Mechanisms Controlling Head Stabilization in the Elderly During Random Rotations in the Vertical Plane

Emily A. Keshner
Sensory Motor Performance Program
Rehabilitation Institute of Chicago

Kelvin J. Chen
Biomedical Engineering Department
Northwestern University, Chicago

ABSTRACT. Frequency-related response characteristics of the mechanisms controlling stabilization of the head in 10 elderly subjects were compared with response characteristics in 8 young adults. Angular velocity of the head with respect to the trunk and EMG responses of 2 neck muscles were recorded in 10 seated subjects during pseudorandom rotations of the trunk in the sagittal plane at frequencies of 0.35 to 3.05 Hz. Subjects were required to actively stabilize their heads with (VS) and without (NV) visual feedback so that voluntary mechanisms and the influence of vision could be tested. Reflex mechanisms were examined when subjects were distracted by a mental calculation task during rotations in the dark (MA). Age emerged as an influential factor in the performance of head stabilization mechanisms, and decrements in performance were even more pronounced in the older as compared with younger elderly subjects. Age effects could be seen in the (a) diminished ability to voluntarily stabilize the head, particularly with the absence of vision, (b) impaired ability to stabilize the head when cognitively distracted, and (c) appearance of a resonant response of the head. Control of head stabilization shifted from reflex mechanisms to system mechanics, probably as a result of age-related changes in the integrity of the sensory systems. The elderly’s system mechanics could not effectively compensate for the disturbances, however, and instability was the result.

Key words: aging, head, neck, neck muscles, neural mechanisms, reflexes, system mechanics

Stabilization of the head in space is essential for maintaining visual stability and vertical orientation during locomotion and other natural movements (Gibson, 1966; Pozzo, Berthoz, & Lefort, 1990; Winter, 1991). If less-than-perfect stabilization occurs, changes and ambiguities can occur in the processing of sensory inputs. This may ultimately affect equilibrium and spatial orientation, resulting in functional consequences such as instability. Such consequences can be exaggerated in an elderly population that has experienced changes in the structure and function of their sensory end-organs (Horak, Diener, & Nashner, 1989). Data on the elderly suggest age-related changes in both the neural and mechanical components of postural control. In the older individual, natural aging involves slow structural deterioration of the nervous system that diminishes and interferes with the detection and perception of stimuli (Skinner, Barrack, & Cook, 1984; Stelmach, Zelaznik, & Lowe, 1990; Teasdale, Bard, LaRue, & Fleury, 1993). Deterioration in labyrinthine detection has been particularly noted (Peterka & Black, 1990; Peterka, Black, & Schoenhoff, 1990a, 1990b; Rosenhall, 1973; Sloane, Baloh, & Honrubia, 1989). Anatomic studies have revealed a gradually decreasing density of labyrinthine hair cell receptors beginning at age 30, and a steeper decline in the number of vestibular receptor ganglion cells beginning around 55–60 years of age (Bergstrom, 1973; Richter, 1980; Rosenhall, 1973). Mechanical changes in the elderly have been observed in the elasticity of their muscles and freedom of movement of their joints, which could alter the natural damping properties of their joints (Aoyagi & Shephard, 1992) and require a change in the coordination of their muscle actions. Indeed, lower limb stabilizing muscle synergies, in response to postural perturbations, have been shown to exhibit a disorganized order of onset in the elderly (Woollacott, 1986, 1993).

Correspondence address: Emily Keshner, Sensory Motor Performance Program, Room 1406 - E809, Rehabilitation Institute of Chicago, 345 East Superior Street, Chicago, IL 60611. E-mail address: eak@nwu.edu

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In the experiments reported here, we examined the frequency characteristics of mechanisms controlling stabilization of the head in the vertical plane in an elderly population. In a previous study of young adults, Keshner, Cromwell, and Peterson (1995; Keshner & Peterson, 1995) found that neural and mechanical control mechanisms participated in producing stabilization of the head when the trunk was randomly rotated in the horizontal and vertical planes. Young adult subjects relied on reflex mechanisms to smooth the transition from control by voluntary mechanisms to that of system mechanics at rotational frequencies above 1.5 Hz. In the vertical plane, reflex mechanisms participated even at rotational frequencies below 1 Hz, a level at which voluntary mechanisms optimally operate (Keshner et al., 1995). Comparison of data from these experiments with those of previous studies of young adult subjects should begin to identify age-related changes in the control of head movement. These changes potentially underlie the increased incidence of instability and falling found in the elderly population (Tinetti & Speechley, 1991).

**Method**

**Subjects**

The equipment and methods used here were identical to those used in a previous experiment by Keshner et al. (1995). Six women and 4 men, ranging in age from 65 to 88 years, gave informed consent to participate in this study. For analysis purposes, we divided the subjects into two groups: young elderly (YE), aged 65 to 73 years, and old elderly (OE), aged 79 to 88 years. Their data were compared with previously published data from 8 young adult (YA) subjects (Keshner et al., 1995). All subjects had no current medical complaints and no history of neurological disorder, falling, or postural instability. All of the elderly subjects were active on a daily basis, performing their activities of daily living (e.g., shopping, gardening, housecleaning) without assistance. Three of the subjects (aged 67, 87, and 88 years old) participated in exercise classes for at least 30 min, 3 days per week. One subject (aged 87 years old) was legally blind in one eye.

**Equipment**

Subjects sat in a rigid chair that provided support to the entire body. Their legs were extended, with a slight bend at the knees, and their feet were supported at approximately the same elevation as the seat. Subjects were securely fastened in the chair with shoulder and lap belts, support cushions, and a chest-level metal gate. Pilot studies comparing trunk and chair angular velocity during chair rotation confirmed that movement of the trunk relative to the chair was insignificant at all test frequencies. Thus, all measures of chair velocity and position were considered to be equivalent to trunk velocity and position. The chair was coupled to a high-torque (500 ft lb), servo-controlled rotatory turntable (Neurokinetics, PA), and we aligned the axis of rotation of the chair with the earth-horizontal axis to produce rotation of the chair in the vertical plane. Subjects were positioned in the chair so that the interaural axis of the head was aligned with the axis of rotation of the chair. The entire apparatus was located in a room that provided complete darkness during the experiment.

A triaxial angular rate sensor (Watson Industries, WI) was attached to a helmet that sat firmly on the subject's head. The sensor on the helmet was positioned next to the temporal lobe, with one of its axes parallel to the interaural axis of the head. The head was free to rotate in all three planes of motion, but measured angular velocity responses of the head indicated that motion was significant only in the pitch plane. Attached to the other side of the helmet was a portable laser pointer, providing a head-referenced visual target. The entire helmet apparatus weighed less than 950 gm. The mass of the helmet was evenly distributed around the circumference of the head, and the center of mass of the helmet was positioned close to the center of mass of the head; thus, when the helmet was worn, the center of mass of the head remained constant. Another laser was mounted on a tripod and provided an earth-fixed visual reference target.

Bilateral electromyographic (EMG) activity from semispinalis capitis (SEMI) and sternocleidomastoid (SCM) muscles were detected by silver–silver chloride electrodes, 4 mm in diameter, spaced 1 cm apart. SEMI was palpated while subjects performed pitch extension with rotation of the head; electrodes were placed on the muscle belly approximately 2 cm below the occipital bone at C1–C2 and 2 cm lateral to the midline. SCM was palpated during contralateral rotations of the head against resistance placed at the chin. Electrodes were placed on the muscle belly approximately one third of its length rostral to its sternal attachment. EMG electrode locations have been verified anatomically and physiologically by a previous study of isometric head stabilization (Keshner, Campbell, Katz, & Peterson, 1989). In our elderly subjects, the skin of the neck tended to fold loosely over the neck muscles, thus presenting greater difficulty in obtaining robust EMG signals. Data were excluded if signal loss was observed in the EMG signals.

**Procedure**

At the beginning of each trial, the chair was positioned such that the subject's trunk was aligned with the earth-vertical plane. Three experimental paradigms were presented, each consisting of five, 40-s-duration trials. Subjects did not report any discomfort, fatigue, or loss of effort in following instructions during the experiment. The mental arithmetic (MA) paradigm was tested first. Subjects were asked to perform a mental task while being rotated in the dark with lasers turned off (tasks, for example, included recall of geographical sites and names of animals). We used the tasks to distract the subject from any voluntary control of head stabilization so that compensatory head movement with respect to the trunk would be the result of reflex and mechan-
cal control. In the visual stabilization (VS) paradigm, subjects were instructed to keep the head-referenced laser target aligned with the earth-stationary laser target while the body was rotated. Before each trial, the head-referenced and the earth-stationary lasers were positioned on the same spot, approximately 1.6 m in front of the subject. By continuously matching the two laser positions, subjects kept their heads stabilized in space, using all available feedback pathways. In the no-vision (NV) paradigm, subjects were rotated in the dark but were instructed to imagine that both lasers were turned on and to match the positions of the imagined targets. In this paradigm, compensation of the head with respect to the trunk occurred in the absence of visual inputs.

Stimulus

A random rotational stimulus was produced by using a sum-of-sines (SSN) waveform consisting of five harmonics having no common divisors but with a common fundamental frequency. The frequency response of the chair to the SSN command signal was composed of the harmonics 7, 15, 29, 43, and 61 of a fundamental frequency of 0.05 Hz, resulting in a frequency range of 0.35 to 3.05 Hz. The velocity response of the chair was 10°/s at 0.35 Hz, 12°/s at 0.75 Hz, 10°/s at 1.45 Hz, 4°/s at 2.15 Hz, and 2°/s at 3.05 Hz. Peak excursion from the vertical was ±12°, and mean excursion was 0°.

Data Collection and Reduction

Data collection for all trials in all paradigms began after one 20-s cycle of stimulation and ended immediately after the next 20-s cycle. Chair (trunk) angular velocity, chair (trunk) position, head angular velocity, and four EMG signals (bilateral SEMI and SPL) were recorded. Kinematic signals passed through an anti-aliasing, 8-pole, low-pass Bessel filter (Frequency Devices, MA) with a 20-Hz corner frequency. The filtered signals were digitally sampled at 50 Hz and stored on a PDP 11/23 computer for later analysis. Analog EMG signals were electronically amplified, bandpass filtered (10–200 Hz), full-wave rectified, and integrated (20-ms time constant) before anti-aliasing filtering and digital storage.

Data Analyses

To determine the response properties of the chair, head, and muscles for each component sine wave of the SSN stimulus, we performed spectral analysis on the recorded signals (head angular velocity, chair angular velocity, chair position, and integrated EMG responses) by using a fast Fourier transform (FFT). The FFT characterizes the response of the signals at discrete frequencies by response vectors with real and imaginary components. Individual subjects’ response variability between trials in each paradigm were small; thus, we used the average of the response vectors of the five trials in each paradigm to characterize a subject’s response in each paradigm. To describe these responses in a more tangible way, we calculated the amplitudes (response vector length) and phases (response vector orientation) of the signals at the discrete frequency components of our SSN.

Chair (trunk) velocity was the input stimulus. Responses of the head with respect to the trunk were defined as changes at the neck. The angular velocity response of the neck was calculated at each SSN component frequency from the vector difference between the head and chair angular velocity response vectors. Results are described as gains and phases of the neck with respect to the trunk. Gain of the neck with respect to the trunk was equal to the neck–chair response vector amplitude ratio; phase of the neck with respect to trunk was equal to the difference between the neck and chair response vector phase angles. A phase of −180° indicated perfect directional compensation of the neck to the chair velocity stimulus so that the head moved in the direction opposite to that of the chair. Phases of 0° indicated that the neck was not compensating and the head was moving in the same direction as the chair. Phase lead of the neck with respect to the trunk was defined as phases between 0° and −180°. Phase lag of the neck with respect to the trunk was defined as phases between −180° and −360°. A gain of 1 and phase of −180° represented perfect compensation of the head (or no movement of the head relative to space).

In a previous study (Keshner et al., 1995) the linearity of the response was tested at similar frequencies and with increasing amplitudes. Even with a four-fold increase in amplitude (from ±5° to ±20°), response gains and phases did not change significantly with stimulus amplitude. Because this amplitude difference was much greater than the differences existing between the two experimental paradigms, we believe that we have been working within a linear range of head–neck responses in this study. Coherence between the neck (head minus trunk) velocity and chair velocity was calculated at each stimulus frequency. In VS, 67% of the responses had a coherence greater than 0.9, and 86% of the coherences fell between 0.7 and 1.0. In NV, 79% of the responses had a coherence greater than 0.9, and 88% of the coherences fell between 0.7 and 1.0. In MA, 83% of the responses had a coherence greater than 0.9, and 96% of the coherences fell between 0.7 and 1.0. These results are consistent with the interpretation of the EMG responses, analogized in a descriptive fashion only, comparing the response of a muscle from each subject across paradigms. Interpretation of individual subject across paradigms. Interpretation of the experimental condition.
1989), and are presented with respect to peak chair position or velocity (Keshner et al., 1995). Thus, response phases around +90° corresponded to peak position backward of the chair (peak extension position of the trunk). Phases of −90° corresponded to peak forward chair position (peak flexion position of the trunk). A muscle response phase of 0° indicated that the muscle was responding in phase with peak forward (trunk flexion) velocity of the chair; a −180° phase indicated a response related to peak backward (trunk extension) velocity of the chair. We collected bilateral muscle EMG signals, but in some subjects there was variability between the response of the right and left muscles. We attributed this to the difficulty in placing and maintaining electrode position during the experiment resulting from slackness of the skin around the neck in the elderly subjects. We always selected the muscle with the greater coherence to the stimulus and higher gains to describe the response.

In the data from the young adult subjects presented here, the angular velocity response of the neck with respect to trunk at each test frequency was calculated as the vector average for the group in each paradigm (Keshner & Peterson, 1995; Keshner et al., 1995). Vector averages for this population of subjects were derived by averaging the real and imaginary components of the FFT of each subject’s responses at each SSN component frequency and then calculating the gain and phase of the average response vector. Because vector averages reflect multiple components, they are not equivalent to statistical averages. Instead, they reflect the two-dimensional (size and orientation) behavior at each frequency. Standard errors of the data were derived by calculating the error vector, composed of standard errors of the real and imaginary components of the response vectors at each stimulus frequency. Gain error was defined as the amplitude of the error vector. Phase error was defined as the difference between the orientation of the error vector and the response vector.

Descriptive statistics are adequate for quantitative comparison of responses described by two interactive characteristics such as gain and phase. Thus, to test whether categorizing population responses by age was appropriate in this experiment, we calculated the scalar distance between the neck vectors of each subject within a group. We entered scalar distances into a Kruskal–Wallis analysis of variance test to determine if, at each frequency of rotation, differences between the age groups were greater than intersubject variability. In other words, were differences between the population’s responses to each paradigm too large to be accounted for by the inherent variability between subjects? In VS and MA, significant differences between the age groups (p < .05 level) emerged at the two highest frequencies (2.15 Hz and 3.05 Hz). In NV, significant differences were present for all but the lowest frequency of rotation (0.35 Hz), thus confirming the decision to independently analyze the population’s responses. Significant effects of each experimental paradigm on each population were examined for each rotation frequency through a 3 x 5 repeat-measures analysis of variance performed on the scalar difference between each subject’s neck vectors in each paradigm. Summary post hoc analyses, using a Tukey–Kramer multiple comparisons test, were performed on these data; statistical significance was accepted at the p < .05 level.

Results

Head Kinematics of Young Elderly Subjects

In Figure 1, the neck gains with respect to the trunk and phases of the YE subjects are plotted. Gains of 1 and phases of −180° indicated good compensation for the chair rotations; thus, the subjects effectively stabilized their heads in space. Although variability among the subjects was evident, the pattern of responses across the frequency range (illustrated by dashed lines between data points) was mostly similar. In the VS paradigm, 2 of the YE subjects (aged 69 and 73 years old) exhibited gains between 0.7 and 1.0 and phases close to −180° at frequencies below 1 Hz. The 65- and 67-year-old subjects had lower gains, but phases close to −180° suggested that these subjects, like the other 2, were compensating for the trunk rotation. The 70-year-old subject showed poor compensation at these low frequencies, with very low gains and phases that lagged the rotation of the trunk. Between 1–2 Hz, compensation was still present in most of the subjects, although the gains of the 67-year-old subject began to drop. The 70-year-old subject continued to deviate the most from the group, with a steep rise in gain and phase. At 2.15 Hz, phases scattered away from compensation. Response gains exhibited a steep ascent, except in the 67-year-old subject. Above 3 Hz, some of the gains were greater than 1, indicating that the head was moving at velocities greater than the chair, as would occur if the system were mechanically resonating. We have observed this phenomenon in our young adult subjects during trunk rotations in the horizontal plane (Keshner & Peterson, 1995) at frequencies above 3 Hz. Only the 67- and 69-year-old subjects were able to restrain their heads from this resonant activity, producing instead the gain drop and phase shift observed prior to the resonant peak response in young adult subjects. The early fall off in gains and phases followed by a steep ascent in the 70-year-old subject would suggest that this individual could not successfully complete the task of head stabilization in this paradigm.

Although gains were somewhat lower when vision was removed from the task of stabilizing the head (NV), across the frequency range, the ability to compensate for the trunk rotations appeared to improve in most of the subjects. Below 1 Hz, gains fell between 0.4 and 0.6 for 3 subjects and around 0.2 for 2 subjects. Although phases were very scattered at the lowest frequency, by 0.75 Hz all but the 65-year-old’s phases were close to −180°, indicating that the subjects were successfully compensating for trunk rotation (although with very small velocities). Between 1–2 Hz, gains increased, whereas most phases stayed near −180°. At 2.15 Hz, the gains were tightly clustered between 0.7 and 1.0 and most of the phases were close to −180°, indicating
improved compensation at this frequency. Mechanical resonance became apparent above 3 Hz, with a rise in gains above 1.0 for 3 of the subjects and a steep drop off in all phases.

At the lowest frequency in MA, all but 2 of the subjects (the 65- and 67-year-old) exhibited gains between 0.3 and 0.8, with phases close to -180°. Compensation improved as the gains of these 3 subjects moved closer to 1, and phases stayed mostly near -180° (the 69-year-old subject was somewhat variable) between 1 Hz and 2.5 Hz. Except in the 69- and 73-year-old subjects whose gains plateaued, steeply ascending gains and descending phases above 3 Hz indicated the presence of a resonant response of the head. The 65- and 67-year-old subjects exhibited ascending gains, with phases starting around -90° and 0°, respectively, and steeply descending throughout the frequency range. This response pattern was similar to that observed in a deeply anesthetized animal (Goldberg & Peterson, 1986) and would suggest that, for these 2 subjects, neural influences were absent in this paradigm.

**Head Kinematics of Old Elderly Subjects**

Responses to VS in the OE group were not markedly different from the YE in 4 of the subjects. Except for 1 subject (aged 79 years old) whose low frequency gain was too low to meaningfully interpret the phase response, most subjects were producing phases near compensatory -180°, with gains between 0.2 and 0.8 (Figure 2) at the lowest frequency. In these subjects, attempts at compensation continued below 2 Hz; above 2 Hz, the steep phase shifts and gains greater than unity were indicative of resonant responses of the head. The 79-year-old subject demonstrated poor compensation throughout the frequency range, exhibiting the continuous rising gains and decreasing phases observed in a

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**FIGURE 1.** Bode plots of neck velocity relative to trunk velocity from 5 YE subjects in three paradigms (see text for abbreviations) elicited during a random sum-of-sines stimulus (0.35 to 3.05 Hz) in the vertical plane. Each plot symbol represents a different subject. We have drawn dashed lines connecting each symbol to better portray the response pattern across frequencies. Responses with phases at -180° indicate head motion that compensates for the trunk perturbation. Responses with phases from -180° to 0° lead compensation; those with phases between -180° and +360° lag compensation; those with phases at 0° indicate head motion in the same direction as the trunk (see labels on lower right plot).
model of a passive mechanical system in the cat head and neck during horizontal rotations (Goldberg & Peterson, 1986). Only the oldest subject (aged 88 years old) was able to delay resonance. Although the gains of this subject were diminishing, phases were maintained close to $-180^\circ$ below 3 Hz.

Response gains of the OE in NV tended to be lower than those of the YE, and the steep drop in phases occurred at a lower frequency. The 79- and 83-year-old subjects exhibited the slowly rising gains observed when there was no neural compensation for the stimulus (Goldberg & Peterson, 1986), but the steady phases below 2 Hz demonstrated that these subjects were attempting to compensate for the trunk rotations. Above 2 Hz, phases began to drop, suggesting that the subjects could no longer compensate for the rotations. Two other subjects (the 87- and the 88-year-old) kept their gains between 0.2 and 0.4 across the frequency range. Low gains coupled with steeply declining phases suggest poor compensation by the 87-year-old subject. The 88-year-old subject managed to attain compensatory phases below 1 Hz that slowly dropped off as frequency increased; stable gain responses in this subject suggested continued attempts at compensation in this paradigm. Only the 82-year-old subject exhibited gains close to 1.0 and phases close to $-180^\circ$ at frequencies below 1 Hz. As frequencies increased, the gains of this subject dropped off between 1–2 Hz and then begin to rise, implying a shift from voluntary to mechanical control over the response (Keshner & Peterson, 1995).

None of the OE subjects exhibited compensatory phases below 2 Hz in MA. In fact, they produced the slowly ascending gains and descending phases expected from a system that was not neurally compensating for the rotational disturbance (data from the 88-year-old subject were lost because of a faulty transducer). Between 2 and 3 Hz, the 79- and 82-year-old subjects produced gains close to 1 and phases close to $-180^\circ$, suggesting that some neural compensation was taking place in this frequency range. The gain
FIGURE 3. Bode plots of the 70-year-old subject's right sternocleidomastoid and right semispinalis muscle EMG responses in the three paradigms. Gain and phase responses are plotted with respect to trunk velocity. Gains correspond to the uncalibrated envelope of the rectified EMG amplitude per °/s trunk velocity. Phases corresponding to peak forward and backward position and velocity with respect to chair rotation are identified in the lower right plot.

FIGURE 4. Bode plots of the 87-year-old subject's right sternocleidomastoid and right semispinalis muscle EMG responses in the three paradigms. Gain and phase responses are plotted with respect to trunk velocity. Gains correspond to the uncalibrated envelope of the rectified EMG amplitude per °/s trunk velocity. Phases corresponding to peak forward and backward position and velocity with respect to chair rotation are identified in the lower right plot.
drop in the 87-year-old subject and the continued increase in gain beyond 1.0 in the 83-year-old subject suggested poor control of the response, even though the phases were not far from -180°. Above 3 Hz, the scattered gains and phases were similar to the response pattern observed when neural control of stabilization was failing (Keshner et al., 1995). One difference between OE and YE seemed to be that the OE showed phases tightly grouped at all but the highest frequency. The YE showed phases that tightly grouped only at the two highest frequencies.

**Muscle EMG Responses**

Figures 3 and 4 show typical muscle EMG responses of the 70- and 87-year-old subjects, respectively. Apparent in each figure are EMG response gains that decreased from their initial value as rotation frequencies climbed to 1 Hz, then increased at frequencies above 1 Hz to values greater than or equal to the low-frequency gains. This U-shaped pattern emerged for almost every subject across all of the paradigms and is a typical response of the neck muscles to random rotations of the trunk (Keshner et al., 1995). EMG output at low frequencies reflected the kinematic differences observed in each paradigm, with response gains that were more often greater in VS than in NV and MA. At high frequencies in SEMI, EMG response gains were similar for all three paradigms.

Two general response strategies could be detected in the phase of the EMG responses. Most often, the neck extensor muscle (SEMI) was matched to the forward position of the chair (-90°) at low frequencies and advanced to a phase matched to forward velocity (0°) or backward position (90°) at high frequencies (Figures 3 and 4). Thus, the muscle was acting to extend the neck so that the head would be held upright in response to the forward motion of the trunk at low frequencies, and was leading this forward motion at the higher frequencies. A 90° phase advance from position to velocity as frequency increases is characteristic of control by the vestibulocollic reflex in decerebrate (Baker, Goldberg, & Peterson, 1985; Bilotto, Goldberg, Peterson, & Wilson, 1982) and alert (Goldberg & Peterson, 1986; Keshner, Baker, Banovetz, & Peterson, 1992) cats, and has been consistently observed in our young adult subjects (Keshner et al., 1995). The flexor muscle (SCM) responded reciprocally, flexing the neck to hold the head upright as the trunk moved backward. SCM response phases were matched to backward position (90°) of the chair at low frequencies and advanced toward backward velocity (180°) as frequency increased. This reciprocal activation pattern was consistently exhibited by 3 YE (the 67-, 70-, and 73-year-olds) and 3 OE (the 83-, 87-, and 88-year-olds) subjects.

The second response strategy did not necessarily appear in all paradigms. It was observed most often in the phase response of the SEMI muscle as activation was matched to backward chair position at low frequencies and advanced toward backward chair velocity in conjunction with the SCM muscle as frequency increased. In Figure 5, comparison of muscle response gains in the NV and MA paradigms with those in VS (where reciprocal activation occurred in this subject) indicated that activation of the antagonist neck muscles in the same direction of trunk rotation was not associated with an increase in EMG response gains. Thus, we cannot assume a significant stiffening of the neck nec-
Comparison With Response Patterns of Young Adult Subjects

Comparison of data from elderly subjects with that from young adult subjects revealed a definite effect of age on the subject's ability to stabilize the head, particularly when combined with an absence of vision or with cognitive distraction. Data presented in Figure 6 are from our previous study of young adult subjects rotated in the vertical plane, with the same stimulus and paradigms used in these experiments (Keshner, Cromwell, & Peterson, 1995). The young adult subjects compensated very well in the voluntary stabilizing paradigms (VS and NV) at frequencies below 2 Hz, as demonstrated by phases of $-180^\circ$ and average gains around 0.5–0.7. Only 1 OE and 2 YE subjects attained such high gains in VS, with phases solidly placed at $-180^\circ$; none of the elderly subjects compensated as well in the NV paradigm. Above 2 Hz, gains of the young adults dropped steeply and the compensatory phases were lost, but unlike the elderly subjects, resonance of the head (as defined by gains greater than unity) was never observed.

Response gains of young adult subjects in MA at 0.35 Hz were below 0.5 and varied widely, but the presence of phases around $-180^\circ$ for all but 2 of the subjects indicated the

![Diagram](image_url)

**FIGURE 6.** Upper half. Bode plots of average neck velocity relative to trunk velocity from 8 young adult subjects in the three paradigms. Neck velocity gain and phase values, elicited during a random sum-of-sines stimulus (0.35 to 3.05 Hz) in the vertical plane, were computed relative to chair velocity. The solid line is the vector average for the group; the shaded area is ±1 standard error of the average. Conventions are the same as in Figure 1 and are stated in the lower right phase plot. Lower half. Bode plots of 1 young adult's right sternocleidomastoid and right semispinalis muscle EMG responses in the three paradigms. Gain and phase responses are plotted with respect to trunk velocity. Gains correspond to the uncalibrated envelope of the rectified EMG amplitude per % trunk velocity. Phases corresponding to peak forward and backward position and velocity with respect to chair rotation are identified in the lower right plot.

*Note.* Portions of this figure are reprinted from an article, “Characteristics of Head and Neck Stabilization in Two Planes of Motion” (Figure 4, p. 91), by E. A. Keshner, G. Peng, T. Hain, and B. W. Peterson, in *Multisensory Control of Posture* (T. Mergner & F. Hlaváček, Eds.), 1995. New York: Plenum Press. Copyright 1995 by Plenum Press.
presence of compensatory activity even in the low frequency range. This low frequency compensatory activity was present in 3 of the YE subjects, but was definitely absent in all of the OE subjects. Young adult subjects exhibited increasing gains and phases clustered more closely about −180° as frequency increased from 1–3 Hz. The YE exhibited increasing gains, but phases did not group near −180° until frequencies between 2–3 Hz were reached. OE subjects exhibited scattered gains with phases near −180° only at frequencies at 2.15 Hz. Above 3 Hz, variability in gains and phases increased for the young adults, but phases were still close to compensatory and did not exhibit the large drops seen in VS and NV. YE subjects did exhibit larger phase drops along with resonant responses of the head. In the OE, phases were very scattered, although gains stayed mostly near 0.5.

Variance between the neck velocity response vectors in the three paradigms produced differing patterns of significance for the three populations. The young adult subjects consistently exhibited significantly greater differences between the VS and MA paradigms than between the VS and NV paradigms across all rotation frequencies (p < .05), confirming the minimal effect of removing vision from this population. At all but the lowest frequency, the YE subjects exhibited significant differences between all comparisons, so VS and MA had significantly greater differences than those observed between VS and NV and between NV and MA. In addition, the YE exhibited significantly greater differences between VS and NV than between NV and MA (p < .05), implying a stronger effect of removing visual feedback from their attempts to produce a head-stabilizing response. OE subjects were the least specifically affected by the different paradigms. Although the analysis of variance suggested the presence of significant differences between paradigms (p < .05), individual comparisons revealed only spurious results. The direction of the responses, which may have been strengthened with a larger sample size, would suggest that this group was demonstrating smaller differences between VS and NV than between the other comparisons.

A typical response of the neck muscles in the young adults is presented for 1 subject at the bottom of Figure 6. Muscle EMG response gains of young adult subjects exhibited the U-shaped pattern of response. SEMI EMG activity was slightly lower in MA than in the other paradigms at frequencies below 1 Hz, but above 2 Hz, no differences were observed. As with most of the elderly subjects, muscle EMG responses were reciprocally activated and began with phases close to trunk angular position. By 1.5 Hz, both muscles advanced to a phase related to velocity. Directional coactivation of the neck muscles was not observed in this population of young adult subjects.

**Discussion**

Even with the small sample sizes, age emerged as an influential factor in the control of head stabilization, and decrements in performance were even more pronounced in the older as compared with younger elderly subjects. Age effects could be seen in (a) the diminished ability to voluntarily stabilize the head, particularly with the absence of vision, (b) the impaired ability to stabilize the head when mentally distracted, and (c) the appearance of a resonant response in some elderly subjects.

Previous studies of young adults performing head stabilization in the horizontal plane (Keshner, Cromwell, & Peterson, 1995) and sagittal plane (Keshner et al., 1995) during random rotations of the trunk determined that the control of head stabilization by voluntary mechanisms was predominant at frequencies below 1 Hz. Lower gains and compensatory phases in the mental arithmetic paradigm suggested that head stabilization was supported by the vestibulocollic and cervico-collic reflexes at these low frequencies. As frequencies increased from 1–3 Hz, rising gains and advancing phases of EMG output, which almost certainly arise from reflex dynamics like those observed in decerebrate cats (Bilotto et al., 1982; Peng, Hain, & Peterson, 1995; Peterson, Goldberg, Bilotto, & Fuller, 1985), appeared to allow reflex mechanisms to damp the responses of the head as voluntary mechanisms began to fail. Thus, the reflex pathways operated independently of, but were complementary to, the voluntary pathways (Forsberg, Grillner, & Rossignol, 1975; Keshner, Baker, Banovetz, & Peterson, 1992). Above 3 Hz, large phase errors were observed in the VS and NV paradigms, whereas responses in MA remained close to compensatory, implying that voluntary efforts to stabilize the head could be counterproductive at higher frequencies.

Elderly subjects were not always capable of voluntarily stabilizing the head, either. Compensatory phases below 1 Hz indicated that the subjects were correctly sensing the direction of trunk rotation and correctly responding with a response of the head in the opposite direction in order to keep the head stable in space. Gains that were less than 1, however, indicated an error in the velocity output of the system. The low, constant gains could indicate a velocity control saturation, where the subject’s maximum voluntary velocity may be less than the velocity input to the trunk. A decrease in muscle strength in the elderly (Whipple, Wolfe, & Amerman, 1987) may explain the inability to move the head at the necessary compensatory velocity. Low gains could also reflect a control strategy that limits the head velocity. Subjects may not feel comfortable moving their heads at higher speeds. Or, a sensory deficit could be implicated, as the response may be interpreted as perfect compensation to a perceived velocity that is proportionally less than the actual velocity.

There is a great deal of literature charting the deficiency of sensory signals in the elderly (Bergstrom, 1973; Peterka & Black, 1990; Peterka et al., 1990a, 1990b; Rosenhall, 1973; Skinner et al., 1984; Sloane et al., 1989; Stelmach et al., 1990), and the amount of undercompensation in gain might be related to the severity of these sensory deficits.
Increasing severity of subtle sensory deficits might also explain why the absence of visual feedback in the NV condition produced fewer compensatory phases in the younger elderly, and reduced gains coupled with phases that were not compensatory in the older elderly subjects. Studies of postural stabilization have emphasized the importance of visual feedback in an elderly population. Quiet sway was observed to increase in the elderly with visual deprivation, and was reduced by about 50% when vision was restored (Pyykkö, Pirkko, & Aalto, 1990; Sheldon, 1963). But the removal of vision alone is rarely destabilizing. Elderly subjects have been found to have much more difficulty balancing than young adults when vision was removed in the presence of an unstable support surface (Manchester, Woollacott, Zederbauer-Hytn, & Marin, 1989; Teasdale, Stelmach, & Breunig, 1991; Whipple, Wolfson, Derby, Singh, & Tobin, 1993; Woollacott 1993). It has been suggested that the visual system may serve as a modulator rather than as a generator of stabilizing responses (Timmann, Belting, Schwarz, & Diener, 1994), fitting in with our findings of compensatory responses that are diminished but not lost.

The increased appearance of the resonant pattern and the greater loss of compensatory responses in the elderly as compared with younger subjects at frequencies above 3 Hz in MA suggests that reflex mechanisms were also not fully capable of compensating for the rotational disturbance. It is possible that the additional mass of the helmet had a greater effect on the weaker head–neck system of the elderly subjects, thereby decreasing compensation. Despite the diminished neural contribution during MA, however, compensatory phases and gains re-emerged in most of our elderly subjects at 2.15 Hz. This frequency falls within the range at which reflex control dominates the head-stabilizing response in young adults (Keshner & Peterson, 1995; Keshner et al., 1995). It is also a particularly functional frequency, because frequencies of 1.5–2.5 Hz have been observed at the head during natural locomotion movements (Grossman, Leigh, Abel, Lanska, & Thurston, 1988; Keshner, Cromwell, & Peterson, 1992). It may well be that at this functional frequency, the system is heavily weighted toward a reflex mode of control and is not reliant upon a summation of sensory signals to elicit the influential control mechanisms.

Some subjects presented what appeared to be directional coactivation of the two neck muscles, the functional result of which should be an increased stiffness of the neck. If the neck were indeed held stiff, gains of zero and phases of 0° should appear in the neck with respect to the trunk response and the resonant frequency should increase. Although such gain and phase parameters were exhibited by the 79-year-old subject at the lowest frequency in the MA paradigm, none of the other subjects that exhibited directional coactivation achieved this stiffening response across the frequency range. Instead, in NV and MA, these subjects produced responses reminiscent of a passive mechanical system (Keshner et al., 1987; Viviani & Berthoz, 1985).

Both passive (i.e., inertial and viscoelastic properties) and active biomechanical factors (i.e., strength and range of motion) contribute to head stabilization at all times. Passive, mechanical properties of the head have been found to exhibit characteristics of a linear, second-order, underdamped system (Viviani & Berthoz, 1985). Such a system will have head response gains increasing with a 40-dB/decade slope with constant phases at frequencies below the resonant frequency of the system. At the resonant frequency, gains will reach a maximum, above 1, and phases will sharply drop toward −180°. In our subjects, the response pattern consisted of a slow rise in gain associated with slowly decreasing, noncompensatory phases as frequency increased, suggesting little to no control by the neural pathways (see, for example, the 65-year-old and 83-year-old in MA, Figures 1 and 2, respectively). It is possible that increasing stiffness was the intent of the coactivation pattern, but either the muscles were not strong enough or the neural pathways were not capable of producing the necessary muscle forces (Keshner et al., 1993). We might hypothesize that subjects who did not attempt coactivation were more reliant upon system mechanics to maintain stability, either because of the failure of neural mechanisms to properly control the response or because these subjects chose not to intervene with the natural mechanical compensation for a destabilized head.

Our data suggest that the control mechanisms for head stabilization are strongly affected by the natural aging process. Changes in the integrity of the sensory systems appear to shift an individual’s reliance from reflex mechanisms to system mechanics. Both active (strength) and passive (viscoelastic) properties of system mechanics change with age (Keller, Spengler, & Hansson, 1987; Whipple et al., 1987); thus, the system is no longer able to effectively compensate for disturbances, and increased instability in the aged population results. Of course, our results depended greatly upon the subject’s willingness to follow our instructions and to attend either to head stabilization or to the mental distraction task. It is possible that the decreased cognitive capacity in the elderly interfered with their ability to perform the mental distraction task equitably, which may explain why these responses were similar to those in NV. It may also be that the absence of a visual target removed any motivation to attend to the stabilization task, making NV more like MA. Finally, although these paradigms demonstrate the integrity of the head-stabilizing system, the task of stabilizing the head in order to focus on a visual object in a natural environment during combined head–trunk motions may produce very different results.

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REFERENCES


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