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# Omni-Directional Human Head-Neck Response

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ABSTRACT

The Naval Biodynamics Laboratory (NBDL) in New Orleans has conducted an extensive research program over the past years to determine the head-neck response of volunteer subjects to impact acceleration. These subjects were exposed to impacts in frontal, lateral and oblique directions. An analysis of a limited number of frontal and lateral tests from a test series conducted in the late seventies with two subjects showed that the observed head-neck dynamics can be described by means of a relatively simple 2-pivot analog system (1,2)*.

The present study extends this analysis to a more recent NBDL test program with 16 human subjects. The database consists of 119 frontal, 72 lateral and 62 oblique tests. The research methodology used for this analysis includes a detailed description of three-dimensional kinematics as well as load calculations near T1 and the occipital condyles. A description of this research methodology and a summary of the major test results will be presented. Special attention is given to the influence of impact severity and impact direction on the head-neck dynamics. It will be shown that a similar analog system as proposed earlier for frontal and lateral impacts, is suitable for all impact directions. Geometrical properties of this analog have been determined by means of newly developed numerical techniques rather than through the graphical techniques that were used earlier. Findings of this analysis will be discussed in view of future omni-directional mechanical neck developments.

THE PURPOSE of this paper is to analyse the head-neck response in frontal, oblique and lateral impacts as measured in a large number of human volunteer tests conducted in 1981 and 1982 at the Naval Biodynamics Laboratory (NBDL) in New Orleans. This study is an extension of a previous analysis of a limited set of earlier NBDL tests (1,2,3).

The major aim of this research program is the development of a mechanical neck for crash dummies with omni-directional biofidelity. Realistic simulation of the neck response in a dummy is of particular importance to get a humanlike dynamical behaviour of the dummy head. Trajectories of the head and the nature of the head contact with vehicle interior or exterior are critically dependent on the dummy's neck design.

Neck performance requirements in the literature appear to be mainly related to the neck response in forward flexion and extension. A review of this literature showed that the existing requirements are not sufficient conditions to ensure a humanlike response (3). Performance requirements in this paper will be defined quite differently namely by means of a 2-pivot analog system proposed in the initial phase of this research program (1,2,3). Detailed geometrical and dynamical properties of this analog system will be presented.

## DATABASE DESCRIPTION

In the tests the subjects are seated in an upright position on a HYGE Accelerator (0.3048 m) and exposed to short duration accelerations simulating frontal, oblique or lateral impacts. The resulting three-dimensional motions of the head and first thoracic vertebral body (T1) are monitored by anatomically mounted clusters of accelerometers and photographic targets. A detailed description of the instrumentation and test methods is provided in (4,5).

In the frontal impact tests, the subjects are restrained by shoulder straps, a lap belt and an inverted V-pelvic strap tied to the lap belt. Upper arm and wrist restraints were used to prevent flailing (5). In addition a loose safety belt around the chest is employed. The same restraint system is used in lateral and oblique tests along with a 25 cm wide chest

* Numbers in parentheses designate references at end of paper.

strap to minimize the load on the right shoulder. In addition a lightly padded wooden board is placed against the right shoulder of the subject to limit the upper torso motion.

SUBSET SPECIFICATION - Testresults for 253 tests have been obtained from NBDL. Out of these test a subset of 109 tests was selected for further analysis. Selection of this subset was based on the following criteria:

- Tests specified by NBDL with errors in sensor or film data or with possible contact of instrumentation with restraint system or subjects chest have been omitted.
- No large forward or lateral bending of the head in the initial position is allowed.
- Tests with large data gaps or data shifts in the photographically derived variables are omitted.
- Only tests with a more severe impact level will be considered: frontal tests > 8 g, lateral tests > 5 g, oblique tests > 7 g.

Appendix A summarizes the tests incorporated in this subset and the most important test characteristics. The subset contains 46 frontal, 31 lateral and 32 oblique tests with 15 subjects. Three of these subjects were exposed to all three impact directions namely subject H00133, H00135 and H00136. For all tests the impact velocity is larger than or equal to 6 m/s.

SLED ACCELERATION - Mean values of the sled acceleration-time histories for the most severe frontal, lateral and oblique tests in the present database are shown in Fig. 1. The oblique sled pulses appear to be more severe than the lateral ones, while the frontal are more severe than the oblique ones. These test conditions are close to the test conditions in the earlier tests with subjects H00083 and H00093 (1,2,3). Only for the oblique tests a slight difference can be observed: a maximum peak sled acceleration of 11.4 g in the present database compared to 9.7 g in the previous tests.

COORDINATE SYSTEMS - Figure 2 illustrates the location of the head and T1 anatomical coordinate systems as defined by NBDL. Both coordinate systems are orthogonal and right-handed. Three-dimensional X-ray techniques were used to specify in each test these coordinate systems relative to head and T1 anatomical landmarks. The initial nominal orientation (i.e. before moving of the sled) of these systems relative to the laboratory and the sled coordinate systems for the 3 impact directions, is illustrated in Fig. 3. More details on the definition of the NBDL coordinate systems are provided in (1,5).

The distance between T1 and head anatomical origin just before the impact (time = 0) is called initial neck length. In the earlier test series (1,2,3) differences in this parameter up to 0.07 m per subject were observed which was explained partly by errors in specification of the T1 coordinate system. These earlier tests, moreover, showed for different tests with the same subject considerable variations in the initial orientation of the T1 coordinate system. In the present database the variations in neck length are found to be smaller namely less than 0.03 m (6), but variations in initial T1 coordinate system orientation still appear to be large. For instance the angle between T1 and laboratory z-axis projected on the plane of impact (i.e. a plane parallel to the laboratory (x,z)-plane) showed deviations for the same subject and same impact direction up to 24 degrees (6). Such deviations can not be fully explained from variations in the initial torso orientation and are consequently attributed in part to errors in the specification of the T1 coordinate system itself.

On the bases of these findings a new corrected T1 coordinate system will be introduced here similar to the one defined for the previous test series. This coordinate system is obtained in the following way:

- For each subject an average initial neck length is determined on the basis of calculated initial neck length values for each test (see Table 1).
- The origin of the T1 coordinate system is shifted vertically with respect to the laboratory in such a way that the initial neck length becomes identical to the subject's average initial neck length.
- Finally the T1 coordinate system is rotated so that it becomes aligned with the

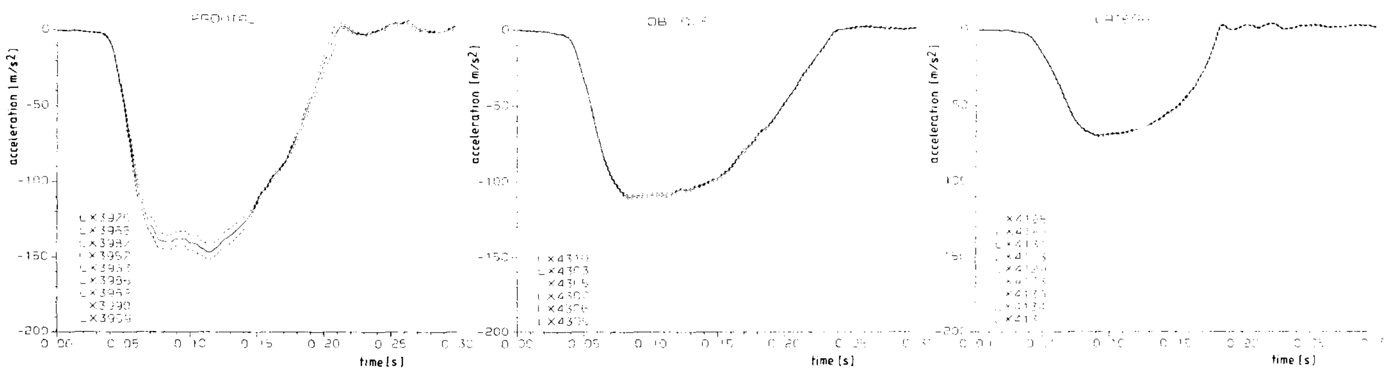


Fig. 1. Mean sled acceleration-time histories for most severe frontal, oblique and lateral impacts (——— mean value, ---- corresponding standard deviation).

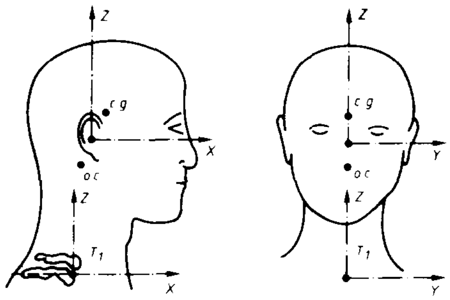


Fig. 2. Location of anatomical coordinate systems according to NBDL (o.c. = occipital condyles, c.g. = center of gravity).

sled (laboratory) coordinate system. As a consequence the orientation of the corrected T1 coordinate system relative to the T1 vertebral body will be dependent on the impact direction. In lateral (oblique) direction it is rotated nominally 90 degrees (45 degrees) with respect to its orientation in frontal direction. Fig. 3 includes the orientation of the corrected T1 coordinate system.

HUMAN SUBJECT ANTHROPOMETRY - At NBDL detailed anthropometric measurements are conducted for each subject as part of the test protocol. Table 1 summarizes the most significant data in this respect. Definitions for these anthropometric variables can be found in (7).

In order to calculate neck load, estimates have to be made for the head mass distribution.

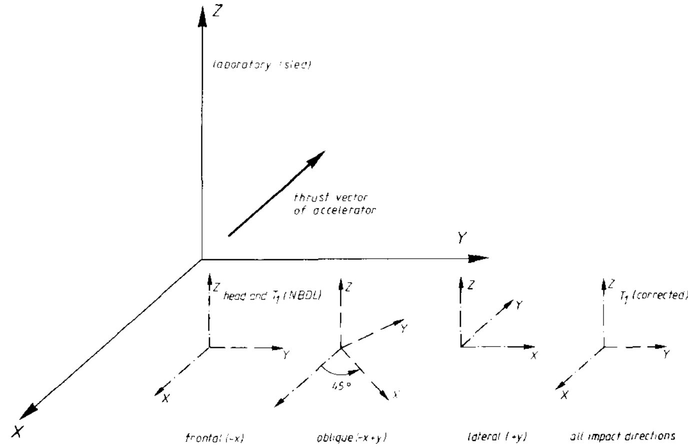


Fig. 3. Nominal initial orientation of head and T1 (NBDL) anatomical coordinate systems relative to laboratory and sled coordinate system in case of frontal, oblique and lateral impacts. Also the orientation of the corrected T1 coordinate systems is included in this figure.

The subjects head mass and the principal moments of inertia will be based on regression equations proposed by McConville et al. (8) using head anthropometry data presented in Table 1. For details on the selected regression equations see (3). The resulting masses and moments of inertia, including a correction for the head instrumentation, are presented in Table 1.

Table 1. Human subject anthropometry.

| | Anthropometric measurements at NBDL | | | | | | | Mass distribution Estimates[2] | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | | | | | | | | | Principal moments of Inertia | | |
| Subject number | Standing height (cm) | Weight (kg) | Sitting height (cm) | Head circumference (cm) | Head breadth (cm) | Head length (cm) | Initial neck length[1] (m) | Mass (kg) | $I_{xx}$ (kgm²) | $I_{yy}$ (kgm²) | $I_{zz}$ (kgm²) |
| H00118 | 185.5 | 73.8 | 97.9 | 57.0 | 14.4 | 20.3 | 0.172 | 4.79 | 0.0266 | 0.0303 | 0.0149 |
| H00120 | 172.6 | 83.0 | 91.1 | 58.6 | 15.6 | 20.2 | 0.172 | 5.14 | 0.0297 | 0.0331 | 0.0169 |
| H00127 | 172.3 | 62.1 | 89.8 | 54.2 | 14.9 | 18.5 | 0.162 | 4.40 | 0.0239 | 0.0252 | 0.0129 |
| H00130 | 180.1 | 72.6 | 94.5 | 56.5 | 14.9 | 19.7 | 0.180 | 4.75 | 0.0266 | 0.0294 | 0.0148 |
| H00131 | 167.0 | 67.6 | 90.0 | 57.5 | 15.4 | 19.6 | 0.156 | 4.98 | 0.0283 | 0.0311 | 0.0160 |
| H00132 | 172.9 | 79.8 | 89.6 | 57.9 | 15.7 | 19.7 | 0.141 | 5.05 | 0.0290 | 0.0319 | 0.0164 |
| H00133 | 161.7 | 61.2 | 86.8 | 56.1 | 14.7 | 19.4 | 0.165 | 4.70 | 0.0259 | 0.0286 | 0.0145 |
| H00134 | 178.3 | 75.3 | 93.0 | 56.6 | 14.4 | 19.4 | 0.158 | 4.81 | 0.0261 | 0.0295 | 0.0151 |
| H00135 | 171.6 | 68.9 | 90.7 | 53.5 | 14.6 | 17.9 | 0.150 | 4.32 | 0.0228 | 0.0240 | 0.0125 |
| H00136 | 185.4 | 88.9 | 92.3 | 56.4 | 15.0 | 19.4 | 0.173 | 4.77 | 0.0266 | 0.0292 | 0.0149 |
| H00138 | 186.0 | 78.9 | 99.2 | 57.1 | 15.4 | 19.8 | 0.174 | 4.87 | 0.0278 | 0.0304 | 0.0154 |
| H00139 | 174.4 | 72.6 | 94.3 | 57.2 | 15.7 | 19.4 | 0.164 | 4.94 | 0.0282 | 0.0306 | 0.0158 |
| H00140 | 177.3 | 86.2 | 94.5 | 56.7 | 15.4 | 19.0 | 0.173 | 4.88 | 0.0273 | 0.0297 | 0.0155 |
| H00141 | 183.3 | 80.7 | 95.6 | 55.4 | 15.2 | 19.2 | 0.175 | 4.57 | 0.0256 | 0.0273 | 0.0138 |
| H00142 | 182.3 | 87.5 | 95.6 | 56.4 | 14.9 | 19.5 | 0.161 | 4.75 | 0.0264 | 0.0292 | 0.0148 |

[1]) Defined as the average value of the initial distance between T1 and head anatomical origin in several tests.

[2]) Including correction for instrumentation.

The orientation of the principal inertia axes, location of center of gravity and location of occipital condyles are assumed to be subject independent. The same values are selected as in the earlier analysis on the basis of data identified in the literature (1,2,3). Table 2 summarizes these data.

Table 2. Subject independent anthropometric data estimated on the basis of data in the literature (relative to head anatomical coordinate system).

| | x (cm) | y (cm) | z (cm) |
|---|---|---|---|
| center of gravity*) | 1.2 | 0 | 2.9 |
| occipital condylar point | -1.1 | 0 | -2.6 |
| principal axis system | rotated -36° about anatomical y-axis (backwards) | | |

*) including correction for instrumentation.

Earlier (see coordinate system section) the anthropometric quantity 'initial neck length' was introduced which was defined as the distance between the T1 and the head anatomical origin. Average values per subject are presented in Table 1 and vary between 0.14 m and 0.18 m. In the previous study (3) one subject (i.e. H00083) showed a significant shorter neck length namely 0.111 m which makes the use of test results from this subject questionable.

T1 DISPLACEMENTS AND ACCELERATIONS - A detailed analysis has been performed of the displacements and accelerations of T1 (6). In agreement with the results of the earlier study (3) the only significant linear displacement of T1 is found in the direction of impact (i.e. along the sled thrust vector). In other words vertical and lateral T1 displacements can be neglected.

Fig. 4 shows for each impact direction separately mean values resulting from the most severe tests for the T1 horizontal acceleration as a function of time. The corresponding standard deviation is incorporated in this figure. It can be seen that the T1 accelerations deviate considerably from the sled accelerations presented in Fig. 1. For all impact directions initially a large spike can be observed due to the interaction between the thorax and the restraint system. As a consequence the peak input acceleration experienced by the head neck system is about twice the peak sled acceleration. Further it follows that the 15 g frontal tests clearly are the most severe tests in terms of input to the head-neck system.

In the previous study a limited rotation of T1 in response to the impact was observed. For the present database such rotations also exist, however, they are considered small enough to be neglected in the remaining part of this paper.

## ANALYSIS OF RELATIVE HEAD MOTIONS: THE 2-PIVOT LINKAGE MECHANISM

This section deals with the motions of the head during the loading phase i.e. up to maximum head excursion. The method used for the analysis in general will be similar to the one used for the previous database. Head motions will be expressed relative to a corrected T1 coordinate system as defined in the preceding section. Since T1 rotations will be neglected here, head motions will be presented with respect to a coordinate system which stays aligned with the laboratory coordinate system.

First an analysis will be made of the occipital condyle trajectories. Appendix B shows a projection of the trajectories on the plane of impact for all tests incorporated in the subset. Results are summarized per subject and per impact direction. It follows that:

- trajectories for different tests with the same subject are quite close to each other;
- maximum head excursions in frontal impacts are slightly larger than in oblique impacts and much larger than in the lateral ones;
- the shape of all trajectories is almost circular.

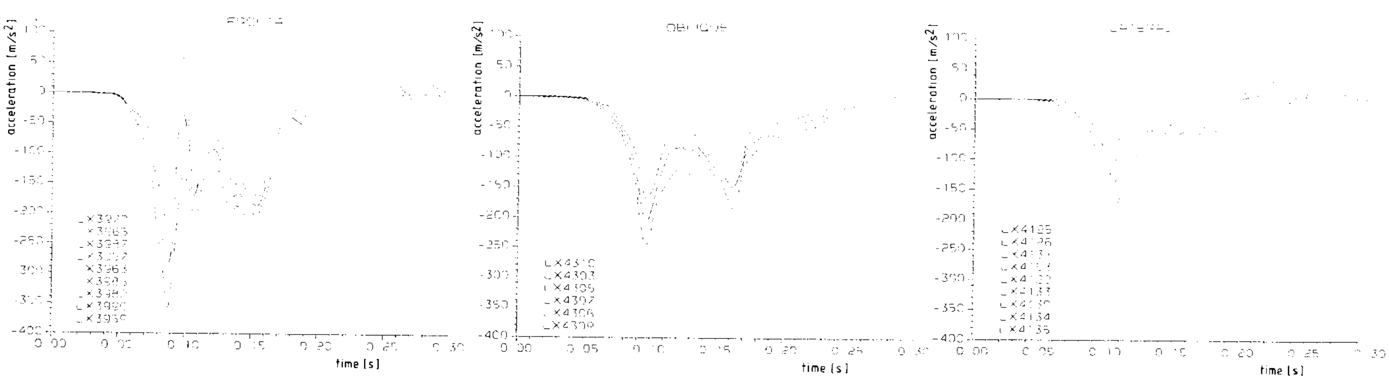


Fig. 4. Mean T1 horizontal acceleration-time histories for the most severe tests in three different impact directions.(—— mean value, --- corresponding standard deviation).

This last finding is the most interesting one. Because displacements of the occipital condyles in a direction perpendicular to the plane of impact are small (less than 0.03 m and without a preferential direction (6)), the 2-pivot linkage mechanism concept proposed in the previous study is also applicable for the present database. Fig. 5 illustrates this linkage mechanism. The upper link represents the head, the middle link the neck and the lower link the torso. The upper pivot is located in the occipital condyles and the lower pivot in the center of the circular arc approximating the occipital condyle trajectories. This lower pivot is a pin joint i.e. a joint with one degree of freedom with the rotation axis perpendicular to the plane of impact. The rotation in this joint is denoted by $\theta$ and is defined as the angle between neck link and z-axis of the corrected T1 coordinate system.

The upper pivot is a joint with two degrees of freedom (universal joint). The first degree of freedom of this joint allows the head link to rotate relative to the neck link in the plane of impact. This rotation angle will be denoted by $\phi$ and is defined here as the angle in the plane of impact between the z-axis of the head anatomical coordinate system and the corrected T1 coordinate system. The second degree of freedom of this upper joint is the rotation $\psi$ of the head about the head anatomical z-axis indicating the head torsion or twist. In frontal impacts this twist motion can be neglected. The rotations of the head link (i.e. head anatomical z-axis) out of the plane of impact will be neglected here because such rotations were found to be very small.

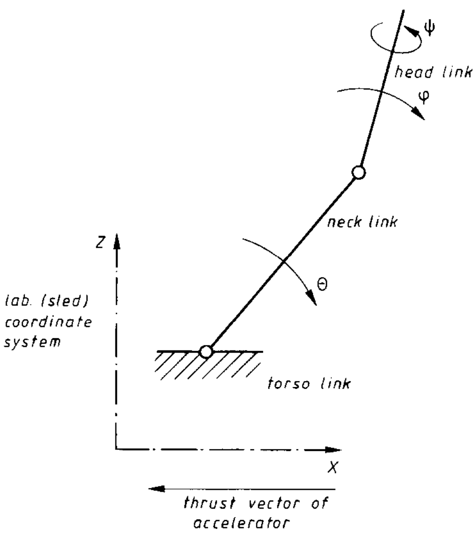


Fig. 5. Analog system for the description of the relative head motions.

NECK LINK LENGTH AND LOWER PIVOT LOCATION - On the basis of the occipital condyle trajectories the neck link length and the lower pivot location will be determined here. Numerical techniques rather than the graphical method applied in the previous study are used in order to improve the accuracy of these estimations. The computer program developed for this purpose allows estimations of an optimal radius and corresponding average center of rotation for a given planar trajectory by means of least-squares estimation techniques (9). The program can also be used to determine the optimal centre of rotation for this given trajectory if the radius has a pre-selected value. Output of the program includes the maximal fitting error in the trajectory (devmax) and the residual standard deviation (sdw).

First the program was applied to estimate the optimal radius for each test separately. Results are presented in Fig. 6 as function of initial neck length. Fig. 7 shows the mean radius as a function of impact severity. Data are given separately for each impact direction. It follows that the radius does not increase for a larger impact severity, whereas the effect of the initial neck length on the optimal radius value is small. The distribution in calculated radii appears to be larger in lateral direction than in oblique or frontal direction which can be explained by the shorter trajectories for lateral impacts. The average optimal radius per impact direction appears to vary slightly: 130.5 mm in frontal, 123.2 mm in lateral and 138.9 mm in oblique direction.

Table 3. Mean values for fitting accuracy in case of free and predescribed radii per impact direction, resulting from numerical estimation techniques.

| Impact direction | Free radius: maximal fitting error (devmax) (mm) | Free radius: residual standard deviation (sdw) (mm) | Predescribed radius: 0.129m: maximal fitting error (devmax) (mm) | Predescribed radius: 0.129m: residual standard deviation (sdw) (mm) |
|---|---|---|---|---|
| Frontal | 6.92 | 1.730 | 7.05 | 2.185 |
| Lateral | 5.13 | 1.271 | 5.41 | 1.565 |
| Oblique | 5.16 | 0.93 | 6.00 | 1.700 |

In the previous study (3) an average radius of 0.125 m was graphically selected for all tests in all impact directions. The average radius resulting from the present database appears to be surprisingly close to this earlier value, namely 0.129 m.

The effect of this fixed radius on the fitting accuracy has been calculated with the least-squares estimation program and is illustrated in Table 3. This table shows average values for devmax and sdw per impact direction resulting from calculations with and without a predescribed radius. It follows that the influence on the goodness of fit in case of an identical radius for all tests is small: the increase in the average standard deviation is less than 0.8 mm. As analyzed in (9) the radius length and the pivot location are highly correlated; this implies that different values for the radius length are largely

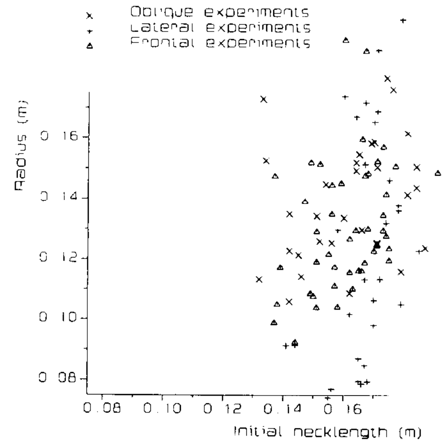


Fig. 6. Optimal radius for occipital condyle trajectories as function of initial neck length and impact direction.

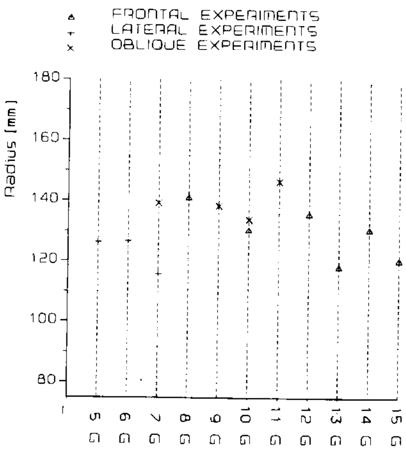


Fig. 7. Mean optimal radius for occipital condyle trajectories as function of impact severity and impact direction.

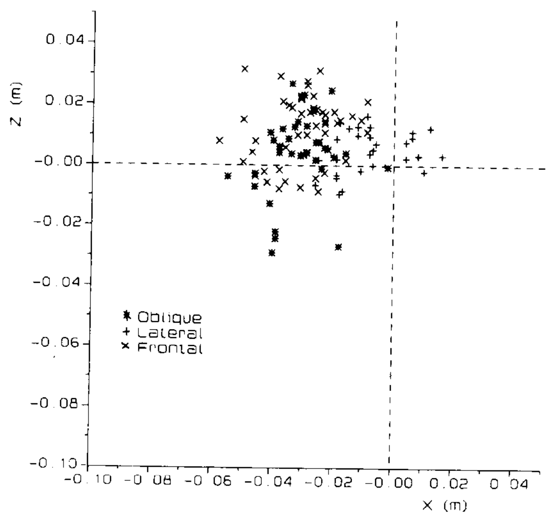


Fig. 8. Lower pivot locations relative to the corrected T1 coordinate system for a fixed neck link length.

compensated for by other optimal pivot values. On the basis of these findings for all tests one fixed value of 0.129 mm will further be used for the neck link length.

The optimal corresponding lower pivot locations for this link length are presented for each test in Appendix B and summarized in Fig. 8. Locations are given per impact direction relative to the (x,z)-plane of the corrected T1 coordinate system. Average values for the x and z coordinates and corresponding standard deviation, per impact direction are summarized in Table 4.

Table 4. Average values and corresponding standard deviations for lower pivot locations per impact direction (relative to the corrected T1 coordinate system).

| Impact direction | x (s.d.) (m) | z (s.d.) (m) |
|---|---|---|
| frontal | - 0.031 (0.011) | 0.011 (0.011) |
| lateral | - 0.008 (0.012) | 0.005 (0.007) |
| oblique | - 0.032 (0.010) | 0.003 (0.014) |

The average z-coordinate appears to be close to zero. Variations in this coordinate are mainly due to differences per subject in initial neck length. The x-coordinate has a negative value in most of the tests i.e. opposite to the direction of motion of the head. For lateral impacts the mean value is close to 0.01 m while for frontal and oblique impacts a value close to 0.03 m is found.

## ANALYSIS OF RELATIVE HEAD MOTIONS: HEAD AND NECK LINK ROTATIONS

Relative head motions will be analyzed here using the linkage concept presented before. Fig. 5 illustrates this mechanism. Three rotations (i.e. degrees of freedom) can be distinguised: head flexion $\phi$, head twist $\Psi$ and neck link rotation $\theta$. The head flexion $\phi$ and head twist $\Psi$ can be calculated directly from the NBDL test results (i.e. the head rotation matrix) and are independent of the selected geometrical linkage parameters.

The definition used for the head twist motion in the present study is slightly different from the one used in the previous study where the orientation of the head was considered to be the result of three successive Euler rotations about the head anatomical x-axis, the (transformed) head anatomical y-axis and the (transformed) head anatomical z-axis, respectively. This last rotation was defined as the twist angle. In the present study a more adequate definition is used: the relative orientation of the head is considered to be the result of two rotations. The first one transforms the head z-axis directly in its final position and the second rotation i.e. the twist motion is a rotation about the (rotated) z-axis (6).

The neck link rotation θ is calculated from the occipital condyle trajectories and the lower pivot location. Data presented here will be based on the calculated radius of 0.129 m and the corresponding test specific optimal lower pivot location.

The magnitude of the angles φ, Ψ and θ in the initial position (time = 0) will be denoted by $\phi_o$, $\Psi_o$ and $\theta_o$, respectively. Maximum values for these rotations are denoted by $\phi_{max}$, $\Psi_{max}$ and $\theta_{max}$. Appendix A summarizes the initial values as well as ($\phi_{max} - \phi_o$), ($\theta_{max} - \theta_o$) and ($\Psi_{max} - \Psi_o$) for 109 tests.

INITIAL VALUES - Table 5 summarizes the mean values per impact direction for $\phi_o$, $\Psi_o$* and $\theta_o$. It follows that initial head twist ($\Psi_o$) and head flexion ($\phi_o$) are close to zero, indicating that the 'average' initial head orientation is close to the orientation of the laboratory coordinate system. The initial neck link rotation $\theta_o$ shows, except for the lateral direction, a significant positive value namely 10.8 degrees in oblique and 17.6 degrees in frontal impacts.

Table 5 Mean values per impact direction for the initial values of the angles φ, Ψ and θ.

| Impact direction | $\phi_o$ (deg.) | $\Psi_o$* (deg.) | $\theta_o$ (deg.) |
|---|---|---|---|
| frontal | 1.8 | -1.0 | 17.6 |
| lateral | 0.8 | 1.2 | 0.7 |
| oblique | -2.9 | -1.3 | 10.8 |

PEAK ROTATIONS AS FUNCTION OF IMPACT SEVERITY - Mean values for ($\phi_{max} - \phi_o$), ($\theta_{max} - \theta_o$) and ($\Psi_{max} - \Psi_o$) as function of impact direction and impact severity are presented in Fig. 9. Both the head flexion and the neck link rotation appear to be strongly dependent on the peak sled acceleration. The influence of the impact direction is most clearly present in the neck link rotation: frontal impacts show larger neck link rotations (for the same impact severity) than oblique ones and oblique impacts show larger neck link rotations than the lateral ones. Finally it can be seen that lateral impacts show a much larger twist than oblique ones.

NECK LINK ROTATIONS AS FUNCTION OF HEAD FLEXION - Fig. 10 presents for the three impact directions the neck link rotation ($\theta - \theta_o$) as function of head flexion ($\phi - \phi_o$). Results for the most severe tests are given. In the initial phase of the motion for all impact directions the head flexion is smaller than the neck link rotation illustrating the translational nature of the initial head motion. This response was also observed in the previous study (3). As

* oblique impacts: $\Psi_o$ - 45°; lateral impacts: $\Psi_o$ - 90°

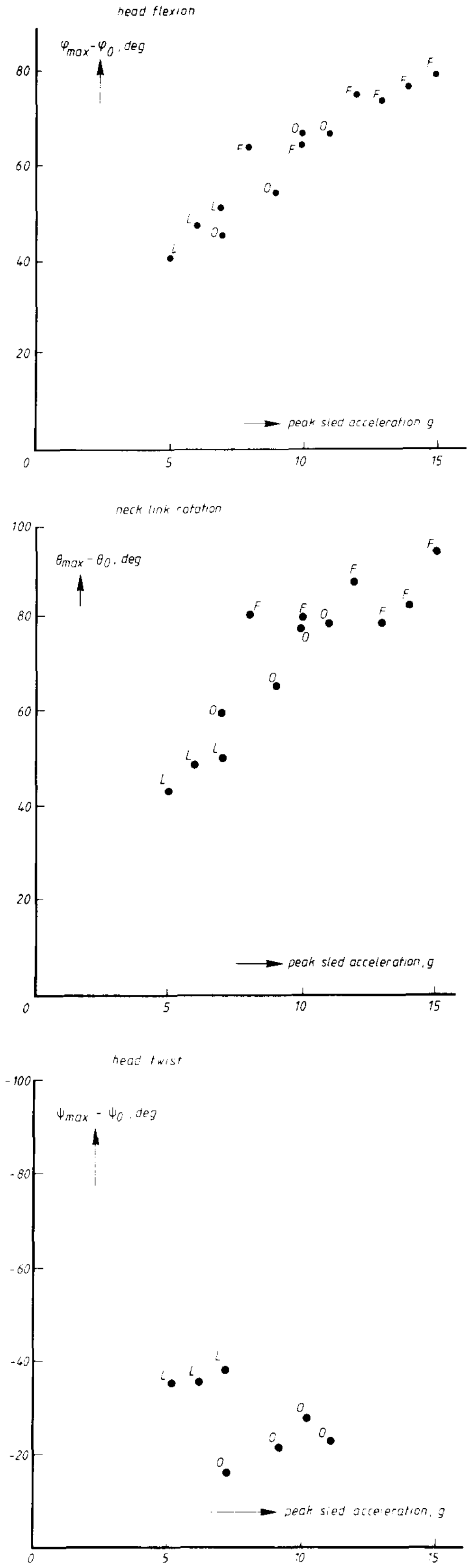


Fig. 9. Mean values for peak rotations as function of impact direction and impact severity.
(F = Frontal, O = Oblique, L = Lateral)

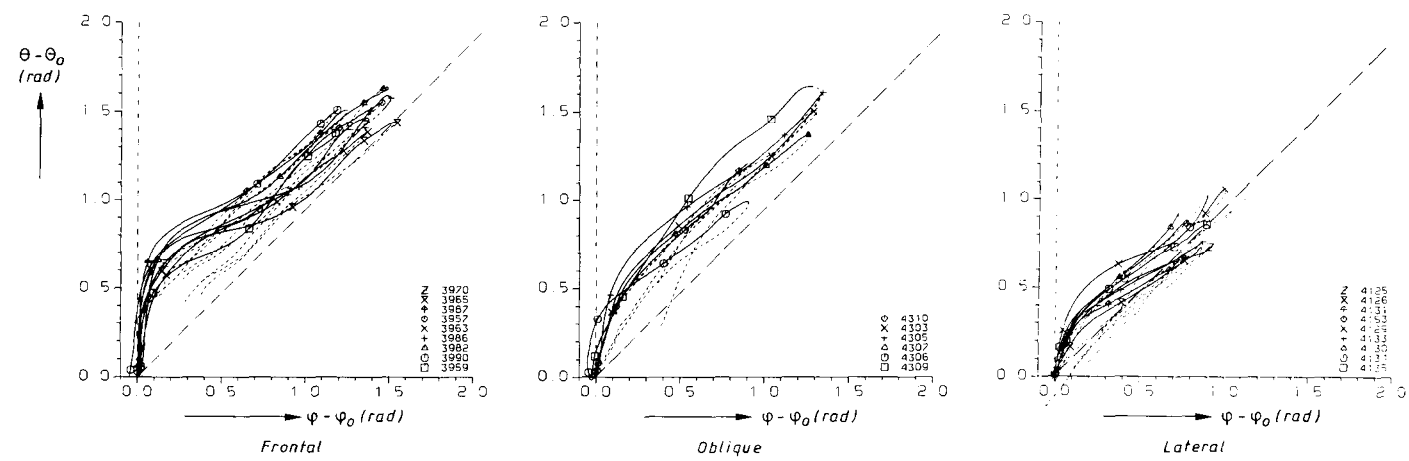


Fig. 10. Neck link rotation $(\theta - \theta_o)$ as function of head flexion $(\phi - \phi_o)$ for the most severe tests as function of impact direction. (———— loading -------- unloading)

soon as the relative angle $(\theta - \theta_o) - (\phi - \phi_o)$ reaches a certain level the head and neck link become more or less locked. For most of the tests this relative angle appears to decrease slightly in the final part of the loading phase.

The maximum relative angle i.e. $((\theta - \theta_o)-(\phi - \phi_o))_{max}$ between head and neck link has been calculated for all tests (see Appendix A). The influence of impact direction and impact severity on this angle is presented in Fig. 11. The impact direction appears to have a significant influence while the effect of impact severity is small. Average values for the upper pivot maximum rotation are 30 degrees in frontal, 20 degrees in oblique and 10 degrees in lateral impacts.

HEAD TWIST AS FUNCTION OF HEAD FLEXION - Fig. 12 shows for the lateral and oblique impact directions the head twist $(\Psi - \Psi_o)$ as function of head flexion $(\phi - \phi_o)$. Results for the most severe tests are presented. It follows that for all tests the head twist is smaller than the head flexion.

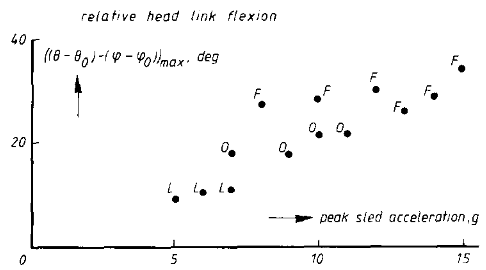


Fig. 11. Influence of impact severity on maximum upper pivot angle (relative head link flexion). (F = Frontal, O = Oblique, L = Lateral).

ANALYSIS OF NECK LOADS

The loads applied by the neck to the head can be calculated using measurements of head acceleration and angular velocity, if the head is regarded as a rigid body and does not come into contact with any other object or body part. Neck load equations used in this study

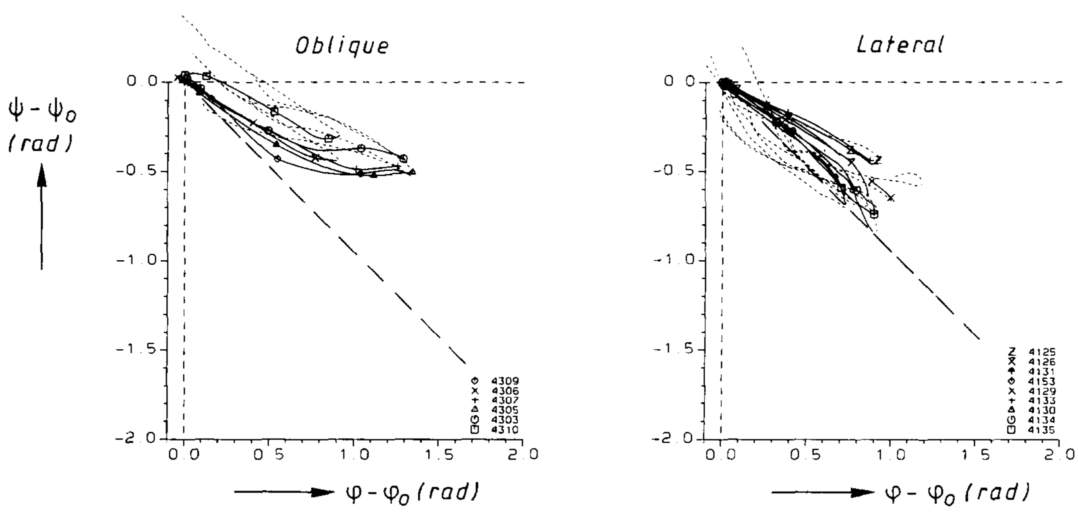


Fig. 12. Head twist $(\Psi - \Psi_o)$ as function of head flexion $(\phi - \phi_o)$ for the most severe tests in lateral and oblique directions. (—— loading --- unloading)

are presented in (1,3,10). Some of the reasons to perform these calculations are:

- they offer an excellent insight in the system's behaviour, for instance with respect to the role of muscle activity;
- load-displacement relations can be used to formulate dummy performance requirements and dummy design specifications;
- it is generally assumed that neck loads correlate quite well with neck injuries.

Another reason to perform such calculations is to estimate the dynamic characteristics of the upper and lower pivot in the analog system proposed in the preceding section. These pivots are located in the occipital condyles and near the T1 origin (i.e. in the center of a circular arc approximating the occipital condyle trajectories). For the calculation of the load in the lower pivot it is assumed that the effect of neck inertia can be neglected.

For all tests incorporated in this study neck loads have been calculated. Detailed results are presented in (6). In this paper some of the most interesting results with respect to the dynamic pivot characteristics will be summarized. Data will be presented for:

- the torque $M_\phi$ at the upper pivot about an axis perpendicular to the impact plane as function at the relative angle between head and neck link $((\theta - \theta_o) - (\phi - \phi_o))$;
- the torque $M_\psi$ at the upper pivot about an axis parallel to the head anatomical z-axis as function of the head twist $(\Psi - \Psi_o)$;
- the torque $M_\theta$ at the lower pivot about an axis perpendicular to the impact plane as function of the neck link angle $(\theta - \theta_o)$.

Torques are defined here as torques applied by the neck to the head ($M_\phi$ and $M_\psi$) or by the torso to the head-neck system ($M_\theta$). Fig. 13 presents the sign conventions for the torques and the degrees of freedom of the analog system.

The calculated torque-rotation characteristics for the most severe tests per impact direction are summarized in Fig. 14. The following observations can be made:

- The occipital condyle torques $M_\phi$ as function of the relative angle between head and neck link confirm the findings in the preceding section for the relative motion in the upper pivot of the linkage: dependent on the impact direction a certain free range of motion exists where the occipital condyle torque is relatively small. As soon as this angle is exceeded the upper pivot gets more or less locked and the occipital condyle torque strongly increases.
- Largest occipital condyle torques $M_\phi$ (i.e. 60-90 Nm) can be observed for the frontal impact tests. These values appear to be slightly larger than calculated from the previous database (2).
- Peak values for the component $M_\psi$ of the occipital condyle torque appear to be much smaller (i.e. 4-6 times) than the component $M_\phi$.
- For the torques at the lower pivot (i.e. $M_\theta$) peaks close to 200 Nm can be observed. As for the occipital condyle torques, largest torque values are found in the most severe frontal impact tests.
- The lower pivot torque characteristics show for frontal impacts an initial spike. The reason for this behaviour is not yet fully understood.

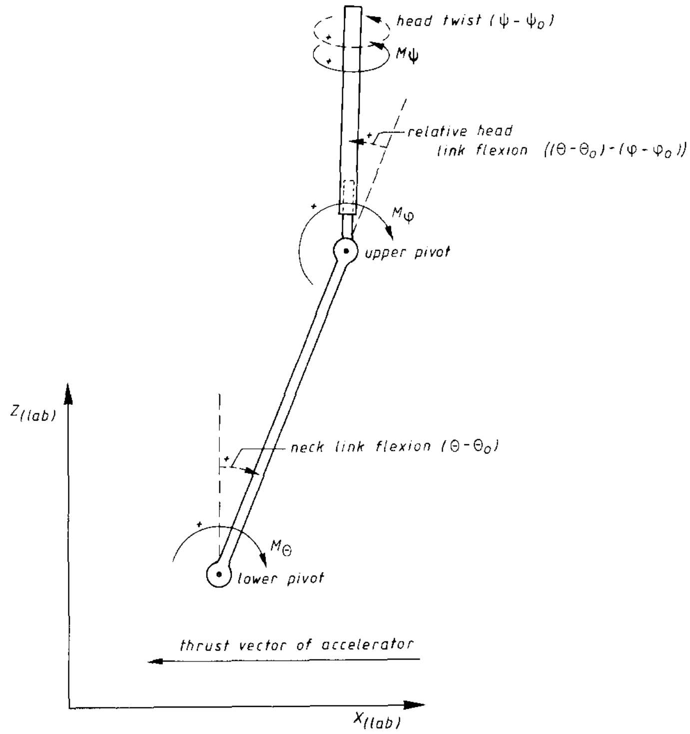


Fig. 13. Analog systems with positive torques and rotations.

The test results presented in Fig. 14 relate to the most severe tests in the database. In general, less severe tests showed an identical shape for the torque-rotation characteristics (6). The effect of test severity on joint stiffness was found to be small. Since, moreover, variations in torque-rotation characteristics between subjects are relatively small, selection of one average joint characteristic seems to be appropriate here. In the following sections estimates for these characteristics will be given:

UPPER PIVOT: RELATIVE HEAD LINK FLEXION - A free range of motion can be observed in this joint which varies per impact direction: about 30 degrees in the frontal, 20 degrees in the oblique and 10 degrees in the lateral direction (see section: "analysis of relative head motions: head and neck link rotations"). The joint stiffness after locking of this joint shows no significant differences per impact direction: this joint behaviour can be quite well approximated by a linear function with a slope

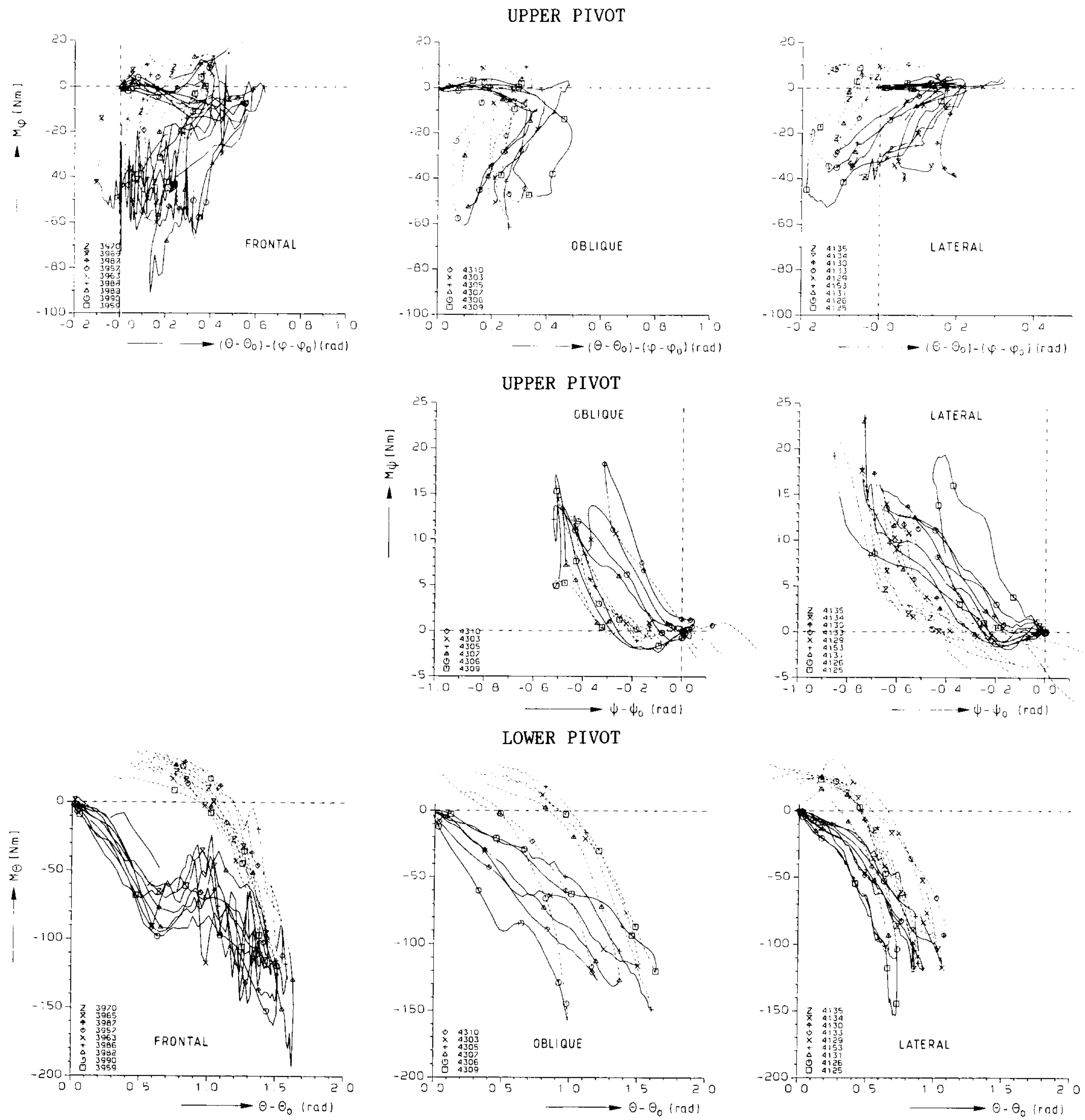


Fig. 14. Upper and lower pivot dynamic characteristics for the most severe tests as function of impact direction (—— = loading --- = unloading).

(i.e. joint stiffness) of 3 Nm/degree.

UPPER PIVOT: HEAD TWIST - A linear function is proposed to characterize the head twist. The stiffness appears to be relatively small and varies per impact direction: about 0.4 Nm/degree in the lateral direction and 0.75 Nm/degree in the oblique direction (note: no torsion motion is present in frontal impacts).

LOWER PIVOT: NECK LINK FLEXION - In order to determine the properties for the lower pivot, first an average torque-rotation characteristic per impact direction has been calculated for the most severe tests. Fig. 15 shows the results of these calculations with corresponding standard deviations. If a linear function is selected to describe this joint behaviour realistic estimates for joint stiffness are as follows:

| | | |
|---|---|---|
| frontal | : | 1.2 Nm/degree |
| oblique | : | 1.5 Nm/degree |
| lateral | : | 2.2 Nm/degree |

The initial peak in the frontal impact characteristics has not been taken into account in this linear approximation.

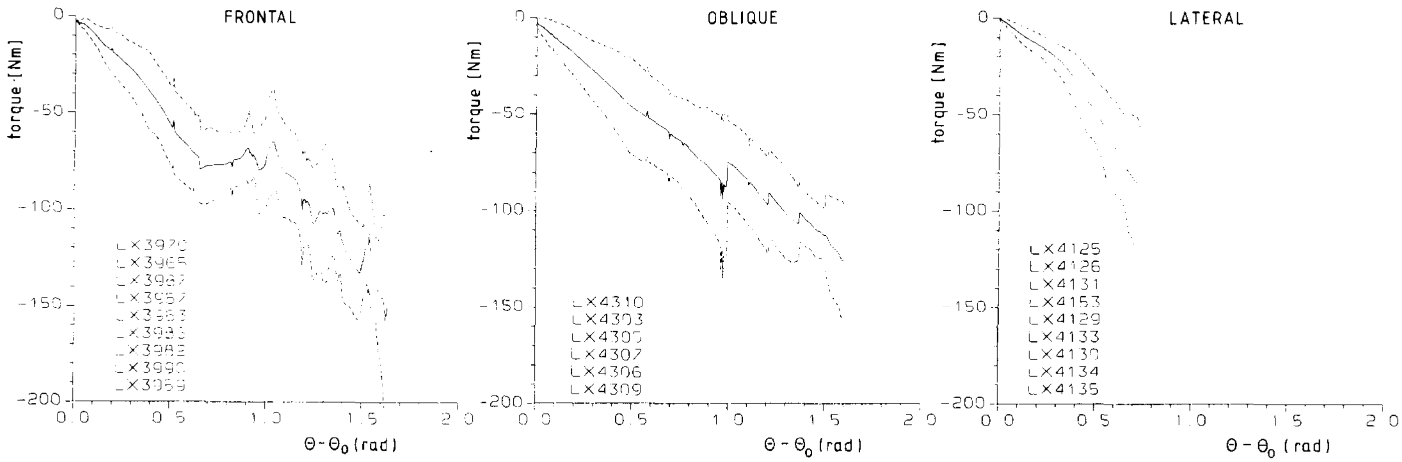


Fig. 15. Average lower pivot torque-rotation characteristics with corresponding standard deviations based on the most severe tests (—— mean value, ---- s.d.).

DISCUSSION

Test results of 253 human volunteer tests have been obtained from the Naval Biodynamics Laboratory in New Orleans. A subset of 109 tests with 15 subjects was selected and partially analyzed in this paper. Special attention was given to the most severe tests in this subset.

This study is an extension of a previous analysis of NBDL tests with two subjects (1,2,3). In this earlier work significant errors were noted in the specification of the T1 coordinate system. In the present database these errors are smaller indicating that the present data are more reliable. On the other hand the present errors are still of such a magnitude that they justified introduction of a corrected T1 coordinate system in a similar way as in the previous study.

The most important finding of the present study is that the observed human head-neck response, like in the previous study, can be represented adequately by a linkage system with 2 pivots. One link represents the head, one link the neck and one link the torso. For each impact direction a separate analog system is proposed. Geometrical parameters of this analog are identical for each impact direction as far as the neck link length and the upper pivot location are concerned. The location of the lower pivot in the torso appears to be slightly different for each impact direction as was illustrated in Table 4 and Fig. 8. Also the initial position of this linkage system varies per impact direction: the initial neck link rotation is almost 18 degrees in frontal impacts, 11 degrees in oblique impacts and close to zero in lateral ones.

For frontal impacts the analog system has 2 and for lateral and oblique impacts 3 degrees of freedom. These degrees of freedom are the neck link rotation in the plane of impact, a head rotation (i.e. head flexion) in the plane of impact and a head twist about the head anatomical z-axis (absent in frontal impacts).

Based on load calculations joint stiffness values have been identified. Different characteristics per impact mode can be observed. For the lower pivot the stiffness in lateral direction is larger than in the oblique direction and the stiffness in the oblique direction is larger than in the frontal direction. For the upper pivot rotation in the plane of impact a free range of motion can be observed, which is about 30 degrees in the frontal, 20 degrees in the oblique and only 10 degrees in the lateral direction.

The 2-pivot analog system has been introduced for the purpose of characterizing the three-dimensional head motion in this type of dynamical events. This analog system should not be considered as a representation of the very complex anatomical structure of the cervical articulations and the associated structures (including musculature). See for instance Huelke (11) and Kapandji (12) for a detailed treatment of the neck anatomy and neck motions.

Torques calculated in this study near the occipital condyles and near T1 represent the resultant torque in the head-neck and neck-torso interface, respectively. These torques are developed by tension forces in the musculature, internal neck compression forces, etc. More detailed neck models are needed to study the load contribution of the various neck structures.

As in the previous studies (1,2,3) it is observed that the initial head rotation is smaller than the neck link rotation followed by a lock up between the head and neck link. This behaviour most likely can be explained by the contribution of the various muscle groups. Also the initial peak in the T1 torque (frontal impacts) might be caused mainly by the musculature (i.e. the large posterior muscle groups of the neck).

In this study separate systems are proposed for the three impact directions. As an alternative, a single three-dimensional linkage could have been proposed to simulate all three impact directions. Such a mechanism would have 5 degrees of freedom: 2 degrees of freedom in the lower pivot and three degrees of freedom in the upper pivot. In such a system the neck link rotation is not limited to a planar motion in the plane of impact. The major problem in defining such a system is the out of mid-sagittal plane location of the lower pivot in oblique impacts. If this lower pivot location can be assumed to be located in the mid-sagittal plane, the formulation of a three-dimensional system is not expected to cause significant problems. Attractive aspects of such a system are, among other things, the possibility to predict the head response in arbitrary directions and the possibility to determine out-of-impact plane motions which were neglected in the present analysis.

The proposed analog system is only valid for low severity impacts, i.e. the NBDL human volunteer test conditions. Additional information should be obtained for higher exposure levels from human cadaver tests. Such tests, moreover, are needed for a better understanding of the contribution of muscle activity. If such data become available, adjustment of the proposed analog system might be necessary.

The proposed analog system implicitly constitutes a performance requirement. As an alternative, performance requirements could have been formulated explicitly by means of kinematic requirements like displacements and accelerations for several impact levels. Such a formulation is desirable for instance when establishing a standard to be used in the laboratory for verifying the dummy response. Such criteria can be derived directly from the NBDL test data or indirectly from the analog system.

Although the developed analog system is not intended as a design principle for a future omni-directional mechanical neck it illustrates some important design principles:

- The basic geometrical set-up of the neck design can be identical for the various impact modes. In other words it is expected that one neck design can be used for all impact directions.
- The stiffness of a dummy neck is dependent on the impact direction. In lateral direction for instance the neck should be more stiff (almost twice as stiff) than in forward direction.
- A feature should be incorporated in the design which suppresses large head flexions but allows relatively large neck link rotations. In order to realize this the effect of the locking mechanism which was identified for the upper pivot, should be approximated in a future dummy neck design.

## CONCLUSIONS

1. 109 human volunteer test with 15 subjects have been analyzed. Tests were conducted in the frontal, the lateral as well as the oblique impact direction.
2. A 2-pivot analog system appears to be adequate to describe the observed head-neck motions.
3. Numerical techniques were developed and applied to determine the geometrical properties of this analog system.
4. Different analog systems per impact mode are proposed of which the most important geometrical parameters are identical.
5. Load calculations were conducted in order to specify the analog system's dynamic pivot properties.
6. Pivot torque-rotation characteristics appear to be stiffer in the lateral than in the frontal direction.
7. Some preliminary guidelines for future dummy neck designs have been presented.

## ACKNOWLEDGEMENT

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Also we would like to acknowledge the University of Nijmegen for preliminary data processing facilities.

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## APPENDIX A

## SUMMARY OF TEST CONDITIONS AND TEST RESULTS

| Subject | No. | Peak sled acc. (G) | Rate of onset (G/s) | Sled vel. change (m/s) | $\phi_o$ (rad) | $\theta_o$ (rad) | $\psi_o$ (rad) | $\phi_{max}-\phi_o$ (rad) | $\theta_{max}-\theta_o$ (rad) | $\psi_{max}-\psi_o$ (rad) | $((\theta-\theta_o)-(\phi-\phi_o))_{max}$ (rad) |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Frontal experiments. | | | | | | | | | | | |
| H00118 | 3886 | 8.2 | 203. | 12.02 | -0.186 | 0.057 | -0.030 | 1.497 | 1.884 | -0.014 | 0.751 |
| | 3903 | 10.2 | 284. | 13.72 | -0.126 | 0.144 | 0.005 | 1.208 | 1.874 | -0.069 | 0.890 |
| | 3985 | 10.3 | 292. | 13.82 | 0.002 | 0.180 | -0.024 | 1.302 | 1.796 | -0.011 | 0.792 |
| | 3920 | 12.3 | 392. | 15.39 | -0.154 | 0.139 | -0.045 | 1.472 | 1.964 | -0.043 | 0.867 |
| | 3958 | 14.6 | 495. | 16.85 | -0.021 | 0.281 | -0.067 | 1.490 | 1.829 | -0.079 | 0.804 |
| | 3969 | 15.4 | 547. | 17.21 | -0.013 | 0.163 | 0.017 | 1.433 | 2.054 | -0.104 | 0.902 |
| H00120 | 3882 | 8.2 | 206. | 11.93 | -0.024 | 0.345 | 0.003 | 0.997 | 1.389 | -0.045 | 0.489 |
| | 3906 | 10.2 | 281. | 13.69 | 0.087 | 0.539 | -0.012 | 0.921 | 1.102 | -0.033 | 0.404 |
| | 3995 | 10.2 | 282. | 13.84 | -0.001 | 0.412 | 0.010 | 1.067 | 1.339 | -0.055 | 0.455 |
| | 3921 | 12.1 | 382. | 15.14 | 0.053 | 0.423 | -0.033 | 1.167 | 1.373 | -0.028 | 0.499 |
| | 3946 | 13.6 | 435. | 16.16 | 0.038 | 0.394 | -0.005 | 1.174 | 1.327 | -0.048 | 0.564 |
| | 3954 | 14.1 | 479. | 16.50 | 0.084 | 0.431 | -0.023 | 1.243 | 1.328 | -0.060 | 0.559 |
| H00127 | 3883 | 8.2 | 206. | 11.99 | -0.035 | 0.322 | -0.008 | 1.034 | 1.439 | -0.043 | 0.471 |
| | 3904 | 10.3 | 277. | 13.76 | -0.094 | 0.386 | -0.017 | 1.237 | 1.380 | -0.059 | 0.431 |
| | 3924 | 12.4 | 387. | 15.43 | -0.008 | 0.352 | -0.030 | 1.200 | 1.451 | -0.057 | 0.505 |
| | 3949 | 13.6 | 445. | 16.09 | 0.029 | 0.447 | -0.068 | 1.186 | 1.343 | -0.047 | 0.405 |
| | 3959 | 14.8 | 467. | 16.84 | 0.091 | 0.393 | -0.120 | 1.179 | 1.387 | -0.013 | 0.392 |
| H00131 | 3894 | 8.4 | 205. | 12.07 | -0.136 | 0.189 | 0.057 | 1.044 | 1.521 | -0.111 | 0.559 |
| | 3908 | 10.2 | 275. | 13.71 | -0.040 | 0.281 | -0.022 | 1.085 | 1.432 | -0.075 | 0.565 |
| | 3999 | 10.3 | 287. | 13.93 | -0.074 | 0.198 | -0.026 | 1.017 | 1.453 | -0.083 | 0.503 |
| | 3926 | 12.1 | 380. | 15.03 | -0.131 | 0.136 | 0.061 | 1.313 | 1.755 | -0.125 | 0.552 |
| | 3987 | 14.5 | 480. | 16.76 | 0.005 | 0.234 | 0.043 | 1.248 | 1.511 | -0.122 | 0.639 |
| | 3990 | 15.4 | 527. | 17.26 | 0.011 | 0.164 | -0.001 | 1.194 | 1.515 | -0.069 | 0.566 |
| H00132 | 3997 | 8.1 | 203. | 12.12 | -0.228 | 0.393 | -0.141 | 0.975 | 1.204 | -0.016 | 0.332 |
| | 3989 | 10.2 | 290. | 13.78 | -0.222 | 0.313 | -0.087 | 1.107 | 1.164 | -0.035 | 0.300 |
| | 3927 | 12.2 | 377. | 15.34 | -0.257 | 0.257 | -0.107 | 1.354 | 1.550 | -0.016 | 0.560 |
| | 3950 | 13.6 | 441. | 16.14 | -0.242 | 0.293 | -0.103 | 1.508 | 1.597 | 0.000 | 0.532 |
| | 3957 | 14.6 | 509. | 16.75 | -0.194 | 0.337 | -0.080 | 1.457 | 1.552 | -0.032 | 0.552 |
| | 3982 | 15.6 | 542. | 17.47 | -0.161 | 0.322 | -0.097 | 1.493 | 1.635 | -0.020 | 0.581 |
| H00133 | 3895 | 8.2 | 206. | 12.00 | -0.135 | 0.298 | -0.012 | 1.364 | 1.382 | -0.027 | 0.405 |
| | 3913 | 10.3 | 274. | 13.87 | 0.060 | 0.315 | -0.035 | 1.235 | 1.391 | -0.067 | 0.455 |
| | 3998 | 10.2 | 285. | 13.71 | 0.104 | 0.294 | -0.018 | 1.018 | 1.335 | -0.034 | 0.443 |
| | 3939 | 12.4 | 382. | 15.36 | -0.138 | 0.204 | -0.005 | 1.463 | 1.488 | -0.046 | 0.451 |
| | 3963 | 14.5 | 476. | 16.69 | -0.020 | 0.259 | -0.059 | 1.368 | 1.391 | -0.008 | 0.414 |
| | 3986 | 15.6 | 538. | 17.31 | -0.054 | 0.149 | -0.004 | 1.506 | 1.589 | -0.083 | 0.541 |
| H00135 | 3898 | 8.3 | 207. | 12.15 | -0.131 | 0.296 | -0.025 | 1.108 | 1.362 | -0.061 | 0.553 |
| | 3916 | 10.3 | 276. | 13.83 | -0.259 | 0.302 | 0.017 | 1.384 | 1.422 | -0.080 | 0.487 |
| | 3941 | 12.5 | 382. | 15.52 | -0.223 | 0.423 | 0.001 | 1.382 | 1.382 | -0.057 | 0.479 |
| | 3955 | 13.6 | 444. | 16.18 | -0.212 | 0.414 | -0.010 | 1.416 | 1.391 | -0.054 | 0.484 |
| | 3965 | 14.6 | 493. | 16.67 | -0.233 | 0.437 | -0.004 | 1.544 | 1.444 | -0.064 | 0.466 |
| | 3970 | 15.6 | 534. | 17.26 | -0.170 | 0.387 | -0.021 | 1.373 | 1.470 | -0.065 | 0.481 |
| H00136 | 3901 | 7.9 | 200. | 11.88 | -0.033 | 0.366 | 0.066 | 1.000 | 1.128 | -0.071 | 0.432 |
| | 3918 | 10.2 | 293. | 14.03 | 0.035 | 0.418 | 0.035 | 1.139 | 1.219 | -0.064 | 0.421 |
| | 3942 | 12.0 | 382. | 15.26 | 0.029 | 0.308 | 0.072 | 1.236 | 1.438 | -0.113 | 0.408 |
| | 3953 | 13.3 | 443. | 16.03 | 0.045 | 0.415 | 0.069 | 1.241 | 1.251 | -0.083 | 0.362 |
| | 3962 | 14.1 | 474. | 16.59 | 0.078 | 0.343 | 0.067 | 1.277 | 1.250 | -0.096 | 0.319 |

SUMMARY OF TEST CONDITIONS AND TEST RESULTS (cont.)

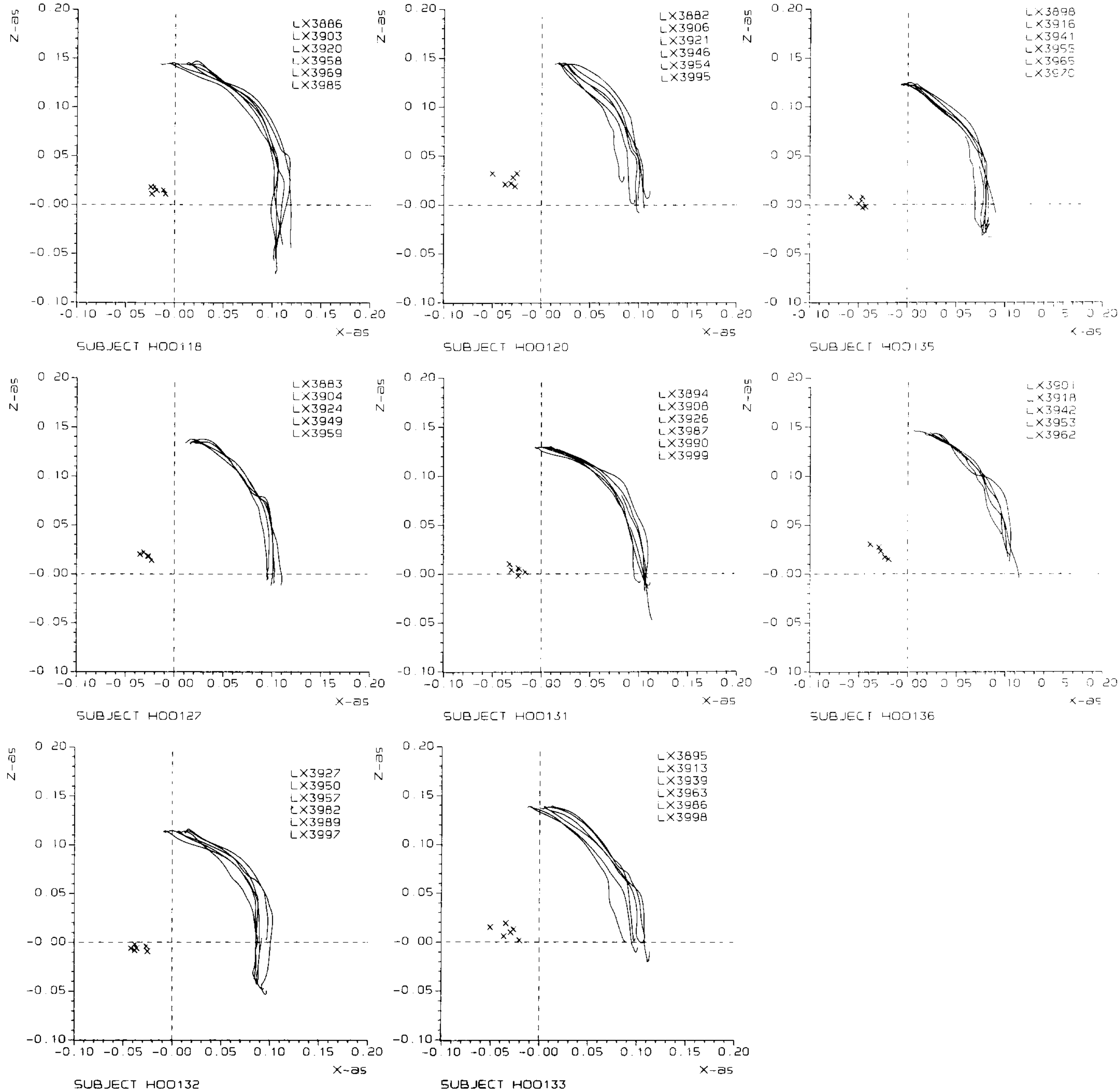
| Subject | No. | Peak sled acc. (G) | Rate of onset (G/s) | Sled vel. change (m/s) | $\phi_o$ (rad) | $\theta_o$ (rad) | $\psi_o - \frac{\pi}{4}$ (rad) | $\phi_{max} - \phi_o$ (rad) | $\theta_{max} - \theta_o$ (rad) | $\psi_{max} - \psi_o$ (rad) | $((\theta-\theta_o)-(\phi-\phi_o))_{max}$ (rad) |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Oblique experiments. | | | | | | | | | | | |
| H00130 | 4235 | 7.0 | 159. | 10.89 | -0.012 | 0.167 | -0.002 | 0.953 | 1.405 | -0.336 | 0.534 |
| | 4301 | 9.3 | 252. | 13.12 | 0.022 | 0.182 | -0.044 | 1.015 | 1.321 | -0.432 | 0.432 |
| | 4286 | 10.1 | 281. | 13.64 | -0.030 | 0.072 | -0.042 | 1.359 | 1.766 | -0.464 | 0.663 |
| | 4309 | 11.3 | 334. | 14.64 | -0.073 | 0.061 | 0.005 | 1.333 | 1.638 | -0.524 | 0.492 |
| H00132 | 4244 | 7.3 | 175. | 11.40 | -0.138 | 0.215 | -0.125 | 0.689 | 0.904 | -0.313 | 0.295 |
| | 4261 | 9.0 | 240. | 12.72 | -0.083 | 0.410 | -0.089 | 0.890 | 0.990 | -0.431 | 0.370 |
| | 4287 | 10.2 | 288. | 13.83 | -0.198 | 0.117 | -0.089 | 1.156 | 1.175 | -0.519 | 0.275 |
| | 4297 | 10.0 | 282. | 13.78 | -0.390 | 0.096 | 0.025 | 1.361 | 1.266 | -0.652 | 0.156 |
| | 4306 | 11.1 | 335. | 14.54 | -0.053 | 0.308 | -0.184 | 0.905 | 0.991 | -0.440 | 0.340 |
| H00133 | 4236 | 7.3 | 171. | 11.36 | -0.051 | 0.245 | 0.001 | 0.778 | 0.864 | -0.169 | 0.263 |
| | 4240 | 9.1 | 236. | 12.95 | -0.028 | 0.165 | 0.007 | 0.959 | 1.070 | -0.241 | 0.279 |
| H00134 | 4237 | 7.2 | 170. | 11.26 | -0.181 | 0.136 | 0.025 | 0.840 | 1.074 | -0.348 | 0.315 |
| | 4264 | 9.3 | 248. | 13.03 | -0.107 | 0.158 | 0.041 | 1.036 | 1.250 | -0.349 | 0.382 |
| | 4298 | 10.1 | 285. | 13.78 | -0.155 | 0.185 | 0.071 | 1.219 | 1.312 | -0.455 | 0.266 |
| | 4307 | 11.4 | 337. | 14.89 | -0.089 | 0.139 | 0.017 | 1.264 | 1.368 | -0.488 | 0.363 |
| H00135 | 4238 | 7.3 | 169. | 11.44 | -0.062 | 0.343 | -0.047 | 0.731 | 0.903 | -0.224 | 0.238 |
| | 4314 | 9.1 | 244. | 12.90 | -0.004 | 0.309 | -0.037 | 0.815 | 1.093 | -0.334 | 0.316 |
| | 4316 | 10.1 | 290. | 13.62 | -0.004 | 0.347 | -0.042 | 0.915 | 1.147 | -0.393 | 0.331 |
| H00136 | 4247 | 7.1 | 166. | 11.11 | 0.018 | 0.242 | 0.057 | 0.749 | 0.832 | -0.437 | 0.160 |
| | 4263 | 9.2 | 247. | 12.98 | 0.068 | 0.093 | 0.068 | 0.935 | 1.041 | -0.605 | 0.270 |
| H00138 | 4241 | 7.2 | 167. | 11.17 | -0.079 | 0.139 | -0.030 | 0.909 | 1.236 | -0.345 | 0.393 |
| | 4265 | 9.2 | 245. | 12.98 | -0.045 | 0.127 | -0.013 | 1.163 | 1.468 | -0.403 | 0.413 |
| | 4296 | 10.1 | 286. | 13.77 | -0.075 | 0.035 | -0.064 | 1.253 | 1.546 | -0.460 | 0.563 |
| | 4305 | 11.4 | 342. | 14.81 | -0.062 | 0.099 | -0.038 | 1.346 | 1.607 | -0.522 | 0.482 |
| H00139 | 4243 | 7.3 | 170. | 11.31 | -0.011 | 0.151 | -0.086 | 0.989 | 1.268 | -0.268 | 0.360 |
| | 4313 | 9.1 | 252. | 12.81 | -0.005 | 0.218 | -0.079 | 1.083 | 1.271 | -0.307 | 0.252 |
| | 4291 | 10.3 | 290. | 13.91 | 0.013 | 0.116 | -0.021 | 1.326 | 1.587 | -0.499 | 0.476 |
| | 4303 | 11.3 | 339. | 14.77 | -0.031 | 0.223 | 0.000 | 1.311 | 1.507 | -0.454 | 0.391 |
| H00140 | 4259 | 7.2 | 172. | 11.18 | 0.066 | 0.242 | -0.040 | 0.588 | 0.951 | -0.225 | 0.410 |
| | 4302 | 9.1 | 248. | 12.96 | 0.104 | 0.358 | 0.008 | 0.677 | 0.902 | -0.356 | 0.256 |
| | 4293 | 10.1 | 286. | 13.76 | 0.035 | 0.198 | 0.084 | 0.886 | 1.145 | -0.502 | 0.335 |
| | 4310 | 11.2 | 334. | 14.69 | 0.009 | 0.172 | -0.069 | 0.916 | 1.210 | -0.317 | 0.324 |

SUMMARY OF TEST CONDITIONS AND TEST RESULTS (cont.)

| Subject | No. | Peak sled acc. (G) | Rate of onset (G/s) | Sled vel. change (m/s) | $\phi_o$ (rad) | $\theta_o$ (rad) | $\psi_o - \frac{\pi}{2}$ (rad) | $\phi_{max} - \phi_o$ (rad) | $\theta_{max} - \theta_o$ (rad) | $\psi_{max} - \psi_o$ (rad) | $((\theta-\theta_o)-(\phi-\phi_o))_{max}$ (rad) |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Lateral experiments. | | | | | | | | | | | |
| HO0133 | 4093 | 5.1 | 105. | 7.11 | 0.020 | 0.093 | 0.016 | 0.671 | 0.628 | -0.331 | 0.124 |
| | 4111 | 6.1 | 132. | 7.25 | 0.033 | 0.007 | -0.003 | 0.939 | 0.847 | -0.454 | 0.097 |
| | 4151 | 6.1 | 130. | 7.17 | 0.057 | -0.005 | -0.032 | 0.849 | 0.743 | -0.468 | 0.168 |
| | 4125 | 7.2 | 164. | 7.02 | 0.057 | 0.107 | -0.042 | 0.920 | 0.734 | -0.460 | 0.168 |
| HO0134 | 4097 | 5.0 | 104. | 7.06 | -0.012 | 0.134 | -0.024 | 0.648 | 0.683 | -0.691 | 0.147 |
| | 4112 | 6.1 | 128. | 7.13 | -0.064 | 0.170 | 0.096 | 0.772 | 0.613 | -0.758 | 0.057 |
| | 4126 | 7.1 | 167. | 6.90 | -0.030 | 0.106 | 0.059 | 0.864 | 0.730 | -0.695 | 0.100 |
| HO0135 | 4095 | 5.2 | 105. | 7.14 | 0.041 | 0.144 | -0.059 | 0.630 | 0.602 | -0.572 | 0.128 |
| | 4114 | 6.1 | 132. | 7.19 | 0.066 | 0.098 | -0.047 | 0.651 | 0.655 | -0.564 | 0.198 |
| | 4131 | 7.3 | 164. | 6.94 | 0.038 | 0.082 | -0.043 | 0.728 | 0.670 | -0.622 | 0.163 |
| HO0136 | 4098 | 5.1 | 106. | 7.03 | -0.030 | -0.064 | 0.099 | 0.642 | 0.712 | -0.652 | 0.155 |
| | 4142 | 6.0 | 131. | 7.07 | -0.007 | -0.069 | 0.096 | 0.771 | 0.820 | -0.734 | 0.259 |
| | 4153 | 7.1 | 157. | 6.86 | -0.016 | -0.075 | 0.121 | 0.912 | 0.889 | -0.861 | 0.218 |
| HO0138 | 4092 | 5.1 | 106. | 7.05 | -0.015 | 0.049 | -0.081 | 0.891 | 0.944 | -0.686 | 0.202 |
| | 4115 | 6.0 | 127. | 7.11 | 0.012 | -0.121 | -0.003 | 0.932 | 1.011 | -0.672 | 0.251 |
| | 4147 | 6.0 | 126. | 7.06 | -0.024 | -0.047 | -0.084 | 0.881 | 0.977 | -0.655 | 0.259 |
| | 4129 | 7.2 | 162. | 6.92 | -0.014 | -0.074 | -0.059 | 1.000 | 1.060 | -0.666 | 0.321 |
| HO0139 | 4100 | 5.1 | 107. | 7.08 | 0.044 | -0.054 | 0.050 | 0.917 | 0.878 | -0.590 | 0.152 |
| | 4118 | 6.1 | 128. | 7.13 | 0.032 | -0.078 | 0.005 | 1.075 | 0.991 | -0.570 | 0.187 |
| | 4144 | 6.1 | 130. | 7.25 | 0.003 | -0.080 | -0.016 | 1.091 | 1.010 | -0.614 | 0.212 |
| | 4133 | 7.2 | 165. | 6.91 | 0.004 | 0.024 | 0.006 | 1.179 | 1.091 | -0.579 | 0.187 |
| HO0140 | 4099 | 5.1 | 108. | 7.11 | 0.029 | 0.069 | 0.056 | 0.598 | 0.784 | -0.520 | 0.186 |
| | 4116 | 6.1 | 128. | 7.20 | 0.021 | 0.067 | 0.109 | 0.694 | 0.848 | -0.665 | 0.181 |
| | 4145 | 6.1 | 127. | 7.20 | 0.005 | -0.009 | 0.105 | 0.577 | 0.788 | -0.576 | 0.211 |
| | 4130 | 7.1 | 161. | 6.89 | 0.018 | 0.009 | 0.127 | 0.727 | 0.924 | -0.701 | 0.203 |
| HO0141 | 4094 | 5.1 | 106. | 7.12 | 0.079 | 0.042 | 0.055 | 0.716 | 0.832 | -0.896 | 0.126 |
| | 4119 | 5.9 | 128. | 7.06 | 0.075 | -0.046 | 0.063 | 0.852 | 0.975 | -0.795 | 0.127 |
| | 4134 | 7.1 | 161. | 6.85. | 0.073 | 0.021 | 0.055 | 0.900 | 1.033 | -0.747 | 0.198 |
| HO0142 | 4104 | 5.1 | 108. | 6.99 | -0.047 | 0.008 | -0.069 | 0.759 | 0.821 | -0.705 | 0.197 |
| | 4120 | 6.1 | 131. | 7.06 | 0.035 | 0.095 | -0.133 | 0.843 | 0.866 | -0.683 | 0.260 |
| | 4135 | 7.2 | 161. | 6.87 | -0.046 | 0.041 | -0.017 | 0.911 | 0.856 | -0.738 | 0.220 |

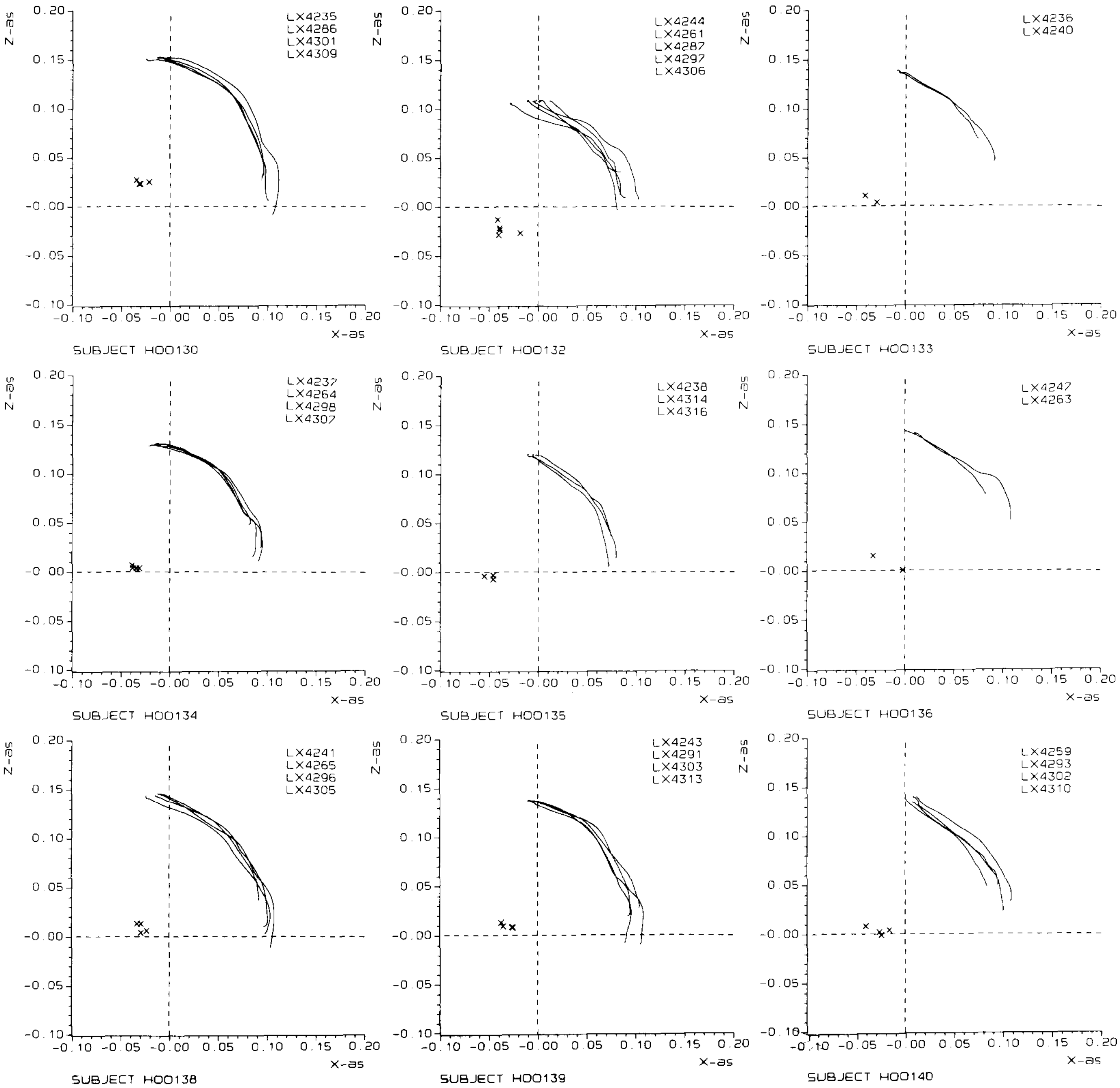
APPENDIX B

OCCIPITAL CONDYLE TRAJECTORIES RELATIVE TO CORRECTED T1 COORDINATE SYSTEM



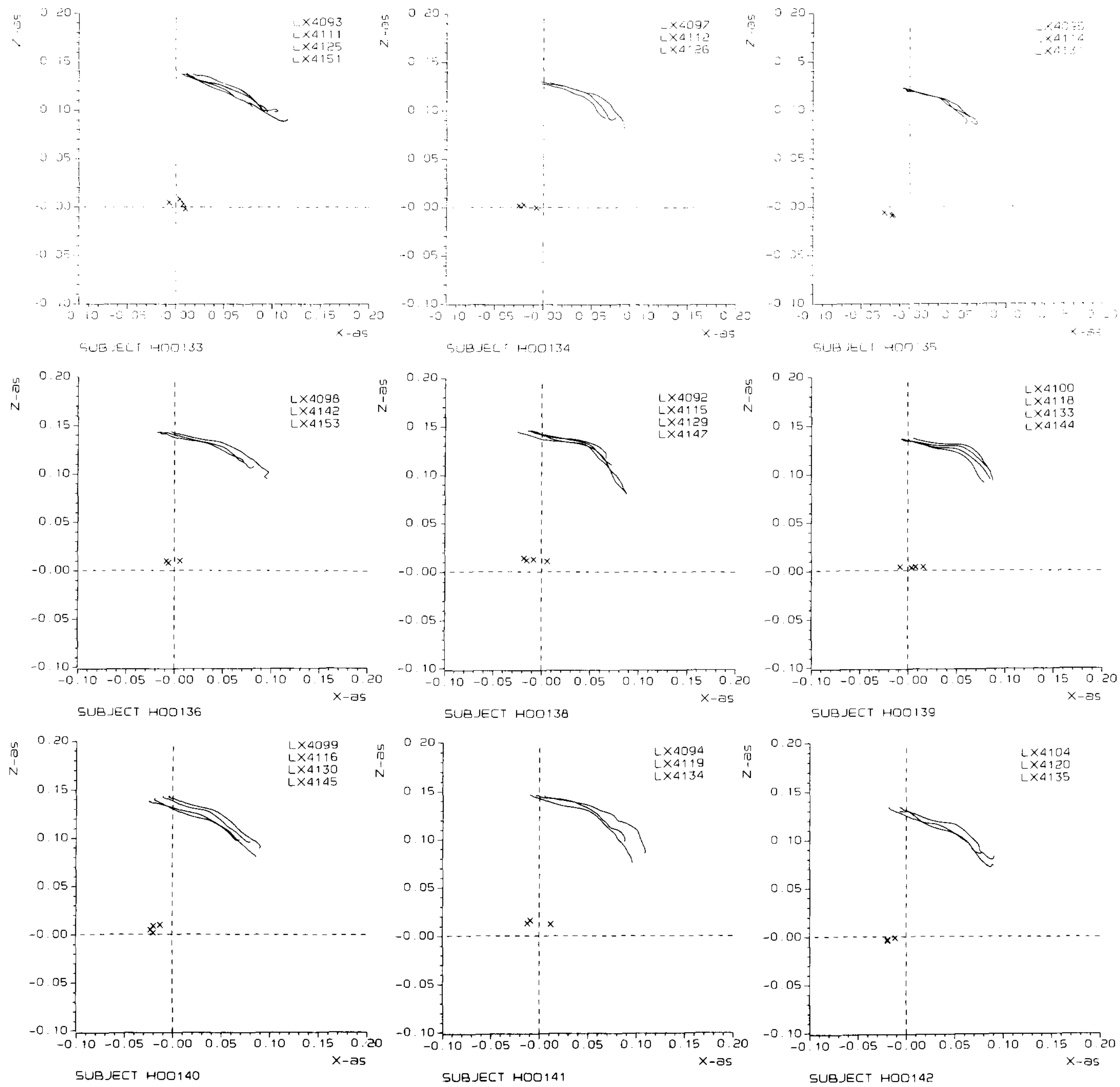

FRONTAL

OCCIPITAL CONDYLE TRAJECTORIES RELATIVE TO CORRECTED T1 COORDINATE SYSTEM (cont.)




OBLIQUE

OCCIPITAL CONDYLE TRAJECTORIES RELATIVE TO CORRECTED T1 COORDINATE SYSTEM (cont.)




LATERAL