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Role of Orientation Reference Selection in Motion Sickness

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STATEMENT OF WORK SUMMARY

The overall objective of this proposal is to understand the relationship between human orientation control and motion sickness susceptibility. Three areas related to orientation control will be investigated. These three areas are 1) reflexes associated with the control of eye movements and posture, 2) the perception of body rotation and position with respect to gravity, and 3) the strategies used to resolve sensory conflict situations which arise when different sensory systems provide orientation cues which are not consistent with one another or with previous experience. Of particular interest is the possibility that a subject may be able to ignore an inaccurate sensory modality in favor of one or more other sensory modalities which do provide accurate orientation reference information. We refer to this process as sensory selection. This proposal will attempt to quantify subjects' sensory selection abilities and determine if this ability confers some immunity to the development of motion sickness symptoms.

Measurements of reflexes, motion perception, sensory selection abilities, and motion sickness susceptibility will concentrate on pitch and roll motions since these seem most relevant to the space motion sickness problem. Vestibulo-ocular (VOR) and oculomotor reflexes will be measured using a unique two-axis rotation device developed in our laboratory over the last four years. Posture control reflexes will be measured using a movable posture platform capable of independently altering proprioceptive and visual orientation cues. Motion perception will be quantified using closed loop feedback technique developed by Zacharias and Young (Exp Brain Res, 1981). This technique requires a subject to null out motions induced by the experimenter while being exposed to various confounding sensory orientation cues. A subject's sensory selection abilities will be measured by the magnitude and timing of his reactions to changes in sensory environments. Motion sickness susceptibility will be measured by the time required to induce characteristic changes in the pattern of electrogastrogram recordings while exposed to various sensory environments during posture and motion perception tests.

The results of this work are relevant to NASA's interest in understanding the etiology of space motion sickness. If any of the reflex, perceptual, or sensory selection abilities of subjects are found to correlate with motion sickness susceptibility, this work may be an important step in suggesting a method of predicting motion sickness susceptibility. If sensory selection can provide a means to avoid sensory conflict, then further work may lead to training programs which could

enhance a subject's sensory selection ability and therefore minimize motion sickness susceptibility.

SUMMARY OF PROJECT STATUS

Three test devices are required for the proposed experiments. They are (1) a moving posture platform, (2) a servo-controlled vertical axis rotation chair with an independently controllable optokinetic stimulator, and (3) a two-axis rotation chair for the generation of pitch and roll motions. The first two devices have been functional for quite some time and are routinely used for both clinical and research testing. The two-axis rotation device has become operational as of mid-August 1990. The development of this two-axis rotator has been a major focus of work and will be described in more detail below.

An important component associated with the two-axis rotator is a computer controlled video system for the measurement of eye movements. This video system for recording horizontal and vertical eye movements has been working for the past six months. We recently added the capability to measure torsional eye movements. The quality of the eye movement recordings are exceptional.

This new ability to record torsional eye movements should add considerable versatility in the design of experiments related to this grant. This is because torsional eye movements are closely associated with the vertical semicircular canals and otolith receptors, which in turn are implicated in the space motion sickness syndrome. In addition, very little is known about the response properties of torsional eye movements as a function of changes in body position with respect to the gravity vector. We have begun a project to characterize the dynamic response characteristics of torsional eye movements during roll rotations about an upright position.

Another experiment in progress involves the determination of the influence of visual, somatosensory, and vestibular motion cues on the control of posture.

An initial set of experiments involving the perceptual feedback technique developed by Zacharias and Young (Exp Brain Res, 41:159-171, 1981) have been completed. These experiments were designed to look for correlations between vestibulo-ocular reflex parameters and the perception of rotation. A paper describing these results is nearing completion.

Four papers describing earlier work on the VOR and posture control function in a large normal population have been accepted for publication and are currently in press in the Journal of Vestibular Research.

TWO-AXIS ROTATOR DEVELOPMENT

The two-axis rotator is a versatile, general purpose stimulator for vestibular and visual-vestibular interaction studies. It consists of two gimbals powered by rotary hydraulic actuators. A single DC torque motor is now available which is interchangeable with either of the hydraulic actuators. The inner gimbal produces yaw axis rotations of the subject. The outer gimbal rotates the subject about a horizontal axis which passes through the subject's ears.

We have completed the essential parts of 5 major projects related to the two-axis rotator in the past several months. These are (1) the installation of various mechanical, hydraulic, electronic, and computer software safety devices and procedures, (2) calibrations of the two-axis rotator motions, (3) tuning of the servo controls for optimum performance, (4) improvements in the data collection and stimulus delivery computer programs, (5) development of an improved system for the video recording and automated analysis of eye movements, including torsional eye movements.

EXPERIMENTS IN PROGRESS

Two experiments are currently being performed. One is an investigation to characterize the influence of visual orientation cues on the control of posture. The second is to measure the dynamic response properties of human ocular torsion in response to roll rotations. The results of both of these experiments will be used to develop a rating of individual subject performance in various reflex and posture control tasks so that a correlation with motion sickness susceptibility can be identified (if the correlation exists).

The Role of Vision in Posture. These experiments are performed on a moving posture platform. The subject stands facing a high contrast visual field. This visual field can be placed in motion by rotating the visual field in an anterior-posterior direction about an axis which passes through the subject's ankle joints. We have been using sinusoidal motions of the visual field at frequencies of 0.1, 0.2, and 0.5 Hz with amplitudes of 1, 2, 5, and 10 degrees presented in random order. In addition, in half of the tests the surface upon which the subject stands is "sway-referenced" in order to alter the somatosensory cues which are available for posture control. Sway-referencing involves the controlled rotation of the platform upon which the subject stands in proportion to the subject's own sway. This results in very little change in the subject's ankle joint angle even though the subject is swaying forward and backward. We record the

subject's anterior-posterior sway at waist and shoulder level. From those measures we estimate the sway angle of the subject's center of mass throughout the trial. A Fourier analysis is used to estimate the average amplitude of the center-of-mass body sway at the stimulus frequency.

Figure 1 shows typical results from a normal subject in response to 0.2 Hz sinusoidal rotations of the visual field at various amplitudes while the subject stood on a fixed surface (left column) and a sway-referenced support surface (right column). Sway-referencing of the support surface refers to a technique which alters the normal relationship between body sway and the rotation of the subject's ankle joint angle. This technique apparently reduces the somatosensory signals available for the control of body sway, and therefore forces a greater reliance on other sensory system information (visual and vestibular in particular). Sway-referencing of the support surface is accomplished by actively rotating the support surface angle in proportion to the subject's sway angle.

Figure 1 shows that this normal subject's sway was only slightly influenced by the "false" visual orientation cues resulting from the sinusoidally rotating visual field when the subject stood on a fixed support surface. Sway increased when the subject stood on the sway-referenced support surface. However when the amplitude of the rotating visual field increased, the subject's sway did not correspondingly increase. This suggests that the subject's somatosensory and vestibular systems in the fixed platform case, and the vestibular system in the sway-referenced case, provided sensory cues which were used by the brain's posture control mechanisms to limit the response to the visual stimulus.

Figure 2 shows the results for a subject with complete bilateral loss of vestibular function during the same conditions. At low amplitudes of the visual field stimulus, the sway of this subject was clearly influenced by the moving visual field. At higher stimulus amplitudes with fixed platform, the bilateral loss subject consistently swayed more than the normal subject. At higher stimulus amplitudes with a sway-referenced platform to reduce somatosensory cues, the bilateral loss subject consistently fell since the subject did not have any source of sensory information which provided an accurate orientation reference.

What is not shown in these figures is that there was a wide range of sensitivities to the visual field motion among the normal subjects. At low stimulus amplitudes, some of the normal subjects showed similar sway amplitudes to bilateral deficit subjects while other normals showed much less. This suggests that there is considerable variation of the behavioral weighting of sensory orientation cues among normal subjects. As this grant work progresses, this variation will provide us with a scale

of performance against which motion sickness susceptibility can be compared.

Ocular Torsion. Rotations of the head about a naso-occipital axis stimulate the vertical semicircular canals and the otolith organs (depending on the orientation of the head with respect to gravity). Signals from these vestibular receptors produce torsional or counterrolling eye movements. As with other aspects of the vestibulo-ocular reflex (VOR), the presumed function is to stabilize images on the retina during head motion in order to insure clear vision. The need to torsionally stabilize eye movements during rolling head movements would seem to be less important than during pitching and yawing head movements since image motion at the eye's fovea, the region of highest acuity, is relatively small during head rolls. Therefore the results of counterrolling experiments which have demonstrated very low gains during static head positions, and relatively low gains during moderate frequency (0.1 to 0.8 Hz), actively generated head rolls seem to confirm the thought that this reflex is not very functionally significant.

Our results show that ocular torsion gains are actually quite large during head motions which resemble those which can occur during natural, everyday movements. That is, during low amplitude (<20 degrees), high frequency (>~1 Hz) head rolls, the gain of the torsional VOR is close to unity. Figure 3 shows the gain and phase responses of the three subjects tested to date. At 2 Hz, 2 of the three subjects had gains above 0.9, and phases were near zero.

As with the posture experiment results, the variability of results among individuals will be a key point of interest in determining if individual variations in reflex function relate to motion sickness susceptibility. As an example of this variability, the torsion measures from 2 subjects are shown in Figure 4 during a 0.2 Hz, ±20° roll rotation. One subject had very little nystagmus while the other had a great deal of nystagmus. In addition, the subject with the least nystagmus also was the one with the largest phase leads in Figure 3. These types of torsional VOR response differences may represent "strategy" differences among individuals in the way that they choose to use available sensory information for the control of compensatory reflexes. Perhaps these differences in strategies are also associated with either more or less successful abilities to avoid motion sickness symptoms when exposed to environments which give conflicting sensory cues to orientation.

PERCEPTUAL FEEDBACK EXPERIMENTS

In 1981, Zacharias and Young presented a method which allowed for the quantification of a subject's perception of rotation under the combined influence of visual and vestibular cues. In this technique, the subject has control over the rotational motion of the chair by adjusting a potentiometer. Subjects are seated in the vertical axis rotation test room with the potentiometer mounted on the arm of the chair. The output of this potentiometer is summed with a velocity command signal from a computer and this summed signal is delivered to the velocity command input of the chair's servo motor. The goal of the subject is to continuously adjust the potentiometer so that he feels like he is not moving. A "perfect" subject would be able to hold himself stationary in space by adjusting the potentiometer so that its output was equal but opposite to the computer's command signal. "Real" subjects do not remain stationary because of the dynamics of their motion perception and motor reaction systems, and because of presumed imbalances in the vestibular receptors.

Relation of Perceptual Feedback to VOR Test Results. The article by Zacharias and Young suggested that the drift of the subject during rotation in the dark, or with subject-referenced vision, might be due to an imbalance in the encoded motion information coming from the two halves of the vestibular system in opposite ears. This is also the interpretation which is generally given to the presence of bias, or directional preponderance observed in tests of horizontal VOR function. We anticipated that there might be a correlation between the drift observed in perceptual feedback tests and the bias recorded in VOR tests. However we have not found any obvious correlation between these two measures. It may be possible that normal subjects have too small a range of bias and drift to provide a reliable correlation. However the bias measured for a given subject does appear to remain consistent over time, as does drift. That is, the reliability of the measurement of these two parameters seems to be fairly good. This would argue that the lack of correlation between these two measures is real, and not an artifact of their limited range, at least in normal subjects. This observation suggests that there are differences between the static (very low frequency) responses of the VOR and the static properties of motion perception. We believe that exploring these differences and their possible association with motion sickness may be productive.

SCIENTIFIC PAPERS AND PRESENTATIONS

An abstract describing torsional VOR dynamics was recently submitted for presentation at the Association for Research in Otolaryngology in February 1991. A copy of the abstract is attached.

Four papers describing the results of our study of VOR, optokinetic reflex, and moving platform posturography from 200 putatively normal subjects are currently in press in the Journal of Vestibular Research. Copies of the four papers are enclosed.

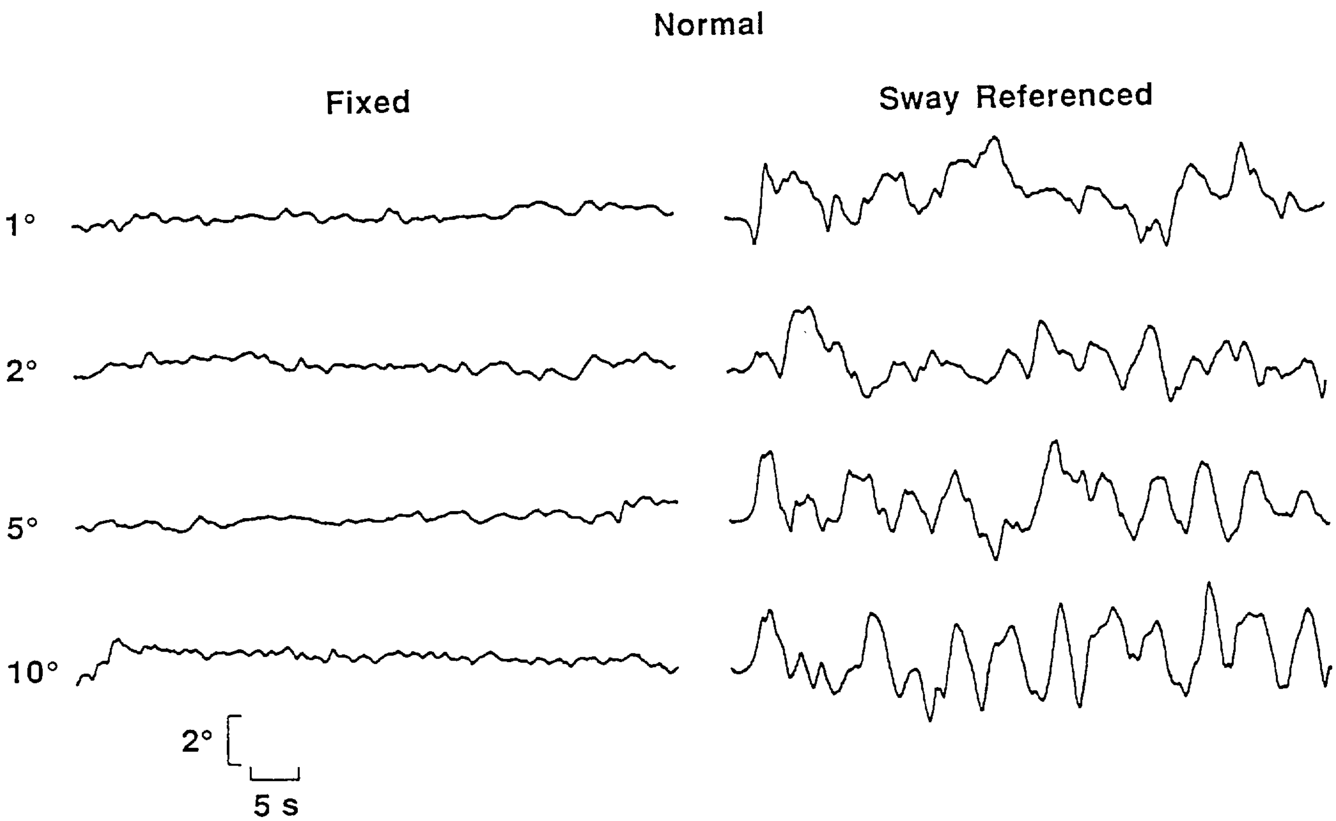

Figure 1. Anterior-posterior center-of-gravity sway angle in response to sinusoidal rotation of a full-field visual surround. Results are for a normal subject standing on a fixed surface (left column of data) and a sway-referenced surface (right column) during 0.2 Hz rotations of the visual surround at amplitudes ranging from ±1° to ±10°.

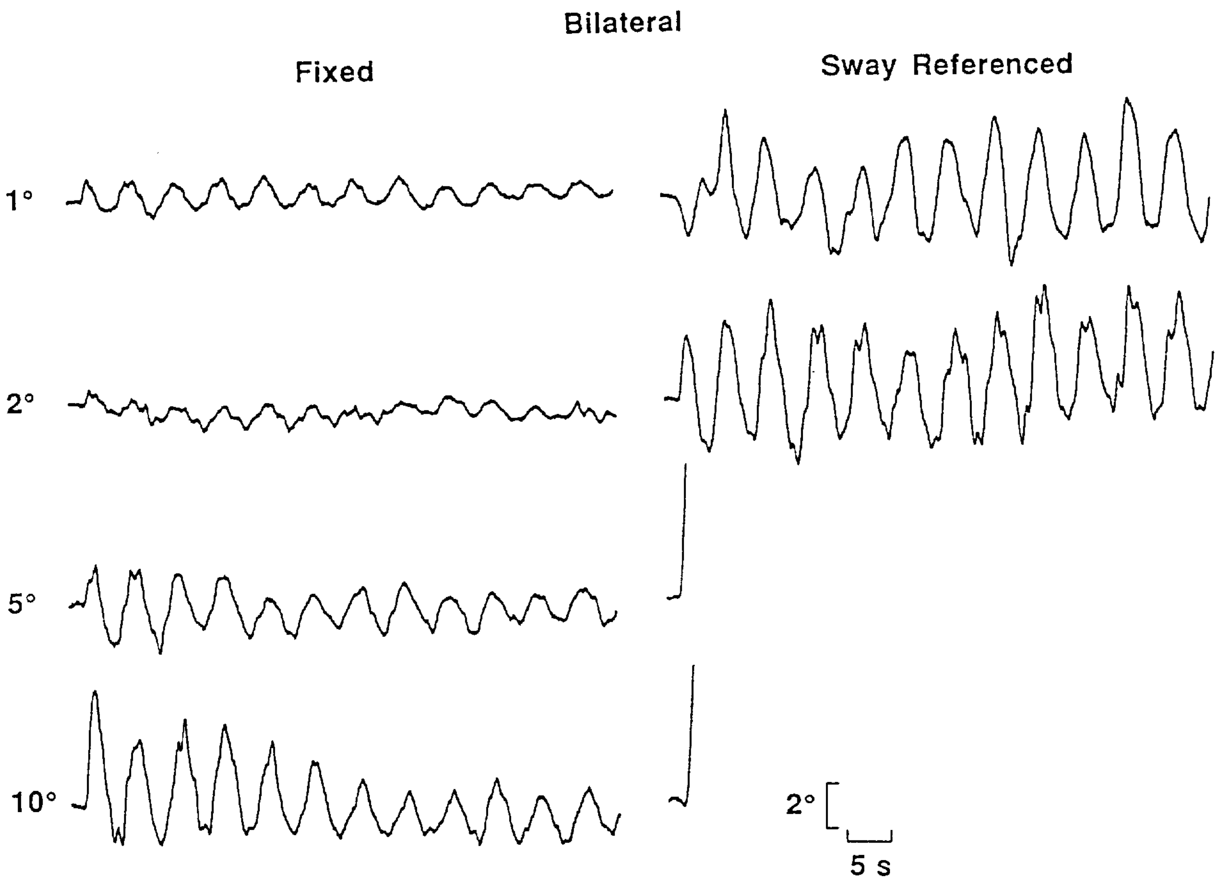

Figure 2. Sway of an abnormal subject with a total bilateral loss of vestibular function responding to the same full-field visual field motions as those in figure 1. Note that the subject fell on the ±5° and ±10° trials when the platform was sway-referenced.

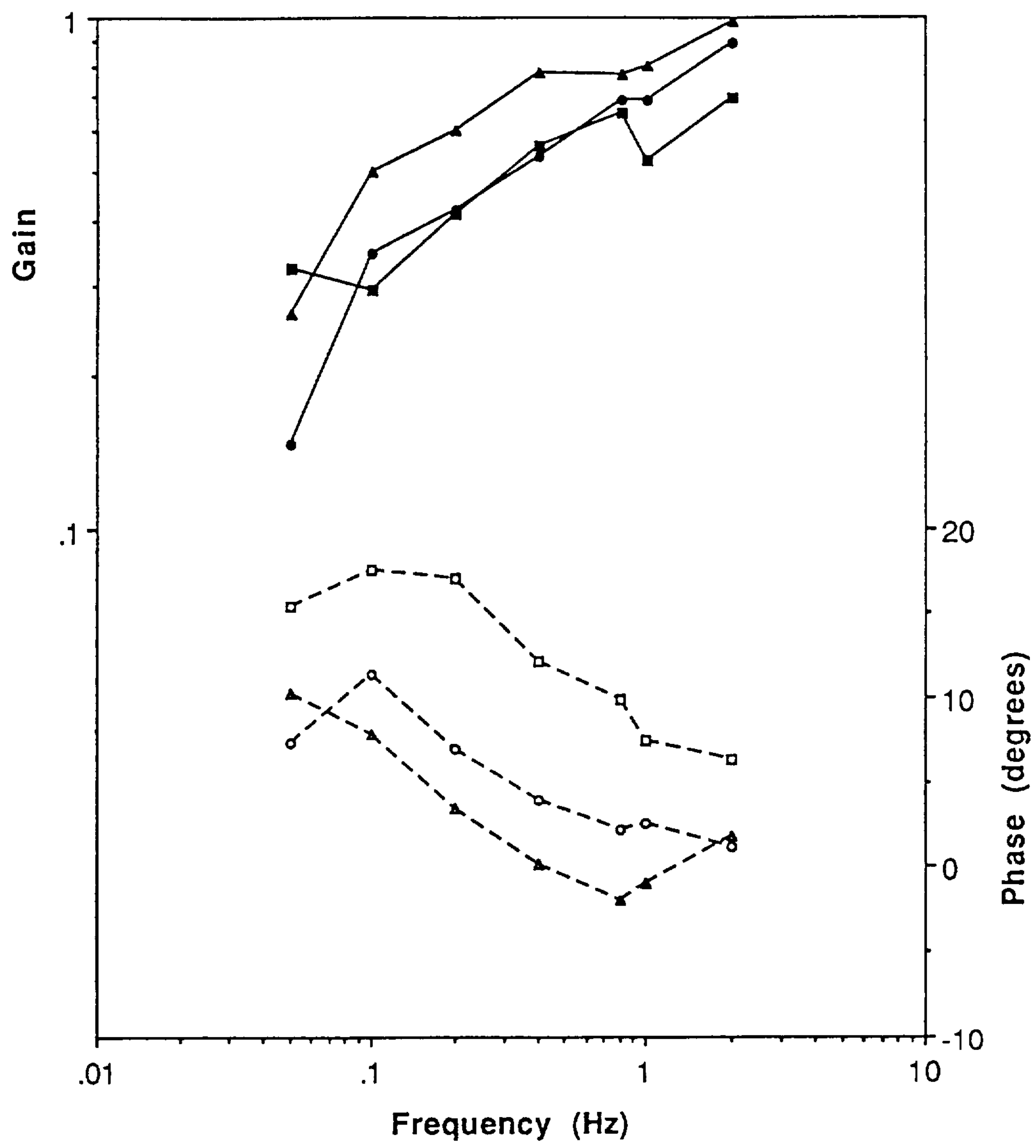

Figure 3. Ocular torsion response dynamics of 3 normal subjects recorded during roll rotations in a dark room while the subjects viewed a single dim fixation light located on the rotation axis. The amplitude (gain = torsional eye velocity/stimulus velocity) and timing (phase with respect to the sinusoidal rotational stimulus) of torsional eye movements changes as a function of the stimulus frequency.

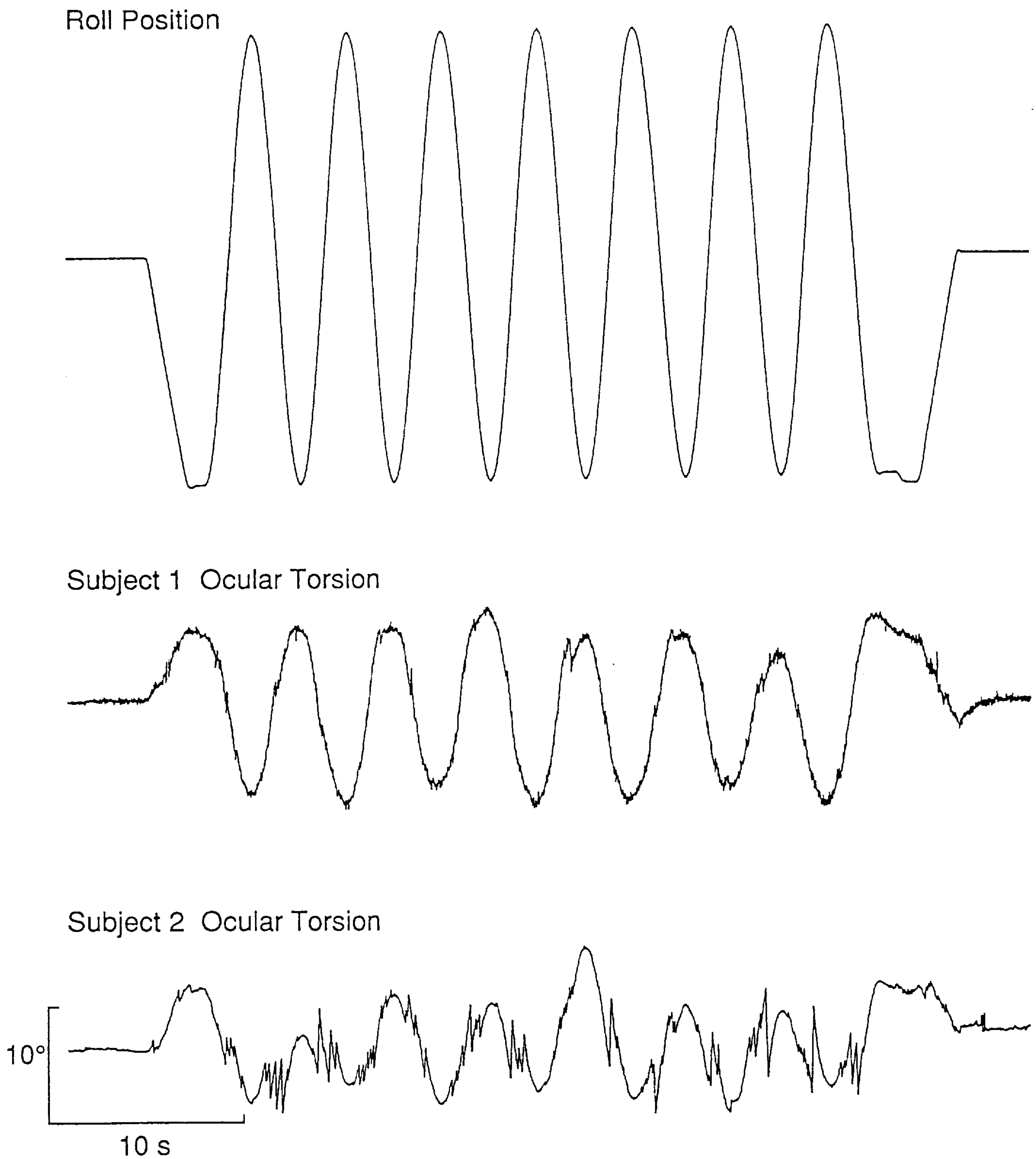

Figure 4. Torsional eye movements of 2 normal subjects recorded during roll rotations in a dark room while the subjects viewed a single dim fixation light located on the rotation axis. The stimulus was a 0.2 Hz sine with ±20° amplitude (top trace). The scale applies to all three traces.

DYNAMIC RESPONSE CHARACTERISTICS OF THE HUMAN TORSIONAL VESTIBULO-OCULAR REFLEX. *R.J. Peterka.
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The torsional vestibulo-ocular reflex (VOR) was measured in three subjects. The stimulus consisted of controlled sinusoidal rotations with amplitudes of ±20° for 0.05 to 0.8 Hz stimuli, ±10° for 1.0 Hz, and ±5° for the 2.0 Hz stimulus. Subjects were rotated in the dark about a naso-occipital axis at the level of the intra-aural axis while viewing a single dim red LED located on the rotation axis about 38 cm in front of their eyes. A small bite-plate mounted video camera recorded eye movements from the right eye under infrared illumination. Each sequential video image (60/s) from a video recording was analyzed off-line by first locating the edges and center of the pupil, and then scanning the intensity of 4 to 6 concentric rings around the iris about midway between the pupil and the sclera. The peak of a cross correlation between the reference iris scan rings obtained at the beginning of each trial and the scan rings from the current video image was used to estimate the ocular torsion. The velocity of the slow phase portions of ocular torsion was calculated and compared to the stimulus velocity in order to calculate torsional VOR gain and phase. Unity VOR gain and 0° phase represent perfect compensatory response dynamics.

Torsional VOR gain generally increased with increasing frequency. Gains at 0.05 Hz ranged from 0.15-0.32 and at 2.0 Hz from 0.69-0.98. At lower frequencies, phase leads were present. Above 0.1 Hz, phases generally declined toward 0° with increasing frequency. Two of the three subjects showed more phase lead at 0.1 Hz than at 0.05 Hz. Phases ranged from 7.4°-15.4° at 0.05 Hz, 7.8°-17.6° at 0.1 Hz, and 1.2°-6.4° at 2.0 Hz.

Previous measures of torsional VOR during active head tilts at frequencies below 1.0 Hz found gains ranging from 0.3 to 0.7 (Ferman et al., Vision Res. 27:811-828, 1987). Although the results presented here were obtained using passive rotations, similar torsional VOR gains were observed at corresponding stimulus frequencies. This study extended the frequency range to 2.0 Hz for the identification of torsional VOR dynamics. At 2.0 Hz, two of the three subjects had gains greater than 0.9 and phases near 0°. This suggests that the torsional VOR can play a significant roll in stabilizing retinal image motion during low amplitude, high frequency head movements.

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AGE-RELATED CHANGES IN HUMAN VESTIBULO-OCULAR REFLEXES: SINUSOIDAL ROTATION AND CALORIC TESTS

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☐ **Abstract – The dynamic response properties of horizontal vestibulo-ocular reflex (VOR) were characterized in 216 human subjects ranging in age from 7 to 81 y. The effects of aging on VOR dynamics and parameter distributions that describe VOR responses to caloric and to sinusoidal rotational stimuli were determined in a putatively normal population. Caloric test parameters showed no consistent trend with age. Rotation test parameters showed declining response amplitude and slightly less compensatory response phase with increasing age. The magnitudes of these changes were not large relative to the variability within the population. The age-related trends in VOR were not consistent with the anatomic changes in the periphery reported by others that showed an increasing rate of peripheral hair cell and nerve fiber loss in subjects over 55 y. The poor correlation between physiological and anatomical data suggest that adaptive mechanisms in the central nervous system are important in maintaining the VOR.**

☐ Keywords – vestibular; eye movements; rotation testing.

Introduction

Age-related changes in the peripheral vestibular organs include loss of hair cells (16), vestibular nerve fibers (4), and Scarpa's ganglion cells (15). The rate of loss of these peripheral vestibular anatomical structures increases in subjects older than ~55 y. If reflex function depends directly on intact peripheral vestibular structures, then we might expect a decline in vestibulo-ocular reflex (VOR) function paralleling anatomical deterioration. Alternatively, if the central adaptive mechanisms remain intact in older subjects, then VOR function may remain relatively stable regardless of peripheral anatomical deterioration.

Literature on age-related changes in vestibular function is limited. Experiments on humans are usually performed on a small number of subjects, and these subjects are typically young adults. The exceptions to this occur in the clinical literature on caloric testing of the VOR (5,11). The results of these studies are rather ambiguous and include ~~both~~ increased, decreased, and unchanged responses with increasing age. Caloric test results are therefore not consistent with the known age-dependent anatomical changes in peripheral vestibular receptors and nerve fibers. Studies on age-related changes in VOR identified using rotation testing are less complete either because the age range of subjects was limited (20,18), or because older analysis methods did not permit detailed characterization of responses to rotation (19). However, small age-related declines have been identified in these studies. Since normal VOR function has not been studied in the same subjects using both caloric and rotation tests, it is possible that the source of this difference between caloric and rotation tests is simply related to the choice of sample test populations.

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The present experiments were designed to characterize the dynamic response properties of human horizontal VOR in a normal population using passive rotational and caloric stimuli. The population was selected to provide results related to the effects of the aging process on these reflexes, and to determine if physiological changes were consistent with anatomic changes which occur with age. Optokinetic reflex and postural control function were also tested in the same subjects on the same day, and are reported in companion papers (12–14).

Methods

Vestibular reflexes were tested in 216 human subjects (90 male and 126 female) aged 7 to 81 y. Ages were approximately uniformly distributed over the entire range. Subjects were required to meet the following criteria:

1. normal age-corrected auditory pure tone responses
2. middle ear reflexes present bilaterally
3. normal middle ear impedance
4. no history of head blows of sufficient magnitude to cause loss of consciousness
5. normal neurologic and otologic physical exams
6. normal corrected vision
7. no history of ototoxic drug use
8. no history of dizziness or disequilibrium
9. moderate or absent use of alcohol with instructions to abstain from alcohol and caffeine 24 h prior to testing
10. no use of psychotropic drugs
11. no history of meningitis, encephalitis, stroke, seizure disorders, diabetes, hypertension, heart disease, or other systemic diseases.

We did not reject subjects based on the results of any of the vestibular, optokinetic, or posture tests performed.

Rotation Tests

Subjects sat in a chair mounted on a 108 N·m velocity servo-controlled motor (Contraves Goerz Corp., Pittsburgh, PA, Model 824) which rotated them about an earth vertical axis. Subjects performed tests of VOR function with eyes closed in a darkened room. Horizontal and vertical eye movements were recorded by electrooculographic (EOG) techniques (bandwidth DC to 80 Hz) using silver/silver chloride electrodes. Horizontal EOG was recorded using bitemporal electrodes, and vertical EOG was recorded by electrodes placed above and below one eye. Stimulus delivery and data collection were controlled by computer (DEC LSI 11/73). Chair tachometer signals as well as horizontal and vertical EOG were digitized and stored for later analysis. Digitizing rates were 200/s for the horizontal EOG and 50/s for vertical EOG and stimulus velocity. Calibrations of the EOG were performed before and after each rotation test.

Rotational stimuli for VOR tests included both a pseudorandom stimulus (14) and single frequency sinusoidal stimuli. Sinusoidal stimuli included 0.05, 0.2, and 0.8 Hz rotations with peak velocities of 60°/s. The duration of sine tests were 100 s (5 cycles) for 0.05 Hz, 45 s (9 cycles) for 0.2 Hz, and 26.25 s (21 cycles) for 0.8 Hz. The first cycle in each data record was considered a transient response and was ignored in the data analysis.

Subjects were given verbal tasks throughout testing to maintain a constant level of alertness. The tasks consisted of alphabetically naming such things as names, places, and foods.

Rotation Test Data Analysis. Eye position data were differentiated to calculate eye velocity. Fast phases of the nystagmus were identified using a method similar to Barnes (3). Curve fits to the remaining slow phase eye velocity data allowed the estimation of VOR response parameters. Curve fits were made to each period of the response. Periods that contained corrupted data were rejected before the final averaging of response parameter values from the remaining periods.

The curve fits to sinusoidal responses were of the form:

$$r(t) = B_r + A_r \sin(2\pi f + P_r) \qquad [1]$$

where B_r is bias in °/s, A_r is response amplitude in °/s, P_r is response phase in degrees, and f is the stimulus frequency. The recorded chair velocity data were separately analyzed to calculate stimulus velocity amplitude, A_s, and phase, P_s. The VOR gain of the reflex is defined as the ratio A_r/A_s, and the phase of the reflex as $P_r - P_s$. Since the VOR is a compensatory reflex, the values of $P_r - P_s$ were close to −180°. For the convenience of working with smaller numbers, a value of 180° was added to $P_r - P_s$ for the VOR test. This is equivalent to inverting the horizontal eye position data.

In order to quantify nonlinear responses, the horizontal eye velocity data was shifted in time by an amount determined by the calculated phase of each period of the response. The time shift was in a direction that brought the response into phase with the stimulus. Slow phase eye velocity was then plotted against stimulus velocity to yield a scatter of points that generally lie along a negatively sloping line. An example is shown in Figure 1. The slope of the line is equal to VOR gain.

A linear VOR response is consistent with equal VOR gains for rotations in opposite directions. One type of nonlinear VOR response, sometimes seen in abnormal subjects, has unequal gains for rotations to the right and left. This type of nonlinearity was quantified by separately calculating the slopes of the eye velocity versus stimulus velocity data for chair rotations to the right and to the left. The slopes were calculated by a ~~least squared error fit~~ of a two-segment line to the data. One line segment was for positive and the other for negative stimulus velocities. The two line segments were constrained to intersect one another at zero stimulus velocity.

The two-part linear curve fit yields three parameters: 1) the reflex gain for slow phase eye movements to the right, G_R; 2) the gain for slow phase eye movements to the left,

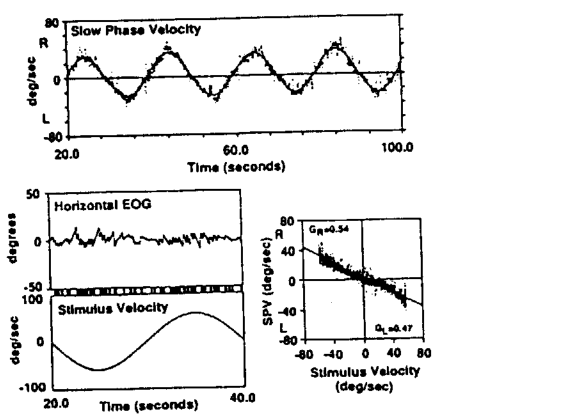

Figure 1. Example of VOR rotation test data. Upper trace shows slow phase eye velocity response to a 0.05 Hz, 60°/s peak velocity sinusoidal rotational stimulus. Solid curve through the data is the curve fit to each cycle. Response gain, phase, and bias are obtained from these curve fits. Lower left trace shows horizontal eye movements evoked by one period of the rotational stimulus. Solid vertical bars under the horizontal EOG trace show the location of fast phase portions of the nystagmus identified in the analysis. Lower right plot shows slow phase eye velocity plotted against stimulus velocity. The two part linear fit is used to measure response symmetry of VOR gain.

G_L; and 3) response offset defined as the eye velocity at zero stimulus velocity. A measure of response asymmetry was calculated according to the formula 100 * $(G_R - G_L)/(G_R + G_L)$. A zero percent asymmetry is consistent with a linear system response where gain is independent of the stimulus direction.

Caloric Tests

Four irrigations of the external ear canals were made using a Brookler-Grams closed loop caloric irrigator. Subjects were in a supine position with head elevated about 30° above horizontal to assure maximal stimulation of the horizontal semicircular canals. The caloric test was not performed on subjects under 12 y, and complete data were not obtained on other subjects who became nauseated or simply chose not to continue the irrigations because of discomfort. Each ear was alternately irrigated for 45 s at 30 and 44°C. Horizontal and vertical eye movements were recorded with EOG techniques identical to those described for rotation tests. Eye movements were recorded during and after each irrigation for a total of 3 min. Horizontal eye movements were analyzed to calculate peak slow phase eye velocity. Caloric responses were quantified by labyrinthine asymmetry (LA), directional preponderance (DP), and average response (AR) measures defined by:

$$LA = \frac{(RW + RC) - (LW + LC)}{RW + RC + LW + LC} \times 100 \quad [2]$$

$$DP = \frac{(RW + LC) - (RC + LW)}{RW + RC + LW + LC} \times 100 \quad [3]$$

$$AR = (RW + RC + LW + LC)/4 \quad [4]$$

where RW, RC, LW, LC are the absolute values of peak slow phase eye velocities recorded during right warm, right cold, left warm, and left cold irrigations, respectively. Subjects were tasked throughout caloric testing to maintain alertness.

Visualization of Trends

In order to visualize trends in scatterplots, a robust locally weighted regression analysis (lowess fit) was used to smooth the scatterplots (6). This smoothing is similar to a moving average filter but is less sensitive to outlying points and allows variable amounts of smoothing. A lowess smoothing parameter of 0.5 and iteration parameter of 2 were used on all data sets.

Data Quality

The overall quality of each rotation and caloric test for each subject was subjectively given a rating of good, fair, or poor. Only good and fair quality data are included in the data summaries in the results section. Quality judgments were based on the standard deviation of response parameters (such as gain, phase, and bias from rotation tests), on the consistency of the responses throughout the duration of the stimulus, and on the accuracy of the eye movement analysis in the separation of slow and fast phases of nystagmus. The actual values of response parameters were not used in judgment of data quality. The test results from about 4% of subjects were rated poor for each test. Poor quality data for one subject on a given test were not used to disqualify ~~other~~ data from the same subject on other tests.

Results

The subjects showed a wide range of responses on all measures of VOR function. Age-related changes were identified in many rotation test response measures, but the magnitude of these changes was not large relative to the variability of the data. Most changes indicated a decline in function. In contrast, no obvious or consistent changes as a function of age were found in caloric test responses. There were no significant differences in reflexes between males and females.

VOR Responses to Rotation

Typical VOR rotation test results are shown in Figure 1. Population statistics for gain, phase, bias, offset, and asymmetry are given in Table 1. The small differences in *N*s are due to data eliminated because of poor quality. The distributions of all parameters were fairly symmetric about their means. Gain increased with increasing frequency and had lower variance at 0.8 Hz as compared to 0.2 and 0.05 Hz. The phase variance also decreased with increasing frequency. The variances of the offset distributions were somewhat less than the variances of the bias distributions at 0.05 and 0.2 Hz, and greater at 0.8 Hz.

Table 1. VOR Response Parameters for Single Sine Stimuli: Mean, SD*, and Percentile Values on Parameter Distributions

	Gain	Phase	Bias	Offset	Asymmetry
Frequency = 0.05 Hz, *N* = 208					
Mean	0.68	10.5	−0.44	−0.35	−0.9
SD	0.15	4.85	3.56	2.90	7.5
2.5%	0.39	0.94	−8.21	−6.11	−18.8
5%	0.44	2.47	−6.34	−4.84	−15.1
25%	0.58	7.54	−2.55	−2.03	−5.7
50%	0.67	10.27	−0.36	−0.32	−0.5
75%	0.78	13.41	1.89	1.46	4.0
95%	0.96	18.10	5.00	3.84	10.8
97.5%	1.02	19.19	6.73	6.46	13.8
Frequency = 0.2 Hz, *N* = 208					
Mean	0.75	1.62	−0.62	−0.35	−1.5
SD	0.16	3.17	2.91	2.32	6.6
2.5%	0.40	−4.27	−6.58	−5.00	−14.4
5%	0.51	−3.58	−5.70	−3.98	−11.7
25%	0.65	−0.36	−2.46	−1.79	−5.9
50%	0.75	1.96	−0.65	−0.33	−1.5
75%	0.85	3.97	1.32	1.09	3.2
95%	0.99	5.95	4.32	3.47	9.7
97.5%	1.02	6.39	4.88	4.49	10.4
Frequency = 0.8 Hz, *N* = 204					
Mean	0.84	0.79	−0.28	0.04	−1.4
SD	0.13	2.59	2.76	3.13	5.9
2.5%	0.59	−3.95	−6.27	−6.78	−14.6
5%	0.62	−3.07	−5.30	−5.20	−11.9
25%	0.76	−0.59	−1.98	−1.62	−4.0
50%	0.84	0.74	−0.21	−0.10	−1.1
75%	0.93	2.05	1.52	1.88	1.6
95%	1.06	5.38	3.91	5.45	8.5
97.5%	1.07	6.99	5.14	6.50	10.3

Units are VOR phase in degrees, bias in °/s, offset in °/s, and asymmetry in percent.

*SD, standard deviation.

Neither bias, offset, nor asymmetry were highly correlated across the three test frequencies. The largest correlation coefficient was 0.71 between bias at 0.2 and 0.8 Hz. Correlations comparing bias at 0.05 and 0.2 Hz, and at 0.05 and 0.8 Hz were 0.56 and 0.52, respectively. The correlation coefficients comparing response offset at the three test frequencies ranged from 0.43 to 0.57. Asymmetry correlations across test frequencies were the lowest of all symmetry measures (range 0.10–0.37).

There were small changes in rotation test gain and phase responses with age (Figure 2). In particular, all gains decreased with increasing age. The gain trend was more consistent at 0.05 Hz than at 0.2 and 0.8 Hz. Phases increased with increasing age at all frequencies tested, although the effect was more pronounced at 0.2 and 0.8 Hz than at 0.05 Hz. The age-related changes in gain and phase were both roughly linear. Linear regression slopes, intercepts, and correlation coefficients are summarized in Table 2. Rotation test measures of response symmetry (the absolute values of bias, offset, and asymmetry) showed no age-related trends.

VOR Responses to Caloric Stimuli

Caloric test results were generally in agreement with those of others who reported normal ranges of LA of about 15 to 25%. Our results were consistent with a 25% upper limit of normal since 95% of our subjects had LA measures below this value. The distribution of AR had a mean of 17.0°/s (±9.0 SD, range 4.5–63.2) and was skewed toward larger values.

Age-related effects on caloric test results were ambiguous. A linear regression curve fit to AR versus age (Figure 3A) showed an average decrease with increasing age. The linear regression had an associated correlation coefficient of −0.15, which was significantly different from zero ($P < 0.05$). However, the lowess fit shown in Figure 3A indicated that a linear regression was probably not an appropriate description of the data. AR de-

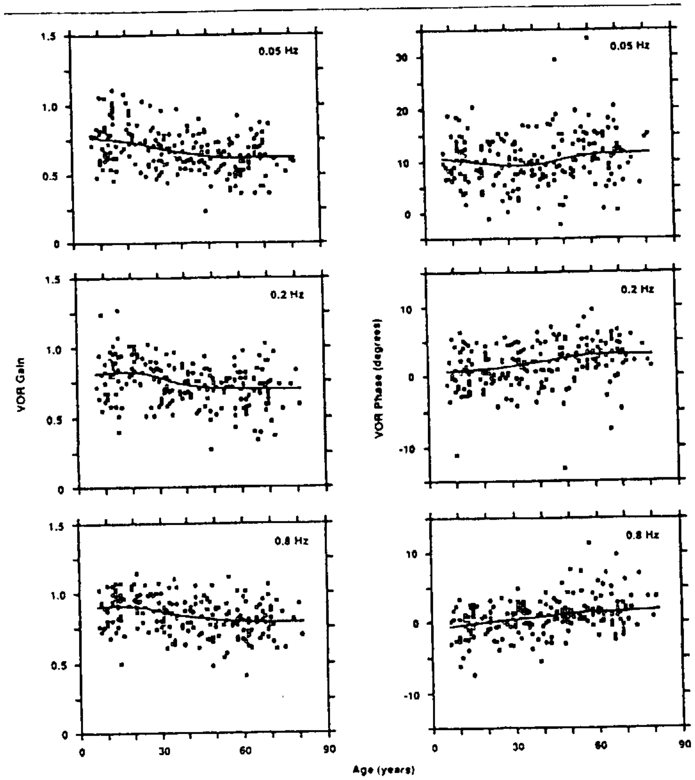

Figure 2. VOR gain and phase as a function of subject age. Data were obtained from sinusoidal rotational stimulation at 0.05, 0.2, and 0.8 Hz. Solid curves are lowess fits.

creased for subjects up to about 40 y, and then increased at a low rate for older subjects.

Figure 3B shows the absolute value of LA versus age. The lowess fit shows essentially no change over the first 6 age decades, and a slight increase in older subjects. Due to the large variance in the data, a much larger sample would be required to determine if the small increase in older subjects was significant.

Correlations among Rotation and Caloric Parameters

Table 3 summarizes correlations among various caloric and rotation test measures. Among rotation test response symmetry measures, bias and offset were highly correlated at all test frequencies. Bias and asymmetry showed moderate correlations (about 0.6) at

Table 2. Age Effects on VOR Rotation Test Gain and Phase Measures

Parameter	Slope (change/yr)	Intercept at 0 years	Correlation coefficient	N
Gain				
@ 0.05 Hz	−0.0030	0.80	−0.39*	208
@ 0.2 Hz	−0.0026	0.85	−0.34*	208
@ 0.8 Hz	−0.0022	0.93	−0.33*	204
Phase				
@ 0.05 Hz	0.029	9.3	0.12	208
@ 0.2 Hz	0.042	0.0	0.27*	208
@ 0.8 Hz	0.050	−1.2	0.39*	204

*Significantly different from zero ($P < 0.05$).

0.05 and 0.2 Hz, but no correlation at 0.8 Hz. There was a small positive correlation between offset and asymmetry at 0.05 Hz, no correlation at 0.2 Hz, and a larger negative correlation at 0.8 Hz.

The correlations between response bias at all test frequencies and caloric DP were about −0.4. Offset and DP showed a similar pattern but with slightly less negative correlations. Asymmetry and DP showed small negative correlations (about −0.2) at 0.05 and 0.2 Hz, but no correlation at 0.8Hz. There was no correlation between LA and DP, or between LA and any of the rotation test symmetry measures.

Caloric AR and rotation test gain and phase measures are known to covary in some vestibular abnormalities (2). Within our putatively normal population, there were only small correlations between these parameters. Correlations between AR and gain at the three test frequencies ranged from 0.22 to 0.31. The correlation between AR and phase was only −0.12 at 0.05 Hz, and less at the other two test frequencies.

Discussion

VOR Parameter Correlations

The pattern of correlation between caloric and rotation test parameters, and among the rotation test parameters themselves apparently depends on rotation test frequency. The asymmetry parameter, which measures the difference in VOR gain for rotations to the right and left, showed the most complex frequency-dependent pattern. For example, the correlations between bias and asymmetry were about 0.6 at 0.05 and 0.2 Hz, but less than 0.1 at 0.8 Hz. The correlation between caloric DP and asymmetry was also higher at 0.05 and 0.2 Hz compared to 0.8 Hz. In addition, there was a shift in the correlations between

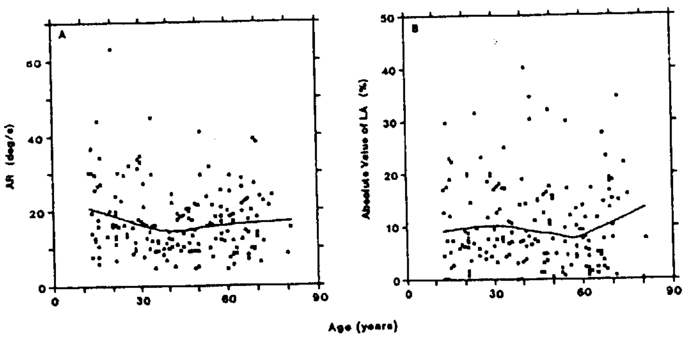

Figure 3. Caloric test AR (A) and LA (B) as a function of subject age. Solid curves are lowess fits.

Table 3. Correlations among Caloric and Rotation Test Response Parameters*

	DP	Bias	Offset	Asymmetry
Frequency = 0.05 Hz				
LA	−0.07	0.08	0.12	−0.01
DP		−0.41	−0.37	−0.25
Bias			0.92	0.58
Offset				0.23
Frequency = 0.2 Hz				
LA	−0.07	0.14	0.13	0.07
DP		−0.39	−0.33	−0.21
Bias			0.83	0.59
Offset				0.06
Frequency = 0.8 Hz				
LA	−0.07	0.13	0.07	0.10
DP		−0.42	−0.41	0.00
Bias			0.85	0.07
Offset				−0.44

*Data are from the 153 subjects who completed all tests with good or fair quality data.

offset and asymmetry from a positive value at 0.05 Hz, to a near zero value at 0.2 Hz, and then to a negative value at 0.8 Hz. Finally, the correlations between asymmetry measures made at different test frequencies were poorer than either the bias or offset parameter correlations across frequency. The poor correlation of asymmetry measures at different test frequencies and the changing relationship of asymmetry to the other caloric and rotation test symmetry measures suggests that the physiological mechanisms which control symmetry are either frequency dependent or that separate physiological factors dominate at different frequencies of head motion.

VOR Changes with Age

We were able to identify small age effects on some VOR response measures. The direction of change of VOR gains was expected. Other age-related changes were not expected. These include increased VOR phase leads with increasing age, and the fact that VOR function measured using the caloric test did not show the same trend as VOR function measured using rotation tests.

Age-related changes in VOR function identified in this study do not follow the same

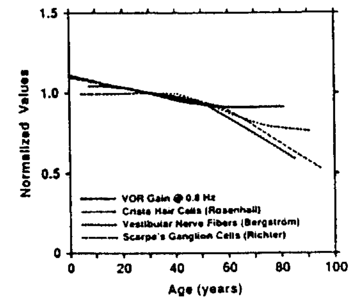

Figure 4. Comparison of age-related changes in VOR gain and peripheral vestibular anatomical data. The 0.8 Hz VOR gain fit is the same as in Figure 2. All curve fits to anatomic data are lowess fits to published data. All fits are plotted on a linear scale normalized to 1.0 at age 30 yr. The normalization factors are 0.87 for 0.8 Hz VOR gain, 6940 crista hair cells, 17450 vestibular nerve fibers, and 18135 Scarpa's ganglion cells.

time course as age-related peripheral vestibular anatomical changes identified by others. Figure 4 shows the lowess curve fit to 0.8 Hz VOR gain versus age plotted along with curve fits to data on human crista hair cell counts (1), vestibular nerve fibers (2), and Scarpa's ganglion cells (3) as a function of age. The ordinate scales are linear and are normalized to their values at a subject age of 30 y. For ages up to about 50 y, there is a gradual decline in both VOR gain and the various measures of peripheral vestibular anatomic components. For the VOR gain this gradual decline continues at about the same rate through the higher age decades. However, the rate of decline of all anatomic measures greatly increases after about age 60, resulting in a divergence between the anatomical and physiological data, with the VOR functioning better in older subjects than would be predicted based on changes in peripheral vestibular anatomy.

Because the subjects of this study were volunteers, it could be argued that the sample of older subjects would be biased in favor of exceptionally healthy elderly individuals who do not reflect the physiological function of a randomly selected population. Several results argue against a strong bias toward exceptionally healthy elderly subjects. First, the auditory pure tone threshold functions of our subjects were consistent with the expected age-related changes (17). Second, extended frequency audiometry (8–20 KHz) was performed on most subjects and showed consistent monotonically declining function with increasing age. Third, the optokinetic reflex time delay showed monotonic increases with age (14). Finally, posture test results (13) in the same subjects showed clear age effects. It seems unlikely that the peripheral vestibular system of these subjects would escape distributed aging processes when other systems did not.

The increases in VOR phase leads at higher frequencies with increasing age were not anticipated. On the surface they would seem to represent a degradation of function since increased phase leads take the system response away from the goal of perfect compensatory eye movements (unity gain and zero phase). Perhaps the phase advance is an artifact of an adaptation that improves overall VOR function. For example, studies of peripheral semicircular canal function in the squirrel monkey have shown that higher gain peripheral nerve fibers have dynamic properties which include phase advances at higher frequencies (7). Phase advances indicate a sensitivity to the velocity of cupula deflection in addition to the cupula position. In contrast, lower gain canal fibers show cupula position sensitivity and therefore, due to the integrating accelerometer characteristics of canal biophysics, the nerve responses are in phase with head velocity at higher frequencies of rotational movements.

On the basis of our results, we might postulate that in young people, low gain tonic canal fibers provide the major contribution to the VOR. As the subject ages and there is a gradual loss of peripheral canal input due to cell death, adaptive mechanisms in the central nervous system may be able to selectively increase the contribution of high gain canal nerve input. The net effect would be to maintain the gain of the VOR at a reasonable level allowing for the generation of adequate compensatory eye movements. However, this mechanism of gain enhancement would be accompanied by the possibly undesirable phase leads associated with the dynamics of the high gain canal fibers. A trade-off may be occurring in favor of maintaining the desirable feature of high response amplitude at the expense of the timing of compensatory eye movements.

This hypothesis may be consistent with the multichannel model of the VOR (9) developed to explain the dynamic properties of VOR adaptation. However, assuming that human and monkey VOR adaptation occurs by similar mechanisms, there are other studies that are not consistent with this hypothesis. Minor and Goldberg (10) have shown that phasic canal fibers do not appear to contribute at all to the VOR of the squirrel monkey. If phasic fibers do not contribute to the VOR, they cannot participate in alterations in VOR dynamics. One might argue that these phasic fibers only contribute to the VOR when they are needed for the adaptive enhancement of the reflex. However, this would be inconsistent with other results that suggest ~~that~~ it is an enhancement of the contribution of the tonic fibers that mediates adaptive increases in VOR gain (8). ← delete "that"

Finally, a hypothesis calling for an increased phasic fiber contribution to the VOR of older subjects may also be inconsistent with anatomical aging results that showed relatively greater hair cell loss on the crest of the crista (16), and the greatest losses of the thick fibers innervating the canal cristas (4). Since the higher gain afferent fibers, at least in the chinchilla, tend to be larger in diameter and to originate from the crest of the crista (1), the selective loss of these cells with increasing age would preclude their participation in VOR gain enhancement.

Current understanding of the mechanisms of VOR adaptation and of anatomical changes in peripheral vestibular receptors does not provide a good explanation of our VOR data. Studies of VOR adaptation, often performed in younger animals, may not adequately characterize changes that occur with

age since the aging process may also affect the functionality of the central neural networks involved in adaptation. An aging adaptive neural network could contribute its own dynamic component to the VOR that differs from the dynamic properties observed in younger animals.

Clinical Significance

The presence of age-related changes in VOR function identified using rotation tests has implications for the assessment of normal function. Part of the variability of VOR response parameters is caused by this age effect. The square of the correlation coefficient gives an estimate of the proportion of variance related to changes with age. The VOR gain versus age measures had correlation coefficients between 0.3 and 0.4. Therefore, approximately 10% to 15% of the variance of gain data is accounted for by the effect of age. Measures of normal vestibular function should account for these age effects.

Since the majority of the observed response variability is independent of age, it is clear that the functional characteristics vary widely within any given age group in a putatively normal population. To the extent that aging effects are deleterious and that our reflex measures accurately characterize the general decline in function, a significant proportion of subjects within any age group look "older" than their chronological ages and may be less functional with regard to their orientation control abilities. One could hypothesize that these subjects would be more susceptible to the development of balance and orientation control problems as their vestibular function further declines with age. Perhaps there is some threshold beyond which the brain's adaptive mechanisms are not able to compensate for the declining function. After this point is reached, subjects may develop dizziness and equilibrium control complaints, or perhaps individuals will restrict their activities so as to avoid situations that stress their remaining capabilities. A longitudinal rather than a cross-sectional study would be required to test this hypothesis.

Although we did not observe large age-related trends in VOR function, it is apparent that central adaptive mechanisms cannot sustain VOR function indefinitely in the face of increasing losses of peripheral receptors and neural substrate. It will be important to extend the age limit of our study to the eighth and ninth decades, and to explore larger amplitude and higher frequency stimuli that more nearly resemble natural head motion. The point at which physiological function begins to follow the anatomical decline will define the effective functional reserve of the central adaptive mechanisms.

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AGE-RELATED CHANGES IN HUMAN VESTIBULO-OCULAR AND OPTOKINETIC REFLEXES: PSEUDORANDOM ROTATION TESTS

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☐**Abstract – The dynamic response properties of horizontal vestibulo-ocular reflex (VOR) and optokinetic reflex (OKR) were characterized in 216 human subjects ranging in age from 7 to 81 y. The object of this cross-sectional study was to determine the effects of aging on VOR and OKR reflex dynamics, and to identify the distributions of parameters that describe VOR and OKR responses to pseudorandom stimuli in a putatively normal population. In general, VOR and OKR response parameters changed in a manner consistent with declining function with increasing age. For the VOR this was reflected in declining response amplitudes, although the magnitude of the decline was small relative to the variability of the data. For the OKR the lag time of the response, probably associated with the time required for visual information processing, increased linearly with age at a rate of about 1 ms per year.**

☐**Keywords – vestibular; visual-vestibular; eye movements.**

Introduction

A great deal is known about age-related performance declines in various visual perception tasks (7). These tasks are usually studied in controlled settings in which oculomotor performance is assumed not to play an important role. However, in natural settings it is obvious that changes in oculomotor performance with age could have a deleterious effect on many aspects of visual perception. In particular, vestibulo-ocular reflex (VOR) and optokinetic reflex (OKR) normally function together to provide clear vision by generating compensatory eye movements that minimize image motion on the surface of the retina during head movements. It is important to know how the VOR and OKR change with age since a degradation in these reflexes could impair the acquisition of visual information during active and passive head movements.

During horizontal head rotations in the light with earth-fixed visual surrounds, the VOR and visual motion information through optokinetic and pursuit tracking systems combine to produce compensatory eye movements that facilitate clear vision by maintaining a fixed gaze direction. The combined VOR and visual tracking reflexes are effective over a bandwidth from DC to several Hertz (9).

Literature on age-related changes in vestibular and oculomotor function is limited. Experiments on humans are usually performed on a small number of subjects, and these subjects are typically young adults. The exceptions to this occur in the clinical literature on caloric testing of the VOR (3), in some work on age-related changes in pursuit tracking ability (16,17), and in responses to constant velocity optokinetic stimuli (10,18).

The present experiments were designed to characterize horizontal VOR and OKR dynamic response properties, and the variability of those responses in normal humans. The population was selected to provide results related to the effects of the aging process on these reflexes. Tests of VOR function using caloric and sinusoidal rotational stimuli, and tests of posture control were also made in the same subjects and are reported in companion papers (12–14).

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Methods

Vestibular and oculomotor reflexes were tested in 216 human subjects (90 male and 126 female) aged 7 to 81 y. Ages were approximately uniformly distributed over the entire range. Details of subject selection are given in a previous paper (13). We did not reject subjects based on the results of any of the vestibular, optokinetic, or posture tests performed.

Rotation Tests

Test conditions for the VOR were identical to those described earlier (13). Additionally, for the OKR the subject was surrounded by a circular cloth cylinder 1.8 m in diameter. The cylinder acted as a projection screen for an optokinetic stimulus. A full field optokinetic stimulus was provided by a pin hole type projector mounted on a 6.8 N·m servo motor (~~Genisco Technology Corp, Model 1100~~) attached to the ceiling directly above the subject's head. The projector produced randomly placed vertical stripes of light against a mostly dark background.

(use update to: Genisco Technology Corp, Rancho Dominguez, CA, Model 1100-5)

Rotational stimuli for VOR tests included both single frequency sinusoidal stimuli [results reported in (13)] and a pseudorandom stimulus. The pseudorandom stimulus consisted of the summation of eight discrete sinusoidal frequencies. The frequency components were selected to minimize corruption of the results of the data analysis due to possible nonlinear responses of the VOR system (19). The eight frequencies were 0.0092, 0.021, 0.046, 0.095, 0.180, 0.388, 0.766, and 1.535 Hz. The nominal amplitudes of these components were all 15.6°/s except the highest frequency component which was 7.8°/s. The highest instantaneous stimulus velocity was about 100°/s. The duration of the stimulus was about 440 s. Transient responses were avoided by recording only the final 327.68 s of the trial.

The OKR of each subject was tested by recording horizontal eye movements evoked by a projected visual stimulus rotated around the stationary subject. The optokinetic projector moved under the control of a pseudorandom stimulus consisting of seven sinusoids with frequencies 0.018, 0.043, 0.092, 0.189, 0.360, 0.775, and 1.532 Hz. The amplitudes of the components were nominally 7.8°/s except the highest frequency component which was 3.9°/s. The peak instantaneous velocity was about 40°/s. The total stimulus duration was about 220 s. Transient responses were avoided by recording only the final 163.84 s of data. Complete OKR data were not obtained on all subjects since the stimulus induced motion sickness symptoms in some subjects, requiring the early termination of the test.

Subjects were given verbal tasks throughout the VOR and OKR rotation tests to maintain a constant level of alertness. The tasks consisted of alphabetically naming such things as names, places, or foods.

Data Analysis

Eye position data were differentiated to calculate eye velocity. Fast phases of the nystagmus were identified and eliminated (11). The Fourier analysis of the remaining slow phase eye movements provided estimates of the response parameters given by the following equation

$$r(t) = B + \sum_{i=1}^{N} A_i \sin(2\pi f_i + P_i) \qquad [1]$$

where B is bias or average slow phase velocity, with units of °/s, N is the number of sinusoidal components in the pseudorandom stimulus, A_i is the response amplitude at the ith frequency f_i, and P_i is the response phase at the ith frequency. A Fourier analysis of the stimulus velocity was performed to calculate the amplitudes and phases of the stimulus waveform. The reflex gains and phases at the N stimulus frequencies were computed as the ratio of response amplitude to stimulus amplitude, and the difference between response phase and stimulus phase, respectively. For the convenience of working with smaller numbers, a value of 180° was added to the calculated phases for tests of the VOR reflex so that unity gain and 0° phase represented perfectly compensatory eye movements.

The algorithms for gain and phase calculations were verified by simulating reflex responses with known electronic circuits. Before analysis, segments of the simulated responses were eliminated to simulate the patterns of missing data caused by the elimination of nystagmus fast phases from real eye movement data.

The gain and phase values of the VOR reflex were fitted with a transfer function equation of the following form

$$H_{vor}(s) = \frac{K_v T_v s}{T_v s + 1} \quad [2]$$

where T_v is an estimate of the VOR time constant (units of seconds), K_v is the VOR gain constant, and s is the Laplace transform variable.

OKR gain and phase data for all subjects were well fit by a three parameter transfer function of the form

$$H_{okr}(s) = \frac{K_o \exp(-T_d s)}{T_o s + 1} \quad [3]$$

where T_o is a time constant with units of seconds, K_o is the OKR gain constant relating slow phase eye velocity to stimulus velocity, and T_d is a time delay parameter with units of seconds describing the lag between visual field movement and eye movement. The $T_o s + 1$ factor represents a lowpass filter which accounts for the declining gain with increasing frequency observed in some subjects. Larger values of T_o are consistent with gain declines beginning at lower frequencies. A value of zero for T_o (ie) the transfer function reduces to $H_{okr}(s) = K_o \exp(-T_d s)$] accounts for subjects whose gain did not decline with increasing frequency. T_o is not the time constant associated with velocity storage and optokinetic afternystagmus (5).

Visualization of Trends

In order to visualize trends in scatterplots, a robust locally weighted regression analysis (lowess fit) was used to smooth the scatterplots (4). This smoothing is similar to a moving average filter, but is less sensitive to outlying points and allows variable amounts of smoothing. A lowess smoothing parameter of 0.5 and iteration parameter of 2 were used on all data sets.

Data Quality

The overall quality of each rotation test for each subject was subjectively given a rating of good, fair, or poor. Only good and fair quality data are included in the data summaries in the results section. Quality judgments were based on the consistency of the responses throughout the duration of the stimulus, and on the accuracy of the eye movement analysis in the separation of slow and fast phases of nystagmus. The actual values of response parameters were not used in judgment of data quality. The test results from about 4% of subjects were rated poor for each test. Poor quality data for one subject on a given test was not used to disqualify ~~other~~ data from the same subject on other tests.

Results

The subjects showed a wide range of responses on all measures of VOR and OKR function. Age-related changes were identified in almost all rotation test response measures, but the magnitude of these changes was not large relative to the variability of the data. Most changes indicated a decline in function. There were no significant differences in reflexes between males and females.

VOR *Responses*

A sample of a typical response to a pseudorandom VOR stimulus is shown in Figure 1. The pseudorandom stimulus evokes a complex eye movement pattern (Figure 1C). However, separation of slow and fast components, and calculation of slow phase eye velocity reveals the underlying compensatory motion (Figure 1B). A spectral analysis of slow phase eye velocity and the recorded stimulus velocity provides measures of response gain and phase as a function of stimulus frequency. Examples of gain and phase data from three subjects are shown in Figure 2. Typically the gain is lower at the lowest test frequency and increases with increasing frequency. In some subjects the gain monotonically increases over the frequency range tested and in others it appears to reach an asymptote. The solid lines through the data points represent curve fits of a two parameter transfer function (eqn 2) to the data of each subject.

The pattern of VOR gain and phase data of most subjects was similar in form to the data in Figure 2A, and was well characterized by the two parameter transfer function model (eqn 2). There were deviations from this pattern which are exemplified by the data from two other subjects in Figures 2B and 2C. The low frequency data in 2B were fit well by the two parameter model but the higher frequency data showed increasing phase leads with increasing frequency. The phases of VOR responses to 0.05, 0.2, and 0.8 Hz sinusoidal rotations were 10.7°, 2.0°, and 6.9°, respectively, for this subject, and therefore confirmed the general pattern. A more accurate curve fit to this data would require a higher frequency lead term in the transfer function. A transfer function of this form would be similar to the one used to describe the dynamic responses of phasic canal afferents in the squirrel monkey (6).

The VOR phase data in Figure 2C were fairly flat and greater than zero across all test frequencies. The phase of responses to sinusoidal stimuli were 3.7°, −2.8°, and 2.1° at 0.05, 0.2, and 0.8 Hz, again confirming the general pattern but with less phase lead than the pseudorandom data. The curve fit identified a long VOR time constant of 44.9 s. However, the two parameter model does not

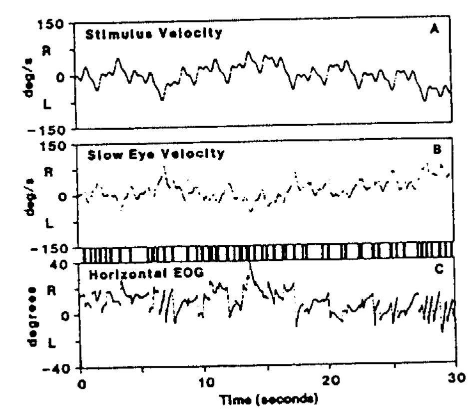

Figure 1. A 30 s sample of eye movements evoked by pseudorandom stimulation of the VOR. (A) shows the subject's rotational velocity, (B) slow phase eye velocity, and (C) horizontal EOG. Vertical bars between (B) and (C) show the locations of fast phases of the nystagmus detected by the computer analysis of the data.

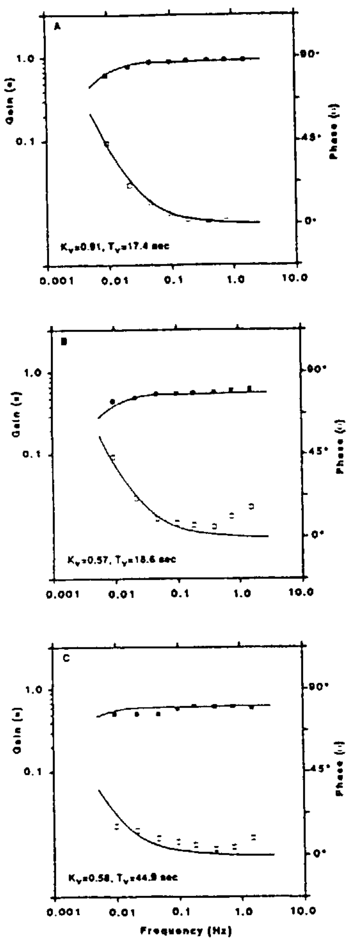

Figure 2. Examples of VOR gain and phase data from three subjects derived from responses to a pseudorandom rotation. Solid line through the data show the transfer function curve fit (eqn 2). Gain and frequency scales are logarithmic.

not consistent w/p. 4 style.

describe either the low or high frequency phase data well. A transfer function fit with an s^k Laplace operator is better able to describe this type of data (1).

VOR gain and phase data from pseudorandom stimulus test results are summarized in Table 1. As with single frequency sinusoidal results, the variance of gain data was similar across all frequencies. The variance of phases data was larger at low frequencies than at high, probably reflecting the variance of the VOR time constant among individuals. The variance of the phase data at 1.535 Hz was larger than the variance at adjacent lower frequencies. This probably resulted from the fact that the stimulus amplitude of the highest frequency component was half that of the lower frequencies resulting in a lower signal-to-noise ratio at the 1.535 Hz test frequency.

The distributions of the VOR gain constants and time constants for the pseudorandom stimulus are summarized in Table 2. The gain constant distribution was symmetric with an average value of 0.72. The VOR time constant distribution was skewed toward longer time constants. Mean value was 24.4 s and median value was 23.0 s. Only two subjects had time constants below 10 s. One of these subjects (time constant = 8.2 s) had a partial unilateral loss of vestibular function as judged by caloric testing. A short time constant is consistent with a unilateral loss of vestibular function (8). However, the other subject (time constant = 7.0 s) had normal caloric test results.

Table 1. VOR Gain and Phase (mean ± 1 SD*) Derived from Responses to Pseudorandom Stimulation

Please add footnote for "SD" for consistency with posture papers.

Frequency (Hz)	Gain	Phase (degrees)
0.0092	0.58 ± 0.14	34.0 ± 8.6
0.021	0.64 ± 0.15	19.4 ± 5.6
0.046	0.69 ± 0.15	11.4 ± 3.8
0.095	0.71 ± 0.16	6.6 ± 2.9
0.180	0.73 ± 0.16	5.1 ± 2.7
0.388	0.75 ± 0.16	3.8 ± 2.7
0.766	0.75 ± 0.16	4.1 ± 3.5
1.535	0.74 ± 0.17	4.4 ± 6.2

*SD standard deviation.

Table 2. VOR Response Parameters for Pseudorandom Stimulus: Mean, SD, and Percentile Values from Parameter Distributions (N = 207)

	Gain constant	Time constant (s)	Bias (°/s)
Mean	0.72	24.5	−0.1
SD	0.16	8.6	2.1
2.5%	0.42	13.6	−5.0
5%	0.48	14.2	−3.8
25%	0.61	18.4	−1.3
50%	0.73	23.1	−0.0
75%	0.81	28.1	1.3
95%	0.97	43.6	3.2
97.5%	1.02	47.4	3.7

Please add space as is on all the tables like this one. (p. 8)

Comparison of VOR Measured by Single Frequency and Pseudorandom Stimuli

If the VOR were a linear system, then gain and phase data obtained from single frequency sinusoidal and pseudorandom stimulation should be identical within the random variability introduced by measurement errors. Statistical comparisons were made between the single frequency (13) and pseudorandom gains and phases and are shown in Table 3. Single sine and pseudorandom results were not significantly different at the lowest frequency (0.05 Hz). Small but consistent differences were evident at higher frequencies. In particular, the single frequency gain was higher at 0.8 Hz than the pseudorandom derived gain, and the pseudorandom phase values at 0.2 and 0.8 Hz were phase advanced by about 3° compared to single frequency sine results. The improved phase response from sine tests may represent a small predictive effect; however, this effect did not apparently carry over to the 0.05 Hz data.

The gain value from the 0.8 Hz test was higher than the pseudorandom test result. This might also be due to a predictive effect. However, in this case it seems likely that the data analysis methods could have contributed to this higher value. During the analysis of the single frequency sine tests, the experimenter had the ability to reject data from stimulus cycles that were obviously corrupted. These corrupted data cycles could easily be identified based on gain, phase, and/or bias values that deviated greatly from the values for other cycles. There were several causes for poor data cycles including transient EMG interference, excessive blinking, gaze deviations from the horizontal plane, inattentiveness to tasking, and failure of the fast phase eye movement detection algorithm. With experience, it became a simple task to detect and correct these problems by rejecting the affected cycles. The net effect usually increased the average gain measure. Eye movement recordings were also transiently corrupted during pseudorandom testing, but we did not have a means of correcting or eliminating these problems and they were therefore aver-

Table 3. Comparison of Single Sine and Pseudorandom Gain and Phase Results*

Parameter	Single sine (Hz)	Pseudorandom (Hz)	Average difference	$N^†$	Significance
Gain	0.05	0.046	−0.011	199	–
	0.2	0.180	0.019	199	‡
	0.8	0.766	0.095	195	‡
Phase	0.05	0.046	−0.3	199	–
	0.2	0.180	−3.2	199	‡
	0.8	0.766	−3.4	195	‡

*Positive differences indicate that the average single sine parameter value was larger than the average pseudorandom parameter value. The average differences listed are corrected for the difference in test frequencies between the single sine and pseudorandom stimuli. Gain and phase corrections were based on the VOR transfer function in eqn 2 with average time constant of 24.5 s and gain constant of 0.72.
†Ns are smaller than those in Table 2 because comparisons were not made if either test had poor quality data.
‡Significant difference (P < 0.05, paired t test).

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aged into the final result. The rejection of corrupted portions of single sine results, but not pseudorandom results, could account for the higher gains measured during sinusoidal rotations.

OKR *Responses*

Typical OKR test results from pseudorandom stimulation for two subjects are shown in Figure 3. Response gain was less than unity in all subjects. The gains of most subjects were approximately flat across the bandwidth of frequencies tested (0.02 to 1.5 Hz) as in Figure 3A. Phases were near 0° at the lowest frequencies and showed monotonic increasing phase lags as frequency increased. Since perfect tracking of the visual stimulus is represented by unity gain and zero phase at all frequencies, subjects demonstrated imperfect tracking in terms of both amplitude (gain) and timing (phase). The major variation on the typical OKR result was the presence of declining gain with increasing frequency in some subjects. Figure 3B shows the OKR transfer function data from one such subject.

The means, standard deviations, and ranges of OKR gain constant, time constant, time delay, and bias are given in Table 4. OKR response bias was near zero for all subjects. Both the gain constant and time delay had approximately symmetric distributions. In contrast, the OKR time constant had a highly skewed and possibly bimodal distribution with about 40% of the values near zero. OKR time constants near zero reflect the fact that OKR gains for these subjects were approximately constant over the frequency range tested.

The OKR pseudorandom stimulus was quite provocative in the initiation of motion sickness symptoms. Twenty subjects requested the termination of testing as a result of the onset of motion sickness symptoms. Approximately an equal number experienced motion sickness symptoms but were able to complete the 220 s duration OKR stimulus. It was not possible to calculate OKR gains and phases from incomplete trials using our current analysis methods. Therefore, it was not possible to test the hypothesis that abnormal OKR responses were related to motion sickness sensitivity in these highly susceptible subjects. However, OKR gains and phases from subjects who reported the onset of motion sickness symptoms but were able to complete the test did not show any obvious differences compared with subjects who did not report symptoms. Also comparisons of VOR rotation test results of OKR motion sickness susceptible subjects with nonsusceptible subjects did not reveal any differences.

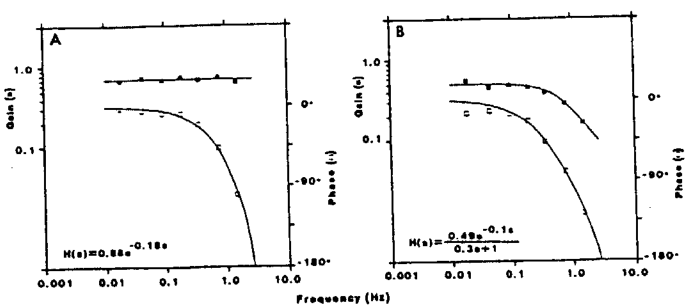

Figure 3. Examples of OKR gain and phase data from two individuals derived from responses to pseudorandom optokinetic stimulation; Solid lines show transfer function curve fits to data. Equations of the curve fits are inset. Gain and frequency scales are logarithmic.

Need to add a period.

Table 4. OKR Response Parameters for Pseudorandom Stimulus: Mean, SD, and Percentile Values from Parameter Distributions* (N = 179 subjects)

	Gain constant	Time constant (s)	Time delay (s)	Bias (°/s)
Mean	0.65	0.080	0.180	−0.1
SD	0.12	0.080	0.043	0.8
2.5%	0.40	0.002	0.099	−1.7
5%	0.47	0.003	0.114	−1.5
25%	0.59	0.008	0.147	−0.7
50%	0.66	0.06	0.187	−0.1
75%	0.72	0.12	0.216	0.3
95%	0.81	0.23	0.248	1.1
97.5%	0.87	0.25	0.253	1.4

Age-Related Changes in VOR *and* OKR

Several VOR and OKR response parameters changed with age (Figures 4 and 5), while the absolute values of VOR and OKR bias did not. Many of the age-related changes showed roughly linear trends. Linear regression slope, intercept, and correlation coefficients are summarized in Table 5. Both VOR time constant (Figure 4B) and OKR gain constant (Figure 5A) increased slightly in subjects up to about 30 y and then decreased with increasing age. The OKR time delay parameter increased with increasing age and showed the clearest age-related trend (r = 0.53 and slope = 1.2 ms/y) of all VOR and OKR parameters.

The age-related change in the OKR time constant was clearly not linear. Data in Figure 5B show that a large proportion of subjects between about 20 and 60 y had OKR time constants close to zero, indicating that their OKR gains were constant across frequency. In contrast, there were very few subjects under 20 y of age and proportionally fewer subjects over 60 who had zero OKR time constants, indicating that on average their OKR gains declined with increasing frequency. The lowess curve fit indicates that age-related trends were minimal for subjects between 20 and 60 y. Subjects under 20 showed an age-related decline in their OKR time constant with increasing age. Subjects over 60 showed an age-related increase in their OKR time constant with increasing age.

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Discussion

Pseudorandom Testing

There are both advantages and disadvantages to the use of pseudorandom stimuli for VOR and OKR testing. An advantage is the concurrent testing of response dynamics over a large frequency bandwidth. The total test

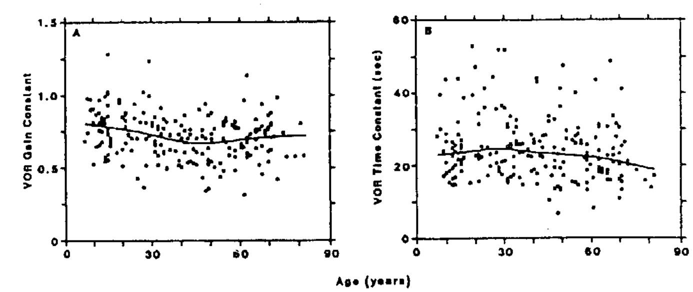

Figure 4. VOR gain constant (A) and time constant (B) parameters as a function of subject age. Parameter values were estimated from transfer function curve fits to gain and phase data obtained from pseudorandom rotation test results. Solid curves are lowess fits.

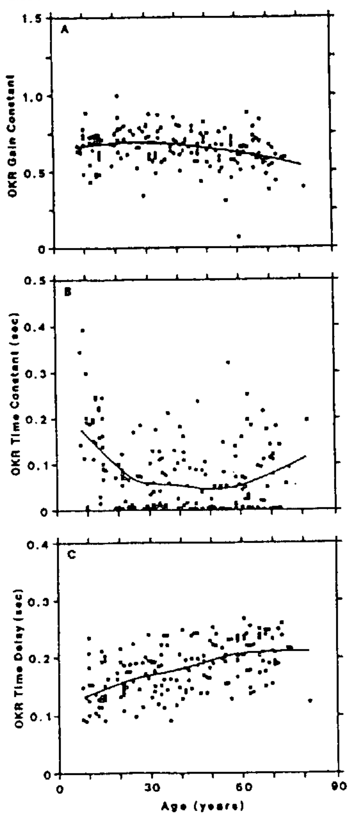

Figure 5. OKR gain constant (A), time constant (B), and time delay (C) parameters as a function of subject age. Parameter values were estimated from transfer function curve fits to OKR gain and phase data obtained from pseudorandom rotation test results. Solid curves are lowess fits.

time is reduced compared to the equivalent testing using single frequency sinusoidal stimuli. In addition, all gain and phase measures obtained from pseudorandom responses are obtained at the same average level of subject arousal. Therefore a subject's reflex gains, which depend on subject alertness, can be compared across test frequencies without worry of possible alertness changes as in sequential testing at individual frequencies. A wide bandwidth stimulus can therefore reveal patterns of gain and phase changes over a range of test frequencies that are not as evident or as reliable when single frequency sinusoids are used. These detailed patterns of gain and phase changes with frequency may contain more clinically useful information than the simple summaries of response dynamics in terms of derived gain constants and time constants. In addition, a pseudorandom stimulus may provide a more "realistic" or "natural" stimulus than more predictable ones, and therefore test the dynamics of the components under conditions similar to everyday conditions.

Pseudorandom testing has the disadvantage that the complexity of the eye movement responses makes it difficult to judge the quality of the responses. The failure to maintain subject alertness throughout the test coupled with the failure to recognize the lack of alertness would bias the gain and phase measures of reflex dynamics.

The similarity between pseudorandom and sinusoidal VOR test results suggests that information from the two test techniques is relatively interchangeable. The most significant difference between VOR responses was the small 3° phase lead of pseudorandom results relative to sines at higher stimulus frequencies. However, others have shown large differences in OKR dynamics when gain and phase data derived from single frequency sinusoidal tests are compared to pseudorandom results (20). In particular, OKR gains from sinusoidal tests are generally higher and show less phase lags at any particular test frequency than pseudorandom results. The simplest interpretation of these differences is that prediction plays an important role in visually driven eye movements. The mechanisms involved in this prediction are poorly understood, but it is known that the brain's predictive mechanisms influence the overall dynamic responses of pursuit eye movements even when pseudorandom visual stimuli are used (2). In particular, the predictive mechanism apparently favors the highest frequency component of the stimulus at the expense of lower frequency components causing the gains of lower frequency components of pursuit eye movements to be depressed while the gain of the highest frequency is enhanced. Although there are differences between pursuit and optokinetic response dynamics to pseudorandom stimulation (20), similar mechanisms to those which occur in pursuit tracking may influence optokinetic responses to full field visual stimuli.

The presence of a predictive mechanism as well as other nonlinear properties of visually-guided eye movements does not necessarily disqualify the use of pseudorandom stimuli for OKR testing. Rather, pseudorandom stimuli may prove to be useful in quantifying the functionality of the predictive mechanism which is likely an important contributor to the control of gaze. Perhaps the larger proportion of young and old subjects with declining OKR gains at higher stimulus frequencies (larger T_o's) may indicate a less functional predictive mechanism for visually-guided eye movements in these age groups.

Table 5. Age Effects on VOR and OKR Response Parameters*

Parameter	Slope (change/yr)	Intercept at 0 years	Correlation coefficient†	*N*
VOR gain constant	−0.0019	0.79	−0.24	207
VOR time constant (s)	−0.62	27.0	−0.15	207
OKR gain constant	−0.0015	0.71	−0.26	179
OKR time delay (s)	0.0012	0.13	0.53	179

* All parameter values that showed significant or nearly significant linear trends with age are listed.
† All correlation coefficients were significantly different from zero ($P < 0.05$).

VOR *and* OKR *Changes with Age*

We were able to identify small age effects on VOR and larger age effects on OKR reflexes. The direction of change of some reflex parameters were expected, such as declining VOR gains, declining OKR gains, and increased time delays in the OKR with increasing age. However, the decreased high frequency OKR gain of the youngest and oldest subjects as compared to middle-aged subjects was not expected.

The rate at which the OKR time delay increased with age was quite large and is similar to the changes found in pursuit latency with increasing age (16,17). If this increased time delay is representative of general changes in the speed of visual system motion processing associated with visuomotor tasks, this could affect tasks such as posture control, which use vision for feedback control. Longer feedback time delays generally contribute to decreased stability and poorer overall performance.

The interpretation of age-related changes in OKR time delay is complicated by the fact that the OKR time constant and time delay parameters are probably not independent. This is because the lag term $T_o s + 1$ in eqn 3 which accounts for the declining gain at higher stimulus frequencies also accounts for some of the phase lag. The larger the OKR time constant, T_o, the more phase accounted for by the lag term, and therefore the smaller the value of the OKR time delay parameter, T_d, required to explain the remaining phase lag. Since the youngest subjects had the largest T_o's this would tend to bias their T_d's toward lower values. The oldest subjects also tended to have larger T_o's which should also bias their T_d's toward lower values. However, Figure 5C shows that, despite this possible bias toward lower values, older subjects still had the largest T_d's in the population. Therefore, although the exact time course of the age-related change in time delay in Figure 5C may be distorted by the interaction with T_o, it is apparent that the oldest subjects did have the largest response delays.

Both younger (<15 y) and older (>65 y) subjects were relatively less responsive to the higher frequency components of the stimulus. The lower OKR responsiveness at higher frequencies could have functional consequences, particularly for older subjects. While it is generally appreciated that visual tracking reflexes improve visual-vestibular generated compensatory eye movements during low frequency head movements, visual motion information is apparently used to improve the dynamics of compensatory eye movements at higher stimulus frequencies associated with natural head movements (15). This would be particularly important for individuals who had VOR phase leads at higher frequencies (Figure 2B). Since older individuals had larger VOR phase leads on average than younger subjects (13), we might expect that the older subjects would need more help from their visual tracking reflexes to correct their imperfect VOR dynamics. However, the sensitivity to optokinetic motion at higher frequencies declined in many older subjects making it less likely that visual tracking reflexes could correct for imperfect VOR dynamics.

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AGE-RELATED CHANGES IN HUMAN POSTURE CONTROL: MOTOR COORDINATION TESTS

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□ Abstract – Postural responses to support surface displacements were measured in 214 normal human subjects ranging in age from 7 to 81 y. Motor tests measured leg muscle electromyograph (EMG) latencies, body sway, and the amplitude and timing of changes in center of pressure displacements in response to sudden forward and backward horizontal translations of the support surface upon which the subjects stood. There were small increases in both EMG latencies and the time to reach the peak amplitude of center of pressure responses with increasing age. The amplitude of center of pressure responses showed no change with age if the amplitude measures were normalized by a factor related to subject height. In general, postural responses to sudden translations showed minimal changes with age, and all age-related trends that were identified were small relative to the variability within the population.

□ Keywords – posturography, EMG, coordination, equilibrium.

Introduction

If the posture of a quietly standing individual is suddenly perturbed by the application of an external force, rapid automatic responses are initiated to maintain postural equilibrium (2). These postural responses produce compensatory muscle contractions beginning about 100 ms following the start of the perturbation. Experimental tests of postural motor coordination typically measure postural reactions to short duration translations or rotations of the support surface under the subject (1,4,7,10). A consistent finding has been a coordinated synergy in which muscle contraction proceeds from distal to proximal leg and trunk muscles following a support surface perturbation.

Various factors associated with the visual, vestibular, and somatosensory systems have been shown to influence or modulate these responses. These factors include support surface condition (8), initial body position (4,9), stimulus velocity and displacement amplitudes (6), galvanic stimulation to the inner ear (12), and availability of visual (11) and proprioceptive cues (5).

The complexity of maintaining upright stance suggests that there is a great deal of functional variability within a normal population as a result of variations in sensory system, central nervous system, and biomechanical function in individuals. Systematic changes may also occur as a result of childhood development and degeneration associated with aging. In order to define the range of normal function, and to identify the nature of any age-related changes in postural motor coordination, we tested a putatively normal population with a wide age distribution.

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Methods

Posture coordination function was tested in 214 human subjects (90 male and 124 female) aged 7 to 81 y. Ages were approximately uniformly distributed over the entire range. Rotation tests of horizontal vestibulo-ocular and optokinetic reflex function, caloric tests, and sensory interaction tests of postural control were measured in these same subjects on the same day, and are reported in companion papers (13–15). Details of subject selection are given in a previous paper (15). Subjects were not excluded from the population based on any vestibular, optokinetic, or posture test results.

Subjects stood on a movable support surface surrounded in front and on two sides by a visual surround that was stationary during motor tests. The visual surround was a box with randomly placed 2 cm black dots on a flat white surface. The average spacing between the dots was ~20 cm, and the distance from the subject to the box was ~50 cm. Support surface motion was controlled by a hydraulic position servo system that could produce forward and backward translations and toe up and toe down rotations. The subject's ankle joints were aligned with the rotation axis located 7.4 cm above the center of the support surface. Force transducers in the support surface recorded vertical forces applied by each of the subject's legs. The anterior-posterior (AP) sway angle (θ_{ap}) of each subject was recorded using a rod attached to a potentiometer. The potentiometer was mounted on a post next to the subject. The end of the rod rested in a V-shaped holder centered on the subject's back at hip level. A voltage proportional to the rotation of the potentiometer was recorded and later transformed using appropriate trigonometric conversions to θ_{ap}.

Tests consisted of five each of forward platform translations, toe down rotations, backward translations, and toe up rotations of the subject's support surface. The subject stood with eyes open viewing the stationary visual surround. Only responses to translations are reported in this paper. Ramp translations were 3 cm in 0.25 s. The support surface returned slowly to the center position following each motion and there was a variable delay averaging 4 s between stimuli. Four EMG's were recorded from the left leg using surface electrodes over the gastrocnemius (G), tibialis anterior (T), hamstring (H), and quadriceps (Q) muscles. EMG's were rectified, low pass filtered at 20 Hz, and sampled at 500 Hz. The latency to the onset of the reflex EMG bursts were estimated from average EMG traces. Latencies were recorded from averaged traces only if the EMG onset times could unambiguously be separated from background activity. Consequently the number of subjects contributing to the data sets in various figures and tables varies.

AP displacement of each subject's center of pressure (CP with units of cm) was calculated for each leg by the following formula:

$$CP = \frac{L(F_f - F_b)}{F_f + F_b} \qquad [1]$$

where L, the length from the ankle joint to the front and to the back force transducers in the platform, was equal to 27.3 cm, and F_f and F_b were the vertical forces recorded by the front and back force transducers during the trial. The center of pressure velocity (CPV in cm/s) was computed from CP by a two point central difference formula. The CP and CPV traces from five trials were averaged, and various peak amplitude and time parameters were measured for each subject (Figure 1). All EMG, CP, CPV, and θ_{ap} times were referenced to the start of platform motion as determined by the earliest deviation of the average CP trace from its baseline.

In order to visualize trends in various scatterplots, a robust locally weighted regression analysis (lowess fit) was used to smooth the scatterplots (3). Lowess smoothing is similar to a moving average but is less influenced by values far from the central tendency of the data. The degree of smoothing is specified by a smoothing parameter (f) between 0 and 1. Larger f values give more smoothing.

When a lowess fit indicated an approximately linear trend, a linear regression analysis was performed along with the calculation of a linear correlation coefficient.

Results

General Response Pattern

Figure 1 shows typical EMG, CP, CPV, and θ_{ap} response patterns for one subject during a 3 cm backward translation. The backward support surface translation results in forward body sway with respect to the platform. In the first 100 ms, the CP movement away from baseline is probably the result of passive properties of body biomechanics combined with artifacts of the platform force recording system. Approximately 110 ms following the start of the translation, the distal leg muscles (gastrocnemius) that oppose the forward body sway begin to contract as evidenced by EMG recordings. The proximal leg muscles (hamstrings) begin to contract about 20–30 ms following the distal muscles. The dorsal leg muscle contractions generate torque about the ankle joint that causes a forward displacement of CP. The onset of the active torque generation (CP_o) begins ~130 ms after the start of platform translation. CP reaches a peak displacement amplitude (CP_a) at about 230 ms (CP_t). Sway (θ_{ap}) reaches a peak (θ_a) at about 260 ms (θ_t) and then returns toward an upright position. The time (CPV_t) of the CPV peak amplitude (CPV_a) occurs between CP_o and CP_t.

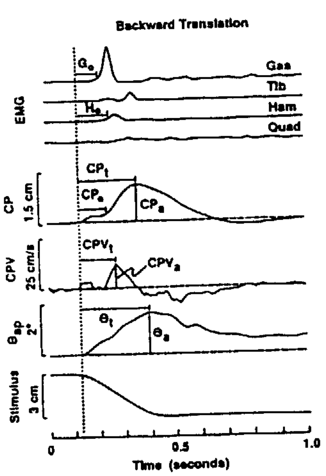

Figure 1. Average EMG, CP, CPV, and θ_{ap} responses of one individual to five consecutive backward support surface translations. Arrows indicate the various amplitude and time parameters used to quantify responses.

Forward support surface translations causing backward sway with respect to the platform initiate contractions of the T and Q muscles. The patterns of sway and changes in CP are similar to those for backward translations, but have opposite sign.

The population statistics describing EMG onset times, CP_o, CP_t, CPV_t, θ_t, CP_a, CPV_a, and θ_a are given in Table 1. The values of all EMG onsets, CP_o, and CPV_t were symmetrically distributed about their means. CP_a and CPV_a distributions were slightly skewed toward larger values. Most values of θ_t for both forward and backward translations were tightly grouped around 260 ms but ~15% of the population had values of θ_t of ~375 ms. θ_t for backward translations also included a scattering of times shorter than 260 ms.

CP_t for both forward and backward translations showed bimodal distributions. For backward translations, 82% of the subjects had both right and left leg CP_t's centered about a mean of 245 ms, 11% had both right and left CP_t's centered about 360 ms, and the remainder of the population had one leg's $CP_t < 300$ ms and the other leg's $CP_t > 300$ ms. For forward translations, 49% of the subjects had both right and left leg CP_t's centered about a mean of 260 ms, 34% had both right and left CP_t's centered about 350 ms, and the remainder of the population had one leg's $CP_t < 300$ ms and the other leg's $CP_t > 300$ ms.

For both forward and backward translations, subjects with shorter CP_t's (<300 ms) had larger mean values of CPV_a and smaller

Table 1. EMG, CP, CPV, and θ_{ap} Parameters (mean ± 1 SD*)

		Backward translation			Forward translation		
	Units	Right	Left	N	Right	Left	N
G_o	ms		115 ± 12.9	182			
H_o	ms		145 ± 22.5	168			
T_o	ms					114 ± 13.4	206
Q_o	ms					126 ± 19.6	147
CP_o	ms	127 ± 13.1	127 ± 14.1	211	137 ± 23.1	137 ± 24.5	212
CPV_t	ms	167 ± 17.2	166 ± 18.2	211	180 ± 39.7	176 ± 32.0	212
CP_t	ms	261 ± 46.5	262 ± 43.7	211	297 ± 56.5	291 ± 55.3	212
θ_t	ms		259 ± 55.9	211		289 ± 44.2	212
CP_a	cm	1.31 ± 0.42	1.31 ± 0.46	211	1.24 ± 0.40	1.23 ± 0.39	212
CP_a/h^2	cm/m²	0.46 ± 0.13	0.50 ± 0.14	198	0.44 ± 0.12	0.43 ± 0.13	199
θ_a	degrees		1.62 ± 0.37	211		1.88 ± 0.31	212

*SD, standard deviation.

mean values of CP_o and CPV_t (all significant at $P < 0.01$) than subjects with longer CP_t's (>300 ms). For backward but not forward translations, mean CP_a were also significantly larger for the short CP_t group. There was no clear relation between the bimodality of the θ_t and CP_t distributions. That is, many subjects with larger CP_t's had smaller θ_t's, and other subjects with smaller CP_t's had larger θ_t's.

The response pattern from three subjects during backward translation and two subjects during forward translation did not allow for accurate estimation of the various center of pressure and sway parameters. In all these cases there appeared to be little or no active torque generated by the subjects.

Age-Related Changes in EMG *Onsets*

With the exception of the quadriceps, EMG onset times generally increased with increasing subject age (Figure 2A–D). Linear fits to the data (Table 2) showed that the rate of change of EMG onset times with age were 0.21 ms/y for G_o, 0.30 for H_o, 0.10 for T_o, and −0.07 for Q_o with linear correlation coefficients of 0.335, 0.267, 0.158, and −0.075 respectively. However, the lowess fits to G_o, T_o, and H_o suggested that there may be an inflection point at roughly age 55 with a larger rate of change for subjects older than 55 y. To compare the rates for younger and older subjects, two part linear fits were made to G_o, H_o, and T_o for subjects younger and older than 55 y with the constraint that the two linear fits intersect at age 55 y. The slopes for younger versus older subjects were 0.17 versus 0.40, 0.14 versus 0.83, and 0.02 versus 0.45 ms/y for G_o, H_o, and T_o respectively. The slowing of motor responses in the older age group was most evident in the T responses since there was a transition from essentially no trend with age for subjects younger than 55 y to a slope comparable to the G_o and H_o data.

The difference between the EMG onset times for the H and G muscles ($H_o - G_o$) during backward translations, and between Q and T muscles ($Q_o - T_o$) during forward translations is plotted as a function of subject age in Figure 2E and F. There was a small increase in the $H_o - G_o$ delay with increasing age (0.17 ms/y with $r = 0.185$). For the $Q_o - T_o$ delay, subjects younger than 20 y tended to have larger $Q_o - T_o$ delays (mean 22.2 ms ± 22.0 SD) than subjects older than 20 y (mean 9.6 ms ± 18.0 SD). The difference in mean $Q_o - T_o$ between these two groups is significant $P < 0.01$). The larger $Q_o - T_o$ delays for younger compared to older subjects is the result of: (1) later Q_o values for younger subjects (Figures 2D), and (2) the upward

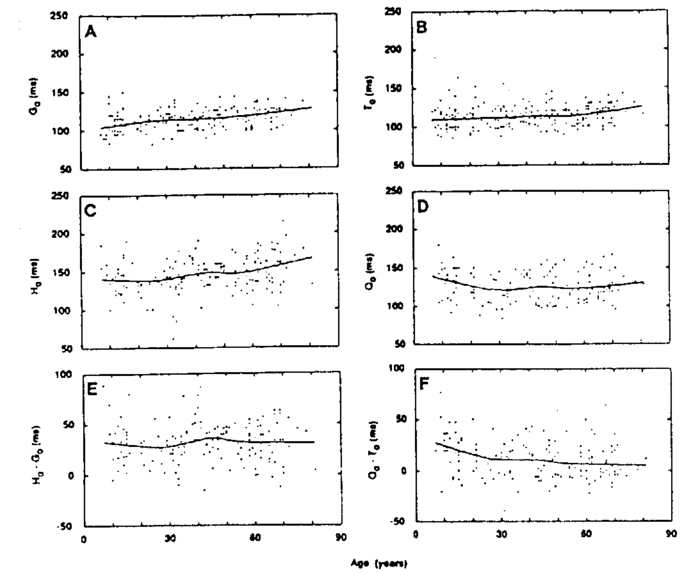

Figure 2. EMG onset times (A, B, C, D) from support surface translations as a function of subject age, and the difference between proximal and distal EMG onset times (E, F) as a function of age. Plots are based on recordings from 182, 168, 206, and 147 subjects for G_o, H_o, T_o, and Q_o, and 145 and 147 for $H_o - G_o$ and $Q_o - T_o$ respectively. Solid lines through data are lowess fits with $f = 0.5$.

trend in T_o with age, particularly for older subjects, coupled with essentially no age trend for Q_o in subjects older than 20 y.

Age Related Changes in CP *and* CPV

Figure 3 shows CP_e, CP_r, CPV_e, CPV_r, and CP_o as a function of age from backward translations recorded from the right leg. In addition, CP_e normalized by dividing by the square of subject height in meters is also plotted. Table 2 summarizes linear regression fits to CP_r, CPV_r, and CP_o data versus age. Linear regressions to CP_r, which had a bimodal distribution, were restricted to the larger group of subjects whose CP_r's were <300 ms.

CP_r, CPV_r, and CP_o for backward translations, and CP_r for forward translations showed small (0.2 ms/y) approximately linear increases with increasing age. CPV_r and CP_o for forward translations did not change significantly with age.

Values of CP_e and CPV_e for subjects older than ~20 y did not show any consistent trend. However, CP_e and CPV_e for subjects younger than 20 y showed large increases with increasing age in both forward and backward translations (Table 2). Normalizing CP_e and CPV_e by the square of the individual subjects' heights (h^2) removed most of the age-related trends in CP_e and CPV_e in the younger subjects indicating that the source of this trend was probably related to changes in body dimensions with growth. Normalization by h^2 also reduced the entire population's variability of CP_e relative to the mean value. For example, the coefficient of variation (CV = SD/mean) of CP_e from right leg backward translations was 0.98 while the CV of CP_e/h^2 was 0.29. The normalization only slightly reduced the CV of CPV_e from 0.44 to 0.41. Normalization of CP_e by h^2 theoretically provides a value proportional to the peak rotational acceleration of the body about the ankle joint (see Discussion).

Right-Left Asymmetry

Table 3 summarizes comparisons between measures of CP_e, CP_r, CPV_e, CPV_r, and CP_o recorded from the right and left legs during forward and backward translations. The CP_e and CPV_e responses from the left leg were significantly larger than the right during backward, but not forward translations. With the exception of forward translation CP_r, in which right side responses were longer than left, there were no significant timing differences between right and left leg responses.

Comparison of EMG *and* CP *Timing*

Table 4 summarizes the correlations between EMG onset times and the various measures of CP and sway times including CP_o, CP_e, CP_r, CPV_r, and θ_r. The average of right and left leg responses of CP_e, CP_r, and CPV_r were used in the calculations. The bimodal distributions of CP_r and θ_r distorted the correlation analysis when data from all subjects were included; therefore, the analysis was restricted to the larger portion of the population with shorter CP_r and θ_r responses.

In general, there were moderate, positive correlations between the various response timing measures. For both forward and backward translations, the largest correlations

Table 2. Linear Regression and Correlation Coefficients for EMG, CP, CPV, and θ_{rp} Parameters versus Age*

	Backward translation				Forward translation			
	Slope (change/yr)	Intercept	r	N	Slope (change/yr)	Intercept	r	N
G_o	0.21	107	0.34§	182				
H_o	0.30	132	0.27§	168				
T_o					0.10	110	0.16§	206
Q_o					−0.07	129	−0.08	147
CP_o	0.16	121	0.26§	211	−0.06	139	−0.06	212
CP_r†	0.15	238	0.19§	172	0.34	239	0.32§	104
CPV_r	0.21	158	0.26§	211	−0.10	182	−0.06	212
θ_r‡	0.31	233	0.15	189	0.08	265	0.19§	171
CP_e								
Age < 20 yr	0.11	−0.27	0.69§	46	0.06	0.15	0.55§	46
Age ≥ 20 yr	−0.004	1.59	−0.16	165	0.001	1.24	0.06	166
CPV_e								
Age < 20 yr	1.26	−2.89	0.52§	46	0.66	1.27	0.47§	46
Age ≥ 20 yr	−0.05	17.4	−0.13	165	0.009	12.0	0.03	166
CP_e/h^2	0.0004	0.47	0.07	198	0.002	0.36	0.33§	199
CPV_e/h^2	0.0017	5.10	0.02	198	0.015	3.54	0.23§	199
θ_e	−0.002	1.71	−0.09	211	−0.002	1.95	−0.11	211

*Units are the same as in Table 1.
†Only includes values with both right and left CP_r's < 300 ms.
‡Only includes θ_r's < 300 ms.
§Correlation coefficients significantly different from zero ($P < 0.05$).

Table 3. Percent Asymmetry and Absolute Difference between Right (R) and Left (L) Motor Response Parameters*

	Units	Backward translation: 100 (R − L)/(R + L)	Backward translation: R − L	Forward translation: 100 (R − L)/(R + L)	Forward translation: R − L
CP_a	cm	−4.3 ± 12.5	−0.12 ± 0.33§	0.4 ± 0.9	0.01 ± 0.27
CPV_a	cm/s	−5.1 ± 14.5	−1.4 ± 4.1§	0.5 ± 12.8	0.0 ± 3.0
CP_t†	ms	−0.3 ± 5.1	−1.4 ± 30.1	1.0 ± 7.0	6.0 ± 41.2‡
CPV_t†	ms	0.2 ± 4.1	0.5 ± 14.0	0.9 ± 7.3	4.4 ± 31.6
CP_a	ms	−0.2 ± 4.1	−0.5 ± 10.4	−0.2 ± 4.7	−0.9 ± 13.6

*All values are mean ± 1 SD. N's are 211 subjects for backward and 212 for forward translations.
†A nonparametric Wilcoxon signed rank statistic was used to test for significant R − L differences from zero for CP_t and CPV_t. A paired variable *t* test was used for the same purpose on all other variables.
‡$P < 0.005$.
§$P < 0.0001$.

were between CP_a and CPV_t. The largest correlation between any EMG and CP parameters was between T_o and CP_a for forward translations. θ_t correlations with other response time parameters were smaller than most other comparisons.

The interpretation of this correlation analysis is potentially problematic since different subsets of the population contributed to different correlation measures. However, a correlation analysis that included only subjects with no missing values gave similar results.

Forward-Backward Translation Comparison

Table 1 shows that CP_a, CP_t, CPV_t, and θ_t response times were larger for forward than backward translations even though T_o and Q_o times were similar, and even slightly shorter than G_o and H_o times. In addition, the timing difference between forward and backward translations increases for parameters that occur later in the normal sequence of motion. That is, EMG timing is similar, CP_a is 10 ms later, CPV_t is 13 ms later, and CP_t and θ_t are ~30 ms later for forward compared to backward translations.

Response amplitude measures also differed between forward and backward translations. Both CP_a and CPV_a were significantly larger for backward translations, and θ_a was larger for forward translations (all $P < 0.0001$, paired *t* test). This pattern is consistent with the generation of less corrective torque on forward translations compared to backward, resulting in larger peak body sway from forward platform motions.

Discussion

Most of the results of motor tests of postural control showed a wide range of what must be considered normal function. In spite of the large variances, age-related changes in function were identified in some response parameters. Although these changes were statistically significant, they were small in magnitude. In particular, the latency of EMG onsets, with the exception of quadriceps, increased with increasing age. In addition, there was evidence that the rate of increase of EMG onset with age was larger for subjects older than ~55 y. This increased rate was most evident in the tibialis muscle. Studies of muscle strength in the elderly (16) have also shown proportionally larger losses in tibialis strength compared to other leg muscles. The loss of strength combined with the slowing of the tibialis muscle response to body perturbations would diminish an individual's ability to control backward sway.

The distal before proximal muscle contraction synergy was observed in most subjects. However, during forward translations, Q_o preceded T_o in ~25% of the subjects. This

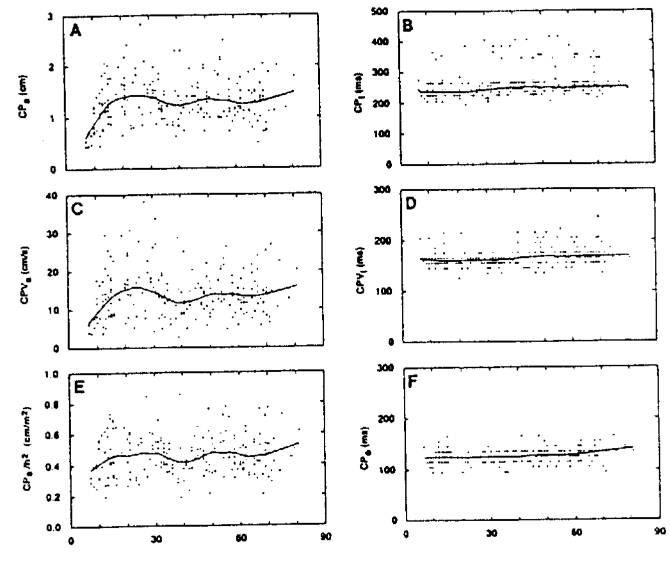

Figure 3. Various motor response amplitude (A, C, E) and time (B, D, F) parameters as a function of subject age. All plots are from backward translation responses from the right leg. Solid lines through data are lowess fits with *f* = 0.3.

may be related to initial knee position that was not carefully controlled. For example, if the knees of some subjects were slightly flexed prior to the translation, an early *Q* contraction would hyperextend the knee and pull the lower part of the trunk slightly forward. A previous study (17) also noted that some subjects had reversed $Q_o - T_o$ timing. However, in that study the reversal was only found in their older subjects. Figure 2F shows that $Q_o - T_o$ reversal occurred across the entire age range, although there was a slightly larger incidence in older subjects.

Normalization of CP_a and CPV_a by the square of subject height (h^2) both removed a large age-related trend for subjects ~20 y, and reduced the variability relative to the mean of CP_a for the entire population. The rationale for this normalization relates to the mechanics of movement. In order to correct for an external perturbation that causes AP sway, a subject exerts a torque, T, about the ankle joint. This torque produces a rotational acceleration, α, according to $\alpha = T/I$ where I is the moment of inertia of the subject. I is related to the mass distribution of the subject relative to the rotation axis (ankle joint). Using the simplifying assumption that all of the subject's mass, m, is located at the center of mass (about hip level), then $I = mr^2$ where r is the distance from the ankle joint to the center of mass. The calculation of CP gives a value proportional to T/m. Dividing CP by h^2 gives a value proportional to T/I

Table 4. Linear Correlation Coefficients Comparing Motor Response Times

	H_o r	H_o N	CP_e r	CP_e N	CPV_f r	CPV_f N	CP_f* r	CP_f* N	θ_f† r	θ_f† N
Backward translation										
G_o	0.38	145	0.37	181	0.37	181	0.33	149	0.22	163
H_o			0.34	167	0.41	167	0.27	138	0.25	148
CP_e					0.61	211	0.40	172	0.31	189
CPV_f							0.50	172	0.25	189
CP_f*									0.36	159

	Q_o r	Q_o N	CP_e r	CP_e N	CPV_f r	CPV_f N	CP_f* r	CP_f* N	θ_f† r	θ_f† N
Forward translation										
T_o	0.34	147	0.54	205	0.48	205	0.12	101	0.23	166
Q_o			0.39	147	0.21	147	−0.08	76	0.08	120
CP_e					0.63	212	0.20	104	0.12	171
CPV_f							0.34	104	0.17	171
CP_f*									0.16	89

*Only includes values with right and left CP_f's < 300 ms.
†Only includes θ_f's < 300 ms.

and to α since r^2 is proportional to h^2. CP_e/h^2 data versus age is fairly constant indicating that the peak angular acceleration in response to a sudden translation changes little with age.

Postural motor coordination tests similar to those described here are being used increasingly for clinical evaluation. For patients with balance disorders, these motor tests serve a function similar to optokinetic and pursuit tests for the evaluation of the ascending visual and visual-vestibular system control of eye movements. That is, they provide information on the integrity of spinal and central nervous system function important for the interpretation of sensory organization tests of posture control (13).

The clinical use of postural motor coordination tests requires an appropriate selection of response parameters and a definition of the range of these parameters in a normal population. Ideally these parameters should have narrow distributions for normal subjects and should be sensitive to abnormalities. This paper does not address abnormal response patterns, but the results do suggest that some of the potential motor response parameters may be difficult to use clinically. In general, the variability of the parameters was large even though age-related trends contributed very little to the variability. Among timing measures, G_o and CP_e for backward translations and T_o for forward translations showed the least variability, followed by Q_o and CP_e for forward, H_o for backward, and CPV_f for both forward and backward translations. Despite their narrow distributions, EMG onset times are problematic in routine clinical tests since they are often difficult to measure.

The bimodal distributions of CP_f and θ_f make them less attractive candidates for clinical functional measures. Although there were differences between some motor response measures related to the bimodal distributions of CP_f, there was no clear indication of the source(s) of these bimodal responses. Perhaps the support surface perturbations evoked different movement patterns in different subjects with some subjects moving like inverted pendulums, while others used more complex motions to maintain their upright posture.

Among response amplitude measures, there appeared to be little range for abnormally low CP_e and CPV_e responses since many subjects in our putatively normal population showed responses only slightly above

the passive-platform artifact level. Different force platform designs with smaller mechanical artifacts might improve the separation of abnormal subjects from normal subjects with low amplitude responses. CP_e and CPV_e values normalized by h^2 were better parameters for comparisons across populations than CP_e and CPV_e alone.

Different mechanical platform systems, instrumentation, data analysis, and particularly stimulus parameters, could influence the conclusions drawn in this paper. As other motor coordination tests with different stimulus conditions are developed, it will be important to consider the possible presence of bimodal parameter distributions, to determine the neural or biomechanical factors that cause these bimodal responses, and to test a large enough population to clearly define the range of normal function.

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AGE-RELATED CHANGES IN HUMAN POSTURE CONTROL: SENSORY ORGANIZATION TESTS

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☐Abstract – Postural control was measured in 214 human subjects ranging in age from 7 to 81 y. Sensory organization tests measured the magnitude of anterior-posterior body sway during six 21 s trials in which visual and somatosensory orientation cues were altered (by rotating the visual surround and support surface in proportion to the subject's sway) or vision eliminated (eyes closed) ~~in various combinations~~. No age-related increase in postural sway was found for subjects standing on a fixed support surface with eyes open or closed. However, age-related increases in sway were found for conditions involving altered visual or somatosensory cues. Subjects older than ~55 y showed the largest sway increases. Subjects younger than ~15 y were also sensitive to alteration of sensory cues. On average, the older subjects were more affected by altered visual cues whereas younger subjects had more difficulty with altered somatosensory cues.

☐Keywords – posturography, vestibular, somatosensory, vision, equilibrium, development.

Introduction

The automatic control of upright stance is an active sensorimotor process that maintains the body's center of gravity over the base of support (the feet). This process requires that deviations of body position from upright be sensed and processed to initiate motor commands that oppose the initial deviation and return the body to an upright position. The vestibular, somatosensory, and visual systems are the main sources of sensory information on body motion. However, situations commonly arise in which information from the various sensory systems is absent or altered even in individuals with physiologically normal sensory function. For example, somatosensory cues from compliant surfaces can be misleading, and visual cues are eliminated when the eyes are closed. In order to maintain postural control under a variety of environmental conditions, motion information from sensory systems must be organized by the central nervous system so that inappropriate or inadequate sensory inputs can be ignored when necessary.

One method for testing postural control, which we will call sensory organization tests, involves postural responses ~~which~~ that occur over tens of seconds to minutes when subjects attempt to stand quietly in various sensory conditions. The simplest sensory organization tests are the clinical standard Romberg tests, which characterize spontaneous postural sway when the subject's eyes are open and closed (3,10). The standard Romberg tests can be extended by altering somatosensory and/or visual motion cues. The alteration of somatosensory and/or visual cues by rotating the subject's visual field and/or support surface in equal proportion to the subject's own sway (11) tests the subject's ability to maintain

Change "which" to "that"

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equilibrium when various combinations of sensory cues are inappropriate or inadequate for orientation to earth vertical. Postural control under these altered conditions may be more difficult than when information is missing, as with eyes closed.

The increased incidence of falls in the older population (18) suggests that one or more of the components required for accurate postural control degenerates with age. Studies that have looked for differences in postural control between young and old adults have generally found these differences (10,19), including increased sway or falls in various sensory organization tests. In addition children show developmental changes in postural control that converge to adult patterns at about age 10 y (6,15). However, the limited scope of these studies having small sample sizes and restricted test paradigms have not clarified either the time course or the mechanism involved in the changes in use of sensory information for postural control with increasing age. We tested a putatively normal population with a wide age distribution using both postural motor coordination (12) and sensory organization tests. In addition, vestibulo-ocular (VOR) and optokinetic reflexes (OKR) were independently tested in the same individuals (13,14) for comparison to postural responses.

Methods

Postural sway under various sensory conditions was measured in 214 human subjects (90 male and 124 female) aged 7 to 81 y. Ages were approximately uniformly distributed over the entire range. Details of subject selection are given in a previous paper (14). No subjects were excluded from the population based on any vestibular, optokinetic, or posture test results.

Body Sway Measurements

The anterior-posterior (AP) sway angle (θ_{ap}, see Figure 1) of each subject was recorded using a rod attached to a potentiometer. The potentiometer was mounted on a post next to the subject. The end of the rod rested in a V-shaped holder centered on the subject's back at hip level. A voltage proportional to the angular displacement of the potentiometer was recorded and later transformed using appropriate trigonometric conversions to θ_{ap}. A second potentiometer mounted at shoulder level recorded AP displacements at the shoulder in the last 65 subjects tested. A measure of hip angle (θ_h, Figure 1) was calculated from AP angles measured at the hip and shoulder. An approximate center-of-gravity AP sway angle (θ_{cg}) was calculated using the following formula:

$$\theta_{cg} = \tan^{-1}\left[\frac{0.860\sin\theta_{ap} + 0.242\sin(\theta_{ap} + \theta_h)}{0.860\cos\theta_{ap} + 0.242\cos(\theta_{ap} + \theta_h)}\right]. \quad [1]$$

This formula was derived assuming the subjects had average body mass distribution and average proportional lengths of various body segments (5). To the extent that the various subjects deviated from average body configurations, the measurement of θ_{cg} would be in error. This error was probably not more than 10% in this population. Sway angles were sampled at 50 Hz.

Test Conditions

Subjects stood with feet separated by about 20 cm on a movable support surface surrounded in front and on two sides by a movable visual surround. The visual surround was a box with randomly placed 2 cm black dots on a white surface. The average spacing between the dots was ~20 cm, and the distance from the subject to the box was ~50 cm. Support surface motion was controlled by a hydraulic position servo system that could produce toe up and toe down rotations about an axis collinear with the subject's ankle joints. Visual surround motion was controlled by a separate hydraulic servo system that rotated the box about the ankle joint axis. Subjects wore a harness attached

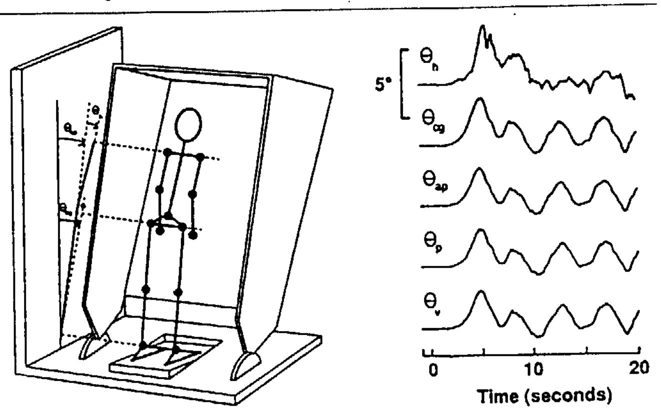

Figure 1. Schematic representation of posture test apparatus and definition of body angles for AP sway in a sagittal plane. The interior of the visual field was white with randomly placed black dots. To the right, body sway angles recorded from one subject during a condition 6 sensory organization test are shown along with traces indicating the sway-referenced motion of the visual field, θ_v, and the support surface platform, θ_p.

to the ceiling to prevent injury in case of a fall. The harness did not impede body motions even during large amplitude sways.

Each subject performed six tests that provided a functional evaluation of the ability of the subject to effectively use vestibular, somatosensory, and visual information in the control of upright posture (11). The subject's task was to maintain an upright stance for 21 s during each of the six conditions with as little postural sway as possible and without moving the feet. The test was rated a fall if the subject required the assistance of the harness to maintain upright stance or if a step was taken in order to prevent a fall into the harness. The six test conditions generally were performed once. In later subjects, a test condition was immediately repeated if the subject fell in that condition. Tabulated results on falls and sway are based on the performance in the first test of each condition.

Conditions 1 and 2 required the subject to stand on a stable surface for 21 s facing an earth-fixed visual field with eyes open and then with eyes closed. The remaining four conditions placed the subject in more demanding sensory environments. These environments were created by rotating the visual field and/or the support surface in equal proportion to θ_{ap}. For example, as the subject swayed forward, the visual field rotated forward about an axis through the ankle joint. Under this condition the normal relationship between body motion and retinal image motion is altered. This is referred to as sway-referenced vision as opposed to the earth-referenced vision in condition 1. The precise relation between the retina and the box depended on the movement patterns that subjects used during sway. The same technique was applied to the support surface by rotating it about the ankle joint in proportion to θ_{ap}. This sway-referenced support condition greatly reduced the change in ankle joint angle as the subject swayed back and forth and therefore altered the somatosensory cues contributing to postural control. The entire sensory test sequence included all six combinations of eyes closed, sway-referenced, and earth-referenced vision and support surface conditions given in Table 1.

Table 1. Sensory Organization Test Conditions

	Test conditions			Sensory information		
Condition	Sensory conflict	Visual reference	Support surface reference	Earth-referenced (accurate)	Sway-referenced (altered)	Absent
1	no	earth	earth	visual, vestibular, somatosensory		
2	no	eyes closed	earth	vestibular, somatosensory		visual
3	yes	sway	earth	vestibular, somatosensory	visual	
4	yes	earth	sway	vestibular, visual	somatosensory	
5	yes	eyes closed	sway	vestibular	somatosensory	visual
6	yes	sway	sway	vestibular	somatosensory, visual	

Body Sway Analysis

Sway data were summarized by two measures: average rectified sway (ARS) and peak-to-peak sway. Both measures were calculated over the final 20 s of the 21 s trials (the visual field and support surface were always earth-referenced during the first second of each trial). For ARS calculations, sway data was normalized by subtracting the average sway values recorded in the first second from the entire sway record. Sway data samples < zero were then rectified (inverted), and the new sway trace was averaged over the final 20 s. ARS often did not reflect how close a given individual was to a fall since, for example, a subject who leaned forward by a few degrees and stayed in that position throughout the remainder of the trial could score the same as a subject who oscillated back and forth during the trial with the peak of the oscillations close to the threshold of a fall. The peak-to-peak sway measure was more indicative of the closeness of sway to fall thresholds.

Movement Strategies

Subjects typically use one of two body motion strategies to maintain upright stance without moving the feet (7). A hip strategy consists of θ_h and θ_{ap} motions that are out of phase. Subjects can be forced to use a hip strategy by asking them to stand on a narrow beam that limits the amount of torque that can be exerted at the ankle. A pure ankle strategy occurs when all motion is about the ankle joint (AP sway angles measured at the hip and shoulder are equal and θ_h is zero). A less pure ankle strategy occurs when there is some motion about the hip joint, but θ_{ap} and θ_h are in phase with each other. In order to quantify the type of body motion, a strategy measure was calculated according to the following formula:

$$\text{strategy score} = \overline{[(\theta_{ap} - \overline{\theta_{ap}})\ (\theta_h - \overline{\theta_h})]} \qquad [2]$$

where the bars over the various terms indicate the average values over time, ie, the strategy score is the average product of zero-meaned θ_{ap} and θ_h calculated over the duration of the trial. The strategy score is negative if θ_h and θ_{ap} are out of phase indicating that the trunk and legs move in opposite directions, positive if they are in phase and the body moves like a whip, and zero when the body moves like an inverted pendulum with no bending at the waist. Since this measure is an average over the entire trial, changes in strategy during the trial would not be correctly characterized by this single measure. In practice, this was not a problem since this putatively normal population did not show marked strategy changes within trials.

Visualization of Trends

In order to visualize trends in scatterplots, a robust locally weighted regression analysis (lowess fit) was used to smooth the scatterplots (4). This smoothing is similar to a mov-

Table 2. AP Sway Measures in Completed Sensory Test Conditions (mean ± 1 SD*)

Condition	N	Average rectified θ_{ap} sway (degrees)	Peak-to-peak θ_{ap} sway (degrees)
1	214	0.26 ± 0.21	0.82 ± 0.44
2	214	0.42 ± 0.33	1.25 ± 0.58
3	189	0.68 ± 0.57	2.78 ± 2.00
4	213	0.74 ± 0.49	2.95 ± 1.97
5	192	1.20 ± 0.48	5.54 ± 2.23
6	155	1.32 ± 0.53	5.75 ± 2.12

*SD, standard deviation.

ing average filter, but is less sensitive to outlying points and allows variable amounts of smoothing. A lowess smoothing parameter of 0.5 and iteration parameter of 2 were used on all data sets.

Results

As visual and somatosensory sensory information were removed and/or altered during the various sensory test conditions, subjects became less stable (Table 2) and falls became more likely. Most subjects did not fall in any condition, but their sway amplitudes increased as they were deprived of orientationally accurate sensory reference information. Some subjects, particularly older subjects, did fall. The pattern of falls was not random, but rather was restricted to certain conditions and combinations of conditions.

Sway Responses

All subjects were on a stable surface with eyes open or with eyes closed (conditions 1 and 2, see Figure 2). No subjects fell in conditions 1 or 2. Postural sway increased in condition 3 when the visual surround rotation was referenced to the subject's sway. The median of the condition 3 sway distribution in Figure 2 was only about one degree higher than condition 2 indicating that most subjects had only slightly more difficulty controlling their posture under the sway-referenced vision condition than with eyes closed. However, the condition 3 distribution is highly skewed toward larger sway amplitudes indicating that a significant fraction of the population had difficulty maintaining their upright posture when visual orientation information was present but sway-referenced. In addition, 30 of 214 subjects (14%) fell in condition 3.

"All subjects had minimal sway standing with eyes open or eyes closed on a stable surface (conditions 1 and 2, see Figure 2)."

Condition 4 provided earth-referenced visual cues but sway-referenced somatosensory cues. This distribution was skewed toward larger sway angles in a similar manner to the condition 3 distribution. ~~However~~ On average, subjects were more stable in condition 4 than in condition 3 since only three subjects out of 214 (1.4%) fell in condition 4.

delete "However"

Visual cues in condition 5 were absent (eyes closed) and somatosensory cues were altered by the sway-referenced support surface. This condition presumably forced a greater reliance on vestibular cues for postural control. Average sway was larger than in any of the previous conditions and was also skewed toward larger values. Twenty-eight of 214 subjects (13.1%) fell in this condition.

Condition 6 was the most difficult of the six conditions. Under this condition, both the visual surround and the support surface were sway referenced. As with condition 5, this condition forced a greater reliance on vestibular cues for postural control. However, the presence of altered visual orientation cues in condition 6 (as opposed to absent vision in condition 5) apparently increased the difficulty of the task. The average sway for subjects who completed condition 6 was larger than in any other conditions, and 70 of 214 subjects (32.7%) fell.

The amount of sway in one condition was generally a poor predictor of the amount of sway in ~~a different~~ another condition. Linear correlation coefficients relating peak-to-peak sway in various combinations of conditions were calculated for the 125 subjects who did not fall in any condition. The largest correlation coefficients (ranging from 0.42-0.54) were between conditions 1 and 2, and among conditions 4, 5, and 6. Correlation coefficients comparing sway in conditions 1 and 4, and 2 and 4 were −0.3. The correlations coefficients for all other paired comparisons were <0.2.

replace "a different" with "another"

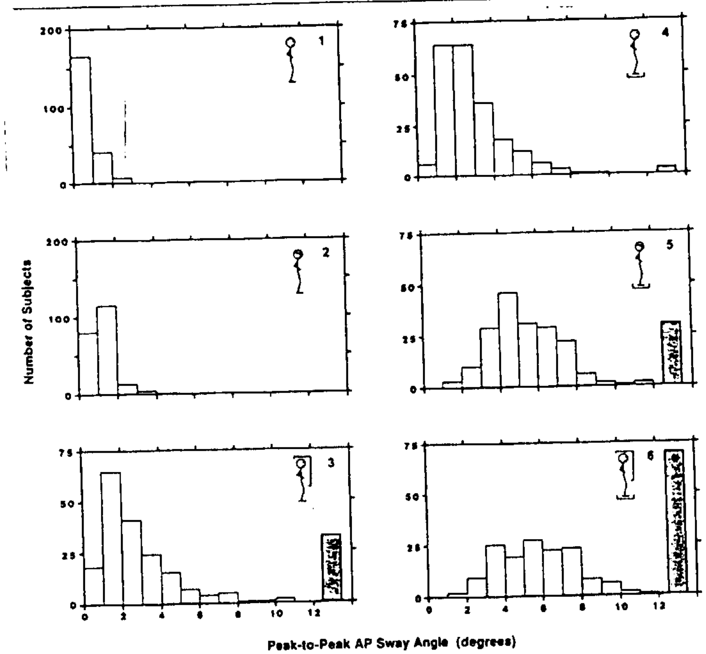

Figure 2. Histograms of peak-to-peak θ_{ap} under the six different sensory test conditions. Gray bars to the right of each histogram indicate the number of subjects who fell in that condition. One subject in condition 3, and 5 subjects in condition 5 had sways greater than 12° but did not fall, and are not included in those respective histograms.

Movement Strategy

The use of an ankle strategy was by far the most common mode of postural sway in these subjects. The 65 subjects whose body motions were measured at both the hip and shoulder were the older portion of the entire population with a mean age of 56.2 y (12.5 SD, range 27 to 81 y). Their mean strategy scores were close to zero and the variances of the scores were small in all six conditions (Table 3). This was confirmed by plotting peak-to-peak θ_{cg} versus peak-to-peak θ_{ep}. For pure hip strategies θ_{cg} and θ_{ep} should be relatively unrelated whereas a pure ankle strategy would have equal θ_{cg} and θ_{ep}, and correlations of −1.0. Correlation coefficients between peak-to-peak θ_{cg} and θ_{ep} data ranged from 0.93-0.98 for the six conditions. Data points were tightly clustered around the line of equal θ_{cg} and θ_{ep}.

"scores"

Fall Patterns

Table 4 summarizes the data on subjects who fell during one or more of the six condi-

Table 3. AP Sway Measures for Subjects Who Completed the Sensory Tests, Sway Measured at Shoulder and Hip (mean ± 1 SD)

Condition	N	Peak-to-peak θ_{ap} sway (degrees)	Peak-to-peak θ_{cg} sway (degrees)	Peak-to-peak θ_h sway (degrees)	Strategy score
1	65	0.75 ± 0.32	0.72 ± 0.34	1.41 ± 0.56	−0.01 ± 0.05
2	65	1.21 ± 0.54	1.33 ± 0.59	1.78 ± 0.81	0.04 ± 0.07
3	46	3.01 ± 2.16	3.28 ± 2.18	3.56 ± 2.09	0.24 ± 0.56
4	65	2.41 ± 1.20	2.22 ± 1.15	2.96 ± 2.10	−0.14 ± 0.46
5	56	5.31 ± 2.45	5.50 ± 2.37	5.53 ± 3.44	0.21 ± 1.38
6	35	5.27 ± 1.86	5.58 ± 2.12	6.08 ± 4.91	0.49 ± 0.95

tions. Falls during sensory test conditions were not random occurrences, but rather were associated with the inability of some subjects to obtain and/or coordinate the sensory information available for the control of posture. Consider subjects who fell in two of the four conditions that presented them with sensory conflict situations. There are six possible combinations of paired falls within the grouping of the four more difficult conditions. If paired falls occurred randomly they would be evenly distributed across the six possible combinations. This was clearly not the case since four of the six combinations of paired falls either were not observed or were rare. That is, no subjects fell in 3-4 and 4-5 paired conditions, and only two subjects fell in the 3-5 and one in the 4-6 combination of conditions. Therefore paired falls were primarily limited to only two of the six possible paired combinations with 12 subjects falling in 5-6 conditions, and 15 falling in 3-6 conditions.

The six subjects who fell in three conditions were also not randomly distributed among the four possible combinations. Rather all six subjects fell in the same set of three conditions which was the 3-5-6 combination. This combination combines the features of the two most common paired condition falls, 3-6 and 5-6 as reported previously (1,2,11). Most of these subjects were older (aged 45, 48, 60, 66, 69, and 70 y).

There was a clear learning effect when sensory tests were repeated immediately following a fall. Thirty-three of the 131 first test falls were repeated. Only 6 of the 33 subjects (18%) fell in the repeated test. The number of repeat test falls for the four sensory conditions where falls occurred were 1 of 8 for condition 3, 0 of 2 for condition 4, 2 of 8 for condition 5, and 3 of 15 for condition 6.

Table 4. Falls in Sensory Test Conditions

	Condition	Number of subjects	% of total (N = 214)
1 Fall	6	36	16.8
	5	8	3.7
	3	7	3.3
	4	2	0.9
Totals		53	24.8
2 Falls	3, 6	15	7.0
	5, 6	12	5.6
	3, 5	2	0.9
	4, 6	1	0.5
Totals		30	14.0
3 Falls	3, 5, 6	6	2.8
1, 2, or 3 Falls		89	41.6

Age-Related Changes

Figure 3 shows peak-to-peak sway and falls as a function of age. Generally, the number of falls increased with increasing age (Table 5). The incidence of falls was lowest for subjects aged 20–40 y. Subjects aged 13–19 y had a high incidence of single condition falls (33%) but low multiple condition falls (11%).

The lowess fits to peak-to-peak sway data in Figure 3 suggest that younger subjects swayed more than subjects in the middle of the age range. A comparison of peak-to-peak sway of 7–15 y olds with 30–39 y olds showed significant differences for conditions 1, 4, 5,

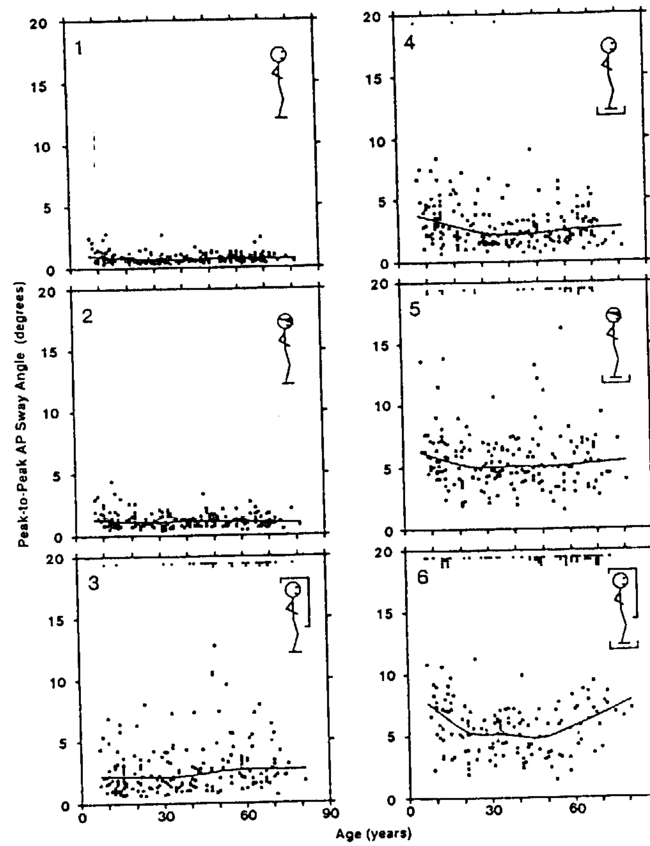

Figure 3. Peak-to-peak θ_{ap} as a function of age for the six sensory test conditions. Solid dots at the top of each graph indicate subjects who fell in that condition. Solid lines are lowess fits to the data for subjects who did not fall during the test.

and 6 ($P < 0.02$, Mann-Whitney nonparametric test). The larger sway of younger subjects was most evident in sensory conditions 4, 5, and 6 suggesting that younger subjects were sensitive to alterations in somatosensory cues.

The occurrence of single condition falls increased rapidly for subjects older than ~45 y, although the incidence of multiple condition falls remained quite stable through the 50's before showing an increase in the 60–70 y

Table 5. Sensory Test Falls Sorted by Subject Age

Age group	Number of subjects	Single falls N	Single falls %	Multiple falls N	Multiple falls %	Total falls N	Total falls %
7-12	21	3	14.3	2	9.5	5	23.8
13-19	27	9	33.3	3	11.1	12	44.4
20-29	28	4	14.3	2	7.1	6	21.4
30-39	32	4	12.5	2	6.3	6	18.8
40-49	32	9	28.1	6	18.8	15	46.9
50-59	26	8	30.8	5	19.2	13	50.0
60-69	35	10	28.6	14	40.0	24	68.6
70 and over	13	6	46.2	2	15.2	8	61.5
Total	214	53	24.8	36	16.8	89	41.6

olds. A possible anomalous result was obtained in the over 70 age group for multiple falls. Their multiple fall rate was less than the fall rate for 60-70 subjects and approximately the same as for subjects 40-60 y. This may be a result of the small sample size of the over 70 age group, an exceptionally healthy condition of this group, or an exceptionally high fall rate for subjects in the 60-70 age group.

Surprisingly, the increased incidence of falls in older subjects in conditions 3 and 5 was not accompanied by a trend toward increasing peak-to-peak θ_{ap} among non-fallers (Figure 3). This is in contrast to condition 6 where both sway and falls increased with age. Since it is not possible to maintain stance with the body's center of gravity outside of its base of support, the theoretical limit of peak-to-peak sway is dependent on foot size and body mass distribution. Since most subjects have a 10° to 12° range of stable AP sway, and peak-to-peak θ_{ap} average 3° and 5.3° for conditions 3 and 5, respectively, it seems that there was some room for the nonfalling population to shift toward larger sways in conditions 3 and 5, and that this shift would be accompanied by an increased number of falls. Although the falls increased, the peak-to-peak sway amplitude of nonfallers did not.

Discussion

Age-related changes in postural control performance were not present in subjects older than ~15 y when they were tested under normal operating conditions during sensory test conditions 1 and 2. That is, when subjects stood on an earth-fixed support surface with eyes open or closed, their sway was small and the oldest subjects performed as well as the younger ones. Since the first two sensory tests are characterized by the presence of multiple sensory system inputs that converge and cooperate in the generation of appropriate postural control responses, it is apparent that subjects are well adapted to an environment without conflicting sensory cues. However, it is also clear from the other sensory test conditions that the "parts" that make up the whole of postural control are not equally functional in all individuals. The prevalence of falls and the wide range of postural sway amplitudes in sensory test conditions 3-6 demonstrate this functional inequality.

Analysis of Fall Patterns

The pattern of falls among subjects who fell in two or more conditions can be logically associated with specific types of peripheral sensory or sensory integration problems. The 15 subjects who fell in the 3-6 conditions were highly sensitive to visual cues which were altered by sway-referencing the visual field. These subjects behaved paradoxically. The fact that they did not fall in condition 5 indicates that they were presumably able to use vestibular cues to properly maintain stable stance since no visual and only altered somatosensory cues were otherwise available. However, when visual cues were present in 3 and 6, they chose to ignore earth-referenced vestibular and somatosensory cues in condition 3 and earth-referenced vestibular cues in condition 6 in favor of the altered visual reference. Although the mechanisms that cause and sustain this preference for a visual reference are not known, it seems likely to be a sensory selection problem rather than a motor coordination problem (11).

The second most common paired falls occurred in conditions 5 and 6 that force subjects to rely primarily on their vestibular systems for postural control since somatosensory and/or visual cues are either absent or altered. This pattern of falls is found frequently in subjects who have peripheral vestibular deficits (8,11). A subject with total bilateral peripheral loss of vestibular function is the extreme form of vestibular deficiency. Patients with bilateral loss as judged from absent caloric and rotation responses invariably fall in conditions 5 and 6 (2,11).

Falls in both conditions 5 and 6 could theoretically also arise from central mechanisms. The central postural control mechanisms perform complex tasks that include both the generation of the correct motor commands to the muscles and the selection of an appropriate orientation reference based on information from several sensory systems that at times may be conflicting. It is possible that peripheral vestibular signals may be normal, but the central mechanisms that make use of this information are faulty. The "fault" may have more than one source. For example, the processing of the sensory information may simply be too slow, in which case the appropriate motor commands never arrive at the muscles or arrive too late to prevent a fall. Alternatively, the central processing that must deal with conflicting sensory information may produce inappropriate responses based on the available sensory signals. These inappropriate responses could drive the system into instability with a resulting fall. A possible example of this may be learning disabled children with normal VOR function who often fall in conditions 5 and 6 (8,16). Finally, motor coordination deficits and muscle weakness could also play a role in 5-6 fallers since these conditions evoke relatively large sway amplitudes in most subjects.

Developmental Postural Changes in Children

Subjects aged 7-15 y had increased peak-to-peak sway amplitudes in all sensory test conditions except conditions 2 and 3. This is consistent with previous results in children aged 2-15 y (15) and 1½-10 y (6,17). Both studies show increased AP sway in sensory condition 1 in the youngest children with a convergence toward adult performance at about age 8-10 y. One of these studies (6) also used sway-referenced tests identical to conditions 3-6 and again found the poorest performance in the youngest children but with incomplete convergence to adult values by age 10 y for all four conditions. Sway-referenced vision (condition 3) results presented here differ from this previous study (6) since the average peak-to-peak sway of the youngest subjects did not differ from adult sway values. However, condition 4, 5, and 6 results presented here agree with the previous findings (6) and extend those results to show that adult performance is not fully attained until ~ age 20 y under these sensory conditions.

In all conditions with altered somatosensory inputs (conditions 4, 5, and 6), subjects younger than ~15 y showed more sway, on average, than middle-aged adults. This suggests that younger subjects rely more heavily on somatosensory cues than do middle-aged and many older adults even when accurate, earth-referenced visual and vestibular cues are available. Many children had sway results compatible with adult sways, while others swayed considerably more than middle-aged adults. This wider range of postural sway for children compared to adults may be associated with differing rates of development of postural control abilities in different children.

Postural Control Changes in the Elderly

Sensory test results showed that most falls occurred in subjects older than ~50 y. In condition 6, the increased number of falls was accompanied by increased sway among nonfallers. However, older nonfallers performed about the same as younger subjects in conditions 3 and 5 even though there were increased falls in the elderly group.

The finding that there is no general increase in sway with age in conditions 3 and 5, but there is an increase in falls suggests that the elderly fallers form one or more subgroups within the elderly population. These elderly fallers apparently either lack information required for postural control or have adopted postural control schemes that are distinct from the remainder of the population and that place them at increased risk for falls in particular sensory environments. However, when redundant sensory cues are available (conditions 1 and 2) and sway amplitudes are greatly reduced, these fallers cannot be distinguished from the remainder of the population.

A simple explanation for these results could be that the average body alignment of these subjects places their center of gravity near a stability limit so that relatively small increases in sway produce a fall (9). If this explanation were correct, then subjects who fell in condition 3 should also fall in other conditions (4, 5, and 6) that increased their sway above the levels in conditions 1 and 2. In general, this pattern of falls did not occur.

Impairments in either sensory system inputs, central nervous system processing, or motor system output could potentially initiate or facilitate the development of postural control schemes that are generally adaptive (judging from the good performance in conditions 1 and 2) but inadequate or nonadaptive in other sensory environments. These impairments might include reduced or altered sensory information, reduced, delayed or absent motor responses, or incorrect patterns of muscle activation resulting in inappropriate and noncompensatory responses. Comparisons of sensory organization test results with postural motor coordination results, and VOR and OKR responses give some insight into the factors that contribute to the age-related decline in postural control in this putatively normal population.

Comparison with Motor Coordination Tests

Neither the amplitude nor timing of postural motor responses to forward and backward platform translations (12) were correlated with the level of sway during sensory tests. In addition, the motor response parameters of subjects who fell during sensory tests were not distinguishable from nonfallers. This would tend to support a hypothesis that sensory system deficits and/or inappropriate central nervous system organization of sensory information are responsible for the increased likelihood of falls independent of motor coordination problems.

However, the amplitude of platform translations used in the motor coordination tests did not produce body sways near fall thresholds, and thus did not require subjects to exert maximal muscular responses. A larger perturbation might have revealed relative muscle weaknesses as well as response timing problems within the subpopulation of fallers. Muscular strength relative to body mass and precise timing of responses might contribute to falls in conditions 5 and 6 where the average level of sway is closer to fall thresholds than in the other four conditions.

Comparison with VOR and OKR Function

There was evidence of VOR and OKR abnormalities in some subjects who fell in two or more conditions. Of the three subjects with the shortest VOR time constants (13,14) one was a 5-6 faller and two were 3-5-6 fallers. The 5-6 faller with a short VOR time constant also had a significant partial unilateral loss of vestibular function in the caloric test. The subject who had the largest OKR time delay (average delay to the onset of eye movement following visual field movement) of any subject tested (268 ms) was also a 5-6 faller. Among subjects over 50 y, two of the three subjects with the lowest OKR gain constants were 3-5-6 fallers, and the other was a 3-6 faller. Finally, the two older subjects with the largest OKR time constants, indicating decreased sensitivity to higher frequency visual field motions, were both 3-6 fallers.

With the exceptions mentioned above, VOR and OKR parameters of most subjects who fell in two or more conditions were not distinguishable from those of subjects who did not fall or fell in only one condition. A comparison of the overall incidence of ex-

treme VOR and OKR parameters (>97.5 or <2.5 percentile points) among subjects who fell in two or more conditions with the incidence of extreme parameters among subjects who fell in no more than one condition showed no significant difference between the groups.

There are at least three possible explanations for the weak correlation between VOR and OKR abnormalities and poor postural control. First, our VOR tests measured primarily horizontal canal function whereas head movements during postural sway primarily stimulate vertical canals and otoliths. To the extent that a vestibular abnormality may only affect one or a limited number of the vestibular receptors in each ear, horizontal VOR and posture results could differ. Second, our OKR tests used horizontal plane visual motion stimuli while the visual system contribution to postural control during sensory tests is associated with the detection of pitch plane movement and with depth cues from the disparity of images on the retina of each eye. There might be a higher correlation of abnormal pitch plane OKR and vergence control responses with postural control deficits than with horizontal plane OKR. Third, differences between our VOR and posture test results could relate to central nervous system problems in the organization of sensory system interactions. Abnormalities in the central nervous system pathways involved in the organization of posture might be specific to the postural control system and therefore would not effect VOR responses.

In conclusion, it is apparent that some equilibrium control deficits exist in a putatively normal population. These deficits are more common in children and subjects older than ~50 y, but are normally masked by the presence of redundant sources of sensory orientation cues. In susceptible subjects, the loss of redundant information can unmask their deficit and cause a sudden loss of postural control.

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