EFFECT OF BASE OF SUPPORT SIZE ON ABILITY TO RECOVER BALANCE

by

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ABSTRACT

Humans' ability to recover balance should depend on the size of the base-of-support (BOS) between the feet and ground. To test this hypothesis, I conducted experiments where subjects (n=15) were released suddenly from an inclined position by means of a tether and electromagnet, and recovered upright stance using the feet-in-place ankle strategy or mixed (hip/ankle) strategy. I varied the size of the available BOS by adjusting the length of a block that the subject stood upon. I found that the maximum angle where subjects were able to recover balance (THETA_MAX) declined from 8.8 to 7.3 deg when BOS decreased from 100% to 75%, and from 7.3 to 5.0 deg when BOS decreased from 75% to 50%. THETA_MAX was 19% larger for the mixed than ankle strategy. However, recovery strategy did not influence the effect of BOS on THETA_MAX. These results confirm that BOS size strongly influences ability to recover balance.

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CHAPTER ONE
INTRODUCTION AND LITERATURE REVIEW

Introduction:

The bipedal posture of the standing human is inherently unstable. The centre of gravity (COG) of the body is located at approximately five-ninths of the body height from the ground over a narrow base of support (BOS). This imposes critical demands on the postural and balance control system [78].

Less than three decades ago, falls and their related injuries were largely considered to be an inevitable consequence of aging. However, over the past 30 years, investigators around the world have contributed a wealth of evidence on the physical and psychological factors associated with falling, affirming the predictability of fall risk [66]. But there still remains a great disconnection between the wealth of evidence supporting fall prevention efforts and real-world practice. The clinical and public health integration of fall prevention programs has lagged behind the research findings.

An important prerequisite to design an effective fall prevention program is improved understanding of biomechanical variables that govern ability to maintain and recover balance [55]. It is well known that falling is associated with a variety of sensory, motor, cognitive, and psychosocial variables. However, risk for falls ultimately depends on the frequency of loss of balance episodes and the ability to recover balance[55]. Fall prevention programs therefore need to evaluate and target each of these areas.

To be able to recover balance requires performing the recovery task rapidly and with substantial strength. Previous studies have shown that both peak lower extremity joint torques and rates of torque development influence the effectiveness of the balance recovery response [55, 60, 61, 79]. In this thesis, I provide a more detailed explanation of how lower extremity joint torque affects balance recovery ability in young women by altering the size of the support surface. My results enhance the understanding of the biomechanical variables that govern postural stability, and ultimately improve our ability to develop fall prevention interventions.

1.1 Epidemiology of fall-related injuries

1.1.1 Definition of fall, balance and postural control

For the purpose of this thesis, I define a fall as "unintentionally coming to the ground or some lower level and other than as a consequence of sustaining a violent blow, loss of consciousness, sudden onset of paralysis as in stroke or an epileptic seizure [15]".

I define "balance" as the ability to maintain the body in equilibrium, and "postural control" as the ability to control the body's position in space for the dual purposes of stability and orientation [58]. Postural control involves complex interactions between musculoskeletal and neural systems (Figure 1).

1.1.2 Incidence of falls and consequence of falls

Falls are the number one cause of injury-related deaths and hospitalizations in Canada [25]. Approximately one in three seniors living in the community will experience at least one fall per year, and up to 50% of these individuals will experience repeated falls. Falls are responsible for about two-thirds of all injury-related discharges from hospital, more than 70% of injury-related days of hospital care (usually to treat fractures of the lower extremities), and more than half of all deaths due to injury for Canadians over the age of 65 [25]. Falls are also the underlying cause of the vast majority of hip fractures in the elderly. Each year, there are 24,000 hip fractures annually in Canada [76].

1.1.3 Cost of falls

Fall-related injury is a serious economic burden. The annual cost of fall-related injuries in Canada is approximately $3.6 billion annually in Canada. In British Columbia, roughly $180 million (85 percent) of the direct cost of elder injuries is attributable to falls [25]. Falls not only create a huge economic burden, but also dramatically change an elderly person's self-confidence, motivation, and ability to function independently. With the increase in our aging population and with increased life expectancy of our elderly, the importance of maintaining mobility is becoming ever more critical. Effective approaches to prevent falls and promote safe mobility and independence need to be developed.

1.2 Risk factors for falls and prevention intervention

Most falls in the elderly are not associated with obvious environmental hazards, but instead occur during the performance of routine activities such as walking, turning, rising, and bending [65]. Individual risk factors for falling include older age, medication usage, specific chronic disease (including arthritis, Parkinson's syndrome, and stroke), and impairments in muscle strength, joint movement, balance, gait, vision, hearing, and cognition [4, 8, 64, 67]. To develop more effective fall-injury prevention techniques, we need a better understanding of how these risk factors affect frequency of loss of balance episodes, and ability to recover balance [55].

1.3 Biomechanics of balance recovery

1.3.1 Centre of gravity and centre of pressure

The centre-of-gravity (COG) of a human body segment is the net location of its centre-of-mass (a balance point representing the mean position of all matter in the body). The COG of the whole body is the weighted average of the COG of each individual body segment. During standing or walking, the centre-of-pressure (COP) is the centroid of the vertical pressure acting between the feet and the ground. In static situations, the COG and COP projection must be vertically aligned, and the vertical component of ground reaction force must be equal and opposite to the weight of the whole body.

However, the human body is never static even during "quiet stance". The horizontal position of the COG constantly moves back and forth or "sways" (Figure 2) and the COP moves in phase with the COG, but with a larger amplitude [77].

1.3.2 Balance recovery strategy

Postural control strategies may be either reactive or predictive, or a combination of both [39]. A predictive postural control strategy might involve a voluntary movement or increase in muscle activity in anticipation of a predicted disturbance. A reactive postural control strategy would involve a movement or muscular response following an unpredicted disturbance. Maki categorized reactive balance recovery strategies into two distinct classes: 1) fixed-support strategies, in which the base of support (BOS, the area that the feet contact with the ground) remains unaltered, and 2) change-in-support strategies, in which the BOS is moved [39]. Sway responses, such as the ankle strategy (Figure 3A), or hip strategy (Figure 3B) are the common fixed-support strategies, while grasping or stepping (stepping strategy) are common change-in-support strategies.

In general, balance recovery strategies are selected according to the size of the postural challenge. The ankle strategy is dominant for the smallest disturbances, whereas the hip strategy is more important when disturbances are larger or when the BOS is limited (Figure 4) [19]. If the COG is moved out of the BOS, a change-in-support strategy must be used to regain balance. The most common balance recovery strategy is the stepping strategy, however subjects have been observed to initiate stepping well before the COP reaches the limit of the BOS [50]. This suggests that both physiological and psychological factors influence step initiation.

1.3.2.1 Ankle strategy and hip strategy

The ankle strategy restores equilibrium by moving the body as a single-link inverted pendulum via moments produced at the ankle joints. It is characterized by early activation of ankle muscles followed by a distal to proximal sequence of muscle activation in the thigh and trunk on the same dorsal or ventral aspect of the body (Figure 3A). The hip strategy, on the other hand, restores equilibrium by moving the body as a double-link inverted pendulum via moments produced at the hip joints. The sequence of muscle activation with the hip strategy is proximal to distal, starting from early activation of trunk muscles to ankle muscles on the same aspect of the body (Figure 3B).

The ankle strategy is most useful for slow balance recovering from small perturbations on a firm, even surface. Normal adults can use an ankle strategy to recover from as much as 8 degrees of forward sway and 4 degrees of backward sway [22]. The hip strategy is useful for larger amplitude perturbations and under conditions where it is difficult to produce ankle torque, such as standing on a narrow beam or compliant surface. Kuo and Zajac [31] developed a computational model to study coordination of ankle and hip strategies. They found that the hip strategy can produce COG accelerations that are three times greater than those generated by the ankle strategy [32]. Thus, the hip strategy permits faster movement of the COG than does the ankle strategy. They also found that the hip strategy required less energy than the ankle strategy to move the COG.

Selection of balance recovery strategy depends on biomechanical, sensory and neuromuscular constraints, as well as on environmental context, and prior experience. When the support surface characteristics are suddenly changed, normal adults continue to use the same strategy initially, and gradually switch to the new strategy, showing a mixture of two strategies before finally adopting the most efficient strategy for the new environmental context [18]. Furthermore, the ankle and hip strategies represent extremes of a response continuum in which subjects may combine hip and ankle strategies in any proportion. Runge et al. [57] computed torques generated at the hip, knee and ankle joints in response to postural perturbations occurring at different velocities. Both hip and ankle torques were present in all the trials for all subjects examined. No pure hip strategy existed. All postural correction movements fell into the mixed strategy with a greater reliance on the ankle joint action for weaker perturbations, and increased contribution from hip joint action as perturbations became stronger. But it still remains unclear how ankle and hip strength influence the effectiveness of the ankle and mixed strategies.

1.3.3 Base of support (BOS)

Daily activities require a person to control the position of the COG over the BOS. In biomechanical terms, most falls are caused by movements (and lack of appropriate corrective actions) that either displace the COG beyond the BOS, as occurs during a trip, or displace the BOS away from a relatively stationary COG, as occurs during a slip.

1.3.3.1. Physical BOS & functional BOS

The "Physical BOS" during standing is the area that the feet contact with the ground. The functional BOS is defined as the proportion of the physical base of support where the body can be supported in a stable manner.

1.3.3.2 Functional BOS

Functional BOS declines with age and is associated with ankle strength [28, 35]. The theoretical basis for this association is illustrated in the following moment balance, which assumes quasi-static conditions for standing balance (Figure 5):

$$T_a - W \times e + F \times p + F_H \times h = 0 \qquad (1.1)$$

where T_a is the ankle torque, F is the resultant vertical force acting on the foot (defined positive if upward), p is the horizontal distance from the ankle joint to the location where F acts (i.e., the location of the COP, defined positive if anterior to the ankle), F_H is the resultant horizontal shear force in the sagittal plane acting on the foot (defined positive if directed posteriorly), W is the weight of the feet, h is the height of the ankle joint above the ground, and e is the distance from foot centre of gravity to the ankle joint.

The foot weight W and the shear force F_H are both small compared to the vertical force F, while the moment arms, e, p, and h are of comparable magnitude [55]. Accordingly, the ankle torque can be estimated as:

$$T_a \approx -F \times p \qquad (1.2)$$

which shows that the ankle torque, T_a is directly proportional to the location p of the COP, and the vertical force F.

In a static situation, the projection of the COG must be located within the functional BOS, and directly in line with the COP. In a dynamic situation, the COG can be displaced outside of the BOS, and stability can be maintained (without stepping) if the horizontal velocity of the COG is appropriate (as in the case of rising from sitting [49]).

Nashner and McCollum [47] investigated the BOS constraints related to foot geometry. They observed that the length of the foot limits the ankle torque that one can generate (Equation 1.2), therefore, in situations where the feet are effectively shortened (such as standing on a narrow bean), there tends to be greater reliance on the hip strategy for maintaining balance. This prediction agreed with experimental result shown in a later study [18]. Lee [35] found that the maximal excursion of the COP during leaning correlated negatively with age for all directions of leaning (forward, backward, right and left). King et al.[28] found that the functional BOS, as measured by COP displacement on a force platform during sustained forward and backward leaning, decreased with age from 60% of foot length in young adults to 42% of foot length in subjects over age 60.

1.3.4 Strength versus speed of response

Previous experimental and epidemiological studies have shown that variables related to both muscle strength and speed of muscle force development associate with measures of postural stability [17, 55, 61, 68] and with risk for falls among the elderly [36, 48]. To guide the development of improved fall prevention programs, we must identify the biomechanical, sensorimotor, and cognitive variables that influence muscle strength and speed of response [17]. These factors are briefly reviewed in the following sections.

1.4 Neuromuscular risk factors for falls

1.4.1 Age changes in muscle strength

The loss of muscle strength with age, even in healthy and physically-active elderly, has long been recognized [1, 33, 53, 59, 60, 75, 81]. Isometric strength peaks at about age 25, and then declines gradually. So that, by age 65, an individual is two-thirds as strong (on average) as they were at age 25 [43]. Aging beyond the sixth decade is marked by a reduction in the number of excitable motor neurons and motor units, and subsequent motor unit remodeling [29]. There is also slowing of motor unit firing rates with age, especially during maximal voluntary contractions [42, 56, 74], and a loss of type I and, more predominantly type II muscle fibers [69, 74]. This influences both strength [69] and speed of ankle torque generation [60].

Such changes have been implicated as major cause of falls in the elderly. For example, Whipple et al. [75] showed that elderly nursing home residents who had a history of falls had lower ankle strength than age-matched nursing home control subjects with no history of falls. The mean doriflexion muscle power of fallers was only 14% of the non-fallers' values.

However, there is strong evidence that strength can be increased in both active and frail elderly who participate in strength-training programs [14, 27], and that such programs can reduce the incidence of falls in elderly subjects [10, 16, 63]. To date, the largest study in this area has been the FICSIT trials (Frailty and Injuries: Cooperative Studies of Interventions Techniques) [52]. This complimentary set of studies found that exercises that included balance training (e.g. Tai Chi) caused a 25% reduction in fall rates [52], while those which focused on muscle strengthening had no effect [52]. It has been suggested that normal age-related loss of muscle strength does not greatly impair the ability to successfully recover balance with feet-in-place and stepping reactions [17, 38, 51]. However, there appears to be a strength threshold below which balance recovery abilities are impaired greatly [51, 80].

1.4.2 Reaction time

Reaction time is the time interval between the onset of a perturbation and the onset of a corrective response (which may be defined, for example, as the onset of torque development or the initiation of movement). Reaction time increases by approximately 25% from age twenty to the sixty [72], and increased reaction time has long been recognized as a risk factor for falling in the elderly [71, 73]. Reaction time is dependent on the nature of the tasks. Reaction times during balance recovery are slower than for a monosynaptic stretch reflex, and faster than for a voluntary movement [18].

Reaction times are categorized into simple and choice reaction time and further fractionated into premotor time and electromechanical delay (motor time) [70]. Premotor time is defined as the time delay from the onset of a perturbation of go cue to the onset of increased muscle activity as measured by electromyography (EMG). It relates to sensory detection of the perturbation, transmission and processing of afferent signals, travel of efferent signals, and the subsequent increase in the number and frequency of action potentials in the muscle [70]. Electromechanical delay represents the subsequent time required to develop increased muscle force in response to an increase in activation [70]. Studies have shown that the electromechanical delays are not affected by age and level of activity. [5, 9, 70]. However, aging does cause slowing of premotor time [70]

1.4.3 Rate of force development in muscle

Balance recovery tasks require the ability to generate an adequate torque in a limited amount of time. Therefore, one's ability to recover balance should depend not only on the torque generation magnitude, but also on how quickly the torque can be developed [17, 55]. By using a combination of experimental and mathematical modelling techniques, Robinovitch et al. [55] compared the relative importance of strength versus speed-of-response variables to balance recovery ability with the ankle strategy. They demonstrated the strong effect on recovery ability of torque generation rate, as well as the magnitude of torque development. Previous studies also found that the rate of torque generation decreases with age. Mackey et al. [37] demonstrated a 16% decrease of the average rate of ankle torque generation in elderly subjects compared with young subjects. Mackey et al. measured the rate of ankle torque generation during upright stance in response to a perturbation by using the ankle strategy. Thelen et al. [60] also found that old adults had maximum rates of ankle torque development that were 30 to 40% slower than young adults.

1.5 Sensory systems

Three major sensory systems are involved in balance and posture control: the visual, vestibular and somatosensory systems. Signals from these sensory systems are integrated and resolved by the central nervous system (CNS) to provide an accurate estimate of the body's orientation, which is necessary for maintaining equilibrium.

Visual inputs provide information regarding the position and motion of the head with respect to surrounding objects; vestibular receptors provide information on the orientation and movement of the head relative to inertia and gravity [58]. Somatosensory afferents include mechanoreceptors in the skin, pressure receptors in deep tissues, muscle spindles, Golgi tendon organs, and joint receptors [58]. These organs provide information about joint position, external loads, and the orientation and movement of the body parts relative to each other [58].

Sensory information is also used for detecting perturbations and triggering compensation to the perturbation. Changes in muscle activation occur as quickly as 90-110 ms following platform movement [46], likely triggered by muscle receptors. Both visual and vestibular inputs trigger slower compensations (185-250 ms), while providing an absolute orientation reference for comparing proprioceptive inputs. However, muscle receptor and visual inputs can provide false information leading to unnecessary postural adjustments which destabilize standing posture. Such inappropriate postural adjustments are observed during movements of a visual surround [34] and rotations of the support surface on which subjects stand [46]. However subjects are able to attenuate these responses after several exposures by comparing misleading sensory inputs to vestibular inputs.

Sensory information affects selection of the balance recovery strategy. Subjects with profound loss of vestibular function are unable to use a hip strategy to stand on narrow beams, in tandem stance, or on one foot [24]. Reduced somatosensory input from the soles of the feet causes an increased reliance on the hip strategy [23]. Horak and her colleagues also found elderly subjects with multiple sensory loss used the stepping strategy for postural correction even in response to very small, slow perturbations in which an ankle or mixed ankle-hip strategy was normally used [20, 22].

1.6 Central neural control of balance recovery

Adaptation by the nervous system occurs in balance control when subjects are repeatedly exposed to external perturbations. For example, in one of Nashner's platform studies [46], subjects stood upon a platform, which could tilt in the sagittal plane about the axis of the ankle joint, translate in the anterior or posterior direction, or perform both motions simultaneously. Task specific differences of reflex function were investigated by experiments in which the role of stretch reflexes to stabilize sway during stance could be altered to be useful, of no use, or inappropriate. This study found that stretch reflex responses were in themselves not sufficient to prevent a loss of balance under large perturbation conditions. In the posterior platform perturbations, the soleus stretch reflex assisted subjects in returning to a stable upright stance. However, in the posterior tilt condition, the soleus stretch reflexes would inappropriately contribute to the loss of balance (appropriate response would be little or no contraction of the ankle muscles), as occurred in the first few trials. However, after 3-5 trials, subjects were able to adopt the correct response by quieting the soleus stretch reflex. Certainly, a higher centre was regulating the reflex gains based on experience and other sensory information.

Central control over response magnitude is important because postural muscle responses are often initiated before the availability of peripheral information characterizing the full nature of the stimulus [21]. The term central set refers to the preparatory state of the nervous system, based on current and expected task conditions, and factors such as arousal and motivation [6]. The ability to utilize prior knowledge, or experience with a predictable stimulus to modify both automatic and voluntary responses has been referred to as the central-set effect [21]. Central set enables descending commands to specify or modulate aspects of a postural response in advance of a perturbation [21], based on prior experience with the expected perturbation and memory of the effectiveness of prior responses.

Muscle activation following a perturbation is influenced by a variety of input signals (descending, intersegmental, and segmental) to motoneuron pools. For example, when a limb is subjected to an external perturbation, a reflex response occurs in the muscles that are stretched. If the movement stretches muscles that are undergoing voluntary contraction, it gives rise to multiple peaks in the rectified and averaged EMG. The first peak (M1) is considered to be due to the monosynaptic spinal stretch reflex because its latency is compatible with monosynaptic activation involving group I_a spindle afferents and γ-motor efferents [11]. The second peak (M2) originates from either the slower conducting secondary (group II) muscle spindle afferent fibers or from afferent terminals in the skin and subcutaneous tissues [11, 41]. There is often a third burst (M3) after muscle stretch, which has been interpreted as a voluntary response because it is of the approximate latency of the reaction time, and its size depends on the central set [11]. In normal subjects, the stretch reflex in leg and trunk muscles is seen at 45-50 ms after a perturbation. This provides little resistive force [46]. A more rigorous voluntary response is seen at 60-80 ms which produces sustained force [26].

1.7 Hypothesis

Previous studies have shown that the functional BOS decreases with age and with declines in ankle strength [28, 35]. However, there is little understanding of how reduced BOS size affects balance ability to recover, and how this depends on the type of balance recovery strategy utilized by individuals. This is an important question for the design of exercise-based fall prevention programs in the elderly.

Accordingly, the goal of the current thesis was to quantify this relationship. We considered that BOS should influence peak ankle torque and therefore ability to recover balance, but the effect should be greater for the ankle strategy than for a mixed strategy that incorporates both the ankle strategy and the hip strategy (figure 6). My experiments were therefore designed to test the following hypotheses: 1) the ability to recover balance decreases as BOS size declines; and 2) the effect is greater in the ankle strategy than in the mixed strategy. To test these hypotheses, we conducted tether release experiments to measure the maximum release angle where participants could recover balance with three different BOS sizes (100%, 75% and 50% of anterior foot length), and using either the ankle strategy or the mixed strategy.

CHAPTER TWO
METHODS

2.1 Participants

Fifteen young female participants participated in the study, having a mean age of 23 ± 5 (S.D.) yrs (range: 19-35 yrs), mean body mass of 56.8 ± 10.9 kg (range: 43.6-74.6 kg), and mean height of 1.7 ± 0.1 m (range: 1.5-1.8 m). Participants were recruited from posting of notices at Simon Fraser University and British Columbia Institute of Technology. Each participant provided informed written consent, and the experiment was approved by the Research Ethics Board of Simon Fraser University.

Participants were screened by telephone interview to ensure they were in good general health with no known neurological or muscular disorders. Eligible participants were scheduled to come to the Injury Prevention and Mobility Laboratory at Simon Fraser University. Ancillary measures were acquired of peak attainable ankle dorsiflexor torque, peak attainable ankle plantarflexor torque, peak hip extensor torque, and peak hip flexor torque, all under isometric conditions. Measures were then acquired of ability to recover balance, as described below in section 2.2. Participants were paid $10 per hour for their participation.

2.2 Protocol

During the balance recovery trials, participants stood on a 12 cm high wooden block mounted on a 90 cm × 60 cm force plate, with their feet at shoulder width and arms crossed against their chest (Figure 7). The participant was inclined in a stationary forward leaning position held by a horizontal tether that attached at one end to an electromagnetic brake (Warner Electric model PB500, South Beloit, IL 61080) and at the other end to a chest harness worn by the participant. The height of the electromagnetic brake was adjusted to equal the height of the tether attachment point on the harness. The instant of tether release was detected as the onset of a sharp decline in the tension measured by a load cell (Sensotec, model 31) located in series with the tether (time required for tether force to decay to 90% of initial force, ~15 ms). To increase the unexpectedness of the perturbation, I inserted a random time delay of 1-3 sec between receiving a "ready" cue from the participant and the time of release. Prior to brake release, I instructed the participants that, upon release of the tether, they should recover a vertical standing position by either a) contracting the muscles spanning the ankles while keeping the hips and knees extended (ankle strategy trials), or by b) flexing the trunk forward and rotating the ankles while keeping the knees extended (mixed strategy trials). To train appropriate body posture and movement, each participant participated in three practice trials for each strategy, which involved a lean angle of about 2 degrees, from which all participants could recover easily.

For both the ankle strategy and mixed strategy, I determined the maximum release angle where participants could recover balance for three different BOS sizes (100%, 75% and 50% of anterior foot length), where anterior foot length is defined as the distance from longest toe to the ankle joint. In all trials, participants stood upon a wooden block mounted on the force plate. In the 100% BOS condition, the participant's feet were completely on the block. In the 75% BOS and 50% BOS conditions, either 75% or 50% of the anterior foot length, respectively, was in contact with the block. Each participant was tested under all three BOS conditions. By manipulating the BOS between these 3 conditions, I was able to control the maximum ankle torque the participant was capable of obtaining during balance recovery, as indicated in Equation 1.2. Before each trial, both the participant and investigator monitored the centre-of-pressure (COP) position and magnitude (via an oscilloscope located at eye level) so that it was close as possible to its position measured during upright standing. I also monitored the tether force and repeated trials where there was an obvious decline in the tether force before its release, indicating anticipation.

To determine the maximum release angle (θ_{max}) for a given condition, I iteratively adjusted the length of the tether and the corresponding lean angle (θ_i) until I identified the maximum value where the participant could recover balance in 3 out of 5 trials (with a resolution of 7 mm in tether length and approximately 0.3 degrees in the lean angle). Each trial took about 30 seconds to complete. A break time of 2 minutes was given every ten trials and between conditions to minimize muscle fatigue. The time interval between consecutive trials was less than one minute, which included the time required to adjust the tether, repeat the experimental instructions, and receive the "ready" cue from the participant.

During each trial, I used a 60 Hz, six camera motion capture system (Qualysis Inc., Glastonbury, CT) to record positions of 16 skin surface markers. These were located at the right and left fifth toe (metatarsal), ankle (lateral malleolus), knee (lateral femoral epicondyle), hip (anterior supeior iliac spine), shoulder (acromium), elbow (radial head), and wrist (junction between ulna and radius). I used a force plate (model 6090H, Bertec Corp., Worthington, OH) to measure the magnitude and point of application of reaction forces between the feet and the ground, at a rate of 960 Hz. I also recorded muscle activity with electromyographic surface electrodes located over the lateral gastrocnemius and tibialis anterior muscle of the dominant kicking leg.

2.3 Data analysis

For each maximum release angle trial, I calculated θ_{max} as the average value of the angle from the vertical to a line in the sagittal plane connecting the average position of the right and left ankles to the position of whole body COG over the 500 ms interval preceding tether release (Figure 7). The baseline value of the angle during quiet standing was subtracted for each trial.

I also calculated temporal variations in ankle plantarflexor torque $T_a(t)$ using the following modified version of Equation 1.1 suitable for dynamic conditions (Figure 5):

$$T_a(t) = W \times e(t) - F(t)p(t) - F_H(t)h(t) + I\ddot{\theta}(t) \tag{2.1}$$

where F is the resultant vertical force acting on the foot (defined positive if upward), p is the horizontal distance from the ankle joint where F acts (defined positive if anterior to the ankle), F_H is the resultant horizontal force in the sagittal plane acting on the foot (defined positive if directed posteriorly), h is the vertical height of the ankle above the ground, I is the foot moment of inertia, $\ddot{\theta}$ is the foot angular acceleration, W is the foot weight, and e is the distance from the foot centre of gravity to the ankle joint.

I calculated the COG stop time as the time interval between tether release and the time of maximum COG displacement in the anterior posterior direction. I calculated the ankle angle as the relative angle between the shin and foot segments, the hip angle as the relative angle between the thigh and trunk segments, the LEG angle as the absolute angle of the shin with respect to the vertical, and HAT angle as the absolute angle of the HAT segment with respect to the vertical. Angles were defined positive in extension (or plantarflexion at the ankle), and joint torques were defined positive if the contracting muscles are extensors (or plantarflexor at the ankle).

Based on these kinetic, kinematic and EMG variables, I calculated the following parameters (Figure 8): ***1)*** baseline ankle torque (T_{A0}) averaged over the 500 ms preceding tether release; ***2)*** peak ankle torque during balance recovery (T_{Amax}); ***3)*** ankle torque generation rate following release (I_A), defined as the slope of a straight line joining torque-time values at the instant ankle torque exceeded T_{A0} by 3 standard deviations and the instant T_A exceeded T_{A0} by 85% of the difference between T_{Amax} and T_{A0}; ***4)*** ankle torque decline rate (D_A) following T_{Amax}, defined as the slope of a straight line joining torque-time values at the instant of T_{Amax} to the instant T_A declined by 85% of the difference between T_{Amax} and T_{A0}. ***5)*** ankle torque response time (ART), calculated as the time interval between tether release and the instant ankle torque exceeded T_{A0} by 3 standard deviations (as measured during the 500 ms before release). ***6)*** premotor time (PMT), defined as the time interval from tether release to the time that lateral gastrocnemius activity exceeded 3 standard deviations above its baseline value measured during the 500 ms interval preceding tether release.; ***7)*** electromechanical delay (EMD), defined as the time interval from the onset of increased gastrocnemius activity to the onset of increased ankle torque (EMD = ART-PMT).

For mixed strategy trials (Figure 8), I calculated the followings additional variables: ***8)*** baseline hip torque (T_{H0}), defined as the average value of hip torque during the 500 ms interval preceding tether release; ***9)*** peak hip torque (T_{Hmax}) during balance recovery; ***10)*** hip torque increase rate (I_H), defined as the slope of a straight line joining the torque-time values at the instant hip torque exceeded T_{H0} by 3 standard deviations; and the instant T_H exceeded T_{H0} by 85% of the difference between T_{Hmax} and T_{H0}; ***11)*** hip torque decline rates (D_H) following T_{Hmax}, defined as slope of a straight line joining torque-time values at the instant of T_{Hmax} to the instant T_H declined to 85% of the difference between T_{Hmax} and T_{H0}; ***12)*** hip torque response time (HRT), calculated as the time interval between tether release and onset of increased hip torque.

2.3 Statistics

I used a two factor repeated-measures ANOVA to determine whether maximum release angles, EMG onset times, and peak magnitudes, onsets, and rates of increase in joint torque associated with BOS size (3 levels) and recovery style (2 levels). I also used Pearson correlations to examine whether maximum release angles associated with EMG onset times, and with peak magnitudes, onsets, and rates of increase in joint torque. All analyses used a significance level of $P \leq 0.05$, and were conducted using the SPSS statistical analysis package (version 11.0, SPSS, Chicago, Illinois).

CHAPTER THREE
RESULTS

3.1 Effect of base of support size and strategy on balance recovery ability

I found that θ_{max} decreased significantly as the BOS size decreased (F=143.2, P<0.001, Table 1, Figure 9A). In the mixed strategy trials, reducing the BOS size from 100% to 75% caused average values of θ_{max} to decline 19%, from 9.8 ± 1.2 deg to 8.0 ± 1.5 deg; (mean difference = 1.8 deg, 95% CI: 1.3 deg to 2.3 deg, t = 7.4, P<0.001). Reducing the BOS size from 75% to 50% caused average values of θ_{max} to decline 30%, from 8.0 ± 1.5 deg to 5.6 ± 1.2 deg (mean difference = 2.4 deg, 95% CI: 1.9 deg to 2.9 deg, t = 9.8, P<0.001). In ankle strategy trials, reducing the BOS size from 100% to 75% caused average values of θ_{max} to decline 16%, from 7.8 ± 0.9 deg to 6.6 ± 1.2 deg (mean difference = 1.2 deg, 95% CI: 0.6 deg to 1.8 deg, t = 4.5, P=0.001), and reducing the BOS size from 75% to 50% caused average values of θ_{max} to decline 33%, from 6.6 ± 1.2 deg to 4.5 ± 1.2 deg (mean difference = 2.2 deg, 95% CI: 1.7 deg to 2.6 deg, t = 10.2, P<0.001).

The mean value of θ_{max} was significantly larger in mixed strategy trials than in ankle strategy trials (F=54.6, P<0.001, Table 1, Figure 9A). In the 100% BOS condition, θ_{max} averaged 20% smaller in the ankle strategy than in the mixed strategy (mean difference = 1.9 deg, 95% CI: 1.2 deg to 2.6 deg, t = 6.0, P<0.001). In the 75% BOS condition, θ_{max} averaged 17% smaller in the ankle strategy than in the mixed strategy (mean difference = 1.4 deg, 95% CI: 0.9 deg to 1.9 deg, t = 5.1, P<0.001). In 50% BOS condition, θ_{max} averaged was 21% smaller in the ankle strategy than in the mixed strategy (mean difference = 1.2 deg, 95% CI: 0.7 deg to 1.6 deg, t = 5.4, P<0.001).

The effect of BOS size on θ_{max} depended on the type of recovery strategy (F= 3.9, p=0.045 for the interaction term (BOS size * recovery strategy)). This reflected that declines in BOS had a slightly bigger effect on θ_{max} in the ankle strategy than the mixed strategy.

3.2 Effect of base of support size and balance recovery strategy on neuromuscular variables

Reductions in BOS size were accompanied by declines in maximum ankle torque, but had no effect on maximum hip torque. Mean values of maximum ankle torque (T_{Amax}) declined as BOS size decreased (F=111, P<0.001, Table 1, Figure 9B, Figure 10). There was no difference in T_{Amax} between ankle strategy and mixed strategy (F=0.2, P=0.9, Table 1, Figure 9B, Figure 10). In the mixed strategy, T_{Amax} declined 22% (from 109.6 ± 23.2 Nm to 85.9 ± 20.5 Nm) when the BOS sized decreased from 100% to 75%, and 28% (from 85.9 ± 20.5 Nm to 61.8 ± 21.0 Nm) when the BOS size decreased from 75% to 50%. In the ankle strategy, T_{Amax} declined 24% (from 112.8 ± 22.0 Nm to 85.5 ± 20.6 Nm) when the BOS sized decreased from 100% to 75%, and 32% (from 85.5 ± 20.6 Nm to 58.3 ± 15.9 Nm) when the BOS size decreased from 75% to 50%. The effect of BOS size on T_{Amax} was unaffected by recovery strategy (F= 0.02, P = 0.9 for the interaction term (BOS size * recovery strategy)). The baseline magnitude of ankle torque (T_{A0}) declined as BOS size decreased on average by 27% from 100% BOS to 75% BOS trials and 38% from 75% BOS to 50% BOS trials (F=25.7, P<0.001), and T_{A0} was larger in mixed strategy than ankle strategy trials by 12% on average (F=10.8, P=0.005).

I also found that the rate of ankle torque generation (I_A) declined as BOS size decreased (F = 16.7, P<0.001, Table 1) and was smaller in the mixed strategy trials than in the ankle strategy trials (F=31.9, P<0.001). The rate of decline in ankle torque (D_A) tended to decrease as BOS size decreased (Table 1) but this trend failed to reach statistical significance (F=3.2, P=0.056), and no difference in D_A was observed between ankle strategy and mixed strategy trials (F=2.0, P=0.2).

Mean values of maximum hip torque (T_{Hmax}) did not change with changes in BOS size (F=2.9, P=0.096, Table 1, Figure 9C, Figure 10). T_{Hmax} was significantly larger in mixed strategy than ankle strategy trials (F=64.4, P<0.001, Figure 9C, Figure 10). Baseline magnitudes of hip torque (T_{H0}) increased as BOS size decreased (F=59.6, P<0.001). The average increase in T_{H0} was 28% from 100% BOS to 75% BOS trials and 1% from 75% BOS to 50% BOS trials. T_{H0} was significantly smaller in ankle strategy than mixed strategy trials (F=18.2, P=0.001). The average decline was 8%. The rate of hip torque generation (I_H) decreased as BOS size decreased in the mixed strategy (F=7.3, P=0.03). The average decrease was 16.1% from 100% BOS to 75% BOS trials and 3% from 75% BOS to 50% BOS trials. The rate of decline in hip torque (D_H) did not associate with BOS size (F=1.4, P=0.3).

The ankle response time (ART) was unaffected by BOS size (F=1.2, P=0.3, Table 1) and balance recovery strategy (F=1.2, P=0.3, Table 1). Similarly, mean values of gastrocnemius premotor time (PMT) were unaffected by BOS size (F=2.1, P=0.2) and balance recovery strategy (F=1.2, P=0.3). Gastrocnemius electromechanical delay (EMD) was also unaffected by BOS size (F = 0.3, P = 0.8) and balance recovery strategy (F=0.02, P=0.9). Hip response time (HRT) was similarly unaffected by BOS size (F=0.1, P=0.9, Table 1) but was significantly shorter in mixed strategy than in ankle strategy trials by 20% on average (F=5.1, P=0.04).

3.3 Kinematic and kinetic differences between ankle and mixed strategies.

Maximum ankle torque and maximum ankle angles (F=771, P=0.4) were not significantly different between ankle strategy and mixed strategy (Figure 11). However, double peaks in ankle torque (approximately 500 ms apart) were commonly observed in mixed strategy trials, but not ankle strategy trials (Figure 11). Participants employed larger hip torque and hip flexions in mixed strategy than ankle strategy trials (Figure 11).

Some degree of hip flexion tended to exist in ankle strategy trials, followed by heel rise (as the subject went up on tiptoes, Figure 12). This caused the ankle to rotate into increased plantarflexion, and the ankle plantarflexor muscles to contract concentrically during balance recovery. The implications of these kinetics are considered further in the Discussion.

The time required to halt forward movement of the COG was shorter in mixed strategy than ankle strategy trials (0.36 ± 0.09 s versus 0.57 ± 0.17 s, F=7.4, P=0.02, Figure 11), but no difference was found across three BOS conditions (F=1.9, P=0.2, Figure 11).

The maximum COG displacement declined with decreases in BOS size (F=273.3, P<0.001), and was significantly larger in mixed strategy trials than in ankle strategy trials (F=31.4, P<0.001). The maximum COP displacement also decreased as BOS size decreased (F=334.2, P<0.001), but was not different in ankle and mixed strategy trials (F=3.7, P=0.08, Figure 13).

3.4 Effect of neuromuscular variables on balance recovery ability

Mixed strategy. In 100% BOS condition in the mixed strategy, I did not find significant correlation between θ_{max} and other variables. However, there were trends towards association between θ_{max} and maximum hip rotation angle (r=0.5, P=0.06), normalized maximum hip torque (r=0.5, P=0.06), and normalized hip torque generation rate (r=0.4, P=0.1). θ_{max} did not correlate with normalized maximum ankle torque (r=0.1, P=0.7), rate of generation of ankle torque (r=0.03, P=0.9), maximum horizontal ground reaction force (r=0.3, P=0.2), ankle torque response time (r=0.1, P=0.7) or hip torque response time (r=-0.2, P=0.4). In the 75% BOS condition in the mixed strategy, θ_{max} correlated with normalized maximum hip torque (r= 0.8 , P=0.04) and rate of ankle torque generation (r=-0.5, P=0.04), but θ_{max} did not correlate with normalized maximum ankle torque (r=0.04, P=0.9), rate of hip torque generation (r=0.06, P=0.8), maximum hip rotation angle (r=0.4, P=0.1), maximum horizontal ground reaction force (r=0.02, P=0.9), hip torque response time (r=-0.4, P=0.2) or ankle torque response time (r=-0.1, P=0.7). In the 50% BOS condition in the mixed strategy, θ_{max} correlated with normalized maximum hip torque (r= 0.7 , P=0.009) and rate of ankle torque generation (r=-0.6, P=0.01), but θ_{max} did not correlate with normalized maximum ankle torque (r=0.07, P=0.8), rate of hip torque generation (r=-0.4, P=0.2), maximum hip rotation angle (r=0.2, P=0.5), maximum horizontal ground reaction force (r=0.2, P=0.5), hip torque response time (r=-0.02, P=0.9) or ankle torque response time (r=-0.3, P=0.3).

Ankle strategy. In the 100% BOS condition in the ankle strategy, I did not find significant correlation between θ_{max} and other variables. θ_{max} did not correlate with normalized maximum ankle torque (r=0.1, P=0.7), rate of ankle torque generation (r=-0.2, P=0.5), maximum horizontal ground reaction force (r=-0.2, P=0.3) or ankle torque response time (r=-0.1, P=0.8). In the 75% BOS condition in the ankle strategy, θ_{max} correlated with maximum horizontal ground reaction force (r=-0.6, P=0.02), but θ_{max} did not correlate with normalized maximum ankle torque (r=0.2, P=0.4), rate of ankle torque generation (r=-0.3, P=0.3), or ankle torque response time (r=-0.2, P=0.4). In the 50% BOS condition in the mixed strategy, I did not find significant correlation between θ_{max} and other variables. θ_{max} did not correlate with normalized maximum ankle torque (r=0.4, P=0.2), rate of ankle torque generation (r=-0.3, P=0.4), maximum horizontal ground reaction force (r=-0.4, P=0.2) or ankle torque response time (r=0.2, P=0.4).

There was no association between foot length (actual or normalized to body height), nor between foot length and normalized maximum ankle torque, in different BOS and strategy conditions. This was likely due to the strong correlation between 1) foot size and body weight (r=0.7, P<0.001), and 2) foot size and body height (r=0.8, P<0.001). This suggested that individuals with larger feet do not necessarily have stronger ankles or better recovery ability.

θ_{max} in the mixed strategy correlated strongly with θ_{max} in the ankle strategy in the 75% BOS condition (r=0.8, P<0.001) and 50% BOS condition (r=0.8, P<0.001), but not in the 100% BOS condition (r=0.3, P=0.3).

CHAPTER FOUR
DISCUSSION

My results indicate that human's ability to recover balance using feet-in-place strategies was limited substantially by BOS size. Regardless of whether my participants used the ankle strategy or the mixed strategy, the maximum release angle (θ_{max}) where they could recover balance declined by approximately 1.14% for every 1% decline in BOS size. These results complement previous studies, which showed that BOS size affects dynamic stability during voluntary (unperturbed) movements [28, 35, 44, 49].

I found that balance recovery ability was affected by BOS size by nearly the same amount in the ankle strategy and the mixed strategy. This reflects that, even in the mixed strategy with hip torque having a primary role in achieving recovery, ankle torque has a major influence on balance recovery ability.

As previously documented, I found that θ_{max} was smaller in the ankle strategy than in the mixed strategy. This complements previous observations of subject tending to use the ankle strategy to recover balance following small perturbations, and the hip strategy following large perturbations [18, 45, 57].

My values of θ_{max} are similar to those reported previously. The mean value of θ_{max} was 7.8 ± 0.9 deg in the 100% BOS condition using the ankle strategy, which was slightly larger than the mean value of 6.9 ± 1.7 deg reported by Robinovitch et al. [55]. The difference may be due to the different angle calculation. In the previous study, the release angle was based on a line connecting the malleolus and acromium markers. In the current study I used a line connecting the ankle and the whole body COG position.

I observed substantial differences in body segment kinematics and kinetics between the ankle strategy and mixed strategy trials. In the ankle strategy, the COG is decelerated solely by the development of increased ankle plantarflexor torque after release. In the more complex mixed strategy, an early increase in hip flexor and ankle plantarflexor torque initiated backward rotation of the leg (plantarflexion about the ankle). In some trials, I observed double peaks in the ankle torque trace. The net effect of this coordinated sequence of hip and ankle torques was to move the COG in the direction opposite to the fall, at a rate that was considerably quicker than observed for the ankle strategy.

Corresponding to this faster deceleration of forward movement, I found that horizontal ground reaction forces (F_x) were larger in the mixed strategy than the ankle strategy, as observed previously by Maki et al. [39] and Horak et al. [18].

Regardless of whether participants used the ankle strategy or mixed strategy, θ_{max} decreased more when BOS declined from 75% to 50% than from 100% to 75%. This was possibly due to the marked reduction in rate of ankle torque generation (I_A) in the 50% BOS condition, when compared to the 75% and 100% BOS condition. These trends suggest that severe reductions in BOS size can impair rate of torque generation as well as peak torque.

I found that θ_{max} was influenced by both peak attainable torque and rate of torque generation. In the ankle strategy, θ_{max} associated with both maximum ankle torque and ankle torque generation rate. However, in the mixed strategy θ_{max} associated with maximum ankle torque and maximum hip torque, but not with ankle torque generation rate or hip torque generation rate. This suggests that torque magnitudes dominate over rate of torque generation in determining performance in the mixed strategy.

I found a nearly linear relationship between maximum ankle torque and BOS size, which makes sense from a biomechanics perspective. In particular, as indicated in the equation 1.2, ankle torque is linearly proportional to COP distance, which is in turn limited by BOS size. Therefore, it is not surprising that a decline of 25% in BOS size would cause a similar size reduction in peak ankle torque.

I also found that the rate of ankle torque generation declined as BOS size decreased. This could be due to a change in postural set (reflecting decreased reliance on ankle plantarflexion muscles in the reduced BOS condition), or a reduction in sensory input from cutaneous receptors on the soles of the feet. These results complement observations by Do et al. (1991), who showed that standing on a reduced support surface led to a striking decrease in the amplitude of the soleus EMG burst following tether release [13], and by Thoumie et al. (1996), who found significant alterations in leg muscle activity after balance perturbation in participants with foot anesthesia [62].

The ankle torque response times observed in this study (which averaged 91±16 ms in the ankle strategy) were similar to the latencies reported previously (99 ± 13 ms) for a similar tether release paradigm [55], and a forward sway on a moving platform study (88 ± 9 ms) [18].

I expected that muscle stiffness (force-length) characteristics could be one source of increased force in the ankle plantarflexor muscles following release. However, the kinematics suggest that this was not a major contributor to muscle force generation. Even in the ankle strategy trials, the ankle joint rotated into increased plantarflexion as opposed to dorsiflexion following tether release (this coincided with backward rotation of the shank, and raising of the heels off the ground). This caused the increase in plantarflexor torque to occur while the plantarflexor muscles were contracting concentrically (see composite traces shown in Figure 11, and the individual trial data shown in Figure 12), as opposed to stretching eccentrically as has been described traditionally for the ankle strategy [17]. This indicates that plantarflexor muscle stiffness did not contribute to the increase in ankle plantarflexor torque following release.

I also expected that the monosynaptic stretch reflex would contribute to increased activation and force generation in the plantarflexor muscles. However, instead of stretching, the plantarflexor muscles shortened after release and there was reciprocal inhibition of activity in the stretching tibialis anterior (dorsiflexor) muscles. This indicates a centrally coordinated response to produce increased plantarflexor torque, the timing of which is consistent with a vestibular or startle reflex response [3].

I conducted experiments with subjects barefoot, in order to eliminate the effect of footwear on balance recovery. Previous studies have shown that shoe tread design, heel height and heel width affect postural stability [30], therefore further studies are required to assess whether these parameters affect performance in my experiments.

Several limitations exist in this study. First, I released participants from a stationary inclined position. In real life, loss of balance episodes also occurs during walking, bending or reaching. While I see little reason for believing my results would not transfer to such situations, further experiments are required to verify this. Second, the magnitude and direction of the perturbation was predictable to the participants in my trials, although I tried to increase the unexpectedness of the release by randomizing the time delay between the ready cue and the release.

Third, although I eliminated obvious hip flexion with the ankle strategy, I found even in ankle strategy trials that there was a small amount of hip flexion, which during the period of ankle torque generation was similar for mixed strategy and ankle strategy trials (in ankle strategy trials, peak hip flexion averaged 9.3 deg, and ranged from 5.9 to 13.0 deg).

Fourth, I manipulated the maximum attainable ankle torque through changes in BOS size, but I did not assess the effect of hip strength on balance recovery ability. More studies are needed to determine the contribution of hip strength to balance recovery in the ankle and mixed strategy.

Fifth, I focused on feet-in-place balance recovery responses, which represent one of several strategies for preventing falls [40, 55]. The most common causes of falls in elderly are trips and slips [2] and stepping is a common technique for balance recovery following a slip or trip and is often invoked before feet-in-place recovery limits are reached [39]. However, there is no evidence that performance on balance recovery tasks that focus on stepping, as opposed to feet in place responses, are more predictive of fall risk. Therefore, experts tend to agree that studies are required on both feet-in-place and stepping responses.

Sixth, I focused on the maximum release angle (θ_{max}) as an indication of the combined influence on balance recovery ability of variables related to strength and reaction time. However, the recovery angle in itself might influence torque generation, due to its effect on the ankle angle, which in turn influences the moment arm and force generating capacity (due to tension-length properties) of the plantarflexor muscles. While this is clearly a limitation of my experimental protocol, I don't regard it as a critical limitation, since the ankle remained near a neutral mid-range position in all trials, and the differences in θ_{max} between the three BOS conditions averaged only 1.7 deg for the ankle strategy, and 2.1 deg for the mixed strategy. The correspondingly small differences between conditions in muscle length should have a minimal effect (much less than BOS size changes) on peak ankle torque.

To my knowledge, my study is the first to illustrate the effect of BOS size on balance recovery ability. I found that BOS size associated with both peak ankle torque and rate of increase in ankle torque, and that regardless of whether the ankle or mixed strategy was used, recovery ability declined by 1.14% for every 1% reduction in BOS size. An important clinical application from this study is the development of a previously-unavailable technique to determine how an individual's ability to recover balance is affected by BOS size. These results have applicability for the design of improved footwear, ladders, and strength training programs to enhance postural stability.

CHAPTER FIVE
CONCLUSIONS

5.1 Overview

Falls are a major cause of injury, especially among the elderly [7, 12, 25]. Clinical studies have shown lower extremity muscle strength is one of the strongest predictors of fall risk in the elderly [54, 81]. This dissertation has established a new methodology for examining how the strength and speed of ankle torque generation affects ability to recover balance. Major findings of my study include the following:

1) BOS size strongly affected ability to recover balance, regardless of whether participants were instructed to use the ankle strategy or the mixed (ankle/hip) strategy. The maximum release angle declined by approximately 1.14% for every 1% decline in BOS size.

2) At a given BOS size, the maximum release angle was larger with the mixed strategy than with the ankle strategy, and this appears due to faster dynamics that more quickly halted downward movement of the COG.

5.2 Future study

Since falls and their related injuries are an increasingly serious health problem in the elderly population, future research work should focus on age-related changes in balance and falls. Given the complex and multifactorial nature of falls in the elderly, future biomechanical research is needed to examine mechanisms that account for age-related declines in balance and understand the cause and prevention of falls and fall-related injuries. Future research also needs to fill the gap between laboratory and clinical research, so that knowledge gained in the laboratory has a positive impact on the health of elderly individuals.

Exercise is an intervention for preventing falls, and the benefits of exercise extend far beyond fall prevention to overall musculoskeletal, cardiovascular, and mental health. Designing an appropriate exercise program to fill the gap between experimental study and clinical application is the long-term goal of this research. It would be beneficial for future studies to address whether a multi-factorial exercise program can enhance speed-of-response and strength variables, and improve ability to maintain and recover balance. There is still substantial uncertainty surrounding whether reaction time (a speed-of-response variable) is amenable to change through an exercise program, so the results of this type of study would be highly valuable.

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FIGURES

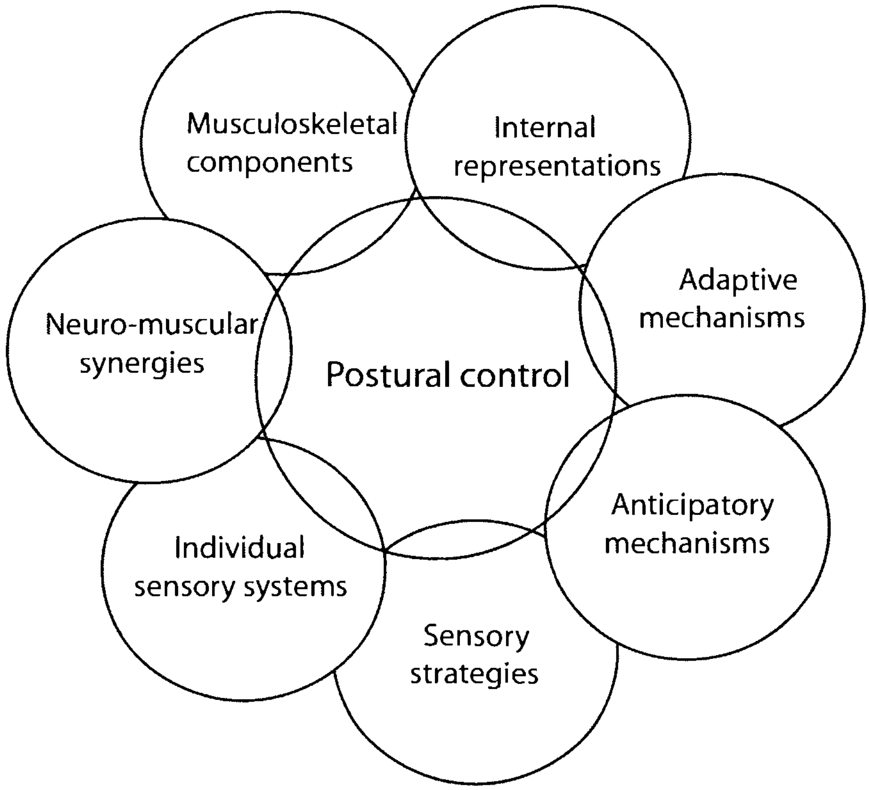


Figure 1 Conceptual model showing systems contributing to postural control (adapted from Shumway-Cook, 2000).

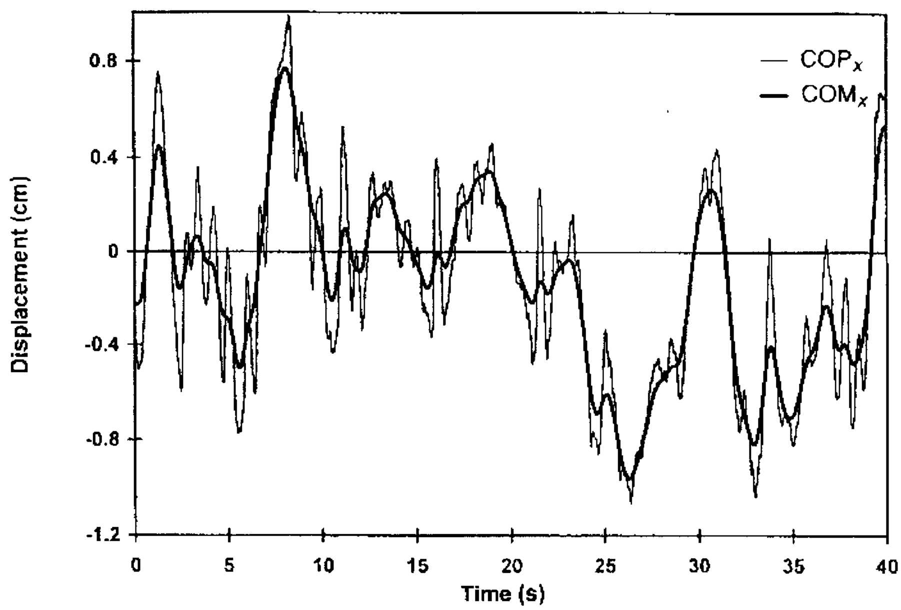
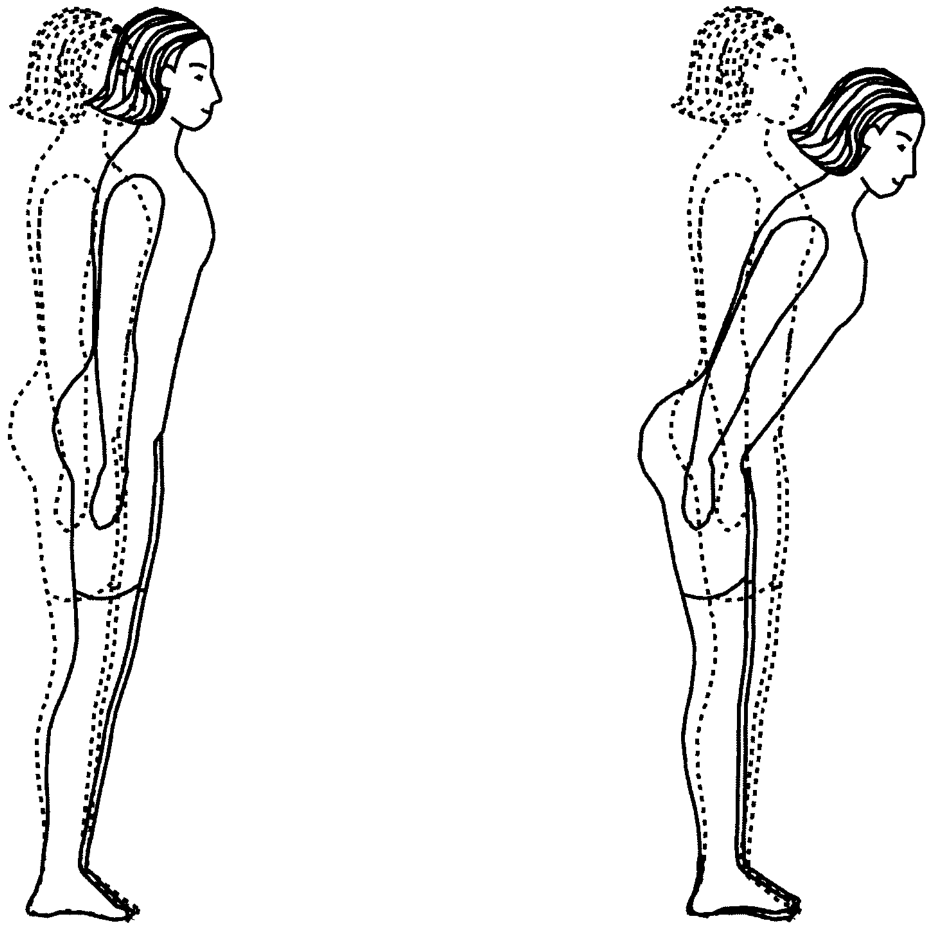


Figure 2 Typical record of the total body centre-of-gravity (referred to here as centre-of-mass, COM) and centre-of-pressure (COP) in the anterior-posterior (x) direction during quiet standing.
The COP magnitude exceeds that of the COM and reversals of direction of the COM are accompanied overshoots of COP. COP is continuously moving anteriorly and posteriorly with respect to the COG with a higher frequency and amplitude (adapted from Winter DA, 1990).



A B

Figure 3 Ankle strategy (A) and hip strategy (B). The ankle strategy restores equilibrium by moving the body as a single-link inverted pendulum about the ankle joints. The hip strategy, on the other hand, restores equilibrium by moving the body as a double-link inverted pendulum about the hip and ankle joints. Restoring torques are applied primarily at the hip.

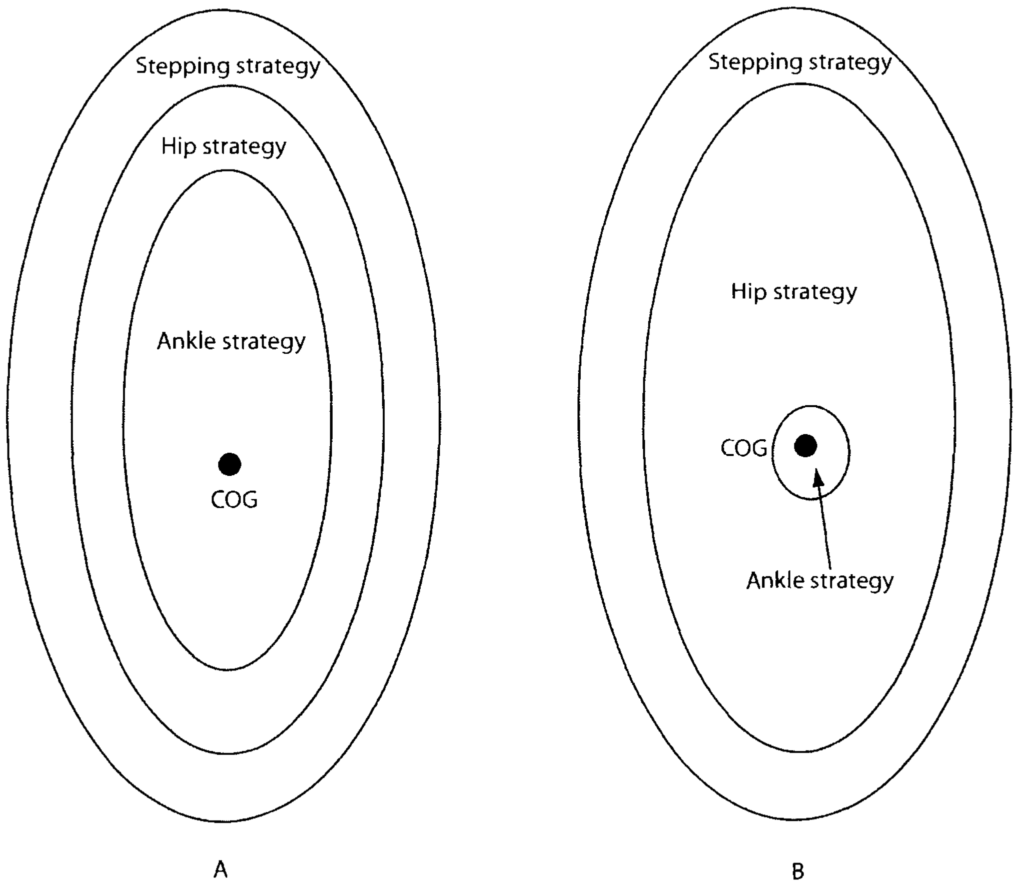


Figure 4 Performance boundaries for motor strategies used to control movements of the whole body centre-of-gravity (COG) in space, while standing on a flat surface (A) versus a narrow beam (B). (adapted from Horak FB, 1992).

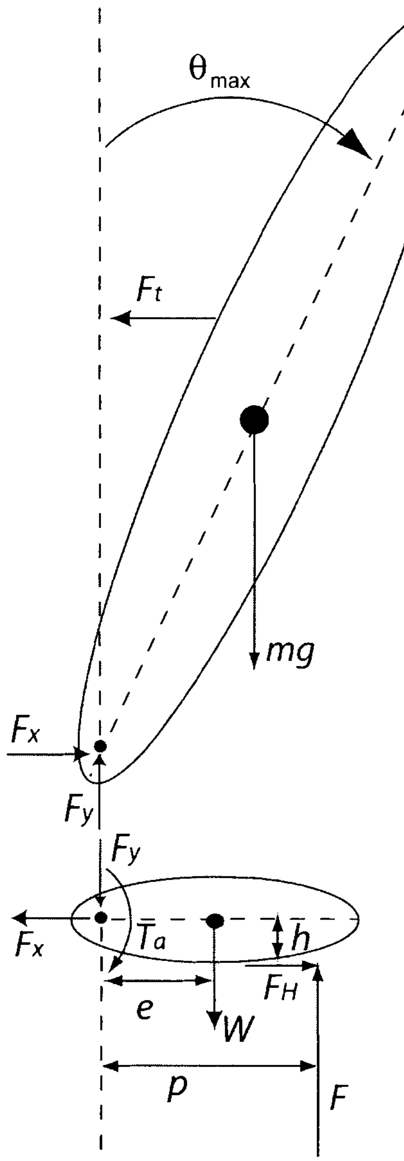


Figure 5 Inverted pendulum model. Body is represented by a one link inverted pendulum of mass mg. In the free body diagram of foot, four forces act on the segment: F (vertical ground reaction force), F_H (horizontal ground reaction force), W (foot weight), F_y and F_X (joint reaction forces acting on the ankle). T_a is the net ankle torque. p (COP), e and h are the moment arms from ankle joint to the vertical ground reaction force, foot COG and horizontal ground reaction force, respectively. In the free body diagram of all body segments above the ankle joint, F_t is the tether force holding participants before release, F_y and F_X are the reaction forces acting on the ankle joint.

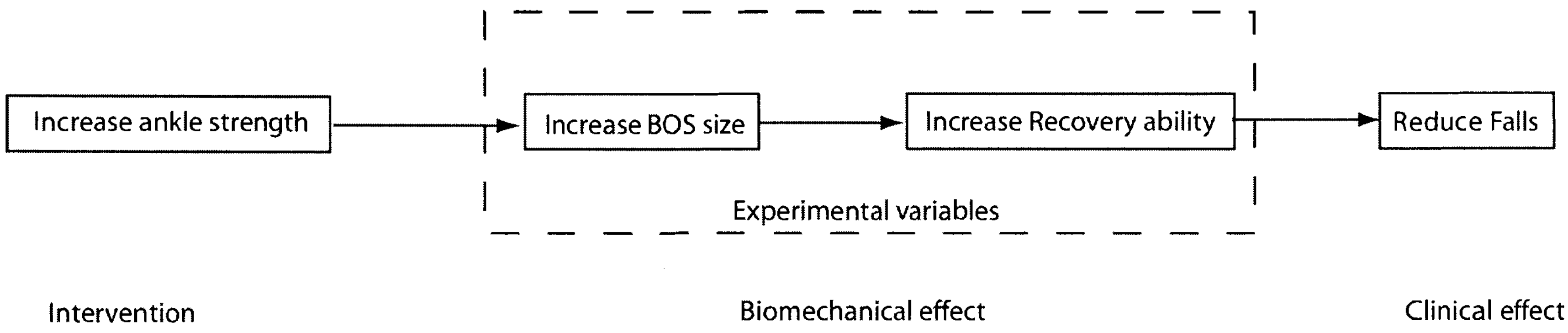


Figure 6 A conceptual model shows how ankle strength could affect balance recovery ability and risk for falls. Ankle strength will increase BOS size, and increase in the ability to recover balance. This should result in reduced risks for falls.

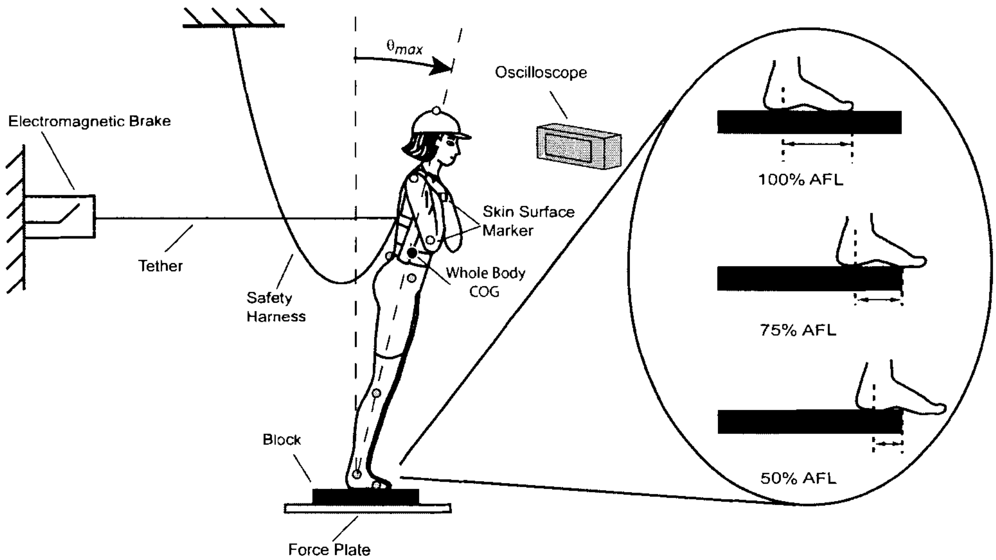


Figure 7 BOS Experiment. Participants were held by a tether in an initially inclined position. Their maximum release angle (θ_{max}) defined as the maximum initial lean angle where they could recover balance after the tether was suddenly released. I conducted experiments for three BOS sizes (100%, 75% and 50% of anterior foot length, and for balance recovery with the ankle strategy and mixed (ankle/hip) strategy.

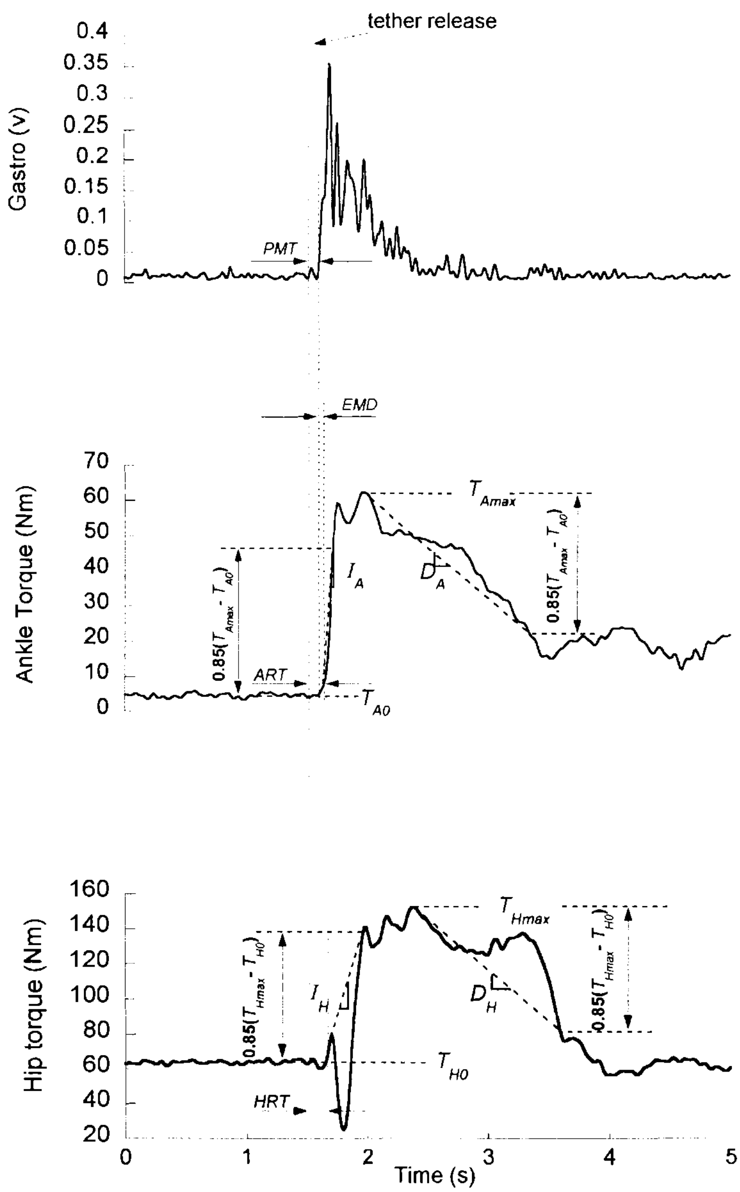


Figure 8 Ankle torque and hip torque characteristics during balance recovery. Tether release was detected by a sharp decline in tether force (shown by the first vertical dashed line). Before tether release, participants adjusted their ankle torque to match the average value during quiet standing (T_{A0}). Following release, there was an ankle torque response time (ART) before the onset of ankle torque generation. Ankle torque was generated at a rate I_A and reached a peak, T_{Amax}, before declining at a rate D_A. ART is composed a gastrocnemius premotor time (PMT) and an electromechanical delay (EMD). The second dashed line indicates the onset of gastrocnemius activity and the third dashed line indicates the onset of ankle torque generation. The second dashed line in the hip torque profile indicates the onset of hip torque generation. All variables in the hip torque profile used the same calculations as those described above for ankle torque. Angles are positive in extension (or plantarflexion at the ankle). Joint torques are positive if they are extensor (or plantarflexor at the ankle).

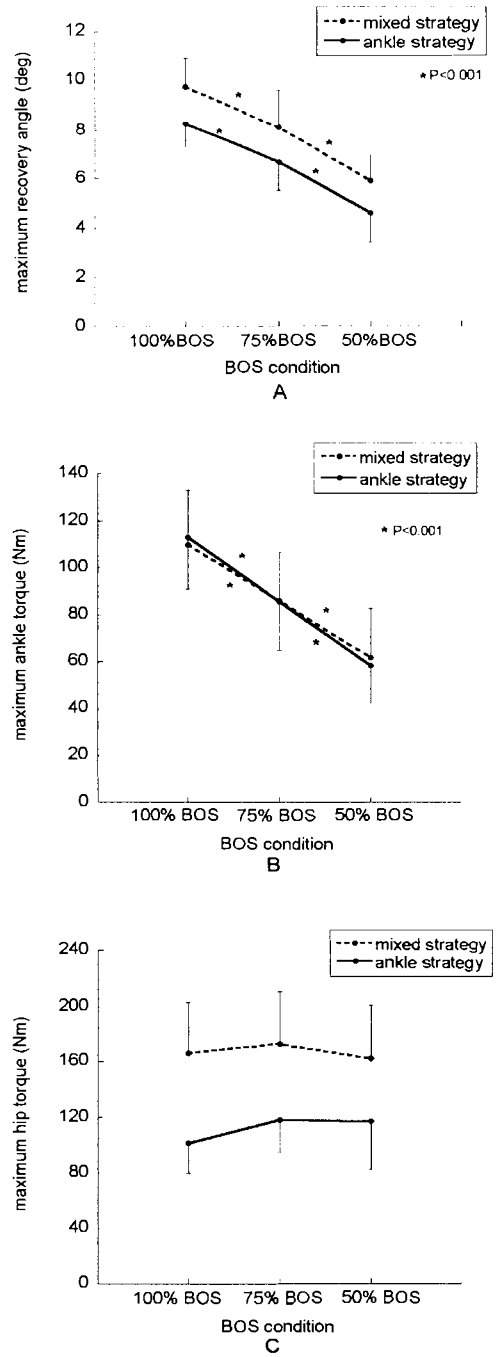


Figure 9 (**A**) Maximum release angle (θ_{max}) decreased as BOS declined (P<0.001) and was larger in the mixed strategy than ankle strategy trials (P<0.001). This suggests that ability to recover balance was influenced by BOS size and balance recovery strategy. Maximum ankle torque (**B**) and maximum hip torque (**C**) in three BOS conditions (100%, 75% and 50%) for ankle strategy and mixed strategy. As BOS sized was reduced, I observed a decrease in maximum ankle torque (P<0.001) in both the ankle strategy and mixed strategy, but no change in hip torque (P=0.096, n=15, error bars show ± one standard deviation).

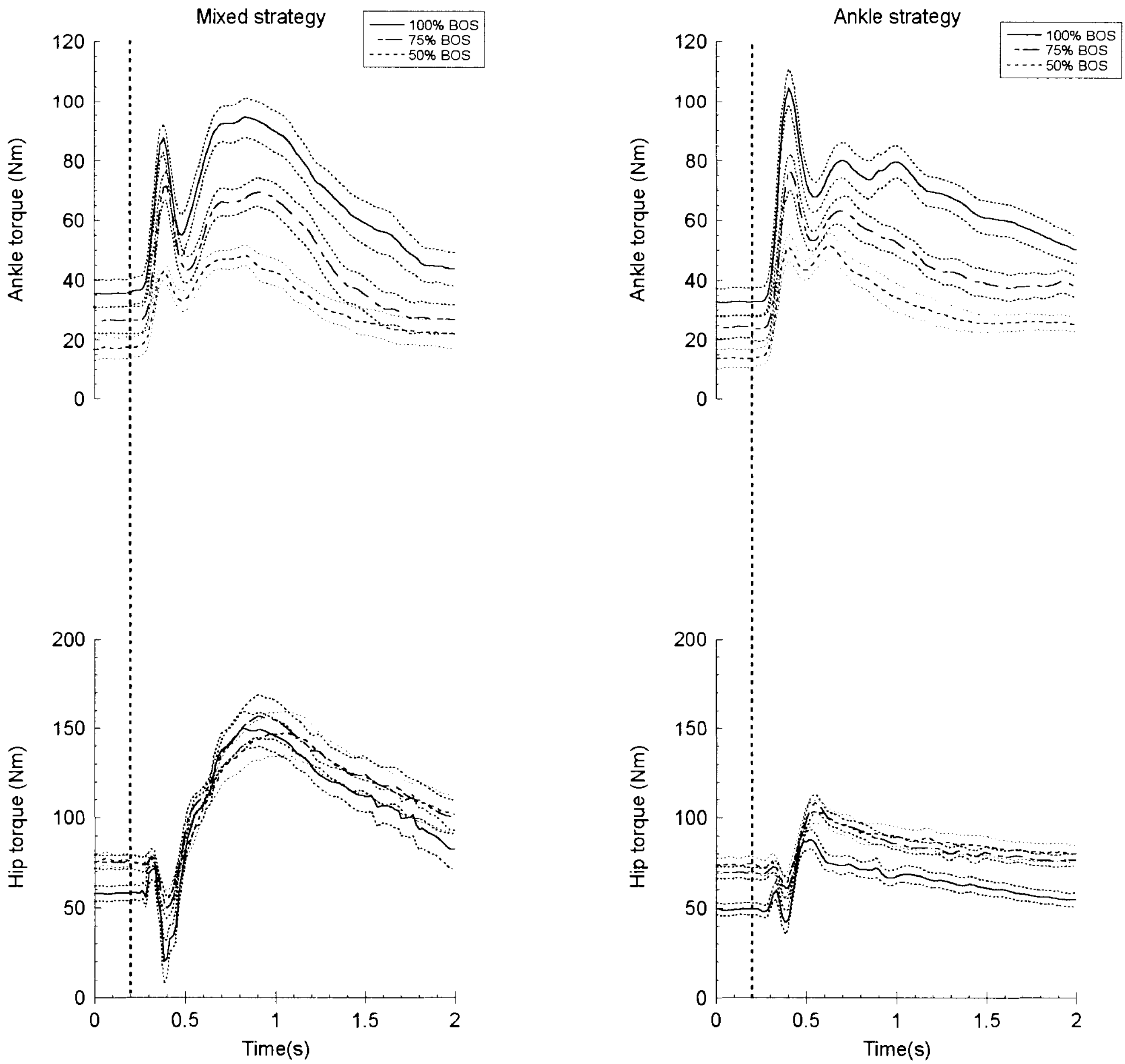


Figure 10 Effect of BOS size (100%, 75% and 50%) on ankle torque and hip torque development. In both ankle and mixed strategies, maximum ankle torque decreased as BOS size decreased. Maximum hip torque was similar regardless of BOS size. (Composite traces show mean ankle and hip torques in the ankle strategy and mixed strategy for each BOS conditions). Dotted lines show ± standard errors, the vertical dashed lines indicates tether release. Angles are positive in extension (or plantarflexion at the ankle). Joint torques are positive if they are extensor (or plantarflexor at the ankle).

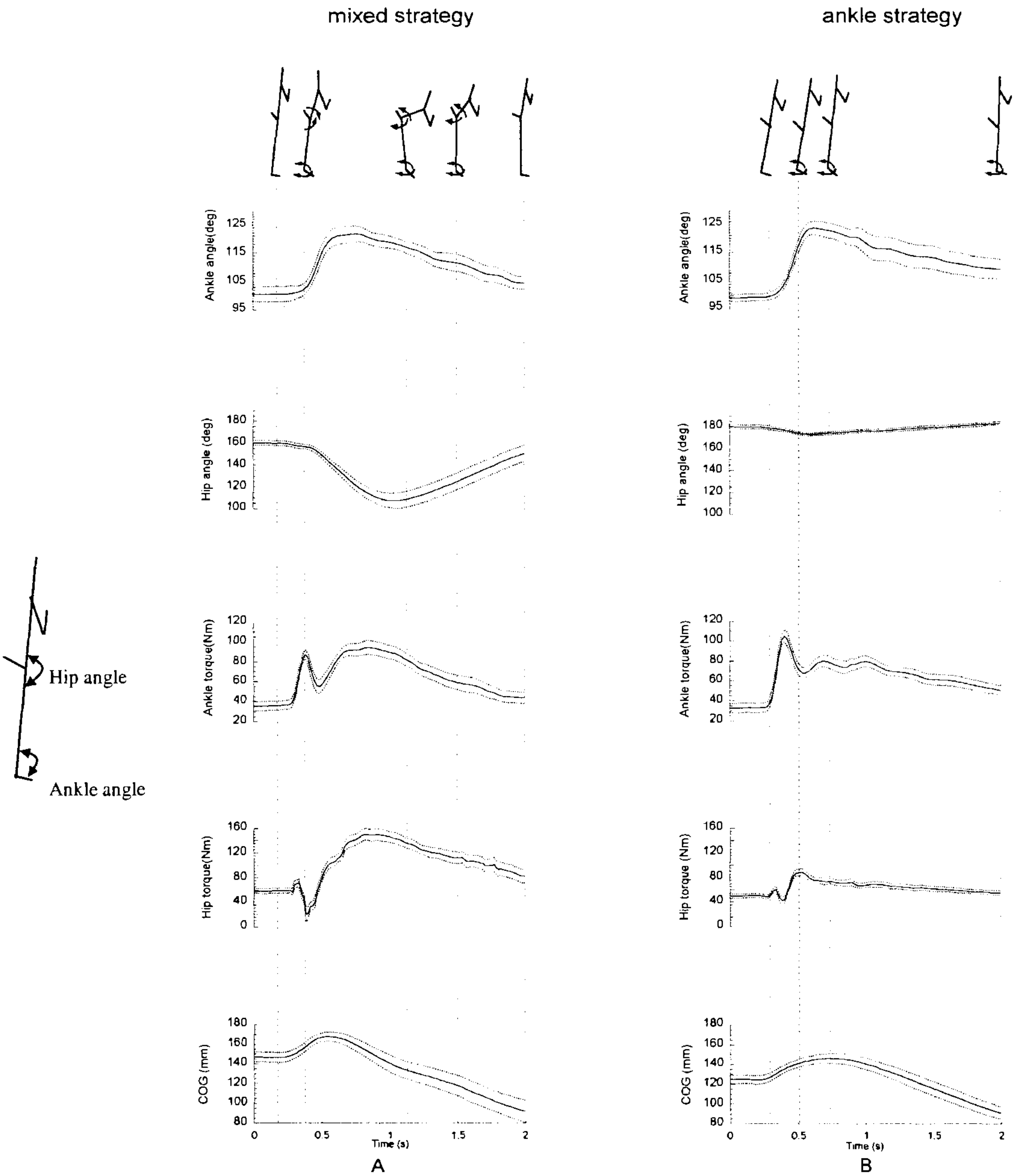


Figure 11 Composite traces (n=15) showing average variations in kinematic and kinetic parameters for the mixed strategy (A) and the ankle strategy (B), in the 100% BOS condition. The top stick figures illustrate typical body positions during balance recovery. After tether release (indicated by the first vertical dashed line), ankle torque began to increase and halt downward rotation of the body, which allowed for return to upright posture. The hip flexor torque (hip torque valley in A) activated almost at the same time as ankle torque to promote rotation of the legs. The coordination of hip and ankle torques contributes to move the COG in the direction opposite to the fall. (The dotted lines in each figure show ± one standard error). Angles are positive in extension (or plantarflexion at the ankle). Joint torques are positive if they are extensor (or plantarflexor at the ankle).

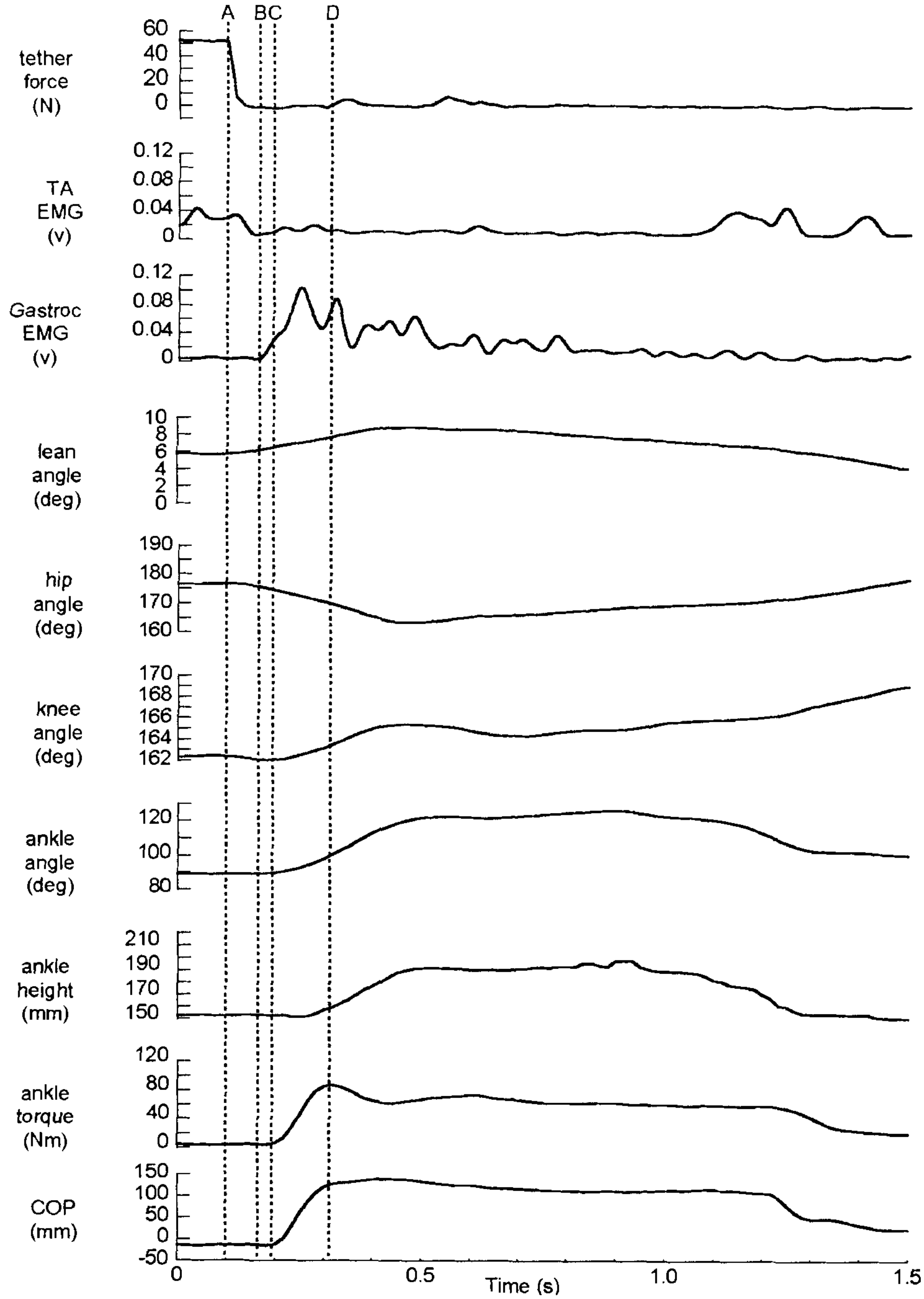


Figure 12 A typical trace showing variations in EMG, kinematic and kinetic parameters for the ankle strategy in 100% BOS condition. A = time to tether release, B = time of increase in gastrocnemius EMG, C = time of increase in ankle torque, D = time of peak torque. Joint angles are positive in extension (or plantarflexion at the ankle). Angles are positive in extension (or plantarflexion at the ankle). Joint torques are positive if they are extensor (or plantarflexor at the ankle).

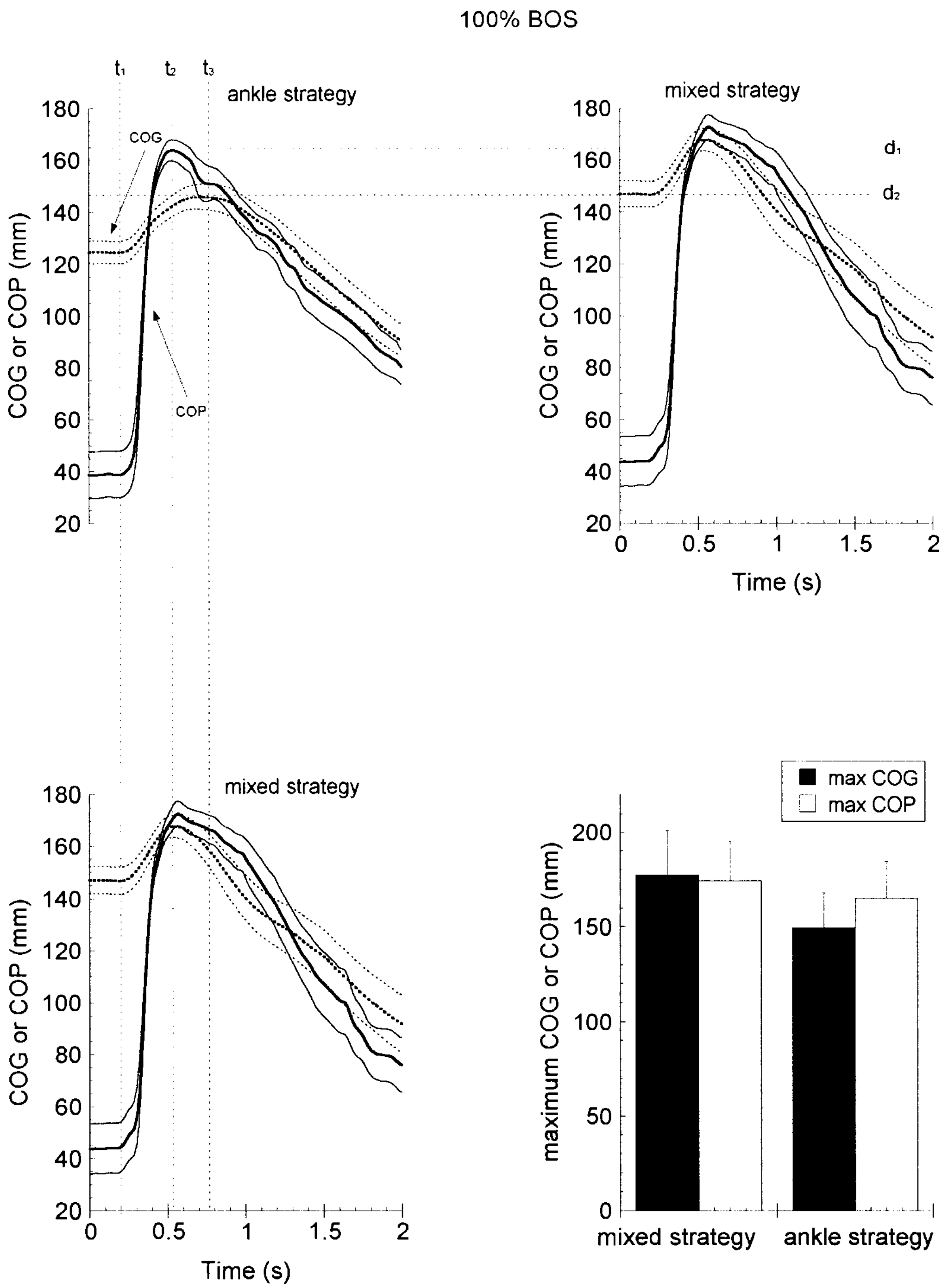


Figure 13 Composite traces (n=15) of COG and COP displacements in the 100% BOS condition. The t_1 symbol indicates the instant of tether release, t_2 indicates the time of maximum COP displacement and t_3 indicates the maximum displacement. d_1 indicates the maximum COP displacement and d_2 indicates the maximum COG displacement (dotted lines and the thinner solid lines in each figure show one standard error). The bar graph shows the maximum COG and COP displacement in each strategy (error bar shows ± one standard deviation).

TABLES

Table 1. Summary Data Results - Mean (± S.D.) parameters values (n=15)

	Ankle strategy			Mixed strategy		
Parameter	100%BOS	75%BOS	50%BOS	100%BOS	75%BOS	50%BOS
θ_{max} (deg)	7.8 ± 0.9	6.6 ± 1.2	4.5 ± 1.2	9.8 ± 1.2	8.0 ± 1.5	5.6 ± 1.2
T_{A0} (Nm)	32.1 ± 18.4	23.3 ± 15.2	13.5 ± 13.9	35.4 ± 18.1	26.2 ± 15.8	17.1 ± 13.6
T_{Amax} (Nm)	112.8 ± 22.0	85.5 ± 20.6	58.3 ± 15.9	109.6 ± 23.2	85.9 ± 20.5	61.8 ± 21.0
I_A (Nm/s)	692.9 ± 296.1	585.2 ± 276.6	358.2 ± 144.8	411.7 ± 258.8	460.3 ± 232.1	252.2 ± 177.7
D_A (Nm/s)	112.3 ± 129.5	73.5 ± 63.7	66.4 ± 38.8	182.6 ± 237.2	83.3 ± 44.6	77.9 ± 79.5
ART (s)	80.9 ± 18.6	87.1 ± 18.1	91.6 ± 16.1	80.3 ± 23.9	75.7 ± 24.2	88.1 ± 18.2
PMT (s)	65.5 ± 23.6	68.6 ± 27.0	79.2 ± 21.5	66.0 ± 23.6	66.8 ± 27.5	68.6 ± 24.1
EMD (s)	10.1 ± 28.7	18.5 ± 25.4	12.4 ± 27.6	14.3 ± 25.5	9.0 ± 30.8	19.6 ± 26.6
T_{H0} (Nm)	48.9 ± 13.1	70.9 ± 13.1	72.2 ± 16.1	57.2 ± 15.7	75.7 ± 13.5	76.5 ± 12.7
T_{Hmax} (Nm)	101.6 ± 21.8	118.5 ± 23.3	117.2 ± 34.6	165.9 ± 37.5	172.7 ± 38.3	162.2 ± 39.3
I_H (Nm/s)	615.1 ± 406.0	430.1 ± 271.8	336.6 ± 200.7	460.2 ± 254.7	386.0 ± 185.0	374.6 ± 238.6
D_H (Nm/s)	182.4 ± 330.6	102.9 ± 125.6	67.1 ± 65.9	86.1 ± 51.5	92.6 ± 56.4	81.1 ± 59.6
HRT (s)	79.2 ± 40.6	101.0 ± 51.5	83.3 ± 37.1	67.4 ± 28.8	87.5 ± 29.1	81.5 ± 68.7
Max COG (mm)	149.4 ± 18.7	112.2 ± 13.6	71.2 ± 19.8	177.2 ± 23.7	131.0 ± 16.7	87.3 ± 16.0
Max COP (mm)	165.3 ± 18.8	124.8 ± 13.8	88.9 ± 13.4	174.3 ± 20.7	128.8 ± 17.6	86.1 ± 17.2
Max F_x (N)	49.1 ± 17.4	45.8 ± 19.6	38.6 ± 14.8	80.6 ± 27.1	64.2 ± 20.5	52.8 ± 18.0
Max F_z (N)	855.5 ± 138.2	865.8 ± 169.5	785.1 ± 155.2	785.3 ± 126.7	768.9 ± 121.9	731.9 ± 127.6
COG stop time (s)	0.57 ± 0.17	0.55 ± 0.27	0.40 ± 0.13	0.36 ± 0.09	0.40 ± 0.18	0.42 ± 0.18