The effect of age on neuromechanical responses to electrical stimulation of superficial peroneal nerve during walking

by

Ryan Brodie
BSc, University of Victoria, 2001
BSc, University of Victoria, 2011

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Supervisory Committee

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Abstract

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In the healthy young, stimulation of superficial peroneal nerve (SPn) cutaneous afferents at the ankle during walking has been shown to elicit functionally relevant neural and mechanical responses that contribute to obstacle avoidance during swing and have been referred to as stumble corrective responses. However, specific age-related differences in the stumble corrective response induced by electrically evoked cutaneous stimulation have yet to be determined. As a confounding contributor to age related changes in dynamic stability during locomotion, neural and mechanical changes in the stumble corrective response may result in a decreased ability to recover from a destabilizing incident and provide key markers of neuromuscular decline. Therefore the purpose of this study was to compare age-dependent differences in responses to electrically evoked stimulation of the superficial peroneal nerve at the ankle during walking in healthy young and elderly groups. Electromyograms (EMG) of the tibialis anterior (TA), soleus (Sol), medial gastrocnemius (MG), biceps femoris (BF) and vastus lateralis (VL) were recorded along with gait kinematics including joint displacement and angular velocity at the ankle and knee as well as toe clearance relative to the walking surface. Overall, the stumble corrective response was preserved in the elderly as evident by significant responses in kinematics and muscle activity that were similar in sign and phase to those seen in the healthy young. However, the magnitude of the kinematic responses and resulting toe clearance in older adults were significantly smaller than in the young. Further, during the swing phase of unstimulated walking cycles, there were reduced knee flexion, plantarflexion and toe clearance in the elderly with corresponding differences in muscle activity. Therefore, smaller kinematic responses to stimulation, in the elderly, superimposed on a different undisturbed gait profile, resulting in reduced toe clearance, reflects early degradation of the stumble corrective response. This early degradation is likely a prodromal sign of increased fall risk. This supports the potential use of cutaneous reflexes in quantifying degradation of neuromuscular control and its contribution to fall risk.

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Acknowledgements

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Dedication

This work is dedicated to Christina and Miles, without whom this achievement would be neither possible nor meaningful.

Chapter 1: Introduction and Review of Literature

Falls account for 74% of injury-related hospitalizations for older adults (Canadian Institute for Health Information, 2011) and cause more than 90% of all hip fractures in this age group (Zuckerman, 1996). In a prospective study of independent, community-dwelling seniors between 60 and 88 years of age, 52% were found to fall within 1 year, with tripping accounting for the greatest percentage of these falls at 59% (Berg et al., 1997).Within one year of fracture 20% of these seniors die. Given the important social and financial consequence of these events it is paramount that we deepen our understanding of potential contributors to the loss of dynamic stability faced by older individuals.

Age-related changes in human biomechanics and motor control quantifiably alter locomotor activities throughout the lifespan (Winter, 1991). For example, ageing is related to differences in the pattern of muscle activity and joint torque during walking such as increased coactivation at the ankles and increased hamstring activity during stance (Schmitz et al., 2009), as well as increased reliance on hip extensor power relative to plantarflexion in stance (DeVita & Hortobagyi, 2000). Studies of reflexes have proven invaluable for understanding functional responses to sensory input during locomotion (Zehr et al., 1997), as well as to characterize deficits in control (Hundza & Zehr, 2007, Zehr & Loadman 2012; Zehr et al., 2012). For example, in healthy young subjects, it has been observed that reflexes result in coordinated neural and mechanical outcomes that serve important regulatory functions during human locomotion (Zehr et al., 1997). That is, reflexes originating from cutaneous nerves of the foot have been shown to assist with obstacle avoidance during the swing phase of walking while reflexes from muscle afferents in the lower limb have been associated with important stance phase corrections (Zehr et al., 1997; Zehr & Stein, 1999). Specifically the response resulting from electrical stimulation of the superficial peroneal nerve at the ankle has been called a stumble corrective response. Originally identified in animal models (Drew & Rossignol, 1987; Forssberg, 1979), this class of responses has since been observed in healthy adults and involves dorsiflexion of the ankle and flexion of the knee during swing phase assisting the swing limb in avoiding the obstacle (Zehr et al., 1997; Van Wezel et al., 1997).

There is evidence of age-related alterations to locomotor relevant reflexes such as a noticeable decrease in the contribution of muscle afferent reflexes during the stance phase of walking that result in altered mechanical responses to perturbations (Chalmers & Knutzen, 2000). Therefore, the study of reflex responses during locomotor activities in older adults may be useful to establish mechanisms of age related neuromechanical differences and present potential markers for deterioration in functional ability related to fall risk. While there has been some research on muscle afferent reflexes in the older adults, there has been much less on the neuromechanical investigation of cutaneous reflexes (Zehr and Loadman 2012). Additionally, while limb kinematic responses have been examined in the young, a comparison of responses in other functional features such as toe clearance, have not been well quantified.

The aim of this review is to provide context for the study of age-related differences in reflex control during gait. This review will examine the basic neural control of locomotion, as well as the changes in the neuromechanics of gait that occur as a result of age, with a focus on how these changes may affect postural stability during normal walking. Finally, the functional role that afferent feedback plays during walking will be reviewed with a focus on age-related differences in the stumble-corrective response to mechanical as well as electrical perturbation.

Motor Control in Human Locomotion

The control of human locomotion is shared among three primary components including central command from higher centres, spinally generated locomotor command and integration at different levels of sensory feedback with these two other mechanisms (Zehr & Stein, 1999). The behaviour of the motor system in response to afferent feedback can be used to reflect on the current state of the neuromuscular system and on the goals of the locomotor task. To place the role of afferent feedback in context, a description of the spinal mechanisms responsible for locomotor pattern generation will be provided.

Locomotor pattern generation in humans and afferent feedback

Rhythmic, patterned activity in the lower limbs has been shown to persist after spinal cord injury (SCI) in humans (Burke, 1999). This has extended even to producing stepping movements which were analogous to normal walking. These movements have been shown to start independently in those with incomplete SCI. Spinal cord stimulation has been used in cases of complete SCI to elicit alternating, reciprocal patterns of excitation in flexors and extensors in the lower limbs, indicating the presence of a central pattern generator (CPG) (Burke, 1999).

Further evidence of CPG contributions to the neural regulation of locomotion in humans is the alternating patterns of flexion and extension in the lower limb seen in neonates during supported treadmill walking. This indicates that there is a neural mechanism for afferent control of walking in place before corticospinal tract is developed sufficiently to allow for influence of cortical motor command (Yang et al., 1998). This neural control is sensitive to afferent sensory input. Infants are able to match their step rate and stride length to the speed of the treadmill (Yang et al., 1998). Similarly, coordination in these spinal pattern generators, studied using split-belt treadmills, showed both interlimb coordination during walking, as well as limb independence in movement patterns in response to various asymmetrical walking speeds (Yang et al., 2005). Stumble-corrective responses evoked with mechanical perturbations are also seen in infant walking, with similar object avoidance strategies as seen in adults. In this, early swing disturbances are followed by a toe clearance response, with increased knee flexion and, in many cases, plantarflexion. Late swing responses were primarily characterized by increased knee flexion (Lam et al., 2003). Together these provide evidence for central, sub-cortical mechanisms responsible both for influencing gait pattern generation, as well as for integrating afferent input with ongoing locomotor activity.

Electrically evoked reflexes during locomotion

A common method for investigating the organization of neural pathways responsible for the integration of sensory feedback with rhythmic movement is through studying the responses to stimulation of various sensory afferents (Burke, 1999). Beginning with work by Sherrington (1906), exogenous sensory input has been known to result in reliable and patterned reflex outputs. Although invasive techniques may provide higher precision results, these methodologies are not available in human research, and thus indirect methods are required. The modulation of responses by changes to the state of the nervous system has been instrumental in understanding the role of afferent feedback and spinal locomotor pattern generation in producing the characteristically rhythmic, patterned and adaptable output (Burke, 1999).

Two common methodologies used to investigate afferent feedback in human locomotion are the electrically evoked Hoffmann (H-) and cutaneous reflex. The H-reflex is the electrical analogue of the natural, monosynaptic stretch reflex. Electrical stimulation is applied percutaneously to a mixed nerve fiber, resulting in recruitment of both α-motor neurons, as well as group Ia afferent fibers innervating muscle spindles (Capaday & Stein, 1986; Zehr, 2002). While reliable and relatively simple to analyze, a limitation of the H-reflex is that changes in muscle characteristics such as length and background motor activity can alter the reflex response in ways which make it difficult to use during dynamic movements (Zehr, 2002). Cutaneous sensory responses provide another approach to understanding spinal regulation of afferent feedback, and can be applied with a consistent intensity more easily than the H-reflex during natural walking.

Cutaneous reflexes during locomotion

Many early studies using a variety of animal preparations, and in particular the cat, have shown that stimulation of cutaneous afferents can interact with the neural pattern generation, resulting in modifications in kinematics (Drew & Rossignol, 1987; Forssberg, 1979; Prochazka et al., 1978). The study of cutaneous reflexes involves percutaneous stimulation of cutaneous receptive fields on the hands or the feet. Specifically, electrical stimulation of these receptive fields excites Aβ, Aδ and C afferent nerve fibers associated with mechanoreceptors in the skin (specifically Merkel disks, Pacinian and Meissner corpuscles, Ruffini endings and free nerve endings) (Zehr & Stein, 1999). Electrical excitation of these skin receptors mimics the neural reaction to external perturbation that contacts the skin. There are a variety of protocols for evoking and recording cutaneous responses to electrical stimulation during different tasks which are described in detail elsewhere (Duysens et al., 1992; Van Wezel et al., 1997; Zehr et al., 1997). In general, trains of non-noxious electrical stimuli are applied to the skin superficial to a peripheral cutaneous nerve, such as the superficial peroneal or the tibial nerve which innervate receptors on the dorsum and the sole of the foot, respectively. Muscle activity is recorded during walking with stimulation delivered psuedorandomly throughout the gait cycle. That is, with a restriction on temporal proximity, generally limiting to between 1-3 step cycles between stimulation events. Average background (undisturbed) muscle activity is subtracted from activity following stimulation to provide a subtracted trace of the ‘reflex’. This response is then considered in the context of the task and phase of movement (Brooke et al., 1997).

The neural pathways of the cutaneous response are polysynaptic and contain a variable number of interneurons modulating the excitability of the motor neuron pool in the spinal cord (Burke, 1999; Zehr, 2006). The result is that a single volley of cutaneous stimuli can produce responses with a great deal of variability in timing, magnitude and even sign. Responses from cutaneous pathways are shown to vary with task as well as phase of movement (Zehr & Kido, 2001). In locomotion, these modulated responses can result in a change in magnitude and sign representing a reversal from excitation to inhibition at different points in the locomotor cycle (Duysens et al., 1992; Duysens et al., 1990). Responses are seen at consistently repeatable latencies related to the synaptic complexity of the underlying pathway connecting the sensory afferent to the motor neuron. P1 (early) responses, occurring at ~50ms in humans are generally small in magnitude and produced with less consistency compared to later responses. P2 or middle latency responses are seen with peaks at ~80ms with duration of ~30ms. These responses are generally greater in magnitude and more consistently repeatable (Baken et al., 2005). As the cutaneous reflex can consist of many, potentially alternating periods of facilitation and inhibition, a further measure of ‘net' or averaged reflex effect can be generated. The average cumulative reflex EMG ($ACRE_{125}$) is defined as the average of subtracted motor output for the 125ms post stimulus. The importance of this measure is in determining how modulated muscle responses may correspond with kinematic outcomes during movement (Zehr & Chua, 2000).

These electrophysiological responses have been further investigated to determine if the modulation patterns they exhibit bear functional relevance in the context or phase of the locomotor task.

Functional relevance of electrically evoked, cutaneous reflexes

The investigation of cutaneous reflex responses has been accompanied by a study of their relationship with movement patterns during walking (Zehr & Haridas, 2003; Zehr et al., 1997), and arm cycling (Haridas & Zehr, 2003; Zehr & Chua, 2000). This involves comparing joint and segment kinematics during movement with the related motor responses as represented by the $ACRE_{125}$ response. To be considered a functionally relevant, the response must bear contextual relevance to the phase and task. What has been shown in these investigations is that phase-modulation of reflexes is functionally appropriate to the goal of the task as well as requirements of the body during each specific phase of movement (Duysens et al., 2004; Zehr & Stein, 1999). Further, in order for responses evoked with electrical stimulation to maintain ecological validity, they should bear resemblance to the responses that are seen when cutaneous receptors are mechanically excited.

Investigation has shown electrically evoked cutaneous responses to be similar in character to those evoked with mechanical perturbation. Electrically evoked responses in cats are qualitatively similar, though reduced in magnitude, to mechanical stimulation (Drew & Rossignol, 1987). This same relationship has also been observed in humans (Zehr & Stein, 1999). The function of these responses in mammals was originally investigated in cats with varying degrees of intact neural control. Electrical stimulation of the hind limb during swing resulted in a patterned response mimicking an object-avoidance and stumble-correction reaction

(Forssberg, 1979; Prochazka et al., 1978). These responses are characterized by passive planterflexion (i.e. inhibition of dorsiflexion), combined with hip and knee flexor excitation. In these studies, cutaneous afferent were shown to provide feedback vital to maintenance of dynamic stability. Of particular importance to this function are responses seen when perturbations fall during phase transitions (i.e. swing-to-stance and stance-to-swing) (Van Wezel et al., 1997; Zehr et al., 1997).

Stumble-correction in response to tripping during locomotion

The motor responses evoked by mechanical perturbations of various types during walking have been used to study the differential reactions between young and old and may provide some insight into the underlying neural differences in these two groups. First is provided a discussion of general motor and mechanical responses to tripping. Following this is a review of responses seen in younger and older adults.

General responses and recovery strategies

Unexpected perturbations to the swing foot during walking present a multitude of stimuli to integrate with efferent motor command in order to respond and maintain balance. As discussed, responses to cutaneous electrical stimulation are similar in pattern to responses to a brush stroke to the same stimulated area (Drew & Rossignol, 1987; Zehr & Stein, 1999). Responses to more realistic mechanical obstruction or perturbation, although similar in character, are greater both in amplitude and complexity due possibly to an increased intensity of stimulus and number of sensory modalities excited by a mechanical intervention (Drew & Rossignol, 1987).

Responses to mechanical perturbation to the swing limb during walking have been broadly classified into two behavioural categories. These are the responses during early swing, generally involving an elevation of the swing limb, and those during late swing, generally involving a lowering and preparation for a subsequent step (Eng et al., 1994; Schillings et al., 2000). Strategy selection is a behavioural outcome that is relatively reliable with responses in the first 25% of swing showing exclusive use of the elevation strategy, those in 55-75% of swing resulting in a lowering strategy, and mid swing responses resulting in a mixture of these two outcomes in neurologically intact young adults (Schillings et al., 2000).

Gross responses characterizing the elevation strategies were described in a study of mechanical tripping in treadmill walking and shown to involve patterned, sequenced activation at the hip, knee and ankle (Eng et al., 1994). 'Early' responses were those in 60-80ms, and 'late' responses in the 110-130ms range. Muscles shown to be active in the early latency include gluteus maximus, medial gastrocnemeus, plantaris, and biceps femoris in the stance phase, acting to stabilize the stance limb, pelvis and trunk. As well, during swing-phase biceps femoris and tibialis anterior showed increased early latency activity, with the rectus femoris showing activity in the late latency period. These muscles work in a coordinated fashion to successively advance the swing limb over the obstacle. Mechanically, this results in a longer stride with increased flexion at the hip, sustained and increased flexion and subsequent rapid extension of the knee as well as greater and sustained dorsiflexion in the swing limb (Eng et al., 1994; Schillings et al., 2000).

Motor responses in the lowering strategy were found to be primarily inhibitory, with hip and knee extensors as well as dorsiflexors showing early latency inhibition. Combined with facilitation responses shown in the hamstrings during swing, the lowering strategy results in characteristically different mechanics from elevation. The kinematic alterations resulting from these motor responses is a slightly temporally longer step cycle in reaction to the purterbation with inhibition of knee extension and dorsiflexion at the ankle. The net result is a shorter step length as the foot is planted as quickly as possible in response to the interaction with an object (Eng et al., 1994).

Schillings and colleagues also investigated the reaction to tripping over an object, which involves feedback from a complex set of interrelated sensory inputs. They enhanced this pertubation through the use of more rigid and heavy obstacles. While these responses had a functional pattern similar to that found in the study of cutaneous reflex pathways, it is difficult to identify afferent pathway-specific responses due to the complexities of the responses. It is known that the responses were not related to the intensity of physical impact with the obstacle, indicating modulation by the nervous system plays an important role in determining these patterned responses (Schillings et al., 2000).

Characteristics of Gait in the Healthy, Older Adults

There has been extensive research into the effects of aging on normal patterns of gait (Barrett et al., 2010; Mills & Barrett, 2001; Winter et al., 1990). These studies have shown the older individuals tend toward a slower walking speed and increased double support phase, both safer and more stable conditions for walking. Systemic changes occur as a result of normal ageing which may have an effect not only on unperturbed walking patterns, but also on the functional ability to react to stimuli. The first broad class of changes that will be discussed are biomechanical constraints and neuromuscular decline that commonly accompany ageing. The result of these is functional modification of gait with respect to kinetics, kinematics and spatio-temporal parameters of gait.

Patterns of reduced muscular strength during unperturbed walking

Among the many changes that occur in the body with age, likely the most functionally noticeable is a gradual reduction in muscular strength and power. After the peak of maximal force capacity is reached in the third and fourth decade of life, it decreases with advancing age (Christ et al., 1992). This leads to incremental functional limitation and the onset of sarcopenia. Sarcopenia is the loss of muscle mass resulting in reduced strength and physical function, and is generally associated with underlying cellular and neuromuscular changes occurring as a result of age (Lang et al., 2010). Using various methods of detection, sarcopenia was seen to have a prevalence of 20% for men and 5% for women at age 65 to 70% for men and 15% for women at age 85 (Van Kan, 2009). Sarcopenia was also found to be significantly correlated with frailty in this population, indicating an association between reduced function and reduced strength capacity (Van Kan, 2009).

It should be noted that sarcopenia does not simply represent the loss of muscle mass resulting in a loss of strength, but also the constellation of underlying factors that may lead to this deterioration (Clark & Manini, 2008; Lang et al., 2010). Specifically, this involves alterations of neurological function, muscle contractile properties and other factors. Changes in muscular fiber type and differential reduction in motor unit strength or tissue loss across muscle groups is reflected in the differential strength changes that occur with age. Focusing on the lower limb, it has been seen that the strength capacity of dorsiflexors is maintained to a greater degree than that of the plantarflexors as one ages (Christ et al., 1992). Given the importance of ankle musculature in phase-transitions during gait, as well as in enabling sufficient toe clearance, it is important to determine if this affects mechanical responses to perturbations during walking.

Kinetics and kinematics during unperturbed walking

Assessment of healthy older gait has shown there to be several classifying features distinguishing 'young' gait from ‘old’, as well as between the healthy old and those predisposed to falls (Kerrigan, et al., 1998; Kerrigan et al., 2000). These characteristics are seen to be distinct from differences that arise with variability in chosen walking speed. Specifically, normal changes in gait due to aging include reduced peak hip extension, increased anterior pelvic tilt, reduced plantarflexion as well as reduced ankle power. Extending to those predisposed to falls, a further reduction in peak hip extension, knee flexion and knee power. Furthering this work, healthy young, old fallers and non-fallers were found to have a progressive reduction in peak hip extensor moment, indicating a progressive reduction of the hip joint range of motion and commensurate limitation on gait function (Kerrigan et al., 2001). This influences the differential role muscles of the lower limb play in controlling gait and how these neuromuscular changes may possibly influence the level and timing of responses to cutaneous electrical stimulation. The resulting kinematic changes are a slower self-selected walking speed due to a shorter step, rather than a quicker step (Winter, 1990).

Toe clearance during gait, while not consistently seen as being affected by age (Winter et al., 1990, Mills & Barrett, 2001) is seen as an important consideration in the determination of falls risk in older adults (Barrett et al., 2010; Hamacher et al., 2011). Though previous research found similar minimum toe clearance between the young and the old during the gait cycle(Levinger et al., 2012; Mills & Barrett, 2001; Winter et al., 1990), there is a paucity of research characterizing differences in toe clearance between young and old at multiple positions across the swing phase.

More subtle changes occur in gait with age, including a reduction in the variability in coordination of intralimb joint kinematics. Winter (1991) found significant reduction in the variability of hip and knee movement indicating a deterioration of dynamic balance control during walking. This behavioural pattern is further illustrated in a reduction in stride-to-stride variability in EMG activity throughout the body. This reduced variability indicates a more consistent pattern of output in the older group, again being associated with a reduction in capacity for dynamic stabilization. Variability in toe clearance is also seen in the elderly old as an indicator of falls risk, though in this case, increased variability was associated with greater risk (Barrett et al., 2010). This indicates that an optimal level of variability in joint kinematics is necessary to maintain dynamic stability and constricted joint kinematics alongside environmental constraints may result in increased variability in minimum toe clearance.

Changes to reflex control with age

In addition to understanding the integration of sensory input during locomotion (Zehr et al 1997), reflexes can be used as a neural probe to characterize deficits in neural control in response to orthopaedic injury (Hundza & Zehr, 2007) or neurotrauma such as stroke (Zehr et al 1998; Barzi & Zehr 2008; Zehr & Loadman, 2012; Zehr et al., 2012; Dragert & Zehr, 2013). As such previous research has begun to characterize age-related differences in reflex control during walking. For example, utilizing the stretch reflex and investigating short and long latency responses, it was found that the old displayed larger long latency responses while short latency responses were unaffected (Obata et al., 2010). Observations made in healthy old and young have also shown reduced H-reflex excitability during stance in the old, indicating impaired contribution of stretch reflex to joint stiffness during walking as well as a possible adaptive mechanism of gait regulation in response to afferent input (Chalmers & Knutzen, 2000; Chalmers & Knutzen, 2002).

While cutaneous reflexes in older adults have not been directly contrasted to the young, cutaneous reflexes have been investigated in an sample of adults with a average age of 64.27 ranging in age from 37-88 years (Zehr & Loadman, 2012). This group was used as a control for comparison to those with stroke and their responses to SP nerve stimulation qualitatively appeared similar to young adults. Responses to mechanical tripping in old participants demonstated several differences in both motor and mechanical output.

Older participants in treadmill-tripping studies show qualitatively similar responses to younger adults, with early swing dominated by an elevating strategy, late swing a lowering strategy and mid swing a mixture of the two (with repeatability found within-subjects) (Schillings, et al., 2005). While the general pattern of motor responses is unchanged in older adults, delays in motor response in the plantarflexors of the support leg show an altered response, also associated with a longer rise time. Older participants were less successful in preventing a fall, primarliy due to lower rates of moment generation, resulting in the requirement for further recovery steps or support to prevent a fall (Pijnappels et al., 2005).

Schillings and colleagues found in their study of 8 older adults (aged 60-73, mean age 65 years) that responses to tripping are qualitatively similar to those in younger adults when perturbations were introduced during early swing and during late swing, the two periods during which there was the greatest consistency of response between groups. Response latencies, however, were delayed in several muscles in the swing limb of older subjects during early swing, including the biceps femoris, tibialis anterior and soleus. Responses in late swing were delayed in the stance biceps femoris only (Schillings et al., 2005). Response amplitudes were found to be mainly similar between young and old as well, with early swing differences found in middle and late latencies in swing leg BF and RF. In late swing, amplitudes were again similar with significant differences seen in middle latency swing TA and stance BF showing an enhanced ability to maintain stability in the stance leg as the swing leg is moved through the corrective pattern (Schillings et al., 2005).

Upon reviewing the age-related changes that occur with natural aging process, it becomes clear there is, in fact, a gap in the literature with respect to a direct comparison of specific alterations in either neural control or functional kinematic outcomes from electrically evoked cutaneous responses between the young and the healthy old. While there is data expressing these responses in the old (Zehr & Loadman, 2012), there has been no direct comparison between young and old aimed at elucidating possible differences in response characteristics. Further, while research does exist investigating responses to mechanical perturbation, these are by definition a complex set of responses involving multiple sensory inputs. In order to address this gap, it would be necessary to isolate the cutaneous afferents in order to determine whether the functional stumble corrective response elicited in the young (Zehr et al., 1997) remains intact in older individuals. Additionally, it is required to show if there are any asymmetrical changes (between muscle groups) in motor or biomechanical (between joints) responses as have been seen in research investigating reflexes through other means (Obata et al., 2010).

Kinematic Measurement with 3D Motion Analysis

Kinematics in reflex studies have traditionally been captured using a direct measure of joint angle through a device such as electrogoniometers (Duysens et al., 1992; Schillings et al., 2000; Yang et al., 1998; Zehr et al., 1997). A primary benefit of these systems is an ability to collect at high sampling rates while maintaining synchroinization with EMG data collection. A limitation of these systems is their limited output of range at a single joint as well as their inability to collect important functional features such as toe clearance and centre of mass movement.

Optoelectric and optical motion capture have become increasingly popular in the assessment and treatment of gait. The primary benefit being whole body capture with little more effort required than more restricted capture. Linear displacement of body segments and specific landmarks, such as the distal position of the lower limb provding a measure of toe clearnce (Levinger et al., 2012), may also provide a new ground for research into the functional nature of electrically evoked responses in walking. Given the sensitivity of reflex studies to errors in measurement and especially measurement timing, though, it is necessary to ensure that data collected will be sufficiently accurate to provide a measure of joint motion at least as valid as electrogoniometers.

The primary tool that will be utilized for kinematic recording and anlysis in the current study is the Vicon T20S (Vicon Motion Systems, Oxford, UK) 3D camera-based motion capture system. It is required, therefore, to show that this system or systems like it are valid recorders of positional information. Furthermore, software used to analyze data produced by these systems must be shown to be capable of capturing the motion of objects moving in both a linear and angular fashion these are keys to assessing characteristics of human movement. DeLuzio et al. (1993) developed a method for validating the positional and angular accuracy of camera based systems. This involved using various models including an analog of the human knee joint and comparing processed motion capture data with a potentiometer. Under these circumstances, the error was less than 1° over a moving range of 38°. This was predicated on very careful calibration of the system before data collection. Although this study focused on movements with a single degree of freedom and thus did not use three dimensional processing of the data, they were able to determine that the camera system has potential to provide a valid representation of a key human movement.

A more specific and rigorous test of camera based systems was completed by Ehara et al. (1997). This study included two Vicon models (Vicon 140 and Vicon 370). The method of validation involved moving an object with known marker distances through the capture volume and then assessing the output of the capture system and its ability to provide marker trajectories at these known distances throughout the trial. In this, the two Vicon systems were able to provide trajectories with mean errors of 1.60 ± 1.82mm and 0.94 ± 0.39mm for the models 140 and 370, respectively.

In further investigations of the accuracy and precision with which a video motion capture system is able to capture linear motion, a sliding device of known properties was placed in the center of the capture space and recorded moving a set of markers at various speeds and in various locations within the capture volume (Everaert et al., 1999). Mean errors produced by this method were 0.34 mm, showing it to be more than acceptable. The limitation in applying this work to gait analysis, however, is that it was focused on accuracy within the context of a very small capture volume (less than $1m^3$). More recent work investigated a similar situation (accuracy within a small volume) and found the Vicon 460 to be highly accurate at locating markers within a space, but that camera positioning and calibration technique had an effect on the accuracy produced (Windolf, Götzen, & Morlock, 2008).

Addressing the models used to provide angular and other information from raw motion capture data was a study by Ferrari et al. (2008), which compared a number of different marker sets and biomechanical models for intra-model correlations between various common measures. This review showed that the biomechanical model used had more of an influence on the accuracy of results than did differences in the marker set. Highest levels of agreement were seen between models based on CAST (Cappozzo et al., 1995), LAMB (Rabuffetti & Crenna, 2004) and Total 3D Gait (T3Dg, Aurion s.r.l., Milan, Italy) systems. With this said, there was strong correlation on many measures between all models with all showing very strong intra-protocol agreement, making them suitable for intrasubject comparisons as well as between subject comparisons. Less agreement was seen in out of plane motion, which is notable when determing kinematic outcome measures for study (Ferrari et al., 2008).

It can be seen from the literature that there is a necessity in using the 3D motion capture system that it be properly calibrated and itself able to produce a valid representation of the actions being captured. With that said, there is little in the way of research investigating the accuracy or precision of these devices in assessing dynamic movement through larger capture volumes, which is presupposed in gait analysis. Many published articles utilizing these systems as a method of validation do not provide references to justify the choice but provide only a manufacturer and less often, a model number.

Summary and conclusions

A combination of neurophysiological and biomechanical changes that occur with age lead to alterations in postural maintenance strategies. This affects both quiet standing (Horak et al., 1989), as well as in response to a perturbation during walking (Schillings et al., 2005). The etiology of these changes is multifactorial, and includes alterations in joint mechanics, strength losses and changes to the neural circuitry underlying the control of movement. Much research has been done in the biomechanics of age-related change to in gait (Winter, 1991). Contributions of reflexes to the neural control of walking are evident from the patterns of task and phase-dependent modulation seen in both motor activity and mechanical outcomes correlated with this motor activity (Zehr et al., 1997). The functional correlate of evoked cutaneous reflexes of the superficial peroneal nerve cutaneous field is the stumble-corrective response, similar to that seen in mechanical perturbation (Zehr & Stein, 1999). Indication of differences in reaction to tripping in older adults make it important to determine how much of this change is ocurring at the neural level.

The neural control of walking has been investigated in younger, healthy adults, but the volume of alterations to human gait that are found with age make this research difficult to generalize to the older population. Further, quantifying the degradation in the neural pathways supporting the stumble correction response could assist in understanding normal decline of the sensorimotor system with aging as well as provide a novel approach to establishing early neuromechanical markers in those individuals with potential future fall risk. To this end, we compared and contrasted responses to electrical stimulation of the superficial peroneal nerve in younger and older adults. We evaluated toe clearance as an end-point determinant of object avoidance, as well as joint kinematics and motor activity of muscles acting at the knee and ankle.

Chapter 2: Manuscript

Introduction

Falls account for 74% of injury-related hospitalization for seniors (Canadian Institute for Health Information, 2011) and in a study of community-dwelling seniors, tripping during walking was observed to account for 59% of falls (Berg et al., 1997). There are age-related differences in human biomechanics and motor control that may contribute to this high prevalence of tripping-related falls in older adults. Examples of these differences include slowed nerve conduction velocity (Noris & Shock, 1953) as well as altered patterns of muscle activity and joint torque during walking such as increased coactivation of muscles at the ankles (Schmitz et al., 2009). There is also an increased reliance on hip extensor power relative to plantarflexion in stance (DeVita & Hortobagyi, 2000). During gait, fall-related trips often occur when the swing foot encounters an obstacle or the ground and the individual in unable to recover from the perturbation. As such, measures of toe clearance (Mills et al., 2008) and response to perturbation (van den Bogert et al., 2002; Weerdesteyn et al., 2005) are considered key factors of fall risk. No age-related differences were seen in mean minimum toe clearance (MTC) in older adults (Mills et al., 2008) nor in some clinical populations (Levinger et al., 2012). However, toe clearance across the swing phase has not been characterized in detail in these populations nor has its relationship to the response to perturbation during gait. Further, reflex control studies have shown age related differences between young and older adults. For example in older adults there is a noticeable decrease in the soleus H-reflex excitability during the stance phase of walking (Chalmers & Knutzen, 2000).

In response to unexpected mechanical tripping during walking young healthy adults produce a stumble corrective response involving plantarflexion of the ankle and flexion of the knee during swing phase assisting the swing limb in avoiding the obstacle (Eng et al., 1994; Schillings et al., 1996; Schillings et al., 2000). In healthy young adults, it has been observed that stimulation of cutaneous afferents also result in coordinated neural and mechanical outcomes that serve important regulatory functions during human locomotion (Duysens et al., 1992; Zehr et al., 1997). Reflexes evoked with electrical stimulation of the superficial peroneal nerve, which innervates the cutaneous receptors on the dorsum of the foot, have been shown to assist with obstacle avoidance during the swing phase of walking similar to the mechanically evoked stumble corrective response. Originally identified in animal models (Drew & Rossignol, 1987; Forssberg, 1979), this class of responses have since been observed in healthy adults, where it is characterized by inhibition of dorsiflexion and facilitation of knee flexion during swing (Van Wezel et al., 1997; Zehr et al., 1997). Preservation of these responses have been established as an important marker in the evaluation of the integrity of neural control after orthopaedic (Hundza & Zehr, 2007) and neurological (Zehr, Loadman & Hundza, 2012; Zehr & Loadman, 2012; Zehr, Fujita, & Stein, 1998) injury. In response to unexpected mechanical tripping during walking, older adults display reduced amplitude ankle and knee kinematic responses supporting a potential degradation of this stumble corrective response (Schillings et al., 2005). These responses, however, involve a complex of different afferent pathways and age-related change to the specific contribution of cutaneous input to the regulation of gait in terms of detailed kinematic and muscular responses has not been specifically quantified.

Understanding potential age-related degradation in neural and kinematic stumble correction responses is vital given their putative functional relevance during walking for obstacle and trip avoidance. Quantifying the degradation in the neural pathways supporting the stumble correction response could assist in understanding normal decline of the sensorimotor system with aging as well as provide a novel approach to establishing early neuromechanical markers in those individuals with potential future fall risk. To this end, we compared and contrasted responses to electrical stimulation of the superficial peroneal nerve in younger and older adults. We evaluated toe clearance as an end-point determinant of object avoidance, as well as joint kinematics and motor activity of muscles acting at the knee and ankle.

Methods

Participants

12 healthy older adults (8 male and 4 female, mean age 76.7 ± 4.8 years) and 17 younger adults (10 male and 7 female, mean age 25.4 ± 5.4 years) free of any known neurological or musculoskeletal impairment or history of metabolic conditions participated in this study. Informed consent was obtained and the study was conducted in accordance with the University of Victoria Human Research Ethics Board.

Protocol

Participants performed a single treadmill walking task of approximately 9 minutes in duration at a self-selected pace. Cutaneous responses were evoked during walking by stimulating the superficial peroneal (SP) nerve on the anterior aspect of the ankle. Muscle activity and kinematics were measured during stimulated and non-stimulated walking cycles. Data from non-stimulated walking trials will be referred to as “undisturbed”.

Nerve stimulation

Percutaneous electrical stimulation was delivered to the superficial peroneal nerve (SPn), innervating cutaneous receptors on the dorsum of the foot. Stimulation was generated by a Grass S88 stimulator (Grass Instruments, AstroMed Inc.) connected in series with a SIU5 isolator and a CCU1 constant current unit. Stimulation consisted of trains of 1ms square-wave pulses (5 x 1ms at 300Hz) at approximately two (2) times the intensity of the threshold for radiating paresthesia over the dorsum of the foot. Stimulation was delivered pseudorandomly throughout the gait cycle and delivered every 1 to 3 full cycles. Approximately 240 stimulus events were captured during a walking trial.

Electromyography (EMG)

Surface EMG recordings were made ipsilateral to the site of stimulation from the tibialis anterior (TA), gastrocnemius, medial-head (MG), soleus (Sol), vastus lateralis (VL) and biceps femoris (BF). Signals were amplified and bandpass filtered at 100-300 Hz (P511, Grass Instruments, AstroMed Inc.).

Kinematics and gait parameters

Kinematics were recorded through the use of an 8 camera Vicon T20S 3D motion analysis system (Vicon Motion Systems, Oxford, UK). Kinematic reconstruction of anatomical landmarks was based upon the 6-degrees of freedom model (Collins et al., 2009), along with anatomical landmarks defined within Visual 3D (C-Motion, Germantown, MD). Joint angle and angular velocity as well as toe height data were collected. Sagittal plane joint angles were calculated for the ankle and knee bilaterally from segment markers for the foot, shank, thigh and pelvis. Joint angular velocity was calculated as the time-differential of joint angular displacement. Toe clearance was determined as the vertical height of the lateral toe (5^{th} metatarsal head) marker relative to the walking surface. The beginning of each gait cycle, corresponding to the timing of heel-strike, was based on the time index when the ipsilateral heel maker reached its lowest vertical point. Swing initiation was taken as the beginning of forward motion of the lateral toe marker.

Data Acquisition and Analysis

EMG data were sampled at 1000 Hz using a 16-bit A/D converter connected to a computer using custom-written LabView software (National Instruments, Austin, TX). Kinematic data were sampled at a rate of 100Hz using the Vicon Nexus 1.7.2 software and analyzed using the Visual 3D software. Kinematic data were interpolated, synchronized for comparison with EMG and Butterworth filtered at 10Hz data using custom-written Matlab (Mathworks, Natick, MA) software.

Post-acquisition, the EMG and kinematic data were partitioned, based on time, into 16 equal bins with bins 1-10 representing stance phase and 11-16 representing swing. EMG and kinematic responses to nerve stimulation (disturbed), aligned to the delivery of stimulation, were each averaged within each step cycle bin. EMG and kinematics recorded without stimulation (undisturbed background) were also averaged within each step cycle bin. Reflex EMG and kinematics (subtracted traces) were calculated by subtracting the averaged undisturbed background data from the data with stimulation (disturbed) at each step cycle bin (10-20 observations per bin).

Cutaneous EMG and Kinematic responses

EMG data were full-wave rectified prior to averaging. Responses were only considered significant if they exceeded a 2-SD band calculated from prestimulus subtracted values. EMG traces were examined for the cumulative average over the 125ms post-stimulus ($ACRE_{125}$) (see (Zehr, et al., 2000)). Mean subtracted and undisturbed (background) EMG (bEMG) were normalized to the maximum average bEMG during the undisturbed walking cycle. Angular displacement responses were taken as the maximal excursion in mean subtracted traces within a window of 70ms-220ms. Angular velocity responses were taken as the peak amplitude of mean subtracted velocity within this window.

Statistics

Descriptive statistics include mean, standard deviation (SD) and standard error of the mean (SEM). Repeated measures analysis of variance was conducted separately for undisturbed background EMG and kinematics (joint angular displacement and velocity and toe clearance), as well as for reflex (subtracted) EMG and kinematics responses to determine significant main effects for age and statistically significant age-bin interactions. Fisher's LSD test was used to investigate *post hoc* significant interactions. Planned comparisons were conducted on reflex EMG ($ACRE_{125}$) responses at bins where there were significant differences between age groups in reflex joint angular displacement for the ankle and knee as determined from analysis of variance and post hoc Fisher's LSD. Specifically, planned comparisons were conducted in TA, MG and Sol at based on significant findings in reflex ankle angular displacement, and in VL and BF based on significant findings in reflex knee angular displacement.

To establish the presence of significant responses to stimulation compared to background in EMG activity and angular displacement and velocity subtracted responses separately within young and old groups, *t*-ratio tests were used. To determine significant responses in toe clearance from background (bin 6) to stimulation, repeated measures analysis of variance and Fisher's LSD on significant age-bin interactions was conducted comparing all bins to bin 6 separately within age groups. Statistical significance for all tests was set at $p<0.05$.

Results

Responses to cutaneous stimulation during walking

Ankle Kinematic and EMG Responses

Average, group ankle kinematic and EMG responses are shown in Figure 1. Based on *t*-ratio analysis, the young displayed plantarflexion responses in ankle angle during swing phase that were significantly different from background (bins 10-15). Similarly older subjects produced significant plantarflexion displacement responses across swing-phase bins (11 and 14).

Upon direct comparison of the age groups (analysis of variance) a reduced amplitude of plantarflexion displacement response was found in the old compared to the young during swing (bins 10, 13 and 15), with mean differences of 3.2°, 1.3° and 1.1° respectively(($F(15, 390)=3.17$, $p<0.001$, Fisher's LSD *post hoc*). Further, during late stance (bin 8), the young had a significant dorsiflexion displacement response from background, which was not evident in the old. The differences noted above are illustrated in representative single subject comparisons in Figure 2c.

Based on *t*-ratio analysis, the young displayed plantarflexion angular velocity responses that were significantly different from zero throughout swing at bins 10-12, 14 and 15. The older adults similarly produced significant plantarflexion displacement at bins 10, 11, 12 and 14. When the groups were directly compared some differences were noted (Significant age/bin interaction ($F(1,15)=3.31$, $p<0.001$), Fisher's LSD *post hoc*). Plantarflexion velocity was significantly slower for the old compared to the young for bins 10, 11 and 15 with mean differences of 58°/s, 47°/s and 30°/s respectively. In addition to having responses of significantly lower velocity responses, the older group (122 ± 4.8 ms) displayed significantly later angular velocity responses at the ankle than the young (109 ± 4.0 ms). Representative comparisons of single subject traces are shown in figure 2d.

Responses in TA were significantly different from background in the young at the stance-swing transition (9-11) and mid-late swing (14-16) and in the old from the stance-swing transition through to the end of swing (9-16) (Figure 1). In the young an additional significant difference from baseline was seen at mid-stance (bin 4). Responses in MG were significantly different from background in the young in stance at bins (1, 3-8, 10) and in the old in stance at bins (2 and 4-7). The young also displayed significant responses in swing bins 11 and 13. Responses in Sol were significantly different from background in the young during stance (bins 4-5, 9-10) and swing (11, 12 and 14-16) and in the old at bins 2, 5 of stance and mid-swing (12).

In TA, a significant main effect for age was found with the old showing greater suppression of TA (7.3 ± 1.0% of peak bEMG) than was seen in the young (4.5 ± 0.9% of peak bEMG) throughout the gait cycle ($F(1,26)=4.30$, $p=0.042$) (see Figure 1). Planned comparisons conducted at bins where joint displacement was significantly different between the old and young (bins 10, 13 and 15) displayed significant differences in TA and MG at mid-swing (bin 13). At this point, the older adults had a significantly greater suppression of TA activity compared to the young ($p=0.016$). This was accompanied by a significant excitation response to stimulation in MG at bin 13 in the young compared with the old. Representative single subject EMG traces can be seen in in Figure 2a and b.

Toe Clearance Response

There was significant age-bin interaction for toe clearance responses ($F(15,345)=1.83$, $p=0.03$, Fisher's LSD *post hoc*). In both groups, in response to stimulation, there was a reduction in toe height in early swing (bin 9 and 10 in young and 11 in the old) and late swing in the young and old (bin 15)(Figure 3). In the young in response to stimulation there was an increase in toe height at mid-swing (bins 12 and 13) while not seen in the old. Peak elevation was seen in the young and the old at bin 12 of 9.8±1.6 mm and 2.6±2.0 mm respectively. Significant differences between the groups were seen at bins 12 and 13 with the young showing significantly greater toe lift than the old of 7.9 mm and 4.2 mm at bins 12 and 13, respectively.

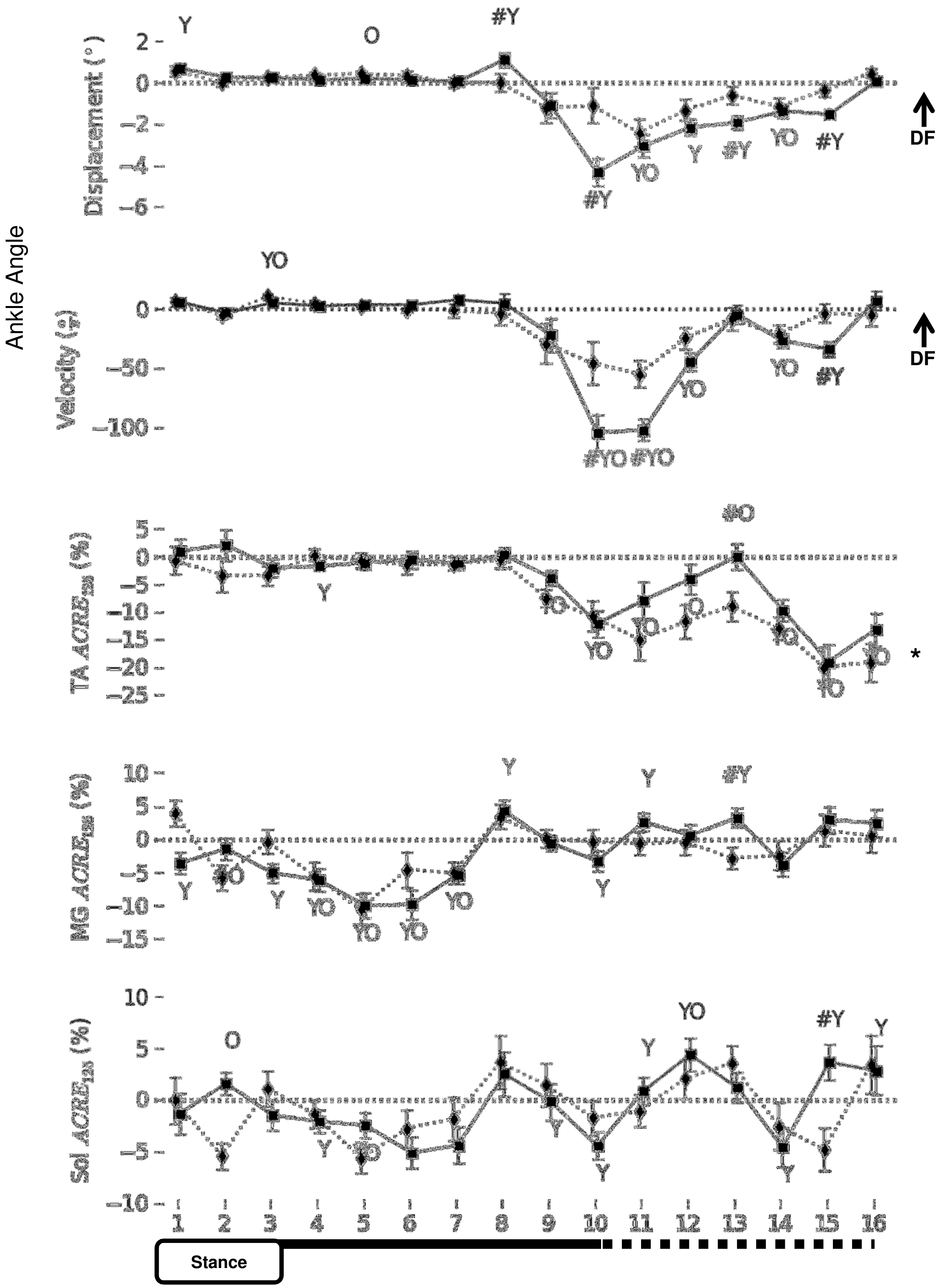

Figure 1. Group mean, subtracted ankle kinematic and EMG responses to stimulation occurring at each phase of the step cycle in old (dashed) and young (solid). EMG values are represented as the average cumulative reflex EMG after 125ms ($ACRE_{125}$) and are normalized to peak muscle activation throughout the step cycle for each subject. O indicates a response significantly different from zero in the Old, Y indicates a response in the young and hash (#) indicates a significant between-group difference at that phase of the gait cycle. (*) indicates a significant main-effect for age across the gait cycle. All tests conducted at α=0.05.

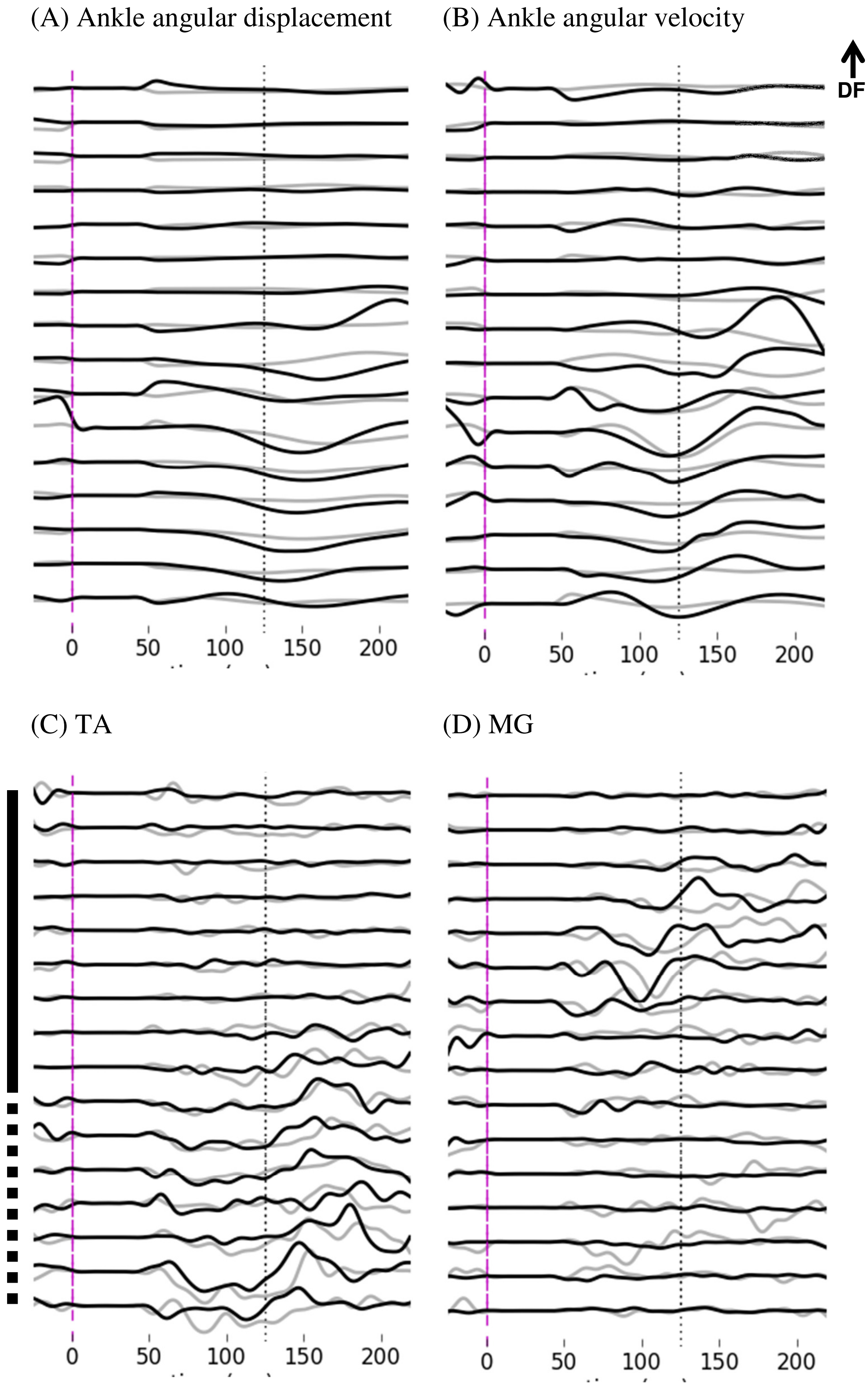

Figure 2. Representative single subject comparisons of normalized, subtracted ankle angular displacement (A) and ankle angular velocity (B) as well as EMG for the (A) TA and (B) MG during each phase of the gait cycle with the young indicated with solid black and old in gray. EMG traces are reported in percent of peak background EMG for the subject. Angular displacement is reported in degrees and angular velocity in degrees per second. Stance phase bins are indicated by a solid, vertical bar with swing phase by a dashed bar.

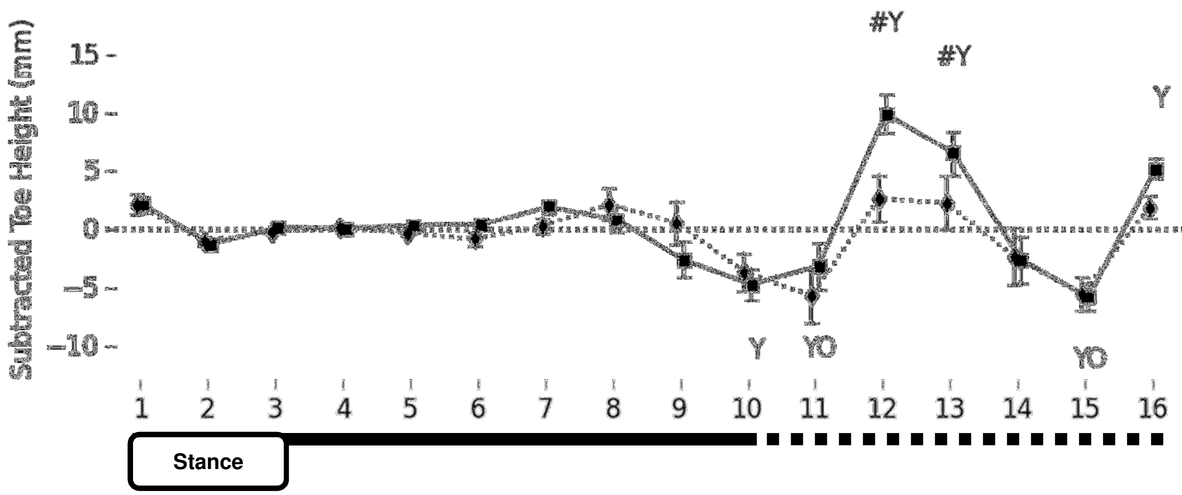

Figure 3. Group mean, subtracted toe height occurring throughout the gait cycle with positive values indicating greater toe height in disturbed cycles compared with controls. O indicates a response significantly different from zero in the Old, Y a response in the young and hash (#) indicates significant between group difference determined through analysis of variance with Fisher's LSD conducted to p<0.05 level.

Knee kinematic and EMG responses

Average kinematic and EMG group responses to perturbation for the knee are shown in Figure 4, with single subject kinematic comparisons shown in Figure 5. Both groups showed a significant knee flexion displacement response from background (based on *t*-ratio analysis from background) throughout the majority of stance and swing. There was a significant flexion displacement response in stance for the young in stance bins (1-4 and 6) as well as for the old (bins 1, 2, 4 and 5). There was a significant displacement response toward flexion in swing for the young (bins 12, 13, 14 and 16) and for the old (bins 12 and 13). Further there was a significant displacement response toward extension in bins 9 and 15 in both the young and old in late stance and late swing respectively. Both groups showed a significant knee angular velocity response toward flexion during swing (bin 11-13 for the young and Bin 12 and 13 for the old) and during heel strike (bin 1 and 2 for the young and bin 1 for the old). Terminal swing showed significant extension response in both groups (bin 15 and 16 for the young and bin 16 for the old).

A significant age/bin interaction was seen in both angular displacement ($F(15,405)=2.35$, $p=0.003$) and velocity ($F(15,405)=1.18$, $p=0.03$) knee responses. Specifically, the old displayed significantly less knee flexion displacement, compared to the young in mid-swing (bins 13 and 14). Knee angular velocity flexion response was also reduced in the old compared to the young in bin 12.

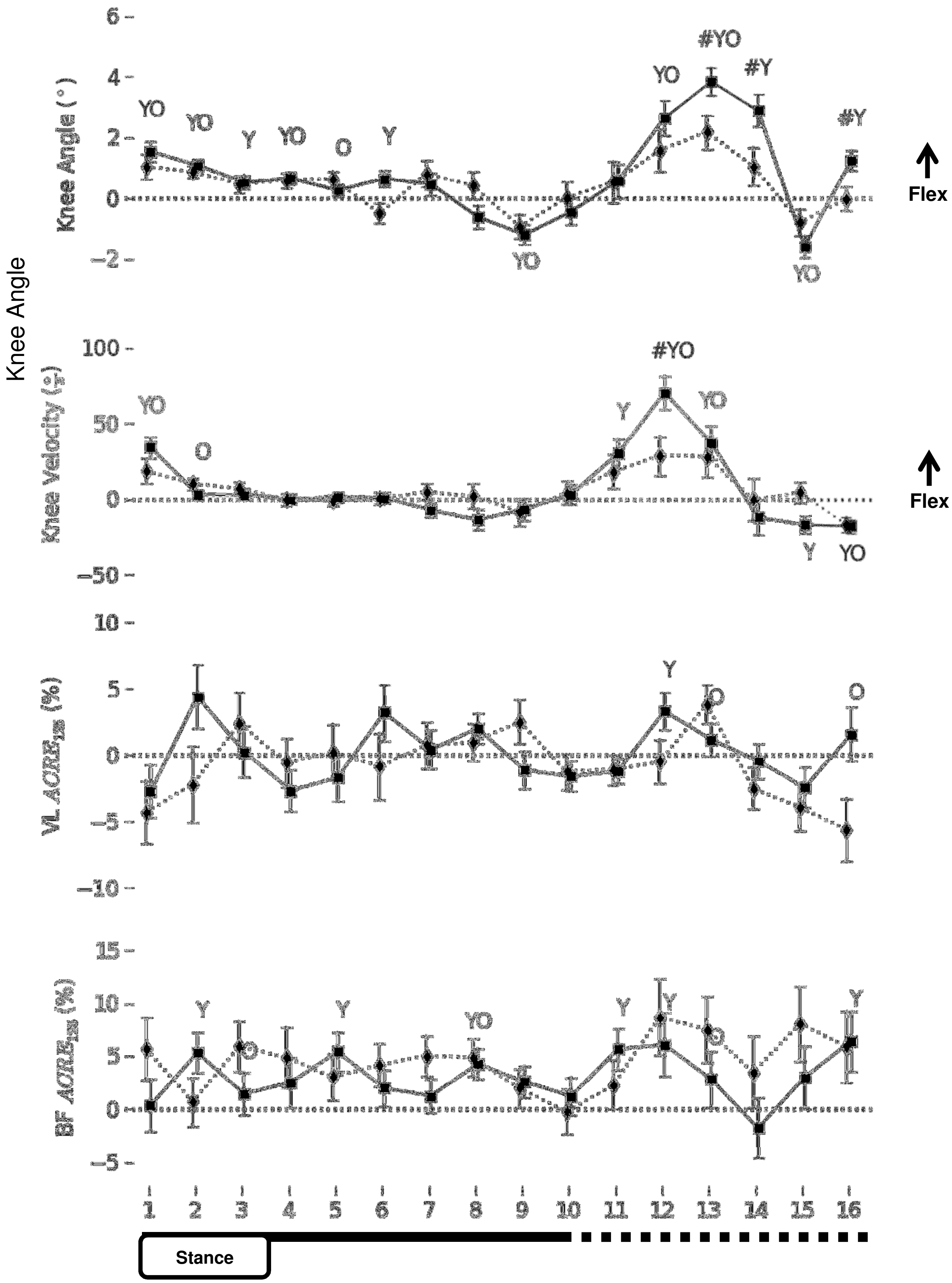

Figure 4. Group mean, subtracted knee kinematic and EMG responses to stimulation occurring at each phase of the step cycle in old (dashed) and young (solid). EMG values are represented as the average cumulative reflex EMG after 125ms (ACRE125) and are normalized to peak muscle activation throughout the step cycle for each subject. O indicates a response significantly different from zero in the Old, Y indicates a response in the young and hash (#) indicates a between group difference determined through planned comparison at that phase of the gait cycle with all tests conducted at α=0.05.

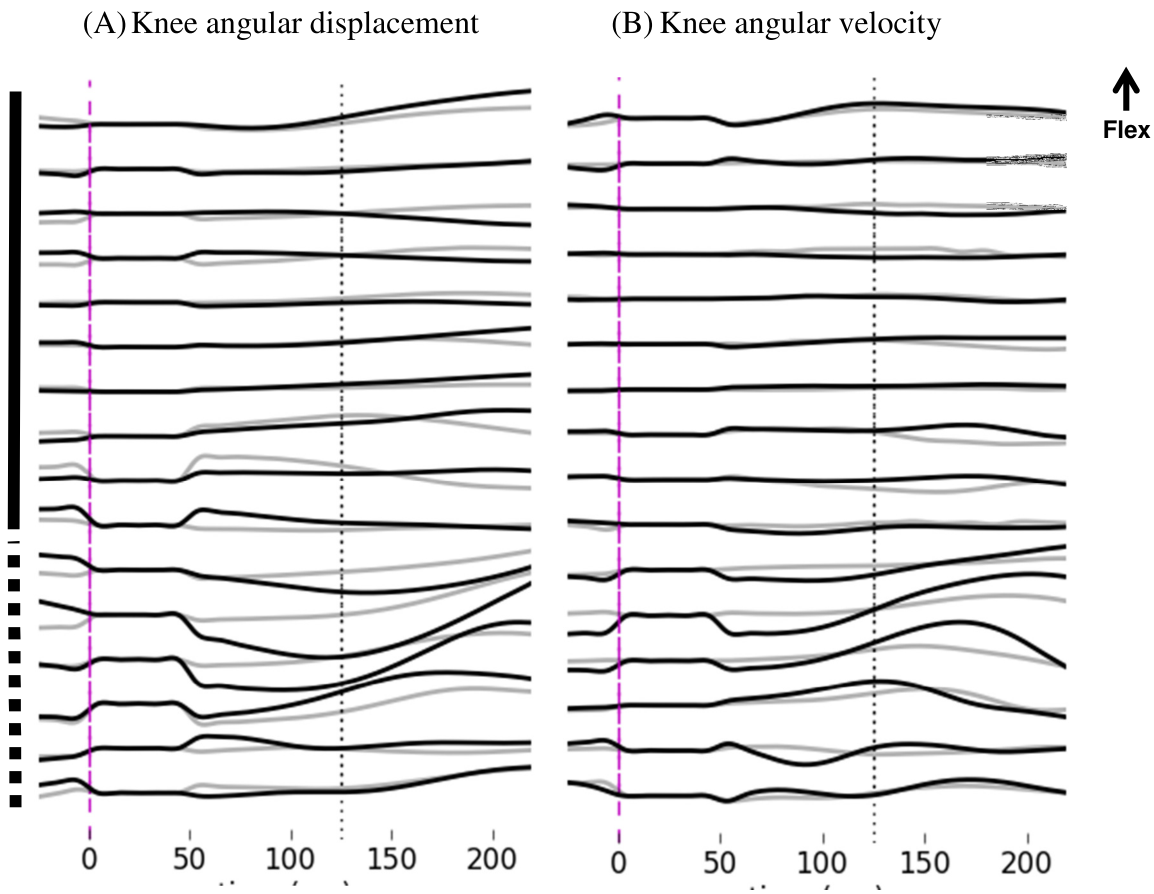

Figure 5. Representative single subject comparisons of, subtracted (A) Knee angular displacement and (B) knee angular velocity during each phase of the gait cycle with the young indicated with solid black and old in gray. EMG traces are reported in percent of peak background EMG for the subject. Angular displacement is reported in degrees and angular velocity in degrees per second.

Background walking pattern

Group averages for kinematic peak range of motion and peak velocities are displayed in Table 1 and group average spatiotemporal parameters of gait indicated in Table 2. Although the older group did walk at a reduced self-selected speed (1.0 $\frac{m}{s}$) compared with the young (1.1 $\frac{m}{s}$), there was no significant correlation in either group between spatiotemporal gait parameters, including gait speed, and joint angular velocities. Furthermore, there was no significant difference between groups in stance time percentage of gait cycle time. It is noted that leg length as well as stride length are significantly different between groups.

Ankle Kinematics and EMG

Group averages of joint angle displacement and angular velocities for the ankle are shown in Figure 6. The direction of ankle displacement and angular velocity was in the same direction at most bins across the gait cycle between young and old. However, the old had greater dorsiflexion displacement through the stance-swing transition at bins 9 and 10 and reduced plantarflexion displacement in early swing at bins 11 – 13 compared to the young ($F(15,360)=8.17$, $p<0.001$, Fisher's LSD *post hoc*). Additionally the old had reduced plantarflexion velocity at bins 9 and 10 and reduced dorsiflexion velocity at bins 12 and 13 compared to the young ($F(15, 405)=6.23$, $p<0.001$, Fisher's LSD *post hoc*). TA activity was significantly greater during early stance (bins 2-4) and mid swing (bin 13 and 14) in the old ($F(15,390)=1.91$, $p=0.02$). Activity was higher during these same phases of the gait cycle in Soleus with older participants showing relatively greater activity during early stance (bins 3, 4 and 7) as well as early swing (bin 12) ($F(15,405)=1.95$, $p=0.02$). In MG, there was no age/bin interaction, but there was a main effect for age (F(1,27)=10.53, p=0.003) indicating a greater level of overall activity in the old compared to the young.

Table 1. Peak mean and SD for joint angular displacement and velocity at the ankle and knee joints during undisturbed walking. (*) indicates values in the young were significantly greater than those in the old (p<0.05).

	Old	**Young**
Peak Range of Motion		
Ankle Dorsiflexion	12.7 ± 4.3°	10.7 ± 3.4°
Ankle Plantarflexion	*9.8 ± 4.7°*	*18.4 ± 6.4°* *
Knee Flexion	*48.6 ± 5.97°*	*53.8 ± 4.0°* *
Peak Joint Velocity		
Ankle PF	*139.0 ± 38.2 °/s*	*176.1 ± 38.2 °/s* *
Ankle DF	*63.0 ± 23.1 °/s*	*98.9 ± 39.7 °/s* *
Knee Flexion	*215.0 ± 29.1 °/s*	*242.7 ± 21.9 °/s* *
Knee Extension	*204.6 ± 38.8 °/s*	*288.8 ± 29.1 °/s* *

Table 2. Group averages for spatial and temporal gait parameters during undisturbed (control) gait cycles. (*) indicates values which are significantly different between groups (p < 0.05)

	Old	**Young**
Walking speed ($\frac{m}{s}$)	*1.0 ± 0.16*	*1.1 ± 0.18* *
Stride time (s)	1.14 ± 0.07	1.17 ± 0.08
Stance time (%)	0.61, ± 0.09	0.63 ± 0.02
Stride length (m)	*1.16, ± 0.15*	*1.29 ± 0.16* *
Peak Toe Clearance (cm)	*8.4* ± 0.7cm	*8.9* ± 1.3cm
Leg Length (m)	*0.81, ± 0.05*	*0.83 ± 0.08* *

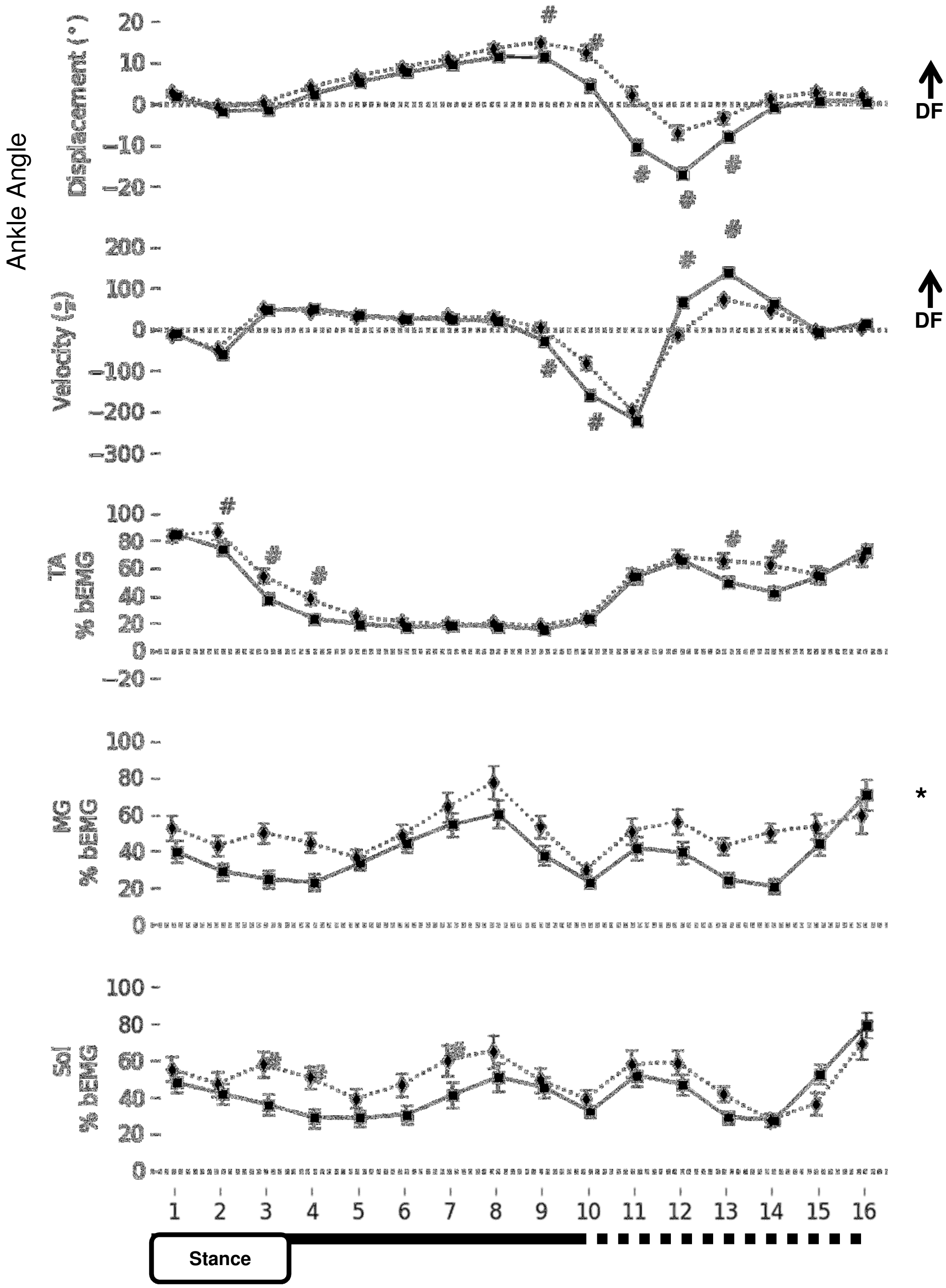

Figure 6. Grouped data representing average undisturbed kinematics values capture during control cycles with solid lines indicating young and dashed indicating old with error bars indicating standard error. (#) indicates a between group difference at that phase and (*) indicates a significant main effect of age with all tests significant at p<0.05 level.

Knee Kinematics and EMG

Group averages of joint angle displacement and angular velocities for the knee are shown in Figure 7. The direction of knee displacement and angular velocity was in the same direction at most bins across the gait cycle between young and old. However, the old showed less knee flexion displacement at bins 10-13 and 16 compared to the young ($F(15,360)=4.89$, $p<0.001$, Fisher's LSD *post hoc*). Additionally the old had reduced knee flexion velocity at bins 1, 9 and 10 and reduced extension velocity at bins 13 -16 compared to the young ($F(15,405)=5.43$, $p<0.001$, Fisher's LSD *post hoc).* Muscles acting at the knee showed higher activation levels during stance and terminal swing in the old compared to the young. VL activity was higher in the old than the young during stance (bin 4) and terminal swing (bin 16) ($F(15,405)=2.35$, $p=0.003$, Fisher's LSD *post hoc*). BF activity was higher in the old than the young during stance bins 2-4 and again leading into heel-strike at bin 16 ($F(15,405)=5.44$, $p<0.001$, Fisher's LSD *post hoc*).

Toe clearance

Group averages of toe height throughout the undisturbed gait cycle are shown in Figure 8. Although there was no significant difference in peak toe height achieved (Table 2), toe lift was significantly reduced after toe off (bins 11 and 12) in the old compared with the young. Greater toe clearance was also seen in late swing and at heel-strike (bins 1 and 16) in the young compared with the old ($F(23,345)=1.83$, $p=0.03$, Fisher's LSD *post hoc*).

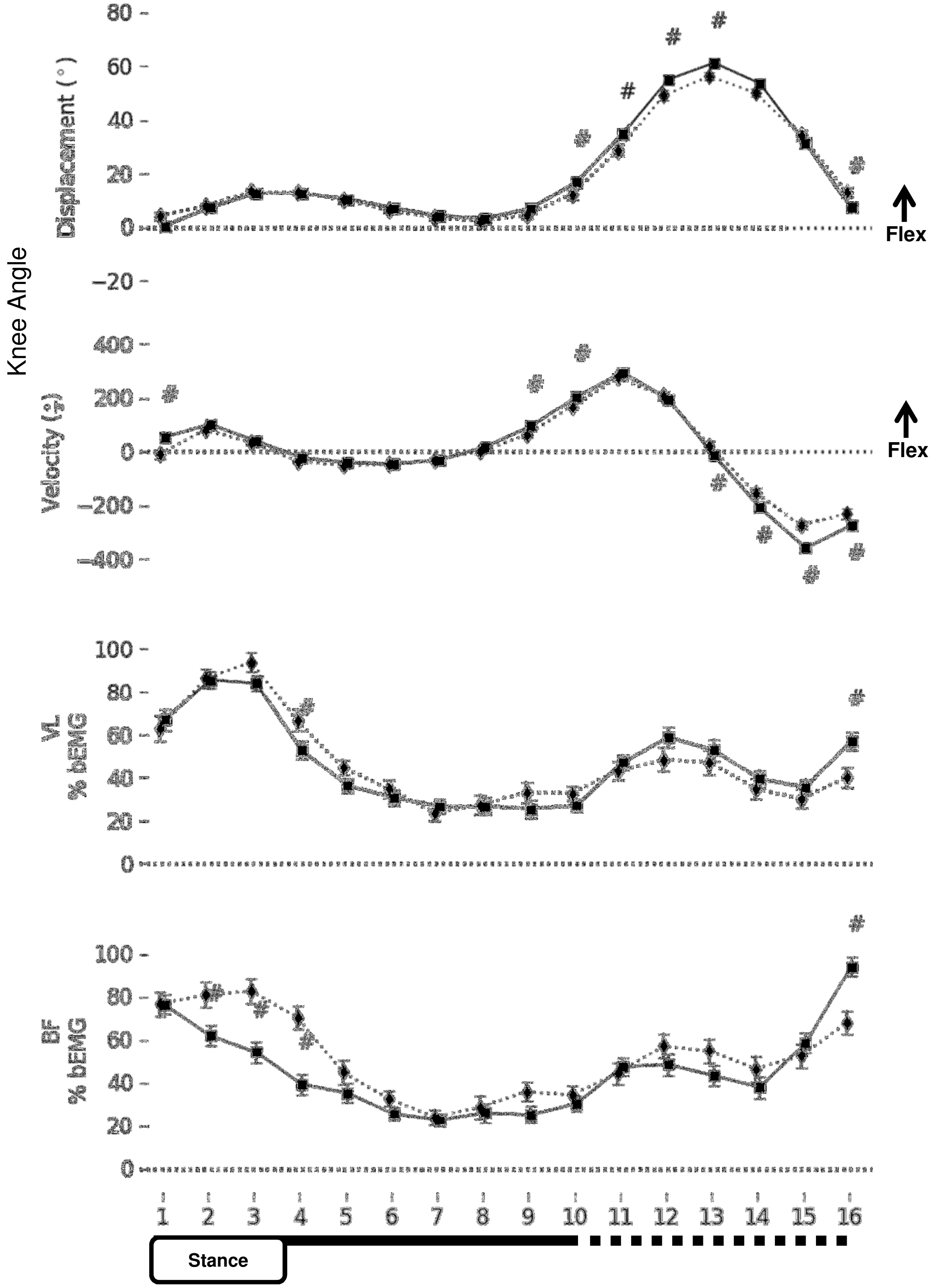

Figure 7. Grouped data representing average undisturbed kinematics values capture during control cycles with solid lines indicating young and dashed indicating old with error bars indicating standard error. (#) indicates a between group difference at that phase.

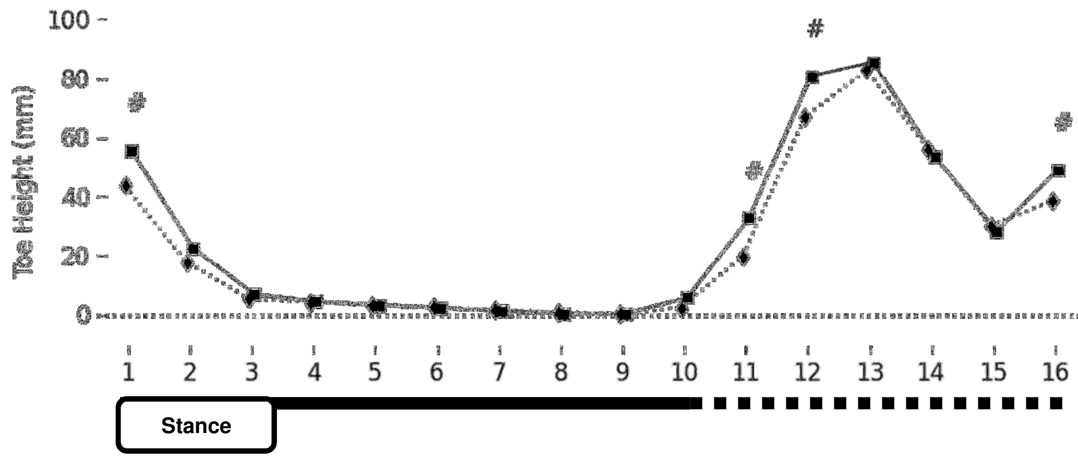

Figure 8. Grouped data representing average undisturbed toe height throughout the gait cycle with solid lines indicating young and dashed indicating old and error bars indicating standard error. (#) indicates a between group difference at that phase of the gait cycle.

Discussion

This paper is the first to directly compare and contrast electrically evoked cutaneous reflexes during walking in young and older adults, and four main outcomes were found. Overall, the responses in joint kinematic and muscle activity are in the same direction from background at similar bins in both groups, demonstrating that elements of the basic neuromechanical stumble correction response are conserved with age. However, the magnitude of the kinematic responses in terms of joint displacement, angular velocity and toe clearance in older adults were significantly smaller than in the young with the old failing to show a significant toe elevation response. There were also significant differences in reflex muscle activity between age groups. Further, during undisturbed walking, there were differences in gait kinematics and muscle activity between the young and old. Smaller kinematic responses superimposed on a different undisturbed gait profiles may reflect early degradation of the functional capacity for object avoidance during gait. This early degradation is likely a prodromal sign of increased fall risk, supporting the potential use of cutaneous reflexes in quantifying degradation of neuromuscular control and its contribution to fall risk.

Gait profile differences between young and old

The general direction of gait kinematics and level of muscle activity in the lower extremities across gait cycle phases were similar in the young and old during undisturbed walking cycles. However there were some important differences. Supporting previous work (Winter et al., 1990), we found reduced ankle and knee range of motion in older participants as well as reduced peak velocities, both of which have been shown to be linked with a redistribution of joint torques favouring proximal musculature (DeVita & Hortobagyi, 2000). In the current study the older adults showed more dorsiflexion at late stance which may be linked with reduced push-off leading into early swing (Winter et al., 1990) and reduced plantarflexion displacement throughout swing.

During swing muscle activity in TA was greater in older adults, when compared to the young. This potentially contributes to relative greater dorsiflexion seen in older adults as well as may relate to the known age-related increase in passive tissue resistance at the ankle (Vandervoort et al., 1992). Current results showed increased activity in the old in all recorded muscles during early stance. This generally elevated motor activity during early stance is indicative of a joint stabilizing strategy and could possibly be associated with a more "tentative" or more stable gait, a tendency seen previously in older adults (Winter et al., 1990). Further, previous research has shown strength losses at the ankle to occur asymmetrically with plantarflexor strength deteriorating more with age than dorsiflexor strength (Christ et al., 1992). This may be partly responsible for the finding of significant increased activity in MG across the gait cycle.

At the knee, the older adults displayed significantly reduced range of flexion and peak flexion and extension velocities in stance and swing. This could result in reduced power output at the knee similar to previously seen reduced mechanical work of the knee during both weight acceptance and push-off phases of stance in healthy older adults (DeVita & Hortobagyi, 2000). Similar to the ankle, these reduced joint kinematics at the knee may be related to a more "tentative" gait pattern with age (Winter, 1991).

Although some studies have shown toe clearance during undisturbed walking to remain relatively unchanged in older adults (Mills & Barrett, 2001; Winter et al., 1990), others found toe clearance to be a relevant factor in differentiating fallers from non-fallers (Barrett et al., 2010). In the current study, the two groups showed a similar pattern of toe clearance; however older adults displayed a reduced height throughout early to mid-swing and again at terminal swing. This difference in toe clearance could result in a greater risk for tripping in older adults. Combined alterations to kinematics at the trunk, hip, knee and ankle could result in reduced toe clearance with the foot acting as the end-point effector for the lower limb kinematic chain. The combined contributions of altered knee and ankle kinematics to reduced toe clearance are demonstrated in this data and create an altered foundation upon which any reflex is superimposed.

Stumble correction response is generally conserved yet blunted with age

Overall, electrical stimulation of the superficial peroneal nerve elicited a similar pattern of kinematic and muscle activity responses, in terms of the direction and bin, in both groups. This demonstrates a general conservation of a functional object avoidance response in older adults, characterized by reduced dorsiflexion at the ankle in early swing supported by suppression of the TA and knee flexion in mid-swing supported by excitation of knee flexors. Ankle angular velocity responses, in both groups showed significant facilitated plantarflexion following toe-off, followed by passive plantarflexion through mid-swing (i.e. reduced dorsiflexion). In TA, inhibitory responses were seen in all swing-phase bins in the old and in all but one in the young. Additionally, both groups displayed suppressive responses in MG in stance-phase, consistent with previous findings (Zehr et al., 1997). Kinematic and EMG responses seen here are consistent with previous studies using electrical stimulation of the SP nerve in healthy young adults (Zehr et al. 1997; Van Wezel et al., 1997). Additionally, studies investigating responses to mechanical perturbation, though involving a more complex group of sensory afferents, also have similar findings (Eng et al., 1994; Schillings et al., 2005).

While the pattern of kinematic and EMG responses were similar between groups, the magnitudes of joint kinematic (displacement and angular velocity) responses were significantly reduced in the older adults compared to the young and some differences were also noted in EMG responses. These differences contributed to a blunted toe clearance response in the older adults compared to the young in mid swing. Thus the obstacle avoidance strategy (toe elevation in mid swing) displayed in the young was not present (i.e. not significantly different from background) in older adults (Figure 3).

Knee flexion displacement and angular velocity responses were significantly less in amplitude in the older group during mid and late swing. Similarly, Schillings and colleagues (2005) found that older participants displayed a blunted knee flexion response leading to a reduced elevation response into mid-swing. This reduced knee flexion seen in older adults in the present study was superimposed on their already reduced knee flexion seen in undisturbed walking cycles further widening the disparity in the degree of knee flexion between the groups. This resultant decreased knee flexion would contribute to the decreased toe clearance response noted in the old relative to the young.

In the ankle, older subjects displayed reduced plantarflexion displacement and velocity responses during early and late swing, similar to that seen by Schillings and colleagues during late swing (2005). This reduced plantar flexion response in the older adults could result from increased ankle joint stiffness (Vandervoort et al., 1992). Alternatively it could result from a combination of the reduced knee flexion response to stimulation coupled with the altered undisturbed gait kinematics. That is, in undisturbed walking cycles during toe-off and swing there is reduced knee flexion (Figure 1) in the old leading to reduced distance between the foot and the ground as seen in the reduced toe clearance reflex in mid swing (bins 12 and 13) (Figure 3). These factors may influence the environment for the presentation of the ankle responses in the old in that there is there is less toe clearance available to support a plantarflexion response of full magnitude (i.e. unblunted) in the old as a unblunted response would further reduced toe clearance and increased risk of tripping. It is noted that even with a decreased plantarflexion reflex at the ankle, the reflex toe clearance in the old is closer to the ground than the young. This means the end-point foot position must be a resultant combination of kinematics at the trunk, hip, knee and ankle.

The current results may suggest a relationship between differential strength losses in ankle musculature and the functional responses to stimulation. As the strength capacity of plantarflexors degrades during normal ageing while the strength of dorsiflexors remains relatively maintained (Christ et al., 1992). Therefore an increased inhibition of TA may be required to mitigate the reduced contribution of plantarflexors to ankle stability during stance and joint stiffness during swing. In spite of increased response to stimulation in dorsiflexor motor output, the functionally relevant responses in the kinematic gait patterns are blunted in magnitude in older adults, indicating that while reflex control mechanisms may be intact, other factors endemic to the ageing process may interfere with execution of the stumble corrective response.

Peak kinematic responses both in terms of angular displacement and plantarflexion velocity occurred at bin 10, in both groups but with reduced amplitude in the old. As these responses are superimposed upon differing undisturbed patterns of motion, specifically reduced dorsiflexion velocity in the old, it may indicate a relationship between limb momentum and the extent of kinematic responses seen in individuals. Decreased momentum of the plantarflexed foot during undisturbed movement in the old may influence the magnitude of the reflex expression.

Clinical implications

Significantly smaller kinematic responses to SP nerve stimulation in older adults may be a sign of early degradation of the stumble corrective response. That is, the functional capacity to respond to perturbations during walking is reduced in older adults. This blunted response could result from a combination of the reduced knee flexion response to stimulation coupled with the altered undisturbed gait kinematics.

Overall, this knowledge adds to a basic understanding of age-related changes in neural function and has translational implications for the diagnosis and prevention of fall risk as well as rehabilitation after falls for the older population. Evaluating the neuromechanical obstacle avoidance response through cutaneous reflex analysis may allow early identification of those at future risk of falls and enable timely presentation of appropriate prevention strategies. Additionally, the restoration of the normal stumble corrective response as well as undisturbed gait properties in the old could be the goal of strength and mobility interventions aimed at improving balance and gait.

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