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PARAMETRIC STUDY FOR CLINICAL EVALUATION OF POSTURAL STABILITY

By

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ABSTRACT

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Balance evaluation parameters were identified which were sensitive to postural instabilities, characterized the subjects' postural responses and served as useful quantitative measures for comparing subjects. Force platform data were collected for normative subjects and one patient during quiet stance in various conditions. Comparisons were conducted between the subject and norms for the following time domain parameters: radial average displacement, radial velocity, ground reaction torque; and the frequency domain parameters: Xcop power, Ycop power, radial power, radial velocity power and ground reaction torque power. Peak frequency and amplitude were used as characteristic measures in the frequency domain. A definition of sensitivity for balance parameters was proposed. Based on this definition, radial average displacement was less sensitive than velocity and ground reaction torque. In the frequency domain, the parameters considered contained valuable information about stability and the type of control strategies used for maintenance of balance. The patient peak frequency was larger than norm for all but one condition.

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CHAPTER I
INTRODUCTION

Postural balance, the body's ability to remain upright, is an essential aspect of everyday life. Often taken for granted, the complex and efficient system that maintains balance also makes possible the successful completion of daily activities such as standing, walking, running, squatting, etc. Without an adequate postural control system the central nervous system (CNS) would not be able to assess movements of the body, or of the surroundings, and perform the necessary muscle response to prevent falling. Thus, if a deficiency in any aspect of the postural control system develops, it is necessary to have a reliable and efficient way to assess balance and decide on appropriate treatments. Development of such an evaluation technique requires knowledge of the working mechanisms of the postural control system.

Postural balance, with assistance from the musculoskeletal system, is controlled by three different mechanisms: the visual, vestibular, and proprioceptive systems. The loss of input from one or more of these systems requires the postural control system to depend on the available input to maintain balance. Although each system has a specialized control of balance, there does exist some overlap in function between them.

Nerve impulses sent to the CNS regarding head position and movement strongly influence the muscle activity involved in maintaining balance. The visual system utilizes surrounding objects as a reference frame to establish a known vertical and provides information about the orientation of the head with respect to gravity. However, the body is also equipped with other mechanisms to sense head position.

The inner ear, in addition to its role in hearing, provides information about the positioning and movement of the head. This information is then utilized to determine what movements may be necessary to maintain balance. Signals sent from the inner ear are also involved in controlling extrinsic eye muscles so that the eye may remain fixed on one object despite movements of the head. Specialized receptor cells which are responsible for this are located within the utricle, saccule, and the ampulae of the semicircular ducts and make up the vestibular apparatus. This system also uses mechanisms within the inner ear to sense both linear and angular accelerations.

In addition to the importance of knowing head position with respect to gravity, it is also essential to know how the whole body is positioned. The proprioceptive system uses receptors in the tissues and joints and pressure receptors in the feet in conjunction with velocity and position sensitive muscle spindles to provide information to the balance system about body orientation and posture. Receptors within the neck provide information about the head relative to the rest of the body. This information is needed since the vestibular system senses head position only. Knowledge of where the head is with respect to the rest of the body will aid in the determination of whether body equilibrium is in a threatening position, as in the case where the whole body is tilted and on one foot but the neck is straight, or when equilibrium is not threatened as when the body is upright and only the head is tilted. Input from each of these systems is used by the central nervous system to signal the necessary muscle response to maintain balance.

With this complicated and integrated system, what happens when one portion of the system does not provide the necessary information and the body is unable to adapt? What can be done to evaluate such problems and determine treatment? As early as 1851, physicians began to take note of patients' observed balance difficulties and investigate the origin of such problems. Studies were begun to obtain an in depth understanding of how balance is controlled and how specific diseases or injuries may impair the ability to maintain balance.

Early evaluations were based solely on observations and comparison of patients with normative subjects performing various tasks. However, with today's advancements in technology, a variety of measurement parameters have been developed, and controversy exists concerning which measures should be used as well as how to interpret them. Some researchers have used external perturbations such as force platform movements to induce a state of imbalance. Others have tried to measure a persons natural sway by having the patient stand in various positions on a force platform. The controversy between which method is more accurate and which measurements should be taken emphasizes the need for more research and consistency in this area of biomechanics. The purpose of this thesis is to explore differences between normal and balance impaired individuals, determine the most informative and reliable means of measuring postural steadiness and develop a feasible testing protocol for use in a clinical setting.

CHAPTER II

LITERATURE REVIEW

For many years, researchers and physicians have been searching for an effective means of assessing postural balance and stability. Surprisingly, methods first used in 1851 by Romberg are still in use today in conjunction with many new ones. Romberg conducted subjective assessments of patients' balance through observation of sway and standing ability. At that time, comparison to "normal" stance was the only means of measuring sway. Thus, the physician relied purely on observation. Romberg used this method of sway observation and expanded upon it by conducting balance evaluations of patients standing with both eyes open and closed. He observed a dramatic increase in postural sway with eyes closed for patients with disease impaired proprioception [23, 113]. Later, this method of balance assessment became standard procedure and was known as the "Romberg Test". This was the beginning of research on postural balance and sparked a great deal of interest in the specific mechanisms of the postural control system of the body.

Since the advancements in balance evaluations by Romberg, many devices and techniques have been developed. These vary from standing on one foot to the use of moving walls and force platforms. However, knowledge of the postural control system is essential before the development of an effective evaluation protocol may begin. Research shows that balance is maintained through the evaluation of afferent information from several inputs, and investigations have been conducted to identify the specific roles played by each input and how each one contributes to the maintenance of balance [12, 21, 55, 88].

2.1 Visual, Vestibular, and Proprioceptive Inputs

Maintenance of postural balance requires three types of input: visual, vestibular, and proprioceptive. Although there is some overlap, Nashner noted in 1970 that the three sensory control loops are each specialized to work within a specific domain of amplitudes and frequencies, indicating this in not a totally redundant system [16, 21]. As will be discussed in section 2.2, the comparison of sway power spectra from patients and normals and the separate determination of the efficiency of the three control loops over a range of frequencies for each type of input is useful in the identification of problem areas and work has been done to define each system's role in postural control.

Proprioceptive Input - The proprioceptive system includes receptors in the tissues of joints, connective tissue, muscles, and cutaneous tissues with elements such as the pressure receptors of the feet and the velocity and position sensitive muscle spindles [50]. These sensors send information to the CNS regarding the location and orientation of each part of the body with respect to a reference frame. Results of experiments investigating afferent input from proprioceptive receptors show that spindle afferents have their working range above 1 Hz. Pressure, joint, and skin receptors are more important in the low frequency range [21].

Visual Input - Vision also plays an important role in the postural control system. Along with the vestibular system, it helps to maintain the head and body's orientation with respect to gravity [50].

Results from experiments with visual perturbations indicated that visual stabilization of posture operates mainly in the low frequency range at and below 0.1 Hz [16, 21].

Vestibular Input - The vestibular system is essential for maintaining head orientation in space with respect to gravity [50]. This system consists of both angular (semicircular canals) and linear (utricular otoliths) acceleration sensors [87]. The working range of the vestibular system has been investigated by many researchers. Nashner [87] predicted the semicircular canals best sense the rate of sway above 0.1 Hz while the otoliths sense sway below this frequency [21].

DeWit [14] investigated postural steadiness in normals, subjects without vestibular function, subjects with weak proprioception but normal vestibular input, and persons with abnormal labyrinth function, such as Menier-Syndrome. The results revealed two types of oscillations: one originating from the proprioceptive muscular system and the other from the vestibular system. The oscillation from the proprioceptive system was characterized by high frequency and small amplitudes. While the vestibular driven oscillation displayed large amplitudes and low frequency.

2.2 Postural Steadiness and Measurement Parameters

Since the introduction of the Romberg test in 1851, evaluations of postural balance have been a routine part of neurological exams. Early studies were primarily based on measurement of static equilibrium during quiet standing. This method of evaluation was termed postural steadiness to avoid confusion with postural stability which consists of responses to perturbations or dynamic movements [98]. Evaluations of steadiness are often conducted to investigate possible sensory impairments by quantifying the amount of postural sway, under the assumption that increased postural sway is associated with increased effort to maintain balance. Simple external disturbances, such as closing of the eyes or assuming an atypical stance, are often applied to decrease the amount of sensory input. However, no disturbances such as force platform movements are applied. Postural sway may be characterized in terms of the following [70]:

1. Ground reaction forces
2. Displacement of the center of pressure (COP)
3. Displacement of the body's center of gravity (CG)
4. Motion of body segments or joints
5. EMG activity in various muscle groups

Unlike early evaluations, such as the Romberg test, which were based purely on observation, the use of modern technology provides these measures of sway which are quantitative and offer greater accuracy.

1. Ground Reaction Forces - Balance is a state of equilibrium where there is no net external force on the body. Therefore, it seems logical to focus on external forces to better understand postural balance and locate problem areas. As the subject stands, forces are exerted onto the floor due mainly to the body weight. In equilibrium, an equal and opposite reaction force is applied to the feet by the floor producing a zero net external force. The three components of these reaction forces consist of two in the plane of the floor and one in the vertical direction. The resultant of these reaction components is known as the resultant ground reaction force.

Various studies have been conducted to investigate the forces involved in maintenance of balance [67]. Early experiments used electric scales to measure vertical reaction forces and weight shifting during stance with eyes open and eyes closed [39, 79]. Thomas and Whitney (1949) employed electric scales to study the A/P movements during normal standing in men [39]. A similar method was also used by Henriksson, et al [39] where two electric scales using strain gauges measured the difference in voltage between the two scales and displayed the difference in pressure exerted by the right and the left foot. However, more commonly used measures today are ground reaction forces obtained through the use of force platforms. Originally the signals were recorded on strip recorders and quantified manually. Thus, ground reaction forces were measured in either the anterior/posterior (A/P), the medial/lateral (M/L), or the vertical directions. Now, force platforms are used to measure 3 orthogonal ground reaction forces and moments about those axes. Some researchers have used commercially available force platforms while others have constructed their own. Specifications for building a vertical force platform to be used as a clinical tool have been presented in the literature [6].

Although some studies have used two adjacent force platforms [98] or scales to measure weight distribution and shifting, most studies are conducted with both feet on only one force platform. Stribley, et al [113] used a force platform to measure ground reaction forces and calculated for the last 50 seconds of a 75 second trial the average of the integrated force signal, which he termed the "steadiness score". The subjects were evaluated in various stances with both eyes open and closed. By steadiness score comparison, results were obtained which indicated no significant difference between men and women or between one-legged stance on the dominant or non-dominant side.

Measurement of the ground reaction forces not only provides information about the loads applied to the body, but it also allows evaluation of the type of balance strategy used based on the sway patterns the subject exhibits. Woolley, et al [134] measured mean A/P shear force and interpreted them as measures of hip sway. In a similar analogy, Barin [4] measured peak to peak horizontal force and normalized by 111.25 N (a theoretical maximum). This measure was used to determine whether hip or ankle strategy was mostly used. A 100% corresponded to complete hip strategy while a 0% corresponded to complete ankle strategy. Balance strategies will be discussed in more detail in section 2.3 on postural stability.

2. Displacement of the Center of Pressure of the Foot - To gain additional information on how the ground reaction forces interact with the feet in maintenance of balance, various attempts have been made at studying the distribution of pressure, defined as force per unit area, on the surface of the foot. Elftman [26] in 1934 expanded on others' past attempts to study the pressure on the feet when walking. He designed a device to record the momentary distribution of pressure over the whole foot during normal gait. Subjects walked across a rubber mat which had pyramidal projections which pressed against a heavy glass plate. The area of contact of the pyramid with the glass increased in proportion to the pressure exerted by the foot. Data were recorded from below the glass plate with a 72 Hz. moving picture camera. This technique provided valuable information about weight transference and pressure distribution, however, more quantitative measures were necessary for clinical applications.

The development of force platforms provided the necessary technology and accuracy for development of new sway parameters to describe the interaction between the ground reaction forces and the foot. The most popular sway parameter, termed the center of pressure (COP), was defined as the point of application of the resultant ground reaction force. The COP may be defined according to two different philosophies: one with a vertical ground reaction torque and the other with the torque parallel to the resultant force [10, 108, 109, 110]. The more preferred method is the one with a vertical torque which is more conceivably possible. Although the parallel resultant force and torque method is mathematically correct, it is difficult to accept an analysis technique which says a person is creating an inconceivable torque by applying forces out of the plane of the force platform.

Research has shown that the COP estimates the movements of the CG which must remain above the support base to prevent falling [100, 111, 118]. Thus, when the COP is used, the entire activity of the postural control system is studied by observing how well the subject can maintain the COP within the foot support base. The COP is the most commonly used measurement of postural steadiness today with various parameters classified into two categories, time or frequency domain.

Time-Domain Measurements - The majority of COP measurements used in postural evaluations today fall under this category. Most commonly studied is the time trace of the COP. However, many different variations are also used. Goldie, Bach and

Evans [31] tabulated the force platform measures used to evaluate steadiness from 1972 to 1989. Among the many variations of evaluating the COP are: the mean COP position [37, 72, 81], average displacement from the mean COP [31, 56, 81, 90, 97, 98, 118, 134], RMS distance from the mean COP [97, 98], the mean velocity of the COP [63, 64, 97, 98, 134], the total excursion (or distance traveled) of the COP [13, 27, 31, 72, 81, 90, 98, 103, 104], the peak-to-peak range [98], the range of the COP (the maximum distance between any two points on the path) [38, 97, 104], and fractal dimension [97, 98]. Fractal dimension relates the number of points on the COP path with planar diameter and the total excursions of the COP. The displacement from the mean COP is sometimes expressed in polar coordinates as an average radial displacement [62, 81]. Soutas-Little, et al [109] investigated further parameters of postural steadiness and preferred the use of COP speed and ground reaction torque due to their sensitivity and easy comparison between conditions and subjects.

In addition to these, some area measures are also used [98]. Area of the COP path can be calculated as an enclosed area [98], circumscribed area, area based on mean distance [54, 63, 64] and 95% confidence ellipse area [37, 97]. In addition, some researchers have used enclosed area as a percentage of base-of-support area [3, 38, 40]. This parameter normalizes the data allowing comparison between conditions with different base-of-support areas and between different subjects.

Hasan, Robin and Shiavi [37] carried this idea even further by having subjects deliberately sway from the ankle a maximum amount in all directions, without stepping or falling, to obtain the individual's maximum COP which he calls the "functional base of support". This area is somewhat smaller than the base of support (defined by the area beneath and between the feet) because the intrinsic foot muscles prevent full body weight bearing by the toes or the back of the heel. The functional base of support was then used to normalize, instead of enclosed area of the feet. Riach and Starkes also measured a patient's maximum sway while maintaining postural stability in the A/P and M/L directions which they termed "stability limits" [137].

Bagchee, and Bhattacharya [3] devised a method of superimposing a subjects foot prints and COP excursions to evaluate one's risk of falls. This method was also used by Riach and Starkes [137] Since the COP must remain within the stability boundary, the area outlined by the feet, those whose COP trace was near the edge of the boundary were at a greater risk of fall and may have decreased postural stability. The results were presented as a possible means of identifying decreased postural stability due to chemical exposure and neurological diseases.

Posturography is the evaluation of sway during upright stance. In posturography, two complementary recording methods are often used. The first, termed a stabilogram (STG), consists of the time series of the center of pressure movements in the A/P and M/L directions. While the second method is termed a statokinesigram (SKG), or "spot stabilogram", and is the total excursion of a subject during a time interval [14, 16, 23, 36, 40, 51, 52, 54, 62, 64, 113, 115]. Terekhov [117] analyzed stabilograms by measuring duration of one oscillation, mean amplitude of oscillation and maximum amplitude of oscillation. Kapteyn and deWit [52] and Kapteyn [54] used SKG results such as area and placement of the excursion within the base of support along with STG frequency analysis in posturography evaluations.

Jeon [48] conducted an experiment to evaluate the sensitivity of the COP measurements during quiet stance and found that the COP displacements were affected by both respiration and eye condition.

The eyes closed to eyes open ratio is often used as in the case of the Romberg Quotient, an index assigned according to how well a patient stands with feet together and eyes closed as compared with eyes open [73, 116]. This concept has been applied to the total COP excursion [114] or the mean amplitude of oscillation [117]. Njiokikjien and van Parys [90] used many of the above COP parameters and performed a comparison with the Romberg Quotient. Results showed that the Romberg Quotient is an easily calculated and reliable parameter to identify proprioception impairments.

Much disagreement has occurred over the length of test duration. Sway test duration in the literature range from a few seconds to a few minutes. However, the typical test duration is within 20-80 seconds. Woolley, et al [134] investigated several COP measures and how they varied over 6 test durations ranging from 10-60 seconds. The data indicated no significant change in the COP parameters used over the different durations. Therefore a 10 second test would give the same results as a 60 second test reducing the possibility of fatigue to the subject. Some researchers have used only central portions of a test to eliminate edge effects and instability as the subject steps onto the plate and during the early portion of the trial [76, 113]. Others have used the entire files for analysis. Hasan, Lichtenstein and Shiavi [36] noticed that a loss of balance by a subject caused 0-3% increase in the area of the COP during double stance. During single stance, area increased 16-38% and velocity measures by up to 10% with loss of balance. Therefore, he developed a method to locate and separate out a loss of balance from the COP plot to be analyzed. A loss of balance was defined as a touchdown of the non-supporting leg, during one-legged stance, in an attempt to regain balance. They also quantified the effects of loss of balance in different conditions. In a review of the 5th International Symposium on Posturography in Amsterdam (1979), Kapteyn, et al [51] expressed the need to standardize platform stabilometry parameters such as test duration, stance, and other methodology aspects to allow comparison between subjects and researchers.

Frequency-Domain Measures - The other type of COP measures, which are very useful but more difficult to read, fall into the frequency-domain category. In one type, a Fast Fourier Transform (FFT) is applied to the COP displacement measurements obtained from the force platform. This provides the frequency spectrum and reveals the total amount of energy in the A/P and the M/L spectra [29, 35, 107]. From the frequency spectrum of the COP movements, some researchers have analyzed peak frequency [35, 73], mean frequency [98], and frequency at which the magnitude decreases 3 decibels (dB) from the maximum frequency [35], while others have conducted qualitative characterization of the spectrum [73], such as identification of the wave forms present in force measurements. The following frequency domain measurements were also calculated for the frequency range of 0.147 - 10.010 Hz [97, 98]: total power, 50% and 95% power frequency [97, 98], centroidal frequency, and frequency dispersion. Mean, standard deviation, and coefficient of variation (CV) have all been used as standard calculations for frequency domain parameters. Soames, Atha and Harding [107] measured the A/P and the M/L components of sway and applied FFT to 4 consecutive minutes of feet together stance, analyzing each minute separately to note changes in the distribution of energy over time.

For clinical evaluations of postural imbalance, determination of the efficiency of each of the three sensory control loops over a range of frequencies may be useful. This may be done by comparing the power spectra of sway in patients with isolated lesions of the visual, the vestibular, or the somatosensory proprioceptive systems with the power spectra of normals [16].

Results by Dichgans, et al [16] showed that in atactic patients, the postural instability frequently exhibits an amplitude peak at about 0.6 Hz in the Fourier power spectrum of the COP. Patients with a cerebellar disease exhibit a peak at 2.5-3 Hz. Thus, the frequency analysis of the COP excursions when compared with normative data can be useful in diagnosis and treatment of patients.

Soutas-Little, et al.[110] took a different approach by conducting an FFT of the torque data. A significant difference was found between the power frequency spectrum of the norm and a Traumatic Brain Injured patient. The power of the norm decayed between 4-5 Hz, while the patient's power was beyond that of the norm, with additional peaks in the 5-10 Hz frequency range.

3. Displacement of the body's Center of Gravity - In addition to the displacement of the COP, center of gravity (CG) displacement has also been studied [123]. A common source of error in evaluation of force platform data is confusion of the COP with the CG. Whereas this confusion might not be serious in a purely static case, it is crucial during accelerated motion [50]. Winter [131] defined the difference between the COP and CG and used an inverted pendulum model to show that "the range of the COP must be somewhat greater than that of the CG". Results of experiments by several researchers have indicated that COP displacement is greater in magnitude and frequency than that of the CG, and that the stabilogram gives an exaggerated impression of the movements of the center of gravity [100, 111, 118].

Displacements of body segments have also been measured and used in conjunction with CG displacements as an indicator of postural sway [93]. Black, et al [7] measured A/P sway of the body center of mass by attaching a potentiometer to the subjects hips with a belt and a system of light rods. In some patients, a second potentiometer was attached to the subject's back at shoulder height to measure the contributions of hip as well as ankle joint motion to the A/P sway.

Tokita, et al [124] measured sway of the head and the CG to compare normals with various categories of patients. The patients, who had different types of impaired sensory systems due to injury or disease, showed definite differences in frequency and direction of sway. Murray, Seireg and Scholz [80] measured vertical supportive forces during squatting. He differentiated between changes in the applied force and changes in the center of gravity of the body. He also investigated the difference between the excursions of the line of gravity and the action line of the vertical supportive force. Hasan, Robin and

Shiavi [37] used fifteen targets to collect kinematic data and force platform data to develop a relationship between COP and the CG data.

4. Motion of Body Segments and Joints - Some researchers felt that kinematic data was essential to evaluate postural steadiness. Angular displacement, velocity, and acceleration of various body segments are all parameters that have been considered. Various devices such as displacement transducers and accelerometers can be used to measure these parameters.

Sway is often characterized by motion or displacement of body segments. Displacement transducers have been attached to the pelvis and hip [5] or to the sacrum and greater trochanter [23, 28] to measure postural movements. Body segment positions in the A/P plane were measured with an opto-electrical movement analyzer and LED targets [83]. Mauritz, Dichgans and Hufschmidta [73] used an electronic goniometer fastened to a belt around the subject's waist or head. Yoshizawa, et al [135, 136] measured head position on the horizontal plane in an on-line real-time fashion by using an ultrasonic distance sensor system. Lord, Clark and Webster [66] used a device consisting of a rod attached at the subject's waist with a pen at the end which recorded movements on graph paper. Another method was developed by Hirashawa [40] to track movement of the gravitational force on human subjects. He named this electro-gravitograph (EGG) and showed quantifiable differences between normal subjects and patients with chronic organophosphate intoxication. Others [30, 83] have used a sensing potentiometer to detect body oscillations due to force plate perturbations such as lower leg rotations [124].

In addition to measurement of body displacements, the calculation of velocity (or speed) and acceleration of the body segments has also provided useful information. Nashner, Schupert and Horak [83] measured velocity of A/P head rotation with an angular rate sensor attached to a helmet. Fernie and Holliday [28] used the motion data, he obtained with transducers, to calculate the mean speed of sway (the length of the locus of sway in unit of time) and the range of movement in the saggital and coronal planes. Accelerometers attached to the subject have been used to measure A/P accelerations of the head, trunk, and waist [118, 124]. Woloszko and Jaeger [132] measured both body angle and acceleration during quiet stance.

5. EMG Activity in Various Muscle Groups - EMG muscle responses are often used in postural evaluations. Many experiments have been conducted to present evidence that during stance, functionally related postural muscles in the legs are activated according to fixed patterns [2, 86]. Others revealed specific movement strategies used in response to perturbations [16, 19, 20, 22, 42, 44, 56, 77, 78, 84, 92, 132]. The organizing principle of mechanical body motion has been extensively studied from the perspectives of biomechanical, neurological sciences, and control systems analysis. Coordination patterns may be found in EMG recordings and the organization of such EMG activities of leg muscles during rapid postural adjustments have been studied by Nashner [83] and others [85]. These results supported hypotheses of hierarchically organized groups of muscles.

2.3 Postural Stability

As discussed earlier, postural steadiness was a measurement of static equilibrium during quiet standing. With respect to balance, equilibrium is defined as a state of posture where there is zero resultant torque or force on the body. Thus, the body is at rest or in uniform motion [50]. However, balance may also be evaluated in terms of postural stability which is the body's response to external perturbations of the postural control system [50]. Postural stability can be either static or dynamic. Static stability is when the body's equilibrium is restored by a force or torque. Dynamic stability means that equilibrium tends to be restored over time. There exists a damping of the velocities such that oscillations about the equilibrium are damped [50]. One's degree of stability is a measure of the body's ability to counteract disturbances to the postural control system such as force platform movements or other negative feedback.

Stability requires that small perturbations give rise only to a small deviation away from equilibrium [50]. According to Black, et al [7], two main components are required for the maintenance of equilibrium: (1) accurate sensory and perceptual information about the body's position, orientation, and movements with respect to external references, and (2) accurate motor system command to correct or maintain the body's position with respect to the earth's vertical, placing the body center of mass over the support base provided by the feet.

Combining head and body movements with muscle response to sway, Nashner, Shupert and Horak [83] investigated head-trunk movement coordination in standing posture. Nashner, Shupert and Horak [89] and Black, et al [7] observed that when exposed to saggital perturbations, a subject may regain equilibrium by two different strategies. The first of these is the "ankle strategy" which, in response to slow translations of the support surface, begins with muscle activity in ankle muscles and then ascends the body to rotate the body around the ankle joint. The second strategy, called "hip strategy", is a response to platform rotations or standing on a narrow beam. This strategy is more complex and involves muscle activity of the body and neck causing hip translations and stable head position [96]. These strategies predict that primary motion occurs at the ankle or hip and consider the knee motionless, ignoring movements of the upper body. However, examination of kinematic data of the responses to force platform perturbations show the strategies to be more complex, involving both knee and upper body motion, multisegmental interactions and inertial influences on the system [110].

Postural steadiness, which quantifies an output of the postural control system, fails to characterize the inputs which cause the sway to occur. Thus, no predictions of response to various inputs can be made. However, with postural stability, the known input allows an input/output model of the postural control system to be identified. Input perturbations can include surface displacements, visual surrounding movements, or the application of other postural perturbations. In most experiments, muscle response, angular displacement of body segments, and COP displacements were measured before, during, and after the perturbations were applied. This was done in hope of defining more clearly the functional role of each sensory input.

A wide variety of different perturbations have been used by various investigators. Maki [70] summarized the perturbation parameters used by investigators from 1957-1984. The objective is to eliminate only one form of sensory input at a time and compare balance with and with out this input.

Vestibular disturbances - Vestibular feedback is essential for orienting the head in space and with respect to gravity [50]. To investigate vestibular contributions to postural stability, static vestibular input is modified by tilting the head forward, backward, or to either side prior to the force platform perturbations. DeWit [14] altered vestibular input by having the subjects shake their head for 10 seconds with eyes closed to stimulate the labyrinth. Still another type of vestibular perturbation was used by Tokita, et al [123] who measured changes in soleus muscle EMG and sway of the CG when galvanic stimulation was applied to the labyrinth of one ear.

Proprioceptive disturbances - Postural perturbations can be applied easily by moving a platform on which a subject stands. This is the most commonly used form of de stabilizing input. The force platform can be translated horizontally in the A/P [77, 98], or in the M/L directions [77]. It may also be rotated about an axis coinciding with the ankle joint axis of rotation [2, 4, 7, 77]. Force plate perturbations are driven by several wave forms. Some are continuous sinusoidal signals, while others are random or pseudo random. Still other force platforms are driven by ramped and pulse signals [30, 88]. A combined input of horizontal translation and ankle rotation has been used by Nashner [87, 88] to evaluate changes in proprioceptive input. This two degree of freedom force platform was translated horizontally then rotated the amount of degrees needed to keep the ankle in a fixed neutral standing position [21, 88]. This eliminated the proprioceptive input from the ankle joint stretch receptors. Ischemic blocking of leg afferents with pneumatic cuffs around each thigh have also been used to decrease proprioceptive input [1, 74]. Nashner, Woollacott and Tuma [84] combined the fixed-ankle angle with horizontal translation, reciprocal vertical displacement, and synchronous vertical displacement inputs.

Most studies in postural control using support rotation and translation or body free fall induce ankle angular velocities which aren't encountered in daily activities. Therefore, Woloszko and Jaeger [132] used perturbation generating step transients to alter the body center of mass without generating high angular velocities at the ankle. He restrained the body with the use of an orthosis which allowed A/P ankle movements only, thus creating a single-link system. Woloszko also dropped weights from the orthosis at random times and measured the muscular response.

Another type of proprioceptive disturbance is foot positioning and spacing. Stribley, et al [113] reported that posture, especially in the M/L direction, was more stable with a large distance between the feet. Various stances, such as tandem (heel to toe), feet apart, feet together, and one-legged stance have been used. Goldie, Evans and Bach [32] developed a reliable and feasible method for testing steadiness in one-legged stance where the non-affected leg was used as a control. This test has potential use in evaluations of unilateral injuries or disorders.

Visual disturbances - Postural response to visual, rather than force platform inputs has also been investigated. Visual disturbances have been applied in the form of closing of the eyes, displaying vertical or horizontal black/white alternating stripes [104], moving rooms and walls, moving lines or patterns on a projection screen, false vertical domes, or a moving visual surround [7]. Also used in studies was the varying of surrounding conditions such as normal lighting, reduced lighting, normal visual acuity, and reduced visual acuity. This provides information on how different environments affect COP excursions. Data support the possibility that having adequate visual stimuli could be an heuristic approach to optimize the role of vision in reducing risk of falling for the elderly. It stresses the need for properly corrected vision and high contrast environments. Bles and deWit [8] used a room which tilted +/- 10^{o} in the M/L direction with an enclosed fixed position force platform to apply visual perturbations. Another moving room which moved horizontally in the A/P direction has been used by Lee and Lishman [60]. Visual stimulation may be induced by simulating an optic flow field with moving walls and moving tunnels on a projection screen [128]. Taguchi [115] used moving stripes displayed on a 270^{o} projection screen which encompassed the stationary force platform. This optokinetic stimulation was applied at various angular frequencies and postural response in terms of A/P and M/L sway was recorded.

Another type of visual input was the "roll vection stimulus" used by Dichgans, et al [16]. This was a visual field which rotated in front of the subject and consisted of a half spherical dome suspended above the force platform. The inner surface of the dome was covered by randomly distributed colored circular patches of different sizes. The dome was rotated about its axis to induce postural sway. DeWit [15] used a vertical light bar in a dark room placed in front of the subject as de stabilizing input. This light bar was rotated +/- 10 degrees to induce sway.

Other types of input create distorted visual feed-back. Yoshizawa, et al [135] used visual feedback where the subject wore the 3D-VD and stood in front of a screen on which a vertical line is drawn. The images of the line, taken by two CCD cameras on the subject's head, are artificially altered by the personal computer and transferred to the 3D-VD. Thus, the subject can watch altered stereoscopic image depending on head motion. This device is called the "visual conflict dome" and has been used by Lehmann, et al [62] and Ingersoll and Armstrong [45]. This dome gives a visual feed-back of vertical lines that changes with the movements of the head. Thus a false sense of vertical is created for the subject. Barin [4] used a visual scene which surrounded the subject on the force platform. The scene rotated according to the subject's sway in order to provide false visual input.

Hlavacka and Litvinenkova [41, 65] have investigated the visual feed-back gain influence on postural control by use of a force platform and monitor. The subject's COP trajectories were displayed on a screen in front of the subject. This allowed the subject to correct his/her balance accordingly and visualize the change. Results showed that in visual feed-back conditions, the postural control, gauged by stabilogram area and velocity, was more stable. The postural movements were quicker (i.e., increased average velocity), and the stabilogram area was smaller.

Other disturbances - In addition to vestibular, proprioceptive, and visual perturbations, other postural disturbances were investigated. Chandler, Duncan and Studenski [11] designed a wall mounted pulley system from which a weight attached to a belt around the subjects waist is dropped along the pulley track providing a de stabilizing force posteriorly. This method of disturbance was also used by Luchies, et al [67] to investigate age effects on step response to impending falls. Also used as de stabilizing inputs were movements of the trunk and of upper extremities such as reaching forward [24].

Mechanical vibrations have been applied to the leg muscles as a form of postural perturbation [25, 34]. Also, Mauritz, Schmitt and Dichgans [75] applied electrical stimuli to the tibial nerves to induce postural sway and study the body's response. Martin, Fletcher and Park [72] investigated effects of hand vibration on postural control and stability. He measured COP excursion while the subject held a vibrating handle (sinusoidal 150 Hz. vibrations). In this experiment alterations of hand proprioceptive cues caused by hand vibrations resulted in a deterioration of balance stability and a change in postural behavior. This information is important for the design of industrial workplaces where hand-held vibrating tools are extensively used.

2.4 Postural Control System Models and Dynamic Equations

Many researchers have taken the control systems approach to understanding the complex process of postural control. Two elements work together to make up the postural control system: neural elements (reflexive and voluntary) and mechanical elements (bones, muscles, ligaments, and tendons). It is a dynamic system (i.e., concerned with motions as it depends on forces and velocities as well as positions) [50]. A control system framework must explicitly describe system outputs, inputs, and input-output relationships. This framework can be used to analyze how the system operates at a given time or how it adapts to changed conditions due to development, disease, trauma, and recovery [61].

Many researchers have developed postural control systems [47, 49, 68, 69, 88, 95]. Johansson and Magnusson [50] presented a model of sensory feedback in control of posture. However, most models were used to predict ankle torque or body sway in the form of COP displacements in response to external disturbances.

In 1984, Shimba and Takeshi [102] derived an equation from the principles of dynamics for a system of particles forming a relationship between force plate data, the center of gravity on the platform, and the moment of momentum of the body about its center of gravity. This equation allows the estimation of the center of gravity path or the time rate of change of the angular momentum of the body. With the use of equilibrium equations, Snijders and Verduin [106] developed a force platform device to measure position of mass CG of erect standing subjects or during lying posture. Barin [4] used measurement of the COP and a mathematical model to estimate peak-to-peak ankle sway angle which was then normalized by 12.5^{o} (a theoretical maximum).

Maki, Holliday and Fernie [69] developed a posture control model which defines relative stability by the degree to which a postural perturbation causes the COP on the feet to approach the limits of the base-of-support. He used a type of external perturbation as input, resulting in a COP displacement output. Small amplitudes, random and pseudo random translation accelerations in the A/P direction were used to define a linear transfer function. Also, three system identification methodologies: ordinary least squares, cross-spectral analysis, and maximum likelihood were used. Then the saturation amplitude, which is the transient perturbation pulse amplitude at which the resulting COP displacement would equal the length of the base-of-support, was predicted.

Body movements were often described by segmented rigid-link mechanics. These models are formulated using the methodology of classical mechanics. Many different approaches have been used. The most common of these is the inverted pendulum. Single or multi-link biomechanical models have been developed which quantify the relationship between body segment angles and joint torques. However, a problem exists with the modeling of damping of perturbations in posture [50].

Gurfinkel [33] in 1973 developed a biomechanical model to quantify the relationship between body segment angles and joint torques. The human body was simplified and modeled as a one-link inverted pendulum with the axis of rotation at the ankle. The model defined difference between the COP and the vertical projection of the CG in terms of the following parameters: sway angles, the mass of the body, and dimension of the body. Concluded from the test was that the upper limit for frequencies which are reflected in the stabilogram with an error less than 10% is 0.2 Hz.

In 1972, Nashner [87] developed a model to describe the control of body sway with only vestibular sensory input. Nashner used the two degree of freedom force platform he created which rotates with body sway to eliminate proprioceptive input thus creating a rigid body that rotates about the ankle joint [88]. The assumed model was allowed movement in the A/P direction only. With this model, he was able to estimate sway angle due to known perturbations and the relative influence of semicircular canals and utricular otoliths, which act as linear and angular accelerometers.

Smith [105] developed an ankle torque equation for postural sway in the A/P direction using a single-link inverted pendulum in 1957. Kadteyn [53] later developed a model for the M/L postural sway. This model used ankle torque derived from stabilometer measurements to predict body sway. A simplified differential equation was formulated and tested in the same way as the equation for A/P sway given by Smith [105]. Both the static and dynamic components of the stability were estimated.

Peeters, Caberg and Mol [95] in 1985 also used the single-link inverted pendulum model. They calculated power gain spectrum, phase angle, and coherence for the frequency relation between ankle joint torque and sway angle. The values derived from the model were compared with experimentally measured ankle torques and body movements. Differences between simulated and measured spectra were mainly noted for high frequencies ($f > 1$ Hz.), and could be attributed to decreasing signal to noise ratio with increasing frequency, sway angle, and the degree of biological coordination between parts of the body.

Modeling of the body by a single-link inverted pendulum uses the force platform for measuring the reactive forces and treats the swaying body as a structure with one equivalent support. However, this type of modeling has been criticized by Valk-Fai [127] and Lestinne (1977). The two reasons for the criticism were the existence of relative motion between adjacent body segments which contribute to body correction, and the failure of the inverted pendulum model to predict corrections between sway magnitude and physique variables such as body height, weight, and due to lack of symmetry of the actual body motion about the vertical axis [76].

Winter [130] modeled frontal plane posture and balance of the total body and head, arms, and trunk during walking with two inverted pendulum systems in 1990. The first is the total body pivoting about the subtalar and ankle joints with the total body CG medial to the foot COP. Thus, the COP accelerates toward the midline during single stance and is shifted to the weight accepting foot during double stance. The second inverted pendulum requiring balance was the head, arms and trunk which pivoted about the support limb hip. Dynamic equilibrium equations were written for both pendulum models. Although this information deals with balance during gait as do many other studies [58, 129], the results will still be useful in the overall understanding of postural control system.

Valk-Fai [127] developed a four-link inverted pendulum model to determine the difference between the measured COP displacement from the ground reaction forces and the calculated CG displacements. Linear potentiometers were used to measure horizontal movement of four points: the knee, femur, shoulders and the head. Angular movements around the ankle joint, the knee, the hip and the neck were then calculated. Results confirmed that COP displacements are greater in magnitude and frequency than displacements of the CG. Also concluded was that larger angular activity occurred at the ankle and hip, rather than at the knee or neck.

Koozekanani, et al [57] developed a four-mass inverted pendulum model which indirectly provided a measure of the movement of the COP of the ground reaction forces. Sway angles were calculated with motion data collected from television cameras. Equations for the COP and CG's were developed and the results compared with COP displacements measured with a force platform. The effects of joint torques produced by muscular contractions, as well as torques produced by external perturbations were also included. Luchies, et al [67] used a nine-link biomechanical model to calculate net reaction joint torques from kinematic and kinetic data.

Hasan, Robin and Shiavi [37] aimed at development of a mathematical model which would use measured COP excursions to estimate movements of the COG and implement novel, clinically relevant measures of postural stability. An eleven-rigid-segment model was used and developed with kinematic data. Barin [4] created a mathematical model of an N-link inverted pendulum atop a triangular foot. The segments were considered rigid bodies with concentrated mass. The study was conducted in order to extend the analysis of error in force platform measurements of postural sway to dynamic testing conditions.

2.5 Clinical Applications

Much research and experimentation has been done to investigate the mechanisms which make up the postural control system. This increase in understanding is very useful in the diagnosis and treatment of patients. In the area of sports medicine, it is important to correctly identify severity of injuries when deciding if an athlete is able to resume competition. In addition, the use of postural sway to quantify balance may be an important tool that can be used in comparing drug effects in young and elderly [37].

Postural stability evaluations are useful with brain-injured patients in determining the area of deficiency and identifying the type of rehabilitation most beneficial for the specific patient. It could also be used to monitor progress over periods of time and rehabilitation. Mizrahi, et al [76] investigated postural stability in post-cerebral vascular accident (CVA) hemiplegic patients and compared these values to normative data. Sway activity was found to be significantly higher in hemiplegics than in normal subjects. Ingersoll and Armstrong [45] evaluated four levels of closed-head-injured patients. The levels were determined by the presence and length of a loss of consciousness upon injury. Results showed that closed-head-injured patients, especially those who experienced long periods of unconsciousness, displayed postural instability. Results of these studies along with that of Lehmann, et al [62], who looked at sway as an indicator of stability in post-traumatic brain injured (TBI) patients, suggest that the contributing factors of instability could be identified and an appropriate rehabilitation program could be developed.

Postural steadiness evaluations have been conducted to evaluate changes due to neurological disease [17, 52, 73, 75, 92, 93]. Dichgans, et al [16] conducted experiments to analyze the stabilizing and de-stabilizing effect of vision in normals and atactic patients. Paulus, Straube and Brandt [94] conducted a study of changes in balance parameters when closing the eyes in subjects with severe deficits of the vestibular and somatosensory systems. In these subjects, a loss of balance occurred within one second after closing the eyes. Comparisons have been done between normals and patients with vestibular lesions [16, 116], cerebellar lesions [16, 17, 73, 116], tabes dorsalis [16], spinal lesions [116], proprioceptive hyperactivity [116], and multiple sclerosis [13, 92]. Results showed that in atactic patients, the postural instability frequently exhibits an amplitude peak at about 0.6 Hz in the Fourier power spectrum. Patients with a cerebellar disease exhibit a peak at 2.5-3 Hz. In addition, patterns of postural movements in patients with vestibular pathologies have shown to be different from those of subjects with normally functioning vestibular systems [2, 7].

Murray, Seireg and Sepic [81] investigated in 1975 for further sensitive parameters to measure postural steadiness. Murray's results showed two characteristics by which normal upright balance can be distinguished from that of patients with neuromuscular or skeletal disability. The first of these was that the enclosed area of the COP positions during sustained weight-shifting was larger for normal men. This indicated a larger area of stability, or security, in normals. The second mark of normal stance was the manner in which upright stability was controlled. The shifts of force and movement for normal subjects, as compared with patients with postural instability, were balanced such that the center of pressure remained remarkably close to its mean position [81].

Simoneau, Cavanagh and Ulbrecht [103] calculated the correlation between different clinical evaluations and total COP displacement. Somatosensory evaluations were found to be most closely associated with postural stability measures in patients with diabetic neuropathy. Since distal sensory neuropathy resulting from diabetes mellitus significantly reduces the ability to maintain a stable stance position, an evaluation technique closely associated with COP measurement would be useful in conjunction with regression equations in predicting postural stability in patients without making a direct posture measurements.

Moffroid, et al [77] proposed the use of postural stability evaluation techniques on patients with lower back pain. Surface EMG was used to measure muscle latency in response to force platform perturbations. Moffroid felt that patients with lower back pain may have delayed postural response due to deconditioning, slowed nerve conduction velocity and/or aberrant muscle recruitment patterns related to postural changes or dysfunctional movement habits.

Postural stability testing is also important for the elderly [98]. There is a need to develop a reliable and appropriate balance evaluation protocol which could be used to identify individuals at risk before they experience a debilitating fall, thereby allowing preventative measures to be taken. Some risk factor evaluations used only questionnaires and clinical evaluations [89, 120, 121] while others have added the use of biomechanical parameters [3, 91]. Bugchee and Bhattacharya [3] used COP excursions near the stability boundary as a sign of decreased stability and increased risk of falls. Tinetti, Williams and Mayewski developed nine risk factors used in evaluation of balance based on the ability to complete various tasks, and identified chronic characteristics associated with falling among the elderly [121].

Postural steadiness evaluations have been conducted to evaluate changes due to age [23, 27, 38, 46, 63, 64, 66, 71, 112, 133]. Some have focused specifically on women's postural stability changes with age [38, 63, 64], while others have studied men [27,]. Panzer, Zeffiro and

Hallett [93] used kinematic and force platform data to identify characteristics of Parkinson's Disease patients separable from aging effects. Tests have been developed to try to classify an elderly patient as a "faller" or "non-faller" [5]. Others have evaluated changes in frequency of falls due to illness and age [122], or by a compilation of various risk factors [9].

Panzer-Decius and McFarland [92] conducted a study to determine which postural stability measurement best correlated to clinical evaluations of multiple sclerosis (MS) patients. An example is the correlation between the number of clinical exacerbations and bursts of enhancing lesions detected by Magnetic Resonance Imaging (MRI) and postural instability. It was determined that postural response latency was a sensitive indicator of motor changes in MS, demonstrating disease activity that is not clinically evident. Measures of postural response appear to be a useful means of monitoring the course of MS and may be useful as a clinical outcome measure.

CHAPTER III
EXPERIMENTAL METHODS

The purpose of this thesis was to develop a feasible and reliable testing protocol which offered useful information in the assessment of balance patients. To insure cost and time effectiveness while preventing patient fatigue, a protocol using only force platform data was developed which could be completed in approximately thirty minutes. The testing protocol parallels that used in previous balance studies by Lehmann, et al. [200].

3.1 Equipment

All data were collected at the Biomechanics Evaluation Laboratory (BEL) at the St. Lawrence Hospital Health Sciences Pavilion in East Lansing, Michigan. The BEL is equipped with three means of data collection: kinetic, kinematic, and electromyographical (EMG).

Kinetic data can be collected on an AMTI Biomechanics Force Platform, model OR6-6-1 which is mounted such that the surface of the platform is flush with the floor. The force platform features graphite-composite construction, has a capacity of 1000 pounds, and a resonant frequency of 500 Hz. The platform uses strain gages to measure the 3 orthogonal ground reaction forces, Fx, Fy, and Fz, and the three components of the moment about the instrument center for a total of six outputs. Strain signals are amplified and sampled at 100 or 1000 Hz. from an A/D Board and converted to forces. The force platform coordinate system and dimensions are as shown in Figure 1.

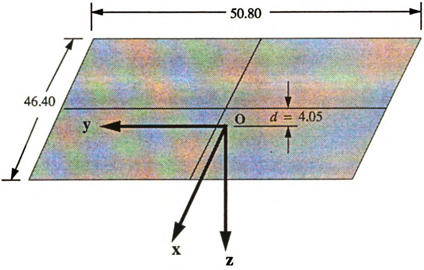


Figure 1. Force platform ccoordinate system and dimensions.

Kinematic data, is obtainable at the BEL through the use of a Motion Analysis System. Four 60 Hz video cameras are used to collect motion data by viewing retroreflective skin targets placed upon the subject's bony landmarks. The four cameras are synchronized by the VP320 video processor and targets are digitized in pixel space. Expertvision (ev3d) software then uses Direct Linear Transformation algorithms to obtain three-dimensional data for each target in the laboratory coordinate system.

The third and final type of data available is EMG, collected by real telemetry on a Transkinetics system with 16 channel capabilities, using Protrace surface silver/silver chloride electrodes on various muscle groups. Files from all three types of data are then stored on a Sun 4/260C work station for analysis. The advantages of having multiple data forms available are great and allow the researcher to obtain a complete understanding of the functioning of different mechanisms within the body. However, the use of all three types of data is very labor intensive, too costly, and not feasible for desired clinical balance tests. Therefore, only force platform data were used for this methodology. A data collection program, "bel4.11", was used to collect and view the data. Bel4.11 also allows the input of the trial description and all header information as well as setting parameters such as trial duration and trigger source.

3.2 Subjects

For this study, human subjects were recruited from Michigan State University and St. Lawrence Hospital under the UCRIHS approval IRB #89-559. Nineteen normative subjects were evaluated in order to obtain base line data. The norm subjects consisted of 9 men and 10 women from the ages 22 to 37, averaging 25.75 years old, who had no connection with the BEL and did not know what parameters were being measured. The norm subjects were given a questionnaire to fill out and were screened for any past neurological problems or losses of consciousness. See Appendix A. The patients evaluated in the study were referred to the BEL by physicians for balance and stability evaluations. Referred patients came with two different diagnoses, either Traumatic Brain Injured (TBI) or exposure to toxic substances.

3.3 Experimental Preparation

Before any testing was done, all subjects were required to read and sign an informed consent form for Michigan State University and St. Lawrence Hospital. A balance and stability questionnaire was given to the patients to obtain background information and details about their present condition (see Appendix A). Each subject was then weighed. A brief explanation of the testing procedure was given to the subject and the specific stances to be used in the test were demonstrated. The subject was also asked which foot he/she would use when kicking a ball. The one stated was recorded as the dominant foot and was placed in front of the other foot during tandem stance.

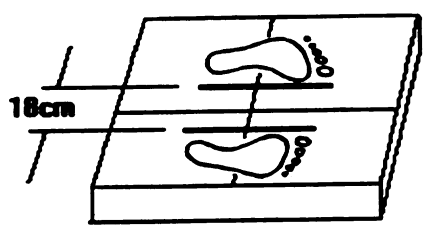
3.4 Testing Procedure

The subject information along with the first trial description was entered into the computer. The condition was announced and the subject stepped onto the plate. A spotter was nearby to insure proper footing and provide assistance if needed. As soon as the subject looked comfortable, the data collection was triggered manually. Data were collected for 20 seconds at a sampling frequency of 100 Hz. The subject was instructed to continue the test for the entire 20 seconds, even if a loss of balance occurred such as shifting of the feet or stepping off the plate. After the 20 seconds, the subject was asked to step off the plate, and the instrument was re-zeroed. The data for the entire 20 seconds was kept and analyzed no matter how much shifting of the feet occurred. All data corresponding to loss of balance was included in order to obtain a complete description of the subject's balance without eliminating valuable information. The test continued by announcing the next condition to the subject according to the following list:

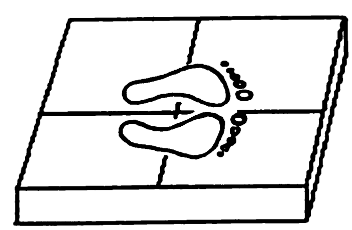
1. feet apart with eyes open
2. feet apart with eyes closed
3. feet together with eyes open

4. feet together with eyes closed
5. feet in tandem with eyes open (dominant foot in front)
6. feet in tandem with eyes closed
7. right-footed stance with eyes open
8. left-footed stance with eyes open.

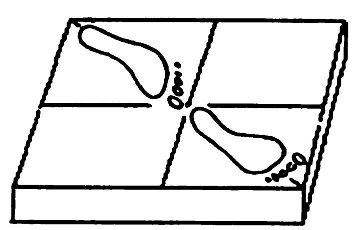
Prior to the beginning of the test, the force platform was prepared with two narrow strips of masking tape as shown in Figure 2. The tape was placed 18 cm. apart, measured from each outside edge of the tape, to insure consistent foot placement in the feet apart conditions between trials and individuals. This was done in order to keep the base of support between patients consistent within the variance in foot size. When using parameters such as total excursion or radial displacement, this consistency in foot placement is necessary to allow comparisons between subjects. One goal of this thesis is to identify parameters such as speed or torque which are unaffected by where on the plate a person is standing. All conditions were performed in stocking feet with arms crossed in front of the body. Having arms crossed eliminated the different arm waving techniques used by subjects to maintain balance and forced the body to rely on strictly hip and ankle strategies. To allow comparison with the work done by others who evaluate the A/P and M/L COP values separately, all conditions were performed facing the negative y direction except for tandem stance which was performed diagonally (see Figure 2.). This allowed the separation of anterior/posterior and medial/lateral sway components for all conditions except tandem. After completion of all 8 conditions, the sequence was repeated 2 more times for a total of 3 trials per condition.



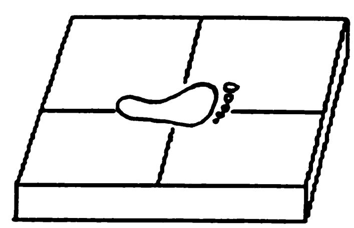
a) FAEO, FAEC



d) FTEO, FTEC



c) TDEO, TDEC



d) RLEO, LLEO

Figure 2. Foot placement protocol.

CHAPTER IV
ANALYTICAL METHODS

4.1 Calculation of the COP

As mentioned previously, the most often used measure of postural stability is the center of pressure. There are two approaches to the definition of the COP.

Vertical Torque Method - In the first approach, consider the measured resultant ground reaction force $\vec{\mathbf{R}}$, and the moment about the instrument center $\vec{\mathbf{M}}_o$. These may be resolved into a single resultant force $\vec{\mathbf{R}}$ with a unique line of action and a vertical torque by the following method [109, 110]:

$$\vec{\mathbf{P}} = X_{COP}\hat{\mathbf{i}}_x + Y_{COP}\hat{\mathbf{i}}_y - d\hat{\mathbf{i}}_z \tag{1}$$

$$\vec{\mathbf{R}} = R_x\hat{\mathbf{i}}_x + R_y\hat{\mathbf{i}}_y + R_z\hat{\mathbf{i}}_z \tag{2}$$

$$\vec{\mathbf{P}} \times \vec{\mathbf{R}} + (GRT)\hat{\mathbf{i}}_z = \vec{\mathbf{M}}_o \tag{3}$$

where $\hat{\mathbf{i}}_x$, $\hat{\mathbf{i}}_y$,, and $\hat{\mathbf{i}}_z$ are unit vectors along the force platform coordinate system, and d is 4.05 cm, the distance between the instrument center and the surface of the force platform (see Figure 1, section 3.1). By equating the $\hat{\mathbf{i}}_x$, $\hat{\mathbf{i}}_y$, and $\hat{\mathbf{i}}_z$ components of each side in equation (3) we are able to solve for the X_{COP} and Y_{COP} positions which are the intercept of the force system resultant with the horizontal surface of the force platform, and the ground reaction torque.

$$X_{COP} = \frac{-(M_y + R_x \cdot d)}{R_z} \quad (4)$$

$$Y_{COP} = \frac{(M_x - R_y \cdot d)}{R_z} \quad (5)$$

$$GRT = \frac{\vec{\mathbf{M}}_o \cdot \vec{\mathbf{R}}}{Fz} \quad (6)$$

where $\bar{\mathbf{P}}$ is the vector from the instrument center to the COP and $P_z = -d$. (see Figure 3.).

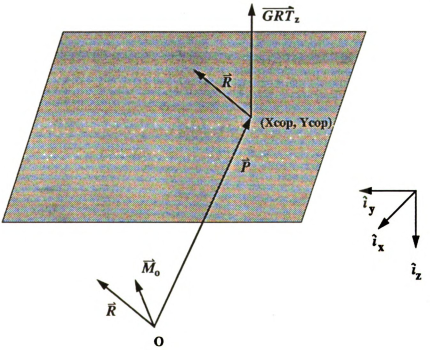


Figure 3. Vertical torque method for calculation of the COP

Parallel Torque Method - Method two involves resolving the force system into a wrench system where the torque $\bar{\mathbf{T}}$ is in the same direction as the resultant force vector [108, 110]. See Figure 4.

$$(\mathrm{T})\hat{\mathbf{i}}_R + \vec{\mathbf{P}} \times \vec{\mathbf{R}} = \vec{\mathbf{M}}_o \tag{7}$$

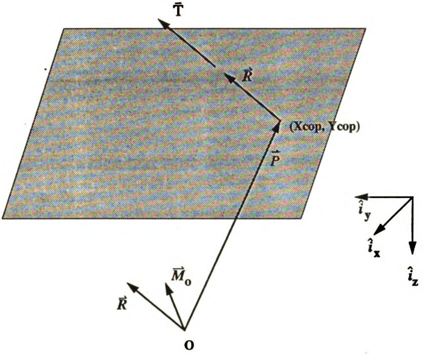


Figure 4. Parallel torque method for calculation of the COP

Equating components in equation (7) as before we obtain the following:

$$X_{COP} = \frac{-(M_y + R_x \cdot d)}{R_z} + \frac{\vec{\mathbf{R}} \cdot \vec{\mathbf{M}}_o}{\vec{\mathbf{R}} \cdot \vec{\mathbf{R}}} \frac{R_y}{R_z} \tag{8}$$

$$Y_{COP} = \frac{(M_x - R_y \cdot d)}{R_z} - \frac{\vec{\mathbf{R}} \cdot \vec{\mathbf{M}}_o}{\vec{\mathbf{R}} \cdot \vec{\mathbf{R}}} \frac{R_x}{R_z} \tag{9}$$

$$(\mathrm{T})\hat{\mathbf{i}}_R = \frac{\vec{\mathbf{R}} \cdot \vec{\mathbf{M}}_o}{\vec{\mathbf{R}} \cdot \vec{\mathbf{R}}} \hat{\mathbf{i}}_R \tag{10}$$

In balance studies, R_x and R_y are small compared to R_z and the two force systems are nearly equal [110]. Thus, the vertical torque method was used for this study.

After calculation of the COP, many different methods of analysis may be employed. These methods of analysis will involve parameters in either the time or frequency domain.

4.2 Time Domain Analysis

One method which has been used by many researchers is the cross plot of the X_{COP} versus the Y_{COP} components to obtain a spot stabilogram, either displayed on a representative force platform or centered at an origin by subtracting the mean values, $\overline{X}$ and $\overline{Y}$. This may also be plotted in a polar coordinate system by the following equations [62, 110]:

$$\overline{X} = \frac{1}{n}\sum_{i=1}^{n} X_{COP}(i) \tag{11}$$

$$\overline{Y} = \frac{1}{n}\sum_{i=1}^{n} Y_{COP}(i) \tag{12}$$

$$X_{COP}(i) = X_{COP}(i) - \overline{X} \tag{13}$$

$$Y_{COP}(i) = Y_{COP}(i) - \overline{Y} \tag{14}$$

$$r(i) = \sqrt{X_{COP}(i)^2 + Y_{COP}(i)^2} \tag{15}$$

obtaining the radial displacement of the COP path from the mean. This is then plotted verses the angle from the x-axis of the force platform obtained by

$$\theta(i) = \tan^{-1}(Y_{COP}(i) / X_{COP}(i)) \tag{16}$$

Radial average displacement (RAD) is often used as an overall measure of sway. The average may be computed using equation (15) and

$$\mathrm{RAD} = \frac{1}{n}\sum_{i=1}^{n} r(i) \tag{17}$$

Radial and average speed may then be calculated using ΔS_i, the distance between the COP at times i and i+1.

$$\Delta S_i = \sqrt{(Xcop_{i+1} - Xcop_i)^2 + (Ycop_{i+1} - Ycop_i)^2} \tag{18}$$

$$S_{total} = \sum_{i=1}^{n} \Delta S_i \tag{19}$$

$$v = f\left(\frac{S_{total}}{n}\right) \tag{20}$$

where f is the collection frequency in Hertz. The speed v(i) will be referred to as radial velocity from this point forward for continuity purposes.

The ground reaction torque (GRT), which is a byproduct of the COP calculation, also contains valuable information. The GRT may be displayed versus a time file. This shows how a patient's GRT varied during the entire test. However, one might wish to describe a patient's overall torque for a specific trial with quantitative parameters. The time series of the GRT measurement is averaged in two ways. The first is an averaging of the raw torque signal termed the "average torque". This value describes the direction of the overall torque. A zero would indicate the subject had equal amounts of torque to the right and to the left. Thus, the averaged signal is zero. A positive average torque corresponds to the patient's twisting to the right, the floor resisting to the left. The inverse is true for a negative average torque. The second method of torque averaging is to rectify the signal, then average it. This measure is termed the "active torque". This value will correspond with the magnitude of the torque. Thus, a person doing more twisting in order to maintain balance will have a higher active torque than someone who does little twisting. The equations used to calculate these torque measures are as follows:

$$\text{Average Torque} = \frac{1}{n}\sum_{i=1}^{n} GRT(i) \tag{21}$$

$$\text{Active Torque} = \frac{1}{n}\sum_{i=1}^{n} \left|GRT(i)\right| \tag{22}$$

4.3 Frequency Domain Analysis

In the frequency domain, the X_{COP} and Y_{COP} displacements may be analyzed with a Fast-Fourier Transform (FFT) algorithm. This produces the power spectrum of the signal and reveals the frequencies present within the COP components [29, 73, 107, 117]. However, in addition to the COP, an FFT analysis may be conducted on the ground reaction torque (GRT) [110] and dr/dt terms also.

To compare and build upon the work of Soutas-Little, et al. [110], a frequency spectrum analysis was conducted for the X_{COP}, Y_{COP}, torque, and radial speed. An FFT was used to obtain the power frequency spectrum for these parameters. The spectrum was then processed to obtain the spectral distribution and plotted versus frequency.

CHAPTER V
RESULTS AND DISCUSSION

Radial average displacement, or more simply combined Xcop and Ycop excursions, was used extensively throughout the literature as an instability parameter. What researchers aim to measure is how one's balance changes with a change in test condition such as a change in stance. However, this is very difficult to measure and therefore, parameters which can be correlated to changes in balance are used instead. It is desirable to use the parameter which will give the most useful information about the specific condition. For this reason, RAD as well at other parameters were evaluated in terms of sensitivity to balance. For this evaluation the sensitivity, β, of a parameter with respect to a change in test condition, Δx, was defined according to

$$\Delta p_i = \beta_i \Delta x \tag{23}$$

$$\Delta p_k = \beta_k \Delta x \tag{24}$$

Sensitivity of i: $$\beta_i = \frac{\Delta p_i}{\Delta x} \tag{25}$$

Sensitivity of k: $$\beta_k = \frac{\Delta p_k}{\Delta x} \tag{26}$$

Here, Δx is the unknown change in balance from one test condition, i, to another, k. The change in the parameter being considered for the same two conditions is Δp, where

$$\Delta p = \frac{p_2 - p_1}{p_1} \tag{27}$$

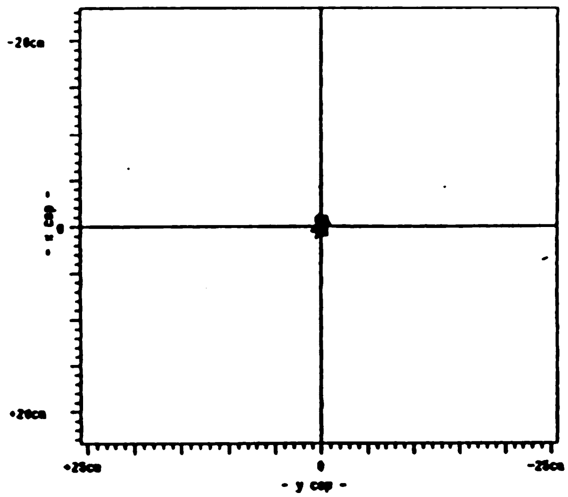
Note that Δx remains the same for all parameters. This is because the subject's change in balance is independent of the measurement parameter. However, Δx is dependent upon the two conditions being tested and is not expected to be the same for all conditions. Δx may be very different for comparison of FAEO and FAEC than for comparison of FAEO and TDEO. Since this difference is unknown, and Δx is not measured, the difference will be apparent by comparison of the sensitivities for these conditions.

The sensitivities β_i and β_k of the two parameters, with respect to the same measure, can be compared by

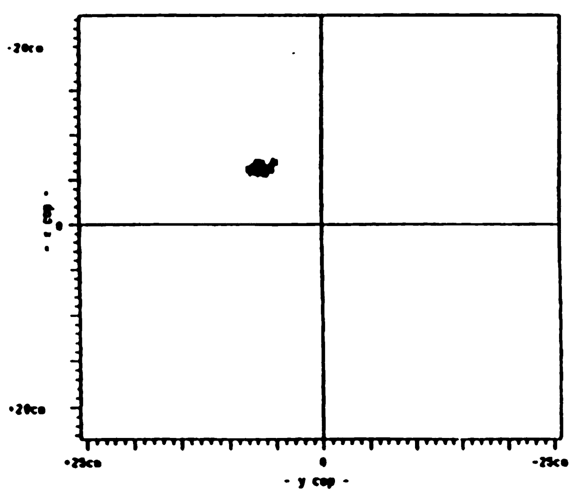
$$\frac{\beta_i}{\beta_k} = \frac{\Delta p_i}{\Delta p_k} \tag{28}$$

Thus, if $\Delta p_i > \Delta p_k$, then $\beta_i > \beta_k$. Likewise, if $\Delta p_i < \Delta p_k$, then $\beta_i < \beta_k$. This allows a sensitivity comparison to be made between any two parameters for a single condition change.

Displacement of the COP provided some useful information about sway, but it failed to accurately characterize a subject's balance in several instances, an example of which is shown in Figures 5 and 6. Figure 5 is a plot of Xcop versus Ycop displacement for two conditions: (a) FTEC, and (b) TDEO for the same subject. Although the patient was observed having difficulty in maintaining balance in tandem stance, the cross plot suggests little or no change in sway between the two conditions. Therefore, the β of the RAD would be low. Calculations show that the RAD actually increased by 14% (β=0.14) from condition a to condition b. Figure 6, however, which is a plot of torque versus time

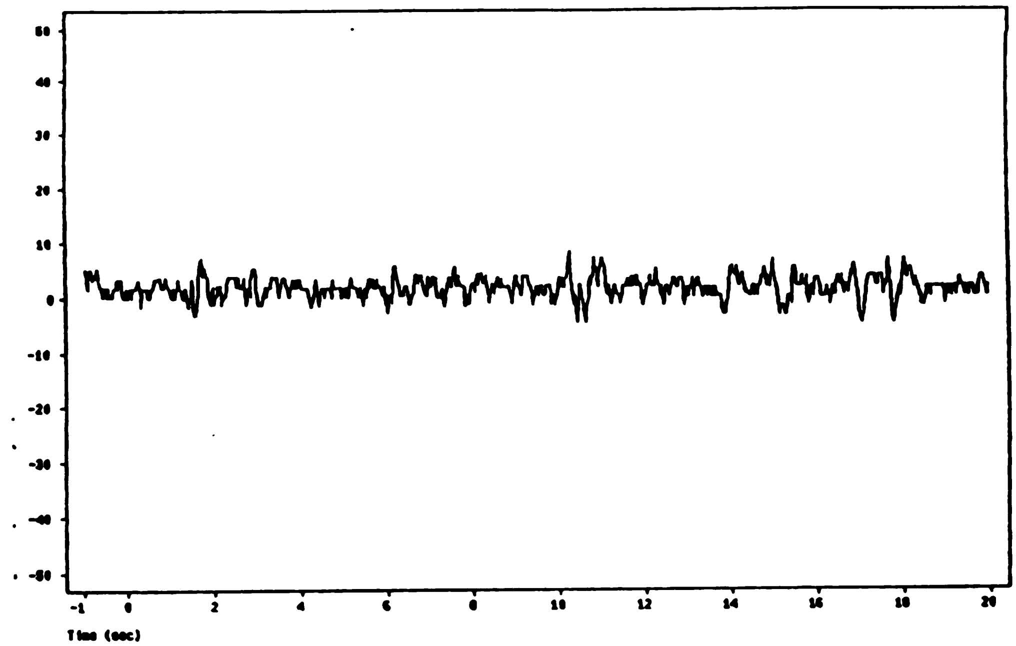


(a) Feet together with eyes closed

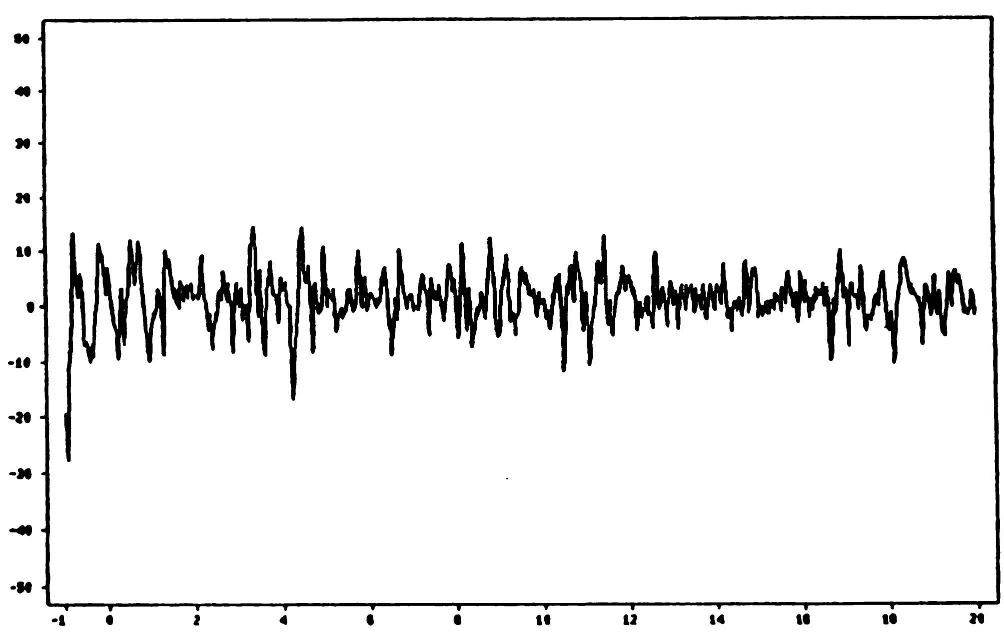


(b) Tandem stance with eyes open

Figure 5. Xcop versus Ycop



(a) Feet together with eyes closed



(b) Tandem stance with eyes open

Figure 6. Ground reaction torque

for the same patient and conditions, shows a torque increase of 56% (β=0.56) from condition a to b. Thus, the torque sensitivity was four times greater than the radial average displacement sensitivity. The velocity parameter increased 45% (β=0.45) from condition a to b and therefore was approximately three times more sensitive than RAD. Therefore, using this definition of sensitivity, RAD was found to be less sensitive than both velocity and torque.

Clearly, if radial average displacement alone was the measure used to determine if a balance difficulty existed in tandem with eyes open, the result would be that no change was present. However, if velocity and torque are also used, there would be evidence that some change in maintenance of balance occurred between FTEC and TDEO. Thus, consideration of only radial average displacement, which is a combination of Xcop and Ycop, could lead to a false conclusion in assessing balance by comparing these two conditions. This illustrates the importance in determining which variable(s) will give the most useful information for balance evaluations.

Another reason why radial average displacement should not be used as a sole means of evaluating balance is because the COP excursions are limited by the area between and under the feet or more accurately, the "functional base of support" as defined earlier [37]. Thus, comparison may not be conducted between a one-footed stance and a wide comfortable stance. Clearly the two bases of support are very different and the limits of COP excursion are different. In addition to intra-subject comparison, inter-subject comparison is also incorrect in some cases due to variance in foot size. One way to combat these comparison obstacles was to calculate COP excursions in terms of a percentage of total base support, or as a percentage of maximum allowable sway without falling [3, 37, 38, 40]. However, this does not eliminate the problem when a balance impaired person maintains balance through small quick ankle movements and has observable difficulty in maintenance of balance, thus does not create large COP excursions.

It has been suggested that dynamic parameters such as velocity and torque are more sensitive than static ones, such as COP displacement or area [109]. This is due to the fact that static parameters do not measure the state of balance at any particular instant in time. They are usually an averaged value over the entire trial duration. The dynamic parameters, however, give information about balance at each time during the trial. Therefore, trends are noticeable in addition to the onset of a fall or presence of a recovery.

The objective of this thesis was to identify balance evaluation parameters that are sensitive to postural instabilities, characterize the subject's postural responses and can be used as a comparison measure, thereby providing a quantitative approach in addition to already conducted qualitative analyses. In order to evaluate patients, a normative population was recruited to provide base line data for comparison. Normative data and results collected from nineteen subjects are shown in Table 1 as a comparison between the calculated and reported normative results found in the literature [62]. Although the analytical approach was identical for the two systems being compared, there is a discrepancy, approximately a factor of two, in the velocity values for all conditions between the reported and calculated mean values. The only difference between the two

Table 1. Comparison between reported and experimental norms

REPORTED AND MEASURED NORMATIVE BALANCE DATA

CONDITION	REPORTED NORM (mean)	REPORTED NORM (SD)	BEL NORM (mean)	BEL NORM (SD)
Comfortable stance, eyes open				
Radial Ave. Disp.(cm)	0.42	0.12	0.52	0.19
Velocity (cm/sec)	1.55	0.39	3.54	0.57
Active Torque (N-m)			0.20	0.23
Comfortable stance, eyes closed				
Radial Ave. Disp.(cm)	0.44	0.09	0.54	0.16
Velocity (cm/sec)	1.69	0.37	3.64	0.51
Active Torque (N-m)			0.22	0.10
Narrow stance, eyes open				
Radial Ave. Disp.(cm)	0.60	0.17	0.76	0.18
Velocity (cm/sec)	1.87	0.42	3.91	0.60
Active Torque (N-m)			0.22	0.08
Narrow stance eyes closed				
Radial Ave. Disp.(cm)	0.72	0.21	0.91	0.21
Velocity (cm/sec)	2.41	0.54	4.47	0.53
Active Torque (N-m)			0.22	0.08
*Tandem stance, eyes open				
Radial Ave. Disp.(cm)	0.65	0.12	0.91	0.29
Velocity (cm/sec)	3.23	0.62	5.61	0.79
Active Torque (N-m)			0.28	0.10
*Tandem stance, eyes closed				
Radial Ave. Disp.(cm)	1.07	0.22	1.23	0.41
Velocity (cm/sec)	5.54	1.46	7.77	1.64
Active Torque (N-m)			0.36	0.13
Right leg stance, eyes open				
Radial Ave. Disp.(cm)			1.03	0.19
Velocity (cm/sec)			6.35	1.01
Active Torque (N-m)			0.35	0.10
Left leg stance eyes open				
Radial Ave. Disp.(cm)			1.02	0.19
Velocity (cm/sec)			6.28	0.92
Active Torque (N-m)			0.31	0.10

* Tandem stance is performed with the dominant foot in front.

systems, which would cause this difference in the velocity and not the RAD was the sampling rate. The reported data were sampled at 50 Hz, while the calculated norms were sampled at 100 Hz. This is one possibility for the error. However, without inspection of the actual program used by Lehmann, et al. [62], this cause is not verifiable. The velocity equations used in this thesis, restated below from Chapter IV, were checked and correctly account for a 100 Hz. sampling rate.

$$\Delta S_i = \sqrt{(Xcop_{i+1} - Xcop_i)^2 + (Ycop_{i+1} - Ycop_i)^2} \qquad (18)$$

$$v = \frac{100}{n}\left(\sum_{i=1}^{n} \Delta S_i\right) \qquad (20)$$

The calculated norms, noted as "BEL norms", were then used as baseline data for evaluation of patients.

Figure 7 is an example of a patient's data compared with normative values for 8 conditions. Radial average displacement, velocity and torque are shown here. The white boxes represent normative values while the gray area is one standard deviation above the norm mean. The black boxes represent patient values with respect to the norm and are labeled with the percentage above norm mean. A patient value lying within the white or gray area is considered to be within normative range and have no apparent balance difficulties. For the patient shown here, all values were above the normative range. This type of plot is very useful for patient reports since it displays the data in a convenient layout for quick comparisons with normative values.

KEY

FAEO:
FEET APART EO

FAEC:
FEET APART EC

FTEO:
FEET TOGETHER EO

FTEC:
FEET TOGETHER EC

TDEO
FEET TANDEM EO

TDEC:
FEET TANDEM EC

RLEO:
RIGHT LEG EO

LLEO:
LEFT LEG EO

EO=EYES OPEN
EC=EYES CLOSED

Numbers represent %above norm mean

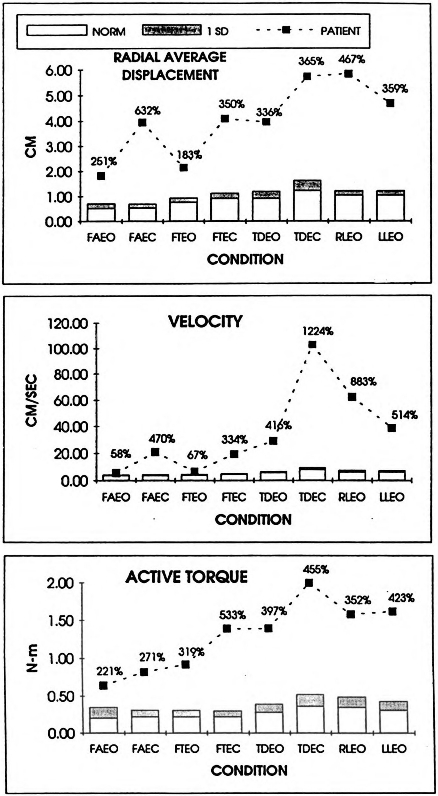


Figure 7. Balance summary chart

In addition to time-domain parameters, five parameters were studied in the frequency domain: Xcop, Ycop, torque, radial displacement, and radial velocity. The spectral distributions for these parameters were calculated and plotted for comparison. A method was devised to isolate the frequency at which the maximum amplitudes occurred in the power spectrums. These values correspond to the frequencies that are most prevalent and used for maintenance of balance.

For interpretation of this data, a model was constructed using fundamental vibration theory. It was necessary to begin with a simple model in order to investigate the variables and their effects on the system. The model considered was a damped second-order single degree-of-freedom (DOF) linear system, with harmonic excitation. Using the model and free body diagram shown in Figure 8, a summation of the forces and Newton's second law, $\sum F = ma$, allows the development of the differential equation of motion for this system. One can then solve for the amplitude, X. The differential equation of motion is

$$m\ddot{x} + c\dot{x} + kx = F_o \sin \omega t \tag{28}$$

See Appendix B for the development of the following equation for gain in terms of frequency ratio.

$$\frac{Xk}{F_o} = \frac{1}{\sqrt{\left(2\zeta\frac{\omega}{\omega_n}\right)^2 + \left(1-\left(\frac{\omega}{\omega_n}\right)^2\right)^2}} \tag{29}$$

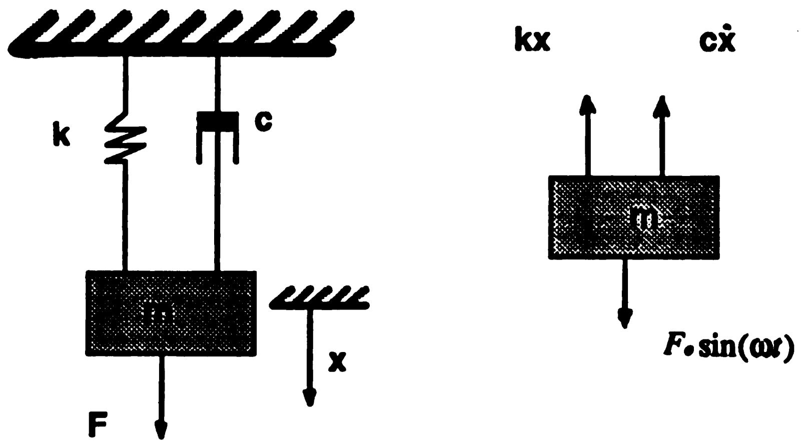


Figure 8. Forced harmonic oscillation 1 DOF model with free body diagram.

The natural frequency, ω_n, is that at which the system oscillates with maximum amplitude. According to vibration theory for this simple model, the natural frequency of oscillation is equal to $\sqrt{k/m}$, [119]. The symbol ζ is the damping ratio which is the ratio between the actual damping and the critical damping, c/c_c. The critical damping is the amount of damping necessary for the limiting case between non-oscillatory and oscillatory motion. When $c = c_c$, $\zeta = 1$ and no oscillations are present. When $c = 0$, $\zeta = 0$ and the system oscillates with a maximum amplitude. This occurs at the natural frequency. Actual postural damping, which is produced by the muscles and soft tissues of the body, is somewhere between 0 and c_c.

Figure 9 is a plot of gain, Xk/F_o, the amplitude ratio of the forcing function and the response, versus frequency ratio. The frequency ratio, ω/ω_n, is the ratio of excitation frequency to the natural frequency. Note that both the peak amplitude and frequency of oscillation depend on the value of ζ. Refer to Equation 28 for the equation relating amplitude and frequency ratio to the damping ratio. For a fixed frequency ratio, the amplitude decreases with increased damping. Likewise, the frequency ratio at which the maximum amplitude occurs will decrease with increased damping. Thus, two important parameters may be investigated, the peak frequency and amplitude.

This simple model may be related to the complex postural control model by correlating both damping and spring control mechanisms with the role played by control mechanisms within the body. The specific values for k and c depend on the amount of elasticity and damping present within the musculoskeletal system. The amount of damping in the postural control system corresponds to the ability to remove energy

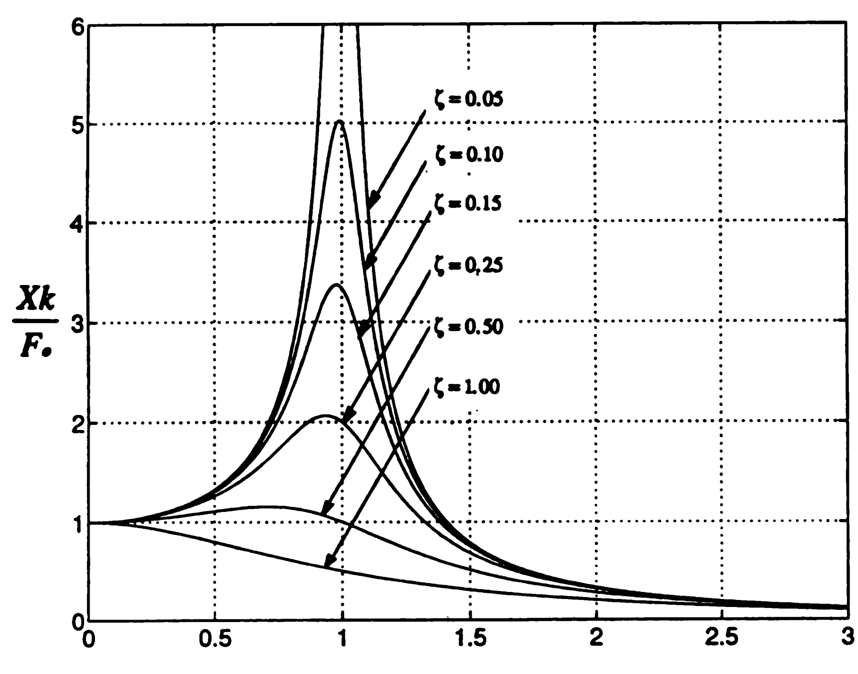


Figure 9. Gain versus frequency ratio plot for 1-DOF model

from the system [119]. By this model and for balance analysis, damping corresponds to the subject's ability to reduce oscillations and maintain balance. If the postural control system has the same parametric relationships as shown in Figure 9, then a person with more control, i.e., damping, will have a lower peak frequency and/or a lower amplitude. Both must be considered since a patient may exhibit a lower peak frequency but higher amplitude.

A dependence exists between the response of the system and the variables k and c, in terms of ω_n and ζ, respectively. This same parametric influence must also apply to more complex models of the human postural control system. The model considered for COP displacement parameters consisted of a double inverted pendulum with two DOF. This is similar to the simple two-link pendulum models used in the literature[130]. The model has one point of rotation at the ankle, and one at the hip, corresponding to the ankle and hip strategy rotations mentioned in section 2.3 on postural stability. The mathematics corresponding to this more complex model were unnecessary for this thesis since the model is used for interpretations only. However, it is important to note that the spring and damping parameters are represented as matrices containing values for each link within the system rather than a single value for the entire system. The model and corresponding modes of oscillation, with spring and damping matrices, denoted as K and C respectively, are shown in Figure 10. All joints are hinge joints and have both spring and damping constraints as shown in Figure 10a.

This system has two modes of oscillation. The first mode consists only of oscillation about the base, as depicted in Figure 10c. This has

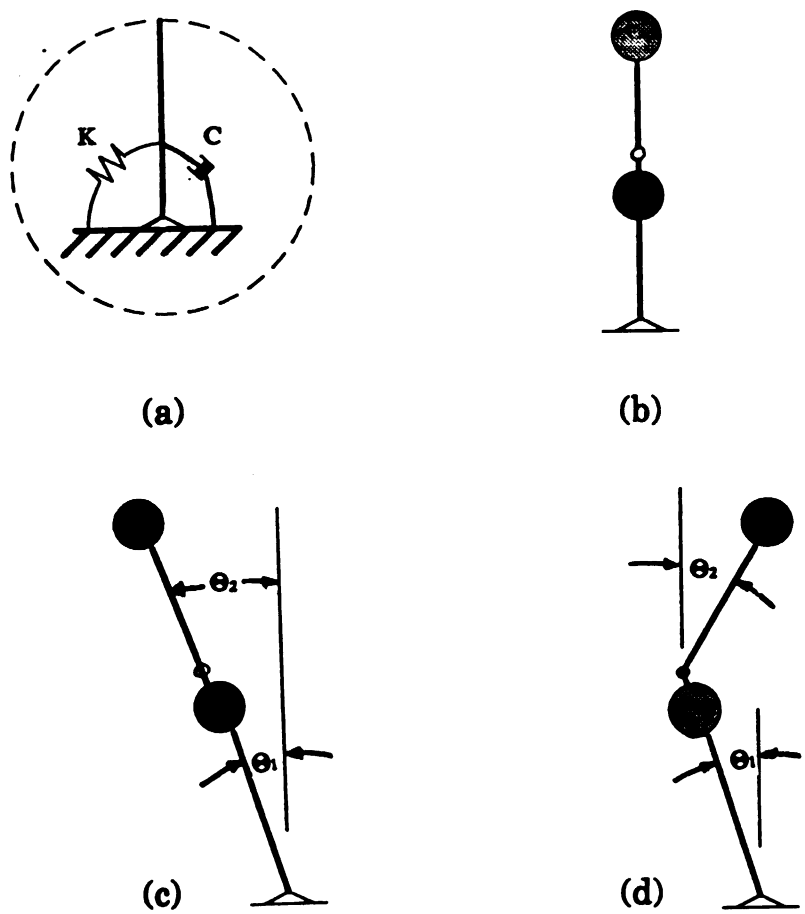


Figure 10. Inverted double pendulum 2-DOF model

two DOF where $\Theta_1 = \Theta_2$. The type of motion associated with this mode has been termed ankle strategy and is where the hip joint remains rigid and the entire body rotates about the ankle joint. The second mode of oscillation is more complex and is shown in Figure 10d. The two degrees of freedom are more noticeable since $\Theta_1 \neq \Theta_2$. They are usually opposite in sign and exactly out of phase. The rotations occurring correspond to rotation about both the ankle and hip joints. The modes of oscillation have been solved for in many vibration text books and show that the first mode is less than the second mode of oscillation [119]. This information is useful later in the interpretation of spectral distribution plots for Xcop, Ycop, and radial power.

The response of this system, like the simple one DOF model depends highly upon both elasticity and damping, and therefore, will characterize the amount of control exhibited by the subject; increased peak amplitude and frequency with decreased damping.

The torque power distribution is a more difficult parameter, conceptually. A different model which allows torsional movements is necessary for interpretation of the torque data. The model consists of a double inverted torsional rod system shown in Figure 11.

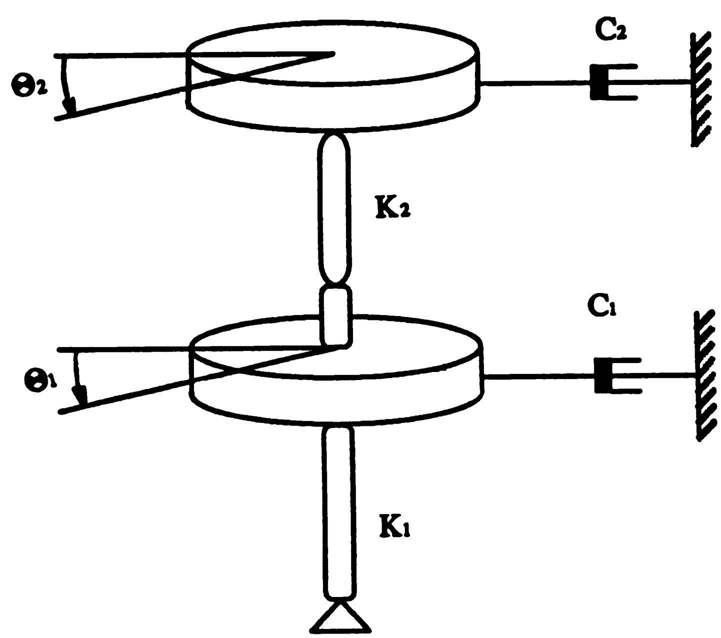


Figure 11. Torsional 2-DOF model for postural stability

The torsional system has two natural modes of oscillation: rotations that are in phase and those that are out of phase. If in phase rotations occur, only ankle strategy is present and the rotation about the hip is greater than rotation about the ankle. However, the frequencies of ankle and hip oscillation are equal. Thus, only one peak will occur in the spectral distribution. If out of phase rotations occur, then two frequencies will be present in the distribution and both modes of oscillation are present, the lower one being ankle rotation.

The final frequency domain parameter considered is radial velocity or the rate of change of the radial displacement. Interpretation of this parameter is even more difficult than the ground reaction torque. The plots are very irregular and peaks occur at very large frequencies. The proposed interpretation of this parameter is as a measure of balance recruitment, assuming a person with good postural control recruits only the control mechanisms necessary to maintain balance. The recruitment is quick, but controlled. However, a subject with reduced postural control most likely overcompensates and recruits more than is required to maintain balance, thus causing an imbalance in the other direction. Hence, more control mechanisms are recruited and postural response becomes more disordered. This interpretation agrees with and is demonstrated in the radial velocity plots. Patients' data displayed much higher peak amplitudes and frequencies and often many peaks occurred suggesting many modes of oscillation and attempts to maintain balance.

This type of model was not intended to predict human motion since the body is a multi-linked system with complex joints and intricate control mechanisms. The model was used only as a basis for interpretation of the data with regards to vibration theory.

The spectral distribution plots for one normative and one balance impaired subject can be found in Figures 12-15 and in Appendix C. Inspection of the COP power plots revealed instances where more than one peak existed. If the double inverted pendulum model in Figure 10 is taken as an anterior view allowing motion in the M/L direction, then movement correlates with the Xcop displacement. M/L motions are allowed at the ankle and hip for a two degree of freedom system. A more complex model would account for additional movements of the body. If the model is turned 90 degrees to a saggital view, movements correspond to the Ycop or A/P motion and one could easily account for a third mode of oscillation by adding a knee joint to the model. Again, a more complex model would allow for knee flexion and extension. This demonstrates the idea that the human body is a multi-link system which may have more than two modes of oscillation.

In addition to the spectral distribution plots, a program was developed which identified the peak frequency and amplitude for each plot. These values are listed in table 2, along with the patient's percentage above norm listed in table 3. The program picked the frequency for the largest peak only. This information was very useful in interpretation of the data and allowed quick comparison between patient and norm. However, if more than one peak was present, there existed a potential for misinterpretation of the data. With only the maximum peak recorded, one has no way of knowing where other peaks occur, if any, which explains the need to include and inspect the frequency distribution plots. The presence of additional peaks may suggest multiple strategies being used for postural control and should not be dismissed.

Some additional interpretation can be made from the spectral plots with regards to balance control strategy. Assuming the correlation between the postural control system and the inverted double pendulum is correct, the first mode of oscillation, ankle strategy, has a lower frequency than the second mode of oscillation, ankle-hip strategy. If only one peak occurs in the spectral distribution, ankle strategy was used. However, if two peaks occur in a spectral distribution, ankle-hip strategy was used and the lower frequency of the two peaks is for ankle strategy, the other peak is for hip strategy.

In table 3a, the patient's percentages above norm are listed for all conditions and parameters. In the FAEO condition, the patient's peak frequency remained below or equal to that of the normative subject for all parameters except dr/dt. However, the peak amplitude was below norm for torque power only. The large amplitudes for this patient suggest balance difficulties are present. Since the radius is made up of the two COP components, the large percent difference in the Ycop contributed to the difference in the radius as well. For this condition, only one large peak occurred in each plot in Figures 12 and 14 and thus the program selected the appropriate data.

As an example of two peaks, the FAEC condition shown in Figure 13 for the norm and Figure 15 for the patient was analyzed. Again, the data collected by the program are listed in tables 2 and 3. Inspection of the plot in Figure 15 reveals the cause of the large frequency shift for the Xcop power from the FAEO to the FAEC

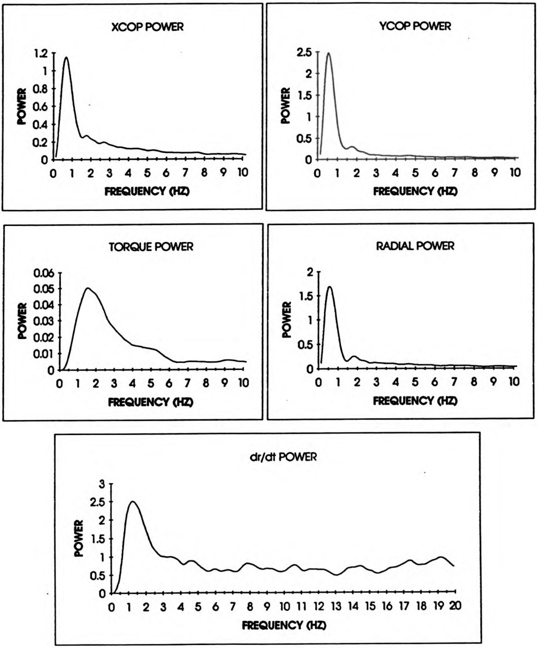


Figure 12. Norm spectral distribution for FAEO condition

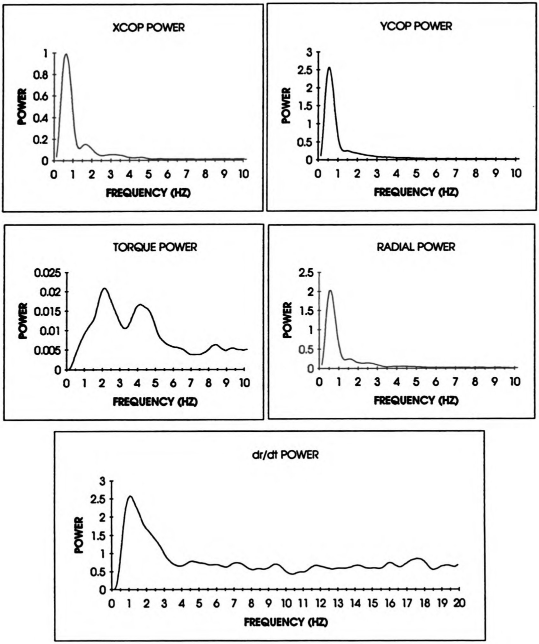


Figure 13. Norm spectral distribution for FAEC condition

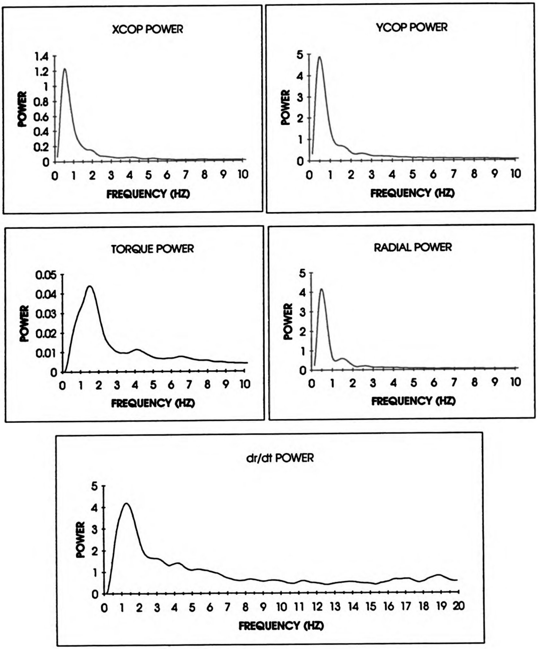


Figure 14. Patient spectral distribution for FAEO condition

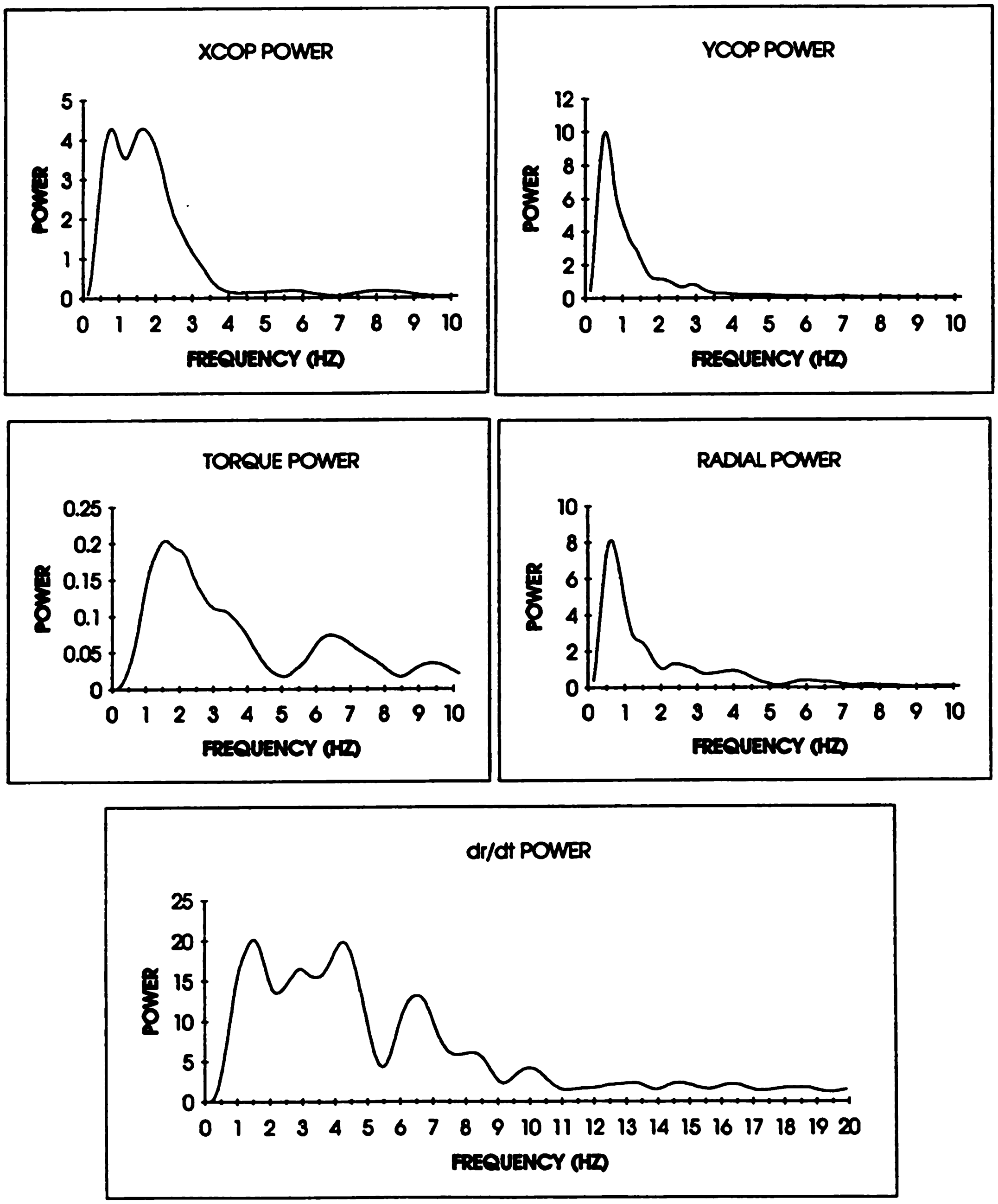


Figure 15. Patient spectral distribution for FAEC condition

Table 2. Patient values compared to norm

	XCOP		YCOP		TORQUE		RADIUS		dr/dt	
CONDITION	NORM	PATIENT	NORM	PATIENT	NORM	PATIENT	NORM	PATIENT	NORM	PATIENT
FAEO	0.68	0.54	0.54	0.54	1.56	1.51	0.59	0.49	1.27	1.32
FAEC	0.63	1.66	0.54	0.54	2.15	1.61	0.59	0.63	1.07	1.51
FTEO	0.68	0.54	0.63	0.68	1.56	1.81	0.68	0.59	1.32	1.42
FTEC	0.93	1.42	0.63	0.68	1.17	1.90	0.59	0.59	2.34	2.59
TDEO	0.59	0.63	0.68	0.59	1.81	2.20	0.59	0.59	1.76	26.37
TDEC	0.63	0.73	0.63	0.73	4.30	3.17	0.54	0.59	2.29	23.39
RLEO	0.63	0.68	0.54	0.63	1.71	1.32	0.63	0.59	3.03	23.00
LLEO	0.68	0.73	0.54	0.63	1.61	1.76	0.59	0.59	1.86	18.26

a) Characteristic frequency comparison

	XCOP		YCOP		TORQUE		RADIUS		dr/dt	
CONDITION	NORM	PATIENT	NORM	PATIENT	NORM	PATIENT	NORM	PATIENT	NORM	PATIENT
FAEO	1.15	1.24	2.47	4.87	0.05	0.04	1.69	4.18	2.51	4.18
FAEC	0.99	4.28	2.57	10.01	0.02	0.20	2.03	8.14	2.58	20.15
FTEO	1.74	3.82	2.17	2.49	0.03	0.09	1.56	2.80	3.67	7.26
FTEC	1.37	9.23	1.79	6.95	0.06	0.33	1.34	9.19	5.00	41.16
TDEO	1.98	27.81	1.38	14.00	0.06	0.24	1.32	24.33	5.33	199.02
TDEC	2.42	16.09	2.92	23.94	0.06	0.34	1.33	26.45	5.93	211.63
RLEO	2.16	30.46	2.20	27.74	0.15	0.31	1.74	20.27	6.32	203.67
LLEO	2.14	12.05	2.59	19.01	0.08	0.43	2.22	13.45	4.43	54.44

b) Characteristic amplitude comparison

Table 3. Patient percentage above norm

CONDITION	XCOP	YCOP	TORQUE	RADIUS	dr/dt
FAEO	-21%	0%	-3%	-17%	4%
FAEC	163%	0%	-25%	7%	41%
FTEO	-21%	8%	16%	-13%	8%
FTEC	53%	8%	62%	0%	11%
TDEO	7%	-13%	22%	0%	1398%
TDEC	16%	16%	-26%	9%	921%
RLEO	8%	17%	-23%	-6%	659%
LLEO	7%	17%	9%	0%	882%

a) Characteristic frequency comparison

CONDITION	XCOP	YCOP	TORQUE	RADIUS	dr/dt
FAEO	8%	97%	-20%	147%	67%
FAEC	332%	289%	900%	301%	681%
FTEO	120%	15%	200%	79%	98%
FTEC	574%	288%	450%	586%	723%
TDEO	1305%	914%	300%	1743%	3634%
TDEC	565%	720%	467%	1889%	3469%
RLEO	1310%	1161%	107%	1065%	3123%
LLEO	463%	634%	438%	506%	1129%

b) Characteristic amplitude comparison

conditions. The second peak amplitude increased dramatically with eyes closed thus producing two dominant peaks. The program identified the second peak amplitude and frequency, even though the peak near 0.54 Hz still exists also. There are two large peaks and two dominant modes of oscillation. Since this is for Xcop which describes M/L sway, the peaks correspond to both the hip and ankle strategies of maintaining balance. Without inspection of the distribution plots, the conclusion could have been drawn that the patient had a much larger peak frequency for ankle only strategy in the FAEC condition, rather than the correct observation that the patient increased both ankle and hip strategy amplitudes with eyes closed. This observation could be very important in determination of proper treatment for the patient.

CHAPTER VI
CONCLUSIONS

The purpose of this thesis was to determine the most informative and reliable means of measuring postural balance, develop a feasible testing protocol for clinical use, and explore differences between normal and balance impaired subjects. Several balance parameters were evaluated, which included experimentally proposed measures such as frequency domain parameters as well as currently accepted ones. In addition, a parameter's sensitivity with respect to a change in condition was defined and used as a comparison measure.

Force platform ground reaction forces and moments were collected for nineteen normative subjects and over one hundred patients with possible balance impairment. Data were then analyzed for one norm and one balance impaired subject, both in the time and frequency domains, to isolate which parameters more effectively identify postural instabilities.

In time domain, the conventional "spot stabilogram" method of balance evaluation, the cross plot of X and Y COP, was shown to have some inadequacies. This parameter was found to be less sensitive, as defined, in identifying balance instabilities than dynamic measures such as radial velocity or ground reaction torque. The dynamic measures also allow for proper comparison within a subject or inter-subject, regardless of condition or foot size.

The frequency distribution of the five balance parameters proved to be another way to evaluate balance. Frequency parameters were interpreted using a dynamic model which consisted of a second order mass-spring-damper system. The model correlated the amount of damping within the system to the amount of postural control exhibited by the subject. Using the proposed model, the frequency distribution plots could be interpreted as to balance strategy according to the number of peaks present, peak frequency, peak amplitude and the number of modes of oscillation present. These values were compared for the balance patient and a norm to evaluate balance instabilities. Relationships were proposed between postural stability and peak frequencies and amplitudes. Patient peak amplitudes were larger than norm for all conditions except one, corresponding to decreased postural stability.

The methodology presented meets all the requirements as a useful clinical evaluation. The test procedure is uncomplicated, yet thorough, and is easily compared with norm and can be presented in a clear manner for physician review. In addition, the parameters were found to be more sensitive than previously used parameters and may characterize each patient's unique postural control system in terms of balance strategy, peak amplitude and peak frequency. The evaluation provided quantifiable data and adds valuable information to qualitative approaches currently used.

In addition to fulfilling the objectives of this thesis, many new areas of research were found. Research should be conducted on identification of postural strategies used in each condition. Addition of EMG data would offer insight to muscle recruitment and allow development of relationships between muscle activity and the various force platform parameters. Kinematic data may serve very useful to obtain relationships between specific body movements and force platform parameters, such as pelvic movement to the ground reaction torque. All three types of data combined together would allow a more complete understanding of the postural control system in quiet stance or in response to perturbations. Initially, this type of research would be very cost and labor intensive and is not currently considered to be a feasible test for clinical use.

The importance of every new clinical tool is the benefit of the test to physicians, therapists, and patients. The information provided by this type of postural evaluation has various applications. Determination of appropriate treatment and rehabilitation focus, evaluation during rehabilitation to monitor progress, and use as a screening device for toxic exposure are just a few of its many applications.

APPENDICES

APPENDIX A

APPENDIX A

Balance and Stability Questionnaire

BIOMECHANICS EVALUATION LABORATORY
St. Lawrence Health Science Pavilion
2900 Hannah Blvd., Suite B-114
East Lansing, MI 48824
517/336-4560

BALANCE AND INSTABILITY QUESTIONNAIRE

Date:______________________________

Name:______________________________ Birthdate:____________ Sex:______

HISTORY:

1. Specifically, what is your main concern or complaint? Please describe the symptoms you are experiencing.

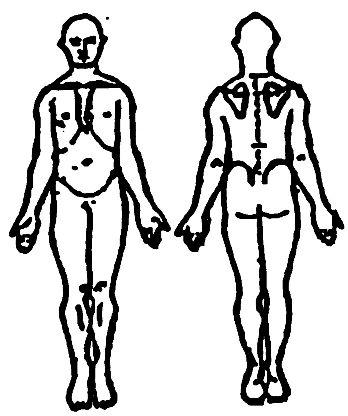
____

____

____

____

2. If pain or numbness is part of your complaint, please shade the painful or numb area(s).



3. When did your symptoms begin? ___at birth ___slowly ___abruptly

Explain: ____

____

____

4. Have you ever had anything similar before? _____yes _____no

Balance and Instability Questionnaire
Biomechanics Evaluation Laboratory
Page 2

5. Have you had a previous muscle, bone, cartilage, ligament, or nerve injury? _____yes _____no

6. Prior to this episode, were you completely symptom free? _____yes _____no

7. Are your symptoms:

Getting better?	_____yes	_____no	___a.m.	___p.m.
Getting worse?	_____yes	_____no	___a.m.	___p.m.
Not changing?	_____yes	_____no	___a.m.	___p.m.

8. Are you better with:

Rest?	_____yes	_____no
Activity?	_____yes	_____no
Neither?	_____yes	_____no

9. During the following activities, are your symptoms:

	Better	Worse	Unchanged
Bending	___yes___no	___yes___no	___yes___no
Lifting	___yes___no	___yes___no	___yes___no
Laying	___yes___no	___yes___no	___yes___no
Sitting	___yes___no	___yes___no	___yes___no
Rising	___yes___no	___yes___no	___yes___no
Standing	___yes___no	___yes___no	___yes___no
Walking	___yes___no	___yes___no	___yes___no

10. Is this complaint most aggravating or debilitating at:

Work?	_____yes	_____no	___a.m.	___p.m.
Daily Activities?	_____yes	_____no	___a.m.	___p.m.
Recreation?	_____yes	_____no	___a.m.	___p.m.

11. Is this complaint causing disruption of sleep? _____yes _____no

12. Do you have any loss of bladder or bowel control? _____yes _____no

13. Have you ever had, or do you now have, any of the following conditions?

Arthritis	_____yes	_____no
Cancer	_____yes	_____no
High Blood Pressure	_____yes	_____no
Heart Problems	_____yes	_____no
Diabetes	_____yes	_____no
Nerve Injury	_____yes	_____no
Cerebral Palsy	_____yes	_____no
Muscular Dystrophy	_____yes	_____no
Multiple Sclerosis	_____yes	_____no

14. Is there any possibility that you are pregnant? _____yes _____no

15. Do you smoke? _____yes _____no

16. Do you use alcohol or recreational drugs? _____yes _____no

Balance and Instability Questionnaire
Biomechanics Evaluation Laboratory
Page 3

17. Have you fallen in the past 12 months? _____yes _____no

Frequency of falls? ____

18. Do you feel muscle weakness, muscle spasticity, or have problems in walking or standing? _____yes _____no

If so, on which side does the problem occur? _____right _____left

Describe: ____

____

____

19. Have you ever had any "loss of consciousness" (LOC)? _____yes _____no

If yes, how long was the LOC? ________________________________
(minutes/hours/days/months)

20. Was there any "post-traumatic amnesia" (PTA)? (Loss of memory after traumatic brain injury with loss of consciousness). _____yes _____no

If yes, how long was the PTA? ________________________________
(minutes/hours/days/months)

21. What type of athletic activities do you do?

	Presently	In the Past
Weight lifting	___	___
Running	___	___
Tennis	___	___
Basketball	___	___
Soccer	___	___
Swimming	___	___
Biking	___	___
Dance	___	___
Snow skiing	___	___
Aerobics	___	___
Other ________________		

PREVIOUS TREATMENT: What previous treatments have recently been tried for this problem?

Surgery	_____yes	_____no
Nerve Blocks	_____yes	_____no
Injections	_____yes	_____no
Brace/Support	_____yes	_____no
Exercise	_____yes	_____no
Traction	_____yes	_____no
Manipulation	_____yes	_____no
Medications	_____yes	_____no

Balance and Instability Questionnaire
Biomechanics Evaluation Laboratory
Page 4

Prosthesis	____yes	____no
Shoes	____yes	____no
Assistive Devices	____yes	____no
Physical Therapy	____yes	____no
Intracranial Pressure Monitoring	____yes	____no
Steroid Administration	____yes	____no
Mannitol Administration	____yes	____no
Other____________		

PREVIOUS TESTS: Which of the following tests have been done for your problem(s)?

X-rays	____yes	____no
Myelogram	____yes	____no
Discogram	____yes	____no
CAT scan	____yes	____no
EMG	____yes	____no
MRI	____yes	____no
Lab Tests	____yes	____no
Nerve Conduction Test	____yes	____no
Work Tolerance	____yes	____no
Biomechanical Test	____yes	____no
Doppler	____yes	____no
Bone Scan	____yes	____no
Other____________		

SURGERIES: List all surgeries you have had:

Type of Surgery	Date

ALLERGIES: List allergies to foods, medications, and other substances.

____

____

MEDICATIONS: List all medications currently being taken.

____

____

____

____

COMMENTS: ____

____

____

____

____

____

THE INFORMATION PROVIDED HERE IS USED IN STRUCTURING YOUR BIOMECHANICS EVALUATION. WE APPRECIATE YOUR RETURNING THIS QUESTIONNAIRE AS SOON AS POSSIBLE <u>PRIOR</u> TO YOUR SCHEDULED APPOINTMENT.

APPENDIX B

APPENDIX B

Mathematical Development of Gain versus Frequency Ratio Expression

The equation of motion can be written from the free body diagram shown in Figure 8 as follows

$$m\ddot{x} + c\dot{x} + kx = F_o \sin \omega t$$

The assumed particular solution for this differential equation will be differentiated twice

$$x = X \sin(\omega t - \phi)$$
$$\dot{x} = \omega X \cos(\omega t - \phi)$$
$$\ddot{x} = -\omega^2 X \sin(\omega t - \phi)$$

where X is the amplitude of oscillation and ϕ is the phase shift. Substitute into the equation of motion to obtain

$$-m\omega^2 X \sin(\omega t - \phi) + c\omega X \cos(\omega t - \phi) + kX \sin(\omega t - \phi) = F_o \sin \omega t$$

Recall,

$$\sin(\omega t - \phi) = \sin \omega t \cos \phi - \cos \omega t \sin \phi$$
$$\cos(\omega t - \phi) = \cos \omega t \cos \phi + \sin \omega t \sin \phi$$

Substitute the above identities and equate the coefficients for $\sin \omega t$ and $\cos \omega t$. Thus obtaining the system equations:

$$X[\cos\phi(k - m\omega^2) + c\omega \sin\phi] = F_o$$

$$X[\sin\phi(m\omega^2 - k) + c\omega \cos\phi] = 0$$

Solving the above system for X, and ϕ (1, 2)

$$X = \frac{F_o}{\cos\phi\left(k - m\omega^2\right) + c\omega \sin\phi}$$

$$X = \frac{F_o / \cos\phi}{k - m\omega^2 + c\omega \tan\phi}$$

From the second system equation we obtain an expression for the phase angle

$$\phi = \tan^{-1}\left(\frac{c\omega}{k - m\omega^2}\right) \qquad (1)$$

$$X = \frac{F_o / k}{\sqrt{\left(1 - \frac{m\omega^2}{k}\right)^2 + \left(\frac{c\omega}{k}\right)^2}} \qquad (2)$$

Define

$$\omega_n = \sqrt{\frac{k}{m}}$$

$$c_c = 2m\omega_n$$

$$\zeta = c/c_c$$

$$\frac{c\omega}{k} = 2\zeta\frac{\omega}{\omega_n}$$

Utilizing the above definitions in equations 1 and 2 , the gain and phase angle expressions become

$$\frac{Xk}{F_o} = \frac{1}{\sqrt{\left(2\zeta\frac{\omega}{\omega_n}\right)^2 + \left(1-\left(\frac{\omega}{\omega_n}\right)^2\right)^2}}$$

$$\tan\phi = \frac{2\zeta\left(\frac{\omega}{\omega_n}\right)}{1-\left(\frac{\omega}{\omega_n}\right)^2}$$

APPENDIX C

APPENDIX C

Spectral Distribution Plots for Norm and Patient

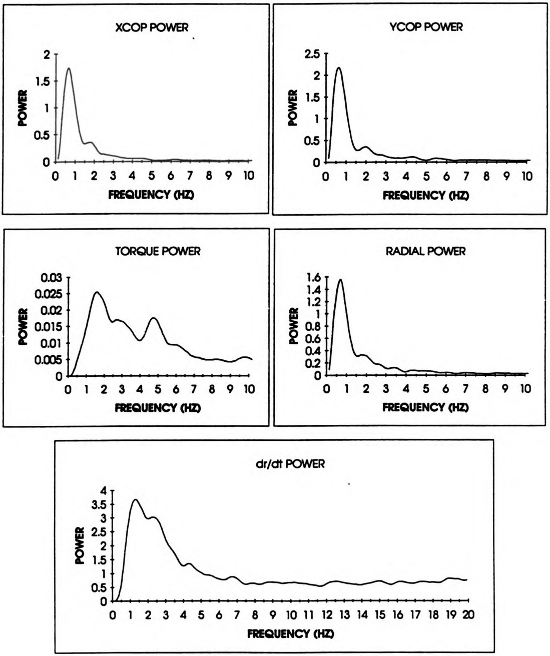


Figure 16. Norm spectral distribution for FTEO

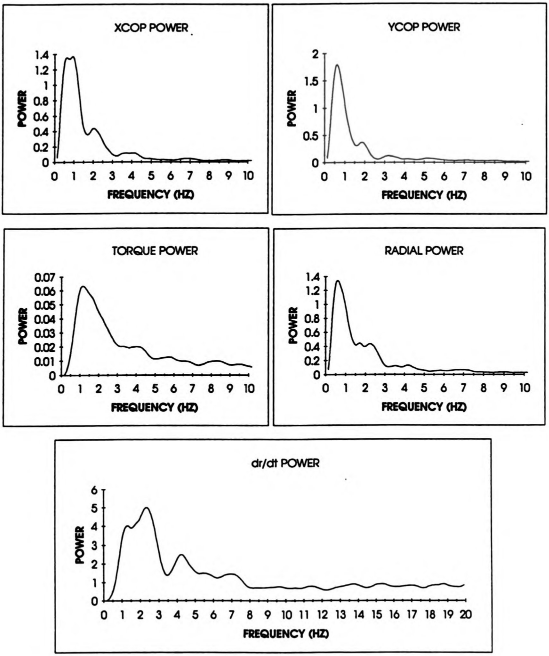


Figure 17. Norm spectral distribution for FTEC

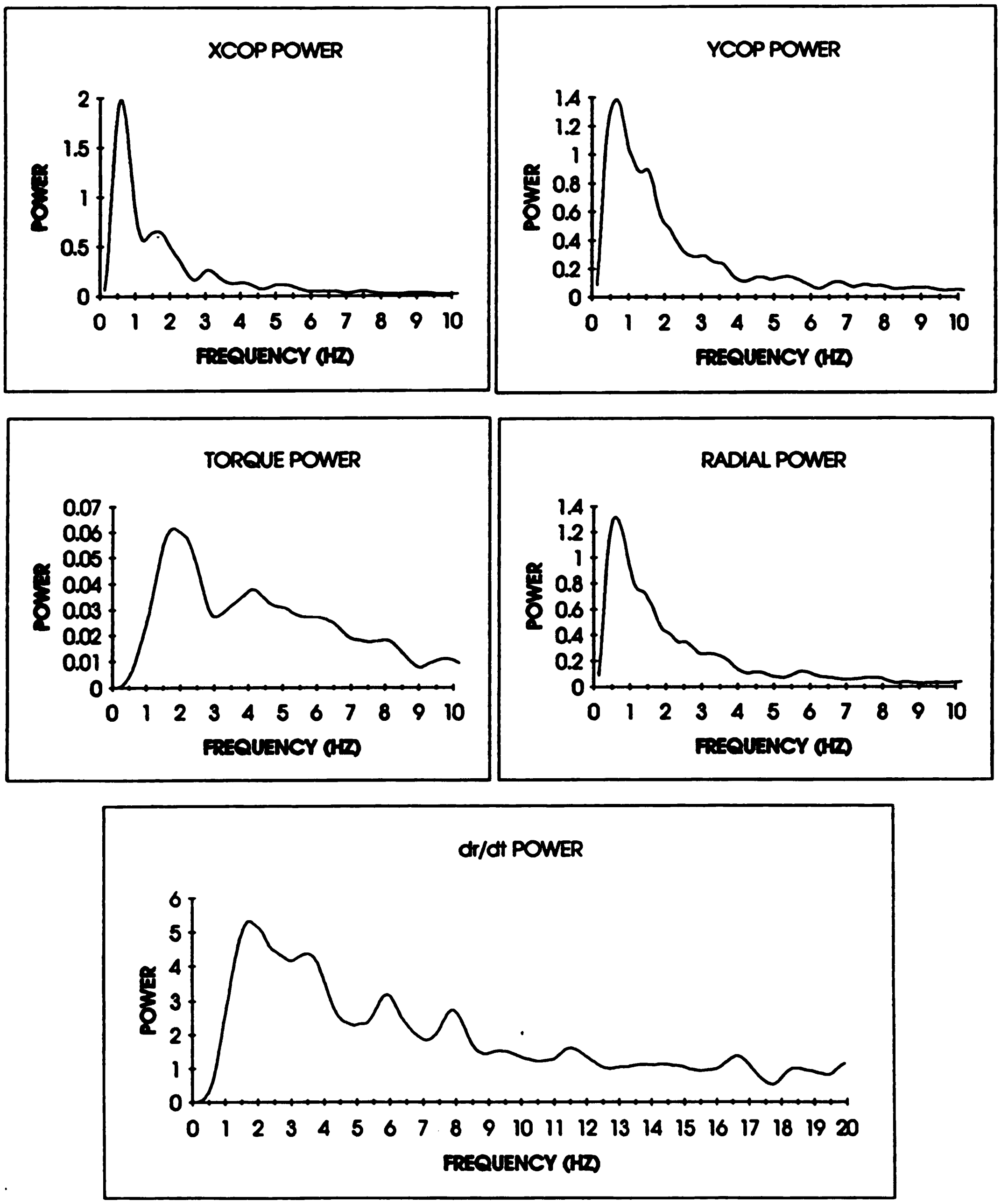


Figure 18. Norm spectral distribution for TDEO

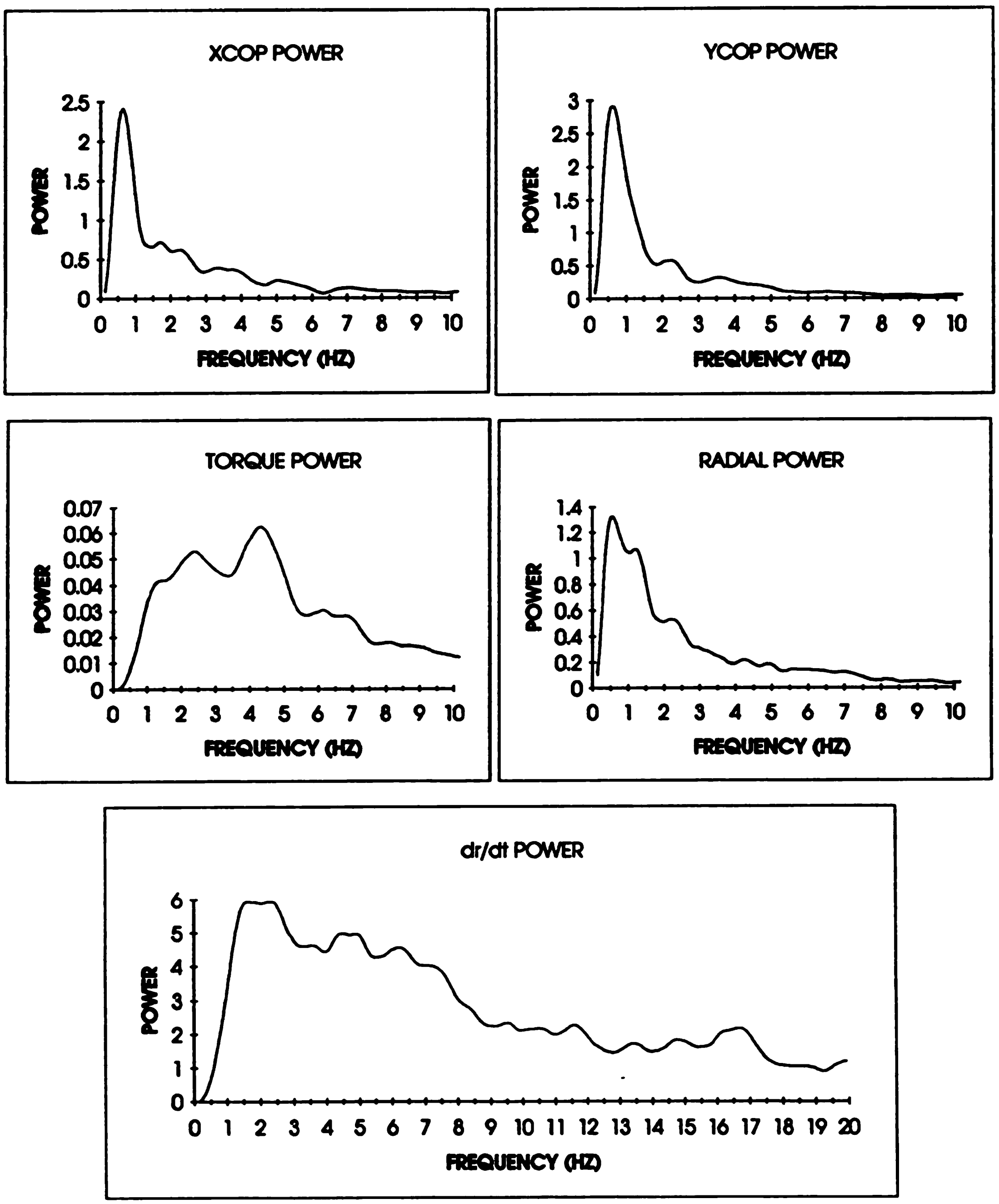


Figure 19. Norm spectral distribution for TDEC

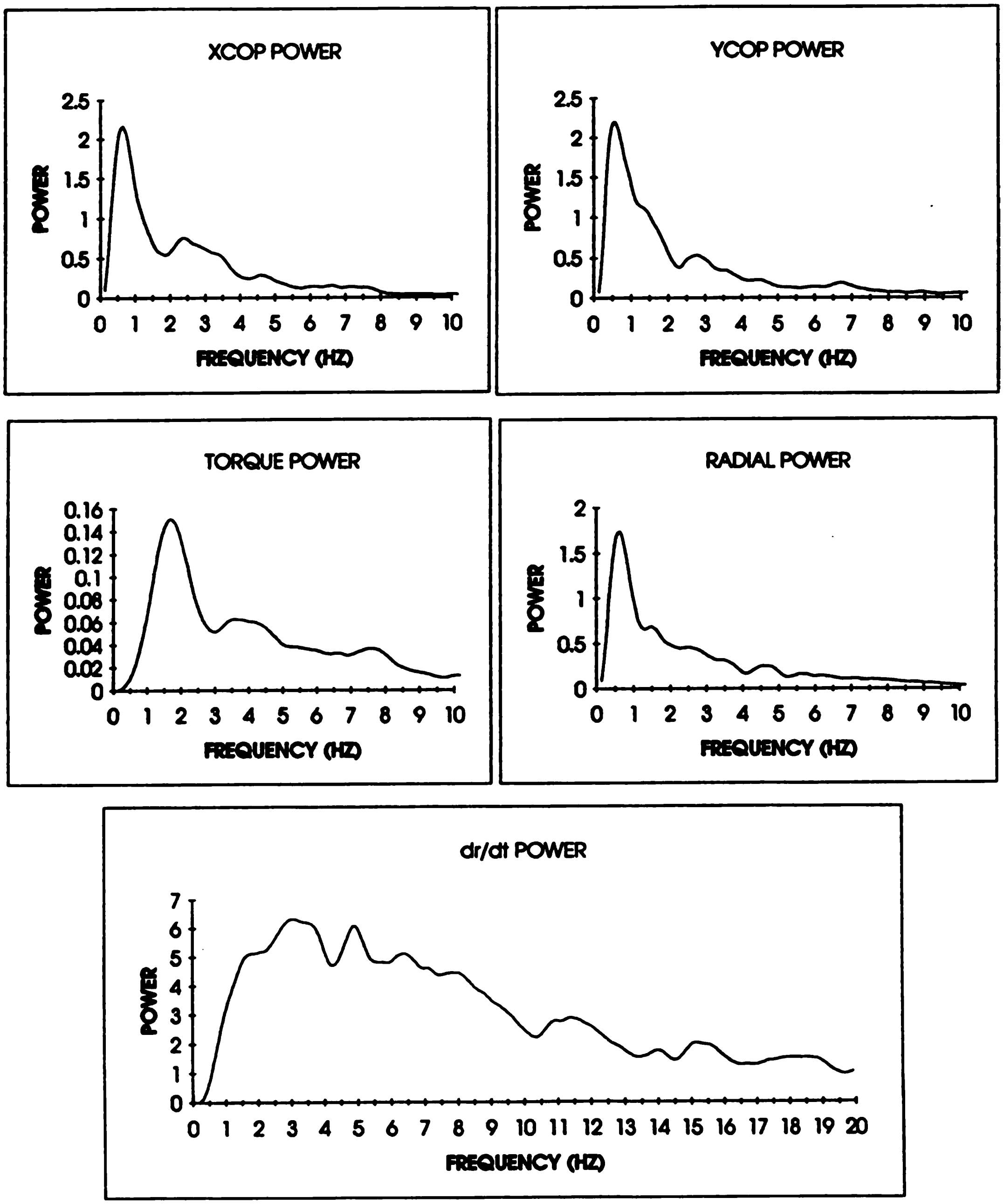


Figure 20. Norm spectral distribution for RLEO

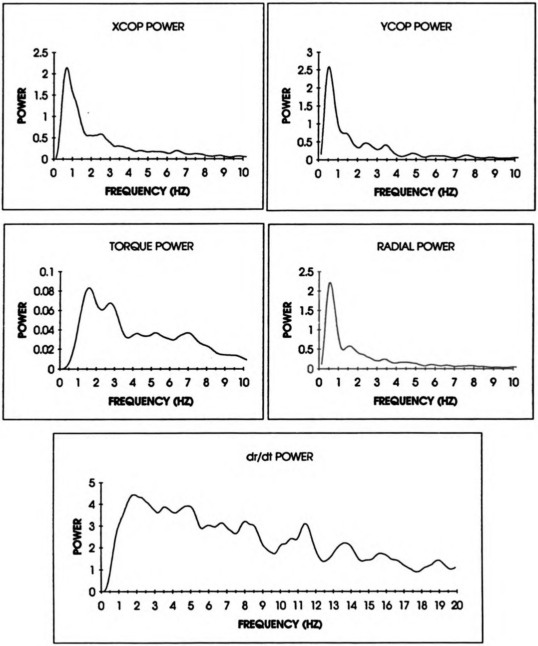


Figure 21. Norm spectral distribution for LLEO

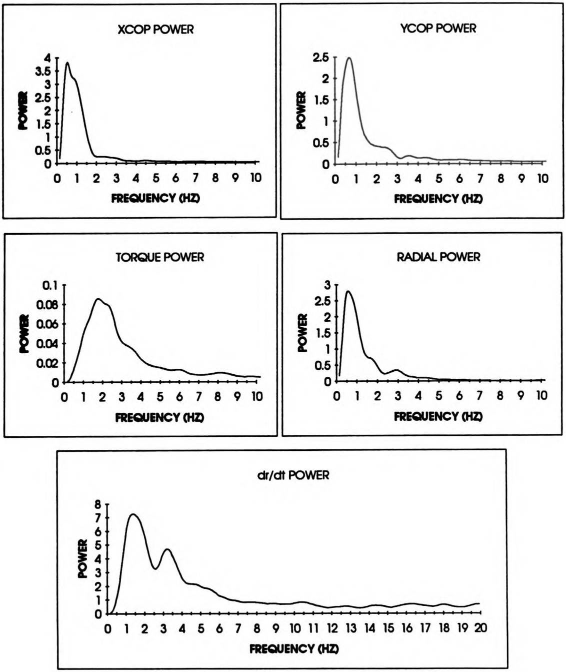


Figure 22. Patient spectral distribution for FTEO

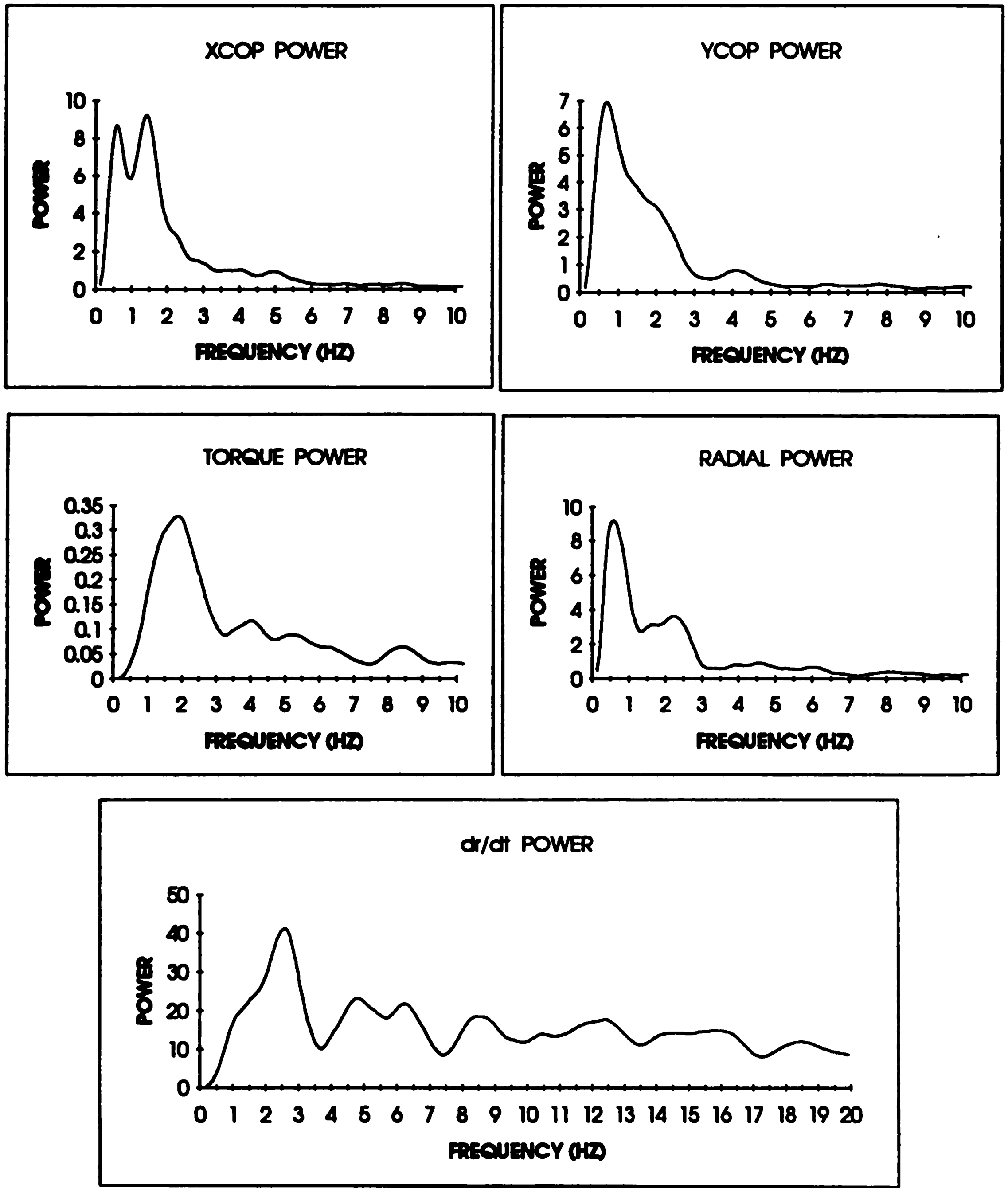


Figure 23. Patient spectral distribution for FTEC

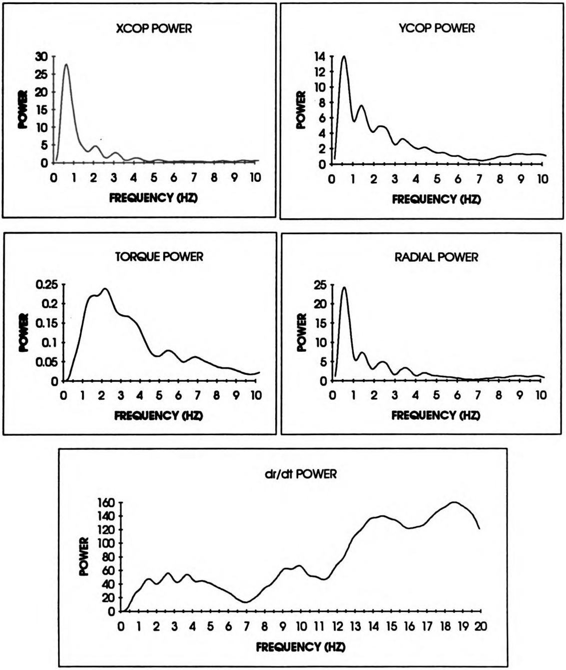


Figure 24. Patient spectral distribution for TDEO

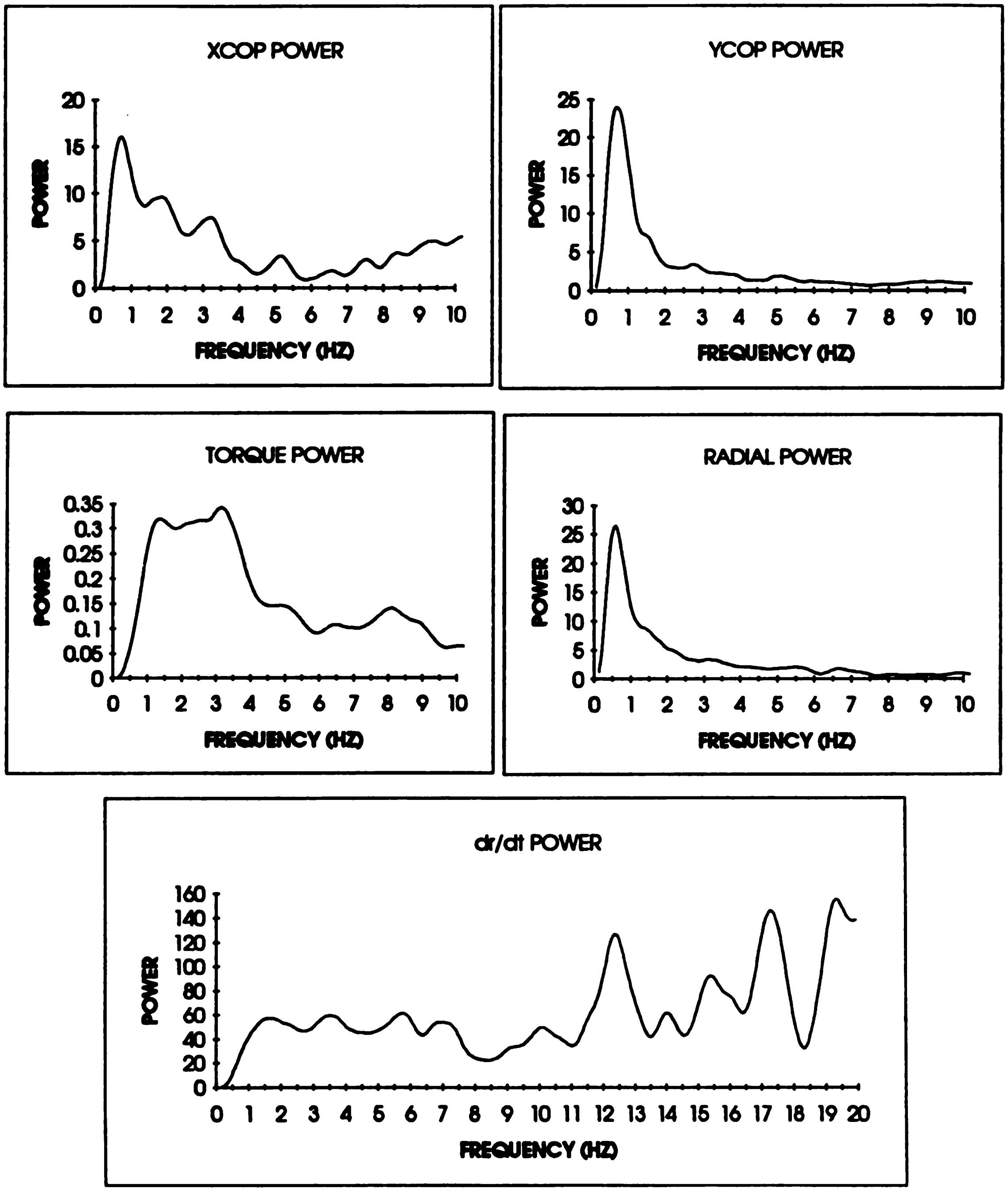


Figure 25. Patient spectral distribution for TDEC

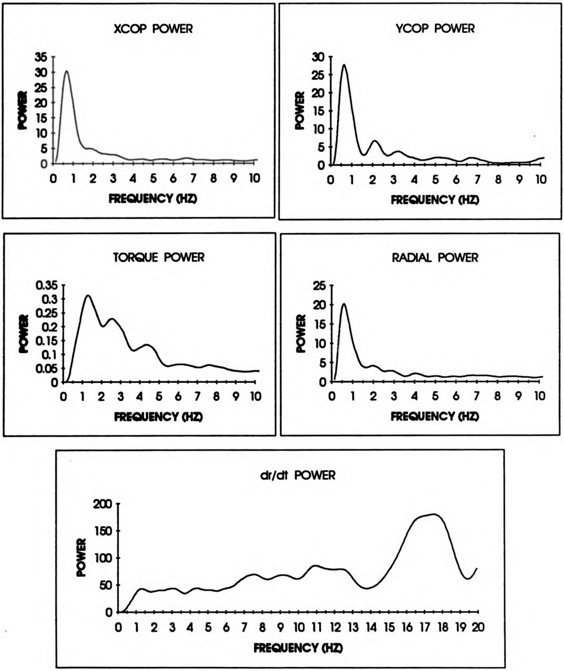


Figure 26. Patient spectral distribution for RLEO

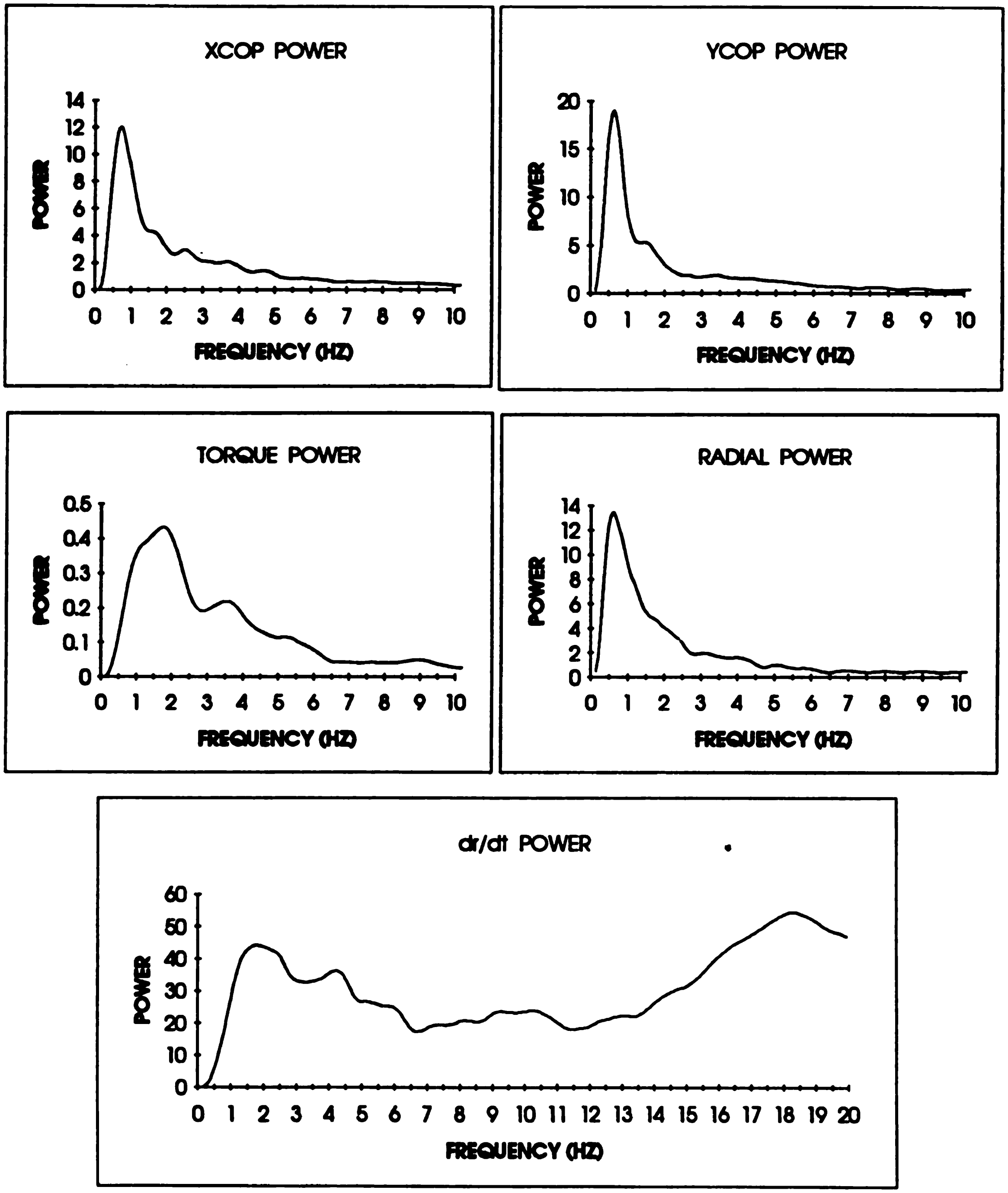


Figure 27. Patient spectral distribution for LLEO

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