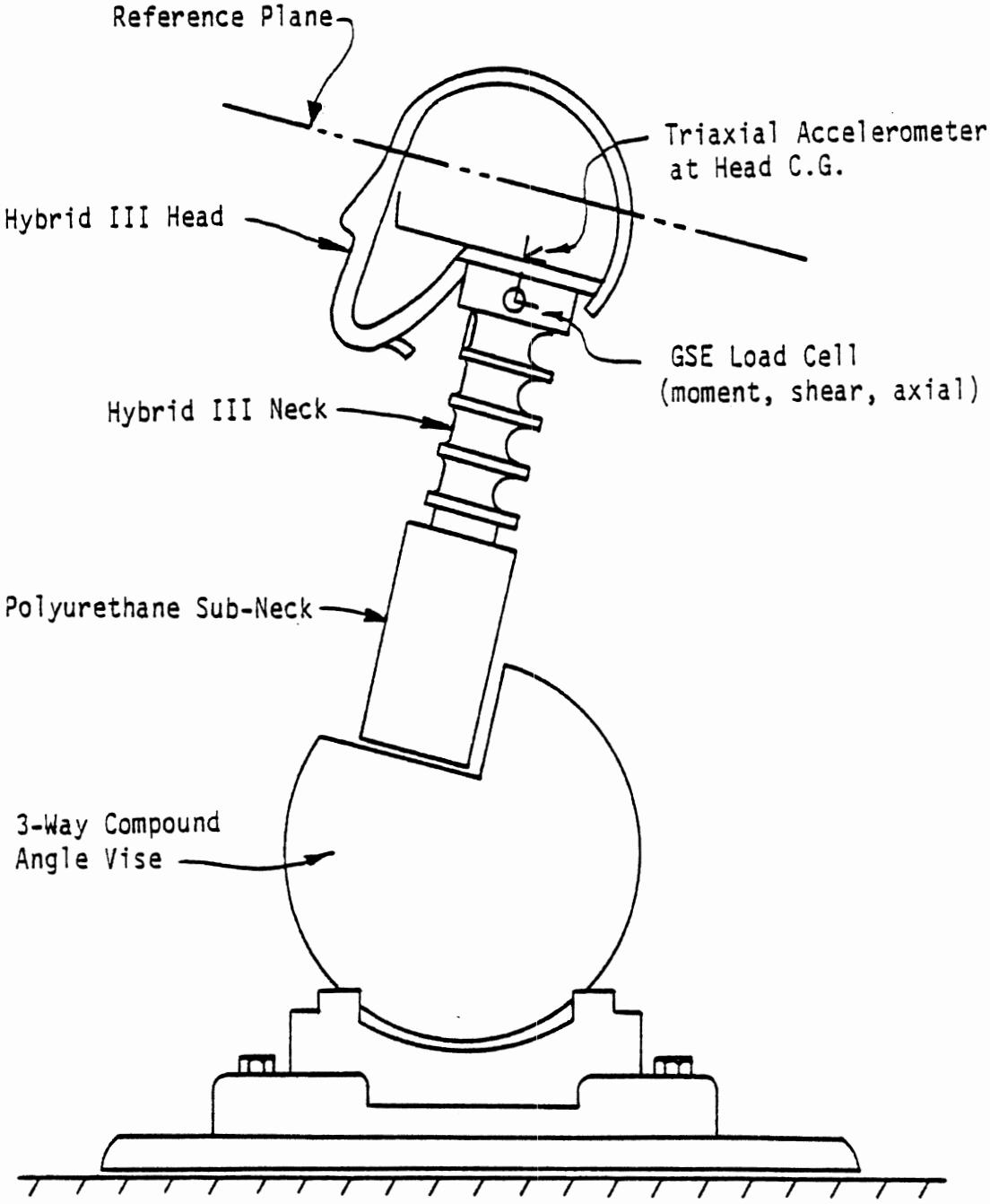


Figure 116 - Compromise Helmet Impact Test System

3.7.2. The Neck

In order to improve and control the response of the head, the chosen neck for the compromise system is carried over from the "best" system described earlier.

Thus, the neck will consist of the GMR Hybrid III neck, mounted on a custom-designed, cylindrical cast of polyurethane. The two "necks" will be interface with an aluminum, "washer" that is not instrumented.

The interface between the upper neck and the head will be indentical to the design of the Hybrid III, so that a "nodding" adjustment is allowed.

Instrumentation of the neck would be limited to measurement of the moment (about the L-R) axis, the S-I axial force and the L-R shear force, using a special load cell designed for this purpose and manufactured by GSE.

3.7.3. The Base

The base specified in this system is the same one proposed in section 3.6.3. for the "best" system. This is a 3-way, compound vise that may be purchased at a relatively low price from amachine shop supply company, or may be obtained from a shop as surplus equipment.

3.7.4. Advantages

When the other two systems are considered, it becomes apparent that the most desirable features are: 1) reasonable effort and cost of development and construction, 2) a controllable neck design that produces a human-like head response, and 3) simplicity of usage in terms of actual testing and data processing.

The proposed compromise design features all these advantages without design specifications, response validation or post-test data processing requirements. The reader is referred to sections 3.5.4. and 3.6.4. for a discussion of the advantages of the least and the most sophisticated system, most of which are offered in the proposed compromise system.

3.7.5. Disadvantages

Very few drawbacks can be pointed out in this compromise system. The most prominent is the elimination of the complete 3-D head motion measurement package, limiting the head response monitoring to translational accelerations at the c.g. (no angular acc.), and the reactions at the condyles to 1 moment (instead of 3) and 2 forces (instead of 3).

As far as the neck/torso is concerned, the use of viscoelastic material (butyl elastomer in the upper neck and polyurethane in the lower neck) is expected to provide some damping; however, this damping may not be sufficient to simulate faithfully the energy absorbing capacity of the human neck and torso. While this limitation is a disadvanyage, it may not be possible to increase the damping without using externally mounted dampers (viscous or friction-type), an option which was initially considered but later dropped.

3.8. ANALYTICAL MODELING

The purpose of using an analytical model for each of the 3 proposed systems is twofold: 1) to determine which system is most likely to produce the desired human-like response, and 2) to use the model to define the mechanical characteristics of the various components in the candidate system, such as the S-I compliance.

In the early stages of this project, it was felt that a sophisticated 3-D crash victim simulater, namely the Calspan 3-D CVS, would be the ideal tool for this purpose. As experience was gained in dealing with this CVS, it became apparent that the model does not have the flexibility required for configuration changes or impact force specification. In addition, the structure of the program and the documentation which accompanies the model is hardly user-oriented.

The alternative is to use a more user-oriented simulator which has both configuration flexibility and allowance for impact force specification. Such advantages are offered in the MVMA-2D crash victim simulator, at the cost of planar motion limitation. Since most of the data that will be used in comparing response is in the saggittal plane, it may be argued that prediction of the off-plane motion cannot be completely validated. Other advantages of the MVMA-2D CVS include its immediate availability and access to this project, and the great success with which it has been able to simulate a car occupant response under various impact conditions.

3.8.1. Simulated Systems

Since the three proposed mechanical devices differ primarily in the mounting interface, there would be only two systems to be simulated. The first one, which consists of the Hybrid III head and neck, is mounted rigidly to the "floor" through a sub-neck which introduces additional axial and bending compliance to the head-neck assembly.

The first model (without a sub-neck element) will not be allowed to have axial compliance, and therefore cannot simulate this characteristic of the actual human head-neck-torso. Any modeling artifice to include such an element amounts to modeling of the second mechanical system which does include a physical equivalent. This would then be redundant effort and, therefore, only one mathematical model will be conceived.

3.8.2. Proposed Model

The model is to be used as a tool for defining the mechanical characteristics of the actual human upper thoracic structure, so that an equivalent component may be incorporated in the test device.

The model consists then of a head (with a mass and a moment of inertia), a neck which has bending characteristics (predefined as those of the Hybrid III neck) with no axial deformation allowed, and a general translational and rotational element representing the unknown upper thoracic complex.

Initial guesses have to be made as to these unknown characteristics. This will be based primarily on T1 and T12 thoracic vertebrae responses, obtained from cadaver testing, either directly from time-histories, or by interpretation of mechanical impedance data.

To exercise the model, the lower point of the sub-neck element would be fixed, while impact force is specified at some point on the head. The motion of the head would then be predicted by the model and compared to monitored responses from actual cadaver tests. To improve the match between predicted and actual response, the translational and rotational characteristics would be modified.

Once a satisfactory match is obtained, it would be reasonable to assume that the values used in the model to describe the mechanical properties of the "sub-neck" approximate those of the actual human structure, and that any sub-neck components to be incorporated in the test device should have equivalent properties.

3.8.3. Alternatives to Modeling

In principal, modeling is an attractive method of simulating a physical system. This effort necessarily involves abstraction of the physical characteristics into mathematical elements. Once the model is refined and tuned, the reverse process is applied to construct a physical structure from the mathematical abstraction. Since both these processes involves approximations, the final outcome (constructed device) may or may not follow the requirements indicated by the model.

Additionally, initial guesses have to be made as to the mechanical properties of the unknown element, namely, those of the proposed sub-neck. These guesses must be based on impedance data and/or response time-histories. It may be argued, then, that going through an analytical model to refine these estimates will produce further estimates and no more.

Since the only component, over which there is control, is the polyurethane sub-neck, and since approximate compliance characteristics have to be obtained from the actual response data, an alternative to the design-by-modeling approach is to built a polyurethane neck to meet these approximate characteristics, then tune the overall structure by conducting actual tests. This would not be too difficult since it may involve the casting of several different sub-necks from polyurethane with different chemical compositions.

3.9. SUMMARY AND RECOMMENDATIONS

Preliminary results of cadaver testing indicate that the upper thorax acts as an energy absorbing structure in S-I impacts. The design criteria for a helmet impact test system should include provision for some damping.

Three systems are proposed that range in sophistication from "minimum cost" to "best possible". The compromise system has the structural advantages of the "best", but has the "minimum" instrumentation required for evaluating a given helmet

The use of the MVMA-2D model is proposed, although argument is made for a direct approach for designing the energy absorbing component of the device.

It is recommended that the proposed compromise system, described in section 3.7., be selected for actual construction, with a sub-neck from polyurethane that has compliance and damping characteristics that approximate the initial guesses obtained from impedance data. This component may be redesigned with paralled guidance from model predictions, device test results as well as additional cadaver testing.

4. PHASE IV - HELMET IMPACT TEST SYSTEM CONSTRUCTION

The helmet impact test system (HITS) selected for actual construction has been proposed and described in the previous chapter. However, as actual prototypes were being tried out, it became obvious that a sub-neck system consisting of a polyurethane neck would not respond to impact as was anticipated.

The design was modified in order to meet the response requirements spelled out by observations and results of actual cadaver tests. The result is a repeatable and reliable test device that would stand up to a demanding testing environment and whose response closely matches observed responses in actual cadaver tests.

In the following sections, key features of the HITS are described, and results of laboratory impact tests are presented to be compared to cadaver test results.

The actual HITS hardware is being delivered to NIOSH, as required by the contract. All engineering drawings and specifications are submitted (under separate cover) to allow exact duplication of the HITS and/or possible future modifications and improvement. Also submitted under separate cover, is an operation and assembly manual which gives detailed step-by-step instructions to test engineers and technicians for the use of the HITS.

4.1. KEY FEATURES OF THE HITS

The design of the HITS was guided by observations from cadaver responses to S-I impact tests. The design philosophy was to incorporate separate mechanical elements which perform different but specific functions. Whenever possible, the latest versions of existing elements were selected; in some cases, however, new components were specifically designed to be incorporated in the HITS.

4.1.1. The Head

The head used in the HITS is the Hybrid III dummy head, developed by General Motors Laboratories for automotive crash testing. This head is the state-of-the-art model of the human head in which the inertial and anthropometric properties are faithfully simulated. The response of this head has been extensevely studied and has been validated against actual cadaver head impact responses.

4.1.2. The Neck

The neck used in the HTIS is also borrowed from the GM Hybrid III dummy. Just as with the head, this neck has been designed to duplicate responses obtained from cadavers as well as from human volunteers.

While this neck can faithfully duplicate human response in pitching motion (flexion - extension) about an L-R axis, its response in lateral flexion about an A-P axis is not well documented. Furthermore, its axial stiffness is so high that the neck exhibits negligible axial deformation either as elongations or compressions.

In actual cadaver S-I impact tests, the primary rotation of the head is about an A-P axis, while other rotations (about the L-R and S-I axes) are secondary. This is a fortunate observation since it allows the acceptance of the Hybrid III neck as a suitable HITS component.

4.1.3. The Thorax

In order to provide axial compliance of the HITS, the neck is mounted on a sub-assembly which allows the plunging of the head/neck up to a maximum of 2.5 inches. This was necessary since such axial motion is not possible with the neck alone.

Based on analysis of impedance curves (presented in chapter 2) and simple models, it was determined that a simple spring and dashpot system would produce results similar to those observed in the cadaver tests. Furthermore, a spring constant of 300-400 lb/in and a critically damped system were suggested by several computer exercises of simple models.

The use of stock compression springs was dictated by the high cost of custom designed springs so that, in order to obtain the desired stiffness and deformation range, six springs were selected as the complient element of the thorax.

Because the neck response in bending is validated, the thorax was designed so that it provides no additional rotation, thereby preserving the already-proven neck bending characteristics.

Finally, cadaver test results indicate that some of the impact energy is absorbed by the spine and upper thorax. The HITS was therefore designed to include friction elements that effectively dissipate some of the kinetic energy. Friction damping was selected as a design compromise over Viscous dampers. The reason in our recent experience with Viscous dampers where, in order to meet certain criteria, a "dashpot" proved to be very troublesome to design and use. The friction elements incorporated here are trouble-free, and provide an acceptable method of energy dissipation.

4.1.4. The Base

Initially, a 3-way compound angle vise was proposed as a mounting base for the thorax, to allow the re-orientation of the head and the control of location and direction of impact.

Because of the impact forces involved, a simpler base was designed to withstand the highest impact forces and moments, and to allow the adjustment of two angles in 5-degree discrete increments.

The base allows therefore adjustments in the "pitch" and "roll" angles in such a repeatable fashion that no standard compound angle vise could. Description and instructions for its use are given in the HITS Operation and Assembly Manual.

4.1.5. Transducers

The HITS includes a multi-channel load cell which fits the Hybrid III head and neck. This neck transducer measures 3 interaction loads between the head and neck, which are the shear force in the A-P direction, the axial force in the S-I direction and the moment about an axis in the L-R direction. These loads are measured at a location which corresponds to that of the occipital condyles in the human head.

This load cell was included because of its availability, even though cadaver tests suggest that the shear force may be ignored in S-I impact cases.

The head has a provision for mounting a triaxial accelerometer package at the head center of mass. The resultant acceleration at the head CG may therefore be computed, even though a single accelerometer in the S-I direction may be sufficient for monitoring the head motion in S-I impact cases.

4.2. TESTING PARAMETERS ADJUSTMENTS

There is a number of parameters which can be adjusted to produce the desired HITS responses. However, two parameters are more important than others; these are:

a) the energy of impact, determined by the weight and drop height of the impactor, and

b) the amount of friction introduced in the system.

Other parameters include the characteristics of the helmet being tested (or those of the padding over the impactor surface) and the location and direction of impact, all of which are variables that are set depending on the objectives of the impact test. Finally, other parameters are fixed and cannot be adjusted, such as the characteristics of the head and neck. For these uncontrollable parameters, it is assumed that they have already been adjusted and that they are at their optimum "setting".

In order to "tune" the device, tests were conducted while varying the two most important parameters: impact energy level and amount of friction.

4.2.1. Effects of Energy Levels

The ANSI Z89.1 standard requires dropping an 8-lb spherical weight from a 5-ft height. This produces an energy level of 40 ft-lb or about 55 J. However, most cadaver tests suggest that even an energy level of 200 J would not produce such high energy level, either the drop height of an 8-lb mass must be increased to over 20 ft, or a weight of 18 lbs may be dropped from a height of 9 ft.

To resolve this dilemma, a series of HITS tests (80H306-311) was conducted varying the drop height while using the same helmet and same weight of 9.55 lb. The pitch and roll orientations of the base were both set to zero. Loads produced in these tests are given in figures 117 - 122, and head accelerations in figures 123 - 128.

Judging by the head accelerations, the HIC values produced (19-152) were too small compared to HIC values (around 300) produced in cadaver tests. This suggests that much higher energy levels would be needed to produce higher HIC values.

When comparing S-I neck loads produced by the HITS tests (4900-7300 N) with those produced in cadaver tests (around 5000 N), it seems that a drop tower with 9.5-lb weight and reasonable heights (5-8 ft) would generate sufficient energy levels to test the HITS.

The fact that the same helmet was repeatedly impacted may explain the lower HIC values and the unexpected lower S-I acceleration in test 80H310. Furthermore, the absence of friction from the system during this series may have effected the response.

To eliminate the effects of repeated impact on the same helmet, tests 80H317, 320,323, and 326 were conducted using a brand new helmet every time, and varying the drop heights of 9.55-lb mass from 5 to 8 ft. load responses, shown in figures 129-132, indicate that the use of a new helmet does not significantly effect the axial loads.

4.2.2. Effects of Friction Damping

To study the effect of friction, the set screws in the two friction blocks were set at torques of 30, 50 and 70 in-lb for three tests 80H312, 313 and 314, respectively. Results are shown in figures 133 - 138. In these tests, the same helmet was used, and the 9.55-lb weight was dropped from a height of 8 ft. These result seem to indicate that friction does not effect the response, although such conclusion was disproved in next series of tests.

In the next series of tests, the same helmet (#1) was used and friction damping was varied from 0 to 45 in-lb by 15 in-lb increments. The drop height was maintained at 7 ft and the 9.55-lb weight was used. Results are shown in fig. 139 - 146 for the four tests 80H347-50. The lowest axial load of 2546N (Test 80H347) corresponds to no-friction setting, while a friction setting of 45 in-lb produced an axial load of 3472N. Therefore, friction does effect the load, if it is low enough as to not "lock" the springs out of the system.

The refinement and adjustment of the system must also be based on head acceleration levels produced during impacts.

4.3. ACCELERATION LEVELS

So far, most head accelerations produced during impacts to the HITS are extremely low. This was true regardless of the type of helmet used (2 types) or the number of repeated impacts delivered to the same helmet.

In order to simulate the impact conditions under which the cadaver tests were actually conducted, it was necessary to test the HITS under the following conditions:

a) No helmet shall be used;
b) some Ensolite padding shall be used;
c) vary the friction (from 0-45 in-lb settings)
d) impact with energy corresponding to an 18-lb weight dropped from 9-ft height.

Using an 18.1-lb weight, with flat impact surface, and a padding of certain thickness of Ensolite (type AH), tests were conducted in search of the "best" drop height that would produce accelerations similar to those observed in cadaver tests.

The effect of padding thickness is demonstrated in fig. 147-150. These are results of two tests where the 19.1-lb weight was dropped from a 9-ft height. The difference is that in test 80H356, a two-inch padding was used , while only one-inch thick padding was used in 80H363. The results suggest that a 1 inch padding produces a much higher acceleration than desired.

If 1-inch padding is to be used, the drop height must be lowered, as was done in tests 80H359 and 360, both run at the same drop height of 5 ft, under identical impact conditions. Results (figures 151-154) indicate that the 5 ft height does not produce acceptable accelerations, while loads are acceptable at this energy level. The optinum drop height would therefore be between 5 and 9 feet.

By interpolating between the two heights, it was determined that a drop height of 6.3 ft would result in approximately 70 G of head acceleration.

The next step was to determine the optimal setting of the friction, for the same drop height of 6.3 ft. The assumption is that the "best" response of the mechanical HITS is one which absorbs the maximum amount of energy by friction, as evidenced by a no-rebound behavior.

Tests 80H374 through 380 were conducted with different settings of the friction. Acceleration responses are shown in fig. 155-161, and load responses are shown in fig. 162-168. From these tests, it was determined that the friction setting of 30 in-lb screw torque is the borderline between a rebound and a no-rebound. This was supported by high speed films taken for this series of tests.

4.4. NON-AXIAL IMPACTS

Several tests were conducted to demonstrate the effects of impacting the HITS at angles other than along the S-I axis, and at locations other than the head vertex.

In these tests, the spherical weight of 9.55 lbs was dropped from a height of 6.33 ft producing impact velocities about 5.5 m/s. The variable parameter in these test was the orientation of the device with respect to the vertical drop line. Results are shown in figures 169-176.

The pitch and roll angles of the base were set as follows:

Test No.	Pitch	Roll
80H382	0°	45°
80H383	0°	30°
80H384	45°	0°
80H385	30°	0°

Location of the impact was the highest point on the helmet when the two angles are set as described.

4.5. SUMMARY

There are several key observations to be made about the behavior of the HITS as a tool for testing the performance of industrial helmets:

a) One should recognize that the human body is a complex system and that one could only hope to approach the human response with mechanical devices.

Another point to be recognized that human tolerance data remains to be firmly established so that the response of the mechanical HITS remains to be interpreted in light of actual tolerance data.

b) The second observation is that a wide range of responses can be produced by the device by controlling external parameters such as the impactor weight and drop height, and internal parameters such as the amount of friction damping introduced into the device and its orientation.

c) An optimun drop height seems to be about 6 feet, using a weight of about 18 lbs. This produces reasonable accelerations but extremely high loads. To reduce the generated loads, a weight of 9 lbs may be used, resulting not only in lower and acceptable loads, but also in lower accelerations. Regardless of the weight used, helmet performance criteria as peak loads and/or peak accelerations must be adjusted based on results from actual cadaver tests; i.e., the results from testing helmets should be scaled to produce meaningfull results.

d) A torque setting of 30 in-lb corresponds to the amount of friction damping which separate the rebound from the no-rebound behavior of the device.

e) The HIC values obtained in testing the HITS were much lower than those judged to be intolerable. This was true regardless of the parameters used in the tests, such as the type of helmet, the amount of padding, the height and weight of the dropping man. This seems to indicate that the head accelerations may not be the proper response upon which helmet performance criteria should be based. Such conclusion is supported by cadaver test results which indicate that neck injury occurs even when HIC valued are significantly below the 1000 level used in automotive crash testing.

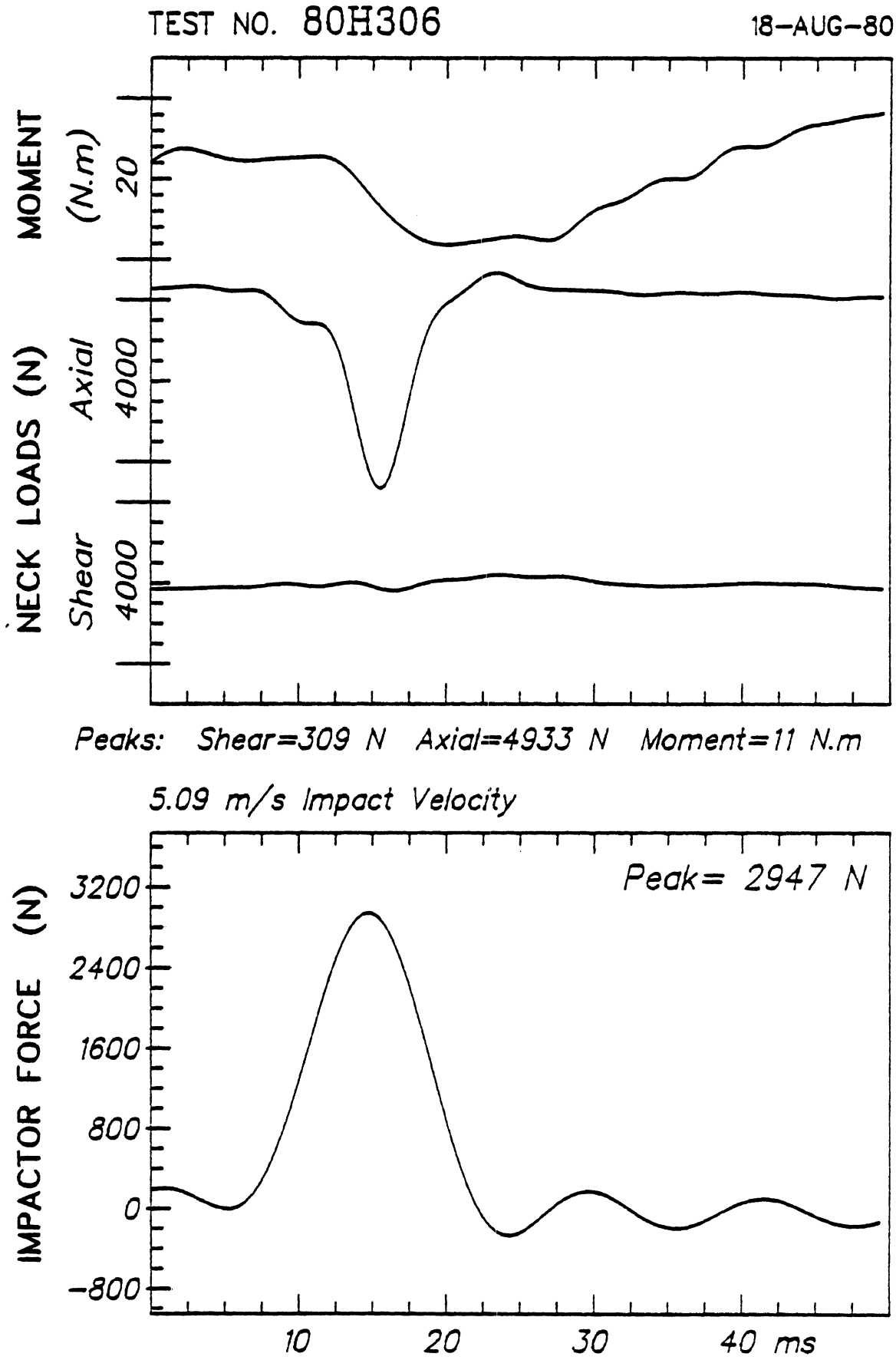


Figure 117: Impactor Deceleration and Neck Loads

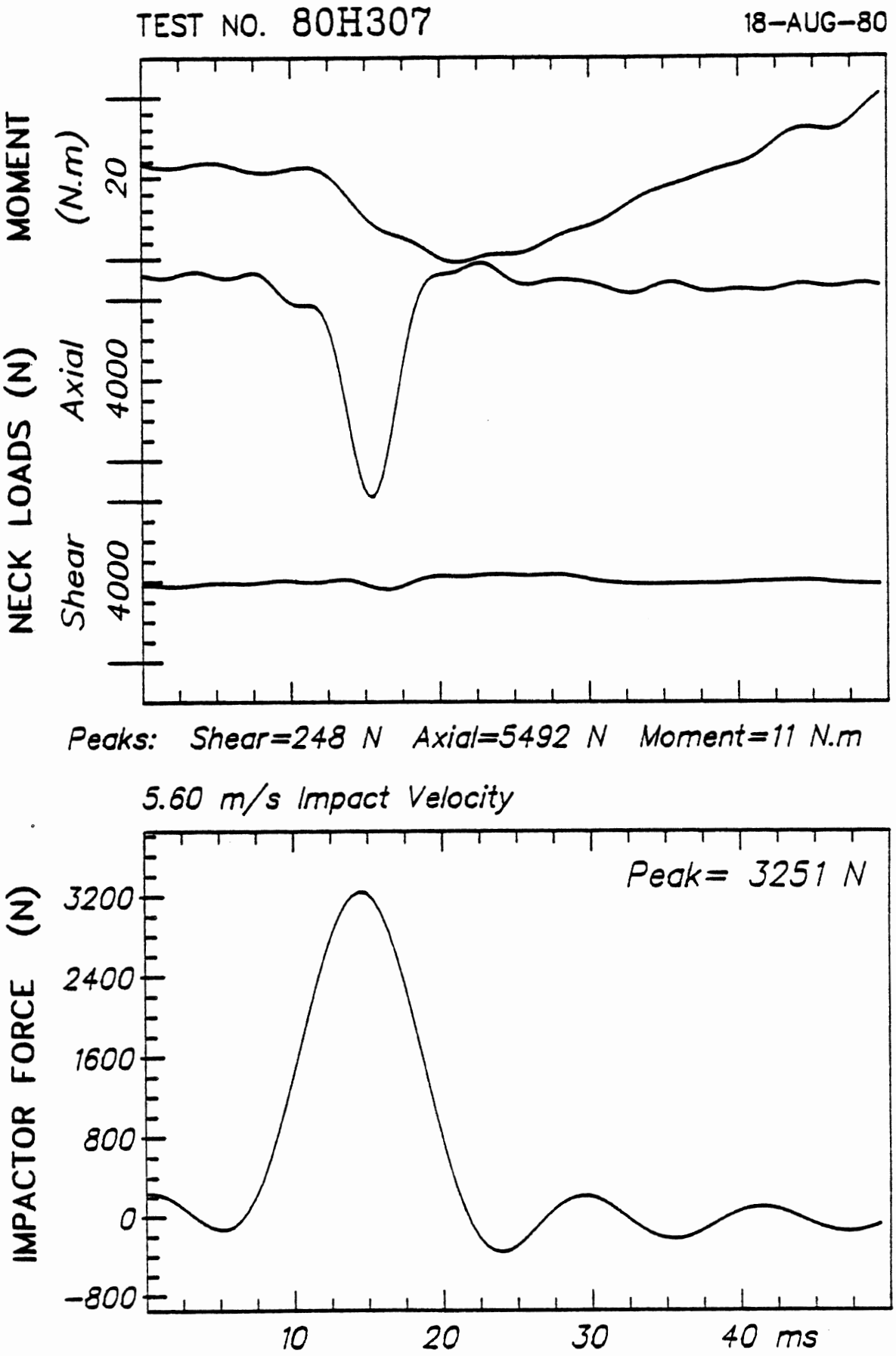


Figure 118: Impactor Deceleration and Neck Loads

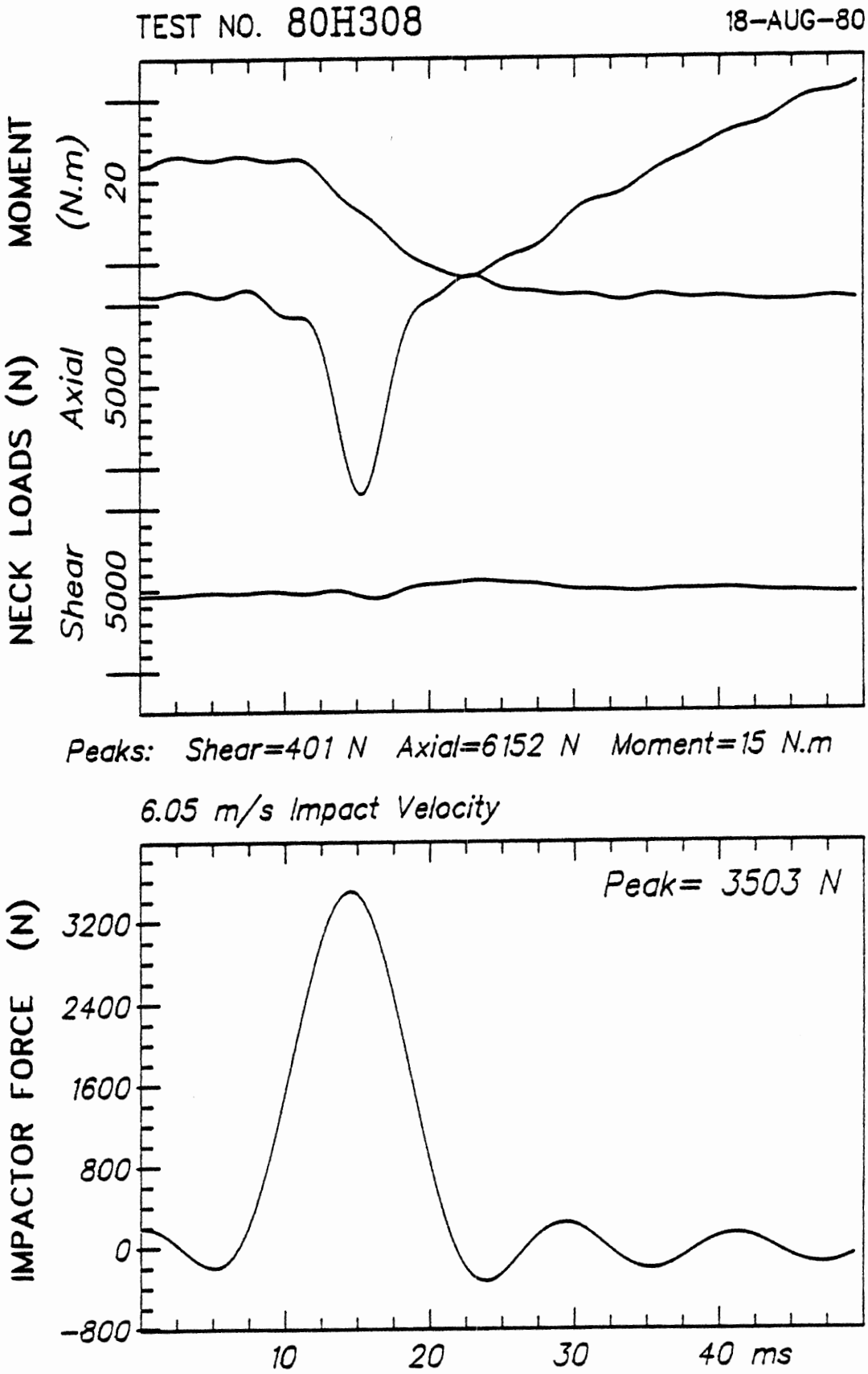


Figure 119: Impactor Deceleration and Neck Loads

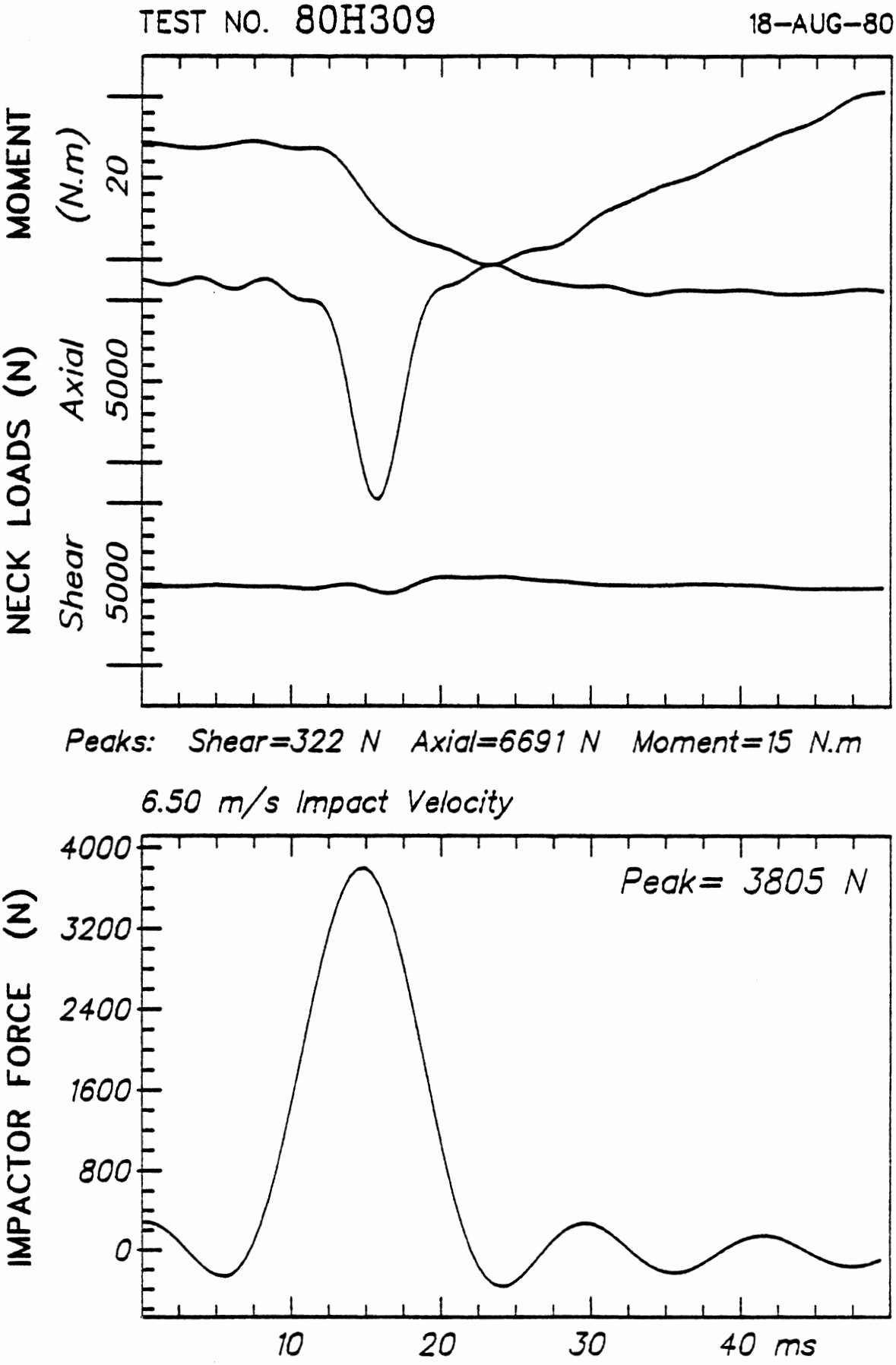


Figure 120: Impactor Deceleration and Neck Loads

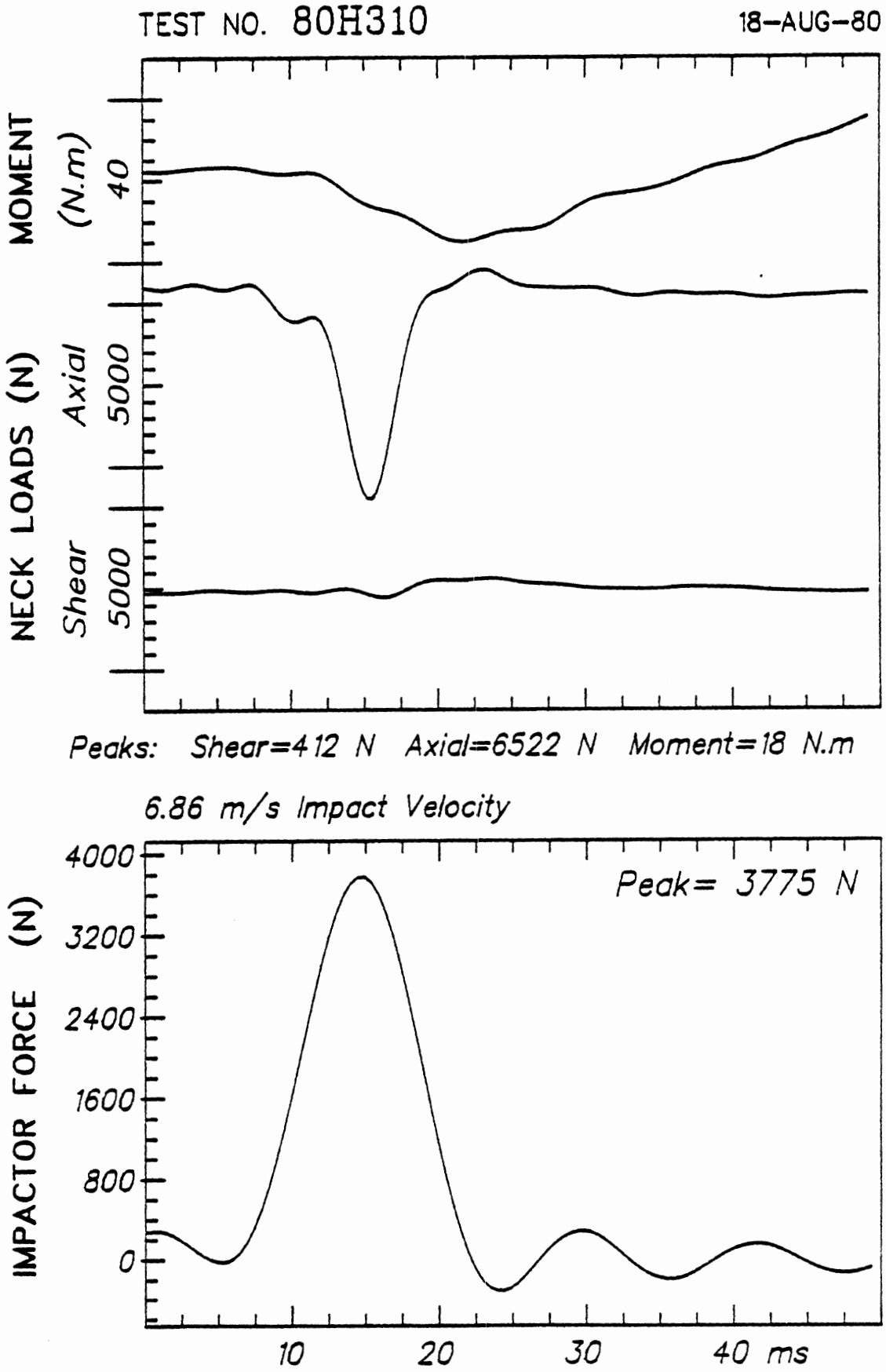


Figure 121: Impactor Deceleration and Neck Loads

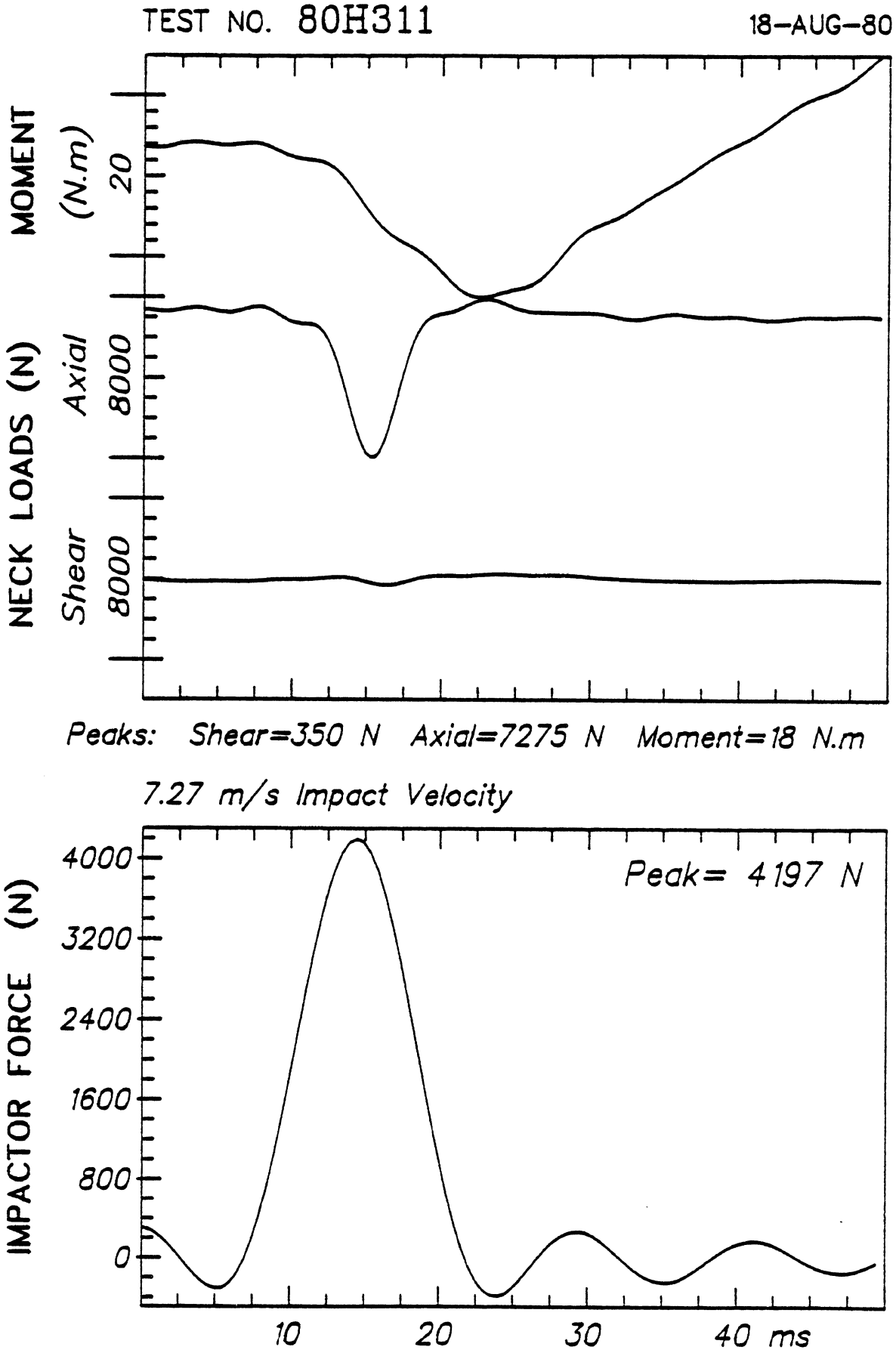


Figure 122: Impactor Deceleration and Neck Loads

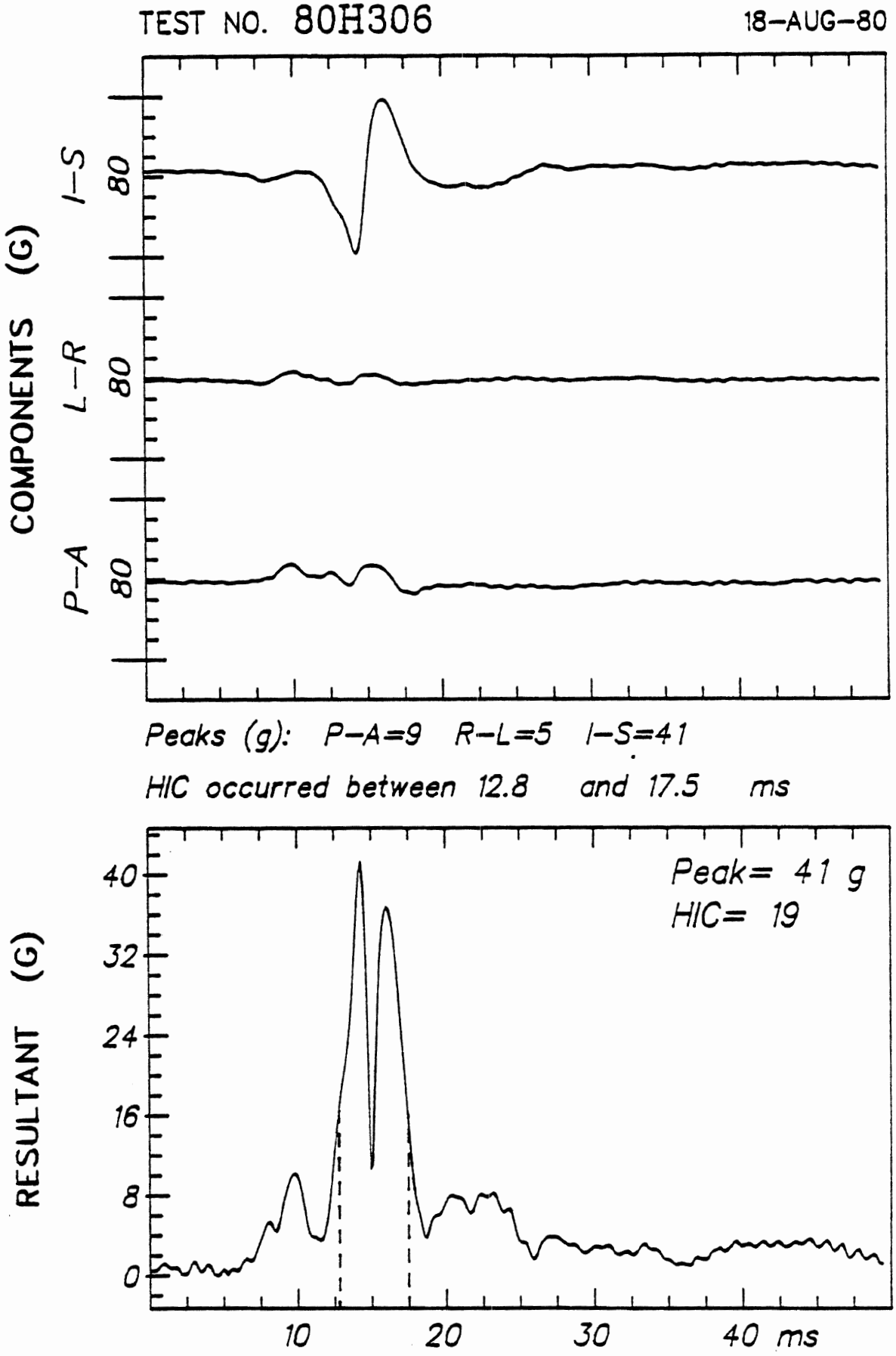


Figure 123: Head Acceleration vs. Time

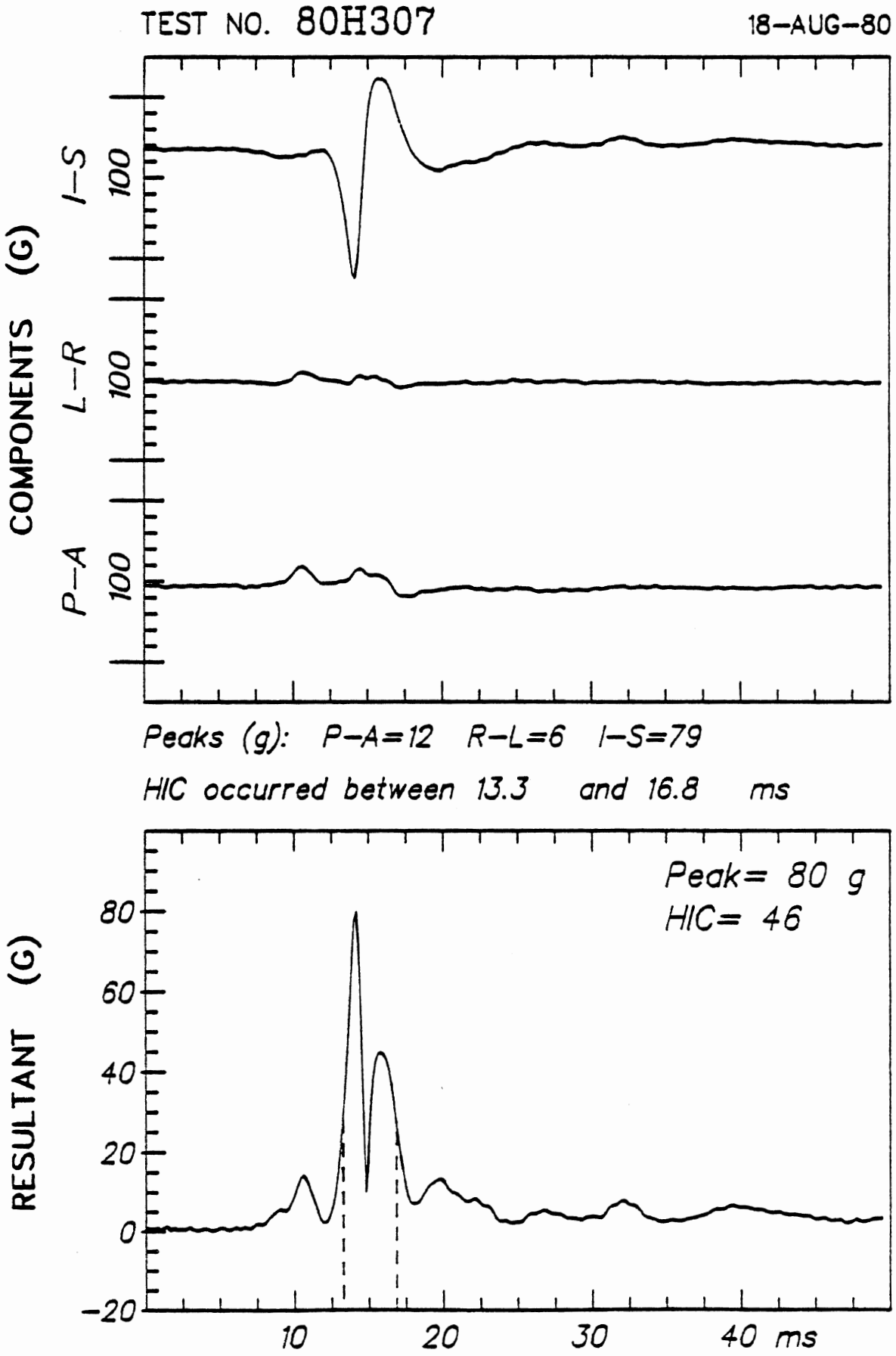


Figure 124: Head Acceleration vs. Time

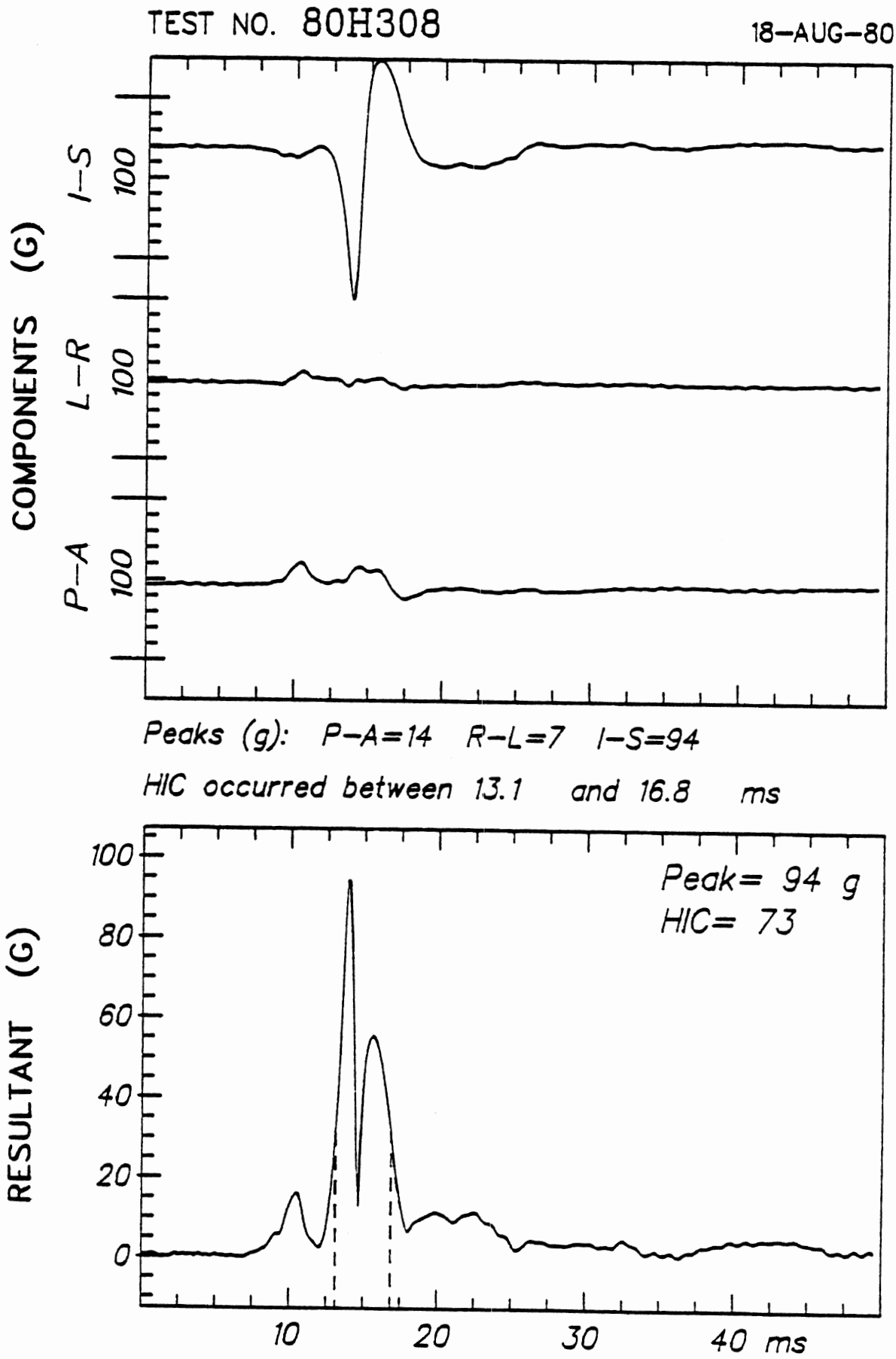


Figure 125: Head Acceleration vs. Time

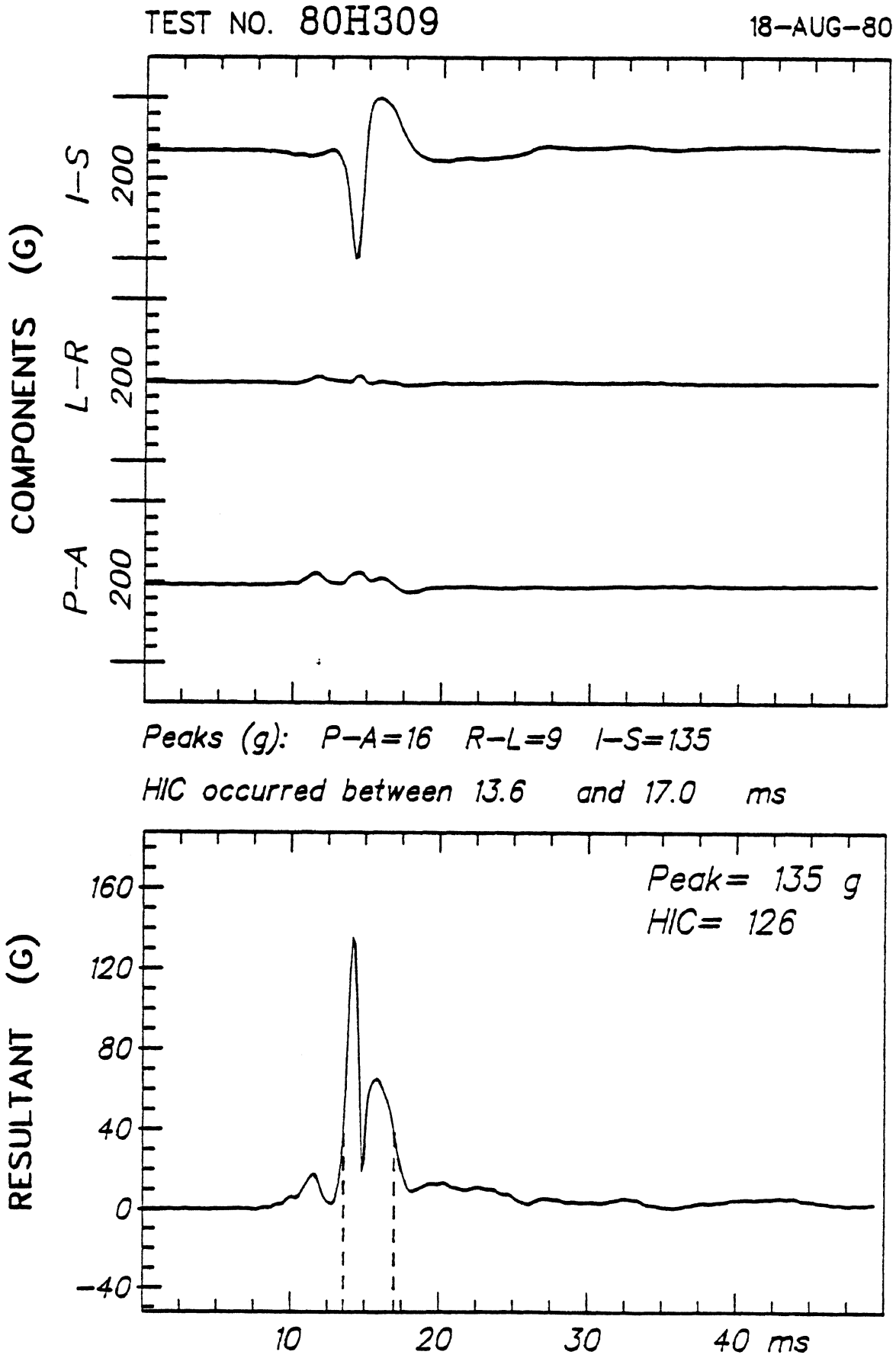


Figure 126: Head Acceleration vs. Time

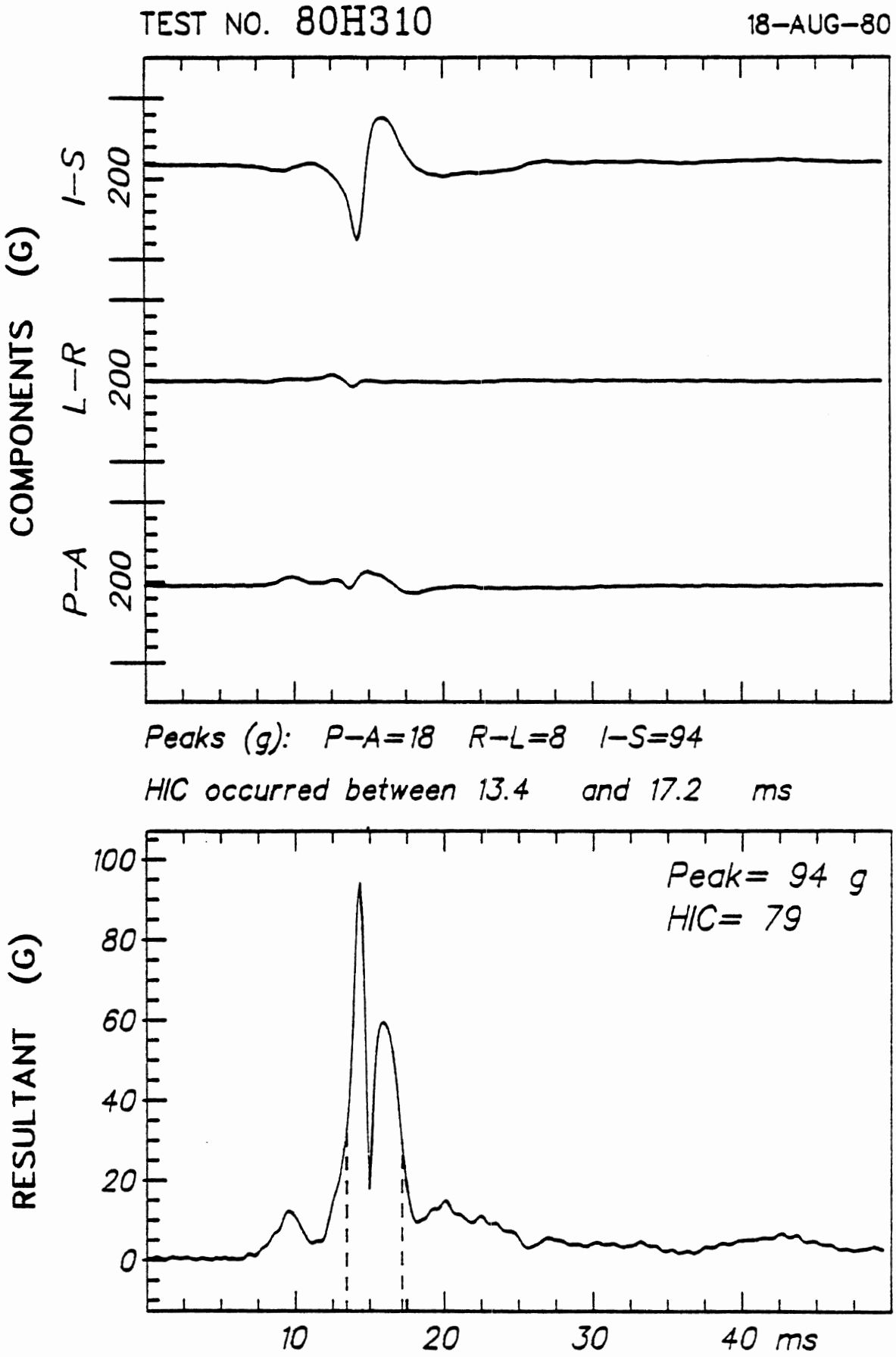


Figure 127: Head Acceleration vs. Time

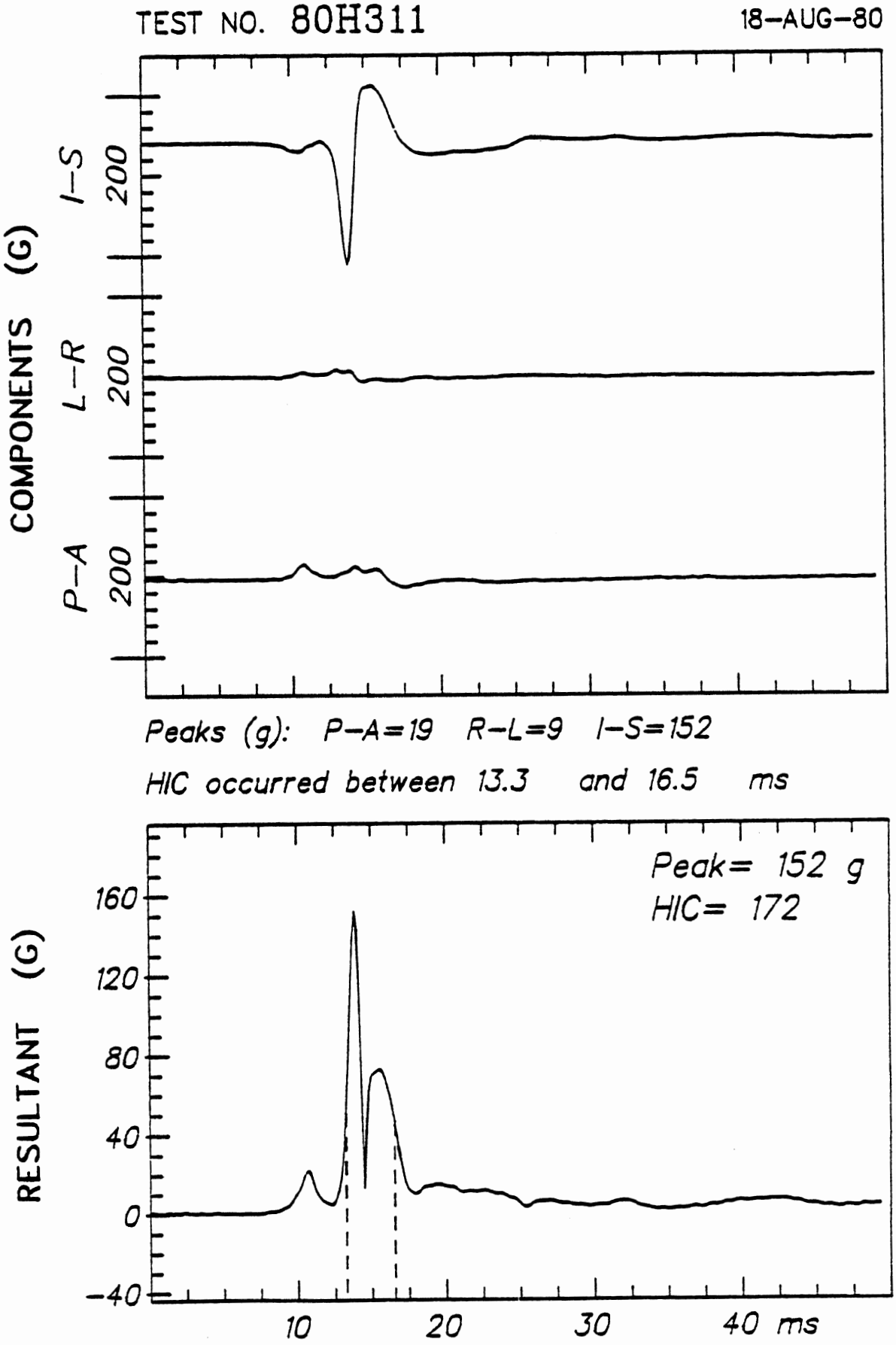


Figure 128: Head Acceleration vs. Time

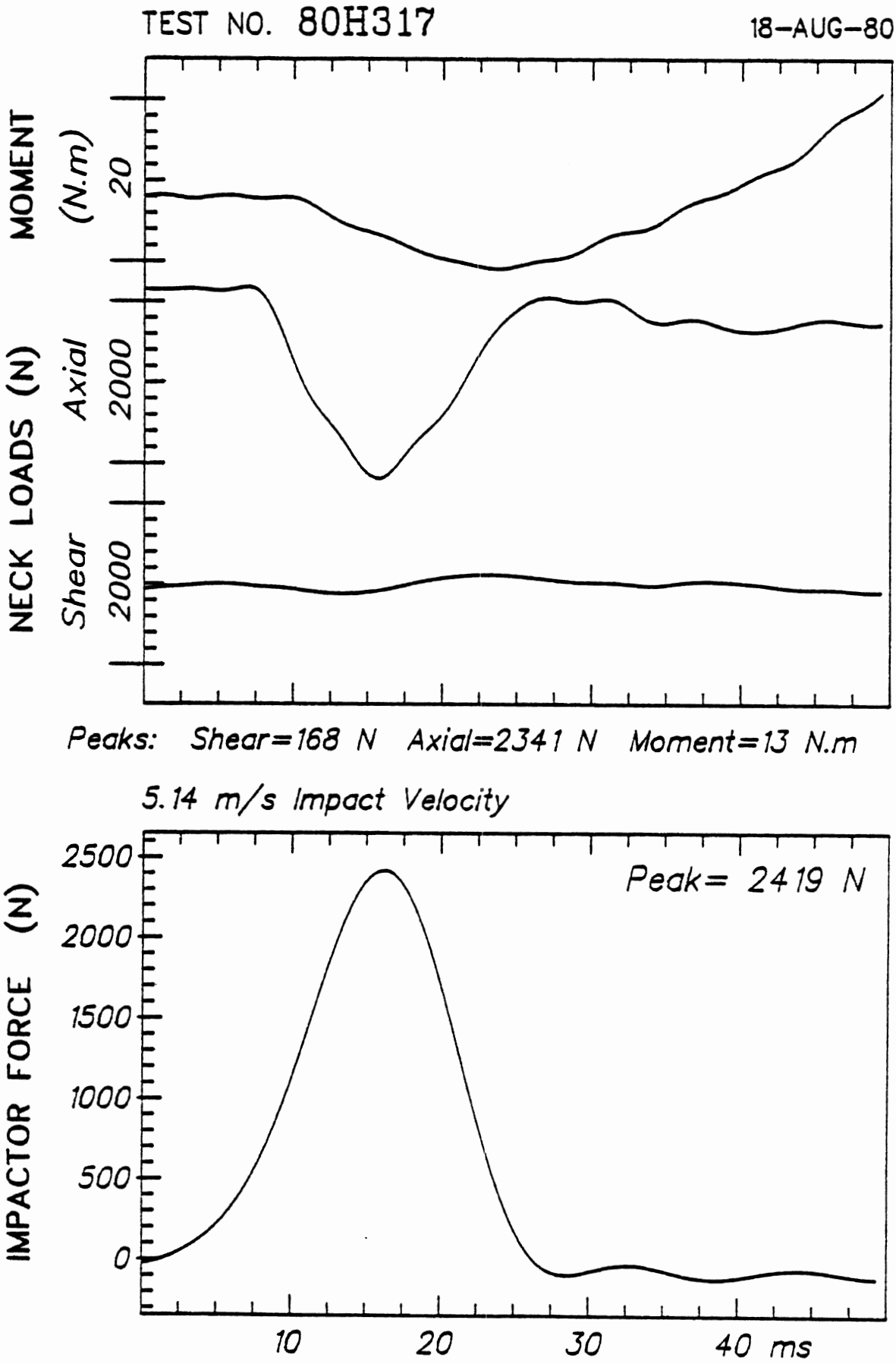


Figure 129: Impactor Deceleration and Neck Loads

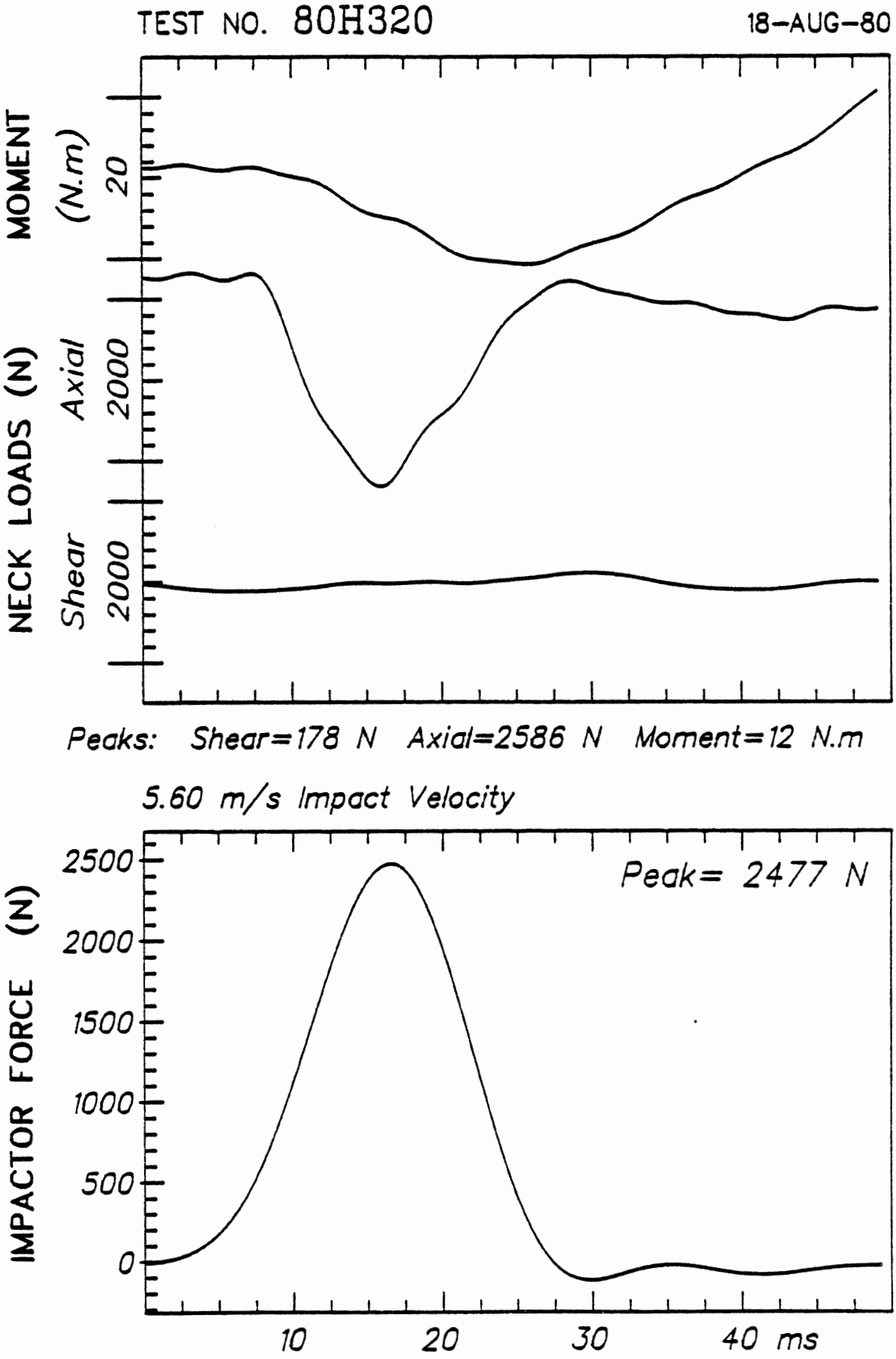


Figure 130: Impactor Deceleration and Neck Loads

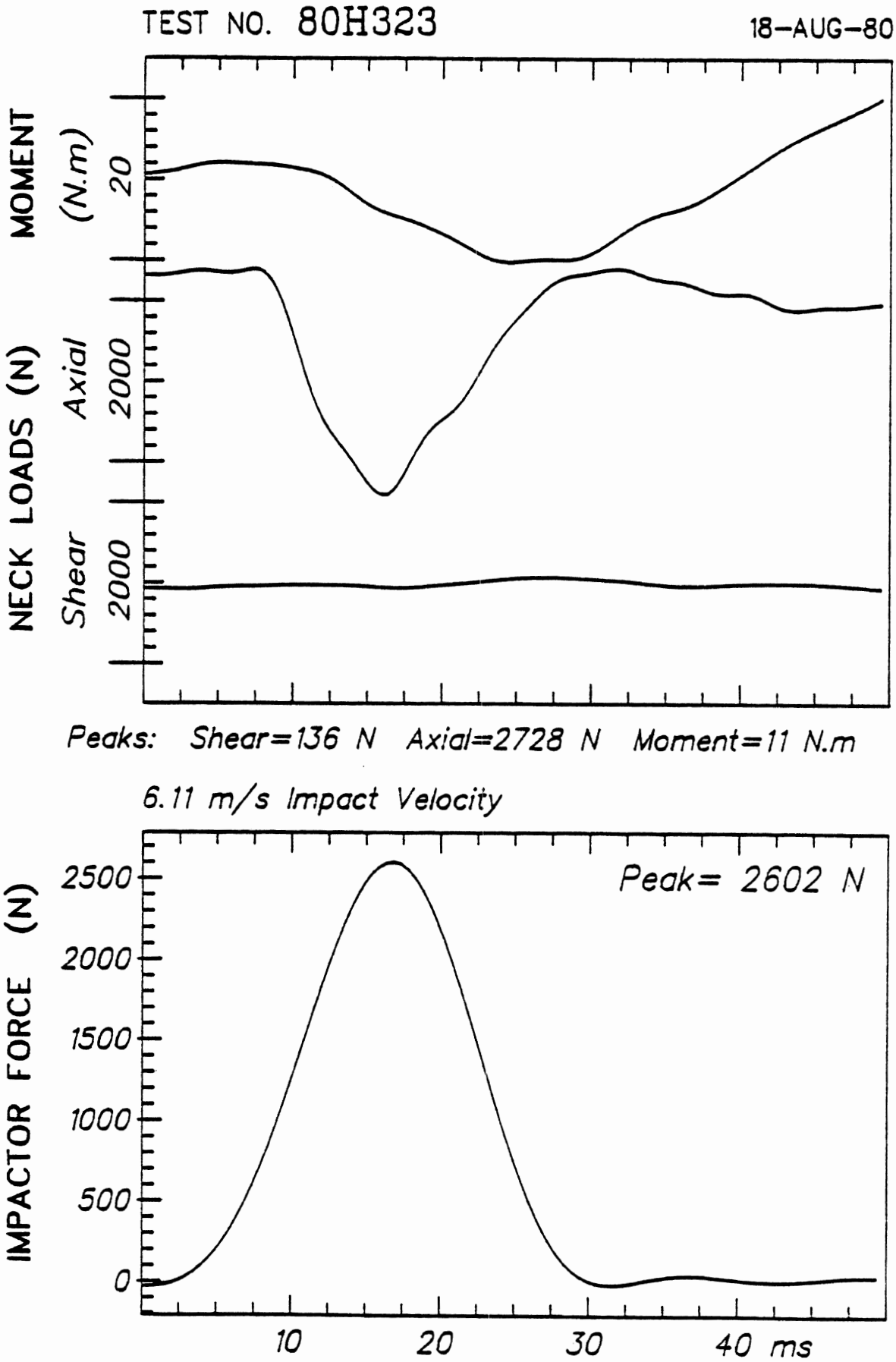


Figure 131: Impactor Deceleration and Neck Loads

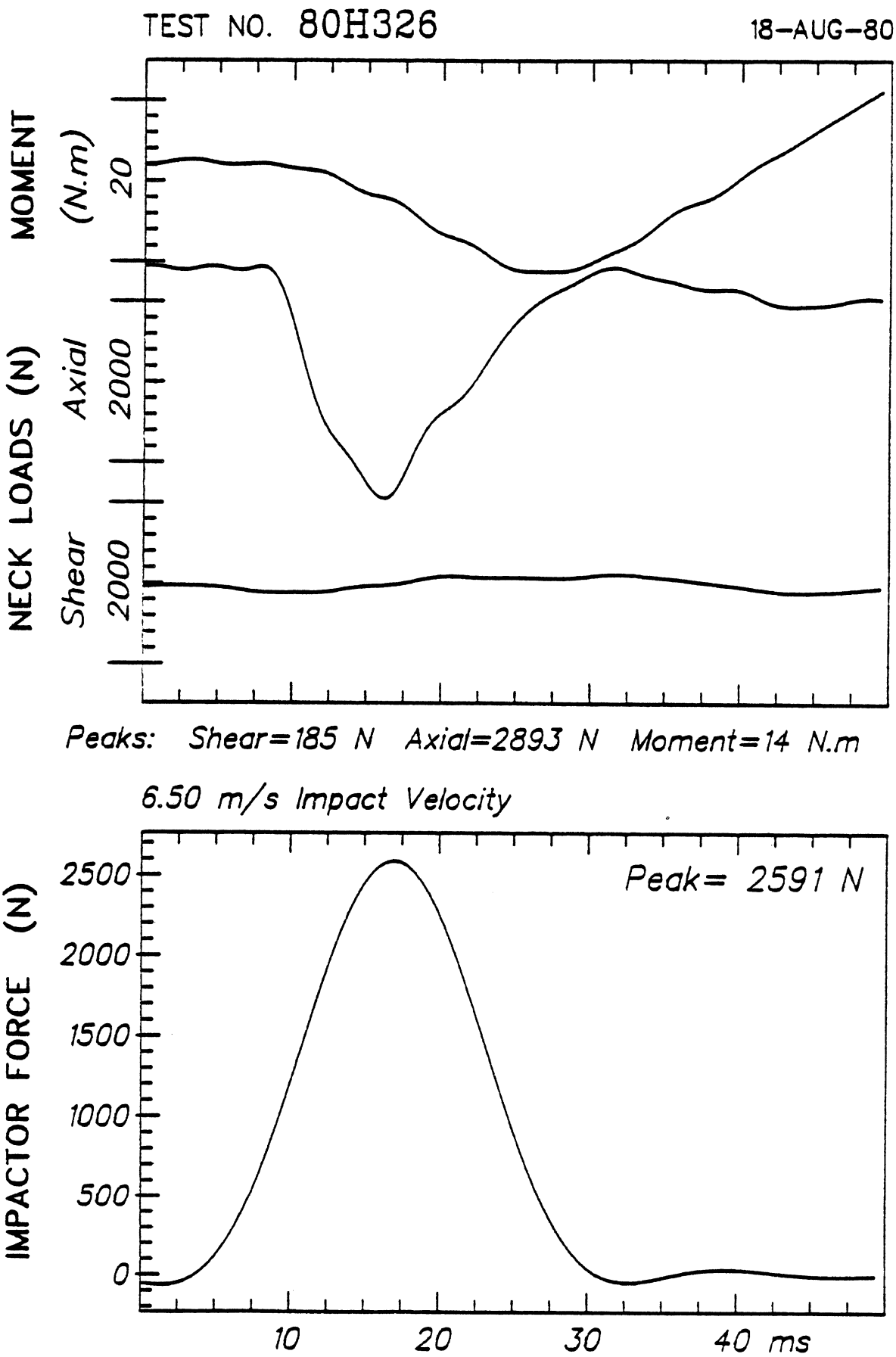


Figure 132: Impactor Deceleration and Neck Loads

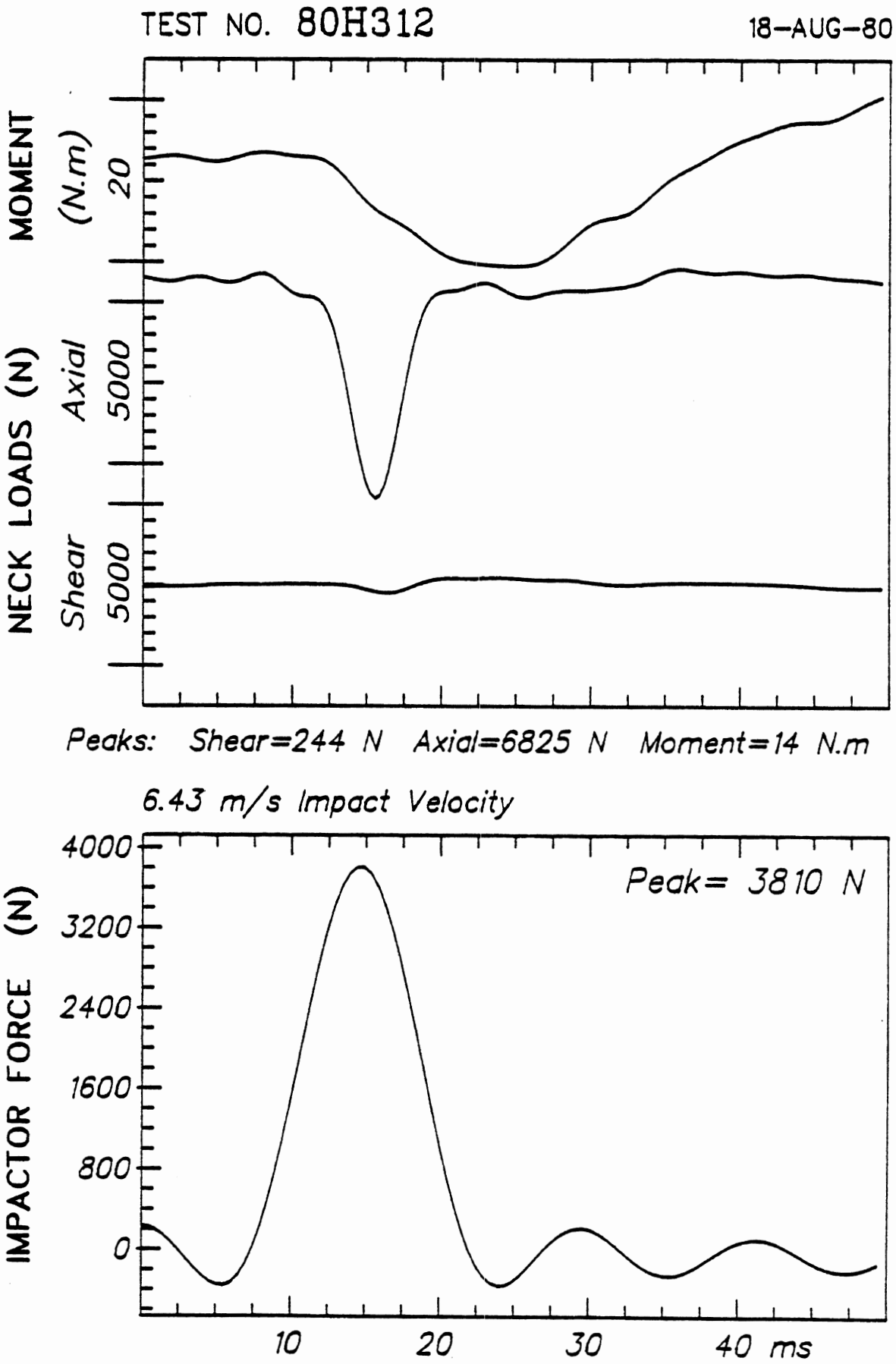


Figure 133: Impactor Deceleration and Neck Loads

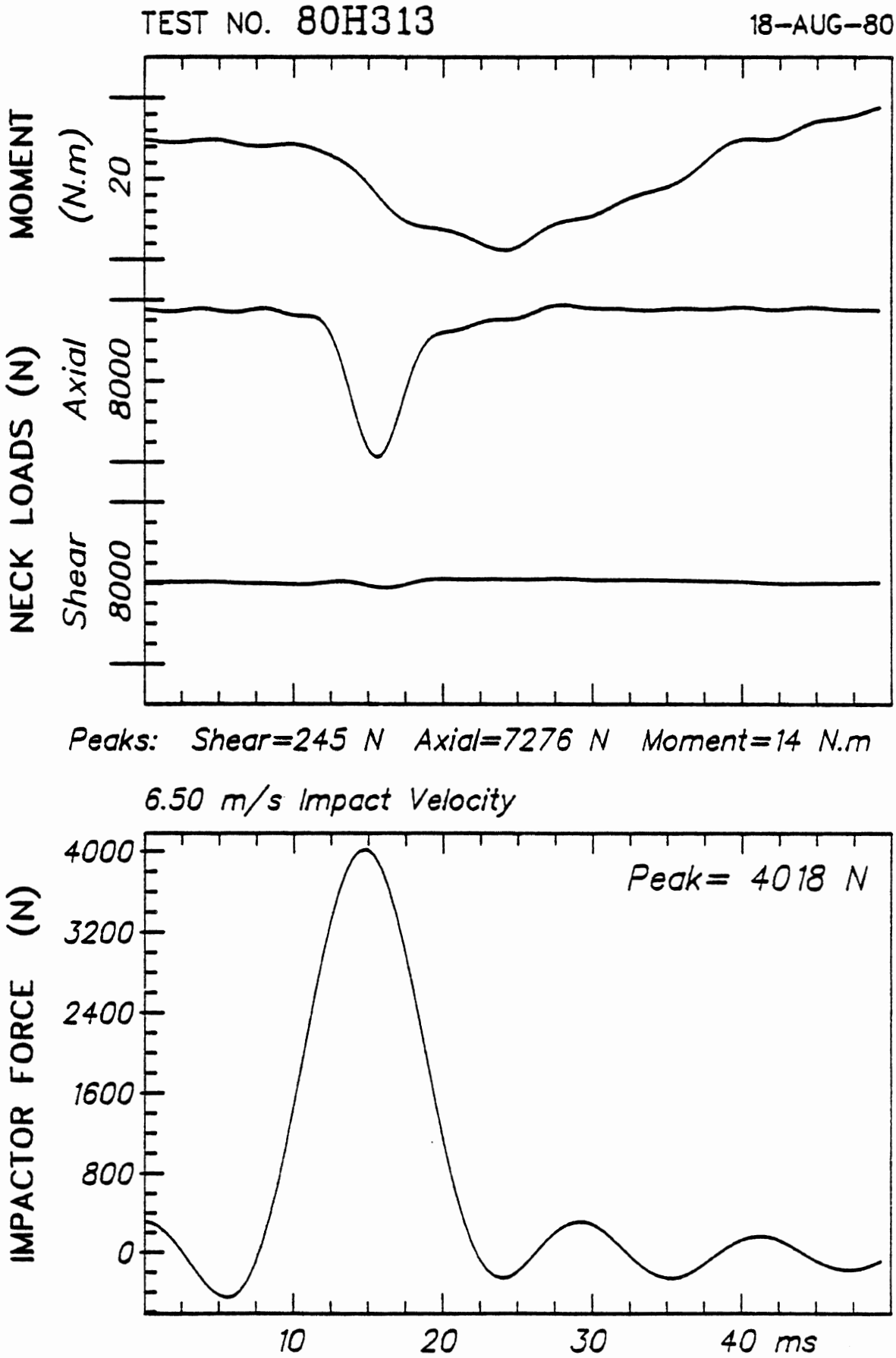


Figure 134: Impactor Deceleration and Neck Loads

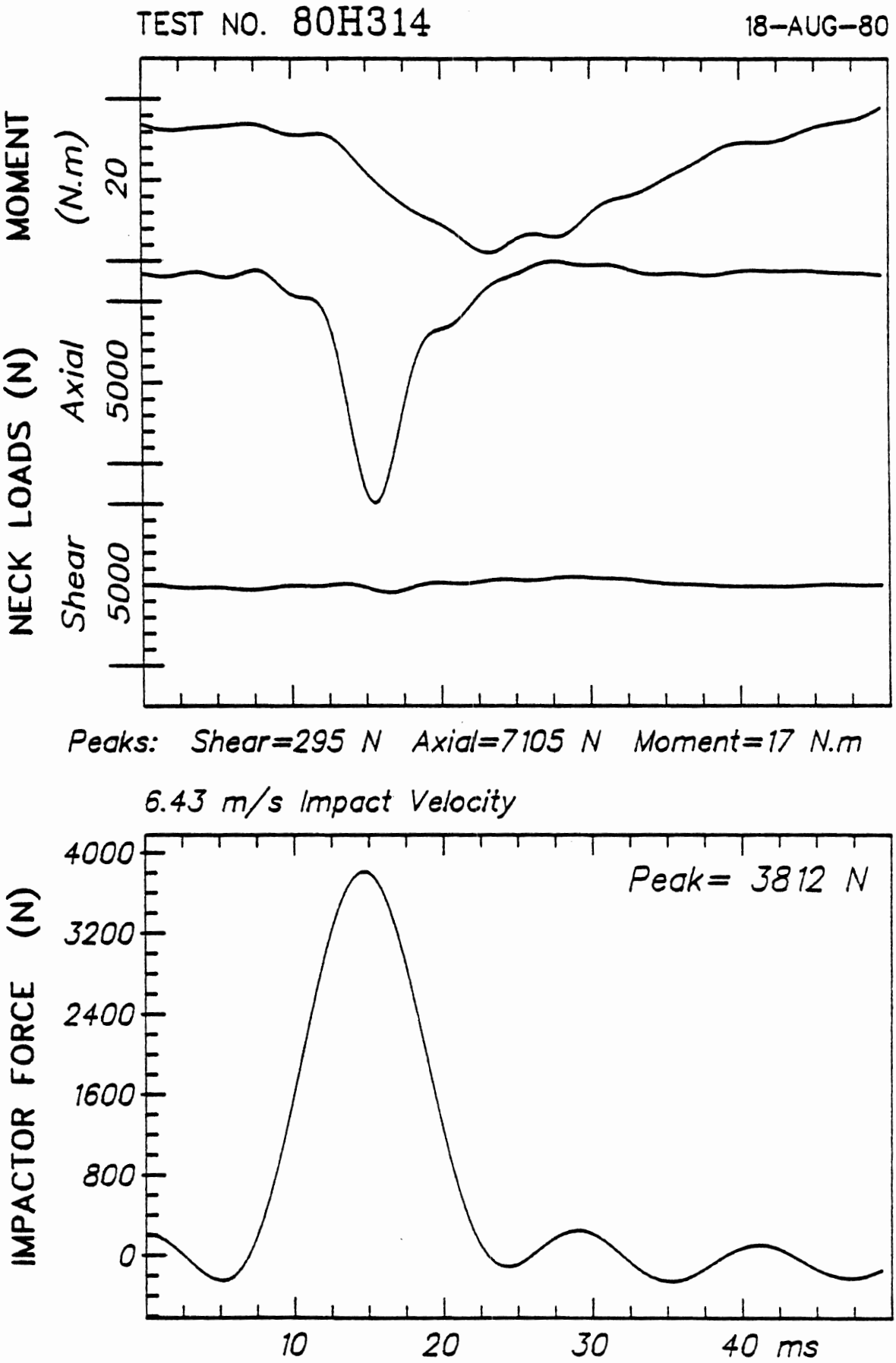


Figure 135: Impactor Deceleration and Neck Loads

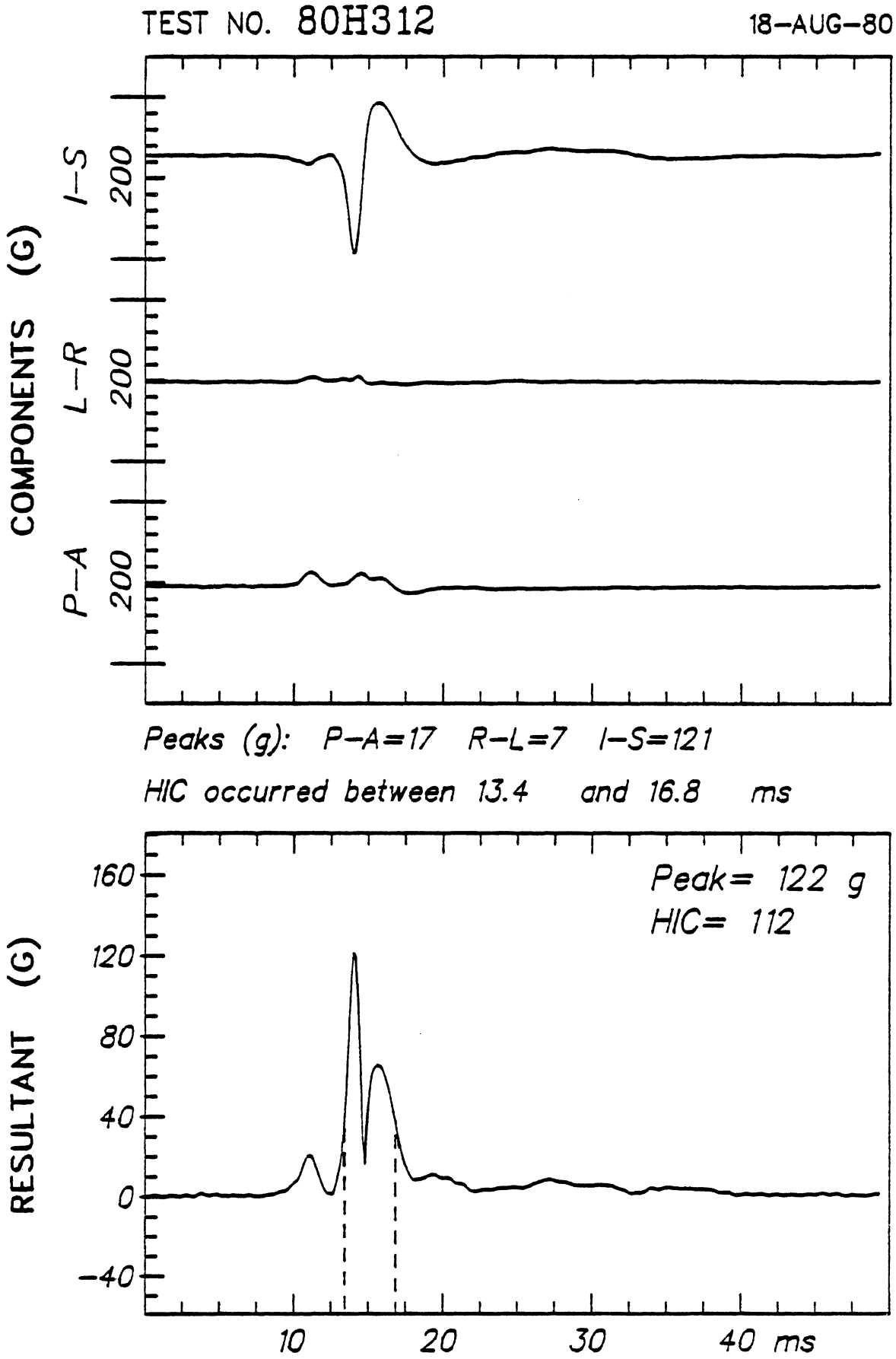


Figure 136: Head Acceleration vs. Time

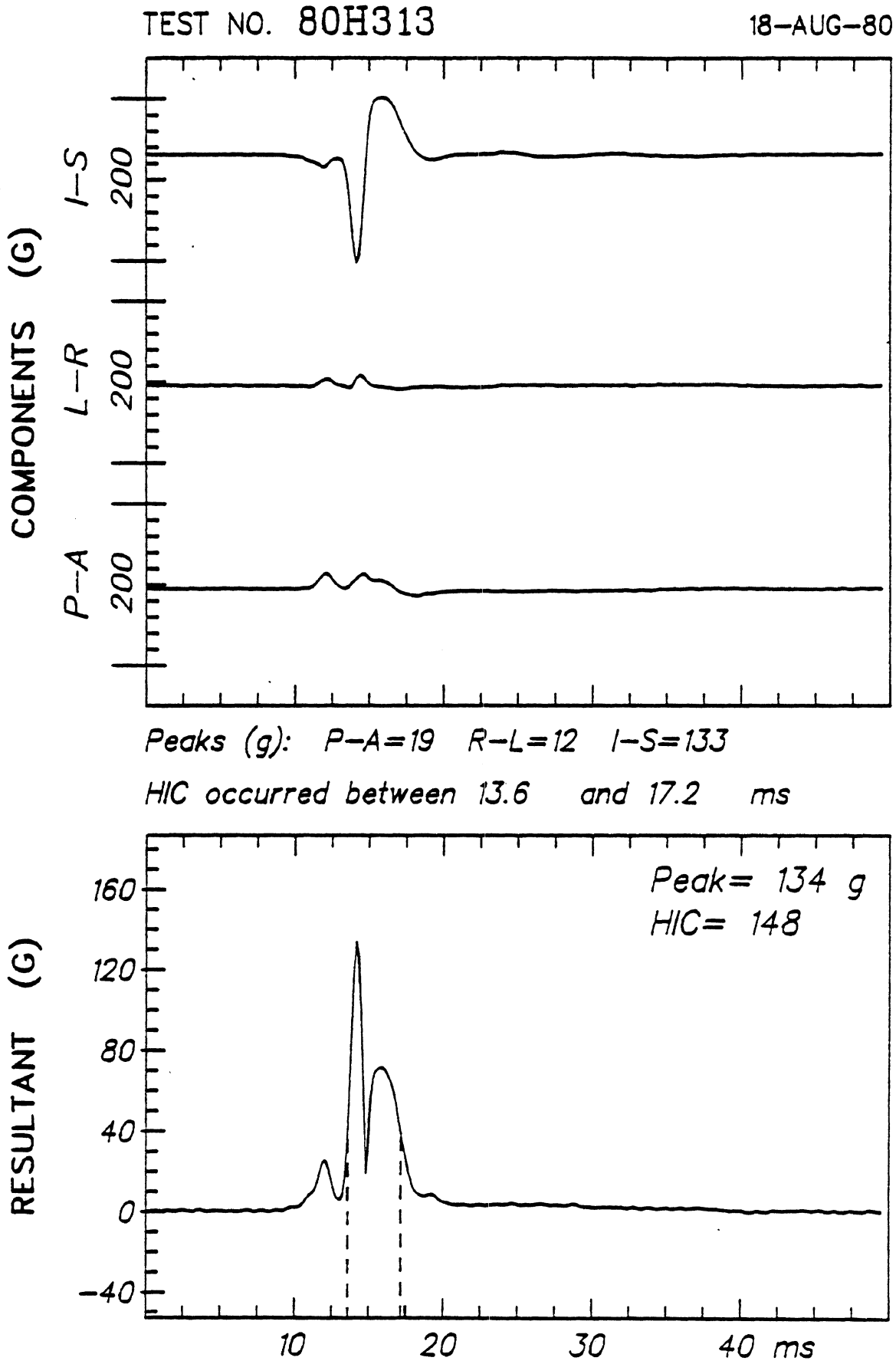


Figure 137: Head Acceleration vs. Time

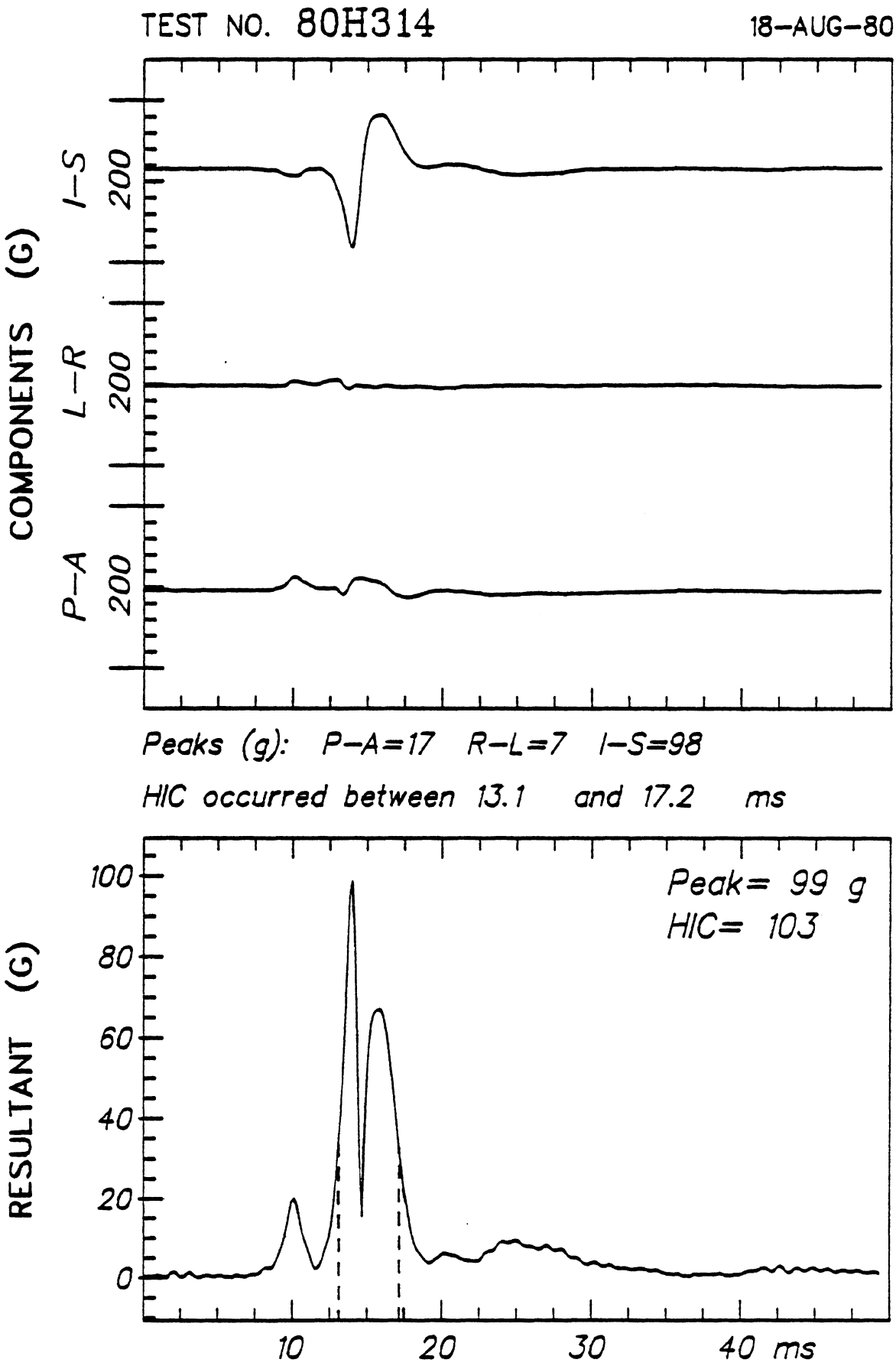


Figure 138: Head Acceleration vs. Time

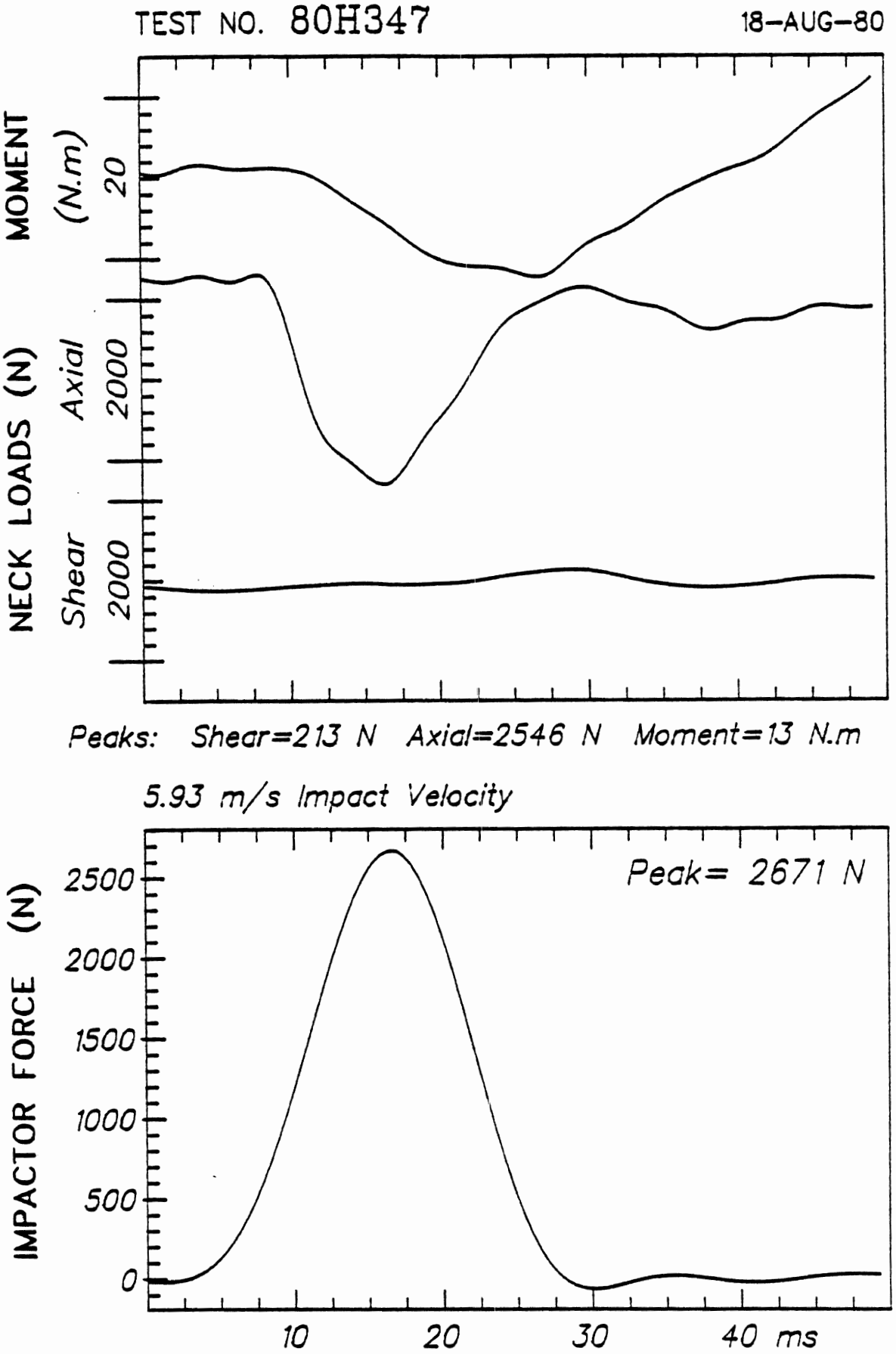


Figure 139: Impactor Deceleration and Neck Loads

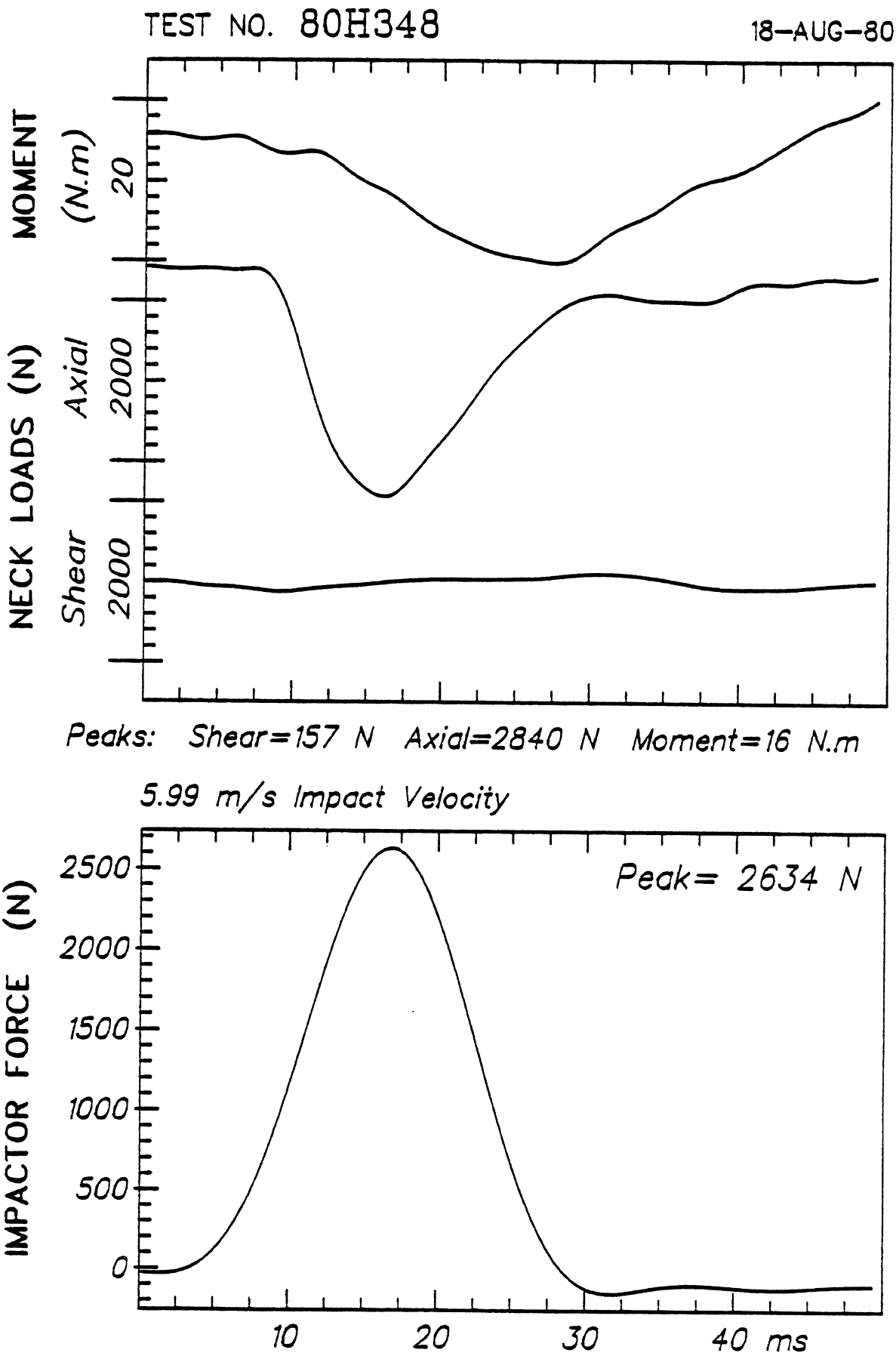


Figure 140: Impactor Deceleration and Neck Loads

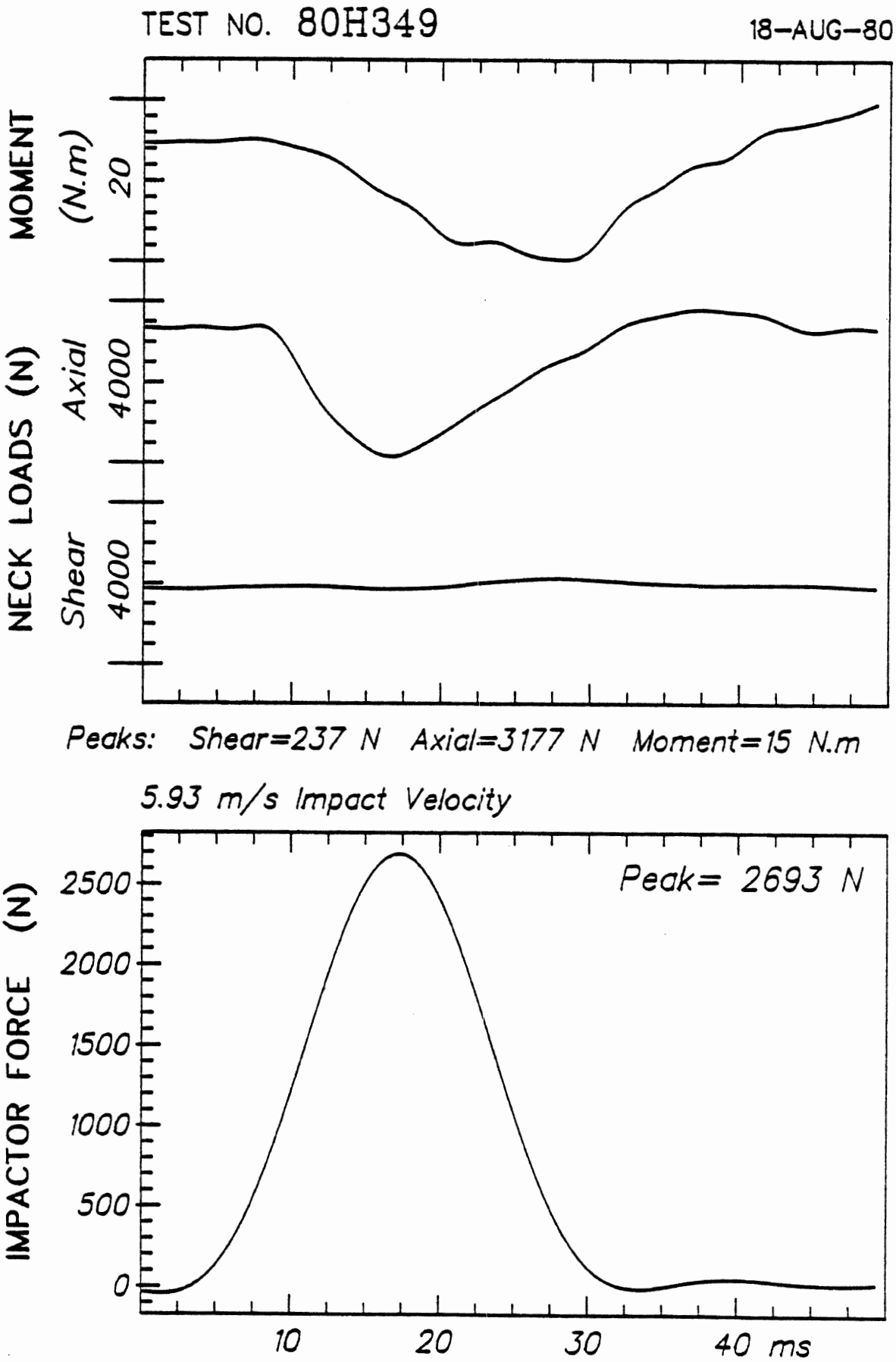


Figure 141: Impactor Deceleration and Neck Loads

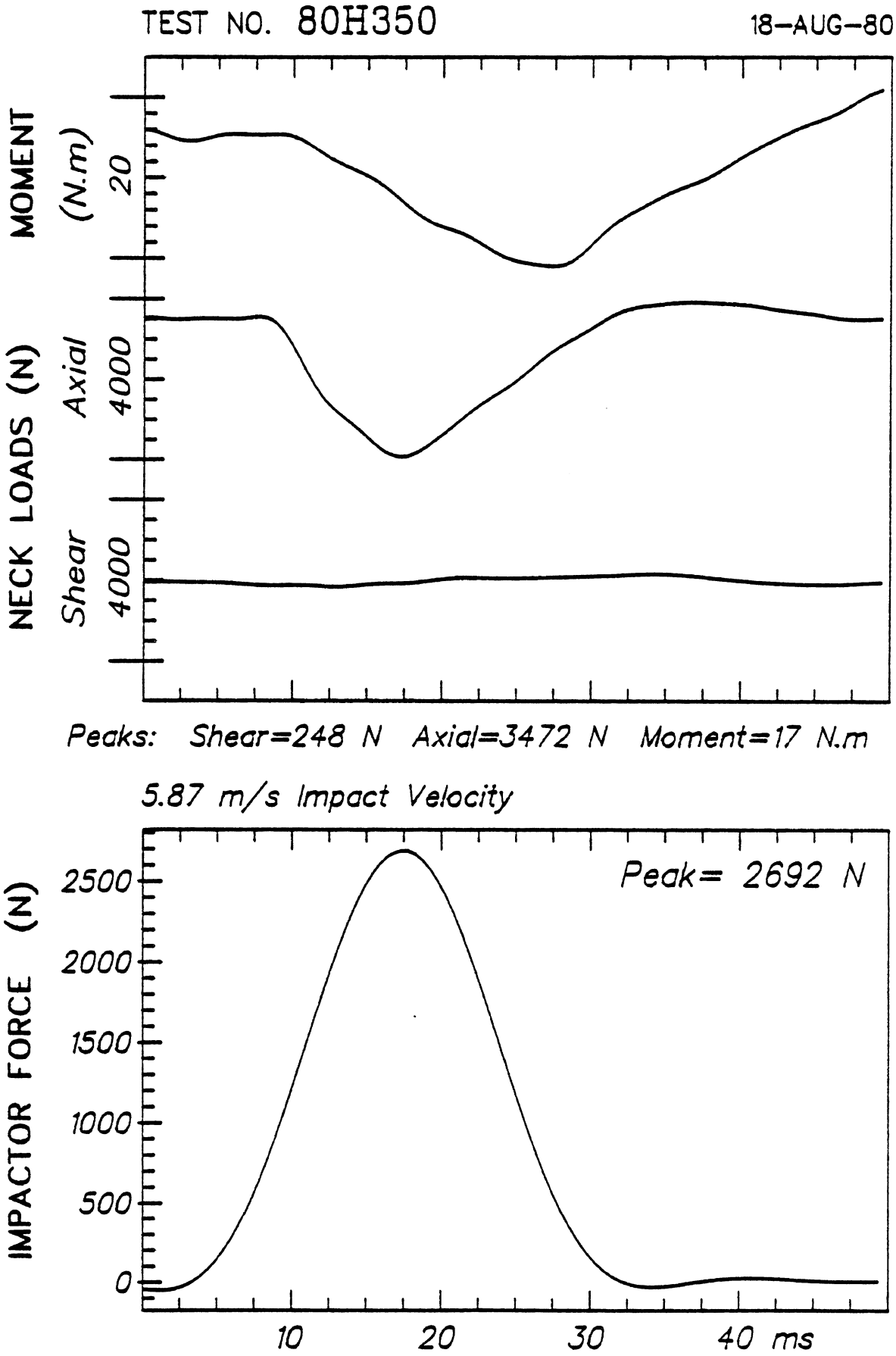


Figure 142: Impactor Deceleration and Neck Loads

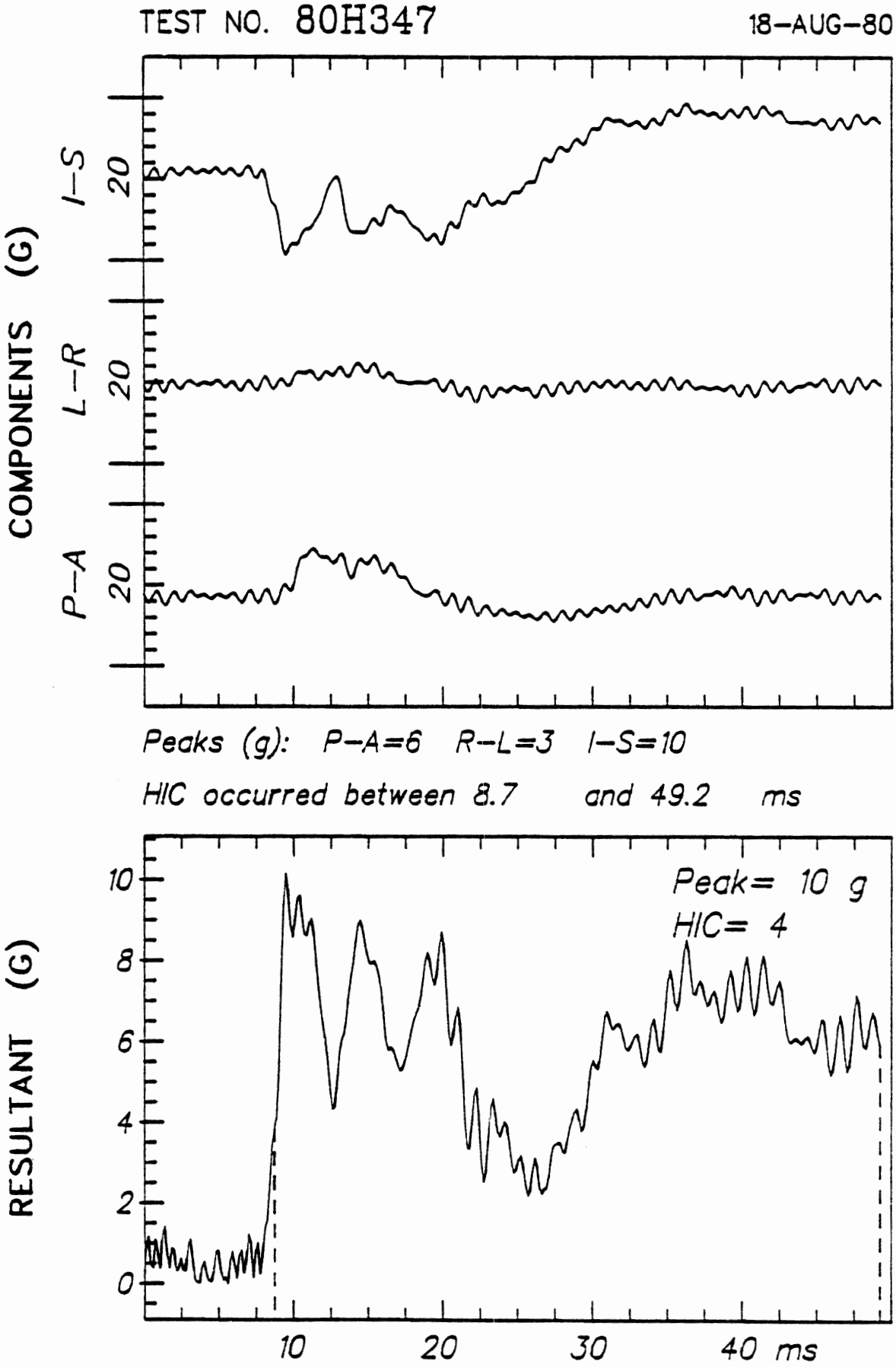


Figure 143: Head Acceleration vs. Time

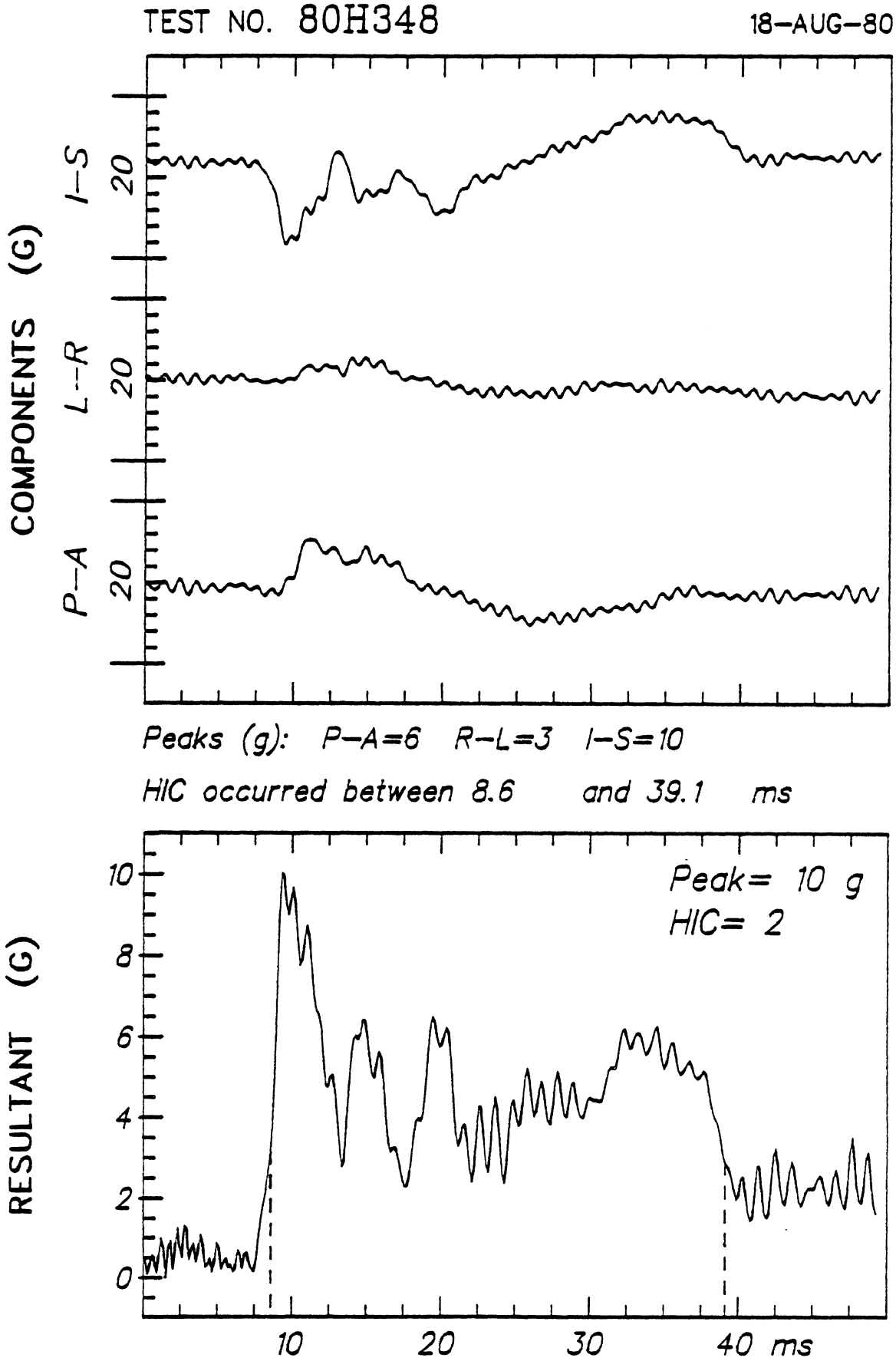


Figure 144: Head Acceleration vs. Time

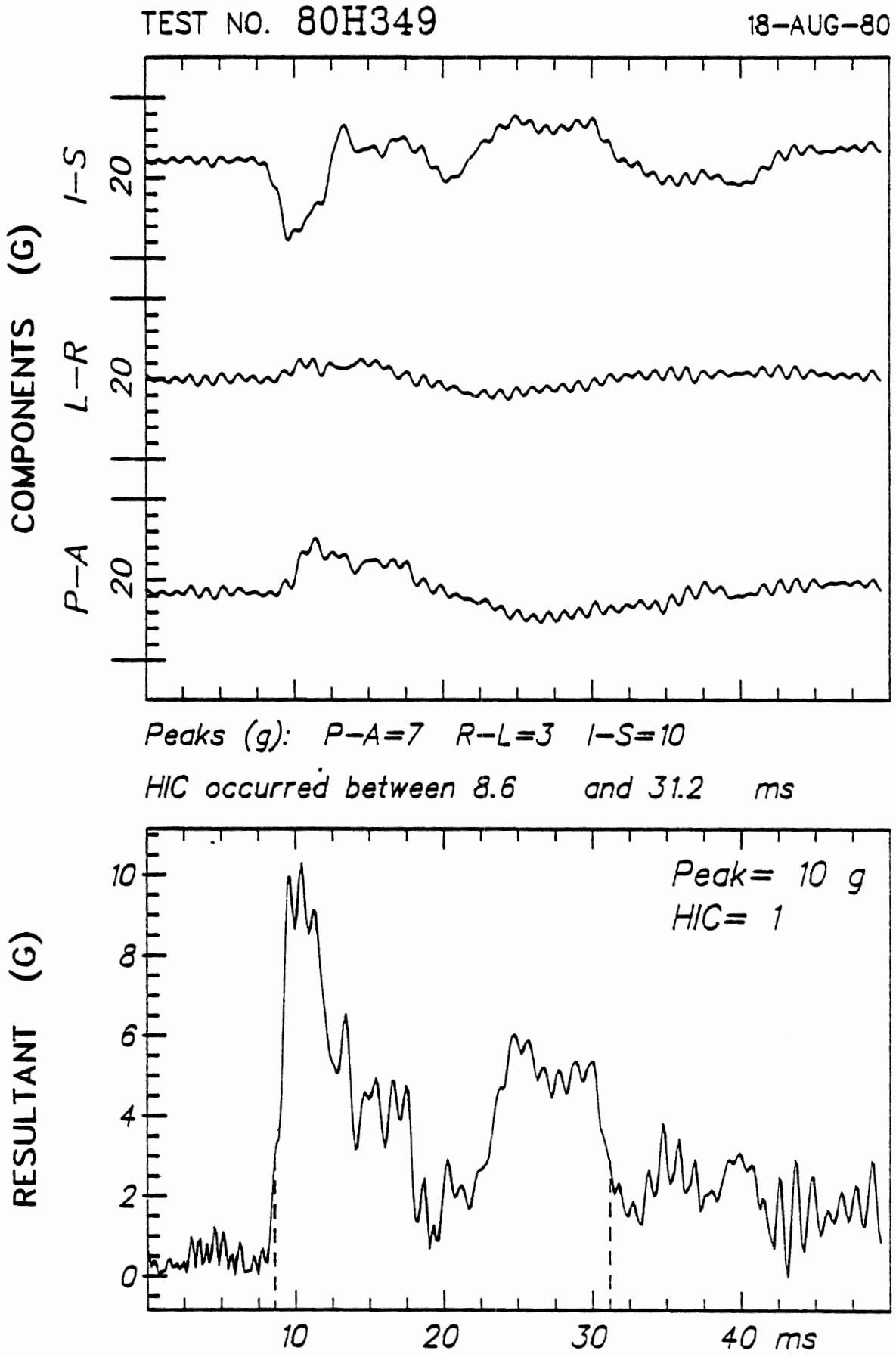


Figure 145: Head Acceleration vs. Time

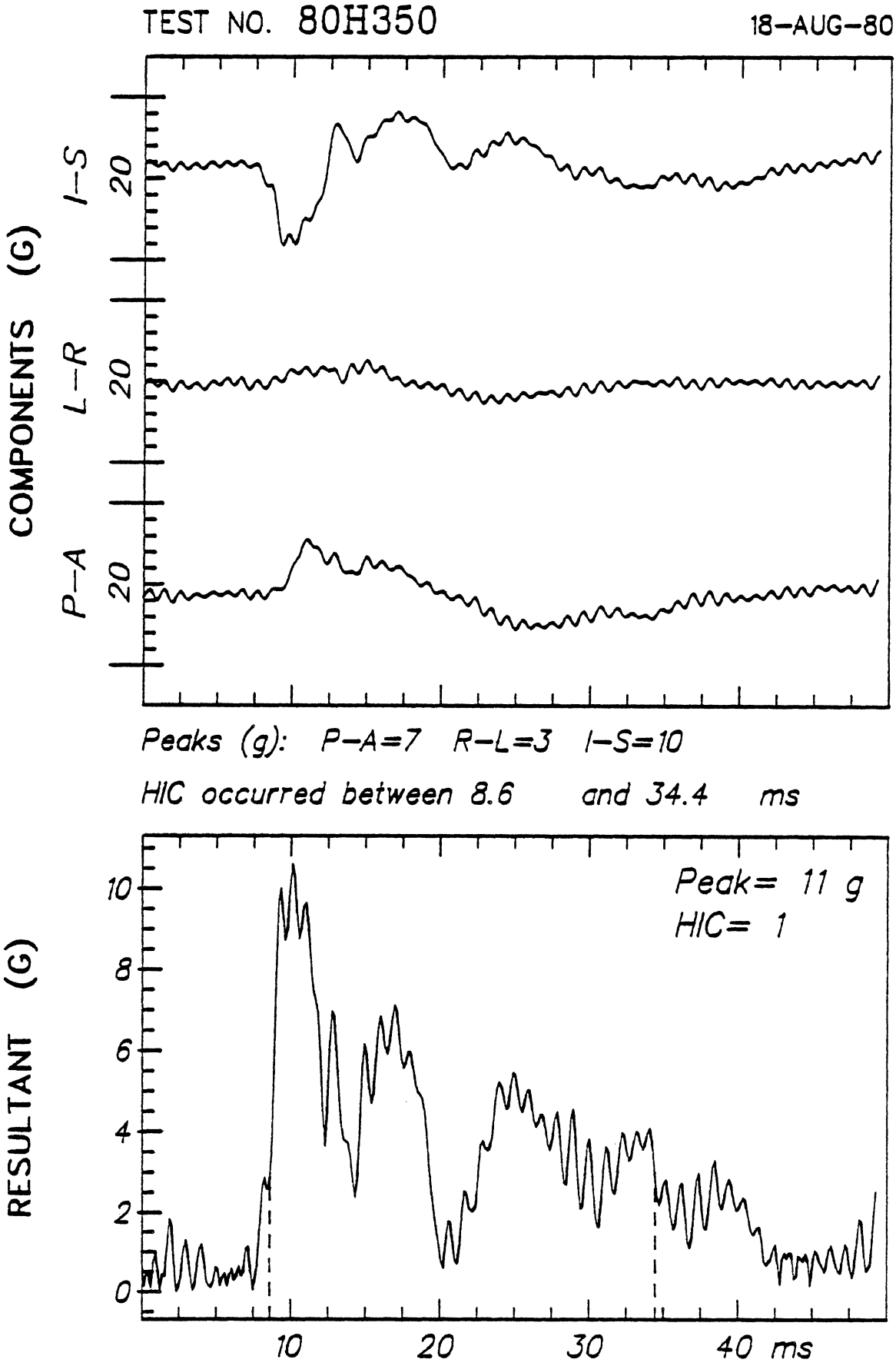


Figure 146: Head Acceleration vs. Time

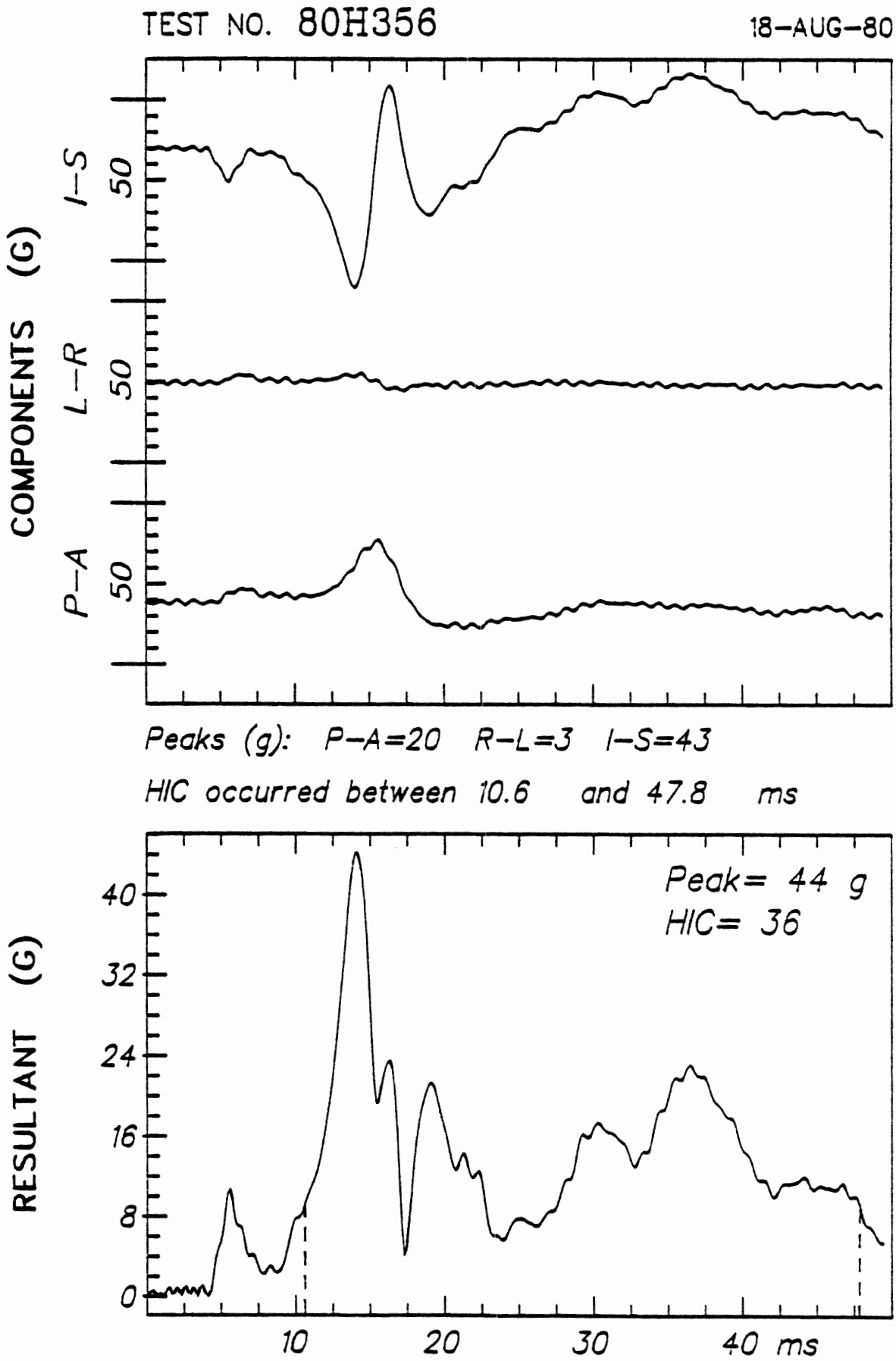


Figure 147: Head Acceleration vs. Time

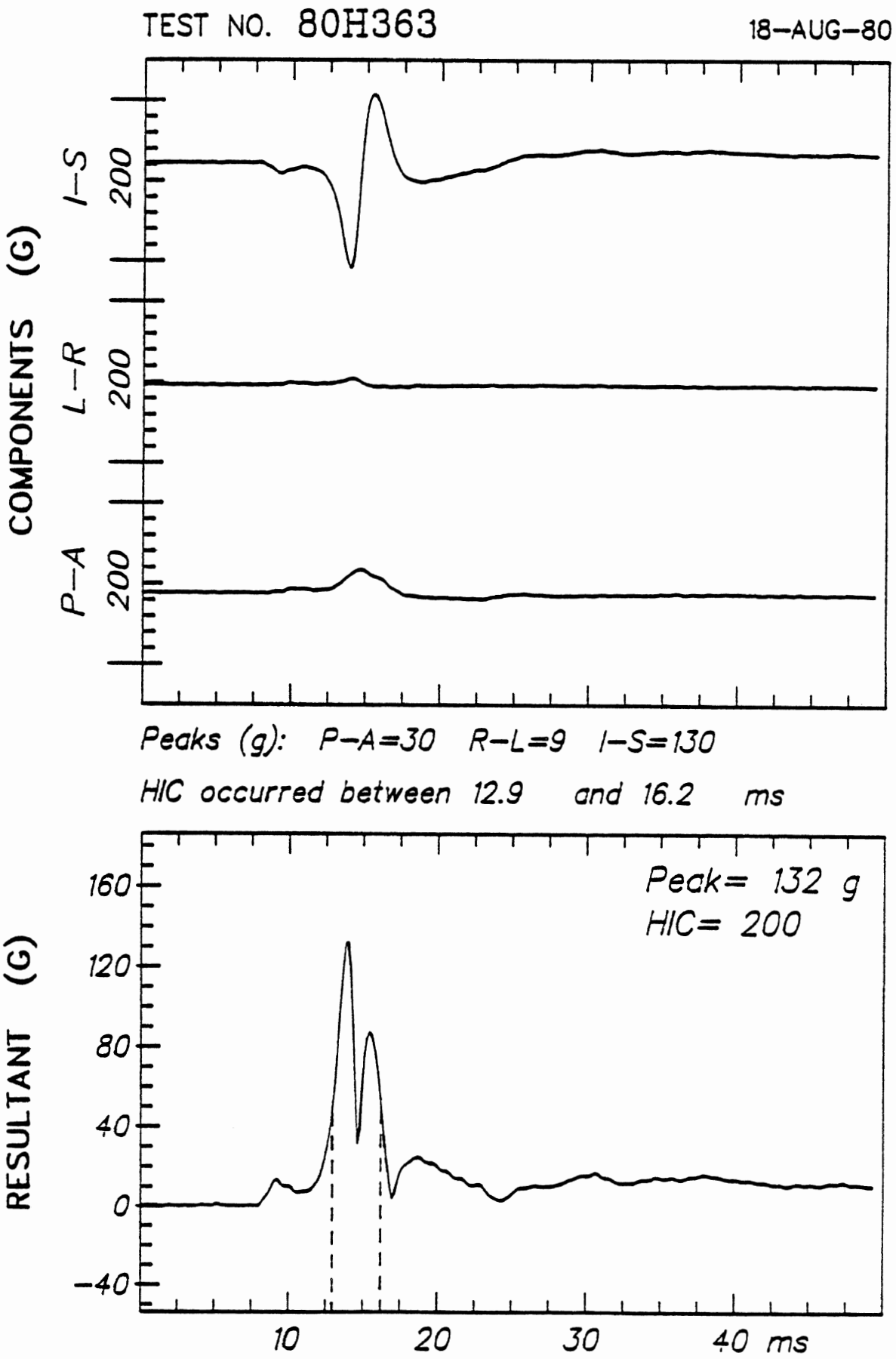


Figure 148: Head Acceleration vs. Time

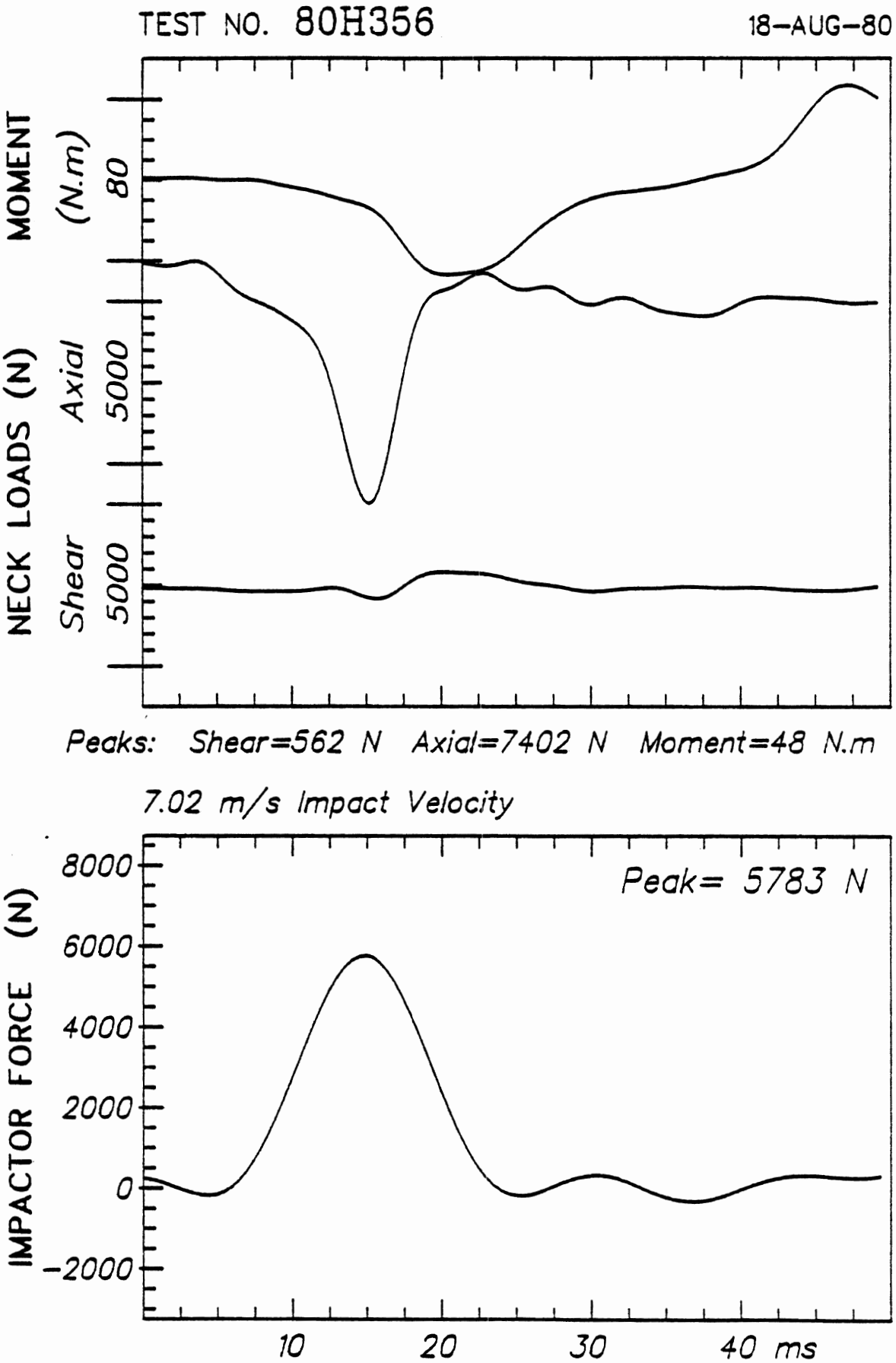


Figure 149: Impactor Deceleration and Neck Loads

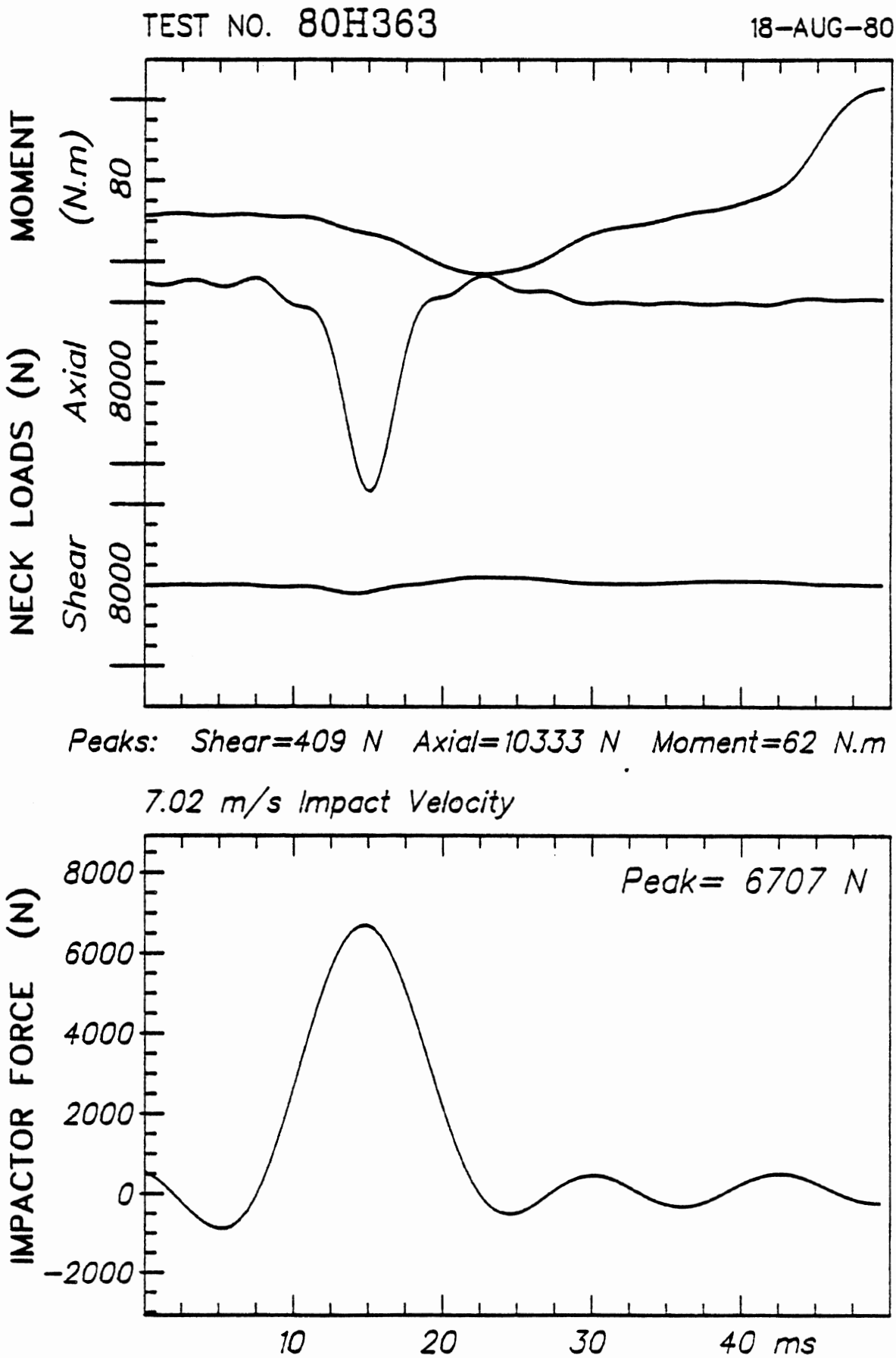


Figure 150: Impactor Deceleration and Neck Loads

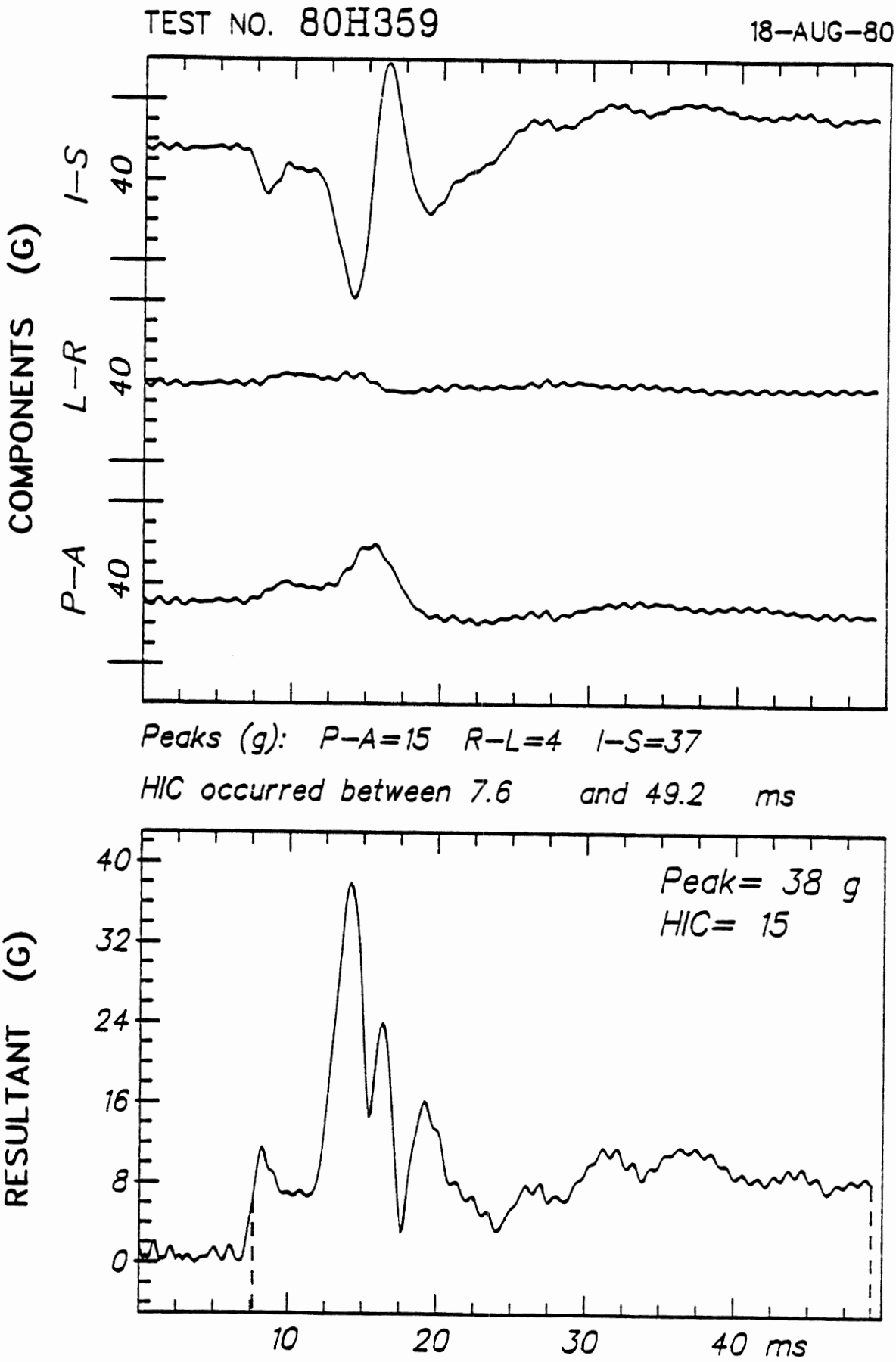


Figure 151: Head Acceleration vs. Time

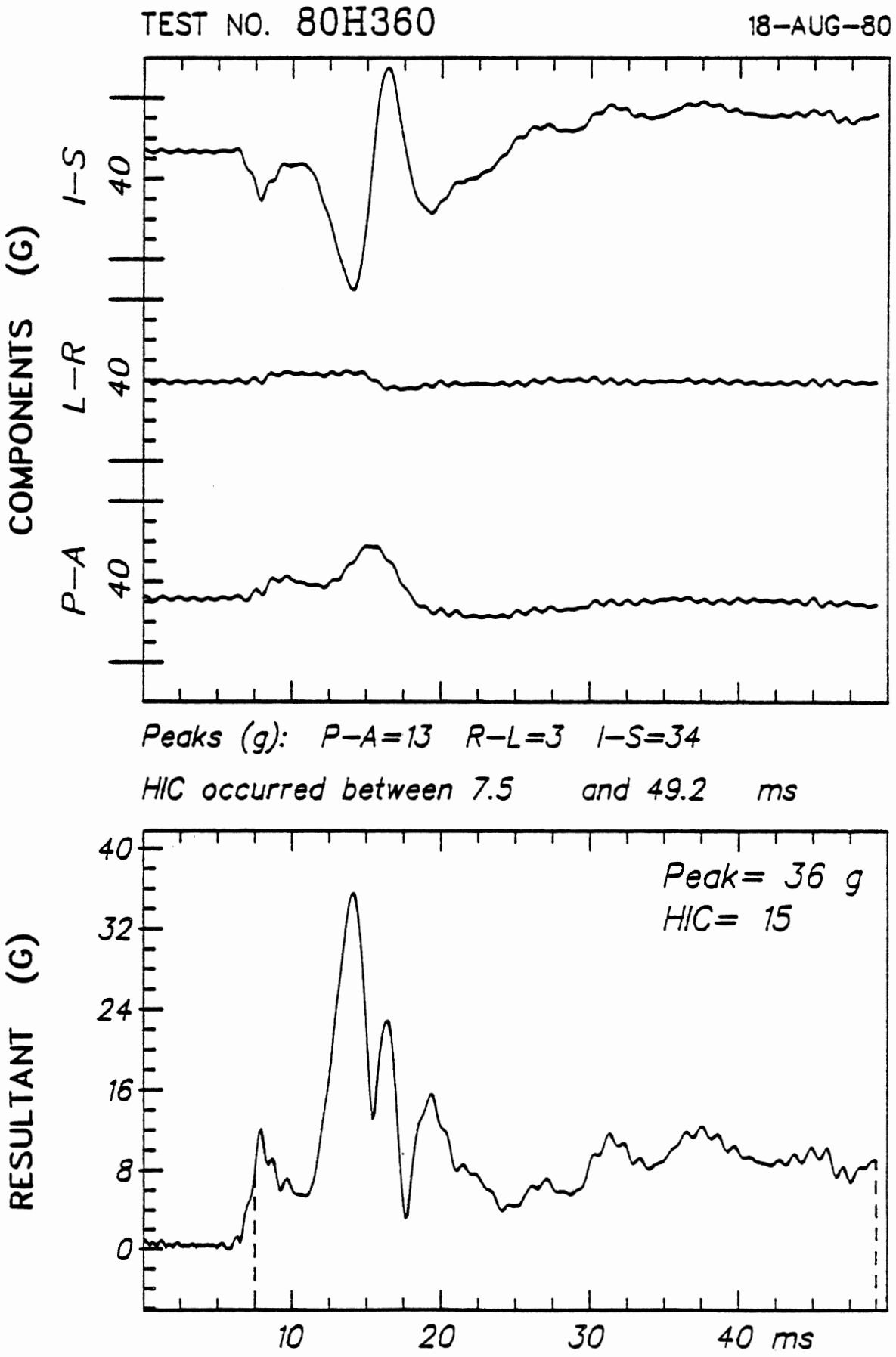


Figure 152: Head Acceleration vs. Time

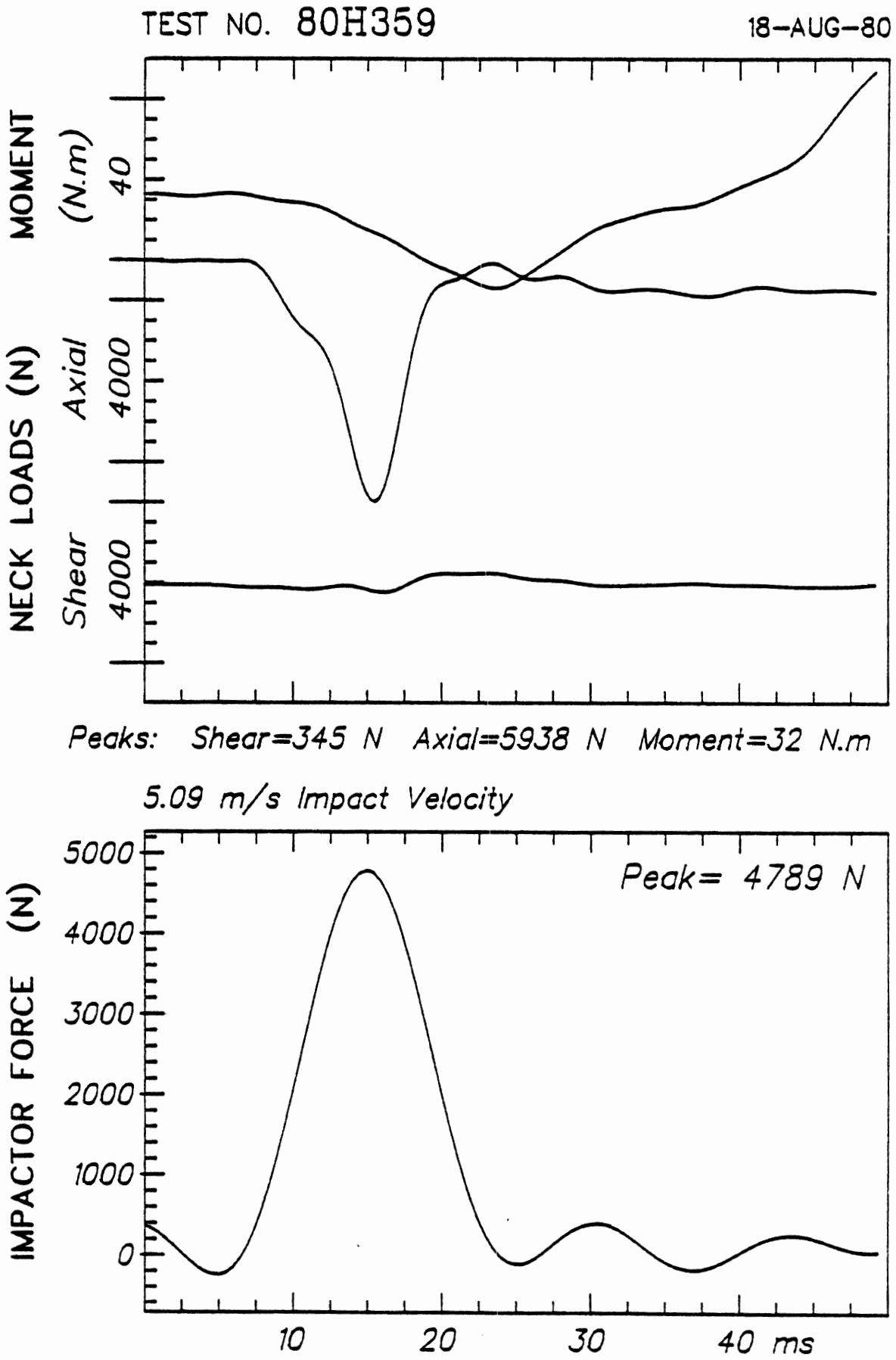


Figure 153: Impactor Deceleration and Neck Loads

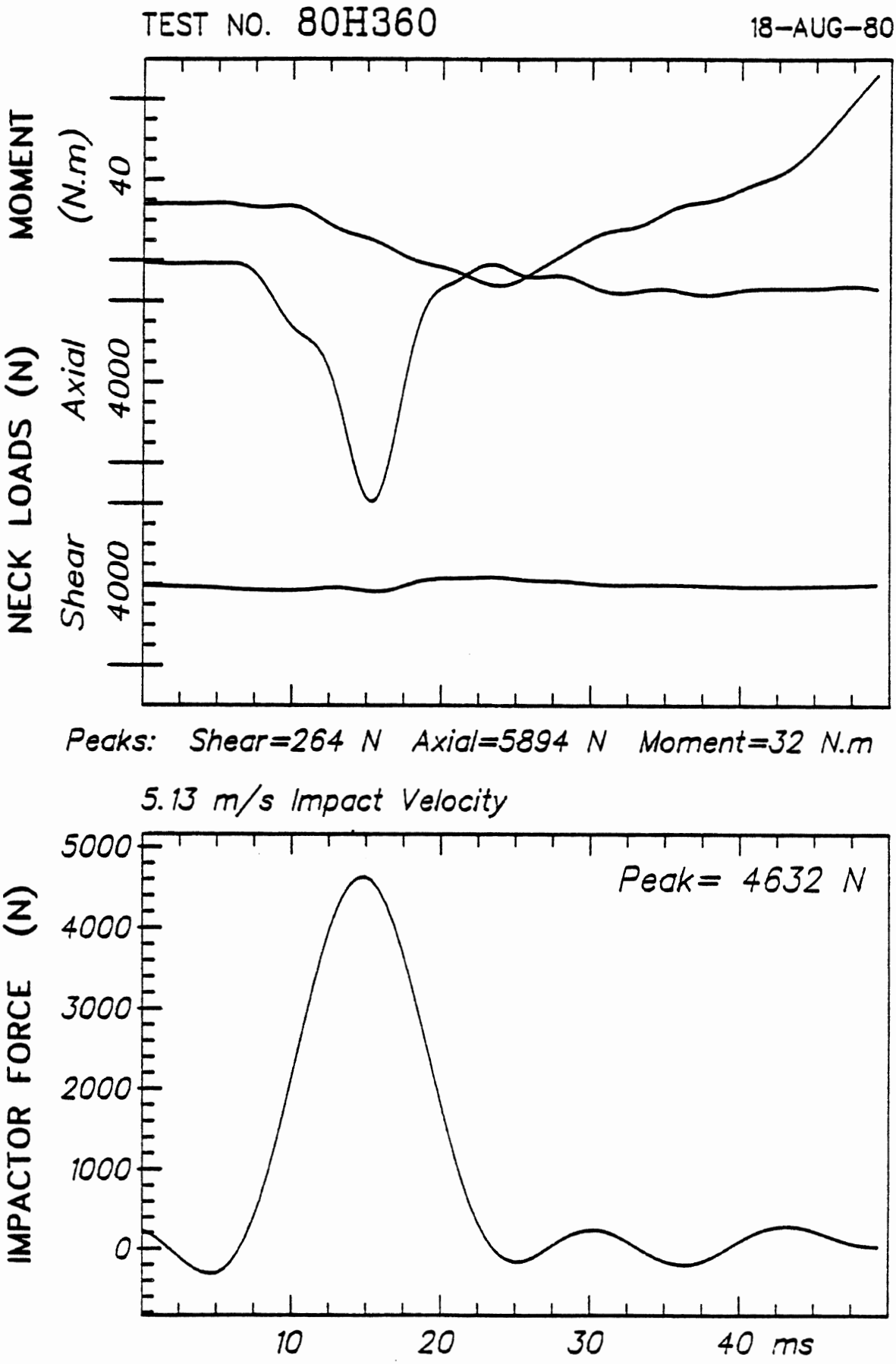


Figure 154: Impactor Deceleration and Neck Loads

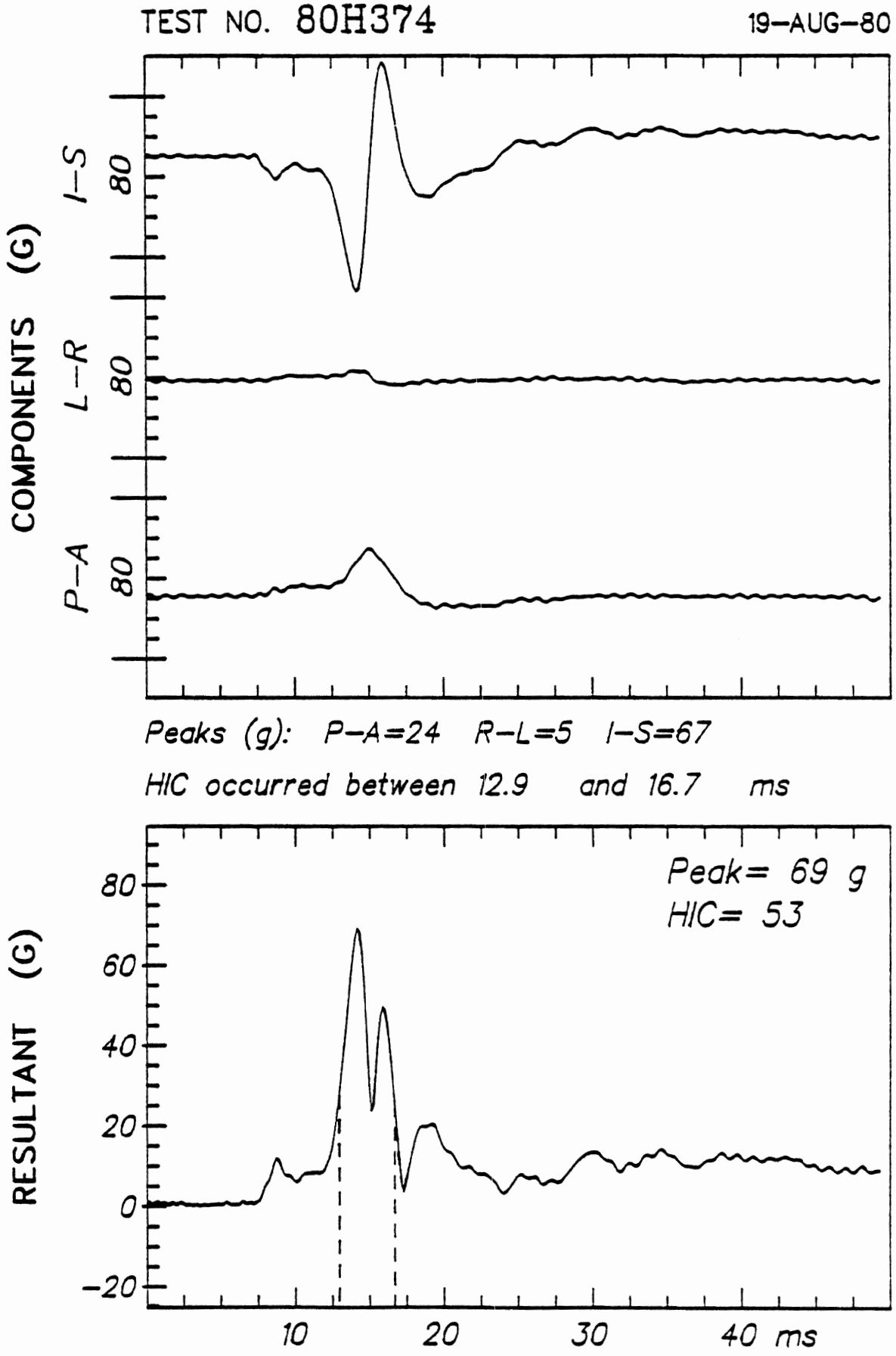


Figure 155: Head Acceleration vs. Time

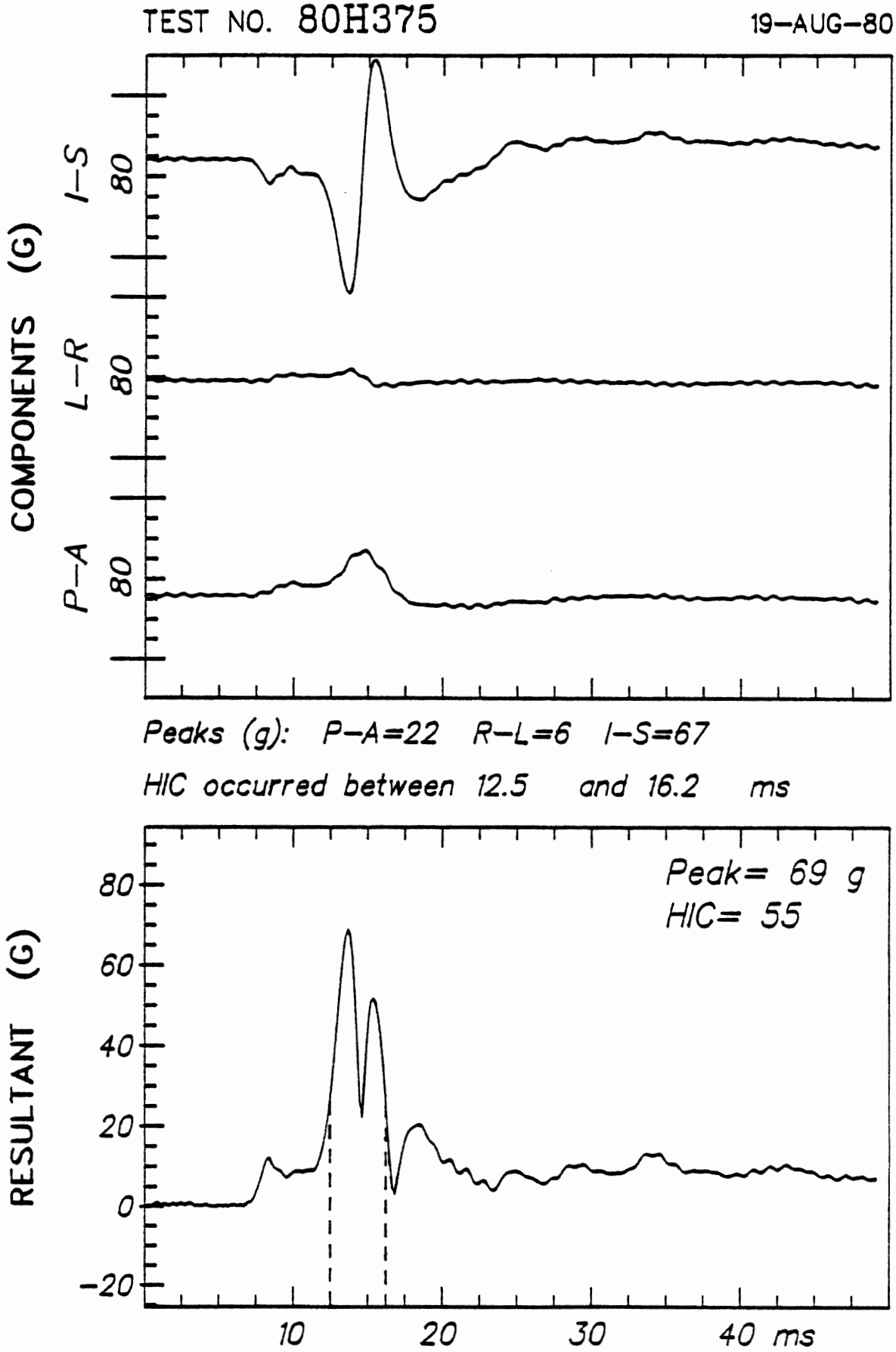


Figure 156: Head Acceleration vs. Time

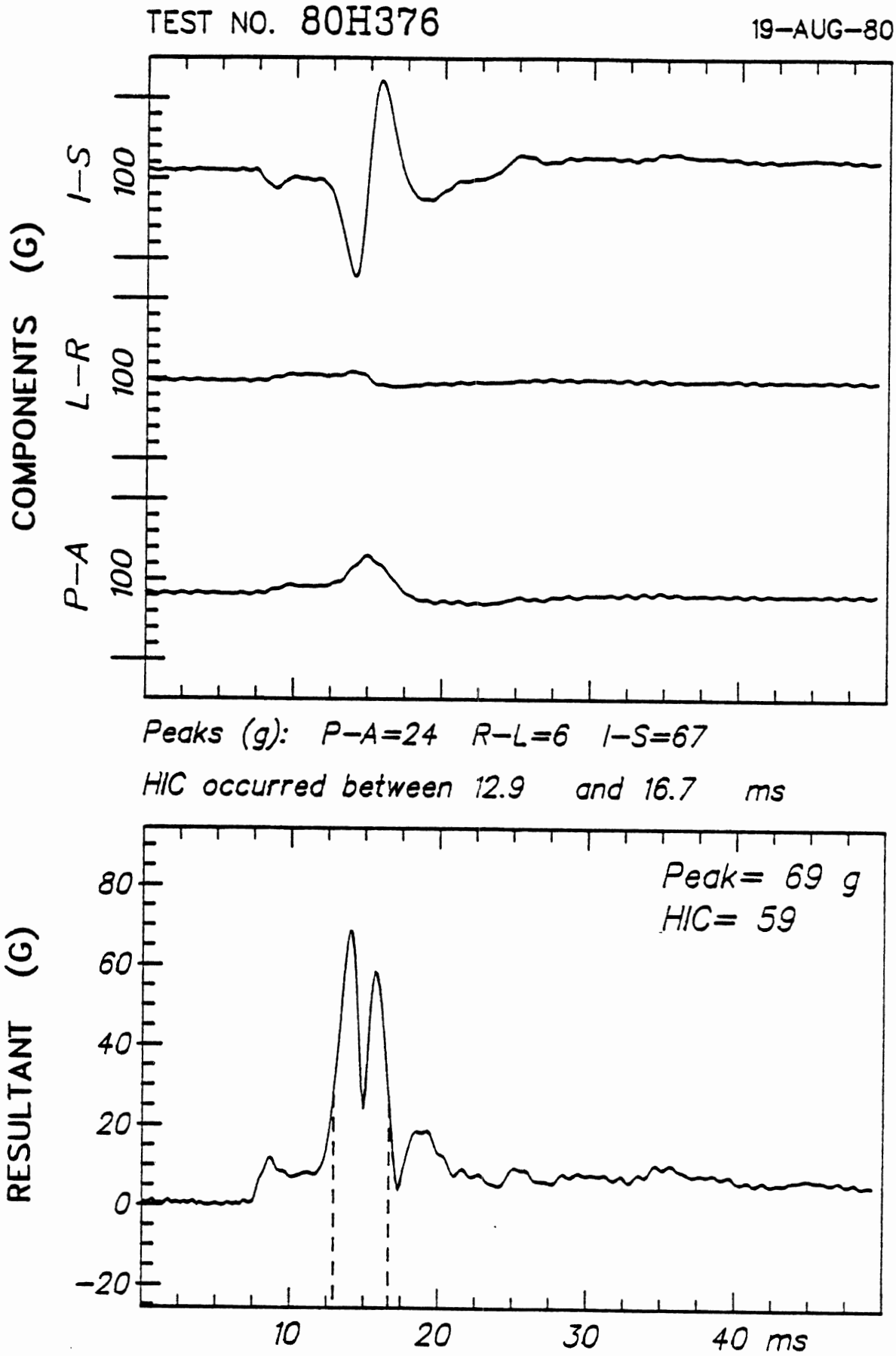


Figure 157: Head Acceleration vs. Time

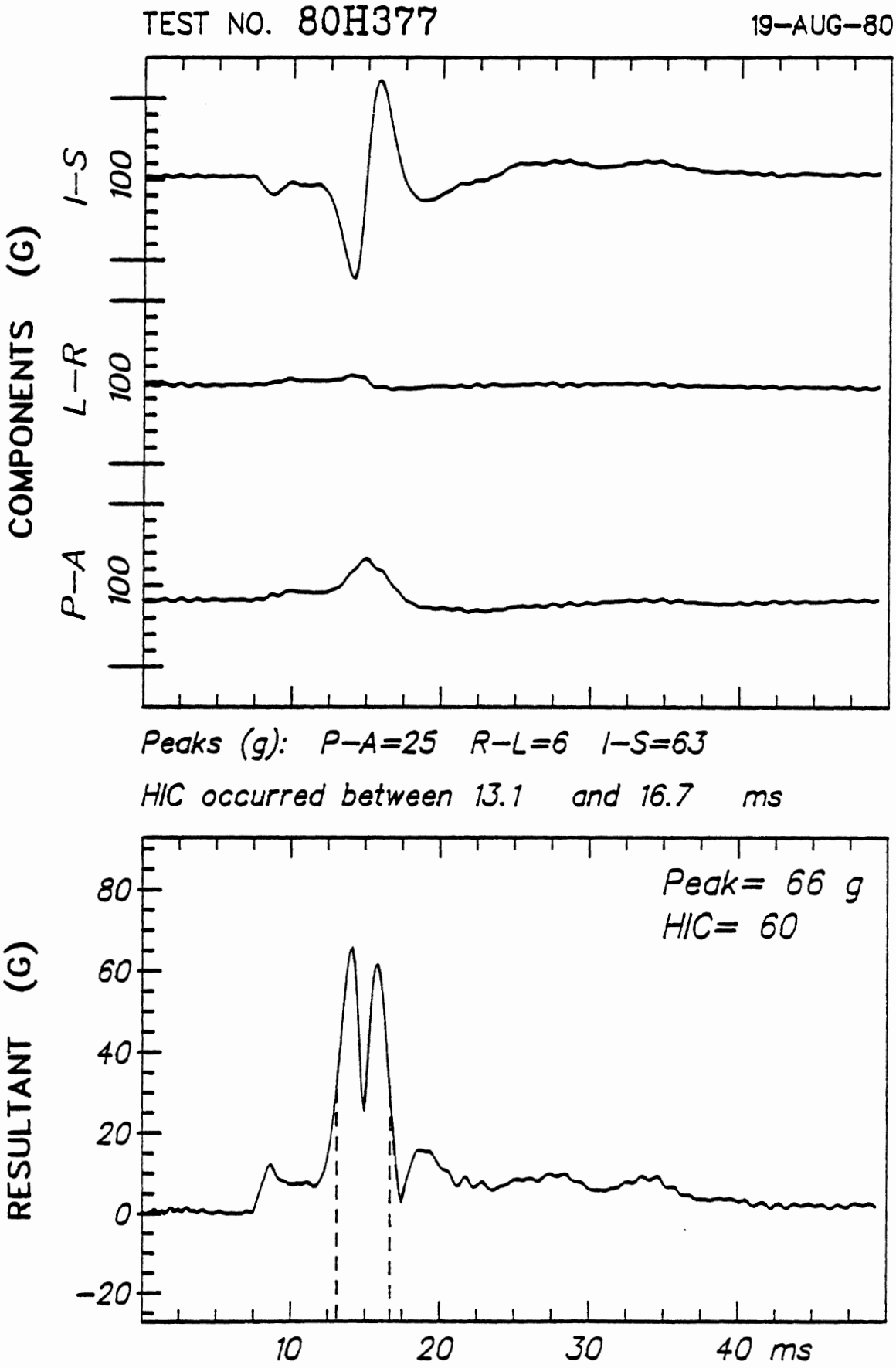


Figure 158: Head Acceleration vs. Time

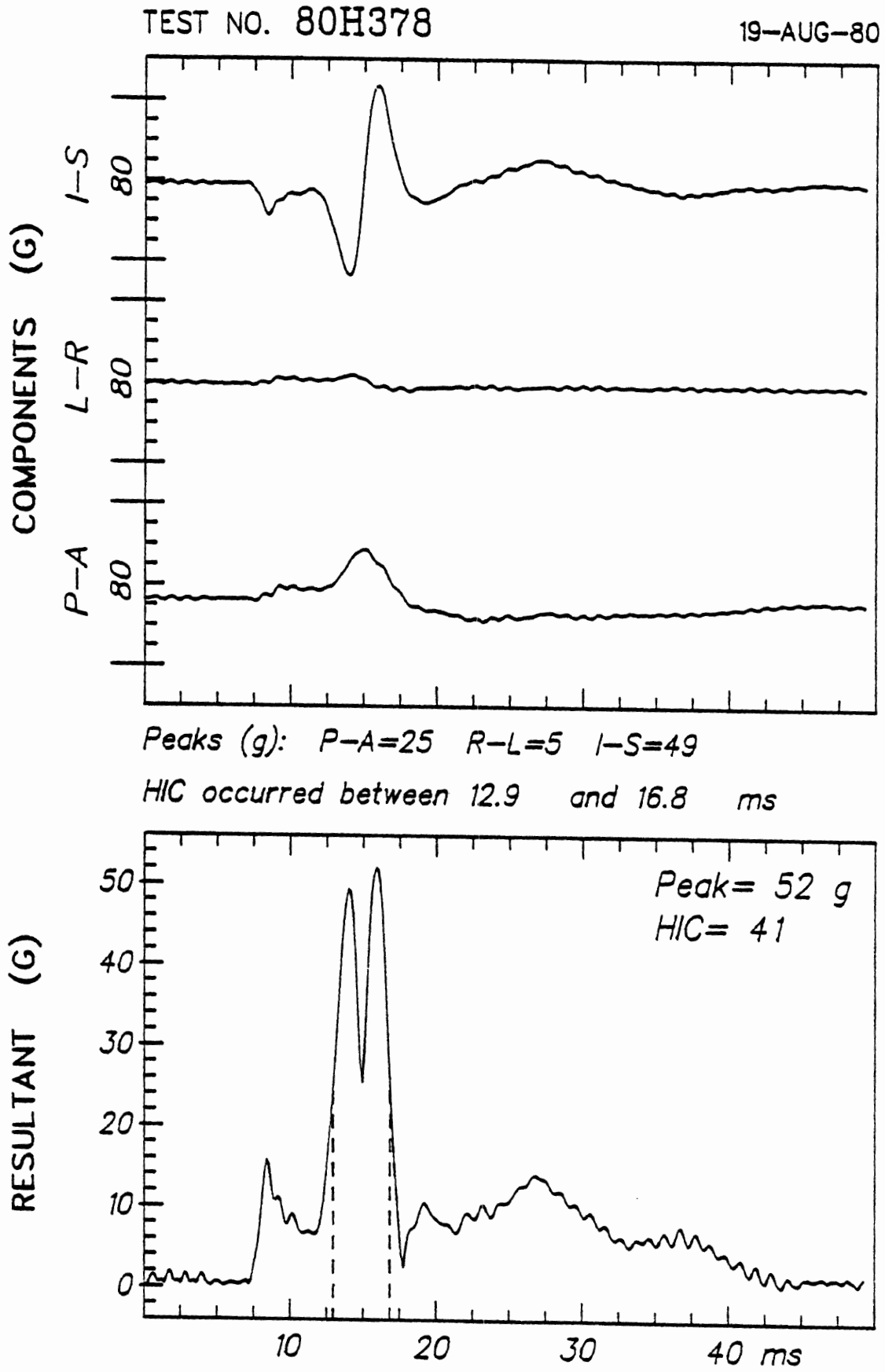


Figure 159: Head Acceleration vs. Time

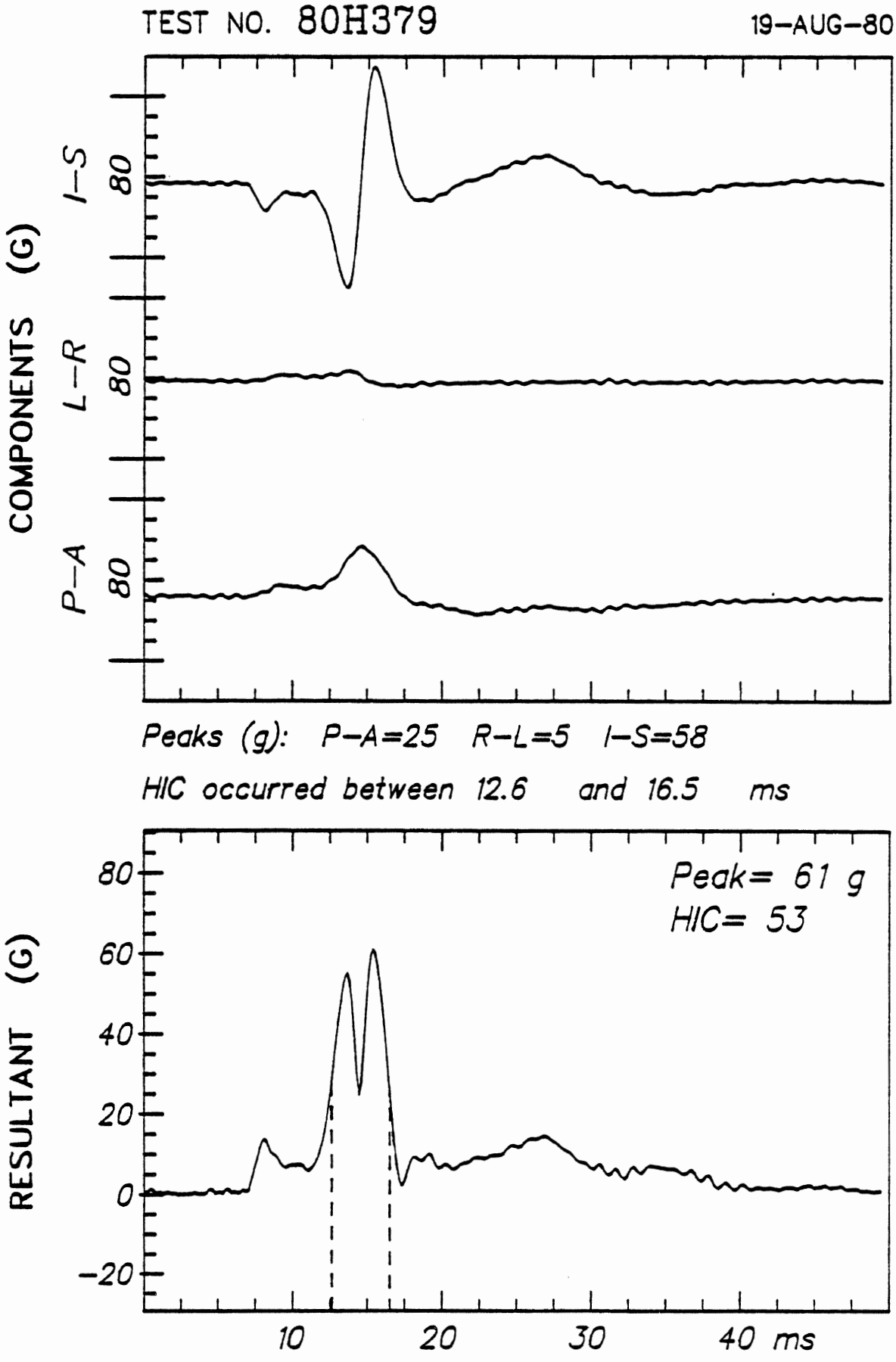


Figure 160: Head Acceleration vs. Time

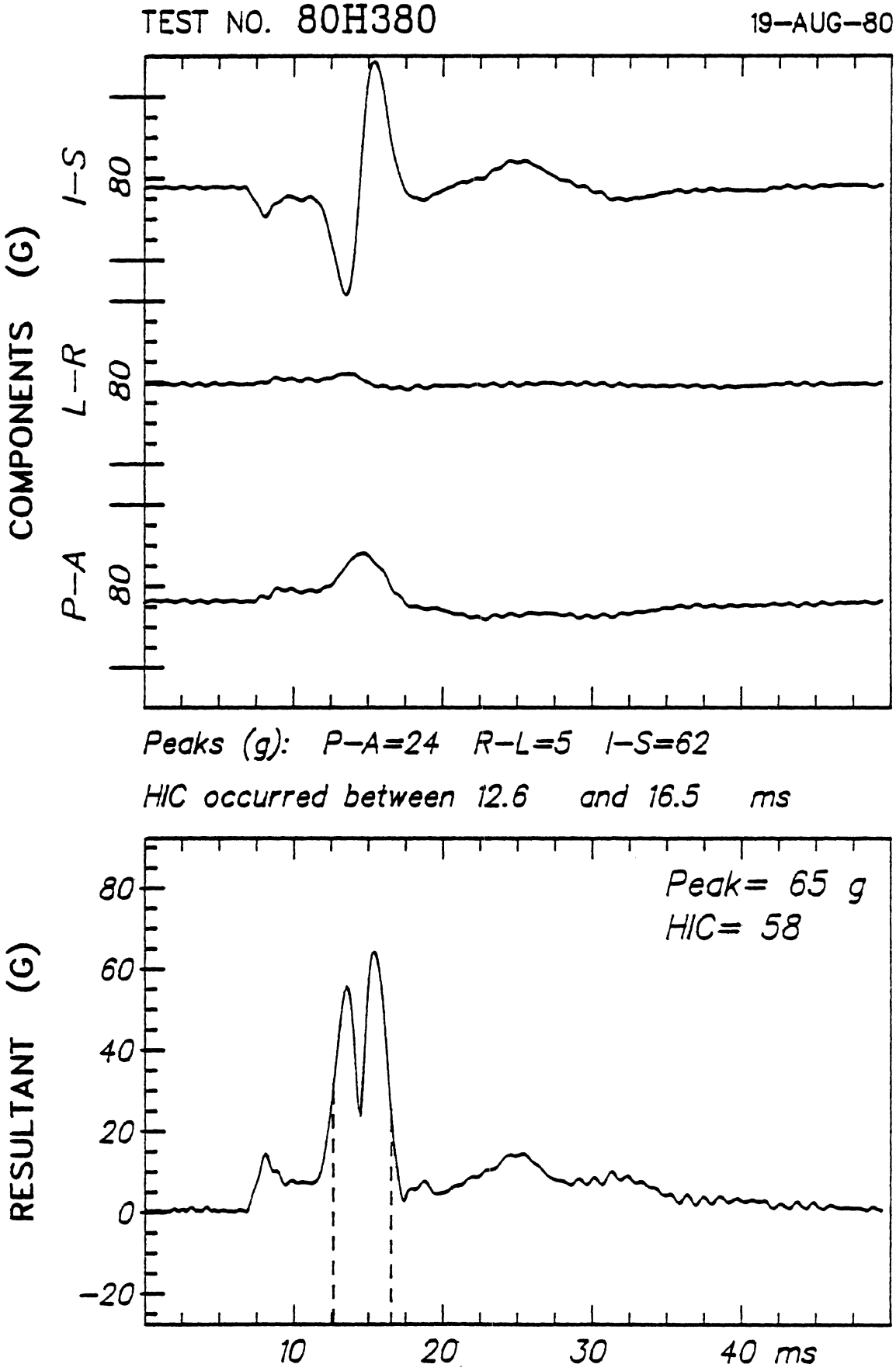
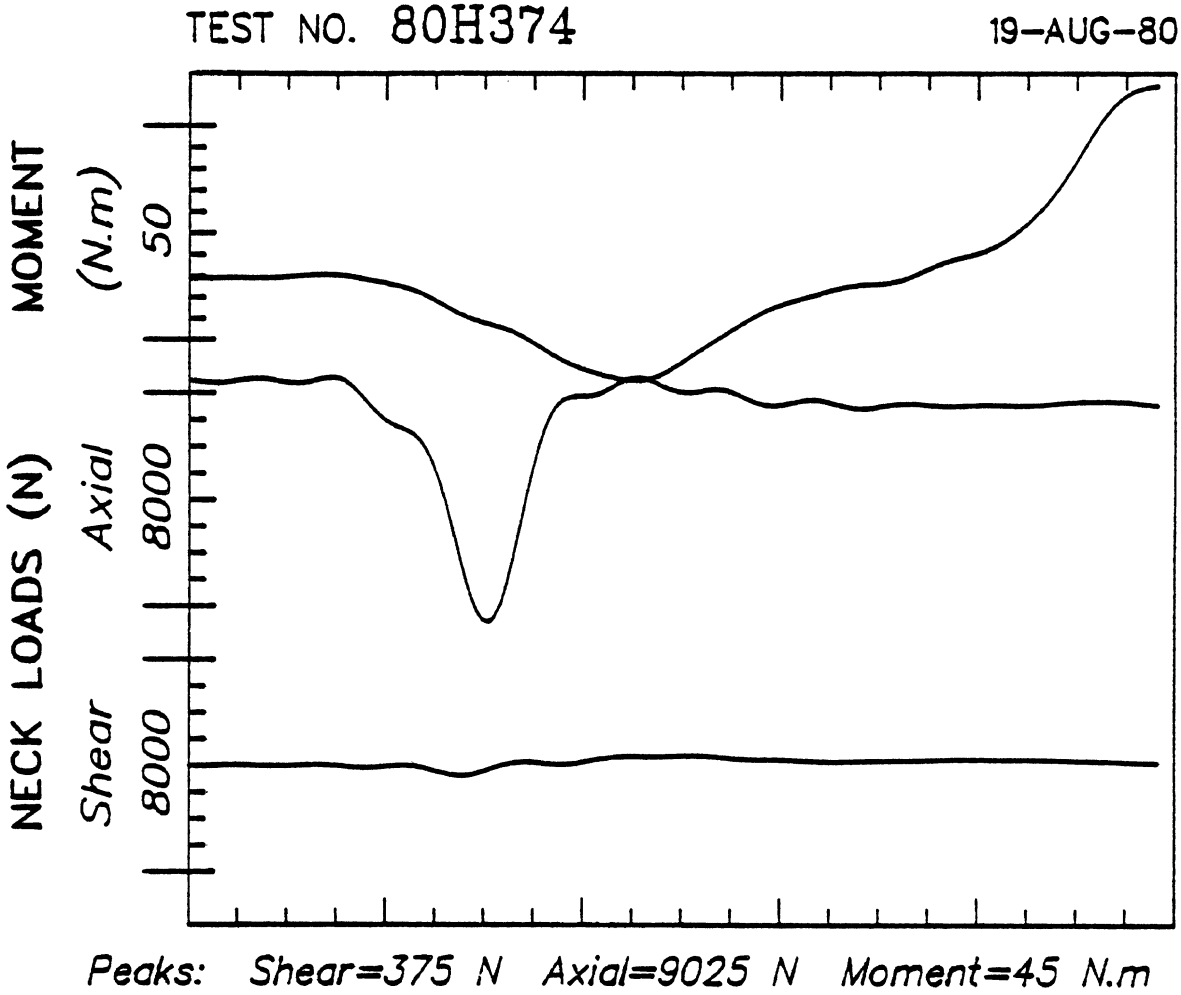


Figure 161: Head Acceleration vs. Time



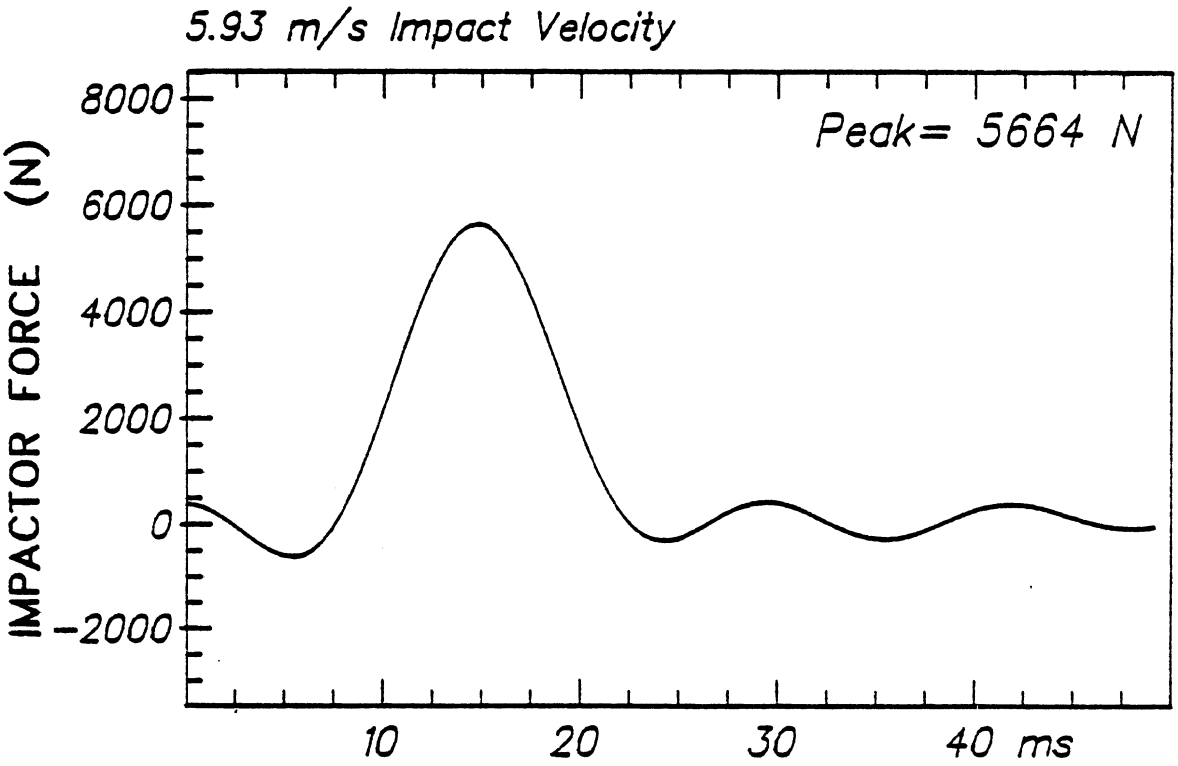


Figure 162: Impactor Deceleration and Neck Loads

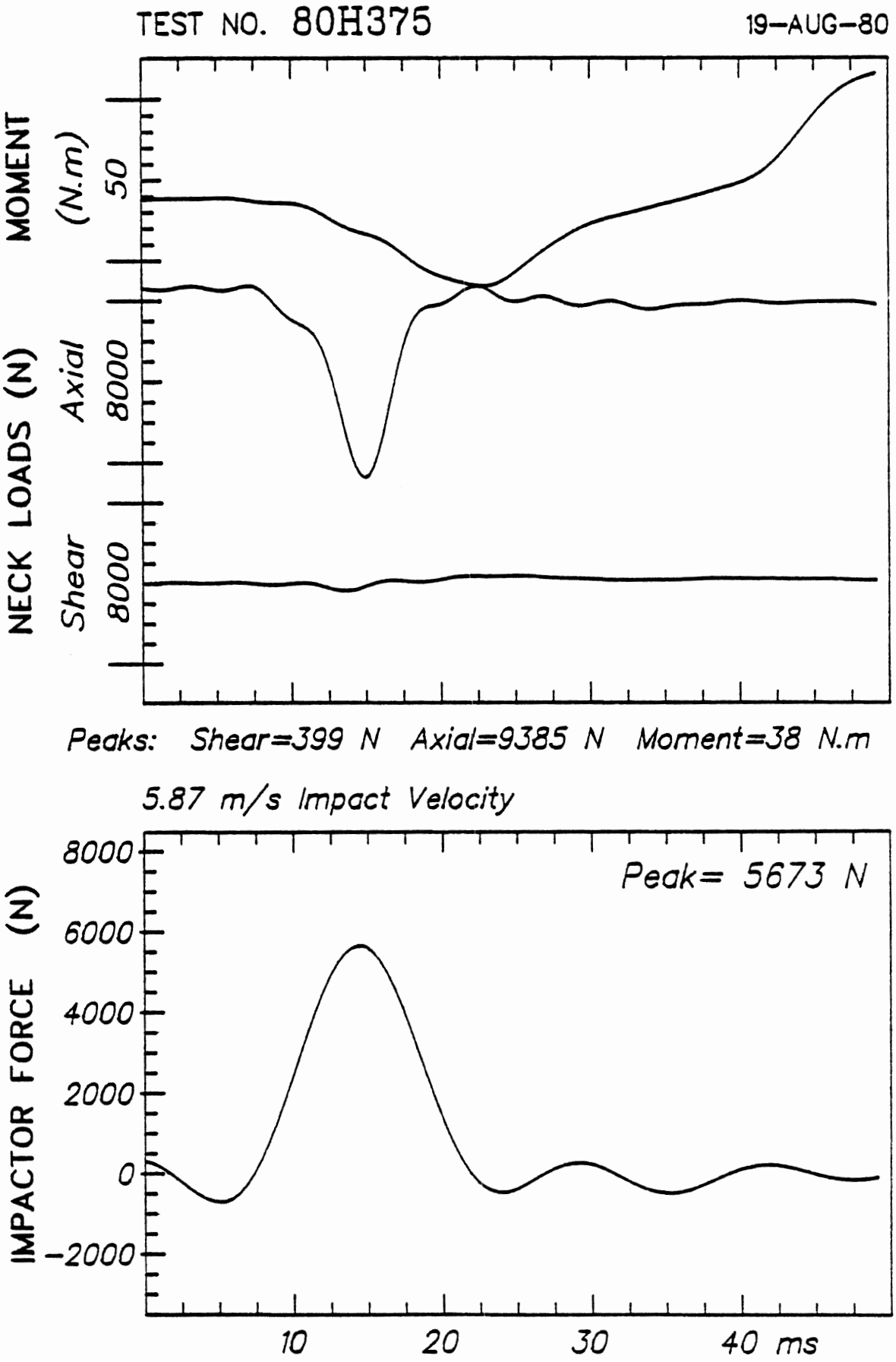


Figure 163: Impactor Deceleration and Neck Loads

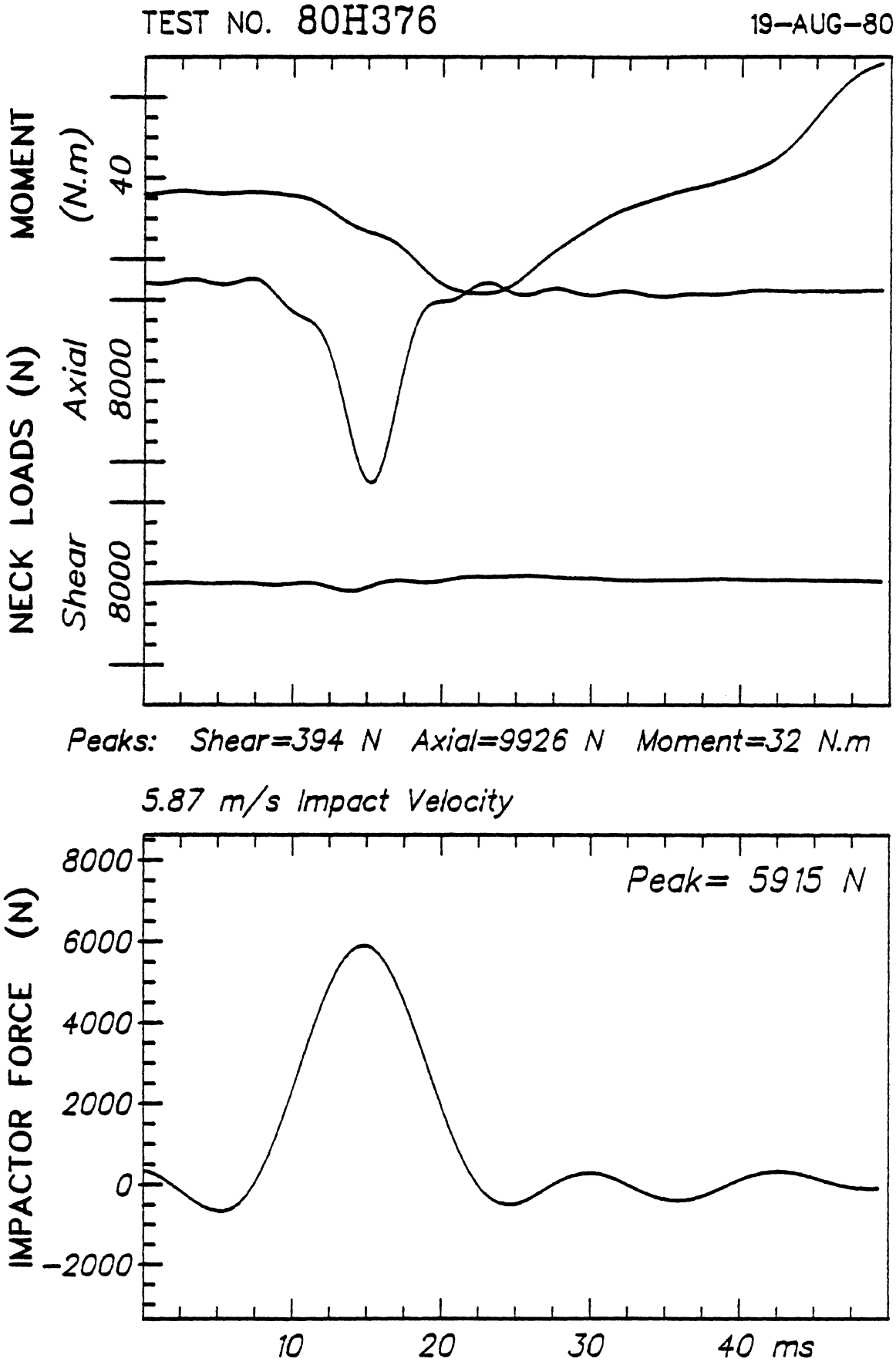


Figure 164: Impactor Deceleration and Neck Loads

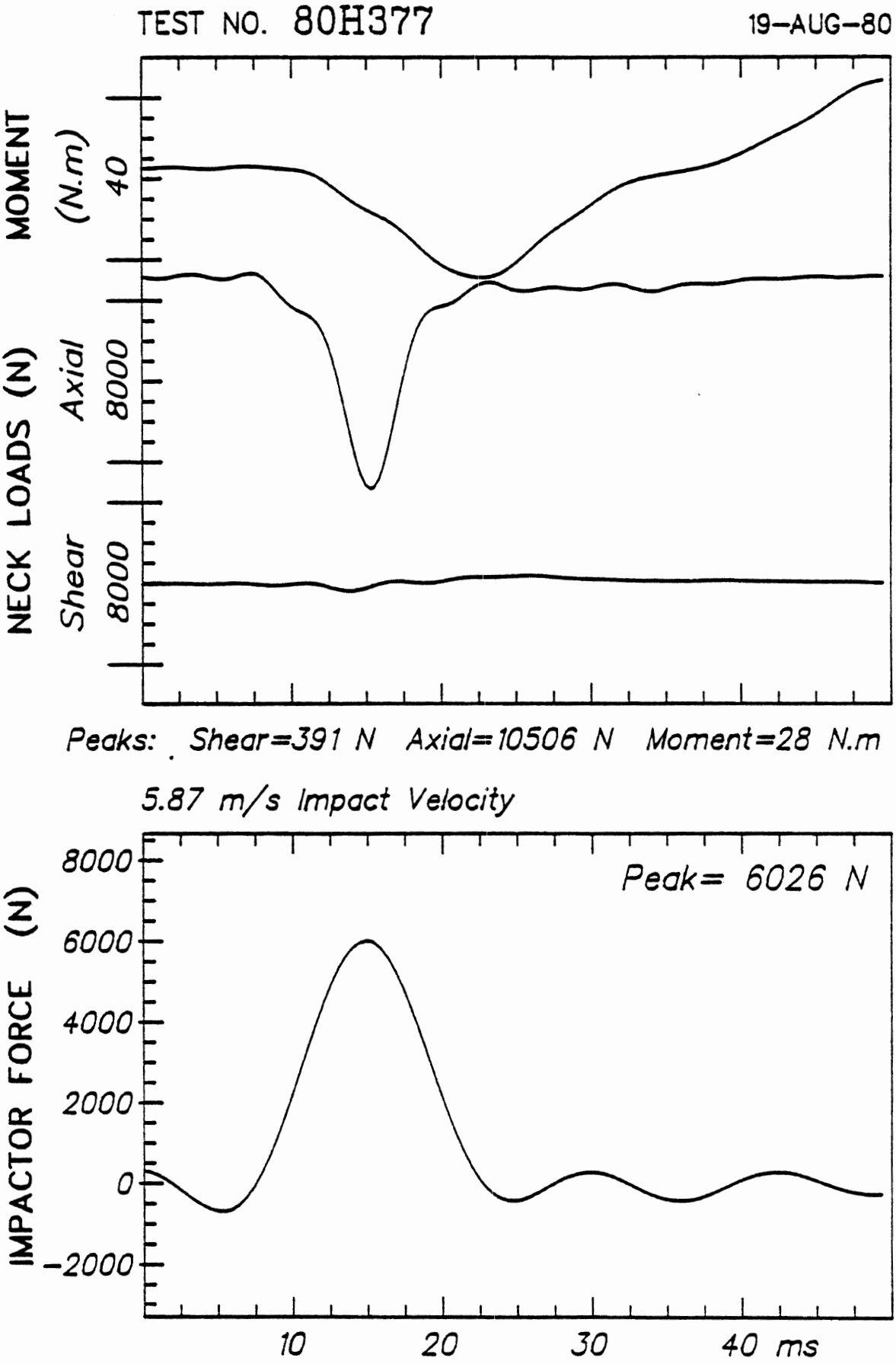


Figure 165: Impactor Deceleration and Neck Loads

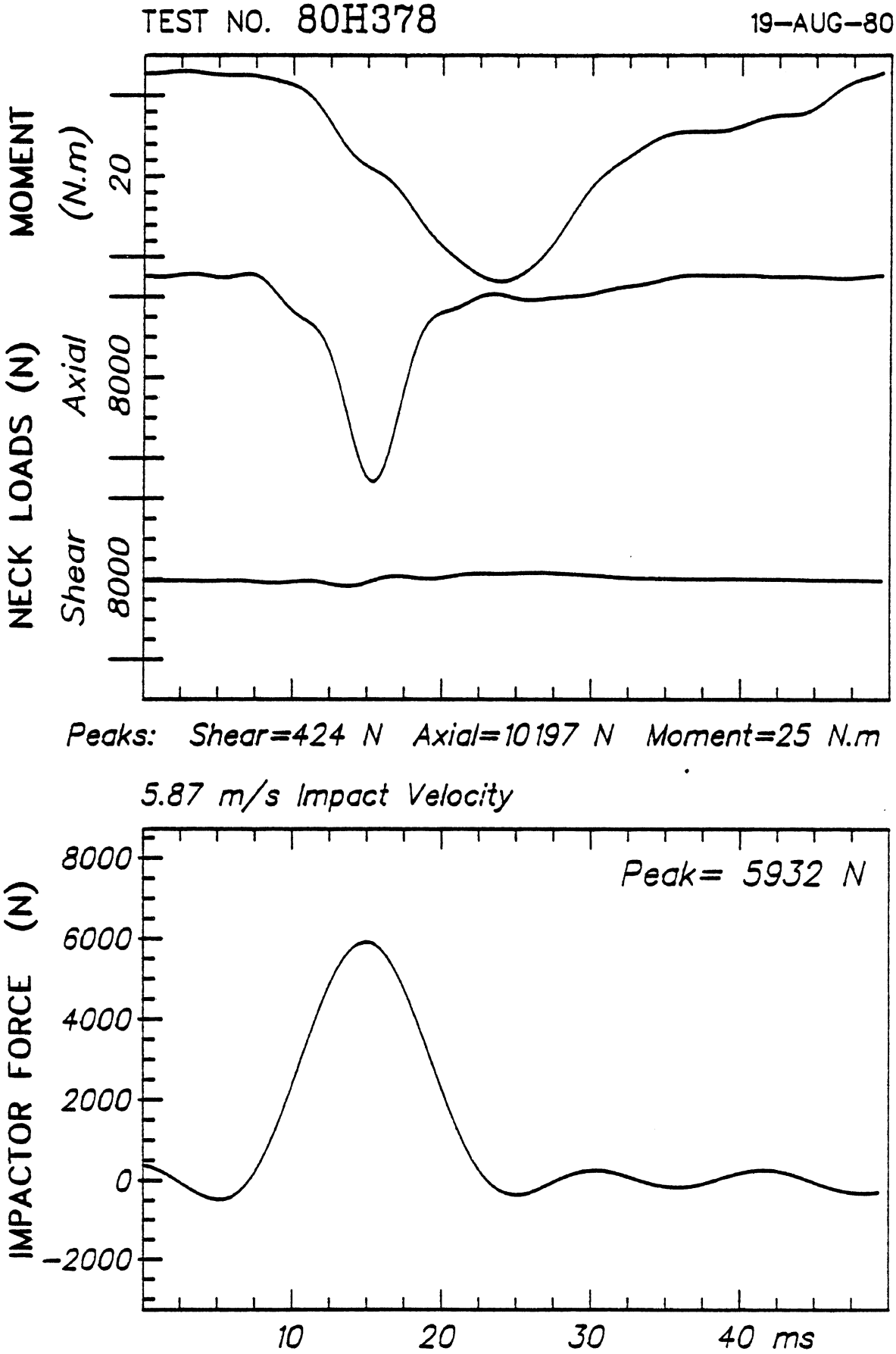


Figure 166: Impactor Deceleration and Neck Loads

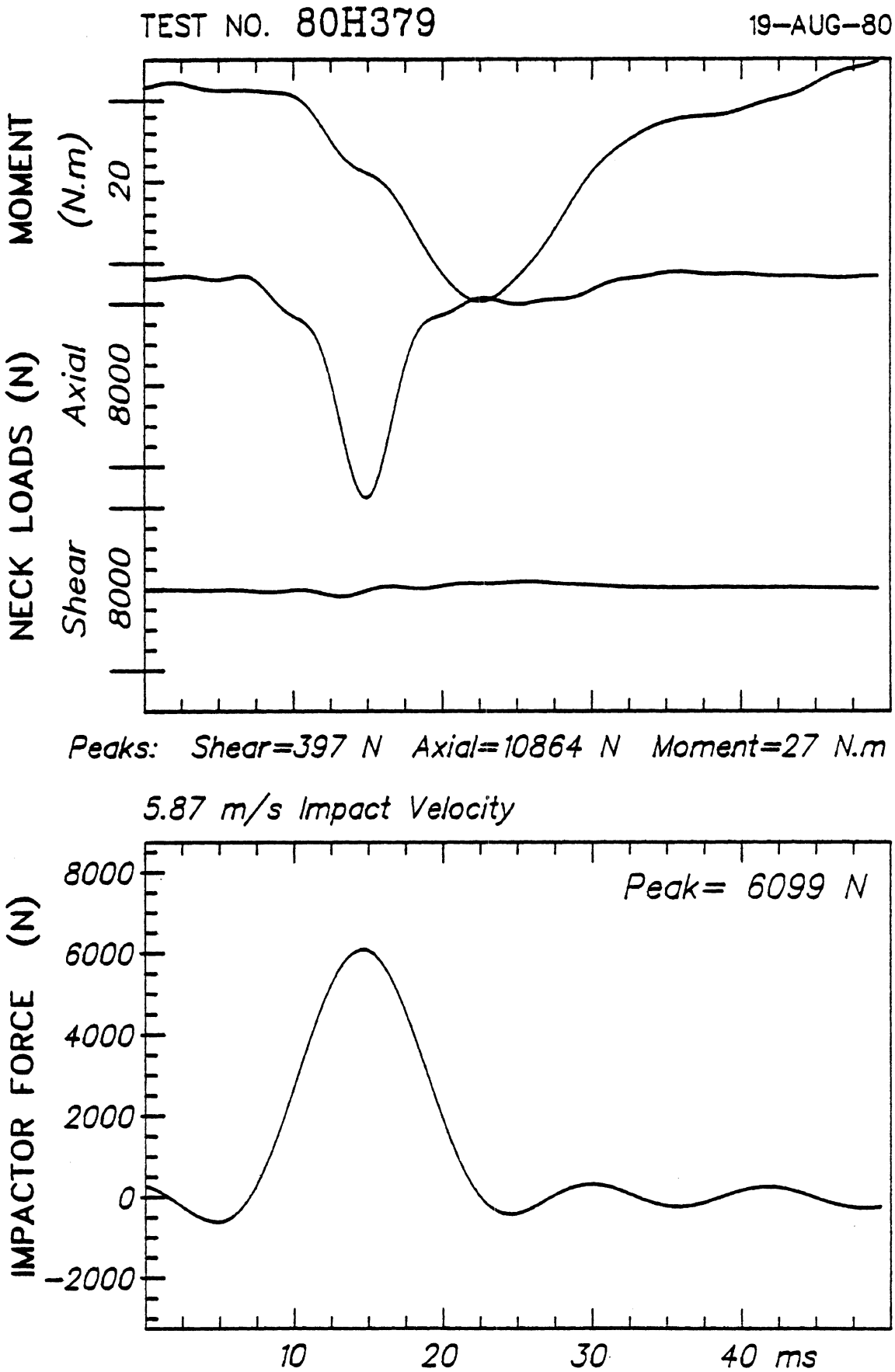


Figure 167: Impactor Deceleration and Neck Loads

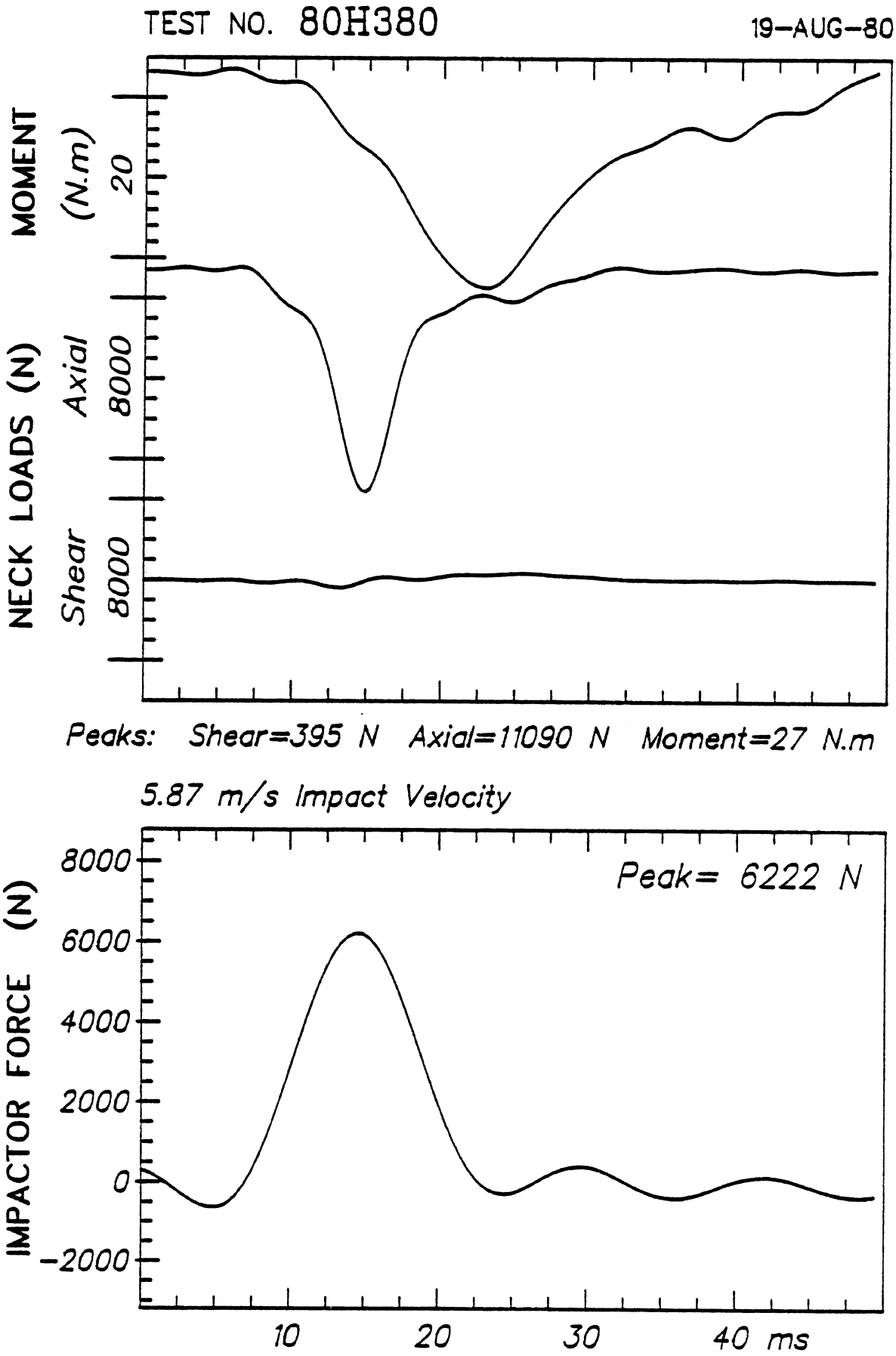


Figure 168: Impactor Deceleration and Neck Loads

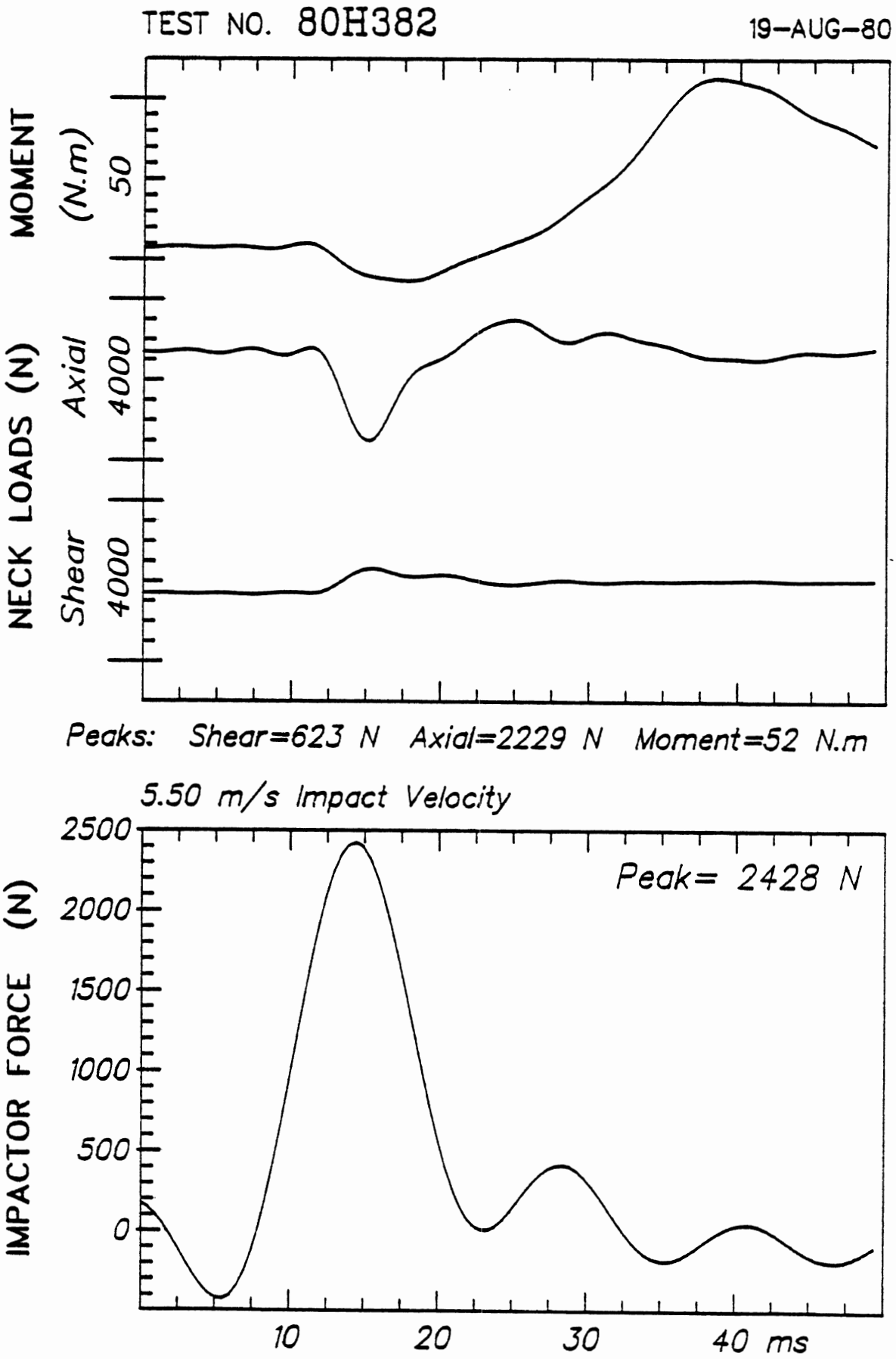


Figure 169: Impactor Deceleration and Neck Loads

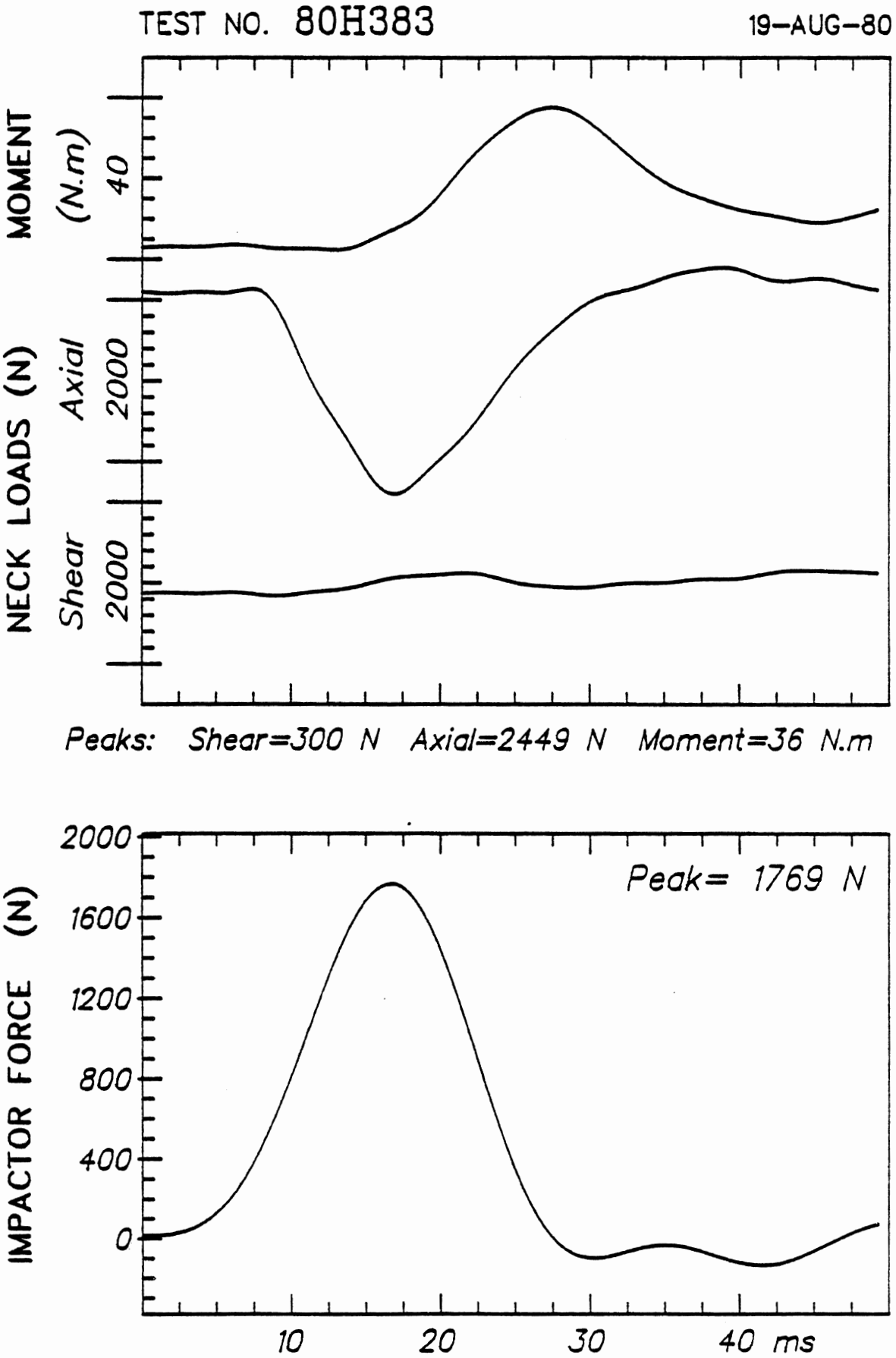


Figure 170: Impactor Deceleration and Neck Loads

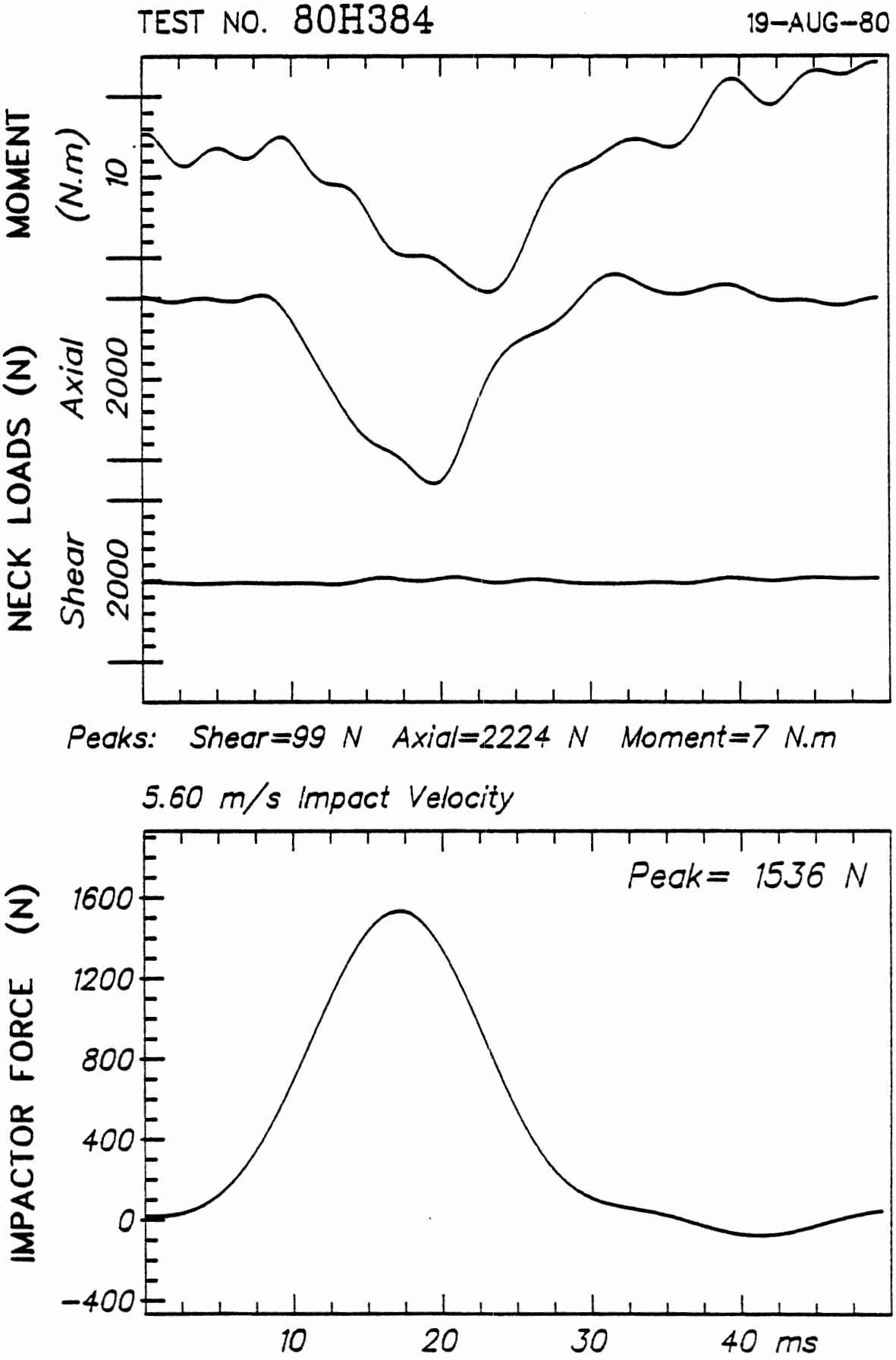
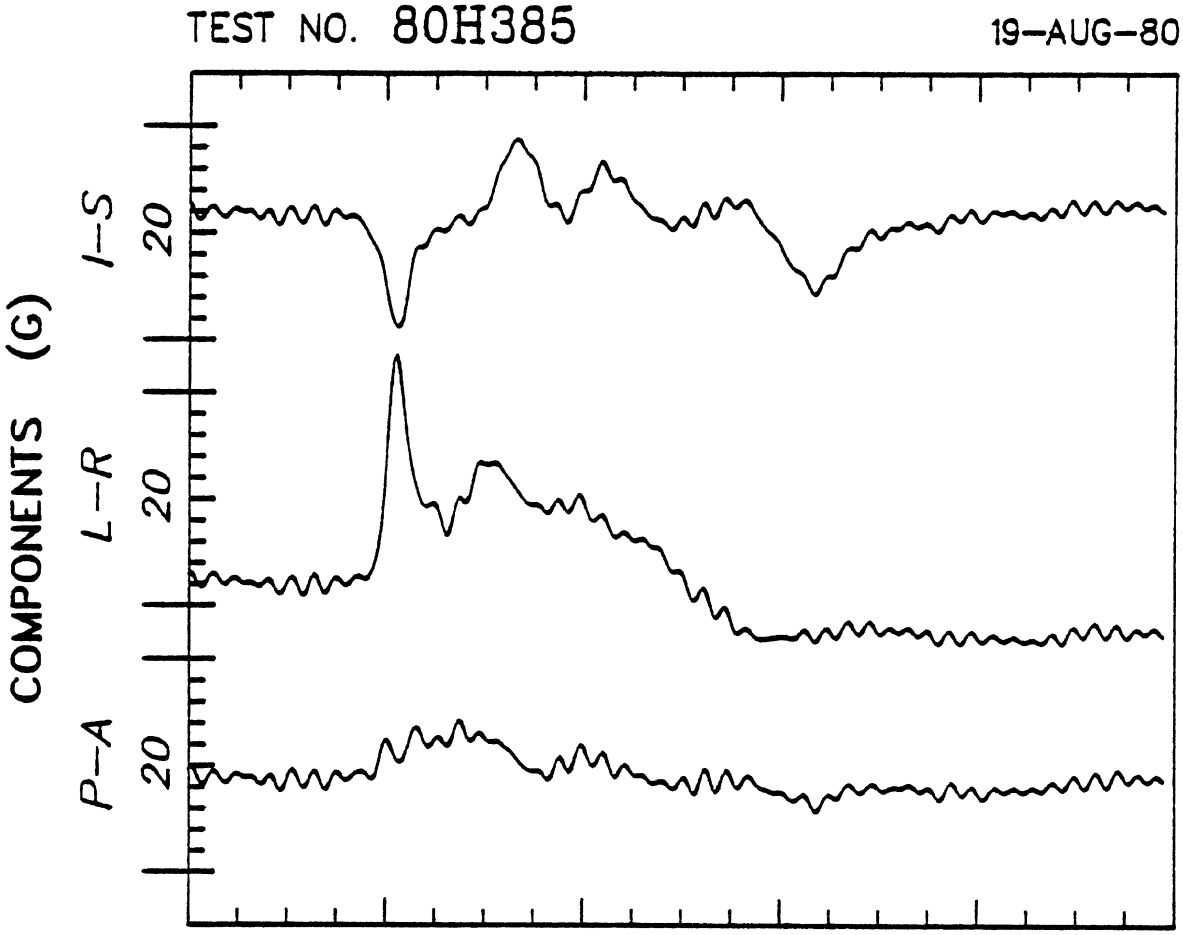


Figure 171: Impactor Deceleration and Neck Loads



Peaks (g): P-A=5 R-L=21 I-S=11

HIC occurred between 9.5 and 49.4 ms

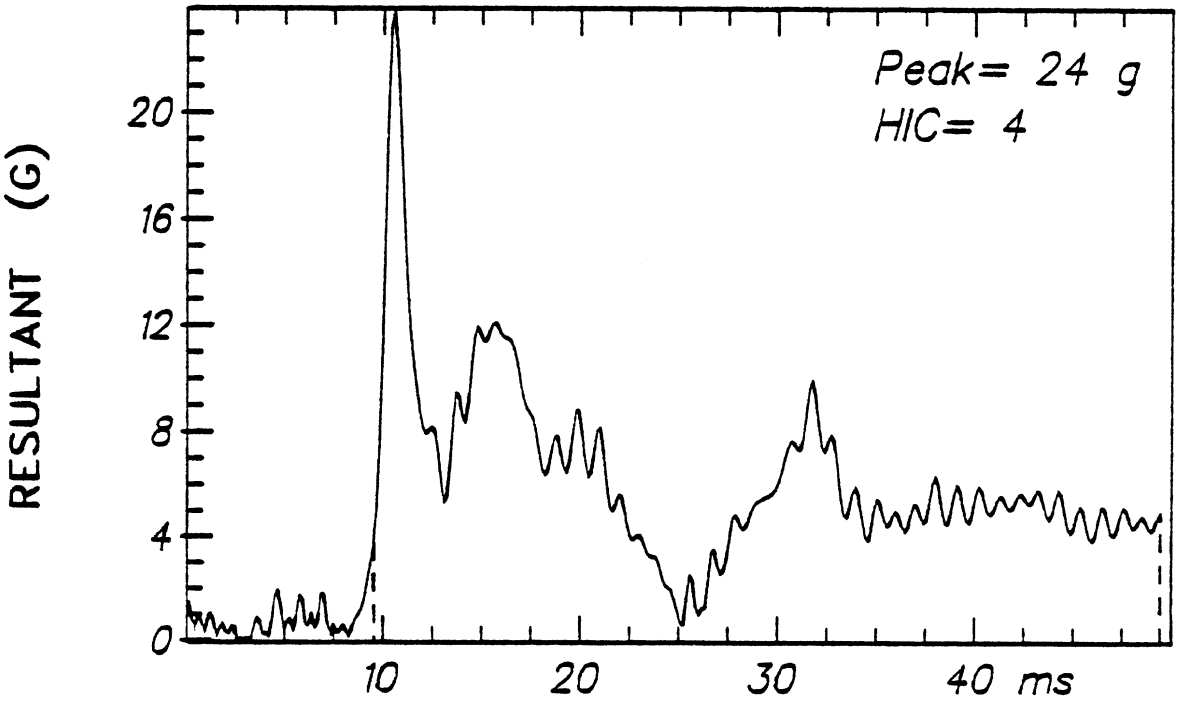


Figure 172: Head Acceleration vs. Time

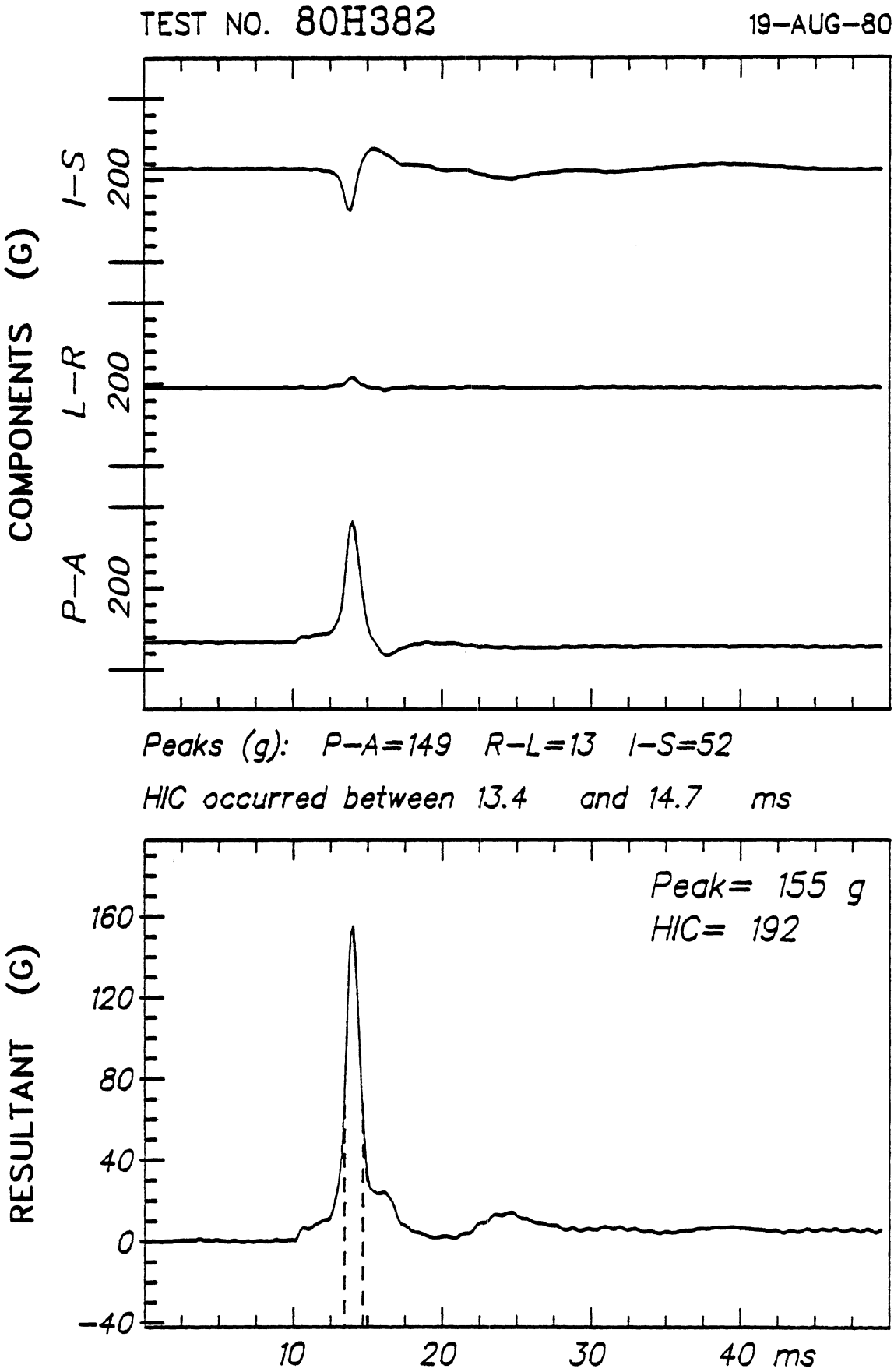


Figure 173: Head Acceleration vs. Time

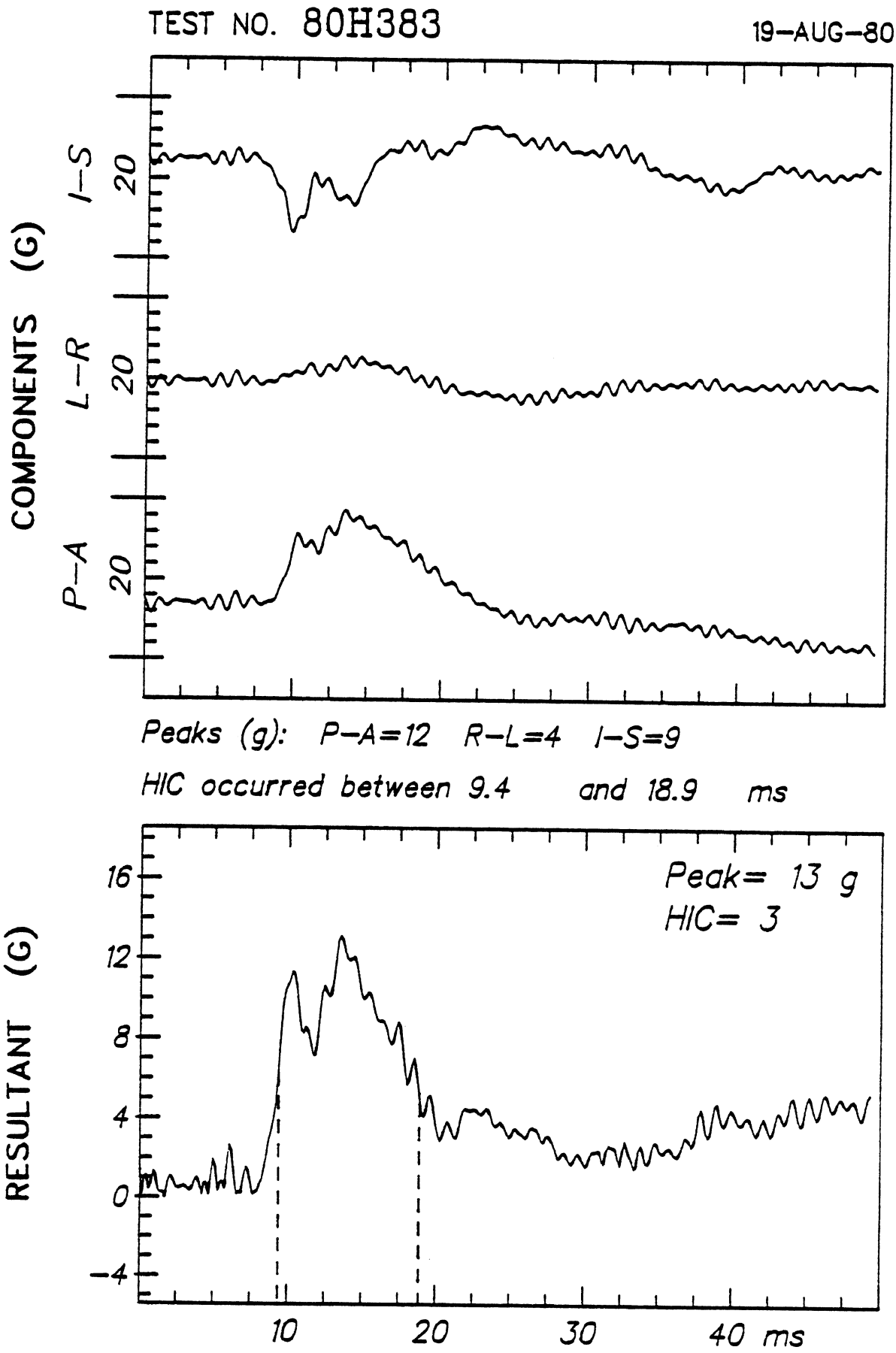


Figure 174: Head Acceleration vs. Time

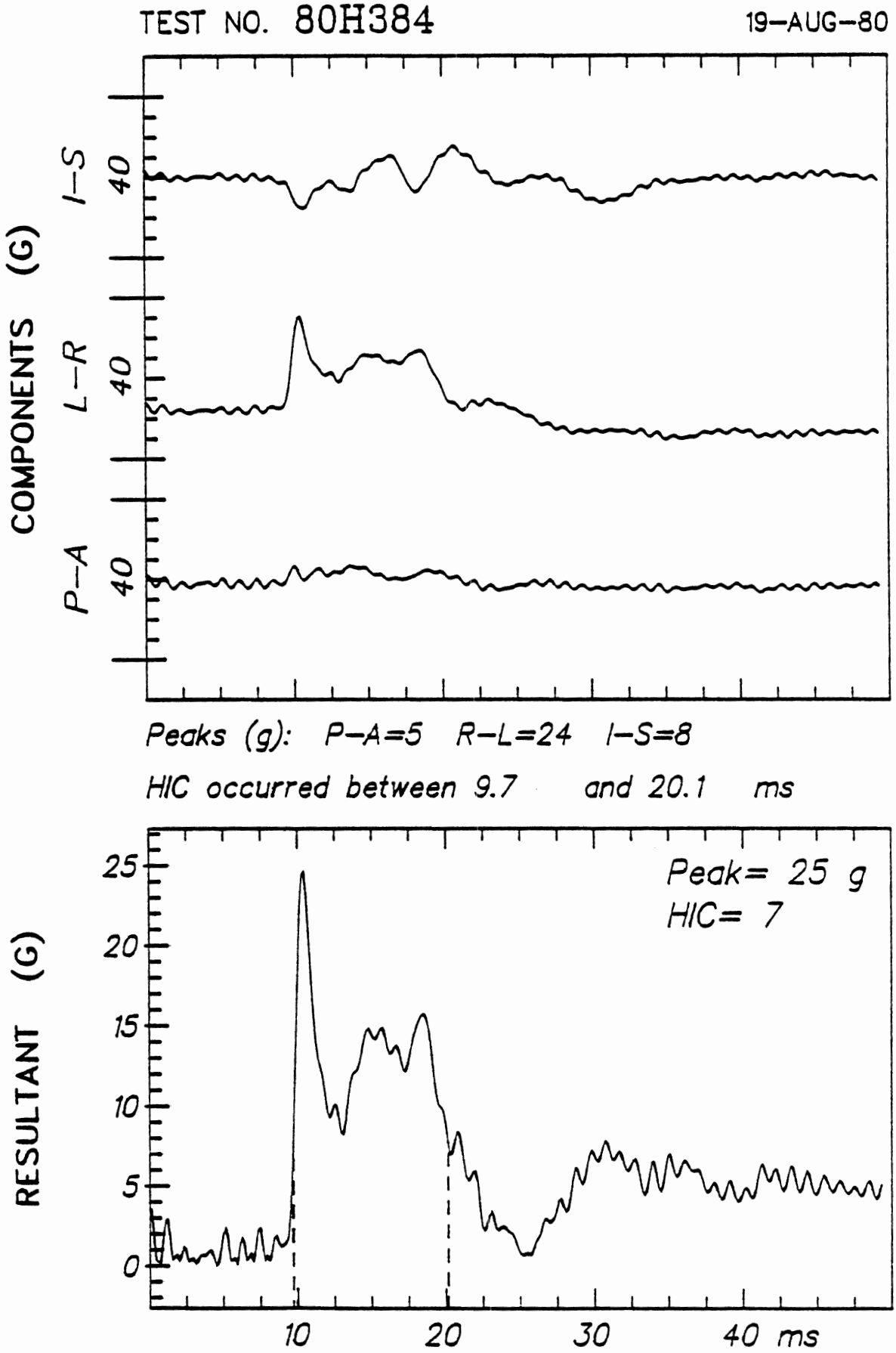


Figure 175: Head Acceleration vs. Time

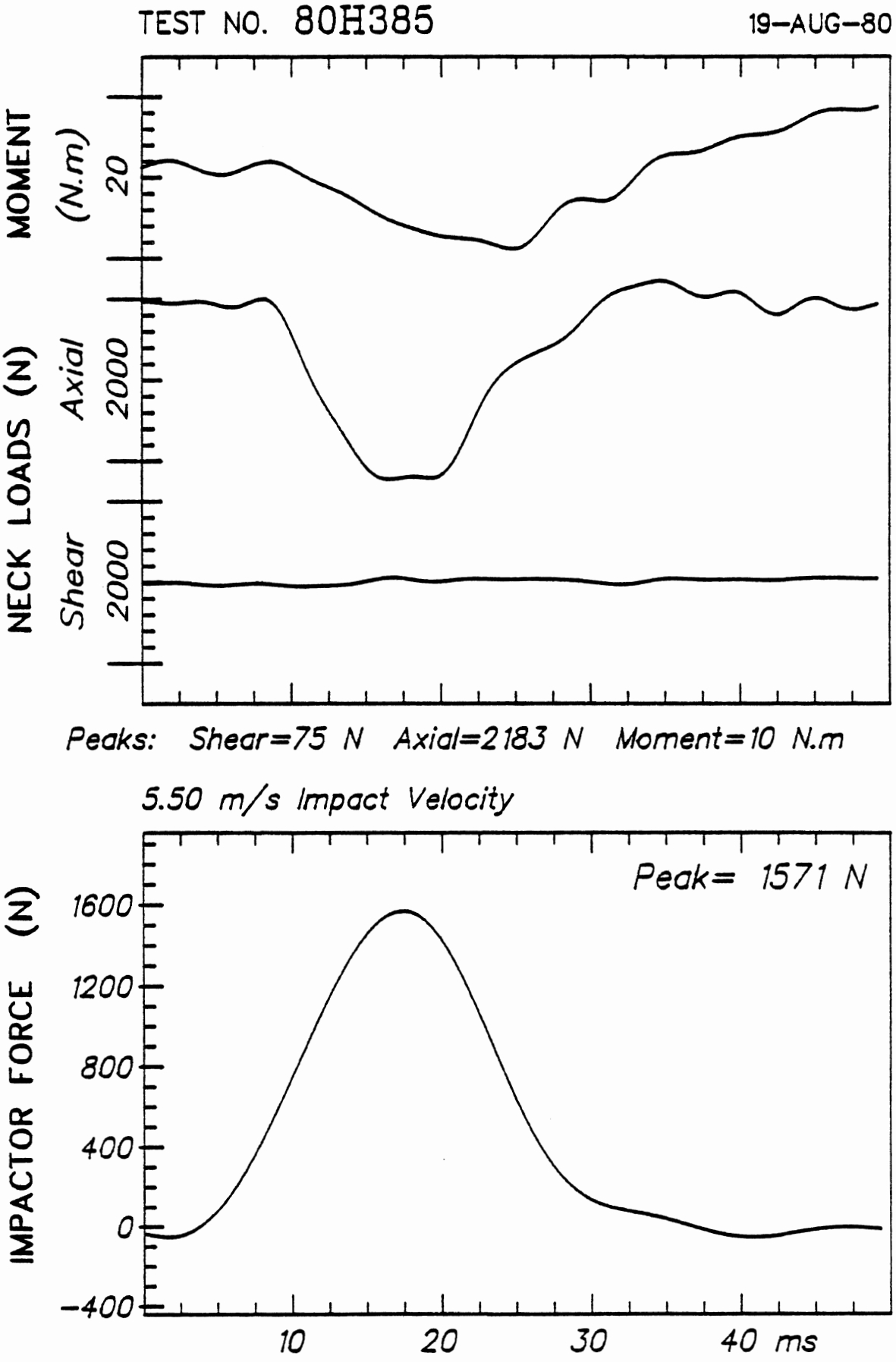


Figure 176: Impactor Deceleration and Neck Loads

5. SUMMARY AND RECOMMENDATIONS

5.1 SUMMARY

The Helmet Impact Test System Development Program has been a complex research program during which the following objectives were met:

(I) The research literature concerning human response and tolerance to S-I impact was surveyed. It was concluded that very little is known about this topic, and that any helmet performance criteria must be based on data yet to be generated.

(II) Five fully instrumented cadaver tests were conducted, primarily to generate response data at impact levels below the estimated tolerance levels. Guidelines for the design of a realistic helmet impact test device were drawn.

(III) Three devices were conceived and proposed, but only one was recommended for actual construction. The advantages and disadvantages of each were spelled out, and the recommended design was defended.

(IV) The actual Helmet Impact Test System (HITS) was designed and constructed. The resulting HITS differed slightly from the proposed one, but the design change was necessary to meet design criteria and requirement spelled out in (III) above.

(V) The HITS operation and assembly instructions manual was written, and delivered as a companion to the device. Complete set of shop drawings were also delivered for possible duplication of the hardware and/or future improvements and modifications.

5.2 RECOMMENDATIONS

Throughout this Final Report, specific observations and conclusions were made. The test results of cadaver S-I impacts are kinematic and dynamic responses and do not include tolerance response. Furthermore, the HITS is a mechanical system with its own limitations. Therefore, care should be exercised when the device is being used, and more importantly, when the resulting HITS responses are being interpreted.