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TRANSFER FUNCTION ANALYSIS OF HUMAN POSTURAL CONTROL

by

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A THESIS

SUBMITTED TO THE FACULTY OF GRADUATE STUDIES AND RESEARCH

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R. E. Rink

Date Oct. 12, 1982

ABSTRACT

Transfer function analysis was investigated as a means of studying and evaluating the postural stability of subjects in an erect stance.

A measurement device was constructed to record concurrently the movement of the center of foot pressure and the sway of the body at three points in the sagittal plane. Subsequent computer analysis using a three segment model of the body produced signals which described the independent horizontal movements of the legs, torso, and the head. Spectral analysis of these signals and the movement of the center of foot pressure produced power spectra and coherences. Transfer function analysis utilized this frequency domain data to provide the relationships which indicated how body sway interacted with movement of the center of foot pressure.

As part of a model of the control system governing human postural motion, these transfer functions were used to reinforce speculation about the neural processes involved in maintaining balance. The gain functions linking the two observations of sagittal body movement indicated a number of resonant peaks. The heights and magnitudes of these peaks varied between the body segments. The lower segment demonstrated the largest resonance, occurring at approximately 2 Hz. The middle segment displayed more peaks of lower magnitudes. The upper segment had even more peaks. Variations between the frequencies of these resonances from

subject to subject tended to obscure the peaks.

The resonant peaks generally grew in magnitude and shifted slightly upward in frequency when the subjects' eyes were closed, as opposed to being open. It was postulated that the increased magnitudes of the resonances contributed to a decrease in stability of the postural control system, and hence a decrease in steadiness of the subjects. The impulse responses of the transfer functions demonstrated oscillations of greater magnitude and longer duration for the tests without vision.

Similar observations were made with the alcohol tests. The advantage of sight for balance control was apparently of less importance for the subjects when they were inebriated than when they were sober.

A small number of neurological patients were tested. The resultant data did indeed indicate abnormalities in postural control, but the sources and patterns of particular conditions must await further study.

A brief study of sagittal body movement as opposed to lateral body movement was initiated. The results concurred with those of the majority of previous researchers. Sagittal movement was found to be more suitable for postural measurements. The transfer functions produced were of greater magnitude with less variance.

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The author expresses his gratitude to the members of the Division of Biomedical Engineering for their aid and support. In particular, along with his computing expertise gained through EEG studies, Dr. Z. Koles provided the data acquisition program and a frequency analysis program. His earlier research in posturography developed the force platform used in this study. N. Ouellette and I. Yamamoto were responsible for much of the electronic and mechanical construction of the measuring system.

The time, which in some cases was considerable during the development of the measuring system, volunteered by the subjects of the tests is also sincerely appreciated.

The author also thanks Dr. J. Russell for the provision and instruction in the use of a breathalyser for the alcohol tests.

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1. INTRODUCTION

Posturography, the study of human posture, has come to be an intensely studied field, as evidenced by the number and variety of investigations over recent years. With widespread interest, it has come to be known by other names, like stabilography, statokinesiography, and posturometry. Posture, or the attitude of the body during movement or station, is recognized as requiring the constant attention of various neural mechanisms. The maintenance of the body's equilibrium is generally paramount in the successful completion of tasks, whether motor or mental, and hence must be attainable independently of conscious control. As described by Wells and Luttgens, "posture is a gauge of mechanical efficiency, of kinesthetic sense, of muscle balance and of neuromuscular coordination."

1.1 Applications

1.1.1 Neurology

The human body is a complex mechanical structure. As a dynamical system, it requires a highly sophisticated control mechanism to maintain balance. The fact that the erect body is continually moving has been acknowledged since the mid-nineteenth century. In 1853 Romberg utilized this phenomenon to indicate tabes dorsalis, a degeneration of the dorsal column of the spinal cord and of sensory nerve trunks. With the feet in the Romberg position, heels and toes together, the patient was asked to close his eyes. The test was positive if the patient could not maintain his balance. Although it is now known that other conditions can cause similar disequilibrium, the Romberg test is still a valid diagnostic tool (see publication by Mayo Clinic and Mayo Foundation).

> "The testing for ability to maintain balance has since become one of the routine procedures in neurological examinations." (Fearing, 1925)

Although Skogland found no characteristic pattern of sway for a number of physical conditions, he recognized the testing of standing sway as "a clinical procedure of value in testing for the integrity of the nervous system." Mitchell and Lewis found it useful for detecting posterior sclerosis. Njiokiktjien and de Rijke concluded that vertebrobasilar insufficiency, which is difficult to diagnose by other physical signs, is quite evident through posturography. Tokita et al used spectral analysis of body movements to differentiate between degrees of damage as well as types of various lesions and diseases. Silferskiold observed a characteristic frequency of oscillation in patients with cerebellar syndrome due to chronic alcoholism, as did Mauritz, Dichgans, and Hufschmidt. Similarly, Honjo, Tanaka, and Sekitani could distinguish between labyrinthine lesions and vestibular nerve disturbances. Buchele and Brandt recognized that characteristics of postural sway common to vertigo and other conditions might yield causes and treatments, as did Nasse et al. Sugano, Takeya, and Kodaira investigated the effects of whiplash, psychosomatic disorders, and anxiety, as well as neurological diseases. Other specific conditions have been investigated by Mano (spinocerebellar degeneration), Gregoric and Lavric (Parkinsonism), de Wit (1973), Draganova et al (vestibular areflexia), and Kapteyn and Bles (Meniere's disease). The effects of learning and behavioural disorders were evident in measurements taken by Njiokiktjien et al (1976). After extensive studies with many different conditions Edwards concluded:

> "The accurate measurement of static ataxia has been found to be more indicative of organic and mental condition than most of the psychomotor tests."

1.1.2 Pharmacology

Many researchers have found posturography invaluable in monitoring the effects of drugs in the treatment of neurological disorders, like Parkinsonism (Cernacek, Brezny, and Jagr, and Stribley et al). Pharmacology has many possible applications for such tests. Soulairac, Baron, and Chavannes found excellent correlation between posturographic measurements and psychometric techniques. The effects of certain substances on the body, particularly the equilibrium, have been studied by Andersson et al. (neuroleptic agents), Baron, Bessineton, and Aymard (alcohol), Begbie (alcohol), Folkerts and Njiokiktjien (L-dopa), Hirasawa (pesticides), Soulairac et al (1973, 1977) (alcohol, vasoactive agents, CO_2, and glucose), Sugano and Tominaga (1973 ,1977) (alcohol, diazepam, betahistine, caffeine, and anti-motion sickness drugs), and Terekhov (1974) (alcohol). Sugano and Takeya provided extensive clinical applications and results.

Indeed, it may be possible to investigate the physical causes of aberrations of personality through posturography. Schilder made many definite connections between the integrity of the postural control system and mental disease.

> "Dizziness occurs therefore in almost every neurosis. The neurosis may produce organic changes in the vestibular sphere. Dizziness is a danger signal in the sphere of the ego and occurs when the ego cannot exercise its synthetic function in the senses."

> "Dizziness is as important from the psychoanalytic point of view as anxiety."

King drew similar conclusions:

> "Some individuals have defective proprioceptive feedback mechanisms, the vestibular component in particular being first underreactive, and second, underactive in its role in the sensorimotor integration process. This deficit, whether genetic, developmental, or the result of trauma, constitutes an important etiological or prodromal

factor in process and reactive schizophrenia." According to Ingham, "all investigations claiming to have found a positive relationship between neuroticism and suggestibility have assessed the latter by means of the Body-sway test."

1.1.3 Orthopaedics

Posturography has seen extensive use in orthopaedics. Snijders used a statokinesimeter to investigate the loading of the spine and its shape. Snijder et al assessed the treatment of scoliosis (lateral curvature of the spine) with stabilography. Sahlstrand, Ortengren, and Nachemson investigated the possibility of postural disequilibrium contributing to idiopathic scoliosis.

Although amputees were found to have adjusted quite adequately with balance, Hellebrandt et al (1950) used posturography to optimize the selection, fitting, and training for use of prostheses. The work of Fernie and Holliday involved a large number of amputees. Murray et al (1975) measured body movement to gauge the adequacy of surgical hip replacement. Tsukimura's subjects were crippled children with whom he evaluated treatment and classified problems.

1.1.4 Other Applications

Aside from abnormal conditions, many researchers have suggested the application of posturographic knowledge to our everyday lives. The use of balance studies in industry to improve working conditions was suggested by Greene and Morris. Sugano and Takeya observed the effects of fatigue. The destabilization caused by loading the body with extra weight has been investigated (Carlsoo; Hellebrandt et al ,1943; Hlavacka and Saling). Others considered the fields of sport medicine (Reicke and Yajin; Rottembourg). Hirasawa and Usui compared the standing abilities of twins.

To demonstrate the effects of environment upon man's static equilibrium, many tests were performed under various adverse conditions. Stress has been seen to increase body movement (Carron; Strelets et al; Tanaka et al). Morgan and Watkins found sway to increase when the subject was measured on the edge of a high roof. Constantinescu et al investigated the effects of magnetic pulses on balance. Homick and Reshke looked into the decrease in steadiness evoked by exposure to the zero gravity of space flight. Migraine et al studied the effects of extended subterranean sojourns. Adolfson et al decided that increased ambient air pressure affected the integrative processes of the brain involved in balance control. Bjerver and Persson noted that hypoxia apparently impaired the coordinating mechanisms of stability. Cigarette smoke was observed to significantly increase postural tremor in separate studies by Boudene et al and by Day.

A definite degeneration of equilibrium with advancing age has been shown in the elderly (Boman and Jalavisto; Murray, Seireg, and Sepic; Overstall et al; Sheldon). Hasselkus and Shambes noted that a better understanding of the needs of the elderly can be used to improve their lifestyle. On the other hand, Casaer, O'Brien, and Proechtl studied the postural control of babies as an indication of abnormal development. Dantin-Gallego and Gomez suggested that via a "global study" of the body, posturography is valuable as an indicator of normality. Terekhov (1976) listed further possible applications.

1.2 History

As noted by Yoo et al, the Romberg test provided no quantitative data. A highly skilled observer was required despite the simplicity of the test procedure. Therefore, it was necessary to devise a better measurement scheme to gauge body movement.

In 1886, Mitchell and Lewis observed the displacement of the head with a calibrated bar mounted horizontally at head level. In 1887, Hinsdale developed an ataxiameter to study the effects of defective muscular coordination that was evident in individuals attempting to maintain a fixed body position. Movements of the head forward or backward and to either side were recorded by threads attached to pens writing on sheets of paper. A pen attached to the head recorded movement on a chart suspended horizontally above the subject. Scales beneath the supporting platform indicated the distribution of the supportng forces.

Many variations of this type of system were designed without much standardization (see Eysenck). In 1922, Miles built the first popularly accepted device. It would be used for the next thirty years. The Miles ataxiameter also involved strings attached to the head, but the accumulated head movements were measured. The threads moved ratchet wheels located on the four sides of the subject. The total movement of the head in each direction was recorded during a battery of tests.

Miles' work was reinforced by the studies of Edwards, Eysenck, Fearing, and Hellebrandt, who contributed the bulk of posturographic knowledge from 1920 to 1950. Although contemporary apparatus has become largely electronic in nature, and much more sensitive, the basic observations have remained unchanged.

To conclude this brief review of pioneering work, the author can only agree with Begbie: "These preliminary experiments have thrown up more questions than answers."

1.3 Physiology

Although no attempt can be made in this research to explain the physiological basis for observations, some insight may be gained by examining previous work which concentrated on such aspects.

1.3.1 Reasons for Motion

The movement of the quietly standing body may be attributed to many factors. Various physiological activities like respiration and cardiac function, inherent vibrations of the mechanical construction, and active control by the central nervous system all contribute to body motion. However, the frequencies observed are often higher than those of the physiological activities, and the high degree of variability resigns passive, uncontrollable elasto-mechanical properties to a secondary role (Aggashyan, Gurfinkel, and Fomin).

The human body is naturally inclined forward about 2 to 3 degrees while standing. Although this puts the calf muscles under continuous tension, Morton recognized the positive aspects of the arrangement. To take advantage of the length of the foot anterior to the ankle for leverage, the center of mass of the body must be anterior to the ankle. If the center of mass was located directly above the ankle, extensive counterbalancing torques in both directions about the joint would be required. A forward lean also facilitates forward motion.

During electromyographic tests of the leg Basmajian noted relatively slight muscle activity during relaxed standing. He remarked on the apparent efficiency of human balance control. In studies of the movement of the body during simple activities, Murray, Seireg, and Scholz were impressed by the minimal exertion associated with the high coordination required to execute tasks. Hellebrandt (1938) attributed the stamina of postural muscles to the phenomenon of sway. Although the extensors of the ankle joint are under constant tension, the movement of the body around that joint is accompanied by varying degrees of muscle contraction. This rotation of motor units allows some parts of the muscle to rest while others work.

Smith came to the same conclusion. He observed periodic variations in intensity of the discharge of action potentials. This 10 Hz variation was considered to be part of an active torque at the ankle which limits the range of motion around the joint while allowing sets of muscle units to relax.

Bonnet et al identified "rhythmic bursts" of 10 Hz activity in the postural muscles of the legs. This was thought to constitute a "dynamic interrogation of the proprioceptors" in order to maintain a communication link between the analytical center of the control system and the peripherals.

"... when an extremely precise regulation of position is necessary and one cannot passively await a detectable position change to trigger regulation commands ... small and periodic disturbances are made for which system response is well-defined and which do not endanger system equilibrium."

This activity is most probably the postural tremor that other researchers have observed (Allum, Dietz, and Freund; Litvintsev, 1972; Rietz and Stiles; Sugano and Takeya).

A byproduct of such muscle activity is the "muscle pump" which aids blood circulation in the leg. Hellebrandt et al (1940) speculated that the fatigue resulting from standing for periods of time results more from cerebral anoxemia than from lack of blood flow or muscle strength in the lower extremities.

1.3.2 Perception of Balance

When likened to a control system, the physiological maintenance of balance may be considered to require three parts: input sensors, a central processor, and motor output. Many interconnections between these parts are possible.

Sensory input is derived from three sources (Dickinson, 1974). Proprioceptors provide awareness of movement through organs in the muscles, tendons, and vestibular apparatus. Some studies have considered muscular and tendonous afferentation as being proprioception, with vestibular sensation being a separate source (de Wit, 1973). This paper will also treat the two sources as being separate for clarity. The traditional five senses belong to the exteroceptors: the eyes, ears, nose, mouth, and skin are sensitive to stimuli originating outside the body. Interoceptors are found in the viscera and sense changes in shape and orientation.

<u>1.3.2.1 Muscular and Tendonous Proprioception</u>

Muscular and tendonous proprioceptors respond to changes in length and tension in the soft tissue at the joints. Deecke et al investigated the modification of vestibular signals by the proprioceptive afferentation of the neck as a means of avoiding postural reflexes or "somatogyral sensation" during head movements. Lewit emphasized the role of proprioceptive information originating from the spinal column.

Proprioception is stongly reinforced by pressure sensation in the soles of the feet. Watanabe et al (1979) studied the effect of the texture of the supporting surface via "plantar mechanoreceptors", obtaining responses more complex than spinal reflexes. The vibrating platform of Shipley and Harley impaired steadiness by interfering with exteroception. Kapteyn (1972a) obtained similar results with foam pads beneath the subject's feet. Under normal circumstances, afferentation from the joints, muscles and feet can control balance (Kapteyn and de Wit).

After enforcing ischemia upon the muscles of the leg, Mamasakhlisov, Elner, and Gurfinkel discovered that the vertical posture was not markedly disturbed. Similarly, Aggashyan et al (1973) observed only an increase in 1 Hz oscillation with the decrease in postural regulation resulting from ischemia. Sensory input from other sources must then act to replace the muscular afferentation. Some researchers have ranked vestibular sensation above muscular and tendonous proprioception (Sugano and Tominaga). Martin (1965) places vestibular function in the predominant role in unstable conditions. (On the other hand, Birren found no relation between vestibular function and body sway.)

1.3.2.2 Vestibular Proprioception

Walsh suggested that the large, low frequency postural movements are controlled by the stretch reflex, while vestibular reactions are limited to smaller, higher frequency movements. Oku et al concurred, where their tests revealed that high amplitude sway was not adequate for testing the vestibular righting reflex since muscular control systems interfered. In studies with postural activity and sleep, Sugawara et al concluded that the vestibular controlling system takes longer to stabilize the tonic postural position than the proprioceptive system.

Fukuda observed that postural muscle responses varied with the position of the head. With similar results, Mano et al described a function of the vestibulo-spinal reflexes which facilitates the stretch reflex of leg muscles.

Nashner has done much work to determine the role of the vestibular apparatus in the control of balance. In 1973 and 1974 he developed a control model involving the two vestibular organs. Angular acceleration above a certain threshold is detected by the semicircular canals, and thus higher frequency movements are monitored. The utricular otoliths handle the detection of linear acceleration, and provide a low frequency signal concerning the orientation of the head.

Galvanic stimulation (Baron et al, 1973; Cernacek and Jagr; Coats; Dzendolet; Fernald and Moore; Kubiczkowa and Skibniewski) and caloric stimulation (Jonsson and Synnerstad) of the vestibular apparatus have been used to further elucidate the role of this mechanism in equilibrium control. Barigant et al used a variety of such tests, including luminous and acoustic stimulation.

1.3.2.3 Exteroception

Some studies have concluded that vision has little effect on balance (Vranken and Willems, Weissman and Dzendolet). After all, a person can stand with his eyes closed. Any increase in body sway may just be due to a reflexive tensing of the muscles. With the eyes closed, the body tends to lean forward further. This is possibly a defence mechanism to ensure that if a fall occurs, it will be forward, and allow some protection by the arms (Gantchev, Draganova, and Dunev). This posture will make the body even more mechanically unstable.

However, the overwhelming majority of work points to some definite dependence of balance on the quality and quantity of visual information. The effect of vision on balance is perhaps the most easily tested of the sensory inputs. An eyes closed test of standing steadiness generally produces about 50% more body movement than the eyes open counterpart. This decrease in equilibrium is also observed gradually with a gradual decrease in light (Kapteyn et al, 1979). Peripheral vision was found to be important as well (Begbie, 1967; Dickinson, 1969; Dickinson and Leonard). De Wit judged peripheral vision to be more effective than central field vision in improving balance.

It is conceded that vision is of less importance to static balance than proprioception. During dynamic body movement though, vision is necessary for balance unless training is undertaken (Dickinson, 1974; van Parys and Njiokiktjien). Bles and de Wit assumed that vision supercedes vestibular sensation under such conditions. Perhaps it is only the many years of practice at standing that enables people to accomplish it without vision. Lee and Aronson observed that sight plays a dominant role in the standing of young children, where a constantly changing body interferes with the 'calibration' of the other sensory organs.

Rather than being a separate source of balance control, vision is regarded as enhancing the performance of other control mechanisms (Dickinson, 1974). The effect of vision appears to be greatest at low frequencies (less than 1 Hz), which indicates involvement with the otoliths (Dichgans et al, Seidel and Brauer). Using optokinetic stimulation, Brandt and Buchele demonstrated some effects on body movement at higher frequencies. They proposed that visual exteroception could alter the latencies of responses to proprioceptive inputs. In this context, Nashner and Berthoz considered vision to act as a moderator of postural action, until confusing sensory input could be resolved.

Vision may serve to gauge the degree of imbalance or the effectiveness of corrective measures undertaken by the body. It inhibits lateral postural movements elicited by galvanic stimulation of the labyrinth (Njiokiktjien and Folkerts). In elderly patients with a bilateral loss of labyrinthic function, vision is required to maintain balance (de Haan). Vision compensates for a lack of exteroceptive input, as induced by foam pads beneath the subject's feet (Amblard and Cremieux, Bles and de Wit) or by a cold foot bath (Orma). Fernie and Halliday noticed a similar increase in dependence upon sight by amputees. Adolfson, Goldberg, and Berghage noted that an increase in air pressure decreased stability, possibly by acting upon the central nervous system. This effect was more obvious with the subject's eyes closed, when vision could not compensate.

Hamann et al decided that since visual stimulation could disrupt balance to some extent, the other sensory systems could not compensate for an absence of stable visual information. Witkin and Wapner noted the difference in effect of an unstable visual field as opposed to a weakened visual field. Many tests have shown the tendency of the body to deviate from true vertical, and hence to greater imbalance, when a visually perceived motion or false vertical reference is introduced (Edwards, 1946; Lestienne et al; Taguchi; Zikmund and Ball). Such visual stimulation involved complex mechanisms of postural reaction, as indicated by long delay times (Oblak, Mihelin, and Gregoric).

It is noted that the contribution to balance control by vision could be largely muscular proprioception from eye movement (Kapteyn et al, 1979). Studies by Tokumasu, Tashiro, and Yoneda indicate that the visual information itself provides the necessary sensory input. Lee and Lishman (1975) maintain that sight is more sensitive to acceleration and position than proprioception or vestibular sensation, and thus acts to finely tune the reaction to such inputs. The instability experienced at great heights may then be explained by the field of view being so distant.

Visual feedback via an oscilloscope generally decreases low frequency sway, with a corresponding increase in high frequency movements (Gantchev, Draganova, and Dunev, 1979; Hlavacka and Litvinenkova). Rather than actually providing visual data for balance control, such feedback may primarily increase the level of alertness or sensitivity of the central balance control system. Increased attention as imposed by visual and mental tasks also decreases body movement (Cantrell; Litvinenkova, Lipkova, and Liska; Seidel and Brauer).

Sound could have a similar effect on balance control. Noise or tones have been found to increase body sway (Bensel, Dzendolet, and Meiselman; Litvinenkova, Lipkova, and Liska). Intermittent auditory distractions were observed to decrease sway (Fearing, 1925) and auditory feedback produced a greater improvement in steadiness (Ratov and Merkulov; Takeya, Sugano, and Ohno). Njiokiktjien (1973) noted that a decrease in low frequency sway during auditory tasks was again accompanied by an increase in high frequency movement.

It is most probably the interplay between all afferentation that provides the body with a sense of position and movement, and hence the ability to maintain equilibrium. Once a pattern of performance has been established, any deviation from this stereotyped postural image produces controlling signals ([illegible]el, Lipshits, and Popov).

1.3.3 Central Control of Motion

Information from the input sensors must be interpreted in some manner so as to initiate an appropriate response. This is accomplished by the central nervous system. The balance control system is heirarchical, but the division of tasks between the levels is not entirely clear.

The cerebral cortex occupies the highest level, being responsible for voluntary movement. Roberts (1978) also refers to the cortex as a "recognition machine", where sensory input is analyzed and suitable patterns of motor response are generated. The cerebellum and brain stem assume supportive roles. Dickinson (1974) attributes the integration of sensory input with cortical movement impulses to the cerebellum. Both authors agree that the dense interconnection between the cortex and the cerebellum results in high coordination between their activities. Thus, a feedback control is achieved to prevent instability in a control system as complex as that required for body balance.

The brain stem serves to modify the output of higher brain centers. The patterns of motor responses, or "motor programs" according to Brooks, are further tailored to adjust for special conditions and to produce coordinated movement. The basal ganglia supply the supportive functions for voluntary movements, such as postural fixation and weight transfer during locomotion. The reticular formation is particularly important in modifying responses. By means of a tonic inhibition system, this formation can decrease the sensitivity of sensory organs. A similar effect is exerted over the spinal reflexes, some of which are involved in postural regulation. Via cortical control, the reticular formation affects the level of arousal of the cortex, thus facilitating the response to particular sensory inputs. De Wit (1972) reported a similar function in the dorsal vestibular nucleus (of Deiters). A somatopographic arrangement of the nucleus divides the human body into two parts. Perception of movement of the top part is most dependent on the vestibular system, whereas the lower part is most dependent on the muscular proprioceptive system of the legs.

The spinal cord relays the neural output to the various efferent nerves. For complex movements the amount of information to be transmitted must be immense. Nashner and Woollacott propose the existence of spinal "function generators" which elicit patterns of muscular action when triggered by commands from the brain. Several such commands would produce locomotor movements. Static balancing would require more primitive responses.

A synergistic effect is observed to negate the effects of internal disturbances. No correlation is observed between body movement and heartbeat (Nilsson). Breathing is not evident in the movement of the center of foot pressure (Sugano and Takeya). Local lesions of the brain result in apraxia; the uncoordinated movement of the body segments.

Breathing then becomes apparent. Gurfinkel and Elner attributed this to some interference with the feed forward control within the cerebrum. Nashner and Woollacott described a coordination between ankle and hip extensor muscles as a stabilizing function called "sway synergy". Marksmen could finely tune this ability. Mauritz, Dichgans, and Hufschmidt defined "fixed intersegmental reaction patterns" as those body movements which balance the movements of other parts of the body.

1.3.4 Initiation of Motion

Once the appropriate excitatory impulses have been transmitted to the specific muscles, the muscles relax or contract to achieve the desired body position. Brooks defines posture as a system of "fixations" or tension movements, in which groups of muscles contract in opposition to each other in order to hold a fixed position.

Nashner (1976) has ascribed four possible functions to postural muscles in maintaining balance. Using a moving platform he measured the various response times of muscle reactions. Initially, the stiffness of the muscle resists immediate movement. Passive resistance to movement is also exerted by the elastic tension of joint ligaments, deep fasciae, and skin (Smith). The myotatic stretch reflex could then follow, 45-50 msec later, fairly quickly since it is a spinal reflex. Then, a "functional stretch reflex" or "triggered response" (Nashner, 1979) was observed after 120 msec. It was postulated that the cerebellum monitors the effects of this reflex action, and adjusts the intensity of subsequent efferent excitation accordingly. The final muscular response depended on vestibular activation, where the longer and more complex neural pathways cause latencies of 180 msec.

The importance of spinal reflexes in postural regulation has been an issue of some contention. Initial investigations such as those by Kelton and Wright discounted the role of the stretch reflex in static balance. Little, if any, muscle activity was recorded. It was thought that the deformations of the muscles during quiet standing were not great enough to elicit the stretch reflex. Litvintsev (1973) concluded that the stretch reflex played no part in postural stability at the ankles or knees, but controlled the body pose, or the articulation angles at the hips and the waist.

The rocking platforms of Gurfinkel, Lipshits, and Popov and Gurfinkel et al (1976) caused ankle rotation that exceeded the threshold of the stretch reflex, but muscle activity was not evident. It was concluded that supraspinal controls raised the threshold. Similar controls maintained body balance during static standing through variation in muscle rigidity. This mechanism was presumed to be sufficient to compensate for slight deviations in posture (Litvintsev, 1972).

However, Thomas and Whitney had calculated that the body was not rigid enough to produce the 10 Hz motion that they observed. Continuous muscle action had to be involved. Subsequent research with more sensitive EMG equipment revealed some muscle activity. In 1973 Gurfinkel concluded that the stretch reflex was the principle mechanism of postural control. Afferentation from sources outside the postural muscles served to alter the "reflex loop gain" through central structures of the nervous system. Interference with muscle afferentation (through fusimotor blockade, ischemia, or vibration) caused varying degrees of imbalance. This hypothesis was supported by other studies (Elner, 1979; Elner et al, 1976; Elner, Popov, and Gurfinkel; Mano et al; Mamasakhlisov, 1979; Shambes).

Burke and Eklund differentiated between sway-induced contractions and the basic stretch reflex because of the supraspinal involvement. Rack assigned the stretch reflex to merely limiting the displacement caused by an unexpected disturbance, after which more adequate measures would be taken through central nervous control. Such control could also be exerted in preparation for an anticipated disturbance (El'ner).

2. SYSTEM DEVELOPMENT

Once the implications of body movement to the integrity of the central nervous system were recognized, researchers attempted to quantify such movement. The maximum amplitude of the sway of the body was easily observable, and served as the basis for the earliest tests. Mechanical apparatus was designed to sum the movements of the body over predetermined periods of time. With tracings of the meanderings of the head, it was possible to calculate areas and mean positions. The advent of the force platform provided similar information regarding the movement of the body as a whole.

2.1 Characteristics of Motion

The movement of the body, whether represented by sway or as displacement of the center of foot pressure, may be considered to occur in two orthogonal planes of the body: the sagittal or anterior-posterior plane and the lateral, frontal, or left-right plane. (Movement in the vertical direction is generally considered as being negligible.) The stabilogram records the one-dimensional movement of the center of foot pressure in either the sagittal or lateral plane. Each plane appears to display different characteristics of balance control. The statokinesigram is the two-dimensional representation of the movement of the center of foot pressure in a horizontal plane.

Using an oscilloscope to provide variations of visual feedback (ie. delayed, reversed, amplified, etc.) Smith and Ramana exaggerated the differences between the postural control systems associated with sagittal and lateral movement. They found that these systems are concerned with the definition of the relation of the eyes, head, and body to the movement of environmental stimuli, as well as maintaining static equilibrium. Other studies have revealed little relation between lateral and sagittal movement (Bensel and Dzendolet; Dickinson, 1974; Kapteyn, 1973; Leifer and Meyer; Litvintsev, 1976; Njiokiktjien).

Sagittal movement appears to be more useful than lateral movement in posturographic studies. The biomechanics of the structure of the body in the sagittal plane demonstrates less stability (Kapteyn, 1973) and therefore demonstrates to a greater degree the resources of the body to compensate for increased demands upon the balance control mechanisms (Nashner, 1971; Nilsson). The simpler biomechanical structure also provides greater reliability for measurements (Seidel et al). Lateral body movement is more dependent on foot position, which is hard to regulate unless a particular stance is enforced (Wright). As a result, sagittal sway power spectra have proven to be more effective in differentiating patients from normal subjects in clinical studies (Mauritz, Dichgans, and Hufschmidt).

Most studies have concentrated on body movement in the sagittal plane. Unfortunately, there has been little standardization in the treatment of posturographic data. Besides amplitudes, areas, and mean positions of body movement, other characteristics that have been investigated include directions, velocities, accelerations, and frequencies.

2.1.1 Variability

The search for new parameters of body movement has been spurred by the problem of variability. Inter-individual and intra-individual variance has been too high to enable foolproof distinction between data obtained from subjects under varying test conditions. Sugano and Tominaga noted changes in postural measurements from season to season, day to day, and morning to afternoon, with up to 20% variance. The range of normal postural movements assumed a Poisson distribution. When data has indicated disturbances in balance, these symptoms were usually visually obvious. As a result, Henriksson et al concluded that the mean position of weight distribution at the feet varied too much among normals to be useful as a test criterion.

Attempts have been made to reduce the high variance through normalization. There is no established relationship between equilibrium and height, weight, or sex. The horizontal height of the center of gravity of the body is very uniform among normal subjects with respect to the total body height (Hellebrandt et al, 1938; Palmer). But, it cannot be stated that taller or heavier subjects will display greater body movement. Some researchers observed less body sway with females, attributing this to their bodies being generally shorter and lighter than those of males (eg. Travis, 1944). However, other researchers found males to be steadier, while some studies showed no difference.

Orma concluded that men and women use different postural mechanisms, which are dependent on muscular strength. Women were thought to use visual cues more effectively as a compensation for less muscular strength (see also Wapner and Witkin; Weissman and Dzendolet).

High variance cannot be accounted for by differences in age. The sense of balance improves from birth into the teen years and starts declining in the forties (Boman and Jalavisto; De Oreo and Wade; Dickinson, 1974; Sheldon; Sugano and Takeya). Between these ages no marked changes in steadiness have been observed in normal populations.

Cureton and Wickens discovered a definite relation between balance and posture, strength, physical fitness, and athletic ability. Joseph and McColl recorded considerable variations in muscle activity in quietly standing subjects. Perhaps, the level of variance of posturographic studies could be reduced by considering muscular ability. The evaluation of this parameter would be a further complication to stability measurements.

2.1.2 Reduction of Variability

For the most part, researchers have tried to reduce variance with modifications of the test procedure and data analysis. An unstable or movable support platform provides a dynamic balancing test. Conscious movements are required to maintain balance, whereas so-called static balance depends solely on nonvolitional responses. By increasing the difficulty of the balancing task, it is hoped that postural disturbances. which may be less evident during the static 'quiet standing' test, will be amplified by the greater demand upon the postural control mechanisms (Azoy; Begbie, 1967; Draganova et al; Gantchev and Popov; Homick and Reshke). Similarly, disturbances to the upper body have been employed to disrupt equilibrium (Kobayashi; Litvintsev). However, it has not been established that a relationship exists between static and dynamic balance (Estep; Travis, 1945). Terekhov's criteria (1978) for the ideal measurement system include comfort and a solid, steady supporting surface.

Some experimenters have divided normals into different groups as a result of high variance (Aggashyan; Dunstone and Dzendolet; Nashner, 1976; Scott and Dzendolet). This would appear to further complicate the analysis procedure. Would unusual results constitute abnormality or a new class of results?

The most popular approach has been to compare data from tests with the subject's eyes closed to 'eyes open' data.

While investigating the compensatory effects of vision in balance control, the test to test variance should also be reduced (Njiokiktjien; Watanabe and Egami). The Romberg's coefficient of Terekhov (1974, 1979) and the Romberg quotient of Hlavacka and Saling involve ratios of posturographic parameters from eyes open (EO) and eyes closed (EC) tests.

2.1.2.1 Frequency Characteristics of Motion

A ratio of two numbers can hardly be expected to differentiate between the very wide variety of observable conditions with the large variances seemingly characteristic of posturographic measurements. Spectral analysis yields values which vary over the frequency range of interest. These values usually indicate the amount of activity or power associated with the process being measured at specified frequencies. It is hoped that a 'spectral signature' can be established as a distinguishing trait of the various conditions of balance disturbance. Aggashyan, Gurfinkel, and Fomin noted that "the vibrations of the body described in the form of stabilograms may be regarded as a steady, ergodic random process. The spectral and correlation methods are a natural apparatus for analysing such processes."

Postural body movements are generally classified as two types with regards to their power spectra. Low frequency, large amplitude movements occur in a band from DC to 2 Hz.

High frequency, small amplitude oscillations range from 5 to 10 Hz.

The low frequency movements are generally considered to represent the actions of the various balance control systems. These movements are irregular in frequency and amplitude but constitute the greater amount of power. Researchers have established frequency bands ruled by the various afferent sources.

The vestibular apparatus and vision initiate movements up to 0.3 Hz. Proprioceptive reactions occur between 0.5 and 1.3 Hz (Aggashyan, Gurfinkel, Mamasakhlisov, and Elner; de Wit, 1972a; Mauritz et al, 1975). Although visual responses exist primarily in the lower range, an increase in power in both bands is observed in the EC test (Leifer and Meyer; Litvintsev, 1972). Movements at higher frequencies (1.5 to 2.5 Hz) are seen to be mechanical oscillations resulting from joint instability (Begbie, 1967).

The oscillatory movements observed in the range from 5 to 10 Hz are generally conceded to be muscle tremors (Halliday and Redfearn; Litvintsev, 1972; Stiles and Rietz). Muscle tremor has been recorded as high as 20 Hz (Rietz and Stiles). Some researchers have identified two different force-generating mechanisms of muscle associated with different frequencies; above and below 6 Hz (Allum, Dietz, and Freund; Sugano and Takeya).

Studies relying on the very low frequencies encountered high variability (Seidel et al; Soames, Atha, and Harding).

Eklund and Lofstedt suggested that fequencies above 1 Hz be recorded to investigate the involvement of neural structures in balance control most effectively. To include the consistent high frequency components of body movement, an upper limit around 16 Hz should be considered (Nilsson; Spaepen, Fortuin, and Willems).

2.1.2.2 Control Systems Modelling

Another approach to reduce variance deals with control systems modelling. An input is related to an output via various processes and feedback loops. Most techniques involve the derivation of differential equations which approximate the performance of the system (see "Biological Systems Analysis and Control Theory" in Clynes and Milsum). In the analysis of biological systems, however, many parameters are difficult or impossible to obtain. The resulting equations are often unwieldy. An alternative is to deduce the underlying control system by observing the output resulting from an observed or forced input.

Gantchev and Popov used a horizontally moving platform to initiate body sway. They then deduced the transfer functions relating perception of the body movement, via the afferents, to muscular response. Bizzo and Baron (also Bizzo et al) used galvanic stimulation of the labyrinths to develop transfer functions relating the movement of the body to the input stimuli.

Nashner (1972, 1973) combined the movement of the center of foot pressure, body sway at the waist, and electromyographs of a leg muscle with a servo-controlled force platform to model the postural control exerted by the vestibular system. Working with Woollacott, Nashner produced a conceptual model, including vision and proprioception as well, with movement of the body in three segments.

Modelling analyses have indeed indicated the complexity of human postural control. De Wit (1973) concluded that no linear correlation between input and output was possible on a postural control model. The central nervous system could selectively concentrate on various sensory inputs as the situation commanded. Houk concurred. Gottlieb and Agarwal noted that such variations in sensitivity to input or magnitude of response were necessary to adapt to changes in the dynamic characteristics of the body (eg. fatigue) or changes in the definition of adequate performance (eg. as for marksmen), as well as to external disturbances.

The last refinement leading up to the proposed method of data processing combined spectral and control systems analyses. Previous work along this line attempted to establish a linear relationship between sagittal and lateral body movements (Brauer and Seidel; Le Roux et al, 1973). In their research using this method, Leifer and Meyer reached the conclusion stated previously. Their results demonstrated the independence of the control systems involved in sagittal and lateral balance.

The present work involves many of the aforementioned 'tricks' to reduce variability by establishing a relationship describing postural control. EO and EC tests are recorded, where the data consists of movement of the center of foot pressure and body sway. Spectral analysis is applied to the data. Transfer functions are produced to formulate the relationship between movement of the body and the subsequent reaction at the feet.

2.2 Development of Model

If one considers the human body to be a simple inverted pendulum, a one segment model, some insight of the control mechanisms involved can be gained. The center of mass of the body is represented by a weight fixed atop a weightless rod which pivots on a fixed hinge, the ankle joint. As gravity acts upon the weight, it tends to tilt the rod as if in the sagittal plane of the body. An appropriate torque acting at the hinge will prevent the pendulum from falling.

In a static situation, the angular displacement of the pendulum from vertical is a sufficient criterion from which to determine the force required to generate this torque. However, in dynamic situations, quantities such as angular velocity and angular acceleration can enter the fray. The monitoring of these quantities and feedback to the force generator can also be used to keep the pendulum erect.

Regularly induced vibration can also reinforce feedforward and/or feedback control (Meerkov). Compensation for external disturbances such as wind and movement of the supporting surface further complicates the control system.

2.2.1 Approximation of the Control System

The simple inverted pendulum representation of the standing human body becomes the basis for the control system model. The movement of the single body segment, as described by position, velocity, and acceleration in terms of angular or linear movement, can be related to the movement of the center of foot pressure as a function of its position and derivatives. The control system may be considered as having an input, the position of the body segment, and an output, the position of the center of foot pressure, at any given instant in time. The data then consists of many such positions observed over a period of time. The Fourier transform (described further in Appendix A) of the output data divided by the transform of the input data yields the transfer function which describes the action of the control system.

In a similar system with multiple independent inputs, each input may pass through a different transfer function to a common summation point to produce a single output. The individual transfer functions may be found by the preceding method if all other inputs not incident to the transfer function in question can be eliminated. In many practical situations, such is not the case. Inputs to the system cannot be 'turned off' without disturbing the control system.

An example of this problem is a three segment model of the human body. Using free body analysis Koozekenani et al derived a number of equations to express the relationships between the forces and torques at the joints and the masses and movements of the body segments. Figure 1 illustrates the parameters of a similar model in the sagittal plane. The equations are numerous and complex. The proposed method of analysis in this study simplifies the situation to some extent.

The distance of the center of foot pressure from a point on the supporting surface immediately below the ankle joint is found by manipulating the equation which sums the forces acting about the ankle:

$$d_c = \frac{T_1 - F_x h + m_f g d_f}{F_y} \tag{1}$$

where F_x = sum of forces resulting from horizontal acceleration of the body segments,

F_y = sum of forces resulting from gravity and vertical acceleration of the body segments,

m_f = the mass of the foot,

g = the acceleration of gravity, and

d_f = horizontal distance from center of mass of the

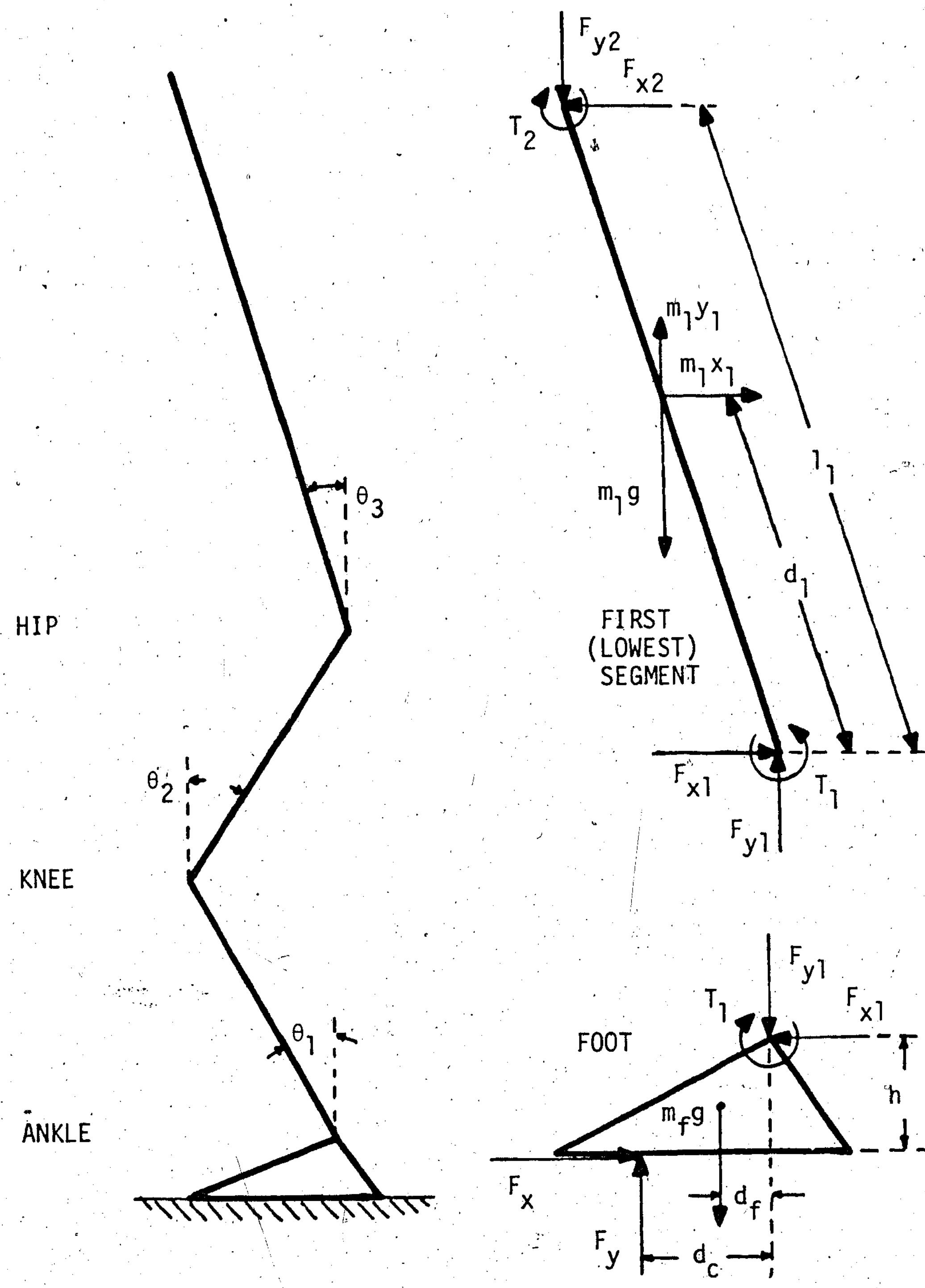

FIGURE 1. Free body analysis of three segment model.

foot to the ankle joint.

The torque, T , at the ankle is balanced by a sum of factors including the moment of inertia of the shank, the vertical and horizontal forces exerted on the shank, the vertical and horizontal forces exerted on the thigh, and the torque at the knee. The next equation is derived from this sum of forces acting upon a single segment rotating about the ankle and defines the relationship:

$$T_1 = -J_1\ddot{\theta}_1 + m_1\ddot{x}_1 d_1\cos\theta_1 + m_1(g - \ddot{y}_1)d_1\sin\theta_1 + T_2 - F_{x2}l_1\sin\theta_1 + F_{y2}l_1\cos\theta_1 \tag{2}$$

where J_1 = the moment of inertia of the first segment,

θ_1 = the angle subtended by the first segment from horizontal,

$\ddot{\theta}_1$ = the angular acceleration of the first segment,

T_2 = the torque at the second joint,

m_1 = the mass of the first body segment,

$\ddot{x}_1$ = the horizontal acceleration of the first segment in the sagittal plane,

$\ddot{y}_1$ = the vertical acceleration of the first segment in the sagittal plane.

F_{x2} = the horizontal force exerted at the second joint,

F_{y2} = the vertical force exerted at the second joint,

d_1 = the distance from the center of mass of the first segment to the first joint, and

l_1 = the length of the first segment.

The torques at higher joints may also be expressed as sums of similar terms. Our preliminary analysis of these equations, however, will deal with a one segment model. Forces and torques with the subscript of 2 in Equation 2 are therefore eliminated. If the body is nearly vertical, as during normal standing, the angles are very small. Hence, $\sin\theta \simeq \theta$, $\cos\theta \simeq 1$, and $x \simeq d\theta$ are valid approximations. With the further assumption that $g \gg \ddot{y}_1$, the torque equation can be approximated by:

$$T_1 \simeq m_1(x_1 g + d_1\ddot{x}_1) - J_1\ddot{x}_1/d_1 \tag{3}$$

The position of the centre of foot pressure can then be determined:

$$d_c \simeq \frac{m_1\ddot{x}_1(d_1 - h) + g(m_f d_f + m_1 x_1) - J_1\ddot{x}_1/d_1}{m_1 g} \tag{4}$$

Further rearranging, where $m_1 x_1 \gg m_f d_f$, yields:

$$d_c \simeq x_1 + (m_1 d_1 - m_1 h - J_1/d_1)\ddot{x}_1/m_1 g \tag{5}$$

For a single segment rotating about one end, the moment of inertia is likely to be expressed:

$$J_1 = K_1 m_1 d_1^2 \tag{6}$$

where $0.25 < K_1 < 0.33$. Equation 5 may then be written:

$$d_c \simeq x_1 + K_1' d_1 \ddot{x}_1 / g \tag{7}$$

where $K_1' = 1 - K_1 - h/d_1$ and is positive but less than unity.

Taking Fourier transforms, the frequency domain expression is shown to be:

$$\begin{aligned} D_c(s) &\simeq (1 + as^2) X_1(s) \\ &= F_1(s) X_1(s) \end{aligned} \tag{8}$$

where $a = K_1' d_1 / g$ and $F_1(s)$ is a function of the frequency dependent variable, $s = j\omega$.

The form of Equation 8 lends itself to transfer function analysis. Figure 2 illustrates the form that the transfer function, $F_1(s)$, would take. At zero frequency the center of foot pressure and the center of mass would coincide in horizontal distance from the ankle. As frequency increases to a frequency, $f_o = 1/2\pi\sqrt{a}$ Hz, the horizontal distance from the ankle to the center of foot pressure is a decreasing fraction of that to the center of mass, but the movements are in phase. At the cut-off frequency, f_o, the magnitude relationship increases monotonically but at 180° out-of-phase.

With the inclusion of multiple body segments in the model, any new terms have proved to be additive. The

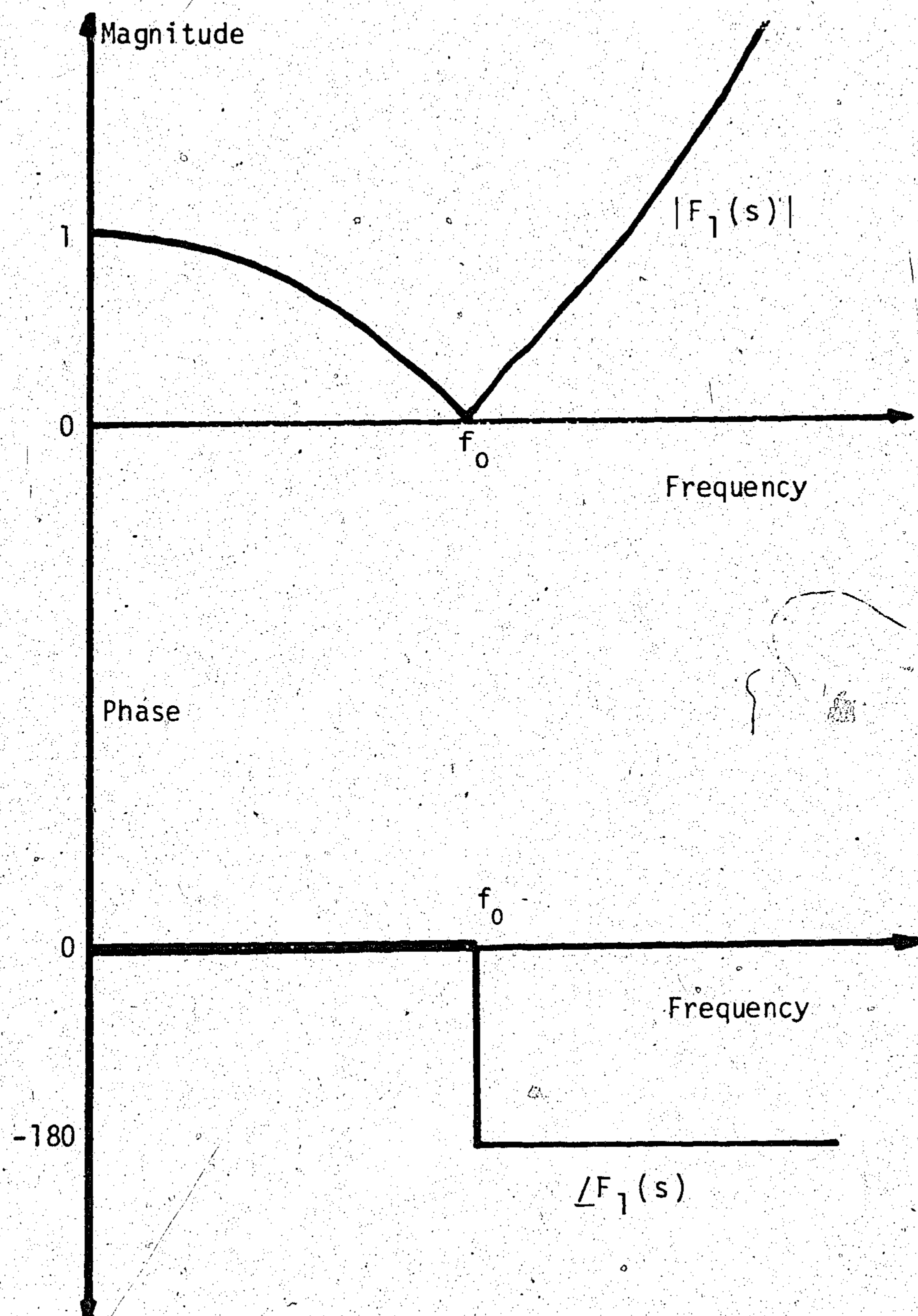

FIGURE 2. Gain and phase relationships for single segment model.

following frequency domain equation was derived in a similar manner as Equation 7, but a two segment model was assumed:

$$D_c(s) \approx \frac{m_1}{m_1 + m_2}\left(1 + \frac{m_2 l_1}{m_1 d_1} + \frac{K_1' d_1 s^2}{g}\right)X_1(s) + \frac{m_2}{m_1 + m_2}\left(1 + \frac{K_2' d_2 s^2}{g}\right)X_2(s) \tag{9}$$

where $K_1' = 1 - K_1 - h/d_1$, and

$K_2' = 1 - K_2 - (h + l_1)/d_2$.

The relation equivalent to Equation 8 for a three segment model can be shown to be of the form:

$$D_c(s) \approx F_1(s)X_1(s) + F_2(s)X_2(s) + F_3(s)X_3(s)$$

This analysis procedure produces three independent inputs, the horizontal movements of the body segments in the sagittal plane. Each input passes through a transfer function which indicates the forces required to move the relative body segment only. The sum of these forces is reflected in the position of the center of foot pressure. This serves as the basis of this study.

The relationship between body sway and center of foot pressure movement was modelled as a set of three filters and a summation block, as depicted in Figure 3. The system was simplified to encompass three input stimuli to which the responses were simply additive. The filter characteristics,

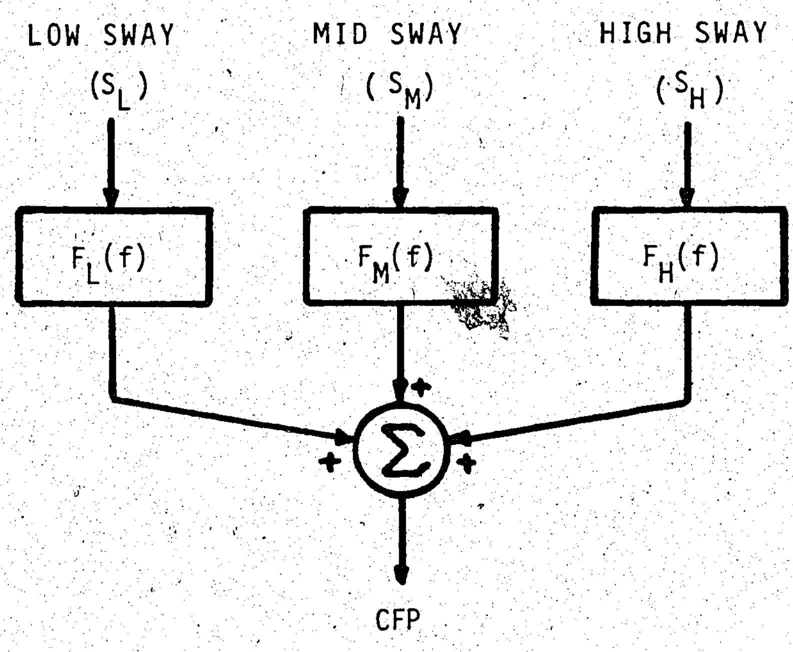

FIGURE 3. Control system model of relationship between sway and center of foot pressure.

each a complex function of frequency, determine the degree to which each body segment contributes to the center of foot pressure movement. A least squares regression analysis conducted at each frequency may be used to estimate the frequency response functions (Jenkins and Watts). Autopower and crosspower spectra (Appendix B outlines the method of calculation which follows the overlapping window approach of Welch) of the three sway components and the center of foot pressure movement are computed. Minimization theory (see Appendix C) yields the following equations relating the frequency components:

$$\begin{aligned} S_{CL} &= F_L S_{LL} + F_M S_{LM} + F_H S_{LH} \\ S_{CM} &= F_L S_{LM} + F_M S_{MM} + F_H S_{MH} \\ S_{CH} &= F_L S_{LH} + F_M S_{MH} + F_H S_{HH} \end{aligned} \tag{11}$$

where S_{aa} = auto power spectra,

S_{ab} = cross power spectra,

F_a = filter characteristic, and

a,b = C, L, M, or H = center of foot pressure, low, mid, or high sway components, respectively.

The transfer functions may now be found by solving the three equations for F_L, F_M, and F_H. The relative contribution of each sway component to movement of the center of foot pressure is then apparent.

2.2.2 Center of Foot Pressure

The most commonly studied aspect of static balance control has been movement of the center of foot pressure. A force platform is utilized to indicate the position of the single vertical force directed at a single point on the supporting surface which is equivalent to the sum of the forces exerted by the feet. Previous studies, some very recent, have considered the position of this force to reflect the horizontal position of the center of mass of the body (Baron; Njokiktjien; Pal'tsev; Ratov and Merkulov; Taguchi; Tokita et al; Tokumasu, Tashiro, and Yoneda). In fact, this is true only if the body is not moving at all.

Thomas and Whitney recognized this fact in 1959. Spaepen, Peeraer, and Willems questioned the validity of stabilograms as indicators of body movement. They found little relation between results from a camera system and those of the force platform. Spaepen, Fortuin, and Willems showed that movement of the center of mass did not match the stabilogram. Stevens and Tomlinson noted that this discrepancy was due to the force platform measuring the "secondary consequences of swaying movements" rather than the sway itself. Other researchers acknowledged this difference (Eklund and Lofstedt; Kapteyn, 1973; Valk-Fai).

With the slightest movement the acceleration and deceleration of the body mass adds to the torques at the body joints which resist the pull of gravity. With increasing frequency of movement, greater torque is required to resist or initiate this movement. The extra torque is generated by increasing the force or by increasing the distance from the joint at which the force is applied. Both means produce the same result on the force platform reading: the excursions of the center of foot pressure are greater than the movements of the center of mass of the body in the horizontal plane. The former movement tends to be an exaggeration of the latter.

Using a single inverted pendulum model, Gurfinkel (1973) calculated the error of using the center of foot pressure movement as an indicator of movement of the center of mass. At a frequency of 0.5 Hz, the displacement of the center of foot pressure was due as much to the forces of acceleration as it was to the displacement of the center of mass. Guersen et al (0.51 Hz) and Massen and Kodde (0.57 Hz) obtained similar results. Van de Mortel, Massen, and Kodde produced an equation describing the relationship between the acceleration measured by the force platform and the actual acceleration of the body.

In the proposed measurement system, the movement of the center of foot pressure is calculated from the recordings of forces at the four corners of a square force platform (some systems use triangular platforms). According to Figure 4 the displacement of the center of foot pressure in the sagittal plane from the center of the platform is given:

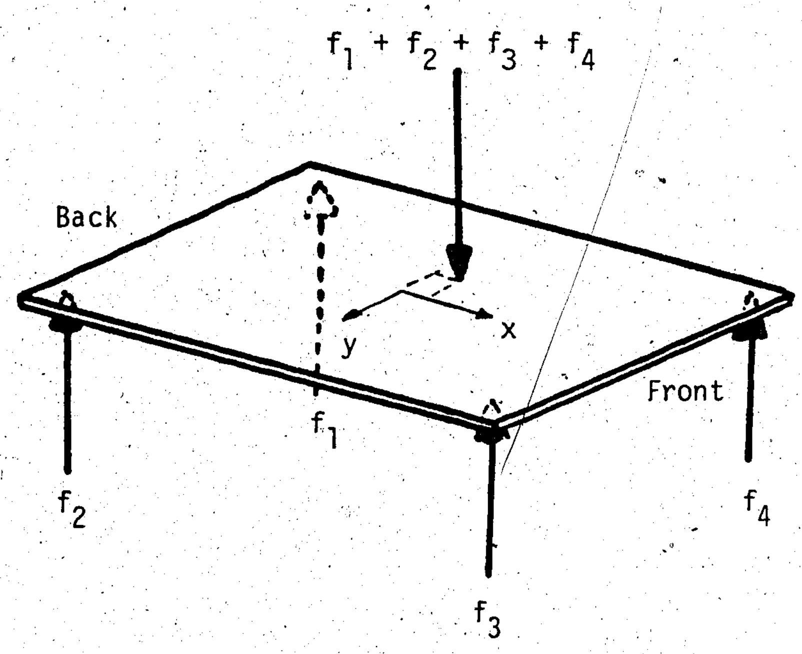

FIGURE 4. Distribution of forces on the force platform.

$$x = \frac{(f_3 + f_4) - (f_1 + f_2)}{f_1 + f_2 + f_3 + f_4} \frac{d}{2} \tag{12}$$

where d is the distance between the front and back points of support. The sum of the four forces in the denominator corresponds to the total weight of the subject, and is assumed to be constant during the testing period.

2.2.3 Sway of the Segmental Body

It has been observed that the human body does not act as a simple inverted pendulum. Aggashyan's spectral analyses showed that the observed frequencies of movement are lower than those calculated for a simple inverted pendulum. Roberts and Stenhouse concluded that a time lag, existing between movements of the head and the hip, was due to some bending at the latter. Two segment models were proposed to include this bending of the body at the waist or hip joint.

Although the lateral malleolus is the obvious point of rotation at the ankle, choosing a singular point of rotation between the legs and the torso is more difficult. There is possible bending from the hip joint all the way up the spinal column to the neck. However, it appears that rotation about the hip joint is mainly a gross movement, and not normally involved in fine postural adjustments. Electromyographic studies have recorded only slight activity in the thigh muscles of quietly standing subjects (Joseph, Nightingale, and Williams; Portnoy and Morin). Similar observations were made at the hip (Joseph). Jonsson and Steen concluded that the thigh muscles would control lateral sway, with little effect on sagittal sway.

Muscle acivity along the spinal column is largely confined to the lower back and abdomen during quiet standing (Joseph and McColl). In studies of movements of parts of the body in relation to one another, Hirasawa (1976) concluded that the pivotal point between the upper and lower body was in the vicinity of the lumbale (in the region of the lumbar vertebrae). Therefore, the height of the second rotational point was measured at the ileac crest, which usually coincides with the upper edge of a belt.

Some three segment models divided the leg in two at the knee (Hemami and Jaswa; KoozeKanani et al). Such studies have concluded that relatively little rotation occurs about this joint (Gantchev, Draganova, and Dunev, 1979; Valk-Fai). Nashner and Woollacott observed relatively little knee rotation even with the knees slightly bent. The representation in a model of the leg as one segment with the knees protracted is therefore valid.

In the present study, the upper body is divided into two segments: the torso (including the arms) and the head. About 70% of the body mass is above the hip joint (Thomas and Whitney; Squire). The torso contains 89% of the mass of the upper body. Its movement should have a significant contribution to the movement of the center of mass of the body. The head, on the other hand, houses the postural control mechanisms of vision and the vestibular apparatus, as well as proprioceptors in the neck joints and muscles. Movement of the head may directly affect postural reaction of the lower body segments, despite not having a great effect on the displacement of the center of mass itself. Nashner (1973) notes that the orientation of the head is important to the operation of the vestibular apparatus. The pivotal point between the torso and the head is chosen by observation.

It is noted that measurements of the joint heights, taken on a number of different occasions, for one individual subject have varied by only 13 mm. Appendix D contains test data which illustrates the minor effect of this source of error.

The adopted model is depicted in Figure 5. Considering independent movements of the three segments, and the resultant torques generated about the joints, the control of stability for this admittedly rudimentary model of the human body becomes increasingly complex. This initial investigation of the control system concerns the relationships between the movements of the three body segments and the displacement of the center of foot pressure in the sagittal plane. It is hoped that these relationships will be of a less variable nature than measurements of body sway or center of foot pressure movement separately.

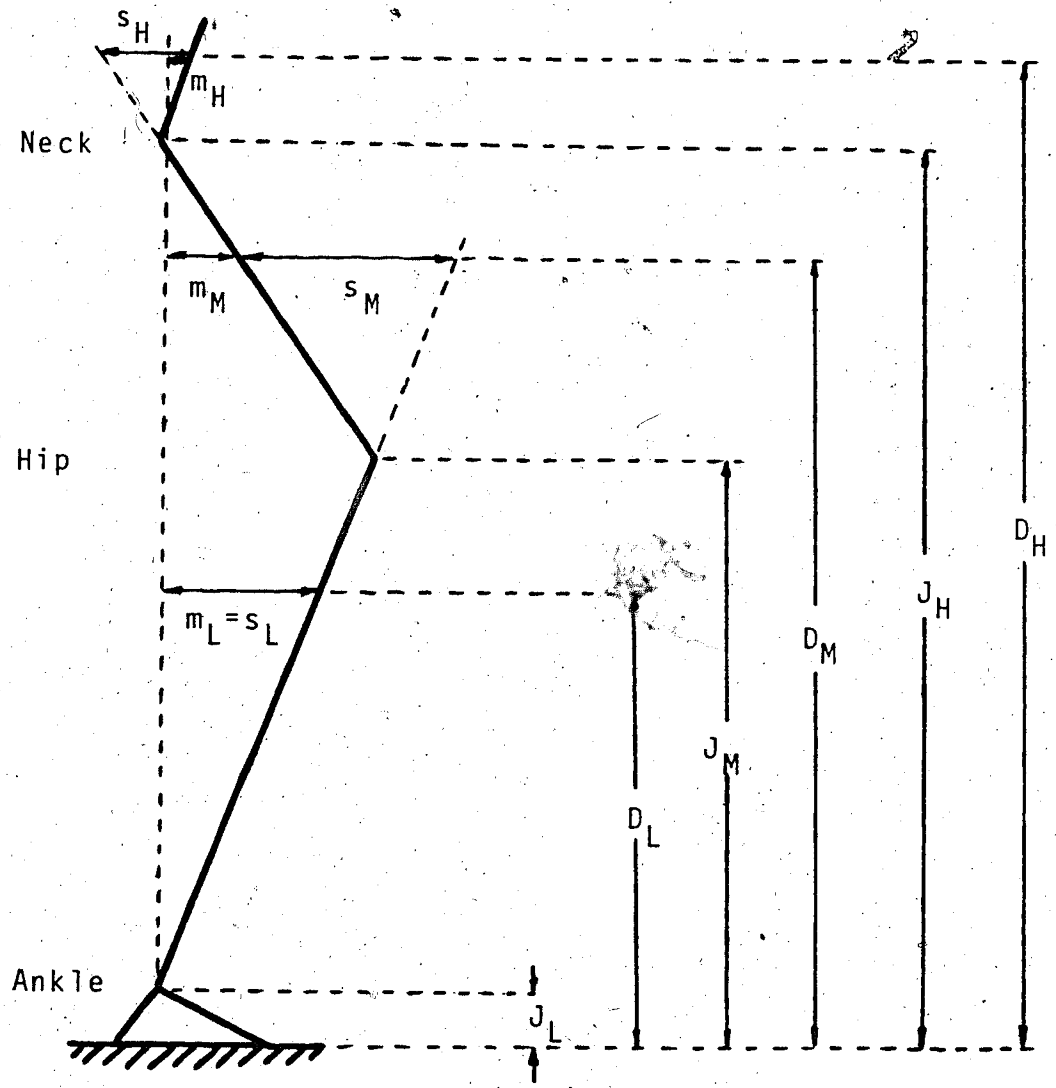

FIGURE 5. Three segment model of human body in sagittal plane.

In order to compute the contribution of each body segmental sway to the movement of the center of foot pressure, each sway component must be isolated from the other two. A large part of the measured sway of the upper body is due to the sway of the lower body. The cancellation of this effect of lower body movement was achieved by translating the sway measured at the low site to the heights of the middle and high measurement sites and subtracting from the values measured at the higher sites. Similarly, the high sway was further reduced by the middle sway translated up to the high measurement site. This process is illustrated in Figure 5 also. The following equations describe the operation:

$$
\begin{aligned}
s_L &= m_L \\
s_M &= m_M - m_L \frac{D_M - J_L}{D_L - J_L} \\
s_H &= m_H - m_M \frac{D_H - J_M}{D_M - J_M} - m_L \frac{D_H - J_L}{D_L - J_L}
\end{aligned}
\tag{13}
$$

where s_a = calculated values of sway,

m_a = measured values of sway,

J_a = height of joint from supporting surface,

D_a = height of detector from supporting surface,

a = L, M, or H = low, mid or high measurement sites, respectively.

The effectiveness of the cancellation was demonstrated by pivotting a rigid board before all three sway detectors simultaneously. After computer processing of the data only the low detector data indicated any appreciable sway.

It is duely noted that this analysis tends to sidestep the problem of noise within the recorded signals. The addition and subtraction of the force platform signals cancels some noise content. Little correlation is seen between the noise in the force platform signals and in the sway detector signals. Therefore, considering the very low power content also, such noise has a negligible effect on the transfer function analysis.

3. APPARATUS

The measurement devices used in recording the movement of the body have varied considerably in complexity and flexibility. From "wabblemeters" (Moss) to "electro-gravitiometers" (Tsukimura) the two most commonly observed parameters have been the movement of the center of foot pressure and the sway of the body. Some studies have included electromyographs, ballistocardiograms, or other physiological data.

3.1 The Force Platform

The force platform is used to indicate the movement of the center of foot pressure. Force sensing elements at the platform supports detect changes in weight distribution on the platform surface. Strain gauges and piezoelectric crystals are the most commonly used force transducers. Although they are more sensitive and linear in response, piezoelectric elements require more complicated, and expensive, circuitry. Dichgans et al (1976), Mauritz et al (1979), and Spaepen et al used piezo-electric force transducers. Other force platforms employed scales (Cureton and Wickens; Hasselkus and Shambes; Waterland and Shambes), mechanical linkages (Cantrell; Sheldon; Travis), or electromagnets (Baron, Molinie, and Vrillac; Hellebrandt and Braun). (See Kapteyn, 1972b or Green and Morris for studies of force platforms.) The existing force platform was constructed with strain gauges which were deemed to be of adequate accuracy and linearity.

Pairs of strain gauges were secured to the top and bottom surfaces of four aluminum cantilever beams which supported the corners of a square aluminum plate (see Figure 6). This pairing of strain gauges in Wheatstone bridges provided a temperature compensated voltage signal proportional to the vertical force exerted on the cantilever beam. The beams and plate were rigid enough so that any deformation was imperceptible. The Wheatstone bridges were DC-coupled to preamplifiers which provided amplification and calibration for the voltage signals. Appendix E gives the circuit details. An 18 kilogram weight in the center of the plate seated it securely on the supporting beams during calibration. After zeroing the preamplifiers, a further 18 kilograms were added to adjust the gains for equal outputs. Sugano and Takeya used a steel ball rolling around in a circle on the platform to calibrate their device. A similar system may be adopted in the future.

As in any mechanical system, the force platform was found to have resonant frequencies, at which responses would be distorted. Verduin et al suggested a motor driven spring or a weight dropped in the center of the platform as means to determine these frequencies. In the present study a linear motor was used to excite the force platform over the frequency range of interest. A resonant peak was observed at 22 Hz for the unloaded platform. With a subject upon the

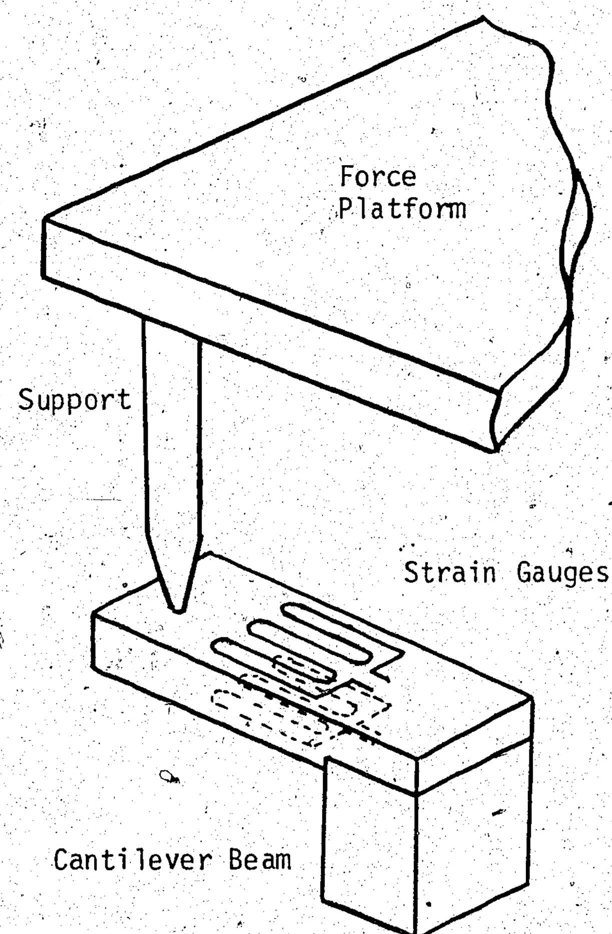

FIGURE 6. Arrangement of strain gauges of force platform.

platform the peak frequency was shifted down to 17 Hz. This information confirmed the lack of mechanical effect of this part of the measuring apparatus upon the readings. The resonant peaks existed above the frequency range of interest.

3.2 The Sway Detectors

Body sway measurements have been procured with a wide variety of equipment. Some of these measurement systems have involved:

1. photographs or television (Boman and Jalavisto; Hemami and Jaswa; Hirasawa; Kapteyn, 1973; Koozekanani et al; Murray, Seireg, and Scholz; Spaepen ,Peeraer, and Willems; Willems and Vranken),
2. displacement potentiometers (Coats; Dzendolet; Fernie and Holliday; Nashner, 1976; Valk-Fai),
3. goniometers or angle potentiometers (Herman et al; Litvintsev; Mamasakhlisov; Mauritz et al, 1979),
4. linear variable differential transformers (Shipley and Harley; Stribley et al),
5. 'displacement transducers' (Tomlinson and Stevens; Viidik and Magi),
6. accelerometers (Miyoshi; Nilsson; Stiles and Rietz; Thomas and Whitney),
7. capacitive plates (Lee and Lishman),
8. ultrasonics (Herman et al; Le Roux et al; Roberts and

Stenhouse),

9. mechanical linkages (Witkin and Wapner; Wright), or

10. hand-held laser on a photodiode (Yoo et al).

The disadvantages of such methods include cost, poor sensitivity (see Kapteyn and de Wit or Shambes), and patient contact which may adversely affect natural movement. Light beams circumvent these problems to a large extent.

A collimated beam impinging upon a photovoltaic detector would produce an output linearly proportional to the area of the detector that was illuminated. A single laser beam or an array of smaller lasers wide enough to detect the expected range of motion was considered impractical due to cost. A system utilizing a single beam swept across the range of motion would encounter complicated mechanical and timing problems.

The present system used a point light source at the focal point of a parabolic mirror to produce the collimated beam. Initial tests found 150 watt mercury arc bulbs to produce too much heat, while not being exceptionally small sources. The adopted system used Oriel xenon arc bulbs (#6341), which produced a 0.18 mm (0.007 inch) round arc while requiring only 2 watts of power. A 1000 volt trigger pulse was required to produce the arc which was then stabilized by 200 volts before the 27 volt running condition was attained. The Oriel point light source power supply (#6342) was inadequately regulated, and the bulbs were switched over to a better regulated power supply after being lit. Shielding was necessary to prevent the high voltage pulse from deregulating this second supply.

The mirrors were made in the optics laboratory of the Department of Electrical Engineering. The parabolic mirrors had a diameter of 108 mm and a focal length of 311 mm. Since only a thin light beam was required, only a small section of the mirror was used. Each mirror thus provided two sections. To further save space, the focal length was folded back with a plane mirror. The light source could then be located immediately below the parabolic mirror section. Figure 7 depicts the source optical system.

The plane mirror swivelled about its longitudinal axis to aid in focussing. The apparent vertical height of the point light source could then be adjusted. The degree of collimation of the light beam depended upon the proximity of the point light source to the apparent focal point of the parabolic mirror. Horizontal travel of the light source from front to back and from side to side completed the focussing adjustments. After passing through a slit the resulting collimated beam was trimmed to 6.4 X 102 mm. These dimensions were judged wide enough to encompass all but grossly abnormal sway.

The light beam was directed onto a 4.6 X 102 mm diffused junction silicon photodiode (Quantrad Corporation Long Line Series #LL4). The diode had +/-5% uniformity and a 1.8 usec response time at 10 volts. The diode was connected as a photovoltaic element in the input stage of a

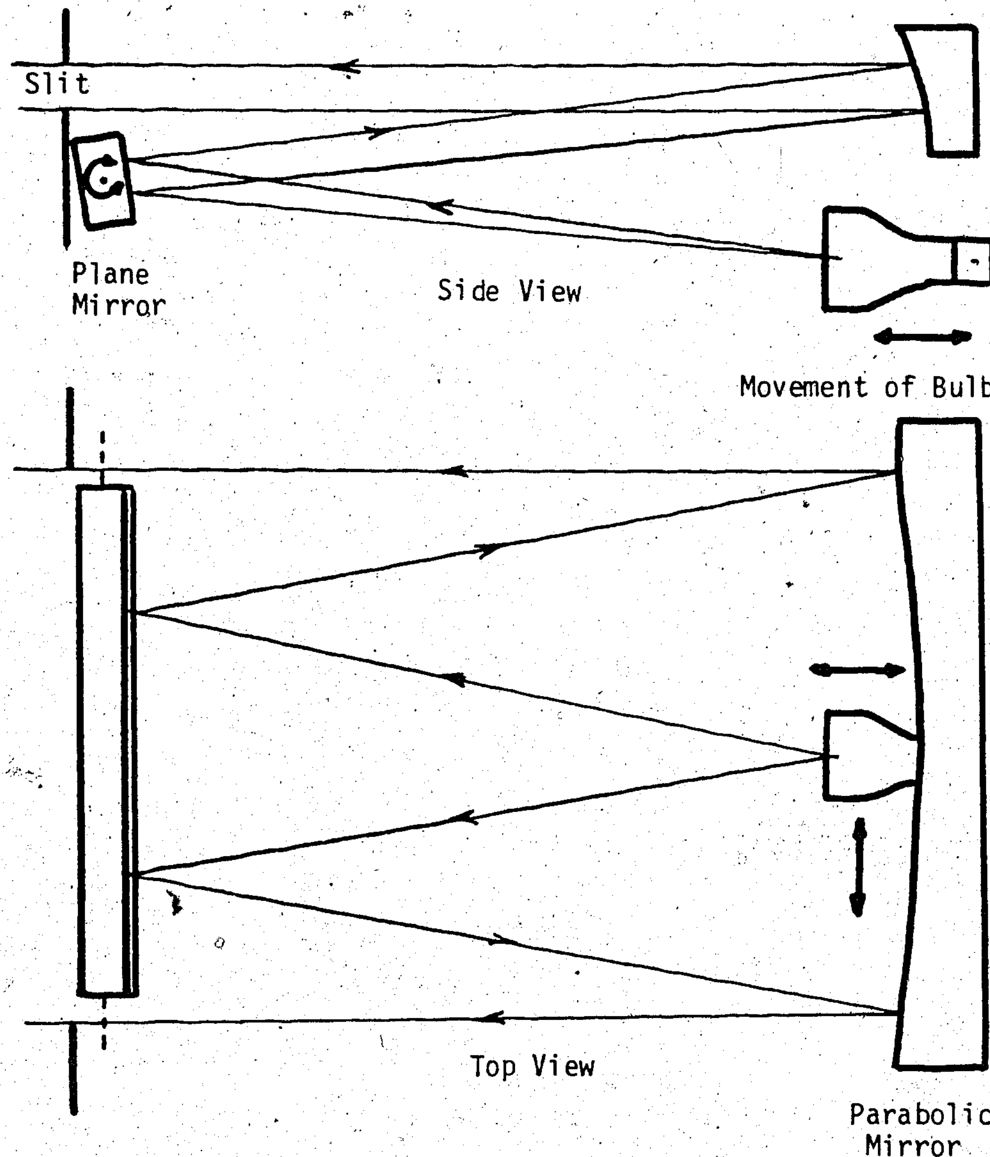

FIGURE 7. Optical system of light source for sway detector.

preamplifier. The output was then linearly proportional to the light incident upon the diode (see Appendix F for circuit details). The diode was recessed in a box, which was painted matte black inside, to reduce the effects of stray light from other sources.

The linearity of the sway detectors was measured by rotating a specially machined disc at a constant speed to sweep through the light beam. The radius of the disc increased linearly with angle from 0 to 180 degrees around the disc, and then again from 180 to 360 degrees. This symmetrical form balanced the disc about its center, providing more uniform rotation. The detector output should then ideally have been a sawtooth wave, the accuracy of which could be measured (see Appendix G).

The light beams were observed to vary in intensity during the day. This variation sometimes exceeded 5% of the average intensity. Calibrations of the sway detectors would considerably reduce this source of error, while allowing the amplifier gains to be freely adjusted before each test. Recorded calibrations would have also accounted for the aging of the light sources and changes in the power supply voltages.

Calibration of the sway detectors was accomplished by rotating an off-centered circular disc in the light beam. The detector output was approximately sinusoidal. By introducing a known reproducible movement to each of the three sway detectors, their relative sensitivities could be compared. During data acquisition these calibrations were recorded and used during the processing of the data from the sway detectors.

Each source and detector set was mounted 70 cm apart on a frame which slided forward and backward on a collar. Each collar was mounted on a vertical column so that the height of the source/detector set between the force platform was continuously adjustable from 66 to 203 cm. The column was set into a heavy wheeled base, upon which the force platform was adjustable forwards and backwards. The entire measuring system was very heavy and sturdy in order to reduce the effects of vibration and deformation. The base was also set upon anti-vibration pads. A set of stairs and a safety bar were provided since the platform was 51 cm above the floor. Figure 8 is a sketch of the complete measurement apparatus.

3.3 Data Acquisition

The seven signals, four from the force platform and three from the sway detectors, were passed through a set of filter/amplifiers. The bandpass filters provided 20 dB/dec (6 dB/oct) attenuation with 3 dB cut-off frequencies at 1, 3, 5, 10, 30, and 50 Hz low pass and 0.1, 0.3, 0.5, 1, 3, and 5 Hz high pass. The high pass filtering served to prevent the large amplitude of very low frequency sway from causing the input signals to be clipped at the analog to digital convertor. The effects of respiratory and cardiac

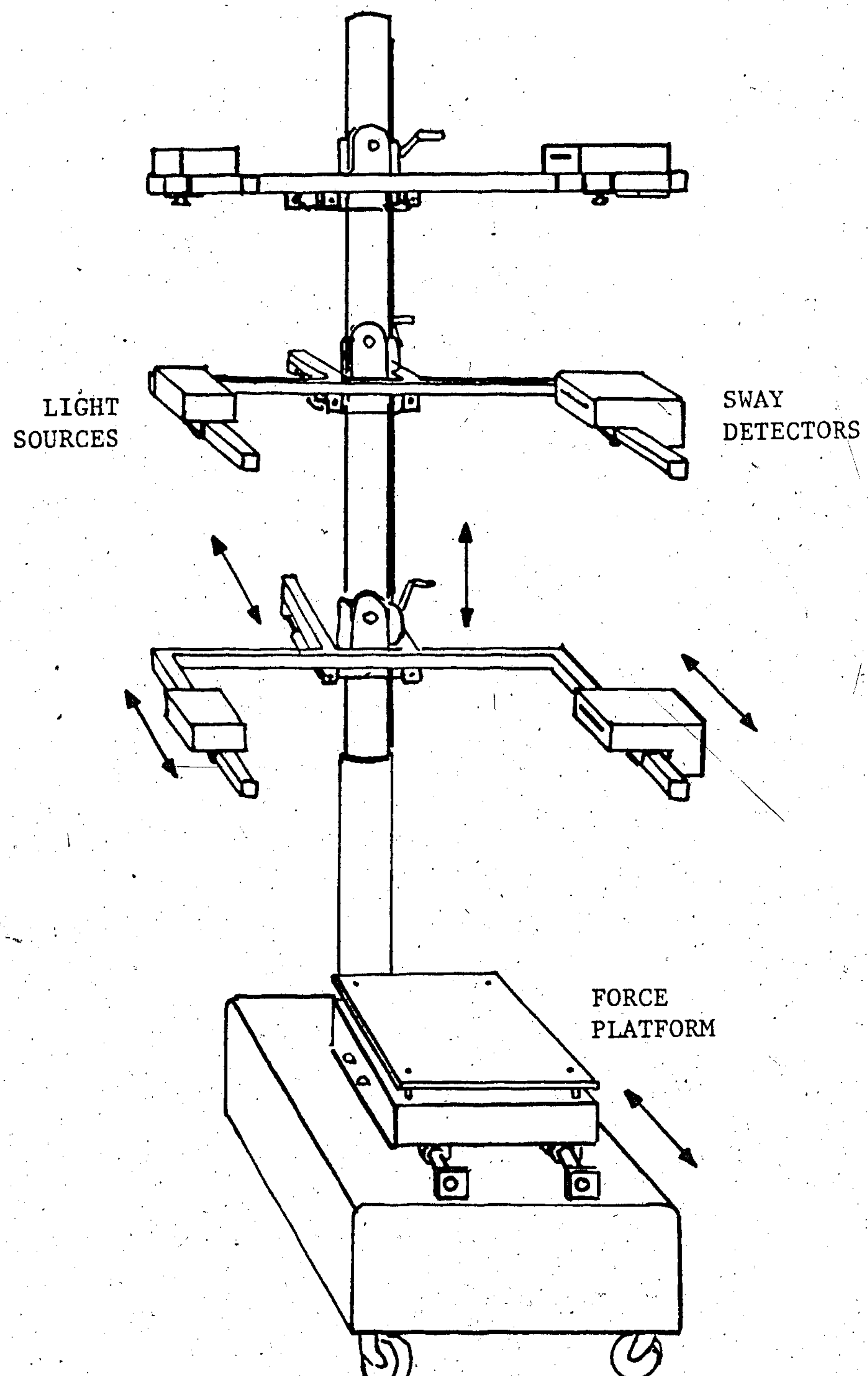

FIGURE 8. Measurement apparatus for body sway and movement of the center of foot pressure. The arrows indicate the possible adjustments.

artifacts, conscious movements, and other low frequency phenomena not of interest were also reduced. Seidel et al (1978) used high pass filters to eliminate "trend-similar sways" evident with variations in foot position and inter-individual performances. Low pass filtering reduced vibration-induced noise and 60 Hz interference via stray light and electrical lines. The amplifier gains were adjusted to produce a maximum signal of +/- 2.5 volts for the H.P. 5610A A/D convertor, which received the seven outputs. The gain was adjustable from 10 to 110.

Each of the data channels were subjected to the same degree of filtering. Therefore, in the transfer function analysis, the filters would not affect the results. The relationships between the channels would not be changed.

The seven channels were alternately sampled at 64 Hz, which was four times the maximum frequency of interest. The Nyquist frequency was exceeded in order to provide a large amount of data and greater frequency resolution for subsequent analysis, if necessary. The digital data was stored on magnetic tape in the form of 16 bit words. The first 6 bits provided channel identification, while the last 10 binary digits constituted the data value. This provided 1024 quantization levels, which resulted in negligible quantization noise with regards to further data processing (Glisson, Black, and Sage). The time required to sample each channel was 10 usec, and all seven channels were sampled in less than 100 usec.

The maximum sampling rate was 12 kHz. The maximum number of channels was 16. Thus, 16 channels could be sampled at 750 Hz each. Within these frequency and channel limits, data could be sampled and stored on digital tape. The data could also be displayed an a CRT, but 500 Hz became the fastest sampling rate. This was the maximum frequency for which both storage and display could have occurred simultaneously. Figure 9 is a schematic of the handling of the signals from the measurement apparatus.

3.4 Data Processing

An HP2100S computer was used for both data acquisition and analysis. The latter task was divided into two parts: processing and display.

Two programs were involved in the initial processing of the data. Program PCS1 performed data reduction and modification. The four force platform signals were combined to provide one signal which indicated the movement of the center of foot pressure in the forward/backward orientation of the platform. This signal was also normalized with respect to weight of the subject. The sway detector signals were equalized using the calibration recordings which preceded every test. The effect of the movements of lower segments of the body was then cancelled, providing three linearly independent sway measurements. These could be expressed either as displacement angles or linear

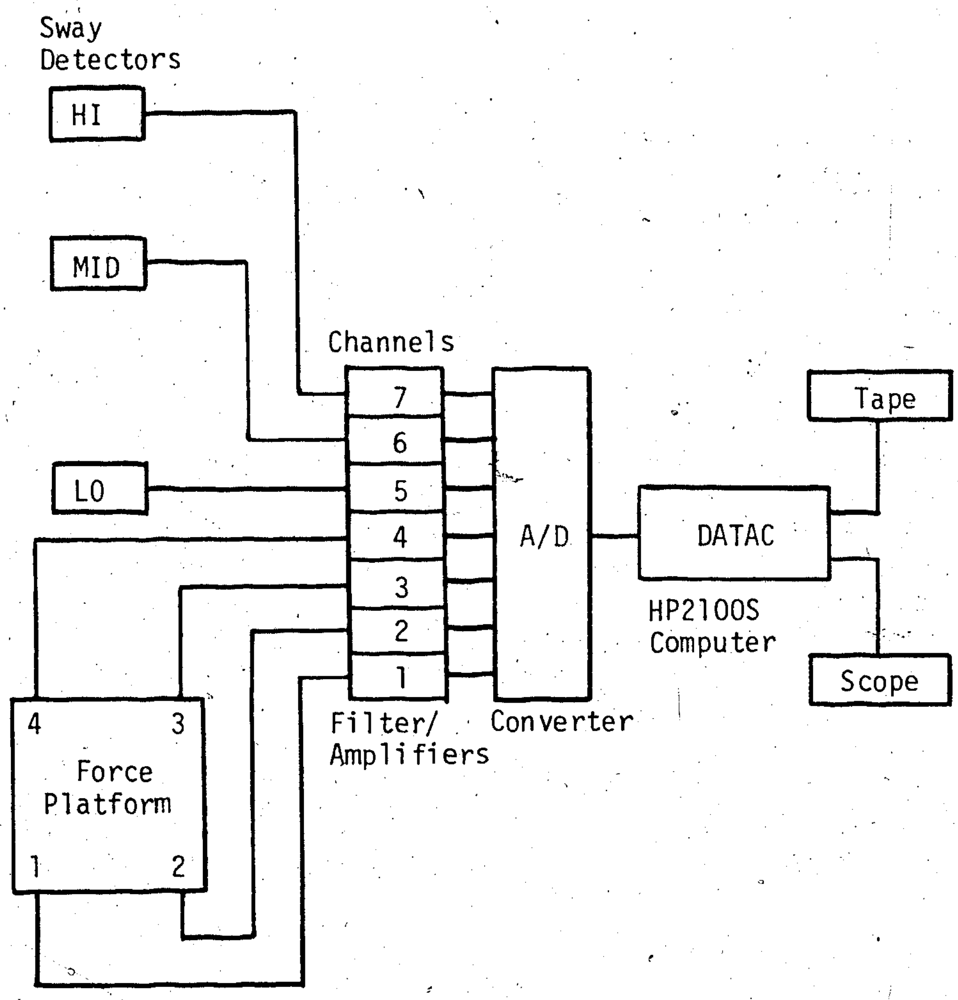

FIGURE 9. Data acquisition schematic.

displacements for each body segment. The four channels of reduced data were written onto magnetic tape.

The sway angle subtended at the leg was much smaller than the sway angle at the head due to the lengths of the body segments involved. This introduced digitization problems. The angles could be arbitrarily scaled, but the shapes of the transfer function curves were dependent upon the multiplicative constants. Displacements were used in this study to avoid the problem of optimizing these constants. Each displacement was transferred to the top of the respective body segment. A marginal decrease in variance over the angular representation was observed.

The data could be displayed as the movement of a three segment mannekin on an oscilloscope screen (similar to Roberts and Stenhouse) with program SMND. However, further processing was required to continue the transfer function analysis.

Program PCS2 provided a frequency analysis of the reduced data. The program used only every second data value, thus reducing processing time and memory space. Initially, the data was digitally low pass filtered to prevent aliasing. However, the filter was seen to reduce some high frequency content of interest. There also appeared to be negligible power at frequencies above 16 Hz even before the filter was applied. The filter has been excluded in recent processing. The data then passed through a 128 point triangular window (see Appendix H) and FFT subroutine (see Appendix A), providing a Fourier transform over the frequency range of 0 to 16 Hz with 0.25 Hz intervals.

With each channel of data represented by real and imaginary Fourier components in the frequency domain, the power and cross spectra were calculated as suggested by Welch (see Appendix B). An overlap of adjacent blocks of data effectively increased the number of available data samples. Each calculation of a transform required 32 new data values. The transformed data was filtered to eliminate spurious coherences due to trends (ie. smoothed), and thus reduce variance. Slowly changing artifacts like amplifier drift or gross disturbances to the system, which escaped the initial low pass filtering, would then be attenuated.

Frequencies which were prevalent in each channel and common to pairs of channels were now evident. The coherence between each of the sways and the center of foot pressure indicated the degree with which each sway was linearly related to the movement of the center of foot pressure. Minimization theory provided a means with which to formulate the transfer functions describing the reactions of the body, as indicated by the center of foot pressure, to the movements of its segments.

Other programs, denoted by GRF and PCS, provided displays of individual and averaged data, respectively. The transfer functions could be specified by real and imaginary components or by gain and phase relationships. Differences and ratios could be calculated also.

The impulse responses of the transfer functions and differences between transfer functions could similarly be calculated and displayed. Computation of the impulse responses served to reduce the two sets of transfer function components to one set of results as well as reveal further details of the control system model. The frequency domain data was smoothed with an asymmetrical Hanning window (in Appendix I the window tailoring process is described) to preserve the more informative frequency characteristics.

Appendix J contains the flow charts for PCS1, PCS2, and PCS6, which averages the impulse responses.

4. EXPERIMENTAL PROCEDURE

4.1 Test Standards

In recording the steadiness of subjects, it was decided that a key requirement was comfort (see Terekhov, 1978). To obtain a characteristic response indicative of a person's balance, it was thought that a natural, unenforced stance was necessary. The Romberg position, with heels and toes together, was used in many studies since it enhanced sway, being inherently less stable. It was also easily reproducible. The 'attention' position, with heels together and feet angled at 45 degrees, exhibited much sagittal sway, while lateral sway decreased with the wider area of support. The 'at ease' position, with heels apart and feet angled outward, produced the least sway in both planes, but was the most comfortable for most subjects.

Since the present study sought a relationship rather than absolute quantities, the amount of sway should have been unimportant. With regards to feet positions, the effect of training or practise of some individuals might have given them an 'advantage' with an enforced stance. Therefore, the 'at ease' position was adopted for testing.

Another consideration was footwear. Again the criterion of comfort was used (Miles, 1922). Most subjects felt more 'natural' wearing everyday shoes as opposed to bare feet or slippers. Also, no special clothing was required. In some cases, a surgical cap was worn to flatten the surface presented by curly hair. It was noted that such flexibility exists in the clinical Romberg test (Mayo Clinic and Mayo Foundation).

Each subject was introduced to the experimental procedure and apparatus. The subject assumed a comfortable stance on the force platform, with the body weight distributed equally on both feet. The subject faced away from the apparatus to reduce psychological effects. Meanwhile, the sway detector heights and amplifier gains were adjusted. The sway detectors were generally raised to maximal heights on the backside profile. The contact between clothing and skin was also considered, where a tight fit was desirable. The posterior surfaces of the thighs or buttocks and the back provided a tighter fit as well as being less prone to muscular disturbances like respiration. Placing the sway detectors to the rear of the subject also reduced the chance of damage to the eyes from the intense ultraviolet light source.

The subject was then allowed to step down and relax, and his/her physical characteristics were noted. The sway detector calibrations were recorded at this time. The subject then resumed the testing position on the platform, with knees locked. Portnoy and Morin recorded a slight increase in leg muscle tension as the body's center of mass shifted forward when the knees were locked. The overall balance was not adversely affected, however. The subject's arms were held in front of the body to prevent interference with the sway detectors and to pull clothing more tightly against the back. The subject was asked to make no conscious movements like talking, coughing, or sneezing.

Watanabe et al (1976) observed that breath-holding decreased body movement, but this was probably the effect of a mental task on balance control. As mentioned earlier, most tests had not revealed a relationship between breathing and body movement (see also Miles, 1922 or Travis, 1945).

The test interval in other studies has varied from 30 seconds to 5 minutes. Bjerver and Persson noted that fatigue became prominant after 5 minutes. Seidel et al observed an increase in low frequency movement with a corresponding decrease in high frequency oscillations of the body as the test period was lengthened. Miles concluded that 2 minutes was an optimal length for a test. The adopted test interval of 2 minutes was used to provide enough data samples (7,680 per channel) and yet avoid fatigue. Rarely did a subject complain about the length of the test. A session was comprised of two or three such intervals: the first with eyes open (EO), the second with eyes closed (EC), and, if a third, with eyes open again.

The amount of visual information available appeared to be critical to the EO test. The field of view should be consistent in illumination, depth, and fullness. Since any movement within the field has been seen to induce sway, such distraction was avoided. Acoustic disturbances were also minimized, especially during an EC test (Miles, 1922).

If the subject felt discomfort or fatigue after a test, a brief rest period was allowed before starting the next test. The EC test was performed under similar circumstances as the eyes open test. The subject was asked to close his/her eyes and, after 5 to 10 seconds of acclimatization, the data was recorded. If the subject was willing, a third test was conducted. A second EO test was administered to possibly demonstrate the effects of fatigue. Again, consistent conditions were maintained.

4.2 Test Groups

The normal subjects had no known neurological disorders, were not under the influence of alcohol or drugs, and were relatively well rested. The primary test group consisted of 20 normal males, ranging in age from 21 to 44 (average 30, standard deviation 7.2). Half of this group wore glasses. A group of 7 normal females were also tested. They were aged from 23 to 32 (average 27, standard deviation 3.6). An EO and an EC test were recorded for each subject. The tests were performed over a period of 9 days.

A 27 year old normal male, not included in the large group, was tested 20 times over a period of 4 days. Three tests were performed on each occasion. Training has been found to improve the performance of subjects in tests of balance (Fearing, 1924b; Holliday and Fernie; Newell and

Wade; Ryan). An increase in steadiness with repeated testing is especially noticeable with more difficult conditions, like EC tests (Dickinson, 1974; van Parys and Njiokiktjien). It is plausible that any decrease in variance that may be expected by using a single subject might be negated by a change in the data due to training.

\A later series of tests were conducted to measure lateral sway as well as sagittal sway on this individual. The subject merely faced one side of the apparatus to enable the sway detectors to register his sideward movement.

The test procedure for a series of alcohol tests was similar. These tests were done in conjunction with psychological and cerebral blood flow tests. The subjects were tested in pairs, first in a sober state. An EO test was followed by an EC test.

In an attempt to reach the minimum state of illegal intoxication, each subject drank an amount of rye whiskey calculated to bring the alcohol level in the blood to 0.08 ml/100 ml. This dosage was equivalent to a 6 to 7 ounce drink. (Begbie, 1967, used 31.5 g./65 kg. body weight.) The subjects were allowed to drink the alcohol in a preferred fashion within an allotted time of 30 minutes.

An hour after finishing their drinks, each subject was submitted to a breathalyzer test to determine the level of intoxication. The three tests, including the posturographic test, were then administered. The breathalyzer and posturographic tests were repeated at later times as the subjects blood alcohol level returned to normal.

Three patients with physical disorders were tested. A 75 year old female with Parkinson's disease, a 42 year old male with a peripheral neuropathy, and a 37 year old male with cortico-spinal tract disease were subjected to EO and EC tests.

5. RESULTS

The results are presented for six sets of tests:

1. a population of 20 normal males,
2. a population of 7 normal females,
3. a normal male tested 20 times,
4. alcohol tests on a population of 7 normal males,
5. 3 patients with neurological disorders, and
6. 6 tests each of sagittal and lateral movement on the subject that was used in the third set of tests.

The statistical treatment of the data is rudimentary. Appendix K provides a brief description of the techniques used.

5.1 Tests of Normal Males

5.1.1 Power Spectra

The averaged power spectra from the EO tests for a group of 20 normal males are shown in Figure 10 following. The shaded areas represent the population contained within the confidence limits of one standard deviation on either side of the average for the movement of the center of foot pressure and the low, middle, and high sways. Figure 11 shows the corresponding EC power spectra.

The high degree of variance is immediately apparent. The vertical and horizontal axes are drawn to the same scale to facilitate comparison. Although the absolute magnitudes

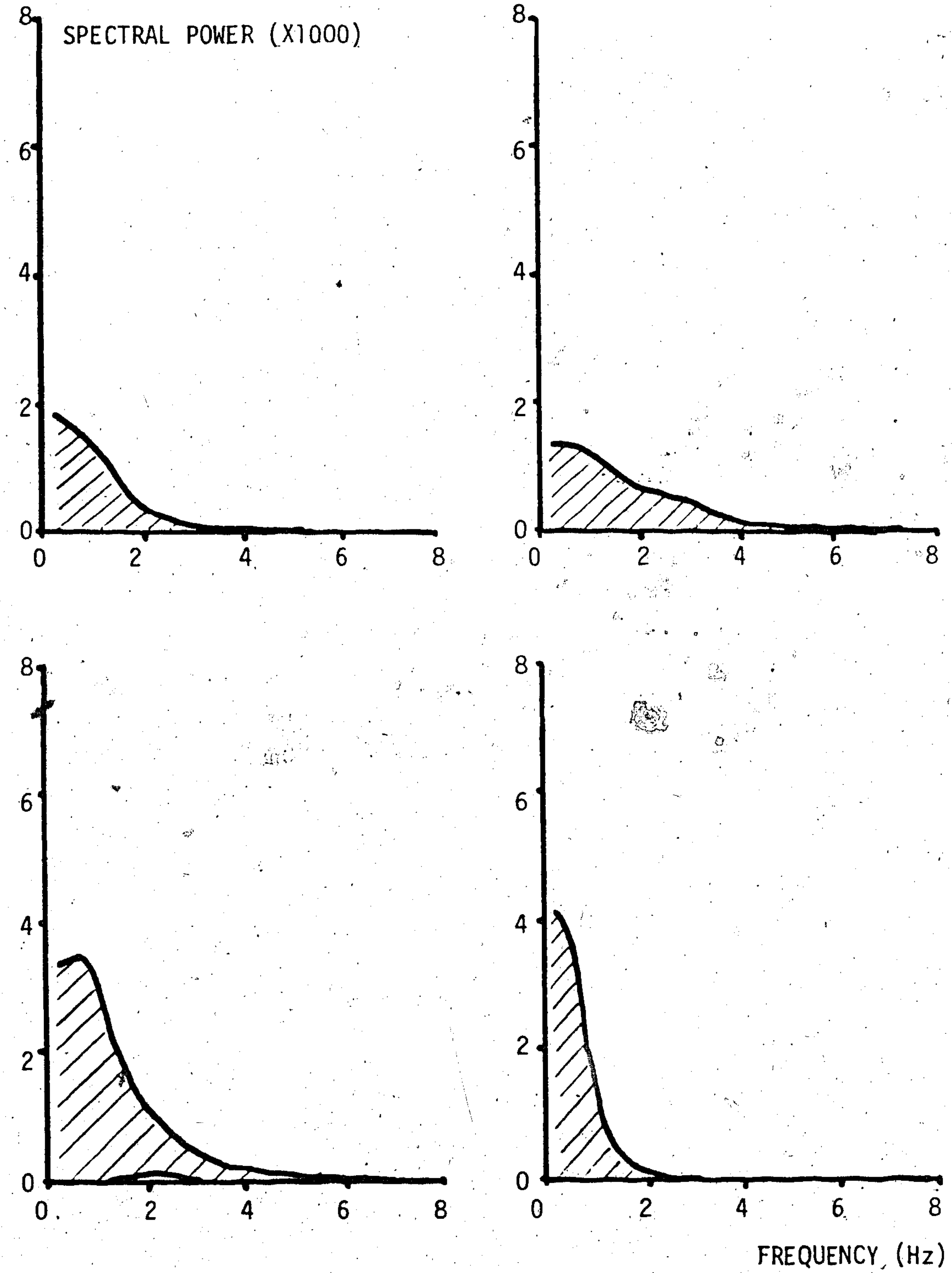

FIGURE 10. 68% confidence limits of averaged power spectra of EO tests of 20 normal males: CFP - bottom left, LO sway - bottom right, MID sway - top left, HI sway - top right.

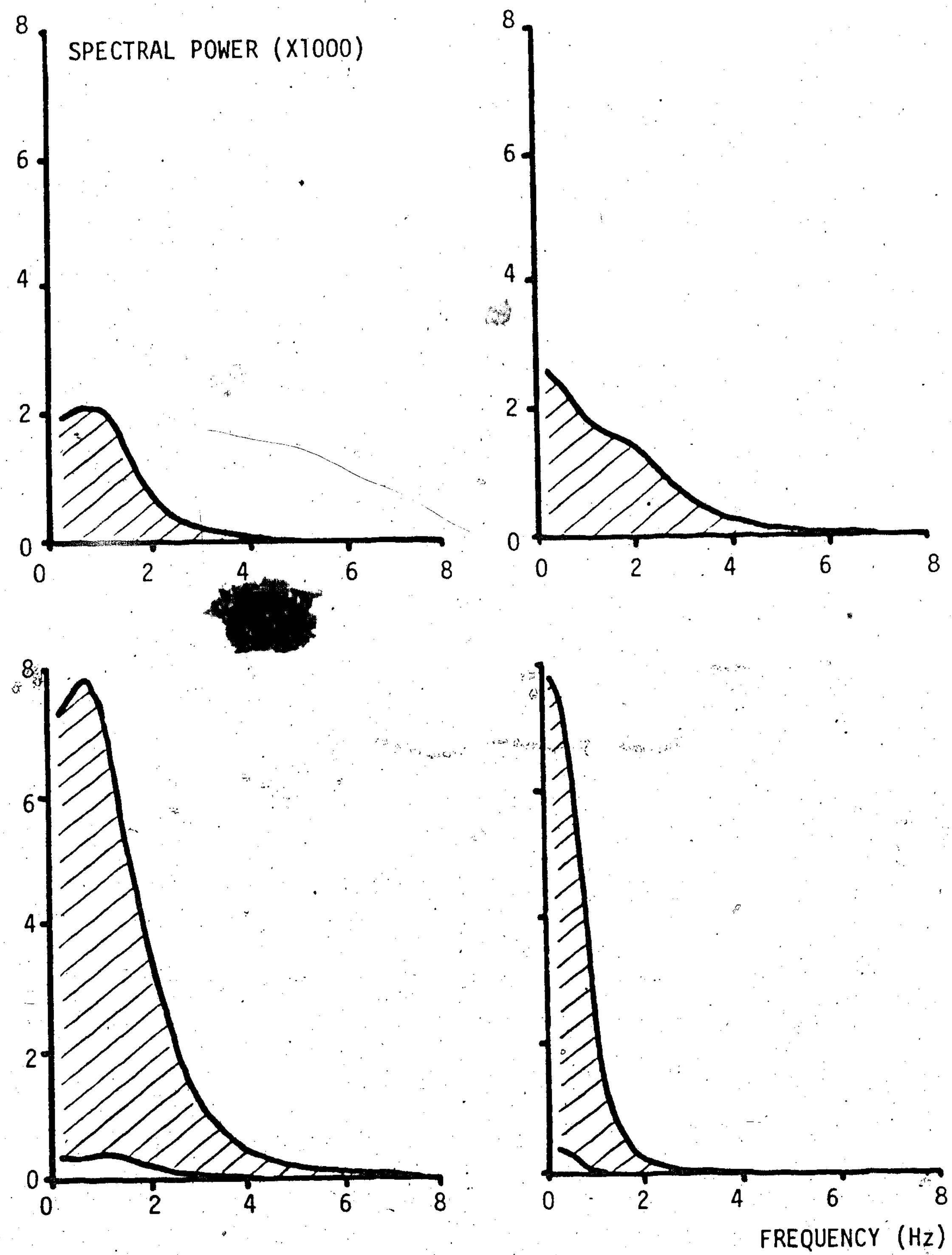

FIGURE 11. 68% confidence limits of averaged power spectra of EC tests of 20 normal males: CFP - bottom left, LO sway - bottom right, MID sway - top left, HI sway - top right.

are unimportant, the graphs do demonstrate the almost complete overlap of the EO power spectra by the EC power spectra. The two conditions would therefore be indistinguishable using these spectral curves.

The spectra do reveal the rapid drop in power above 1 Hz. The upper frequency limits (determined asymptotically) for the four body movements under the EC condition are approximately: center of foot pressure at 3.0 Hz, low sway at 1.3 Hz, middle sway at 2.5 Hz, and high sway at 4.0 Hz. The EO condition yields slightly lower frequency limits.

The relative magnitudes of the spectra of the body movements indicate that the center of foot pressure and the legs exhibit much more low frequency activity than the torso and the head. It is noted that where the spectra generally double in magnitude for the EC test over the EO test, the middle sway spectrum increases relatively little.

Although many observations can be made from these plots, they are quite nondescript when higher frequencies are considered. The logarithms of the EO power spectra, as Figure 12 shows, offer a mediocre improvement. The frequency range of the graphs is extended to 16 Hz and it is evident that the spectral power is concentrated within the lower quarter of the frequency range.

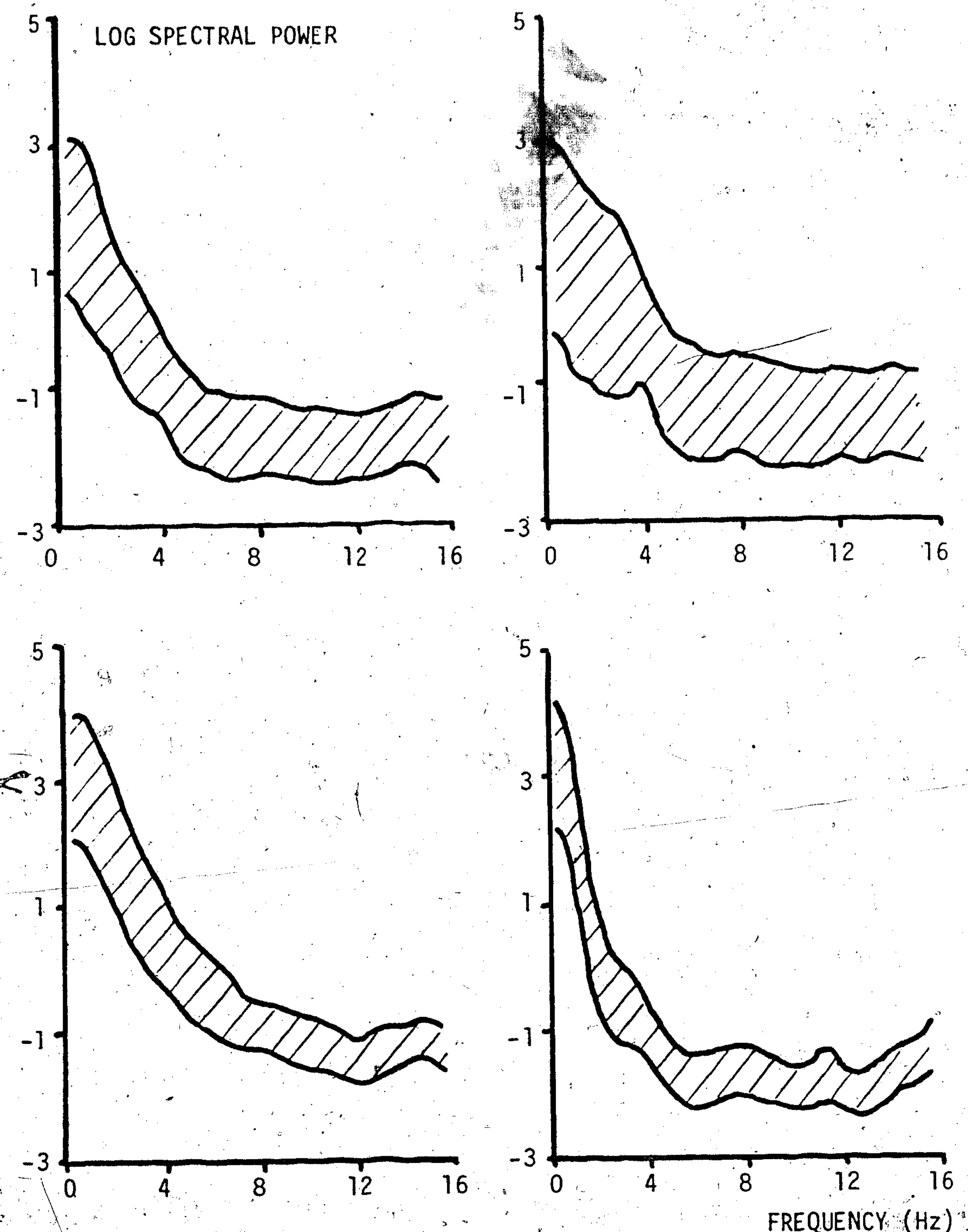

FIGURE 12. 68% confidence limits of averaged logarithms of power spectra of EO tests of 20 normal males: CFP - bottom left, LO sway - bottom right, MID sway - top left, HI sway - top right.

5.1.2 Coherence

The coherences (as defined in Appendix C) between the sway components and the movement of the center of foot pressure for the EO and EC tests are displayed in Figure 13 overleaf. The shapes of the curves are more interesting than those of the power spectra, but there is little difference between the two conditions. Since a normalizing process is involved (ie. range limited from 0 to 1), the information contained by the relative magnitudes of the signals is ignored.

5.1.3 Transfer Functions

The averaged transfer functions which describe the relationships between the sway components and the movement of the center of foot pressure are depicted in the following sets of graphs. Each set of graphs is arranged vertically. The bottom graph corresponds with the transfer function of the low or leg sway component. The middle graph corresponds with the middle or torso sway component. The top graph corresponds with the high or head sway component.

5.1.3.1 Real and Imaginary Components

The averaged real and imaginary components of the transfer functions for the EO and EC data from 20 normal males are shown in the next two sets of graphs. The shaded areas of Figure 14 indicate the confidence limits (as defined by one standard deviation) of the real components.

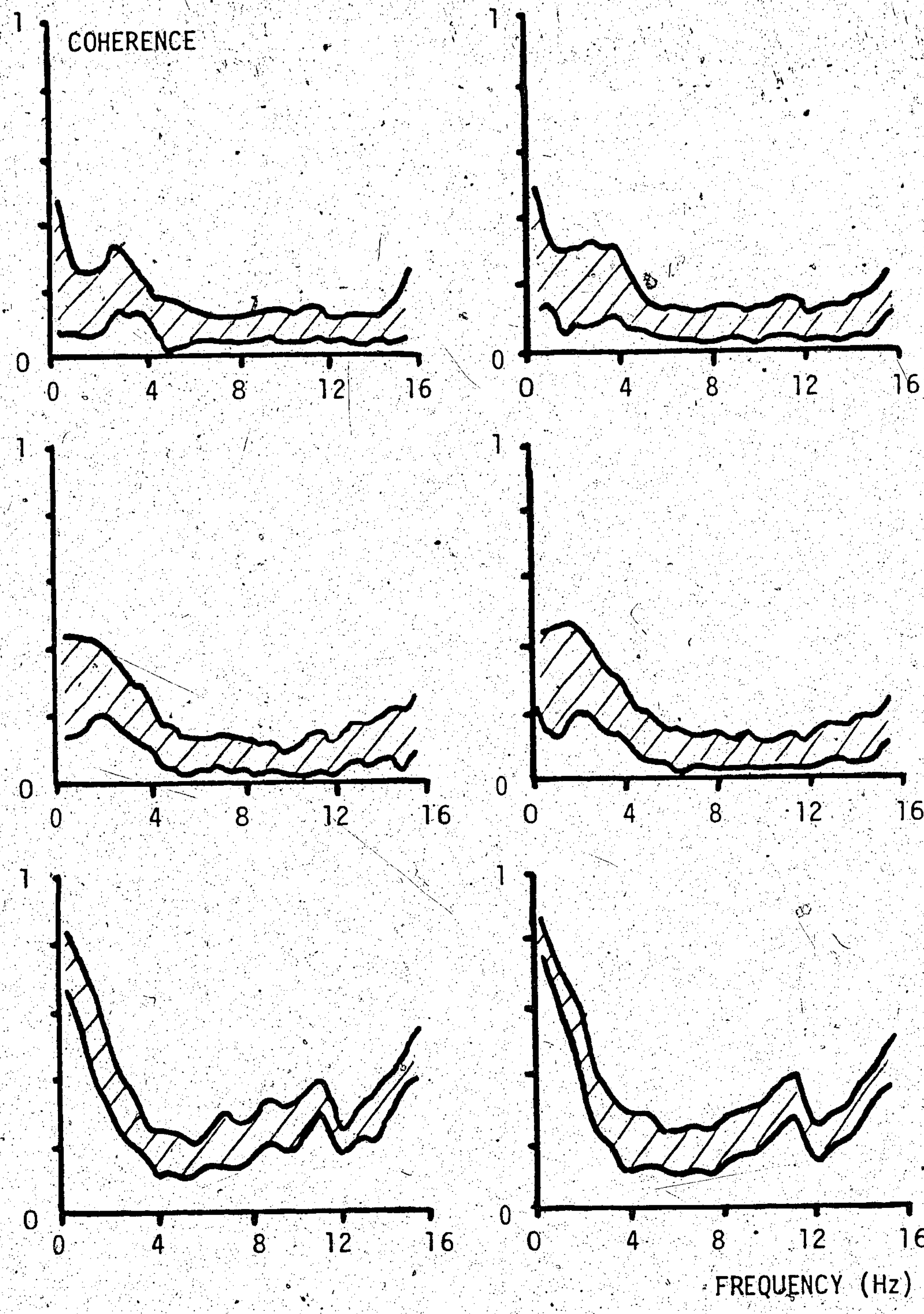

FIGURE 13. 68% confidence limits of averaged coherences of EO (left) and EC (right) tests of 20 normal males: LO sway and CFP - bottom, MID sway and CFP - middle, HI sway and CFP - top.

Figure 15 displays the imaginary part. The magnitudes are necessary only for comparative purposes between the graphs. The shapes of the curves are perhaps more important.

Some advantages of the transfer function analysis are obvious. A reduction in data is observed. The four power spectra and the three coherences (64 points each) are essentially combined to produce six transfer function components (64 points each). The confidence bands appear to be narrower, indicating a decrease in variance. The transfer function curves also appear to be more descriptive, perhaps yielding more information about the control systems in question.

The real component of the low transfer function for the EO test remains positive over the entire frequency range. The predominant feature is a peak around 1.5 Hz. A valley at 12.4 Hz is also consistent. Smaller peaks and valleys are observed at other frequencies in the curves of individuals, but averaging tends to obscure these characteristics.

The main feature of the real component of the middle sway is a valley at 2.4 Hz. Most of the features of the top graph are obscured by the confidence limits, but many peaks and valleys are evident in plots of individual data.

The real components of the EC transfer functions demonstrate a 50% increase in the amplitudes and a slight increase in the frequencies at which the peaks and valleys occur. The imaginary components also demonstrate this to

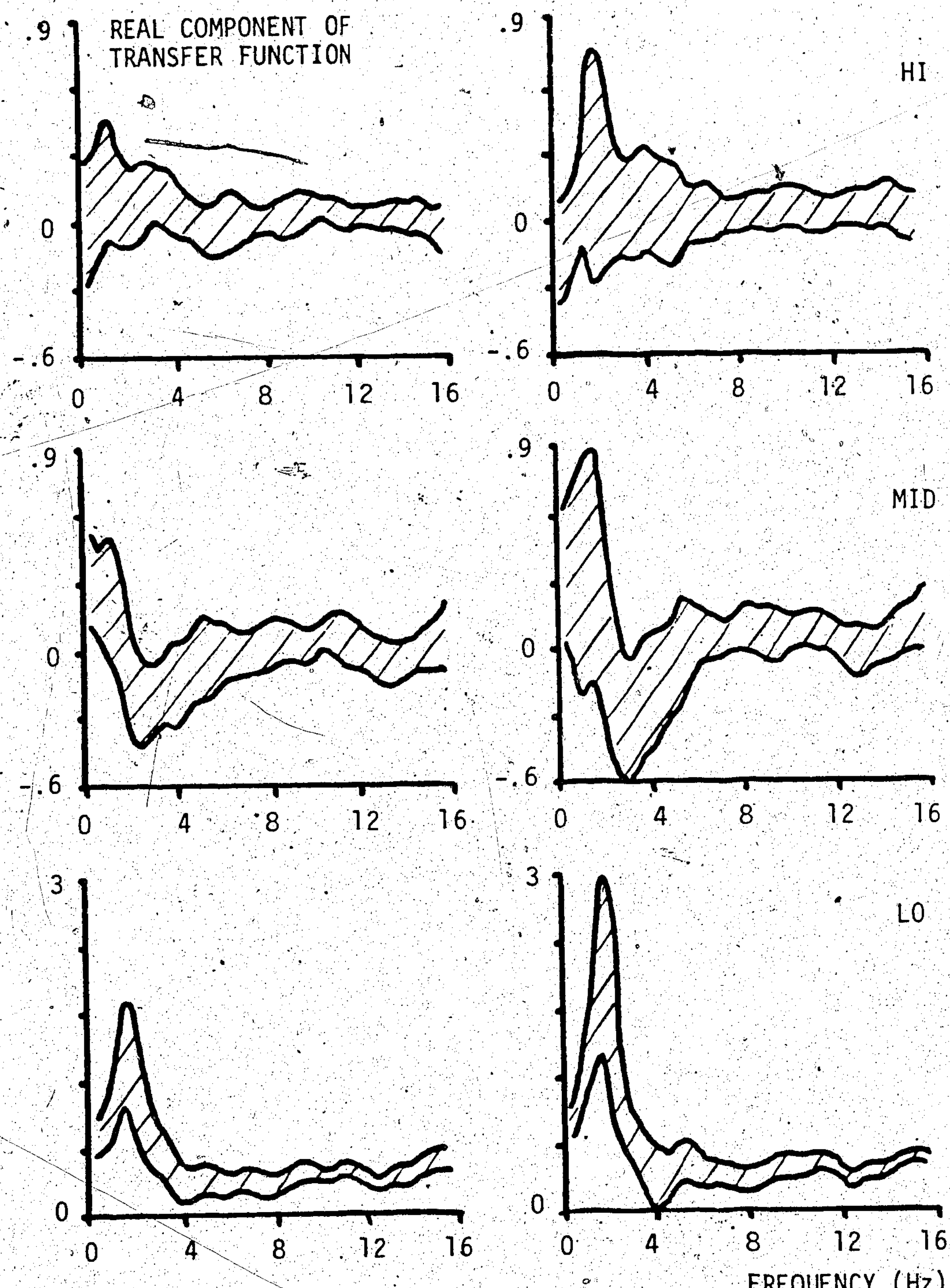

FIGURE 14. 68% confidence limits of averaged real components of transfer functions from EO (left) and EC (right) tests of 20 normal males.

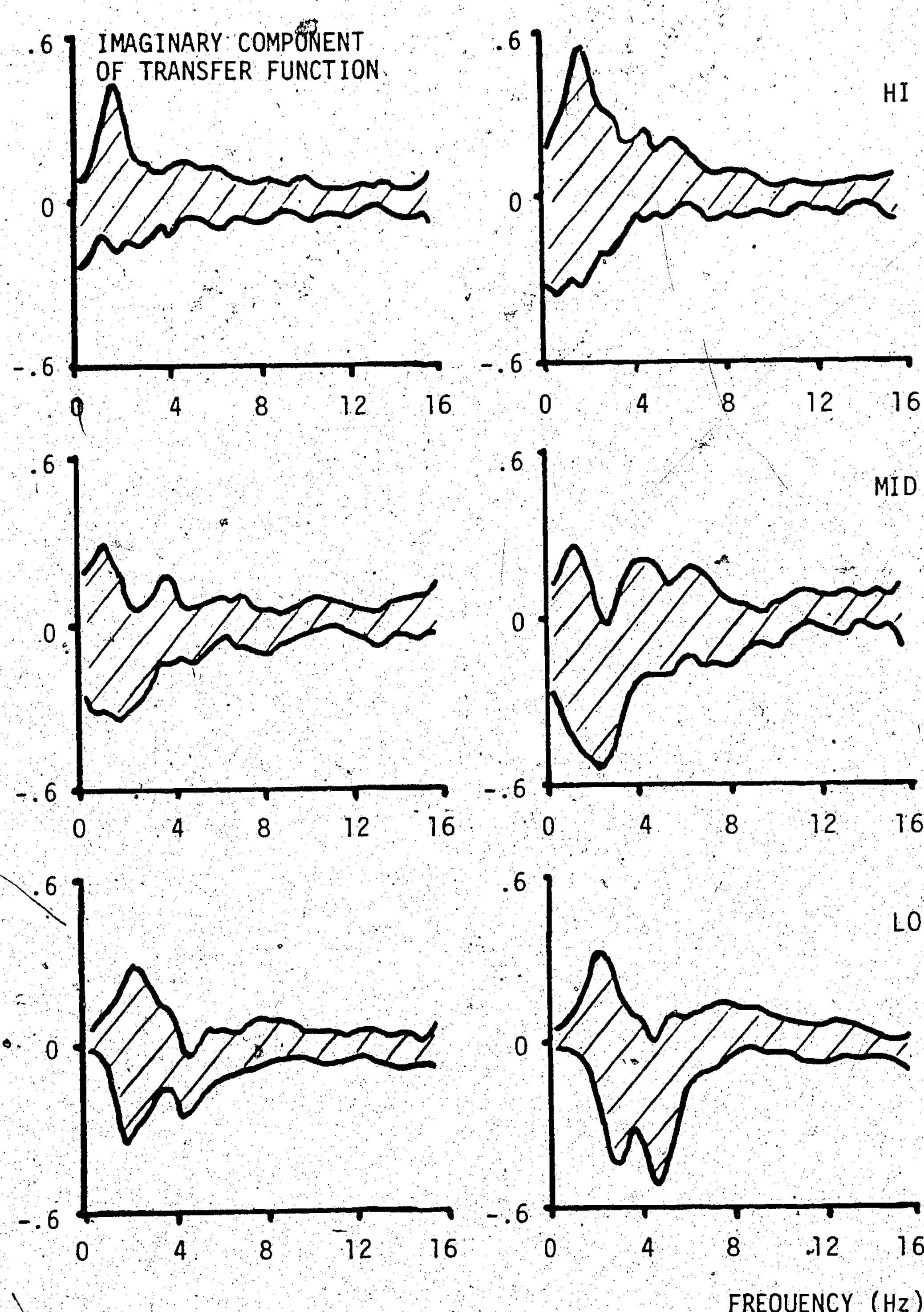

FIGURE 15. 68% confidence limits of averaged imaginary components of transfer functions from EO (left) and EC (right) tests of 20 normal males.

some degree. The general shapes of the curves do not appear to change appreciably.

Averaging the differences between the individual components of EC and EO data did produce a reduction in variance. Only once in the plots of differences between the real components of Figure 16 did the confidence limits of one standard deviation yield a consistent relationship. The real component of the EC low transfer function is greater than that for EO conditions between 1.25 and 2.5 Hz. A pair test (see Appendix K) indicates that this difference is significant at a level of probability of <0.05% (one sided test). Such high significance levels exist whenever both confidence limits are either negative or positive. Less obvious but highly significant (p<1%) differences occur in a number of other bands in all the real and imaginary components. For example, there appears to be a trend for the real component of the EO low transfer function to exceed that of the EC test at higher frequencies.

5.1.3.2 Gain and Phase Relationships

The real and imaginary transfer function components were pooled to produce the average gain and phase relationships. Figure 17 displays the gains for the EO and EC data. The magnitude of the low transfer function is seen to be 4 times greater than that of the middle or high gains. The EC gains are 1.5 times greater than the EO gains. The middle gain displays more prominent peaks than the low gain.

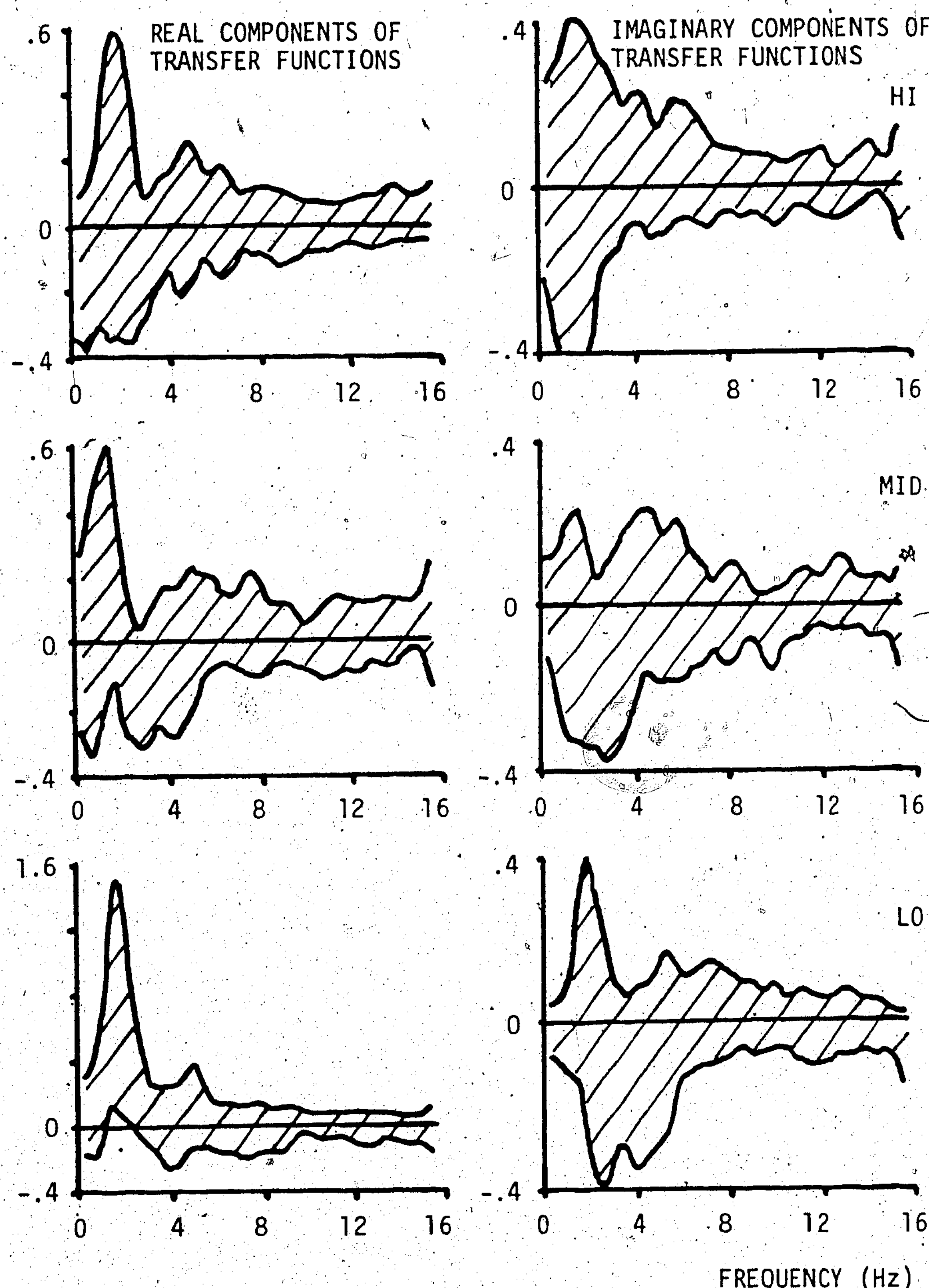

FIGURE 16. 68% confidence limits of averaged differences between real (left) and imaginary (right) components of transfer functions from EC and EO tests of 20 normal males.

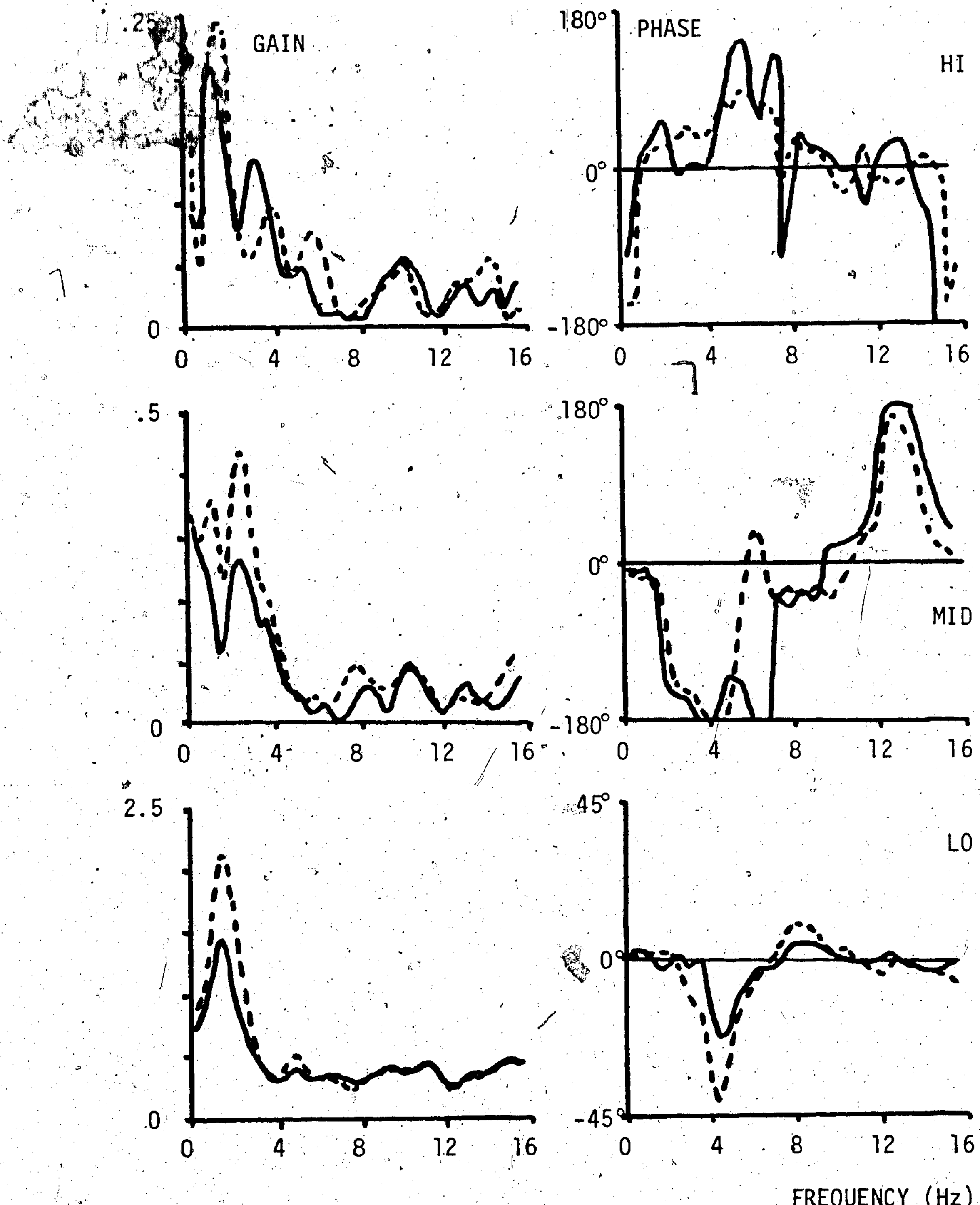

FIGURE 17. Average values of gain (left) and phase (right) components of transfer functions from EO (solid line) and EC (broken line) tests of 20 normal males.

The high gain appears to have even more of these resonant frequencies, which become more pronounced during the EC test.

The gain is averaged between 1.0 and 2.5 Hz for each subject in Table 1 for the EO and EC tests. Interpolation of the highest and two adjacent gain values produces the frequency of the peak gain. The subjects' vital statistics are also recorded. No correlation is apparent.

The values of the real and imaginary components often approach zero. The resultant phase values then change very rapidly. For different subjects these discontinuities occur at different frequencies, causing large variances. Only the low transfer function displayed phase confidence limits which yielded useful information. The average phase curves provided by the pooled data are preferable. Figure 17 also contains the resultant phase responses of the transfer functions for the EO and EC tests. The sudden shift in phase of the lower transfer function at 4.5 Hz is the most prominent feature.

There is an apparent phase lag between low body sway and center of foot pressure movement at this frequency for EO conditions. This lag increases for the EC test and occurs at 4.25 Hz. There is less distinction between EO and EC phases for the upper body.

The phase relationship between middle body sway and movement of the center of foot pressure rapidly drops to a phase lag of 180° at 4 Hz from about 0° at very low

SUBJECT	AGE (yr)	HEIGHT (cm)	WEIGHT (kg)	EO LO GAIN Average	EO LO GAIN Peak Freq	EC LO GAIN Average	EC LO GAIN Peak Freq
1 NB	29	178	67	1.34	1.83	2.59	2.21
2 BF	22	173	67	0.88	1.47	1.53	1.72
3 LF	27	171	61	0.87	1.31	0.98	1.65
4 DK	22	192	88	1.27	1.51	1.83	1.93
5 WR	24	189	86	1.34	1.42	1.26	1.51
6 BH	25	190	77	1.62	1.70	2.30	1.91
7 PC	37	172	70	0.70	1.35	2.04	2.03
8 GG	22	177	75	1.10	1.53	1.05	1.74
9 DW	39	193	101	1.19	1.87	1.09	1.67
10 LT	21	189	73	0.92	2.10	1.39	1.74
11 KW	23	191	86	1.25	1.59	1.58	1.74
12 TH	33	183	69	0.50	1.56	1.11	1.73
13 KC	44	180	88	1.32	1.73	2.42	1.85
14 PB	33	183	92	1.15	1.55	2.41	1.79
15 EY	37	174	71	2.23	1.88	2.63	2.02
16 RT	24	172	62	1.47	1.90	2.02	1.79
17 ZK	36	179	81	0.90	1.49	0.93	1.64
18 DO	32	180	79	1.54	1.57	2.17	1.80
19 RB	30	173	74	1.19	1.52	1.29	1.54
20 RS	41	173	77	1.38	1.46	1.82	1.80
AVERAGE	30	181	77	1.21	1.62	1.72	1.78
S.D.	7	8	10	0.37	0.21	0.57	0.18

TABLE 1. Vital statistics and average transfer function values (1.0 to 2.5 Hz) for the 20 normal male subjects.

frequencies. This phase lag decreases to 0° around 8 Hz and becomes a 180° phase lead at 12 Hz. The torso is generally observed to move in a direction opposite to the movement of the legs, and this change in phase is most interesting.

At the head, the phase relationship is marked by an initial phase lag at very low frequencies. The lag rapidly dwindles to a slight phase shift by 1 Hz.

5.1.4 Impulse Responses

The impulse responses of the transfer functions describe the responses of the three filters of the control system model to impulse excitations (ie. sudden, non-sustained movements of the body segments). The real and imaginary components are combined to produce an impulse response via an inverse Fourier transform. It has been found that 256 transform points produce adequate resolution and only 64 points of the output are necessary to convey information about each transfer function. A further reduction in data is possible when differences between impulse reponses are considered.

The averaged EO and EC impulse responses for 20 normal males are shown in Figure 18 overleaf. The shapes of the low and middle responses are well defined and distinctive. The fact that the two curves are not similar indicates that the tapering of the real and imaginary components of the transfer function has not adversely affected the impulse responses. The shape of the high response is concealed by

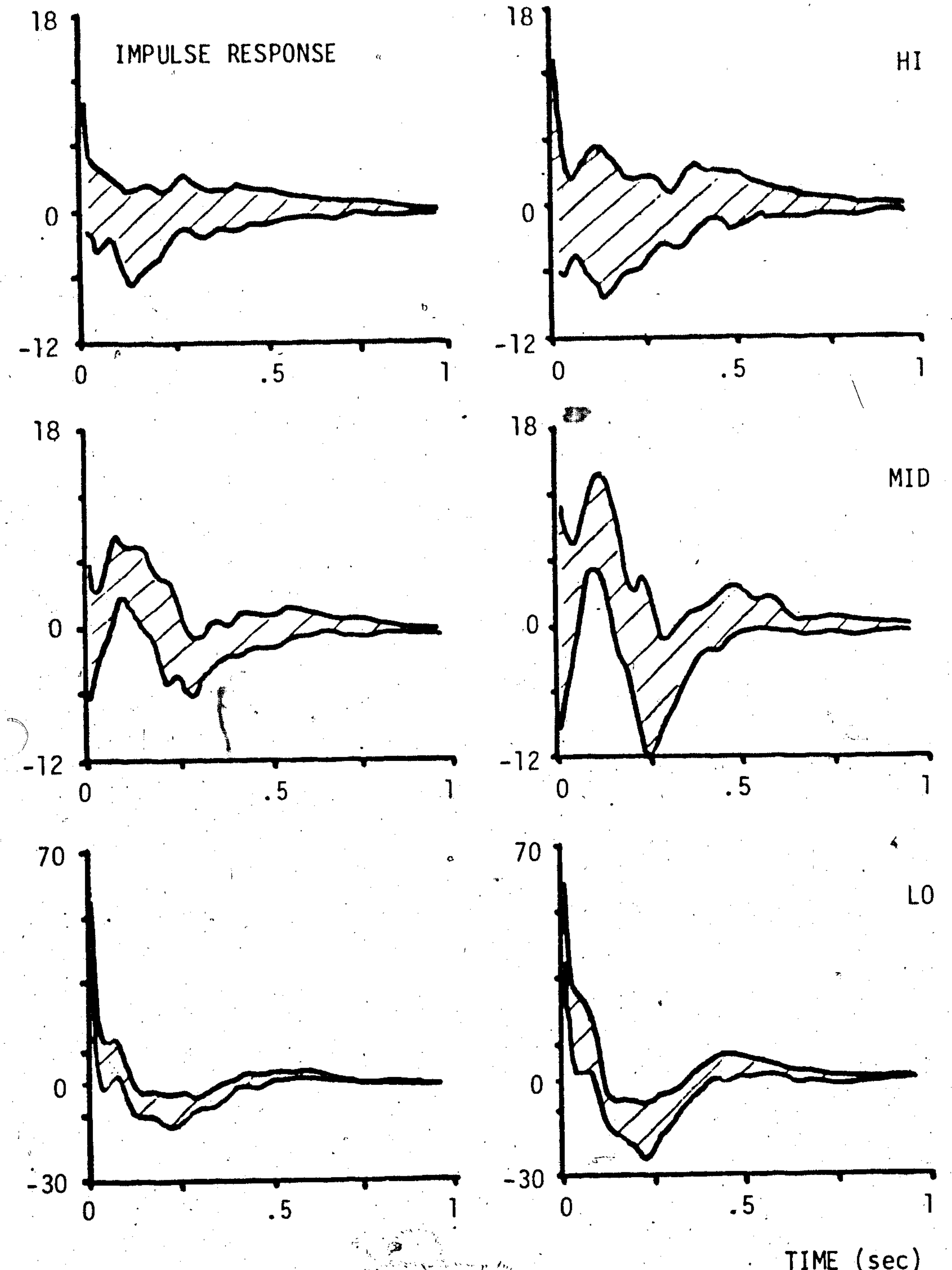

FIGURE 18. 68% confidence limits of averaged impulse responses of transfer functions from EO (left) and EC (right) tests of 20 normal males.

the confidence limits.

The averaged EC impulse responses display the same general form as the averaged EO impulse responses. The magnitudes of the low and middle EC curves are approximately 50% greater than the corresponding EO curves. This is further demonstrated by the averaged differences between the EC and EO responses, which Figure 19 displays. The difference between the low impulse responses shows three frequency intervals where the EC curves are consistently greater in magnitude than the EO curves. As mentioned before, these differences are significant at the $p<0.05\%$ level. The difference between the middle impulse responses displays two such intervals. Again, high variance obscures similar trends within the high impulse responses.

The average values of the EO impulse responses are plotted to different scales on the same coordinates in Figure 20 following. There is some resemblance of the curves to those of an underdamped second order differential equation. The low and high impulse responses might correspond to the solution. The middle impulse response might correspond to the first derivative of the solution. The EC impulse responses are similarly displayed.

Chow and Jacobson used a mathematical analysis of a two segment model to produce impulse responses very similar to those displayed in Figure 20. Viviani and Terzuolo also observed oscillatory characteristics in their studies of muscle response.

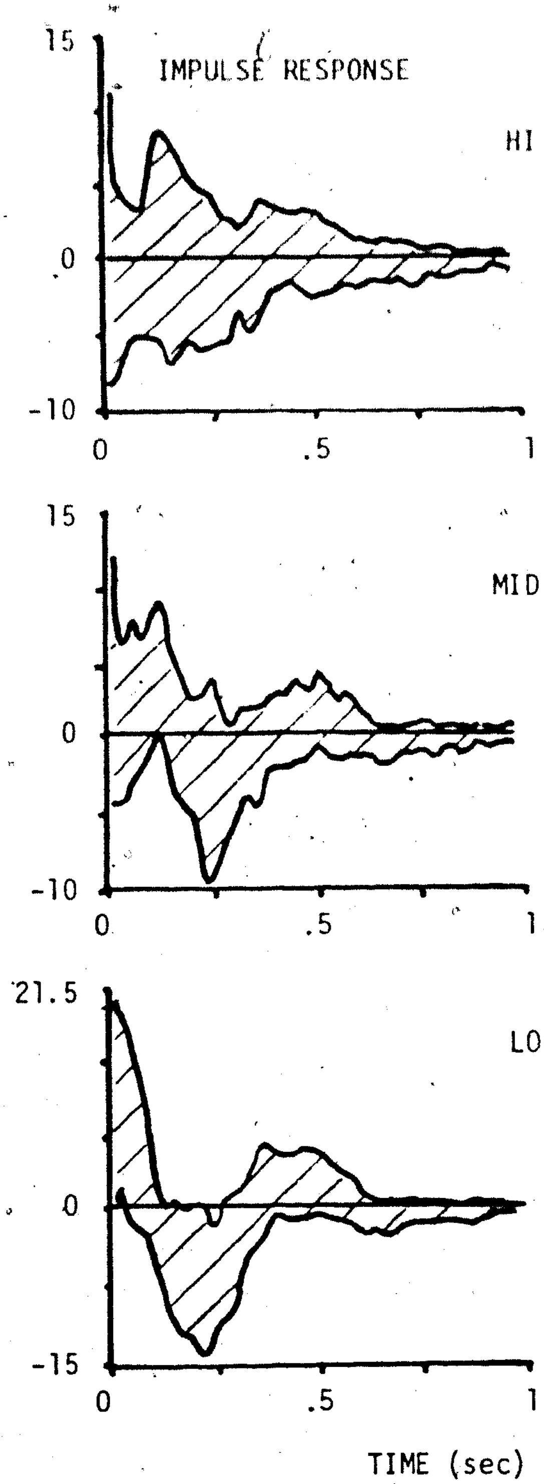

FIGURE 19. 68% confidence limits of averaged differences between impulse responses of transfer functions from EC and EO tests of 20 normal males.

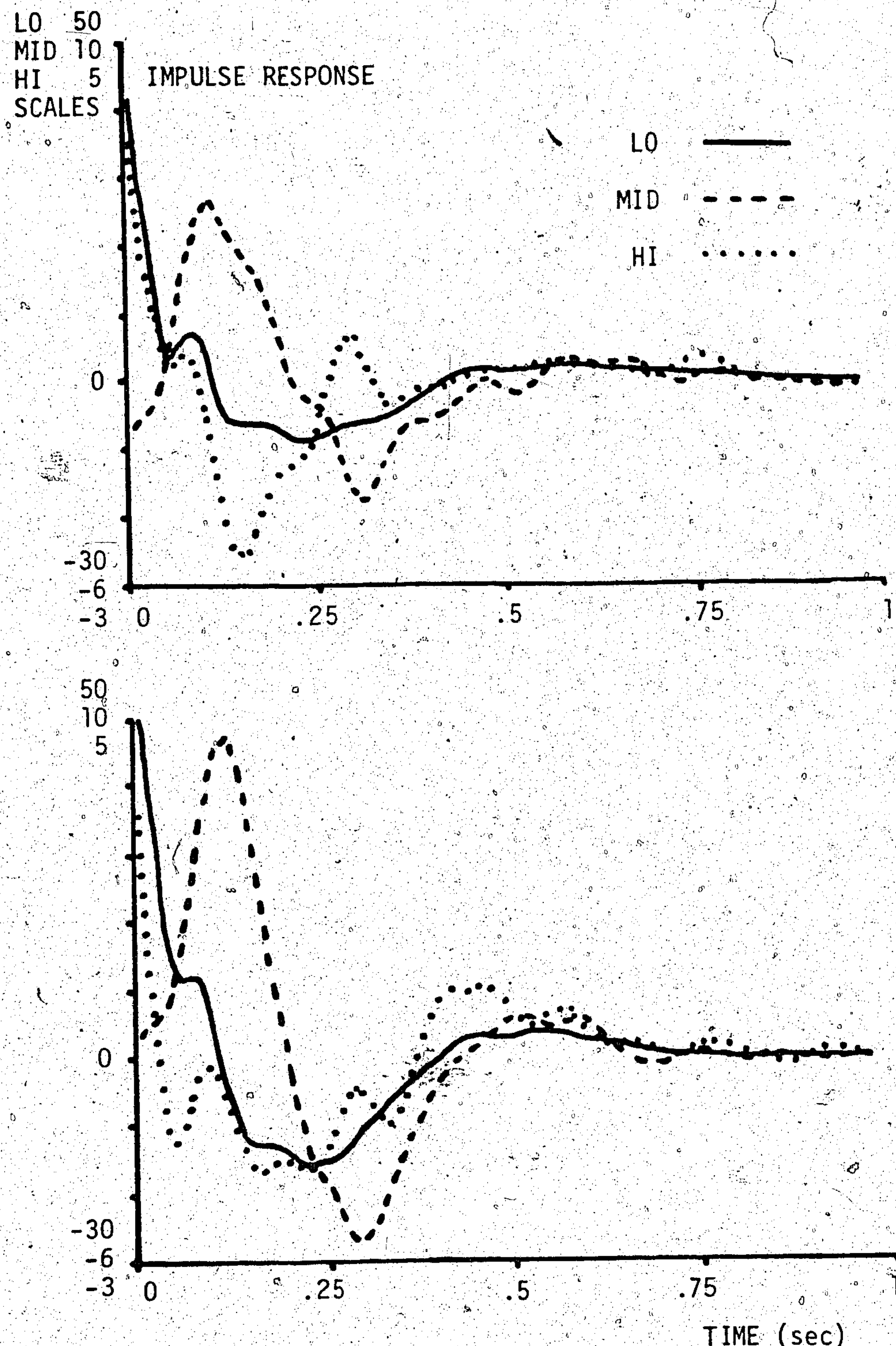

FIGURE 20. Average values of impulse responses from EO (top) and EC (bottom) tests of 20 normal males: LO - solid line, MID - broken line, HI - dotted line.

5.2 Tests of 7 Normal Females

Little difference is observed between the results from male and female populations. It is not possible to distinguish the sexes with any of the suggested comparisons between the results. Table 2 presents the gains averaged between 1.0 and 2.5 Hz for the EO and EC tests along with the subjects' vital statistics.

5.3 20 Tests of One Subject

The repeatability of the measurements and analysis is investigated by testing an individual normal male a number of times under the same conditions. The power spectra display a marked decrease in variance over the previous groups of tests. The variance of the coherence decreased slightly. The same general comparisons between the EO and EC results are observed.

5.3.1 Real and Imaginary Components

The overall shapes of the curves of the real and imaginary components of the transfer functions are similar to those of the previous groups of tests. With a decrease in variance, however, the definition of the curves improves, as shown in the graphs of Appendix L. It is interesting to note that while the average of the real and imaginary components of the repeated tests do fall within the confidence limits of the results for the population of

SUBJECT	AGE (yr)	HEIGHT (cm)	WEIGHT (kg)	EO LO GAIN Average	EO LO GAIN Peak Freq	EC LO GAIN Average	EC LO GAIN Peak Freq
1 CM	24	178	65	0.92	1.39	1.55	1.76
2 MC	31	178	69	1.66	1.65	1.68	1.70
3 JS	26	1.73	62	1.46	1.85	1.71	1.84
4 EH	32	168	58	1.42	1.50	1.39	1.61
5 JA	23	163	60	0.83	1.28	1.66	1.76
6 PC	26	164	56	1.33	1.66	1.62	1.74
7 CZ	24	161	60	1.68	1.81	1.41	1.60
AVERAGE	27	169	61	1.33	1.59	1.57	1.72
S.D.	3	7	4	0.34	0.21	0.13	0.09

TABLE 2. Vital statistics and average transfer function values (1.0 to 2.5 Hz) for the 7 normal female subjects.

normal males, the equivalent confidence limits of the repeated tests do not.

The confidence limits of the averaged real components for the EO and EC tests demonstrate that the features of the EC curves are more pronounced and occur at slightly higher frequencies than those of the EO curves.

The imaginary components for the EO and EC tests display similar features. The frequencies of these distinguishing features in the EO and EC results do not generally coincide.

The averaged differences between the EC and EO components reflect a large improvement in consistency. The data indicates a number of frequency bands where the EC components differ significantly ($p < .05\%$) from the EO counterparts.

5.3.2 Gain and Phase Relationships

The gain and phase relationships are again produced from the average real and imaginary components of the test series. The corresponding graph in Appendix L displays the EO and EC gain curves concurrently. The peaks of the EC gain are consistently of greater magnitude than those of EO gain. The number of peaks increases for the transfer functions related with the higher body segments.

The phase curves do not display a marked difference between EO and EC conditions. A larger negative phase shift at 4 Hz appears to be characteristic of the low EC transfer function.

5.3.3 Impulse Responses

The averaged impulse responses from twenty tests of one individual demonstrates a decrease in variance from the previous tests on many subjects. Again, the general shape of the impulse responses does not change when the subject closes his eyes.

The differences of the EC and EO impulse responses provide an improvement in the separation between the impulse responses observed in the two conditions. At all three measurement sites there are instances where highly significant differences ($p<0.05\%$) occur.

5.4 Alcohol Tests

The differences between the EC and EO tests for a group of 7 subjects is presented in the series of graphs in Appendix M. The two sets of transfer functions are derived from tests of the subjects while sober and then intoxicated. From the curves, it is apparent that the inebriation of the subjects had a greater affect upon the sway of the upper body. In particular, the frequencies at which significant differencies occur between the EC and EO conditions are not common to both sober and inebriated tests. Where a significant difference in the real components of the middle transfer function for the sober subjects is noted at 1.25

Hz, a similar difference is observed at 2.75 Hz for the inebriated subjects.

Plots of the confidence limits for the difference between the real components derived from the intoxicated and sober tests under EO and EC conditions and the corresponding imaginary component differences display many frequency intervals where consistent relationships exist.

The impulse responses are found to yield little information.

5.5 Neurological Patients

Although four patients were tested, the results for only three of them are presented in Appendix N. The excluded patient, who suffered from cerebral ataxia, had great difficulty in performing the tests. Little data was obtained. The impulse responses that were produced from the data varied greatly from the normal impulse responses. The remaining three patients exhibited results which differed more subtly from the normal population, which was expanded to include the normal females. These patients are identified by their afflictions in the following sections.

5.5.1 Parkinson's Disease

The female with Parkinson's Disease, although brought to the test in a wheel chair, demonstrated the least unsteadiness while performing the tests. At no time did she appear to be in danger of losing her balance, and a full test was completed. The data obtained was satisfactory.

The impulse responses for the EO and EC tests indicate that the lower body sway is not markedly affected. The magnitudes of the oscillations due to the middle body segment are increased considerably in both test conditions. The most interesting comparison is found at the head. With vision an abnormally large oscillation is observed. The EC impulse response appears to be normal, however.

5.5.2 Peripheral Neuropathy

This patient also had little trouble completing the tests. The data obtained was complete and satisfactory. The EO and EC impulse responses produced from the data are further testimony to his apparent stability.

The curves do not disclose any gross deviation from the normal curves. Some interesting observations may be made, however. The EO impulse response is greater in magnitude than the EC response for the lower body segment. The responses for the middle body segment appear to occur more quickly (ie. display a negative time delay) than the averaged normal responses. Again, the EO impulse response at the head has a greater magnitude than the EC response.

5.5.3 Cortico-spinal Tract Disease

The third patient had difficulty in maintaining his balance. Since this problem was not adequately anticipated during the calibrations, some of his movements produced signals which were too large for the A/D convertor. Also, his body often moved beyond the limits of the light beam of the sway detector. Therefore, the data that was obtained was less than ideal in quantity and quality. The impulse responses do not appear to reflect this shortcoming.

Again, the low impulse responses are close to being within the confidence limits of the normal population. It is the upper body segments which display impulse responses that indicate abnormality, especially during the EO test. Large and sustained oscillations are predominant at the head, and noted to a lesser degree at the chest.

5.6 Sagittal versus Lateral Sway

A study of the sways in the two orthogonal planes of movement of the erect body was conducted to reinforce, or disprove, the findings of previous researchers. In general, the former purpose was realized.

A 27 year old normal male was tested on 12 occasions, EO and EC, where the measurements alternated between sagittal and lateral sway. The subject did note that the immediacy of the apparatus during the measurement of lateral sway may have had some effect upon his performance.

Unfortunately, this problem could not easily be circumvented. The effects are not expected to be great. Appendix O contains the pertinent graphs.

5.6.1 Power Spectra and Coherence

The confidence limits of the averaged logs of the EO power spectra illustrate the larger power content of sagittal body movement. The power distributions over the frequency range are very similar. The approximate average ratios between the power spectra for the sagittal and lateral tests are: 30:1 for the center of foot pressure, 100:1 for the low sway, 3:1 for the mid sway, and 10:1 for the high sway. The variances are comparable.

Greater coherence is evident for sagittal body movement. The variance of the relationship for the low sway component is also notably smaller.

5.6.2 Transfer Function Components

The EO tests demonstrate that sagittal body movement produces results which yield real transfer function components of greater magnitude with less variance than the lateral measurements. For sagittal body movement the average coefficient of variance from 1 to 2 Hz for the real component of the low transfer function is 50% smaller than the lateral movement counterpart. The imaginary components are well-defined for both planes. Dissimilarities between the transfer functions of the two tests are more apparent with the real components.

The differences between the real components of the EC and EO tests for both sagittal and lateral body movement are generally of greater magnitude for the sagittal plane. Some interesting trends are noted in the lateral movement of the upper body.

The impulse responses exhibit greater magnitudes for sagittal body movement. The lateral impulse responses are also of markedly different shapes. This again demonstrates that the windows used at various stages in the signal processing do not seriously affect the shapes of the curves. The differences between the EC and EO impulse responses reveal that the sagittal curves clearly demonstrate greater reliability than the lateral curves.

6. DISCUSSION

6.1 Evaluation of Results

Transfer function analysis has been successfully used to study postural body movement. Consistent relationships were established between the movements of each of the three segments of the model of the body and the movement of the center of foot pressure. It is proposed that these relationships or transfer functions are more reliable in assessing the postural integrity or normalcy of a subject than the mere measurements of body movement alone. It also appears that this treatment of the data reveals more information about the postural control system than previous studies.

The results support the measurement of sagittal sway over lateral sway as being more likely to indicate postural disturbances. The transfer functions derived from sagittal body movement had greater magnitudes with less variability. The curves representing the sagittal transfer function components or impulse responses were generally better defined. The differences between the EC and EO results were generally more consistent in the sagittal plane. Although a new model should have been utilized for body movement in the lateral plane, it is believed that the results are valid for comparative purposes.

Results have verified that the data processing scheme did not influence the output enough to obscure the distinguishing traits. Other window functions are being considered to replace the triangular data window, however, since processing time is not a great factor.

Variance has apparently been reduced. Van Parys and Njiokiktjien measured three different parameters for 62 subjects during EO and EC tests. They found an increase in sagittal sway for the EC test over the EO test with 37% of the subjects. Lateral sway increased for 50% of the subjects. The mean length of the sway trajectory increased for 75% of the subjects. In the present series of tests of 27 subjects, the 2 Hz peak in the gain of the low transfer function was seen to increase in amplitude for 85% of the subjects. For the remaining 15%, an increase in the gain just above this peak frequency in the middle transfer function was observed in all cases.

Njiokiktjien and de Rijke measured the area enclosed by the trace of the movement of the center of foot pressure from a force platform. Their studies of 30 sec EO and EC tests with a group of 50 normal males yielded a median EO value of 81 mm^2. The median of the differences between EC and EO values was 44 mm^2. Using the provided values of the 95% confidence limits for the mean, the mean EO area was calculated to be 82 mm^2 with a standard deviation of 43 mm^2. The mean EC - EO value was 41 mm^2 with a standard deviation of 69 mm^2. The respective coefficients of variation were

0.53 and 1.67. In the present study, for a group of 20 males, the corresponding coefficients of variation were 0.38 and 0.86, where the real component of the low transfer function at 2 Hz was evaluated. If the peak values were used, the coefficients were 0.32 and 0.72. Apparently, this aspect of the relationship between foot pressure and body sway is significantly less variable in normal males than the measurement of foot pressure alone (Koles and Castelein).

The extraction of information from these relationships between movements of the body segments and the center of foot pressure relies on previous work to some extent. However, the potential of the proposed system appears to be much greater than most experiments where single measurements are considered.

The body obviously does not sway rigidly, as a one segment pendulum about the ankle in the sagittal plane. Observation of the sway measurements which have been corrected for lower body movement indicates that sway of the legs in one direction is accompanied by sway of the torso in the opposite direction. The head tends to move in the same direction as the legs.

The relative contributions of each of the sway components to the movement of the center of foot pressure are not proportional to the weights of the body segments involved. The sway of the legs is seen to have the greatest effect upon the center of foot pressure. Sway of the torso has less effect, but more than the sway of the head in the sagittal plane. This is as expected, since more strength, and hence, more muscular control are required for the lower joints.

The shapes of the gain functions reveal resonant peaks. These resonances become more numerous for the higher joints. Such resonances indicate that the center of foot pressure is more sensitive to the sway components within certain frequency bands. Rotation of the body segments around the higher joints involves lighter masses, and therefore, higher frequencies of movement are possible.

Referring to Equations 1 to 10, the shapes of the gain functions are quite unexpected. The one segment model produces a transfer function with no poles. Such a transfer function produces infinite gain at high frequencies and no resonant peaks. Intuitively, one expects this relationship to exist between the movements of a single body segment and the center of foot pressure.

With a multiple segment model the counterbalancing effects of the upper body segments may cancel the effects of movements of the lower body segments at some frequencies. Such zeroes are seen in the results. The lack of output above 4 Hz is due to little body movement at such frequencies. This is realized in ~~the~~ calculation of the transfer functions since the matrix method of solution introduces poles into the solutions of the equations.

Taking the center of mass of a 190 cm tall subject to be at 55% of his height, the single segment model produces a zero at approximately 0.6 Hz. This figure is not inconsistent with the recorded data. A gain minima and negative phase shift are observed at 4.0 Hz.

Pal'tsev and Aggashyan proposed a mathematical model encompassing servosystems that controlled stability. A "supporting" system maintained muscle tone while a "counteracting" system provided dynamic control in response to disturbances. With varying time delays and degrees of negative feedback this model produced resonant peaks also.

With regards to the 10 Hz tremor that previous researchers have noted, a peak at this frequency is observed in most of the gains and real components of the transfer functions for both sagittal and lateral body movements. This is as expected, and is most noticeable at the head where the tremor is comparable in amplitude to other muscular control.

The lack of vision does not change these general relationships. The magnitudes of the transfer function components and impulse responses increase for the EC tests, but the shapes of the curves tend not to change. Therefore, the plots of the differences between EC and EO results resemble the EO curves.

The increase in oscillatory behaviour as displayed by the impulse responses indicates a decrease in steadiness of the erect stance. This can be explained as a decrease in stability of the postural control system when visual information is eliminated. The gain relationships of the transfer functions show increases in the magnitudes of the resonant peaks for the EC tests. The foot pressure sensitivity appears to increase for different frequencies of motion of the body segments. These frequencies are slightly higher than those observed for the corresponding peaks in the EO data.

Alcohol imbibition is also seen to have deleterious effects upon stability. The resonant peaks generally become larger. However, the sensitivity of the balance control system to visual input is altered. The effects of the lack of vision decrease. For the lower body segment, this phenomenon is most noticeable at frequencies above 3 Hz. At the head, the difference between EC and EO data is reduced considerably, especially at lower frequencies (<5Hz). If vision is a major contributor to control of head movement, it appears that the foot pressure sensitivity to head movement has decreased. If alcoholic intoxication affects visual accuity, the processing of visual input, and/or the sensitivity of muscles to the afferentation stemming from such processing, these observations are as expected.

It is difficult to draw conclusions from the data obtained from the neurological patients. Larger populations are required to determine whether specific conditions exhibit consistent patterns. All three of the studied conditions were seen to reduce stability. The strength of the ankle musculature and inertia of the body probably conceal abnormality in the transfer function of the lower body segment. Disturbances were more noticeable at the torso and head. Oscillations and changes in response times of the impulse responses appeared to be the prime indicators of problems.

6.2 Directions

Large variances are still worrisome. Further efforts are necessary to reduce variance and to define more clearly the shapes of the transfer function components. The extreme variance observed with the phase relationships has been reported previously (Soechting) and may be beyond redemption. Alternative measurement schemes have been proposed. As suggested by Koles and Castelein, the horizontal forces of reaction at the feet could also be considered. Thomas and Whitney measured the "reactionary force" in the sagittal plane and likened it to a high pass filtered, 180° out-of-phase version of the sagittal movement of the center of foot pressure.

The force platform is subjected to horizontal as well as vertical forces. These horizontal forces have been measured in a few previous studies (Dichgans et al; Herman et al; Roberts and Stenhouse; and Thomas and Whitney). The two types of forces are interrelated, although the horizontal forces are about ten times smaller than the vertical forces (Spaepen, Vranken, and Willems).

As the analysis of Koozekenani et al illustrated, there is an effect on the center of foot pressure due to the horizontal force exerted on the platform. Equation 1 contains an element of torque due to this horizontal force acting about the ankle. This study initially assumed this torque to be negligible relative to the torque generated about the ankle by vertical forces.

E.V. Gurfinkel's mathematical analysis of the center of foot pressure and its relation to movement of the center of mass included the horizontal forces. He suggested that measurements in the horizontal plane indicated only those forces due to body movement. Most force platforms respond to movement of the center of foot pressure, which entails the position of the center of mass as well. The force platform of Thomas and Whitney was used to record only the forces due to muscular action. The support points of the platform were at the height of the ankle.

Massen et al designed a similar platform. The vertical forces were observed to be larger than with a standard force platform. The torque at the ankles was thought to be eliminated since the plane of support passed through the ankles. The torques generated at higher joints were presumed to have little effect.

The present force platform can be made free to move forwards and backwards. It is possible to therefore measure horizontal forces in the same plane as the sway measurements. Further development of the system will probably involve construction of a new force platform to measure both horizontal and vertical forces. Plans also include an ultrasonic system to detect sway. This should reduce the size and weight of the apparatus, increase the sway measurement range, eliminate the problem of component short-term aging (as with light sources), and reduce the sensitivity of the system to building vibration. Another improvement might involve a method of calibrating the sway detectors and the force platform together. Hence, a known sway would produce a predetermined displacement of the center of foot pressure.

6.3 Conclusions

The proposed method of measuring postural stability has been shown to reduce the high variance associated with previous studies. Some effectiveness in distinguishing various conditions (i.e. EO and EC, sober and inebriated, normal and abnormal due to disease) has been demonstrated. With the recommended changes and further research with larger test groups, these encouraging results will improve. It is hopeful that the measurement of postural stability will become a valuable clinical diagnostic tool.

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APPENDIX A

Power Spectra

Analog signals are often described by the frequency components which make up the continuous waveform. The Fourier transform of the signal provides this information. The inverse Fourier transform can be used to produce the analog signal from its frequency spectrum. For a signal, $x(t)$, where $-\infty < t < \infty$, and its Fourier transform, $X(f)$, where $-\infty < f < \infty$:

$$X(f) = \int_{-\infty}^{\infty} x(t) \exp(-j2\pi ft) \, dt \tag{A.1}$$

$$x(t) = \int_{-\infty}^{\infty} X(f) \exp(j2\pi ft) \, df \tag{A.2}$$

where $j = \sqrt{-1}$ and $\exp(p)$ is the exponential function.

The action of Equation A.1 may be likened to an infinitely narrow, variable bandpass filter which sweeps the frequency range and indicates the power associated with the signal at each frequency (Bergland). Similarly. Equation A.2 reconstructs the signal from its frequency components. However, since only finite amounts of signal are available for analysis at any one time, the exact Fourier transform cannot be evaluated.

An analog signal may be adequately represented by evenly spaced (in time), discrete samples of that signal if the sampling rate is at least twice the highest frequency observed within the signal. This is the Nyquist criterion. If there is an appreciable amount of frequency content above this Nyquist frequency, the phenomenon of aliasing produces spurious frequency components within the range of interest.

Corresponding relationships exist between the time domain and the frequency domain for digital signals as with analog signals. For a digital signal, x(k), where k = 0,1,2,...,N, and its discrete Fourier transform (DFT), X(i), where i = 0,1,2,...,N:

$$X(i) = \frac{1}{N}\sum_{k=0}^{N-1} x(k)\ \exp(-j2\pi ik/N) \tag{A.3}$$

$$x(k) = \sum_{k=0}^{N-1} X(i)\ \exp(j2\pi ik/N) \tag{A.4}$$

The DFT is seen to be a periodic representation of the signal in the frequency domain. This periodicity is due to the exponential term:

$$w(p) = \exp\{(-j2\pi/N)p\} \tag{A.5}$$

The period is equal to the sampling frequency. The exponential term also creates symmetry. The real component of the DFT of a real sampled signal is symmetrical about a frequency equivalent to half the sampling frequency. The imaginary component is antisymmetrical about this frequency.

Equations A.3 and A.4 involve much computation for large numbers of data values (N^2 complex additions and multiplications for N data values). The fast Fourier transform (FFT) is a development of the DFT which markedly reduces the number of calculations ($N \log_2 N$). The Cooley-Tukey algorithm utilizes recursive equations. The result is possibly more accurate than that of the DFT since the fewer computations involve less round-off error (Monro).

Where the DFT uses all of the input data terms for calculating each Fourier coefficient of frequency, the FFT factors the number of data values and transforms these shorter sequences of data. Redundancies involving the exponential term are eliminated. By writing intermediate results over unneeded values, the amount of storage required is reduced.

Consider the Cooley-Tukey algorithm for the FFT:

$$X(i) = \frac{1}{N} \sum_{k=0}^{N-1} x(k)\, w(-ik) \tag{A.6}$$

where $i,k = 0,1,2,\ldots,N-1$. When N is a power of 2, i and k may be represented by binary numbers. For example, for N = 8, then $i,k = 0,1,2,3,4,5,6,7$ and:

$$\begin{aligned} i &= 4i_2 + 2i_1 + i_0 \\ k &= 4k_2 + 2k_1 + k_0 \end{aligned} \tag{A.7}$$

where i_2, i_1, i_0, k_2, k_1, and k_0 are either 0 or 1. Then:

$$X(i_2, i_1, i_0) = \sum_{k_0=0}^{1} \sum_{k_1=0}^{1} \sum_{k_2=0}^{1} x(k_2, k_1, k_0)$$

$$w\{-(4i_2 + 2i_1 + i_0)(4k_2 + 2k_1 + k_0)\}$$

(A.8)

Noting that $w(8) = 1$, this equation simplifies to:

$$X(i_2, i_1, i_0) = \sum_{k_0=0}^{1} \sum_{k_1=0}^{1} \sum_{k_2=0}^{1} x(k_2, k_1, k_0)\ w(-4i_0k_2)$$

$$w\{-(4i_1 + 2i_0)k_1\}\ w\{-(4i_2 + 2i_1 + i_0)k_0\}$$

(A.9)

The indicated groupings determine the order in which the summations should occur, such that the exponential terms will contain only variables involved in that particular summation:

$$A_1(i_0, k_1, k_0) = \sum_{k_2=0}^{1} x(k_2, k_1, k_0)\ w(-4i_0k_2)$$

$$A_2(i_0, i_1, k_0) = \sum_{k_1=0}^{1} A_1(i_0, k_1, k_0)\ w\{-(4i_2 + 2i_0)k_1\}$$

$$A_3(i_0, i_1, i_2) = \sum_{k_0=0}^{1} A_2(i_0, i_1, k_0)\ w\{-(4i_2 + 2i_1 + i_0)k_0\}$$

$$= X(i_2, i_1, i_0)$$

(A.10)

Each summation entails eight operations for each of the two values of the summation variable. Note that the input data samples are used only in the first summation. Successive summations may thus overwrite previous data.

Further savings in computation are obtained by symmetry: $w(0) = -w(4)$, $w(1) = -w(5)$, etc.

The reordering of the DFT operations by the FFT scrambles the frequency domain representation of the sampled signal. The correct sequence is restored simply. Each term is represented by a binary number by position. A reversal of the order of the binary digits indicates the correct position of the terms within the output sequence.

Monro presents a more flexible version of the FFT. Whereas the Cooley-Tukey algorithm factors the sampled data points with binary coefficients, Monro's version uses factors (or radices) that may be chosen to handle data lengths other than powers of two. Letting $N = ABC$, any data point may be represented by the factors i_a, i_b, and i_c:

$$i = ABi_c + Ai_b + i_a \tag{A.11}$$

where $i_a = 0, 1, \ldots, A-1$,

$i_b = 0, 1, \ldots, B-1$, and

$i_c = 0, 1, \ldots, C-1$.

The output sequence may be similarly partitioned:

$$k = BCk_c + Ck_b + k_a \tag{A.12}$$

where $k_a = 0, 1, \ldots, A-1$,

$k_b = 0, 1, \ldots, B-1$, and

$k_c = 0, 1, \ldots, C-1$.

Equation A.3 may then be written:

$$X(i_c, i_b, i_a) = \sum_{k_a=0}^{A-1} \sum_{k_b=0}^{B-1} \sum_{k_c=0}^{C-1} x(k_c, k_b, k_a)$$

$$\exp\{-j2\pi(ABi_c + Ai_b + i_a)(BCk_a + Ck_b + k_c)/N\} \tag{A.13}$$

To simplify further manipulation of the equation, a function is defined: $e(p) = \exp(-j2\pi p)$, and:

$$X(i_c, i_b, i_a) = \sum_{k_a=0}^{A-1} e(k_a i_a/A)\ e(k_a i_b/AB)$$

$$\sum_{k_b=0}^{B-1} e(k_b i_b/B)\ e\{i_c(k_a + k_b A)/ABC\} \sum_{k_c=0}^{C-1} e(k_c i_c/C) x(k_c, k_b, k_a) \tag{A.14}$$

Monro refers to equation A as the Sande version. This particular arrangement of terms allows for a more efficient computer implementation than other versions.

As in the Cooley-Tukey algorithm, the inner summations have been simplified by removing variables from the pertinent terms. The resulting expression may be viewed as a series of transforms over subsets of data. The term immediately preceding each summation is referred to as a "twiddle factor". Each series thus differs from being exactly a DFT by this value.

The inner summation operates upon the input data samples. The middle summation is then applied. An advantage of the Sande version is the simple "twiddle factor" associated with this summation. Assuming that the first summation over C values has been completed, with the associated "twiddle factor", the summation over B values are then evaluated with the second "twiddle factor". Choosing a new radix B', the other radices are redefined according to:

$$A' = A/B'$$
$$C' = BC$$
$$A'B'C' = N \tag{A.15}$$

With such a change in factors, it is possible to calculate the entire transform within the inner summation. Starting with factor $C = 1$, Equation A.14 recursively sums over B values until factor $A = 1$. If $N = B^k$, the total number of complex operations will be kN or $N \log_B N$.

The choice of a value for B is facilitated by the periodic nature of the "twiddle factor". A value of 2 or 4 provides changes in the exponent of π or $\pi/2$. Therefore, the complex multiplication only involves "twiddle factors" that are either real or imaginary.

In the actual implementation of his FFT, Monro points out further details which facilitate the computation. He also notes some pitfalls.

APPENDIX B

Spectral Estimation

The Fourier transform of a sampled signal may be used to produce an estimate of the power spectrum of that signal. Where $X(f)$ is the transform of $x(t)$, the spectral estimate is given:

$$S'(f) = \frac{1}{T} |X(f)^2| = \frac{1}{T} X(f)\ X*(f) \tag{B.1}$$

The complex conjugate of $X(f)$ is denoted as $X*(f)$. The sample length is T.

The accuracy of the spectral estimate may be improved by evaluating the average of the spectral estimates or "periodograms" for a number of independent subsets of data. This smoothed spectral estimate, using subsets of data of length, L, is given:

$$S''(f) = \frac{L}{T} \sum_{i=1}^{T/L} S'_i(f) \tag{B.2}$$

Two conflicting criteria are encountered at this point. A large number of data points per transform is desirable to reduce variance and to acquire adequate frequency resolution. The width of the central peak of the data window determines the minimum pulse width which can be detected. This is dependent on the number of data points per transform. However, many subsets of data samples will also reduce the variance of the spectral estimate via averaging. A suitable compromise is achieved by overlapping the subsets of data samples. The decrease in variance is not as great as with a similar number of non-overlapping subsets because the independence between subsets is violated.

An optimal overlap of adjacent data samples depends somewhat on the shape of the data window. Data points of equal importance are weighted unequally. A reasonable overlap for many common data windows is one half the length of a subset of data (Welch). However, if the computing time is available, a greater overlap will further reduce the variance. In this case, a three quarter length overlap is suggested.

Jenkins and Watts observed that such data subdivision is equivalent to smoothing the sample spectrum with the spectral window:

$$W(f) = L\{\sin(\pi fL)/\pi fL\}^2 \tag{B.3}$$

The corresponding lag window in the time domain is triangular. Thus, to further reduce the variance of the inverse transforms, a triangular smoothing function may be applied to the Fourier spectra.

The resolution of the power spectrum may be improved by artificially increasing the amount of data. This is accomplished by padding the data with zeroes.

APPENDIX C

Transfer Function Analysis

Assuming a linear relationship exists between the input and the output power spectra of a system, the transfer function describing this relationship may easily be found:

$$|F_i(f)| = \sqrt{S_{oo}(f)/S_{ii}(f)} \tag{C.1}$$

$$F_i(f) = S_{io}(f)/S_{ii}(f) \tag{C.2}$$

where $F_i(f)$ = transfer function (of frequency, f),

$S_{ii}(f)$ = auto power spectrum of input,

$S_{oo}(f)$ = auto power spectrum of output, and

$S_{io}(f)$ = cross power spectrum of input and output.

A measure of the degree of linearity between two signals at each frequency is coherence. Conversely, a lack of coherence indicates a nonlinear relationship and/or the presence of contaminating noise. Coherence may be expressed as the ratio of the above two estimates of the frequency response of the system. Equation C.2 involves the cross power spectrum of the input and output signals and takes into account only those output components linearly related to the input. Equation C.1 contains the auto power spectrum of the output signal, which will contain a bias error as well if the output is affected by sources other than the input (Clynes and Milsum). Since the magnitude of the cross power term will be less than or equal to the auto power term, the coherence will be less than or equal to unity:

$$\mathrm{Coh}_{io}(f) = \frac{|S_{io}(f)|}{\sqrt{S_{ii}(f)\ S_{oo}(f)}} \leq 1 \tag{C.3}$$

As the degree of linearity increases and the amount of contaminating noise decreases, the coherence approaches unity.

When more than one input to the system is considered, the output may be linearly related by varying degrees to each input. Since Equation C.2 involves phase as well as magnitude, it is used in further analysis. The definition of cross correlation provides the required relationships. For a system with three inputs and one output the equations are given by Marmarelis and Marmarelis as:

$$\begin{aligned}
S_{12}(f) &= F_2(f)S_{22}(f) + F_3(f)S_{23}(f) + F_4(f)S_{24}(f)\\
S_{13}(f) &= F_2(f)S_{23}(f) + F_3(f)S_{33}(f) + F_4(f)S_{34}(f)\\
S_{14}(f) &= F_2(f)S_{24}(f) + F_3(f)S_{34}(f) + F_4(f)S_{44}(f)
\end{aligned} \tag{C.4}$$

where 1, 2, 3, and 4 denote the output and the three inputs, respectively.

The minimization theory of Goodman uses these relationships and Cramer's rule to solve for the transfer functions at each frequency. The three equations in the three unknown variables, the transfer function coefficients, may be solved to minimize the mean-squared error between the output and the weighted sum of the three inputs.

APPENDIX D

Joint Measurement Error

The heights of the joints of a subject were measured with a tape measure. Although this scheme is not exceedingly accurate, a study of the resultant error reveals that it is adequate. A typical set of test data is submitted to computer-processing a number of times using 27 different sets of joint parameters in this study. The heights are varied by 13 mm. The variances due to this source of error are illustrated in Figure D.1 for the real components and in Figure D.2 for the impulse responses of the transfer functions. These variances are compared to those observed for 20 tests of the same subject using a single set of joint heights.

The worst case results from varying the ankle height. This adversely affects the upper two sway components. However, the lateral malleolus presents the easiest point at which to measure a joint height.

The upper joint, at the neck, is difficult to observe, but this measurement does not affect the calculation of the sways of the lower body segments. Some of the considerable variance observed in the analysis at the head is undoubtedly due to the inaccuracy of this joint height.

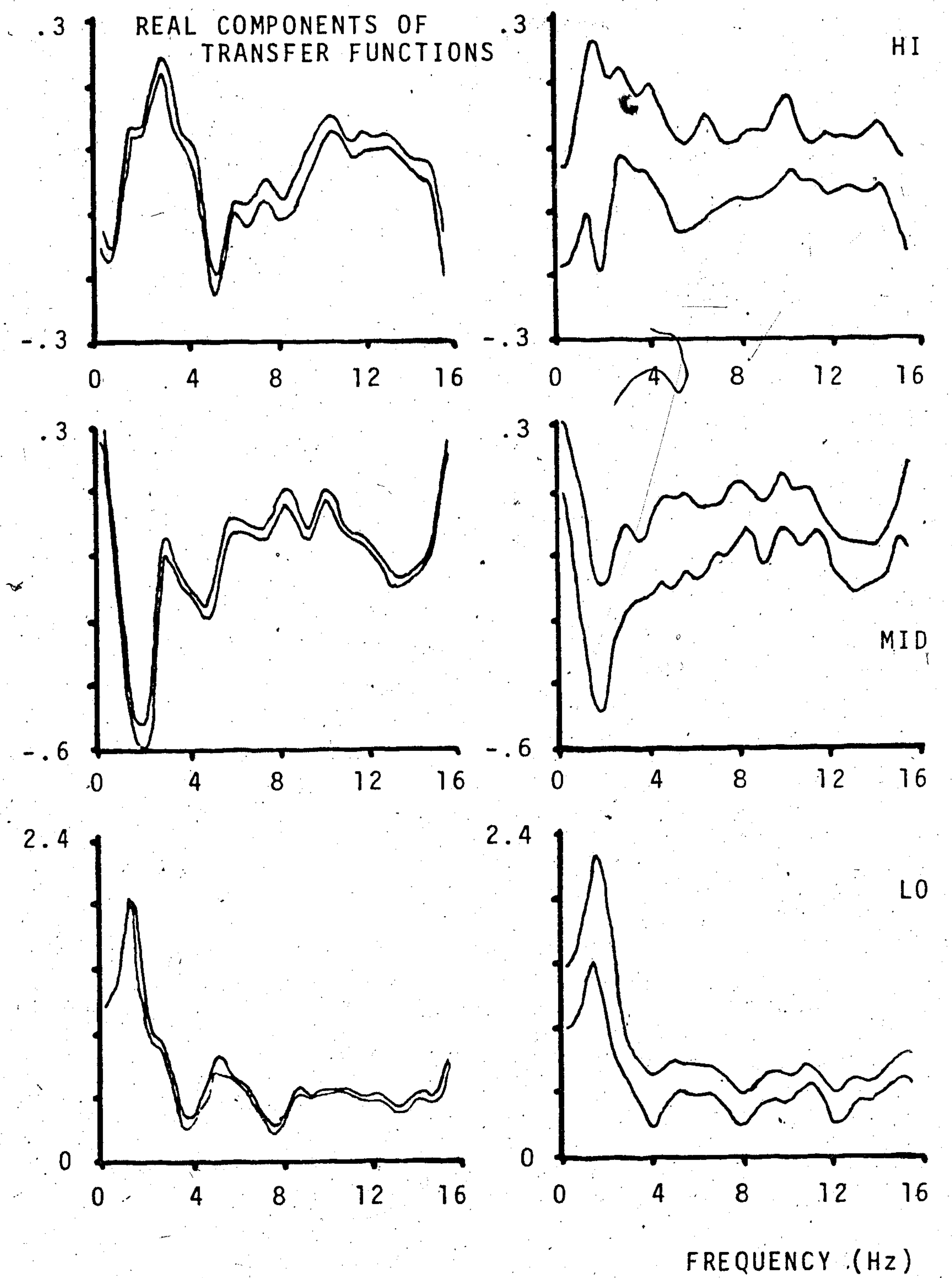


FIGURE D.1. 68% confidence limits of averaged real components of transfer functions for one test with 20 various joint heights (left) and for 20 tests with one set of joint heights (right).

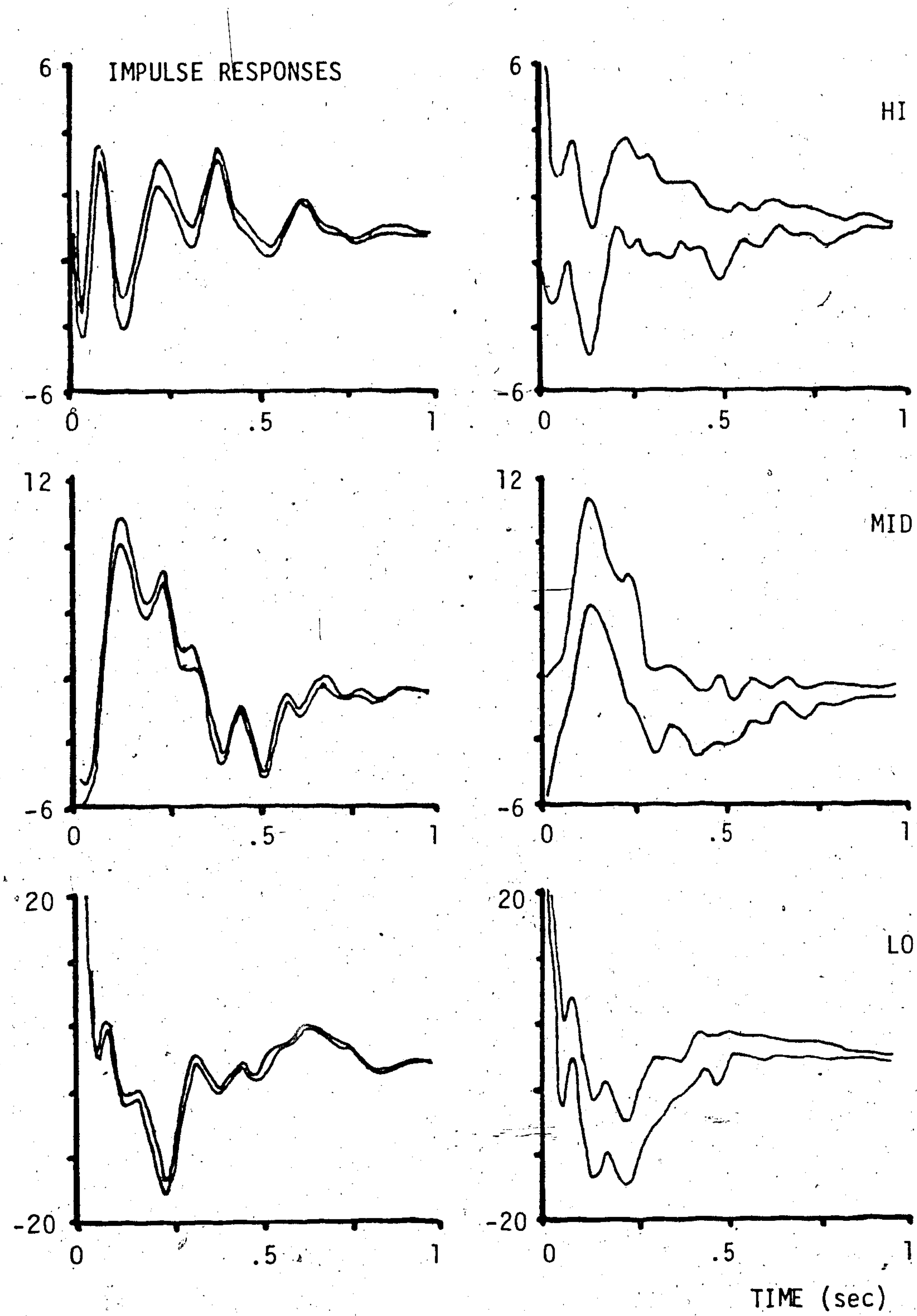


FIGURE D.2. 68% confidence limits of averaged impulse responses of transfer functions for one test with 20 sets of various joint heights (left) and 20 tests of the same subject using one set of joint heights.

APPENDIX E

Force Platform Circuitry

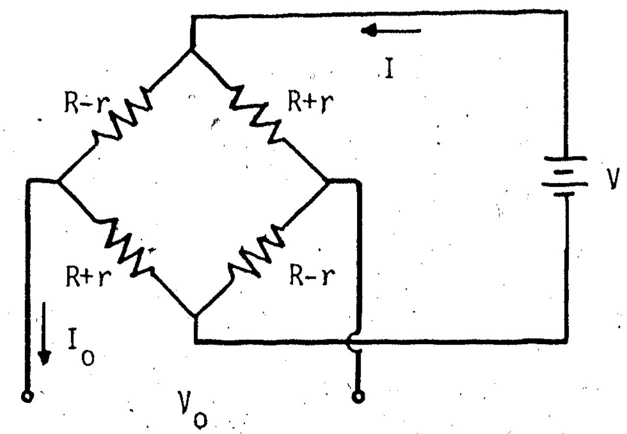
The basic strain gauge circuit consists of a Wheatstone bridge with varying resistances in each of the branches. The strain gauges are arranged so that while one pair is deformed and increases in resistance, the resistances of the other pair decrease with deformation. In this case, such a relationship exists with one pair of strain gauges fastened to the top surface of the supporting cantilever and the other pair fastened to the bottom surface. The circuit arrangement is depicted in Figure E.1a. The nondeformed resistance is denoted by R, with a deformation producing a change in resistance of r. A DC voltage, V, is applied across two opposite corners of the bridge. The output voltage, V_o, indicates the amount of deformation.

If a large load resistance is used, the output current, I_o, may be assumed to be negligible, compared to the input current, I:

$$I \simeq V/\{(R+r + R-r)//(R-r + R+r)\} = V/R \tag{E.1}$$

The output voltage is then given:

$$V_o \simeq \frac{R+r - R-r}{R+r + R-r} V = V\, r/R \tag{E.2}$$



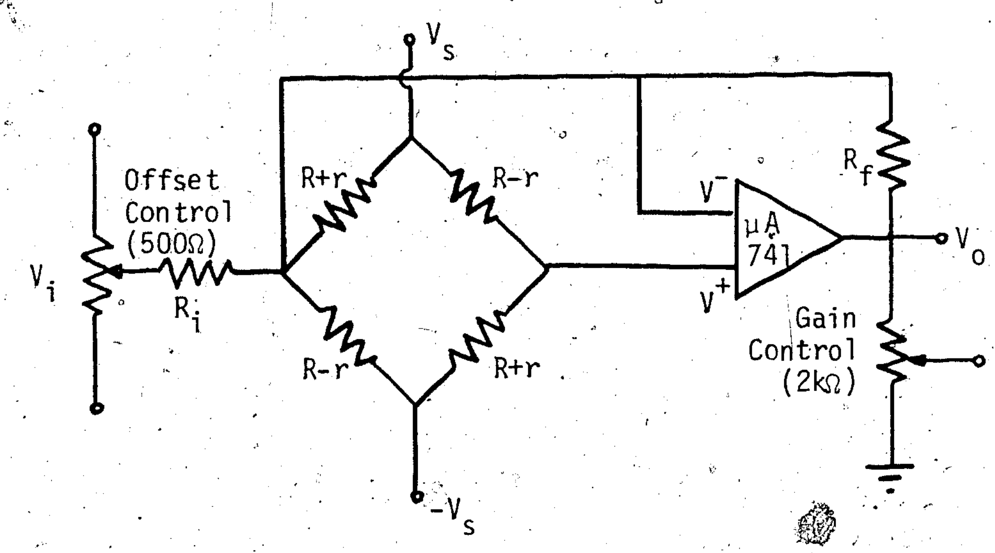


FIGURE E.1. a)Standard strain gauge Wheatstone bridge (top).
b)Strain gauge circuit used in force platform (bottom).

The actual circuit used was a modification of this bridge design, as shown in Figure E.1b. This circuit provides greater sensitivity to changes in the resistances of the strain elements. It was designed by Dr. Z. Koles. The operation of the circuit can be explained as follows. Assuming ideal operational amplifier characteristics, the input bias currents may be considered to be negligible and the input voltage levels approximately equal.

$$V^+ = -V_s + \{V_s - (-V_s)\}\,(R+r)/(R+r+R-r) = V_s r/R \simeq V^- \tag{E.3}$$

Summing the currents at the input node of the bridge:

$$\frac{V_i - V^-}{R_i} + \frac{V_0 - V^-}{R_f} + \frac{V_s - V^-}{R+r} + \frac{-V_s - V^-}{R-r} = 0 \tag{E.4}$$

Combining these two equations and simplifying produces:

$$V_0 = V_s\{4R_f r/(R^2 - r^2) + (1 + R_f/R_i)r/R\} - V_i\, R_f/R_i \tag{E.5}$$

If the resistance change, r, is treated as a fractional change, c, of the undeformed resistance, R, then c = r/R. The equation may be rewritten:

$$V_0 = V_s\{(4R_f/R)\, c/(1 - c^2) + c(1 + R_f/R_i)\} - V_i\, R_f/R_i \tag{E.6}$$

If $c^2 << 1$ and $4R_f/R_i >> 1 + R_f/R_i$, this simplifies to:

$$V_o \simeq V_s 4cR_f/R - V_i\, R_f/R_i \tag{E.7}$$

For this circuit, Micro-Measurements strain gauges (type EA-13-250MQ-350) were used. The circuit resistance values were: R = 33K, R_i = 120K, and R_f = 33K. Using these values:

$$V_o = 5700c - 0.3V_i \tag{E.8}$$

to better than 5%. For a maximum change of c = 0.1, an output of 570 volts is produced, assuming a null offset voltage.

APPENDIX F

Sway Detector Circuitry

The sway detector circuit consists of a photovoltaic diode modulating a preamplifier. The circuit (see Figure F.1) is a suggested application from the manufacturer (Quantrad Corporation). The photodiode is represented by a current source, i_D, with a resistance, R_D. The noise bandwidth and upper frequency response are determined by capacitor, C:

$$f = 1/2\pi RC \tag{F.1}$$

The output voltage is expressed:

$$V_o \approx i_D R \tag{F.2}$$

The operation of the circuit is that of a transconductance amplifier. With the input impedance of the operational amplifier being much larger than that of the feedback loop or the diode, the feedback and diode currents are approximately equal. With AC coupling and low frequency excitation of the diode, the output is dependent upon the diode current and the feedback resistor. The bias resistor is set equal to the feedback resistor to minimize offset.

Using the uA741 operational amplifier and component values of R = 470K and C = 0.27nF, a corner frequency of 1254 Hz and a sensitivity of 0.3 uA/uW of light results in

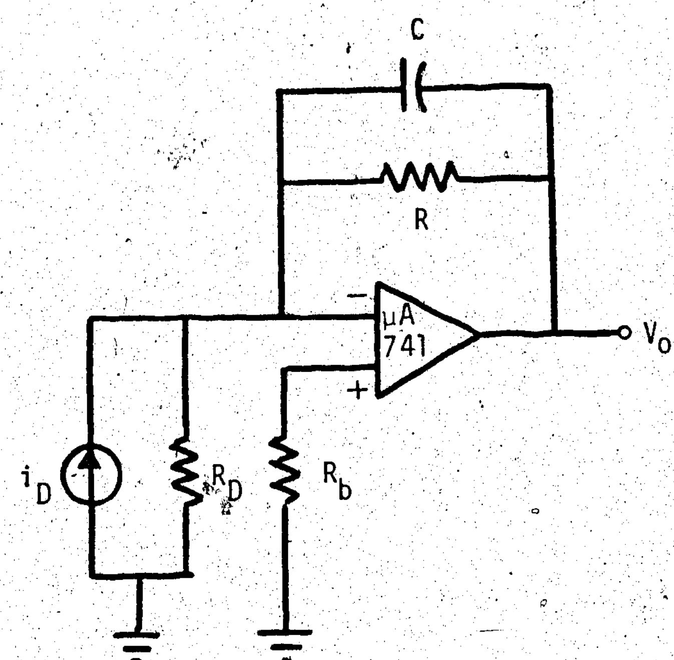


FIGURE F.1. Photovoltaic diode circuit of sway detector.

an adequate light detection circuit. The sensitivity of the detector to displacement is in the range of 38 mvolts/mm.

APPENDIX G

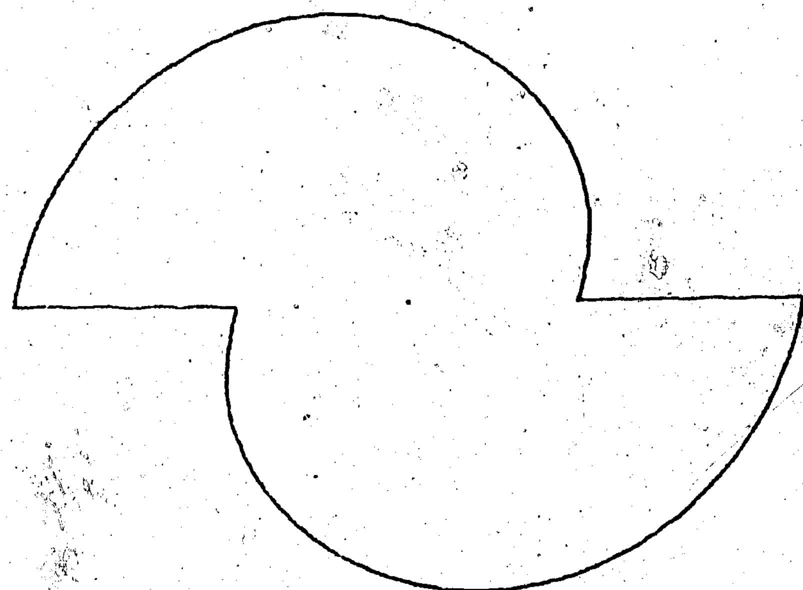
Sway Detector Linearity

A fast, repeatable test of the linearity of the sway detectors is necessary to demonstrate that the measurement system will produce the same output for the same input (movement) at any location within the light beam. Such a test would also be useful in focussing the light beams or monitoring the condition of the light sources.

The verification of the linearity of the sway detectors was achieved by occluding the light beam at a uniform rate and recording the output. Ideally, a graph of the output of the sway detector circuit should produce a straight line of voltage versus width of the light beam. The slope of this line equals the sensitivity of the sway detector to movement. This assumes that the response of the sway detector is uniform over the frequency range of interest.

To sweep the light beam of the sway detector, an aluminum disc (3.2 mm thick) was machined to tolerances of 0.05 mm. The shape of the disc, as shown in Figure G.1a, is defined by two arcs. The radius of each arc increases linearly with the angle subtended by the arc. A DC motor turns the disc at a constant speed, aided by the flywheel action of the symmetrically balanced disc.

The output of the detector preamplifier is recorded without filtering, which would distort the waveform. Subsequent computer analysis indicates the degree of



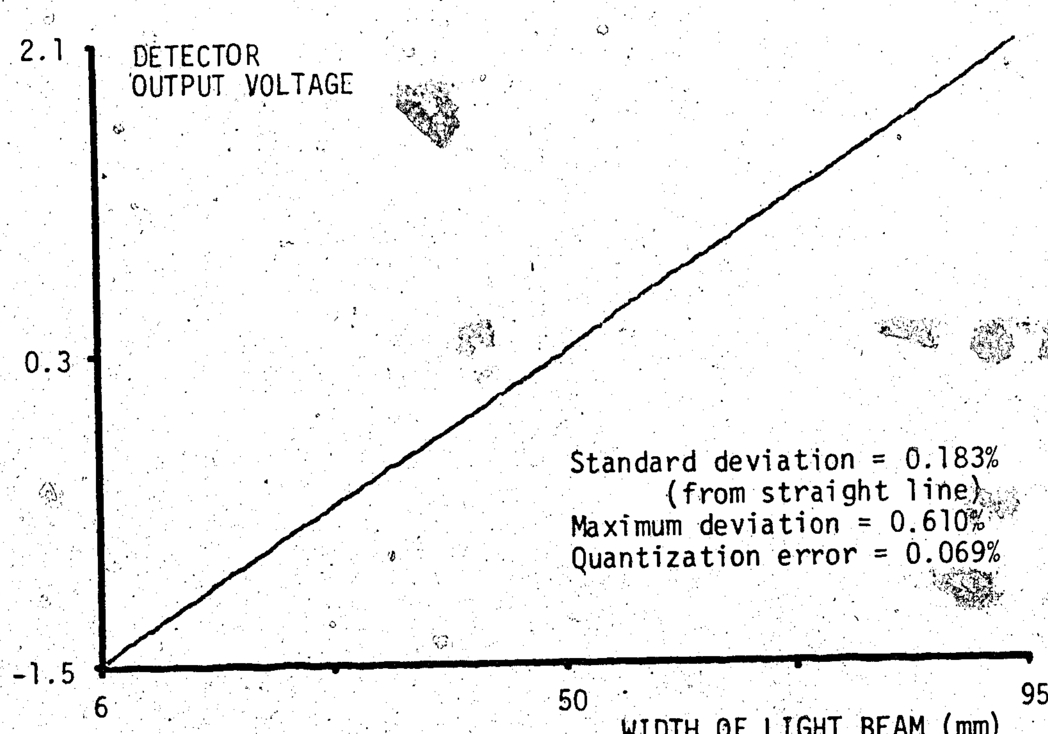


FIGURE G.1. a) Shape of disc to test linearity of sway detectors (top).
b) Curve from test of linearity of sway detectors (bottom).

nonlinearity by calculating the error between the output and a least squared fit straight line. A typical test curve is shown in Figure G.1b. The high degree of linearity is apparent, and the statistical values confirm this observation.

Quantization error introduces a degree of nonlinearity since the discrete samples cannot generally reproduce a straight line. The comparison of sample-to-sample slopes to the average slope may have been a better representation of linearity, but quantization error would have contributed a very large error since few quantization levels existed between consecutive samples.

APPENDIX H

Data Windows

The FFT is often used to produce the convolution of two waveforms. Convolution in the time domain becomes simple term-by-term multiplication in the frequency domain. Conversely, multiplication in the time domain corresponds to convolution in the frequency domain.

A sample of a continuous signal may be considered to be the product of a signal of infinite length with a data window. In the case of a block of N sampled data points with unaltered amplitudes, the data window is rectangular. The transform of the data sample is then equivalent to the convolution of the frequency domain representations of the continuous data with the rectangular data window. The Fourier transform of the data window is referred to as a frequency window.

The transform of the rectangular window reveals the problem of leakage. The sharp cutoff of the window in the time domain may be considered to add high frequency components to the data, while also limiting the low frequencies which may be adequately represented. The transform is characterized by a central peak with smaller peaks or sidelobes on either side. Each frequency component of the data sample is thus represented by a central peak, and sidelobes which interfere with neighbouring frequency components. To reduce the degree of interference or leakage, the shape of the data window is changed (ie. "window carpentry").

The triangular or Bartlett data window is regarded as being adequate for many applications (Welch). It requires relatively little computation to utilize, while significantly improving upon the rectangular window. A graphical comparison between the two windows is provided by Figure H.1. The scaling of the data values causes some information to be lost. The peak amplitude of the frequency window decreases as a result. The gradual reduction of the data severely curtails leakage. The wider peak decreases the frequency resolution to some extent.

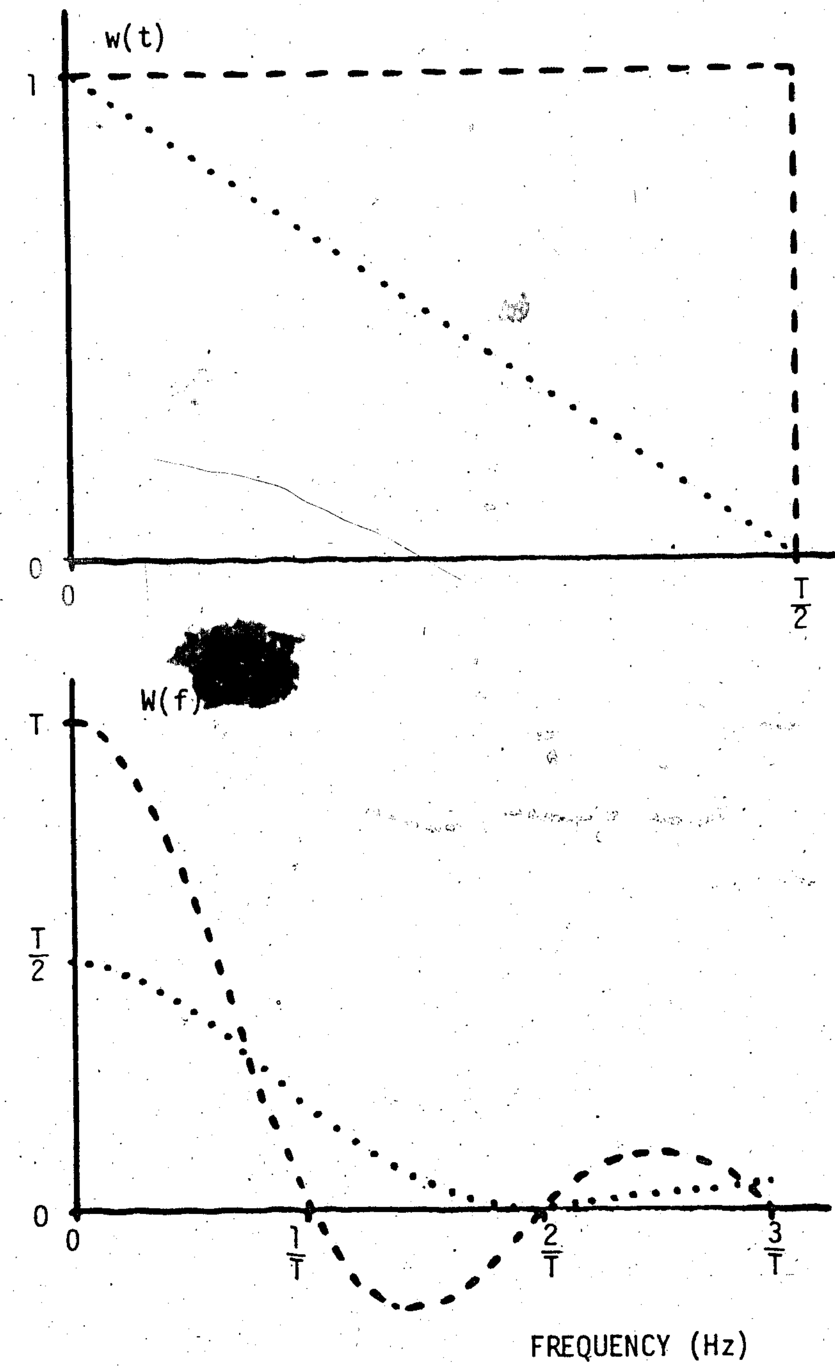


Figure H.1. Data windows (top) and corresponding frequency windows (bottom): rectangular window - broken line, triangular window - dotted line.

APPENDIX I

Spectral Windows

The frequency domain counterpart of the data window is the spectral window. The inverse transform of the spectral window (in the time domain) is referred to as a lag window. The use of spectral windows is based on the same principles as those of data windows; to reduce the effects of spurious or undesired spectral characteristics. When computation time is not as great a factor, as in displaying processed data on a screen or printer, a more complex window shape may be employed.

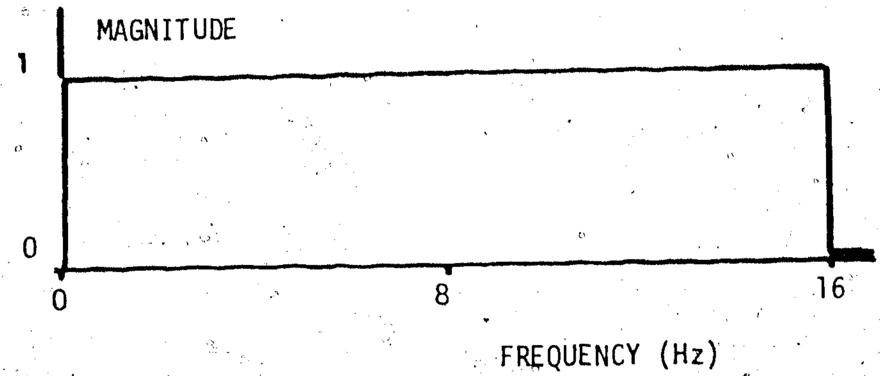
The cosine squared window is also called the cosine-bell, Hanning or Tukey window. Jenkins and Watts conclude that this window is quite adequate for most data analysis. The flat central portion of the extended cosine squared window serves to improve the bandwidth of the window. The inverse FFT of the modified spectrum can then reproduce sharper peaks within the impulse response. This window was further tailored in the present version to suit the particular spectra which were observed in the transfer function analysis.

The first spectral point (at DC) is of no interest. The measurement system is AC coupled. However, the peaks around 2 Hz are the most prominent and consistent of the results. To preserve this information, the spectra were tapered only below this frequency. The spectral information above 8 Hz was observed to be of less importance. The full benefits of the cosine squared spectral window are used above this higher frequency.

The results of this "window carpentry" are illustrated in the following figures. Each window is applied to a "white noise" spectrum, where the amplitude at each frequency point is unity. The inverse FFT produces the impulse response of the modified spectra.

The rectangular window of Figure I.1 demonstrates the 'worst case', where the spectrum is not tapered. The lack of the lowest frequency component is seen to introduce a decaying offset, as the impulse response is slow to settle upon zero amplitude. In Figure I.2 the cosine squared window is seen to reduce the sidelobes beyond the first one considerably. The decrease in low frequency content has shifted the main peak and first lobe downwards, as AC coupling would do to a signal with low frequency content. The extended cosine squared version of Figure I.3 behaves similarly to the rectangular window, with a faster attenuation of the side lobes. The gradual attenuation of the low frequencies by the window has reduced the shift seen in the rectangular window.

The nonsymmetric extended cosine squared window demonstrates a compromise of the previous two versions. The side lobes are rapidly attenuated while the loss of low frequency information is minimized. This is depicted in Figure I.4.



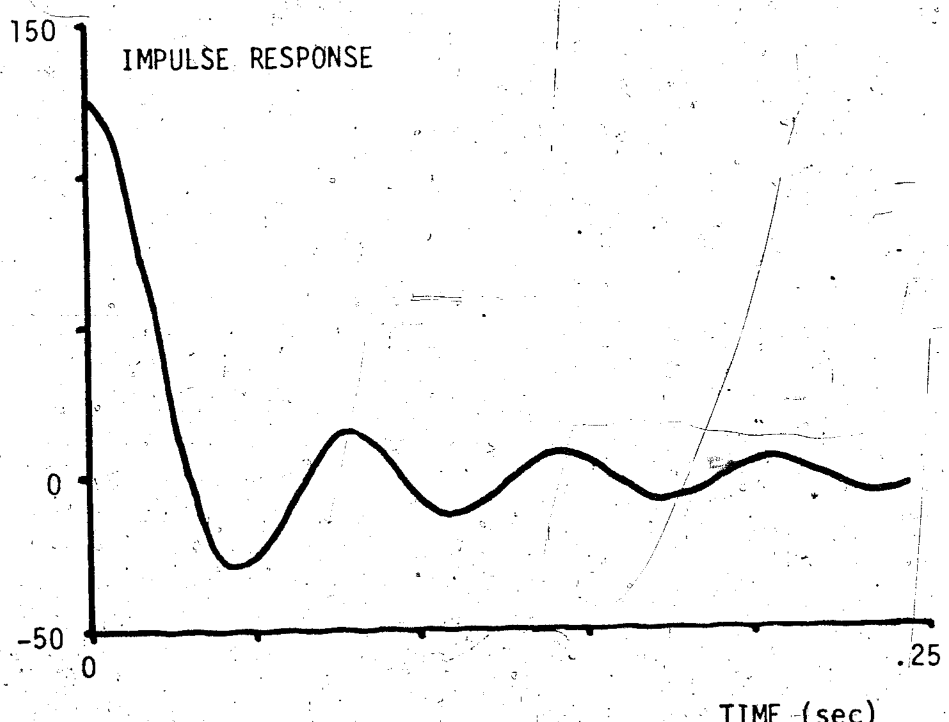


FIGURE I.1. Rectangular spectral window (top) with the resulting effect upon a "white noise" spectrum after an inverse FFT (bottom).

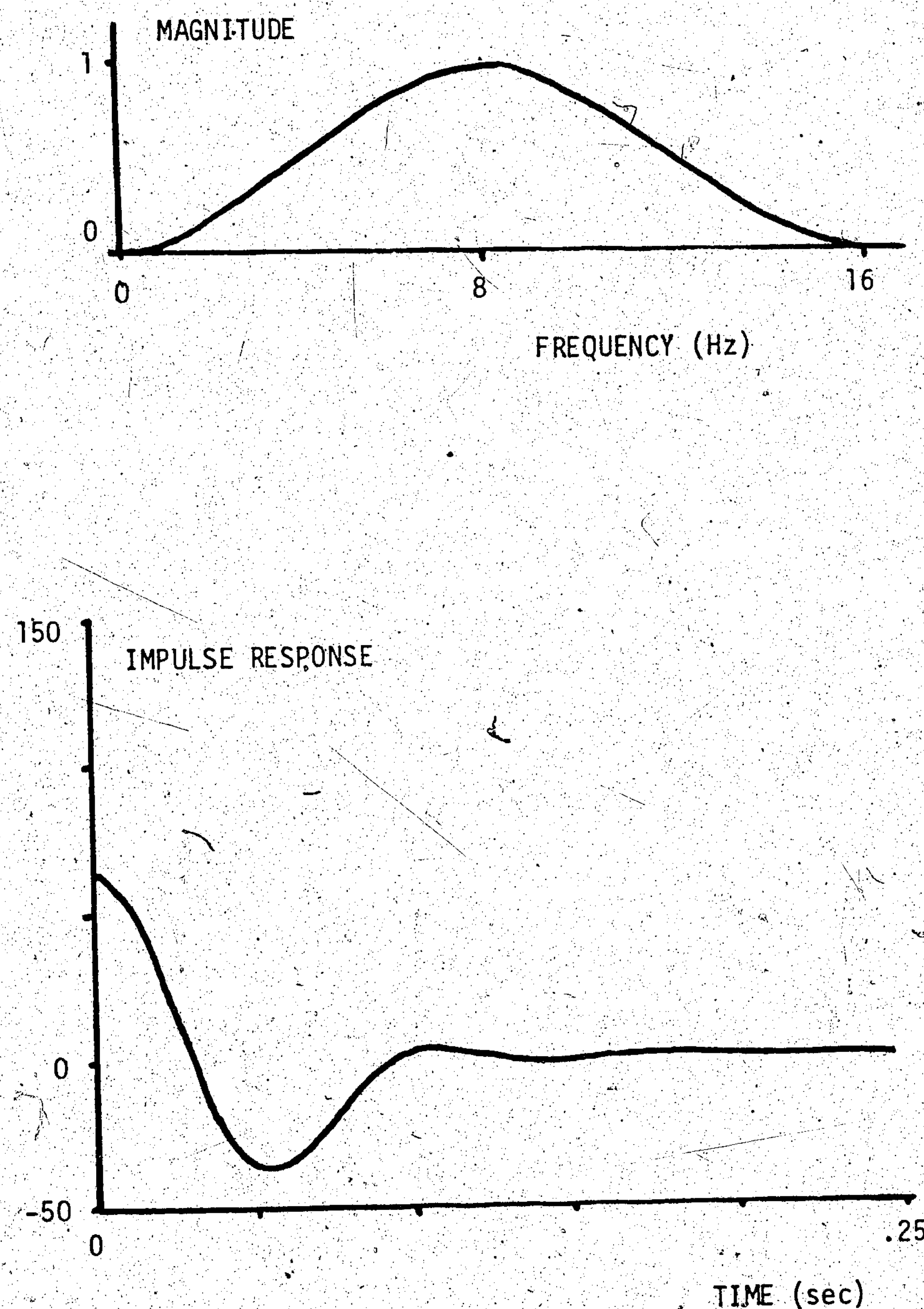


FIGURE I.2. Cosine squared spectral window (top) with the resulting effect upon a "white noise" spectrum after an inverse FFT (bottom).

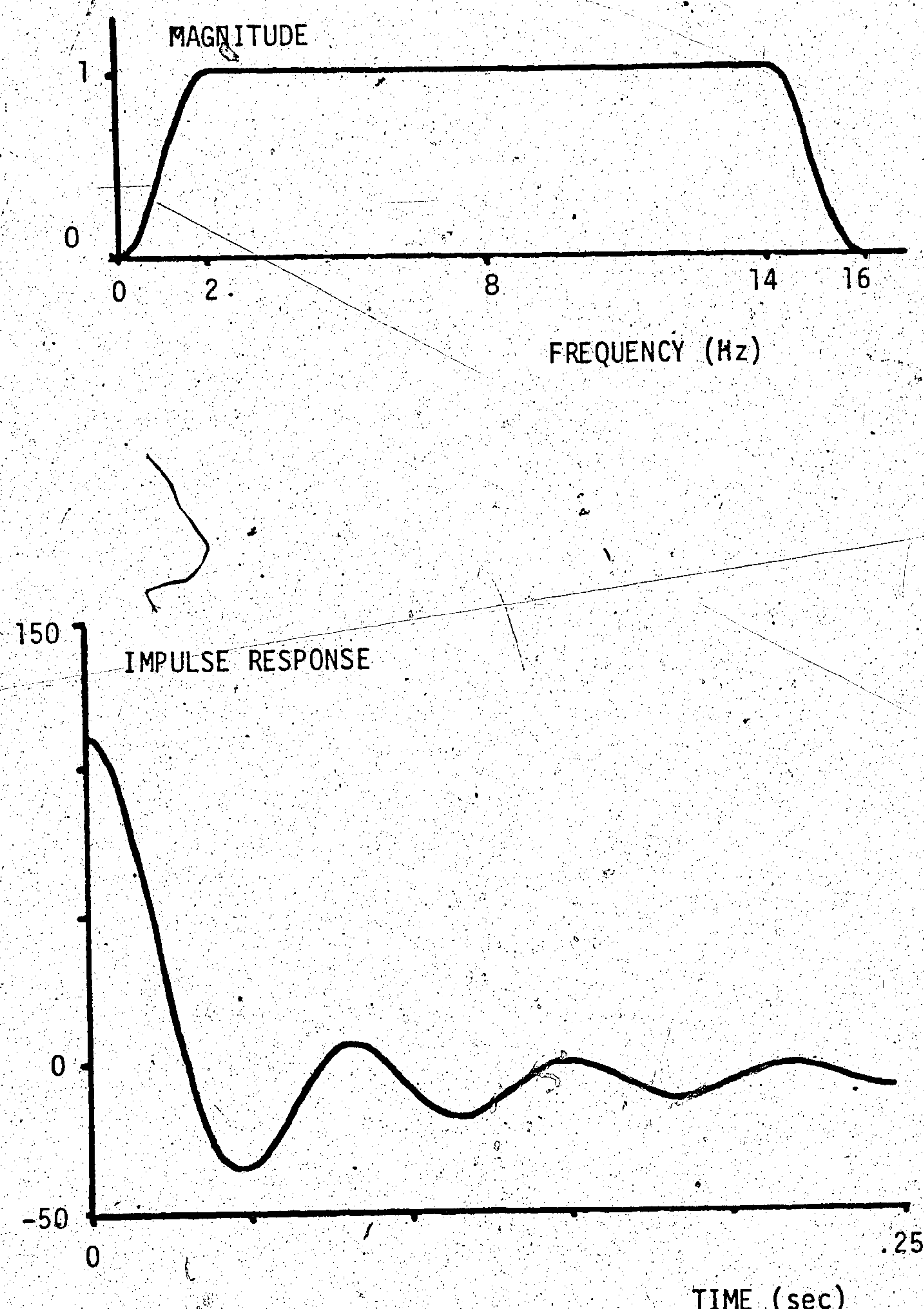


FIGURE I.3. Extended cosine squared spectral window (top) with the resulting effect upon a "white noise" spectrum after an inverse FFT (bottom).

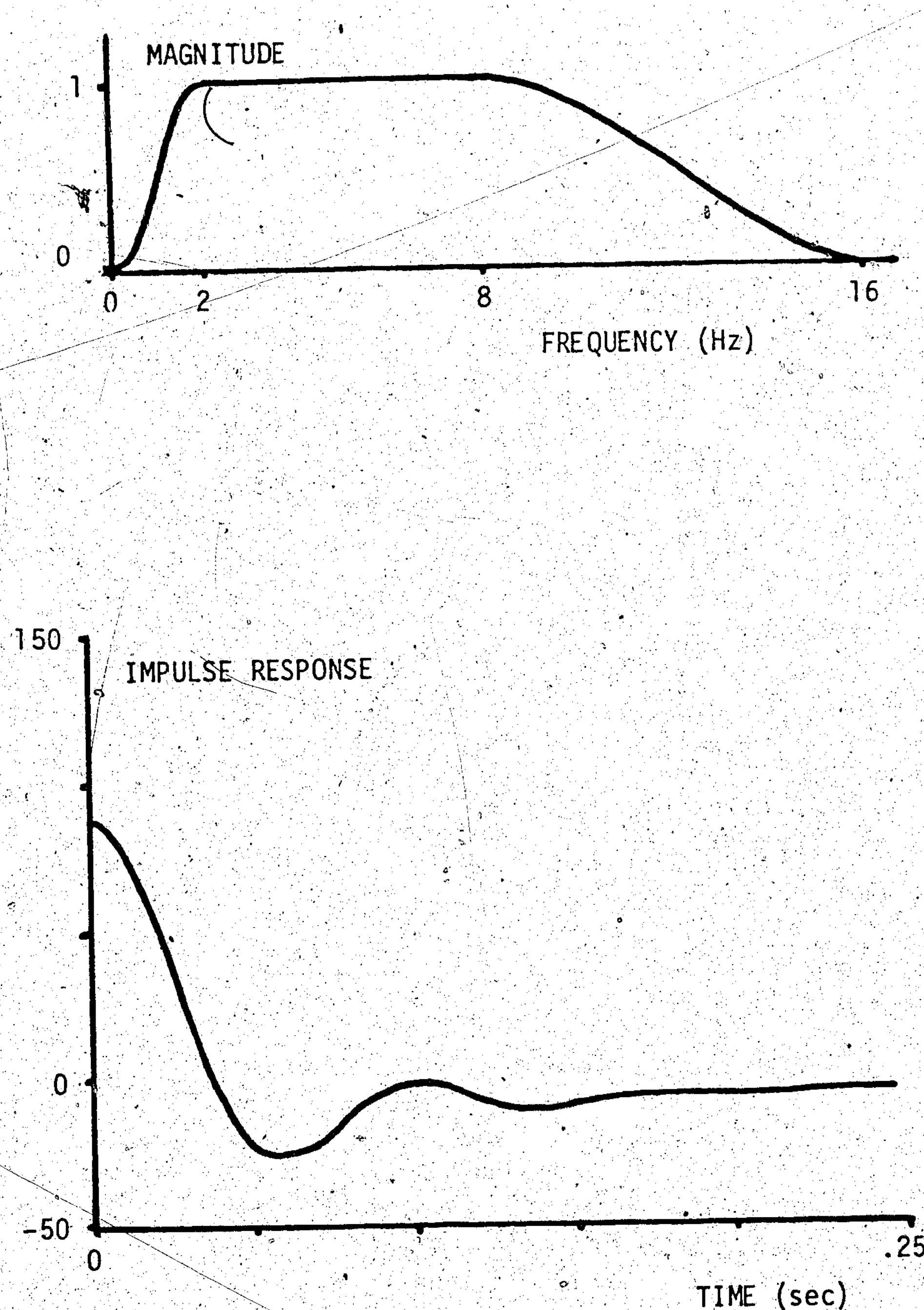


FIGURE I.4. Nonsymmetric extended cosine squared spectral window (top) with the resulting effect upon a "white noise" spectrum after an inverse FFT (bottom).

APPENDIX J

Flow Charts

The flow charts for three computer programs are presented on the following pages. The language is exclusively HP Fortran. PCS1 in Figure J.1 reads the data from magnetic tape, processes this data, and writes the reduced results onto magnetic tape. PCS2 in Figure J.2 reads the processed data from the magnetic tape, performs the transfer function analysis, and stores the results in disc resident files. PCS6 in Figure J.3 is similar to the remaining PCS programs in that the disc resident data is read and an averaging process occurs. With this particular program the output is in the form of the confidence limits of the averaged impulse responses of the transfer functions.

The abbreviations used within the flow charts are explained below:

AD - advance the magnetic tape a specified number of records

BA - backspace the magnetic tape a specified number of records

Cal - calibration data file

Calib - subroutine for providing equalization constants for sway data

EOF - end of file

Err - erroneous input

FFT - fast Fourier transform

Fname - the identifying filename which precedes data on tape or which locates disc resident data

FP - floating point

FS - file search by filename

FSCH - subroutine which locates data on magnetic tape using the filename

Int - integer

PC - parameter change which allows program variables to be changed

RW - rewind the magnetic tape to the starting position

ST - stop program execution

WR - write data onto magnetic tape

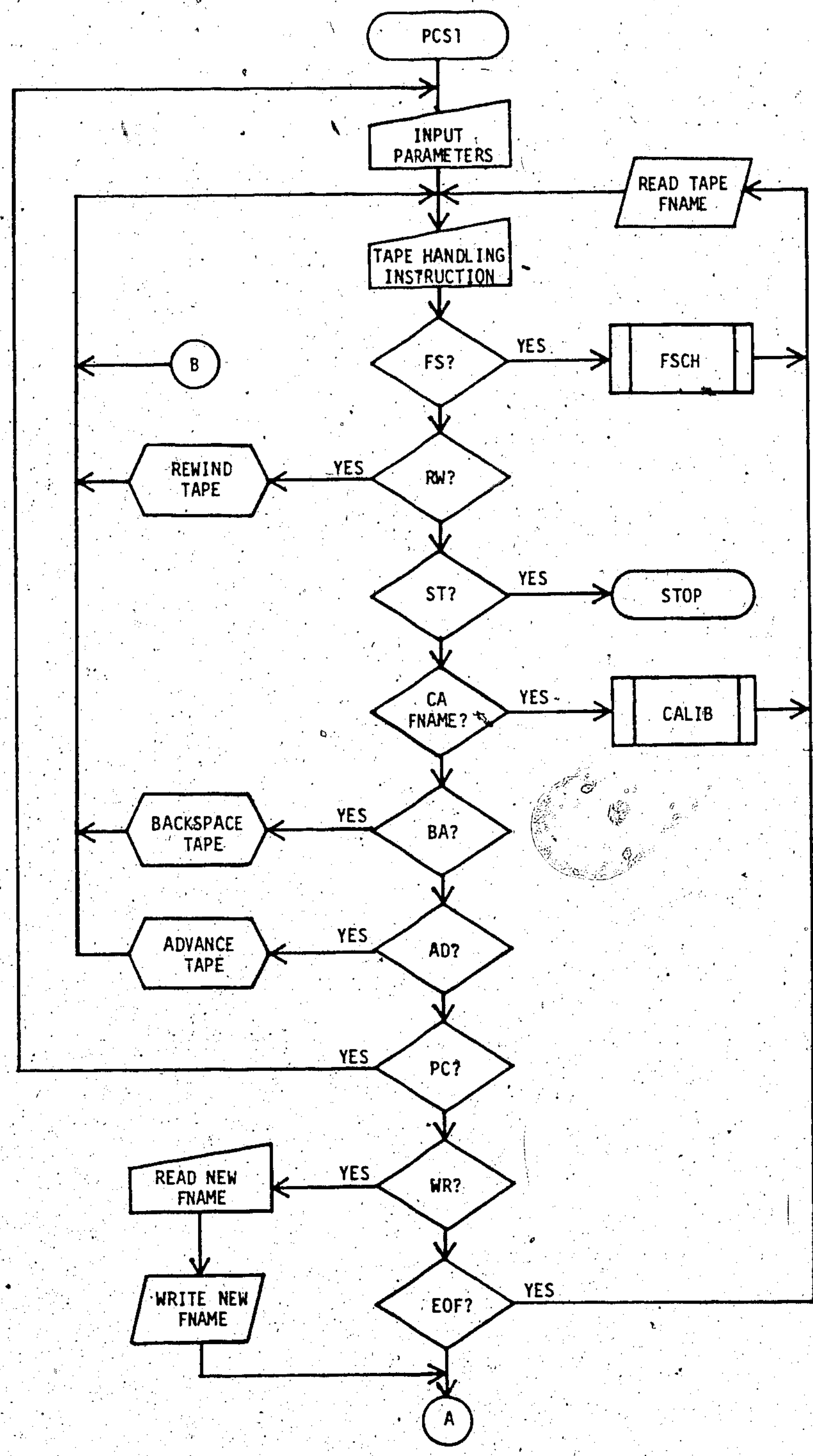


FIGURE J.1. Flow chart for data reduction program, PCS1.

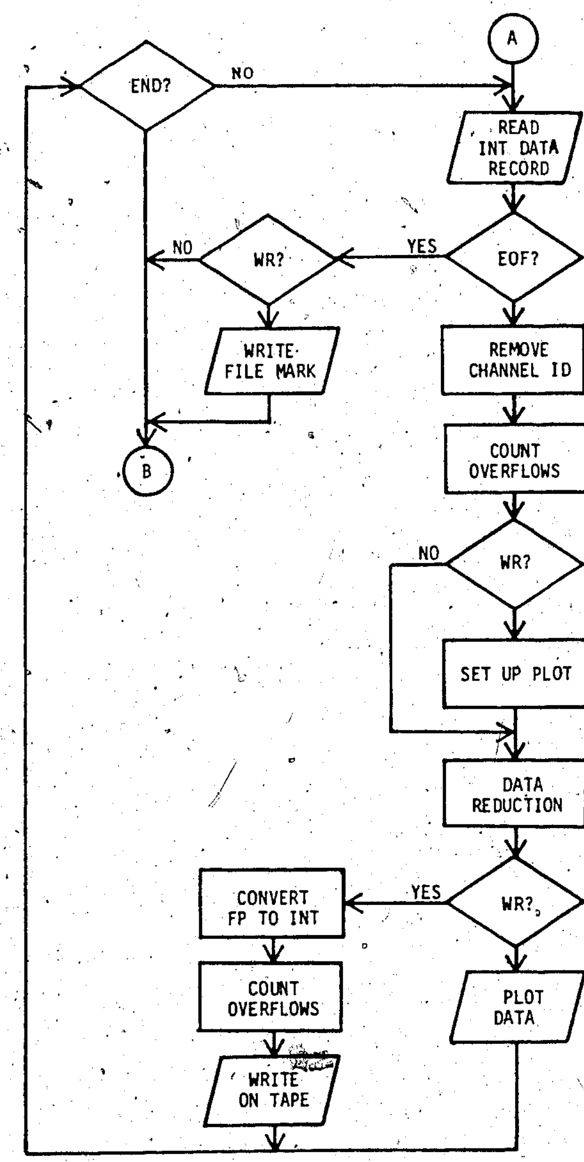


FIGURE J.1. (continued)

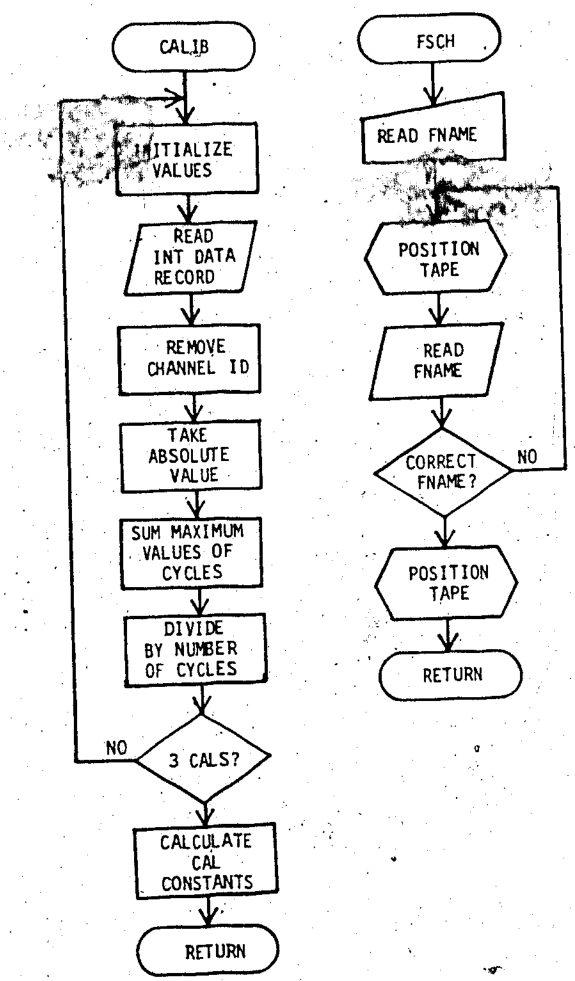


FIGURE J.1. Subroutines for program, PCS1.

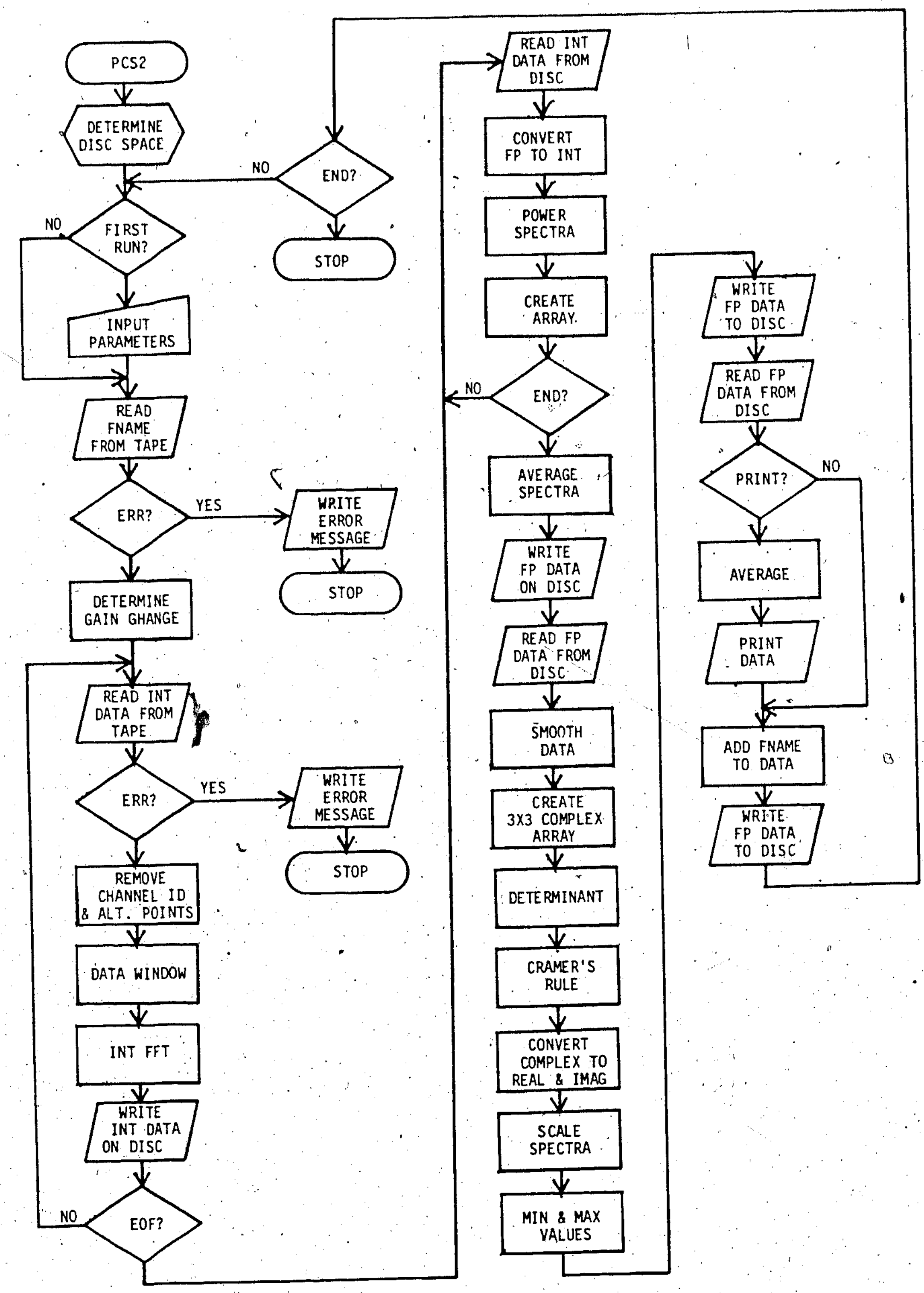


FIGURE J.2. Flow chart of transfer function analysis program, PCS2.

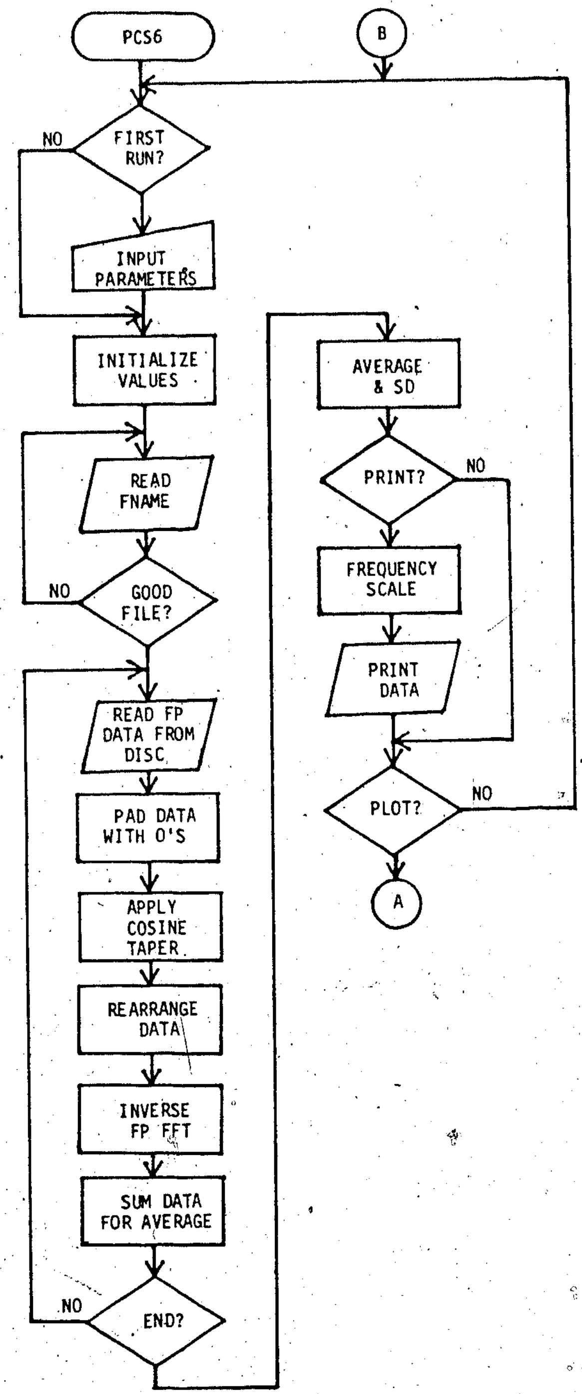


FIGURE J.3. Flow chart of impulse response averaging program, PCS6.

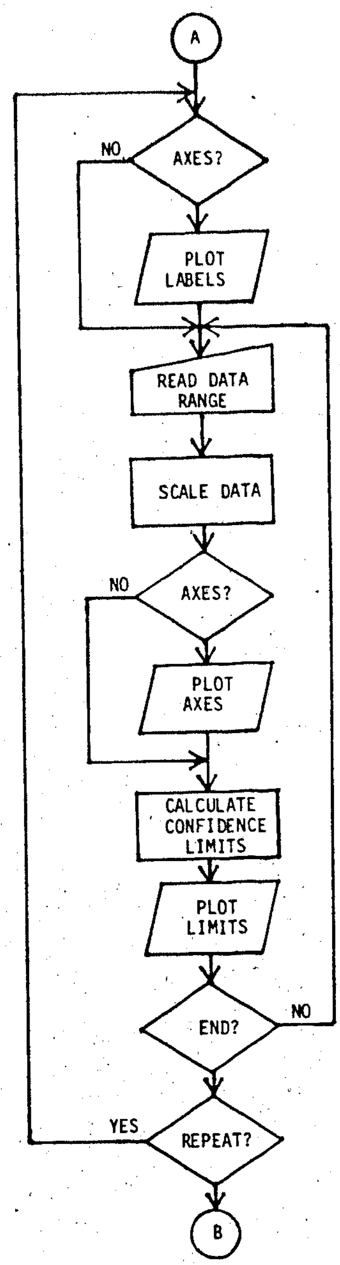


FIGURE J.3. (continued)

APPENDIX K

Statistical Tests

Assuming a normal distribution, the test results may be simply represented by the spread of the data. This may be expressed by confidence limits symmetrically centered about the mean. A graph of the confidence limits presents an effective demonstration of the magnitude of the data and the corresponding variance. If the confidence interval is equal to one standard deviation, approximately 68% of all test results will fall within the limits. The coefficient of variation is the ratio of the standard deviation to the mean.

The t test is utilized to provide a numerical comparison between sets of data. This statistical test is well suited to small samples from normal distributions. Since, in this case, it is important to detect deviations from standard or normal test data, the differences between two sets of data are often studied. Such pair tests or paired comparisons (see Guttman, Wilks, and Hunter) indicate whether one set of data is consistently higher or lower in magnitude than another. Although the pairs of data values may not be independent, the differences between pairs of data values are independent of each other.

The procedure involves a two-sided test of the hypothesis that the mean of the differences of two samples is greater or less than a mean of zero in a normal population. If the mean of the differences falls outside the confidence intervals around zero at a specified level of significance, then the comparison may be said to be significantly different from zero. The two sets of data are then significantly different, at the specified level, from each other.

Given a sample of differences, $x_i = y_i - z_i$, where $i = 1,\ldots,N$, with a mean, $\overline{x}$, and a standard deviation, s, the critical region for the test is governed by the expression:

$$\frac{|\overline{x}\sqrt{N}|}{s} > t_{N-1;a/2} \tag{K.1}$$

where a is the level of significance. For N = 20, the t tables provide the following limits for the critical region around a zero mean:

a	50%	20%	10%	5%	2%	1%	.1%
t	.688	1.328	1.729	2.093	2.539	2.861	3.883

The level of significance is decreased by half if a one-sided test is used. In this case, it is known that the differences between data values will be all positive or all negative. Therefore, only one side of the distribution is considered.

APPENDIX L

Results of Tests of One Normal Male

The following sets of graphs present the results from 20 EO and EC tests of one normal male. Figure L.1 and Figure L.2 display the averaged real and imaginary components respectively. Figure L.3 shows the differences between these components as derived from EO and EC tests.

Figure L.4 shows the average values of gain and phase for the EO and EC transfer functions. The impulse responses for the EO and EC tests in Figure L.5 are followed by the differences between these impulse responses in Figure L.6.

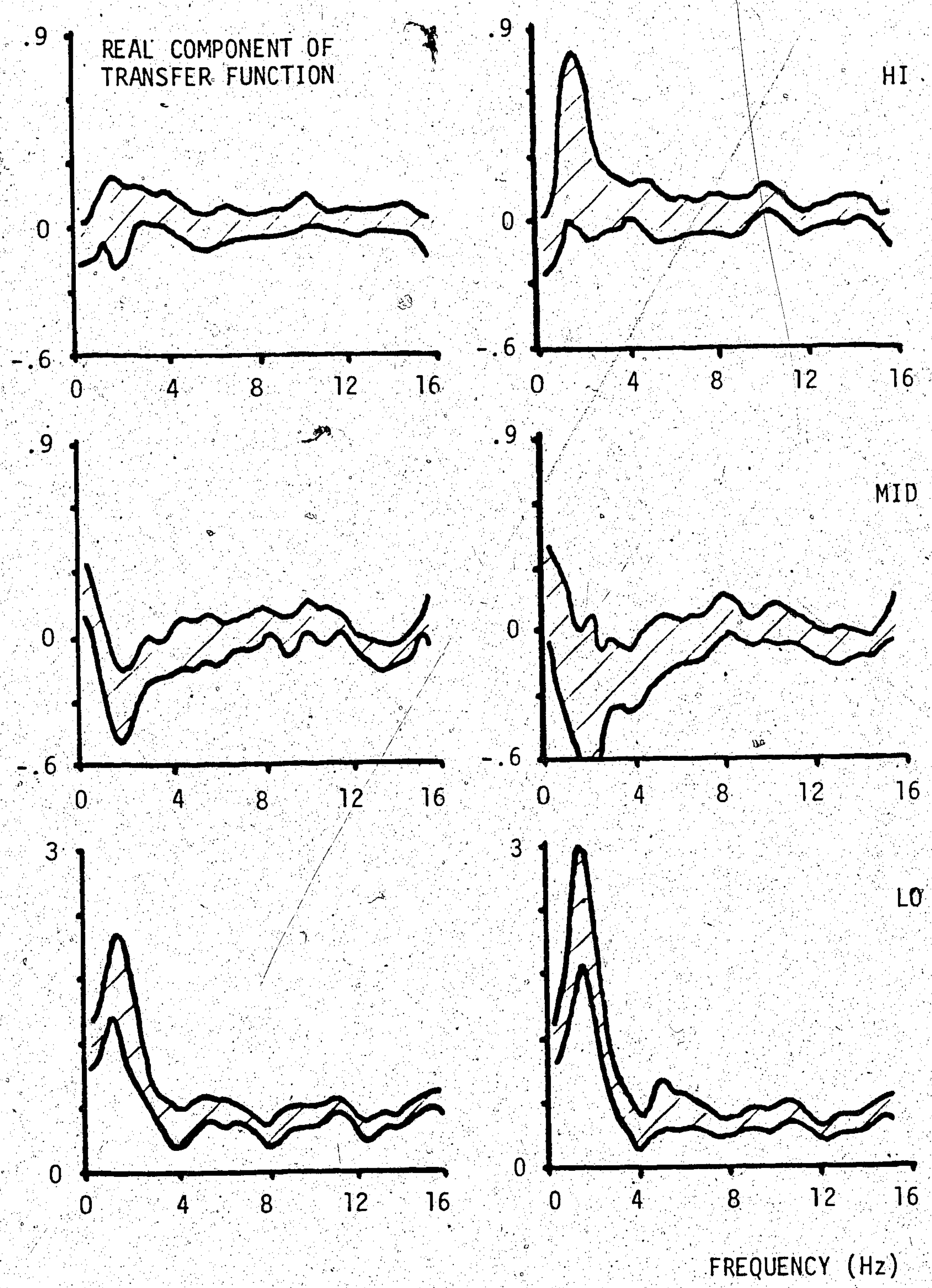


FIGURE L.1. 68% confidence limits of averaged real components of transfer functions from 20 EO (left) and EC (right) tests of one normal male.

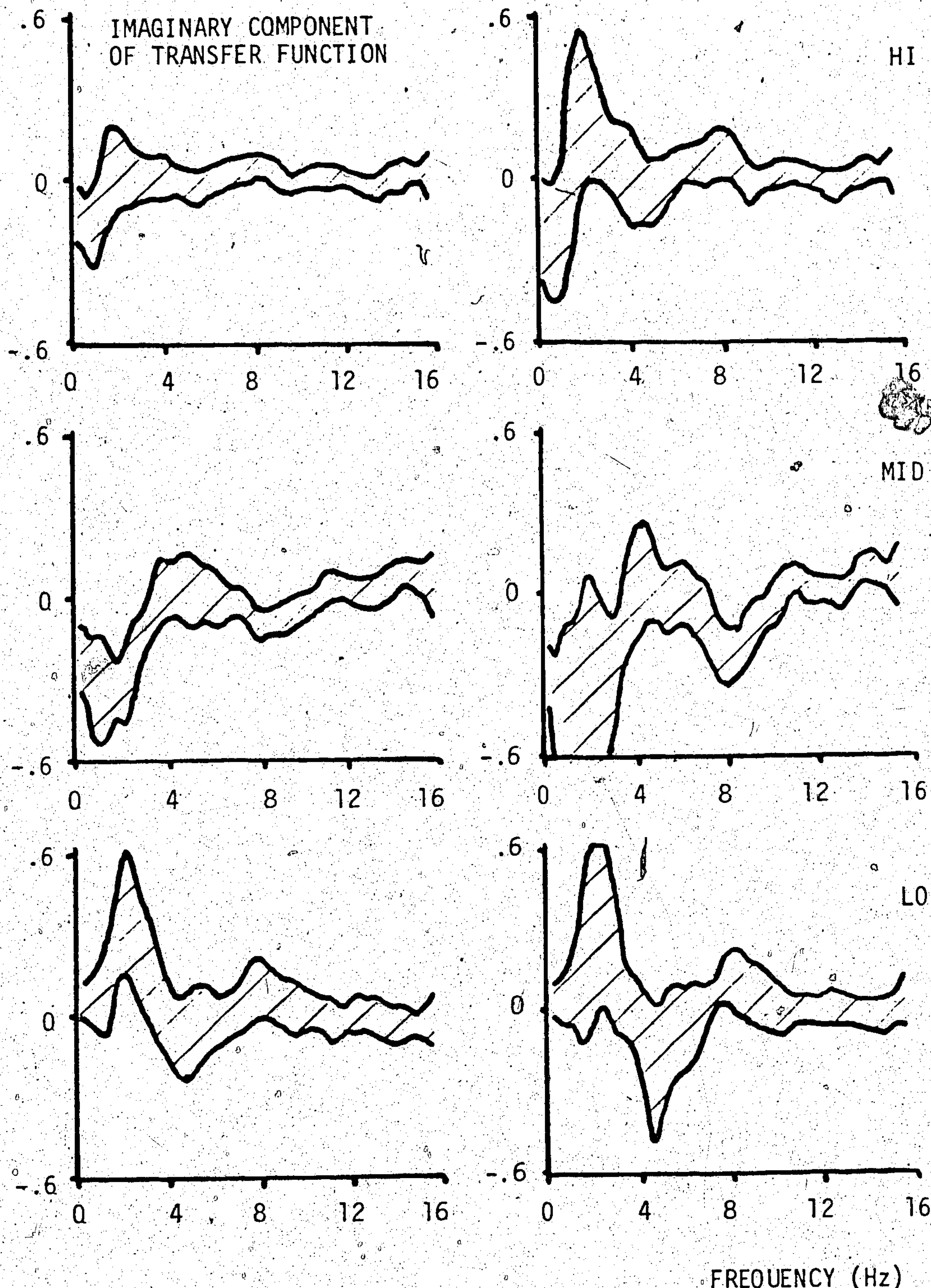


FIGURE L.2. 68% confidence limits of averaged imaginary components of transfer functions from 20 EO (left) and EC (right) tests of one normal male.

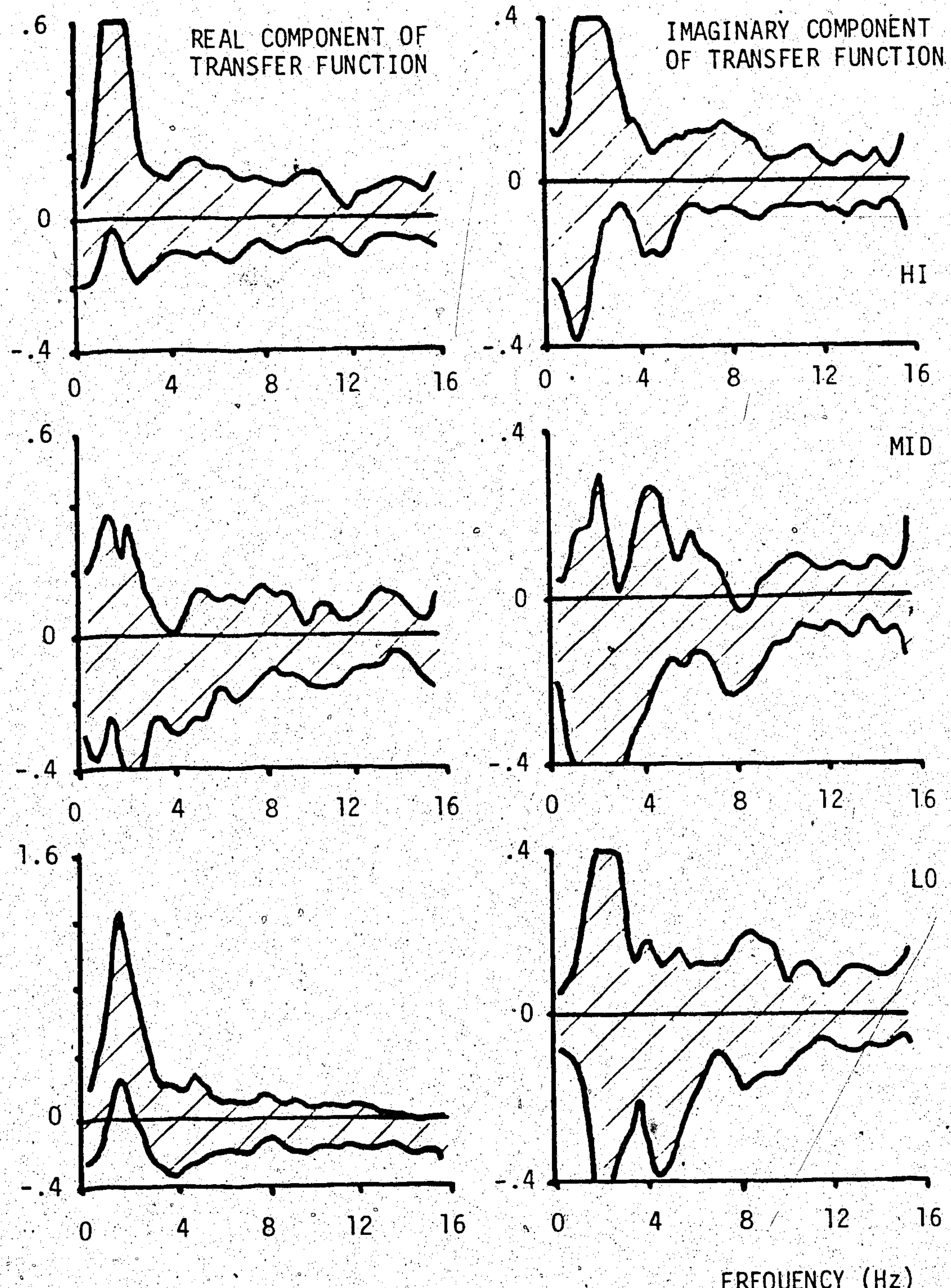


FIGURE L.3. 68% confidence limits of averaged differences between real (left) and imaginary (right) components of transfer functions from 20 EC and EO tests of one normal male.

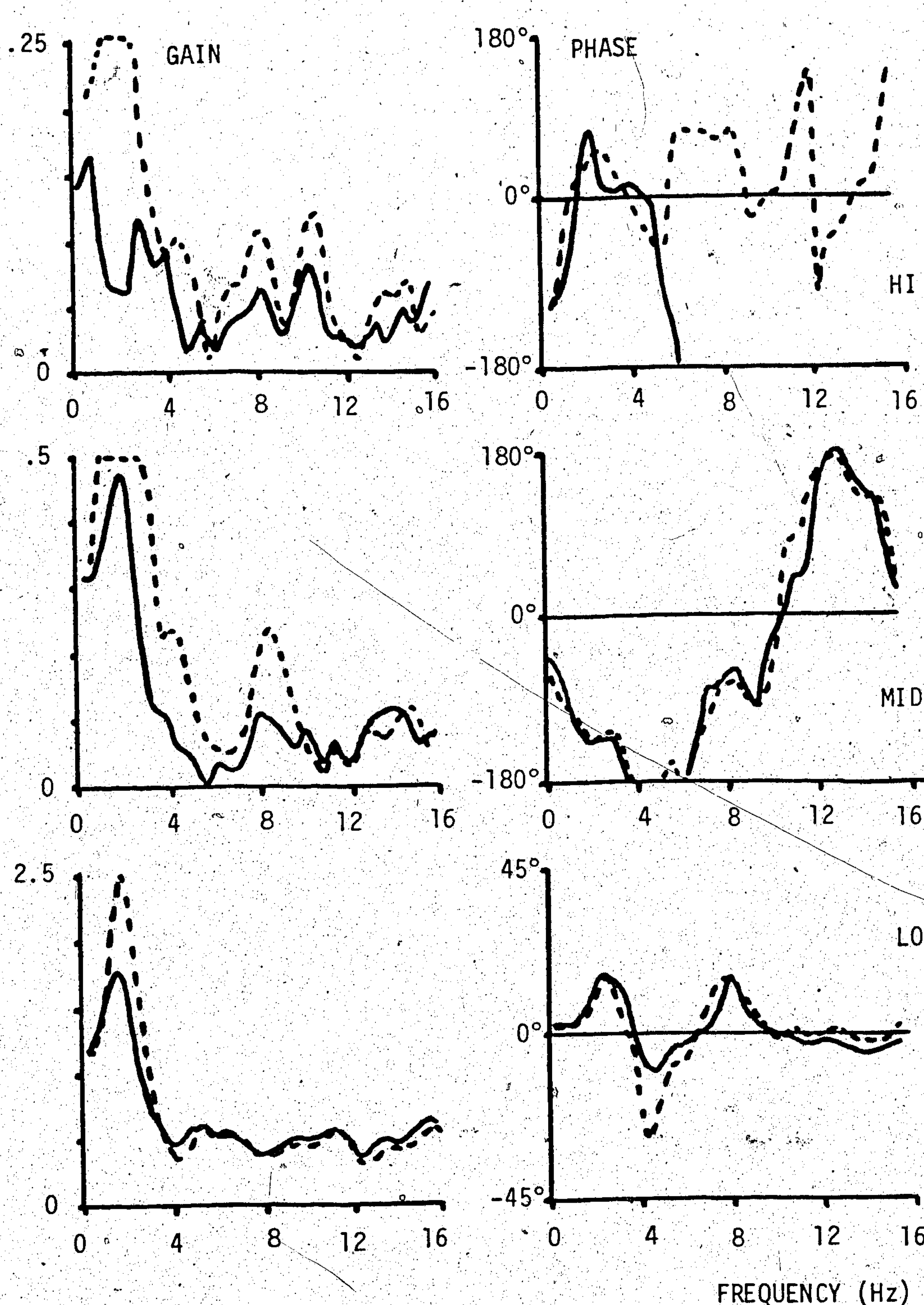


FIGURE L.4. Average values of gain (left) and phase (right) components of transfer functions from 20 EO (solid line) and EC (broken line) tests of one normal male.

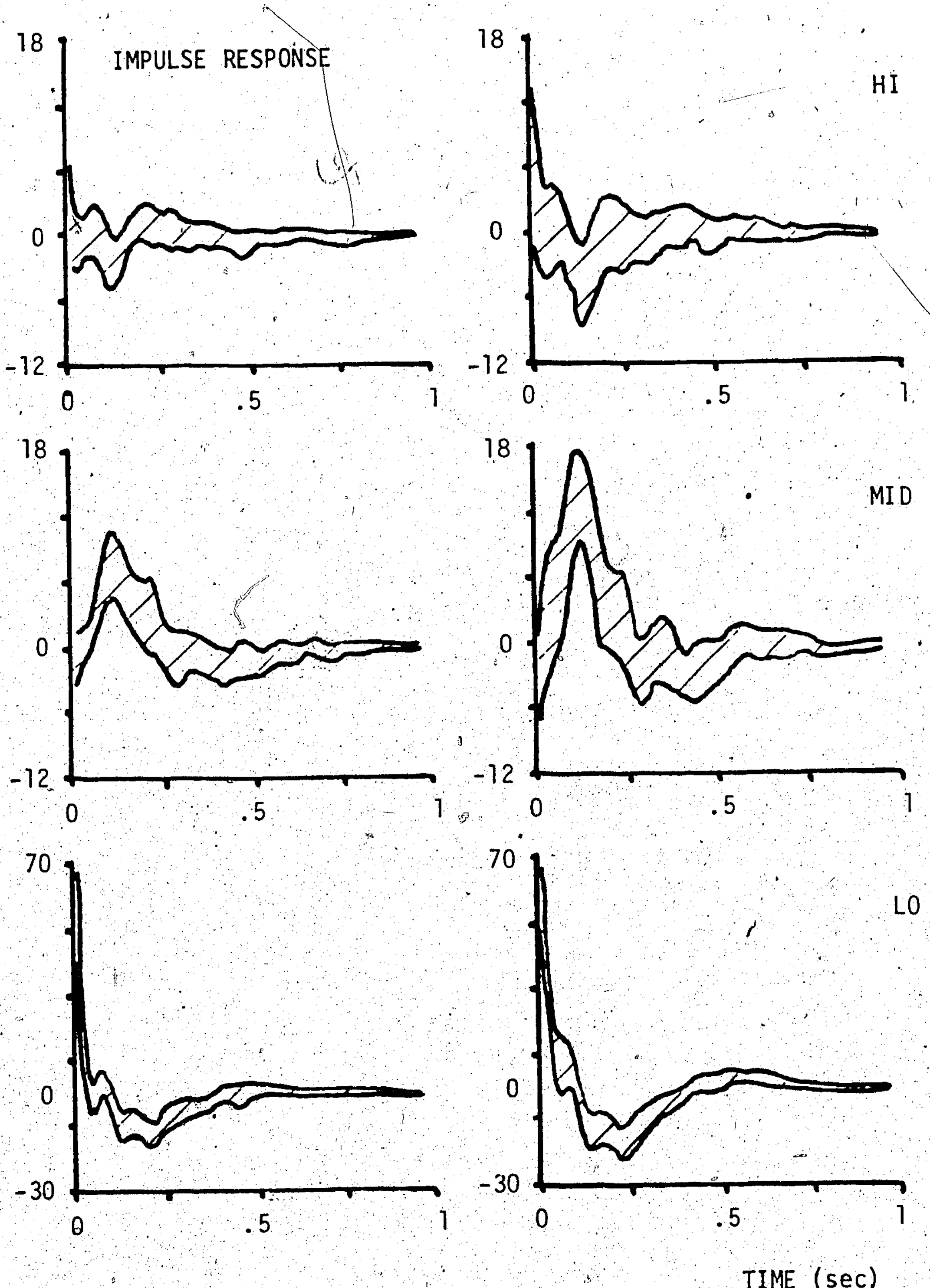


FIGURE L.5. 68% confidence limits of averaged impulse responses of transfer functions from 20 EO (left) and EC (right) tests of one normal male.

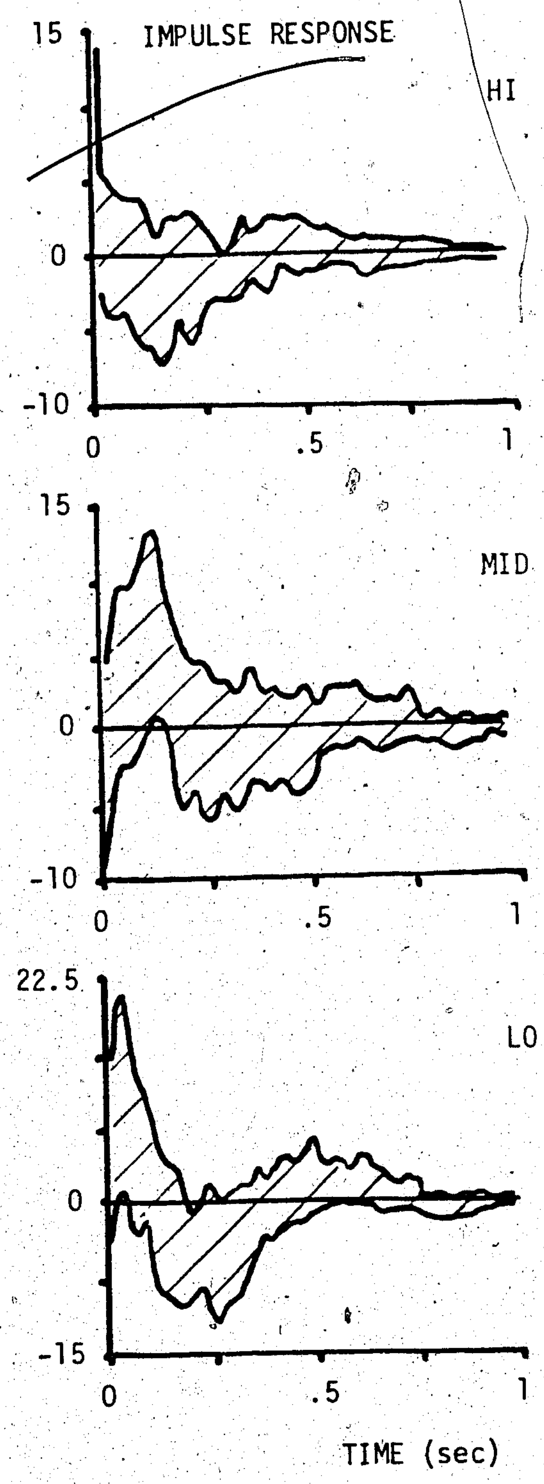


FIGURE L.6. 68% confidence limits of averaged differences between impulse responses from 20 EC and EO tests of one normal male.

APPENDIX M

Results of the Alcohol Tests

The results of the alcohol tests are displayed as differences between the real and imaginary components to illustrate the relative effects upon these parameters. Figure M.1 and Figure M.2 compare the EC - EO differences for sober and then intoxicated subjects. Figure M.3 and Figure M.4 take the differences between EO and EC tests for sober and then intoxicated subjects.

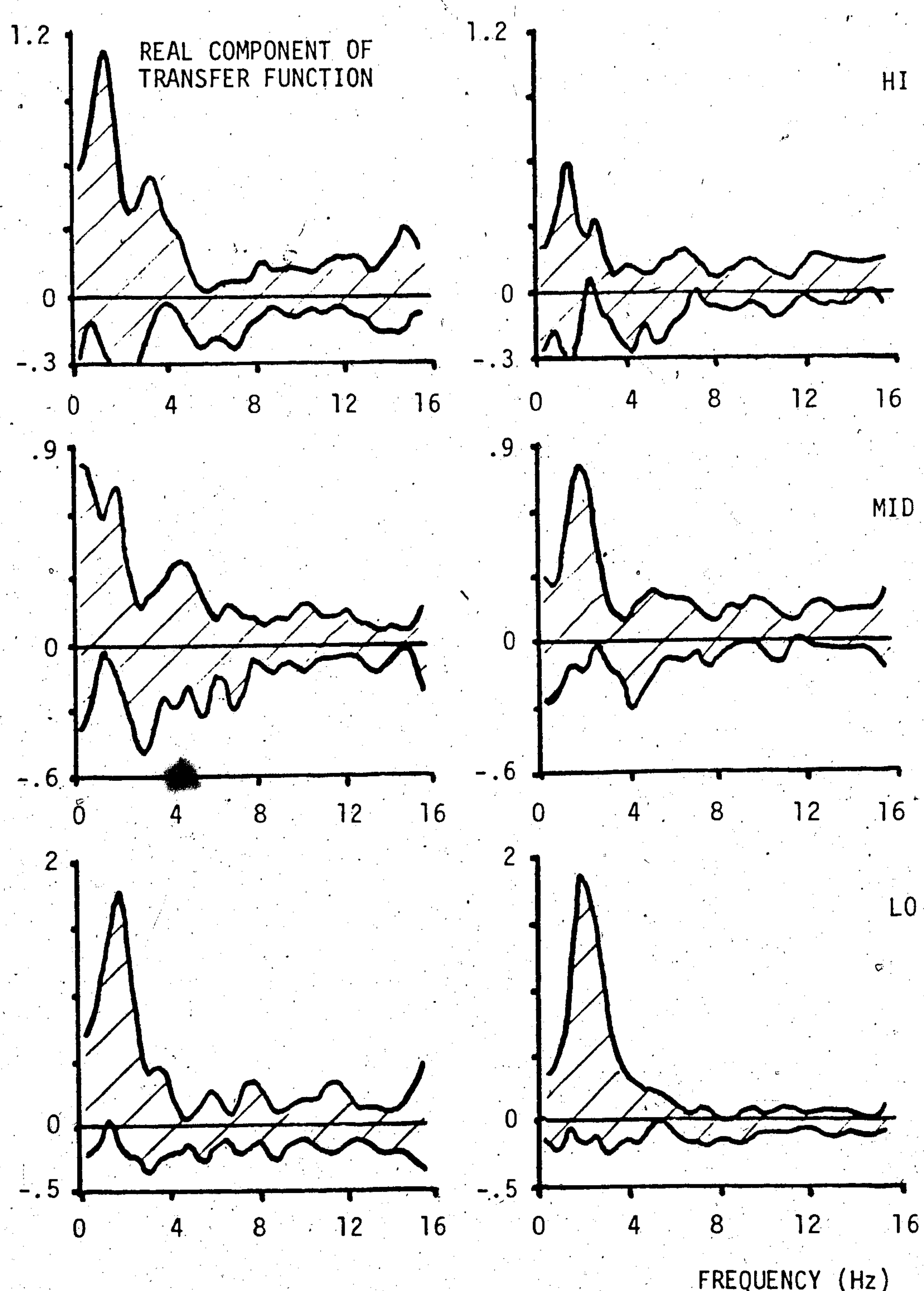


FIGURE M.1. 68% confidence limits of averaged differences between real components of transfer functions from EC and EO tests of 7 sober (left) and then intoxicated (right) normal males.

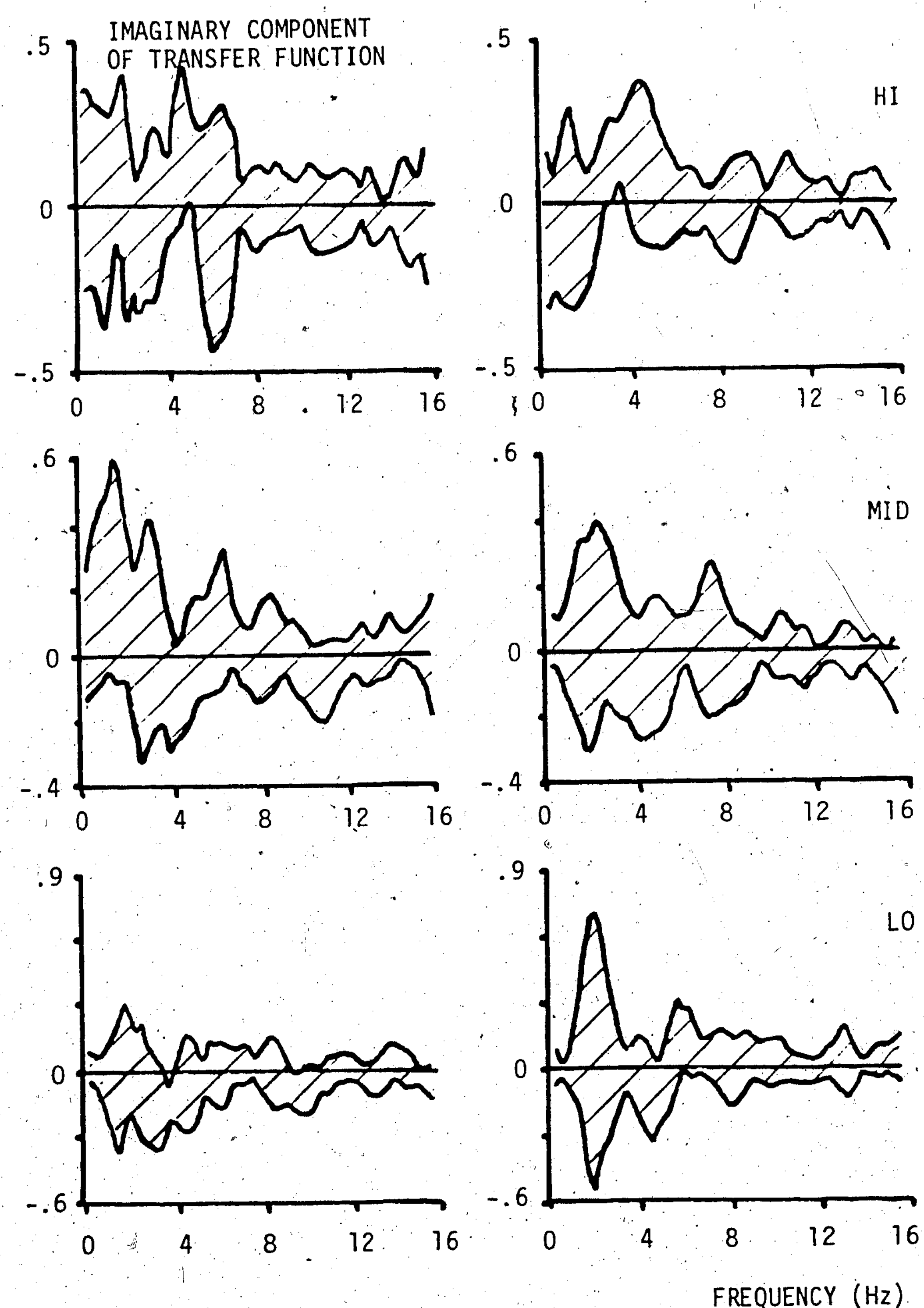


FIGURE M.2. 68% confidence limits of averaged differences between imaginary components of transfer functions from EC and EO tests of 7 sober (left) and then drunk (right) males.

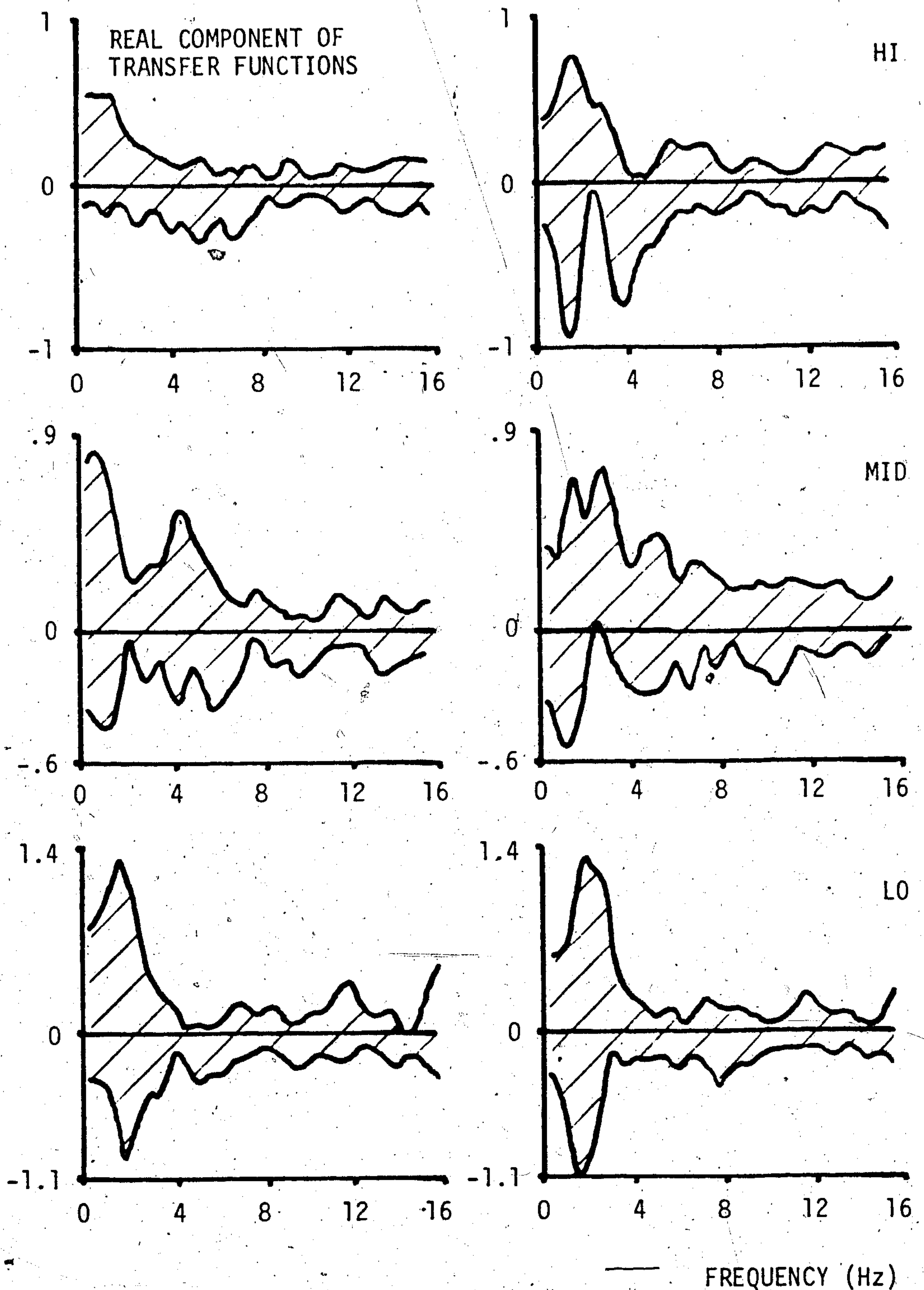


FIGURE M.3. 68% confidence limits of averaged differences between real components of transfer functions from EO (left) and EC (right) tests of 7 normal males, intoxicated and sober.

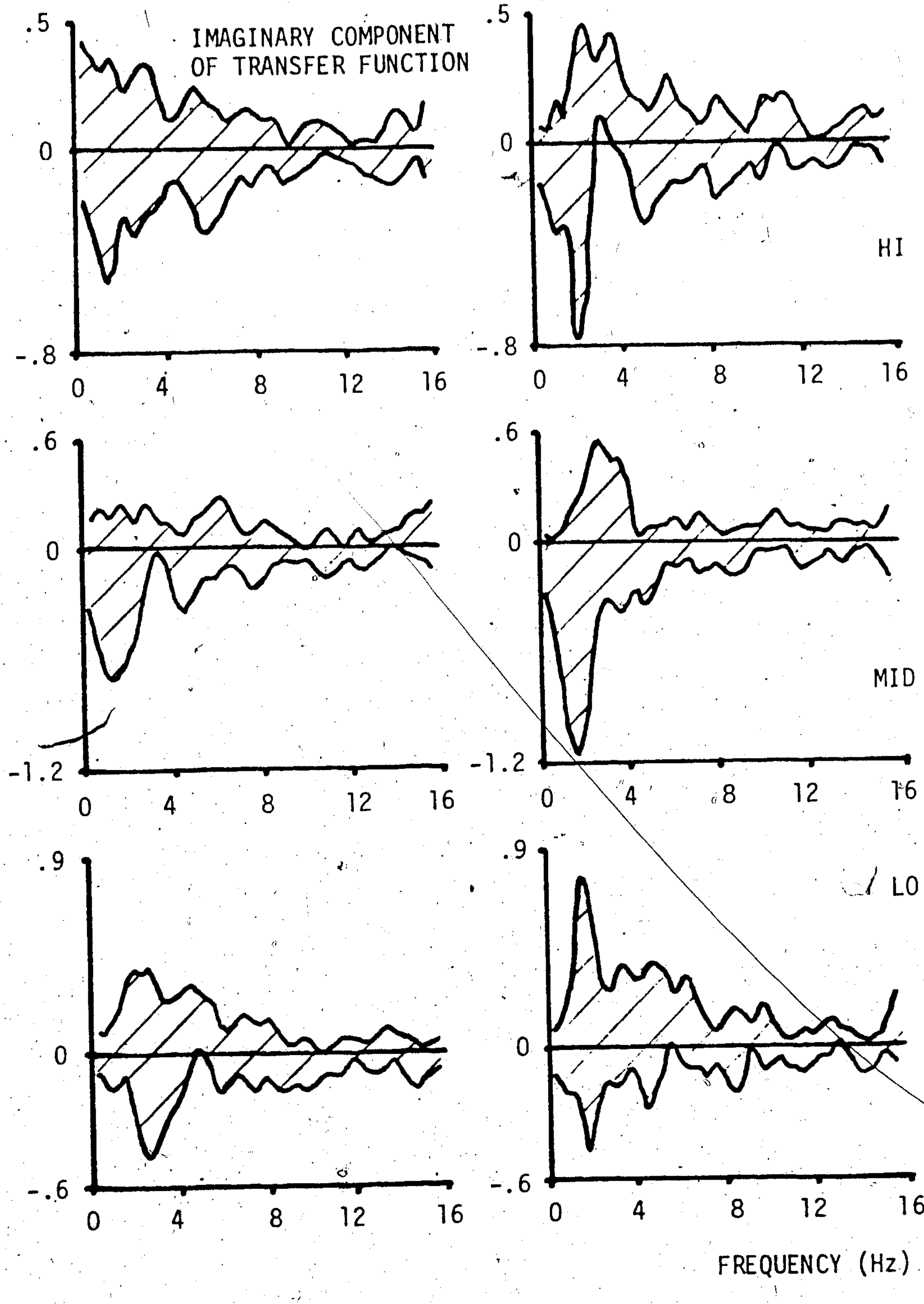


FIGURE M.4. 68% confidence limits of averaged differences between imaginary components of transfer functions from EO (left) and EC (right) tests of 7 males, intoxicated and sober.

APPENDIX N

Results of Tests of Neurological Patients

The EO and EC impulse responses from the tests of the three neurological patients are used to depict possible effects of their conditions upon their balance by comparing them to averaged normal curves. Figure N.1 deals with Parkinson's disease. Figure N.2 deals with a peripheral neuropathy. Figure N.3 deals with spino-cortical tract disease.

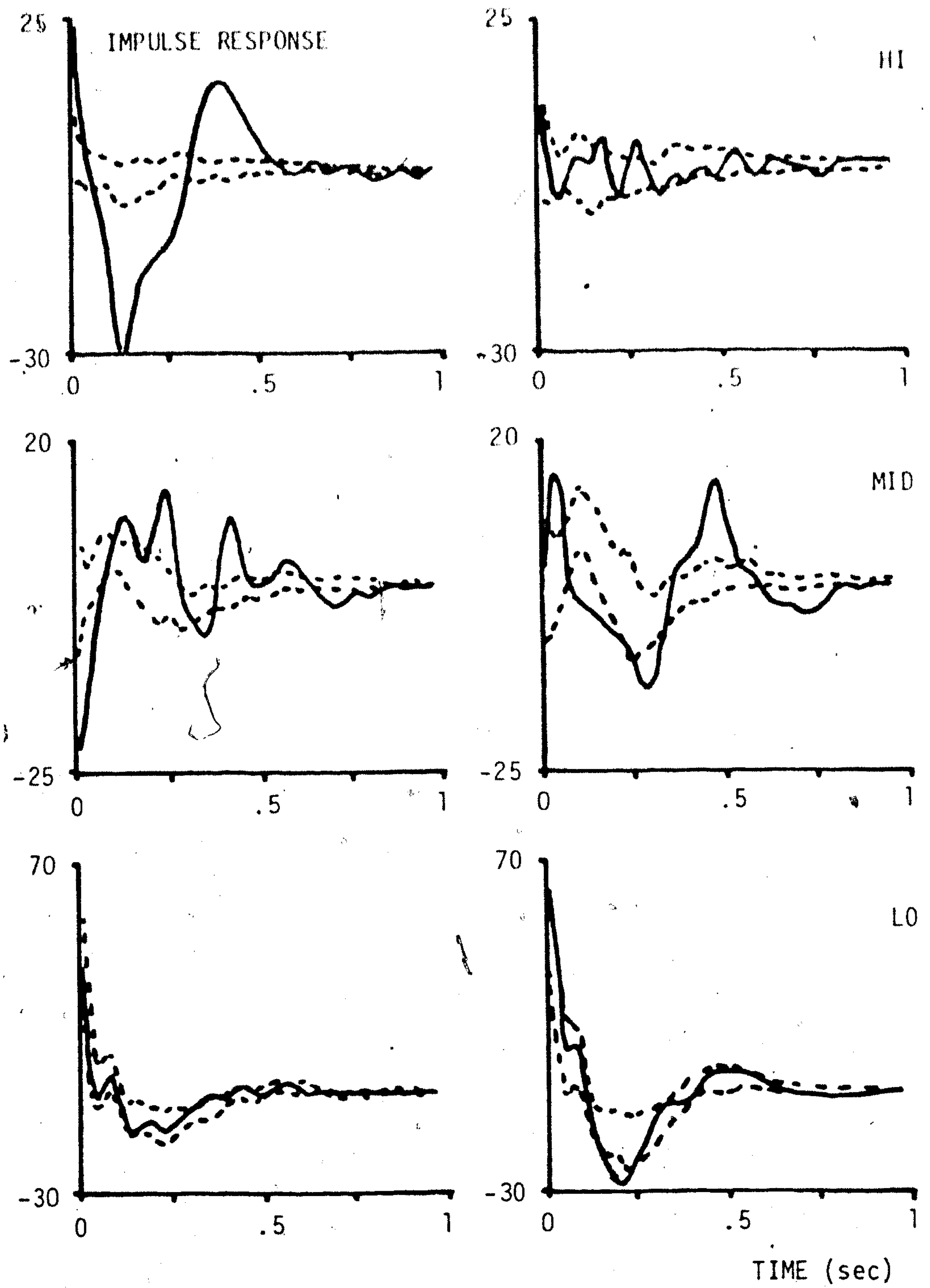


FIGURE N.1. Impulse responses of transfer functions from EO (left) and EC (right) tests of a female with Parkinson's disease (solid line) compared to 68% confidence limits of the corresponding averaged impulse responses from 27 normal subjects (broken line).

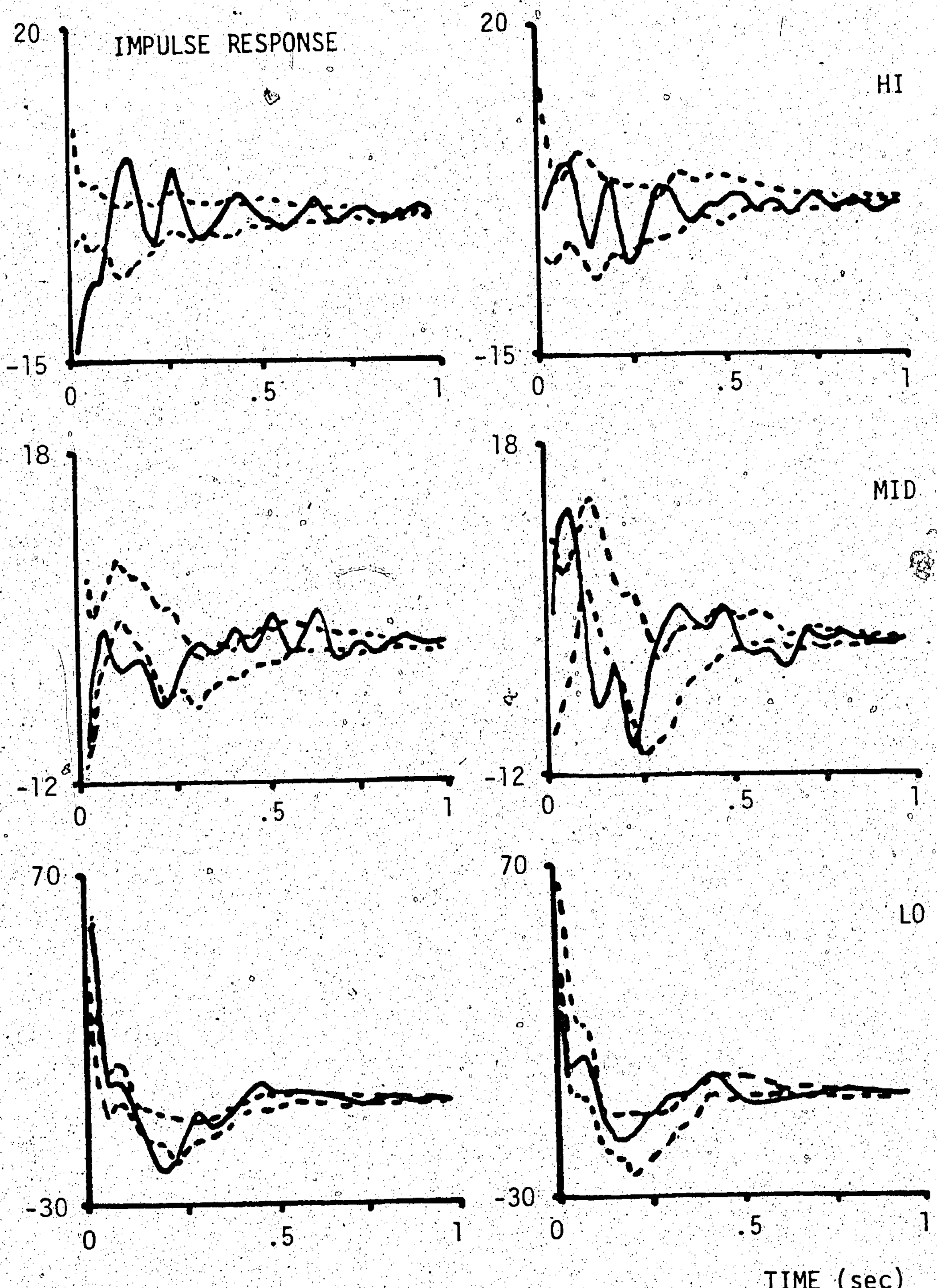


FIGURE N.2. Impulse responses of transfer functions from EO (left) and EC (right) tests of a male with a perioheral neuropathy (solid line) compared to 68% confidence limits of the corresponding averaged impulse responses from 27 normal subjects (broken lines).

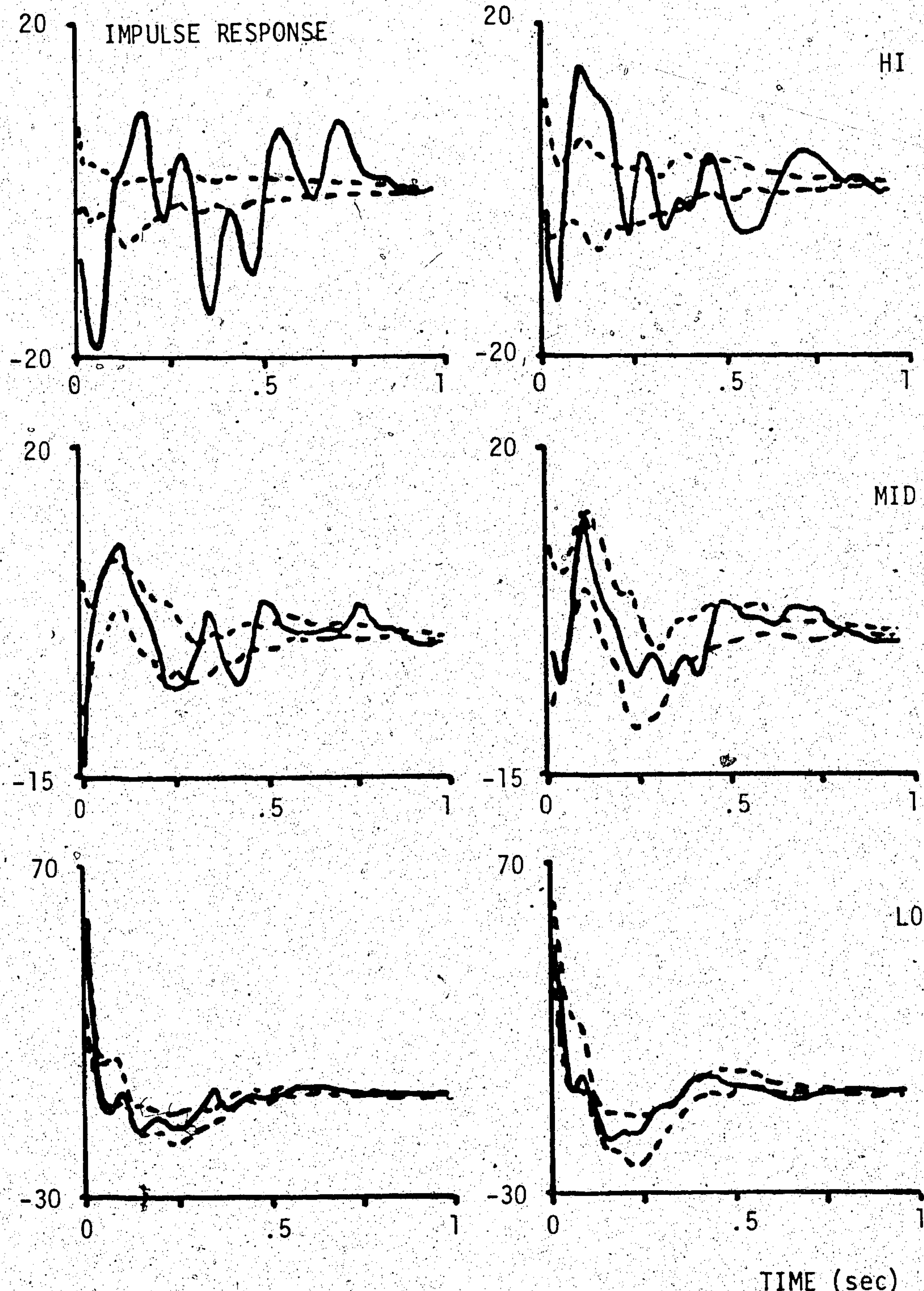


FIGURE N.3. Impulse responses of transfer functions from EO (left) and EC (right) tests of a male with spino-cortical tract disease (solid line) compared to 68% confidence limits of the corresponding averaged impulse responses from 27 normal subjects (broken lines).

APPENDIX O

Results of Sagittal and Lateral Sway Tests

The following graphs serve to compare processed data obtained from posturographic tests performed in two orthogonal planes of the body. Figure O.1 compares the power spectra. Figure O.2 shows the coherences derived from the two types of movement. These graphs are drawn to the same scales to demonstrate the difference in magnitudes.

The scales of the following graphs just fit the curves to compare the variances of the results from sagittal and lateral sway tests. Figure O.3 and Figure O.4 use EO test results in the form of real and imaginary transfer function components respectively. The differences between EO and EC real components for the two planes are depicted in Figure O.5.

Figure O.6 compares the sagittal and lateral impulse responses obtained by averaging the test results. The curves of Figure O.7 are derived from the differences between EC and EO tests.

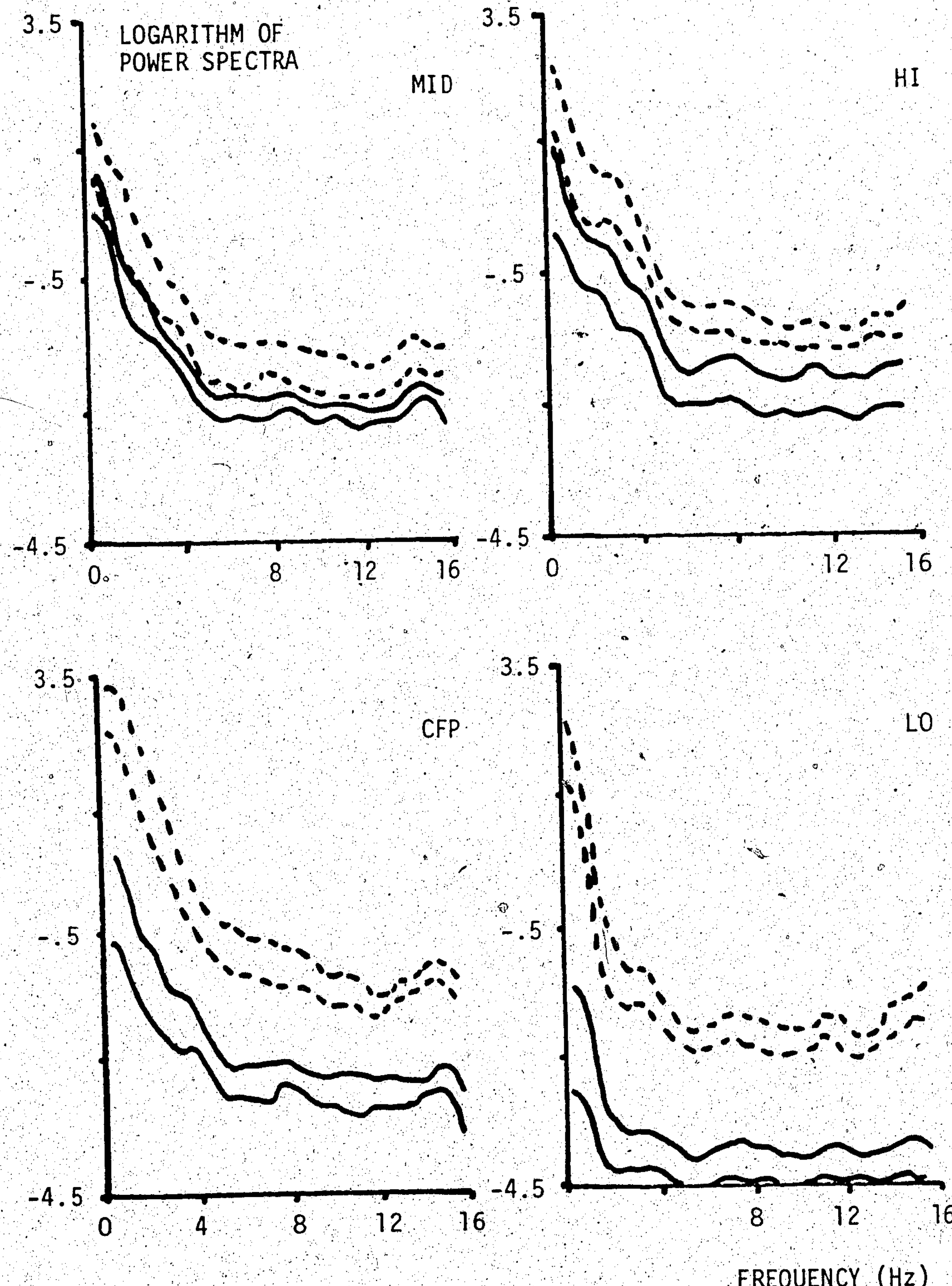


FIGURE 0.1. 68% confidence limits of averaged logarithms of power spectra of EO tests of lateral (solid lines) and sagittal (broken lines) body movement.

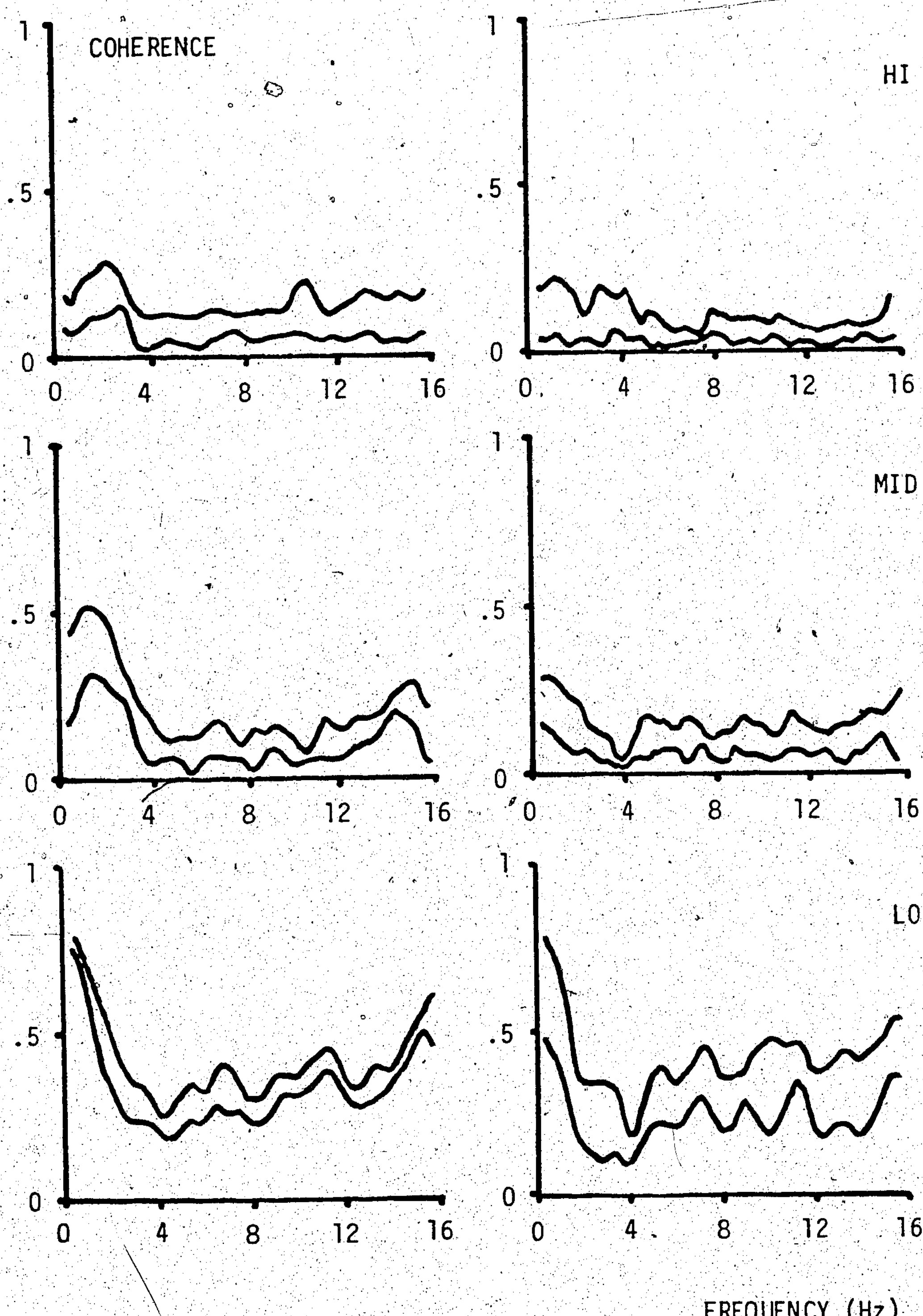


FIGURE 0.2. 68% confidence limits of averaged coherences of 6 EO tests of one normal male for sagittal (left) and lateral (right) body movement.

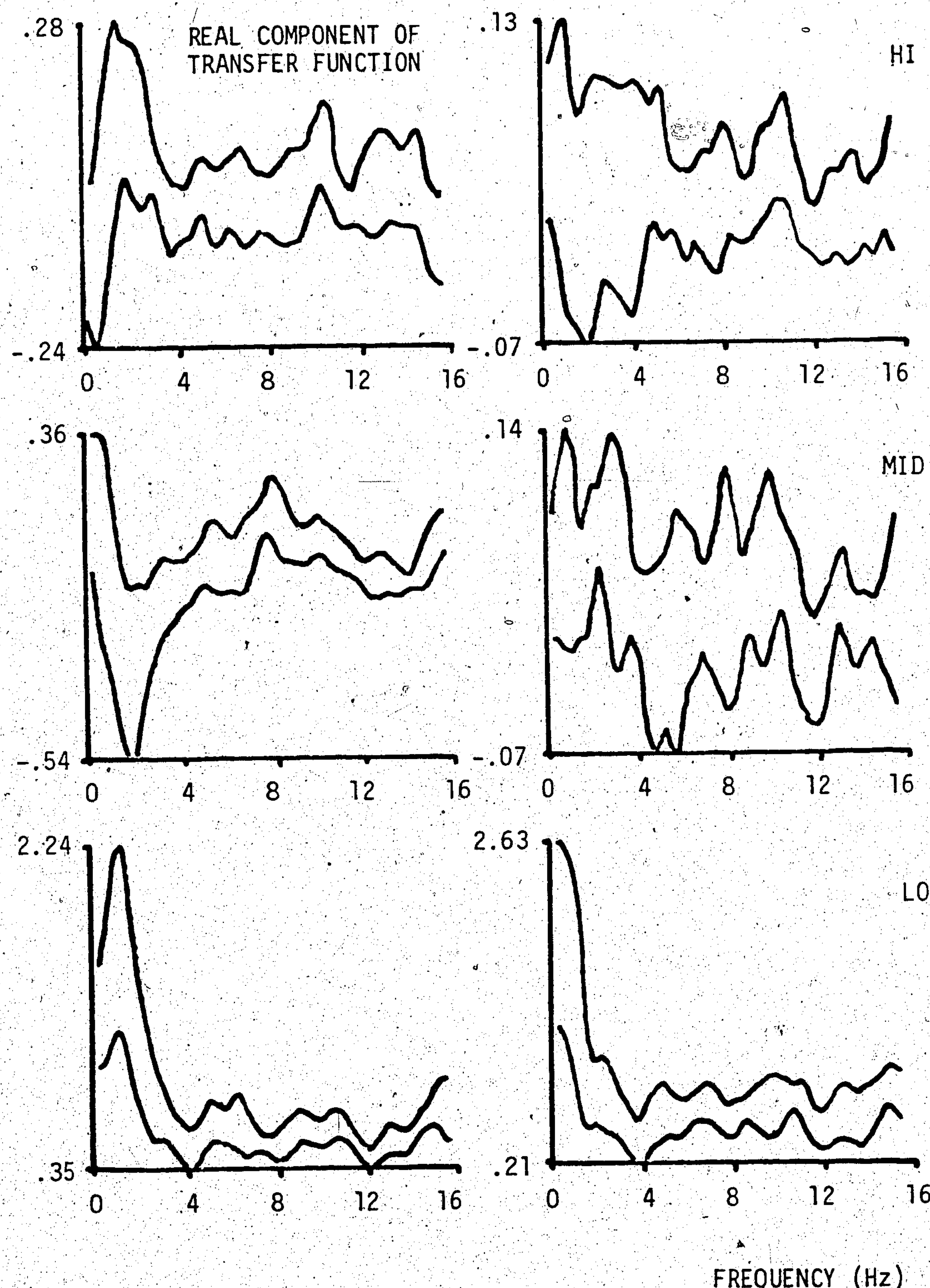


FIGURE 0.3: 68% confidence limits of averaged real components of transfer functions from 6 EO tests of one normal male for sagittal (left) and lateral (right) body movement.

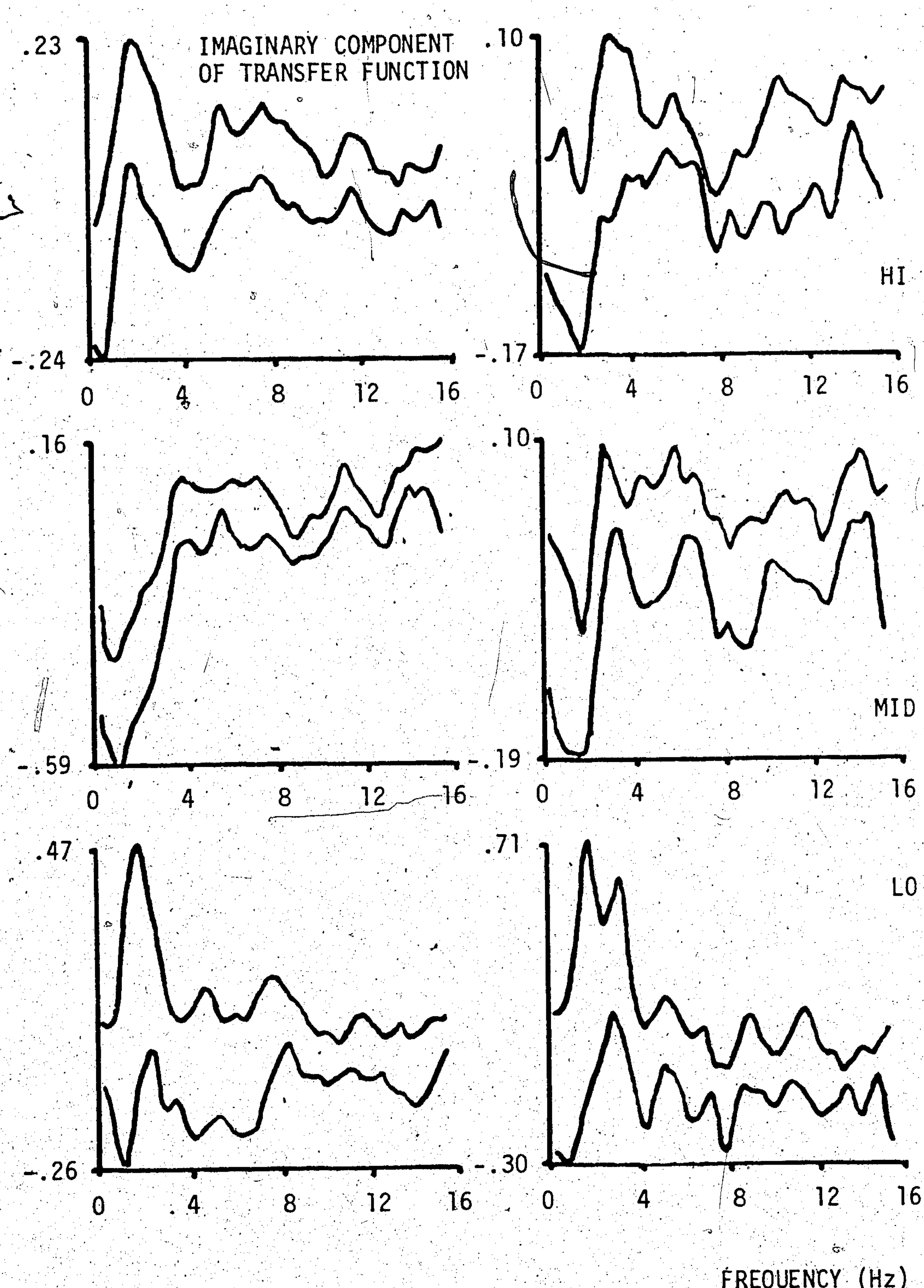


FIGURE 0.4. 68% confidence limits of averaged imaginary components of transfer functions from 6 EO tests of one normal male for sagittal (left) and lateral (right) body movement.

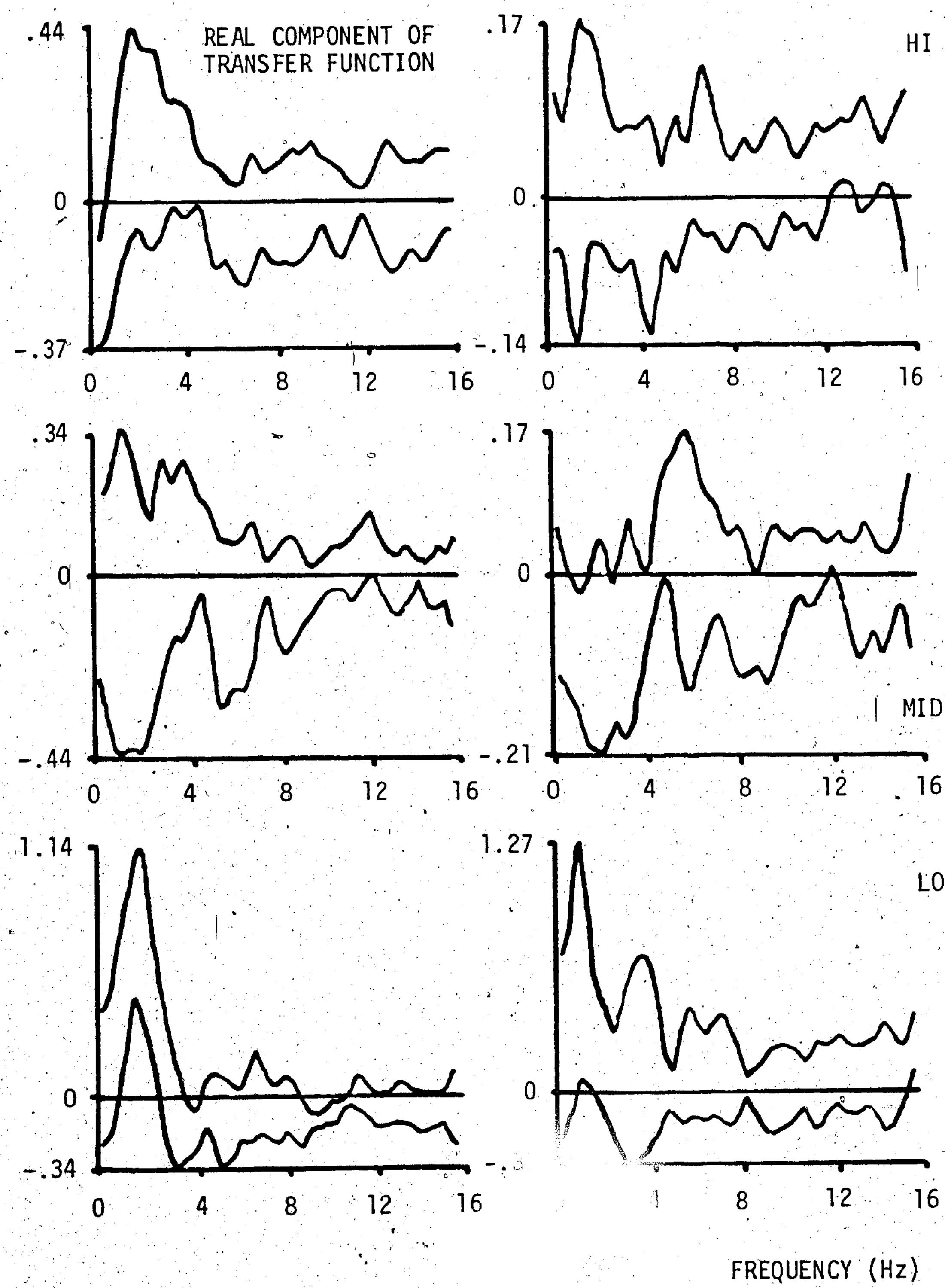


FIGURE 0.5. 68% confidence limits of averaged differences between real components of transfer functions from 6 EO and EC tests of a normal male for sagittal (left) and lateral (right) sway.

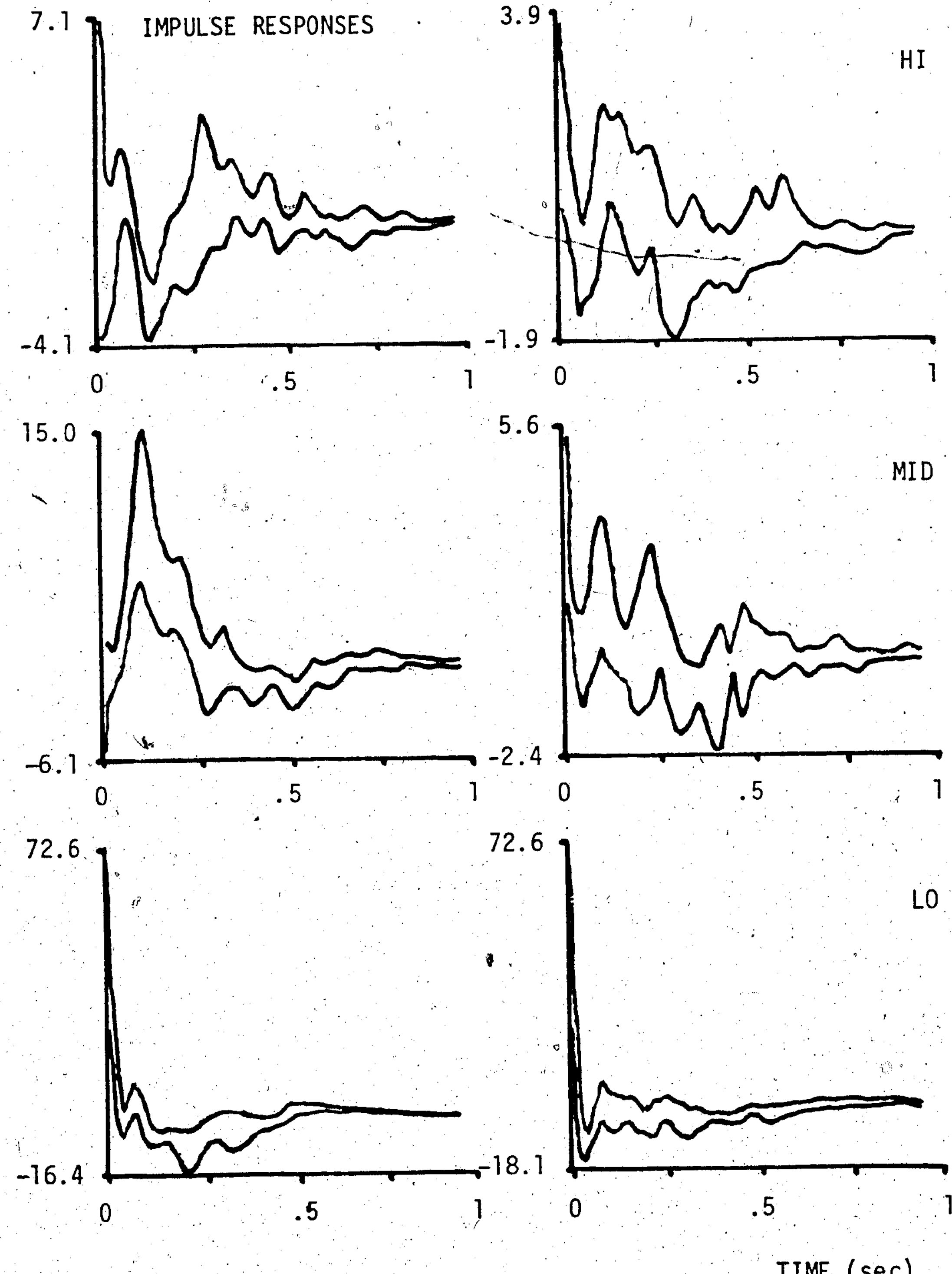


FIGURE 0.6. 68% confidence limits of averaged impulse responses of transfer functions from 6 EO tests of a normal male for sagittal (left) and lateral (right) body movement.

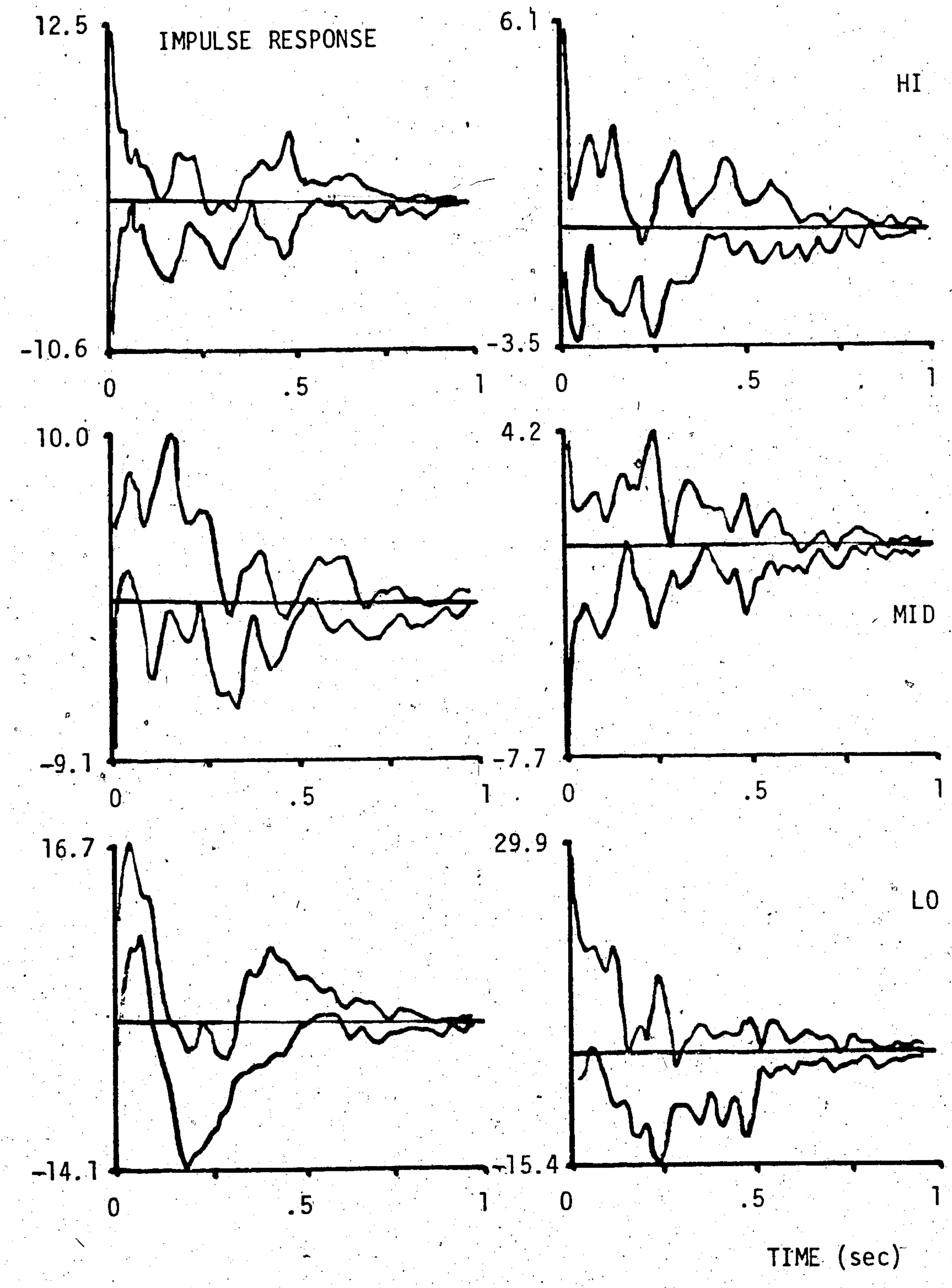


FIGURE O.7. 68% confidence limits of averaged differences between impulse responses of transfer functions from 6 EC and EO tests for sagittal (left) and lateral (right) body sway.