A KINEMATIC AND KINETIC ANALYSIS OF POSTURAL PERTURBATION

A Thesis in
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by
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Abstract

This study was designed to quantify postural reaction to a sudden toe-up rotational perturbation by twenty-two healthy young adults (eleven males, eleven females, average age of 26 years, height of 1.72 m, and weight of 649.62 N) with no physical or physiological conditions that would compromise their ability to react to the perturbation. The goal was to achieve an understanding of the mechanical requirements necessary for maintenance of balance and stance.

The human body was modeled as a system of five rigid segments. The act of reacting to a movement of the ground that suddenly dorsiflexed the ankle was monitored by a video analysis system operating at 60 Hz. Special care was taken to maximize marker resolution by covering the action with several cameras each zoomed in on part of the body. Segmental kinematics and kinetics were calculated after numerically filtering at 6 Hz and differentiating the digitized location of retroreflective markers attached to the subjects’ skin. Because no ground reaction data was available, kinetic calculations were performed in a head to toe direction.

Analysis of the segmental reactions to the postural perturbation revealed that: (1) All segments and joints studied were utilized in a wave like motion and the reaction was not limited to a hip or an ankle strategy. (2) The horizontal components of the ankle joint reaction force and knee joint reaction force were of approximately the same magnitude and approximately twice the magnitude of the hip joint reaction force. The vertical component of the ankle joint reaction force was larger than the knee joint reaction force, which was greater than the hip joint reaction force. The ankle joint produced the largest moment in response to the platform perturbation while the knee moment was larger then the hip moment. (3) With some reservations the larger the magnitude of the sudden angular ground perturbation distance the greater was the joint angular displacement at all joints. (4) The faster the angular velocity of the rotational perturbation the larger the joint angular displacement and peak force. Even though the peak knee torque did not increase in reaction to faster angular velocities of perturbation, the peak torque of the ankle and hip did. (5) The torques required to successfully maintain stance following a toe-up perturbation were well within the capabilities of the elderly. If any joint was to be considered the "weak link" it would be the ankle joint.
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Chapter 1
Introduction

1.1. Introduction.

Posture, the position and orientation of individual body segments, is linked to balance, the maintenance of the position of the center of gravity (CG) or the center of mass (CM) over the base of support. With few exceptions, a change in posture will change the location of the CM, and without exception, a change in the location of the CM will be accompanied by a change in posture. For many years, scientists have been striving to understand how humans control posture in order to maintain balance in various environmental conditions, and how this control is affected by aging, disease and different psychological and physiological conditions. For example, researchers have studied how vestibular deficits (Allum, Keshner, Honegger and Pfalz, 1988; Horak, Nashner and Diener, 1990; and Horstmann and Dietz, 1988), multiple sclerosis (Panzer-Decius and McFarland, 1992), tardive dyskinesia (van Emmerik et al., 1993), Parkinson’s disease (Allum, Keshner and Wuehrich, 1988; Beckley, Bloem, Dijk, van Roos and Remler, 1991; Bloem, 1992; Diener, Dichgans, Guschtlbauer, Bacher, Rapp and Langenbach, 1990; Dietz, Berger and Hortmann, 1988; and Dietz, Zijlstra, Assiairite, Trippel and Berger, 1993), Friedreich’s disease (Diener, Dichgans, Bacher and Guschtlbauer, 1984), spinal and supraspinal lesions (Diener, Ackermann, Dichgans and Guschtlbauer, 1985) and diabetes (Cavanagh, Simoneau and Ulbrecht, 1993) influence balance. However, few scientists have scrutinized the dynamics of posture and balance control in individuals that are not subjected to abnormal influences. One exception is the work of Nashner and McCollum (1985). These researchers have suggested that different strategies be used to balance in different environments and during environmental changes. However, they based their conclusions on analysis of EMG and force plate data, which provide only a limited view of human balance control. Recently, several scientists have started to directly investigate the mechanics of posture using whole body models (Alexander, Shepard, Gu and Schultz, 1992; Gu, Schultz, Shepard and Alexander, 1996; Romick-A llen and Schultz, 1988; and Yang, 1990). Unfortunately, these studies were limited to net joint torque peaks and latencies in response to mild horizontal ground movements. This author has been unable to find any studies that provide a more thorough dynamic analysis of human response to perturbations nor any dynamic analysis of responses to rotational perturbations.

This dissertation used a mathematical model of the human body to calculate full body dynamics in the sagittal plane of healthy young adult subjects reacting to an unexpected rotational movement of the ground on which they stood. The ground rotations were in the sagittal plane about an axis coincident with the ankle joints of each subject. The magnitudes of the perturbations were varied to span the range of perturbations seen in the literature. The movements of the subjects were measured by combining the video data of two synchronized cameras. Kinematics and kinetics were calculated and analyzed based on a five-link model of the human body.

1.2. Significance.

An improved understanding of postural dynamics will advance understanding of the basic function of human muscular control mechanisms. This basic understanding will lead to an improved evaluation of how aging and/or disease influence balance control. The premise for this assertion is that changes in vision, vestibular and proprioceptive abilities, physical activity, hormonal levels, alcohol and medication intake, motivation and other psychological and physiological conditions all express themselves as changes in biomechanical response. However, one cannot fully understand how these factors affect posture without first fully understanding the biomechanics of posture without these additional influences. Sutherland, Kaufman and Moitoza in their chapter of human walking state: “A principle of science and of medicine in particular, is that one must understand the normal or natural history of a studied phenomenon before attempting to describe and study the pathological or abnormal.” (1994, p. 25). A thorough understanding of the biomechanics of posture is indispensable to an understanding of the consequences of various human and environmental conditions. A basic understanding of the mechanical factors required for balance will lead to better therapeutic methods for improving posture and gait, and to strategies that can be used to
prevent falls. Because correction of postural problems often involves mechanical interventions (e.g., surgery, braces and collars), the understanding of postural dynamics is also necessary in prosthetic design, in physical and occupational therapy formulation and in surgery planning. Expertise in the biomechanics of the human response to the environment will aid in ergonomic and safety design considerations that are important in many settings, and critical in hospitals and housing for the elderly. To the point, a better knowledge of a human’s response to the environment may lead to improvement in quality of life.

1.3. Specific aims.

The goal of this dissertation was to explore the mechanical factors involved in maintaining posture following an unexpected rotational support surface perturbation. The rotations were at two velocities (approximately $30^\circ$/s and $60^\circ$/s) and two amplitudes (approximately $5^\circ$ and $10^\circ$). The response was video taped at 60 Hz and retroreflective markers indicating the location of various anatomical landmarks were digitized. The video data, along with anthropometrical data, allowed the calculation and subsequent analysis of linear and angular position, and joint force and moment.

1.4. Hypotheses.

1.4.1. Kinematics of strategy.

Knee rotation will occur in response to a sudden unexpected rotation of the ground of $5^\circ$ to $10^\circ$ at $30^\circ$/s to $60^\circ$/s. Therefore, the subjects’ strategy will be to use all lower extremity joints in response to the perturbations and not limit themselves to an ankle or hip strategy.

1.4.2. Joint moments and forces.

To maintain balance following a sudden unexpected upward rotation of the ground ranging from $5^\circ$ to $10^\circ$ at $30^\circ$/s to $60^\circ$/s, the subjects will produce greater net ankle reaction forces and moments than net knee reaction forces and moments, and will produce greater net knee reaction forces and moments than net hip reaction forces and moments.

1.4.3. Reaction to platform angular displacement.

The greater the sudden angular ground perturbation distance, the larger the joint angular displacement at all joints. However, the distance of angular perturbation will have no effect on the peak moment at any joint.

1.4.4. Reaction to platform angular velocity.

The faster the angular velocity of the rotational perturbation the larger will be the joint angular displacements, net joint forces and moments.

1.4.5. Joint torque required to maintain stance.

In reaction to the sudden rotational perturbation, the subjects will generate joint torques that are lower than the capabilities of elderly individuals as reported in the literature. That is, elderly individuals should not have difficulty in maintaining stance following a rotational perturbation due to a lack of muscular strength.

1.5. Limitations.

The following limitations were recognized:

Twenty-two healthy young adult subjects

In this study, postural control characteristics were studied in 22 healthy young adults. A health history and an evaluation of visual, vestibular and cutaneous mechanoreceptor function were obtained for
all subjects. The kinematics and kinetics of postural stability under several specific conditions were quantified.

**Net joint forces and moments**

The joint forces and moments calculated were the net joint reaction forces and moments (Winter, 1990). These were not separated into their component forces and moments due to individual joint surfaces, ligaments, tendons or other components of the joint structure. Although there can be tremendous differences between bone on bone vs. net joint force, it is the net joint force that results in an acceleration and therefore a change in posture.

**Normal sensory and motor function**

It was assumed that all subjects had normal sensory function and motor capabilities. Thus, electro-diagnostic tests were not conducted.

**Clinical tests of vestibular function**

Assessment of vestibular function was limited to a few clinical tests of spontaneous nystagmus and saccadic eye movement. The equipment required to measure eye movement by electro-oculography was not available.

**Posture in response to rotational perturbation**

This study was limited to the assessment of posture in response to sudden rotational perturbations. It has been shown that reactions to different environmental conditions are specific; thus, extrapolation of these findings to other environmental conditions is limited.

**Two-dimensional five-segment model of body**

The body was modeled as five rigid segments [foot, shank, thigh, trunk (includes arm, forearm-and-hand) and head] connected by four one degree of freedom (DOF) hinge or pin joints (ankle, knee, hip and neck). Imbedded in this model was the assumption that the left and right sides acted symmetrically.

**Rigid segments**

It was assumed that the segments were perfectly rigid and did not deform under loads. Since displacement due to deformation is negligible compared to movements of the markers due to their attachment to soft tissue, no loss is incurred with this assumption. See soft tissue movement section.

**Fixed axis of rotation**

It was assumed that the joint centers remained fixed relative to the joint markers despite the fact that most of the joints in the human body do not have a fixed axis of rotation. For example, knee joint motion is comprised of translation and rotation, and therefore the knee joint axis moves with respect to any coordinate system fixed in either the thigh or the shank. In addition, each person’s knee joint has a unique geometry. To be precise, separate modeling of each individual’s geometry would be necessary. The calculations and data collection required to characterize the actual geometry of the joint are involved. The gains in precision were thought to be too small to justify the added effort in this study.

**Transmission of forces**

It was assumed that internal forces were directly transmitted across articulated surfaces, and that articulating surfaces maintained contact at the joint center. Therefore, no torques due to offsets (non-co-linearity of forces) would exist.

**Soft tissue movement**

Displacement of the skin and other soft tissues during segmental motion allowed the markers to move relative to bony landmarks. In motions with high accelerations, marker relative motion may have been significant. In addition, soft tissue and fluid movement causes a shift in the CM and a change in the rotational inertia of a segment. This is true regardless if the segment is accelerating or not. If the segment is accelerating, inertia will cause relative displacements of tissue and a redistribution of mass within the segment. Gravity will act to redistribute mass in a non-rigid segment. Therefore, a segment’s CM location and rotational inertia properties will vary with orientation with respect to gravity. See rigid segment section.

**Body segment parameters**

Estimations of body segment parameters were determined using generalized anthropometrical formulas found in the literature.
**Video data at 60 Hz**

The body segmental motion was recorded using two synchronized video cameras each recording at a rate of 60 Hz. Thus, it was assumed that all motion had a maximum frequency component of 30 Hz or less.

**Joint calculations top down**

The joint force and moment calculations were performed from the head down because the forces between the platform and the subjects’ feet were not known.
Chapter 2
Review of Literature

2.1. Introduction.
Standing erect is one of the most basic and fundamental tasks performed by humans. The terms “basic and fundamental” are used here to indicate that erect stance is a prerequisite for many other tasks and not to imply that standing is an easy or simple task. “Either consciously or unconsciously, humans spend most of their waking hours adjusting their positions to the type of equilibrium best suited to the task” (Luttgens and Wells, 1982, p. 386). For humans, standing upright requires balancing a multi-segmental inverted pendulum through the coordination of a redundant system of muscles. To perform this coordination, the central nervous system (CNS) integrates information from sensors that make up the somatosensory, vestibular and visual systems that monitor the constantly changing status of the body. If the body is out of balance, the action or inaction of appropriate muscles in the skeletal system is modified through alterations in the signals sent down their associated motor units. If effective coordination of information received from the millions of sensors with the commands sent through millions of motor units does not make stance difficult enough, the architecture of the human body makes upright stance more difficult for the CNS for two reasons. First, a standing human is a relatively tall structure balanced on a relatively small base of support. Approximately two-thirds of your body mass including some delicate organs are precariously balanced some distance from the ground (about 2/3 of our height) over two spindly structures, the legs, which provide a narrow base of support (Winter, Patla and Frank, 1990, p. 31).

Second, humans do not stand straight. In an ideal situation, the CM of each body segment would lie directly above the center of the joint below it so that no torques would be generated by the non-co-linearity of the gravitational forces acting on the segment and the reaction force acting through the joint center. However, Steindler (1955) showed that, in stance, the joint center and the CM of each segment do not normally lie on a single line parallel with the force of gravity. The bones of the skeleton are too irregularly shaped to stand upon each other and maintain an upright posture without some outside help (Thibodeau and Patten, 1993). The most obvious example of this can be seen in a sagittal view of the spine in which each vertebra does not sit directly above the one below it. Furthermore, humans stand with a slight forward lean of four to five degrees as measured by the angle between vertical and a line through the total body CM and center of the ankle joint (Dichgans and Deiner, 1989). Ligaments and other joint structures balance some of the torques produced by this natural “mal-alignments”. Nevertheless, some muscular activity is needed to maintain erect posture. To appreciate the role of muscle in stability, consider the contrast between world-class weight lifters who can support more than 1500 N overhead to individuals without muscular control, for instance individuals with cerebral palsy or paralysis who cannot support the weight of their own body, or newborn babies who cannot support their heads.

A prerequisite to the understanding of the biomechanics of human stance requires the definition of and a short discussion of the relationships between a few basic physical principals.

2.2. Equilibrium, Stability, Balance and Posture.
In the posture literature, there exist many definitions of equilibrium, stability and balance due to the diversity of the scientific backgrounds of researchers studying posture. To eliminate any confusion the definitions and relationships between equilibrium, stability, balance and posture need to be discussed in some detail.

For an object to be in equilibrium, two conditions must be met:
(1) Sum of all external forces acting on an object must equal zero, and
(2) sum of all external moments acting on an object must equal zero.

Given these two conditions, an object is in equilibrium when it is at rest or in uniform motion (i.e., when it has constant linear and angular momentum) (Hemami and Stokes, 1983).
Stability is a term used to differentiate three different kinds of equilibrium: stable equilibrium, unstable equilibrium and neutral equilibrium. A body is stable if, after the body is slightly displaced from equilibrium, it returns to the initial equilibrium condition. For instance, stable equilibrium exists when an object under the influence of gravity is disturbed its center of gravity is raised. To illustrate this, consider a point mass at the bottom of a depression (figure 2.2.1a). If the mass were disturbed or displaced to the left or right and then released, the external forces acting on the mass (gravity and the reaction force due to contact with the surface) would tend to move the mass toward the bottom of the depression, and the point would return to the bottom center of the depression after the disturbance. At the bottom of the depression the external forces acting on the mass balance (equilibrium), while at any other point, the external forces are unbalanced in a direction that drives the mass back to the equilibrium position. An object is in a situation of unstable equilibrium if, upon being slightly displaced from equilibrium, it moves away from that equilibrium (figure 2.2.1b). In this situation, the external forces are balanced only when the mass is at the top. At any other point, the external forces are unbalanced in a direction that drives the mass away from the equilibrium position. In the third situation (figure 2.2.1c) the external forces balance regardless of the position of the point mass. This situation illustrates an example of neutral equilibrium.

From these examples, one can see that an object, under the influence of gravity, is in a condition of:

- stable equilibrium if the center of gravity of the object rises with displacement from equilibrium
- unstable equilibrium if the center of gravity of the object drops with displacement from equilibrium
- neutral equilibrium if the center of gravity of the object does not rise or fall with displacement from equilibrium

If all the external forces acting on the object are conservative (as in the previous example of a particle under the influence of gravity and without the influence of any frictional forces) a potential energy function \((U)\) can be defined such that:

\[
F_x = -\frac{\partial U}{\partial x} \quad F_y = -\frac{\partial U}{\partial y} \quad F_z = -\frac{\partial U}{\partial z} 
\]  
(Equation 2.2.1)

Equation 2.2.1 means in words that the net external force \((F_x, F_y, F_z)\) is pointed in the direction \((\partial x, \partial y, \partial z)\) of decreasing potential energy \((-\partial U)\), or an increase in potential energy with a change in position will
be met with an oppositely directed force (F\textsubscript{x}, F\textsubscript{y}, or F\textsubscript{z}). At points where \( \frac{\partial U}{\partial x} \) equals zero (points where there is no change in potential energy with position), F\textsubscript{x} will equal zero, and the particle will be in equilibrium in the x direction (similarly for the y and z directions). The derivative of the potential energy will be zero when the potential energy is at an extreme value (maximum or minimum) or when the potential energy is constant. Therefore, when the potential energy (U) is at a minimum \( \frac{\partial^2 U}{\partial x^2} \) is positive so \( \frac{\partial^2 U}{\partial x^2} \) is negative and therefore \( \frac{\partial F}{\partial x} \) is negative. Thus, any displacement from this position (\( \partial x \)) must be accompanied by a force in the opposite direction, or a restoring force that tends to return the particle to the equilibrium position. Another way of looking at the situation is that because force and displacement are oppositely directed in conditions of stable equilibrium, work must be done on the particle by an external force to change its position, resulting in an increase in the particle’s potential energy (figure 2.2.1a).

When U is at a maximum, \( \frac{\partial^2 U}{\partial x^2} \) is positive, the particle is in an unstable equilibrium, and any displacement from this position results in a repelling force tending to move the particle away from the equilibrium position. If a particle is in unstable equilibrium, work will be done by the particle in changing the position of the particle. A position change results in a decrease in potential energy (figure 2.2.1b).

When U is constant, \( \frac{\partial^2 U}{\partial x^2} \) is zero, the particle is in a neutral equilibrium and the particle can be displaced without incurring a restoring or repelling force (figure 2.2.1c).

Note that the equilibrium conditions act independently along each direction or coordinate. Therefore, a particle may simultaneously be in stable, unstable and neutral equilibrium. For example, a particle may be in stable equilibrium with respect to the x coordinate, unstable with respect to the y coordinate and neutral with respect to the z coordinate.

Because the human body is not a particle or a point mass without volume its relationship with respect to equilibrium depends on its orientation or posture. Rather than a point mass, a more representative characterization of the human body is a series of linked rigid bodies. For a single rigid body, the equilibrium conditions are evaluated at the CM. While for a series of rigid bodies, the equilibrium conditions are evaluated by a weighted sum of the rigid bodies or at the CM of the collection of all linked bodies. For a single rigid body or object, both translational and angular displacements may change the position of the body’s CM and therefore its potential energy. Consider rotations of three differently shaped rigid bodies, under the influence of a gravitational force, and resting on a flat surface (figure 2.2.2). If an angular displacement from equilibrium causes the CM of the rigid body to rise (an increase in potential energy), the gravitational force will generate a “restoring” torque toward equilibrium. In this instance, the rigid body is in a stable equilibrium (figure 2.2.2a). If an angular displacement causes the CM to drop, the potential energy decreases and the gravitational force will pull the body away from equilibrium. Here the body is in an unstable equilibrium (figure 2.2.2b). If an angular displacement does not alter the height of the CM the potential energy does not change with displacement and the body is in a neutral equilibrium (figure 2.2.2c).
Like a particle, the shape of the surface with which a rigid body rests also determines whether a rigid body is in a stable, unstable, or neutral equilibrium condition. In figure 2.2.2, if each rigid body were to slide along the flat surface without any rotation there would be no change in potential energy and all three rigid bodies would be in neutral equilibrium. However, if each object were placed and then translated along a curved surface such as that of figures 2.2.1a and 2.2.1b, change in potential energy would occur and the object would be in stable or unstable equilibrium depending on whether the CM rose or fell respectively. Thus, the stability of a particle is determined solely by the shape of the surface on which it rests, while the stability of a rigid body is dependent on both the shape of the surface and on the shape of the rigid body. It is possible to vary the equilibrium of a rigid body by modifying the shape of the surface on which it rests. For example, it is possible to construct a surface for which a rotating square will be in neutral equilibrium rather than stable or unstable equilibrium (figure 2.2.3).

Instead of evaluating the vertical displacement of a rigid body’s CM to determine its stability, one can evaluate the horizontal relationship between the rigid body’s CM and its base of support. An object’s base of support includes all points of contact with the ground as well as all points within a border drawn around the most extreme contact points. For example, the base of support for a three-legged chair includes the area between the legs in addition to the area under each leg (figure 2.2.4 left), and the base of support for a human standing on two feet encompasses the area under both feet as well as the area between them (figure 2.2.4 right). Note that this is the maximum area of the base of support for a human. The area under and between both feet is the base of support only if all the joints in the body are locked (Howe and Oldham, 1997). For example, if the ankle joints are not locked then the area of contact between the talus and tibia defines the area of the base of support.
Equilibrium requires that a vertical line (line of gravity) pass through both the CM and base of support of the object. That is the horizontal position or coordinates of the CM must be within those of the base of support. Once the line of gravity passes beyond the margin of the base of support, stability is lost and a new base of support must be established or the object will fall.

At this point, the term center of pressure (CP) must be defined. Center of pressure is the weighted average or center of all the contact forces between an object and its environment. The CP will be located at the geometric center of the base of support if the contact forces are equal at every point of the base of support or they are symmetric about the geometric center of the base of support. That is the CP is located at the center of the base of support if the forces under each foot of the three-legged chair or the human, in figure 2.2.4, are equal. If the force on one of the feet (or part of one of the feet) is larger than that on the others the CP will be biased toward the area of high force. The CP can also be defined as the location of the point of application of the concentration of all the external contact forces. That is, a mechanically equivalent situation exists if all the contact forces were concentrated and applied to a rigid body at its CP. This is similar to the concept of applying the sum of the gravitational forces at the CM or CG of an object.

Returning to figure 2.2.2b, the base of support of the triangle is a point and does not change location with rotation. Thus with rotation of the triangle the location of the CM moves laterally, while the CP (which is coincident with the base of support) does not. Hence with increasing rotation of the triangle, the horizontal distance between the CM and the CP increases, and the non-co-linearity between the gravitational forces concentrated at the CM and the contact forces concentrated at the CP increases. Thus a destabilizing torque develops and increases with increasing rotation of the triangle. The triangle in this situation is unstable, and its stability decreases with increasing rotation. In the condition depicted in 2.2.2a, any rotation will move both the CM and the CP laterally in the same direction. However, the CP initially moves to a greater extent than does the CM. The torque generated by the non-co-linearity of the contact and gravitational forces will tend to force the triangle back to equilibrium. Note that after the initial rotation where the CP moves from the center of the base of the triangle to the apex, it stops moving and remains at the apex. That is with continued rotation, the CP remains stationary while the CM moves horizontally toward it. Thus, the torque tending to twist the triangle back to equilibrium decreases with additional rotation. At the instant when the CM passes above the CP, the torque ceases to act to restore the triangle to its original equilibrium but away from it. At this point, the CM has passed beyond the original base of support and the triangle is no longer in the realm of stable equilibrium. In the neutral equilibrium condition of figure 2.2.2c, the CM and the CP move equal distances with rotation. Thus, determination of stable, unstable and neutral equilibrium can be made by analyzing the relative horizontal motion of the CM and the CP. Stable equilibrium occurs when the CM and the CP move in the same lateral direction and the CM moves less than the CP. Unstable equilibrium will occur when the CM and the CP move in opposite directions, or they move in the same direction with the magnitude of the CM displacement being larger than the CP displacement. Neutral equilibrium exists when the CM and the CP have equal displacements (both magnitude and direction).

By combining this information, some generalizations of the degree of stability can be made. For instance, if a stable situation exists the wider the base of support or the lower the CM of an object, the more stable the object is. Consider a quadrilateral that is rotated in a uniform gravitational field as shown in figure 2.2.5. The height of its CM is H, the distance between its CM and the base of support is h and the length of the base of support is b.
After rotation, the potential energy changes by $\Delta U$ which is proportional to $\Delta H = H_f - H_i$ or

$$\Delta U \propto (h \sin \theta_f + \frac{b}{2} \cos \theta_f) - (h \sin \theta_i + \frac{b}{2} \cos \theta_i) \quad \text{(Equation 2.2.2)}$$

Rearranging terms

$$\Delta U \propto h(\sin \theta_f - \sin \theta_i) + \frac{b}{2}(\cos \theta_f - \cos \theta_i) \quad \text{(Equation 2.2.3)}$$

The constant of proportionality is the force due to gravity. With the limits $0 \leq \theta \leq 90$ and $\theta_f \leq \theta_i$, $(\sin \theta_f - \sin \theta_i) \leq 0$ and $(\cos \theta_f - \cos \theta_i) \geq 0$. Therefore, holding the distance between the base of support and the CM constant (h) while increasing the base of support (b) increases $\Delta U$ and increases stability.

Another way of analyzing the relationship between base of support and CM is to consider the amount of rotation an object has to go through before the CM is no longer over the base of support. Consider the three rectangles in figure 2.2.6. The first and second rectangles have the same base of support, but the height of the CM is higher in the first rectangle. The first rectangle is less stable than the second because, the amount of rotation that would have to occur for the CM to no longer be over the base of support is less than for the second rectangle ($\theta_1 < \theta_2$). The height of the CM is the same in the second and third rectangles. However, the base of support is larger in the third rectangle and thus, the amount of rotation that would have to occur for the CM to no longer be over the base of support is larger in the third rectangle than it is in the second ($\theta_2 < \theta_3$).

In addition to the size of the base of support, the shape of the base of support and the location of the CM projection on the base of support also influence stability. The larger the range of movement that the projection of the CM has within the area of the base of support, the greater the stability. Thus, if the projection of the CM is near an edge of the base of support there will be minimal stability in the direction toward the close edge and maximal stability in the direction away from the close edge. That is, one can increase stability in one direction by trading it for stability in another. For instance, if a person is expecting or anticipating a disturbance to their stance in one direction, they will lean in this direction to maximize the distance the CM will have to be moved before it passes outside the base of support.

The mass of an object also influences its stability in two ways. The first is the constant of proportionality between displacement ($\partial y$) and the change in potential energy ($\partial U$). For an object under the
influence of gravity, the larger the mass of the object, the larger will be the gravitational force and the larger the change in potential energy with displacement. Therefore, an unstable object will be more unstable with increasing mass and a stable object will be more stable with increasing mass. The second manner in which mass influences the stability of an object is through friction. As a first approximation, the force of friction \( F \) is proportional to the normal force \( N \) acting on the object by the surface, on which it is resting or sliding. That is, \( F \leq \mu N \) where \( \mu \) is coefficient of friction \{depends on the characteristics of the surfaces in contact and on whether there is relative motion (kinetic friction) or not (static friction)\}. The inequality \( \leq \) indicates that the frictional force \( F \) may be less than \( \mu N \) in static situations. The amount of frictional force present will, in part, determine if an object will rotate or just slide when pushed. For example, a body resting on a horizontal surface that generates a large frictional force will have a larger tendency to rotate when pushed, therefore both the shape of the object and the shape of the surface with which it lies will both influence the stability of the object. If the frictional force is too low to rotate the object then the shape of the object is not a factor in its stability. Also ignoring the situation in which the object rotates, the greater the frictional force the greater the stability of a resting object in that it requires a greater force to get it sliding and upon removal of that force the quicker the object will return to rest. As a general example of the influence of mass on stability, consider two identically shaped cardboard boxes: one filled with telephone books the other one empty. It is not hard to imagine the empty box with less mass and thus less stability tumbling and sliding down a street due to a gust of wind. However, it is more difficult to imagine the box filled with phone books (more mass, more stability) blowing down the street unless acted on by gale force winds. Note that friction is a non-conservative force and equation 2.2.1 no longer holds, therefore there is no easy method to classify an object as being in stable, unstable, or neutral equilibrium. However, the conditions for equilibrium still hold. Specifically: \( \Sigma F = 0 \) and \( \Sigma M = 0 \) (bold indicates vector).

Care must be taken in applying these concepts to the human body. Besides gravitational and frictional forces, there are muscle forces that are non-conservative, and at sub-maximal efforts, independent of position. To satisfy the equilibrium conditions as a segment rotates from vertical, muscle must act to generate forces to counter balance gravitational forces. Equilibrium can be maintained over a wide range of postures by modifying the muscular forces so that the torques and forces acting on a segment balance. Murray, Serig and Sepic (1975) and Stribely, Alberts, Tourtelotte and Cockrell (1974) found a broad range of upright postures within which body weight can be shifted and equilibrium still be maintained. For example, one can maintain equilibrium of the trunk even though the waist is bent 0°, 45°, or 90°, provided the muscle forces balance those of gravity.

Consider the simplified model by Crisco and Panjabi (1990) (figure 2.2.7.).

![Figure 2.2.7. Inverted pendulum model with torsional spring and bilateral linear springs.](image)

In this model, a rigid column of height \( h \) hinged to a rigid surface represents a segment. The ligamentous structure is modeled as a torsional spring of stiffness \( \tau_1 \) and the bilateral activation of muscle is modeled by a linear spring of stiffness \( \tau_m \). Since the equilibrium condition is \( \sum M = 0 \).
F_h \sin \theta + \tau_l + \tau_m = 0. \quad \text{Thus} \quad \theta = \sin^{-1}\left(\frac{\tau_l + \tau_m}{-F_h}\right)

Where the sum of the torque due to the ligaments ($\tau_l$) and the torque due to the muscular action ($\tau_m$) and the couple of the ground reaction force and gravity acting on the CM ($F_h \sin \theta$) equals zero. Thus equilibrium can be maintained at different angles ($\theta$) by altering the muscular action ($\tau_m$).

Other aspects to consider in the stability of humans include psychological and physiological factors. “There is a positive relationship between one’s physical and emotional state and the ability to maintain balance under difficult circumstances” (Luttgens and Wells, 1982, p. 395). As an example, consider the difference between crossing a level parking lot and crossing a level, but tightrope bridge over a gorge with a swirling river with hungry alligators in it. Mechanically the two situations may be equivalent and there should be no difference in the stability of the two situations. Nevertheless, the second condition is much more difficult because of the “psychological instability” created by the height and hungry alligators.

2.3. Center of Mass, Center of Gravity and Center of Pressure.

To assess postural control, researchers have studied the movement of center of mass (CM), center of gravity (CG) and center of pressure (CP). At times, some have used CM, CG and CP interchangeably. However, they are not equivalent and care should be taken to differentiate between them. They represent the average of different physical attributes and therefore relate to different aspects of balance control. Winter (1990) has pointed out that since these represent different entities their trajectories represent different aspects of balance. Patla et al. (1989) added that utilization of the differences between CM and CP paths might yield additional information on an individual’s balance capabilities. The CM is the centroid of the mass in a body. The CG is the centroid of the gravitational forces acting on all the points throughout a body. In a uniform gravitational field, the CM and the CG are located at the same point and can be used interchangeably. In the remainder of this dissertation, the assumption of uniform gravity will be made and CG and CM will be used interchangeably. In general, there is no one sensory system that monitors the location of the CM. The location of the CM is more or less calculated or deduced from the integration of the information from the visual, vestibular and somatosensory systems. Thus, while from a purely mechanical point of view the task of stance may be to keep the CM over the base of support (as defined by gravity), there is no direct physiological measure of the CM and thus the body must be either inferring the location of the CM from its multiple sensory inputs or it is maintaining stance via another set of rules. Keep in mind that the CM (the centroid of mass, the average location of all the mass in a system) is a mechanical entity derived to simplify calculations of dynamics of a system of particles, a rigid body or a system of rigid bodies. Although many theories exist, the “human law of balance” has not been discovered. The CP is the centroid of the normal contact forces acting on points on the surface of a body. Thus, the CP is always located on the surface of an object, while the locations of the CM and CG are not so restricted. For instance, the CG and CM, in some cases, can be located outside the volume of an object or a multi-segment body. Additionally unlike CM, the body through the cutaneous receptors can directly perceive CP.

While the location of the CP is easily calculated from force plate or pressure platform measurements, determination of the location of the CG from these measurements is prone to error. To determine the horizontal displacement of the CG from force plate data, the ground reaction force must be integrated twice with respect to time.

\[
CG_d = \int\int R(t) \, dt = \frac{1}{2} R t^2 + K_1 t + K_2 \quad \text{(Equation 2.3.1)}
\]

where

- $CG_d$ = the displacement of the center of gravity
- $R(t)$ = the resultant ground reaction force
- $K_1$ = constant of integration = initial velocity $v_0$
- $K_2$ = constant of integration = initial position $r_0$
A limitation of this method is that the constants of integration are known only if the initial conditions \( (v_0, r_0) \) are known. In most cases, they are not (Winter, Patla and Frank, 1990). Even in stance, the body is constantly moving and accelerating, and therefore the constants of integration cannot be set to zero and are unknown. King and Zatsiorsky (1998) and Zatsiorsky and King (1997) have developed a method for determining the constants of integration based on a postulation that when \( F_h = 0 \) (horizontal component of the ground reaction force) the CM and CP are vertically aligned. A more accurate method of determining the CG movement is from anthropometry and kinetics of the body (Winter, Patla and Frank, 1990). It is from anthropometric measurements that one determines the location of the CM of each segment of the body. Kinetic measures of each segment are taken from film, video or other instrumentation and combined with the anthropometric determination of the location of the segmental CMs. The location of the CM of the total body is then simply calculated through a weighted average of the locations of each individual segment’s CM location. The weights of the averages are the masses of the individual body segments.

In a static situation, an object is in equilibrium if its CP lies directly below its CG. Thus, in a static situation, the location of the CG can be found by projecting upward from the CP location. This method is sometimes used in anthropometrics to locate the CG of body segments. However, as stated before human stance is never static. There is always some movement (e.g., the action of the heart and lungs will cause movement in the body and fluctuation in ground reaction forces). Consider the situation first proposed by Winter (1990), in which an individual is swaying at the ankles (figure 2.3.1). In this figure, the ground reaction force \( \vec{F} \) is in blue and acts through the center of pressure (CP). The force of gravity (red line) drops vertically from the center of gravity (CG). Angular velocity (\( \omega \)), angular acceleration (\( \alpha \)) and torque (\( \tau \)) are depicted by curved red, blue and green arrows respectively. In this figure, the lengths of the arrows do not reflect the magnitudes of the vectors, however the direction of the vectors is represented by the direction of the arrows. For example, if there is no acceleration of the CG upward, the ground reaction force and the force of gravity should be equal in length but opposite in direction. The arrows representing the force of gravity are elongated to illustrate offset with the CP or ground reaction force. Initially, the body’s CG is ahead of the CP and its angular velocity (\( \omega \)) and angular acceleration (\( \alpha \)) are clockwise. To simplify matters, consider only the vertical component of the ground reaction force. This is a gross simplification, as the horizontal component of the ground reaction force may produce a considerable torque about the CM, because of a long lever arm. Nevertheless, to maintain balance, the CP is moved forward. Let the CP be moved to the end of its range of motion, the toes. This creates a counter-clockwise torque about the CM that generates a counter-clockwise angular acceleration. If the torque is large enough to reduce the angular velocity to zero before the projection of the CM progresses beyond the toes (the forward limit of the CP) the rotation will reverse and the body will approach an upright position. With no change in the location of the CP, the body will continue to accelerate and rotate counter-clockwise past vertical. To bring the body back to vertical, the CP must be shifted behind the projection of the CG so that a clockwise torque is created and the angular acceleration acts to slow and reverse the angular velocity. Again, the reversal of angular velocity must occur before the projection of CG passes the heel, the rear limit of CP. Therefore, it is possible for the projection of the CG to be at the toes while the CP is under the heel (or vice versa). In addition, the range of the CG is not limited to the boundary of the base of support as is the range of the CP. In the case of lost balance, the CP would still be constrained to be under the foot, while the CG projection would not. This occurs naturally in walking and running.
To further demonstrate the problems of using CP as a measurement of posture or balance, consider the following two illustrations (figure 2.3.2). First, consider a person standing first on his toes, then moving to his heels. In this case, the CP has moved from one extreme to the other with little change in posture or balance of the body. Second, consider a person balancing on his toes while bending at the hips. Here there is a drastic change in posture with no change in CP.

Gu, Schultz, Shepard and Alexander (1996) have found experimental evidence of the difference between CM and CP movements. They found that CP excursions were significantly larger, and varied at a significantly higher frequency than CM excursions for elderly and young adult subjects reacting to four different perturbation conditions.

Despite the difference between CP and CG, postural sway has traditionally been quantified by measures of CP. The CP signal has been analyzed in both the time and frequency domains. In the time domain, CP has been quantified in terms of path length or range (e.g., Diener, Dichgans, Bootz and Bacher, 1984), surface area (e.g., Horak, 1992) and root mean square (RMS) displacement (e.g., Maki and
In the frequency domain, fast Fourier transforms (FFT) have been used to determine the power frequency spectrum of the CP movement. The power spectrum analysis is useful because different ranges of the frequency spectrum have been linked to each of the three sensory systems that control posture. The working range of the visual system is below 0.3 Hz (Diener, Dichgans, Bruzek and Selinka, 1982), the muscle spindles from the somatosensory system are active at frequencies above 1 Hz (Diener, Dichgans, Guschlbauer and Mau, 1984), while the otolith organs and semicircular canals of the vestibular system work at frequencies below 0.1 Hz (Nashner, 1971) and at 1 Hz (Mauritz and Dietz, 1980) respectively.

Nevertheless, biomechanists have not decided on a gold standard measurement of sway. In part this is due to the fact that smaller amounts of sway do not always reflect better balance control (Patla, Frank and Winter, 1990; Overstall Exton-Smith, Imms and Johnson, 1977). One can also come up with hypothetical sway patterns as Tang and Woollacott (1997) have done that demonstrate instances in which different individuals will be considered to have better balance control depending upon how one characterizes their sway. Thus, the situation is similar to the statement often jokingly made about statistics, specifically: One can prove anything that one wants to with sway patterns (statistics).

At this point it is prudent to mention that balance in the sagittal plane is different from balance in the frontal plane. First, depending on the foot placement the dimensions of the base of support are most likely to be different. Second, the geometry of the body is quite different in each of the two planes. If an individual is standing with feet astride, the joints all line up in the sagittal plane while they are separated in the frontal plane. Thus balance control in the sagittal plane consists of adjusting the angular positions or moments at each joint; while balance control in the frontal plane consists of adjusting the distribution of weight on each pair of joints (Frank, Winter and Craik, 1996). If the individual assumes a stance in which the feet are in-line one in front of the other the situation reverses. Now balance in the frontal plane involves joint angular positions and moments, and balance in the sagittal plane consists of redistribution of force. Unless explicitly stated, the former situation (feet astride, symmetrical joints lining up along a ML axis) is assumed in the following discussions.

### 2.4. Measurements of Balance and Posture

In addition to CP measurements, most studies of posture record EMG signals from the lower extremity and trunk muscles. As with CP measurements, EMG measurements cannot be easily related to posture or balance. At best EMG measurements can only tell us which muscles are active and at what times. In determining muscle timing from EMG signals, one must take into account the temporal disassociation between the EMG signal and force (termed the electromechanical delay (EMD)). For example, the EMG signal in the cat soleus begins approximately 72 ms before force production and stops approximately 109 ms before the end of force production (Nigg and Herzog, 1994). Two other factors make the EMG posture relationship difficult to define. First quantitative relationship between EMG signals and muscle forces are presently not defined. Second, the link segment structure of the human body makes relating muscle force and body motion difficult.

Lippold (1952) and Moritani and devVries (1978), among others, have tried to quantify the EMG-force relation in isometric contractions. Despite the ideal conditions of isometric contractions, many different relationships have been found between muscle force and EMG signal. Bigland-Ritchie et al. (1980) noted that different muscles possess different EMG force relationships. Additionally muscle is capable of producing different forces depending on its length (length-tension relationship), type of muscle (type I, type II, etc.) and geometry (cross sectional area, pennate, multi-pennate, etc). Part of the difficulty in determining a relationship between force and EMG may be due to differences in recording, processing and analyzing the force and EMG signals. For instance, it is very difficult to get reproducible results after an electrode has been removed and replaced. In addition, strict isometric muscle contraction is difficult to achieve because although a joint or joints that a muscle crosses may not move, the muscle and its associated tissues may stretch during the muscle action. Thus, although the action is isometric from a joint perspective, it is not isometric from the perspective of the muscle. Difficulties increase in dynamic situations. Other factors that need to be taken into account in a dynamic measurement of the EMG-force relationship include speed of muscle movement, direction of movement (force-velocity relationship) and changing muscle electrode geometry. A further problem concerns the ability to measure the length change
of a muscle. The easiest method of determining muscle length is through calculations based on joint angle measurements, however elasticity of muscle and tendons, non-rotary movements of joints and movement of other joints also play a role.

Even if the force of a muscle were determinable from EMG signals, the transfer of muscle force to postural changes is not straightforward. First, as a muscle acts there are geometrical variations in lever arms that change the relationship between muscle force and the resulting torque on a segment or segments. Second, the body is a linked system, therefore the changing relationship between muscle action and segmental action not only involves the segments in which the muscle directly inserts or originates from but all the segments of the body. That is in posture, the whole body must be taken into consideration. The body is a multi-degree of freedom (DOF) linked system in which one has to take into account dynamic coupling (Zajac, 1987; Gordon, Zajac, Khang and Loan, 1988). Because of inertial and dynamic coupling, a single muscle affects the motion of all the joints and segments in the body, not just the joints it crosses or the segments it attaches to. “As the body is multi-segmental almost every movement a healthy person makes will set off a reaction throughout many parts of the body, so that the parts are acting in harmony and a balanced posture achieved.” (Galley and Forster, 1987, p. 86). For example, in posture, the ankle flexors act to accelerate, in addition to the ankle joint, the knee, the hip and all the joints of the body (Gordon, Zajac and Hoy, 1986; Hollerbach and Flash, 1982; Zajac, Gordon and Hoy, 1986). Further, it can be shown that in some posture configurations, the muscle contributing most to the movement of a joint may not be a muscle that crosses that joint. Thus, a muscle may act not to affect the motion of a joint or joints it crosses, but to alter the motion of a joint in another part of the body. Furthermore, the purpose of agonistic and antagonistic co-contraction may not just be to stiffen a joint, as generally thought (Hogan, 1982), but to improve the dynamics of the multi-joint system as a whole (Zajac, 1987). Thus, because of this dynamic coupling, it is very difficult to correlate EMG activity to mechanical events.

All of the difficulties in determining a relationship between EMG and muscle force do not lie with the measurement of EMG signals. Muscle force measurements have their limitations too. There are two general approaches used to determine the force and moments acting on segments of the human body. The first is to surgically implant strain, pressure, or force gauges at strategic locations in the body and directly measure the force. Rydell (1965, 1966) was the first to make direct measurements of the articular forces in a joint. He implanted a Hear and Austin-Moore hip prosthesis of modified construction, containing wire resistance strain gauges, in two patients. The forces and moments at the hip joint during level walking were recorded (the greatest force was 3.3 times body weight). Another example of direct measurement is Komi’s work in 1990, using buckle transducers on animals and humans to determine the mechanics of muscles. In essence, a buckle transducer is a ring and a bar. The tendon is looped through the ring and the bar is placed through the loop formed by the tendon. The tendon is released and allowed to meet the bar. The bar is instrumented with strain gauges. As the muscle acts, tension is built up in the tendon, the bar is bent and the strain gauges are stretched. Measurement of muscle force is determined by relating the strain in the bar, to the tension in the tendon, to the tension in the muscle or muscles acting on the tendon. At present, the buckle transducer is limited to use in long tendons like the Achilles tendon (tendon calcaneus). In the case of the Achilles tendon, there are three muscles that insert into it (gastrocnemius, soleus and plantaris), and the buckle transducer is incapable of distinguishing the actions of individual muscles or compartments within a muscle. The resolution of a transducer depends on the tendon and the number of muscles connected to it. Other types of sensors have been used to quantify forces in tendons and ligaments. They include: foil strain gauge transducers (White and Raphael, 1972), liquid metal strain gauge transducers (Brown et al., 1986), Hall effect transducers (Arms, Johnson and Pope, 1983) and implanted force transducers (Holden, Grood and Cummings, 1991 and Xu, Butler, Stouffer, Grood and Glos, 1992). These transducers have the same limitations as the buckle transducer. Several other researchers have investigated measurements on assemblies of skeletal parts and cadavers with muscles and ligaments replaced by mechanical linkages incorporating transducers (Bloiuint, 1956; Denham, 1959; Inman, 1947; Merchant, 1965; Strange, 1963 and Williams, 1964). Because of the invasive nature of these direct measurement methods, the subject pool is small. Although the measurements are direct, prostheses are not exact duplicates of the joints they replace. Buckle transducers alter the path of tendons, and implementation can damage surrounding tissues. Therefore, the results obtained are not representative of a
normal situation, and the shortcomings have to be carefully considered. Despite these shortcomings, they are the best that one can presently achieve.

The second method is to mathematically model the human body, or segments under investigation, to estimate the forces and moments acting on the segments.

For a complete description of the behavior of the legs of the human mechanism in walking knowledge of the forces and moments at the joints is essential. The determination of these forces and moments requires either measurement or calculation of the mass distribution of the leg displacements and accelerations of the leg segments through one complete cycle, and the reaction forces of the ground on the foot (Bresler and Frankel, 1950, p. 29).

Elftman (1939) was a pioneer of photographic analysis of gait. Elftman used Fischer’s (1906) data of body characteristics with his own measurement of segmental orientation and ground reaction forces to calculate joint moments. He limited his investigation to forces and movements in the mean plane of progression. Bresler and Frankel (1950) were the first to look at the time changes of the components of the moments of the lower limb in three dimensions. They used a force plate developed by Cunningham and Brown and two film cameras to record the displacements of body markers and the ground reaction forces. They obtained curves showing the variation with time of the three force components and the three moment components at the ankle, the knee and the hip joints during walking on a level surface. Since then, many researchers including Cappozzo, Figura, Marchetti and Pedotti (1976); Cappozzo, Leo and Pedotti (1975); Hatze (1977); McLeish and Charnley (1970); Morrison (1968, 1969, 1970a, 1970b); Paul (1965, 1966a, 1966b, 1971); Pedotti, Krishnan and Stark (1978); Serig and Arvikar (1989) and Strange (1963) have researched the forces and moments of the lower extremities during different activities.

When an attempt is made to define a model of the human skeletal link system, it is soon realized that considerable simplifications have to be adopted if the procedure is to lead to feasible results (Hatze, 1977, p. 799).

These models have as their independent variables, values that are easily measured or calculated, such as position, velocity and acceleration of individual body segments as well as the external forces and moments acting on these segments. The resultant forces and moments at each joint are determined by modeling each segment as a rigid body (link segment model), equating the sum of external forces acting on the body with the product of the mass of the body and the acceleration of its CG, and equating the sum of external moments acting on the body with the moment of inertia and angular acceleration of the body.

In order to calculate the forces and moments acting on joints, one must know the location of the joint centers and centers of masses of the segments with respect to external markers. The traditional approach has been to derive this data from cadavers. In 1957, Barter used the results of the cadaver work of Braune and Fisher (1889) and Dempster (1955) to develop regression equations to predict segment weights from the over all body weight. Because only body weight was used as a predictor of segmental mass, Barter’s equations would produce the same results for two individuals of the same weight, but of different size and shape. In addition, he failed to determine the CM or the moment of inertia. Clauser, McConville and Young (1969) developed a set of regression equations to calculate segment masses, volumes and centers of mass. Their regression equations utilized in addition to subjects weight, other variables, such as segment length, diameter and circumference. Chandler, Clauser, McConville, Reynolds and Young (1975) measured six cadavers to develop their regression equations. They measured segment weights, volumes, densities, CM location, principal moments of inertia and angular orientation of the principal axes relative to anatomical axes. These cadaver averages are, for the most part, based on elderly Caucasian men. Unfortunately, the anthropometric values obtained are not meant to be extrapolated to other populations, especially specialized populations (e.g., elite female gymnasts, football linemen, and sumo wrestlers). Anthropometric data has also been generated from mathematical models (Hanavan, 1964 and Hatze, 1977), scanning techniques such as gamma rays and axial tomography (Brooks and Jacobs, 1975; Huang and Wu, 1976; and Zatsiorsky and Seluvanov, 1985) and kinetic measurements (Danis, 1980). Optimally, one desires a method that is flexible enough to fit individual subjects, is inexpensive in equipment, supplies and time, and is safe and is accurate. None of the above methods meets all of the criteria. Each has its advantages and disadvantages.
Many methods exist for developing biomechanical models of the human body and there are other papers that review the various methods in detail (e.g., King, 1984 and Reid and Jensen, 1990). It is important to note that assumptions must be made. The accuracy of any model is completely dependent on the nature of the validity of the assumptions used to create it.

In contrast to the abundance of studies of the dynamics of gait, few studies have been devoted to in depth analysis of body segmental kinematics and kinetics in response to postural disturbance. Most studies include only ground reaction forces and EMG signals (See appendix four for a table of posture studies in the literature). In 1851, Rhomberg was the first to research standing posture (Thyssen et al., 1982). He compared the sway of subjects standing with their eyes open to sway with their eyes closed. Sheldon (1963) made direct plots of postural movements with a pen attached to a metal frame that was fixed to the shoulders of subjects he was studying. The Wright ataximeter (Overstall, Exton-Smith, Imms and Johnson, 1977) and potentiometers attached to anatomical landmarks through long poles were later used to measure displacements of different body segments. More recently, force plates have been used to record the CP location and movements of standing individuals. Most of the studies listed in appendix four use ground reaction forces coupled with EMG data to quantify stance and reactions to postural perturbations.

More comprehensive studies of postural dynamics have been done by Alexander, Shepard, Gu and Schultz (1992); Gu, Schultz, Shepard and Alexander (1996); Hughes, Chandler, Schenkman and Studenski (1992); Romick-Allen and Schultz (1988); Woolacott, Inglin and Manchester (1988). Woolacott, Inglin and Manchester (1988) analyzed movements of individual body segments in response to anterior and posterior platform translations, and compared them with the onset latencies of the neck muscles. Romick-Allen and Schultz (1988) measured the reactions of eleven young adult males to a forward acceleration of 0.18 g. These measurements were then input into a nine or twelve segment whole body sagittal plane model (twelve segments were used if the subject stepped in response to the perturbation, otherwise, the nine segment model was used), and the net joint torques were calculated. Hughes, Chandler, Schenkman and Studenski (1992) studied sixteen adults who experienced randomly sequenced forward and backward horizontal platform perturbations. They measured EMG latencies at selected muscles and displacements of the ankle, knee, hip, shoulder and neck. Alexander, Shepard, Gu and Schultz (1992) and Gu, Schultz, Shepard and Alexander (1996) studied 24 young adults and fifteen elderly men (mean ages 26 and 72, respectively). Four tasks were presented to these individuals: (1) standing on an anteriorly accelerating platform, (2) standing on a stationary narrow beam, (3) standing on an anteriorly accelerating narrow beam and (4) standing on an unstable platform. They measured maximum rotation, time to first rotation, time to first rotation response, direction of initial rotation and time to first rotation reversal of individual body segments. Excursion of the CP and total body CM, peak ground reaction forces, total body angular momentum about the ankle and the peaks of the joint torques were also measured or calculated. Besides the experimental research cited above, there have been several theoretical projects. The goal of this type of work has been to perfect computer models rather than to analyze posture. Computer models of posture first appeared about two decades ago and were generally based on single mass, single DOF inverted pendulums (Geurssen, Altena, Massen and Verduin, 1976; Gurfinkel, 1973; Kodde, Caberg, Mole and Massen, 1982; McGhee and Kuhner, 1970; Peters, Caberg and Mole, 1985). For instance, McGhee and Kuhner (1970) used a planar inverted pendulum to study the relative roles of ankle and hip torques in the control of posture. Golliard and Hemami (1976) extended this work by including lower extremity masses and knee motion. Koozekani, Stockwell, McGhee and Firozmand (1980) used a four mass linkage restricted to the sagittal plane to predict the CP and ground reaction forces during stance. Onyshko and Winter (1980) used a seven segment Lagrangian model to simulate body motion. Riley, Mann and Hodge (1990) developed an eleven-segment model in three dimensions to compare the location of the projection of the whole body CM with the measurement of CP location by a force plate. In summary, there are many researchers working to understand whole body mechanics (See King, 1984 for further review), however, few are applying this work to understanding postural dynamics and balance control.

2.5. “Ideal” Standing Posture.

Defining ideal posture is not easy. One can refer to a position that requires the least effort to maintain, puts the least amount of strain in the ligaments, bones and joints or maintains the CM over the
base of support (Thibodeau and Patten, 1993). However this definition is too simplistic in that only mechanical factors are taken into consideration. The literature on posture generally defines an ideal posture as one in which a line parallel to gravity passing through the center of gravity also passes through the following landmarks:

- the mastoid process
- just anterior to the shoulder joints
- just posterior to the hip joint center
- just anterior to the knee joint center
- approximately 0.05 to 0.06m anterior to the ankle joint

(Basmajian and DeLuca, 1985; Galley and Forster, 1987; Kendall, McCreary and Provance, 1993; Klausen, 1965; Luttgens and Wells, 1982; Woodhull, Maltrud and Mello, 1985) (Figure 2.5.1).

Figure 2.5.1. An idealized erect posture.

“Ideally, these links should be so stacked that the line of gravity passes directly through the center of each joint between them. But even in man, this ideal is only closely approached and is never completely reached and then only momentarily” (Basmajian and De Luca, 1985, p. 255). Note in figure 2.5.1, the axes of the vertebral column, hip, knee and ankle lie on either side of, but not along, the vertical line passing through the CM. Thus, passive equilibrium is not possible, and some muscular activity is necessary for stance. Still, the amount of muscular activity required to maintain an upright posture is only “slight or moderate” (Basmajian and De Luca, 1985, p. 255).

Kendall, McCreary and Provance (1993) believe that the orientation of the pelvis and head must also be defined in idealized posture. They define a neural pelvis orientation as having the anterior superior spines and the pubic symphysis in the same vertical plane. The head is in a neutral position when a vertical line passes through both the mastoid process and a point slightly posterior to the apex of the coronal suture. Kendall, McCreary and Provance (1993) also indicate that the vertical line must pass through the bodies of the lumbar vertebrae and bodies of most cervical vertebrae. Thus in the...

...standard posture the spine presents the normal curves, and the bodies of the lower extremities are in ideal alignment for weight bearing. The “neutral” position of the pelvis is conductive to good alignment of the abdomen and trunk, and of the extremities below. The chest and upper back are in a position that favors optimal function of the respiratory organs. The head is erect in a well balanced position that minimizes stress on the neck musculature (Kendall, McCreary and Provance, 1993, p. 71).

A word of caution must be included at this point. The terms “standard posture” or “normal posture” may have come from the work of Braune and Fischer (Rasch, 1989). Their major premise was that an understanding of the location of the CM was key to an understanding of the forces that are
developed by muscles in movement. To determine the location of the CM, they froze four cadavers and fixed them to a flat surface. The planes of the center of gravity were then determined. Braune and Fischer labeled the original position of the cadavers’ “normalstellung”. They intended this term to indicate that this was their reference position from which their measurements were to be based. Subsequently, many have interpreted “normalstellung” to be the normal standing position that one should try to achieve. Comfortable relaxed standing was called “bequeme haltung”, meaning comfortable hold.

The work of R. A. Flick pointed out differences between Braune and Fischer’s “normalstellung” and real world situations (Rasch, 1989). For example, he pointed out that because the cadavers were recumbent when frozen the lumbar vertebra exhibited less of a lordosis than would be present if they were standing. In stance, the CG would be anterior to the location that was determined by Braune and Fischer. Flick also noted that no one posture is common for all people. This is backed by the work of modern anthropometrists (Rasch, 1989). Thibodeau and Patton (1993) note that systems other than the musculoskeletal system may influence posture. Postural differences may arise from differences in:

- **anthropometry**
- **physiology**
- **sociology or culture**
- **psychology and**
- **environment.**

Normal posture is difficult to define because every person has a unique anthropometric profile (Howe and Oldham, 1997). For instance, individuals with lax ligaments tend to stand with hyperextended knees and hips and with exaggerated curves of the spine (Largon and Gould, 1974). Observations of dancers have shown that even within groups of similar background (age, race, gender etc.) there exist individual differences. Laws and Harvey (1994) found that in any balanced position, different dancers find that slightly different positions of the body work best, due to differences in their individual body structure, flexibility and comfort. Adrian and Cooper (1989) pointed out that individuals will experience different stress patterns depending on their unique anatomy. Metheny (1952) agreed with this viewpoint. He felt that there is no single best posture for all individuals, and that for each person the best posture is one in which the body segments are balanced in the position of least strain and maximum support. Included in anthropometric differences are gender differences such as differences in stance due to differences in body proportions, distribution of mass and differences in joint angles and ranges of motion (e.g., Q angle). Physiological parameters influence posture through the digestive, excretory and endocrine systems that shift levels of nutrients and other components of the body. Postural effects of these systems can be observed in individuals with obesity, osteoporosis or muscle cramps. The respiratory and circulatory systems also alter posture with their continuous activity. The effect of breathing and the heartbeat can be measured in an individual standing on a force plate. Rasch (1989) noted that sociological and cultural factors influence posture. Hewes (1957) has identified over 100 different postures ranging from standing, squatting, kneeling and sitting. For example, at least one fourth of the population of the world crouches in a low squat to rest and work. In addition, a stork like stance in which the sole of one foot is placed against the support leg near the knee is a common resting position of people in Africa, India, Australia and South America. This type of one leg posture is often called the “Niolitic stance” meaning belonging to the Nile. Hewes (1957) believes that warm climate, bare feet and unknown cultural reasons worked together to develop this manner of stance. Rank of an individual in a group is another sociological factor that may influence posture. A higher-ranking individual is more likely to have a more erect stance while a subordinate individual may stand with their head bowed and their eyes looking downward. Cultural norms dictate different postures for men and women. This is especially true in the seated posture whereby women normally adopt a closed leg or crossed leg position while men are not so restricted. Cultural requirements of clothing may restrict or alter the posture of some either through physical means or through modesty considerations. Psychological factors may also affect stance. For instance, happy confident individuals will tend to stand differently than depressed self-conscious individuals. Environmental factors such as height, slope of ground, stability of supporting surface, wind, rain, heat and cold will inject differences in one’s posture.

So far, it has been assumed that posture is static, however it is not. Even if the feet remain stationary, the CG moves randomly within an area called the “limit of stability” (Nashner, 1997) or
“functional reach” (Duncan, Weiner, Chandler and Studenski, 1990). The limits of stability for a normal adult standing on a flat, firm surface with their feet adjacent and spaced comfortably apart is a cone with an elliptical cross section. The anteroposterior dimensions of the limits of stability are approximately 12.5° while the medial lateral limits are approximately 16° (with feet approximately 4 inches apart) (Nashner, 1997; Blaszczyk, Lowe and Hansen, 1994; Nashner and McCollum, 1985; Schieppati et al., 1994). The limits of stability will vary inversely with the height of an individual and directly with their foot length and foot spacing (i.e. height of CM and size of base of support). How much movement a person exhibits within their limits of stability varies between individuals. In general it is thought that the smaller amount of sway exhibited the better off, or the less prone a person is to falling, however in some individuals (e.g. individuals with peripheral neuropathy due to diabetes), an increased sway may be used to increase the strength of environmental stimuli to improve their ability to stand without falling (Simoneau, 1992).

In conclusion, a perfect or even a normal posture does not exist; posture is dependent upon one’s build. Individuals will have different locations for anatomical landmarks, and distribution of mass. “At best, the location of the line of gravity should be used as a general indicator of good posture. It might be more realistic to determine a ‘normal zone’ within which the line of gravity might reasonably be expected to lie...” (Luttgens and Wells, 1982, p. 429). Part of the rationale behind the concept of “ideal posture” is a posture in which each segment is balanced vertically upon the segment below it and that muscles and ligaments are required to counter balance minimal gravitational torques. This reasoning ignores two facts (Luttgens and Wells, 1982): (1) In a fatigued posture, in which the muscles have relaxed, the segments become more “poorly” aligned and the ligaments are left to maintain posture. (2) Even in the most well aligned posture, some torque is present due to, (a) the spine being situated close to the posterior surface of the body, (b) the chest creating a constant anterior torque, (c) the feet being more anterior than posterior and (d) the spinal column being curved anteroposteriorly.


The ability to maintain stability in stance or to recover from a perturbation requires the precise coordination of multiple links, joints and muscles. According to some, the neuromuscular system uses a “postural sensorimotor coordination”, or a “movement strategy” to maintain stability (Nashner, 1997; Horak, 1992). In general, one can respond to a perturbation in an infinite number of ways given enough time. According to Nashner (1997), there are three categories of movements related to postural activity: myotatic stretch reflex, automatic postural movements and voluntary postural movements.

The myotatic stretch reflex is the first mechanism to respond to the external perturbation. The major role of the myotatic stretch reflex is unconscious regulation of muscle contractile forces. These reflexes are localized to the point of stimulus and are highly stereotyped with latencies fixed at 35-40 ms (Nashner, 1977). These reflexes may be productive or counter productive to the task of maintaining stance. For example, when nodding off, stretching in the neck musculature produces a productive reflex that rights the head, whereas a toes up tilt stretches the triceps surae, which elicits a stretch reflex that further destabilizes the individual.

Automatic postural movements are the first functionally effective responses that consistently aid in regaining postural stability (Nashner, 1976, 1977, 1997; Nashner, Woollacott and Tuma 1979). Automatic postural movements are instigated by abrupt, unexpected stimuli, and occur at fixed latencies (90 to 100 ms for EMG with an additional 20 to 40 ms for muscle force onset (Marsden, Merton and Morton, 1973)). Automatic responses generally have stereotypical patterns called postural strategies (described shortly). The time of the latency suggests brainstem and cortical structure participation (Melvill-Jones and Watt 1971, Diener et al., 1985). Additional indications of brainstem and cortical involvement are indicated by research of responses in individuals with brain and spinal cord lesions (Horak, Nashner and Diener, 1990). The initiation and direction of automatic postural movements are triggered by somatosensory and proprioceptive signals (Horak, Nashner and Diener, 1990; Diener, Horak and Nashner, 1988; Allum, Honegger and Schicks, 1994; Inglis, Horak, Schupert and Ryczewicz, 1994; Nelson, DiFabio and Anderson, 1995; Jackson, Epstein and Del’Amme, 1995; Shupert, Horak and Black, 1994; Horak, Nutt and Nashner, 1994). Visual and vestibular inputs are integrated with recent experiences to influence the response amplitude and pattern (Diener, Horak and Nashner, 1988; Wolfson, Whipple, Amerman and Kleingberg 1986; Nashner and Berthoz, 1978; Allum, Honegger and Schicks, 1994). Recent
experiences, not conscious or unconscious decisions determine the strategy that is used (Horak and Nashner, 1986; McCollum, Horak and Nashner; 1986).

Decisions first come into play with voluntary postural movements. The muscle latency for a voluntary response is at least 150 ms depending on the complexity of the task and the attention of the individual. This does not mean that the voluntary postural movement necessarily follows or is in reaction to an event. Voluntary postural movements generally occur before an anticipated event in order to establish an appropriately stable base of support or to initiate counter balance movements necessary to support or aid the execution of the event. In fact, voluntary actions with the potential for disturbing balance are delayed so that the anticipatory voluntary postural movements can be initiated to achieve the necessary support or counter movements necessary (Nashner, 1997).

As stated previously, three general postural strategies have been proposed for the correction in anterior-posterior sway in normal adults: an ankle strategy, a hip strategy and a stepping strategy (Do, Breniere and Brenguier 1982; Horak and Nashner, 1986). The ankle strategy shifts the body’s CM by rotating the body about the ankle joints with little or no movement about the hip or knee joints (figure 2.6.1, left). There is also a shift of the center of pressure forward or backward under the feet as the toes or the heels are pressed more or less into the ground. Here the muscles activate in a distal to proximal order. During an unexpected displacement in the anterior direction, the loss of stability triggers a distal to proximal activation of the plantar flexor, knee flexor, gastrocnemius, hamstring and trunk muscles. Conversely, during a posterior displacement of the CM the anterior muscles are activated in a distal to proximal order (ankle dorsiflexors, quadriceps and trunk flexors). In the hip strategy, a person tries to maintain balance by flexing and extending at the hips (figure 2.6.1, center). In this strategy, the muscles are activated simultaneously or in a proximal to distal sequence. The mixed strategy is a combination of the hip and ankle strategies (not shown). The stepping strategy is different from the ankle, hip and mixed strategies, because it’s primary goal is to change the size and location of the body’s base of support, while the other strategies serve to move the body’s CM (figure 2.6.1, right). The stepping strategy may take on a number of forms (a single step, a series of small steps, hops, skips or stumbles). Another commonly practiced strategy that is frequently ignored in the literature is grabbing. In the grabbing strategy, one uses the hand or hands to secure the body by clutching or touching a solid object. This may be considered an extension of the stepping strategy where instead of the foot being used to change the base of support the hand, arm or other body part is used to change the base of support.

Figure 2.6.1. Ankle, hip and step strategies.

Different strategies are used under different conditions. The strategy used depends on mechanical constraints, the available sensory information, the environmental context and prior experience (Horak, 1992). For example, the ankle strategy is used to maintain balance when the surface is firm and wide and the perturbation is slow and small (Nashner, 1989; Horak, 1992). If the perturbation is large or fast, or the support surface is narrow (e.g., railroad track) or soft (e.g., foam, sand), the hip strategy is usually used
(Nashner, 1989; Horak, 1992). The stepping and grabbing strategies are usually only used when the ankle and hip strategies are inadequate. For example, the ankle and hip strategies are useless once the body’s CM has passed outside the limits of the base of support. In these cases, only a stepping or grabbing strategy can prevent a fall.

People have particular regions or limits in which they can successfully use a particular strategy (Horak, 1992). Figure 2.6.2 is an idealized map of the relationship between the projection of the body’s CM on the ground and the strategy used to maintain balance. One can think of the areas as regions of increasing environmental stress. Low or mild environmental stress is located in the center and the stress increases radially. Each individual has their own borders that are dependent on anthropometrics, muscle strength and motivation, as well as other factors. In the center, where the conditions are moderate, the ankle strategy is used. The ankle strategy reaches its limit when the torque at the ankle joint starts to rotate the foot. Because the ankle torque is counterbalanced by a couple generated by the ground reaction force and the force due to the weight of the body acting at the ankle joint, the size of the torque that can be tolerated depends on the distance between the ankle joint and the center of pressure and the weight of the individual.

Outside the ankle strategy limit, which is reached when ankle torque alone is insufficient to return the body to equilibrium, a person will add horizontal shear forces by including rapid hip movements (Nashner, 1989, 1997; Horak, 1992). Note that unlike the ankle strategy, the hip strategy is transient and cannot counterbalance a constant offset between the gravitational force acting through the body’s CM and the center of pressure (ground reaction force). If this strategy is insufficient, the base of support must be moved or increased by stepping or grabbing. Note that the areas are more stable in certain directions than in others. The direction of an axis of maximum stability will depend on foot placement and strength. Figure 2.6.2 depicts the situation in which the feet are placed side by side. Here the axis of maximum stability is parallel to the medial lateral axis while the axis of minimum stability is parallel to the anterior-posterior axis. If the feet are offset, the primary axes of stability will be skewed. Additional asymmetries will result from differences in the ability to generate muscular force and the flexibility (passive or structural constraints) in different directions. Note that the use of ankle and hip strategies only really applies to reactions to postural imbalances in the sagittal plane. Reactions to lateral imbalances or imbalances along other planes or axes have not been defined or classified.

Figure 2.6.2. Horak’s strategy areas (1992).

In addition to ambient environmental conditions, postural strategies are used according to an individual’s prior experience (Horak, 1992). When subjected to a novel environmental condition, an individual’s initial reaction is to use a strategy that was used successfully in the past. If the initial strategy of reaction is not the most efficient, a subject will modify their strategy upon subsequent exposures to the new environmental conditions until an efficient reaction or strategy is achieved (Horak and Nashner, 1986). During this evolutionary period, a mixture of strategies may be used.

The coordination, initiation and execution of a strategy are influenced by the age of the individual. Stelmach, Phillips, DiFabio and Tensdale, 1989; Woollacott, 1990; Woollacott et al., 1988) have shown that the initiation of a muscular response to a balance displacement is delayed in healthy older individuals. The lower leg muscles of these older individuals took 20 to 30 ms longer to react when compared to younger controls. Stelmach and Worringham (1985) point out that since the postural reflexes are generally slower the older individual will be further from equilibrium before corrective action is initiated and therefore a stronger response will be required to correct the situation. The longer latencies seem to occur more frequently in the tibialis anterior than in the gastrocnemius (Inglin and Woollacott, 1988). In addition to the longer latency, there may be occasional disruptions in the sequencing of the activation of muscles in
response to postural disturbances (Stelmach et al., 1989). Similar changes have been observed in the elderly in situations of voluntary adjustments to posture in anticipation of movement (Mankovskii et al., 1980; Inglis and Woollacott, 1988). For example postural adjustments used to stabilize the body before lifting one foot off the ground or in initiating walking. Manchester et al., (1989) found that older individuals seemed to prefer to use the hip strategy more than younger individuals do in response to postural disturbance. Two explanations for this preference have been given. First is that the older individuals have less muscular strength at the ankles and may not be able to produce sufficient ankle torque to correct the postural disturbance. The second explanation related to a decrease in proprioception in the ankle. It is possible that small displacements from equilibrium do not provide sufficient amounts of sensory information to detect a perturbation (Stelmach and Sirca, 1987) and therefore a more drastic strategy has to be used as the individual is further out of equilibrium when the displacement is detected. Luchies, Alexander, Schultz and Ashton-Miller (1994) have found differences with age in the stepping strategy. In general, younger individuals took a single step while the older individuals took multiple steps in response to a postural perturbation. The steps taken by the older individuals were shorter with less ground clearance. They concluded that the older individuals were being more conservative in their response. With multiple steps the range of motion of the joints is less and there are more opportunities to make further adjustments to correct for any mistakes in the initial response.

How or if one maintains balance is partly determined by how far one is from equilibrium. For example, if the projection of the CM is not too far outside the area of support the acceleration away from a balanced position is not very large. The closer one is to equilibrium, the greater the time one will have to recover. Conversely, the farther from equilibrium a person is, the more urgent the recovery process becomes. This in of itself is obvious, however the large effect that small deviations from balance have on the time that one has to react may not be. Two factors act to increase the urgency with which one has to act when further out of balance. The first is quiet obvious, the further one is out of balance the closer one is to falling and the further they have to go to return to equilibrium. The second factor is a little less obvious. With increasing lean there is a greater destabilizing torque generated by the gravitational and ground reaction couple. To illustrate this point, consider the human body to be modeled as a single segment of length (l) as done by Laws and Harvey (1994). Initially the body is placed at an initial angle (θ₀) from the vertical (figure 2.6.3). The force of gravity and the ground reaction force are not co-linear, and therefore place a torque on the body.

![Figure 2.6.3. Effect of different initial angles.](image)

The torque is:

\[ \tau = mgR_c \sin \theta = I \alpha = mR_g^2 \alpha \]  

(Equation 2.6.1)

where

- \( \tau \) = torque on body,
- m = mass of body,
- g = acceleration due to gravity,
- \( R_c \) = distance between point of support and CM,
θ = angle between body and vertical, 
I = rotational inertia of body through axis through point of support = \( \sqrt{I/m} \), 
\( \alpha = \) angular acceleration of body and 
R_g = radius of gyration. 

If the angle is small \( \sin \theta = \theta \) and 
\[ \tau = mgR_c \theta = I\alpha = mR_g^2 \frac{d^2\theta}{dt^2} \]  
(Equation 2.6.2)

Solving for \( \theta \):
\[ \theta = \theta_o \cosh \left[ \frac{gR_c}{R_g^2} \right] \sqrt{\frac{1}{t}}. \]

For example, if we have a 2m tall person the CM would be roughly a little more than half way up. Let \( R_c = 1.1 \text{ m} \) and as a first approximation, let \( R_g = R_c \). Thus \( \theta = \theta_o \cosh[3.13t] \) (table 2.6.1).

### Table 2.6.1. Effect of different initial angles on displacement from vertical.

<table>
<thead>
<tr>
<th>Initial angle (degrees)</th>
<th>0.5</th>
<th>1.0</th>
<th>2.0</th>
<th>3.0</th>
<th>4.0</th>
<th>5.0</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.00</td>
<td>0.50</td>
<td>1.00</td>
<td>2.00</td>
<td>3.00</td>
<td>4.00</td>
<td>5.00</td>
</tr>
<tr>
<td>0.25</td>
<td>0.66</td>
<td>1.32</td>
<td>2.64</td>
<td>3.97</td>
<td>5.29</td>
<td>6.61</td>
</tr>
<tr>
<td>0.50</td>
<td>1.25</td>
<td>2.50</td>
<td>4.99</td>
<td>7.49</td>
<td>9.99</td>
<td>12.48</td>
</tr>
<tr>
<td>0.75</td>
<td>2.64</td>
<td>5.28</td>
<td>10.56</td>
<td>15.83</td>
<td>21.11</td>
<td>26.39</td>
</tr>
<tr>
<td>1.00</td>
<td>5.73</td>
<td>11.46</td>
<td>22.92</td>
<td>34.38</td>
<td>45.84</td>
<td>57.29</td>
</tr>
<tr>
<td>1.25</td>
<td>12.51</td>
<td>25.02</td>
<td>50.04</td>
<td>75.07</td>
<td>&gt;90</td>
<td>&gt;90</td>
</tr>
<tr>
<td>1.50</td>
<td>27.35</td>
<td>54.70</td>
<td>&gt;90</td>
<td>&gt;90</td>
<td>&gt;90</td>
<td>&gt;90</td>
</tr>
<tr>
<td>1.75</td>
<td>59.81</td>
<td>&gt;90</td>
<td>&gt;90</td>
<td>&gt;90</td>
<td>&gt;90</td>
<td>&gt;90</td>
</tr>
<tr>
<td>2.00</td>
<td>&gt;90</td>
<td>&gt;90</td>
<td>&gt;90</td>
<td>&gt;90</td>
<td>&gt;90</td>
<td>&gt;90</td>
</tr>
</tbody>
</table>

At the beginning, the angle doubles in only a quarter of a second, or increases by a factor of eight in approximately one second. If an individual’s CM was off by only 0.06m, or they were tilted by 4°, they would have to react quickly, for if nothing were done they would be leaning by almost 40° after one second.

### 2.7. Senses Involved in Balance.

Mechanically balance can be defined as the activity of regulating the CM with respect to the base of support. Sensing the location of the CM is done indirectly through integration of the sensory information provided by the visual, vestibular and somatosensory (tactile, pressure, joint and muscle receptors and proprioceptors), since none of these inputs directly measures the CM position. Vision provides information as to the orientation of the eyes and the head in relation to surrounding objects and the orientation of the body parts within the field of view. The vestibular system indicates the orientation of the head with respect to the force of gravity and linear and angular accelerations of the head. The somatosensory system provides information about the interrelationship of segments through joint and muscle sensors, and the interaction of the body with the environment through skin pressure, pain and vibration sensors. Because one or more of the senses may provide misleading information the individual must constantly alter how these inputs are used. Their importance or weight is constantly changed according to the conditions. For example, one must ignore the visual perception that one is falling backwards when standing next to a large truck that suddenly moves forwards. A period of momentary disorientation or unsteadiness may last for a fraction of a second while the brain determines that it is the truck that is moving and not the body. This process of selecting and combining appropriate sensory
information is called “sensory organization” (Nashner, 1997). In addition to utilizing the redundant information about the location of the CM provided by multiple sensory input individuals can utilize these redundancies to compensate for lack of information due to deficits in a sensory system caused by disease or injury. Additionally the body is very adaptable in how it interprets information. For instance after wearing prism glasses that flip the image on the retina for a period of time the brain can adapt by flipping the image again as it processes it.

As will be discussed in the section on falls (see section 2.8) it is not the loss of one specific sensory system that causes problems, but subtle degradation of multiple systems. For example, individuals without vision can stand and walk. Problems begin when there are small deficits in multiple systems. Since these individual deficits are small they are sometimes difficult to detect. Additionally the body’s ability to compensate for the loss adds to the difficulty in detection. One method of isolating sensory inputs to balance is through tests that combine different environmental conditions in an organized fashion. For example, tests that combine normal (fixed), eyes closed (blind folded), sway referenced visual and sway-referenced support surface conditions (Nashner, 1989, 1997; Schumway Cook and Horak, 1986; Jackobson, Newman and Kartush, 1993). Thus, a subject’s incorrect response to a perturbation in a certain condition may not be due to deficiencies in sensory information, but an inability to correctly organize the sensory information or may be a combination of both.


2.8.1. Definition of Falls.

Recurrent falls are a marker of frailty, immobility, and acute and chronic health impairment in older persons. Falls, in turn, contribute to functional decline by causing injury, activity limitation, fear of falling, and loss of mobility and independence (Nevitt, 1997, p. 13).

There are falls, and then there are falls. How a fall is defined will influence what risk factors are associated with it, and subsequently what action(s) (if any) can be taken to prevent them, or at least decrease the chance that they will occur. There is large variance in the significance of falls. A fall may be the first indication of an acute problem such as an infection or postural hypotension, it may indicate progression of a chronic disease, or it may simply be a consequence of the normal aging process. On the other hand, a fall may just indicate that the individual was overwhelmed by environmental conditions that could not be mastered by anyone regardless of their age, strength or physical ability.

Gibson (1987), defined a fall as an:

...event which results in a person coming to rest inadvertently on the ground or other lower level, and other then a consequence of the following: sustaining a violent blow, loss of consciousness, sudden onset of paralysis, as in a stroke, or an epileptic seizure (p.4).

Some believe that, certain types of falls, such those caused by being struck by an automobile, those occurring in vigorous activities, or falls caused by loss of consciousness have different causes and should be considered different phenomena (Nevitt, 1997). In addition to defining and differentiating between various types of falls, one must be careful to define the individual(s) of interest. For instance, even within the elderly population there are different groups that can be differentiated according to gender, health and home or dwelling type (community, nursing home, hospital). The risk factors associated with the falls in each of these groups are different (Nevitt, 1997; Sudarsky, Masdeu and Wolfson, 1997; Downton, 1996; Northridge, Nevitt and Kelsey, 1996; Northridge et al., 1995; Studenski et al., 1994; Judge, Underwood and Gennosa, 1993; Topper and Holliday, 1993; Speechly and Tinetti, 1991). Each group has intrinsic strengths and weaknesses, and face different environmental problems. Therefore, one cannot treat all elderly individuals the same or use the same methods to effectively and efficiently reduce their fall risk. Keeping this in mind, one should be careful in interpreting research data. In addition to differences in defining falls and populations or subjects, other methodological problems introduce variability into the data. For example, many hospital studies rely on data obtained from incident reports, which may over- or under-report falls. Sutton, Standen, and Wallace (1994) found that 35% of accidents were not documented in incident reports. On the other hand, institutions may over report falls because of a fear of liability and differences in how each defines a fall (Tideikssar, 1997). In a community setting
researchers usually rely on the faller’s interpretation of the incident. Additionally the terms “incidence” and “prevalence” have been found to be used in error in some studies (Cumming, Kelsey and Nevitt, 1990). Specifically “incidence” refers to the number of new cases of a disease in a population over a period of time, while “prevalence” is defined as the number of existing cases of a disease in a given population at a specific time (McDonough, 1994). In addition, results are reported in different “units” (Downton, 1987) which may or may not be convertible. For instance, one cannot readily convert between falls per day and falls per 100 hospital admissions or falls per 10,000 patient days and falls per hospital bed.

Specificity is a very important concept in training for athletics. Athletes have limited time and energy to train for their particular sport or event. Thus, athletes, coaches and trainers are constantly driven to optimize training programs to acquire the finest performances. For example, marathon runners spend the greatest proportion of their training time doing some type of endurance running. While studies may show that marathoners adapt to anaerobic interval training or can improve their strength through weight training these adaptations may not be functional to the athletes’ chosen activity. In addition to taking up time that may be better spent, it may actually decrease the athlete’s endurance performance. Analogously, elderly individuals can be considered athletes in training with specific performance goals in mind. For the elderly individual, that goal might be walking across a room or down the hall, getting in and out of bed or tub, or on and off a chair or toilet. In the same manner as the athlete and coaches working with the athletes, the elder individual and the physical therapists and doctors working with the elderly have a limit to their time and energies and want to optimize the outcome of their efforts. Thus, it is important for them to recognize what training will best fit their needs and most efficiently help them to meet their goals.

2.8.2. Frequency of Falls.

The exact frequency of falls is difficult to determine as not all falls are reported or recorded even if an injury occurred (Downton and Andrews, 1991). Falls that do not result in serious injury are commonly not reported. Downton (1996) hypothesized that some falls were not reported because the individual supposed that falling was an inevitable consequence of advancing age, or that they may lose their independence (e.g., placed in a nursing home) if they admitted falling. With this caveat in mind, both retrospective (Wickham, Cooper, Margetts and Barkder, 1989; Blake et al., 1988; Campbell et al., 1981; Prudham and Evans, 1981) and prospective (O’Laughlin, Robitaille, Boivin and Suiissa, 1993; Campbell, Borrie and Spears, 1989; Nevitt, Cummings, Kidd and Black; 1989; Tinetti, Speechley and Ginter, 1988) studies found similar fall frequencies in the elderly. Not surprisingly, they found that the fall rate increased with age. These studies found that approximately 25 to 30% of those 65+, 35% of those 70+, and 32 to 42% of those 75+ years old fell at least once in a given year. This results in approximately 7 million falls a year (Thapa and Ray, 1996). For reasons unknown, women seem to be more prone to falls than men (Downton, 1996; Nevitt, Cummings, Kidd and Black, 1989). Interestingly, one study found the situation reversed in hospitalized populations. That is, in hospitals men in tended to fall more than women do (Berry, Fisher and Lang, 1981). Again, why this occurs is not known. Fall rates are, regardless of gender, generally higher in hospitals or in nursing homes (especially in those institutions with extremely frail populations) (van Dijk et al., 1993; Rubenstein and Josephson, 1992; Rubenstein et al., 1988; Baker, O’Neill and Karpf, 1984). Four studies found the incidence of falls to be about 140 per 100 patient-years in nursing homes, 165 per 100 patient-years in hospitals, and 400 per 100 patient-years for nursing home residents with dementia (van Dijk et al., 1993; Rubenstein and Josephson, 1992; Meyers et al., 1991; Rubenstein et al., 1988). An individual’s past experience and history have been shown to be related to their fall risk. For example, Nevitt et al. (1989) and Evans, Prudham, and Wandless (1979) found the likelihood of a person falling increases to 60 to 70% if they had fallen in the previous 12 months. Other researchers that have found an increased risk of falls with a history of falls include: O’Loughlin, Robitaille, Boirin and Suiissa (1993); Cwikel (1992); Ryynanen, Kivela, Honkanen, and Laippala (1992a); Meyers et al. (1991); Campbell, Borrie, and Spears (1989); Nevitt, Cummings, Kidd, and Black (1989); Tinetti, Speechley and Ginter (1988); Janken, Reynolds and Sweich (1986).

2.8.3. Injury and Death due to Falls.

Fortunately all falls do not result in serious injury. Keeping in mind that not all falls are reported and that failure to report is probably disproportionately greater for falls that do not result in an injury, it has
been estimated that between 10 and 25% of falls in the elderly result in a serious injury (Tinetti and Speechley, 1989; Tinetti, Speechley and Ginter, 1988), and nearly half of these injuries are fractures (5 to 10% of all falls) (Nevitt, Cummings and Hudes, 1991; Nevitt et al., 1989; Tinetti and Speechley, 1989; Rubenstein et al., 1988; Tinetti, Speechley and Ginter, 1988). At least two studies have found that older women are more likely to suffer an injury following a fall (Sjorgen and Bjornstig, 1991; Sattin et al., 1990). As will be reported later this is opposite to the gender difference for fall related deaths. As was true for fall incidence, injury and death due to falling is seen to increase with age. Thap and Ray (1996) detected a 40 to 60-fold increase between ages 25 and 75. Sattin et al. (1990) reported injury rates increased steadily after individuals reach 65 years old to 138 per 100 for men and 159 per 100 for women greater than 84 years old. Contrary to expectations, one study found that individuals who have multiple falls seem to be less likely to suffer fractures, and consequently they have a lower fracture rate per fall (Baker and Harvey, 1985). However, Nevitt et al. (1991) in a one-year prospective study of 325 individuals with an average age of 70.3 years, found that the risk of injury was associated with the number of falls a person had. They also found that the risk of injury was greater in individuals with neuromuscular and cognitive deficiencies and in those who partook in certain activities (stair climbing, turning around and reaching for objects). Tinetti (1987) and Nevitt, Cummings and Hudes (1991) found that while individuals will fall performing such activities as rising from a chair and other transfer activities, or stooping or bending, these falls had a low risk of injury associated with them. This may be because of the relatively small drop in height associated with these falls. When a fracture does take place, it usually occurs in the extremities, pelvis, ribs or on the face (Nevitt and Cummings, 1992). Along with fractures, 5% of falls result in soft tissue injuries that require immediate medical care (O’Laughlin et al., 1993; Tinetti, Clauss and Liu, 1992; Nevitt, Cummings and Hudes, 1991; Rubenstein et al., 1988; Tinetti et al., 1988; Tinetti and Speechley, 1981). These injuries include hematomas, head injuries with loss of consciousness, dislocations and other joint injuries, lacerations, and sprains (Tinetti, Clauss and Liu, 1992; Sattin et al., 1990). Between 30 and 50% of falls result in minor soft tissue injuries, and the remainder (~35 to 60%) causes no or insignificant physical injury (Nevitt, 1997).

The direction of a fall influences the probability of a distal forearm fracture. Conventional thought is that a forward fall is accompanied by a protective extension of the arms in an effort to break the fall (Melton, Chao and Lane, 1988). The result is a break in the bone along with the break in the fall. However, there is some evidence that falling backward may produce more wrist fractures (Nevitt et al., 1993), as a person will generally fall on their hands prior to landing on their coccyx, sacrum or ischium. The risk of distal forearm fracture seems to be related to bone integrity. In women, the incidence of distal forearm fractures rises at or near menopause, and then reaches a plateau by age 65-70 when their frequency is surpassed by hip fractures (Nevitt and Cummings, 1993; Melton et al., 1988).

Aside from death, one of the most devastating outcomes of a fall is a hip fracture. Although hip fractures make up a small percentage (1-2%) of all fractures (Sattin, 1992; Nevitt et al., 1989; Tinetti et al., 1988), approximately 200,000 hip fractures are recorded in the United States each year (Sudarsky, 1990), 660 per 100,000 older persons (Jacobson et al., 1990). The rate of occurrence for hip fractures has been found to double with each decade after the age of 50 (Cummings et al., 1985). They calculated that an individual has a 15% chance of getting a hip fracture if they were female and 5% if male. This is equal to the likelihood of the female contracting breast and uterine cancer and the male prostate cancer (Cummings et al., 1985). Several researchers have found that women are more likely then men to have a hip fracture (Jacobson et al., 1990; Kelsey and Hoffman, 1987; Wolinsky and Fitzgerald, 1994; Downton, 1996). However, one study found that the ratio of falls to hip fractures was higher in men than in women (Cumming and Klineberg, 1994b). While another study found that regardless of a difference in fracture rates with gender, the two genders show a similar rate of hip fracture increase with age (Hedlund and Lindgren, 1987). Other researchers have found that Caucasian women have the highest incidence of hip fracture, while Caucasian men are second (Vogt, 1995; Farmer et al., 1984). Blacks in general have a relatively low incidence of fracture while Hispanics and Asians are somewhere in the middle (Jacobson et al., 1990; Silverman and Madison, 1988). Melton (1988) found that around 90% of all hip fractures were the result of a fall. The remaining 10% occur spontaneously in weight bearing because of osteopenia (Tideiksaar, 1997). Many studies have indicated that low bone mass increases ones risk of experiencing a hip fracture (Greenspan et al., 1994; Kelsey et al., 1992; Cummings et al., 1990). Factors other than low
bone mass, gender and race that seem to increase a person's risk of hip fracture include: stature or height of fall (Hayes et al., 1993; Mayo et al., 1993; Nevitt et al., 1993; Grisso et al., 1991), falling sideways as opposed to front or back (Greenspan et al., 1994; Hayes et al., 1993; Nevitt et al., 1993); loss of protective reflexes (Nevitt et al., 1993), falls caused by a loss of consciousness as compared to non-syncopal falls (Nevitt et al., 1991); history of previous falls (Cumming and Klineberg, 1994; Lau and Donnan, 1990; Wolinski and Fitzgerald, 1994), lower extremity dysfunction, neurological conditions, barbiturate use and long acting benzodiazepines (Grisso et al., 1991; Ray et al., 1987), slight or thin build (Greenspan et al., 1994; Grisso et al., 1991; 1994; Malmivarra et al., 1993; Cummings and Nevitt, 1989; Pruzansky, Turano, Lucky and Senie, 1989; Kelsey and Hoffman, 1987). Increased fat and muscle build surrounding vulnerable areas of the hip may be capable of absorbing the impact of a fall and thus decreased the risk of hip fracture. In addition, some believe that the higher weight may mechanically stimulate the bone into maintaining its mass and structural integrity.

In addition to fractures, many older men and women suffer head injuries as a result of falls (Pentland et al., 1986). Head injuries do not seem to prefer either gender (Sattin et al., 1990; Galbraith, 1987). In addition to subdural hematomas, bruises and lacerations are common results of impacts to the head in a fall (Tideiksaar, 1997). Although not direct results of the fall, dehydration, pressure sores, rhabdomyolysis (acute potentially fatal muscle degradation) (Marcus et al., 1992), and psychological trauma may result from a long lie. It has been estimated that approximately 50% of fallers need help getting up (Nevitt, Cummings and Hudes, 1991) and that between 8 and 20% of falls by older persons living independently in the community result in lies lasting greater than an hour (Campbell et al., 1990).

Unfortunately, permanent or long-term disability is too often a consequence of a fall. The disability may be physiological, psychological or both. For example, almost 25% of the falls suffered by elderly living in the community result in a limitation of activity because of injury or fear of falling (Nevitt, Cummings and Hudes, 1991; Tinetti, Speechley and Ginter, 1988; Vellas et al., 1987), and approximately 20% of restricted activity in the elderly is due to falling (Kosorok et al., 1992, National Center for Health Statistics, 1987). Sometimes it is not the individual who fell that is the problem, but family members and other caregivers may over react and become over-protective and place restrictions on the faller (Tideiksaar, 1997).

This fear may be so severe that it leads to an inability to walk (Murphy and Isaacs, 1982). Additionally, an individual may have a fear of falling even if they have no history of falls (Downton and Andrews, 1990). Tinetti and Powell (1993) define fear of falling as a lasting concern about falling that leads to an individual avoiding activities that they are capable of performing. Approximately 10 to 50% of those elderly individuals living in the community with a recent fall have a fear of falling (Arfken, Lach, Birge and Miller, 1994; Franzoni et al., 1994; Baraff et al., 1992; Downton and Andrews, 1990; Tinetti and Speechley, 1989). Tideikssar (1997) believes that many will be reluctant to admit to a fear of falling and thus the prevalence reported in literature may be low. The fear of falling is seen to increase with age (Arfken, Lach, Birge and Miller, 1994) and seems to be more prevalent in women (Arfken et al., 1994; Walker and Howland, 1991; Cwikel, Fried and Galinsky, 1990; 1989). However, there are some that have not seen this gender difference (Downton and Andrews, 1991). Just the threat or chance of falling may be enough. For example, Arfken et al. (1994) found that half of community dwelling older persons who were fearful had not fallen in the prior 12 months. Rubin and Cummings (1992) found that women who have learned about osteoporosis and the possible consequences of a fall may develop an intense fear of falling. This is analogous to the natural fear one experiences when standing next to the edge of a deep precipice even though they may not have fallen off one before. The fear of falling may not be detrimental if it matches this analogy and represents a rational response to danger. For frail individuals a “healthy” fear will drive the individual from situations that may be beyond their capability or capacity, while a lack of fear will allow the person to get into unwanted and dangerous circumstances. Fear may also cause the individual to become preoccupied with fall avoidance to an extent that it effects their mobility (Franzoni et al., 1994; Maki, Holliday and Topper, 1991). It has also been reported that older persons with a fear of falling exhibit increased sway on postural performance tests (Maki et al., 1991). The association between fear of falling and sway becomes more significant under conditions of decreased visual input (for example, eyes are closed) (Baloh et al., 1994). It is suggested that persons with poor balance and fear of falling may feel anxious about their instability and thereby display anxiety related effects on balance, resulting in
instability (Maki et al., 1991). Typical gait of an individual that fears falling is hesitant and irregular; sometimes these individuals rush or surge from support to support (Tideiksaar, 1997).

In addition to falls resulting in fear, older persons may experience depression, shame, loss of confidence or self worth, anxiety or anger (Tideiksaar, 1997). These feelings may result from a feeling of helplessness and loss of self-sufficiency. They may take the fall as an omen indicating the beginning of the end. In addition, they may worry about the strain or added burden that they will impose on other family members. The anger may develop from the loss of their independence and self-sufficiency that they have experienced for decades, or a role reversal (now dependent on their children rather than being a provider).

Unfortunately, the effects of a fall are not always short term. Grisso et al. (1992) found that 40% of the people who fell and were treated in an emergency room, continued to be in pain or have a disability two months later, and 16% were still experiencing pain seven months after the incident. Many researchers have found that falls directly or indirectly place an individual’s overall quality of life into a self perpetuating downward spiral (Tinetti et al., 1994; Wolinsky, Johnson and Fitzgerald, 1992; Dunn, Rudberg, Furner and Cassel, 1992). Kiel et al. (1991) discovered that fallers as compared to non-fallers, and multiple fallers as compared to individuals who fell only once, had larger declines in their functional abilities. For example, approximately 75% of individuals who ambulated independently prior to a hip fracture needed assistance to walk for the first 6 to 12 months after fracture (Jette et al., 1987; Marottoli, Berkman and Cooney, 1992; Magaziner et al., 1990). One study found that 20% to 30% of individuals who lived independently in the community before having a hip fracture from a fall were still in long term care one year after the injury (Ray, Griffin and Baugh, 1990). It has been estimated that between 30% and 60% of patients with hip fractures are discharged from the hospital directly to nursing homes (Borkhan and Quirk 1992; Sattin et al., 1990; Fitzgerald, Moore and Dittus 1988). Up to 40% continue to reside in the nursing home 6 months later (Bonar, Tinetti, Speechley and Conney, 1990; Sattin et al., 1990; Fitzgerald, Fagen, Tierney and Dittus, 1987; Palmer et al., 1987) and as many as 57% after a 1 year period (Fitzgerald and Dittus, 1990; Fitzgerald, Moore and Dittus, 1988). Sattin et al. (1990) found that they were moved to a nursing home following their stay in the hospital if they did not die first. Gyrfå, Aimes and Ashley (1997) and Overstall (1985) found that only 50% those admitted to a hospital because of a fall were living one year later, and Lucht (1971) found that 10% died before they had a chance to be released.

While the vast majority of falls do not result in death, people do die from falls. Baker et al. (1992) and Baker, O’Neill and Karpf, (1984) found that falls represent the sixth leading cause of death in persons 65 and older. If one limits the scope to deaths due to unintentional injuries, falls are second only to death due to automobile accidents in those 65 to 75 years and first in those over age 75 (National Safety Council, 1990). DeVito et al. (1988) found that of all U.S. deaths in 1984 5.1/100,000 people died from falls; in individuals 65+ the rate increased to 31.3/100,000. Accidental injury is the sixth leading cause of death in people older than 65, with falls being the major cause of injury in this age group (Sattin, 1992; Imms and Edholm, 1981). As can be inferred from these studies the risk of death from a fall increases with age.

The rate for fall related deaths differ by gender, by race, by residence and by study. One study found that more than 50% of injury related deaths in women were related to falling, while approximately 40% of injury related deaths in men were related to falling (Sattin, 1992). Kohn et al. (1991) found that older men had a greater mortality rate from falls than did women. Sattin et al. (1990) stated that men were twice as likely as women were to die from a fall. Baker et al., (1992) hypothesized that this difference may be in part to more men than women falling from large heights such as off ladders. However, others believe that older men may just be more frail then older women and therefore are unable to recover as well from the fall injury (Kohn et al., 1991; Campbell, Borrie and Spears, 1989). In another study, Rishe et al. (1991) reported that older men were more likely than older women to die from a fall, possibly because of the higher rate of death in men following a hip fracture. The death rate after a femoral neck fracture can be up to 20% during an acute hospital stay (Greatorex, 1988), and continues to be greater than normal for a considerable time afterward (Downton, 1996). As mentioned previously, falls seem to be an indicator of increased frailty and risk of dying (Campbell et al., 1985), as an increase in the number of falls have been observed in elderly residing in institutions just before death (Gyrfe, Aimes and Ashley, 1977). The time of year influences the risk, with deaths from falls being greater in winter (Hemenway and Colditz, 1990). In general, the fall risk may be increased in winter months because of adverse environmental conditions such as slippery surfaces and lower temperatures (Nevitt, 1997; Sattin, 1992). Campbell, Spears, Borrie and
Fitzgerald (1988) found that older women experienced an increased rate of falls during the winter when temperatures dropped to one degree Celsius or below. They attributed the increased risk of falls to defects in thermoregulatory mechanisms and hypothermia. Baker et al. (1992) found that Caucasians are twice as likely to die from a fall as are black individuals. A popular explanation for this is that the incidence of osteoporosis and risk of hip fractures in Caucasians is higher (Griffin, Ray, Fought and Melton, 1992; Cummings, Kelsey, Nevitt and O'Dowd, 1985). There exist differences within the United States in fall mortality depending if one lives in an urban vs. rural area or northern vs. southern states. Baker et al. (1992) found that mortality was higher in urban areas as compared to rural areas, while Hogue (1980) found the difference between mortality in those located in the New England area higher than those in Southern states. The reason for these differences are unknown, but may be related such factors as variations in climate, differences in the manner that different regions report deaths and differences in the activity of the populations studied.

From death certificates, the National Center of Health Statistics tabulated over 8,000 deaths resulting from falls in the elderly in the United States (Nevitt, 1990). In a later work, Nevitt (1997) stated that this number may be an under-representation. Death certificates usually do not include falls as a cause or even a contributing factor of death. More frequently, pneumonia or other disease will be listed (Tideiksaar, 1997). Wild, Nayak and Isaacs (1981a) found 26% of deaths of community dwelling elderly were related to falls, however none of the death certificates indicated this.

Two other factors that seem to relate to mobility and falls are “long lies” and hip fractures. Tinetti, Liu and Claus (1993) and Vellas et al., (1987) found long lies to increase the risk of death independent of physical injury. This is not surprising, since a long lie is an indication of the overall frailty of an individual. Hip fractures may also be an indication of frailty. However, there is a positive note associated with the relationship between hip fractures and mortality. This is that the statistics seem to be improving. Older studies have shown a mortality rate as high as 50% one year following a hip fracture, while more recent studies have shown rates from 14 to 36% (Tideiksaar, 1997). With respect to hip fractures and death the presence of co-factors such as chronic diseases, poor cognitive status, postoperative complications, male gender, and whether or not the person is placed in an institution or able to return to a community residence following the fracture all increase the chances of death following a hip fracture (Tideiksaar, 1997).

2.8.4. Risk Factors for Falls.

It is well known that people fall. However, it is not well known how to prevent or minimize falls or the injuries and deaths resulting from them. Often it is not just one factor alone, but the combination of many factors that predisposes a person to fall. Ordinarily, corrections in muscular tension (or joint torques) are constantly being made according to sensory information of posture and environmental conditions. Posture and environmental conditions are noted by visual, vestibular (cochlea and semicircular canals) and proprioceptor/somatosensory (muscle spindles, Golgi tendon organs and joint and skin receptors) systems. Adjustments in muscular tone are made according to the results of the central processing that occurs in the cerebellum, brainstem, basal ganglia, sensorimotor and association cortex and spinal cord (see reviews by Patla, 1995; 1996). Adjustments are also made according to internal influences such as experience, mood and arousal or attention along with the information provided by the sensory system.

Poor balance and falls often reflect the combined impact of several abnormalities, such as the combination of poor vision, loss of peripheral sensory function and muscle weakness. In addition, activity related and environmental factors, such as tasks that stress postural abilities and poor lighting may interact with intrinsic (host) susceptibilities to precipitate a fall (Nevitt, 1997, p. 19). Teasing out the contributions of the various components is difficult because of redundancy of sensory input, the crosstalk or influence that each factor has on the other factors and the plasticity or evolving nature of the “balance system(s)”. Part of the problem lies with the existence of wide variability in structure and function between individuals even within a specific well defined group (i.e. by age, gender, health status...). These differences between individuals within in a group may be greater than the differences between the groups. For instance, Dowton (1996) found differences between some fallers’ and non-fallers’ characteristics to be small, and in some cases, there was a large overlap between the groups. Making the problem even more difficult is the fact that there are almost an infinite number of potential causes of falls, including determinants that may have no direct relation to balance such as cardiac
arrhythmia, epilepsy, or diabetes (Downton, 1996). Furthermore, interactions between the various factors may exist, leading to great difficulty in identifying any individual relationship. Consequently, it is difficult to develop functionally relevant tests or measurers of fall risk. For example, an easy experimental measure is sway during stance, however few individuals fall while standing “still” and the association between static postural control or any sway measure and falls is not yet clear (Downton, 1996; Tang and Woollacott, 1996). Patla et al. (1989) point out that individuals with better static balance control do not necessarily have better dynamic balance control and vice versa. Thus, to adequately evaluate an individual’s balance capabilities; a wide range of balance situations needs to be tested. However, inclusion of all possibilities is impractical if not impossible.

2.8.4.1. Intrinsic Risk Factors.

Even though proprioceptive (or somatosensory), visual and vestibular input are crucial for detecting the relationship between one’s CM and base of support, scientific studies have not found a consistent relation between these factors and falls (Nevitt, 1997). As discussed before, this may be due to the inability of any sensory organ, collection of organs, or system to directly measure the location of the CM of the body. Furthermore, the redundancies and interactions of other systems hide the relationship between a sensory system and balance. One sensory system can compensate for small deficiencies (due to disease, injury, or age) in one or more of the other sensory systems. The CNS is also capable of altering or correcting its interpretation of sensory input to match the circumstances. For instance, the CNS to achieve correct results can adjust images that are altered by lenses or prisms. Also, the discrepancy in the relation between somatosensory or proprioceptive impairment and fall risk may be due to the fact that clinical measures are too crude to detect functionally important differences (Nevitt, 1997).

In his review of literature, Nevitt (1997) rated intrinsic risk factors of falling. A risk factor was strongly related to falls if most studies he reviewed found a positive association between the risk factor and falling. A moderate rating was given to factors that were associated with falling in several but not all studies, while a weak rating was given to those factors in which a relationship to falling was found in just one or two studies, or if studies were conflicting and inconsistent in their findings. A summary of his findings is given in Table 2.8.4.2.1.
Table 2.8.4.2.1. Intrinsic risk factors of falls in the elderly.

<table>
<thead>
<tr>
<th>STRONGLY RELATED TO FALLS IN THE ELDERLY</th>
</tr>
</thead>
<tbody>
<tr>
<td>• age</td>
</tr>
<tr>
<td>• gender</td>
</tr>
<tr>
<td>• activities</td>
</tr>
<tr>
<td>• history of falls</td>
</tr>
<tr>
<td>• gait kinematics (speed, step length)</td>
</tr>
<tr>
<td>• lower leg strength</td>
</tr>
<tr>
<td>• cognitive impairment</td>
</tr>
<tr>
<td>• specific diseases (e.g., Parkinson’s disease)</td>
</tr>
<tr>
<td>• medications</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>MODERATELY RELATED TO FALLS IN THE ELDERLY</th>
</tr>
</thead>
<tbody>
<tr>
<td>• low or high level of physical activity or exercise</td>
</tr>
<tr>
<td>• laboratory tests of balance with altered sensory or support conditions</td>
</tr>
<tr>
<td>• qualitative gait abnormalities</td>
</tr>
<tr>
<td>• hip or knee pain or reduced range of motion</td>
</tr>
<tr>
<td>• foot problems</td>
</tr>
<tr>
<td>• impaired visual acuity</td>
</tr>
<tr>
<td>• depression or anxiety</td>
</tr>
<tr>
<td>• arthritis</td>
</tr>
<tr>
<td>• stroke</td>
</tr>
<tr>
<td>• dementia</td>
</tr>
<tr>
<td>• incontinence</td>
</tr>
<tr>
<td>• antidepressants</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>WEAKLY RELATED TO FALLS IN THE ELDERLY</th>
</tr>
</thead>
<tbody>
<tr>
<td>• laboratory tests of static balance (quiet standing)</td>
</tr>
<tr>
<td>• impaired knee or plantar reflexes</td>
</tr>
<tr>
<td>• slow reaction time</td>
</tr>
<tr>
<td>• impaired contrast sensitivity or depth perception</td>
</tr>
<tr>
<td>• visual perceptual error</td>
</tr>
<tr>
<td>• impaired lower extremity sensory function</td>
</tr>
<tr>
<td>• frontal cortex or release</td>
</tr>
<tr>
<td>• cerebellar, pyramidal, extrapyramidal</td>
</tr>
<tr>
<td>• postural hypotension</td>
</tr>
<tr>
<td>• cardiovascular</td>
</tr>
</tbody>
</table>

Modified from Nevitt (1997).

Rubenstein and Josephson (1997) combined the information from several studies to categorize the causes of falls and their relative frequencies (Table 2.8.4.2.2). The mean percent was calculated from the total of all the falls reported in all the studies combined (i.e., portion of 3,628 total falls).
Table 2.8.4.2.2. Causes of falls in elderly adults in the literature.

<table>
<thead>
<tr>
<th>cause</th>
<th>mean (%)</th>
<th>range (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>accident or environment related</td>
<td>31</td>
<td>1.5 - 53</td>
</tr>
<tr>
<td>gait or balance disorder or weakness</td>
<td>17</td>
<td>4.3 - 39</td>
</tr>
<tr>
<td>dizziness or vertigo</td>
<td>13</td>
<td>0.3 - 30</td>
</tr>
<tr>
<td>dorp attack</td>
<td>9</td>
<td>0.3 - 52</td>
</tr>
<tr>
<td>confusion</td>
<td>5</td>
<td>0.1 - 14</td>
</tr>
<tr>
<td>postural hypotension</td>
<td>3</td>
<td>0.2 - 24</td>
</tr>
<tr>
<td>visual disorder</td>
<td>2</td>
<td>0.5</td>
</tr>
<tr>
<td>syncope</td>
<td>0.3</td>
<td>0.3</td>
</tr>
<tr>
<td>other causes*</td>
<td>5</td>
<td>0.2 - 21</td>
</tr>
<tr>
<td>unknown</td>
<td>5</td>
<td>0.2 - 21</td>
</tr>
</tbody>
</table>

*includes arthritis, acute illness, drugs, alcohol, pain, epilepsy and falling from a bed.

In another review of 16 studies in the literature, Rubenstein and Josephson (1997) listed eight risk factors related to falls and calculated the number of studies that reported significant association, relative risks and odds ratios for each (Table 2.8.4.2.3).

Table 2.8.4.2.3. Risk factors for falls from sixteen studies in the literature.

<table>
<thead>
<tr>
<th>risk factor</th>
<th>significant association/total</th>
<th>mean RR (OR)</th>
<th>range</th>
</tr>
</thead>
<tbody>
<tr>
<td>weakness</td>
<td>11/11</td>
<td>4.9 (8)</td>
<td>1.9 - 10.3</td>
</tr>
<tr>
<td>balance deficit</td>
<td>9/9</td>
<td>3.2 (5)</td>
<td>1.6 - 5.4</td>
</tr>
<tr>
<td>gait deficit</td>
<td>8/9</td>
<td>3.2 (5)</td>
<td>1.6 - 5.4</td>
</tr>
<tr>
<td>visual deficit</td>
<td>5/9</td>
<td>2.89 (9)</td>
<td>1.1 - 7.4</td>
</tr>
<tr>
<td>mobility limitation</td>
<td>9/9</td>
<td>2.5 (8)</td>
<td>1.0 - 5.3</td>
</tr>
<tr>
<td>cognitive impairment</td>
<td>4/8</td>
<td>2.4 (5)</td>
<td>2.0 - 4.7</td>
</tr>
<tr>
<td>impaired functional status</td>
<td>5/6</td>
<td>2.0 (4)</td>
<td>1.0 - 3.1</td>
</tr>
<tr>
<td>postural hypotension</td>
<td>2/7</td>
<td>1.9 (5)</td>
<td>1.0 - 3.4</td>
</tr>
</tbody>
</table>

where:

significant association = number of studies with significant association between the risk factor and falls
total = total number of studies examining that factor
RR = relative risk (prospective studies)
OR = odds ratios (retrospective studies)
number in parentheses indicates the number of studies that reported relative risks or odds ratios
It can be argued with some confidence that almost all the risk factors listed in Tables 1, 2 and 3 are related to one another.

...balance in the elderly ... [is a] ... highly complex set of overlapping sensorimotor, musculoskeletal, psycho emotional and perceptual functions. Confounding this complexity are diverse additional factors such as age, gender, unmanifested disease, learning, activity level, genetic endowment, socioeconomic status, risk- taking behavior and fitness. (Whipple, 1997, p. 355).

Similar statements can be found in nearly every research publication on balance. With this in mind the following sections review and summarize what is known about the various factors that influence (or at least are correlated) to falling and injuries from falls.

2.8.4.1.1. Age.

From table 2.8.4.1.2, one can see that various studies found a strong association between old age and the risk of falling (Ryynanen et al., 1993; Meyers et al., 1991; Nevitt et al., 1989; Blake et al., 1988; Tinetti et al., 1988). Fortunately, increased age per se may not predispose a person to falls, but the increased risk of falling may be due to other factors that correlate highly with age. For instance, other risk factors such as activities of daily living, mobility, physical performance and lower extremity strength tend
to decrease with age. Thus, it may be that the increased risk of falls is only indirectly linked to age through these confounding factors. Research has yet to answer this question (Whipple, 1997).

2.8.4.1.2. Activity:

A major difficulty is the confounding nature of inactivity and age, and unfortunately, almost one-half of the elderly do not engage in regular physical activity (Wagner, et al., 1992). Blair, Brill and Kohl (1988) found that physical activity decreases with age. As Pescatello and Judge (1995) indicated, there are many physiological alterations associated with age, which are very similar to the physiological alterations associated with disuse (Table 2.8.4.1.2). Judge (1997) summarizes the situation very well by noting that “declines in functional status are related closely to chronic disease burden, but physical activity may delay disability by delaying the onset of chronic disease and by maintaining endurance and body composition (muscle mass, percentage body fat and bone mineral density (p. 394)).”

Table 2.8.4.1.2. Similarities between age and disuse caused physiological changes.

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Aging</th>
<th>Disuse</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Body composition</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>lean body mass</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td>fat mass</td>
<td>up</td>
<td>up</td>
</tr>
<tr>
<td>bone mass</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td><strong>Cardiovascular function</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>VO₂max</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td>cardiac output (resting, maximal)</td>
<td>down, NC</td>
<td>down</td>
</tr>
<tr>
<td>stroke volume, resting</td>
<td>down</td>
<td>down, NC</td>
</tr>
<tr>
<td>stroke volume, maximal</td>
<td></td>
<td></td>
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<tr>
<td>heart rate, resting</td>
<td>up, NC</td>
<td>up</td>
</tr>
<tr>
<td>heart rate maximal</td>
<td>down</td>
<td>down, NC</td>
</tr>
<tr>
<td>baroreceptor function</td>
<td>down</td>
<td>down, NC</td>
</tr>
<tr>
<td>A-VO₂ difference</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td><strong>Musculoskeletal system</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>muscle fiber number and size (type II &gt;type I)</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td>muscle strength</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td>capillary density</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td>muscle oxidative capacity</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td><strong>metabolic/hematological systems</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>basal metabolic rate</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td>exercise induced hyperthermia</td>
<td>up</td>
<td>up</td>
</tr>
<tr>
<td>glucose tolerance</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td>insulin sensitivity</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td>calcium balance</td>
<td>down</td>
<td>down</td>
</tr>
<tr>
<td>cholesterol levels (low density lipoprotein)</td>
<td>up</td>
<td>up, NC</td>
</tr>
</tbody>
</table>

Notes: disuse data were taken from studies of hypokinesia, immobilization, bed rest, or weightlessness.
NC = no change in a majority of studies.

As seen in table 2.8.4.1.2, there are many similarities between human responses to aging and to inactivity. A “Catch 22” relationship exists between activity and risks for falls and injuries. Some studies have found that increased activity places the elderly in more compromising and potentially dangerous situations and thus increases their chance of falling and injury (Speechley and Tinetti, 1991; Finsen, 1988). However there are others who have found that individuals who are more active tend to maintain their neuromuscular function, protective reflexes and cognitive function (Molloy, Richardson and Grilly, 1988), tend to have more “efficient” postural responses and have a lower fall rate (Gabell, Simons and Nayak,
1985) additionally increased physical activity insulates one against fracture (Law, Wald and Meade, 1991; Wickham et al., 1989; Boyce and Vessey, 1988).

Another caveat is that the term activity lacks specificity. Consider the daily activities of an elderly rancher, contractor, lawyer and architect. The daily activities of each of these individuals are different. This is true even if we look at a specific activity such as walking. Contrast the requirements of a rancher walking across a soft uneven field with that of a lawyer walking across a smooth firm hardwood floor of a courtroom.

2.8.4.1.3. Gender.

As alluded to earlier one’s fall risk may be related to one’s gender as found in several studies (Downtown and Andrews, 1991; Hornbrook et al., 1991; Campbell, Borrie et al., 1990; Campbell, Spears, and Borrie, 1990; Blake et al., 1988, Sorock and Shimkin, 1988; Tinetti et al., 1988). For example, Overstall (1985) found that in individuals between the ages of 65 and 69, 15% of the males fell and 30% of the females fell. The percentages increased to 33% and 44% respectively in an older group (80 to 84 years). This finding is not universal. Several studies have found that the difference in fall prevalence associated with gender begins to disappear after 75 (Luukinen, Koski, Hiltunen and Kivela, 1994) and by 84 or 85 is nearly equal (Svensson, Rundgren and Landahl, 1992; Woodhouse et al., 1983). The reasons for these gender differences are not all physical. For example, Cwikel (1992) found that women reported falling twice as often then men. Campbell, Spears and Borrie (1990) found that fall rates were equal in the two genders however, the men tended to deny the fact that they had fallen.

2.8.4.1.4. Nervous System.

Commencing at about age 20 the brain weight declines (Duara et al., 1985), the cortex thins (Earnest et al., 1986), and the primary motor and sensory cortex lose neurons (Cech and Martin, 1995). Approximately ten years later, at about age 30, the hippocampus shows a decrease in neurons (Ball, 1977; Konigsmark and Murphy, 1970). During this time (between the ages of 30 and 40) the cerebral volume declines by 11% (Yamamura et al., 1980). Several studies have found greater neuron loss in the higher order association areas as compared to the primary motor or visual cortex, and in the frontal and temporal lobes more than in the parietal lobes with age (Kemper, 1984; Ball, 1977; Brody and Vijayashanker, 1977). Haug et al. (1984) and Haug (1984) found that the total number of neurons in the cortex does not change but the neurons do shrink in size with age. The basal ganglia (Whitbourne, 1985) and brain stem nuclei (Konigsmark and Murphy, 1970; Moatamed, 1966) show little or no change as a result of aging. At present, there is no conclusive evidence of glial cell changes with aging (Cech and Martin, 1995), which would have an indirect effect on neuron function if nerves were unable to receive nutrients. Even with all these structural changes Cotman and Holets (1985) state that few overall changes exceed 25% of the total area, except in diseases states, and only in the last few months of life. Many studies have shown biochemical changes accompany age. Decreases in neurotransmitter synthesis enzymes and receptor sites in the CNS and PNS have both been recorded (Rogers and Bloom, 1985; Rowe and Troen, 1980). An exception to the general decrease in neurotransmitters is the increase in norepinephrine levels that has been observed in older subjects (Katzman and Terry, 1983). This may be a functional response to the decrease in receptor sensitivity with age (Whitbourne, 1985). Cerebral blood flow has been shown to decrease in some studies and not in others. The difference in results may be related to differences in sensory function or the presence or absence of arteriosclerosis (Cech and Martin, 1995). For example, Dura et al. (1985) found different results depending if they controlled for sensory input to the brain. They found differences with age if the data was not normalized and no differences if the data was.

A functional result of these changes is that older individuals may have to concentrate on or pay more attention to balance than do younger individuals. Several investigators have reported an association between certain cognitive factors and postural sway. Stelmach, Zelaznik and Lowe (1990) studied postural sway in a group of healthy older and younger individuals and found that following an arm swinging task and math task performed concurrently, the older persons exhibited greater sway then the younger persons and took longer to recover from postural instabilities. Teasdale, Bard, LaRue and Fleury (1993) examined young and older persons performing an auditory reaction time task while maintaining an upright posture, while at the same time altering visual and surface conditions. They found that as sensory information or
input decreased, the postural task became harder for older people, and required more of their attention. These studies suggest that secondary attention demanding tasks may interfere with the control of posture in older persons. Several have found that a delay in the voluntary onset of postural muscles in response to anticipatory balance disturbance may be affected by cognitive factors (i.e., decreased attention levels) (Woollacott and Manchester, 1993; Zattara and Bouisset, 1986).

In general, the conduction velocity of peripheral nerves decreases with age (Stelmach and Worringham, 1985; Schaumburg et al., 1983), which increases the lag between the stimulus and the response, leading to a decreased ability to respond to stress. The result is that measures that are more drastic will have to be utilized, and errors may never be corrected. Sensory nerve impulse speed has been shown to decrease after age 30 (Buchta et al., 1984), while motor nerve conduction velocity decreases by one m/s per decade after age 15 to 24 (Schaumburg et al., 1983). These changes result in the slowing of simple reaction times (Stelmach and Worringham, 1985). Birren (1979) found a 20% increase in reaction time in 60-year-old subjects compared to 20-year-old subjects. Light and Spirduso (1990) found that age has a negative effect on complex reaction time especially with increasingly difficult tasks. Laufer and Schweitz (1968) calculated that about 4% of the decrease in reaction time with aging is attributed to motor nerve conduction velocity and about 10% to a decrease in sensory nerve conduction velocity. Some of the differences in reaction time may be due to changes in muscle along with changes in nerve. If reaction time is separated into pre-motor time (PMT, the time between the stimulus and the EMG activity), and movement time (MT, time between the EMG onset and the movement) the muscular component can be separated from the neural component. Studies have shown that both are reduced with aging (Cech and Martin, 1995). On a positive note, Spirduso (1980, 1975) found that active older individuals have faster reaction times than sedentary older individuals. Unfortunately, Panton et al. (1990) did not find that exercise significantly improved reaction time.

Declines in sensation begin in adulthood and progress with age. These changes do not always directly result in a functional degradation. In part this is due to development of cross-talk between the different sensory systems or a plurality in ways the human body monitors balance and posture, which sometimes makes it difficult for researchers to measure the status of a particular sense. In addition, perceptions of stimuli change with age. For example, physiological or mechanical reasons cannot entirely explain one’s adaptation to pain with age. Given this, it is generally found that older adults show a decrease capability to detect touch, vibration, temperature and pain (Williams, 1990; Kenshalo, 1977).

The threshold for light touch increases in the hands and the feet with age after 40 (Williams, 1990), while the threshold for vibration begins to decline at age 50 (Steiness, 1957).

The loss of vibratory sensation and joint position has been found to be greater in the lower rather than the upper extremities (Olney, 1985; Skinner, 1984; Brocklehurst, 1982; Potvin, et al., 1980; Kokmen, 1978). Generally, it would be thought that the loss of vibration sensation would be attributable to the decline in number of pacinian corpuscles with age. However, Kenshalo (1977) believes the loss of vibration sensation is due to decreased nerve conduction. Several investigators have found an association between declines in vibratory sense and instability; however, the results vary widely among different age groups, and between older men and older women. For instance, Brocklehurst et al. (1982) reported an inverse relationship between increased postural sway and impaired lower extremity vibration sense in women aged 75 to 84 years, but not in women aged 65 to 74 or beyond 85 years. MacLennan et al. (1980) showed a significant association between vibration sense and sway in women aged 75 to 84 years, but no association in women aged 65 to 74 or men aged 65 to 84. Era and Heikkinen (1985) found a positive association between sway and vibration sense in men aged 51 to 55 years but not in men aged 71 to 75 years. In an effort to explain these differences, Olney (1985) postulated that despite the fact that vibration and proprioception are mediated via similar peripheral pathways, it is possible that the neurons involved in vibration sense decline or degenerate independently from proprioceptive function, and hence do not necessarily demonstrate age related declines in a given individual.

Proprioceptive input, as evidenced by decreased joint position sense and cutaneous vibratory sensation, has been shown to decline with age (Olney, 1985; Skinner, 1984; Potvin, 1980; Kokmen, 1978; Whanger and Wang, 1974). For example, Kaplan et al. (1985) found that women lose proprioception of the knee with age, while Kokmen et al. (1978) could not find any loss in finger movement perception. As a general rule of thumb Williams (1990) found that lower extremity joint proprioception thresholds were
double in individuals older than 50 when compared with individuals younger than 40, but he could not find any major changes in joint kinesthesia (perception of active joint motion). MacLennan et al. (1980) measured proprioception in the toes, but did not find an association with increased body sway. Duncan, Studenski, Chandler and Prescott (1992) in a study of older individuals found a strong relationship with impaired vibration sense and sway, but not passive joint movement. Lord, Clark and Webster (1991) found that poor joint position sense of the toes was associated with increased body sway. Woollacott et al. (1982) showed that older persons had greater postural sway when ankle proprioception was eliminated. Skinner et al. (1984) found that the decrease in joint position sense was greatest in slow movements. Simoneau et al. (1996) found subjects with diabetic sensory neuropathy had a significant loss of ankle movement perception. Although some of their subjects had cutaneous sensory loss secondary to diabetic neuropathy, Simoneau et al. (1996) were unable to directly attribute the ankle joint position loss to the loss in cutaneous sensation. Simoneau et al. (1994) found that subjects with diabetic sensory neuropathy but not with diabetes per se, had more instability as measured by quiet standing on a force platform. They hypothesized that this increased sway was utilized by the subjects with sensory neuropathy to generate a stronger input signal. Brocklehurst, Robertson and James-Groom (1982) found that this decrease in joint position sensation was not correlated with postural sway in older adults. Other mechanisms may hypothetically compensate for loss in these individuals.

Cervical spine mechanoreceptors have been shown to contribute to static postural sensation (Wyke, 1979) as they monitor differences in movement between the torso and head. Cervical joint anesthesia produces a subjective feeling of unsteadiness while standing and walking (deJong et al., 1977). Older individuals with cervical spine disorders seem to experience similar sensations (Downton, 1996).

With age, small temperature changes become undetectable. For example, Navari and Sheehy (1986) found that many elderly were unable to distinguish temperature differences of 5° C. Compounding their troubles is the inability of elderly individuals to regulate their body temperature through sympathetic redirection of blood flow (Cech and Martin, 1995). In combination, these two factors may lead to the elderly experiencing mild hypothermia in cooler rooms.

Superficial pain perception probably diminishes with age but is highly dependent on the individual (Cech and Martin, 1995). On the other hand, deep pain perception shows a definite decline with age (Katzman and Terry, 1983).

Visual System.

Most research indicates that balance control is regulated primarily by peripheral vision (Leibowitz and Shupert, 1985; Stoffregen, 1985; Amblard and Carblanc, 1980). However, Paulus et al. (1984) found that foveal or focal vision plays a role in balance control too. They found that artificially reduced visual acuity increased their subjects’ postural sway. Simoneau et al. (1992) similarly found that experimentally reduced visual acuity resulted in greater postural sway while changes in lighting and spatial frequency of visual surroundings minimally degraded postural sway. The same authors found in a previous study that degraded visual acuity had a significant effect on cadence, foot placement and foot clearance of 35 elderly women walking down stairs (Simoneau et al., 1991). Information from the eyes seems to be particularly helpful in abnormal or novel stance situations (Lee and Lishman, 1975) and when proprioceptive input (joint, skin and muscle receptor input) is removed (Fernie and Holliday, 1978). Sway generally increases with deterioration of visual input (Gerson, Jarjoura and McCord, 1989; Lord, Clark and Webster, 1991) as do fractures (Felston et al., 1989). Nevertheless, Brocklehurst, Robertson and James-Groom (1982) found no correlation between objective visual impairment and sway. All visual attributes do not contribute equally to balance control. To illustrate, decreased contrast sensitivity rather than poor visual acuity was found to be associated with falling, implying that the ability to distinguish fine detail is not as important as the ability to discriminate edges or perceive objects (Lord, Clark and Webster, 1991). With respect to balance control, contrast variances that are oriented vertically are more beneficial (Simoneau et al., 1992). Moreover, the elderly in general (Brownlee et al., 1989) and fallers in particular (Tobis, Nayak and
the amount of light reaching the retina, and an inability of the lens to focus properly accounts for much of this change. Adding to the problem is the trend of increased dependence on the visual input for balance with increased age (Fernie, Gryfe, Holliday and Llewellyn, 1982). Unfortunately, assessments made by elderly individuals of their visual capability do not always correspond well with objective measures of visual function (Milne, 1979) and therefore they may not always be aware of a problem. For example, Gabel, Simons and Nayak (1985) could not document any more or less visual problems in fallers compared to non-fallers, however fallers were more likely not to be wearing prescription glasses.

Visual acuity (the ability to detect subtle differences in shapes) declines as a function of age. There is a gradual decline in visual acuity, or the resolving ability of the eye, prior to the sixth decade of life, followed by a more rapid decline between the ages of 60 to 80 years (Pitts, 1982; Tideiksaar, 1997). Acuity may decline as much as 80% by the ninth decade (Eisner, Fleming, Caruso and de Monasterio, 1988). Consequently, healthy older persons require about three times as much contrast as younger persons for the detection of objects in the environment. Contrast sensitivity also seems to be closely related to the ability to detect and discriminate objects in a naturally cluttered environment (Owen, 1985). An increased thickening and loss of elasticity and a reduction in retinal illumination have all been suggested a possible factors responsible for a decrease in visual acuity and contrast sensitivity (Elworth, Larry and Malmstrom, 1986). Moreover, the loss of acuity and contrast sensitivity is more evident under conditions of low illumination (Lampert and Lapolice, 1995). An age related decline in color sensitivity (the ability to discriminate between certain colors) has been found by some (Eisner et al., 1987) to be the result of a thickening and yellowing of the lens. Cool colors especially blues, greens and violets are particularly difficult to distinguish. Warm colors such as reds, oranges, yellows are much easier to differentiate and are sometimes used to enhance the contrast of objects such as step edges from their backgrounds. Errors in color discrimination are more noticeable where the amount of illumination is decreased (Tideiksaar, 1997). A decline in depth perception (the ability to judge distances and relationships among objects in the visual field) can develop as a consequence of age as well (Christenson, 1990). Declines in light sensitivity (the ability of the eyes to adjust to varying levels of dim and bright light) are also noticeable with advancing age (Eisner et al., 1987). Dark adaptation (the ability to adjust to low levels of illumination) is especially affected (Christenson, 1990). Williams (1990) found that while a person in their teens needs only six to seven minutes to adapt to darkness, an person in their 80s may need more than 40 minutes. A decline in the diameter of the pupil (i.e., senile miosis), which limits the amount of light reaching the retina, and an inability of the lens to focus properly accounts for much of this change. The ability of older persons to distinguish objects under conditions of excessive brightness is impaired as well. Older person display a greater sensitivity to glare and decline in glare recovery (Christenson, 1990). A thickening or opacity of the lens (which diffuses incoming light) and degenerative
changes in the cornea have been suggested as primary causes of glare sensitivity. This glare can cause particular problems if it is off the floor (light reflecting off waxed floors) which may cause an inappropriate walking pattern or inability to see objects on the ground. A shiny surface may result in the person incorrectly perceiving the floor surface as slippery and consequently they will have a cautious gait. An age related decline in the extent of the visual field is also seen (Christenson, 1990; Johnson and Keltner, 1986). Johnson and Keltner (1986) reported a visual field loss of 3 % to 3.5 % for persons between 16 and 60. The prevalence doubled for people aged 60 to 65, and nearly redoubled for individuals over 65. The peripheral and upper visual fields are the most affected. Much of the restriction of visual fields is due to optical factors (senile miosis or decreased pupil size) and mechanical problems such as realization of the upper eyelid and loss of retrobulbar fat, which results in the eyes sinking deeply into the orbits. Visuospatial function (the ability to match and integrate the positions of stimuli and objects in space) has been shown to deteriorate with age. Wahlin, Backman, Wahlin and Lindblad (1993) studied visuospatial ability and spatial orientation in a sample of 219 healthy older persons ranging in age from 75 to 96. They found an age related decline in both visuospatial abilities and spatial orientation. Together, these changes can lead to problems with the visual perception of objects in the environment and may lead to falling.

The eyes receive information on the location of objects in the environment, characteristics of the surface (e.g., rough, smooth, slippery, compliant, firm) upon which one is walking or standing, and the arrangement of body segments. This information can be utilized to plan ahead to control movement and prepare for any difficulties that one might encounter or try to avoid. Subsequently, when visual input is available, older people are able to adapt to a loss of stability as well as younger individuals; however when visual cues are absent, particularly on compliant ground surfaces, they have more difficulty. Some investigators feel that older individuals rely more on visual information (Bohannon et al., 1984; Potvin et al., 1980). When Pyykkö, Jantti and Aalto (1990) studied healthy men and women 85 and older, they found that visual deprivation had a significant effect on postural stability. Manchester, Woollacott, Zederbauer, Hylton and Marin (1989) as well as Judge, King, Whipple, Clive and Wolfson (1985) found that stability in older persons was significantly decreased under conditions in which vision and proprioceptive input were limited. Several studies have demonstrated an association between increased body sway and poor visual acuity (Dornan, 1978) particularly with near vision (Liechtenstein, Shields, Shiari and Berger, 1988). Lord et al. (1991) measured visual acuity and contrast sensitivity in a group of persons aged 59 to 97, and found that visual acuity and contrast sensitivity were not associated with body sway when persons were standing on a firm base; however when the subjects were standing on an absorbive foam surface body sway was associated with poor visual acuity and contrast sensitivity. Diminished sensitivity to lower frequency spatial information, which is mediated by the peripheral field of vision, may influence balance as well (Leibowitz, Rodemer and Dichgans, 1979). A narrowing of visual fields deprives the older person of that part of the field most sensitive to movement (Stelmach and Worthingham, 1985) Paulus et al. (1984) found that decreased peripheral vision was associated with greater sway in the anteroposterior direction. However, with deprivation of visual information and altered proprioception, both anteroposterior and lateral sway increased markedly (Ring, Nayak and Isaacs, 1989). Interestingly, lateral sway during stance with eye closure alone did not increase with age.

2.8.4.1.6. Vestibular System.

The vestibular system indicates the position of the head with respect to gravity and movement with respect to an inertial reference frame. The otoliths (utricle and saccule) respond to linear acceleration of the head while the semicircular canals respond to angular acceleration. Note that since the planes of the canals are not exactly orthogonal to one another or to the anatomical planes of the head the nervous system must make some adjustments in its interpretation of their signals (Zatsiorsky, 1998). Motion of the head that is sensed by the vestibular system elicits postural reflexes (a righting reflex) as well as a reflexive movement of the eyes in a direction opposite to the movement of the head. The visual reflex aids in ocular fixation during body motion. Degradation of the vestibular system includes loss of vestibular hair cells (20 % to 40 % by age 70) and decreases in the number of vestibular nerve fibers (Ochs et al., 1985; Bergstrom, 1973; Rosenhall and Rubin, 1975; Rosenhall, 1973). Bergstrom (1973) discovered that by 75 years of age the number of myelinated vestibular nerve fibers is about 40 % of that of the young. Common sense would lead one to believe that these structural changes would lead to a decrease in vestibular function. Some
believe that the vestibular system is over built or that there is considerable plasticity in the system. Lord et al. (1991) failed to find an association between vestibular testing and body sway even though the large number of older persons studied had evidence of vestibular impairment. This agrees with Nashner (1971) who found that the otoliths are not initially involved in detection of body sway. Thus, these age related degradations in the structure of the vestibular system may not have any functional effect. On the other hand, Woollacott, Inglis and Manchester (1988) showed that the inability of older people to stand while using primarily vestibular inputs (decreased visual and proprioceptive feedback) may in part indicate impaired vestibular function. Chandler Duncan and Studenski (1990) studied a group of people aged 18 to 85 with unilateral peripheral vestibular loss and found greater problems of balance when visual and proprioceptive inputs were altered. Black et al. (1983) tested the postural control of persons with vestibular deficits, and found that when proprioceptive and visual inputs were eliminated only those individuals with no vestibular function lost their balance. The vestibular system is an aid to balance control in situations when visual and somatosensory systems are compromised (Ghez, 1991; Goldberg, Eggers and Gouras, 1991; Nashner, 1982). One such situation occurs when a moving visual surround and support surface creates an impression of movement in the “eyes” of the visual and somatosensory systems. There is a lower limit or threshold to the vestibular system (Bronstein, 1986) below which only the information from the somatosensory or visual systems is useful in maintaining balance.

No correlation has been shown between vestibular function and sway as measured with a Wright ataximeter (Downton, 1996). Greater sway has been found to accompany sound induced hearing loss (Juntunen et al., 1987) and there seems to be an association between self-reported hearing problems and balance difficulties in the elderly (Gerson, Jarjoura and McCord, 1989). Drachman and Hart (1972) showed that vestibular abnormalities can lead to dizziness, but the relationship to falls is not clear because dizziness may have many causes (Downton, 1996). The differences in experimental results may be because much of the experimental evidence of the functional capacity of the vestibular system must be inferred, because of the difficulty in isolating the vestibular system (Stelmach, 1985; Woollacott, 1982). For example, Brocklehurst et al. (1982) employed a seat tilting device to test vestibular responses to rotations, however they found that proprioceptive input from the buttock area may have influenced their results. Rotating table experiments were also utilized by Brocklehurst, Robertson and Games-Groom, (1982). They also found an age-related decline in vestibular function.

2.8.4.1.7. Bone.

Bone continuously changes its structure and composition throughout life. These changes may be expressed as osteopenia, osteomalacia, or osteoporosis. Osteopenia is decreased calcification or density of bone, or reduced bone mass due to inadequate osteoid synthesis (McDonough, 1994, p. 732). Osteomalacia relates to a disease characterized by gradual and painful softening and bending of the bones, due to osteoid tissue which has failed to calcify (McDonough, 1994, p. 731). Osteoporosis is a reduction in the quantity of bone or atrophy of skeletal tissue; resulting in bone trabeculae that are scanty, thin and without osteoclastic resorption (McDonough, 1994, p. 732). Part of the loss in bone is due to the body’s attempt to maintain its serum calcium level. In both men and women, intestinal absorption of calcium declines with age, thus to maintain homeostasis calcium must be supplied from bone stores. This affects both cortical and cancellous bone. In post-menopausal women cancellous bone becomes more sensitive to parathyroid hormone, increasing it’s rate of re-absorption (MacKinnon, 1988). A general clinical rule of thumb in classifying osteoporosis is bone mass two standard deviations below that of healthy young individuals (Borner et al., 1988). With age, bone generally loses both mass and strength. After 35 to 40 years, bone re-absorption may begin to exceed bone formation (Thibodeau, 1987; LeVeau and Bernhardt, 1984; Whitbourne, 1985; Martin and Brown, 1998). Cancellous bone loss begins in the third decade while cortical bone loss begins in the forth decade of life (Borner et al., 1988). Race and gender influence bone change. In general, slight, sedentary, Caucasian females are most susceptible to osteoporosis. African-American adults seem to be able to retain bone better then Caucasian adults (Cech and Martin, 1995), and women begin to lose bone mass earlier than men do and have more difficulty in retaining it (Duncan and Parfitt, 1984). Bone loss in men is approximately 0.5 % per year. While, Raab and Smith (1985) found that women lose 1 % of bone mass per year prior to menopause. The rate of bone loss increases to 2 to 4 %
for the first 4 to 5 years after menopause. After this, bone loss in women returns to 1% per year.

McKinnon (1988) feels that one needs at least 40% of normal bone strength to withstand mechanical loading due to activities of daily living. If the 40% level is breached a “spontaneous” fracture may result. The spine, proximal femur, and wrist seem to be the most susceptible to “spontaneous” fractures (Cech and Martin, 1995).

Age related bone loss can be the result of many factors including: age, decreased activity, poor nutrition, metabolic changes or disorders and hormonal changes. Metabolic and hormonal changes were briefly described above. A lifelong practice of good nutrition and exercise will help generate bone mass early in life and maintain it in the later years. Many studies have shown that weight bearing exercise can prevent bone loss and may even increase bone mass (Krall and Dawson-Huges, 1994; Nelson et al., 1994; Breslau, 1992; Martin and Brown, 1989; Smith et al., 1989; Borner et al., 1988; Dalsky et al., 1988; MacKinnon, 1988; Ayalon, Simkin, Leichter and Raitman, 1987; Chow Harrison and Notarius, 1987). Conversely, bone loss will occur in the absence of weight bearing exercise (Nelson et al., 1994). Extending this idea a bit further, many believe that the age related bone loss is due primarily to the decrease in activity level with age (Martin and Brown, 1989; Pickles 1989; MacKinnon, 1988; Borner et al., 1988; Smith, 1995). Fortunately, bone seems to react positively to exercise regardless of the age of the individual (Talmage, Stinett, Landwehr, Vincent and McCartney, 1986).

Along with changes in bone thinning, cracking in cartilage and vertebral disks occur with age. Osteoarthritis, the degeneration of articular cartilage, affects approximately 70% of all people at some point in time (Gradisar and Porterfield, 1989). Osteoarthritic changes in cartilage include increased joint friction and a degradation of the cartilage’s ability to absorb and redistribute joint forces. Thus, the underlying bone is subject to greater forces. Consequently, bony spurs may develop. The net result is limited painful movement. For example, James and Parker (1989) found an age related reduction in hip and knee joint ROM in individuals over 70. Vandervoort et al. (1986) found a similar age related reduction in the ROM at the ankles. In this study, the females lost more range of motion than did the males. This decreased ROM leads to functional impairments such as difficulty in transfer tasks such as getting up from or onto a chair, toilet or bed, as well as hindering the ability to climb stairs (Cech and Martin, 1995; Bergstrom et al., 1985). The pain and limitation in ROM may also result in non-standard gait patterns and decreased balance. Moreover, the intervertebral disks lose water and the annulus fibrosus undergoes fibrotic changes. The rate of water loss is greatest in adults between 20 and 40 years old, after which it slows to a more normal level (Cech and Martin, 1995). The net result of these changes is that the intervertebral disk becomes thinner and less flexible. This loss in back flexibility has been found by Tinetti et al. (1986) to increase a person’s risk of falling. Decreased ROM has been shown to occur with aging (Walker et al., 1994) and can be improved with exercise (Munns, 1981). However, it is not clear whether this improved ROM will aid in fall prevention or not. In one study Murray, Kory and Clarkson (1969) showed that the elderly tend to actively use only a portion of the ROM that can be achieved passively in clinical tests. Nevertheless, a decreased ROM may limit a person’s ability to perform a task or alter the manner in which they perform that task. Thus, decreased ROM at one joint may cause another joint to move more that it would otherwise. This chain reaction effect may place undo stress on the body or place the person in a compromising situation (e.g., one that requires more strength or skill to maintain balance). Still, exercises to improve ROM are not enough. Increased ROM gained through stretching and exercise are only beneficial to an individual when the increased ROM is utilized in daily activity. Increased flexibility may have a beneficial effect by reducing injury from a fall whether or not it helps reduce the risk of falls themselves. As was true for strength, there probably exists a critical threshold of flexibility, however I have found no research to support this hypothesis.

In combination, bone and joint changes may lead to alterations in posture. As Aiken (1998) points out one of the more noticeable changes that occurs with aging is the loss of height and a stooped posture. A stereotypical picture of an older individual is one who is stooped over with thoracic kyphosis, flattened lumbar spine, flexed hips and knees, anteriorly inclined or rotated trunk and head. The stooped posture may be the result of osteoporosis, vertebral compression fractures or degradation of vertebral disks. A decreased ability to balance and a fear of falling which results in the individual trying to increase their stability by lowering their CM may also contribute to stooped posture. Unfortunately in lowering the CM, it may be moved closer to the edge of the base of support, thus decreasing stability in that direction. The
stopped posture may make walking more difficult and require a greater muscular contraction in order to support the crouch. The result is that the crouched stance and mobility is less efficient and more fatiguing than a more upright one. It is important to realize that the stooped posture may not be due to any physical mechanisms, but may be due to a depressed health and or mental state.

2.8.4.1.8. Muscle.

The ability to sense and perceive one’s status quo is necessary but not sufficient to maintain balance. To maintain balance, one must be capable of correcting an imbalance. Adequate strength and joint ROM are needed for effective postural control (e.g., Whipple et al., 1993; Studenski, Duncan and Chandler, 1991). With age, individuals allow themselves to be further out of balance before they respond, thus the muscular force required to correct the problem increases (Stelmach and Worringham, 1985). Unfortunately the strength of skeletal muscles required for postural control and locomotion decreases with age and inactivity (e.g., Tang and Woollacott, 1996; Fiatarone and Evans, 1993; Bassey, Bendall and Pearson, 1988; Vandervoort and McComas, 1986; Aniansson, Zetterberg, Hedberg and Henriksson, 1984; Grimby and Saltin, 1983; Larsson, Grimby and Darlsson, 1979). Peak muscular strength is typically reached at the age of 20 for men and a few years earlier for women. Muscle mass declines by an average of 0.4 % per year beginning at age 25 until age 50. As stated previously, the loss may begin earlier for women (Grimbly et al., 1982). After the age of 50, the rate of muscle loss increases to 1 % per year (Cech and Martin, 1995). Larsson et al. (1979, 1982) and Asmussen (1981) feel that strength remains relatively constant and does not decline until after age 50. By age 70, individuals only have about 25 to 30 % of their muscle mass and 50 % of their strength (Cech and Martin, 1995). Astrand and Rodahl (1986) found that the loss in strength is a little more severe. They state that the strength a 65 year old person to be 75 to 80 % of that of a 20 to 30 year old, with a larger decline to about 60 % in leg and back muscles and to 70 % in arm muscles from 30 to 80 years of age.

Loss of muscular strength may be due to a decrease in the number of fibers, a decrease in the cross section of fibers or both. It has not been resolved which occurs with aging (Schultz and Lipton, 1982), however Young (1984) believes that most people experience a decrease in muscular strength due to a decrease in both the number and the size of muscle cells. There are several hypotheses on the structural changes that take place with age (Cech and Martin, 1995). One theory is that for type IIb fibers the diameter decreases while the number of fibers remains the same; for type IIa fibers the number decreases. A second theory is that both IIa and IIb fibers decrease in number and diameter. Thus, the ratio IIa:IIb remains constant while the ratio I:II increases. In both theories, the percentage of the ratio of I:II fibers increases (Cech and Martin, 1995). Additionally, there are biochemical changes with age. For instance, with age there is a general decrease in all muscle enzymes. Parallel to the decrease in relative amount of type II fibers, there seems to be a greater reduction in anaerobic enzymes. The functional result of all these changes is that there is a reduction in strength due to morphologic changes and a reduction in endurance due to biochemical changes (Cress et al., 1984). Because the different basic fiber types are subjected to different degrees of functional demands, it should not be surprising that different muscles show different rates and magnitudes of decline (i.e. changes in type I:type II ratio).

The muscle losses may be the result of changes in the vascularisation or innervation of muscles. One theory proposes that the muscular changes that occur with aging are related to the changes in the peripheral nervous system. Schultz and Lipton (1982) argue that there may be a “functional denervation” of muscle in which the nerve to a muscle deteriorates as a result of inactivity. This hypothesis stems from the observation that some of the age related changes in muscle resemble those found in denervated muscle (Cech and Martin, 1995). In addition, the neuromuscular junction (NMJ) changes with age. The synaptic vesicles increases and bunch together, the synaptic cleft enlarges and fills with a thickened basal lamina, and the plasma membrane of the NMJ thickens and loses its folding. The second theory is that the deterioration of muscle is related to decreased blood flow to the muscles with age (Cech and Martin, 1995). This decreased blood flow to muscles may also be the result of inactivity. This theory is supported by the observation of decreased number of capillaries in muscles of older individuals. In addition to loss of blood flow, the flow of nutrients into and metabolic wastes out of muscle may be hindered with age due to an increase in the basement membrane thickness around both the capillary and the myofiber (Cech and Martin,
Also there is an increase in the amount of connective tissue and increased cross-linking between collagen that may explain the increase in “stiffness” of muscles in older individuals. Both strength and speed of contraction have been found to decline with age (Cech and Martin, 1995). The reasons they give for these changes are (tonic contraction):

- latency period
- decreased nerve conduction velocity,
- clumped vesicles in the NMJ slows and decreases the amount of Acetylcholine (ACh) released into the NMJ,
- the wider gap increases the transit time of the ACh to the postsynaptic membrane,
- increased basement membrane material decreases ACh diffusion,
- decreased folding spreads receptors for ACh binding
- possible decrease in conduction velocity of sarcolemma
- contraction period
- disorganization of myofilament lattice
- Calcium activated ATPase activity is decreased
- decreased levels of stored creatine phosphate
- reduced uptake of calcium by the sarcoplasmic reticulum
- relaxation period
- amount of AChE at the NMJ is decreased (increased time to hydrolyze ACh)

In addition to age related changes in muscle, toxins, systematic metabolic disturbances and endocrine disorders (Elble, 1997), disease and disuse may drive muscular alterations. Disease effects on muscle will not be discussed here except to mention that diseases commonly associated with muscle weakness include hypo- and hyperthyroidism, polymyalgia rheumatica, polymyositis, osteomalacia and neuropathies. The effect of lack of muscle activity on muscle loss is dramatically illustrated in situations of bed rest or immobilization. In these extreme situations, muscle can lose approximately 1% to 3% of its strength per day of immobility (Halar and Bell, 1988). In addition, lower extremity muscles seem to lose strength about twice as fast as those of the upper extremities (Harper and Lyles, 1988). As people get older, their physical activity patterns tend to change and this change is seen to be one of the more important factors driving muscular degradation (Aniansson et al., 1981; Aniansson and Gustafsson, 1981; Aniansson et al., 1978).

As an indication that activity levels influence strength levels, Dook, James and Henderson (1997) found that aging female athletes have greater strength than their sedentary counterparts. Independent of athletics per se, several studies have shown that regular exercise in general will improve the endurance and strength of older individuals. Chapman et al. (1972) found that elderly subjects could reach the strength levels of younger subjects through exercise. DeVries, (1970) and Sidney and Shephard (1978) have shown strength and endurance improvements for older adults in response to resistance training. Charette et al., (1991) and Stamford (1988) also found that the very old (up to 90 years old) are capable of increasing muscle strength. Frontera et al. (1988) found that they could increase knee extensor and flexor strength 107% in 60 to 72 year olds, and induce muscle hypertrophy. Fiatarone et al. (1989) found after an eight-week weight-training program the average strength gain was 174% in a group of ten frail nursing home residents ranging in age from 87 to 96 years old. The increase in strength paralleled improvements in tandem gait speed and time taken to rise form a chair. Roberts (1989) found that participation in a six-week program of aerobic walking, among other things, improved lower extremity strength. Fisher, Pendergast and Calkins (1991) examined the effects of a six-week resistance-training program on a group of nursing home residents (age 60 to 90 years). They found that strength was improved 15%. While this was not as large an improvement as seen in other studies (possibly due to the short duration of study), they did find that the improvement in strength was still evident four months after training. In addition, many of the subjects in this study felt that they were more capable in moving about. Judge, Underwood and Gennosa (1993) found that in a group of older individuals (71 to 97 years old), exercise improved leg strength and increased walking speed up to 30%. Fiatarone et al. (1994) found that in a group of 100 frail nursing home residents (mean age 87, approximately 1/3 in their 90s), found that muscle strength increased by 113% in the residents who underwent resistance training. Moreover, the individuals increased their walking speed and their ability to climb stairs. Therefore, some of the individuals who required a walker for ambulation were
now able to walk with a cane. Pyka, Lindenberger, Charette and Marcus (1994) found that in a group of 68 year-olds, resistance training improved hip extensor strength by 30% and hip flexor strength by 97%. Many other studies exist showing that elderly individuals can maintain and increase their strength and muscular capabilities through exercise. How these training effects are related to balance and fall risk are explored in the rehabilitation section of this chapter.

Not all studies have found such positive results from exercise in elderly (e.g., Hanson, Agostinucci, Dasler and Creel, 1992; Morey et al., 1991). Possible reasons for the negative findings include, at too low an intensity, insufficient exercise or study duration to see effect, poor compliance with study protocol and a ceiling effect (little or no room for improvement). The last factor may in part explain the general finding that larger gains in strength are seen in individuals that are initially very frail. Thus, there exists a diminishing returns factor. The aptitude an elderly individual has for improving strength appears to be inversely related to the starting strength of the individual (Henderson, White and Eisman, 1998).

In addition to the general finding that training elderly can improve their muscle function, many studies indicate that stronger individuals function better. Imms and Edholm (1981) found a correlation between poor quadriceps strength and slow gait velocity and small steps. Wolfson et al. (1985) found fallers to have, in addition to differences in walking and balance ability, a lack of dorsiflexion strength (10% of control values). Fallers in this study also had an increase in vibratory threshold, diminished sural nerve conduction velocity and abnormal H-reflexes. Whipple et al. (1987) found isokinetic strength of the knee and ankle were related to the incidence of falling in age and gender matched groups. The deficit was greatest during faster (120°/s) movement then it was during slower movements (60°/s). The greatest declines were found in those muscle groups associated with balance: the knee flexors and extensors and the ankle plantar and dorsiflexors. They concluded that ankle dorsiflexor strength compared to all muscles was decreased the most in fallers compared to non-fallers. Hyatt et al. in 1990, found that muscle strength correlated with clinical measures of functional status, and was associated with manual dexterity. Gehlsen and Whaley (1990) compared a group with a history of falls and one with no history of falls. They found the group with a history of falls were significantly poorer in static balance (eyes open and eyes closed), leg strength (combined hip extension, knee extension and ankle plantar flexion) and hip and ankle flexibility. Studenski, Duncan and Chandler (1991) found that ankle strength was related to history of falling. Wilder (1992) calculated that 64% of the variation seen in the gait characteristics (e.g., step length and time in double support) of elderly adults was accounted for by muscle strength. Cunningham et al. (1993) compared older individuals living independently with those living in nursing homes. They found that the individuals living independently had significantly greater flexibility, increased strength, greater daily activity and faster self selected walking speed. Judge, Underwood and Gennosa (1993) found that quadriceps strength was also correlated to balance. Several controlled studies have identified quadriceps muscle weakness as an important risk factor for falls (Lipsitz et al., 1991; Campbell et al., 1989; Nevitt et al., 1989; Robbins et al., 1989; Tinetti et al., 1986). Such lower extremity weakness and disability may in turn lead to gait abnormalities, postural instability and falls. However, as Frank, Winter and Craik (1996) note, research has so far indicated only that a relationship between strength and balance and fall risk exists. “The finding that frail elderly are less fit does not suggest that intervention will improve function. Intervention studies seem the only way to examine the underlying clinical assumptions which guide the treatment of the older person” (p. 297). There is a clinical assumption that increased balance, range of motion, torque generating capacity, and improved cardiovascular fitness are associated with improved functional performance. Frank, Winter and Craik (1996) point out that increased muscular strength may not directly improve balance, however it may allow a person to make more mistakes. Thus, a person can theoretically lean over further, or move out of balance more and still be able to catch themselves.

2.8.4.1.9. Physiology.

Not all improvements in fall risk may come from strength per say. It is apparent that individuals with lower work capacities (e.g., low VO2max) will have a lower tolerance for activity and will fatigue more readily (Frank, Winter and Craik, 1996; Brown, 1992). Studies have shown that elderly individuals can improve their VO2max and adapt to endurance training (e.g., Hagburg et al., 1989, Verg et al., 1985; Heath et al., 1981). While some have found adaptation to endurance training to be minimal for older
individuals (Wilmore et al., 1970). Thomas et al. (1985) found that the best predictors of an elderly subject’s VO\textsubscript{2max} after 1 year of training is the initial VO\textsubscript{2max}. Thus, some of the mixed results seen in the literature may be due to differences in starting points of the subjects tested. Heath et al. (1981) calculated that trained master athletes (males) have a decline in VO\textsubscript{2max} of 5 % per year, whereas untrained older adults have a decline of 9 % per year. Therefore, even if one cannot improve endurance capacity, one should be able to at least slow its decline. Since it is likely that elderly become more susceptible to falling with increased physical fatigue, improving VO\textsubscript{2max} should have a positive influence on fall risk. Whether this is the case or not has yet to be proven.

### 2.8.4.1.10. Gait

Many researchers have found impairments in mobility to be linked to a greater risk of falling. Harris and Kovar (1991) found that no individuals aged 75 to 84 with a walking limitation were 10 times more likely to fall than comparably aged individuals without walking limitations. In general, stride length is observed to be shorter in healthy active elderly (Winter et al., 1990), sedentary elderly (Ferrandez, Pailhous and Durup, 1990) and fallers and non-fallers (Guimaraes and Isaacs, 1980) than in normal adults. This stride length reduction has been observed to correlate with diminished gait velocity in the elderly (Elble, Hughes and Higgins, 1992). Other researchers who have found reduced walking speed in fallers include: Campbell et al. (1989), McClaran, Forette, Hervy and Bouchacourt (1991) and Nevitt et al., (1989). The combined short stride and slow gait velocity have been reported more often in fallers when compared to non-fallers (Wolfson et al., 1990) and for hospitalized elderly (Guimaraes and Isaacs, 1980). Elderly living independently or in the community show less of a decrease in velocity and stride length than those living in institutions (Guimaraes and Isaacs, 1980). In addition to decreased stride length and velocity, Guimaraes and Isaacs (1980) found that elderly individuals had a narrower base of support and increased variability in their gait. Hageman and Blanke (1986) did not find any stride width difference in two groups of women, one 20 to 35 years old and the other 60 to 84 years old. They did find significantly smaller values of step length, stride length, ankle range of motion, pelvic obliquity and velocity in the older women.

Elble et al., (1991) found that although gait was slower it was not “abnormal”. If walking speed was matched in normal and elderly subjects, stride length, cycle duration, duration of double support, step weight, joint angular displacements and toe and heal clearance were similar in the two groups. Elble et al. (1992) found this to be also true for individuals with symmetrical neurological gait disturbances (sometimes called senile gait). That is, when gait was compared at a self-selected pace differences were observed but when normals were instructed to walk at a speed that matched the individuals with the gait disorder, the differences disappeared. Differences were almost entirely attributable to stride length. The research of Larish et al. (1988) supports this. They found that younger subjects took longer strides at the faster speeds, while the older adults compensated by a greater increase in their stride frequency. They found that the difference in stride length appeared only at faster walking speeds (i.e., when the functional capacity of the older adult became stressed). This difference in stride length may be caused by a weaker push off and decreased energy generation by the ankle plantar flexors observed in the elderly by Winter et al. (1990).

Winter et al. (1990) found a higher covariance in the hip and knee moment patterns in elderly walkers. They attributed this high covariance to difficulties in balance during walking. Yack and Berger (1993) found that the accelerations of the trunk during walking were higher in elderly with stability problems as compared to younger and older normals.

### 2.8.4.1.11. Medications

Many researchers are concerned about the effects that drugs have on the risk of falls (see reviews by Thapa and Ray, 1996; and Tideiksaar, 1997). The relationship between drug use and falls is hard to evaluate in part because it is difficult to separate the effects of the disease for which the drug is prescribed from the effect of the drug (Downton, 1996; Campbell, 1991; Tinetti et al., 1988). For example, Tinetti et al. (1988) found an association between sedative use and falling, but if the presence of cognitive impairment was taken into consideration the risk of falling associated with sedative use was markedly decreased. In addition, a large fraction of elderly take both prescribed and over the counter drugs and there
is an age-related change in pharmaco-kinetics or -dynamics (Downton, 1996, Thapa and Ray, 1996; Kelly and O’Malley, 1992). Nevertheless, Prudham and Evans, (1981) found that taking any drug may increase the likelihood of falling. This was supported by Granek et al. (1987) and Tinetti et al. (1988). This is a problem for the elderly because although individuals over the age of 65 constituted only 12 % of the U. S. population in 1988, they received 29 % of all prescriptions (Ray, Gurwitz, Decker and Kennedy, 1992). More then 80 % of the individuals in this age range took one or more prescriptions (Moeller and Mathiowetz, 1989). Another study found that in the United States up to 93 % of individuals older than 65 take at least one prescription or over the counter medication (Hale, May, Marks and Stewart, 1997). The extent of drug use among older persons in hospitals and nursing homes is even greater. Nolan and O’Malley (1988) found an average of seven medications per individual in these groups. Psychotropic, hypnotic, diuretic, anti-hypertensive and anti-Parkinsonian medications, may contribute to falls by decreasing alertness, depressing psychomotor function, or causing weakness, fatigue, dizziness, or postural hypotension (Ray and Griffin, 1991; Nevitt, Cummings, Kidd and Black, 1989; Studinski et al., 1994). Several studies have found that fall risk is high in individuals taking drugs with extended half lives, and increases with the number of medications a person takes (Blake et al., 1988; Buchner and Larson, 1987; Campbell et al., 1989; Tinetti et al., 1986). Thus, an individual that needs to take several medications further increases their fall risk (Blake et al., 1988; Nevitt, 1991; Campbell, Borrie and Spears 1989; Granek et al., 1987; Tinetti et al., 1988; Tinetti, Williams and Mayewski, 1986). Non-compliance poses additional problems as both hyper-compliance (taking more medication than is needed) and hypo-compliance or non-compliance (taking less medication, not taking medication, or not finishing a prescription) can also lead to adverse medication effects and to falls. One must also note that not all studies have found a relationship between falls and drug use (e.g., Bates, Pruess, Sooney, Platt, 1995; Janken, Reynold and Swiech, 1986; Morse et al., 1985; Perry 1982; Sehested and Severin-Nielsen, 1977). The relationship between fall risk and alcohol use is mixed. McConnell and Matteson (1988) found an association while Campbell et al. (1989), Nevitt et al. (1989), O’Loughlin et al. (1993), Teno et al. (1990) and Tinetti et al. (1988) did not. Because alcohol abuse in the elderly has been estimated to be somewhere between 15 and 50 % (Bristow and Clare, 1992; Curtis Geller, Stokes, Levine and Moore, 1989) and has been shown to increase the risk of falls (Honkanen et al., 1983) the null results are surprising. Several theories have been postulated. One is that individuals who fall may refrain from drinking (Campbell, 1991), while only those of good health imbibe (O’Loughlin et al., 1993), or that those who use alcohol and fall die before reaching an old age (Tinetti et al., 1988). These ideas are in part supported by Waller (1978) who found that less than 10 % of elderly fallers have histories of heavy drinking. Lastly, as not all people report falls, all people may not report alcohol use (Teno et al., 1990).

2.8.4.1.12. Disease.

Falling may be the first indication that a person has a disease. Diseases that have been associated with increased fall risk include diabetes mellitus, Parkinson’s disease, previous stroke, osteoarthritis, dementia and depression (King, 1997), in addition to other diseases that affect the visual, neuromuscular and cardiovascular systems (Tideiksaar, 1997). Many studies have found that individuals who have fallen generally have some type of illness or disease (e.g., Campbell et al., 1990; Cwikel, 1992; Lipsitz, Jonsson, Kelley and Koestner, 1991; Nevitt et al., 1989; Robbins et al., 1989; Rubenstein, Robbins, Josephson, Schulman and Osterwell, 1990; Svensson et al., 1991; Tinetti, Williams and Mayewski, 1986; Tinetti et al., 1988). Rubenstein et al. (1988) estimated that approximately 55 % of falls are related to some type of medical condition including effects of medication. Of this 55 %, 83 to 88 % of the falls are related to chronic diseases. While Wolinsky, Johnson and Fitzgerald (1992) found a positive association between repetitive falling, rapid deterioration of health and increased mortality. This association between mortality and falling may not be a direct one as many of the deaths may occur several weeks or months after the fall (Morfitt, 1983, Waller, 1978; Wild, Nayak and Isaacs, 1981a) and the death may be the result of fall related complications (e.g., immobility, and co-existing conditions (e.g., pneumonia, heart failure, ulcers)(Tideiksaar, 1997). The most commonly reported medical conditions that affect lower extremity function and increase fall risk include:

- arthritis (Blake et al., 1988; Buchner and Larson, 1987; Granek et al., 1987; Meyers et al., 1991; Robbins et al., 1989; Tinetti et al., 1988)
• foot impairment (Blake et al., 1988; Nevitt et al., 1989; Tinetti et al., 1988)
• stroke or hemiplegia (Campbell et al., 1989; Mayo et al., 1989; Nevitt et al., 1989; Yasumura et al., 1994)
• peripheral neuropathy (Richardson, Ching and Hurvitz, 1992)
• Parkinson’s disease (Campbell et al., 1989; Granek et al., 1987; Koller, Glatt, Veter-Overfield and Hassanein, 1989; Nevitt et al., 1989).

While not directly related to lower extremity dysfunction, urinary dysfunction or incontinence has been found to be associated with an increased risk of falling (Barker and Mitteness, 1988; Campbell et al., 1989; Janken and Reynolds, 1987; Janken et al., 1986; Mayo et al., 1989; Nevitt et al., 1989; Rapport et al., 1993; Robins et al., 1989; Stewart, Moore, May, Marks and Hale, 1992; Tinetti et al., 1988). This may be due to the fact that toilet transfer tasks are an activity in which falls often occur, or that rushing to the bathroom may put the individual at risk, one of the proposed reasons for increased risk of falling with laxative use.

2.8.4.1.13. Emotional State.

Not all balance problems result from peripheral deficiencies. Postural responses may become ineffective due to deterioration of central neurological integration of the sensory and motor systems (Stelmach and Worringham, 1985; Woollacott, 1993; Alexander, 1994). A few researchers have investigated the association between falls and slowed reaction time, reflexes and other nonspecific neurological features, however no conclusions could be reached (Nevitt, 1997). Several researchers have found an association between impaired cognitive function, dementia, sleep disturbances, depression, fall related anxiety and falls (Tideiksaar, 1997). These problems may directly affect postural control through impaired judgment, spatial disorientation, behavioral changes and decreased attention to environmental hazards (Nevitt, 1997).

2.8.4.1.14. Many factors contribute to falls.

Many researchers have discovered that fall risk increases with increases in the number of intrinsic risk factors (Granek et al., 1987; Robbins et al., 1989; Tinetti et al., 1986; 1988). For instance, Tinetti et al. (1986) found that in addition to increasing fall risk with an increasing number of risk factors, no subject with three or fewer factors fell more than once, while every subject with seven or more factors fell two or more times. Robbins et al. (1989) used three factors to predict one-year fall risk (hip weakness, unstable balance and taking four or more medications). They found that 12% of those who had one of the factors fell while 100% of those with three factors present fell. Tideiksaar (1997) warned that there may be many elders with one or more risk factors that do not fall while others with similar risk factors fall frequently. Thus, these risk factors may be necessary but not sufficient to cause a fall in all older people.

There are numerous other problems that may predispose an elderly individual to falls, including gender, history of falls, specific diseases, cognitive impairments, fear of falling, motivation etc. (see table 2.8.4.2.1 for more suggestions, note this is probably not an all inclusive list). Nevitt (1997) found that a number of medical conditions (arthritis (especially in the joints of the lower extremities), dementia, stroke, Parkinson’s disease and cataracts) were positively associated with the risk of falls in one or more studies. Individuals who have had a stroke, even if overt deficits have been resolved, have a greater risk of falling (Downton and Andrews, 1991). Parkinson’s disease is a particular problem because of gait disturbances and the difficulty in initiating and stopping movement. In addition, the medications taken by individuals with Parkinson’s disease (e.g., L-dopa, bromocriptine) are not conducive to balance. Morris et al. (1987) found that the incidence of falls increased by a factor of three in a group of individuals with dementia compared to controls. Diabetes may increase an individual’s risk of falling because of the development of peripheral neuropathy, hypoglycemia related to insulin or oral hypoglycemic drugs, or retinal degradation. Falls resulting from these conditions maybe due to the direct effect these diseases have on the neuromuscular system or indirectly through physical de-conditioning. A similar relationship is seen between strength, activity level and risk of falling. As discussed in detail in section 2.8.4.1.8, strength may be negatively correlated to fall risk because active individuals tend to be stronger than inactive ones.

One difficulty in assessing whether or not a person is susceptible to falls is that many factors can cause a fall. For example, an individual may not have any observable intrinsic factors such as increased body sway...
in stance, but may be susceptible to tripping. Thus, a measure like sway which gives an indication of an individual’s balance capability does not measure fall susceptibility as completely because other external, environmental or extrinsic factors may influence the likelihood of falling.

2.8.4.2. Extrinsic Risk Factors.

Unfortunately in many studies of the environmental contributions to falls the control group is not exposed to the same environment (Nevitt, 1977). With this in mind, it has been estimated that environmental factors are the sole or major cause of 1/3 to ½ of falls in community living individuals (Kellogg International Work Group on the Prevention of Falls by the Elderly, 1987; Tinetti, Speechley and Ginter, 1988; Rubenstein et al., 1988; Rubenstein and Josephson, 1992; Nevitt, Cummings and Hudes, 1991), and falls resulting in injury (Morfitt, 1983). This percentage most likely increases with age as the number and severity of intrinsic factors increases (Downton, 1996). In other words, there is crosstalk between factors regardless if they are intrinsic or extrinsic. For example, healthy compared to frail or impaired elderly individuals, fall less frequently, and in general, only in situations where there are environmental hazards. O’Laughlin et al. (1993) found that active individuals had a decreased intrinsic risk of falls. However, those who were very active increased their fall risk. Because these individuals have greater mobility many of these environmental hazards were “self generated” or were due to a wider range of physical environments that they encountered (e.g., climbing ladders, running, sports) (Kellogg International Work Group on the Prevention of Falls by the Elderly, 1987; Tinetti, Speechley and Ginter, 1988; Nevitt, Cummings, Kidd and Black, 1989; Northridge, Nevitt, Kelsey and Link, 1995; Speechley and Tinetti, 1991). For example, falls in the more active individuals were more likely to occur away from home (Hornbrook et al., 1991; Reinsch et al., 1992; Kellogg International Work Group on the Prevention of Falls by the Elderly, 1987). On the other hand, individuals who are sick will tend not to leave the home and thus do not have as much of an opportunity to fall outside the home. Most falls occurring in a home take place in the bedroom, bathroom, living room, kitchen and on stairways (Campbell et al., 1990; DeVito et al., 1988; Downton and Andrews, 1991; Schelp and Svanstrom, 1986) while most falls in the hospital or nursing home setting occur in the bedroom and bathroom (Fleming and Pendergast, 1993; Lake and Biro, 1989; Svensson et al., 1992). The reason for the high frequency of falls in these locations may not be because these places are hazardous, but because the individuals frequent these places often. Another explanation for the high rate of falls in these areas is that some individuals will fall in familiar or “safe” environments (e.g., bedroom, bathroom) while performing routine activities (e.g., walking, standing, reaching) (Nevitt et al., 1989; Sjorgen and Bjornstig, 1991; Rubenstein and Josephson, 1992; Northridge, Nevitt, Kelsey and Link, 1995; Speechley and Tinetti, 1991; Lipsitz, Jonsson, Kelley and Koestner, 1991; Fleming and Pendergast, 1993). This may account for the finding that only about 5% falls in elderly occur as a result of hazardous activity (Tinetti, Speechley and Ginter, 1988), but result from activities such as bending over and picking an object off of the floor, or reaching to place a object on a high shelf. A familiar and safe environment may help to reduce the risk of falls but will not eliminate it. One can only hope to recognize the environmental factors that increase the risk of falling and remove or at least minimize them. If they cannot be eliminated or removed, it is still important that they be recognized. Studenski et al. (1994) found that some individuals increase their fall risk through a lack of concern or consideration of their abilities and environmental conditions. That is, some elderly individuals continue to be active and disregard hazards despite having lost some balance and gait ability. These individuals may have a very high fall risk. Speechly and Tinetti (1991) state that interventions to prevent falls need to be tailored to the capabilities of each individual, which may improve or deteriorate with time. Interventions can be aimed at the environment through removal of hazards (e.g., removing or anchoring rugs, improving lighting), or they can be aimed at the individual through modification of their activities. Nevitt (1997) notes that caution must be used when modifying an individual’s lifestyle.

...A balance always must be struck between reduction in risk and maintenance of mobility, quality of life, and independence. Although severe curtailment of activities might decrease falls in the short term by reducing exposure, over the long-term reduced self-confidence and physical decondition serves only to increase the risk of falls and functional decline (Nevitt, 1997, p. 27). The most commonly listed environmental hazards are:

- inadequate lighting
• poor stairways (either in design or repair)
• obstacles (e.g., electrical cords, rocks and twigs in the yard or on a hiking trail)
• slippery floors (e.g., unsecured rugs, non-skid bathtub surface)
• soft compliant surfaces (e.g., soft sand or dirt, grassy fields)
• low temperature

Inadequate lighting magnifies the effects of poor visual function (Cullinan et al., 1979). Lighting may add to the physical difficulty that one encounters with poor stairways (Downton, 1996). The type of floor covering can influence mobility in ways that are unexpected or match common sense. For example, one study found that vinyl floors were more difficult to negotiate than carpet for elderly hospital in-patients (Willmott, 1986). Thus, while a carpeted floor may be more compliant and provide an obstacle to catch ones toe, a vinyl surface may be slippery or at least appear slippery (shiny, thus indirectly altering gait) or it may be “grippy” thus catching an individual’s toe or abruptly stopping the foot at initial contact, especially if the individual is wearing crape soles. In addition, the individual may be afraid to walk on a vinyl surface as compared to a carpeted surface due to fear of the relative results of a fall on the respective surfaces. Soft compliant surfaces such as the soft sand at beaches presents two difficulties. First, footing is variable except near the shoreline where the waves have smoothed the surface. Second, the non-compliant surface lowers the force with which an individual can use to counter any imbalances, and the soft surface may reduce the sensory feedback necessary to make balance decisions. Interestingly, environmental temperature has been shown to affect the risk of falling in women (Campbell et al., 1988). Bastow, Rawlings and Allison, (1983) found that this may occur only in thin or under-nourished women. In general, poor nutrition has been found to be more common in fallers than in non-fallers (Vellas et al., 1992), and in those who suffered a hip fracture in a fall than in those who did not (Delmi et al., 1990). With respect to low temperatures, one can expect a degradation of both sensory and motor function. In addition, the ground may be covered with ice.

As an aside Tideikssar (1997) pointed out that some caution must be used in reading some studies of environmental hazards, due to errors or biases in the reporting of the falls. That is a majority of falls is not observed, and thus the cause of a fall must be guessed at. Thus, the cause of the fall is reported based on the opinion of what hazards may be present rather than the actual mechanism of the fall. Therefore, falls may be either over or under reported depending on the faller’s recollection and willingness to admit to hazardous conditions being present. Similarly, a health care professional may tend not to admit that a hazardous condition(s) existed. On the other hand, the faller, their family members and caregivers may want to blame the environment for a fall rather than admit or recognize the fact that the faller may be having some intrinsic difficulties.

2.8.4.3. Risk Factors for Injurious Falls.

As stated previously, not all falls result in injuries, and the more a person falls does not mean that the likelihood that they are to be injured increases. Therefore, the risk factors for injury from falls are different from the risk factors for falls themselves. Unfortunately most studies in this area have limited their focus only to whether the fall results in a fracture (Nevitt, 1977), thus the risk factors for other types of injuries are not well understood.

Approximately 350 J to 770 J is converted from gravitational potential energy to kinetic energy in a fall from a standing position to the ground given an arbitrary range of standing height of 1.65m (5’ 5”) to 1.82m (6’) and corresponding range of mass of 45 kg (100 lb.) to 86 kg (190 lb.). If all this energy was then suddenly absorbed (as in an impact), by any bone in an elderly woman’s body, it would break, for according to Hayes, Piazza and Zysset (1991) this is an order of magnitude greater than the energy need to fracture almost any bone in an elderly woman. However recall from the section on frequency of injury in falls, that only between 5 % and 10 % of falls result in fractures. The reasons that all falls do not result in a fracture include (Nevitt, 1997):
• orientation of fall
• distance or height of fall
• protective responses
• energy absorption by soft tissue
• energy absorption by environment
• differences in bone density

First, the fall must be oriented “correctly” for the impact to fracture the bone. For instance, if one does not land on the femur during a fall there is little chance that the femur will be broken. The greater the distance one falls, the greater the kinetic energy just before impact, and the greater the chance of bone fracture (Nevitt and Cummings, 1993; Greenspan et al., 1994; Hayes et al., 1993). If the falling individual can slow the fall down by grabbing an object on the way down, or by hitting another object (e.g., falling on a bush) prior to ground impact the risk of fracture decreases (Nevitt and Cummings, 1993). The amount and rate of energy absorbed by the bone is reduced by skin, fat, muscle and clothing that cover the bone. Thus, obese individuals have a reduced risk of hip fractures due to fall impacts (Griffin et al., 1992). This has been put to practical use by Lauritzen, Peterson and Lund (1993) who found that wearing protective pads over the hips reduced the risk of hip fractures. Unfortunately, the device was not always well tolerated by individuals. In a similar manner, compliant surfaces may lower the rate and peak amount of energy that is absorbed by bone (Nevitt and Cummings, 1993). Unfortunately, compliant surfaces may increase the difficulty of balance and gait. To address this concern Cassalino (1995) has developed a unique flooring with a cross section not unlike corrugated cardboard that is firm until an individual falls upon it. At this point, it flexes and decreases the risk of bone fracture by increasing the time and area of impact. The risk of most fractures increases with decreasing bone density, independent of age in general (Seeley et al., 1990), and specifically in those who fall (Nevitt and Cummings, 1993). After age 50, bone density declines approximately 1 % per year, on average (Steiger et al., 1992). Many elderly women have a bone density at least two standard deviations below the mean for normal young women (Melton et al., 1992). On the positive side Nevitt (1997) points out that most falls do not result in a fracture even in women with lower than normal bone density.

2.8.4.4. Fall Prevention Training.

Many studies have found that elderly individuals can improve their balance and gait thorough training and exercise. For example, Fansler, Pott and Shepard (1985) found that older women increased their balance performance on one-leg stance tests after five days of practicing this position. Woollacott, Shumway-Cook and Nashner (1986) found that while older individuals initially lost their balance in perturbations, they were able to maintain their balance during subsequent exposures to the same conditions. Thus, simple exposure or practice will improve balance skill. Probst (1989) found that hip abductor strength in older females was related to their ability to stand on one leg. Roberts (1989) found that participation in a six-week program of aerobic walking improved balance, attributing this to improved lower extremity strength, coordination and flexibility. Conright et al. (1990) showed that a simple walking program involving frail, chair bound nursing home residents walking at least once a week to their tolerance could improve gait and balance. Fiatarone et al. (1990) found a positive relationship between gait velocity and ankle plantar strength in older nursing home residents and that with strength gains came a decreased time to rise from a chair and improved walking performance. Hopkins, Murich, Hoeger and Rhodes (1990) showed that a low impact aerobic dancing program was effective in improving balance performance in older women. Two groups (Judge, Underwood and Gennosa, 1993 and Judge, Lindsey, Underwood and Winsemius, 1993) investigated the effects of specific types of exercise on improving lower extremity strength, balance and walking speed. The interventions in these two studies included flexibility exercises, balance exercises, resistance exercises, walking and Tai Chi movements. Through the various exercises they found that they could improve subjects muscular strength significantly (26 % or 30 % depending on study), increase walking speed and improve balance.

A innovative method of balance training involves the use of Tai Chi. Tai Chi consists of a series of individual dance like movements stressing slow graceful movement and awareness of body weight and alignment. Several researchers investigating the value of Tai Chi in the health of older persons have shown that persons participating in Tai Chi exhibit significantly better postural control (Province et al., 1985; Tse and Bailey, 1992). The movements of Tai Chi may be effective in improving neuromuscular function (quadriceps strength joint flexibility). Two studies comparing computerized balance training and Tai Chi showed mixed results (Wolf et al., 1997; Wolf et al., 1996). The balance training reduced postural sway but did not reduce the rate of falls while Tai Chi did not reduce postural sway but did reduce the rate of falls in the subjects studied. A possible explanation for this discrepancy may be that with Tai Chi training
the skill of posture is improved, thus allowing the individuals to successfully perform tasks with more variety. That is with improvement of skill one should be able to successfully perform a task stance in this case with a wider range of postures. Two studies (Judge, Underwood and Gennosa, 1993; Judge, Lindsey, Underwood and Winsemius, 1993) also incorporated Tai Chi, however they also used other training modalities and it is difficult to tease out the effects related specifically to Tai Chi.

Several researchers have failed to demonstrate an effect of exercise on measures of balance (Crilly et al., 1989; Lichtenstein, Shields, Shiva and Burger, 1989; Topp, Mikesky, Wigglesworth, Holt and Edwards, 1993) or gait (Brown and Holloszy, 1991), or even reducing the occurrence of falls (Mulrow et al., 1994). Brown and Holloszy (1991) using subjects (60 to 71 years) and a combination of resistance and balance exercise found no gait velocity improvement and only minor strength improvements. Sauvage et al. (1992) examined the effects of a moderate to high intensity strengthening and aerobic exercise program on gait and balance in a group of de-conditioned nursing home men (73 years). After completion of a twelve-week program consisting of lower extremity weight training and stationary training, the individuals demonstrated significant improvements in stride length and gait velocity, although their balance was not affected. More recently, research performed on 79 elderly individuals reported in two papers (Owings et al., 1998 and Pavol et al., 1998). When investigators eccentrically, concentrically and isometrically tested strength at the ankle and knee joints and measured the subject’s static posture and reaction to a postural perturbation, they did not find any correlations between strength measures and balance measures.

Whipple’s (1997) review of 25 studies on training older adults to improve their balance, found that approximately 40 % of the studies resulted in no significant improvement in one or more balance parameters measured. He did find that despite the training specificity rule, there were some abilities that improved in a “universal manner” regardless of the training modality (e.g., the ability to stand on one leg improved with almost any training method). The one-legged stance time was the most frequently used measure of outcome. It was used approximately three times more than any other measure, and no other test was used with any consistency. The successful outcome studies contained certain common qualities in their training interventions. They had (1) high body or head velocity, which frequently required fast interactive movements of body segments and eyes in a horizontal plane; (2) larger amplitude vertical movements of the center of gravity which required task associated thigh and hip strength, and (3) endurance activities during full body weight bearing. One surprising finding of Whipple’s was that training seemed to be more unsuccessful in frailer subjects (p < 0.025) despite other studies finding that frailer subjects tended to respond to training. Hypothetically this may be because frailer subjects are undernourished or burdened with disease and thus are incapable of responding to further stress. Judge, Underwood and Gennosa, (1993) and Judge, Lindsey, Underwood and Winsemius, (1993) point out that there may be a critical threshold for muscle strength below which functional impairment occurs.

Whipple (1997), points out that results of training are specific to the type of training that was performed. Unfortunately, he found that most studies only provided abbreviated descriptions of the training procedures or used generalized terms such as coordination, traditional, flexibility, dance like and postural or describe movements that were probably highly specific. Because balance and falls encompass many factors (sensorimotor, musculoskeletal, psychological, social, economical and environmental factors among others that all interact to a greater or lesser degree), training for balance and fall prevention is difficult. On cannot train everything very well. This in part explains some of the failures of training studies. After dividing balance into thirteen major categories (some of which were further divided into subcategorizes), Whipple (1997) found that many studies stressed one factor in the intervention while using another factor as a measure of outcome. Several studies fail to match the training modality with the balance criterion being tested. In such a case, a positive study result depended on the amount of positive interaction the two factors had. One thing that Whipple did not note was the possibility that training could have had a negative effect even if it positively influenced the measure of outcome if it also negatively influenced another (other) factor(s) to a greater extent. This would be similar to the interaction that is seen between endurance training, strength training and jumping ability. That is endurance training has been shown to have a negative effect on an individual’s jumping ability while strength training (provided its ballistic enough) has a positive influence. In addition strength and endurance training together are counter productive. Although training generally only influences one category the average number of categories per study was only two. Just three research studies used as many as four categories of balance in their training
and only one utilized eight. In addition to under representation of the broadness of balance within studies, there was an under representation of balance challenges even when the studies were combined. That is, there are many more challenges to balance than have been studied in training experiments. Specifically while falls may arise from hazards and/or unexpected events, training in the studies that Whipple reviewed did not address these issues. In general, the more successful studies used more strenuous training methods.

Given the general trend of deterioration of balance control ability and increased risk of falling with advancing age, the question of whether this deterioration can be slowed, prevented or counterbalanced by training naturally arises. While many studies have shown that many characteristics of the elderly can be improved and maintained with training (e.g., strength, VO\textsubscript{2}), studies have yet to show how to train an individual to prevent or minimize falls and fall related injuries.

**2.9. The Effect of Muscular Strength on Balance.**

Although the activity of individual muscles during stance has been monitored through many EMG studies, with two exceptions the strength that is needed to maintain posture has not been studied. So far, research indicates that, within limits, the strength of the muscles is not related to balance ability in healthy subjects (Alexander, Shepard, Gu and Schultz, 1992; Gu, Schultz, Shepard and Alexander, 1996). In these two studies, the posture of 26 healthy young adults (mean age 26 years) and 15 healthy elderly adults (mean age 72 years) was studied in four tasks. The four tasks included standing with feet flat on a platform that accelerated anteriorly, standing on a 0.11m wide beam, standing on the 0.11m wide beam while it accelerated anteriorly, and standing for 10 s on a spring board that was designed to rotate 0.8\textdegree for every 1.0\textdegree of whole body sway. The accelerations for the flat translation and beam translations were 1.67m/s\textsuperscript{2} for 0.033m and 0.89m/s\textsuperscript{2} for 0.017m respectively. Alexander, Shepard, Gu and Schultz (1992) and Gu, Schultz, Shepard and Alexander (1996) modeled the body by seven rigid, pin joint connected links (one each for the two feet, two shanks, two thighs, trunk, head and neck, two upper arms and two forearms with hands). They assumed sagittal plane symmetry based on the symmetry they observed in kinematic observations (Alexander, Shepard, Gu and Schultz, 1992). In addition to time histories of the torques at the ankle, knee, hip, neck, shoulder and elbow joints, they measured the total body angular momenta about the ankle joint, the excursions of the centroid of the ground reaction force and the total body CM. They calculated the multi-body dynamics using both a “head and arms- down” and a “foot up” approach. They found that the foot support torques calculated from the “head- and arms- down” method from ten randomly selected trials ranged from 3.3 to 10.1 Nm with a mean error of 5.7 Nm. They found the mean peak angular momenta about the ankle joint to range from 1.7 kg\cdot m\textsuperscript{2}/s in the two standing tasks to 6.0 kg\cdot m\textsuperscript{2}/s in the two translation tasks (see table 2.9.1).
Table 2.9.1. Population means of the peak total angular momenta about the ankle joint (kg·m²/s). Adapted from Gu, Schultz, Shepard and Alexander (1996).

<table>
<thead>
<tr>
<th></th>
<th>Flat translation</th>
<th>Beam standing</th>
<th>Beam translation</th>
<th>Spring standing</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>counter clockwise</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>young</td>
<td>6.00 (1.56)</td>
<td>1.21 (1.00)</td>
<td>3.27 (0.98)</td>
<td>0.92 (0.43)</td>
</tr>
<tr>
<td>elderly</td>
<td>5.45 (1.07)</td>
<td>1.74 (1.06)</td>
<td>2.95 (0.72)</td>
<td>0.92 (0.43)</td>
</tr>
<tr>
<td><strong>clockwise</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>young</td>
<td>3.17 (1.38)</td>
<td>1.20 (0.91)</td>
<td>1.44 (0.78)</td>
<td>0.96 (0.46)</td>
</tr>
<tr>
<td>elderly</td>
<td>3.32 (2.08)</td>
<td>1.74 (1.14)</td>
<td>2.36 (1.51)*</td>
<td>0.95 (0.51)</td>
</tr>
<tr>
<td><strong>amount arrested</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>young</td>
<td>2.84 (1.83)</td>
<td></td>
<td>1.83 (1.35)</td>
<td></td>
</tr>
<tr>
<td>elderly</td>
<td>2.12 (2.29)</td>
<td></td>
<td>0.60 (1.25)**</td>
<td></td>
</tr>
<tr>
<td><strong>amount arrested (percent of CCW peak)</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>young</td>
<td>44 (31)</td>
<td></td>
<td>50 (34)</td>
<td></td>
</tr>
<tr>
<td>elderly</td>
<td>37 (40)</td>
<td></td>
<td>22 (37)</td>
<td></td>
</tr>
</tbody>
</table>

Standard deviations in parentheses.

The angular momentum arrested is equal to the difference between the first counterclockwise and the first clockwise peak.
Elderly different from young *(p<0.02) **(p<0.05).

Unfortunately, they did not report absolute joint torques. Rather they gave deviations from reference values in which the reference values were the torques of the mean body configuration over the two seconds prior to the perturbations (Alexander, Shepard, Gu and Schultz, 1992). They reasoned that the reference joint torques were probably small because the horizontal moment arm from any given joint to the CM of the body segment was small when the subject was in the reference position (Gu, Schultz, Shepard and Alexander, 1996). They found the reference ankle joint torques to be approximately 7.3 Nm when the subject was in an upright standing position, and 14.4 Nm when a subject leaned forward and shifted the total body CM 0.01m anteriorly. The mean maximum changes from reference torque values were at most 38.8 Nm (elderly knee extension). Note that the values reported in tables 2.9.2 and 2.9.3 are for both sides of the body combined for the ankle, knee and hip joints. Further, they found that the largest torque changes occurred at the ankles and the knees in response to flat translation. Note that the peaks for beam translation are greater than that for the beam standing, and they are less than that for flat translation. Unfortunately, the accelerations experienced in the translation tasks were different.
<table>
<thead>
<tr>
<th>Table 2.9.2. The maximum changes in joint flexor torque (Nm). Adapted from Gu, Schultz, Shepard and Alexander (1996).</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>flat translation</strong></td>
</tr>
<tr>
<td></td>
</tr>
<tr>
<td>ankle (DF)</td>
</tr>
<tr>
<td>young</td>
</tr>
<tr>
<td>37.1 (11.0)</td>
</tr>
<tr>
<td>elderly</td>
</tr>
<tr>
<td>37.1 (9.1)</td>
</tr>
</tbody>
</table>

| **beam standing**                                            |
|                                                                 |
| ankle (DF) | knee | hip | neck |
| young      |      |     |      |
| 12.0 (5.4)  | 9.0 (5.22) | 5.6 (3.3)  | 0.20 (0.12)  |
| elderly    |      |     |      |
| 17.3 (7.8)** | 16.9 (8.1)** | 12.4 (9.9)***** | 0.44 (0.42)***** |

| **beam translation**                                         |
|                                                                 |
| ankle (DF) | knee | hip | neck |
| young      |      |     |      |
| 19.1 (7.2)  | 9.8 (6.8)  | 12.7 (5.6) | 0.44 (0.28) |
| elderly    |      |     |      |
| 23.2 (7.1)  | 16.4 (13.0)* | 18.3 (6.5)****** | 0.68 0 (0.40)* |

| **spring standing**                                          |
|                                                                 |
| ankle (DF) | knee | hip | neck |
| young      |      |     |      |
| 10.2 (5.0)  | 7.1 (3.5)  | 3.6 (1.8) | 0.18 0 (0.08) |
| elderly    |      |     |      |
| 11.5 (4.6)  | 7.8 (3.4)  | 4.7 (2.4) | 0.18 (0.07) |

Standard deviations in parentheses.
To obtain the torques per side, all quoted values except those of the neck should be divided in half.
Elderly different from young * (p >0.05) ** (p >0.01) *** (p >0.001) **** (p >0.002) ***** (p >0.02)
****** (p >0.005).

<table>
<thead>
<tr>
<th>Table 2.9.3. The maximum changes in joint extensor torque (Nm). Adapted from Gu, Schultz, Shepard and Alexander (1996).</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>flat translation</strong></td>
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<tr>
<td></td>
</tr>
<tr>
<td>ankle (PF)</td>
</tr>
<tr>
<td>young</td>
</tr>
<tr>
<td>16.6 (11.6)</td>
</tr>
<tr>
<td>elderly</td>
</tr>
<tr>
<td>19.5 (15.2)</td>
</tr>
</tbody>
</table>

| **beam standing**                                            |
|                                                                 |
| ankle (PF) | knee | hip | neck |
| young      |      |     |      |
| 11.5 (6.3)  | 10.1 (5.2)  | 5.84 (3.41) | 0.22 (0.11)  |
| elderly    |      |     |      |
| 17.9 (7.3)** | 17.5 (10.5)*** | 11.5 (8.4)** | 0.41 (0.37)* |

| **beam translation**                                         |
|                                                                 |
| ankle (PF) | knee | hip | neck |
| young      |      |     |      |
| 13.9 (7.8)  | 22.6 (9.4)  | 11.1 (3.7) | 0.40 (0.27) |
| elderly    |      |     |      |
| 17.4 (11.5) | 25.7 (5.4)  | 16.3 (13.4) | 0.62 (0.42) |

| **spring standing**                                          |
|                                                                 |
| ankle (PF) | knee | hip | neck |
| young      |      |     |      |
| 10.4 (5.0)  | 7.14 (3.79) | 3.65 (1.80) | 0.19 0 (0.08) |
| elderly    |      |     |      |
| 10.5 (3.8)  | 8.27 (3.96) | 4.47 (2.44) | 0.19 (0.11) |

Standard deviations in parentheses.
To obtain the torques per side, all quoted values except those of the neck should be divided in half.
Elderly different from young * (p<0.05) ** (p<0.05) *** (p<0.002).

Gu, Schultz, Shepard and Alexander (1996) found that torques developed at major body joints in response to horizontal perturbations are small compared to the available joint strengths in both young and
elderly subjects. They concluded from these tasks that whatever difficulties with postural balance the healthy elderly may have in response to modest perturbations, those difficulties do not arise because of deficits in muscular strength. They are the only researchers that have so far looked at the dynamics of postural control when stance is perturbed. No one has researched the torques required to maintain balance in response to rotational perturbations.

In addition to the posture studies, biomechanical studies of a more strength intensive task, rising from a seated position, have mixed results. Chun et al. (1992) showed that the muscle strength in most adults and elderly is adequate while Hughes, Myers and Schenkman (1996) found a small group of elderly with functional impairment (defined as the inability to descend four consecutive stairs without use of a handrail for added support) had trouble rising from a seated position with out the aid of their hands. Hughes, Myers and Schenkman (1996) found that the torque generated at the knee joint was the limiting factor (in contrast to the task of standing balance where the ankle joint experiences the highest magnitude torques). Simoneau (1992) demonstrated that weakness resulting from diabetic neuropathy had no statistically significant effect on postural sway. In spite of these findings, most physical therapy is designed to increase muscular strength (Schultz, 1992).

The following tables consist of normative data of ankle, knee and hip joint strengths from a review of 49 studies in the literature. The tables list the net joint torques in Nm however, many of the papers reported their results in different units. The appropriate conversions were made. In some cases, the results were not reported in units of torque but of force. In these cases, the force was converted to a torque by assuming that the force was applied to lever arms that were arbitrarily assigned to be 0.10 m for the foot, 0.44 m for the shank and 0.42 m long for the thigh. Other studies reported results in units of mass. In these instances, it was assumed that the proper results could be achieved by multiplying by the acceleration of gravity (g = 9.81 m/s² = 32.2 ft/s²). The results of the studies were divided into groups by the angular velocity of the joint tested, age and gender. A young adult designation includes individuals between 17 and 39 years, while an older adult designation indicates individuals over 60 years old. Note the wide variation that exists in the literature. The variation may be due to differences in subjects (i.e., health, frailty, and activity) as well as methodological factors (i.e., different equipment, movement velocities, positions or ranges). Unfortunately, there is no way to normalize the data for comparison.

Table 2.9.4. Net ankle dorsiflexion torques (Nm) from studies reported in literature.

<table>
<thead>
<tr>
<th>study</th>
<th>angular velocity</th>
<th>young adult</th>
<th>older adult</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>females</td>
<td>males</td>
</tr>
<tr>
<td>Amundsen (1990)(1)</td>
<td>0 °/s</td>
<td>16</td>
<td></td>
</tr>
<tr>
<td>Clarke (1966)(1)</td>
<td>0 °/s</td>
<td>32</td>
<td></td>
</tr>
<tr>
<td>Oberg et al. (1987)</td>
<td>0 °/s</td>
<td>52</td>
<td></td>
</tr>
<tr>
<td>Sepic et al. (1986)</td>
<td>0 °/s</td>
<td>74</td>
<td>78</td>
</tr>
<tr>
<td>Fugl-Meyer (1981)</td>
<td>30 °/s</td>
<td>20</td>
<td>25</td>
</tr>
<tr>
<td>Weldon et al. (1988)</td>
<td>30 °/s</td>
<td>29</td>
<td>22</td>
</tr>
<tr>
<td>Oberg et al. (1987)</td>
<td>60 °/s</td>
<td>36</td>
<td></td>
</tr>
<tr>
<td>Whipple et al. (1987)</td>
<td>60 °/s</td>
<td></td>
<td></td>
</tr>
<tr>
<td>average ± std. dev.</td>
<td>42 ± 21</td>
<td>38 ± 28</td>
<td>34 ± 17</td>
</tr>
<tr>
<td>range</td>
<td>20-74</td>
<td>16-78</td>
<td>22-46</td>
</tr>
</tbody>
</table>

Note: (1) Data were given in lbs. in study. Converted to Nm by assuming a foot length of 0.10 m.
Table 2.9.5. Net ankle plantar flexion torques (Nm) from studies reported in literature.

<table>
<thead>
<tr>
<th>study</th>
<th>angular velocity</th>
<th>young adult females</th>
<th>young adult males</th>
<th>older adult females</th>
<th>older adult males</th>
</tr>
</thead>
<tbody>
<tr>
<td>Amundsen (1990)(1)</td>
<td>0 %/s</td>
<td>95</td>
<td>121</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Clarke (1966)(1)</td>
<td>0 %/s</td>
<td></td>
<td></td>
<td>28</td>
<td></td>
</tr>
<tr>
<td>Falkel (1978)</td>
<td>0 %/s</td>
<td>58</td>
<td>87</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Fugl-Meyer et al. (1979)</td>
<td>0 %/s</td>
<td></td>
<td></td>
<td>139</td>
<td></td>
</tr>
<tr>
<td>Fugl-Meyer et al. (1980)</td>
<td>0 %/s</td>
<td>109</td>
<td>163</td>
<td>83</td>
<td>128</td>
</tr>
<tr>
<td>Oberg et al. (1987)</td>
<td>0 %/s</td>
<td>210</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sale et al. (1982)</td>
<td>0 %/s</td>
<td></td>
<td></td>
<td>114</td>
<td></td>
</tr>
<tr>
<td>Sepic et al. (1986)</td>
<td>0 %/s</td>
<td>131</td>
<td>129</td>
<td>82</td>
<td>100</td>
</tr>
<tr>
<td>Strobbe (1982)</td>
<td>0 %/s</td>
<td>81</td>
<td>126</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Vandervoort et al. (1989)</td>
<td>0 %/s</td>
<td>135</td>
<td></td>
<td>39</td>
<td></td>
</tr>
<tr>
<td>Farkel (1978)</td>
<td>30 %/s</td>
<td>45</td>
<td>71</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Fugl-Meyer (1981)</td>
<td>30 %/s</td>
<td>63</td>
<td>96</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Fugl-Meyer et al. (1979)</td>
<td>30 %/s</td>
<td></td>
<td></td>
<td>123</td>
<td></td>
</tr>
<tr>
<td>Gerde et al. (1985)</td>
<td>30 %/s</td>
<td></td>
<td></td>
<td>78</td>
<td>139</td>
</tr>
<tr>
<td>Oberg et al. (1987)</td>
<td>60 %/s</td>
<td>128</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Puhl et al. (1982)</td>
<td>30 %/s</td>
<td>60</td>
<td>80</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Weldon et al. (1988)</td>
<td>30 %/s</td>
<td>87</td>
<td></td>
<td>52</td>
<td></td>
</tr>
<tr>
<td>Whipple et al. (1987)</td>
<td>60 %/s</td>
<td></td>
<td></td>
<td>16</td>
<td></td>
</tr>
<tr>
<td>average ± std. dev.</td>
<td></td>
<td>100 ± 46</td>
<td>106 ± 36</td>
<td>67 ± 20</td>
<td>96 ± 56</td>
</tr>
<tr>
<td>range</td>
<td></td>
<td>45-210</td>
<td>28-163</td>
<td>39-83</td>
<td>16-139</td>
</tr>
</tbody>
</table>

Note: (1) Data were given in lbs. in study. Converted to Nm by assuming a foot length of 0.10m.
Table 2.9.6. Net knee flexion torques (Nm) from studies reported in literature.

<table>
<thead>
<tr>
<th>study</th>
<th>angular velocity</th>
<th>young adult</th>
<th>older adult</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>females</td>
<td>males</td>
</tr>
<tr>
<td>Amundsen (1990)(2)</td>
<td>0°/s</td>
<td>110</td>
<td>160</td>
</tr>
<tr>
<td>Aniansson et al. (1983)</td>
<td>0°/s</td>
<td>42</td>
<td>73</td>
</tr>
<tr>
<td>Borges (1989)</td>
<td>0°/s</td>
<td>169</td>
<td>301</td>
</tr>
<tr>
<td>Campney et al(1965)(2)</td>
<td>0°/s</td>
<td>127</td>
<td>196</td>
</tr>
<tr>
<td>Clarke (1966)(2)</td>
<td>0°/s</td>
<td></td>
<td>179</td>
</tr>
<tr>
<td>Houtz et al. (1957)(2)</td>
<td>0°/s</td>
<td>57</td>
<td></td>
</tr>
<tr>
<td>Kuta et al. (1970)</td>
<td>0°/s</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Moffroid et al. (1969)</td>
<td>0°/s</td>
<td>65</td>
<td>65</td>
</tr>
<tr>
<td>Murray et al. (1977)</td>
<td>0°/s</td>
<td>118</td>
<td>91</td>
</tr>
<tr>
<td>Murray et al. (1980)</td>
<td>0°/s</td>
<td>117</td>
<td>76</td>
</tr>
<tr>
<td>Murray et al. (1985)</td>
<td>0°/s</td>
<td>71</td>
<td>50</td>
</tr>
<tr>
<td>Schudder et al. (1980)</td>
<td>0°/s</td>
<td>170</td>
<td></td>
</tr>
<tr>
<td>Smidt (1973)</td>
<td>0°/s</td>
<td>62</td>
<td></td>
</tr>
<tr>
<td>Strobbe (1982)</td>
<td>0°/s</td>
<td>62</td>
<td>100</td>
</tr>
<tr>
<td>Williams (1959)(2)</td>
<td>0°/s</td>
<td></td>
<td>153</td>
</tr>
<tr>
<td>Borges (1989)</td>
<td>12°/s</td>
<td>100</td>
<td>155</td>
</tr>
<tr>
<td>Aniansson et al. (1983)</td>
<td>30°/s</td>
<td>38</td>
<td>68</td>
</tr>
<tr>
<td>Costain et al. (1984)</td>
<td>30°/s</td>
<td>109</td>
<td></td>
</tr>
<tr>
<td>Gilliam et al. (1979)</td>
<td>30°/s</td>
<td></td>
<td>149</td>
</tr>
<tr>
<td>Goslin et al. (1979)</td>
<td>30°/s</td>
<td>48</td>
<td>74</td>
</tr>
<tr>
<td>Imwold et al. (1983)</td>
<td>30°/s</td>
<td></td>
<td>107</td>
</tr>
<tr>
<td>Puhl et al. (1982)</td>
<td>30°/s</td>
<td>110</td>
<td>165</td>
</tr>
<tr>
<td>Smith et al. (1981)</td>
<td>30°/s</td>
<td></td>
<td>174</td>
</tr>
<tr>
<td>Murray et al. (1980)</td>
<td>36°/s</td>
<td></td>
<td>78</td>
</tr>
<tr>
<td>Murray et al. (1985)</td>
<td>36°/s</td>
<td>57</td>
<td>32</td>
</tr>
<tr>
<td>Dibiezzo et al. (1985)</td>
<td>60°/s</td>
<td>70</td>
<td></td>
</tr>
<tr>
<td>Weldon et al. (1988)</td>
<td>60°/s</td>
<td>89</td>
<td>51</td>
</tr>
<tr>
<td>Whipple et al. (1987)</td>
<td>60°/s</td>
<td></td>
<td>31</td>
</tr>
<tr>
<td>Housh et al. (1984)</td>
<td>180°/s</td>
<td>80</td>
<td></td>
</tr>
<tr>
<td>average ± std. dev.</td>
<td>84 ± 34</td>
<td>135 ± 60</td>
<td>64 ± 34</td>
</tr>
<tr>
<td>range</td>
<td>38-169</td>
<td>62-301</td>
<td>32-122</td>
</tr>
</tbody>
</table>

Note: (2) Data were given in lbs. in study. Converted to Nm by assuming a leg length of 0.44 m.
Table 2.9.7. Net knee extension torques (Nm) from literature studies.

<table>
<thead>
<tr>
<th>study</th>
<th>young adult</th>
<th>older adult</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>velocity</td>
<td></td>
</tr>
<tr>
<td></td>
<td>females</td>
<td>males</td>
</tr>
<tr>
<td>Amundsen (1990)(2)</td>
<td>0°/s</td>
<td>204</td>
</tr>
<tr>
<td>Aniansson et al. (1980)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Aniansson et al. (1983)</td>
<td>0°/s</td>
<td>103</td>
</tr>
<tr>
<td>Aniansson et al. (1986)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Clarke (1966)(2)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Dannenskiold et al(84)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Houtz et al. (1957)(2)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Kuta et al. (1970)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Mendler (1967)(2)</td>
<td>0°/s</td>
<td>236</td>
</tr>
<tr>
<td>Moffroid et al. (1969)</td>
<td>0°/s</td>
<td>104</td>
</tr>
<tr>
<td>Murray et al. (1977)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Murray et al. (1980)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Murray et al. (1985)</td>
<td>0°/s</td>
<td>98</td>
</tr>
<tr>
<td>Peterson et al. (1960)</td>
<td>0°/s</td>
<td>130</td>
</tr>
<tr>
<td>Schudder et al. (1980)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Sipila (1991)(2)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Smidt (1973)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Strobbe (1982)</td>
<td>0°/s</td>
<td>106</td>
</tr>
<tr>
<td>Viitasalo et al(1985)(2)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Williams (1959)(2)</td>
<td>0°/s</td>
<td></td>
</tr>
<tr>
<td>Borges (1989)</td>
<td>12% /s</td>
<td>183</td>
</tr>
<tr>
<td>Aniansson et al. (1980)</td>
<td>30% /s</td>
<td></td>
</tr>
<tr>
<td>Aniansson et al. (1983)</td>
<td>30% /s</td>
<td>82</td>
</tr>
<tr>
<td>Aniansson et al. (1986)</td>
<td>30% /s</td>
<td></td>
</tr>
<tr>
<td>Costain et al. (1984)</td>
<td>30% /s</td>
<td>179</td>
</tr>
<tr>
<td>Gilliam et al. (1979)</td>
<td>30% /s</td>
<td></td>
</tr>
<tr>
<td>Goslin et al. (1979)</td>
<td>30% /s</td>
<td>110</td>
</tr>
<tr>
<td>Imwold et al. (1983)</td>
<td>30% /s</td>
<td>194</td>
</tr>
<tr>
<td>Puhl et al. (1982)</td>
<td>30% /s</td>
<td>140</td>
</tr>
<tr>
<td>Smith et al. (1981)</td>
<td>30% /s</td>
<td></td>
</tr>
<tr>
<td>Murray et al. (1980)</td>
<td>36% /s</td>
<td>194</td>
</tr>
<tr>
<td>Murray et al. (1985)</td>
<td>36% /s</td>
<td>116</td>
</tr>
<tr>
<td>Dibrezzo et al. (1985)</td>
<td>60% /s</td>
<td>131</td>
</tr>
<tr>
<td>Nordesjo et al. (1978)</td>
<td>60% /s</td>
<td>125</td>
</tr>
<tr>
<td>Weldon et al. (1988)</td>
<td>60% /s</td>
<td>139</td>
</tr>
<tr>
<td>Whipple et al. (1987)</td>
<td>60% /s</td>
<td></td>
</tr>
<tr>
<td>Housh et al. (1984)</td>
<td>180% /s</td>
<td>107</td>
</tr>
<tr>
<td>average ± std. dev.</td>
<td></td>
<td></td>
</tr>
<tr>
<td>range</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Note: (2) Data were given in lbs. and converted to Nm by assuming a leg length of 0.44 m.
Table 2.9.8. Net hip flexion torques (Nm) from studies reported in literature.

<table>
<thead>
<tr>
<th>study</th>
<th>angular velocity</th>
<th>young adult females</th>
<th>young adult males</th>
<th>older adult females</th>
<th>older adult males</th>
</tr>
</thead>
<tbody>
<tr>
<td>Amundsen (1990)(3)</td>
<td>0°/s</td>
<td>83</td>
<td>139</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cahalan et al. (1989)</td>
<td>0°/s</td>
<td>105</td>
<td>167</td>
<td>86</td>
<td>166</td>
</tr>
<tr>
<td>Clarke (1966)(3)</td>
<td>0°/s</td>
<td>387</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Elkins et al. (1951)(3)</td>
<td>0°/s</td>
<td>122</td>
<td>185</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Jensen et al. (1971)(3)</td>
<td>0°/s</td>
<td>25</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Markhede et al. (1980)</td>
<td>0°/s</td>
<td>120</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Smidt et al. (1983)</td>
<td>0°/s</td>
<td>150</td>
<td>150</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Strobbe (1982)</td>
<td>0°/s</td>
<td>126</td>
<td>185</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Viitasalo et al. ('85)(3)</td>
<td>0°/s</td>
<td>306</td>
<td>212</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Williams et al. ('59)(3)</td>
<td>0°/s</td>
<td>243</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cahalan et al. (1989)</td>
<td>30°/s</td>
<td>91</td>
<td>152</td>
<td>67</td>
<td>113</td>
</tr>
<tr>
<td>Strobbe (1982)</td>
<td>30°/s</td>
<td>97</td>
<td>190</td>
<td></td>
<td></td>
</tr>
<tr>
<td>average ± std. dev.</td>
<td>118 ± 25</td>
<td>186 ± 92</td>
<td>77 ± 13</td>
<td>164 ± 50</td>
<td></td>
</tr>
<tr>
<td>range</td>
<td>83-150</td>
<td>25-387</td>
<td>67-86</td>
<td>113-212</td>
<td></td>
</tr>
</tbody>
</table>

Note: (3) Data were given in lbs. in study. Converted to Nm by assuming a thigh length of 0.42 m.

Table 2.9.9. Net hip extension torques (Nm) from studies reported in literature.

<table>
<thead>
<tr>
<th>study</th>
<th>angular velocity</th>
<th>young adult females</th>
<th>young adult males</th>
<th>older adult females</th>
<th>older adult males</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cahalan et al. (1989)</td>
<td>0°/s</td>
<td>95</td>
<td>160</td>
<td>82</td>
<td>156</td>
</tr>
<tr>
<td>Clarke (1966)(3)</td>
<td>0°/s</td>
<td>419</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Jensen et al. (1971)(3)</td>
<td>0°/s</td>
<td>39</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Markhede et al. (1980)</td>
<td>0°/s</td>
<td>248</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Smidt et al. (1983)</td>
<td>0°/s</td>
<td>325</td>
<td>325</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Strobbe (1982)</td>
<td>0°/s</td>
<td>97</td>
<td>190</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Viitasalo et al. ('85)(3)</td>
<td>0°/s</td>
<td>420</td>
<td>277</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cahalan et al. (1989)</td>
<td>30°/s</td>
<td>110</td>
<td>177</td>
<td>101</td>
<td>157</td>
</tr>
<tr>
<td>Smith et al. (1981)</td>
<td>30°/s</td>
<td>276</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>average ± std. dev.</td>
<td>157 ± 112</td>
<td>250 ± 125</td>
<td>92 ± 13</td>
<td>197 ± 70</td>
<td></td>
</tr>
<tr>
<td>range</td>
<td>95-325</td>
<td>39-420</td>
<td>82-101</td>
<td>156-277</td>
<td></td>
</tr>
</tbody>
</table>

Note: (3) Data were given in lbs. in study. Converted to Nm by assuming a thigh length of 0.42 m.

2.10. Summary.

Stability, balance, equilibrium and posture are interdependent entities, each dependent on one another as well as on other factors such as the individual’s mental state, environment, heredity and culture. Each individual holds their body according to how they feel emotionally, their perception of the environment, their body structure and physiology that was handed down by their ancestors, and in a manner that is socially consistent and acceptable. In addition to these factors, mechanical quantities such as CM, center of gravity, center of pressure and base of support interact through the laws of mechanics to place limits on the variability in postures an individual can attain. Muscular strength, joint range of motion, segmental mass and geometry also interact to define the strategies one uses to maintain posture. Thus, it is impossible to understand posture and the strategies with which one tries to achieve balance without an
understanding of the mechanics involved. Only then can one hope to scrutinize the effects of different physiological and orthopedic conditions on balance and develop methods to alleviate any problems.
Chapter 3
Methodology

3.1. Subject Description.
Twenty-two subjects with ages ranging from 19 to 35 years old were recruited from the undergraduate and graduate populations at the Pennsylvania State University (table 3.8.1). Those willing to participate were given a preliminary screening by telephone (Appendix C). Those who passed the initial screening (exclusion criteria in Appendix A) were invited to the Center for Locomotion Studies (CELOS) to continue in the study. At CELOS, the subjects went through a more thorough medical history and clinical testing (Appendix D). Included were tests of visual function, vestibular function, deep tendon reflexes, vibratory perception, touch-pressure sensation and range of motion of the ankle joint.

Visual function was tested using a standard visual acuity test chart illuminated at approximately 100 candelas/m$^2$. The subjects were asked to stand 20 feet from the chart and read the line of letters corresponding to 20/20. If the subject failed to read this line, they were asked to read progressively larger lines until they were successful. If the subjects were successful in reading the 20/20 line, they were asked to read progressively smaller lines until they were unsuccessful. The visual acuity of the subject was the smallest line that they could successfully read. This test was performed with both eyes open, only the left eye open and only the right eye open.

3.3. Vestibular Function Testing.
All subjects were tested for gross vestibular dysfunction by four tests as described by Herdman (1989). Each subject was tested for spontaneous nystagmus (spasmodic, slow drift of the eye followed by a rapid correction) while focusing on a stationary target, and while tracking a moving target. Saccadic eye movements were searched for when the subject tracked a moving object with their eyes while their head was held still, and when the subjects fixed their eyes on a stationary target while their head was slowly rotating left and right or up and down.

Deep tendon reflexes of the patellar and Achilles tendons were stimulated using a reflex hammer, according to standard clinical techniques. The response was scored as being either present or absent. In the absence of a response, reinforcement through clasping hands and tensing arm muscles was used to facilitate the reflex response.

Vibration sensation was assessed using a fixed-frequency (60 Hz) variable amplitude biothesiometer (Bio-medical Instrument Company, Newbury, Ohio) on the heel, fifth metatarsal head, first metatarsal head and the hallux of both feet. Each subject was tested while they were laying in a prone position with the knee flexed 90$^0$ and the plantar surface of the foot horizontal. The bio-thesiometer was placed on the foot so that its weight generated a standard contact pressure. Sensation was tested by starting the bio-thesiometer at zero and slowly increasing the vibration amplitude until the subject felt the vibration. The amplitude was increased slightly. Then the stimulus was decreased until the subject could no longer perceive any vibration. The process was repeated two more times and the upper and lower limits of perception were recorded.

3.6. Touch-Pressure Sensation Testing.
Semmes-Weinstein (S-W) monofilaments (North Coast Medical, Inc.) were used to evaluate touch-pressure sensation (Weinstein, 1962). The monofilaments are nylon fibers of different gauges. The greater the thickness, the greater the force required to buckle the filament. The fibers that were used were
graded as 4.17, 5.07 and 6.10 which is the log_{10} of 10 times the first order (C shape deformation) buckling force in mg. That is, the monofilaments required a force of 1.48 g, 11.75 g and 125.89 g respectively to cause them to buckle. Perception was measured on the heel, fifth metatarsal head, first metatarsal head and the hallux using a forced choice protocol developed by Sosenko et al. (1990). For each site, the subject was tested using the monofilaments in descending order. At each site, a filament was applied at time A or at time B. The subject had to indicate at which time they thought that the monofilament was applied. If a correct answer was given, the next lowest monofilament was used and the process repeated. If the subject gave a wrong answer, the subject was tested again with the same monofilament. The failure to sense a 6.10 monofilament at one of the forefoot regions was defined as loss of protective sensation (Simoneau, 1992). The testing was performed, on both feet, with the subjects lying prone, knee flexed 90°, and plantar surface of the foot, horizontal.

3.7. Range of Motion of the Ankle Joint.

Range of movement of the ankle (dorsiflexion and plantar flexion) was measured using standard clinical methods. To measure ankle plantar flexion, the subjects were in a supine position with their knees fully extended and the subtalar joints in neutral position. Both ankles were plantar flexed as far as possible by the examiner. Ankle dorsiflexion was measured with the subjects in two positions 1) with the subjects lying supine with the knee fully extended and 2) with the subjects lying supine and the knee flexed 90°. The ankle was dorsiflexed as far as possible by the examiner. The vertex of the goniometer was positioned over the lateral malleolus center with the one arm aligned with the midline of the fibula (using the head of the fibula for reference) and the other arm aligned parallel to the inferior aspect of the calcaneus.

3.8. Subject Summary.

Half of the subjects were male and half were female. The subjects ranged from 1.53m to 1.86m in height and from 478.7 N to 772.1 N in weight. The subjects’ characteristics are listed in table 3.8.1. Blood pressure (BP) measurements were used to screen for diastolic and systolic drops when a subject stood from a seated position. Visual acuity was measured by the standard Snellen test, and the score reported was the size of the smallest line the subject could read at a distance of 20 feet. The subjects were tested with corrective eyewear, if needed, as the test was a screen for a subject’s ability to see and not a test of their innate visual system. No subject was below 20/25. A few subjects even possessed an uncorrected acuity of 20/10. The lateral malleolus height, the distance from the center of the right lateral malleolus to the floor, was measured and used to adjust the platform so that its axis of rotation coincided with the ankle joint. Each subject was screened for insensate feet by using a biothesiometer. Vibration scores were obtained for the left and right heel, fifth metatarsal head, first metatarsal head and hallux. Ankle dorsiflexion range was measured in degrees with the knee in two positions: full extension and flexed 90°. The ankle plantar flexion range was measured with the knee fully extended. The ankle range of motion measurements were performed to eliminate any subjects from the study whose ankle range of motion limited their ability to stand on the rotating platform.
Table 3.8.1. Subject characteristics.

<table>
<thead>
<tr>
<th></th>
<th>units</th>
<th>mean</th>
<th>std dev</th>
<th>max</th>
<th>min</th>
</tr>
</thead>
<tbody>
<tr>
<td>age</td>
<td>years</td>
<td>25.91</td>
<td>4.96</td>
<td>35.00</td>
<td>19.00</td>
</tr>
<tr>
<td>male</td>
<td>number</td>
<td>11</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>female</td>
<td>number</td>
<td>11</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>height</td>
<td>m</td>
<td>1.72</td>
<td>0.10</td>
<td>1.86</td>
<td>1.53</td>
</tr>
<tr>
<td>weight</td>
<td>N</td>
<td>649.62</td>
<td>84.99</td>
<td>772.05</td>
<td>478.72</td>
</tr>
<tr>
<td>sitting BP</td>
<td>mmHg</td>
<td>106.36</td>
<td>7.21</td>
<td>124.00</td>
<td>94.00</td>
</tr>
<tr>
<td>standing BP</td>
<td>mmHg</td>
<td>106.64</td>
<td>9.47</td>
<td>120.00</td>
<td>90.00</td>
</tr>
<tr>
<td>visual acuity</td>
<td>pts</td>
<td>17.14</td>
<td>4.03</td>
<td>25.00</td>
<td>10.00</td>
</tr>
<tr>
<td>lat. mal ht</td>
<td>cm</td>
<td>6.51</td>
<td>0.80</td>
<td>8.00</td>
<td>5.00</td>
</tr>
<tr>
<td>heel vib l</td>
<td>units</td>
<td>3.82</td>
<td>1.06</td>
<td>6.00</td>
<td>2.00</td>
</tr>
<tr>
<td>heel vib r</td>
<td>units</td>
<td>3.52</td>
<td>0.89</td>
<td>6.00</td>
<td>2.50</td>
</tr>
<tr>
<td>fifth vib l</td>
<td>units</td>
<td>3.77</td>
<td>1.21</td>
<td>8.00</td>
<td>2.50</td>
</tr>
<tr>
<td>fifth vib r</td>
<td>units</td>
<td>3.82</td>
<td>1.27</td>
<td>7.50</td>
<td>2.50</td>
</tr>
<tr>
<td>first vib l</td>
<td>units</td>
<td>3.86</td>
<td>1.26</td>
<td>7.50</td>
<td>2.00</td>
</tr>
<tr>
<td>first vib r</td>
<td>units</td>
<td>4.07</td>
<td>0.81</td>
<td>5.50</td>
<td>3.00</td>
</tr>
<tr>
<td>hallux vib l</td>
<td>units</td>
<td>6.68</td>
<td>2.50</td>
<td>11.00</td>
<td>3.00</td>
</tr>
<tr>
<td>hallux vib r</td>
<td>units</td>
<td>6.73</td>
<td>3.01</td>
<td>15.50</td>
<td>3.50</td>
</tr>
<tr>
<td>ank dors l</td>
<td>degrees</td>
<td>14.41</td>
<td>3.91</td>
<td>21.00</td>
<td>9.00</td>
</tr>
<tr>
<td>knee @ 180</td>
<td>degrees</td>
<td>13.27</td>
<td>3.18</td>
<td>20.00</td>
<td>8.00</td>
</tr>
<tr>
<td>ank dors r</td>
<td>degrees</td>
<td>20.95</td>
<td>5.08</td>
<td>32.00</td>
<td>14.00</td>
</tr>
<tr>
<td>knee @ 90</td>
<td>degrees</td>
<td>19.05</td>
<td>5.73</td>
<td>32.00</td>
<td>8.00</td>
</tr>
<tr>
<td>ank plant l</td>
<td>degrees</td>
<td>43.73</td>
<td>5.76</td>
<td>58.00</td>
<td>33.00</td>
</tr>
<tr>
<td>ank plant r</td>
<td>degrees</td>
<td>40.73</td>
<td>5.60</td>
<td>54.00</td>
<td>30.00</td>
</tr>
</tbody>
</table>


Following the history and the clinical examination, the subjects changed into shorts, and retroreflective markers were placed on the skin over anatomical landmarks for the postural control tests (See section 3.10 Body Segment Model for the location of the markers and general link segment model description). The markers were small plastic spheres (figure 3.9.1) covered with 7610 Scotch® high gain retroreflective tape (3M Packaging Systems Division). The covered balls were hot glued to a small square of vinyl. The vinyl allowed the markers to be fixed to the subjects with cloth tape.
The subjects were then placed in a safety harness that permitted full freedom of movement yet prevented them from hitting the ground if they lost their balance and fell. Surface silver/silver chloride EMG electrodes (Therapeutics Unlimited, model #GCS67, IA) were placed on the lateral gastrocnemius. This, along with visual information from video cameras was used to ensure that the subjects maintained a standard relaxed position for the 10 s before perturbation and did not make any anticipatory postural adjustments prior to the motion of the platform. Although the platform did make noise when it rotated, there were no signals or other indicators prior to perturbation that cued the subjects when the platform was going to rotate. Thus, even though the subjects knew that the platform was going to move and after their first perturbation had a general idea of what the perturbation would be like, they did not know when each perturbation would be. The task was a simple reaction to a more or less known stimulus. Unfortunately, in an everyday situation, a perturbation would not be expected nor would its characteristics be known. A more realistic experiment would have been to allow each subject only one perturbation. This experimental design has the drawback of being very expensive with respect to subject resources for if just the first perturbation was used there would have only been five trials for each condition. Although subjects fell more often on the first perturbation especially if the first perturbation was condition one (full displacement/full power), the successful reactions were not noticeably different from the others. Close examination of the graphs of the individual trials shown in Appendix F shows that the first trial is not an outlier from the rest indicating that the first reaction does not have any unique characteristics.

Each trial started with the subject standing on a platform that was adjusted so that its rotational axis was co-linear with the ankle axis (figure 3.9.2). The subjects stood as still as possible with their arms folded across their torsos, and their feet positioned so that the heels were 0.27m apart and the mid-line of each foot was parallel to the sagittal plane (figure 3.9.3). Their knees and hips were fully extended, and their heads held erect. Their eyes were focused on a target directly in front of them at eye level. The target was part of a visual surround that enclosed the platform and provided a consistent visual environment for all trials of all subjects. The visual surround was composed of alternating dark gray and white vertical stripes, a pattern that has been shown to be of benefit in balance (Simoneau, 1992). As indicated before, the subjects held this position for at least 10 s before a sudden toe-up platform rotation occurred.
The dynamics of each subject’s response to four different conditions were recorded (table 3.9.1).

Table 3.9.1. Rotational conditions.

<table>
<thead>
<tr>
<th>Condition</th>
<th>distance (deg)</th>
<th>velocity (deg/sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>9.71</td>
<td>65.06</td>
</tr>
<tr>
<td>2</td>
<td>5.86</td>
<td>38.56</td>
</tr>
<tr>
<td>3</td>
<td>9.81</td>
<td>47.46</td>
</tr>
<tr>
<td>4</td>
<td>5.84</td>
<td>30.44</td>
</tr>
</tbody>
</table>

The platform perturbations were produced by a DC motor driven ball screw actuator (Industrial Devices Corporation, H Series #H105B, CA). A microprocessor-based controller (Industrial Devices H3851, CA) governed movement of the actuator. The rather odd combination of displacements and velocities was a result of the nature of the actuator and its programmable controller. A primary objective of the experiment was to have the subjects experience a variety of displacements distributed throughout the range of perturbations reported in literature (see Appendix E). However, while the angular displacement of the platform was directly controlled by altering the linear displacement of the actuator, the velocity and the acceleration of the platform perturbation could only be indirectly controlled through altering the “power” and time parameters of the controller. Therefore, the angular velocity of the platform was the velocity that the platform reached when the actuator was programmed to move at maximum power and at 50% power. Unfortunately, the velocity also depended on the distance that the actuator moved (table 3.9.1). Initially, the actuator was programmed to move the platform a distance of ten degrees and five degrees. However, when the platform moved five degrees the velocity of the platform was too slow to sufficiently perturb the subjects even at the highest power setting. Increasing the displacement to six degrees increased the platform’s velocity so that it was within the range of velocities seen in the literature. Note that the actual displacements achieved in the four conditions were slightly less than ten and six degrees. This discrepancy was due to the actuator moving the platform a smaller distance when it was loaded with a subject than when it was not loaded. The actuator was programmed to move five and ten degrees when unloaded resulting in a slightly smaller displacement when loaded with a subject. Although there was a difference between unloaded and loaded platform displacements, there were no significant differences in the platform displacements when subjects of different weights stood on the platform. That is, the platform displaced all subjects 9.71° in condition one, 5.86° in condition two, 9.81° in condition three and 5.84° in condition four. Similarly the peak velocity of rotation was constant regardless of the subject load. Because velocity
was controlled by the power setting of the actuator and there was an interaction between power and distance it was impossible to program the actuator to move the platform at a “clean” velocity such as 40, 50 or 60°/s, or to set up a testing block with only two displacements and two velocities.

Table 3.8.2 lists the programs used to generate the four conditions. As an example, the first program instructs the actuator to accelerate to the programmed power in 0.01 of a second (AC 0.01), decelerate from the programmed power in 0.01 of a second (DE 0.01), move at a velocity produced by 100% power (VE 100), tilt the platform 10° (DA 0.585), move according to the previous specifications (GO), pause for 10 s (TD 10), move at a velocity produced by 25% power, move the platform to horizontal (DA 0.641) and move according to the previous specifications (GO). Note that the distance of tilt had to be expressed in linear terms. The other programs had a similar structure.

Table 3.9.2. Actuator programs.

<table>
<thead>
<tr>
<th>condition 1</th>
<th>condition 2</th>
<th>condition 3</th>
<th>condition 4</th>
</tr>
</thead>
<tbody>
<tr>
<td>AC</td>
<td>0.01</td>
<td>0.01</td>
<td>0.01</td>
</tr>
<tr>
<td>DE</td>
<td>0.01</td>
<td>0.01</td>
<td>0.01</td>
</tr>
<tr>
<td>VE</td>
<td>100</td>
<td>100</td>
<td>50</td>
</tr>
<tr>
<td>DA</td>
<td>0.585</td>
<td>0.007</td>
<td>0.585</td>
</tr>
<tr>
<td>GO</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>TD</td>
<td>10</td>
<td>10</td>
<td>10</td>
</tr>
<tr>
<td>VE</td>
<td>25</td>
<td>25</td>
<td>25</td>
</tr>
<tr>
<td>DA</td>
<td>0.641</td>
<td>0.641</td>
<td>0.641</td>
</tr>
<tr>
<td>GO</td>
<td>0.641</td>
<td>- 0.641</td>
<td>0.641</td>
</tr>
</tbody>
</table>

AC: time of positive acceleration (s).  
DE: time of negative acceleration (s).  
VE: velocity, power of actuator (percentage).  
DA: end position (actuator coordinates).  
0.641 = horizontal platform.  
0.585 = 10° tilt when unloaded.  
0.007 = 6° tilt when unloaded.  
GO: move according to the above conditions.  
TD: time delay (s).

The order of conditions was randomly assigned to each subject. Each individual was presented with each condition ten times during the experiment. No instructions on how a subject was to react were given, except, that they were told to maintain posture as best they could without moving the arms or taking a step.

3.10. Body Segment Model.

For the dynamic analysis, the body was modeled as a series of five interconnected rigid segments. The model consisted of a head-neck segment, a trunk segment, which incorporated the upper arms, forearms and hands, a thigh segment, a shank segment and a foot segment (figure 3.10.1). The thigh, shank and foot segments combined the left and right sides. The joints included the neck, hips, knees and ankles.
Each segment was considered to be a rigid slender rod defined by a line connecting its proximal and distal endpoints. Each joint was modeled as a pin or hinge joint with one degree of rotational freedom and no translational freedom. A two dimensional sagittal plane analysis of the dynamics was justified by the assumption that rotations about the anteroposterior and longitudinal axes were negligible and even if they existed they would have little influence on balance in response to a rotational perturbation in the sagittal plane. The symmetry assumptions are based on the fact that inter-limb postural responses are seen when two limbs are used for support. In addition, postural responses are seen in the contralateral limb when only one limb is used for support in posture or gait (Dietz, 1992; Marsden, Morton and Morton, 1983). Unfortunately, this response may be the reason for many gun accidents. If the hand that is used for support is somehow perturbed, an automatic response in the opposite hand is elicited which if holding a gun may cause it to fire. An argument against this assumption would include the point that humans are not structurally symmetrical. The left and right sides of the body do not have the same length, mass, strength, flexibility etc... Most obvious are the asymmetries produced by the heart, lungs and intestines. In addition, people generally have a preferred side with which they perform tasks (i.e., left or right handed, and left or right foot preference and proficiency). Regardless of the lack of true symmetry, it can be argued that either side will provide a good representation of sagittal plane motion. Alexander, Shepard, Gu and Shultz (1992) noted in their posture work that there were no kinematic differences between the two sides in reactions to sagittal plane translational perturbations.

The location of the CM and the radius of gyration of each segment was calculated, based on the digitized location of retroreflective markers and from anthropometric data from Winter (1990), (table 3.10.1).
Table 3.10.1. Marker placement and body segment parameters.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Proximal marker</th>
<th>Distal marker</th>
<th>Seg. wt</th>
<th>CM/length proximal</th>
<th>Radius of gyration of seg. length</th>
</tr>
</thead>
<tbody>
<tr>
<td>Foot</td>
<td>Lateral malleolus*</td>
<td>5th metatarsal</td>
<td>0.0145</td>
<td>0.5</td>
<td>0.475</td>
</tr>
<tr>
<td>Shank</td>
<td>Femoral condyle</td>
<td>Lateral malleolus*</td>
<td>0.0465</td>
<td>0.433</td>
<td>0.302</td>
</tr>
<tr>
<td>Leg</td>
<td>Greater trochanter</td>
<td>Femoral condyle</td>
<td>0.100</td>
<td>0.433</td>
<td>0.323</td>
</tr>
<tr>
<td>Trunk**</td>
<td>Greater trochanter</td>
<td>C7 - T1</td>
<td>0.497</td>
<td>0.50</td>
<td>0.503</td>
</tr>
<tr>
<td>Upper arm</td>
<td>Glenohumeral joint</td>
<td>Olecranon</td>
<td>0.028</td>
<td>0.436</td>
<td>0.468</td>
</tr>
<tr>
<td>Forearm-and-hand</td>
<td>Olecranon</td>
<td>Ulnar styloid</td>
<td>0.022</td>
<td>0.682</td>
<td>0.468</td>
</tr>
<tr>
<td>Head and neck***</td>
<td>C7 - T1</td>
<td>Auditory meatus</td>
<td>0.081</td>
<td>1.00</td>
<td>0.495</td>
</tr>
</tbody>
</table>

* The location of the lateral malleolus was determined from data obtained in a static calibration.

** The location of the CM of the trunk was determined by a weighted sum of the trunk, upper arm, forearm-and-hand CMs. The trunk, upper arm, forearm-and-hand complex was considered one segment. Thus, any trials with arm movement were excluded.

*** A line drawn from the auditory meatus to the zygomatic arch determined the angular position of the head. The location of the CM, and the length of the radius of gyration were determined by the markers at C7 - T1 and at the auditory meatus.

The angular position of each segment, except the head, was defined to be the angle between the horizontal and the line passing through the segment’s proximal and distal markers. The angular position of the head was defined by the angle between the line passing though the marker at the auditory meatus and the marker at the zygomatic arch, and the horizontal. The joint angles were defined by the difference between the distal and proximal segmental angles. Angular velocities and accelerations, and the moments at each joint followed this convention. Thus, as viewed in the sagittal plane from the right side of the subject, a positive displacement, velocity, acceleration or torque was counter-clockwise and corresponded to neck extension, hip extension, knee flexion and ankle plantar flexion (see figure 3.10.2).
Figure 3.10.2. Angle definitions.

With this convention, the knee operates in a manner opposite to that of the ankle and the hip. Winter (1991) avoided this confusion by reversing the direction of positive rotation for the coordinate system at the knee. Thus, positive rotations are extensions at each joint and negative rotations are flexions at each joint. However, flipping the coordinate system at the knee joints creates as many problems as it solves. Although it assigns a positive direction to an action at a joint that would extend the leg, it creates problems in the anterioposterior direction. With Winter’s convention positive rotation at the ankle and the hip would move the CM of a standing individual posteriorly, while a positive rotation of the knee would move it anteriorly. With a system in which the rotational coordinates of the knee are not reversed, a positive rotation of any joint would move the body’s CM backwards and a negative rotation would move the CM forwards. Therefore, although balance does depend in part on the height of the body’s CM (or how extended the legs may be, as stressed by Winter’s system) a more important consideration is the horizontal relationship between the base of support and the CM. For this reason, Winter’s convention for reversing the knee coordinate system will not be utilized here.

The location of the lateral malleolus was blocked by the platform during the perturbation experiments. Thus, to determine its location, a standing calibration was performed with markers placed on the femoral condyle, tibial tuberosity and lateral malleolus (figure 3.10.3).

Figure 3.10.3. Calibration of lateral malleolus.

The position of the lateral malleolus marker was determined from the location of the femoral condyle and the tibial tuberosity in the following way.
\[ r_2 = \sqrt{(x_{lm} - x_{fc})^2 + (y_{lm} - y_{fc})^2} \]

where
- \( r_2 \) = vector from femoral condyle to lateral malleolus,
- \( x_{lm}, y_{lm} \) = x and y coordinates of the lateral malleolus and
- \( x_{fc}, y_{fc} \) = x and y coordinates of the femoral condyle

\[
\theta = \arctan(2(x_{lm} - x_{fc}, y_{lm} - y_{fc}) - \arctan(2(x_{tt} - x_{fc}, y_{tt} - y_{fc}))
\]

where
- \( \theta \) = angle between \( r_1 \) and \( r_2 \)
- \( r_1 \) = vector from femoral condyle to tibial tuberosity,
- \( x_{tt}, y_{tt} \) = x and y coordinates of the tibial tuberosity

\[
\phi = \arctan(2(x_{tt} - x_{fc}, y_{tt} - y_{fc}) + \theta
\]

where
- \( \phi \) = angle between \( r_2 \) and horizontal

\[
x_{lm} = x_{fc} + r_2 \cos \phi
\]

\[
y_{lm} = y_{fc} + r_2 \sin \phi
\]

Equation 3.10.1

3.11. Experimental Measurements.

The dynamic responses of the subjects to the different platform rotations were quantified by measurements and calculations of:

- **Maximum platform angular displacement and time to maximum angular displacement;**
  largest angular displacement from the starting angle of the platform and time from the start of the platform movement to the maximum platform displacement.
- **Platform movement after perturbation;**
  movement of platform after being displaced by the actuator due to subjects pushing down on it.
- **Maximum platform angular velocity and time to maximum angular velocity;**
  largest angular velocity of platform and time from the start of the platform movement to the maximum platform velocity.
- **Maximum and minimum platform angular accelerations and times to maximum and minimum angular accelerations;**
  maxima and minima of the angular acceleration of the platform and time from the start of the platform movement to the angular acceleration maxima and minima.
- **Platform jerk;**
  maximum platform acceleration divided by time to maximum platform acceleration.
- **Body CM displacement;**
  maximum body CM displacement, CM displacement after maximum, time to initial and maximum CM displacement.
- **Joint angular kinematics;**
  ankle, knee and hip angular displacements and time to initial and maximum angular displacements.

The velocities and accelerations were calculated from the video data by the central difference method. In addition, the following reactions were calculated at each of the joints.

- **Joint reaction forces;**
  anterior-posterior and superior-inferior ankle, knee and hip forces and time to peak joint forces.
- **Joint reaction torques;**
  ankle, knee and hip torques and time to peak joint torques.
- **Relation between joint torques and joint movements;**
  relationships between the net joint torque and the angular movement at a joint and the segment proximal to the joint.

The forces and moments were calculated, using the classic link segment method developed by Bresler and Frankel (1950). The joint forces were calculated from the head down as opposed to the traditional manner of foot up, because no ground reaction force data were available. Bobbert, Schamhardt and Nigg (1991) have performed similar calculations to successfully determine the ground reaction force.
from positional data. Additionally Alexander, Shepard, Gu and Schultz (1992) calculated the multi-body dynamics using both a “head-and-arms down” and a “foot up” approach. They found that the foot support torques calculated from the “head-and-arms down” method from ten randomly selected trials ranged from 3.3 to 10.1 Nm and had a mean error of 5.7 Nm. Thus, the forces on the head at the neck joint were equal to the mass of the head, times the acceleration of the CM of the head, minus the gravitational force on the head (mass of the head times the acceleration due to gravity):

\[ F_{\text{neck}} = m_{\text{head}} a_{\text{head}} - m_{\text{head}} g. \]  

Equation 3.11.1

The force on the trunk at the hip joint was equal to the masses of the arm, forearm-and-hand and trunk segments, times the acceleration of their respective CMs, minus the weight of each of these segments, plus the force on the trunk at the neck. Note because \( F_{\text{neck}} \) is the force on the head at the neck “the force on the trunk at the neck” is equal to \(-F_{\text{neck}}\). Similar conversions have to be accomplished in subsequent joints.

\[ F_{\text{hip}} = m_{\text{arm}} a_{\text{arm}} - m_{\text{arm}} g + m_{\text{hand}} a_{\text{hand}} - m_{\text{hand}} g + m_{\text{trunk}} a_{\text{trunk}} - m_{\text{trunk}} g + F_{\text{neck}}. \]  

Equation 3.11.2

The force on the thigh at the knee joint was equal to the mass of the thigh, times the acceleration of the thigh, minus the weight of the thigh and plus the thigh force at the hip

\[ F_{\text{knee}} = m_{\text{thigh}} a_{\text{thigh}} - m_{\text{thigh}} g + F_{\text{hip}}. \]  

Equation 3.11.3

The force on the shank at the ankle joint was found through a similar calculation:

\[ F_{\text{ankle}} = m_{\text{shank}} a_{\text{shank}} - m_{\text{shank}} g + F_{\text{knee}}. \]  

Equation 3.11.4

Once the joint forces were calculated, the net joint moments were calculated. Again the process started from the top of the body and continued down. The moment at the neck joint was equal to the rotational inertia of the head, times the angular acceleration of the head, minus the torque due to the force on the head at the neck acting through a lever arm, equal to the difference between the location of the neck joint center and the CM of the head. Again, the convention is that forces and moments at a joint are acting on the segment proximal to the joint. Thus \( +\tau_{\text{neck}} \) is the net joint torque acting on the head at the neck and \(-\tau_{\text{neck}} \) the net neck joint torque acting on the trunk.

\[ \tau_{\text{neck}} = I_{\text{head}} a_{\text{head}} + r_{\text{head-neck}} \times F_{\text{neck}}. \]  

Equation 3.11.5

The moment at the hip was then calculated using the torque calculated at the neck, plus the rotational inertia of the trunk and upper limbs, times the angular acceleration of the trunk and upper limbs, plus the torque due to the forces acting on the trunk at the neck and the hip:

\[ \tau_{\text{hip}} = I_{\text{trunk&upper limbs}} a_{\text{trunk&upper limbs}} + \tau_{\text{neck}} + r_{\text{neck-trunk}} \times F_{\text{neck}} + r_{\text{hip-trunk}} \times F_{\text{hip}}. \]  

Equation 3.11.6

Similar calculations were made to determine the torque at the knee and the ankle:

\[ \tau_{\text{knee}} = I_{\text{thigh}} a_{\text{thigh}} + \tau_{\text{hip}} + r_{\text{hip-thigh}} \times F_{\text{hip}} + r_{\text{knee-thigh}} \times F_{\text{knee}}. \]  

Equation 3.11.7

\[ \tau_{\text{ankle}} = I_{\text{shank}} a_{\text{shank}} + \tau_{\text{knee}} + r_{\text{knee-shank}} \times F_{\text{knee}} + r_{\text{ankle-shank}} \times F_{\text{ankle}}. \]  

Equation 3.11.8


Dynamic measurements of each subject were acquired through digitization of 60 Hz video of the sagittal plane. The digitization was performed on a Peak Performance™ data acquisition system installed on a 386 PC. The actual process of digitization will not be described here. However the interested reader can refer to Peak Performance™ manuals for details. To increase spatial resolution, two cameras were used to record the motion of the entire body (figures 3.8.3, 3.12.1 and 3.12.2). The view of one camera included the head, arm and trunk segments, while the second camera’s view concentrated on the lower extremities. The resolution was further increased by rotating the cameras so that the long axis of the rectangular field of view was vertical. The cameras were gen-locked so that time matching the data from each camera was not difficult. The gen-locking assured that the frames of video data from each camera were being exposed simultaneously. However, each videotape was digitized separately and thus, the digitized data had to be synchronized. Finding corresponding frames of video data were achieved by
locating a synchronization mark that was electronically placed in the corner of simultaneous frames of video data. Along with synchronization of the videotapes, combination of the data involved transforming digitized data from camera to global or lab coordinates. This transformation was performed in three steps: magnification, rotation and translation. Magnification was necessary to correct for the different lens magnifications of the two cameras as well as to transform the data from units of pixels to meters. Differences in x and y scaling (aspect ratio) was corrected for by Peak Performance™ before its output of data in pixels. The magnification factor was derived from the ratio between the distances between calibration markers on a calibration stick (that was placed in the same plane as the subject’s marker when the subject was standing on the platform) and the distance in pixels between these same markers as seen by each camera. The locations of the calibration markers were designated to have the global coordinates shown in table 3.12.1.

Table 3.12.1. Global coordinates of calibration markers (m).

<table>
<thead>
<tr>
<th>marker</th>
<th>x - coordinate</th>
<th>y - coordinate</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.000</td>
<td>1.258</td>
</tr>
<tr>
<td>2</td>
<td>-0.003</td>
<td>1.006</td>
</tr>
<tr>
<td>3</td>
<td>-0.001</td>
<td>0.755</td>
</tr>
<tr>
<td>4</td>
<td>-0.002</td>
<td>0.515</td>
</tr>
<tr>
<td>5</td>
<td>-0.010</td>
<td>0.257</td>
</tr>
<tr>
<td>6</td>
<td>0.000</td>
<td>0.000</td>
</tr>
</tbody>
</table>

Rotation was necessary because the cameras were tilted on their sides therefore, digitization coordinates were approximately 90° from the lab coordinates. In addition, rotation served to eliminate errors made in aligning the cameras before data collection. Alignment was achieved by rotating the cameras’ coordinate systems so that the calibration markers were as vertical as the markers in the global system. Translation was needed to place the top half of the body on the bottom half, or more generally, to align the camera and global coordinate systems so that the calibration markers had the same x- and y- coordinate values in each system.

The location of the lateral malleolus in each frame was also calculated at this time from the lateral malleolus calibration data and the bottom camera data.

Figure 3.12.1. Experimental setup rear view.
This “raw” positional data were then filtered in a low double pass zero phase shift Butterworth filter with an overall cutoff frequency of 6 Hz. The subject’s anthropometric measurements were input into the link segment model, and the location of the segmental and body CM and joint angles were calculated. This position and angle data were differentiated using a central difference algorithm to yield velocities and accelerations. This, in combination with the mass derived from the link segment model, was used to calculate the joint forces. The joint forces were used, along with the kinematic data and the inertial values from the link segment model, to calculate the joint moments. The linear and angular momenta of the segments and of the total body were also calculated. After calculating the kinematic and kinetic variables, the data were analyzed. Synchronization was necessary to compare the trials. This was achieved by noting the first frame in which the fifth metatarsal marker moved. The assignment of the beginning of a trial was confirmed by checking the velocity and acceleration calculations. Once the beginning frame of each trial was determined, average reaction curves were generated and analysis performed.

A representative curve of each subject’s reaction to a given perturbation condition was calculated by taking the average of all the reactions (e.g., joint angle, torque) at each frame. The general reaction to a specific condition was then determined by taking a frame by frame average of the subject’s representative curves. Therefore, first the trials of a subject were averaged together to form a representative curve for each subject’s reaction to a perturbation condition. These subject representative curves were then averaged (mean) together to form an experiment representative curve of the reaction to a perturbation. The event magnitudes (e.g. ankle maximum torque) were determined by taking the mean of the event magnitudes from each individuals representative curve. The variance in these values was expressed as the standard deviation of these values. Similarly, the time of each event (e.g., the time at which ankle maximum torque occurred) was determined by finding the mean of the event times from each individual’s representative curves, and the variance was the standard deviation of these event times.

As is inevitable with any statistical representation of data some information is lost in its calculation. In addition, the resulting statistical value that is calculated may not represent a value that was ever achieved or even achievable. For example, the average number of children in a certain population could be 2.5, a number of children that is not physically achievable. Another issue with statistics is that outliers have strong influences on results. This may be indirectly a concern with the method of data reduction used in this dissertation. The complication arises in the time averaging of data where the values are averaged at each frame. Unfortunately, any two curves may not be in synchronization with each other (thus the standard deviation in the events). Thus, a peak in one curve may occur in a frame in which another curve is decreasing. This lack of synchronization will tend to flatten and broaden the calculated representative curve. The result will be similar to that of a low pass filter. The variance in the timing of events will cause the representative curve to widen. The lack of synchronization will mean that at the peak time of one curve, there will be other curves with lower magnitude values. The more the various curves differ from one another, the greater effect these factors will have. Of course, if the curves were identical, there would be no need to average let alone perform multiple trials. As was true in the 2.5 children example, the resulting time averaged representative curve may not be physically possible.

Figure 3.13.1 summarizes the steps in the processing of data following digitization. Following calibration, the top and bottom camera (including the ankle calibration data) data were transformed into global or lab coordinates, and the location of the lateral malleolus was calculated. Then the data from the two cameras were synchronized and combined into a single file. The position data were then filtered and input to the line segment model. Also input to the link segment model were the anthropometric measurements of each subject. Using this model, the location of the CM and the joint angles were calculated at each frame of data. The velocity and acceleration of the various positions and angles were then computed using a central difference algorithm. The position, velocity, acceleration and inertial characteristics from the model were used to calculate the joint forces. The forces were in turn used in the calculation of the joint moments. The linear and angular momenta of the segments and of the body as a whole were calculated using the model parameters and the linear and angular velocities calculated in the central difference step.

Mean and variance were determined for each of the position, velocity acceleration, force and moment variables for each condition for each subject. The resulting magnitudes and timing of events are presented in the next chapter.

![Data processing flowchart](image_url)
Chapter 4
Results

4.1. Subject Description.
A total of 22 college students with ages ranging from 19 to 35 years old participated in this study. Half of the subjects were male and half were female. The subjects ranged from 1.53m to 1.86m in height and from 478.7 N to 772.1 N in weight. The subjects’ characteristics are listed in table 3.8.1. Blood pressure (BP) measurements were used to screen for diastolic and systolic drops when a subject stood from a seated position. Visual acuity was measured by the standard Snellen test, and the score reported was the size of the smallest line the subject could read at a distance of 20 feet. The subjects were tested with corrective eyewear, if needed, as the test was a screen for a subject’s ability to see and not a test of their innate visual system. No subject was below 20/25. A few subjects even possessed an uncorrected acuity of 20/10. The lateral malleolus height, the distance from the center of the right lateral malleolus to the floor, was measured and used to adjust the platform so that its axis of rotation coincided with the ankle joint. Each subject was screened for insensate feet by using a biothesiometer. Vibration scores were obtained for the left and right heel, fifth metatarsal head, first metatarsal head and hallux. Ankle dorsiflexion range was measured in degrees with the knee in two positions: full extension and flexed 90°. The ankle plantar flexion range was measured with the knee fully extended. The ankle range of motion measurements were performed to eliminate any subjects from the study whose ankle range of motion limited their ability to stand on the rotating platform. For full details of the methods of measurement, see Chapter 3.

4.2. Inertial Characteristics of the Subjects.
Table 4.2.1 lists the mass, in kilograms, of each segment of each subject as calculated by the model (Winter, 1990). The number in parentheses under each segment column title indicates whether a segment’s mass is a combination of both the left and right sides. For instance, the number (2) in the third column indicated that the foot segmental mass was the mass of both feet. The row labeled “factor” contains the numbers that were multiplied to the body mass to achieve the segmental masses. The “total” column contains the sum of the segmental masses, and the “difference” column demonstrates how the model accounted for 100% of the subjects’ mass by showing there was no difference between the body and the total columns.
Table 4.2.1. Segmental mass (kg).

<table>
<thead>
<tr>
<th>subject</th>
<th>body mass (kg)</th>
<th>feet (2)</th>
<th>shanks (2)</th>
<th>thighs (2)</th>
<th>trunk (1)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>69.3</td>
<td>2.01</td>
<td>6.44</td>
<td>13.86</td>
<td>34.44</td>
</tr>
<tr>
<td>2</td>
<td>77.7</td>
<td>2.25</td>
<td>7.23</td>
<td>15.54</td>
<td>38.62</td>
</tr>
<tr>
<td>3</td>
<td>59.4</td>
<td>1.72</td>
<td>5.52</td>
<td>11.88</td>
<td>29.52</td>
</tr>
<tr>
<td>4</td>
<td>73.5</td>
<td>2.13</td>
<td>6.84</td>
<td>14.70</td>
<td>36.53</td>
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<tr>
<td>5</td>
<td>60.4</td>
<td>1.75</td>
<td>5.62</td>
<td>12.08</td>
<td>30.02</td>
</tr>
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<td>1.74</td>
<td>5.57</td>
<td>11.98</td>
<td>29.77</td>
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<tr>
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<td>12.48</td>
<td>31.01</td>
</tr>
<tr>
<td>8</td>
<td>75.5</td>
<td>2.19</td>
<td>7.02</td>
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<td>37.52</td>
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<td>9</td>
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<td>33.75</td>
</tr>
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</tr>
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<td>14</td>
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<td>15.34</td>
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<td>7.32</td>
<td>15.74</td>
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<td>16</td>
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<td>2.12</td>
<td>6.81</td>
<td>14.64</td>
<td>36.38</td>
</tr>
<tr>
<td>17</td>
<td>76.4</td>
<td>2.22</td>
<td>7.11</td>
<td>15.28</td>
<td>37.97</td>
</tr>
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<td>1.95</td>
<td>6.25</td>
<td>13.44</td>
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<td>5.51</td>
<td>11.86</td>
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<td>2.12</td>
<td>6.81</td>
<td>14.64</td>
<td>36.38</td>
</tr>
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<td>57.5</td>
<td>1.67</td>
<td>5.35</td>
<td>11.50</td>
<td>28.58</td>
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<tr>
<td>22</td>
<td>56.5</td>
<td>1.64</td>
<td>5.25</td>
<td>11.30</td>
<td>28.08</td>
</tr>
</tbody>
</table>
Subject sixteen was unable to complete the perturbation testing because of dizziness. All information collected from this subject was potentially flawed and was not analyzed or included in the results.

Table 4.2.2 lists the moments of inertia of each segment about each segment’s CM in the sagittal plane in kg·m² as calculated by the model (see table 3.10.1 for body segment parameters) (Winter, 1990). The number in parentheses under each segment is based on the assumption in the model that the subjects acted symmetrically to the perturbations and that the left and right sides can be combined. For example, the number indicated that the foot moment of inertia is the moment of inertia of both feet. Because of the proportionally small mass and longitudinal axis of rotation of the fore-arm and hand segment, the segmental moment of inertia for the fore-arm and hand was assumed negligible.
Table 4.2.2. Segmental moment of inertia about each segment’s CM in the sagittal plane (kg·m²).

<table>
<thead>
<tr>
<th>subject</th>
<th>body mass (kg)</th>
<th>feet (2)</th>
<th>shanks (2)</th>
<th>legs (2)</th>
<th>trunk (1)</th>
<th>upper arms (2)</th>
<th>head and neck (1)</th>
</tr>
</thead>
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<tr>
<td>1</td>
<td>69.3</td>
<td>0.011</td>
<td>0.126</td>
<td>0.256</td>
<td>2.980</td>
<td>0.134</td>
<td>0.020</td>
</tr>
<tr>
<td>2</td>
<td>77.7</td>
<td>0.011</td>
<td>0.132</td>
<td>0.248</td>
<td>3.531</td>
<td>0.146</td>
<td>0.043</td>
</tr>
<tr>
<td>3</td>
<td>59.4</td>
<td>0.009</td>
<td>0.080</td>
<td>0.144</td>
<td>2.282</td>
<td>0.070</td>
<td>0.027</td>
</tr>
<tr>
<td>4</td>
<td>73.5</td>
<td>0.011</td>
<td>0.120</td>
<td>0.231</td>
<td>3.531</td>
<td>0.124</td>
<td>0.022</td>
</tr>
<tr>
<td>5</td>
<td>60.4</td>
<td>0.010</td>
<td>0.092</td>
<td>0.180</td>
<td>2.388</td>
<td>0.098</td>
<td>0.026</td>
</tr>
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<td>0.092</td>
<td>0.198</td>
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<td>0.149</td>
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<tr>
<td>8</td>
<td>75.5</td>
<td>0.011</td>
<td>0.138</td>
<td>0.311</td>
<td>2.990</td>
<td>0.122</td>
<td>0.030</td>
</tr>
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<td>0.008</td>
<td>0.076</td>
<td>0.156</td>
<td>2.420</td>
<td>0.091</td>
<td>0.025</td>
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<tr>
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<td>0.006</td>
<td>0.080</td>
<td>0.168</td>
<td>2.162</td>
<td>0.073</td>
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<td>0.080</td>
<td>0.164</td>
<td>2.103</td>
<td>0.081</td>
<td>0.019</td>
</tr>
<tr>
<td>12</td>
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<td>0.097</td>
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<td>2.389</td>
<td>0.078</td>
<td>0.022</td>
</tr>
<tr>
<td>13</td>
<td>48.8</td>
<td>0.006</td>
<td>0.056</td>
<td>0.103</td>
<td>1.628</td>
<td>0.050</td>
<td>0.013</td>
</tr>
<tr>
<td>14</td>
<td>76.7</td>
<td>0.011</td>
<td>0.117</td>
<td>0.219</td>
<td>3.455</td>
<td>0.133</td>
<td>0.031</td>
</tr>
<tr>
<td>15</td>
<td>78.7</td>
<td>0.012</td>
<td>0.136</td>
<td>0.263</td>
<td>3.358</td>
<td>0.119</td>
<td>0.038</td>
</tr>
<tr>
<td>16</td>
<td>73.2</td>
<td>0.010</td>
<td>0.122</td>
<td>0.263</td>
<td>3.165</td>
<td>0.122</td>
<td>0.022</td>
</tr>
<tr>
<td>17</td>
<td>76.4</td>
<td>0.014</td>
<td>0.116</td>
<td>0.251</td>
<td>3.002</td>
<td>0.117</td>
<td>0.031</td>
</tr>
<tr>
<td>18</td>
<td>67.2</td>
<td>0.010</td>
<td>0.100</td>
<td>0.173</td>
<td>2.654</td>
<td>0.094</td>
<td>0.031</td>
</tr>
<tr>
<td>19</td>
<td>59.3</td>
<td>0.008</td>
<td>0.088</td>
<td>0.198</td>
<td>2.380</td>
<td>0.096</td>
<td>0.017</td>
</tr>
<tr>
<td>20</td>
<td>73.2</td>
<td>0.010</td>
<td>0.112</td>
<td>0.218</td>
<td>3.568</td>
<td>0.073</td>
<td>0.035</td>
</tr>
<tr>
<td>21</td>
<td>57.5</td>
<td>0.008</td>
<td>0.066</td>
<td>0.171</td>
<td>2.070</td>
<td>0.073</td>
<td>0.013</td>
</tr>
<tr>
<td>22</td>
<td>56.5</td>
<td>0.007</td>
<td>0.072</td>
<td>0.175</td>
<td>1.687</td>
<td>0.070</td>
<td>0.019</td>
</tr>
</tbody>
</table>

4.3. Video Resolution.

To determine the expected resolution of the experiment, a pilot study was performed in which markers were placed on a steel I-beam. The arrangement replicated the marker placement on a 1.80m individual. The marked I-beam was placed in the location where the subjects would be in the experiment. One hundred fifty frames of video data were digitized and processed in the same manner that was used for the experimental data. The mass of the I-beam individual was arbitrarily set at 68.18 kg. Each marker generated circles consisting of 30 pixels in which the centroid was found. This led to a maximum standard deviation in position of 0.00m (i.e. less than 0.005m), in velocity of 0.00m/s (i.e., less than 0.005m/s), and in acceleration of 0.03m/s². Note that the angular acceleration variability was the largest due to error propagation in double differentiation. However, since the magnitude of position velocity and acceleration increase the standard deviations represent almost equivalent percentages of each value (e.g., 0.4 %, 0.83 % and 0.93 % of the range of angular position, angular velocity and angular acceleration of the platform respectively). The most variable angle was the head angle due to the close proximity of the markers on the zygomatic arch, and the auditory canal and the shoulder. The results of the pilot study are summarized in table 4.3.1. The force and moment resolutions may pose more of a problem. As will be reported shortly anterior and posterior joint forces typically had a magnitude of 10 N, therefore standard deviation represented approximately 10 % of this value. The 3.37 Nm moment resolution approached 25 % of the smallest joint torque range (12.87 Nm in the hip in condition four, see section 4.17).
Table 4.3.1. Video resolution.

<table>
<thead>
<tr>
<th></th>
<th>standard deviation</th>
<th>unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>position</td>
<td>0.00</td>
<td>m</td>
</tr>
<tr>
<td>linear velocity</td>
<td>0.00</td>
<td>m/s</td>
</tr>
<tr>
<td>linear acceleration</td>
<td>0.03</td>
<td>m/s²</td>
</tr>
<tr>
<td>angle</td>
<td>0.04</td>
<td>deg</td>
</tr>
<tr>
<td>angular velocity</td>
<td>0.54</td>
<td>deg/s</td>
</tr>
<tr>
<td>angular accel.</td>
<td>11.59</td>
<td>deg/s²</td>
</tr>
<tr>
<td>force</td>
<td>1.17</td>
<td>Newton</td>
</tr>
<tr>
<td>moment</td>
<td>3.37</td>
<td>Nm</td>
</tr>
</tbody>
</table>

4.4. Kinematics.

4.4.1. Platform Kinematics.

Because the reactions of the body depended on the motion of the platform, an accurate description of the kinematics of the platform motion is critical to the analysis of the subject’s reactions to perturbation. Figure 4.4.1.1 contains four graphs of the angular displacement of the platform. Each graph indicates the mean displacement (red line) and plus and minus two standard deviations (green and blue lines respectively) of all the trials of each condition. Toe-up tilt of the platform is indicated by increasing angles on the graph. The graph in the upper left corner is of condition one in which the actuator was set on maximum power and displacement. The graph in the upper right is of condition two, in which the actuator at maximum power drove the platform through only “half displacement”. The third graph in the lower left corner depicts condition three trials in which the platform was driven by the actuator running at 50 % power through the “full” displacement. The last graph shows the platform displacement for condition four trials. Here the actuator moved the platform half the displacement at 50 % power.
As can be seen in figure 4.4.1.1, the movement of the platform was not uniform within or between conditions. The lack of uniformity within a condition was due to the inability of the actuator to hold its final position. The actuator tightly regulated the position of the platform while it was moving it, as indicated by the small deviation of the lines in the initial phases of each graph. However, once the actuator moved the platform its preprogrammed displacement it did not lock it in that position. It was possible for the subjects to push the platform down. Unfortunately, the subjects did so to different degrees in each trial. The variability of the position of the platform following perturbation made comparison of the kinematics and kinetic reactions difficult after the platform reached its destination. Therefore, it was decided to limit analysis to the first 1.5 seconds after the perturbation.

Lack of uniformity between conditions was due to interactions between displacement, power setting, velocity and acceleration. Comparisons and contrasts between conditions are more clearly depicted in figure 4.4.1.2. In this figure, the mean displacement from each condition was plotted on the same graph. In condition one and condition two in which the power of the actuator was maximum, the platform rotated rapidly until just before its maximum displacement. The motion had one phase to it. In conditions three and four, the actuator power was set at 50 %, and the platform motion had two phases. The first phase was a rapid movement similar to that of conditions one and two. This lasted for approximately one half of the
total displacement. Following this initial rapid phase was a phase of slower rotation that lasted until the final displacement was reached.

Table 4.4.1.1 gives the mean maximum platform displacement of all trials, of all subjects, for each condition. The subjects were perturbed by approximately 9.7° in conditions one and three, and by approximately 5.9° in conditions two and four. The small standard deviations indicated that the actuator was consistent in its displacement of the platform, regardless of the load applied by the different subjects. The lack of difference between maximums in conditions one and three, and conditions two and four indicated that while the power setting influenced the velocity and acceleration it had no effect on the maximum displacement of the platform.

Table 4.4.1.1. Maximum platform displacement (deg) and time to maximum displacement (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>displ</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>9.7 (0.6)</td>
<td>0.37 (0.05)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>5.9 (0.3)</td>
<td>0.48 (0.06)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>9.8 (0.7)</td>
<td>1.20 (0.03)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>5.8 (0.4)</td>
<td>0.98 (0.02)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

Table 4.4.1.1 also lists the time to maximum platform displacement. Not surprisingly, it took longer for the platform to reach its maximum displacement in conditions three and four where the setting of
the actuator was at 50 % power, than it did in conditions one and two where the setting was full power. However, it was surprising to find that the platform took slightly longer to travel 3.8° less in condition two than in condition one, even though in both conditions the power was set at 100 %. This demonstrates the interaction between displacement and velocity of the platform movement. Presumably, accuracy of displacement was the highest priority. The velocity of movement was sacrificed when the actuator approached its programmed displacement so that it could precisely achieve its goal.

To investigate the amount that the subjects pushed the platform down following the toe-up perturbations, the difference between the maximum perturbation angle and the platform angle 3.3 s after the initiation of the perturbation was calculated. Table 4.4.1.2 indicates the mean post-perturbation displacement for each condition.

Table 4.4.1.2. Platform displacement after perturbation (deg).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>mean</th>
<th>std dev</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>5.2</td>
<td>1.8</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>1.7</td>
<td>1.5</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>6.6</td>
<td>1.9</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>2.6</td>
<td>1.4</td>
</tr>
</tbody>
</table>

The displacement in all conditions was significantly different from zero (p ≤ 0.05). In addition, the amount of movement following a perturbation was highly variable as indicated by the relatively large standard deviations (approximately 30 % to 90 % of the total movement), and the spread in the graphs in figure 4.4.1.1.

Figure 4.4.1.3 depicts the angular velocity (deg/s) of the platform in each of the four perturbation conditions. By convention, an upward rotation of the platform produced a positive angular velocity.
Figure 4.4.1.3. Platform angular velocity.
Larger velocity indicates a faster upward tilt.
Red = mean of all trials in a condition.
Green = mean + 2 standard deviations of all trials in a condition.
Blue = mean - 2 standard deviations of all trials in a condition.

Figure 4.4.1.4 combines graphs of the mean platform velocity in the four conditions. The graph clearly shows the one- and two-phase motions. In condition one and two, the full power conditions, the velocity peaks and then drops to 0°/s. In conditions three and four, the half power conditions, the angular velocity does not drop directly to zero. In these conditions, it remains at about 4.0°/s for 0.7 to 1 s before stopping. The platform motion due to the subjects pushing it down can be seen as a negative velocity beginning approximately 1.0 s after the initiation of the perturbation. Note that these negative velocities are never very large. Their magnitude is only a few deg/s.
Figure 4.4.1.4. Platform angular velocity

Larger velocity indicates a faster upward tilt.
Each line represents mean of all trials in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

Each of the four perturbation conditions provided a different maximum velocity, although the actuator was programmed at only two different power levels. As can be seen in figures 4.4.1.3 and 4.4.1.4, and in table 4.4.1.3, the maximum velocity for condition one was $65.1^{\circ}/s$, for condition three it was $47.5^{\circ}/s$, for condition two it was $38.6^{\circ}/s$ and for condition four it was $30.4^{\circ}/s$. As stated previously, the velocity achieved depended on the displacement of the platform as well as on the power setting. This was evident in the pilot study when the platform was programmed to move $5^{\circ}$ and very little velocity was achieved even at 100 % power. In spite of the difficulty in setting a specific actuator velocity, the velocity produced by the actuator was consistent. For example, in condition one, the standard deviation in the maximum velocity was only $5.7^{\circ}/s$ in spite of the different loads placed on it by the various subjects. Regardless of the maximum velocity or the displacement velocity interaction, the time it took for the platform to reach maximum velocity was fairly constant (mean 0.15 sec).

Table 4.4.1.3. Maximum platform angular velocity (deg/s) and time to maximum angular velocity.

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>velocity</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>65.1 (5.7)</td>
<td>0.16 (0.01)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>38.6 (2.8)</td>
<td>0.14 (0.01)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>47.5 (0.7)</td>
<td>0.15 (0.00)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>30.4 (3.1)</td>
<td>0.13 (0.01)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.
Figure 4.4.1.5 shows angular acceleration after the start of platform displacement in each of the four conditions. Figure 4.4.1.5 indicate that following a period of rapid toe-up positive acceleration and negative acceleration in the first half second, there were no significant accelerations. The magnitudes of the positive and negative accelerations were approximately equal in magnitude, with the negative accelerations tending to be slightly smaller.

Condition one      Condition two
full displacement/full power      half displacement/full power

Condition three     Condition four
full displacement/half power      half displacement/half power

Figure 4.4.1.5. Platform angular acceleration.
Larger acceleration indicates an upward tilt.
Red = mean of all trials in a condition.
Green = mean + 2 standard deviations of all trials in a condition.
Blue = mean - 2 standard deviations of all trials in a condition.

In figure 4.4.1.6, the mean platform angular acceleration for each of the four conditions is overlaid on the same graph. These means are calculated from the accelerations of all trials of a single condition. The magnitudes of the positive accelerations were slightly larger than the negative accelerations. Note that the peak positive accelerations occurred at relatively the same time after the initiation of perturbation while the peak negative accelerations did not. This should not be surprising as the platform had different distances to travel and traveled these distances at different velocities. The conditions in which the actuator moved the least reached their peak negative accelerations first. As can be seen in figure 4.4.1.6, peak negative acceleration occurred earlier in conditions two and four than in conditions one and three. Thus, the peak negative acceleration was independent of the type of displacement (one phase or two phase) that the platform underwent. Furthermore, the negative acceleration peak corresponded to the end of the first phase of motion in all four conditions. Therefore, from an acceleration as well as a velocity viewpoint, the
motion of the platforms in effect only had one phase of motion. Additionally there was no appreciable acceleration associated with the motion due to the subjects pushing the platform down following the perturbation. Also not the small negative accelerations at 0.93 s and 1.15 s indicating the cessation of upward movement in conditions 3 and 4.

![Platform angular acceleration graph](image)

Figure 4.4.1.6. Platform angular acceleration.
Larger acceleration indicates an upward tilt.
Each line represents mean of all trails in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

As shown in table 4.4.1.4, the maximum acceleration was different in each condition. Condition one perturbed the subjects with the greatest positive acceleration (771.0°/s²). Conditions two and three perturbed the subjects with a maximum acceleration of 528.9 and 571.5°/s² respectively. Condition four perturbed the subjects the least, with a maximum acceleration of 445.7°/s². As with the velocity, the acceleration of the platform was not directly controllable. The controller allowed only the length of the acceleration period to be specified in a program rather than the actual acceleration of the platform. Thus, the acceleration depended upon the maximum velocity of the platform, which depended on the distance the actuator traveled, and the power programmed into the actuator. All acceleration periods were programmed to last 0.01 s.
Table 4.4.1.4. Maximum angular acceleration (deg/s²) and time to maximum angular acceleration.

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>accel</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>771.0 (76.8)</td>
<td>0.08 (0.00)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>528.9 (36.8)</td>
<td>0.07 (0.01)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>571.5 (44.3)</td>
<td>0.07 (0.01)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>445.7 (56.4)</td>
<td>0.07 (0.00)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

The times to maximum angular acceleration were essentially the same although the maximum accelerations in the four conditions were not the same. Thus, the subjects were jerked (change in acceleration with time) more in conditions two and three (mean jerk = 7555.7 and 8164.3º/s³ respectively) than in condition four (mean jerk = 6367.1º/s³). However, the largest jerk was produced in condition one, in which it took 0.08 s for the platform to reach a maximum acceleration of 771.0º/s² (mean jerk = 9637.5º/s³). A summary of platform jerk by condition can be found in table 4.4.1.5.

Table 4.4.1.5. Platform jerk (deg/s³).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>jerk</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>9637.5</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>7555.7</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>8164.3</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>6367.1</td>
</tr>
</tbody>
</table>

As can be seen in table 4.4.1.6, the negative accelerations of the platform were as uncontrollable as the positive accelerations. However, at the low power settings there were essentially no differences between the negative accelerations. Again, acceleration was dependent on displacement and velocity of the platform, and was not under the direct control of the experimenter.

Table 4.4.1.6. Minimum angular acceleration (deg/s²) and time to minimum angular acceleration (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>accel</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-621.5 (70.5)</td>
<td>0.20 (0.01)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-375.5 (38.5)</td>
<td>0.16 (0.01)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-478.3 (54.7)</td>
<td>0.20 (0.01)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-397.1 (67.4)</td>
<td>0.16 (0.01)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

Surprisingly, time at which minimum angular acceleration occurred was the same for the conditions in which the maximum displacements were similar (table 4.4.1.6). One would expect the minimum angular acceleration to occur earlier in conditions with equal displacements but higher velocities. That is, the time to minimum acceleration was the same in conditions one and three, where the platform traveled roughly 9.76º, and the times to minimum acceleration were the same in conditions two and four, where the platform rotated approximately 5.9º. As stated previously, the time of minimum angular acceleration indicates the end of the first phase of movement. In most regards it can be considered to be the time in which the perturbation stopped, even though the platform continued to move for a considerable time after the minimum acceleration in conditions three and four, (which have a two phase movement). That is, the platform did continue to move in these conditions, however with negligible velocity and acceleration.
In summary, there were negligible differences in the dynamics of the platform perturbation within the trials in a specific condition, regardless of the differences in loads applied to the platform by each of the subjects. The only dynamic difference within a condition occurred during the period following the perturbation when the subjects pushed the platform down to different degrees. A summary of the platform characteristics is given in Table 4.4.1.7.

Table 4.4.1.7. Mean platform dynamic characteristics by condition.

<table>
<thead>
<tr>
<th>condition</th>
<th>distance</th>
<th>velocity</th>
<th>Pos accel</th>
<th>Neg. accel</th>
<th>jerk</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>9.7</td>
<td>65.1</td>
<td>771.0</td>
<td>-621.5</td>
<td>9637.5</td>
</tr>
<tr>
<td>2</td>
<td>5.9</td>
<td>38.6</td>
<td>528.9</td>
<td>-375.5</td>
<td>7555.7</td>
</tr>
<tr>
<td>3</td>
<td>9.8</td>
<td>47.5</td>
<td>571.5</td>
<td>-478.3</td>
<td>8164.3</td>
</tr>
<tr>
<td>4</td>
<td>5.8</td>
<td>30.4</td>
<td>445.7</td>
<td>-397.1</td>
<td>6367.1</td>
</tr>
</tbody>
</table>

Distance in degrees, velocity in \(\text{o/s}\), acceleration in \(\text{o/s}^2\), jerk in \(\text{o/s}^3\). Each line represents mean of all trials in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

The distance that was programmed into the platform resulted in a slightly smaller displacement when the subjects were on the platform than when the platform was unloaded. Despite this, the displacement was invariant irrespective of the load placed on the platform by the different subjects. This was true for velocity and acceleration as well. However, the velocity and the acceleration were dependent on the programmed velocity and displacement. Table 4.4.1.8 shows the displacement and velocity as programmed into the actuator, the actual peak velocity, positive acceleration, negative acceleration and jerk achieved by the platform. The conditions are arranged according to the magnitude of the actual peak velocity, positive acceleration, negative acceleration and jerk. Fortunately, the order of the conditions was the same. The order of the conditions (one, three, two and four) indicates the sensitivity of the perturbation. It also indicates that the displacement of the platform had a greater influence on the actual velocity, positive acceleration, negative acceleration and jerk than did the programmed power.

Table 4.4.1.8. Mean platform dynamic characteristics by magnitude.

<table>
<thead>
<tr>
<th>cond</th>
<th>prog displ</th>
<th>prog power</th>
<th>actual displ</th>
<th>vel</th>
<th>pos accel</th>
<th>neg accel</th>
<th>jerk</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>100</td>
<td>100</td>
<td>9.7</td>
<td>65.1</td>
<td>771.0</td>
<td>-621.5</td>
<td>9637.5</td>
</tr>
<tr>
<td>3</td>
<td>100</td>
<td>50</td>
<td>9.8</td>
<td>47.5</td>
<td>571.5</td>
<td>-478.3</td>
<td>8164.3</td>
</tr>
<tr>
<td>2</td>
<td>50</td>
<td>100</td>
<td>5.9</td>
<td>38.6</td>
<td>528.9</td>
<td>-375.5</td>
<td>7555.7</td>
</tr>
<tr>
<td>4</td>
<td>50</td>
<td>50</td>
<td>5.8</td>
<td>30.4</td>
<td>445.7</td>
<td>-397.1</td>
<td>6367.1</td>
</tr>
</tbody>
</table>

Prog displ = programmed displacement in percentage.
Prog power = programmed power in percentage.
Actual displ = actual displacement in degrees.
Vel = actual velocity in \(\text{o/s}\).
(pos/neg) accel = actual (positive/negative) acceleration in \(\text{o/s}^2\).
Jerk = \(\text{o/s}^3\).
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

It must also be noted that the differences between conditions were not equal. Specifically while the increase in velocity from condition four to condition two and from condition two to condition three was
roughly equal at 8.2 deg/s and 8.9 deg/s respectively, the jump from condition three to condition one was roughly twice as great at 17.6 deg/s. The jumps in positive and negative acceleration and the jumps in jerk had a similar pattern of progression in which the first jump (from condition four to condition two) was twice as large as the second jump (from condition two to condition three), and the last jump (from condition three to condition one) was the largest (table 4.4.1.9).

Table 4.4.1.9. Increase in platform dynamic characteristics.

<table>
<thead>
<tr>
<th>initial condition</th>
<th>final condition</th>
<th>velocity</th>
<th>positive accel</th>
<th>negative accel</th>
<th>jerk</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>2</td>
<td>8.2</td>
<td>83.2</td>
<td>21.6</td>
<td>1188.6</td>
</tr>
<tr>
<td>2</td>
<td>3</td>
<td>8.9</td>
<td>42.6</td>
<td>102.8</td>
<td>608.6</td>
</tr>
<tr>
<td>3</td>
<td>1</td>
<td>17.6</td>
<td>199.5</td>
<td>143.2</td>
<td>1473.2</td>
</tr>
</tbody>
</table>

Velocity in deg/s, acceleration in deg/s², jerk in deg/s³.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

Figure 4.4.1.7 and table 4.4.1.10 show that as the platform’s angular velocity increased, the subject’s joint angular displacements increased. Note that the ankle and the hip were affected more than the knee. While the ankle and the hip underwent a 4.3° and a 5.4° increase respectively, the knee was limited to a 1.2° increase with increased platform velocity. This was not surprising given the structural limits posed by the ligamentous structures such as the anterior cruciate, oblique popliteal and the collateral ligaments and the tendon from the long head of the gastrocnemius. These joint structures acted to limit the knee’s range of motion in extension, whereas the ankle and the hip were not as restricted. The hip, as its reaction is primarily one of flexion, would essentially be limitless in its motion, while the ankle would be limited to 12° to 15° of dorsiflexion in its response by the Achilles’ tendon (and triceps surae), the posterior deltoid ligament, the posterior talofibular ligament and the broader anterior portion of the trochlear surface meeting the ankle mortise. At this point, the ankle joint would begin to act like the knee utilizing passive joint structures to stop its motion. Thus, without these passive structures to limit knee range of motion in extension, increases in knee displacement would probably occur with increased perturbation velocity, as did other joint displacements.

Figure 4.4.1.7. Platform velocity and joint displacement.
Table 4.4.1.10. Platform velocity and joint displacement.

<table>
<thead>
<tr>
<th>condition</th>
<th>platform velocity</th>
<th>ankle displacement</th>
<th>knee displacement</th>
<th>hip displacement</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>30.4</td>
<td>3.40</td>
<td>1.86</td>
<td>4.13</td>
</tr>
<tr>
<td>2</td>
<td>38.6</td>
<td>4.09</td>
<td>2.51</td>
<td>5.60</td>
</tr>
<tr>
<td>3</td>
<td>47.5</td>
<td>5.94</td>
<td>2.78</td>
<td>7.92</td>
</tr>
<tr>
<td>1</td>
<td>65.1</td>
<td>7.68</td>
<td>3.06</td>
<td>9.49</td>
</tr>
</tbody>
</table>

Velocity in deg/s, displacement in deg.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

The time to peak of each of the characteristics is listed in Table 4.4.1.11. Table 4.4.1.12 lists the order of the conditions. The time order of the various dynamic characteristics was not as straightforward as the rank of the peak magnitudes of the dynamic characteristics. Given that the times of some peaks were not appreciably different, the order of the velocity and positive and negative accelerations can be adjusted, so that these characteristics occur first in condition two, second in condition four, third in condition three and fourth in condition one. This ordering was different from the order of the time of peak displacement, which was condition one, two, four and three.

Table 4.4.1.11. Time to peak platform dynamic characteristics.

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>velocity</th>
<th>positive accel</th>
<th>negative accel</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.37</td>
<td>0.16</td>
<td>0.08</td>
<td>0.20</td>
</tr>
<tr>
<td>2</td>
<td>0.48</td>
<td>0.14</td>
<td>0.07</td>
<td>0.16</td>
</tr>
<tr>
<td>3</td>
<td>1.20</td>
<td>0.15</td>
<td>0.07</td>
<td>0.20</td>
</tr>
<tr>
<td>4</td>
<td>0.98</td>
<td>0.13</td>
<td>0.07</td>
<td>0.16</td>
</tr>
</tbody>
</table>

Times are in seconds.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

Table 4.4.1.12. Order of peaks of platform dynamic characteristics by condition

<table>
<thead>
<tr>
<th>order</th>
<th>displ</th>
<th>velocity</th>
<th>positive accel</th>
<th>negative accel</th>
</tr>
</thead>
<tbody>
<tr>
<td>first</td>
<td>1</td>
<td>2</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>second</td>
<td>2</td>
<td>4</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>third</td>
<td>4</td>
<td>3</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>fourth</td>
<td>3</td>
<td>1</td>
<td>1</td>
<td>1</td>
</tr>
</tbody>
</table>

Conditions in which order is interchangeable are not separated by a line.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

4.4.2. Body Center of Mass Kinematics.

The motion of the body CM has been used in many studies in the past as an indicator of balance or posture. It is easily measured and is a good indicator of overall balance, however, it has significant limitations. For example, the motion of the individual segments can have a major impact on the maintenance of balance. This subject was covered in detail in the review of literature in Chapter 2.
Figure 4.4.2.1 shows the body CM displacement in reaction to the toe-up perturbations. Convention dictates that increasing values of the CM displacement indicate anterior displacement of the body’s CM while decreasing values indicate posterior displacement. As can be seen in figure 4.4.2.1, the body’s CM did return towards its initial position after perturbation, but the return was not complete. Determination of whether the CM returned to the pre-perturbation position was difficult because of the highly variable post-perturbation movement of the platform.

<table>
<thead>
<tr>
<th>Condition one</th>
<th>Condition two</th>
</tr>
</thead>
<tbody>
<tr>
<td>full displacement/full power</td>
<td>half displacement/full power</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>time (seconds)</th>
<th>displacement (meters)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.1</td>
<td>-0.06</td>
</tr>
<tr>
<td>0.2</td>
<td>-0.04</td>
</tr>
<tr>
<td>0.3</td>
<td>-0.02</td>
</tr>
<tr>
<td>0.4</td>
<td>0</td>
</tr>
<tr>
<td>0.5</td>
<td>0.02</td>
</tr>
<tr>
<td>...</td>
<td>...</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>time (seconds)</th>
<th>displacement (meters)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.1</td>
<td>-0.06</td>
</tr>
<tr>
<td>0.2</td>
<td>-0.04</td>
</tr>
<tr>
<td>0.3</td>
<td>-0.02</td>
</tr>
<tr>
<td>0.4</td>
<td>0</td>
</tr>
<tr>
<td>0.5</td>
<td>0.02</td>
</tr>
<tr>
<td>...</td>
<td>...</td>
</tr>
</tbody>
</table>

Figure 4.4.2.1. Body CM displacement (positive values indicate anterior displacement of the CM).
- Red = mean of all trials in a condition.
- Green = mean + 2 standard deviations of all trials in a condition.
- Blue = mean - 2 standard deviations of all trials in a condition.

Figure 4.4.2.2 gives a direct comparison of the effect of the different conditions on the displacement of the CM. As with most of the kinematics and kinetics, the CM’s reaction to condition one was the most pronounced. The effect on the subjects was a bit less pronounced in condition three while conditions two and four had the smallest effect.
Table 4.4.2.2. Body CM displacement (positive values indicate anterior displacement of the CM).
Each line represents mean of all trails in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

Figure 4.4.2.2. Body CM displacement (positive values indicate anterior displacement of the CM).
Each line represents mean of all trails in a condition.

Table 4.4.2.1 indicates the mean maximum of the CM displacement in each condition. If displacement of CM was the only measure, condition one provided the subjects with the strongest perturbation. Condition three was second, condition two was third and condition four was last.

There was no real difference between the conditions in the time it took for the CM to be maximally displaced from its original position. In all conditions, it took a mean of 0.81 s for the CM to reach its maximum displacement.

Table 4.4.2.2 indicates the degree to which the subjects returned to the position they held before perturbation. All CM displacements following perturbation were significantly different from zero (p ≤ 0.05). However, the subjects did not return all the way back to their original postures even in conditions of relatively low severity. Expressed as a percentage of perturbation distance, the CM returned 63 % of the way back in condition one, 68.1 % of the way in condition two, 42.3 % of the way in condition three and 58.8 % of the way in condition four. Part of the discrepancy between the initial and final CM positions may be due to the differences in initial and final platform positions. Nevertheless, the subjects did try to...
return to their original position as expressed by the post perturbation displacement of the subject’s CM and of the platform.

Table 4.4.2.2. Body CM displacement (m, final - maximum posterior displacement).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>mean</th>
<th>% return</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>0.021 (0.011)</td>
<td>63.6</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>0.015 (0.013)</td>
<td>68.1</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>0.011 (0.009)</td>
<td>42.3</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>0.010 (0.007)</td>
<td>58.8</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

There was no measurable difference in the times to initial CM displacement (table 4.4.2.3). A mean of 0.14 s passed between the beginning of platform motion and the initial CM displacement.

Table 4.4.2.3. Time to initial CM displacement (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>mean</th>
<th>std dev</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>0.14</td>
<td>0.02</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>0.14</td>
<td>0.02</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>0.15</td>
<td>0.02</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>0.15</td>
<td>0.02</td>
</tr>
</tbody>
</table>

In summary, the subjects reacted to the toe-up perturbation by shifting their CM posteriorly from 0.017 to 0.033m. The size of the CM motion did parallel the severity of the kinematics of the platform motion, however neither the timing of the CM displacement nor the degree to which the subjects returned their CM to its original location were related to the kinematics of the perturbation. Each subject’s CM only returned approximately half of the way back to its original position. There did not seem to be any reason based on platform kinematics to account for the differences in the amounts that the subjects returned their CM to its initial position following perturbation. Thus, there was no indication that the subjects were more careful about CM placement or about stability following a perturbation of greater severity than following one of less severity.

4.4.3. Ankle Angular Kinematics.

Figures 4.4.3.1 and 4.4.3.2 show the time course of the ankle angle in each of the four perturbation conditions. The ankle angle was defined to be the acute angle between the foot and the shank segments. Thus, dorsiflexion is indicated by a decrease in the ankle angle while plantar flexion is indicated by an increase in the ankle angle.

The initial reaction of the ankle to the platform toe-up rotation was dorsiflexion. This was followed immediately by a slight ankle plantar flexion. The ankle never did return to its pre-perturbation state. This was not surprising since the platform was at a different angle after the perturbation than it was prior to perturbation. However, unlike the other joints there was not even a tendency for the ankle to move toward its pre-perturbation configuration. Nor was there an increase in variance with time as was seen in the platform displacement curves. Thus, the act of driving the platform down after the perturbation was accomplished through a fixed ankle and not through ankle plantar flexion.
Figure 4.4.3.1. Ankle angular displacement (larger angles indicate plantar flexion).

Red = mean of all trials in a condition.
Green = mean + 2 standard deviations of all trials in a condition.
Blue = mean - 2 standard deviations of all trials in a condition.
Each line represents mean of all trails in a condition.

Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

There were differences in the time it took for the ankle to dorsiflex maximally (except between conditions two and three). It took the ankle the longest to dorsiflex maximally in condition one (0.25 s), even though this was the condition in which the platform reached the maximum tilt the quickest. It took 0.24 s for the ankle to dorsiflex maximally $7.7^\circ$ in condition two and $5.9^\circ$ in condition three. The time to maximum platform angle in condition two was 0.48 s, and in condition three it took 1.20 s. It took roughly
the same amount of time (0.24 s) for the ankle to reach maximum dorsiflexion in conditions two and three although the ankle displacement was different in these two conditions. In no condition did the ankle dorsiflex as much as the platform was rotated, nor did the peak platform displacement and the peak ankle displacement coincide in time in any condition. Thus, some of the platform rotation was transferred to the other segments. Table 4.4.3.2 compares the amount of absorption of the platform movement by the ankle with the platform peak displacement and the ankle peak displacement. The last column lists the time at which the ankle reached its peak dorsiflexion.

Table 4.4.3.2. Ankle kinematics.

<table>
<thead>
<tr>
<th>cond</th>
<th>displ</th>
<th>power</th>
<th>platform</th>
<th>ankle</th>
<th>absorption</th>
<th>peak time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>9.7</td>
<td>7.7</td>
<td>79.2 %</td>
<td>0.25</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>5.9</td>
<td>4.1</td>
<td>69.3 %</td>
<td>0.24</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>9.8</td>
<td>5.9</td>
<td>60.6 %</td>
<td>0.24</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>5.8</td>
<td>3.4</td>
<td>58.6 %</td>
<td>0.21</td>
</tr>
</tbody>
</table>

Platform = peak platform displacement in degrees.
Ankle = peak ankle angle in degrees.
Peak time = time of peak dorsiflexion in seconds.

The magnitude of the ankle displacement paralleled the severity of the perturbation characteristics. That is, the order of the magnitude of ankle dorsiflexion was: condition one, condition three, condition two and condition four. However, the amount of platform displacement absorbed by the ankle did not follow the severity of the perturbation. The greatest absorption occurred in condition one, which was the most dramatic perturbation. In condition two, the ankle absorbed more of the perturbation than it did in condition three, which was a more severe perturbation. There was only a small difference in the amount of ankle dorsiflexion and platform absorption between conditions three and four.

4.4.4. Knee Angular Kinematics.

Despite the contention in the literature that the knee plays only a minor role in posture mechanics, the present findings show that the knee does play a dynamic role in maintenance of balance in response to perturbation. Figures 4.4.4.1 and 4.4.4.2 graph the time course of the knee response to the platform’s rotation in the four conditions. The knee angle was defined such that full extension of the knee was zero degrees and any flexion of the knee would result in a positive angle. As expected, the subjects’ reaction was to more fully extend the knee in response to the platform movement. As can be seen in the first three conditions, the knee tended to quickly return toward its original orientation by about 0.5 s after the platform perturbation. Recall from section 4.4.1 that this was before the platform reached its maximum angle.


Condition one
full displacement/full power

Condition two
half displacement/full power

Condition three
full displacement/half power

Condition four
half displacement/half power

Figure 4.4.4.1. Knee angular displacement.

Red = mean of all trials in a condition.
Green = mean + 2 standard deviations of all trials in a condition.
Blue = mean - 2 standard deviations of all trials in a condition.
Each line represents mean of all trails in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

Table 4.4.4.1 shows that the amount of knee extension was small. This was not surprising since in relaxed stance the subjects’ knees were almost fully extended to the end of their range of motion and had little room to extend. There were no differences in the subjects’ reactions at the knees to the different perturbation conditions. Conditions one, two, three and four all perturbed the subjects’ knees by a mean of 2.6°. Also, note from the figures 4.4.4.1. and 4.4.4.2 and the table 4.4.4.1 that the timing of the knee motion was almost identical regardless of the condition.

Table 4.4.4.1. Knee angular displacement (degrees, maximum extension - initial angle, positive is flexion), time of beginning of knee displacement (s) and time to maximum extension (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>displ</th>
<th>beg. time</th>
<th>max time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-3.1 (2.6)</td>
<td>0.08 (0.02)</td>
<td>0.28 (0.06)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-2.5 (1.8)</td>
<td>0.07 (0.02)</td>
<td>0.30 (0.05)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-2.8 (1.8)</td>
<td>0.08 (0.02)</td>
<td>0.30 (0.04)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-1.9 (1.3)</td>
<td>0.07 (0.03)</td>
<td>0.36 (0.13)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

The knee was used less to absorb the larger perturbations. When the knee displacements were normalized with respect to the amount of platform displacement, the knee ranged between 31.6 % (condition three) and 49.3 % (condition four). This was in the reverse order of the magnitude of the different velocities or accelerations of the platform in the conditions of equal displacement. The knee began its rotation at the same time after the beginning of the perturbation regardless of the severity of the perturbation. The mean time to start of knee rotation was 0.08 s. However, there was a
difference in conditions in the time to maximum knee extension. It took 0.36 s for the knees to maximally flex in condition four, while it took 0.29 s in conditions one, two and three.

4.4.5. Hip Angular Kinematics.

Figures 4.4.5.1 and 4.4.5.2 show the initial 1.5 s of the angular position of the hip in the four different conditions. The hip angle was defined to be the angle between the thigh and trunk segments measured on the anterior side. Therefore, decreasing angles represent hip flexion, and increasing angles represent hip extension. Regardless of the condition, the general shape of the hip’s angular displacement was similar. Initially the hip extended a small amount before flexing. The flexion was not held constant as the hip drifted toward its original orientation by a gradual extension. As will be seen later, there was no difference between the conditions in the time of initiation of hip movement or in the time of maximum flexion. The only differences between conditions were the magnitudes of the hip’s displacement. Unlike the knee joint, the hip returned toward its original configuration much more slowly.

Condition one                  Condition two
full displacement/full power    half displacement/full power

Condition three                 Condition four
full displacement/half power    half displacement/half power

Figure 4.4.5.1. Hip angular displacement (positive indicates hip extension).
Red = mean of all trials in a condition.
Green = mean + 2 standard deviations of all trials in a condition.
Blue = mean - 2 standard deviations of all trials in a condition.
Differences were observed in the amount of hip movement with variation in platform maximum tilt, but not with variation in actuator power level (table 4.4.5.1). Thus, conditions one and three perturbed the subjects’ hips by $\sim 8.8^\circ$, and conditions two and four perturbed the subjects’ hips by $\sim 4.9^\circ$. Differences in the amount of hip movement in conditions one and three, or between two and four could not be found. Although differences in timing of peak displacement, velocity, acceleration and jerk existed between conditions one and three and between conditions two and four, none of these factors influenced the maximum displacement of the hip. When the hip displacements are normalized with respect to the amount of platform displacement, the hip displacement approached 90% of the platform displacement (89.34% in conditions one and three and 83.59% in conditions two and four).

Table 4.4.5.1. Hip angular displacement (degrees, maximum hip flexion - initial) and time (s) to start of hip angular displacement, and time of maximum hip angular displacement.

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>displ</th>
<th>start time</th>
<th>max time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-9.5 (5.6)</td>
<td>0.20 (0.04)</td>
<td>0.52 (0.07)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-5.6 (2.4)</td>
<td>0.19 (0.05)</td>
<td>0.57 (0.09)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-7.9 (4.2)</td>
<td>0.19 (0.05)</td>
<td>0.56 (0.12)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-4.1 (2.4)</td>
<td>0.18 (0.05)</td>
<td>0.58 (0.14)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

There was no difference between the conditions in the time to initial hip flexion, nor difference between the conditions in the time of maximum hip flexion. The mean time to initiation of hip flexion was 0.19 s, while the mean time to maximum hip flexion was 0.56 s.
4.4.6. Joint Angular Kinematics.

In condition one, the foot rotated 9.7° with the platform during perturbation (figure 4.4.6.1). This caused the ankle to dorsiflex, which limited shank counter-clockwise motion to 3.2°. The shank rotation did not begin until approximately 0.1 s after the start of platform rotation. The ankle stopped moving at 0.25 s following perturbation after dorsiflexing 7.7° from its pre-perturbation configuration. At this time, the foot and shank were both rotating in a counter-clockwise direction with the same velocity. Soon thereafter, the platform had reached its full displacement and had stopped rotating. Thus, the foot stopped rotating however the shank continued to rotate in the counter-clockwise direction. At this point, the ankle had reversed its motion and was plantar flexing. This plantar flexion was of a small magnitude (a bit more than 1°) and a long duration (until 0.62 s after the beginning of the platform perturbation or 0.37 s after peak ankle displacement). Finally, the shank slightly and slowly rotated clockwise while the foot remained stationary. During all of this, the thigh had started its movement in a clockwise direction 0.23 s after the perturbation. Following the 0.23 s, the thigh reversed direction and rotated in a counter-clockwise manner, while the shank was rotating counter-clockwise also extending the knee. Knee extension continued for 0.05 s after the thigh reversed its direction of rotation, because the shank was still rotating in the counter-clockwise direction with a higher velocity. From this point, the knee flexed as the thigh rotated counter-clockwise faster than the shank. At approximately 0.77 s after the perturbation, the shank and thigh stopped rotating and the knee stopped flexing (at 1.5° from stance). The thigh segment was the only segment to have a bi-phasic displacement pattern, causing the knee and hip to also have bi-phasic displacement curves. The hip initially extended; peaking at 0.20 s, which was slightly before the ankle reached its asymptote. This extension was slight (0.6°) and was due to the thigh rotating in a clockwise direction faster than the trunk. The trunk continued to rotate in a clockwise direction, while the thigh reversed its motion at 0.23 s. At this point, the thigh and trunk were rotating in opposite directions (thigh counter-clockwise, trunk clockwise) causing the hip to flex. The trunk clockwise rotation peaked after 0.48 s at 5.7° from its pre-perturbation attitude, while the hip continued to flex until 0.52 s due to continued thigh counter-clockwise rotation. The thigh continued its rotation until about 0.77 s. By this time, the hip was extending due to trunk counter-clockwise rotation.

Figure 4.4.6.1. Subject one’s reaction to condition one.
Horizontal position shifted with time.
Time between figures is 0.08 s.

The reactions to the other three perturbation conditions were similar in pattern. There were only slight changes in timing and magnitudes, the details of which have already been specified in their respective sections. Figure 4.4.6.2 shows the mean joint reactions and the platform displacement in condition one. Each line represents the mean reaction of all subjects in all trials of condition one. Each angle was initially set to zero to illustrate their displacement or deviation from their pre-perturbation values.
Figure 4.4.6.2. Joint motion and platform displacement from their initial positions. Each line represents the mean of all trials of all subjects in condition one. Condition one = full displacement/full power. Positive is:
- dorsiflexion of the platform
- plantar flexion of the ankle
- flexion of the knee
- extension of the hip.

When sorted by magnitude of displacement (table 4.4.6.1), the knee reacted the least, the ankle next and the hip the greatest, regardless of the perturbation condition. This was true even if the angular displacement of the joint was normalized with respect to the displacement of the platform. Therefore, within a specific condition the motion in response to perturbation was greatest at the hip, next at the ankle and least at the knee.
Table 4.4.6.1. Effect of platform magnitude on joint motion (sorted by condition, displacement and normalized displacement).

<table>
<thead>
<tr>
<th>condition</th>
<th>joint</th>
<th>platform displacement</th>
<th>joint displacement</th>
<th>normalized displacement</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>knee</td>
<td>9.7</td>
<td>-3.0</td>
<td>31.2 %</td>
</tr>
<tr>
<td>1</td>
<td>ankle</td>
<td>9.7</td>
<td>7.7</td>
<td>79.2 %</td>
</tr>
<tr>
<td>1</td>
<td>hip</td>
<td>9.7</td>
<td>-9.5</td>
<td>97.8 %</td>
</tr>
<tr>
<td>2</td>
<td>knee</td>
<td>5.9</td>
<td>-2.5</td>
<td>41.7 %</td>
</tr>
<tr>
<td>2</td>
<td>ankle</td>
<td>5.9</td>
<td>4.1</td>
<td>69.3 %</td>
</tr>
<tr>
<td>2</td>
<td>hip</td>
<td>5.9</td>
<td>-5.6</td>
<td>94.9 %</td>
</tr>
<tr>
<td>3</td>
<td>knee</td>
<td>9.8</td>
<td>-2.5</td>
<td>25.4 %</td>
</tr>
<tr>
<td>3</td>
<td>ankle</td>
<td>9.8</td>
<td>5.9</td>
<td>60.6 %</td>
</tr>
<tr>
<td>3</td>
<td>hip</td>
<td>9.8</td>
<td>-7.9</td>
<td>80.8 %</td>
</tr>
<tr>
<td>4</td>
<td>knee</td>
<td>5.8</td>
<td>-1.8</td>
<td>31.9 %</td>
</tr>
<tr>
<td>4</td>
<td>ankle</td>
<td>5.8</td>
<td>3.4</td>
<td>58.6 %</td>
</tr>
<tr>
<td>4</td>
<td>hip</td>
<td>5.8</td>
<td>-4.1</td>
<td>71.2 %</td>
</tr>
</tbody>
</table>

Displacement = peak angular displacement in degrees.
Normalized displacement = percentage of platform displacement.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

The timing of the rotations of the joints, and the platform and the translation of the CM in condition one is shown by expressing each as a percentage of their range of motion in figure 4.4.6.3. A definite sequence of motion exists. Although the displacements, velocities and accelerations of the platform varied between the conditions, the subjects’ reaction to a perturbation was of a constant order (table 4.4.6.2 and figure 4.4.6.4). The events started with the movement of the foot and progressed up the body; beginning with the start of the ankle rotation, continuing with the start of the knee rotation, initiation of CM displacement, start of the hip rotation, peak ankle displacement, peak knee displacement, peak hip displacement and finishing with the maximum CM displacement. The ankle and knee reached their respective peaks, before the majority of the hip and CM displacements occurred. The knee was the only joint that returned nearly to its pre-perturbation position within the first second. There was one exception to the constant event order. The maximum platform angle occurred before the minimum hip angle and low CM displacement in conditions one and two, and after the minimum hip angle and low CM displacement in conditions three and four. The order of events gave the reaction of the body a wave or whip like appearance. This leads one to believe that each segment’s response was a reaction to activity in the segment below and not a preemptive or proactive response.
Figure 4.4.6.3. Selected displacements plotted as a percentage of their range.
Each line represents the mean of all trials of all subjects in condition one.
Condition one = full displacement/full power.

Table 4.4.6.2. Order of important displacement events by condition.

<table>
<thead>
<tr>
<th>condition one</th>
<th>condition two</th>
<th>condition three</th>
<th>condition four</th>
</tr>
</thead>
<tbody>
<tr>
<td>full displacement</td>
<td>half displacement</td>
<td>full displacement</td>
<td>half displacement</td>
</tr>
<tr>
<td>full power</td>
<td>full power</td>
<td>half power</td>
<td>half power</td>
</tr>
<tr>
<td>start ank rot</td>
<td>start ank rot</td>
<td>start ank rot</td>
<td>start ank rot</td>
</tr>
<tr>
<td>start knee rot</td>
<td>start knee rot</td>
<td>start knee rot</td>
<td>start knee rot</td>
</tr>
<tr>
<td>init CM displ</td>
<td>init CM displ</td>
<td>init CM displ</td>
<td>init CM displ</td>
</tr>
<tr>
<td>start hip rot</td>
<td>start hip rot</td>
<td>start hip rot</td>
<td>start hip rot</td>
</tr>
<tr>
<td>min ank ang</td>
<td>min ank ang</td>
<td>min ank ang</td>
<td>min ank ang</td>
</tr>
<tr>
<td>min knee rot</td>
<td>min knee rot</td>
<td>min knee rot</td>
<td>min knee rot</td>
</tr>
<tr>
<td><strong>max plat ang</strong></td>
<td><strong>max plat ang</strong></td>
<td><strong>min hip ang</strong></td>
<td><strong>min hip ang</strong></td>
</tr>
<tr>
<td>min hip ang</td>
<td>min hip ang</td>
<td>low CM displ</td>
<td>low CM displ</td>
</tr>
<tr>
<td>low CM displ</td>
<td>low CM displ</td>
<td><strong>max plat ang</strong></td>
<td><strong>max plat ang</strong></td>
</tr>
</tbody>
</table>

As shown in figure 4.4.6.4 and table 4.4.6.3, even when all the conditions were combined, the order of rotation events paralleled their distance from the platform. The ankle started rotating in all conditions before the knee started rotating, which started rotating before the initial CM displacement, etc. The only movement that was not accomplished in a specific order was the maximum platform displacement. The maximum platform angle in conditions one and two occurred before the peak hip displacements and the peak CM displacements, which in turn preceded the maximum platform angle in conditions three and four.
Figure 4.4.6.4. Time of rotational landmarks (platform rotation begins at time zero).
Because the start of the ankle motion was equivalent to the start of perturbation, the time between the start of ankle motion and the start of knee motion, and the time between the start of ankle motion and the start of hip motion were equivalent to the time of initial knee motion and the time of initial hip motion respectively. These results were reported previously in sections 4.4.4 and 4.4.5. The tie between the initiation of knee rotation and initiation of hip rotation is shown in table 4.4.6.4. The differences between the time of initiation of the knee flexion and the time of initiation of hip flexion were not significantly different between conditions (p > 0.05, table 4.4.6.4). The mean time between the initiation of knee flexion and the initiation of hip flexion was 0.12 s. The difference was significantly different from zero (p ≤ 0.05) in all conditions.

<table>
<thead>
<tr>
<th>event</th>
<th>condition</th>
<th>time (s)</th>
<th>std dev (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>start knee rot</td>
<td>2</td>
<td>0.07</td>
<td>0.02</td>
</tr>
<tr>
<td>start knee rot</td>
<td>4</td>
<td>0.07</td>
<td>0.03</td>
</tr>
<tr>
<td>start knee rot</td>
<td>3</td>
<td>0.08</td>
<td>0.02</td>
</tr>
<tr>
<td>start knee rot</td>
<td>1</td>
<td>0.08</td>
<td>0.02</td>
</tr>
<tr>
<td>init CM displ</td>
<td>2</td>
<td>0.14</td>
<td>0.02</td>
</tr>
<tr>
<td>init CM displ</td>
<td>3</td>
<td>0.14</td>
<td>0.02</td>
</tr>
<tr>
<td>init CM displ</td>
<td>1</td>
<td>0.14</td>
<td>0.02</td>
</tr>
<tr>
<td>init CM displ</td>
<td>4</td>
<td>0.15</td>
<td>0.02</td>
</tr>
<tr>
<td>start hip rot</td>
<td>4</td>
<td>0.18</td>
<td>0.05</td>
</tr>
<tr>
<td>start hip rot</td>
<td>2</td>
<td>0.19</td>
<td>0.05</td>
</tr>
<tr>
<td>start hip rot</td>
<td>3</td>
<td>0.19</td>
<td>0.05</td>
</tr>
<tr>
<td>start hip rot</td>
<td>1</td>
<td>0.20</td>
<td>0.04</td>
</tr>
<tr>
<td>min ank ang</td>
<td>4</td>
<td>0.21</td>
<td>0.01</td>
</tr>
<tr>
<td>min ank ang</td>
<td>2</td>
<td>0.23</td>
<td>0.02</td>
</tr>
<tr>
<td>min ank ang</td>
<td>3</td>
<td>0.24</td>
<td>0.01</td>
</tr>
<tr>
<td>min ank ang</td>
<td>1</td>
<td>0.25</td>
<td>0.04</td>
</tr>
<tr>
<td>min knee rot</td>
<td>1</td>
<td>0.29</td>
<td>0.04</td>
</tr>
<tr>
<td>min knee rot</td>
<td>2</td>
<td>0.29</td>
<td>0.04</td>
</tr>
<tr>
<td>min knee rot</td>
<td>3</td>
<td>0.30</td>
<td>0.04</td>
</tr>
<tr>
<td>min knee rot</td>
<td>4</td>
<td>0.35</td>
<td>0.07</td>
</tr>
<tr>
<td>max plat ang</td>
<td>1</td>
<td>0.37</td>
<td>0.05</td>
</tr>
<tr>
<td>max plat ang</td>
<td>2</td>
<td>0.48</td>
<td>0.06</td>
</tr>
<tr>
<td>min hip ang</td>
<td>1</td>
<td>0.52</td>
<td>0.07</td>
</tr>
<tr>
<td>min hip ang</td>
<td>3</td>
<td>0.56</td>
<td>0.12</td>
</tr>
<tr>
<td>min hip ang</td>
<td>2</td>
<td>0.57</td>
<td>0.09</td>
</tr>
<tr>
<td>min hip ang</td>
<td>4</td>
<td>0.58</td>
<td>0.14</td>
</tr>
<tr>
<td>low CM displ</td>
<td>2</td>
<td>0.79</td>
<td>0.13</td>
</tr>
<tr>
<td>low CM displ</td>
<td>3</td>
<td>0.80</td>
<td>0.11</td>
</tr>
<tr>
<td>low CM displ</td>
<td>1</td>
<td>0.81</td>
<td>0.11</td>
</tr>
<tr>
<td>low CM displ</td>
<td>4</td>
<td>0.83</td>
<td>0.23</td>
</tr>
<tr>
<td>max plat ang</td>
<td>4</td>
<td>0.98</td>
<td>0.02</td>
</tr>
<tr>
<td>max plat ang</td>
<td>3</td>
<td>1.20</td>
<td>0.03</td>
</tr>
</tbody>
</table>

Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.
Table 4.4.6.4. Time between start of knee rotation and start of hip rotation (hip - knee).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ power</th>
<th>mean</th>
<th>std dev</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full full</td>
<td>0.12</td>
<td>0.04</td>
</tr>
<tr>
<td>2</td>
<td>half full</td>
<td>0.13</td>
<td>0.04</td>
</tr>
<tr>
<td>3</td>
<td>full half</td>
<td>0.12</td>
<td>0.04</td>
</tr>
<tr>
<td>4</td>
<td>half half</td>
<td>0.11</td>
<td>0.06</td>
</tr>
</tbody>
</table>

In summary, it took on average 0.00, 0.08 and 0.12 s for the ankle, knee and hip respectively to react to the platform perturbation. Each time of initiation for a joint was the same regardless of the perturbation condition. Therefore, the subjects initiated their reaction with identical timing to each perturbation regardless of the severity.

Table 4.4.6.5 summarizes the times of the peak rotational displacements of the platform and of each of the joints, during the four conditions. The platform peak time in each condition was significantly different from each other ($p \leq 0.05$). Despite this, the time of peak ankle displacement in conditions two and three were not significantly different, the time of peak knee displacement in conditions one, two and three were not significantly different and the time of peak hip displacement in all conditions were not significantly different. Thus, the timing of the hip and the knee were relatively insensitive to the differences in the conditions. Even the ankle joint showed some independence from the platform timing.

Table 4.4.6.5. Time to maximum displacement in seconds.

<table>
<thead>
<tr>
<th>condition</th>
<th>platform ankle knee hip</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.37 0.25 0.29 0.56</td>
</tr>
<tr>
<td>2</td>
<td>0.48 0.24 0.29 0.56</td>
</tr>
<tr>
<td>3</td>
<td>1.20 0.24 0.29 0.56</td>
</tr>
<tr>
<td>4</td>
<td>0.98 0.21 0.35 0.56</td>
</tr>
</tbody>
</table>

Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

In all four conditions, the ankle and the knee reached their greatest angular displacements before the platform achieved its maximum tilt as indicated by a negative value in table 4.4.6.6. Only the hip, in conditions one and two, reached its peak angular displacement after the platform fully tilted. All differences in all conditions were significantly different from zero ($p > 0.05$). Thus, the joint reactions, with the exception of the hip, were completed prior to the perturbation traveling its full course.

Table 4.4.6.6. Time between maximum platform displacement and maximum displacement of joints (s, joint – platform).

<table>
<thead>
<tr>
<th>condition</th>
<th>platform ankle knee hip</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.00 0.10 0.08 0.19</td>
</tr>
<tr>
<td>2</td>
<td>0.00 0.24 0.19 0.08</td>
</tr>
<tr>
<td>3</td>
<td>0.00 0.96 0.91 0.64</td>
</tr>
<tr>
<td>4</td>
<td>0.00 0.77 0.63 0.42</td>
</tr>
</tbody>
</table>

Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

In all conditions, the maximum knee extension and the maximum hip flexion followed the maximum dorsiflexion of the ankle (table 4.4.6.7). Positive values indicate that the knee and the hip reached maximum displacements after the ankle did in all conditions. Condition one was the only condition in which the time between the ankle minimum and the knee minimum was not significantly
different from zero ($p > 0.05$). All of the conditions had significantly different hip and ankle times of peak displacement ($p \leq 0.05$). There was no significant difference between conditions one, two and three in the time between maximum ankle displacement and maximum knee displacement ($p > 0.05$). The mean difference in these conditions was 0.04 s. The time between the ankle maximum displacement and the hip maximum displacement in conditions one and four were significantly different from each other ($p \leq 0.05$). However, none of the other pairs of conditions showed a significant difference between maximum hip displacement and maximum ankle displacement. The mean time between the ankle and hip maximum displacements was 0.32 s.

Table 4.4.6.7. Time between maximum ankle displacement and maximum displacements of the knee and hip (s, joint - ankle).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>ankle</th>
<th>knee</th>
<th>hip</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>0.00</td>
<td>0.02</td>
<td>0.29</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>0.00</td>
<td>0.05</td>
<td>0.32</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>0.00</td>
<td>0.05</td>
<td>0.32</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>0.00</td>
<td>0.14</td>
<td>0.35</td>
</tr>
</tbody>
</table>

The hip reached maximum flexion 0.24 s after the knee reached maximum flexion in all conditions (table 4.4.6.8). There was no significant difference between the conditions ($p > 0.05$). However, in all the conditions, the time between maximum knee and maximum hip displacements was significantly different from zero.

Table 4.4.6.8. Time between maximum knee displacement and maximum displacement of the hip (s, hip - knee).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>knee</th>
<th>hip</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>0.00</td>
<td>0.27</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>0.00</td>
<td>0.27</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>0.00</td>
<td>0.27</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>0.00</td>
<td>0.21</td>
</tr>
</tbody>
</table>

Consequently, after the ankle reached its peak displacement, the knee reached its peak 0.04 s later in the first three conditions and 0.15 s later in condition four, and the hip reached its peak displacement 0.32 s after the ankle peak. Thus, there was a wave like progression or structure of the body’s initial and peak reaction to the toe up perturbations.

4.5. Joint Forces.

4.5.1. Ankle Force.

4.5.1.1. Anterior-Posterior Ankle Force.

Figure 4.5.1.1.1 shows the net ankle force in the x, horizontal or anterior-posterior direction acting on the proximal or shank segment. A positive force was directed to the subject’s anterior while a negative force was directed to the subject’s posterior. Comparison of the four graphs showed that each condition resulted in very similar force time curves (also see figure 4.5.1.1.2). At first glance, it appeared that there was only a simple constant of proportionality separating the force curves of the four conditions, albeit a different constant for force than for time. However, comparison of the peaks in tables 4.5.1.1.1 and 4.5.1.1.2 did not support this contention.

In all conditions, the first force reaction at the ankle was to push the shank anteriorly. This force peaked at 6.80 N at 0.07 s and ended 0.10 s after the perturbation started. Following this brief anterior thrust, there
was a substantially larger and longer posterior thrust. The second peak occurred at approximately 0.17 s after the start of the perturbation regardless of the condition. However, the magnitude of the posterior drive varied with condition. Condition one peaked at 49.13 N. In condition four, the peak was a little more than half of the peak in condition one, while the peaks of conditions two and three were sandwiched in between (see table 4.5.1.1.1).

<table>
<thead>
<tr>
<th>Condition one</th>
<th>Condition two</th>
</tr>
</thead>
<tbody>
<tr>
<td>full displacement/full power</td>
<td>half displacement/full power</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>time (seconds)</th>
<th>force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.22</td>
<td>0.4</td>
</tr>
<tr>
<td>0.45</td>
<td>0.65</td>
</tr>
<tr>
<td>0.87</td>
<td>1.08</td>
</tr>
<tr>
<td>1.30</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Condition three</th>
<th>Condition four</th>
</tr>
</thead>
<tbody>
<tr>
<td>full displacement/half power</td>
<td>half displacement/half power</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>time (seconds)</th>
<th>force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.22</td>
<td>0.4</td>
</tr>
<tr>
<td>0.45</td>
<td>0.65</td>
</tr>
<tr>
<td>0.87</td>
<td>1.08</td>
</tr>
<tr>
<td>1.30</td>
<td></td>
</tr>
</tbody>
</table>

Figure 4.5.1.1.1. Anterior-posterior ankle force acting on the shank.

- Positive = force directed anteriorly.
- Negative = force directed posteriorly.
- Red = mean of all trials in a condition.
- Green = mean + 2 standard deviations of all trials in a condition.
- Blue = mean - 2 standard deviations of all trials in a condition.
Figure 4.5.1.1.2. Anterior-posterior ankle force acting on the shank.
Positive = force directed anteriorly.
Negative = force directed posteriorly.
Each line represents mean of all trails in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

Table 4.5.1.1.1. Maximum posterior ankle force (N) and time to maximum posterior ankle force (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>49.13 (16.19)</td>
<td>0.17 (0.02)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>36.48 (14.56)</td>
<td>0.17 (0.01)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>43.66 (14.55)</td>
<td>0.17 (0.01)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>28.12 (10.36)</td>
<td>0.17 (0.03)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

While the posterior peak force occurred at the same time after the initiation of perturbation in each condition, the conditions differed in the time at which the force switched directions. The order of reversal was opposite to the order of the magnitude of the peak posterior force. That is, condition four reversed first, condition two was next, condition three followed and condition one was last. In condition four, the force oscillated for approximately 0.10 s, before maintaining a constant positive or anterior direction at 0.38 s following perturbation initiation. The longer it took to reach the peak anterior force, the greater the magnitude. Thus, condition four peaked first, but the magnitude of the peak was the lowest, while the peak of condition one was the largest and the last (table 4.5.1.1.2). This was the inverse order of flipping. The anterior ankle force ceased in the same order as the peak anterior ankle force.
Table 4.5.1.1.2. Maximum anterior ankle force (N) and time to maximum anterior ankle force (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>16.15 (8.61)</td>
<td>0.55 (0.19)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>9.86 (5.92)</td>
<td>0.60 (0.64)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>12.71 (7.19)</td>
<td>0.52 (0.14)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>6.65 (10.36)</td>
<td>0.45 (0.20)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

4.5.1.2. Superior-Inferior Ankle Force.

The superior-inferior or y-component of the ankle force varied about a value equal to the weight of the subjects’ bodies minus the weight of the subjects’ feet. Because each subject’s weight was different, the results of the subjects’ reactions to a specific condition were expressed as deviations from the average pre-perturbation value. The average was used to eliminate any deviation present in the semi static, pre-perturbation stance of each individual.

Figures 4.5.1.2.1 and 4.5.1.2.2 show the time history of the deviation of the superior-inferior or y-component of the force acting on the distal end of the shank. Each graph represents an increase in force as a positive deflection and a decrease in force as a negative deflection. At all times, the force was directed up against the shank. A downward directed force would only exist if the force deviated negatively an amount equal to the weight of all the body segments above the ankle. Unlike the anterior-posterior force component at the ankle, the pattern of the superior-inferior component was not consistent over the four conditions. Conditions one and four had a positive, negative, positive, and no deviation pattern. In conditions two and three, the second positive peak was negligible and there was a second negative peak. In condition one and in condition three, the pattern took longer to evolve than it did in conditions two and four. The oscillation was spread out the greatest in condition one, next in condition three. Conditions two and four had similar time distributions. The relatively large variance in the y ankle force response in each of the conditions was due to the variance in the masses of the subjects.
Condition one  
full displacement/full power  

Condition two  
half displacement/full power  

Condition three  
full displacement/half power  

Condition four  
half displacement/half power  

Figure 4.5.1.2.1. Superior-inferior ankle force deviation (acting on the shank).
Red = mean of all trials in a condition.
Green = mean + 2 standard deviations of all trials in a condition.
Blue = mean - 2 standard deviations of all trials in a condition.
Differences in ankle superior-inferior force reaction did not manifest themselves until just prior to the first positive deviation peak. The differences between conditions began to emerge as conditions two, three and four reached their respective peaks at 0.08 s (condition two peaked at 22.1 N, condition three at 23.2 N and condition four at 20.3 N). Condition one reached its peak of 25.4 N ~0.02 s later.

From this point, conditions two and four followed essentially parallel paths until they separated just after their first negative peaks (condition two reached -13.1 N, condition four -14.7 N). At this time, condition one and condition three had just returned to their initial pre-perturbation values. Soon thereafter, conditions one and three reached their lowest values (~46.2 N and -21.4 N respectively) and conditions two and four were at a positive peak (2.7 N and 5.6 N respectively). In condition three, the second local maximum barely had a positive magnitude (0.2 N). After this point, most deviations from the pre-perturbation average were negligible and for most intents and purposes, all fluctuation ceased by 0.05 s after perturbation initiation regardless of the condition. The peaks and times to peaks of the ankle superior-inferior force deviation are summarized in tables 4.5.1.2.1 and 4.5.1.2.2.

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>25.4 (28.0)</td>
<td>0.10 (0.15)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>22.1 (12.6)</td>
<td>0.08 (0.11)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>23.2 (13.0)</td>
<td>0.08 (0.12)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>20.3 (8.0)</td>
<td>0.08 (0.08)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.
Table 4.5.1.2.2. Minimum deviation of the superior-inferior ankle force (N) and time to minimum superior-inferior force deviation (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-46.2 (36.6)</td>
<td>0.27 (0.06)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-13.1 (13.4)</td>
<td>0.18 (0.10)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-21.4 (18.1)</td>
<td>0.25 (0.08)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-14.7 (7.2)</td>
<td>0.17 (0.10)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

Subjects with larger mass will tend to have a greater influence on the ankle superior-inferior deviation curves than would subjects with smaller mass. An alternative method for combining the subject’s data would have been to first normalize the forces by dividing by the body weight minus the weight of the feet. The results of this process (figure 4.5.1.2.3) did not differ qualitatively from the previous method of using the deviation from the pre-perturbation force value. However, it was decided not to use the normalized data in the analysis because it was much easier to interpret units of Newtons than it was to interpret the unit less ratio of net joint force to (body - feet) weight. For the sake of consistency, the knee force was normalized with respect to body mass minus the mass of the shanks and feet, and the hip force was normalized by dividing by the body mass minus the masses of the thigh, shanks and feet. Neither of these differed quantitatively from their respective deviation curves and were not reported.

Figure 4.5.1.2.3. Anterior-posterior normalized ankle force acting on the shank (force/body - feet weight). Each line represents mean of all trails in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

4.5.2. Knee Force.
4.5.2.1. Anterior-Posterior Knee Force.

The net knee anterior-posterior force was reported as the horizontal component of the force acting on the thigh at the knee. As with the anterior-posterior ankle force component, a positive knee force was directed anteriorly while a negative knee force was directed posteriorly. Figure 4.5.2.1.1 shows the average and spread of the horizontal component of the net knee force in the four conditions separately, while figure 4.5.2.1.2 shows the average horizontal component of the net knee force in the four conditions together.

Figure 4.5.2.1.1. Anterior-posterior knee force (acting on the thigh).
Positive = force directed anteriorly. Negative = force directed posteriorly.
Red = mean of all trials in a condition.
Green = mean + 2 standard deviations of all trials in a condition.
Blue = mean - 2 standard deviations of all trials in a condition.
Figure 4.5.2.1.2. Anterior-posterior knee force (acting on the thigh).
Positive = force directed anteriorly.
Negative = force directed posteriorly.
Each line represents mean of all trials in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

The anterior-posterior force curves of the knee were very similar to those of the ankle (see section 4.5.1.1). The major difference between the force at the ankle and knee joints was that the anterior-posterior force at the knee had a slightly smaller magnitude than that at the ankle. There was very little difference between the timing of the ankle and knee inflection points. In condition four, the knee force did not oscillate direction, as did the anterior-posterior ankle force. Here it reached a plateau at -1.6 N just before switching direction. Additionally, the order of direction reversal was not the same at the knee as it was at the ankle. For the knee, the order was: condition one, three, two and four. The particulars of the magnitude and timing of the inflections are given in tables 4.5.2.1.1 and 4.5.2.1.2.

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-44.4 (11.6)</td>
<td>0.18 (0.02)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-31.9 (12.0)</td>
<td>0.17 (0.01)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-36.3 (10.0)</td>
<td>0.18 (0.02)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-24.9 (9.1)</td>
<td>0.17 (0.02)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.
Table 4.5.2.1.2. Peak anterior knee force (N) and time to knee anterior peak force (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>15.2 (8.5)</td>
<td>0.57 (0.18)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>9.0 (4.6)</td>
<td>0.47 (0.22)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>13.0 (7.2)</td>
<td>0.50 (0.16)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>6.3 (2.4)</td>
<td>0.43 (0.19)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

4.5.2.2. Superior-Inferior Knee Force.

As was done for the superior-inferior ankle force, the y-component of the force at the knee acting on the thigh was analyzed by calculating the time history of the deviation of the force from its pre-perturbation value. As before, a positive deviation indicated that a larger magnitude force was acting upward on the thigh, and a negative deviation indicated that a smaller magnitude force was acting upward on the thigh than was initially present (figures 4.5.2.2.1 and 4.5.2.2.2). At no time did the negative deviation have a larger magnitude than the weight of the body minus the weight of the shank and foot segments. Therefore, the superior-inferior component of the knee force was always directed up. As will be shown later, the pattern of knee deviation closely matched that of the deviation at the ankle. The magnitude at any point was only slightly less than that of the deviation at the ankle.
Figure 4.5.2.2.1. Superior-inferior knee force deviation (acting on the thigh).
Red = mean of all trials in a condition.
Green = mean + 2 standard deviations of all trials in a condition.
Blue = mean - 2 standard deviations of all trials in a condition.
Figure 4.5.2.2.2. Superior-inferior knee force deviation (acting on the thigh).
   Each line represents mean of all trails in a condition.
   Condition one = full displacement/full power.
   Condition two = half displacement/full power.
   Condition three = full displacement/half power.
   Condition four = half displacement/half power.

The tables 4.5.2.2.1 and 4.5.2.2.2 list the magnitudes and times of the inflection points.

Table 4.5.2.2.1. Maximum deviation of the superior-inferior knee force (N) and time to maximum deviation of the superior-inferior knee force (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>23.3 (25.6)</td>
<td>0.10 (0.16)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>21.2 (11.9)</td>
<td>0.08 (0.11)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>21.1 (11.7)</td>
<td>0.08 (0.11)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>18.8 (7.2)</td>
<td>0.08 (0.08)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

Table 4.5.2.2.2. Minimum deviation of the superior-inferior knee force (N) and time to minimum deviation of the superior-inferior knee force (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-42.1 (8.5)</td>
<td>0.28 (0.06)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-11.6 (4.6)</td>
<td>0.37 (0.09)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-18.7 (7.2)</td>
<td>0.25 (0.08)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-13.0 (2.4)</td>
<td>0.17 (0.10)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.
4.5.3. Hip Force.

4.5.3.1. Anterior-Posterior Hip Force.
As shown in figures 4.5.3.1.1 and 4.5.3.1.2 the anterior-posterior hip force followed a similar pattern to that of the ankle and the knee.

<table>
<thead>
<tr>
<th>Condition one</th>
<th>Condition two</th>
</tr>
</thead>
<tbody>
<tr>
<td>full displacement/full power</td>
<td>half displacement/full power</td>
</tr>
</tbody>
</table>

![Graphs showing hip force over time for different conditions](image)

Figure 4.5.3.1.1. Hip anterior-posterior force (acting on the trunk).
Positive = force directed anteriorly. Negative = force directed posteriorly.
Red = mean of all trials in a condition.
Green = mean + 2 standard deviations of all trials in a condition.
Blue = mean - 2 standard deviations of all trials in a condition.
Figure 4.5.3.1.2. Hip anterior-posterior force (acting on the trunk).
Positive = force directed anteriorly.
Negative = force directed posteriorly.
Each line represents mean of all trails in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

As with the horizontal component of the forces on the knee and ankle, a negative sign indicated that the force was directed posteriorly, while a positive sign indicated that it was directed anteriorly. The timing of the inflections and intercepts of the horizontal force component at the hip was almost identical to that at the ankle and at the knee while the peak magnitudes were slightly smaller. As with the anterior-posterior forces at the ankle and at the knee, the hip anterior posterior force began in the anterior direction. This was the briefest and lowest magnitude period. The peak was about 7 N and occurred at 0.07s after initiation of the platform perturbation. The transition from anterior to posterior force occurred at 0.10 s in each condition. At this point, the anterior-posterior hip force curves began to deviate from one another. The posterior force of condition four peaked first at 0.17 s at -15.0 N. Condition two was next at -18.4 N at 0.17 s. Condition three was third (-21.1 N, 0.18 s) and condition one was last (-27.5 N, 0.22 s). The results are summarized in tables 4.5.3.1.1 and 4.5.3.1.2.

Table 4.5.3.1.1. Peak posterior hip force (N) and time to peak posterior hip force (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-27.5 (7.7)</td>
<td>0.22 (0.05)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-18.4 (8.1)</td>
<td>0.17 (0.06)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-21.1 (5.4)</td>
<td>0.18 (0.05)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-15.0 (6.3)</td>
<td>0.17 (0.07)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.
Following the posterior peak, the hip force decreased in magnitude and eventually flipped direction. In all conditions, except condition one, this transition was smooth occurring at 0.42 s after the initiation of platform rotation. In condition one, the transition from posterior to anterior occurred first (0.33 s). However, the switch was only temporary as the force was switching back to a posterior direction at about the time that the hip forces in the other conditions were going from posterior to anterior. This second period of posteriorly directed hip force in condition one was very brief. At 0.47 s the hip force in condition one made its final transition to the anterior direction. As was true for the magnitude of the posterior force peak, condition one produced the largest anterior force peak. Condition three produced the second largest, condition two the third largest, while condition four was the smallest (table 4.5.3.1.2).

Table 4.5.3.1.2. Peak anterior hip force (N) and time to peak anterior hip force (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>10.2 (6.0)</td>
<td>0.58 (0.29)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>7.2 (3.1)</td>
<td>0.62 (0.26)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>9.2 (4.0)</td>
<td>0.65 (0.17)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>5.8 (3.0)</td>
<td>0.63 (0.32)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

4.5.3.2. Superior-Inferior Hip Force.

The deviation of the hip’s inferior superior force components is shown in figures 4.5.3.2.1 and 4.5.3.2.2. The pattern of deviation from the pre-perturbation force was similar to that of the ankle and the knee. The major distinguishing feature of the superior-inferior hip force was the smaller magnitude. As was true for the other joints, the negative deviation did not become greater than the pre-perturbation force, thus the hip force always had an upward or superior component to it.
Figure 4.5.3.2.1. Superior-inferior hip force deviation (acting on the trunk).

Red = mean of all trials in a condition.
Green = mean + 2 standard deviations of all trials in a condition.
Blue = mean - 2 standard deviations of all trials in a condition.
Figure 4.5.3.2.2. Superior-inferior hip force deviation (acting on the trunk).
Each line represents mean of all trails in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

The values and timing of the various peaks are given in tables 4.5.3.2.1 and 4.5.3.2.2.

Table 4.5.3.2.1. Peak inferior-superior hip force component maximum (N) and time to peak inferior-superior hip force component maximum (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>18.1 (20.2)</td>
<td>0.10 (0.15)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>16.3 (9.6)</td>
<td>0.08 (0.11)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>16.6 (9.2)</td>
<td>0.08 (0.11)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>15.0 (5.5)</td>
<td>0.08 (0.09)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

Table 4.5.3.2.2. Peak inferior-superior hip force component minimum (N) and time to peak inferior-superior hip force component minimum (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>force</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-32.1 (27.6)</td>
<td>0.28 (0.06)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-9.9 (10.2)</td>
<td>0.37 (0.10)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-12.8 (12.9)</td>
<td>0.25 (0.08)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-9.9 (5.5)</td>
<td>0.17 (0.10)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.
4.5.4. Relation Between Joint Force and Platform Perturbation Velocity.

4.5.4.1. Relation between Anterior-Posterior Joint Force and Platform Perturbation Velocity.

Figures 4.5.4.1.1, 4.5.4.1.2 and 4.5.4.1.3 along with tables 4.5.4.1.1, 4.5.4.1.2 and 4.5.4.1.3 show that the anterior-posterior component of the net joint force increased in magnitude with platform velocity. Note the posterior joint force is expressed as a negative value due to the convention established by the laboratory coordinate system. Therefore, an increasing posterior joint force was one that was increasingly negative. Unlike the joint displacement reaction to increasing platform perturbation velocity, in which, the knee had a different general reaction than did the ankle and the hip, the anterior-posterior joint force response to increasing platform perturbation velocity was similar at all joints. The maximum posterior, maximum anterior force and force range of each joint increased with platform perturbation velocity. The ankle had a range of 34.77 N at the low platform velocity of 30.40 deg/s and had a range of a little less than double that (65.28 N, an increase of 30.51 N) at the high platform velocity of 65.10 deg/s. This 88 % increase in anterior-posterior force range was mostly due to an increase in the magnitude of the posterior ankle joint force (21.01 N). There was only a 9.50 N increase in the anterior ankle joint force with increasing velocity. The knee range almost doubled too, from 31.13 N to 59.56 N (a 28.43 N increase). Again, the increase in range was primarily due to the increase in posterior knee force magnitude. The hip increased 16.91 N in its horizontal force range (an 81 % increase). Similarly, the force range was increased primarily by an increase in posterior hip force magnitude. Note that while the magnitude of the posterior joint force increase with velocity was greater than the magnitude of the anterior joint force increase, the percentage increase of the anterior ankle and knee joint forces were greater (143 % and 141 % respectively) when compared to posterior ankle and knee joint force percentage increases (75 % and 79 % respectively). That is the ankle and knee anterior joint force reactions were doubled with the increase in speed while the posterior reactions did not double with the increase in speed. The hip experienced similar percentage increases in its anterior (75 %) and posterior (83 %) joint force reactions. Also of interest is the effect that a joint’s distance from the platform has on the force-velocity relationship. Specifically the magnitude of the increase in joint reaction force whether it is the anterior force, posterior force or range of the joint force, decreases with increasingly superior joints. This effect was probably due to energy losses as the wave of platform perturbation traveled up the body. Each structure or each segment had the opportunity to absorb some of the energy of the propagating wave leaving less for the more superior structures and segments.
Figure 4.5.4.1.1. Platform velocity and maximum posterior joint force.

Table 4.5.4.1.1. Platform velocity and maximum posterior joint force.

<table>
<thead>
<tr>
<th>condition</th>
<th>platform velocity</th>
<th>ankle force</th>
<th>knee force</th>
<th>hip force</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>38.60</td>
<td>-36.48</td>
<td>-31.86</td>
<td>-18.37</td>
</tr>
<tr>
<td>3</td>
<td>47.50</td>
<td>-43.66</td>
<td>-36.27</td>
<td>-21.05</td>
</tr>
<tr>
<td>1</td>
<td>65.10</td>
<td>-49.13</td>
<td>-44.40</td>
<td>-27.54</td>
</tr>
</tbody>
</table>

Velocity in deg/s, force in Newtons.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

Figure 4.5.4.1.2. Platform velocity and maximum anterior joint force.
Table 4.5.4.1.2. Platform velocity and maximum anterior joint force.

<table>
<thead>
<tr>
<th>condition</th>
<th>platform velocity</th>
<th>ankle force</th>
<th>knee force</th>
<th>hip force</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>30.40</td>
<td>6.65</td>
<td>6.28</td>
<td>5.80</td>
</tr>
<tr>
<td>2</td>
<td>38.60</td>
<td>9.86</td>
<td>8.94</td>
<td>7.24</td>
</tr>
<tr>
<td>3</td>
<td>47.50</td>
<td>12.71</td>
<td>12.92</td>
<td>9.18</td>
</tr>
<tr>
<td>1</td>
<td>65.10</td>
<td>16.15</td>
<td>15.16</td>
<td>10.17</td>
</tr>
</tbody>
</table>

Velocity in deg/s, force in Newtons.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

Figure 4.5.4.1.3. Platform velocity and joint force range (difference between anterior and posterior force).

Table 4.5.4.1.3. Platform velocity and joint force range (difference between anterior and posterior force).

<table>
<thead>
<tr>
<th>condition</th>
<th>platform velocity</th>
<th>ankle force</th>
<th>knee force</th>
<th>hip force</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>30.40</td>
<td>34.77</td>
<td>31.13</td>
<td>20.81</td>
</tr>
<tr>
<td>2</td>
<td>38.60</td>
<td>46.34</td>
<td>40.80</td>
<td>25.61</td>
</tr>
<tr>
<td>3</td>
<td>47.50</td>
<td>56.37</td>
<td>49.19</td>
<td>30.23</td>
</tr>
<tr>
<td>1</td>
<td>65.10</td>
<td>65.28</td>
<td>59.56</td>
<td>37.71</td>
</tr>
</tbody>
</table>

Velocity in deg/s, force in Newtons.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

4.5.4.2. Relation between Superior-Inferior Joint Force and Platform Perturbation Velocity.

The relationship between the velocity of the platform perturbation and the superior-inferior joint force as given in tables 4.5.4.2.1, 4.5.4.2.2 and 4.5.4.2.3 was a bit more complicated than the relationship between the platform velocity and the horizontal component of the net joint forces. There was virtually no increase in the maximum superior-inferior joint force with increase in platform perturbation velocity especially when the velocity was increased from 38.60 deg/s to 47.50 deg/s (figure 4.5.4.2.1 and table 4.5.4.2.1). There was even a tendency for the maximum knee superior-inferior force to decrease in response to this particular increase in platform velocity. The changes in the minimum superior-inferior joint force reaction were more dramatic especially with higher velocities. The hip, knee and ankle reacted in almost parallel manner (figure 4.5.4.2.2 and table 4.5.4.2.2). Initially there was no change in, or only a slight tendency to decrease the magnitude of the minimum force, however, beginning with the second
velocity jump (from 38.60 deg/s to 47.50 deg/s) the magnitude increased and it increased to an even greater extent with the last velocity jump (from 47.50 deg/s to 65.10 deg/s). Because of the slight change in the maximum superior-inferior joint force, the joint force range response to increasing velocity was fundamentally a mirror of the change in the minimum joint force reaction.

![Figure 4.5.4.2.2. Platform velocity and maximum superior-inferior joint force. Red = ankle joint. Green = knee joint. Blue = hip joint.](image)

Table 4.5.4.2.2. Platform velocity and maximum superior-inferior joint force.

<table>
<thead>
<tr>
<th>condition</th>
<th>platform velocity</th>
<th>ankle</th>
<th>knee</th>
<th>hip</th>
</tr>
</thead>
<tbody>
<tr>
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<td>30.40</td>
<td>20.33</td>
<td>18.83</td>
<td>15.04</td>
</tr>
<tr>
<td>2</td>
<td>38.60</td>
<td>22.09</td>
<td>21.16</td>
<td>16.38</td>
</tr>
<tr>
<td>3</td>
<td>47.50</td>
<td>23.19</td>
<td>21.13</td>
<td>16.56</td>
</tr>
<tr>
<td>1</td>
<td>65.10</td>
<td>25.35</td>
<td>23.25</td>
<td>18.18</td>
</tr>
</tbody>
</table>

Velocity in deg/s, force in Newtons.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

![Figure 4.5.4.2.3. Platform velocity and minimum superior-inferior joint force. Red = ankle joint. Green = knee joint. Blue = hip joint.](image)
Table 4.5.4.2.3. Platform velocity and minimum superior-inferior joint force deviation by magnitude.

<table>
<thead>
<tr>
<th>condition</th>
<th>platform velocity</th>
<th>ankle</th>
<th>knee</th>
<th>hip</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>38.60</td>
<td>-13.07</td>
<td>-11.56</td>
<td>-9.94</td>
</tr>
<tr>
<td>3</td>
<td>47.50</td>
<td>-21.38</td>
<td>-18.70</td>
<td>-12.84</td>
</tr>
<tr>
<td>1</td>
<td>65.10</td>
<td>-46.16</td>
<td>-42.14</td>
<td>-32.10</td>
</tr>
</tbody>
</table>

Velocity in deg/s, force in Newtons.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

Figure 4.5.4.2.6. Platform velocity and superior-inferior joint force range.

Table 4.5.4.2.6. Platform velocity and superior-inferior joint force range.

<table>
<thead>
<tr>
<th>condition</th>
<th>platform velocity</th>
<th>ankle</th>
<th>knee</th>
<th>hip</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>30.40</td>
<td>34.99</td>
<td>31.85</td>
<td>24.96</td>
</tr>
<tr>
<td>2</td>
<td>38.60</td>
<td>35.16</td>
<td>32.72</td>
<td>26.32</td>
</tr>
<tr>
<td>3</td>
<td>47.50</td>
<td>44.57</td>
<td>39.83</td>
<td>29.40</td>
</tr>
<tr>
<td>1</td>
<td>65.10</td>
<td>71.51</td>
<td>65.39</td>
<td>50.28</td>
</tr>
</tbody>
</table>

Velocity in deg/s, force in Newtons.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

4.5.5. Joint Force Summary.

4.5.5.1. Anterior-Posterior Joint Forces.

Figure 4.5.5.1.1 is a composite of the ankle, knee and hip anterior-posterior force in each of the four conditions. Note that the shapes of each curve were almost identical, especially with respect to timing of inflection and transition points. The only distinguishing factors were the magnitude of the peak posterior force, and to a lesser extent the magnitude of the peak anterior force. In all conditions, the ankle peak forces were larger than the knee peak forces, which were greater than the hip peak forces. Thus, with respect to the joint anterior-posterior force, the body did not react like a whip with a wave of force traveling up it. Instead, it reacted more like a rigid bar with the force acting almost instantaneously at each joint.
However, the more superior the joint the lower the magnitude of the horizontal component of the joint force.

**Condition one**
- full displacement/full power

**Condition two**
- half displacement/full power

<table>
<thead>
<tr>
<th>time (seconds)</th>
<th>force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.22</td>
<td>0.43</td>
</tr>
<tr>
<td>0.65</td>
<td>0.87</td>
</tr>
<tr>
<td>1.08</td>
<td>1.30</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>time (seconds)</th>
<th>force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.22</td>
<td>0.43</td>
</tr>
<tr>
<td>0.65</td>
<td>0.87</td>
</tr>
<tr>
<td>1.08</td>
<td>1.30</td>
</tr>
</tbody>
</table>

**Condition three**
- full displacement/half power

**Condition four**
- half displacement/half power

<table>
<thead>
<tr>
<th>time (seconds)</th>
<th>force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.22</td>
<td>0.43</td>
</tr>
<tr>
<td>0.65</td>
<td>0.87</td>
</tr>
<tr>
<td>1.08</td>
<td>1.30</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>time (seconds)</th>
<th>force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.22</td>
<td>0.43</td>
</tr>
<tr>
<td>0.65</td>
<td>0.87</td>
</tr>
<tr>
<td>1.08</td>
<td>1.30</td>
</tr>
</tbody>
</table>

Figure 4.5.5.1.1. Anterior-posterior force (acting on the segment superior to joint. Positive = anterior force, negative = posterior force).

Each line represents mean of all trails in a condition.

### 4.5.5.2. Superior-Inferior Joint Force

As shown in figure 4.5.5.2.1 the hip, knee and ankle vertical force components were in almost perfect synchronization. The timing of each joint peak and intercept matched that of the other joints. The actual magnitudes of the deviation from pre-perturbation values were larger at the ankle than at the knee, and larger at the knee than at the hip because of the relatively larger mass that each successively inferior joint has to support. Interestingly the subjects responded differently in condition one than they did in the other three conditions. This was surprising since a similar difference did not exist in the anterior-posterior component of the joint force (section 4.5.5.1).
Figure 4.5.5.2.1. Superior-inferior force deviation acting on the segment superior to joint. Each line represents mean of all trails in a condition. Red = ankle force. Green = knee force. Blue = hip force.

4.6. Joint Torque.

4.6.1. Ankle Torque.

Due to symmetry, the torque generated at one ankle was half the value calculated and plotted in the graphs (figures 4.6.1.1 and 4.6.1.2). In addition, the torque shown is the half of the reaction pair at the ankle that acts on the distal end of the shank. To find the net torque acting on the foot, switch the sign or direction of the torque on the distal end of the shank. Positive moments indicate that the net moment acting at the ankle was in the plantar flexion direction. As can be seen in the graphs, each condition produced a similar reaction. Despite differences in the displacements, velocities and accelerations of the platform in the four conditions, the timing of the ankle torque was essentially identical. The magnitudes of the ankle torques, not the timing, differed between conditions. Initially, the net moment at the ankle was positive, a plantar flexion moment, (average 36.5 Nm, table 4.6.1.1) reflecting the body’s response to having both the knee joint and the body’s CM anterior to the axis of rotation of the ankle joint. Differences in initial values were due to slight differences in the initial stance of the subjects prior to platform perturbation, and were not significant (p > 0.05). Upon platform movement, the plantar flexion torque decreased by a maximum of 6.2 Nm in condition one to a maximum of 8.4 Nm in condition three in the first 0.08 s (table 4.6.1.2).
This allowed the ankle to dorsiflex. Thus, the foot rotated along with the platform while the shank and the rest of the body remained relatively stationary. Quickly the ankle moment increased to a peak of 55.7 to 71.3 Nm 0.1 s later (table 4.6.1.3). This represented an increase in plantar flexion torque from 17.7 Nm (minimum, condition four) to 32.8 Nm (maximum, condition one). During this period, the shank was driven backwards (along with the rest of the body). The torque following the perturbation tended to settle at a lower value than the torque prior to the perturbation. Initially this was to slow the posterior rotation of the shank (table 4.6.1.4). Once the person stopped moving, the ankle plantar flexion torque required to maintain stance was lower than it was prior to perturbation due to the posterior movement of the body during the perturbation (see body CM displacement section 4.4.2). The difference between the pre-perturbation ankle torque and the post-perturbation ankle torque minimum was similar for conditions with similar platform displacements. In conditions one and three where the perturbation was approximately 10°, the difference between the torques was 25.2 Nm on average. In conditions two and four where the platform was displaced only half as much, the average difference in ankle torque between pre-perturbation and post-perturbation minimum was only 16.6 Nm.

<table>
<thead>
<tr>
<th>Condition one</th>
<th>Condition two</th>
</tr>
</thead>
<tbody>
<tr>
<td>full displacement/full power</td>
<td>half displacement/full power</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Condition three</th>
<th>Condition four</th>
</tr>
</thead>
<tbody>
<tr>
<td>full displacement/half power</td>
<td>half displacement/half power</td>
</tr>
</tbody>
</table>

Figure 4.6.1.1. Ankle torque (positive is plantar flexion).  
Red = mean of all trials in a condition.  
Green = mean + 2 standard deviations of all trials in a condition.  
Blue = mean - 2 standard deviations of all trials in a condition.
Figure 4.6.1.2. Ankle torque (positive is plantar flexion).

Each line represents mean of all trails in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

Table 4.6.1.1. Ankle initial torque (Nm, positive is plantar flexion) and time to ankle initial torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>38.5 (10.1)</td>
<td>0.00</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>34.6 (5.2)</td>
<td>0.00</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>34.8 (8.1)</td>
<td>0.00</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>37.9 (17.2)</td>
<td>0.00</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

Table 4.6.1.2. Ankle initial minimum torque (Nm, positive is plantar flexion) and time to ankle initial minimum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>difference</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>32.3 (10.9)</td>
<td>-6.2</td>
<td>0.08 (0.02)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>27.4 (7.6)</td>
<td>-7.3</td>
<td>0.08 (0.02)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>26.5 (9.8)</td>
<td>-8.4</td>
<td>0.08 (0.02)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>31.2 (10.0)</td>
<td>-6.8</td>
<td>0.08 (0.02)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.
Difference = difference from initial ankle torque.
Table 4.6.1.3. Ankle maximum torque (Nm, positive is plantar flexion) and time to ankle maximum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>difference</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>71.3 (18.5)</td>
<td>32.8</td>
<td>0.18 (0.02)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>58.3 (12.6)</td>
<td>23.6</td>
<td>0.18 (0.02)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>62.0 (13.5)</td>
<td>27.2</td>
<td>0.18 (0.03)</td>
</tr>
<tr>
<td>4</td>
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<td>half</td>
<td>55.7 (13.3)</td>
<td>17.7</td>
<td>0.18 (0.01)</td>
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</tbody>
</table>

Standard deviations are in parentheses.
Difference = difference from initial ankle torque.

Table 4.6.1.4. Ankle minimum torque (Nm, positive is plantar flexion) and time to ankle minimum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>difference</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
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<td>-25.9</td>
<td>0.68 (0.20)</td>
</tr>
<tr>
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<td>half</td>
<td>full</td>
<td>14.9 (13.1)</td>
<td>-19.8</td>
<td>0.63 (0.20)</td>
</tr>
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<td>full</td>
<td>half</td>
<td>10.3 (13.7)</td>
<td>-24.5</td>
<td>0.70 (0.11)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>24.4 (9.8)</td>
<td>-13.5</td>
<td>-</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.
Difference = difference from initial ankle torque.

4.6.2. Knee Torque.

The knee torque was calculated using the same conventions that were used for the ankle torque. Specifically, the torque generated at one knee was half the value calculated and plotted in the graphs (figures 4.6.2.1 and 4.6.2.2). Also, the torque shown is the half of the reaction pair at the knee that acts on the distal end of the thigh. To find the net torque acting on the shank, switch the sign or direction of the torque on the distal end of the thigh. Positive moments indicate that the knee torque was attempting to flex the knee.

In the periods of stance before and just after platform perturbation, there was a meager extension torque at the knee. This is shown in the beginning and at the end of the graphs in figures 4.6.2.1 and 4.6.2.2 and in table 4.6.2.1. During these periods, the knee was not fully extended and a small torque was required to keep the joint from flexing under the weight of the body. The subjects’ first reaction following perturbation was to increase this extension torque (first dip in graphs, also see table 4.6.2.2). This, along with inertial effects drove the knee further into extension than existed in the relaxed stance (see knee displacement section 4.4.4). Following the increased extension there was a longer period of flexion moment. Unlike the initial increased extension, the magnitude of the flexion torque depended on the severity of the perturbation. The flexion torque may have been due to active contraction of the flexor muscles, inertial factors, or to the ligamentous structures at the end of the knee’s range of motion. The flexion torque was not a steady reaction as is clearly indicated by the dip in the curves at approximately 0.35 s following perturbation in condition one, 0.30 s in condition three and 0.23 s in the other two conditions (table 4.6.2.4). The first hump of the double humped flexion torque was always the larger of the two. As seen in table 4.6.2.3, condition one produced the largest absolute flexion torque as well as the largest deviation from the initial knee torque. Condition three produced the second largest flexion torque; however it did not produce any greater deviation from the initial knee torque than was produced by condition two. Condition four produced the smallest torque in both absolute and relative terms. Following this initial peak, there was a drop in torque of 2.6 to 7.1 Nm. Unlike most other kinematic and kinetic characteristics, condition three rather than condition four produced the smallest drop. Condition four almost drove the subjects to generate an extension moment at this point (peak of -0.3 Nm which was not significantly less than zero, p > 0.05). The second peak was smaller in magnitude than the first peak, although, except for condition one, the differences were negligible (table 4.6.2.5). With respect to the timing of the fluctuations of the knee moment, conditions two and four were essentially identical.
Condition one was similar to the others until after 0.20 s after the perturbation. At this time, the peaks arrived a bit later than they did in conditions two and four. Condition three similarly lagged behind conditions two and four but not as much as condition one did. As was true for the net torque at the ankle joint, the final torque at the knee joint was slightly different from that before the perturbation. The post-perturbation knee torque was slightly more extensive than it was pre-perturbation.

<table>
<thead>
<tr>
<th>Condition one</th>
<th>Condition two</th>
</tr>
</thead>
<tbody>
<tr>
<td>full displacement/full power</td>
<td>half displacement/full power</td>
</tr>
</tbody>
</table>

Figure 4.6.2.1. Knee torque (positive is flexion).
- Each line represents mean of all trials in a condition.
- Red = mean of all trials in a condition.
- Green = mean + 2 standard deviations of all trials in a condition.
- Blue = mean - 2 standard deviations of all trials in a condition.
Table 4.6.2.1. Knee initial torque (Nm, positive is flexion) and time to knee initial torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-1.3 (12.3)</td>
<td>0.00</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-4.8 (9.1)</td>
<td>0.00</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-3.1 (11.9)</td>
<td>0.00</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-5.1 (12.3)</td>
<td>0.00</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

Table 4.6.2.2. Knee initial minimum torque (Nm, positive is flexion) and time to knee initial minimum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>difference</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-6.5 (11.6)</td>
<td>-5.2</td>
<td>0.07 (0.04)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-10.0 (11.3)</td>
<td>-5.2</td>
<td>0.07 (0.06)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-8.9 (11.7)</td>
<td>-5.8</td>
<td>0.07 (0.04)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-11.1 (11.1)</td>
<td>-6.1</td>
<td>0.07 (0.04)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.
Difference = difference from initial knee torque.
Table 4.6.2.3. Knee maximum torque (Nm, positive is flexion) and time to knee maximum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>difference</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>14.3 (12.8)</td>
<td>15.6</td>
<td>0.25 (0.09)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>6.2 (11.2)</td>
<td>11.0</td>
<td>0.18 (0.08)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>8.8 (11.8)</td>
<td>11.9</td>
<td>0.18 (0.10)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>4.0 (13.1)</td>
<td>9.0</td>
<td>0.18 (0.08)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

Difference = difference from initial knee torque.

Table 4.6.2.4. Knee second minimum torque (Nm, positive is flexion) and time to knee second minimum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>diff</th>
<th>drop</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>7.2 (11.5)</td>
<td>8.5</td>
<td>7.1</td>
<td>0.35 (0.08)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>3.0 (13.4)</td>
<td>7.9</td>
<td>3.2</td>
<td>0.23 (0.08)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>6.2 (11.4)</td>
<td>9.3</td>
<td>2.6</td>
<td>0.30 (0.09)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-0.3 (12.3)</td>
<td>5.0</td>
<td>4.3</td>
<td>0.23 (0.06)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

Diff = difference from initial knee torque.

Drop = difference from maximum knee torque.

Table 4.6.2.5. Knee second maximum torque (Nm, positive is flexion) and time to second maximum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>difference</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>8.5 (11.5)</td>
<td>9.8</td>
<td>0.40 (0.07)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>5.8 (12.3)</td>
<td>10.6</td>
<td>0.33 (0.09)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>8.1 (12.2)</td>
<td>11.2</td>
<td>0.37 (0.08)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>3.7 (11.8)</td>
<td>8.8</td>
<td>0.33 (0.08)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

Difference = difference from initial knee torque.

4.6.3. Hip Torque.

The hip torque was calculated using the same conventions that were used at the ankle and the knee. The torque shown in figures 4.6.3.1 and 4.6.3.2 is double the torque at one hip and the half of the reaction pair that acts at the hip on the distal end of the trunk. Positive values indicate that the hip torque was acting to rotate the trunk in a counterclockwise direction or to extend the hip.

Figures 4.6.3.1 plot the hip torque reaction to the toe-up perturbation in each of the four conditions. Prior to perturbation, there was a small hip flexion moment (table 4.6.3.1). Upon toe-up rotation of the platform, the hip flexion moment varied about its initial value (tables 4.6.3.2, 4.6.3.3 and 4.6.3.4). The magnitude of the increase in flexion torque was approximately 3 Nm and the oscillation was of approximately equal magnitude (2.76 to 3.66 Nm). The size and timing of the oscillation was the same regardless of the condition. The oscillation was quickly followed by a decreased hip flexion moment, and in some conditions, a slight extension moment. The largest hip flexion moment occurred in condition one, followed by condition three. In condition one, the peak extension moment was approximately equal in magnitude to the initial flexion moment. Conditions two and four did not produce a hip flexion moment that was significantly greater than zero (p > 0.05) (table 4.6.3.5). Following the peak, the hip torque slowly returned toward its pre-perturbation flexion moment. As was true for the other joints, the initial hip condition was not achieved again.
Condition one
full displacement/full power

Condition two
half displacement/full power

Condition three
full displacement/half power

Condition four
half displacement/half power

Figure 4.6.3.1. Hip torque (positive is extension).
Red = mean of all trials in a condition.
Green = mean + 2 standard deviations of all trials in a condition.
Blue = mean - 2 standard deviations of all trials in a condition.
Figure 4.6.3.2. Hip torque (positive is extension).
Each line represents mean of all trails in a condition.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.
Table 4.6.3.1. Hip initial torque (Nm, positive is extension) and time to hip initial torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-12.3 (7.1)</td>
<td>0.00</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-13.2 (7.9)</td>
<td>0.00</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-11.3 (6.9)</td>
<td>0.00</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-12.6 (6.5)</td>
<td>0.00</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.

Table 4.6.3.2. Hip initial minimum torque (Nm, positive is extension) and time to hip initial minimum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>difference</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-15.3 (7.2)</td>
<td>-3.0</td>
<td>0.08 (0.02)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-15.9 (7.0)</td>
<td>-2.7</td>
<td>0.08 (0.02)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-14.4 (7.0)</td>
<td>-3.2</td>
<td>0.08 (0.02)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-16.0 (7.4)</td>
<td>-3.1</td>
<td>0.08 (0.03)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.
Difference = difference from initial hip torque.

Table 4.6.3.3. Hip first maximum torque (Nm, positive is extension) and time to hip first maximum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>difference</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-12.5 (7.1)</td>
<td>-0.2</td>
<td>0.15 (0.03)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-12.6 (7.0)</td>
<td>0.6</td>
<td>0.15 (0.03)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-11.0 (7.3)</td>
<td>0.3</td>
<td>0.15 (0.03)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-12.3 (7.5)</td>
<td>0.2</td>
<td>0.15 (0.03)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses.
Difference = difference from initial hip torque.

Table 4.6.3.4. Hip second minimum torque (Nm, positive is extension) and time to hip second minimum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>difference</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-15.6 (7.3)</td>
<td>-3.3</td>
<td>0.22 (0.03)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-15.4 (7.0)</td>
<td>-2.2</td>
<td>0.22 (0.03)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-13.9 (7.1)</td>
<td>-2.6</td>
<td>0.22 (0.02)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-14.4 (7.2)</td>
<td>-1.6</td>
<td>0.22 (0.03)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses. Difference = difference from initial hip torque.

Table 4.6.3.5. Hip second maximum torque (Nm, positive is extension) and time to hip second maximum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>difference</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>12.7 (13.8)</td>
<td>25.5</td>
<td>0.42 (0.05)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>1.4 (10.0)</td>
<td>14.6</td>
<td>0.38 (0.07)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>8.5 (11.9)</td>
<td>19.8</td>
<td>0.40 (0.06)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-3.1 (8.5)</td>
<td>9.5</td>
<td>0.38 (0.06)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses. Difference = difference from initial hip torque.
Table 4.6.3.6. Hip third minimum torque (Nm, positive is extension) and time to hip third minimum torque (s).

<table>
<thead>
<tr>
<th>condition</th>
<th>displ</th>
<th>power</th>
<th>torque</th>
<th>difference</th>
<th>time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>full</td>
<td>full</td>
<td>-6.4 (11.8)</td>
<td>5.9</td>
<td>0.82 (0.09)</td>
</tr>
<tr>
<td>2</td>
<td>half</td>
<td>full</td>
<td>-9.2 (12.3)</td>
<td>4.0</td>
<td>0.72 (0.05)</td>
</tr>
<tr>
<td>3</td>
<td>full</td>
<td>half</td>
<td>-6.1 (11.8)</td>
<td>5.1</td>
<td>0.70 (0.06)</td>
</tr>
<tr>
<td>4</td>
<td>half</td>
<td>half</td>
<td>-9.5 (11.8)</td>
<td>3.0</td>
<td>0.73 (0.05)</td>
</tr>
</tbody>
</table>

Standard deviations are in parentheses. Difference = difference from initial hip torque.

4.6.4. Joint Torques.

When analyzed in concert (figures 4.6.4.1 and 4.6.4.2), all three of the joint torques initially fell to their first minima and rose to their first maxima in parallel. However, because the knee joint operates in a manner opposite to that of the ankle and the hip, a drop in the knee moment indicated a tendency toward knee extension while a drop in the hip and ankle moments indicated a tendency toward hip flexion and ankle dorsiflexion respectively. Conversely, an increase in joint torque would tend to drive the knee into flexion, the hip into extension and the ankle into plantar flexion. However, as far as the anterior-posterior motion of the CM was concerned, the knee did not operate in opposition to the hip and the ankle. This meant that a positive moment at each joint would tend to move the body’s CM posteriorly, and a negative moment at each joint would tend to move the body’s CM anteriorly. At approximately 0.13 s following the start of the platform perturbation, the net joint torques had returned to their initial values. The hip peaked at this time while the ankle continued toward a stronger extension and the knee reversed and generated a net flexion torque. The hip moment continued oscillating until approximately 0.27 s. By this time, the ankle had reached its peak flexion torque and was well on its way back to its pre-perturbation value. The knee at this point had reached its peak flexion torque and was in the midst of a brief period of oscillation. The ankle reached and passed its initial value at approximately 0.30 s after the perturbation started. At this point, the moment at the knee was still in flexion while the hip moment was just reaching or about to become an extension moment. The ankle moment continued to decrease until it reached a minimum at a little more than 0.60 s after the trial started. At this time, both the knee and hip were within a 0.10 s of reaching their flexion and extension peaks respectively. The knee fell off first and returned to and continued beyond its initial torque value. The hip approached its initial torque but never quite got there. In general, the ankle was the first to change from its initial moment. The moment at the ankle also showed the greatest deviation from initial values. The hip was the last to react and showed the next greatest magnitude of change from initial values.
Figure 4.6.4.1. Ankle, knee and hip moments.
Each line represents mean of all trials in a condition.
Red = ankle moment (+ = plantar flexion moment).
Green = knee moment (+ = flexion moment).
Blue = hip moment (+ = extension moment).

The next series of graphs, (figure 4.6.4.2) shows how each net joint torque deviated from initial or pre-perturbation values.
Condition one
full displacement/full power

Condition two
half displacement/full power

Condition three
full displacement/half power

Condition four
half displacement/half power

Figure 4.6.4.2. Joint moments-deviation from initial.
Ankle moment (red, up = plantar flexion, down = dorsiflexion)
Knee moment (green, up = flexion, down = extension)
Hip moment (blue, up = extension, down = flexion).
Each line represents mean of all trails in a condition.

The range of torques required at the ankle was almost two times greater than that at the knee or the hip although there was no change in direction (the ankle always produced a plantar flexion moment, while the hip and knee produced both extension and flexion moments, table 4.6.4.1). The range of the hip was slightly larger than that of the knee in all conditions except for condition four where the knee range was larger than the hip by a few Nm.

Table 4.6.4.1. Range of joint torques by conditions.

<table>
<thead>
<tr>
<th>joint</th>
<th>condition one</th>
<th>condition two</th>
<th>condition three</th>
<th>condition four</th>
</tr>
</thead>
<tbody>
<tr>
<td>ankle</td>
<td>58.9</td>
<td>43.4</td>
<td>51.7</td>
<td>33.2</td>
</tr>
<tr>
<td>knee</td>
<td>26.8</td>
<td>16.2</td>
<td>17.7</td>
<td>15.1</td>
</tr>
<tr>
<td>hip</td>
<td>28.3</td>
<td>17.3</td>
<td>22.9</td>
<td>12.9</td>
</tr>
</tbody>
</table>

As can be seen in table 4.6.4.2 the ankle torques were greater than those at either the hip or the knee except in condition one. Only in condition one were the hip and knee maximum greater than the minimum ankle torque.
Table 4.6.4.2. Order of peak joint torque magnitudes within conditions.

<table>
<thead>
<tr>
<th>condition one</th>
<th>torque (Nm)</th>
<th>direction</th>
</tr>
</thead>
<tbody>
<tr>
<td>ankle high</td>
<td>71.3 (18.5)</td>
<td>plantar flexion</td>
</tr>
<tr>
<td>ankle init</td>
<td>38.5 (10.1)</td>
<td>plantar flexion</td>
</tr>
<tr>
<td>knee high</td>
<td>14.3 (12.8)</td>
<td>flexion</td>
</tr>
<tr>
<td>hip high</td>
<td>12.7 (13.8)</td>
<td>extension</td>
</tr>
<tr>
<td>ankle low</td>
<td>12.4 (13.8)</td>
<td>plantar flexion</td>
</tr>
<tr>
<td>knee init</td>
<td>-1.3 (12.3)</td>
<td>extension</td>
</tr>
<tr>
<td>knee low</td>
<td>-6.5 (11.6)</td>
<td>extension</td>
</tr>
<tr>
<td>hip init</td>
<td>-12.3 (7.1)</td>
<td>flexion</td>
</tr>
<tr>
<td>hip low</td>
<td>-15.6 (7.3)</td>
<td>flexion</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>condition two</th>
<th>torque (Nm)</th>
<th>direction</th>
</tr>
</thead>
<tbody>
<tr>
<td>ankle high</td>
<td>58.3 (12.6)</td>
<td>plantar flexion</td>
</tr>
<tr>
<td>ankle init</td>
<td>34.6 (5.3)</td>
<td>plantar flexion</td>
</tr>
<tr>
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<td>14.9 (13.0)</td>
<td>plantar flexion</td>
</tr>
<tr>
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</tr>
<tr>
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</tr>
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</tr>
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<td>-10.0 (11.3)</td>
<td>extension</td>
</tr>
<tr>
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<td>flexion</td>
</tr>
<tr>
<td>hip low</td>
<td>-15.9 (7.0)</td>
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<table>
<thead>
<tr>
<th>condition three</th>
<th>torque (Nm)</th>
<th>direction</th>
</tr>
</thead>
<tbody>
<tr>
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<td>62.0 (13.5)</td>
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</tr>
<tr>
<td>ankle init</td>
<td>34.8 (8.0)</td>
<td>plantar flexion</td>
</tr>
<tr>
<td>ankle low</td>
<td>10.3 (13.7)</td>
<td>plantar flexion</td>
</tr>
<tr>
<td>knee high</td>
<td>8.8 (11.8)</td>
<td>flexion</td>
</tr>
<tr>
<td>hip high</td>
<td>8.5 (11.9)</td>
<td>extension</td>
</tr>
<tr>
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<td>-3.1 (11.9)</td>
<td>extension</td>
</tr>
<tr>
<td>knee low</td>
<td>-8.9 (11.7)</td>
<td>extension</td>
</tr>
<tr>
<td>hip init</td>
<td>-11.3 (6.9)</td>
<td>flexion</td>
</tr>
<tr>
<td>hip low</td>
<td>-14.4 (7.1)</td>
<td>flexion</td>
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<table>
<thead>
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<th>direction</th>
</tr>
</thead>
<tbody>
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</tr>
<tr>
<td>ankle init</td>
<td>37.9 (17.2)</td>
<td>plantar flexion</td>
</tr>
<tr>
<td>ankle low</td>
<td>24.4 (9.8)</td>
<td>plantar flexion</td>
</tr>
<tr>
<td>knee high</td>
<td>4.0 (13.1)</td>
<td>flexion</td>
</tr>
<tr>
<td>hip high</td>
<td>-3.1 (8.5)</td>
<td>flexion</td>
</tr>
<tr>
<td>knee init</td>
<td>-5.1 (12.3)</td>
<td>extension</td>
</tr>
<tr>
<td>knee low</td>
<td>-11.1 (11.1)</td>
<td>extension</td>
</tr>
<tr>
<td>hip init</td>
<td>-12.6 (6.5)</td>
<td>flexion</td>
</tr>
<tr>
<td>hip low</td>
<td>-16.0 (7.4)</td>
<td>flexion</td>
</tr>
</tbody>
</table>
4.6.5. Relation between Joint Torque and Platform Rotational Velocity.

Various characteristics of joint torque responses to increasing platform perturbation are graphed and listed in figures 4.6.5.1, 4.6.5.2 and 4.6.5.3 and tables 4.6.5.1, 4.6.5.2 and 4.6.5.3. The general net torque reaction was to increase with increasing platform velocity except at the knee, where there was no increase. The largest increase in any knee torque characteristic was the 6.5 Nm increase in the knee torque maximum, however since the initial knee torque varied by 4.4 Nm this increase was not significant. There were other exceptions too. Specifically, the initial torque of every joint (red line) did not vary with platform velocity increases, nor was there a change in the first minimum torque of any of the joints. There were increases in the joint torque reactions with velocity.

In general, there was an equal increase in the magnitude of the minimum and maximum ankle torques. The ankle’s maximum increased by 15.0 Nm, while the minimum decreased by 12.4 Nm making the increase in the range with which the ankle had to respond increase by 27.4 Nm with the increase in speed. This was an 87.80 % increase in the reaction.

![Figure 4.6.5.1. Platform velocity and ankle torque.](image)

Red = initial torque. Green = first minimum torque.
Blue = maximum torque. Yellow = minimum torque.
Pink = torque range.

<table>
<thead>
<tr>
<th>condition</th>
<th>platform velocity</th>
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<th>first min</th>
<th>max</th>
</tr>
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</tr>
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Table 4.6.5.1. (continued) Platform velocity and ankle torque.

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</table>

Velocity in deg/s, torque in Nm.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

As stated previously, there was little change in the knee torque minimums, maximums or range of response to increases in platform velocity. The initial pre-perturbation knee torque varied by 4.4 Nm, while the first minimum increased only 0.8 Nm, the first maximum by 6.5 Nm, the second minimum by 3.55 Nm, the second maximum by 2.5 Nm and the range by only 5.7 Nm. Because of the equality of the variations, one cannot say that changes in platform perturbation velocity had any effect on any of the knee torque characteristics.
Table 4.6.5.2. Platform velocity and knee torque.

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<td>-5.2</td>
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<td>-5.8</td>
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<table>
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<th>dev from max</th>
<th>second max</th>
<th>range</th>
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<td>2.6</td>
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<td>17.7</td>
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<td>7.1</td>
<td>9.8</td>
<td>20.8</td>
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</table>

Velocity in deg/s, torque in Nm,
Second min|dev from max = value of second minimum | difference between max and second minimum.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

The increase in hip torque range was solely due to an increase in the second maximum from -3.1 Nm to a 12.7 Nm, an increase of 15.8 Nm. The other characteristics only ranged from a low of 1.6 Nm for the first minimum to 3.4 Nm for the third minimum. The range of the initial value was 1.9 Nm, thus rendering any of the other differences meaningless. The increase in the range with platform velocity was slightly less, 15.4 Nm, than the increase in the second maximum. This difference was due to variability between trials.

Figure 4.6.5.3. Platform velocity and hip torque.
Red = initial torque.
Green = first minimum torque.
Dk. blue = first maximum torque.
Yellow = second minimum torque.
Pink = second maximum torque.
Lt. blue = third minimum.
Brown = torque range.
Table 4.6.5.3. Platform velocity and hip torque by magnitude.

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<thead>
<tr>
<th>cond</th>
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<th>initial min</th>
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<th>first max</th>
<th>second min</th>
<th>second max</th>
<th>third min</th>
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<td>-14.4</td>
<td>-11.0</td>
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<td>-12.5</td>
<td>-15.6</td>
<td>12.7</td>
<td>-6.4</td>
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</table>

Velocity in deg/s, torque in Nm.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

4.6.6. Relation between Joint Torques and Joint Movements.

4.6.6.1. Ankle Torque and Movement.

Figure 4.6.6.1.2 contains four graphs. The first graph is of the horizontal distance between the ankle joint and the projection of the CM (figure 4.6.6.1.1). It is essentially the same graph as figure 4.4.2.2 except that the origin of this graph is centered at the ankle rather than at reference marker one. The second graph simply multiplies the average body weight of the subjects (649.6 N) by the horizontal difference between the CM and the ankle, to get the moment due to this offset. Note a more accurate representation would not include the feet. Thus, the position of the CM of the body minus the feet would be used to calculate the lever arm and the weight of the body above the ankle would be the force. However, because the feet are such a small component of the body this calculation was not necessary. The third graph of the ankle torque is a repeat of figure 4.6.1.2. The fourth graph indicates the difference between the moment at the ankle and the moment due to the offset between the ankle and the CM.
Figure 4.6.6.1.1. Ankle torque relations.
CG = center of mass or gravity
G = gravity line
CP = location of center of pressure
GRF = ground reaction force
R = distance between CP and GRF
horizontal difference between projection of CM and ankle

moment due to CM-ankle difference

<table>
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<th>difference (meters)</th>
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<td>0.00</td>
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<table>
<thead>
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<td>10</td>
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<tr>
<td>1.08</td>
<td>0</td>
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<tr>
<td>1.30</td>
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ankle torque
difference between ankle torque and torque due to CM-ankle offset

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<td>20</td>
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<td>0</td>
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<tr>
<td>1.08</td>
<td>-10</td>
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<tr>
<td>1.30</td>
<td>-20</td>
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<table>
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<th>moment (Nm)</th>
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<tbody>
<tr>
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<td>10</td>
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<td>1.08</td>
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<td>40</td>
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</table>

Figure 4.6.6.1.2. Ankle torque (positive is plantar flexion).
Each line represents mean of all trails in a condition.
Red = Condition one = full displacement/full power.
Green = Condition two = half displacement/full power.
Lt. Blue = Condition three = full displacement/half power.
Dk. Blue = Condition four = half displacement/half power.

Each graph in figure 4.6.6.1.3 shows the ankle torque due to the CM-ankle offset in blue and the calculated net ankle torque in red. As can be seen in each of the four graphs, the first reaction to the platform perturbation was a brief decrease in the ankle plantar flexion torque. This corresponded to a brief period in which the horizontal distance between the CM and the ankle decreased as the CM moved slightly posterior. An anterior movement of the CM was accompanied by a striking increase in the net ankle moment, due in part to the reverse in CM displacement as it now moved anteriorly and the CM-ankle distance increased. Next, the net ankle moment decreased below its initial pre-perturbation net planar flexion value. This decrease was not of the same magnitude as the previous increase in plantar flexion torque, however it was much longer in duration. In fact, the net moment at the ankle never returned to the value of plantar flexion that it had before perturbation. The decrease in the distance or CM-ankle lever arm and associated torque followed this general trend. This paralleled Winter’s (1990) argument of the difference between CP and CM location. In this case, it was the difference between the net ankle moment and the moment due to the ankle-CM offset.
In comparing the net ankle torque with that required to offset the CM displacement, the first response is less than what is needed at the ankle. Thus, there would be a slight tendency for the body to fall forward. This anterior motion is observed as a slight hump in the first graph of figure 4.6.6.1.2. The ankle reacts to this by increasing the plantar flexion moment. In turn, the body responds by falling backwards. This backward motion is quickly halted by decreasing the ankle torque faster than the CM-ankle offset, except in condition four where they reasonably balance one another. In the other three conditions, a balance is not achieved until approximately 1.25 s after the perturbation.

Figure 4.6.6.1.4 shows the highly antiparallel nature of the relationship between the ankle moment and displacement in reaction to the toe-up platform perturbations. The moments and displacements in these graphs are represented as deviations from their respective initial or pre-perturbation values normalized with respect to the range of values. That is:

\[
\text{percentage} = \left( \frac{\text{value} - \text{initial}}{\text{range}} \right) \times 100
\]

The blue curve indicates the time course of the ankle displacement while the red curve follows the ankle moment. A downward-sloped curve reflects a dorsiflexion moment and a decrease in the net plantar flexion moment (can be viewed as a tendency towards a dorsiflexion torque). The first reaction to the platform perturbation was a dorsiflexion of the ankle. This was initially met by a slight decrease in the net ankle flexion torque. With continued platform perturbation and concurrent ankle dorsiflexion, the ankle
stiffened as the net ankle plantar flexion moment increased. This increase in ankle plantar flexion moment acted to drive the shank (figure 4.6.6.1.4) as well as the rest of the body (as represented by the body’s CM, figures 4.4.2.2 and 4.6.1.6) in the posterior direction. However, the effect of the ankle’s increased stiffness on the body was relatively small, as the shank was displaced by only 2° to 3° and the CM by only 0.01 to 0.02m. In addition, the increased ankle stiffness did little to maintain the pre-perturbation ankle configuration. Thus, with platform perturbation, the ankle dorsiflexed and absorbed a great deal of the platform rotation allowing the shank and the body to remain upright. This was shown before in table 4.4.3.2 where it was pointed out that ankle dorsiflexion absorbed from 58.62 % to 79.18 % of the platform rotation. Considering that the ankle plantar flexion torque increased by a maximum of only 32.8 Nm for both ankles or 16.4 Nm at each ankle (an amount roughly equivalent to the torque used to maintain stance prior to perturbation) it was not surprising to find that the ankle absorbed most of the platform perturbation. The net ankle torque returned to its pre-perturbation value a few hundredths of a second after the ankle joint reached its peak displacement. The net ankle torque continued to decrease and eventually reached a value 13.5 to 25.9 Nm lower than its initial value. At this point, the net ankle torque was acting to slow the posterior rotation and displacement of the body. The net ankle torque stabilized at a new point that was less than that which was required initially. This primarily reflected the decreased net torque required at the ankle to offset the torque generated by the body not being perfectly aligned over the ankle joint as represented by the smaller horizontal distance between the ankle joint center and the CM that existed after the perturbation (see figure 4.6.6.1.3).

<table>
<thead>
<tr>
<th>Condition one</th>
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</thead>
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</tr>
</tbody>
</table>

![Graph](image1)

<table>
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<tr>
<th>Condition three</th>
<th>Condition four</th>
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</thead>
<tbody>
<tr>
<td>full displacement/half power</td>
<td>half displacement/half power</td>
</tr>
</tbody>
</table>

![Graph](image2)

Figure 4.6.6.1.4. Ankle moment (red, up = plantar flexion, down = dorsiflexion) and angle (blue, up = plantar flexion, down = dorsiflexion). Each line represents mean of all trails in a condition.
Condition one  
full displacement/full power  

Condition two  
half displacement/full power  

Condition three  
full displacement/half power  

Condition four  
half displacement/half power  

Figure 4.6.6.1.5. Ankle moment (red, up = plantar flexion, down = dorsiflexion) and shank angle (blue, up = counterclockwise, down = clockwise). 
Each line represents mean of all trails in a condition.
4.6.6.2. **Knee Torque and Movement.**

As with the previous figures, the knee moment and displacement are plotted with respect to their normalized deviation from pre-perturbation values with time. Figure 4.6.6.2.1 shows that the relationship between the knee moment and displacement, as was true for the relationship between the ankle moment and displacement, was antiparallel. Here an upward slope is a flexion and a downward slope is an extension movement or torque. Thus, the subjects were trying to maintain the knee angle by countering any displacement with an opposing torque. Note that initially there was a net extension moment that occurred before any knee movement. This was a response to the initial ankle dorsiflexion moment. Following the small extension moment was the relatively long (~0.40s) and large flexion moment that occurred in reaction to the knee extension. The knee extension began and ended slightly before the knee flexion torque. As the knee returned to and passed beyond its initial angle to a net flexion, the net moment at the knee reversed and proceeded to extension. For the most part, the knee remained in a more flexed position than it was before the platform perturbation and correspondingly maintained a net extension torque.
Figure 4.6.6.2.1. Knee moment (red, up = flexion, down = extension) and angle (blue, up = flexion, down = extension).

Each line represents mean of all trails in a condition.

The relationship between the net knee moment and the angular displacement of the thigh was much weaker than the relationship between the net knee moment and the angular displacement of the knee (figure 4.6.6.2.2). Here down indicates an extension torque and a clockwise displacement, while up indicates a flexion torque and a counterclockwise displacement. The weakness of the knee thigh relationship is in part due to the fact that the thigh displacement depends on the ankle joint as much as it depends on the knee joint. That is:

\[ \text{thigh angular displacement} = \text{platform angular displacement} + \text{ankle angular displacement} + \text{knee angular displacement} = \text{shank angular displacement} + \text{knee angular displacement} \]

The portions that do have a strong relationship were the first hump of the knee flexion moment, which opposed the thigh’s counterclockwise displacement, and the knee extension moment that countered the thighs’ counterclockwise rotation.
<table>
<thead>
<tr>
<th>Condition one</th>
<th>Condition two</th>
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<td>half displacement/full power</td>
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</table>

<table>
<thead>
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<th>time (seconds)</th>
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<td>1.30</td>
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Figure 4.6.6.2.2. Knee moment (red, up = flexion, down = extension) and thigh angle (blue, up = counterclockwise, down = clockwise). Each line represents mean of all trials in a condition.

### 4.6.6.3. Hip Torque and Movement.

The following figures 4.6.6.3.1 and 4.6.6.3.2 depict the relationship between the hip moment (red curves) and the hip displacement or trunk angular displacement (blue curves). The moments and displacements are plotted as functions of their deviation from initial conditions normalized with respect to the range of the deviation. Down indicates a flexion torque or displacement of the hip or a clockwise displacement of the trunk while up indicates an extension torque or displacement or a counterclockwise displacement of the trunk. Unlike the ankle and the shank, or the knee and the thigh, the hip and the trunk have very similar displacement patterns. Thus, little additional information is gained from examining the hip moment thigh angle curves (figure 4.6.6.3.2), however they are included for completeness. Here again, the body tried to maintain a homeostasis by meeting any movement at the hip with an opposing moment. In addition, after the reaction to the perturbation the subjects maintained a slightly flexed hip and thus needed to maintain a greater hip extension torque than was initially present before the perturbation.

The trunk ankle displacement, which results from the combination of the angular movements at the ankle, knee and hip, summarizes the overall effect of the combined movements of the three joints. That is:

\[
\text{trunk angular displacement} = \text{platform angular displacement} + \text{ankle angular displacement} + \text{knee angular displacement}
\]
+ hip angular displacement
trunk angular displacement = shank angular displacement
+ knee angular displacement
+ hip angular displacement

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<th>Condition two</th>
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<td>full displacement</td>
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<td>/half power</td>
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</table>

Figure 4.6.6.3.1. Hip moment (red, up = extension, down = flexion) and angle (blue, up = extension, down = flexion). Each line represents mean of all trails in a condition.
Figure 4.6.6.3.2. Hip moment (red, up = extension, down = flexion) and trunk angle (blue, up = counterclockwise, down = clockwise).

Each line represents mean of all trails in a condition.

4.6.7. Comparison of Joint Movements with Joint Strength Norms.

No joint response exceeded the averages of norms for net joint torque (table 4.6.7.1, see section 2.7). The magnitude of the average maximum net joint torque produced in reaction to the platform (usually found in condition one) is given along with the average net joint torque strength norms found in the literature (tables 2.7.4 through 2.7.9). To make these comparisons, the torque values calculated in this experiment were halved because they represented the torque produced by both left and right joints. For example, the maximum plantar flexion torque produced at the ankle was 71.29/2 = 35.64 Nm. The experimental results are compared to the norms through expressing the experimental results as a percentage of the norm. No ankle dorsiflexion was generated in response to the platform perturbation, thus the designation not applicable (NA) was placed in the corresponding table squares. The ankle plantar flexion moment response was the closest to meeting the maximum joint torques reported in literature however, it only slightly exceeded half of the maximum ankle plantar flexion of older adult females. Knee and hip reactions only barely exceeded five percent of the torque a person could generate, in most groups except older adult females, in which the net joint torques amounted to barely more than ten percent of their maximum capabilities. In other words, according to the literature norms an individual should have the ability to produce 20 times the torque than was required at the knee and hip, and 10 times the torque for older adult females.
Table 4.6.7.1. Comparison of joint moments (Nm) with average joint strength norms (Nm).

<table>
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<tr>
<th></th>
<th>ankle dorsiflexion</th>
<th>ankle plantar flexion</th>
<th>knee flexion</th>
<th>knee extension</th>
<th>hip flexion</th>
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<td>135 (60)</td>
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Standard deviation is in parentheses.
Percentage = maximum data values expressed as a percentage of literature norms.
Young adult = 17 to 39 years old. Older adult = 60 or older.

4.7. Summary.

Twenty-two young adults age 19 - 35 were screened for sensory and musculoskeletal factors that may be detrimental to normal balance and posture (table 4.1.1). After passing the screening, the subjects were placed on a rotational platform and their balance was perturbed by a toe-up rotation. The kinematics of the platform (5.8 to 9.8°, 65.1 to 30.4°/sec, and 771 to 445°/s², table 4.4.1.7) spanned the range of perturbations seen in literature (Appendix E). Unfortunately, the displacement, velocity and acceleration characteristics of the platform’s motion could not be controlled independently, and consequently the effects of each of these platform characteristics on balance could not be investigated. The subjects reacted to the platform perturbation by shifting their CM posteriorly 0.017 to 0.033m (table 4.4.2.1). While the magnitude of the CM displacement did parallel the severity of the platform perturbation, the timing did not. Nor was there any indication that the subjects were more careful about CM placement following a perturbation of greater severity than following one of less severity. The reaction to the platform perturbation traveled up the body in a wave like manner. The magnitude of the ankle displacement paralleled the severity of the perturbation as the foot followed the platform’s motion while the shanks’ motion was relatively insignificant (table 4.4.3.1). While the subjects’ ankles were able to absorb a large proportion of the platform perturbation, the amount of platform displacement absorbed was not proportional to the severity of the platform perturbation (table 4.4.3.2). The knee kinematics were small but not meaningless as angular displacement was 1.9 to 3.1° or in the range of 25 to 30 % of the platform or ankle displacement (table 4.4.4.1). The magnitude of the hip displacements were closer to the magnitude of displacements seen at the ankles (table 4.4.5.1). When sorted by magnitude of displacement the knee reacted the least, the ankle next, and the hip the greatest, regardless of the perturbation condition (table 4.4.6.1). The range of torques, or the magnitude of the torque adjustment required to maintain balance, was the greatest at the ankle and least at the knee (table 4.6.4.1). The timing of the net joint moments like the joint displacements proceeded up the body in a wave like manner. The ankle joint moment reaction was generally first and the hip generally last, while the knee joint moment bridged the gap between them (figure 4.6.4.1 and 4.6.4.2). No joint response exceeded the averages of the published net joint torque normals (table 4.6.7.1). In comparing the torques generated by the subjects in this study with the average of the joint torque norms, one finds that an average individual should possess a minimum of three times the ankle strength and ten times the knee and hip strength required to maintain balance after a perturbation similar to the one in this study. If an individual had only the strength of the lowest reported
norms, they would not have sufficient ankle plantar flexion strength to withstand the platform perturbation without falling or additional support. However, it seemed that the lowest values were unrealistically low. They indicate that these individuals would be incapable of standing or rising on the balls of their feet.
Chapter 5
Conclusions and Recommendations for Future Research

5.1. Introduction.

For many years, scientists have been striving to understand how humans control posture in order to maintain balance. Studies have attempted to investigate how changes or differences in various factors such as: vision, proprioception, vestibular function, physical activity, hormones, drugs, emotion, motivation and mental state affect balance control. Because variation in each of these factors results in changes in the biomechanical response, measurements of the biomechanics of balance and posture control will not only provide insight on how humans in general control balance but will also yield specific information on an individual's balance capabilities, strengths and weaknesses. Unfortunately, there exist two problems in utilizing biomechanics in this manner. First, the body is adaptable and it utilizes all its resources to different degrees to maintain balance. Thus, for example, if vision is unable to provide useful information about posture and balance the body will then recruit more heavily from other sources of information. Conversely, if another source of information is corrupted or unavailable then the body will adapt and utilize the information from the visual system to a greater extent. Therefore, it is difficult to related changes in the biomechanics of balance back to a deficit or change in a particular factor. The second and more fundamental problem is that there exists no baseline from which to measure biomechanical change. Only a few scientists have investigated the dynamics of posture and balance control of humans that are not subject to additional influences. These studies were limited to net joint torque peaks and latencies in responses to mild horizontal ground movements. Without this baseline one cannot begin to measure the contributions of various sensory systems, pharmaceuticals and exercise or training programs to balance. Therefore, the goal of this work was to study the dynamics of the reaction to rotational perturbations to posture in healthy young subjects without any physical attributes that could compromise their ability to respond. The mechanical and strength requirements quantified in this study define a baseline from which one can investigate the effects that various human physical, physiological and psychological conditions have on an individual’s ability to maintain posture. Twenty-two subjects completed this study. They were placed on a motor driven platform, and their reaction to a sudden sagittal plane toe-up rotation of the platform was video taped. Four toe-up rotation conditions, that were representative of those used in previous studies of reactions to postural perturbation (displacement ranged from 5.8 to 9.7 degrees and velocity ranged from 30.4 to 65.1 degrees per second), were presented to the subjects in a random order. The subjects repeated each of the four conditions ten times (40 trials per subject). The videos were digitized, and the dynamics of the platform and subjects’ segments and joints were calculated using a link-segment model that was individualized to each subject based upon their anthropometric data. Based on the results presented in Chapter 4 the following conclusions were made:

5.2. Hypothesis One: Kinematics of Strategy.

Significant knee rotation will occur in response to a sudden unexpected rotation of the ground of 5 to 10 degrees and 30 to 60 degrees per second. Therefore, the subjects’ strategy will be to utilize all lower extremity joints in response to the perturbations and not limit themselves to an ankle or hip strategy.

Consideration of section 4.4.4 of knee angular kinematics shows us that there was knee angular displacement in all four conditions, albeit the amount of knee extension was small (only 3.1°, 2.5°, 3.8°, 1.9° of extension from the pre perturbation position in conditions one through four respectively). Nevertheless, in all four conditions the displacement was significantly different from zero (p < 0.001 in each case). More importantly, when compared to the magnitude of the angular displacements of the platform and the other joints, the motion at the knee is meaningful. If expressed as a percentage of the platform displacement, the knee absorbed 42 % and 49 % of the platform angular displacement in conditions two and four, and 31 % and 38 % of the platform displacement in conditions one and three respectively. In comparison to the hip and the ankle, the knee was displaced approximately half as far as
mounted markers to markers fixed to bone through pins (e.g., Lafortune, 1992). Thus, most researchers
motion. For example, a great deal of research has been published comparing the motion of the skin
active muscular contraction of the quadriceps muscles is necessary for this to occur. According to Gray’s
moves, and fainting in soldiers asked to stand steadfast for long periods. With this in mind, locking the
reposition their feet and take small steps. These movements are not due to a lack of balance, but are used to
maintain comfort. Consider sitting still or lying still. In these two positions, sitting and lying, balance is
not difficult yet people move (even when asleep). Movement is used to redistribute stress and allow and
aid blood circulation. As counter examples consider pressure sores developed in individuals who cannot
move, and fainting in soldiers asked to stand steadfast for long periods. With this in mind, locking the
in a later edition of the same book indicate that the locking of the knee is not achieved in stance, and that
active muscular contraction of the quadriceps muscles is necessary for this to occur. According to Gray’s
Anatomy (1980), the knee extends until a balance is reached between the torques acting to extend the joints
(line of body weight passing in front of the joint rotation axis) and passive structures (stretching ligaments,
increasing congruence and compression of articular surfaces, increasing tension in extra articular tissues
posterior to the joint). More specifically, the passive tissues involved include: parts of both cruciate
ligaments, the tibial and fibular collateral ligaments, the oblique and arcuate popliteal ligaments, the skin
and fasciae (i.e., passive tension in the hamstring and gastrocnemius muscles) posterior to the joint capsule,
and the anterior parts of the menisci. The actual position is just short of the fully close packed position.
Thus, neither muscular energy nor a closed packed position is needed to maintain the stability of the knee.
With this in mind, locking the knees would be unnecessary and inefficient especially of active muscle is
needed to maintain it. This also allows some flexibility or cushioning with which to absorb some energy in
such a situation as a platform perturbation. It is also indicated in Gray’s Anatomy (1980) that besides
active contraction of the joint extensors, a fully extended closed packed position of the knee occurs only in
asymmetrical postures, during forward sway, heavy loading of the trunk, or during a powerful extension of
the leg. As an unscientific and informal test I have sneaked behind several individuals while they were
standing and pulled lightly posteriorly on the knee of the leg that seemed to be supporting the majority of
the person’s body weight (as indicated by knee extension and pelvic obliquity). There was always some
give, some additional posterior movement, or extension of the knee in response to the posterior force. This
occurred even in an individual who stood with a hyperextend knee.

Regarding the timing of the joint motions, the subjects responded in a manner similar to a whip
with a wave of displacement traveling up their bodies from the feet to the head (figures 5.2.1 and 5.2.2). In
all conditions, the ankle displacement preceded the knee displacement, which preceded the hip
displacement. The wave motion of the body served to dissipate the energy of the perturbation over the
body and lasted for a relatively long time (~2s of CM motion following perturbation). Rather than
concentrating or limiting the reaction to a specific structure or set of structures, the whole body acted to
absorb the energy of the perturbation, thus lessening the stress that any single structure had to bare. By
propagating the perturbation of the platform (kinetic energy) as a wave pulse from the feet through the
body to the head, the time with which the body has to dissipate the energy increased, thus lessening the
force that was required to generate the same impulse or eliminate the angular and linear momentum
generated by the platform. Energy dissipation also occurs through motion of muscle, other soft tissues and
fluids relative to bone. At this time, the significance of the soft-fluid component in posture (and other
activities) is not known. At present, the vast majority of work in biomechanics is focused on measuring the
motion of the skeleton or assumes that the body can be modeled as linked rigid segments, and most
research concerning the motion of other tissues is directed at improving the measurement of skeletal
motion. For example, a great deal of research has been published comparing the motion of the skin
mounted markers to markers fixed to bone through pins (e.g., Lafortune, 1992). Thus, most researchers

these joints were. Specifically the knee was displaced 32 %, 45 %, 48 % and 45 % as far as the hip, and
was displaced 40 %, 61 %, 95 % and 55 % as far as the ankle was displaced in conditions one through
four respectively. This knee movement occurred despite the limitations placed on it by passive tissues. Considering that the knee extended approximately two degrees in response to the relatively mild platform perturbation presented in the trials of condition four (5.8°, 30.4 deg/s, 397 deg/s/s peak) it is doubtful that the subject’s knees were fully locked (i.e., screw home joint configuration with the ligaments pulled tight, and joint surfaces fully compressed in a closed packed position). If the subjects had locked their knees in pre-perturbation stance, the two to three degrees of extension would not have occurred. While subjects may not have locked their knees because they were anticipating the platform perturbation, this is probably not the case. First, people do not normally stand still. That is they shift their weight from foot to foot, reposition their feet and take small steps. These movements are not due to a lack of balance, but are used to maintain comfort. Consider sitting still or lying still. In these two positions, sitting and lying, balance is not difficult yet people move (even when asleep). Movement is used to redistribute stress and allow and aid blood circulation. As counter examples consider pressure sores developed in individuals who cannot move, and fainting in soldiers asked to stand steadfast for long periods. With this in mind, locking the knees would be counter productive. Rasch and Burk (1974) in their kinesiology book and Grabner (1989) in a later edition of the same book indicate that the locking of the knee is not achieved in stance, and that active muscular contraction of the quadriceps muscles is necessary for this to occur. According to Gray’s Anatomy (1980), the knee extends until a balance is reached between the torques acting to extend the joints (line of body weight passing in front of the joint rotation axis) and passive structures (stretching ligaments, increasing congruence and compression of articular surfaces, increasing tension in extra articular tissues posterior to the joint). More specifically, the passive tissues involved include: parts of both cruciate ligaments, the tibial and fibular collateral ligaments, the oblique and arcuate popliteal ligaments, the skin and fasciae (i.e., passive tension in the hamstring and gastrocnemius muscles) posterior to the joint capsule, and the anterior parts of the menisci. The actual position is just short of the fully close packed position. Thus, neither muscular energy nor a closed packed position is needed to maintain the stability of the knee. With this in mind, locking the knees would be unnecessary and inefficient especially of active muscle is needed to maintain it. This also allows some flexibility or cushioning with which to absorb some energy in such a situation as a platform perturbation. It is also indicated in Gray’s Anatomy (1980) that besides active contraction of the joint extensors, a fully extended closed packed position of the knee occurs only in asymmetrical postures, during forward sway, heavy loading of the trunk, or during a powerful extension of the leg. As an unscientific and informal test I have sneaked behind several individuals while they were standing and pulled lightly posteriorly on the knee of the leg that seemed to be supporting the majority of the person’s body weight (as indicated by knee extension and pelvic obliquity). There was always some give, some additional posterior movement, or extension of the knee in response to the posterior force. This occurred even in an individual who stood with a hyperextend knee.

Regarding the timing of the joint motions, the subjects responded in a manner similar to a whip
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activities) is not known. At present, the vast majority of work in biomechanics is focused on measuring the
motion of the skeleton or assumes that the body can be modeled as linked rigid segments, and most
research concerning the motion of other tissues is directed at improving the measurement of skeletal
motion. For example, a great deal of research has been published comparing the motion of the skin
mounted markers to markers fixed to bone through pins (e.g., Lafortune, 1992). Thus, most researchers
regard the relative motion of the non-skeletal tissue as a “noise” to be eliminated or at least minimized rather than a functionally significant aspect of motion. There are few exceptions to this generalization; these are mostly studies involved in impacts (martial arts punches and kicks, foot contact in running or walking) or in shear stress (studies of blister formation or diabetic ulceration). The reason for this focus stems primarily from the rigid link assumption made in the models used to approximate humans. Regardless it is important to at least recognize the existence of the relative motion between bone and other tissues during periods of segmental motion, and note that this motion may significantly influence the dynamics of segments. The manner in which soft tissue hangs on bone and fluid pools in a segment may introduce errors in static situations as well. For example, the location of the CM of a segment will shift with respect to a local coordinate system fixed to the bone. The location of the CM will even change with different orientations of the segment as the tissue and fluid move under the influence of gravity, contraction of muscles, pull from tissues that comprise the neighboring segments, and influences of other external forces (e.g., clothing, braces, sports equipment, ground). Because, some segments are composed of multiple bones, such as the foot, hand and trunk, one also needs to carefully consider which bone the local coordinate system is to be fixed to or divide the segment into multiple segments.

Figure 5.2.1. Joint angular and CM linear displacements plotted as a percentage of their range. Each line represents the mean of all trials of all subjects in condition one. Condition one = full displacement/full power.

5.3. **Hypothesis Two: Joint Reaction Moments and Forces.**

To maintain balance following a sudden unexpected upward rotation of the ground ranging from 5 to 10 degrees and from 30 to 60 degrees per second, the subjects will produce greater net ankle reaction forces and moments than net knee reaction forces and moments, and will produce greater net knee reaction forces and moments than net hip reaction forces and moments.

5.3.1. **Joint Reaction Forces.**

With respect to the superior-inferior component of the joint reaction forces, the ankle reaction force was always larger than the knee reaction force, which in turn was always larger than the hip reaction force. This occurred because more mass was superior to the knee than the hip, and more mass was superior to the ankle than to the knee, rather than due to any factor related to the platform perturbation. In general,
the ankle reaction force was greater than the knee reaction force by an amount equal to the mass of
the subject’s two shanks (44.5 N to 71.8 N), and the knee reaction force was greater than the hip reaction
force by an amount equal to the mass of the subject’s two thighs (95.8 N to 154.4 N) (see table 4.2.1). The
difference in reaction force between two joints also depended on the acceleration of the segments’ CM, but
the acceleration of the segments’ CM was negligible in the vertical direction. If one considers the deviation
of the superior-inferior component of the joint reaction force from its initial value (figure 4.5.4.2.1), the
response at each joint was essentially identical.

The horizontal or anterior-posterior component of the net knee and ankle joint reaction forces were
comparable and considerably greater than the horizontal or anterior-posterior component of the net hip joint
reaction force throughout the four conditions (section 4.5.4). This occurred despite the similarity in the
time course of each joint’s force reaction to the perturbation. At all three joints, the anterior-posterior
component of the joint reaction force was first directed anteriorly for a brief period (peaking at ~0.07 s, and
lasting ~0.10 s regardless of condition). In these first brief moments, the anterior-posterior component of
the joint reaction force was essentially identical regardless of the condition in which it occurred or the joint
from which it originated. With the passage of time, the horizontal component of the knee and ankle
reaction forces increasingly separated themselves from the horizontal component of the hip reaction force.
At the peak of the posterior force reaction, the knee and the ankle force reactions were approximately one
and a half and two times respectively the magnitude of the posterior reaction force at the hip (average: 168
% for the knee and 193 % for the ankle). The peaks occurred at approximately the same time in each joint
in each condition (~0.18 s following perturbation initiation). The ankle and the knee continued to respond
much more vigorously than did the hip. In the interval between 0.35 s and 0.50 s after the beginning of the
perturbation, the hip was relatively un-responsive while the horizontal component of the joint reaction force
at the ankle and the knee were approximately 10 N in the anterior direction. By the time the hip generated
a 10 N anterior reaction force, the ankle and knee reaction forces had climbed approximately another 5 N.
Despite the differences in magnitude, the shapes of all the joint reaction force curves were similar reflecting
 simultaneity of the joint force reactions. Following the peak anterior force reaction at each joint, the
magnitude of each joint reaction converged to a common level. From 0.75 s after the start, there were no
appreciable differences in the anterior-posterior component of the joint reaction forces. In summary, the
ankle showed the greatest response with respect to the anterior-posterior component of the joint reaction
force, the knee responded with only a slightly lower magnitude and the hip’s response in the anterior-
posterior direction was significantly lower than that of the ankle’s or the knee’s.
In conclusion, the net ankle reaction force was greater than the net knee reaction force which was greater
than the net hip reaction force in the superior-inferior direction only, because there was more mass above
the ankle than the knee and more mass above the knee than above the hip. Even with the differences in
weight that each joint had to support, the deviation or adjustments that each joint had to make in the
superior-inferior component of the reaction force were more or less identical. In considering the anterior-
posterior component of the joint reaction forces, the ankle had to make the greatest adjustments while there
was no difference between the adjustments made by the knee and the hip.

5.3.2. Joint Moments.

If only the absolute value of the peak joint moments were considered, the greatest net torques were
produced at the ankle joint, the torques produced at the knee were next, while the torques generated at the
hip were the smallest in magnitude (see table 4.6.4.2 and figures 4.6.4.1 and 5.3.1). More precisely, in
descending order of magnitude, the peak moments were:
• ankle maximum (plantar flexion),
• ankle minimum (plantar flexion),
• knee maximum (flexion),
• hip maximum (extension in conditions one, two and three, flexion in condition four),
• knee minimum (extension) and
• hip minimum (flexion).

With the exception that the minimum magnitude of the torques produced at the ankle was smaller than the
maximum torques produced at the knee and hip in condition one.
Figure 5.3.1. Ankle, knee and hip moments.
Condition one = full displacement, full power.
+ = ankle plantar flexion moment (ankle extension moment).
+ = knee flexion moment.
+ = hip extension moment.

Another way to evaluate this hypothesis is to examine the change in net joint torque at a joint in response to the platform perturbation. To maintain stance, what adjustments or deviations from the static stance net torque had to be made at each joint (table 4.6.4.1 and figures 4.6.4.2 and 5.3.2)? If one only considers the peak torque deviation, the torques deviated the greatest at the ankle, and the adjustment at the hip was larger than that at the knee in all conditions.

Interpretation of the torques with respect to balance control is difficult. The difficulty arises from the geometry of the body. The ankle, knee and hip joints act in parallel with respect to controlling the anterior posterior (AP) motion of the body (e.g., positive net moments: ankle plantar flexion, knee flexion and hip extension, all move the body’s CM anteriorly). In adjusting the height of the body’s CM, the knee acts in opposition to the hip (e.g., positive net moments: flexion of the knee and extension of the hip, have opposite effects on the body’s CM vertical movement). The vertical motion of the body’s CM in response to the ankle is dependent on the initial state of the ankle joint. If the ankle is neutral, both plantar flexion and dorsiflexion act to lower the CM, if the ankle is in dorsiflexion, dorsiflexion will lower the CM while plantar flexion will raise it, and if the ankle is in plantar flexion, then plantar flexion will lower the CM and dorsiflexion will raise the CM. Interpretation of the relationship between joint torque and balance control is further made difficult because different postures require different net torques at each joint to maintain that given posture. Thus, increases or decreases in net joint torque may reflect a change in posture as well as a drive to maintain balance. As evidence, consider the torque (figure 5.3.1) or the deviation from initial net joint torque (figure 5.3.2) at 1.5 seconds after the perturbation. Since this is a period of relatively little body CM acceleration the difference in torque values must be due to differences in posture. Nevertheless, at 1.5 seconds after platform perturbation, the net ankle and knee torques decreased, while the net hip torque increased.
5.4. Hypothesis Three: Reaction to Platform Angular Displacement.

The greater the sudden angular ground perturbation distance, the larger the joint angular displacement at all joints. However, the distance of angular perturbation will have no effect on the peak moment at any joint.

A disclaimer needs to be made concerning the subjects’ response to a particular kinematic variable of the platform perturbation (see end of section 4.4.1). Not explicitly stated in the hypothesis was the assumption that the actuator would be capable of producing different displacements with similar, if not identical velocities, accelerations and jerks. For example, the velocities in conditions one and two were to be equal, as were the velocities in conditions three and four. That is, because the kinematic variables were not independent, it is impossible to say that differences in any one of the kinematic variables (in this case differences in angular displacement of the platform) induced differences in the subjects’ reaction to the rotational platform perturbation. Fortunately as shown in tables 4.4.1.7 and 4.4.1.8, when ranked by the severity of any of the dynamic characteristics the order was the same. Specifically, condition one produced the largest displacements, velocities, accelerations and jerks, condition three produced the second greatest platform dynamic characteristics, condition two produced the third greatest (with the exception of negative acceleration) and condition four produced the smallest platform dynamics (with the exception of negative acceleration). Unfortunately, the differences between the conditions were not equal (table 4.4.1.9). For example, while the increase in velocity from condition four to condition two and from condition two to condition three was roughly equal at 8.2 deg/s and 8.9 deg/s respectfully, the increase from condition three to condition one was approximately twice as large (17.6 deg/s).

This having been said, the results indicate that the angular displacement of the platform did have an effect on both the angular displacement and the peak moment of each joint. In general, the larger displacements (conditions one and three) produced greater reactions regardless of the variable (displacement, moment, etc.) than did smaller displacements (conditions two and four). Specifically, the
larger displacement conditions (one and three) produced larger joint angular displacements and net joint moments than did the smaller displacement conditions (two and four).

In conclusion, hypothesis three was only half supported. Increases in perturbation displacement increased both joint motion and moments. However, since other kinematic variables also increased with increases in displacement it cannot be ruled out that a perturbation characteristic other than displacement generated the changes in the joint reactions.

5.5. **Hypothesis Four: Reaction to Platform Angular Velocity.**

The faster the angular velocity of the rotational perturbation the larger will be the joint angular displacements, net joint forces and moments.

Similar to hypothesis three, a disclaimer needs to be made concerning the subjects’ response to a particular kinematic variable of the platform perturbation (see end of section 4.4.1). Because the kinematic variables were not independent, it is impossible to say that differences in any one of the kinematic variables (in this case differences in angular velocity of the platform) induced differences in the subjects’ reaction to the rotational platform perturbation.

In section 4.4.1 (figure 4.4.1.7 and table 4.4.1.10 reproduced below), it was shown that as the platform’s angular velocity increased, the subjects’ joint angular displacements increased. Note that the ankle and the hip were affected more than the knee. While the ankle and the hip underwent a 4.3° and a 5.4° increase respectively, the knee was limited to a 1.2° increase with increased platform velocity. This was not surprising given that the ligamentous structures of the knee such as the anterior-cruciate, arcuate popliteal, oblique popliteal and the collateral ligaments and the tendons and fascia from the semimembranous semitendinosus biceps femoris and gastrocnemius act to limit the knee’s range of motion in extension, whereas there are no corresponding structural limits to the reactions to the platform perturbation at the ankle and the hip.

![Platform velocity and joint displacement](image)

**Figure 4.4.1.7.** Platform velocity and joint displacement.

Red = ankle joint.
Green = knee joint.
Blue = hip joint.
Table 4.4.1.10. Platform velocity and joint displacement.

<table>
<thead>
<tr>
<th>condition</th>
<th>platform velocity</th>
<th>ankle displacement</th>
<th>knee displacement</th>
<th>hip displacement</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>30.4</td>
<td>3.40</td>
<td>1.86</td>
<td>4.13</td>
</tr>
<tr>
<td>2</td>
<td>38.6</td>
<td>4.09</td>
<td>2.51</td>
<td>5.60</td>
</tr>
<tr>
<td>3</td>
<td>47.5</td>
<td>5.94</td>
<td>2.78</td>
<td>7.92</td>
</tr>
<tr>
<td>1</td>
<td>65.1</td>
<td>7.68</td>
<td>3.06</td>
<td>9.49</td>
</tr>
</tbody>
</table>

Velocity in deg/s, displacement in deg.
Condition one = full displacement/full power.
Condition two = half displacement/full power.
Condition three = full displacement/half power.
Condition four = half displacement/half power.

The maximum posterior force, maximum anterior force and force range of each joint increased with platform perturbation velocity. The increase in the range of each joint’s force reaction was primarily due to an increase in the magnitude of the posterior joint force with increased velocity. However, the percentage increase in force with velocity was greater for the anterior ankle and knee force. The distance between a joint and the source of the perturbation, the platform, influenced the force-velocity relationship. Specifically the magnitude of the increase in joint reaction force whether it was the anterior force, posterior force or range of the joint force, decreased with increasingly superior joints. This effect was probably due to energy losses as the wave of platform perturbation traveled up the body.

Recall that the superior-inferior joint forces were reported as deviations from normative or pre-perturbation values (refer to sections 4.5.1.2, 4.5.2.2 and 4.5.3.2). There was virtually no increase in the maximum superior-inferior joint force with increase in platform perturbation velocity (figure 4.5.4.2.1 and table 4.5.4.2.1). The changes in the minimum superior-inferior joint force reaction were more dramatic especially with higher velocities. Initially, there was no change in or only a slight tendency to decrease the magnitude of the minimum superior-inferior joint force with increased platform perturbation velocity. With the second velocity increase (from 38.60 deg/s to 47.50 deg/s, from condition two to condition three), the magnitude of the minimum superior-inferior joint force decreased. It increased to an even greater extent with the last velocity increase between conditions three and one (from 47.50 deg/s to 65.10 deg/s). Because of the slight change in the maximum superior-inferior joint force, the joint force range response to increasing velocity was fundamentally a mirror of the change in the minimum joint force reaction.

The general net torque reaction was to increase with increasing platform velocity except at the knee, where there was no increase (section 4.6.5). The results revealed an equal increase in the magnitude of the minimum and in the maximum ankle torques. The ankle’s maximum increased by 15.02 Nm, while the minimum decreased by 12.40 Nm making the range with which the ankle had to respond increase by 27.42 Nm with the increase in speed. This was an 87.8 % increase in the reaction. There was little change in the knee torque minimums, maximums or range of response to increases in platform velocity. Because of this, one cannot conclude that changes in platform perturbation velocity had any effect on any of the knee torque characteristics. The increase in hip torque range was solely due to an increase in the second hip torque maximum from 3.10 Nm to 12.67 Nm, an increase of 15.77 Nm. Changes in the other characteristics were meaningless.

5.6. **Hypothesis Five: Joint Torque Required to Maintain Stance.**

In reaction to the sudden rotational perturbation, the subjects will generate joint torques that are lower than the capabilities of elderly individuals as reported in the literature. That is, elderly individuals should not have difficulty in maintaining stance following a rotational perturbation due to a lack of muscular strength.

No joint response exceeded the averages of the net joint torque norms (see sections 2.9 and 4.6.7). The net moment about the ankle joint was the closest to meeting the maximum joint torques reported in literature, however it was only 53 % of the lowest maximum ankle plantar flexion torque (older adult females). Knee and hip reactions only barely exceeded five percent of the torque a person could generate in most groups. In older adult females, the hip flexion joint torques generated in reaction to the platform
perturbation were barely more than ten percent of older adult female maximum capabilities. Therefore, according to the norms reported in the literature an individual should have the ability to produce 2 to 3 times the torque required at the ankle and 10 to 20 times the torque required at the knee and hip to maintain stance following perturbation.

In the above discussion, the influences of co-activation of antagonistic muscles or tension in passive structures were not considered. In both this experiment and the norms reported in literature, the torque values presented were net joint torques. Therefore, the reported values represent the lower limit of the torque generated by a prime mover or agonistic muscle or muscle group, and the actual torques needed may be much larger depending upon the amount of stabilizing and antagonistic activity that occurs around that joint. Nevertheless, this does not seem to be a point of contention. Although there is no proof, one may postulate that the stabilizing activity and antagonistic activity would be a small factor in the normative data because these values are usually generated in isolation either with a jig or some type of isokinetic dynamometer, while the perturbation net torques are similar to a free weight exercise in which antagonistic co-contraction may be a significant factor. Therefore, the normative net torque data would tend to be closer to the agonistic torque, and the net torque reaction would be less then the agonistic torque. Given that the limits of the ankle range of motion were not reached in the perturbation, passive structures around the ankle contributed little to the ankle torque at any point in time. At the knee and hip, passive structures would only contribute to extension torque. Because the toe-up perturbation flexed the hip and extended the knee into the end of its ROM, passive structures at the hip made only a small contribution to the hip joint torque, while passive structures at the knee made an increasingly stronger flexion contribution to the net knee torque. Thus, muscles that act to flex the knee, such as the hamstrings and gastrocnemius, would not contribute as much.

Consider the body to have an additional joint: one that exists between the ground (or support) and the metatarsal heads (for simplicity name this the toe joint). One can model this toe joint as a frictionless hinge joint with its axis parallel to the axis of rotational perturbation. This model will neglect the rolling of the foot on the ground that would translate the joint axis, and the unequal lengths of the metatarsals (or equivalently the different distances between the metatarsals and the ankle joint), which would rotate the axis from the frontal plane. Because the toe joint is a frictionless hinge joint with an axis perpendicular to the sagittal plane, and there are no physical structures crossing this joint (i.e., muscles, tendons, ligaments), no torque about this axis can be applied to the ground by the body or foot. The foot can only generate torque in the vertical direction through couple composed of shear forces. This contrasts Nashner’s (1997) supposition that one manner of maintaining balance is through ground reaction torques at the foot-ground interface. Nashner further indicates that the ability to generate these torques is limited by the length of the base of support, and therefore narrow bases of support (e.g., balance beams) limit the body’s ability to generate this torque. However, a narrow support does limit the ability of the body to maneuver the CP and thus manipulate the offset between the line of action of the ground reaction force (acting through the CP) and the force of gravity (acting through the body’s CM).

Modeling the interaction of the body with the ground as a hinge joint across which no torque can be generated, leads to a limitation in the ability of the muscles of the body to support the body in an upright position. To illustrate, consider eliminating the toe joint by fixing the foot to the ground (e.g., bolting the shoe to the ground, a condition in which torque can be applied to the ground about a ML axis). In this case, the mechanical limitations that the CM be above the base of support, and that torques cannot be transmitted to the foot from the ground (and visa versa) do not exist. Now the body could lean over and beyond the toes. The amount of lean would be limited only by the strength of the muscles (and other associated structures). With the toe joint, the foot will lose its ability to support the body when the CM passes in front of it as the heel will be pulled off the ground. Just before the heel leaves the ground, the CP will have reached its anterior limit just under the toe. This allows for an estimation of the maximum strength needed to maintain balance, which is the amount of muscle force needed to rise on one’s toes. By summing the moments about the ankle joint, we find that the muscle force needed to lift or support the body on one’s toes, must generate a torque just greater then the torque produced by the ground reaction force (GRF) acting at the toes (Figure 5.6.1.).
Now consider the situation of an individual standing on a platform that suddenly rotates (toes up, Figure 5.6.2). If the individual fixes the ankle joint through muscular action, the force needed is equivalent to that needed in the previous situation to support the body on one’s toes provided that the axis of rotation of the platform is at the toe joint (Figure 5.6.2 center). If the axis of rotation of the platform is not at the toe joint, the GRF depends on the weight of the subject and the acceleration of the body’s CM (Figure 5.6.2 right). As a first approximation, we can ignore horizontal accelerations for at least two reasons. First, the torque produced by horizontal accelerations will be small because the perpendicular lever arm to the ankle joint is small. Second, the horizontal accelerations themselves are negligible while the platform is horizontal. Thus, the muscle torque required is:

\[ \tau = (mg + ma_v)d_h = m(g + a_v)d_h \]

where
- \( m \) = mass of the body
- \( g \) = acceleration due to gravity
- \( a_v \) = vertical acceleration of the CM of the body
- \( d_h \) = horizontal distance between the ankle joint and the toe joint
Here the vertical acceleration of the CM of the body depends on the vertical (tangential) acceleration of the toe joint and is decreased by ankle dorsiflexion, knee flexion and other movements of the body. Thus, dorsiflexion of the ankle, flexion of the knee, hip, etc. will lower the force that the plantar flexor muscles have to produce. In addition, if the individual can begin (before perturbation) with their CM over the toes, the perturbation of the platform will not tilt the body. Therefore, in the situation where the platform tilts while the body does not tilt, the muscular strength required to remain balanced is equivalent to that required to stand on one’s toes.

To approach the situation from a slightly different perspective, consider the strength requirements for standing on one’s toes on the edge of a platform (Figure 5.6.3 second from left). Suppose that just before standing on the edge there existed a block that supported the heel and the rest of the foot (Figure 5.6.3 second from left), which suddenly dropped away (Figure 5.6.3 second from left). If sufficient muscle force, exists, the person will not fall. The muscle force that is sufficient is that required just to stand on the edge of the platform. Now picture a similar situation with several narrow blocks that fall after initially supporting the foot (Figure 5.6.3 second from right). The requirements of the individual have not changed and the person will not fall if they generate enough force to stand on their toes. If we continue this analogy to the limit where infinitely thin blocks are used, we reach a situation equivalent to a platform that pivots at the toe joint (Figure 5.6.3 right). Therefore, the muscle strength required to maintain balance on a platform that pivots at the toe joint is the same as that required to stand on the edge of a ledge. The only difference between this rotating platform and the ones used in perturbation experiments is that in experiments, an upward acceleration of the body is produced because the rotational axis of the platform is not coincident with the rotational axis of the toe joint.
Figure 5.6.3. Progression of ground falling away to upward platform tilt.

Therefore a functional test of whether or not one has the strength to balance during a toes up perturbation are toe raises. The toe raise approximation: (1) takes into account that the toe rise will produce an upward acceleration of the body (2) requires that muscles other than the ones at the ankle act to support the whole body. It does not take into account the difference between the self-generated movement of toe raises and the external or reaction generated movement produced by the tilting platform. Thus, while the strength requirements can be completely tested using this simple test, the motor control requirements cannot.

In conclusion, all but individuals with excessively weak muscles have the strength necessary at each joint to generate the torques needed in response to a toe-up rotational perturbation of the ground. Of all net joint torque reactions, ankle plantar flexion was the closest to being greater than the capacity as defined by normative data in the literature. Thus, the ankle joint plantar flexion had the smallest safety factor. Given that the experimental results showed that a maximum torque of 35.6 Nm was produced in reaction to the platform perturbation, the average weight of the subjects in this experiment was 649.6 N, and assuming that the lever arm of the foot is 0.10m, the capability of a person to lift their heel off the ground and stand on the toes of one foot, or the ability to stand on a stair supported only by the ball of a foot would give an individual a safety factor of two. That is, they would posses the ability to generate twice the plantar flexion torque that is required to successfully react to a toes-up tilt. Similar calculations indicate that the torque required at the knee in reaction to the platform rotation is of the same magnitude as the torque required to hold the shank in horizontal position while lying down. For example, given the subjects averaged 650 N, a foot would weigh 650 x 0.0145 = 9.4 N, and the shank 650 x 0.0465 = 30.2 N (table 3.10.1). Arbitrarily assigning the length of the shank to be 0.36 m, a torque at the knee of (9.4 x 0.36) + (30.2 x 0.36 x 0.433) = 8.1 Nm would be required to hold the shank horizontal. To ensure that a person has a safety factor of two at the knee they would have to be capable of holding their shank horizontal with a 20 N weight attached to the foot. This would add a torque of 20 x 0.36 = 7.2 Nm. Consequently, a simple test of having an individual support their shank in a horizontal position with a common 2.2 kg weight plate attached to the foot would indicate if the person had adequate strength to remain standing after a rotational perturbation. Whether this torque was a flexion or an extension torque would only depend if the person were lying prone or supine. Similar calculations at the hip show that the hip torque required to respond to the toe-up perturbation is a factor of ten less than the torque required to hold the trunk horizontal when bending at the waist even neglecting the head, arm, forearms and hands (τ = 650 x 0.497 x 0.25 = 80.8 Nm).
5.7. Recommendations for Future Research.

In this experiment, the peak torque production at each joint of the lower extremities in response to a rotational perturbation of the ground was compared to the strength of elderly individuals as reported in the literature. There are limitations to this comparison, most of which are related to what is generally termed “specificity”. That is, the kinematics and kinetics of these tasks that were used to arrive at strength norms were different from the task of balancing in response to a rotational perturbation. In short when performing a specific task one will generate a force or torque which may or may not be related to one’s strength. Before continuing any further strength needs to be defined. Kulig, Andrews and Hay (1984) define strength as the maximum amount of force produced by a muscle or muscle group at a site of attachment on the skeleton. Zatsiorsky (1995) for the most part concurred. However, he noted the many factors that influence the ability of a muscle or muscle group to produce force and that the force might vary throughout a movement. Therefore, he defines strength as the maximum of the maximum force (F_{mm} or “maximum maximorum force”) one can produce under optimum conditions. Some of the factors that establish optimum conditions include:

- direction of muscle action (eccentric, concentric, isometric), state of pre-loading,
- length-tension, torque-angle, force-time characteristics,
- intrinsic muscle geometry (pennation, muscle-tendon length), fiber type and
- motor neurological and motivational factors.

A muscle’s force capability varies with the direction and speed of movement. A muscle has the greatest potential to generate forces eccentrically and the least ability concentrically while the capability of a muscle to produce force isometrically falls between that of eccentric and concentric actions. The concentric force capability is greatest at low velocities, while eccentric force production may increase with velocity. One mechanism underlying these differences exists in the ability of the muscle to develop cross bridge links between the actin and myosin filaments. The ability to form cross bridge links decreases with speed of movement therefore force capacity decreases. With eccentric action, the external force has to break the cross-bridge bonds. Thus, the muscle does not have to act as hard to resist the external force and it has the ability to counteract larger forces than it would otherwise.

History influences a muscle’s ability to act. Pre-loading increases force production through stretching the elastic components of muscle (storage of potential elastic energy), which can be released with the concentric action of the muscle (Komi, 1986, 1992, Huijing, 1992). To be most effective, the pre-loading should involve fast-twitch muscles, result in mild strain and should occur just prior to the muscle action (< 0.9 s) (Komi, 1984). Another indication that muscle force production is dependent on its history comes from the observation that the isometric force capability of a skeletal muscle is lower following shortening when compared to a purely isometric reference action (Abbott and Aubert, 1952, Marechal and Plaghi, 1979, Sugi and Tsuchiya 1988, Edman et al., 1993, Edman, 1996, Herzog, Leonard and Wu, 1998). The mechanism for this force drop is not understood. However, it is hypothesized that the force reduction is associated with inhibition of cross-bridge attachments in the new zone formed by the shortening of muscle (Herzog, Leonard and Wu, 1999).

The length-tension relationship is related to the structure of the sarcomere and its ability to generate cross bridges. At short muscle lengths, the cross bridges begin to interfere with one another. If a muscle continues to shorten, the actin filaments will run into the Z-lines. At long muscle lengths, the region of overlapping of myosin and actin filaments decreases, resulting in a reduction in the ability of cross bridges to form and therefore reduces force production. Some of this lost force production is made up passively by stress in response to the strain of the parallel elastic components of muscle. Fortunately, few movements of the body place muscles at these extreme lengths and the length-tension factor is generally negligible (Hamill, and Knutzen, 1995). The exceptions are two joint muscles. The result of a muscle action at a joint will depend on the dynamics of the other joint(s) that the muscle crosses (Riener and Edrich, 1999). Consider the effect of hip position (or movement direction) on one’s ability to flex (semimembranosus, semitendinosus, long head of biceps femoris) or to extend (rectus femoris) the knee. Note that “off axis rotations” of joints may also influence force production. This is illustrated by the difference in one’s ability to flex the elbow when the forearm is supinated and when it is pronated. In addition, the ability of the hip muscles to internally rotate the thigh changes with hip flexion (Delp, Hess,
Hungerford and Jones, 1999). Some have argued that the division into length-tension and force-velocity characteristics is an artificial one and that all three have to be simultaneously taken into consideration (Baratta, Solomonow, Nguyen and D’Ambrosia, 2000). Therefore, a muscle’s characteristics should be described in the context of a force-length-velocity relationship.

The torque-angle relationship combines the changing force-length relationship of muscles with the changing effective lever arm of a muscle that occurs with variations in joint position. Few muscles act along straight lines and many act across anatomical “pulleys” that alter the direction of their pull and grant them a mechanical advantage they would not have otherwise (e.g. patella and other sesamoid bones). One factor that many neglect to include in this relationship is the fact that the moment arm of the external force also normally changes with position of a segment (Delp, Hess, Hungerford and Jones, 1999). For example, as a segment rotates, its relationship with external forces generated through gravity (the weight of a dumbbell or the weight of the segment itself) changes. For instance, a dumbbell held in the hand will not provide much resistance to elbow flexion if the forearm is vertical. However, if the forearm is horizontal, the weight of the dumbbell will act perpendicular to the forearm and its lever arm will be of maximum length.

The force curves across a joint are the result of the superposition of the action of several muscles each with different characteristics. Since several muscles may be involved in a particular task, and a task can involve several positions and velocities (both speed and direction) the influence of each muscle may vary through the task. At one point, a muscle may be an active contributor to the motion, at another point it may act as a stabilizer and at another point may act to inhibit the motion. In other words, a muscle may act concentrically, eccentrically and isometrically. In addition, a muscle acts on all of the joints of the body, even ones that it does not cross, due to transmission of kinetic energy. As an example, consider a rope that is held at one end in the hand. One may consider the rope to be made up of an infinite number of joints and segments none of which is crossed by a muscle in the body. Nevertheless, the rope (and its joints) can be made to move and flex by muscles in the forearm flexing and extending the wrist. Therefore, coordination of muscles is essential to strength production. That is, even if an individual has the needed strength capacity in individual muscles, they may not have the skill to correctly utilize this capacity. Thus, a weaker individual with more skill may be "stronger" than a stronger individual with less skill.

The force-time characteristics of a muscle depend on both mechanical and neurological factors. Mechanically a muscle-tendon complex is not rigid. Therefore, it takes time for a muscle to pull up any slack and build tension. Additionally there is an electrical-mechanical delay (difference between nerve impulse and force production in muscle). It also takes time for the nervous system to deliver an electrical signal to the muscle. Before a voluntary action can occur, a need for the action has to be perceived and recognized, decisions have to be made (either consciously or sub-consciously) and the response signal has to be delivered. Each of these takes a finite amount of time. Therefore, one may not have the capacity to generate maximal force (let alone any force) in events that occur in a very short period. For instance, it has been estimated that in many explosive activities the movement time is less then the time it takes one to reach a peak isometric force (0.3 s to 0.4s) (Zatsiorsky, 1995). In this situation, the rate of force development, not the absolute force capacity is the key factor in the successful completion of a task. Strong individuals do not necessarily posses the ability to generate force quickly (e.g., not all strong individuals are capable of throwing a 90 mph fastball).

By Newton’s third law, forces always act in pairs. The relevance of this to strength production is illustrated by throwing a ping-pong ball. In this situation it is impossible for a person to produce a force anywhere near his capability because the ping-pong ball will not provide the reaction forces due to its low inertia. Therefore, in order to generate a large force one must interact with something substantial that can respond with an equal and opposite large force. This in part leads the operational definition of strength as the heaviest possible weight that one can move in one repetition. Hardly ever is strength defined with respect to the largest acceleration that one can impart on an object in one repetition, however this definition may be more appropriate in the context of performing tasks quickly.

Many other factors such as pain, injury, sickness or fatigue can also alter a person’s ability to produce force. Even if all the right conditions exist, the individual’s motivation may not be there, and maximal force will not be produced. Unless the muscles are externally stimulated, maximal action requires that the individual possesses the will to generate the necessary effort. Other factors such as the interaction
of muscle spindles and the reflex due to Golgi organs influence the neural input to muscle during some
actions (e.g. the reaction to a sudden perturbation). The myotatic stretch reflex is responsible for activating
muscles to oppose a sudden stretch, or more precisely, lengthening of the muscle. The Golgi organ acts to
inhibit high forces generated in tendons by inhibiting muscle action. Thus, the resulting effect of a
movement, a reaction to a perturbation for instance, is in part a result of the stretch reflex acting to activate
muscles and the Golgi organs acting to inhibit muscle action.

Therefore given the preceding discussion, there are many limitations in comparing the strength
capabilities of elderly individuals with the peak torques produced by young individuals who successfully
balanced on a perturbing platform. The primary problem is the lack of an instrument or protocol to
measure the strength of individuals in a manner that mimics the characteristics of the balance task. Other
more traditional methods using free weights, weight machines, dynamometers or other means of generating
resistance will allow the subjects to have as their goal maximizing force/torque production. However, they
all have a limited ability to replicate the kinematics and kinetics of the perturbation task. One method
would be to utilize the perturbation task itself. That is, by progressively increasing the difficulty of the
perturbation challenge one could gain insight into the capabilities of the individuals tested. However
because the goal of the task is to maintain balance not to output maximum force or torque, strength may not
be measurable in this manner. Nevertheless, the goal here was not to quantify the strength of individuals
but to measure the dynamics utilized in an ideal situation (define a baseline), in subjects with no
neurological or muscular deficits. As part of the dynamic measurements the net joint torques at each of the
major lower extremity joints was calculated. The net joint torques were then compared to the strength
values (norms) reported in the literature for elderly individuals. The theory was that if the strength values
of the elderly were lower than the torques utilized by the subjects used in this experiment to balance then
lack of strength could be identified as a hindrance to the maintenance of balance. However if the
capabilities of the elderly do exceed the torques utilized in this experiment yet they were having difficulty
balancing then more research needs to be done to determine what is limiting the elderly from performing to
their potential. Then the problem is not whether or not they can produce the required torque to balance, but
rather what is it about the balance task that prevents them from producing the required torque. Is it the
speed with which the torque needs to be produced? Is it an inability to effectively coordinate their muscles
to perform the task? Is it a fear of the situation?

The next logical steps to take in the research of balance, posture and falling would be to: (1)
enlarge the focus to the kinematics and kinetics of falling as well as balancing, (2) expand the subject
population to include elderly non-fallers and fallers, (3) examine the effect of different somatotypes on
balance, (4) investigate the effects of different mechanical aids to balance (5) examine the posture and
falling in different planes and in asymmetric stance and (6) develop training methods to minimize falls and
fear of certain environmental situations.

Now that the kinematics and kinetics required to maintain balance in response to a toe-up postural
perturbation have been defined, research needs to focus on ascertaining the kinematics and kinetics of
falling. This means analyzing trials in which the perturbation caused the individual to fall. Given this
information, one will be able to compare and contrast the two to pinpoint any kinematic or kinetic
characteristic(s) that define or determine if a person is going to or is likely to succumb to a perturbation. If
these characteristics do exist, perhaps simple clinical tests can be developed to identify individuals with a
potential for falling. Once potential fallers have been identified, practical methods for personal
modification to lessen the chance that the individual will fall may be created. Training programs should be
designed to focus on the specific weaknesses an individual may have that would predispose them to a fall
and ignore items that would not. For example, if an individual could raise themselves easily on the ball of
their foot, and hold their leg horizontal, strength training would provide little additional protection from
falling. Such an individual’s time would be better spent on skill or coordination development. Based on
the results in this experiment it seems unlikely that an individual’s strength limits their ability to balance, at
least in reaction to a sagittal plane toe-up perturbation. Success in balance must therefore depend on
coordination or skill. This is not without precedent, as participation in activities of art, some dance, some
sport and activities of everyday living often depend more on coordination, finesse and skill than on
strength. Therefore, in order to try to prevent a person from falling, the focus should be on teaching the
skill of balance rather than on strength development. Development of training methods to increase an
individual’s ability to maintain balance even in unexpected or unfamiliar situations should be the focus of future research. This might lead to balance or fall avoidance training programs in which individuals walk on safe but highly unstable surfaces such as very thick carpet, uneven mats, a soft sandy beach, a thick high grass field, or with special unstable shoes in which balance is difficult, falls likely, yet chance of injury remote. In addition to the skill of balance maintenance, training should include hazard recognition, avoidance and minimization. With such training it is possible that the fear of certain situations will be minimized, for, if a person has a fear of falling, they will not react in an efficient or appropriate manner. Consider an individual who has just learned to ride a bicycle, who has found himself or herself in the situation of going very fast down a steep hill. They have the capability to balance and ride the bicycle, but because of the fear of falling, they stiffen up and act in an inefficient or inappropriate manner that may result in crashing or falling. More to the point, there are everyday situations such as steep stairs and the approach or exit of an escalator, in which a person acts inappropriately due to fear. In such instances, the situation or hazard may not be avoidable, but may be minimized by training the individual to act in an efficient manner with little fear. The dangers of falling can also be minimized through training individuals how to fall without injuring themselves as is done in martial arts training. Thus, if a hazard is not recognized or avoided and the individual does happen to fall, the results need not be so debilitating.

Research on the kinematic and kinetic reactions of older adults should parallel the work done on young adults. The population should be expanded to investigate if there are any age related dynamical differences between the young and the old that would pre-dispose one to falls. It might be true that such characteristics commonly associated with aging such as lost stature (lower CM, increased stability) higher body mass, lower lean mass (harder for muscle torques to alter segmental motion), and decreased flexibility in joints (increased passive resistance to perturbation) may actually improve stability and decrease the likelihood of falling.

At present research of the effects of mass variation and redistribution has not been conducted. While there are studies that indicate lower risk for hip fracture with increased body weight (see Chapter 3) no study has indicated the effect on fall risk imposed by this increased mass. Neither have studies indicated the effect of different body types, relative proportion of lean and fat mass, strength weight ratio, body height, or other anthropometric (mechanical) variables on fall or injury risk. For example, the ability to rise on one’s toes is easier for a lighter individual than it is for a heavier individual. Thus, the weight of an individual may allow extrapolation of the rule of thumb that individuals have enough strength at the ankles to balance if they can raise themselves on their toes without error. However because the mass of a lean and a robust individual is distributed differently, extrapolation may not work. Regardless, the effect of mass and mass distribution (or somatotype) on human balance needs to be studied.

A forth area of research would focus on devices that modify an individual’s kinematic and kinetic characteristics so that falling is more difficult. Already in existence are canes, crutches and walkers that increase stability through the expansion of an individual’s base of support. Other devices that increase personal stability need to be researched. For instance, while a pole similar to that used by tightrope walkers might be beneficial to those in need of balance aid, it might be impractical, especially on a crowded city street, subway or soccer stadium. A device that might be of practical use might be a personal gyroscope. Provided power and weight issues could be solved, a gyroscope firmly attached to the trunk may provide the additional stability needed to prevent a fall. It could also be utilized as a feedback mechanism. A device such as this might be very beneficial to those with a vestibular deficit. A simpler device such as a balance or a bubble-level may serve as a feedback mechanism just as well. In the future, research may develop the use of feedback to aid or replace other defective sensory organs or systems. Kinesthesia may be provided by electrical goniometers or three dimensional position sensing devices that utilize infrared or electromagnetic radiation (e.g., Bird of Prey), and proprioception may be provided by insole pressure devices such as those made by F-scan or Emed. Presently two groups of researchers, Streight and Cavanagh (unpublished) at the Pennsylvania State University and Steve Robinovitch at University of California, San Francisco are taking a slightly different approach by researching devices that lessen the damage an individual might sustain from a fall. Streight and Cavanagh are looking into dual density floors which can be installed in hospitals, nursing homes or even individual residences, that act as a soft cushion when fallen upon yet maintain a firm surface to allow smooth rolling of carts and wheelchairs and as well as a stable surface for walking. The Berkeley group is researching personalized protection pads that protect
body parts susceptible to damage in a fall (i.e., hip). Each pad has a unique ability to modify and improve a certain balance related characteristic, so that even if there are no universal or even group kinematic or kinetic characteristics differences related to susceptibility to falling, each individual can be provided with a personalized array of devices that will optimize their ability to be stable.

Most studies on posture (especially perturbed posture) are limited to the sagittal plane. There are a few frontal plane studies. Yet falling is not strictly limited to these planes. For example, in the sagittal plane the posture most often studied is one in which the ankles are aligned. While this is experimentally convenient, this position is a special and unique postural situation. Here all joints are aligned in the sagittal view, and the base of support is the smallest it can be in the AP direction (save reduced support conditions or when one stands on a fraction of one’s foot). With another stance, the feet and joints of the lower extremities are offset (even the hips may be offset of the pelvis when rotated about the body’s longitudinal axis). Taking this a step further, one can consider there to be a continuum between the AP and ML axes depending on the rotation of the body (with the complication that the ankle and knee joints are primarily hinge joints). Consider the following example. First, stand with the feet side by side (Figure 5.7.1.a). Then, rotate the entire body by pivoting on the heels or the balls of the feet about a vertical axis in either direction (Figures 5.7.1.a-d). After rotating ~90° one obtains a posture in which the feet are in line one in front of the other (Figure 5.7.1.d). Reversing the direction of rotation (Figures 5.7.1.e-h) one returns to the side by side foot position (Figure 5.7.1.h). Continuing with the reversed rotation (Figures 5.7.1.h-l) one obtains a stance in which the feet are in line again but the opposite foot is in front (Figure 5.7.1.l).

![Figure 5.7.1](image-url)

Figure 5.7.1. Progression posture from ankles aligned in sagittal plane to aligned in frontal plane (top view).
Finally, all this research has to be put into practice in an efficient manner by modifying factors that can be effectively modified. First, the factors relevant to fall risk and injury need to be identified. As indicated in Chapter 3, a great deal of work has gone into this, however most research only reports correlations between a single or a few factors and fall (injury) risk and does not establish causation. Secondly, while many studies have found that individuals of all ages can train and adapt to training to different degrees, the studies are generally limited to the relationship between a training protocol and falling risk factors. The effect of training on fall and injury prevention is then inferred. Therefore, while a person may train properly and adapt to the exercises, they may not actually decrease their fall risk. The situation is similar to an athlete who trains hard in practice yet performs poorly in competition, a musician who rehearses well yet plays poorly in concert and a student who hits the books hard yet cannot apply his knowledge well in a practical exam. More closely related to balance, is the frequent situation of a physical therapist increasing an individual’s range of motion through stretching exercises without the individual utilizing the range of motion in activities of daily living. The missing piece of the puzzle is the connection between practice and performance. The individual at risk of falling, like the athlete, musician and student, may have the technical and physical skills necessary to perform a task yet may not have the ability to apply these skills to real life situations.

5.8. Summary.

The subjects reacted to the toe-up perturbation by shifting their CM posteriorly from 0.017 to 0.033m. The size of the CM motion paralleled the severity of the platform perturbation, however, neither the timing of the CM displacement nor the degree to which the subjects returned their CM to its original location paralleled the platform perturbation strength. There was no indication that the subjects were more careful about CM placement or about stability following a perturbation of greater severity than following one of lesser severity. The small amount of CM displacement was accomplished through the subjects dorsiflexing their ankles almost the same distance that the platform rotated, extension of the knee and flexion of the hips. These actions proceeded up the body in a wave like manner with the motion of each segment or joint following the motion of the segment or joint inferior to it. As the motion traveled up the body it was dissipated so that the large movement of the platform and the foot was followed by a smaller motion of the shank, a still smaller motion of the thigh and an even smaller motion of the trunk. Unlike segmental motion, joint motion did not follow a pattern of decreasing magnitude with increasing distance from the platform. Instead, the knee’s range of motion was limited by ligamentous and other joint constrains to only a couple of degrees. The ankle and the hip were displaced approximately five and four times as much respectively.

In response to this motion of the various segments of the body, joint torques were established almost instantaneously. In general, the ankle was the first to change from its initial moment. The ankle
also showed the greatest deviation from initial values. The hip was the last to react however, it deviated from pre-perturbation values more than did the knee. The range of torque required at the ankle was almost two times that of the knee or the hip even though there was no change in direction (the ankle always produced a plantar flexion moment, while the hip and knee produced both extension and flexion moments). The range of the hip displacement was only slightly larger than that of the knee. The magnitude of the ankle torque paralleled the movement of the projection of the CM on the base of support, or the horizontal distance between the CM and the ankle joint. Anterior movement of the CM was accompanied by increases in ankle torque, while posterior movements were accompanied by decreases. With platform perturbation and concurrent ankle dorsiflexion, the ankle stiffened (the amount of torque applied at the ankle approximately doubled). Fortunately, the effect of the increased stiffness was relatively small as it did little to maintain the pre-perturbation ankle angle, the shank was displaced by only two to three degrees, and the CM displaced by only 0.01 to 0.03m. The increased ankle stiffness still allowed the ankle joint to absorb 58.62 % to 79.18 % of the platform rotation. The torque response at the knee was directly opposite to that of the knee displacement. With perturbation, the knee was thrown into extension and then a slight flexion. The torque at the knee correspondingly reacted in opposition. Similarly, the hip as with other more distal joints opposed movement with a counterbalancing torque. As the hip flexed in response to the perturbation, an extensor torque was generated at the hip to oppose it.
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Appendix A
Exclusion Criteria

Subjects were excluded if they had:

- any medical conditions that might interfere with lower extremity function and posture
- a history of major surgical procedures, major fractures, or other musculoskeletal problems likely to affect lower extremity function or posture
- acute weakness of the lower extremity musculature
- any marked lower extremity deformity or abnormal joint range of movement
- any arthritic conditions causing frequent pain, discomfort and difficulty in walking or standing
- Any clinically apparent significant abnormalities of the back, hip knee, ankle or foot
- any clinically apparent problems with gait or posture
- a history of neurological disease likely to affect lower extremity function or posture, including, but not limited to peripheral nerve compression, focal neuropathies, radiculopathies, strokes and signs of Parkinson’s disease
- any significant visual impairment
- visual acuity less than 20/100
- loss of binocular vision
- presence of double vision
- abnormal clinical evaluation of vestibular function
- a history of psychological/psychiatric conditions requiring medical attention
- use of any medication that affects the central or peripheral nervous system or alters muscular function or may affect postural stability (Table A1.1)
- a history of drug or alcohol dependency treated or untreated.
- a diastolic drop of more than 10 mmHg or if a systolic drop of more than 20 mmHg was measured. Subjects were excluded if their systolic blood pressure dropped more than 30 mmHg, or if there were visible signs of postural hypertension.
Appendix B
Consent For Clinical Research Study
The Pennsylvania State University

Title of Investigation: “Kinematic and Kinetic Analysis of Posture Following Rotational Perturbation.”
Investigators: John D. Henley
Peter R. Cavanagh

Date: August, 1994 - October, 1994

This is to certify that I, ________________________________, have been given the following information about my participation as a volunteer in a scientific study by John D. Henley.

1. Purpose of the Study:

It has been explained to me that the purpose of this study is to investigate the forces produced at the joints in response to a tilting movement of the ground.
I understand that the study will be made up of subjects between 18 and 35 years old and who do not have any physical problems that would hurt their ability to stand. Ten (10) males and ten (10) females will be tested.

2. Procedures to be Followed:

The experiment will take about three (3) hours. During this time I will be asked questions about my health, my vision and my vestibular system, my feeling on the bottom of my foot, the range of movement of my joints, the strength of my muscles and my reflexes will be tested. Then, I will be asked to stand quietly on a platform, with my feet comfortably separated side-by-side and my arms folded across my body. I will be asked to stare at a target placed on the wall directly in front of my eyes. Reflective markers that make it easy to find landmarks in a videotape will be attached to the several places on my body with tape and EMG sensors (metal plates that detect the electrical activity in the muscles) will be attached to my leg. In order for the sensors to work properly the area under the EMG sensors will be cleaned and shaved.

Volunteer ___________________________ Date ___________________________ Investigator ___________________________ Date ___________________________
The platform will be tilted up at different speeds and for different amounts. Four different types of tilting movements of the ground will be conducted ten times each for a total of 40 trials. The sequence of these movements will be in a random order (equal likelihood that any type of tilting movement could occur at any time).

In response to the platform movement, I will try to stand as still as possible and not move my feet or arms unless I absolutely have to. During the test, a harness will be attached to me at the shoulders to protect me from falling. During the experiment, I will be closely watched by the principal investigator in order to prevent any injury from occurring. Mary Becker, RN, will also be present during the test to provide help if necessary.

3. Discomforts and Risks:

I understand that there is minimal risk for me during this study. None of the tests are invasive. Only a non-allergic tape will be used to apply the reflective markers and EMG sensors to my skin. The EMG sensors simply measure the electrical activity produced naturally in my muscles and do not output any electrical current of their own. Furthermore, the testing attempts to duplicate things that happen in normal daily activities (slippery floor, moving bus, etc.).

During this experiment, I will wear a safety harness around my shoulders which is attached to the ceiling and has been used for several years at CELOS.

4. a. Benefits to Me:

Except for the examination of how well I stand after the ground tilts up, I understand that I will not directly benefit from this study.

b. Potential Benefits to Society:

I understand that my contribution to this study will help scientists learn how people stand-up and how to prevent them from falling down. For example, up to 50% of elderly report a recent fall. Therefore, the results of this study may be very helpful to them.

________________________________________  __________________________________________
Volunteer                     Date                     Investigator                     Date
5. **Alternative Procedures Which Could be Utilized:**

Previous research has measured the reaction to tilting ground movements by measuring the electrical activity produced when muscles contract and by watching how people move their bodies. However, these measurements are not able to determine how much force and twist is produced at the joints.

6. **Time Duration of the Procedures and Study:**

I understand that this study will take about three (3) hours of my time for one visit to the Center for Locomotion Studies.

7. **Statement of Confidentiality:**

Your participation in this research is confidential. Only the investigator and his assistants have access to your identity and the information that can be associated with your identity. In the event of publication of this research, no personally identifying information will be disclosed.

8. **Right to Ask Questions:**

A copy of the directions and a description of any risks and discomfort have been provided to me and have been fully explained to me by a member of the research team, and I understand the explanation. I have been given the opportunity to ask any questions I may have, and all such questions or inquiries have been answered to my satisfaction.

**If I have any questions about this study, I can contact John Henley, Peter R. Cavanagh, Ph.D., or Mary Becker, RN, at (814) 865 - 1972. Their offices are located in Room 10 Intramural Building, Penn State University, University Park, PA 16802.**

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<th>Investigator</th>
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9. **Compensation:**

I understand that, in the event of injury resulting from this investigation, neither financial compensation nor free medical treatment is provided for such injury by the University. Questions regarding this statement or your rights as a subject of this research should be directed to the Office for Regulatory Compliance in 115 Kern Building, University Park, PA (814) 865-1775.

10. **Voluntary Participation:**

I understand that my participation in this study is voluntary, and that I may withdraw from this study at any time by notifying the investigators. My withdraw from this study or my refusal to participate will in no way affect my care or access to medical services.

This is to certify that I consent and give permission for my participation as a volunteer in this program of investigation. I understand that I will receive a signed copy of this consent form. I have read this form, and understand the content of this consent form.

Date: ________________ Subject’s Signature: ___________________

I, the undersigned, have fully explained the investigation to the above volunteer.

Date: ________________ Investigator’s Signature: ___________________
Appendix C
Telephone Interview

Name: ___________________________ Subject Number: ____________
DOB: __________________________ Age: ______________
Address: __________________________ __________________________

Telephone: __________________________
Best time to call: __________________________
Date recruitment letter sent: __________________________
Did subject call:________________________ Date and time: __________________________
Date and time of attempts to reach the subject: __________________________

Briefly, describe the study to the person:
This experiment is designed to measure the mechanics, that is the forces and moments generated by the human body during sudden and unexpected ground movements. If you participate in this study, you will be asked detailed questions about your physical and psychological health and habits. You will be given a physical examination and you will be placed upon a movable platform. Your reactions to the moving platform will be video taped and measured by force sensors in the platform. The whole procedure will take about 2 hours.
Are you still interested in participating in this study? Y/N
Do you have any questions about the study? Y/N
Best time to participate (days and time)?: __________________________

Height: ______ (ft’ in”)

Weight: ___________ (lbs.)

Gender: M/F
Smoke > 7 cigarettes (cigar or pipe)/week within the last 3 months? Y/N
If no, have you ever smoked on a regular basis? Y/N
If yes, when did you stop? __________________________ Do you currently use tobacco, such as snuff or chew at least once a day? Y/N Have you ever used tobacco such as snuff or chew? Y/N
If yes, when did you stop? __________________________
How many alcoholic beverages do you drink in an average week? ____________
Over the last two years, has there been a sustained period of time (at least three months) when you drank more in an average week than the amount you just reported to me? Y/N
If yes, how many alcoholic beverages did you drink in an average week during that time?
When was this period?

Have you ever fallen when you were drinking?

How many caffeine-containing (coffee, coke, Pepsi) beverages do you drink in an average day?
Medical History
Are you taking any medications? Y/N
Names and actions:

Do you use recreational drugs? Y/N

Are you currently under the care of a physician for any ongoing medical or psychological problem? Y/N
If yes, explain:
Have you been diagnosed as having a mental or emotional disorder? Y/N
If yes, explain:

Do you currently have any type of infection? Y/N
If yes, explain:

Do you have any heart problems? Y/N
If yes, explain:

Do you have high blood pressure? Y/N
If yes, explain:

Do you have any lung or respiration problems? Y/N
If yes, explain:

Do you have problems with ringing in your ears? Y/N
If yes, explain:

Have you had any significant medical problems within the last 10 years? Y/N
If yes, explain:
Have you had any broken bones, surgery, or injury to your legs and feet?? Y/N
If yes, explain:

Do you have arthritis? Y/N
If yes, does it cause frequent pain or discomfort when you walk or stand? Y/N
If yes, explain:

Do you have any muscle, joint, or nervous system problem (for example stroke, TIA, Parkinson, etc.)? Y/N
If yes, explain:

Do you have any allergies (drug, food, bee sting others)? Y/N
If yes, explain:
Do you have good sensation in your feet? Y/N
If no explain:

Do you have pain in your feet without reason? Y/N
If yes, explain:

Do you have any painful blisters or cuts that hurt when you walk or stand? Y/N
If yes, explain:

Do you have any condition that affects you when you walk? Y/N
If yes, explain:
Do have any back pain? Y/N
If yes, explain:

Do you have any temporary injuries (shin splints, pulled muscles etc.)? Y/N
If yes, explain:

Do you use eyeglasses or contact lenses? Y/N
If yes, explain:

Do you have any damage to your retina (detached retina, muscular degeneration etc.)? Y/N
If yes, explain:

Do you ever see double when you are not tired? Y/N
If yes, explain:

Do you have any other problems that significantly affect your vision and are not corrected by glasses or contacts? Y/N
If yes, explain:

Have you ever had surgery to correct your vision? Y/N
If yes, explain:

Do you have any other eye problems? Y/N
If yes, explain:

When would it be possible for you to come in for an initial visit?
Date: ______________
Time: ______________
Appendix D
Medical History and Physiological Testing

Name: ____________________________ Subject Number: ____________
Date: ________________ Time: ________________
DOB: ________________ Age: ________________
Gender: ________________

Examine the subject’s gait as they walk in, to determine their suitability for the study: OK/NOT OK

Introduction
This experiment is designed to measure the mechanics, that is the forces and moments generated by the human body during sudden and unexpected ground movements. If you participate in this study you will be asked questions about your physical and health and habits. Your lower extremities will be examined and you will be placed upon a movable platform and your reactions to the moving platform will be video taped and measured by force sensors in the platform. The whole procedure will take about 2 hours.

Copy of consent form given to subjects: ____________
Sign consent form: ____________

Medical history
Are you taking any medications? Y/N
Names and actions:

Have you had any alcohol today? Y/N
If yes, explain:

Have you had any caffeine beverage today? Y/N
If yes, explain:

Smoke? Y/N
If yes, explain:

Do you have problems with ringing in your ears? Y/N
If yes, explain:

Have you had any significant medical problems within the last 10 years? Y/N
If yes, explain:

Have you had any broken bones, surgery, or injury to your legs and feet? Y/N
If yes, explain:

Do you have arthritis? Y/N
If yes, explain:

Do you have any muscle, joint or nervous system problem (for example, stroke TIAs, Parkinson, etc.)? Y/N
If yes, explain:

Do you have any allergies (drug, food, bee sting, others)? Y/N
If yes, explain:

Do you have any painful blisters or cuts that hurt when you walk or stand? Y/N
If yes, explain:

Do you have any condition that affects you when you walk? Y/N
If yes, explain:
Do you have any back pain?  Y/N
If yes, explain:

Do you have any temporary injuries (shin splints, pulled muscles, etc.)?  Y/N
If yes, explain:

Do you use eyeglasses or contact lenses?  Y/N
If yes, explain:

Do you have any other eye problems?  Y/N
If yes, explain:

Blood pressure sitting (after three minutes):  ________________
Blood pressure standing (after one minute):  ________________
Height:  __________ (ft’ in”) barefoot
Weight:  __________ (lbs.)

Eye examination:
Eye disease Y/N
If yes, explain:
Visual acuity:  __________

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Foot examination:
Height of lateral malleolus (cm)  ______

Blisters, sores, corns, callus: Y/N
Notes:

Nails: Y/N
Notes:

Monofilaments:
1 = 4.17 (mean + 2 SD)
2 = 5.07 (protective sensation)
3 = 6.10 (loss of protective sensation)
4 ≥ 6.10

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<td>knee at 90</td>
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<td>knee at 89</td>
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<td>ankle</td>
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### Appendix E

#### Postural Perturbation Studies

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<th>Dist</th>
<th>Vel</th>
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<td># of Trls</td>
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<td>8</td>
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<td>6 deg/s</td>
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<td>Nashner (1977)</td>
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<td>5</td>
<td>1.8°</td>
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Appendix F

Subjects’ Kinematics And Kinetics

In this appendix are graphs of the reactions of individuals to each trial. The first graph shows a subject’s reaction to each trial of the first condition. The second, third, fourth and fifth graphs show the time averages of the reactions to each condition. That is the trials combined by calculating the average of each trial reaction at each frame. Reaction at each frame of data is plotted. The reactions that are included in this appendix are the ankle angle, knee angle, hip angle, ankle moment, knee moment, and hip moment.
F.1. Ankle Angles

Figure F.1.1.1. Subject 1 ankle angle in response to condition one (full displacement/full power) each line represents a single trial

- Condition one (full displacement/full power)
- Condition two (half displacement/full power)
- Condition three (full displacement/half power)
- Condition four (half displacement/half power)

Figure F.1.1.2. Subject 1 ankle angle.
Figure F.1.2.1. Subject 2 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.1.2.2. Subject 2 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.3.1. Subject 3 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

- **Condition one (full displacement/full power)**
  - Angle (deg): 70, 75, 80, 85, 90, 95, 100
  - Time (s): 0, 50, 100, 150

- **Condition two (half displacement/full power)**
  - Angle (deg): 70, 75, 80, 85, 90, 95, 100
  - Time (s): 0, 50, 100, 150

- **Condition three (full displacement/half power)**
  - Angle (deg): 70, 75, 80, 85, 90, 95, 100
  - Time (s): 0, 50, 100, 150

- **Condition four (half displacement/half power)**
  - Angle (deg): 70, 75, 80, 85, 90, 95, 100
  - Time (s): 0, 50, 100, 150

Figure F.1.3.2. Subject 3 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.4.1. Subject 4 ankle angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)

![Graph showing ankle angle response to condition one](image1)

Condition two (half displacement/full power)

![Graph showing ankle angle response to condition two](image2)

Condition three (full displacement/half power)

![Graph showing ankle angle response to condition three](image3)

Condition four (half displacement/half power)

![Graph showing ankle angle response to condition four](image4)

Figure F.1.4.2. Subject 4 ankle angle (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.5.1. Subject 5 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.5.2. Subject 5 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.6.1. Subject 6 ankle angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.6.2. Subject 6 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.7.1. Subject 7 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.7.2. Subject 7 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.8.1. Subject 8 ankle angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.8.2. Subject 8 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.9.1. Subject 9 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.1.9.2. Subject 9 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.10.1. Subject 10 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power) vs. Condition two (half displacement/full power)

Figure F.1.10.2. Subject 10 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.11.1. Subject 11 ankle angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.11.2. Subject 11 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.12.1. Subject 12 ankle angle in response to condition one (full displacement/full power) each line represents a single trial

<table>
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<tbody>
<tr>
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Condition three (full displacement/half power) | Condition four (half displacement/half power)

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<td><img src="image3" alt="Graph" /></td>
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Figure F.1.12.2. Subject 12 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.13.1. Subject 13 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.13.2. Subject 13 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.14.1. Subject 14 ankle angle in response to condition one (full displacement/full power) each line represents a single trial

Figure F.1.14.2. Subject 14 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.15.1. Subject 15 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.15.2. Subject 15 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.17.1. Subject 17 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.17.2. Subject 17 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.18.1. Subject 18 ankle angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.18.2. Subject 18 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.19.1. Subject 19 ankle angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.19.2. Subject 19 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.20.1. Subject 20 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.20.2. Subject 20 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.21.1. Subject 21 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.21.2. Subject 21 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.22.1. Subject 22 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.22.2. Subject 22 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.1.23.1. Subject 23 ankle angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.23.2. Subject 23 ankle angle. (Positive = dorsiflexion, Negative = plantar flexion)
F.2. Knee Angles

![Diagram of knee angles over time for different conditions]

Figure F.2.1.1. Subject 1 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)    Condition two (half displacement/full power)

![Graphs showing knee angle over time for condition one and condition two]

Condition three (full displacement/half power)    Condition four (half displacement/half power)

![Graphs showing knee angle over time for condition three and condition four]

Figure F.1.1.2. Subject 1 knee angle (larger angles indicate flexion)
Figure F.2.2.1. Subject 2 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.1.2.2. Subject 2 knee angle (larger angles indicate flexion)
Figure F.2.3.1. Subject 3 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)

Condition two (half displacement/full power)

Condition three (full displacement/half power)

Condition four (half displacement/half power)

Figure F.2.3.2. Subject 3 knee angle (larger angles indicate flexion)
Figure F.2.4.1. Subject 4 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)

Condition two (half displacement/full power)

Condition three (full displacement/half power)

Condition four (half displacement/half power)

Figure F.2.4.2. Subject 4 knee angle (larger angles indicate flexion)
Figure F.2.5.1. Subject 5 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.5.2. Subject 5 knee angle (larger angles indicate flexion)
Figure F.2.6.1. Subject 6 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.6.2. Subject 6 knee angle (larger angles indicate flexion)
Figure F.2.7.1. Subject 7 knee angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.7.2. Subject 7 knee angle (larger angles indicate flexion)
Figure F.2.8.1. Subject 8 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)

Condition two (half displacement/full power)

Condition three (full displacement/half power)

Condition four (half displacement/half power)

Figure F.2.8.2. Subject 8 knee angle (larger angles indicate flexion)
Figure F.2.9.1. Subject 9 knee angle in response to condition one (full displacement/full power) each line represents a single trial.

<table>
<thead>
<tr>
<th>Condition One</th>
<th>Condition Two</th>
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<tbody>
<tr>
<td>(full displacement/full power)</td>
<td>(half displacement/full power)</td>
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</table>

Figure F.2.9.2. Subject 9 knee angle (larger angles indicate flexion)
Figure F.2.10.1. Subject 10 knee angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.10.2. Subject 10 knee angle (larger angles indicate flexion)
Figure F.2.11.1. Subject 11 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)                  Condition two (half displacement/full power)

Condition three (full displacement/half power)               Condition four (half displacement/half power)

Figure F.2.11.2. Subject 11 knee angle (larger angles indicate flexion)
Figure F.2.12.1. Subject 12 knee angle in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.2.12.2. Subject 12 knee angle (larger angles indicate flexion)
Figure F.2.13.1. Subject 13 knee angle in response to condition one (full displacement/full power) each line represents a single trial

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<thead>
<tr>
<th>Condition one (full displacement/full power)</th>
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<tr>
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<table>
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<th>Condition four (half displacement/half power)</th>
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<tr>
<td><img src="image3" alt="Graph" /></td>
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</table>

Figure F.2.13.2. Subject 13 knee angle (larger angles indicate flexion)
Figure F.2.14.1. Subject 14 knee angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.14.2. Subject 14 knee angle (larger angles indicate flexion)
Figure F.2.15.1. Subject 15 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.15.2. Subject 15 knee angle (larger angles indicate flexion)
Figure F.2.17.1. Subject 17 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.17.2. Subject 17 knee angle (larger angles indicate flexion)
Figure F.2.18.1. Subject 18 knee angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.18.2. Subject 18 knee angle (larger angles indicate flexion)
Figure F.2.19.1. Subject 19 knee angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.19.2. Subject 19 knee angle (larger angles indicate flexion)
Figure F.2.20.1. Subject 20 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.20.2. Subject 20 knee angle (larger angles indicate flexion)
Figure F.2.21.1. Subject 21 knee angle in response to condition one (full displacement/full power) each line represents a single trial.

<table>
<thead>
<tr>
<th>Condition one (full displacement/full power)</th>
<th>Condition two (half displacement/full power)</th>
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</thead>
<tbody>
<tr>
<td><img src="image1.png" alt="Graph" /></td>
<td><img src="image2.png" alt="Graph" /></td>
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<table>
<thead>
<tr>
<th>Condition three (full displacement/half power)</th>
<th>Condition four (half displacement/half power)</th>
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<tbody>
<tr>
<td><img src="image3.png" alt="Graph" /></td>
<td><img src="image4.png" alt="Graph" /></td>
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Figure F.2.21.2. Subject 21 knee angle (larger angles indicate flexion)
Figure F.2.22.1. Subject 22 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.22.2. Subject 22 knee angle (larger angles indicate flexion)
Figure F.2.23.1. Subject 23 knee angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.2.23.2. Subject 23 knee angle (larger angles indicate flexion)
F.3. Hip Angles

Figure F.3.1.1. Subject 1 hip angle in response to condition one (full displacement/full power) each line represents a single trial

Figure F.3.1.2. Subject 1 hip angle (positive indicates extension)
Figure F.3.2.1. Subject 2 hip angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.3.2.2. Subject 2 hip angle (positive indicates extension)
Figure F.3.3.1. Subject 3 hip angle in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.3.3.2. Subject 3 hip angle (positive indicates extension).
Figure F.3.4.1. Subject 4 hip angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.3.4.2. Subject 4 hip angle (positive indicates extension)
Figure F.3.5.1. Subject 5 hip angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.3.5.2. Subject 5 hip angle (positive indicates extension)
Figure F.3.6.1. Subject 6 hip angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)

Condition two (half displacement/full power)

Condition three (full displacement/half power)

Condition four (half displacement/half power)

Figure F.3.6.2. Subject 6 hip angle (positive indicates extension)
Figure F.3.7.1. Subject 7 hip angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  
Condition two (half displacement/full power)  
Condition three (full displacement/half power)  
Condition four (half displacement/half power)  

Figure F.3.7.2. Subject 7 hip angle (positive indicates extension)
Figure F.3.8.1. Subject 8 hip angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)

Condition two (half displacement/full power)

Condition three (full displacement/half power)

Condition four (half displacement/half power)

Figure F.3.8.2. Subject 8 hip angle (positive indicates extension)
Figure F.3.9.1. Subject 9 hip angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.3.9.2. Subject 9 hip angle (positive indicates extension)
Figure F.3.10.1. Subject 10 hip angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power) Condition two (half displacement/full power)

Condition three (full displacement/half power) Condition four (half displacement/half power)

Figure F.3.10.2. Subject 10 hip angle (positive indicates extension)
Figure F.3.11.1. Subject 11 hip angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.3.11.2. Subject 11 hip angle (positive indicates extension)
Figure F.3.12.1. Subject 12 hip angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.3.12.2. Subject 12 hip angle (positive indicates extension)
Figure F.3.13.1. Subject 13 hip angle in response to condition one (full displacement/full power) each line represents a single trial

Figure F.3.13.2. Subject 13 hip angle (positive indicates extension)
Figure F.3.14.1. Subject 14 hip angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)

Condition two (half displacement/full power)

Condition three (full displacement/half power)

Condition four (half displacement/half power)

Figure F.3.14.2. Subject 14 hip angle (positive indicates extension)
Figure F.3.15.1. Subject 15 hip angle in response to condition one (full displacement/full power) each line represents a single trial

**Condition one (full displacement/full power)**

![Graph showing hip angle over time for condition one](image1)

**Condition two (half displacement/full power)**

![Graph showing hip angle over time for condition two](image2)

**Condition three (full displacement/half power)**

![Graph showing hip angle over time for condition three](image3)

**Condition four (half displacement/half power)**

![Graph showing hip angle over time for condition four](image4)

Figure F.3.15.2. Subject 15 hip angle (positive indicates extension)
Figure F.3.17.1. Subject 17 hip angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power)    Condition two (half displacement/full power)

Condition three (full displacement/half power)    Condition four (half displacement/half power)

Figure F.3.17.2. Subject 17 hip angle (positive indicates extension)
Figure F.3.18.1. Subject 18 hip angle in response to condition one (full displacement/full power) each line represents a single trial

Condition one (full displacement/full power) Condition two (half displacement/full power)

Figure F.3.18.2. Subject 18 hip angle (positive indicates extension)
Figure F.3.19.1. Subject 19 hip angle in response to condition one (full displacement/full power) each line represents a single trial

Figure F.3.19.2. Subject 19 hip angle (positive indicates extension)
Figure F.3.20.1. Subject 20 hip angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)

Condition two (half displacement/full power)

Condition three (full displacement/half power)

Condition four (half displacement/half power)

Figure F.3.20.2. Subject 20 hip angle (positive indicates extension)
Figure F.3.21.1. Subject 21 hip angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.3.21.2. Subject 21 hip angle (positive indicates extension)
Figure F.3.22.1. Subject 22 hip angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)                  Condition two (half displacement/full power)

Condition three (full displacement/half power)               Condition four (half displacement/half power)

Figure F.3.22.2. Subject 22 hip angle (positive indicates extension)
Figure F.3.23.1. Subject 23 hip angle in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.3.23.2. Subject 23 hip angle (positive indicates extension)
F.4. Ankle Moments

Figure F.4.1.1. Subject 1 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)

Condition two (half displacement/full power)

Condition three (full displacement/half power)

Condition four (half displacement/half power)

Figure F.4.1.2. Subject 1 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.2.1. Subject 2 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  |  Condition two (half displacement/full power)

Figure F.4.2.2. Subject 2 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.3.1. Subject 3 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.4.3.2. Subject 3 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.4.1. Subject 4 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.4.4.2. Subject 4 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.5.1. Subject 5 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.4.5.2. Subject 5 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.6.1. Subject 6 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.4.6.2. Subject 6 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.7.1. Subject 7 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

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<thead>
<tr>
<th>Condition one (full displacement/full power)</th>
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</table>

Figure F.4.7.2. Subject 7 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.8.1. Subject 8 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.4.8.2. Subject 8 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.9.1. Subject 9 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.4.9.2. Subject 9 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)

Condition one (full displacement/full power)

Condition two (half displacement/full power)

Condition three (full displacement/half power)

Condition four (half displacement/half power)
Figure F.4.10.1. Subject 10 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.4.10.2. Subject 10 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.11.1. Subject 11 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

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<tr>
<th>Condition one (full displacement/full power)</th>
<th>Condition two (half displacement/full power)</th>
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<td><img src="image1" alt="Graph" /></td>
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Figure F.4.11.2. Subject 11 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.12.1. Subject 12 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Figure F.4.12.2. Subject 12 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.13.1. Subject 13 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.4.13.2. Subject 13 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.14.1. Subject 14 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.4.14.2. Subject 14 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.15.1. Subject 15 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.4.15.2. Subject 15 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.17.1. Subject 17 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Figure F.4.17.2. Subject 17 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.18.1. Subject 18 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.4.18.2. Subject 18 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.19.1. Subject 19 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

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<thead>
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Condition three (full displacement/half power)  
Condition four (half displacement/half power)

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<tr>
<th>Condition three (full displacement/half power)</th>
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</table>

Figure F.4.19.2. Subject 19 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.20.1. Subject 20 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.4.20.2. Subject 20 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.21.1. Subject 21 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Figure F.4.21.2. Subject 21 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.22.1. Subject 22 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.4.22.2. Subject 22 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
Figure F.4.23.1. Subject 23 ankle moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)                  Condition two (half displacement/full power)

Condition three (full displacement/half power)                  Condition four (half displacement/half power)

Figure F.4.23.2. Subject 23 ankle moment (Positive = dorsiflexion, Negative = plantar flexion)
F.5. Knee Moments

Figure F.5.1.1. Subject 1 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.5.1.2. Subject 1 knee moment (Positive = flexion, Negative = extension)
Figure F.5.2.1. Subject 2 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.5.2.2. Subject 2 knee moment (Positive = flexion, Negative = extension)
Figure F.5.3.1. Subject 3 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power) | Condition two (half displacement/full power)
---|---

Condition three (full displacement/half power) | Condition four (half displacement/half power)
---|---

Figure F.5.3.2. Subject 3 knee moment (Positive = flexion, Negative = extension)
Figure F.5.4.1. Subject 4 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Figure F.5.4.2. Subject 4 knee moment (Positive = flexion, Negative = extension)
Figure F.5.5.1. Subject 5 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.5.5.2. Subject 5 knee moment (Positive = flexion, Negative = extension)
Figure F.5.6.1. Subject 6 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)

Condition two (half displacement/full power)

Condition three (full displacement/half power)

Condition four (half displacement/half power)

Figure F.5.6.2. Subject 6 knee moment (Positive = flexion, Negative = extension)
Figure F.5.7.1. Subject 7 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.5.7.2. Subject 7 knee moment (Positive = flexion, Negative = extension)
Figure F.5.8.1. Subject 8 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.5.8.2. Subject 8 knee moment (Positive = flexion, Negative = extension)
Figure F.5.9.1. Subject 9 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power) | Condition two (half displacement/full power)

Condition three (full displacement/half power) | Condition four (half displacement/half power)

Figure F.5.9.2. Subject 9 knee moment (Positive = flexion, Negative = extension)
Figure F.5.10.1. Subject 10 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.5.10.2. Subject 10 knee moment (Positive = flexion, Negative = extension)
Figure F.5.11.1. Subject 11 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.5.11.2. Subject 11 knee moment (Positive = flexion, Negative = extension)
Figure F.5.12.1. Subject 12 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.5.12.2. Subject 12 knee moment (Positive = flexion, Negative = extension)
Figure F.5.13.1. Subject 13 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.5.13.2. Subject 13 knee moment (Positive = flexion, Negative = extension)
Figure F.5.14.1. Subject 14 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  
Condition two (half displacement/full power)  
Condition three (full displacement/half power)  
Condition four (half displacement/half power)  

Figure F.5.14.2. Subject 14 knee moment (Positive = flexion, Negative = extension)
Figure F.5.15.1. Subject 15 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  
Condition two (half displacement/full power)

Condition three (full displacement/half power)  
Condition four (half displacement/half power)

Figure F.5.15.2. Subject 15 knee moment (Positive = flexion, Negative = extension)
Figure F.5.17.1. Subject 17 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.5.17.2. Subject 17 knee moment (Positive = flexion, Negative = extension)
Figure F.5.18.1. Subject 18 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.5.18.2. Subject 18 knee moment (Positive = flexion, Negative = extension)
Figure F.5.19.1. Subject 19 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.5.19.2. Subject 19 knee moment (Positive = flexion, Negative = extension)
Figure F.5.20.1. Subject 20 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power) Condition two (half displacement/full power)

Condition three (full displacement/half power) Condition four (half displacement/half power)

Figure F.5.20.2. Subject 20 knee moment (Positive = flexion, Negative = extension)
Figure F.5.21.1. Subject 21 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.5.21.2. Subject 21 knee moment (Positive = flexion, Negative = extension)
Figure F.5.22.1. Subject 22 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.5.22.2. Subject 22 knee moment (Positive = flexion, Negative = extension)
Figure F.5.23.1. Subject 23 knee moment in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.5.23.2. Subject 23 knee moment (Positive = flexion, Negative = extension)
F.6. Hip Moments

Figure F.6.1.1. Subject 1 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)

Figure F.6.1.2. Subject 1 hip moment (Positive = extension)
Figure F.6.2.1. Subject 2 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)          Condition two (half displacement/full power)

Condition three (full displacement/half power)      Condition four (half displacement/half power)

Figure F.6.2.2. Subject 2 hip moment (Positive = extension)
Figure F.6.3.1. Subject 3 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.3.2. Subject 3 hip moment (Positive = extension)
Figure F.6.4.1. Subject 4 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.4.2. Subject 4 hip moment (Positive = extension)
Figure F.6.5.1. Subject 5 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.5.2. Subject 5 hip moment (Positive = extension)
Figure F.6.6.1. Subject 6 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.6.6.2. Subject 6 hip moment (Positive = extension)
Figure F.6.7.1. Subject 7 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.7.2. Subject 7 hip moment (Positive = extension)
**Figure F.6.8.1.** Subject 8 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

**Figure F.6.8.2.** Subject 8 hip moment (Positive = extension)
Figure F.6.9.1. Subject 9 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

<table>
<thead>
<tr>
<th>Condition one (full displacement/full power)</th>
<th>Condition two (half displacement/full power)</th>
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</table>

<table>
<thead>
<tr>
<th>Condition three (full displacement/half power)</th>
<th>Condition four (half displacement/half power)</th>
</tr>
</thead>
</table>

Figure F.6.9.2. Subject 9 hip moment (Positive = extension)
Figure F.6.10.1. Subject 10 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)

Condition two (half displacement/full power)

Condition three (full displacement/half power)

Condition four (half displacement/half power)

Figure F.6.10.2. Subject 10 hip moment (Positive = extension)
Figure F.6.11.1. Subject 11 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Figure F.6.11.2. Subject 11 hip moment (Positive = extension)
Figure F.6.12.1. Subject 12 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.12.2. Subject 12 hip moment (Positive = extension)
Figure F.6.13.1. Subject 13 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.13.2. Subject 13 hip moment (Positive = extension)
Figure F.6.14.1. Subject 14 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.14.2. Subject 14 hip moment (Positive = extension)
Figure F.6.15.1. Subject 15 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

<table>
<thead>
<tr>
<th>Condition one (full displacement/full power)</th>
<th>Condition two (half displacement/full power)</th>
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<tbody>
<tr>
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<td><img src="image2.png" alt="Graph" /></td>
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<tr>
<td>Condition three (full displacement/half power)</td>
<td>Condition four (half displacement/half power)</td>
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<tr>
<td><img src="image3.png" alt="Graph" /></td>
<td><img src="image4.png" alt="Graph" /></td>
</tr>
</tbody>
</table>

Figure F.6.15.2. Subject 15 hip moment (Positive = extension)
Figure F.6.17.1. Subject 17 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.17.2. Subject 17 hip moment (Positive = extension)
Figure F.6.18.1. Subject 18 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power) Condition two (half displacement/full power)

Condition three (full displacement/half power) Condition four (half displacement/half power)

Figure F.6.18.2. Subject 18 hip moment (Positive = extension)
Figure F.6.19.1. Subject 19 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.19.2. Subject 19 hip moment (Positive = extension)
Figure F.6.20.1. Subject 20 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.20.2. Subject 20 hip moment (Positive = extension)
Figure F.6.21.1. Subject 21 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.21.2. Subject 21 hip moment (Positive = extension)
Figure F.6.22.1. Subject 22 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.22.2. Subject 22 hip moment (Positive = extension)
Figure F.6.23.1. Subject 23 hip moment in response to condition one (full displacement/full power) each line represents a single trial.

Condition one (full displacement/full power)  Condition two (half displacement/full power)

Condition three (full displacement/half power)  Condition four (half displacement/half power)

Figure F.6.23.2. Subject 23 hip moment (Positive = extension)
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