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Labyrinthine contribution to static equilibrium.

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LABYRINTHINE CONTRIBUTION TO STATIC EQUILIBRIUM

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Abstract

13 anthropometric body measures (plus 20 measures derived from the original 13) were obtained from 77 normal male Ss who were also tested for 4 min on a standing sway apparatus. A power spectral density (PSD) analysis was applied to each resulting sway waveform. Multiple regression prediction equations were computed between the 33 body measures and power values at 13 separate frequencies from the PSD curves. It was found that the body characteristics measured in this experiment could consistently account for approximately 55% of the variability in the PSD sway patterns. The prediction equations derived from these 77 Ss were then used to predict the mean PSD sway curve of 6 labyrinthine defective (L-D) Ss solely from knowledge of the group's mean body measures. The predicted (or control) curve was then compared to the actual curve obtained from the 6 L-D Ss. It was found that the L-D Ss sway significantly more ($p < .05$) at the lower sway frequencies (i.e., 0.15 Hz to 0.40 Hz) than that predicted for these Ss. One possible physical model for standing sway was proposed and a transfer function was derived for this model.

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The human vestibular or labyrinthine sensory system refers to the non-acoustic section of the inner ear and is located just dorsal to the cochlea or acoustic portion of the inner ear. The labyrinth is generally assumed to be a receptor for control of both static and dynamic aspects of the postural mechanisms that are involved in equilibrium.

The inner ear consists of a series of membranous canals lying within corresponding hollow sections (the bony canals) of the spongious or petrous portion of the temporal bone (Anson, 1967; Davies and Davies, 1962). The membranous canals are filled with endolymphatic fluid which bathes the true receptor organs of both the labyrinthine and cochlear portions of the inner ear. The space between the membrane and the bony wall is filled with perilymphatic fluid and supporting connective tissue.

The labyrinth is anatomically, and perhaps functionally divided into two major areas. The most dorsal area is comprised of the three semi-circular canals (superior, posterior, and horizontal) which lie in three mutually orthogonal planes. These canals respond maximally to angular accelerations in their respective planes as has been shown by direct observation (Dohlman, 1944; Steinhausen, 1933), by electrophysiological evidence

(Ross, 1936; Lowenstein and Roberts, 1949; Gernandt, 1950; Adrian, 1943), and by theoretical considerations (Summers, et al. 1943). In this capacity the canals serve as centers that regulate kinetic equilibrium by sending impulses to proprioceptive areas in the CNS. Signals from the canals also control movements of the eyes so that visual fixation is unimpaired during changes in the position of the head or body (Ewald, 1892).

The vestibule is the second area of the membranous labyrinth and is composed of two closely related structures, the utricle and the saccule. The receptor portion of the utricle (utricular macula) is situated on the floor of the utricle with its most anterior aspect extending up on the anterior wall of the membranous utricle. This extension, then, lies in a plane nearly perpendicular to the floor of the utricular macula. Fischer (1956, p. 15) has named this portion of the macula, the macula utriculi accessoria. The shape of the utricular macula, for a human subject standing upright, can best be represnted by extending the hand forward in a horizontal plane with the palm up and the fingers slightly curled upward (Jongkees, 1967).

The receptor organ of the saccule, the saccular macula, lies medial to the utricle and at right angles to the plane of the floor of the utricular macula. It thus runs parallel and vertical to the more horizontally positioned utricular macula.

The epithelial layer of the macula of the utricle contains the receptive hair cells and the supporting cells. The hairs of the hair cells are embedded within an over-lying gelatinous mass containing the small otolithic crystals. The whole anatomical structure of the macula system makes it ideally suited for detection of tangential stimulations (i.e., forces parallel to the surface of the macula) rather than to perpendicular stimulations (Jongkees, 1967). When subjected to tangential stimulations, the heavier overlying gelatinous mass, suspended within the endolymphatic fluid, tends to remain in position while the base of the hair cells move with the body to which they are attached. This action produces a shearing force on the embedded hairs and bends them. The bending of the hairs is thought to be the adequate proximal stimulus for these receptor organs. The external, or distal, stimulus which produces these forces is linear acceleration. Such linear acceleration forces are produced either by changes in head position relative to gravity, by centrifugal accelerations, or by linearly accelerated or progressive movements. All of these conditions, when translated into the proximal stimulus, behave as identical forces (Jongkees, 1967).

The function of the saccular macula, although structurally identical to the macula of the utricle, is still not completely known. Several investigators (McCabe and Lawrence, 1958; Ashcroft and Hallpike, 1934; Ross, 1936) have proposed that the saccule may function in receiving acoustic stimuli although a number of recent studies have strongly supported the notion that the saccule, like the utricle, is important for maintainance of equilibrium, especially in its influence on head and eye movements (von Bekesy, 1966; Jongkees, 1950; Lowenstein and Roberts, 1948).

The reflexes that are initiated from the otolithic organs in response to linear movements act to maintain or establish erect posture. Such reflex responses as the vestibular placing reaction (Bard, 1937), righting reflexes (Warkentin and Carmichael, 1939), pulsion reflexes (Fischer, 1928), and trunk and limb responses to head position changes (Roberts, 1967b) are all thought to be initiated at least partially by stimuli from both the labyrinthine structures and the muscle receptors of the neck and limbs.

In addition to detecting linear accelerations and supplying stimuli for postural adjustments, the otoliths are also responsible for maintaining general muscular tone throughout the body. The otolithic tonic control of body musculature is inferred from findings which show loss of extensor tone after the inner ears of patients have been removed (McNally and Tait, 1936) and decreased oxygen consumption following bilateral labyrinthectomy (Ciurlo, 1936). Direct electrophysiological evidence indicating a tonic discharge from the otoliths will be presented in the next section.

The important aspect of otolithic function for the present study is the detection of small deviations in head and body positions (tilt) and the effectiveness of the reflexive discharge in controlling postural adjustments that occur in response to these small tilt deviations.

That the utricle and the saccule function in detecting small body tilts is strongly supported by numerous electrophysiological studies. Adrian (1943) recorded from the lateral vestibular nucleus (the first central projection of the utricular fibers in the cat and found that neurons were increasingly stimulated as the head was slowly tilted away from the normal position. Adrian also noted that these neurons showed little or no adaptation to a maintained tilt.

Other studies employing similar methods to Adrian's for recording otolithic discharge have found comparable results (Ross, 1936; Lowenstein, 1956; Lowenstein and Roberts, 1949; Rupert, Moushegian, and Galambos, 1962). While the molar type of stimulation used in these studies does not directly implicate the otolith organs as the specific receptor, a recent study by Gernandt (1970) using air puff stimulation delivered directly to the surface of the exposed utricular macula has produced results in good accord with the earlier findings. Gernandt used single cell recording techniques to record from units located in the lateral vestibular nucleus and identified four different types. Three types demonstrated spontaneous firing, but their responses to stimulation were different. The first type increased as stimulation increased, the second showed an initial burst of firing which rapidly changed to a state of complete inhibition, and the third displayed complete inhibition when stimulated. The fourth type of unit had no spontaneous firing, but showed firing proportionate to the intensity of the stimulus.

The accumulated electrophysiological evidence suggests that at least one of the functions of the otolith organs is to signal positional information. The types of units that have been identified seem well suited to convey this type of information in that (1) they show spontaneous discharge (2) they respond specifically to tilt and (3) at least some types show no adaptation to prolonged tilts. It seems justified to consider that the reflexes from these organs help maintain erect posture during slight shifts in the body's center of gravity. This is accomplished by the organ's acute sensitivity to tilt (2 - 4 deg from vertical; Guiallorotti, 1966; Clark and Graybiel, 1964; Gescheider and Wright, 1965) and the rapid postural reflexes that accompany this detection.

There is, however, evidence disclaiming such high sensitivity of the otoliths to tilt. The studies obtaining thresholds of 2 - 4 deg were performed with the S sitting in a normal environment. Results from studies where the S make~ ~rientation determinations while immersed in water demonstrate that the otoliths are very inefficient as gravity sensors especially when the head is held in a fixed position. Immersion in water eliminates almost entirely the kinesthetic and cutaneous cues since these are equally distributed about the body when the Ss are at a neutrally bouyant depth (15-25 ft.)

Stigler (1912) and Nelson (1968) have both reported that <u>S</u>s, after being rotated under water, could not point straight upwards with any degree of accuracy. The error of estimates ranged from 0 deg to 180 deg with the average error being 15 deg. Brown (1961) obtained similar results, demonstrating again the low sensitivity of the otoliths, but he also noted a finding which may indicate that the sensitivity of the otoliths during water immersion may be an artifact. If <u>S</u>s were allowed to move their heads after rotation, estimates became much more accurate. The question of otolithic sensitivity becomes very important in theoretical considerations as to whether or not the otoliths are involved in static equilibrium performance. The available evidence concerning otolithic thresholds is contradictory, but if the lowest estimates are considered as being correct, then theoretically the otoliths could be involved in detecting the small body oscillations that occur during static postural equilibrium.

The study presented here represents a new application of the well-known power apectral density measure of a random process in an attempt to ascertain if the labyrinthine system is involved in contributing to normal erect posture.

Static equilibrium has traditionally been defined as the ability of man to maintain a normal erect body posture while standing still for a period of time. An S's ability to maintain static equilibrium has been determined by a variety of methods, all of which utilize some methods of recording standing sway. Historically, according to Luciani (1915), Hinsdale (1887) was the first investigator to measure standing sway. Hinsdale fastened a piece of cardboard to the top of the S's head and stretched a piece of smoked paper over the cardboard. Swaying movements were registered as the smoked paper rubbed against a small needle positioned above it. This method and variations of it were in general use until Miles (1922) developed a mechanical pulley system (ataxiameter) which automatically tabulated amount of sway in both lateral and anterior-posterior directions. The ataxiamter was designed so that four small threads were attached to the center of a helmet worn by the S and passed outward to the corners of a square wooden frame positioned at the level of the S's head. The threads passed over small unidirectional adder wheels located at each corner of the frame. These wheels were turned by the threads, which were pulled as the S swayed. The counters were automatically increased as the wheels turned, thereby tabulating total amount of sway in arbitrary units.

Recently, electrical recording techniques were utilized by Dzendolet (1963) to record standing sway. The recording apparatus developed by Dzendolet consisted of a flexible tubing system which extended from the top of an army helmet liner worn by the S. The tubing was attached to a wiper blade which activated a microtorque potentiometer. Amplitude of sway movements could then be recorded on an oscillograph as voltage deflections. Other similar methods have been used (Travis, 1945; Jarrige, 1968), but the criticism of all these techniques is that they do not reflect total body sway, but reflect instead a combination of body movements and spurious head movements. Bensel and Dzendolet (1968) attempted to overcome this problem by utilizing a power spectral density (PSD) analysis of standing sway movements. Ss in the Bensel and Dzendolet experiment were required to stand upon a platform. Four electrical strain gauges were placed under the platform, and those on opposite sides were connected as elements of two separate Wheatstone bridge circuits. This arrangement, like the apparatus developed by Miles, allowed partitioning of any sway movement into its anterior-posterior and lateral components.

The PSD analysis as utilized by Bensel (1967), Bensel and Dzendolet (1968), and Weissman (1970) in standing sway, is applied to a waveform that does not show characteristic or periodic changes (i.e. it is non-recurring). The PSD measure gives the average power[1] contained in a given frequency band of the waveform, and is generally represented as a plot of PSD *vs* frequency. Since the initial waveform in these studies is a direct reflection of the *S*'s sway movements, the derived PSD curve can be considered as a measure of total body sway. In addition to proprioceptive adjustments, the measure may also be sensitive to the contributing effects of such sensory systems as the otolithic, cutaneous, visual, and auditory as well as to numerous non-sensory factors.

The sensory processes generally considered as contributing to normal upright balance, and the central nervous system process (CNS) itself, seem to be largely reflexive (Jongkees, 1967, p. 155). Information received by the CNS signalling the position of the body in space comes from many sensory systems. Presumably, this entire "pool" of information is integrated by the CNS and sent to the compensatory effector organs via the final common path.

Standing sway can thus be considered as a series of small compensatory responses made in response to a pool of sensory information. One problem in using standing sway as a dependent measure is that the degree to which each of the sensory systems contributes is largely unknown.

Fearing (1924a) stated the problem:

> Sway may be considered as (1) a reflex response initiated by nervous impulses arising in the non-acoustic labyrinth, (2) a conscious response dependent upon kinesthetic, cutaneous, visual and auditory cues, and (3) a response in part dependent upon reflex connections with receptors in the ampullae and vestibule and in part dependent upon voluntary responses to the sensory cues outlined in (2). (p. 94).

Available research indicates that Fearing may have placed excessive influence on the contribution of the labyrinthine system. The evidence presented below is the result of attempts to separate the effects of the various sensory systems which contribute to static equilibrium.

<u>Labyrinthine Effects</u>

The evidence previously presented supporting the role of the otoliths in detecting body tilts and their sensitivity to these small deviations indicates, on a theoretical basis, that the otoliths can contribute to static equilibrium. A basic assumption of this study is that the otolith organs do affect the process of maintaining upright balance. Miles (1922), Gates (1918), and Bolton (1903), however, dismissed the labyrinth as a contributor to standing sway and considered the response as purely a function of muscles working against gravity. Miles (1922) believed that the slight deviations in tilt that occurred during sway were too small to initiate labyrinthine compensatory reflexes, and, if these impulses were initiated, they would be but vague, weak impressions. Miles' calculations showed that a man 173 cm tall who sways with deviations as great as 30 mm subtends a tilt angle of about 1 deg. In order for the otoliths to detect this deviation they must be at least as sensitive, if not more sensitive than current threshold data of 2 deg-4 deg indicates.

The earliest work undertaken on a labyrinthine effect, and perhaps the study upon which these early investigators based their opinions, was done by Hinsdale (1887) who ran 17 deaf mutes in a static equilibrium test. The deaf mutes compared favorably in total amount of body sway with normal adults, and demonstrated no problems in standing upright with their eyes closed even though they were presumably missing the labyrinthine sense organs. Although these results indicate a lack of labyrinthine influence, Edwards (1942) found that a few deaf mutes swayed excessively, especially with eyes closed. The overall mean sway for deaf-mutes with eyes open was quite normal, but the increase in sway with eyes closed was much greater than that found with normal Ss. The results obtained by Edwards show that vision may adequately compensate for a loss in labyrinthine structures under normal conditions.

Two unpublished studies which support a labyrinthine role in standing sway have come from this laboratory. Following a procedure developed by Dzendolet (1963), both Bensel (1967) and Scott and Dzendolet (1970) applied sinusoidal electrical stimulation bilaterally to the mastoid processes where the vestibular portion of the eighth cranial nerve runs most superficially. Bensel (1967) found that sway movements across a wide range of frequencies were unaffected except for a specific increase at a narrow frequency band which corresponded to the frequency of the stimulus input. Similarly, Scott and Dzendolet (1970) found that sway movements reflected the form of the sinusoidal electrical stimulus, although showing a small phase lag.

This result was obtained by employing a technique of the general type used in computers of average transients to average sway responses over a four minute period. These two studies indicate that sway can be strongly influenced by activation of the labyrinthine receptors (or their neurons) during a standard static equilibrium test.

Visual Effects

Results showing that visual cues can increase stability come from numerous studies (Hinsdale, 1887 and 1890; Hancock, 1894; Miles, 1922; Bullard and Brackett, 1888; Edwards, 1942; Travis, 1945; Fisher, Birren, and Leggart, 1945) which unanimously show a decrease in total sway ranging from 10% to 86% when the eyes are open. The reported decreases were greater in the anterior-posterior direction than in the lateral direction. Miles (1922) does report one set of results in contrast to the general findings reported above. The data from [illegible] aviation candidates showed an increase in sway with the eyes open. However, other results from the same paper by Miles showed that, with a highly practiced S, sway suddenly improved by 45% when visual cues were not blocked. Recently, Weissman (1970) has also found that total power under the PSD waveform was less when vision was allowed, but was not significantly different ($p > .05$) from the eyes closed condition.

Auditory Effects

Auditory cues may also play a role in determining total amount of sway. Bensel, Dzendolet, and Meiselman (1968) showed that amplitude of sway was significantly increased in a 5 second period immediately following presentation of a tone. von Bekesy (1935) reported increased head movements to intense (92 dB) tones. Husband (1934) reported greater sway with all types of music than during silent periods. The above two variables, vision and audition, while influencing sway to some degree, can be experimentally eliminated or controlled as cues during standing sway.

The process of analyzing the contribution or effect of other sensory systems to equilibrium becomes more complex. Cutaneous cues (i.e., tactile receptors on the feet, pressure gradients from unequal distribution of clothing, and pressure from the visceral organs) and kinesthetic cues are less easy to separate or eliminate.

Cutaneous Cues

Results supporting a contribution of tactile cues of the feet to sway are equivocal. Miles (1922) has reported that pressure gradients on different parts of the soles of the feet are useful in controlling equilibrium. Miles comments that heavy anesthesia of the soles has "long been known" to markedly increase swaying, but cites no reference source for this statement.

It is interesting to note that Miles (1922) reported that an S, highly practiced on the sway apparatus, was more stable in bare feet than in shoes. It would seem that the increase in sensitivity to sway from cutaneous cues of the bare feet more than offsets the presumed increase in support obtained from leather shoes.

Hinsdale (1887) found greater sway without shoes for an S who was not highly practiced. The differences may be explained by the degree of training but no general conclusions can be drawn concerning the effects of tactile cues from these two limited studies.

Proprioceptive Cues

The proprioceptive component of the postural mechanisms is unquestionably the predominant influence in standing sway. Wendt (1951) has stated his views on the importance of the musculature in maintaining upright support: "The vestibular apparatus is the second most important source of stimuli for postural tonus and body posture. Only the muscle receptors take precedence." (p.1204).

Results supporting the validity of Wendt's statement come from two electrophysiological studies. Anderson and Gernandt (1956), while recording from the ventral root of the cat, found that even strong efferent discharges from the vestibular system could be inhibited by a muscular contraction. Gernandt (1970) recorded unit activity in the lateral vestibular nucleus of the cat. An electric shock to the sciatic nerve and an air puff to the exposed utricle were paired at varying intervals in a classical conditioning paradigm. The stimulus to the sciatic nerve inhibited the utricular test response for a period of 200 msec. However, a strong volley of impulses from the utricle was not affected by sciatic stimulation if it arrived at the lateral vestibular nucleus first. Gernandt concluded that t' bisensory convergence between utricular and ascending propriospinal impulses revealed that utricular activity is strongly and dominantly under proprioceptive control, except in special cases when the impulses from the otoliths are unusually strong.

Psychophysical studies using the oculogravic illusion as the dependent measure further emphasize the importance of proprioceptive stimuli in maintaining posture. The oculogravic illusion is "an apparent motion and an apparent displacement of a fixed visual field when an S is exposed to a change in magnitude and direction of gravitonertial force" (Clark, 1967, p. 332).

Operationally, the illusion is produced by seating an S in a chair at the end of a centrifuge arm and applying an acceleration. The S's task, while in darkness, is to set a luminous rod to the horizontal during the applied acceleration. For normal Ss either the rod appears to turn in a clockwise direction about a central pivot point, or it is displaced upwards. These two phenomena occurring either together or separately are generally large effects and last as long as acceleration is applied to the S.

The most likely explanation for the oculogravic illusion is that the S uses the resultant vector from the two applied accelerations of gravity and centrifugal force to orient himself to the horizontal. The resultant force is then considered to be the gravity vector and perceptions of the true horizontal change depending upon the difference between true gravity and the new resultant. There is also ocular counterrolling present during applied accelerations or tilts and this is thought to accurately reflect otolithic function. The eyeballs rotate about a fronto-occipital axis and such rotation may aid the oculogravic illusion.

Labyrinthine defective (L-D) *S*s when exposed to this type of experimental situation do not experience the oculogravic illusion (Graybiel, 1952; Graybiel and Clark, 1965). However, Clark and Graybiel (1966), using both normal and L-D *S*s, found no significant differences between the two groups in their perception of the visual horizontal if the *S*s were standing, instead of sitting, during the applied accelerations. Their data suggest that, when available, tactile and kinesthetic information for the normal *S*s is the dominant influence in the perception of the visual horizontal.

In a study testing the accuracy of settings to the postural vertical, both normals and L-D *S*s showed consistent improvement over a series of successive settings, and there were no differences between the two groups. This result indicates that either the postural cues are sufficient for making settings to the vertical or that there is some central compensation in the L-D *S*s for the loss of the labyrinthine structures.

This statement is also supported from the results of the water immersion studies mentioned previously. The low sensitivity of the otoliths for detecting the vertical during immersion tests, and the accuracy of those settings in air, suggest that the kinesthetic sense is very important in maintaining upright posture.

All of these studies confirm the importance of postural cues in maintaining upright balance. The proprioceptive cues are the dominant information in bodily posture; the other non-proprioceptive cues may influence or modify the basic information but not unless these competing cues are unusually strong or well defined.

In this regard, it seems difficult to make statements supporting the dominance of one contributing sensory system over another. For instance, the oculogravic illusion is a case where cues from the labyrinth become more important to the S than the few available proprioceptive and visual cues, and the S thus experiences the apparent movement of the rod. The oculogravic illusion can be interrupted, however, when the target rod is presented within a full visual field (Graybiel, 1952; Witkin, 1950). This condition apparently supplies sufficient visual cues to override the dominant influence of the labyrinth.

Non-sensory factors such as differences in sex, athletic activities and abilities, foot position, type of footwear, practice effects, physiological rhythms, respiration rate, age, mental set or attentional processes, and physical attributes (height, weight, etc.) have also been investigated as possible contributors to standing sway.

Sex Differences

Findings regarding sex differences have all tended to show that females are considerably more stable than males, and that males demonstrate greater variability in sway responses (Hinsdale, 1887; Hancock, 1894; Miles, 1922; Fearing, 1924a; Bensel, Dzendolet, and Meiselman, 1968). The same general result was obtained even when sexes were grouped according to weight (Travis, 1945).

Athletic Activities

Fearing (1924a) polled 116 *S*s as to their athletic abilities and activities and found that there was no relationship between these activities and amount of sway. Wyrick (1969) found that the strength of the dorsi and plantar flexor muscles of the foot was not significantly related to any of the three static balancing tests given to the Ss.

Foot Position

There are three commonly used positions of the feet when testing standing sway. The most stable position is with the feet placed parallel to each other and at least 20 cm apart, whereas the least stable is with the heels and toes placed together (Romberg position). A position which gives stability intermediate to the above two is the V-position or 45 deg position. The S stands with his heels together and the inside angle of his feet at 45 deg (Miles, 1922; Fearing, 1924a). Fearing (1924a) found that the anterior-posterior sway is always greater than the lateral sway regardless of the foot position employed. He also noted that the Romberg position is the most susceptible to practice effects, probably because it is the most unaccustomed of the positions. The S, therefore, has no aids built up from past experience and must depend upon the raw data offered by the receptors. For these reasons, Fearing believed that the Romberg was the best foot position for studying static equilibrium. Miles (1922), on the other hand, felt that the 45 deg position gave all Ss the same advantage and the sway responses were less variable than with the Romberg.

Footwear

The wearing of shoes during static equilibrium tests was a standard procedure used by most of the early investigators. The studies comparing sway results with and without shoes have had mixed findings. Bensel, Dzendolet, and Meiselman (1968), testing without shoes, found that the difference between anterior-posterior and lateral sway was much less than the differences reported in these earlier studies (Fearing, 1924a; Edwards, 1942). Bensel, et al. believed that the shoes used in the previous research supplied added support, especially in the anterior-posterior direction, so that the body's center of gravity could be shifted forward to a greater extent without loss of balance. Absolute comparisons between these studies are difficult to make because of the different response measures, and it may just as well be that the obtained differences were due to an increase in lateral sway without shoes. This methodological problem is complex and shoes serve only to confound the results. Body sway testing should be done either in bare feet or in stocking feet so that tactile information is not reduced, and each S uses those cues that are a natural part of his postural system.

Respiration Rate

Respiration may also affect sway behavior. During inspiration the chest cavity expands, throwing the center of gravity forward. The head then moves back to counterbalance these changing forces. Sway, especially if measured from the head level, should be markedly affected by the rate of respiration (Travis, 1945). Travis found that total sway increased after an S performed mild exercise. The results were not attributed to a fatigue effect, but only to increased respiration after the exercise. Travis later ran six half minute sway periods which alternated breathing with no breathing, and found sway was slightly reduced with the no breathing condition. Miles (1922) recorded sway responses on a kymograph and noticed rhythmic fluctuations that seemed to follow respiration rate. The fluctuations, however, did not disappear when breathing was temporarily discontinued. Nevertheless, Miles felt that the regular wave patterns may have resulted from slight reflex contractions or alternating changes of muscle tonus associated with respiration. Respiration rate ranges from 12 to 18 cycles per minute for an adult male. This rate is about 0.2 or 0.3 cycles/second, which is within the frequency range of the PSD waveform. Should respiration rate be reflected in the PSD curve, an increase in power would appear at the frequency band corresponding to the breathing rate.

Physiological Rhythm

There has been very little research on sway behavior and physiological rhythm. Fearing (1924a) tested one female S daily for a total of 134 days. He found a large decrease in sway at the onset of menstruation. Total sway gradually increased as the cycle progressed and became maximum a few days before the next menstrual flow. These data suggest that female participants should be tested while at the same stage of their menstrual cycle, and that within subject testing over a number of days may seriously confound results.

Practice Effects

Fearing (1924a) observed an effect of practice in the study described above. Total sway and sway variability decreased steadily, the decreases being less with the 45 deg position than with the Romberg position. In a different study, Fearing (1924b) found that practice effects for total amount of sway were more pronounced with the Romberg position, but not significantly different from the 45 deg foot position. However, when analyzed by direction, decreases in sway in the anterior-posterior direction was greater for the 45 deg position, while the larger decrease in lateral sway occurred with the Romberg position.

It has been shown that sway in the anterior-posterior direction is greater with the 45 deg foot position than with the Romberg position, and also that lateral sway is greater with the Romberg than with the 45 deg foot position. Generally, the effect of practice, then, is greatest on the combination of foot position and direction that normally gives the greatest amount of sway (Fearing, 1924b). In this respect lateral sway in the 45 deg foot position would be least affeected by practice and this position should be preferred when recording lateral sway.

Miles (1922) also reported a practice effect from data on 12 adults who were tested for 21 consecutive days. The data, presented in three one-week blocks, showed a 13% improvement between the first and second block and an additional 5.5% improvement between the second and third block.

Negative findings on the effect of practice have been reported both by Fisher, Birren, and Leggart (1945) and Edwards (1942). Begbie (1966) reported no practice effects when Ss were tested with eyes closed, while Dickenson and Leonard (1967) reported improvement in stability over a number of trials if minimal visual cues were allowed. These studies suggest that practice may have its effects through an increase in ability to utilize visual cues for body orientation.

Miles (1922) noted that the average decrease observed due to practice effects was less for sway responses than for other motor tasks. It may be that sway is already a highly practiced function and that practice has its effects only on the relatively unfamiliar standing position imposed upon the S during testing (Fearing, 1924a). If this is true, then it is reasonable to assume that practice effects are the result of an increased ability to respond to kinesthetic, tactile, labyrinthine, and other sensory cues such as the visual effect hypothesized above.

Fatigue

Edwards (1941) found that fatigue greatly increased the total amount of sway. Ss were deprived of sleep for 100 consecutvie hours and the sway responses measured after the deprivation period were more frequent and more variable than those measured just prior to the deprivation period. Travis (1945) and Begbie (1966) both found decreased stability after mild exercise. Travis attributed the effect to increased respiration, while Begbie thought his result to be a genuine fatigue effect. Bensel and Dzendolet (1968) found that both anterior-posterior and lateral sway increased at all frequencies (0.05-4.0 Hz) when male Ss were tested after 20 minutes of standing.

Age

Sway responses are not greatly influenced by age. The general finding (Hindsale, 1887; Hancock, 1894; Edwards, 1942) has been that total sway is extremely high in young children (3 - 7 years), and then gradually decreases and remains consistently low and stable from age 14 through age 70. The steadiest age group is the 25 to 29 year olds.

Attention

Mental alertness states may also affect sway as unpublished results from our laboratory have shown. The total power in the PSD waveform varies considerably while Ss are performing different mental tasks during the sway period. A 'reverie' condition produced the largest amount of sway, while those Ss involved in mental arithmetic swayed the least. Brown (1966) found that vestibular nystagmus was greater when Ss were performing a mental arithmetic task than when encouraged simply to relax. Collins and Guedry (1962) also reported an enhancement of vestibular induced nystagmus while Ss were engaged in silent mental arithmetic. They also noticed a failure of the nystagmus to decline in strength during prolonged periods of angular accelerations while the Ss were mentally doing arithmetic. Fearing (1924b) found that his sway data was considerably higher than that reported by Miles (1922). The only difference between the two studies was that Ss in the Miles experiment silently counted during the sway period, whereas Fearing imposed no such mental task.

These results are in good accord with our own data. Mental alertness states may function as a type of arousal process with the general trend of greater or better performance in those responses controlled by the labyrinth. This conclusion is supported by those studies which show an increase in nystagmus and a decrease in body sway when Ss are required to perform mental tasks. An alternative explanation to these results is that the basic reflex mechanisms involved in postural adjustments act more efficiently to reduce the amount of oscillation than the more consciously controlled impulses from the higher nervous centers. This explanation at first appears improbable since it seems to violate biological efficiency in adding a control system that results in a deterioration of performance. The exertion of voluntary control over the reflexive postural system is advantageous during complex body maneuvers. However, the mental calculation of arithmetic problems may serve to divert the attention of these centers and allow the basic reflex mechanisms to act singly in controlling the relatively simple process of maintaining upright posture.

Body Characteristics

Finally, the physical characteristics of a S (height, weight, center of gravity, etc.) may also determine the extent and pattern of body sway. Miles (1922) found that taller and heavier Ss show greater unsteadiness since swaying occurred principally from the ankles. Miles artificially increased the height and then the weight of an S and found that sway increased with both conditions. Fearing (1924a) also found that height had a somewhat greater effect on sway than did weight. His results showed correlations of .22 and .21 for height, and lateral and anterior-posterior sway, respectively. Similar correlations regarding weight were .15 and .22, respectively. Travis (1945), using the Romberg position, found very low negative correlations between height, weight, and foot length with total sway. Bensel, Dzendolet, and Meiselman (1968) found a large (r=.42) correlation between the product of the center of gravity and the weight, and a male S's mean anterior-posterior sway amplitude, while the correlation of that product with lateral sway was considerably reduced (r=.01). They concluded that lateral sway more accurately reflected the basic processes acting to maintain an upright posture than did anterior-posterior sway, since the anterior-posterior sway was more highly correlated with physical body characteristics.

The current research was carried out in the form of the following two main experiments.

Experiment I

This experiment was designed to investigate more completely than previous research the influence of body characteristics on sway behavior. Earlier work was limited to finding correlations between the total amount of sway and variables such as height, weight, foot length, and center of gravity. The general result of this earlier research indicated that sway in the anterior-posterior direction was highly correlated with simple physical body characteristics, and that lateral sway remained relatively independent of such variables. Since lateral sway is implicated as the response most sensitive to the basic mechanisms involved in upright posture, it is desirable to have available information that specifies the influence of body characteristics on this response. A series of 13 anthropometric measurements were taken on each individual and 20 more variables were derived from the original 13. These 33 variables were then correlated with 13 discrete frequency values from the PSD curve for lateral sway, and also with a measure of total sway (total power under the curve). The purpose of this study was to isolate the factors contributing significantly to lateral sway. If a multiple regression equation can be computed from the data obtained in this study, then a reasonable representation of the PSD sway curve can be predicted for use as a control for the labyrinthine defective Ss in Experiment II.

Experiment II

This study attempts to determine whether or not the intact otoliths are an integral part of the postural system. The multiple regression equations derived from the data of 77 normal Ss in Experiment I were used to predict the expected PSD curve of the L-D group tested in this experiment. The same measuring and testing procedures applied to the normal group was also used on the L-D group. The mean anthropometric measures for the L-D group were used to calculate a mean PSD sway curve, and this predicted curve was then compared to the actual averaged curve obtained from the four minute sway period. Expected results obtained from these comparisons are (1) that the otoliths contribute to the maintenance of upright balance, and (2) where, in terms of sway frequency, the otoliths exert their control on the postural system.

One way of conceptualizing the postural control system is in terms of a series of positive and negative feedback mechanisms working on the anti-gravity muscle groups, each of which in turn is a small, yet highly complex servo-mechanism. These automatic lower level processes are probably stabilized by the visual, labyrinthine and cutaneous systems which may represent part of the high level (conscious) control or may, in part, also contribute on a reflexive level. These stabilizing influences work to feed information to the leg and trunk muscles to reduce unwanted oscillations that occur during a period of standing upright. This brief sketch of how cybernetics can be applied to the postural system will be expanded in the discussion section.

Method

Subjects

Experiment I. Anthropometric measures and sway records were obtained from 77 normal, young, male adults. The majority of participants were drawn from a population of students enrolled in an introductory psychology course. Participation for these Ss was on a voluntary basis as this study was one of many available to the student to fulfill a course requirement. Other normal Ss were obtained, on a volunteer basis, from the graduate student population at this University. Graduate student participation was solicited so that the sample would reflect a wider range of ages than that attainable with only freshman students. Only Ss who reported no history of fainting spells, serious foot or leg injuries, inner ear infections, or serious head injury (concussion or fracture) were selected for the experiment.

Experiment II. Labyrinthine defective Ss were obtained through confidential letters mailed by the University of Massachusetts Communication Disorders Clinic to patients with reported or suspected labyrinthine defects. Additional L-D Ss were obtained from the ENT clinic at Westover Air Force Base in Chicopee, Mass.

Diagnostic reports were requested from each patient's attending physician. These reports indicated the extent of the labyrinthine involvement. In addition to these reports, each L-D S was asked a series of questions and several tests were administered to confirm the presence of a labyrinthine disorder. In all, 4 male and 2 female Ss were obtained.

Apparatus

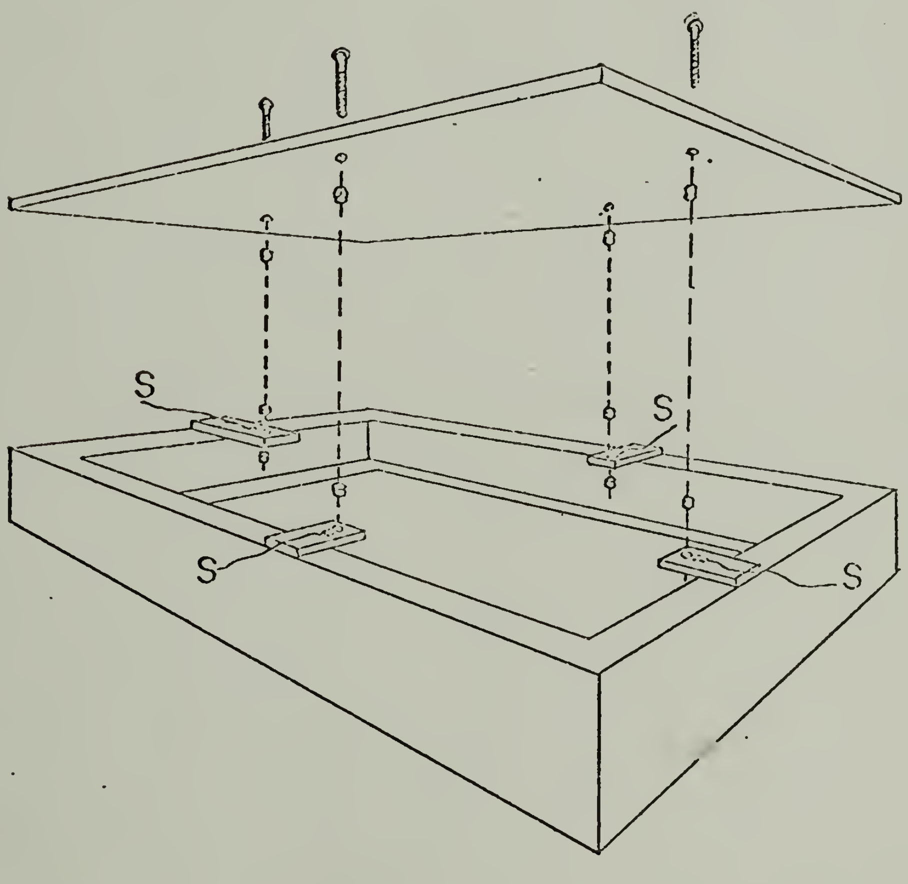
Anthropometric measures were taken with a standard anthropometer (Lifkin Co.- Model 7112 ME), and a centimeter ruler. Height and weight were taken on a sliding weight balance scale (Detecto-Scales, Inc.- Model 239). The center of gravity (CG) of each S was taken on a balancing board (244 cm long, 28 cm wide, and 2.6 cm thick) which rested on an iron pipe that was used as the fulcrum point. The bottom portion of the board near its center was fitted with a steel plate (53.5 cm by 28 cm) so that the balancing board was easily moved across the iron pipe when balancing the S. The board was initially balanced at the center point by appropriate weighting at each end. This gave the center of gravity for the board (CG_B). The S was then placed on the board with his feet flush with one end, and the board and the S were balanced. The center of gravity for the entire system (CG_S), the board and the S was thereby obtained. The center of gravity of the S (CG_I) was then calculated from the equations given in Appendix A.

The sway transducer was a square wooden platform 2 cm thick and 68 cm along each side. The platform was supported at the center of each side by the ends of four horizontally positioned steel bars which extended under the platform. The platform was firmly fastened to these bars by a machine screw through each bar. The other ends of the supporting bars were rigidly fastened to a framework of welded steel positioned below the platform. Strain gauges were placed on all four bars, and those on opposing bars were connected as elements of two separate Wheatstone bridge circuits. A diagram of the sway platform is presented in Figure 1. For this experiment, only the right and left strain guages were wired to the recording instruments as only lateral sway data was collected. The output from the bridge circuit was led into two ganged amplifiers (Hewlett-Packard- Model 2470A) and then recorded on magnetic tape. Each lateral sway signal, thus recorded, was then sampled every 0.2 sec. and converted to digital form using a PDP 8/I general purpose digital computer with an analog-to-digital converter. This sampling method gave 1200 data points for the four minute sway record. The converted data was then analyzed on a CDC 3600 computer with an autocorrelation function being computed first. The PSD function (Milsum, 1966) was obtained via the computer as the Fourier transform of the autocorrelation function. Taking the Fourier transform of the autocorrelation function places the sway signal in the frequency domain rather than the time domain. For uniformity and ease of comparison, all PSD values are plotted in decibel units in this thesis. ($10 \log_{10}$ PSD)

<u>Procedure</u>

The following body measures were taken from each *S*: (1) weight, (2) height, (3) buttock-knee length, (4) bideltoid diameter (shoulder breadth), (5) gluteal furrow height, (6) kneecap height, (7) sub-malleolus breadth, (8) medial malleolus height, (9) footlength, (10 heel breadth, (11) ventral instep breadth, (12) center of gravity, and (13) age. All length and breadth measures were taken in, or converted to, centimeters. The weight of each subject was taken in pounds and converted to Kg. Footlength, heel breadth, and ventral instep breadth measures did not conform to the standard anthropometric foot measures, but were taken with the feet in a position which corresponded to the foot position used during the

S - Strain gauge



THE SWAY TRANSDUCER
PLATFORM

Fig. 1

sway test.

A complete description of all these measures is presented in Appendix A. The 13 original measures were then used to derive 20 additional variables, 15 of which were proportionate measures (e.g., leg length was represented as a percentage of total body length, and footlength was likewise determined as a proportion of the ponderal index). The 20 derived measures are listed below: (1) center of gravity minus medial malleolus height (CG-MMH), (2) center of gravity times weight (CGxWgt), (3) moment of inertia of a cylinder pivoted at one end, $(\frac{ml^2}{3})$, (MOM IN), (4) gluteal furrow height divided by height (GFH/Ht), (5) ponderal index, (height divided by $\sqrt[3]{\text{weight}}$), (PondIx), (6) gluteal furrow minus medial malleolus height (GF-MMH), (7) kneecap height divided by height (Kne/Ht), (8) medial malleolus height divided by height (MMH/Ht) (9) buttock-knee length divided by height (Bkn/Ht) (10) gluteal furrow height minus medial malleolus height divided by height (G-M/Ht), (11) center of gravity minus medial malleolus height divided by height (C-M/Ht), (12) center of gravity divided by height (CG/Ht), (13) submalleolus breadth divided by ponderal index (Sub/PI), (14) submalleolus breadth divided by weight (Sub/Wt), (15) foot length divided by ponderal index (FtL/PI), (16) heel breadth divided by ponderal index (HeB/PI), (17) ventral instep breadth divided by ponderal index (Ins/PI), (18) footlength divided by height (FtL/Ht), (19) heel breadth divided by height (HeB/Ht), (20) ventral instep breadth divided by height (Ins/Ht).

After the body measures were obtained, S was shown the platform and asked to remove his shoes. Instructions were read to each S which asked him to stand quietly on the platform in a relaxed manner with his hands clasped loosely in front of him. S was asked not to move his feet and to keep his weight evenly distributed upon both feet. Once standing on the platform, the S was blindfolded and his feet were quickly positioned so that the heels were together and the feet spread at 45 deg to one another. Attention was controlled by requiring each S to mentally multiply the seven and eight tables during the sway period. Recordings began immediately and lasted for 4 min, and this 4 min sample was analyzed according to the procedure outlined in the apparatus section above.

Results

Experiment I

Anthropometric measures. The means, the sample standard deviations, and the range of scores obtained from the 77 normal Ss were compared to the data obtained by Hertzberg, Daniels, and Churchill (1954) on 4,000 Air Force flying personnel in 1950. The data from the Hertzberg, et al. and from the present study are presented in Table 1. Inspection of Table 1 shows that the present data are in good accord with that obtained from the earlier, and much larger, sample. The difference scores show small, but consistently higher, means for all measures, and a slightly larger variability for the data in this study. Although there are no noticeably large discrepancies in the mean and the standard deviation scores, the range values indicate that the Hertzberg et al. group encountered more extreme types than those Ss who volunteered for the present study. The greater range of body types found in the Hertzberg et al. study may be due to the larger sample of Ss that was measured. Alternatively, the fact that Ss in the present study were volunteers makes it likely that fewer of the extreme types would sign up for such an experiment.

The mean PSD curve for the 77 normal Ss is presented in Figure 2. The slope of this curve can be expressed in dB/octave. Visual inspection of this

Table 1

Table comparing the mean, standard deviation, and range of anthropometric measures obtained in the present study and from a WADC Tecnical report (Hertzberg, Daniels, and Churchill, 1954) based on data from more than 4,000 Air Force flying personnel.

Ek N=77 WADC (1950) N > 4,000	MEAN			STANDARD DEVIATION			RANGE	
Variable	Ek(1970)	WADC(1950)	d	Ek(1970)	WADC(1950)	d	Ek(1970)	WADC(1950)
Weight	75.78 Kg	74.39	1.39	10.59	9.48	1.11	53.6-101.1	47.2-120.4
Height	176.35 Cm	175.54 Cm	.81	7.21	6.19	1.02	160-194	151-197
Buttock-Knee Length	60.02	60.00	.02	2.69	2.70	-.01	54-67	47-70
Shoulder Diameter	45.94	45.41	.53	2.52	2.30	.22	40-53	37-58
Gluteal-Furrow Height	81.57	80.18	1.39	4.63	4.12	.51	67-89	64-94
Kneecap Height	52.92	51.35	1.57	3.06	2.61	.45	41-57	40-59
Medial Malleolus Height	8.82	8.77	.05	.56	.53	.03	7.2-10.4	6.6-10.9
*Foot Length	25.99	26.67	-1.08	1.34	1.48	-.14	23.4-29.5	22.5-31.1

*Foot length as measured in this experiment does not conform to the standard anthropometric procedures for obtaining foot-length measures. The lower mean value obtained in the present study reflects this procedural difference. See Appendix A.

Figure 2

Graph of mean PSD sway curve for 77 normal *Ss*. Power is in dB units *vs* frequency of sway waveform.

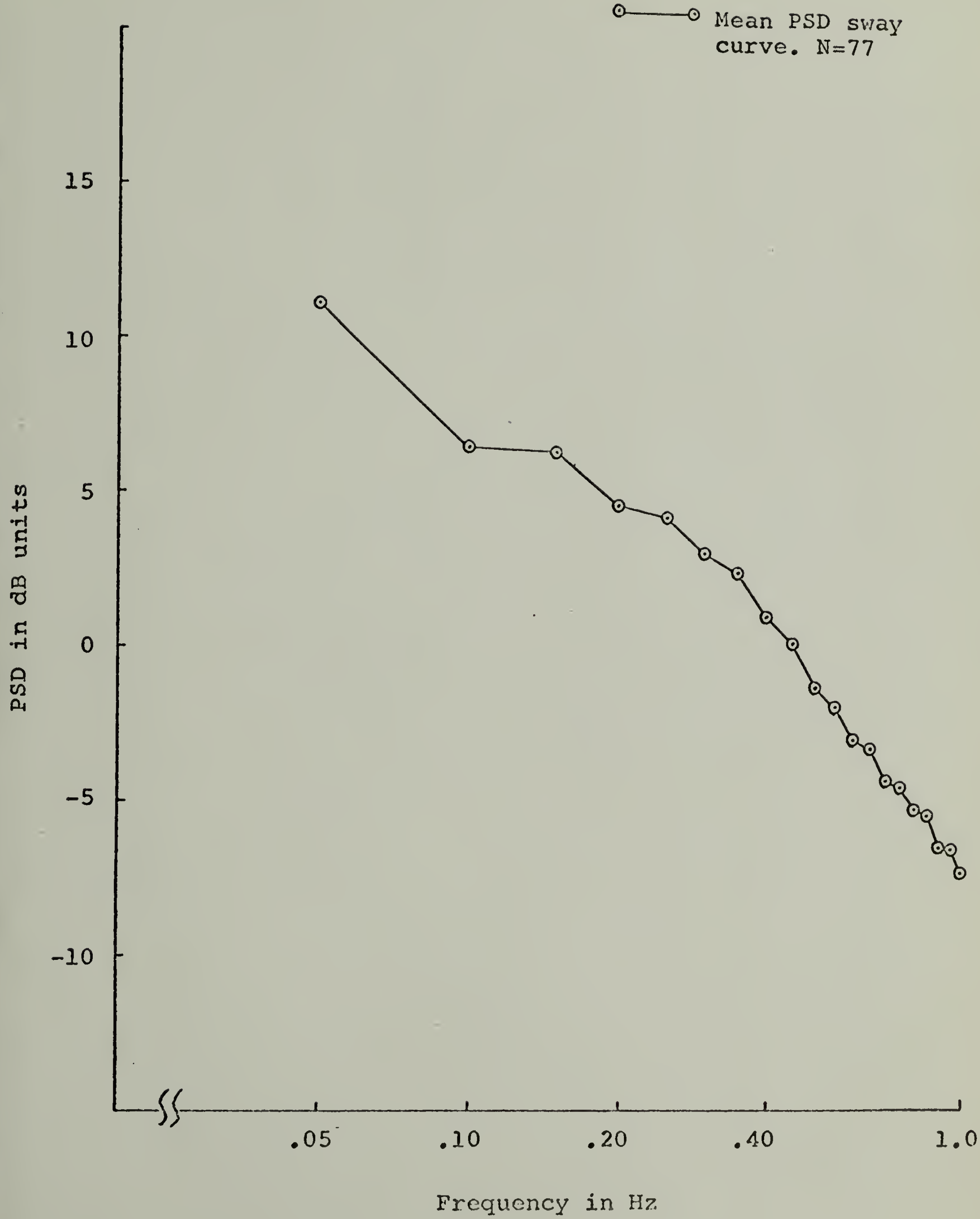

Mean PSD sway curve. N=77
15
10
5
0
-5
-10
PSD in dB units
.05
.10
.20
.40
1.0
Frequency in Hz

Fig. 2

curve shows that it is not a single linear fuction, but may be composed of two or more linear components. One pair of somewhat arbitrary regression lines which may be fit to the data are given in Figure 3. This analysis, though arbitrary, was chosen because of the similarity of the shape of this function to curves obtained from theoretical analysis of control systems as outlined below.

According to Milsum (1966) a biological control system, when analyzed in terms of PSD and plotted in a log-log manner, will have a slope of 0.0 dB/ octave if it is a proportional system ($C\theta=d$), and, at higher frequencies, will take on a slope of -6.0 dB/octave if it follows the equation for a first order[2] system ($b\frac{d\theta}{dt}$ $C\theta=d$). It can be seen from Figure 3 that the present data shows that the postural mechanisms may be considered a first order control system.

The linear portion of the fitted curve from 0.10 Hz to 0.21 Hz. has a slope of 0.0 dB/octave. The portion of the curve from 0.21 Hz to 1.0 Hz has a slope of -6.1 dB/octave, and when these two lines are extrapolated they cross at 0.21 Hz. One point (0.05 Hz) is considerably misplaced from these two slope lines. This point may represent a portion of yet another system

Figure 3

Graph of one possible pair of best fitting straight lines to the data points of the mean PSD sway pattern obtained from 77 normal male Ss. The data point at 0.05 Hz indicates that another system may be involved at very low frequencies.

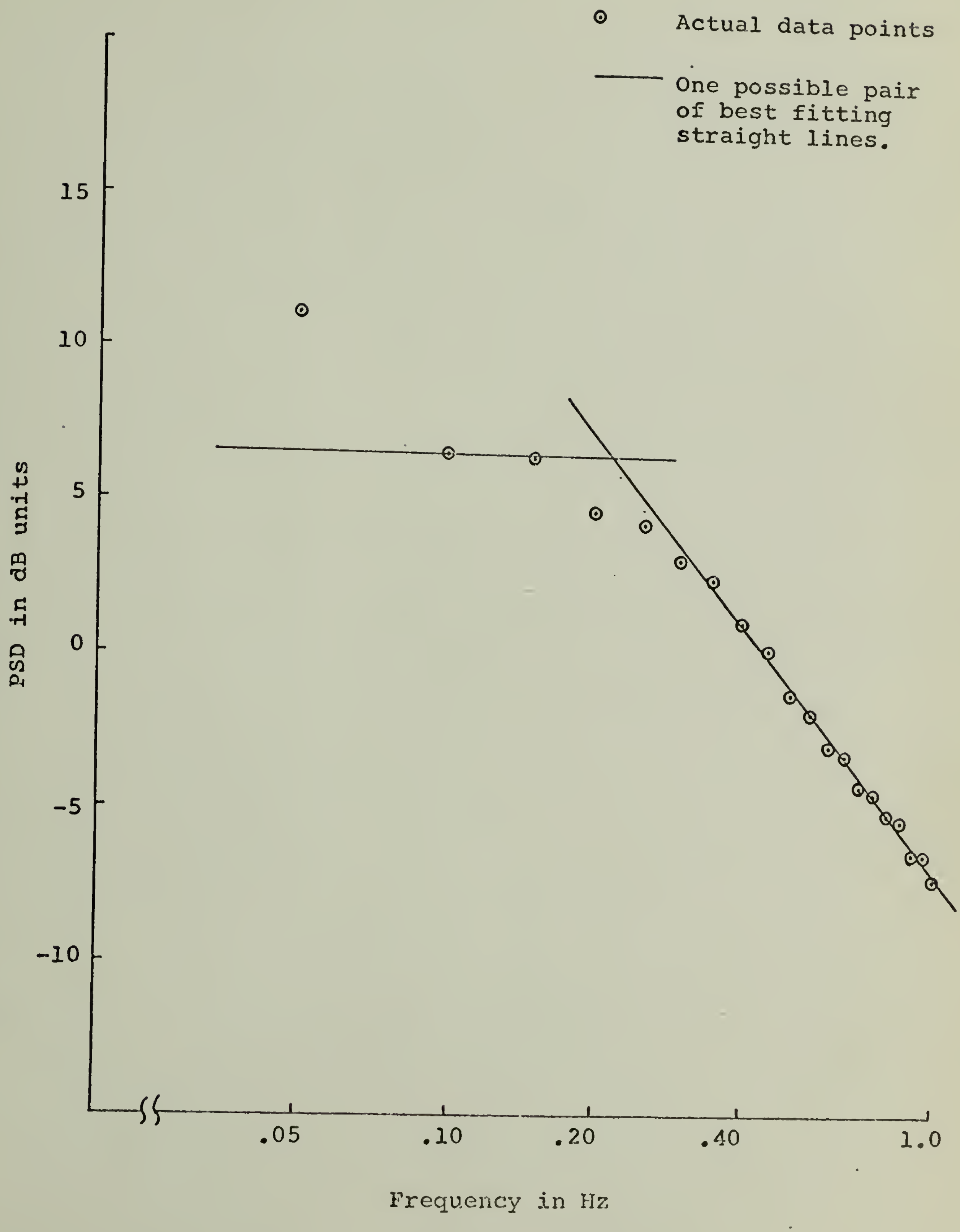

Actual data points
One possible pair of best fitting straight lines.
15
10
5
0
-5
-10
PSD in dB units
.05
.10
.20
.40
1.0
Frequency in Hz

Fig. 3

involved in the postural mechanism. However, the present method yielded insufficient data at the lower frequencies to warrant further discussion of this point.

A multiple correlation (R) was computed between the 33 body variables and the PSD values at 13 separate sway frequencies and for the total power under the waveform. The 13 frequencies were 0.05 Hz, 0.10 Hz, 0.15 Hz, 0.20 Hz, 0.25 Hz, 0.30 Hz, 0.40 Hz, 0.50 Hz, 0.60 Hz, 0.70 Hz, 0.80 Hz, 0.90 Hz, and 1.0 Hz. A multiple regression equation of the form $Y' = a + b_1X_1 + b_2X_2 .... + b_pX_p$ was computed at each of these frequencies from the data obtained in Experiment I. The variables entering into the equation were arbitrarily selected as those which could account for at least 1% of the total variance. This cutoff was determined after a small sample correction was made on the original R values (Guilford, 1956, p. 398). The $\underline{b}_i$ weights of each independent variable forming the equations for each frequency and for total power are presented in Table 2. Table 3 presents those variables that entered at least 3 equations and the frequency of the order in which they entered. The general trends evident from Table 3 show that moment of inertia for a cylinder pivoted at one end (a situation chosen as being a first

Table 2

Table presenting the b_i weights for the multiple regression equations derived from the data of Experiment I.

	TotPow	0.05Hz	0.10Hz	0.15Hz	0.20Hz	0.25Hz	0.30Hz
Const.	56.990	19.548	9.352	-23.020	-6.803	-14.606	-6.404
Height	-1.664						
Buknee							
Should				[illegible]	-.335		
Glufur						-.950	
Knecap			1.191	.750			
Submal			-13.849				
FootLe		1.115					
Instep			-1.374		-1.235		
C.Grav	.574		1.081				
Age	-.283		-.442	-.373	.437	-.378	.527
CG-MMH						.868	
CGxWgt							
MomIn	.084	.019	.094	,025	.054	.031	.038
GF-MMH			-1.611				
GFH/Ht					-115.443		-187.535
Kne/Ht		178.545					
G-M/Ht	-141.293	-232.645		-247.677			
C-M/Ht				159.350	144.682		176.962
CG/Ht							
PondIx	3.867		-2.163				
Sub/Wt			1184.194		156.908		
FtL/PI							
Ins/PI							-25.358
FtL/Ht	113.188						
Ins/Ht							

	0.40Hz	0.50Hz	0.60Hz	0.70Hz	0.80Hz	0.90Hz	01.0Hz
Const.	19.293	-20.671	-6.776	-23.853	-136.787	-44.897	-43.948
Height							
BuKnee							
Should		.640			.702	.670	
Glufur							
Knecap							.703
Submal							
FootLe							
Instep							
C.Mom.		.799	.732	.737		1.017	
Age	-.347						
CM-MMH					[illegible]		
CM-Wgt					-.010		
MomIn	.033	.007	.018	.015	.116	.002	.032
GF-MMH							
GFH/Ht	-181.910		-200.116		-188.295	-169.909	
G-M/Ht		-140.301		-155.362			-212.086
C-M/Ht	124.822						180.295
CM/Ht					346.227		
PondIx							
Sub/Wt							
Ftl/PI					-55.232		-67.306
Ins/PI							
Ftl/Ht						-193.393	
Ins/Ht	-134.984	-96.238					
Kne/Ht		-100.707		-40.450			

Table 3

Table presenting those variables that entered at least three regression equations, and the frequency of the order in which each variable entered the equations.

Variable	Number of Times Entered 1st	2nd	3rd	4th	5th	6th	7th
Should				1	1	1	
Knecap				1	1		1
CGrav			2	2	2		
Age				2	3	2	
MomIn	14						
GFH/Ht		3	2				1
Kne/Ht		2	1				
G-M/Ht		3		2		1	
C-M/Ht			4		1		

approximation to an individual standing upright) is the variable entered first and into all the equations. The five other variables which enter most frequently into the 14 regression equations are: (1) age, which enters 7 equations; (2) center of gravity, which enters 6 equations; (3) center of gravity minus the medial malleolus height divided by height, which enters 5 equations; and (5) gluteal furrow height minus the medial malleolus height divided by height, which also enters into 6 equations. Other than these 6 variables, there were 3 variables which entered into three of the equations, 6 variables which entered into 2 of the equations, 10 variables which entered into only 1 equation, and 8 of the variables did not enter any of the equations.

The simple correlations between each of the 33 physical variables and the 13 frequencies plus total power are presented in Table 4. The correlation matrix for total power was inspected to see if the correlations obtained in this study were of the same magnitude as those found in previous studies. The correlation of weight with total sway was .62, and height with total sway was .42. The same correlation for center of gravity times weight was .64. These correlations are much higher

Table 4

Table of simple correlations for each body measure and total power under the waveform.

VARIABLE		TP	.05	.10	.15	.20	.25	.30	.40	.50	.60	.70	.80	.90	1.00
Weight	1	.62	.57	.55	.63	.62	.66	.63	.66	.57	.62	.58	.55	.56	.56
Height	2	.42	.34	.41	.44	.43	.41	.42	.45	.49	.52	.54	.52	.53	.54
BuKnee	3	.37	.32	.36	.41	.40	.38	.37	.35	.36	.42	.44	.42	.45	.46
Should	4	.48	.46	.39	.45	.45	.52	.50	.56	.49	.50	.50	.47	.47	.46
GluFur	5	.20	.12	.22	.25	.25	.21	.22	.19	.22	.26	.30	.31	.31	.34
Knecap	6	.31	.30	.37	.40	.37	.35	.30	.27	.28	.33	.36	.40	.42	.44
SubMal	7	.26	.26	.28	.30	.30	.30	.27	.29	.21	.23	.24	.20	.20	.19
MedMal	8	.40	.41	.34	.38	.29	.31	.28	.37	.37	.40	.41	.37	.38	.37
FootLe	9	.46	.41	.35	.42	.36	.38	.34	.38	.37	.37	.35	.29	.30	.29
Heel B	10	.23	.10	.15	.20	.21	.26	.29	.24	.25	.27	.28	.25	.25	.27
Instep	11	.35	.31	.19	.29	.24	.29	.26	.25	.25	.29	.32	.30	.27	.26
C.Grav	12	.41	.30	.42	.44	.45	.43	.43	.43	.46	.50	.51	.54	.55	.56
Age	13	.02	.10	.04	.05	.07	.06	.03	.11	.07	.16	.14	.13	.16	.15
CG-MMH	14	.37	.26	.40	.42	.43	.41	.41	.40	.43	.47	.48	.51	.52	.54
CGXWgt	15	.64	.56	.58	.66	.65	.68	.65	.67	.60	.65	.63	.61	.61	.62
MOM In	16	.65	.57	.60	.68	.67	.68	.66	.69	.64	.69	.67	.64	.65	.65
GF-MMH	17	.16	.07	.19	.22	.22	.18	.19	.15	.18	.22	.26	.27	.28	.31
GFH/Ht	18	-.14	-.19	-.10	-.09	-.09	-.12	-.12	-.20	-.21	-.17	-.12	-.09	-.09	-.05
Kne/Ht	19	.06	.13	.16	.19	.13	.13	.05	-.04	-.07	-.01	.02	.10	.12	.15
MMH/Ht	20	.15	.22	.09	.12	.02	.05	.02	.10	.06	.07	.07	.05	.05	.02
BKn/Ht	21	-.02	.02	-.03	.01	.00	.02	-.02	-.09	-.13	-.08	-.08	-.08	-.06	-.05
G-M/Ht	22	-.16	-.23	-.12	-.11	-.09	-.13	-.12	-.22	-.22	-.18	-.14	-.10	-.09	-.05
C-M/Ht	23	-.06	-.13	-.02	-.03	.02	.01	.01	-.07	-.09	-.07	-.08	.01	.00	.02
CG/Ht	24	-.03	-.08	.00	.00	.02	.02	.01	-.05	-.08	-.05	-.06	.02	.01	.03
PondIx	25	-.21	-.26	-.15	-.22	-.21	-.27	-.23	-.24	-.12	-.14	-.09	-.07	-.07	-.07
Sub/PI	26	.33	.35	.31	.37	.36	.39	.35	.37	.24	.27	.25	.21	.21	.20
Sub/Wt	27	-.44	-.42	-.36	-.44	-.42	-.48	-.47	-.49	-.46	-.49	-.45	-.44	-.44	-.45
FtL/PI	28	.54	.53	.40	.51	.46	.52	.45	.49	.39	.40	.36	.29	.30	.29
HeB/PI	29	.31	.20	.21	.28	.28	.35	.37	.32	.28	.30	.30	.26	.26	.28
Ins/PI	30	.41	.40	.24	.36	.32	.40	.35	.35	.26	.31	.30	.27	.25	.24
Ftl/Ht	31	.17	.20	.02	.09	.01	.06	-.01	.02	-.03	-.08	-.11	-.18	-.18	-.20
HeB/Ht	32	.06	-.05	-.01	.03	.04	.10	.14	.06	.05	.06	.07	.05	.04	.06
Ins/Ht	33	.02	.04	-.18	-.08	-.13	-.06	-.09	-.13	-.18	-.15	.13	-.14	-.18	-.20

than previous findings have shown. Weight, height, and center of gravity times weight, despite their high correlations with total power, probably did not enter into the multiple regression equations because of their strong intercorrelations with other variables that could account for the same, or even more, variability in the sway patterns, specifically, moment of inertia.

The correlation of age with total power under the waveform is .018. The correlations of age with the 13 separate frequencies range from .025 to .164. The small size of these correlations suggests that age has a negligible positive influence on body sway, but can consistently account for approximately 1% of the total sway variance since it remains independent of body build variables. For this reason, age was included in 7 of the multiple regression equations.

Moment of inertia, which consistently accounts for about 50% of the variance of the sway scores at all frequencies, has a simple correlation with total sway of .653.

In order to test the regression equations, the mean body measures of 6 additional normal male Ss were used in the multiple regression equations to predict the mean sway curve of these Ss. The mean body measures for these Ss are presented in Table 5. Inspection of Table 5 shows that, on the average, the Ss in this sample were 10 Kg heavier and 14 years older than the Ss measured in Experiment I. No other large differences were noted in the physical measures. The actual sway scores, predicted scores, the difference scores, R, the variance, and value of 1.96 standard error of estimates, and the 95% confidence intervals are presented in Table 6, and for convenience are also shown graphically in Figure 4. The actual scores, with the exception of 0.50 and 0.80 Hz, are all within the 95% confidence interval. The prediction equation for 0.80 Hz does not seem to be accurate and it may be that the prediction at this frequency reflects a high sensitivity to the variable CG/Ht or to the variable CGxWgt, or to both, since these two variables did not contribute to the equations at any of the other frequencies. The value at 0.50 Hz is .17 dB lower than the lower value of the 95% confidence interval. On the whole, these results indicate that it is possible to predict an average sway curve for a group of people solely from knowledge of their mean body measures.

The equations that were derived from the data obtained on male Ss were also used to predict sway scores for normal females. Body measures and sway scores

Table 5

Table of mean body measures for 77 normal Ss, 6 normal test Ss, 2 normal female Ss, and 6 L-D Ss.

	Normal N=77 ♂	Normal N=6 ♂	Normal N=2 ♀	L-D N=6 ♂ ; ♀
Weight	75.78Kg	74.75Kg	60.11Kg	73.94Kg
Height	176.34Cm	175.26Cm	167.64Cm	171.77Cm
Should	60.02	59.20	56.45	59.02
BuKnee	45.94	45.98	41.45	45.70
Glufur	76.67	75.28	70.45	73.37
Knecap	48.02	46.80	42.85	46.32
Submal	5.88	6.45	6.25	6.32
Medmal	8.82	8.48	7.90	8.37
FootLe	25.99	25.30	24.10	24.62
Heel B	8.93	9.02	9.30	8.82
Instep	28.30	27.78	26.90	27.27
C.Grav	100.72	100.47	94.05	98.87
Age	20.39yrs	29.50yrs	24.50yrs	31.17yrs

Table 6

Table presenting actual sway scores, predicted sway scores, difference scores, multiple R, variance, and the 95% confidence intervals about the predicted scores at the 13 sway frequencies and for total power for 6 normal test Ss.

	Actual Y	Predict Y	d	R	R^2	95%	Interval Low	High
TP	1.63	1.78	.15	.747	.557	1.28	0.50	3.06
.05	9.01	10.76	1.75	.666	.443	2.09	8.67	12.85
.10	5.09	5.89	.80	.715	.512	2.08	3.81	7.97
.15	5.01	4.91	-.10	.719	.517	1.93	2.98	6.83
.20	3.91	4.06	.15	.729	.532	1.87	2.19	5.93
.25	3.29	3.27	-.02	.742	.550	1.63	1.64	4.90
.30	2.01	1.51	-.50	.738	.545	1.70	-0.19	3.21
.40	0.56	0.34	-.22	.770	.593	1.56	-1.22	1.90
.50	-2.27	-0.54	-1.73	.763	.583	1.56	-2.10	1.02
.60	-3.81	-2.59	1.22	.761	.580	1.68	-4.26	-0.91
.70	-5.70	-4.07	1.63	.734	.538	1.63	-5.70	-2.44
.80	-6.05	-2.85	3.20	.762	.581	1.59	-4.44	-1.26
.90	-6.78	-5.60	1.18	.751	.565	1.62	-7.22	-3.98
1.0	-7.96	-6.59	1.37	.757	.573	1.54	-8.12	-5.05

Figure 4

Graph of the predicted scores for 6 normal test _Ss_, the 95% confidence intervals (two-tailed), and the actual sway scores for 13 frequencies (PSD in dB units _vs_ frequency).

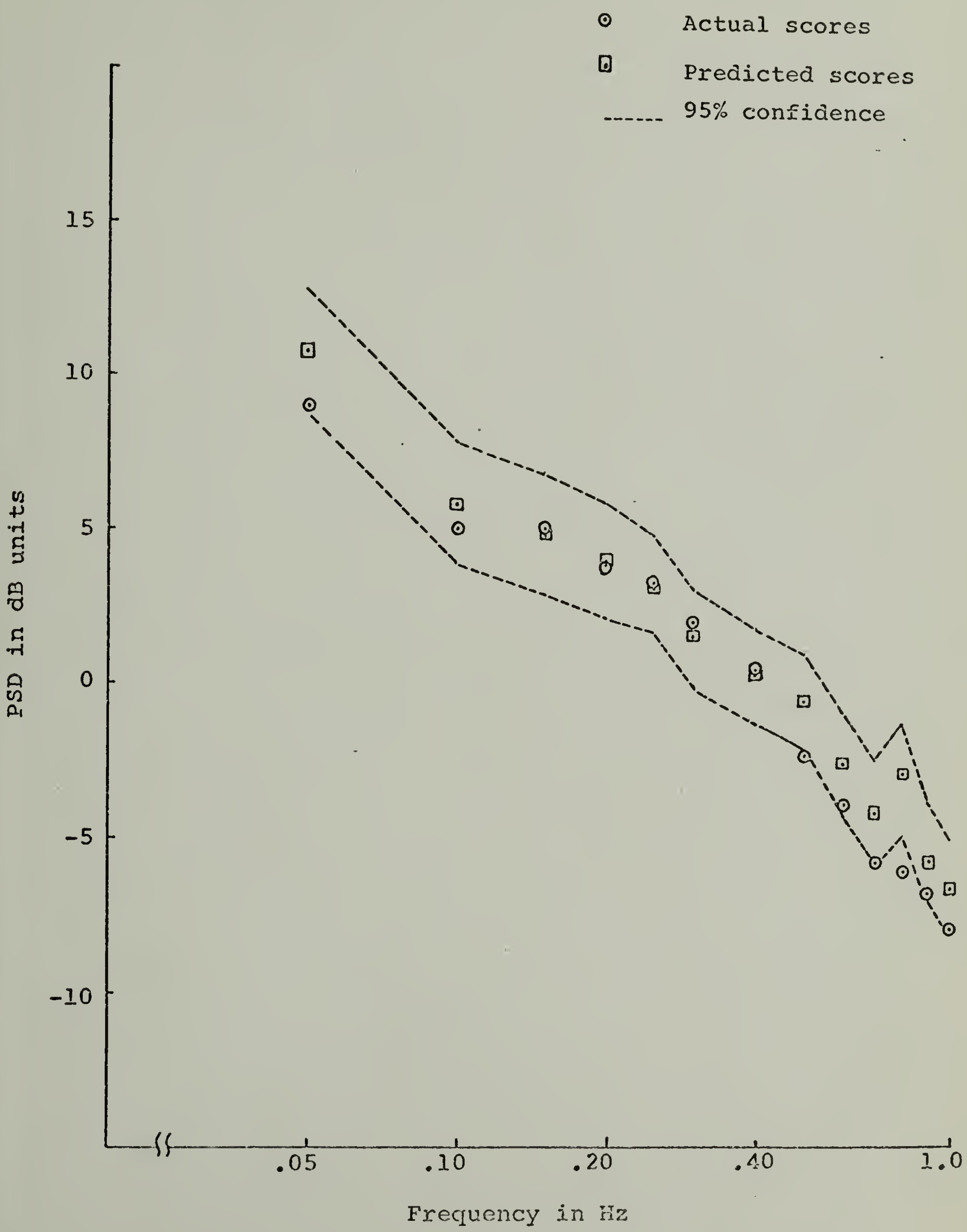

Actual scores
Predicted scores
95% confidence
PSD in dB units
15
10
5
0
-5
-10
.05
.10
.20
.40
1.0
Frequency in Hz

Fig. 4

were obtained from 2 normal female Ss. The mean body measures for the normal females are also presented in Table 5. These measures show that the average female S is distinctly smaller than the average male S and this smaller size difference is reflected by the low predicted sway scores as shown in Table 7 and in Figure 5. Nevertheless, even with this weak test of the prediction equations for female Ss, the equations appear to have some validity in predicting female scores, as evidenced by these results which show that each of the actual sway values fell within the predicted 95% confidence intervals.

Experiment II

Results from Experiment I demonstrated that it is possible to predict a range of values within which actual sway scores for small samples of both male and female Ss will fall. For this reason the data from the 4 male and 2 female L-D Ss were combined, and their actual scores were compared to those predicted by the multiple regression equations. The mean body values obtained from this sample of 6 L-D Ss were very similar to the mean of the male sample used to derive the prediction equations (Table 5). The most noticeable differences were that the L-D Ss were 2 Kg

Table 7

Table showing actual, predicted, and difference scores, plus the 95% confidence intervals about the predicted scores for 2 normal female *Ss*.

	Actual Y	Predict Y	d	95% Interval	
TP	1.15	0.78	-.37	-1.43	2.99
.05	9.53	7.98	-1.55	4.35	11.60
.10	4.75	5.43	0.68	1.83	9.03
.15	4.06	1.77	-2.29	-1.57	5.10
.20	2.10	1.49	-.61	-1.75	4.73
.25	1.07	0.72	-.35	-2.10	3.53
.30	-.39	-.86	-.47	-3.81	2.08
.40	-2.40	-2.30	0.10	-5.00	0.41
.50	-3.54	-4.29	-.75	-6.99	-1.59
.60	-5.76	-5.95	-.19	-8.85	-3.04
.70	-6.82	-7.20	-.38	-10.02	-4.38
.80	-7.44	-7.44	0.00	-10.19	-4.68

Figure 5

Graph of predicted scores for 2 normal female Ss, the 95% confidence interval (two-tailed), and the actual sway scores for the 13 frequencies (PSD in dB units *vs* frequency).

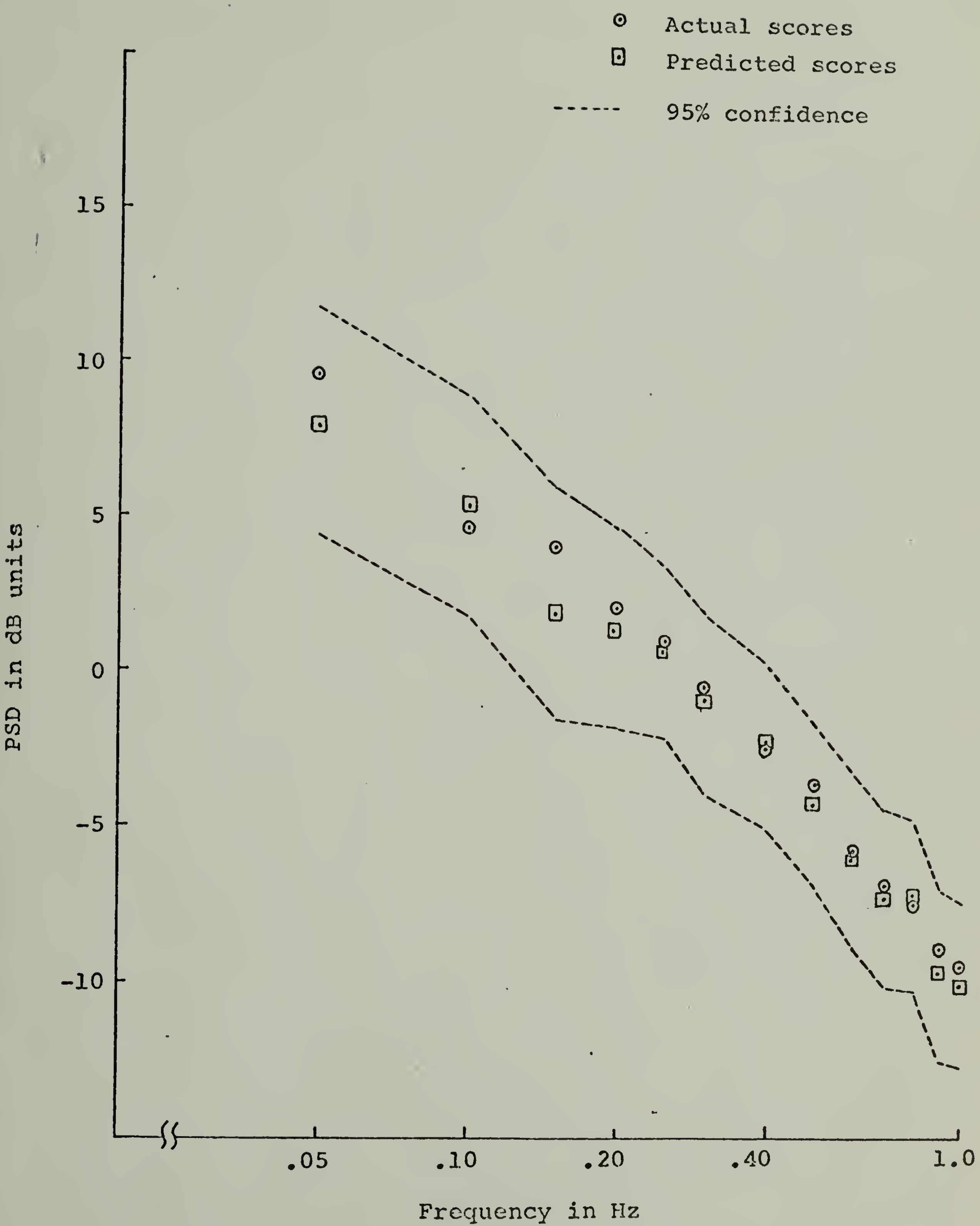

Actual scores
Predicted scores
95% confidence
15
10
5
0
-5
-10
PSD in dB units
.05
.10
.20
.40
1.0
Frequency in Hz

Fig. 5

lighter, 5 cm shorter and 11 years older.

Table 8 gives the actual scores, the predicted scores, the difference scores, and the 95% confidence intervals for the 13 frequency values plus total power under the curve. The same data is also presented in Figure 6, which shows that the power at frequencies 0.15, 0.20, 0.25, 0.30 and 0.40 Hz are significantly different ($p<.05$) from the predicted values. All other values fall within the confidence interval established by plus or minus 1.96 standard error of estimate from the predicted values. The shape of the curve indicates that the proportional and the first order system are still present. However, the two slope lines, when extrapolated, cross at 0.185 Hz instead of 0.21 Hz found for the 77 normal Ss. The slope of the L-D curve from frequency 0.185 to 1.0 Hz, as determined visually, is -6.0 dB/octave.

Since the differences between the mean body measures of the 6 L-D Ss and the 77 normal Ss were small, the mean curve for the normal Ss would, in addition to the predicted curve, be a reasonable representation of a "normal" or control curve for thr L-D Ss given that they had normally functioning labyrinths. This comparison is presented in Figure 7 which shows that the curve for the L-D Ss closely

Table 8

Table showing actual, predicted, difference scores, and the 95% confidence interval about the predicted scores for 6 L-D Ss.

	Actual Y	Predict Y	d	Interval Low	High
TP	4.46	2.26	-2.20	0.98	3.54
.05	12.05	10.59	-1.46	8.50	12.69
.10	8.38	7.33	-1.05	5.25	9.41
.15	8.49	4.42	-4.07	2.50	6.35
.20	6.63	3.60	-3.03	1.73	5.47
.25	5.77	2.72	-3.05	1.10	4.35
.30	4.32	0.83	-3.49	-0.86	2.53
.40	1.69	-0.22	-1.87	-1.78	1.34
.50	0.05	-1.37	-1.42	-2.93	0.19
.60	-2.19	-3.27	-1.08	-4.95	-1.59
.70	-3.09	-4.78	-1.69	-6.40	-3.15
.80	-3.98	-3.34	0.64	-4.93	-1.75
.90	-5.61	-6.25	-.64	7.87	-4.64
1.0	-6.80	-6.54	0.26	-8.07	-5.00

Figure 6

Graph of predicted sway scores, its 95% confidence interval, and the actual sway scores for 6 L-D _S_s at 13 separate frequencies (PSD in dB units _vs_ frequency).

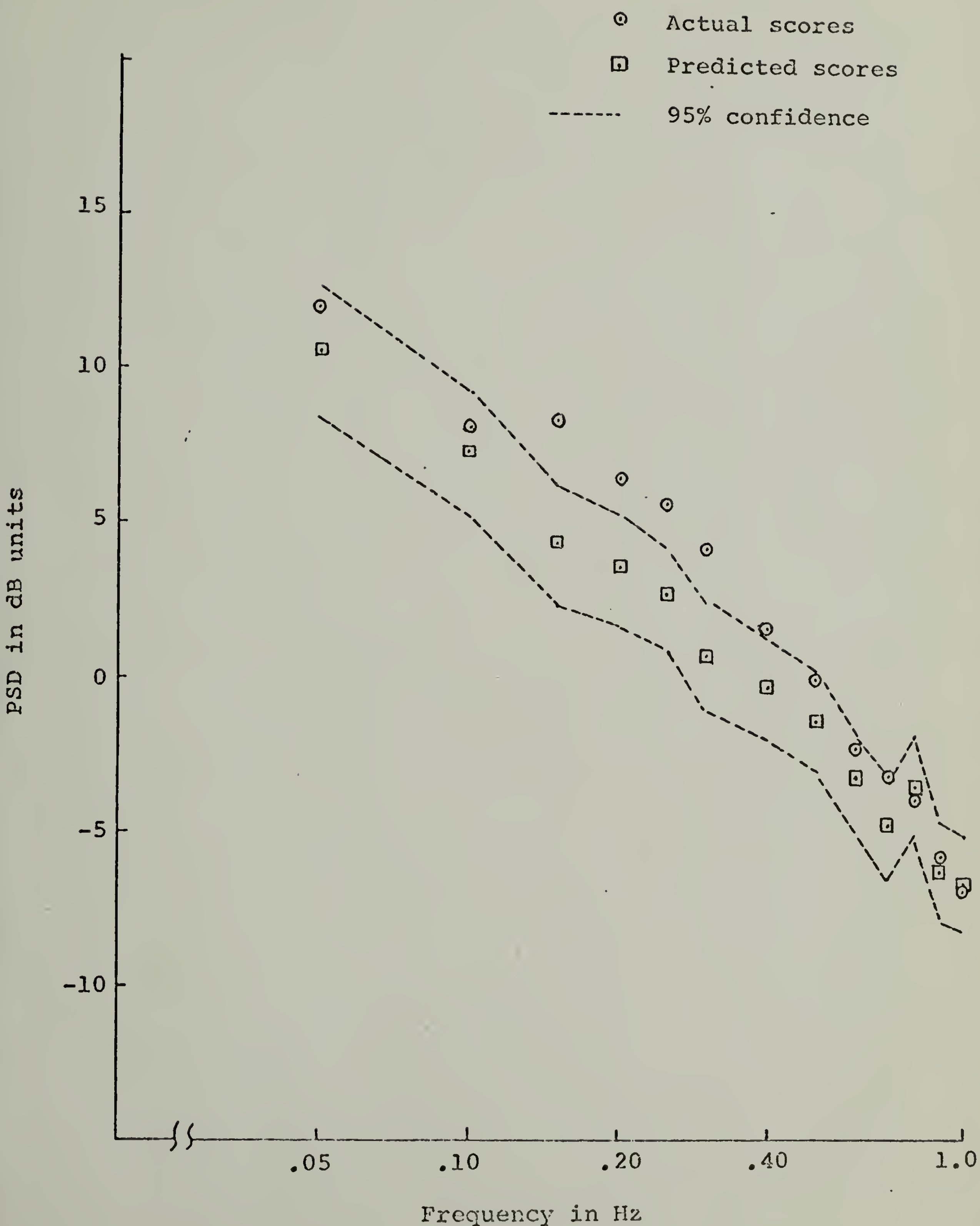
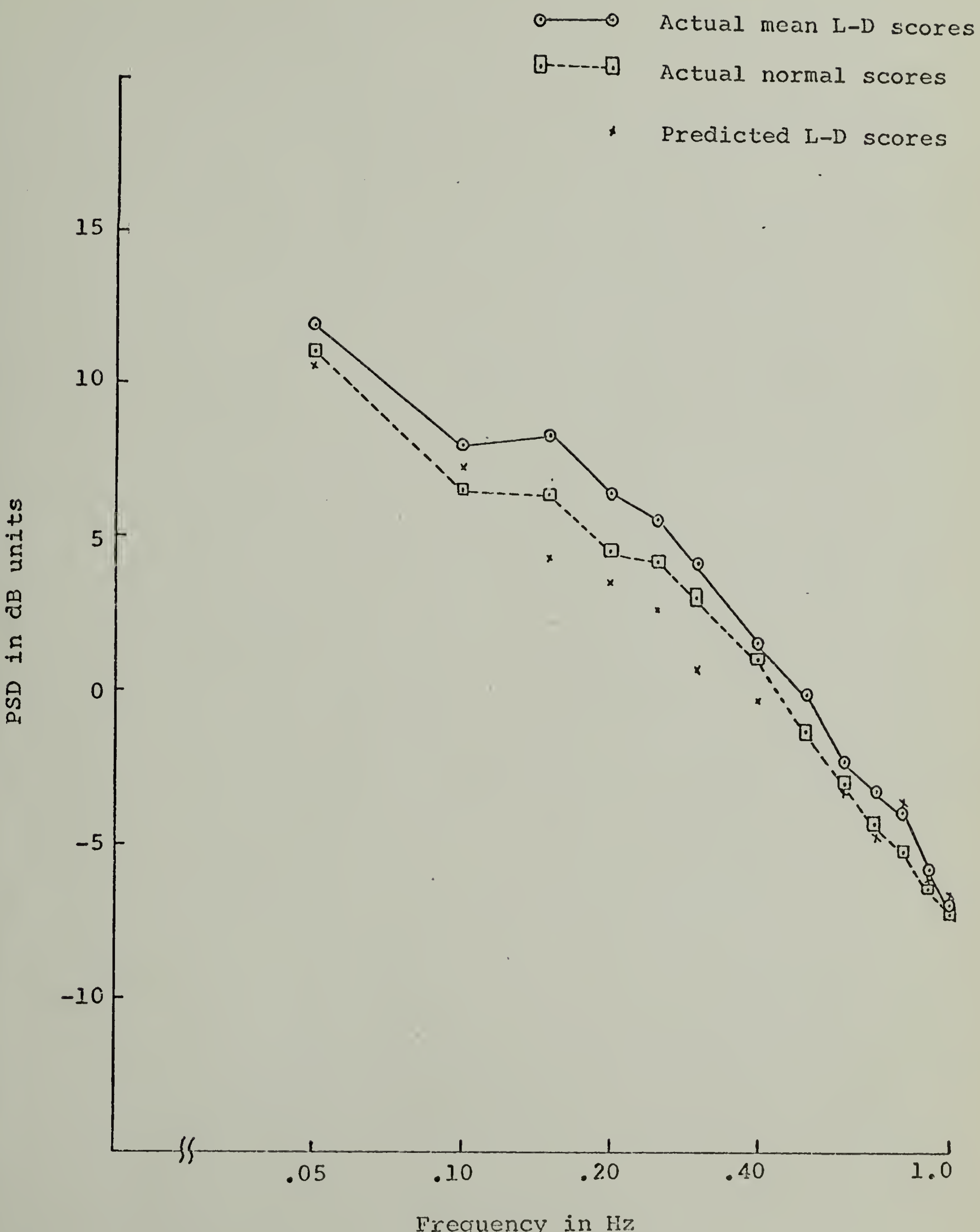

Actual scores
Predicted scores
95% confidence
15
10
5
0
-5
-10
PSD in dB units
.05
.10
.20
.40
1.0
Frequency in Hz

Fig.6

Figure 7

Graph depicting the actual mean sway scores for the 77 normal Ss and the 6 L-D Ss. The predicted scores for the L-D Ss are also presented (power in dB units vs frequency).



Actual mean L-D scores
Actual normal scores
Predicted L-D scores
15
10
5
0
-5
-10
PSD in dB units
.05
.10
.20
.40
1.0
Frequency in Hz

Fig. 7

closely parallels the sway pattern for the normal Ss. The average increase in power is greatest (1.8 dB) at the frequencies lying within the range of the proportional system, while the increase in power at the 7 lowest frequencies averages .75 dB. The predicted scores for the 6 L-D Ss are also presented in Figure 7, and are slightly lower than the mean curve for the normal subjects. This result is expected since the mean L-D body scores are also slightly lower.

Discussion

Early findings using standing sway as a response measure established a few general trends concerning variables which contribute to, or influence, sway. The trends established by previous investigators have been questionable due to the poor methodological controls used in those studies. The study presented here institutes a number of procedures designed to control as many variables as possible which have been previously investigated as influencing body sway. In this respect, physiological rhythm and sex differences were controlled by using only the data from male Ss to derive the multiple regression equations in Experiment I. Furthermore, all Ss were tested using the same foot position, and wearing equivalent footwear (i.e., none). Each S was blindfolded, and, to control attention, was required to mentally calculate arithmetic problems. All Ss were naive and, to avoid practice effects, only the first four minutes of data were used to compute the PSD sway patterns.

Respiration rate and cutaneous cues were the two major variables not controlled in this study. As mentioned before, if respiration rate influences sway, its effects would be reflected in the sway curve at 0.2 to 0.30 Hz, and should be the subject of a future study.

The obtained sway curves reflect the basic postural mechanisms, plus the influences of physical characteristics, age and respiration rate. Results from Experiment I show that it is possible to predict, with approximately 55% accuracy, PSD sway patterns for a small group of Ss solely from knowledge of their physical dimensions, providing that certain factors are adequately controlled.

The simple correlations of individual body characteristics with a measure of total sway are considerably higher than those obtained from previous research. Fearing (1942a) found the correlation between weight and sway to be .15, while the same correlation in the present study was .62. Likewise, Fearing found that the r for height and sway was .22, while the correlation obtained for the present data was .42. These large differences can only be explained by assuming that the recording methods employed in the two studies are not measuring equivalent responses.

Bensel, et al. (1968) obtained an r of .01 between the product of the center of gravity and weight, and the mean amplitude of the lateral sway response for male Ss. The r obtained for the same correlation in this study was .64. This large difference is most likely the result of the differences, again, in the criterion response measures.

Edwards (1942) found that age has no effect on body sway, since sway responses remain stable from age 14 through age 70. The low correlation found for age and sway here (r=.018) partially confirms the findings of Edwards for an age group spanning 12 years (18-30 years). However, the distribution of age in the present study is positively skewed and the correlations thereby obtained may not be a true representation of the actual correlation for a population of Ss in this age range.

The body variable that correlates most highly with each sway frequency is moment of inertia. This variable has an average simple correlation of .65 with each frequency tested. The grouping of individuals according to their moments of inertia would be a simple and fast procedure, since it involves obtaining only weight and height measures on each S.

Sway prediction based solely on the moment of inertia variable would be much less accurate than the complete regression equations presented in Table 2, and therefore not as useful for most research. However, a simpler method which uses only five basic body measures is suggested in Table 1 of Appendix B. These five measures (and the variables derived from them) were selected after examining the complete multiple regression equations and determining those variables which removed the greatest amount of variability and, at the same time, entered the greatest number of regression equations. Overall, this simpler method can account for an average of 44% of the sway variability which is 11% less than the full regression equations.

Results from Experiment II, in contrast to earlier findings, show that L-D Ss sway more than normal Ss, and that this increase in sway occurs predominantly at the lower frequencies. The conclusion that L-D Ss sway more at the lower frequencies is supported by the results of three separate comparisons. In the first, the mean PSD sway curve of the L-D Ss was compared to the sway curve predicted by the multiple regression equations from measures of their physical characteristics. The second and third compared the mean curve of the L-D Ss to the mean curve of the 77 normal Ss and to the 6 normal test Ss. The power at the lower frequencies for the L-D sway curve was greater than for any of the three comparison curves.

The 6 L-D Ss evidence no difficulties in maintaining the upright sway position, with eyes blindfolded, during the 4 minute test session. There was only one near fall during the 24 minutes of total sway time for the group. There were no reports from the L-D Ss of general muscular weakness or fatigue. As a result of these observations, the sway records of the L-D Ss were assumed to be valid representations of a normal postural system without the influence of labyrinthine impulses.

Each L-D S was tested twice, the second test being conducted with the S's eyes open. This procedure was adopted so that "eyes closed" sway records would include only the first four minutes of sway from naive Ss and thus make the data comparable to the data of

Experiment I. The "eyes open" data, for methodological reasons, is not presented in the body of the text, but is included in Figure 1 of Appendix B. The "eyes open" test was performed after a five minute rest period. The two opposite effects of fatigue and practice may combine to make these curves reasonably representative of what would have been expected had the two conditions been properly counterbalanced. If the "eyes open" curve is taken to be reasonably accurate, then the general conclusion can be made that vision has an influence on sway at all frequencies tested. The effect is a decrease in sway but the overall shape of the curve has not changed.

The original problems of Experiment II have been satisfactorily answered. The otolithic organs are an integral part of the natural postural system and, in this role, they appear to exert their influence specifically at the lower frequency range of the PSD sway curve. The present data also suggest that it is possible to proceed with proposing a physical model of how the otoliths interact with the postural system to affect sway. Of course, any proposed physical model must also be supported by known or suspected physiological and neurophysiological correlates.

The observation that L-D Ss can maintain balance with their eyes closed suggests that the PSD curve for this group represents only the effects of the proprioceptive system incombination with the tactile cues of the feet. Since the general shape and slopes of the sway curves for both normal Ss and L-D Ss are similar, it can be assumed that the first order or linear system proposed in Figure 3 (and the PSD sway curve itself) is predominantly a measure of the effects of both proprioceptive and tactile cues.

The assumption that the normal PSD sway curve is predominantly proprioceptive may have some validity since two other investigators have found evidence, and proposed linear or first order transfer functions, for the isometric twitch of striated muscle (Gilson, 1944), and for a model of the monosynaptic reflex arc (Jones, 1969). The physiological aspects of the proprioceptive system are well known and a brief sketch of this system is given below.

As mentioned earlier, the anti-gravity muscles and their receptors provide the predominant control for standing equilibrium. The muscles (extrafusal fibers) that supply the force and support the weight of the body are intimately associated with the smaller stretch receptors or muscle spindles (intrafusal fibers). The stretch receptors act to register their own degree of stretch. These small muscle spindles are divided into three areas: the two end areas are innervated by gamma efferent fibers and contain contractile elements whereas the middle area of the spindle is the receptor proper and signals (via the alpha afferent fibers) the state of tension of the muscle spindle established either by its own contractile elements or by the action of the larger extrafusal fibers in which the spindle is imbedded. The muscle spindles are arranged in a self-excitatory or positive feedback system such that the afferent output from the receptor portion excites the contractile areas, causing them to contract. As the spindle contracts, the stretch receptor is stimulated to fire at a greater rate, and, in turn, continues to excite the contractile elements. The self-excitation is checked by a built-in negative feedback system. The same afferent impulses that excite the gamma fibers also act to excite the alpha efferent neurons that supply the extrafusal fibers surrounding the muscle spindles which excite them. The ensuing extrafusal contraction release tension on the muscle spindle and the stretch receptor ceases firing

(Stanley-Jones, 1960).

In passive or static equilibrium the pull of gravity causes the anti-gravity muscles and their associated muscle spindles to stretch. The stretch is registered and signalled to the spinal cord. Simultaneously, the contractile elements of the spindles and the large extensor muscles are stimulated to contract. The contraction of the muscle spindle further excites the receptor portion, which in turn acts to increasingly stimulate the large alpha efferent fibers. This insures that these fibers will fire and cause the extrafusal fibers to contract in order to counter the effect of gravity. The contraction of the extrafusal fibers then inhibits the positive feedback loop of the stretch receptor system by releasing the tension on them. A schematic diagram of the muscle-stretch receptor arc is presented in Figure 8.

The combination of positive and negative feedback of the muscles and their stretch receptors constitutes the basic control of equilibrium. The system, however, is not very finely adjusted and oscillation is introduced by the process of correction and countercorrection. Refinement, stabilization, or damping of this basic process occurs principally from those sensory cues

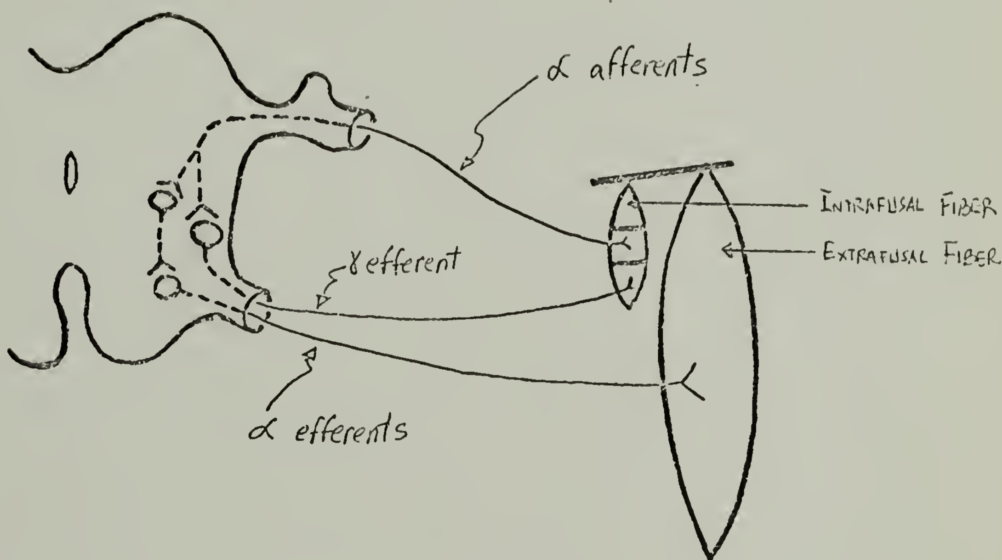

α afferents
INTRAFUSAL FIBER
EXTRAFUSAL FIBER
γ efferent
α efferents

Figure 8

already discussed, i.e. labyrinthine, visual, and cutaneous. (Lowenstein, 1966).

A control system model has been derived (see Appendix D) in which the cutaneous cues are included with the proprioceptive aspects of posture. This model was assumed since these two sensory systems were not separated and both were present in the data from these experiments.

The model has shown that the value 8.13 dB (see Appendix D) is the value of the power that would be predicted for the L-D Ss' mean curve at the asymptotic section of the curve (i.e. for frequencies of .05- .20 Hz). The actual obtained value for the L-D Ss is 8.5 dB and the prediction from the proposed model closely agrees with the actual data. The model appears to be a good first approximation in describing actual processes involved in standing equilibrium. The quantities a, K_1 and K_2 and the values for the break frequencies and asymptotes can be useful tools in future model buildi-- where the interactions of other sensory systems with standing equilibrium are studied. For convenience these quantities are presented again in Table 1 of Appendix B.

Summary

In summary, the major results from this study show

(1) that approximately 55% of the variability in sway patterns can be attributed to differences in the physical characteristics of the individual tested, (2) that sway responses can be predicted from knowledge of these physical characteristics, and (3) that the prediction procedure is a valid method for establishing "normal" or control curves where these cannot be readily obtained. In addition, data from the L-D group indicate that these _S_s consistently sway more at the lower frequencies than normal _S_s and also more than what would be predicted had L-D subjects had properly functioning otoliths.

Furthermore, the idea that the otoliths do not contribute to standing equilibrium because of their low sensitivity and weak conscious impressions does not seem to be tenable. The threshold data for otolith stimulation indicates that 2 - 4 deg of tilt is necessary in order for the normal person to perceive that he has been tilted. The current data, however, shows that the otoliths play a more continuous role in their influence on the postural mechanisms. The impulses that are sent to the spinal cord during static equilirium are reflexive and they may rarely reach consciousness, except when the tilt becomes large (2 deg - 4 deg).

This concept of otolith function also explains why the typical L-D S has no difficulty in maintaining upright balance while standing stationary with eyes closed, even though he may not be able to walk alone on a dark night without falling. It would seem from this anecdotal information that the small tilt deviations incurred during static equilibrium are handled on a tonic reflexive level, while the larger tilts found during dynamic movements are relayed not only to the spinal cord, but also to the higher nervous centers.

Finally, it has been shown that a control system approach to standing equilibrium can be a fruitful framework in which to analyze the various influences that affect sway.

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Footnotes

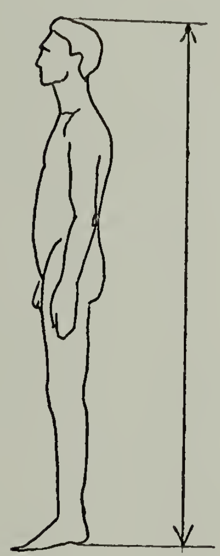
1. The term "average power" is used regardless of the dimensional units of the varying quantity being measured. The useage is generally accepted even though it is strictly correct in only certain cases (e.g., when the quantity is the voltage across a one ohm resistor).

2. This presumes a white noise signal input (i.e., equal amplitude signal at each frequency). The input signal for body sway is presumed to arise from oscillations produced by random fluctuations in muscle tonus, respiration, etc.

APPENDIX A

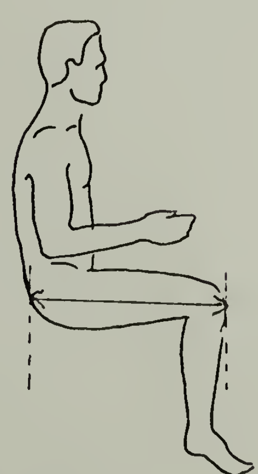
WEIGHT

Weight is measured with S in stocking feet and underwear.



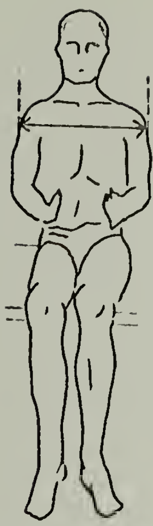
HEIGHT

S stands erect in stocking feet with eyes level. Measure vertical distance from floor to top of head.



BUTTOCK-KNEE LENGTH

S sits erect with lower part of leg hanging over edge of chair. Upper section of leg is parallel to chair. Measure distance from back of buttock to furthermost extension of knee.

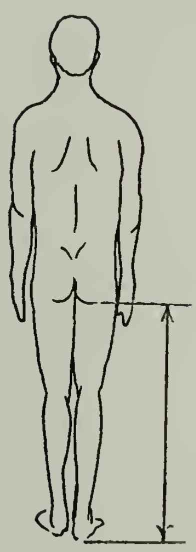


SHOULDER BREADTH

S sits erect, his upper arms hanging at his sides and his forearms extended horizontally. Measure the horizontal distance across the maximum lateral protrusion of the deltoid muscles.

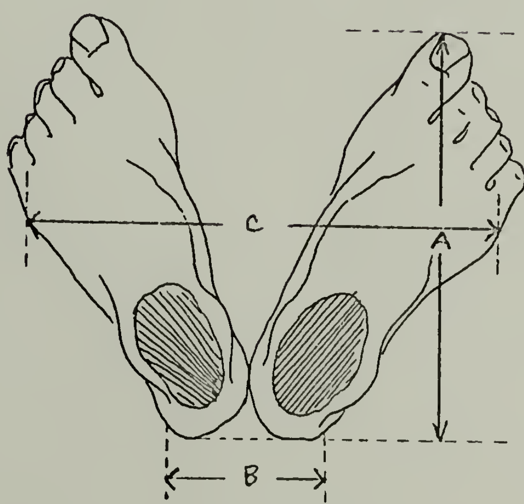
GLUTEAL FURROW HEIGHT

S stands erect. Measure vertical distance from floor to lowest point where buttock curve and back of thigh join.



MEDIAL MALLEOLUS HEIGHT

S stands with weight equally distributed on both feet. Measure from floor to most projecting point of medial malleolus bone of the right foot.



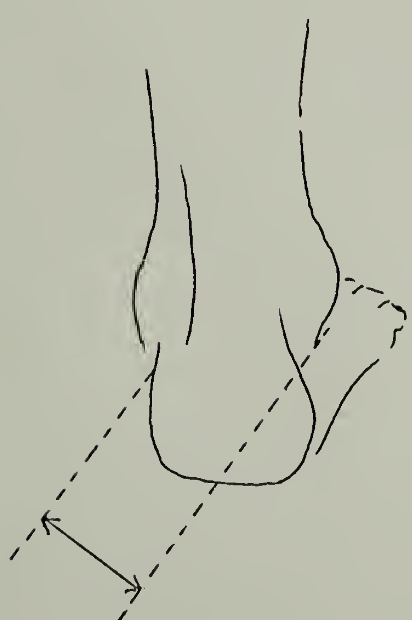
FOOT MEASURES

A=instep breadth
B=heel breadth
C=foot length

S stands in the 45^0 foot position used in standing sway. All measures are as shown.

KNEECAP HEIGHT

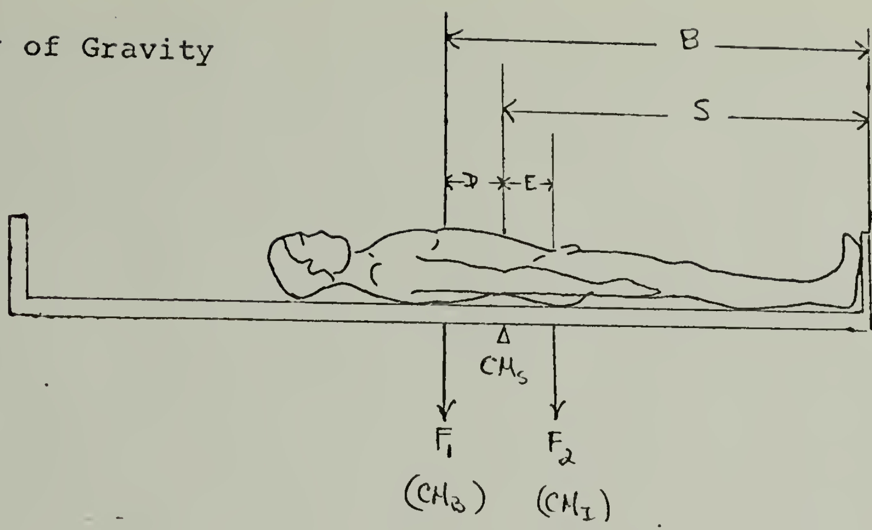
S stands erect. Measure vertical distance from floor to uppermost point of right kneecap.



SUBMALLEOLUS BREADTH

Breadth measured from the back of the foot at the level just below the malleolus bone.

Center of Gravity



Since the forces on each side of the fulcrum (CG_s) must be equal, then

$$F_1 = F_2$$

where $F_1 = Wt_B \times D$ and $F_2 = Wt_I \times E$ and

Wt_B = weight of the board and Wt_I = weight of the individual

then: $$Wt_B \times B = Wt_I \times E$$

and: $$E = \frac{Wt_B \times D}{Wt_I}$$

and: $$D = B - S$$

then substituting: $$E = \frac{Wt_B \times (B-S)}{Wt_1}$$

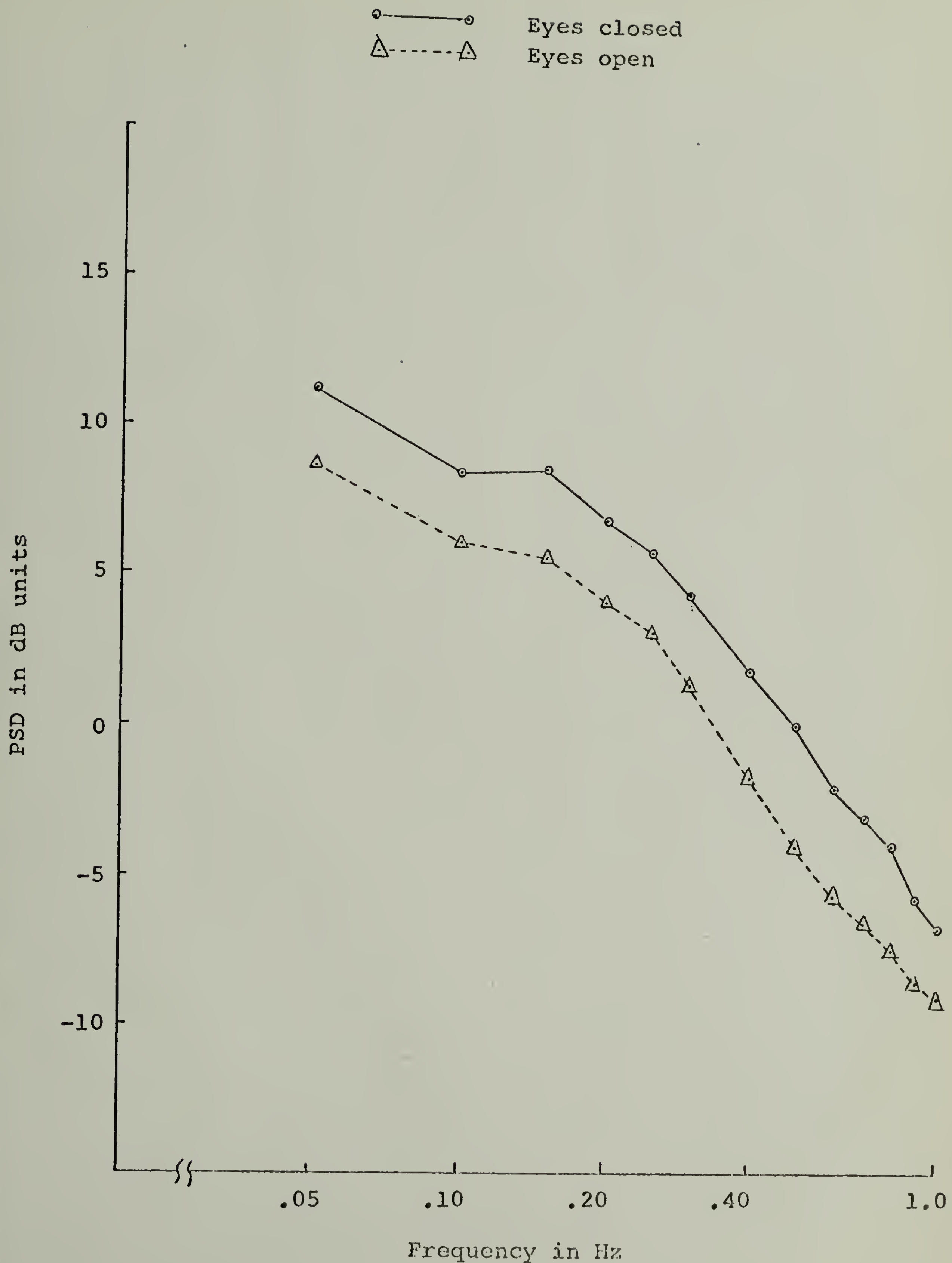
and finally, knowing E, it is possible to find CG_I by the equation: $$CG_I = S - E$$

or, $$CG_I = \frac{S(1-Wt_B)-Wt_B B}{Wt_I}$$

Appendix B

Figure 1

Graph of the mean PSD curve of "eyes open" condition <u>vs</u> "eyes closed" condition for the 6 L-D <u>S</u>s.



Eyes closed
Eyes open
15
10
5
0
-5
-10
PSD in dB units
.05
.10
.20
.40
1.0
Frequency in Hz

Fig. 1B

Table 1

Table presenting the b_i weights for the simplified multiple regression equations derived from the data of Experiment I.

	TP	.05	.10	.15	.20	.25	.30
Const	14.77	50.18	9.87	16.43	-18.03	-13.13	-14.65
MomIn 1	.02	.03	.02	.03	.03	.03	.03
C-M/Ht 2					106.08	102.51	101.61
GFH/Ht 3			-149.07		-124.64	-127.47	-128.79
G-M/Ht 4	-69.23	-130.18		-74.51			
CGrav 5			.51				

	.40	.50	.60	.70	.80	.90	1.0
Const	2.65	3.67	-6.77	-17.06	-29.62	-33.27	-50.06
MomIn 1	.03	.01	.02	.02	.01	.01	.01
C-M/Ht 2	79.65						
GFH/Ht 3	-152.48	-205.01	-200.12	-166.19	-175.32	-175.31	
G-M/Ht 4							-156.73
CGrav		.712	.73	.68	.85	.87	.87

Table 2

Derived values for variables in the proposed physical model for standing sway.

Quantity	Value
$G(jw)$ normal	7.0 dB
$G(jw)$ L-D	8.5 dB
f_{br} normal	0.21 Hz
f_{br} (L-D)	0.185 Hz
a	1.16
K_1	1.96
K_2	.05

Appendix C

```
      PROGRAM PREDICT
      DIMENSION VM(40),Y(30),R(30),SEE(30),RSQ(30),RL(30),RH(30),
     1V(50,50),SE(30)
    5 FORMAT (4X,F6.2,1X,F5.2,1X,9(F4.1,1X),F5.1,1X,F2.0)
    6 FORMAT (44X,*RANGE*)
    7 FORMAT (10X,*Y*,7X,*R*,8X,*RSQ*,5X,*95 PC *,5X,*LOW*,5X,*HIGH*,/)
    8 FORMAT (8X,F6.2,3X,2(F6.4,2X),2(F6.2,3X),F6.2)
    9 FORMAT (3X,I2)
   10 FORMAT (3X,2(F9.2,1X))
      READ(60,9)N
      DO 15 I = 1,N
   15 READ(60,5) (V(I,J),J = 1,13)
      DO 100 I = 1,N
      V(I,1) = V(I,1)*0.45454
      V(I,2) = V(I,2)*2.54
      V(I,14) = V(I,12)-V(I,8)
      V(I,15) = V(I,12)*V(I,1)
      V(I,40) = V(I,2)*V(I,2)
      V(I,16) = (V(I,40)*V(I,1))/3000.
      V(I,17) = V(I,5)-V(I,8)
      V(I,18) = V(I,5)/V(I,2)
      V(I,19) = V(I,6)/V(I,2)
      V(I,20) = V(I,8)/V(I,2)
      V(I,21) = V(I,3)/V(I,2)
      V(I,22) = (V(I,5)-V(I,8))/V(I,2)
      V(I,23) = (V(I,12)-V(I,8))/V(I,2)
      V(I,24) = V(I,12)/V(I,2)
      V(I,39) = V(I,1)**.3333
      V(I,25) = V(I,2)/V(I,39)
      V(I,26) = V(I,7)/V(I,25)
      V(I,27) = V(I,7)/V(I,1)
      V(I,28) = V(I,9)/V(I,25)
      V(I,29) = V(I,10)/V(I,2 )
      V(I,30) = V(I,11)/V(I,25)
      V(I,31) = V(I,9)/V(I,2)
      V(I,32) = V(I,10)/V(I,2)
      V(I,33) = V(I,11)/V(I,2)
  100 CONTINUE
      DO 150 J = 1,33
      VM(J) = 0.0
      DO 140 I = 1,N
      VM(J) = VM(J)+V(I,J)
  140 CONTINUE
      B = N
      VM(J) = VM(J)/B
  150 CONTINUE
      Y(1)=56.990-(1.604*VM(2))+(.574*VM(12))-(.263*VM(13))
     1+(.084*VM(16))-(141.293*VM(22))+(3.807*VM(25))+(113.188*VM(31))
      Y(2) = 19.546+(1.115*VM(9))+(.019*VM(16))+(178.545*VM(19))
     1 -(232.645*VM(22))
      Y(3)=9.352+(1.191*VM(6))-(13.849*VM(7))-(1.374*VM(11))
     1 +(1.081*VM(12))+(.094*VM(16))-(1.611*VM(17))-(2.163*VM(25))
     1 +(1184.194*VM(27)) -(.442*VM(13))
      Y(4)=-23.020+(.750*VM(6))-(.373*VM(13))+(.025*VM(16))-(247.677*
     1VM(22))+(159.-50*VM(23))
      Y(5)=-6.893-(.335*VM(3))-(1.235*VM(11))-(.437*VM(13))
```

```
     1 +(.054*VM(16))-(115.443*VM(18))+(144.682*VM(23))+(156.908*VM(27))
      Y(6)=-14.606-(.950*VM(5))-(.378*VM(13))+(.808*VM(14))+(.031*VM(16
     1))
      Y(7)=-6.494-(.527*VM(13))+(.038*VM(16))-(187.535*VM(18))
     1 +(176.962*VM(23))-(25.158*VM(30))
      Y(8)=19.293-(.347*VM(13))+(.033*VM(16))-(181.910*VM(18))
     1 +(124.822*VM(23))-(134.984*VM(33))
      Y(9)=-(20.671)+(.640*VM(4))+(.799*VM(12))+(.007*VM(16))
     1 -(100.707*VM(19))-(140.301*VM(22))-(96.238*VM(33))
      Y(10)=-6.176+(.732*VM(12))+(.018*VM(16))-(200.116*VM(18))
      Y(11)=-23.853+(.737*VM(12))+(.015*VM(16))-(40.450*VM(19))
     1 -(155.362*VM(22))
      Y(12)=-136.787+(.702*VM(4))-(.010*VM(15))+(.116*VM(16))
     1 -(188.295*VM(16))+(346.227*VM(24))-(55.232*VM(28))
      Y(13)=-44.897+(.670*VM(4))+(1.017*VM(12))+(.002*VM(16))
     1 -(169.909*VM(18))-(173.393*VM(31))
      Y(14)=-43.948+(.703*VM(6))+(.032*VM(16))-(212.086*VM(22))
     1+(180.295*VM(23))-(67.306*VM(28))
      R(1)=.7406
      R(2)=.6655
      R(3)=.7152
      R(4)=.7193
      R(5)=.7293
      R(6)=.7417
      R(7)=.7383
      R(8)=.7700
      R(9)=.7634
      R(10)=.7014
      R(11)=.7337
      R(12)=.7021
      R(13)=.7514
      R(14)=.7509
      SE(1)=4.663
      SE(2)=7.009
      SE(3)=7.439
      SE(4)=6.927
      SE(5)=6.634
      SE(6)=6.051
      SE(7)=6.298
      SE(8)=6.118
      SE(9)=6.030
      SE(10)=6.470
      SE(11)=5.991
      SE(12)=6.146
      SE(13)=6.127
      SE(14)=5.877
      DO 30 I = 1,14
      SE(I) = SE(I)/2.
      Y(I) = Y(I)/2.
   30 CONTINUE
      DO 200 I = 1,14
      SEE(I) = SE(I)*(1.-(R(I)*R(I)))**0.50
      SEE(I) = SEE(I)*1.96
  200 CONTINUE
      DO 210 I = 1,14
      SEE(I) = SEE(I)/(B**.50)
```

```
210 CONTINUE
      DO 220 I = 1,14
      RSQ(I) = R(I)*R(I)
220 CONTINUE
      DO 240 I = 1,14
      RL(I) = Y(I)-SEE(I)
      RH(I) = Y(I)+SEE(I)
240 CONTINUE
      WRITE(61,5)
      WRITE(61,7)
      WRITE(61,8)(Y(I),R(I),RSQ(I),SEE(I),RL(I),RH(I),I = 1,14)
      DO 250 J = 1,33
250 WRITE(61,10)(V(I,J),I = 1,N),VM(J)
      STOP
      END
```

```
1 [illegible]
2 [illegible](1[illegible]),[illegible](1[illegible]),[illegible](1[illegible]),[illegible]([illegible]),[illegible]([illegible])
3 [illegible]
4 [illegible]=1
5 [illegible]=5
6 [illegible] [illegible]=1,[illegible]
7 [illegible], ([illegible]([illegible]),I=[illegible], )
10 [illegible]=[illegible]+5
15 [illegible]=[illegible]+5
20 CONTINUE
30 DO 40 I=1,1[illegible]
31 [illegible](I)=[illegible]([illegible](I))
40 [illegible](I)=[illegible](I)/1[illegible].
45 [illegible]=[illegible]
50 [illegible]=[illegible].[illegible]
51 [illegible]=1[illegible]
52 [illegible]=0.[illegible]
53 [illegible]=50
54 [illegible]=[illegible]+1
55 [illegible]=1
56 [illegible]=0
60 [illegible]([illegible],[illegible]/1[illegible]/[illegible],/,1[illegible],
[illegible],I4,[illegible],I3,[illegible]
[illegible],/,[illegible],I5,[illegible]
[illegible],[illegible].3,[illegible])
70 [illegible](/,[illegible] = [illegible],[illegible]11.[illegible],/)
80 [illegible] 60,[illegible],[illegible],[illegible]
90 [illegible]=0.0
100 DO 110 I=1,[illegible]
110 [illegible]=[illegible] (I)
120 [illegible]=[illegible]/1[illegible].
130 [illegible] 70,[illegible]
140 [illegible]=[illegible]
150 [illegible]=[illegible]/[illegible]
160 [illegible]=[illegible]
170 [illegible] [illegible] [illegible]=1, [illegible]
180 [illegible]=[illegible]-1
190 L=[illegible]-[illegible]
200 [illegible]=0.0
210 DO 230 I=1,L
220 [illegible]=I+[illegible]
230 [illegible]=[illegible]+ (I)[illegible](I[illegible])
240 [illegible]=0.[illegible]
250 [illegible] 260 I=[illegible],L
260 [illegible]=[illegible]+ (I)
```

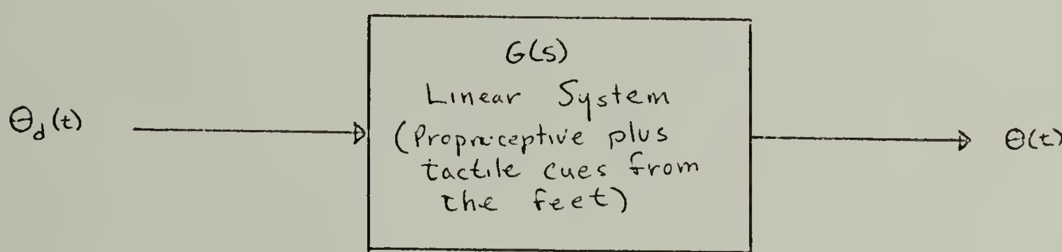
```
270 [illegible]=[illegible]
280 [illegible]I( )=( [illegible]-[illegible]-[illegible]*[illegible])/[illegible]
290 CONTINUE
300 [illegible](1 ,[illegible]7.3,[illegible]20.4)
310 [illegible](1 ,[illegible],10 ,[illegible])
320 DO 370 I=1,[illegible]
330 J=I-1
340 [illegible]=[illegible](J)
350 [illegible]=[illegible]/(5.0)
360 [illegible]=[illegible]I(I)/[illegible]I(1)
360 CONTINUE
370 CONTINUE
380 [illegible] 510
390 [illegible] 520
400 DO 500 J=1,[illegible]
410 [illegible]=0.0
420 [illegible]=J-1
430 [illegible]=[illegible]-1
440 [illegible] =
450 DO 490 [illegible]=1,
460 [illegible]=I
470 [illegible]=(3.14159[illegible])/[illegible]
480 [illegible]=1.0-[illegible]/[illegible]
490 [illegible]=[illegible]I([illegible]+1)*[illegible]([illegible])
500 [illegible](J)=([illegible]I(1)+[illegible].0*[illegible])/5.0
510 [illegible](//,[illegible])
520 [illegible](1[illegible],4[illegible],[illegible],12,[illegible])
530 DO 5[illegible]0 I=1,21
540 J=I-1
550 [illegible] = J
560 [illegible]=[illegible]
570 [illegible](I)=[illegible]([illegible](I))[illegible]
580 [illegible] 520, [illegible], [illegible](I), [illegible](I)
590 [illegible](4[illegible],[illegible]10.4)
600 [illegible]=0.0
610 DO 620 I=[illegible],21
620 [illegible]=[illegible]+0.0[illegible](I)
630 CONTINUE
640 [illegible]=[illegible]
641 [illegible] 470
640 [illegible](1 ,[illegible])
650 [illegible]660,
660 [illegible](1[illegible],[illegible] = 0.05 [illegible] 1.0 [illegible] = [illegible],[illegible]12.6)
670 CONTINUE
680 END
[illegible] [illegible]
```

APPENDIX D

In the following model the cutaneous cues are included with the proprioceptive aspects of posture since these two systemes are extremely difficult to separate and both were present when the data from these experiments were collected.

The following model first defines the transfer function for the basic unit of the postural system (the muscle-muscle spindle system and the effects of tactile cues from the feet), then, with added assumptions, develops a model that includes otolithic feedback. Finally, the model is used to predict how Ss without otolithic feedback will sway at the very low frequencies.

A transfer function (G(S)) for the basic unit of postural equilibrium can be written as follows:



where:

$\Theta_d(t)$ = desired angle of body with respect to gravity (i.e. zero) with random disturbances caused by changes in muscle tonus, respiration, etc.

$\Theta(t)$ = obtained or actual angle of sway.

G(s) = The linear transfer function for the combination of processes found in the proprioceptive and tactile aspects of posture. No assumptions of how these two systems intereact are made except that the general form of this linear transfer function is (K_1 / s = + a), i.e., a first order system.

a = constant

K_1 = constant

K_2 = constant

The transfer function of a system is defined as the output of that system divided by the input, and the system can be completely defined by this relationship:

$$a\frac{d\Theta}{dt} + b\Theta = c\Theta_d$$

(by definition this example is a first order system because no second or higher order derivative is present.)

Taking the Laplace transform (Ritow, 1965) of the above differential equation (i.e., $d/dt \longrightarrow s$):

$$as\Theta + b\Theta = c\Theta_d$$

by algebra:

$$\Theta(as + b) = c\Theta_d$$

$$\frac{\Theta}{\Theta_d} = \frac{c}{as + b}$$

dividing numerator and denominator by b:

$$\frac{\Theta}{\Theta_d} = \frac{(c/b)}{(a/b)S + 1}$$

The ratio on the right side of the equation is called the transfer function, G(s), which is of the general form: $K_1/\ S+a$.

If the input into the system, $\Theta_d(t)$, is a sinusoidal signal, then let $s \rightarrow jw$.

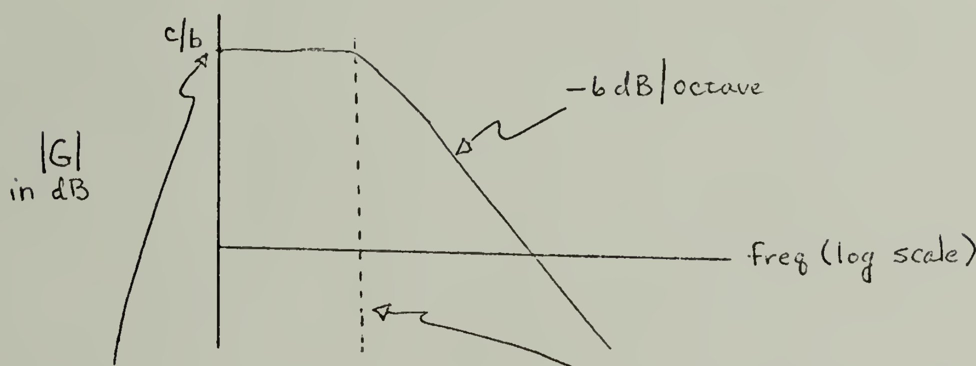
where: $j = \sqrt{-1}$

and, $\omega = 2\pi f$ breakpoint

and the transfer function becomes:

$$G(j\omega) = \frac{c/b}{1 + j\omega(a/b)}$$

From this formula a Bode plot can be constructed of which just the magnitude curve $|G|$ is diagrammed below.



The height of the horizontal portion of the curve is found by asking: What is the value of G(jw) for extremely small values of f (and therefore, for $w \rightarrow 0$)

$$G(j\omega) = \frac{c/b}{1 + 0} = c/b$$

This is called the "break frequency". Note: it is found by setting the co-efficients of the real and imaginary terms in the denominator of G(s) equal to each other. (i.e., 1 and jw (a/

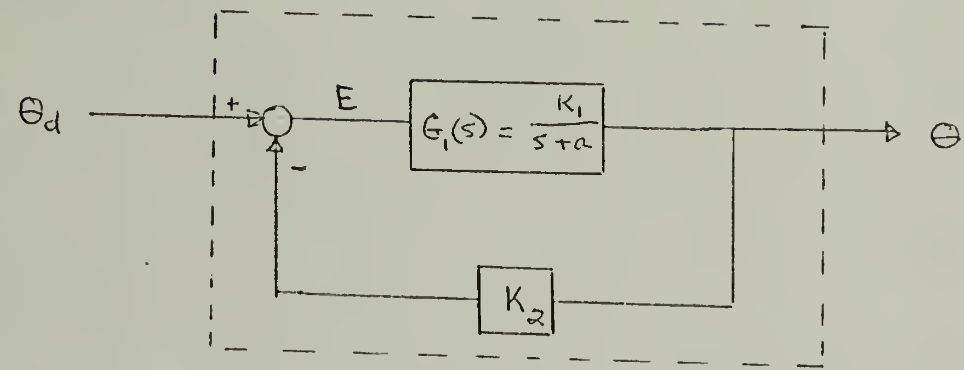
$$1 = \omega(a/b)$$

$$1 = 2\pi f_{br} \, (a/b)$$

$$f_{br} = \frac{b}{2\pi a}$$

The next assumption is that the labyrinthine system detects any θ , and the impulses thus initiated tend to "feedback" on the input of the basic postural mechanism to reduce the effects of random disturbances. This assumption is supported by the present data which obtained a large increase in power when the labyrinthine organs were absent.

One possible otolith-muscle control system can be diagrammed as follows;



where:

K_2 = a proportional constant representing the effect of the otoliths on the input signal to the basic postural unit.

○ = a mixer or differential where the stabilizing effects are removed from θ_d.

E = the resultant input into the basic postural system after the labyrinthine feedback has been removed from θ_d. ($E = \theta_d - K_2\theta$).

The transfer function for the entire otolith-muscle system (dotted lines in Figure 8) can be derived as follows:

$$\frac{\theta}{E} = G_1(S)$$

where:

$$E = \theta_d - \theta K_2$$

and substituting,

$$\frac{\theta}{\theta_d - \theta K_2} = G_1(S)$$

$$\theta(1 = G_1(S)K_2) = G_1(S)\theta_d$$

$$\frac{\theta}{\theta_d} = \frac{G_1(S)}{1 + G_1(S)K_2}$$

and since $G_1(s) = K_1 / s + a$,

$$G_2(S) = \frac{\dfrac{K_1}{s + a}}{1 + \dfrac{K_1K_2}{s + a}}$$

and multiplying the numerator and denominator of the right side by $(s + a)$, the transfer function for the entire system is now given by:

$$G_2(S) = \frac{K_1}{S + a + K_1K_2}$$

and letting $s \longrightarrow j\omega$

$$G_2(j\omega) = \frac{K_1}{j\omega + (a + K_1K_2)}$$

This is the final form of the postulated transfer function that describes the postural system model (i.e., proprioceptive-cutaneous with labyrinthine feedback).

The actual input to the postural system is unknown, since the input results from internal stimuli. The assumption is made that the input is random noise with equal amplitude at all frequencies (i.e., white noise) and, therefore, the mean PSD sway curve is assumed to be a true representation of the magnitude curve ($|G|$) of the Bode plot.

The quantities _a_, K_1, and K_2 can be derived from the transfer function of the proposed model by using values obtained from the mean PSD curves of both the normal and the L-D _S_s. After deriving values for these constants, then the height (in dB) of the proportional component of the mean PSD sway curve for the L-D Ss can be calculated from the equation

$$G_2(j\omega) = \frac{K_1}{j\omega + a + K_1K_2}$$

but the values for a, K_1 and K_2 must first be calculated. Break frequency for normal Ss must be used to calculate a. As before (page 125):

$$\omega = (a + K_1K_2)$$

$$2\pi f_{br} = a + K_1K_2$$

$$f_{br} = \frac{a + K_1K_2}{2\pi}$$

and for the L-D Ss there would be little or no feedback from the labyrinth (i.e. $K_2 = 0$) and there should theoretically be a corresponding reduction in the value of the break frequency. Values obtained from the two curves (See Figure 3D) give:

$$f_{br}(\text{L-D}) = \frac{a}{2\pi} = 0.185 \text{ Hz}$$

and solving for a

$$a = 1.16$$

and to find K_1:

$$f_{br}(\text{normal}) = \frac{a + K_1K_2}{2\pi} = 0.21 \text{ Hz}$$

$$0.21 = \frac{1.16 + K_1K_2}{6.284}$$

$$K_1K_2 = (o.21)(6.284)-1.16$$

$$K_1K_2 = 0.16$$

and substituting a and K_1K_2 into the basic equation for the proposed transfer function for normal Ss,

$$G_2(j\omega) = \frac{K_1}{j\omega + (1.16 + .16)}$$

and, again, at low frequencies $\omega \to 0$ and the equation becomes

$$G_2\ (j\omega) = \frac{K_1}{1.32}$$

The asymptote of the mean sway curve for the normal Ss is 7.0 dB or 7.0 dB = 20 $\log_{10} \left| G_2(j\omega) \right|$ or $\log_{10} \left| G_2(j\omega) \right| = .35$

and $\left| G_2(j\omega) \right| = 2.24$

Solving for K_1:

$$K_1 = (\left|G(j\omega\right|)\ (1.32)$$

$$K_1 = 2.96$$

and finally, knowing the value of a, and K_1, we can determine the height of the asymptote of the L-D S's mean PSD sway curve in dB by substituting:

$$\left| G_2(j\omega) \right| = \frac{K_1}{j\omega + a + K_1K_2}$$

but since $K_2 = 0$,

$$G_2(j\omega) = \frac{K_1}{j\omega + a}$$

and at low frequencies we obtain:

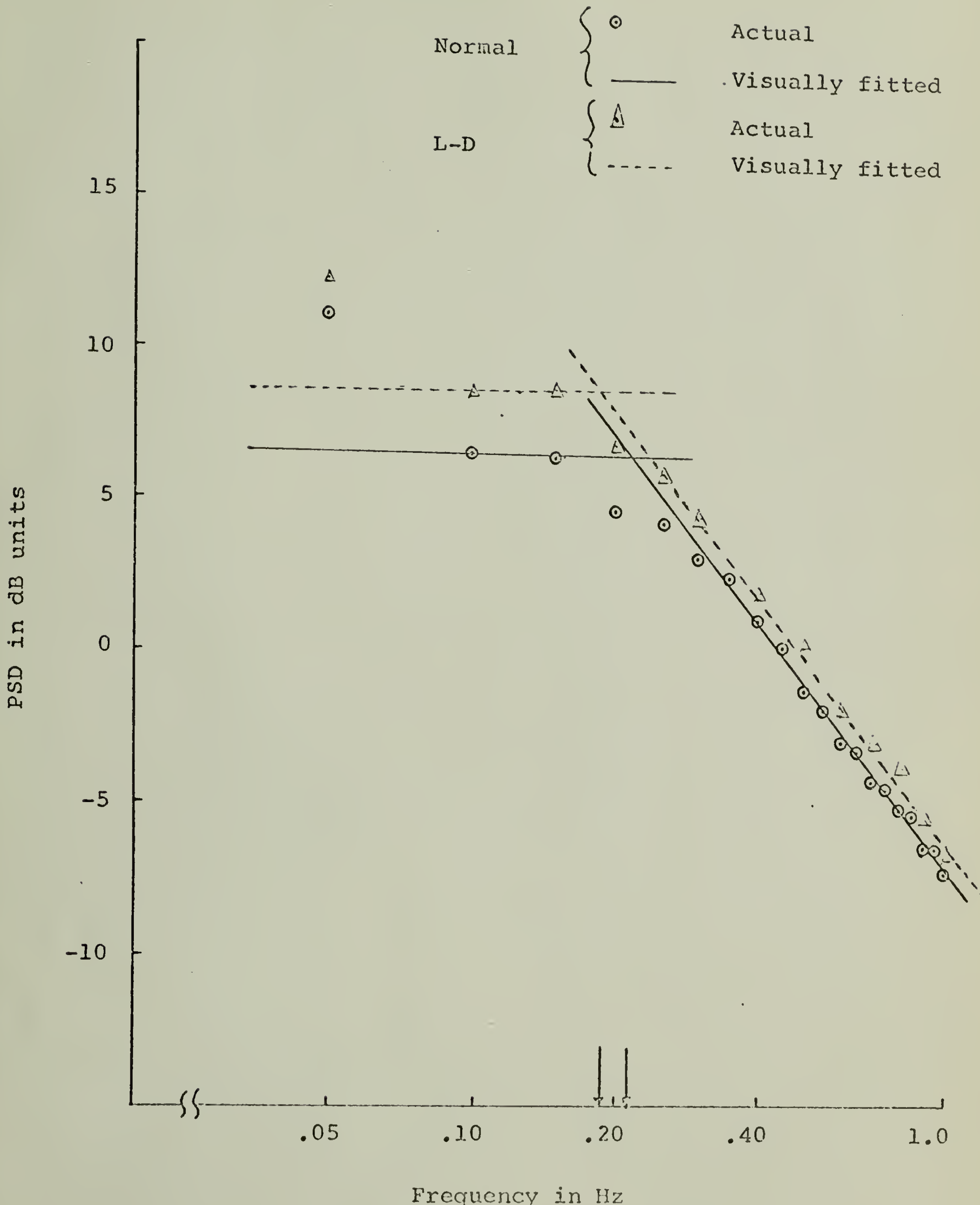

Normal
Actual
Visually fitted
L-D
Actual
Visually fitted
15
10
5
0
-5
-10
PSD in dB units
.05
.10
.20
.40
1.0
Frequency in Hz

Fig. 3D

$$\left|G_2(jw)\right| = \frac{K_1}{a}$$

$$= \frac{2.96}{1.16}$$

$$= 2.55$$

and

$$G_2(jw) = 20 \log_{10} 2.55$$

and finally the power at the asymptotic portion of the sway curve for L-D Ss is:

$$G_2(jw) = 8.13 \text{ dB}$$