Neuromuscular Coordination during Slope Walking

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Neuromuscular Coordination during Slope Walking

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LIST OF ABBREVIATIONS

ANOVA	Analysis of variance
ASIS	Anterior superior iliac spine
CNS	Central nervous system
BF	Biceps femoris muscle
BSP	Body segment parameter
CF	Contralateral foot
CMC	Adjusted coefficient of multiple correlation
COG	Center of gravity
COP	Center of pressure
CPG	Central pattern generator
CV	Coefficient of variability
DSCT	Dorsal spinocerebellar tract
EMG	Electromyogram
FR	Resultant Force
GI	Gait initiation
GM	Gluteus maximus muscle
GRF	Ground reaction forces
GTO	Golgi tendon organ
HO	Heel off
HS	Heel strike
MG	Medial gastrocnemius muscle
MLR	Mesencephalic locomotor region
POI	Points of interest

RET	Reticulospinal
RF	Rectus femoris muscle
RUB	Rubrospinal
SCI	Spinal cord injury
SD	Standard deviation
SF	Strike foot
SLR	Subthalmic locomotor region
SM	Semimembranosus muscle
Sol	Soleus muscle
SRCP	Spinoreticulocerebellar path
ST	Semitendinosus muscle
TA	Tibialis anterior muscle
TO	Toe off
TS	Toe strike
VES	Vestibulospinal
VM	Vastus medialis
VSCT	Ventral spinocerebellar tract

SUMMARY

The biomechanics and muscle activity of forward and backward slope walking was investigated in humans to gain additional insight into neural control strategies. An adjustable instrumented ramped walkway was constructed and validated. Kinematic, ground reaction force, and muscle activity data were collected from nine subjects walking at three grades (0%, 15%, and 39%) for each of four conditions (forward upslope and downslope and backward upslope and downslope). The changes observed in the data were generally progressive from 0% to 15% to 39% grade. During forward downslope walking the joint moment pattern at the knee changed significantly, power absorption increased, and changes in the muscle activity patterns corresponded directly to changes in joint mechanics. During forward upslope walking, the hip joint moment pattern changed significantly, power generation increased, and changes in the muscle activity pattern were not directly related to changes in the joint moments at all joints. The muscle activity pattern data suggest that modifications to the level walking control strategies were necessary during slope walking. Backward slope walking was used to further explore these findings. Backward upslope and forward downslope kinematics and kinetics were similar, as were those from backward downslope and forward upslope walking. However, power generation increased during upslope walking tasks and power absorption increased during downslope walking tasks, and the changes in muscle firing patterns were more similar for these tasks than for those with similar kinetics. Increased power generation required compensatory muscle activity at adjacent joints that was not directly related to the moments at those joints; increased power absorption did not require

such compensatory activity, and muscle activity was directly related to the joint moments. Overall, these data suggest that changes in the control strategy and/or modifications of the level walking control strategy are strongly influenced by the power demands of a task. The characterization of forward and backward slope walking presented here is novel and has important implications for many patient populations; knowledge of the task mechanics may be used to develop or improve physical therapy and rehabilitation exercise programs as well as the design of replacement and/or assistive devices.

CHAPTER 1

INTRODUCTION

Locomotion on sloped surfaces is a challenge in our daily environment that places unique demands on the neuromuscular system. Several studies have therefore used this paradigm to investigate the neural control of locomotion in quadrupeds and have reported muscle activity patterns that are different from those observed during level walking (Gregor et al., 2001; Smith and Carlson-Kuhta, 1995; Smith et al., 1998). For example, hip and ankle flexor activity was observed during stance in downslope walking in the cat, while hip extensors were inactive; the opposite activity is typically observed during level walking (Smith et al., 1998). These findings conflict with the traditional idea of a single Central Pattern Generator (CPG) controlling all forms of locomotion, and instead suggest that sloped surfaces require specialized responses from the nervous system, which have been modeled as task-specific CPGs (Smith et al., 1998). Though the presence of a CPG in humans is open to debate (Duysens and Van de Crommert, 1998; Yang et al., 2004), results from these quadruped studies may be extended to humans: one would expect that the changes in mechanical demands during slope walking would require a change in the neural control strategy, which would result in modifications in the neuromuscular patterns observed during level walking. In this work, the term 'control strategy' refers to the output from the nervous system used to accomplish a task, which integrates input from several sources and results in the pattern of muscle activity observed during a task.

The question that then follows is *how* does the neuromuscular system adapt to changes in the locomotor environment, i.e. changes in mechanical demand? Using the

CPG theory as a starting point, we can assume that the human nervous system uses a basic control strategy to accomplish the task of level walking. Because overground locomotion is a task that humans do quite naturally and almost subconsciously, we can also assume that the control strategy is fairly robust and is modified only when markedly influenced by supraspinal and/or sensory feedback signals. For example, the sudden appearance of an obstacle may elicit a feedback signal (either visual, from seeing the object, or proprioceptive, from contact with the obstacle) that would trigger termination of the basic locomotor pattern and onset of the muscular response necessary to recover from the perturbation (Pijnappels et al., 2004, 2005). Similarly, responding to a change in terrain, such as the introduction of a grade in the walking surface, may require a response from the nervous system related to the grade intensity. Assuming that the basic muscle firing pattern is inadequate to handle such a change in terrain, one could envision two strategies the control system could use to meet this type of locomotor challenge: the control system could either change the level of muscular activity within the basic muscle firing pattern, or it could alter the timing muscle firing pattern. If the first strategy was used, the burst durations of the muscles would be the same as during level walking, but the magnitudes of the activity would be increased or decreased as the grade changed. If, instead, the second strategy was used, the burst durations and the burst magnitudes would change. As these examples demonstrate, studying the changes that occur in the muscle firing patterns during slope walking provides insight changes in the control strategy.

However, just knowing how the nervous system adapted to the task may not completely explain *why* certain changes in the muscle activity patterns were necessary. To better understand the neuromuscular system's response to the task, knowledge of the changes in mechanical demands during slope walking, compared to level walking, would be needed. The advantage of this more comprehensive approach (studying *both* the muscle activity and the gait mechanics) can be illustrated using results from quadruped model as an example. Although the muscle activity observed in quadrupeds during slope walking led to the proposal of specific control models, these models were not sufficient to completely understand the task. For example, Smith et al. (1998) found that select single joint hip extensors were inactive during stance in downslope walking, and predicted there would be "no extensor torque at the hip during the stance phase of downslope walking." However, a distinct hip extensor moment is reported for the first 25% of stance during downslope walking at 50% grade (Gregor et al., 2005). A more careful review of Smith's data reveals that the biarticular hip extensors show some activity in early stance, which would then contribute to the extensor moment. In the same paper, the authors observed a decreasing duration of ankle extensor activity and concluded that the duration of the ankle extensor moment must likewise decrease (Smith et al., 1998). In contrast, Gregor et al. (2001, 2005) show an ankle plantarflexor moment for the entire stance phase during downslope walking. Smith and colleagues did not measure activity in the medial gastrocnemius or soleus muscles, which are also plantarflexors and may be contributing to the plantarflexor moment for the remainder of stance. These examples reflect the limitations of predicting joint torques from muscle activity patterns, especially considering that the set of muscles tested is necessarily incomplete and that there are other factors that contribute to the joint moment, such as the inertial properties of the limb segments. The reverse danger also exists. For example, it is difficult to predict muscle activity patterns from joint moments because of the large number of muscles at each joint, the many muscles that span multiple joints, the variable moment arms during movement, and differences in force production during shortening and lengthening contractions. One reason this combination of data is so powerful is that the data are inherently linked: muscle activity controls limb motion (kinematics), which causes the limb to exert forces on the ground; the reaction forces cause moments about the joints (kinetics), which must be counteracted by muscle activity. The power that is generated or produced at the joint is related to the direction of length changes in the muscles acting at the specified joints. The integration of the environmental mechanical demands, limb dynamics, and EMG data can provide substantial insights into the control system; therein lies the strength of the approach used in this study.

Changes in the magnitude and/or duration of muscle activity patterns could be caused by changes in supraspinal signals and/or feedback to the control system, or could indicate a change in the basic control strategy, equivalent to the proposed switch to task-specific CPGs in quadrupeds. We acknowledge that distinguishing between these possibilities in humans would be very difficult. Fully characterizing the motor output and movement biomechanics, however, is a crucial first step in revealing important information about the control system. Synchronous analysis of the two data sets will provide a more complete representation of motor output and may help explain what mechanical factors contribute to changes in the muscle activation patterns. For example, the data obtained during slope walking will indicate whether or not changes in the joint kinematics and joint kinetics (moments and powers) have the same influence on the muscle activity patterns. Once the relationship between the motor output and muscle activity is known, the question of what control mechanisms are used during the task can be addressed.

To date, slope walking paradigms have not been used to investigate neural control strategies in humans. There is general consensus, however, that alterations in the pattern and magnitude of joint kinematics, kinetics, and electromyographic (EMG) activity between tasks, such as walking on different slopes, may provide some insight into control strategies used by the nervous system (Winter and Eng, 1995). In fact, it has been said that "quantifying limb dynamics … provides a powerful tool in our armamentarium for revealing the mechanisms of neuromotor control" (Zernicke and Smith, 1996). While there is a dearth of information on slope walking, other challenging tasks have been used for the purpose of evaluating neural control strategies in humans. For example, Brunt and colleagues (1999) observed similarities in peak ground reaction forces (GRF) and EMG latencies and durations during gait initiation and initiating movement by stepping over an obstacle, leading them to conclude that a similar motor program may govern both tasks. A similar conclusion was drawn in another investigation on a variety of forward-oriented movements, including gait initiation (Crenna and Frigo, 1991). In contrast, differences in the joint kinetic and EMG patterns for time-reversed backward walking and forward walking led to the conclusion that the motor program for forward walking must be modified to control backward walking, in spite of the strong similarities in joint kinematic profiles (Grasso et al., 1998; Thorstensson, 1986; Winter et al., 1989). Riener et al. (2002) presented GRF, kinematics, and kinetics during stair walking at 24˚, 30˚ and 42˚ inclines (45%, 58%, and 90%, respectively) that suggested there is also a specific motor program for stair climbing. Similarly, Earhart and Bastian (2000) observed that subjects stepping on a wedge utilized different motor programs, described using interjoint coordination and EMG patterns, for steeper 20˚ (36%) and 30˚ (58%) grades than for 0˚ and 10˚ (18%) grades. These authors also identified a transition grade at 15˚ (27%) where either program could be used. The findings from these tasks suggest that slope walking could also require movement control strategies different from those for level walking that are not yet well understood.

Data on gain mechanics during slope walking have been presented only sporadically in the literature (Kuster et al., 1995; Lange et al., 1996; Leroux et al., 1999; Redfern and DiPasquale, 1997; Tokuhiro et al., 1985). The different walking grades investigated and experimental techniques used, and especially the lack of kinetic data during upslope walking and of EMG data during downslope walking, prevent us from using these data to draw conclusions about neural control. **Therefore, the overall goal of this project was to gain insight into human neural control strategies by using up- and downslope walking to challenge the neuromuscular system and by synthesizing an analysis of lower limb biomechanics and muscle activity patterns during these tasks.** The underlying hypothesis was that during these tasks the control strategy, and therefore the motor output, is driven by changes in mechanical demands, such as joint moments and joint power. In other words, it was expected that changes in joint kinetics and the relative amounts of power absorption and generation during slope walking would present a perturbation to the control system, requiring changes in the muscle activation patterns used for level walking in order to successfully achieve upslope or downslope walking.

The task of slope walking was chosen as a perturbation to the neuromuscular system because of the reports on slope walking in the quadruped model and because it is related to other tasks that have been used to study neural control in humans, such as wedge-stepping (Earhart and Bastian, 2000) and the transition step to an inclined surface (Prentice et al., 2004). In addition, it is a task where the mechanical demands can be scaled by changing the incline of the walking surface, an advantage that has been reported in stair-walking experiments (Riener et al., 2002). By changing the mechanical demands, different sensory feedback mechanisms can be evaluated. For example, during upslope walking the body must be lifted up against the force of gravity; this presumably requires increased muscle force and power generation, which is commonly related to concentric muscle activity in the primary extensor muscles. This idea is supported by Minetti et al. (2002), whose data suggests that during upslope walking at steep grades muscle activity is entirely concentric (Minetti et al., 2002). This increased power generation is most likely related to increased force-dependent sensory feedback. By contrast, during downslope walking power is absorbed to resist collapse as the body is pulled down the slope, which may or may not affect length-dependent sensory feedback from the muscle spindles (Gregor et al., 2005). The importance of length-feedback during downslope walking has been demonstrated in cats, where a disruption of the stretch reflex in the triceps surae resulted in kinematic deficits at the knee and ankle (Abelew et al., 2000). Finally, additional and more novel perturbations could be introduced by having participants walk backwards up and down the different slopes. Although backward walking is a task that is practiced much less often than forward walking, it has been suggested that the control system for these tasks is the same in a quadruped model (Buford and Smith, 1990). The evidence that forward and backward walking in humans are driven by the same movement control strategy is not as conclusive. A study using human infants suggested that a single CPG could account for multiple directions of walking (Lamb and Yang, 2000), and studies with adults have come to the same conclusion based on similarities in muscle activity patterns during forward and backward walking (Van Deursen et al., 1998; Winter et al., 1989). However, other studies on human adults reported variability in the muscle firing patterns and did not necessarily support this idea (Grasso et al., 1998; Thorstensson, 1986). The interest in backward walking as a task that can elucidate neural control strategies suggested that backward slope walking might also be a useful task for this purpose. Given these characteristics of forward and backward slope walking, the goal of this project was achieved through the following aims.

Aims

The **first aim** of this project was to study the task of downslope walking; to describe the task we would first quantify the biomechanics, and then determine the relationship between the kinetics and muscle activity. From the limited data reported in the literature, it was expected that both the temporal pattern and peak magnitude of the joint moments would change during downslope walking (Redfern and DiPasquale, 1997). Because the joint moments arise from the muscle activity (as is discussed in Chapter 2), it was expected that the pattern and magnitude of muscle activity would also change. Specifically, it was *hypothesized that changes in the pattern and magnitude of the muscle activity would be directly related to the changes in the pattern and magnitude of the joint* *moments during downslope walking*. This would be tested using correlations of the moment and muscle activity duration and magnitude. Based on the joint angles and moments presented by Redfern et al. (1997), it was also expected that power absorption would increase at all joints during downslope walking (Redfern and DiPasquale, 1997), and therefore that the relative amount of eccentric muscle activity would increase. This finding would suggest that length-dependent sensory feedback from the muscle spindles would increase. Because Abelew et al. (2000) showed that such length feedback was necessary for quadrupeds to achieve normal kinematics during downslope walking, but not in level or upslope walking, it was thought that this increased feedback might play a role in the control strategy for human downslope walking. Although the increased length feedback and its role in changing the control strategy could not be implicitly tested in this project its potential impact was considered.

The **second aim** of this project was to study the task of upslope walking, using the same methods employed in the first aim. There were no prior data regarding the kinetics of upslope walking, but there was some information reported in the literature about upslope walking muscle activity (Leroux et al., 1999; Tokuhiro et al., 1985). The majority of the changes in the muscle activity during upslope walking occurred in the burst magnitudes, rather than in the burst durations. Using the muscle activity data as a first approximation for the joint moments (recognizing the limitations discussed earlier), it was expected that the changes in the joint moments would be predominately changes in magnitude rather than in pattern. Specifically, it was *hypothesized that changes in the magnitude of the muscle activity would be directly related to the changes in the magnitude of the joint moments, but that the patterns of both the joint moments and* *muscle activity would remain unchanged*. Again, this relationship would be tested using correlations of the moment and muscle activity magnitudes and durations. As already discussed, it was expected that power generation would increase during upslope walking (Minetti et al., 2002), and therefore that the relative amount of concentric muscle activity would increase. This finding would suggest that force-dependent feedback from the golgi tendon organs (GTO) would increase. Again, the potential impact of this increased feedback on the control strategy was considered although it could not be implicitly tested. Also embedded in these two aims was how the intensity of the joint moments and muscle activity would be affected by changes in the slope: would the change in surface grade invoke a gradual change, or would there be an all-or-nothing effect with a transition grade, as seen in the wedge-stepping?

The **third aim** of the project was to use the task of backward slope walking to further test the movement control strategies employed by the nervous system for forward slope walking. In order to present this information most clearly, Aim 3 was divided into three parts: i) to define the relationship between the forward and backward slope walking tasks, ii) to analyze the task of backward upslope walking, and iii) to analyze the task of backward downslope walking. Part 1 of this aim was to establish the relationship between forward and backward slope walking. Because the biomechanics of backward slope walking have not been analyzed it was unclear how they would compare to the biomechanics of forward slope walking. Based on comparisons between forward and backward level walking, *it was hypothesized that the joint kinematics and joint moments would be similar and the joint power would be reversed between backward upslope walking and forward downslope walking, and between backward downslope walking and* *forward upslope walking* (Grasso et al., 1998; Thorstensson, 1986; Winter et al., 1989). After a relationship between forward and backward slope walking was established, an analysis of the biomechanics and muscle activity during backward slope walking would be completed. This analysis relied on the same techniques used in the analysis of forward slope walking. Part 2 of this aim studied backward upslope walking, which was intended to parallel the investigation of forward downslope walking from Aim 1 (Chapter 5). Part 3 of this aim investigated backward downslope walking, as a complement to the study of forward downslope walking in Aim 2 (Chapter 6). It was expected these investigations of backward upslope and downslope walking would provide further insight into the robustness of the movement control strategies for slope walking. One definition for the 'robustness' of a movement control strategy is its ability to govern multiple tasks, or perturbations of its primary task. The authors that studied backward walking were attempting to show that the forward walking movement control strategy was robust enough to also control backward walking (Grasso et al., 1998; Thorstensson, 1986; Winter et al., 1989), a concept that has been suggested to be true in infants (Lamb and Yang, 2000). However, their data show differences in the muscle firing patterns for forward and backward level walking, indicating it is unlikely that the same control strategy is valid for both walking directions. Assuming that there are different control strategies for forward and backward walking, it is then possible to test the 'robustness' of *each* control strategy by using the slope perturbation. In other words, if the grade is changed to +15% during forward and backward walking (in the same direction), requiring forward upslope walking and backward downslope walking, do the same changes occur in the initial control strategy? Is that also true when the grade is changed to -15%, requiring forward downslope walking and backward upslope walking? Because the changes in the joint kinetics were expected to be similar between these corresponding tasks, the changes in the control strategies with slope were also expected to be the same. As stated in Part 1 of this Aim, it was expected that the power requirements (and therefore the feedback signals) would be reversed between these corresponding tasks. For example, during forward downslope walking power absorption and length feedback would be emphasized; during the kinematically and kinetically similar task of backward upslope walking, power generation and force feedback were expected to be emphasized. It was expected that the movement control strategies would be robust in spite of the expected changes in power requirements, such that the relationship between muscle firing patterns and joint kinetics would be similar for locomotor tasks with similar patterns of joint kinetics. This line of reasoning and the hypotheses from forward downslope and forward upslope walking led to the following expectations for backward slope walking: For backward upslope walking, it *was hypothesized that changes in the pattern and magnitude of the muscle activity would be directly related to the changes in the pattern and magnitude of the joint moments.* For backward downslope walking it was *hypothesized that the changes in the magnitude of the muscle activity would be directly related to the changes in the magnitude of the joint moments*.

These three aims were used to address the general goal of probing the neural control strategies used during challenging locomotor tasks. Although it is difficult to eliminate any particular form of feedback, the chosen tasks emphasize one type, therefore providing different conditions for studying control mechanisms. In addition, the third aim uses tasks with kinematics and kinetics similar to the first two aims, but reverses the type of feedback that is emphasized, allowing further probing of the robustness of the control strategies used in forward slope walking.

Significance

The significance of this project, first, is in the exploration of the unique relationship between mechanics and muscle activity during a challenging task. Biomechanics and EMG are studied in isolation far too often, which detracts from a true understanding of the function of the nervous system. Integrating these data has the potential to provide insight into how neural control strategies change in response to specific changes in environmental demands. A better understanding of these new control strategies in normal subjects has implications many different patient populations. For example, lower limb amputees fitted with prosthetic feet have a difficult time ascending and descending ramps (Gailey and Clark, 2004). Knowledge of the biomechanics and muscle activity in normal intact subjects can be used in the design of lower-limb prostheses and in the programming of computerized prosthetic components. Such understanding and knowledge can also be used in designing rehabilitation regimes for spinal cord injury (SCI) patients and those with neurological disorders that affect movement, such as Parkinson's disease.

This project is also significant because of the gain of fundamental knowledge about the biomechanics and neural control of forward and backward slope walking. As mentioned, forward slope walking is a task where neither the mechanical or muscular outputs have been completely described in the literature for healthy humans. Before an assessment of any patient population can be undertaken, these baseline data must be available. Understanding how the nervous system meets the demands of slope walking in normal subjects can provide a basis for better understanding the deficits caused by other afflictions, such as stroke, cerebral palsy, and neuropathy. The data presented here, therefore, are critical for any group desiring to use slope walking for clinical treatments or evaluations. Backward slope walking is a task that has been given negligible attention in the biomechanics literature, in spite of its increasing popularity as a rehabilitation exercise (Cipriani et al., 1995). The data presented here are novel and provide a solid foundation for understanding the demands of backward slope walking, which is critical for clinicians, especially as backward slope walking continues to be prescribed as a rehabilitation exercise for patients suffering from knee pain or recovering from knee surgery.

Limitations

The most significant limitation of this project is the extent to which we can draw any definite conclusions about the structure of the locomotor control system. By studying the joint mechanics and muscle activity we can understand the summed output of the nervous system, but we learn nothing about the complex neural circuits producing the commands. Is there a CPG in humans? Is the CPG output overridden by supraspinal commands during slope walking? This research cannot answer such questions, although it can lead to a better understanding of what environmental factors and feedback sources influence changes in the output from the human locomotor control system.

In addition, there are limitations inherent in the biomechanics methods used in this project. For example, the joint moment data gives the net moment due to the activity of all the muscles acting at the joint (see Chapter 2 for more details). EMG signals indicate which muscles are active, and therefore are contributing to the joint moment. However, the relationship between the muscle activation and the amount of force produced by the muscle is not well defined and changes depending on the conditions of the contraction, so the exact contribution of each muscle to the joint moment is unknown. In addition, it is impossible to monitor all the muscles of the lower limb using EMG, so there could be additional muscles contributing to the moment, leading to an overestimation of the contributions of the monitored muscles. There is also an inherent delay between the muscle activation as measured by EMG electrodes and the actual force production (typically estimated at ~40ms), which leads to a small error when comparing the timing of EMG bursts to the joint moment patterns. Another limitation lies in the net joint power, which is determined from the kinematics and joint moments and is often used to infer whether the major muscles at the joint are acting eccentrically or concentrically. For example, if there is an extensor moment and the joint is flexing, then power is absorbed and it is assumed that the extensor muscles are acting eccentrically. However, it could be that the muscle is passively lengthening, or that the muscle-tendon complex is lengthening overall although some muscle fibers are actively shortening. In addition, the joint power, because it is a *net* quantity, does not give information about what the other muscle groups are doing. In the previous example, the conditions of the flexor muscle contractions are unknown. Because it is difficult to calculate muscle lengths in humans during locomotor tasks some of these questions cannot be resolved. These limitations are inherent in any biomechanical analysis of human gait, however, and are considered minor in comparison with the useful information which can be gained from such an analysis.

CHAPTER 2

BACKGROUND

Historical Perspective

More than a century ago there was a surge of interest in determining the nature and locus of the control mechanism for locomotion, and extensive animal studies were performed to this end. As a result, in 1911 T. Graham Brown was able to write that “the act of progression is one richly co-ordinated” (Brown, 1911). His report outlined some work of his contemporaries, such as Sherrington, Freusberg, and Philippson, and discussed the peripheral stimuli that contributed to controlling locomotion via the “neural centres.” More importantly, however, he wrote that it had “long been known” that spinal animals also exhibited progression movements, which implied that some mechanism in the spinal cord was sufficient to cause the “act of progression.” Brown’s own experiments with cats further supported this idea by showing that rhythmic muscle activity occurred even in spinalized, deafferented preparations (Brown, 1911, 1914) (for a detailed review see (Duysens and Van de Crommert, 1998)). These findings led Brown to propose the existence of a “central mechanism . . . consisting of antagonistic centers, augmented” by the “proprioceptive mechanism” to adapt the locomotor pattern to the environment (Brown, 1911).

Eventually this “central mechanism” evolved into the concept of a central pattern generator (CPG) located in the spinal cord that generates basic synergies for locomotion, which are integrated with sensory feedback and supra-spinal signals to achieve precise amplitude and timing control of movement (Figure 1) (Grillner, 1981). Grillner (1981)

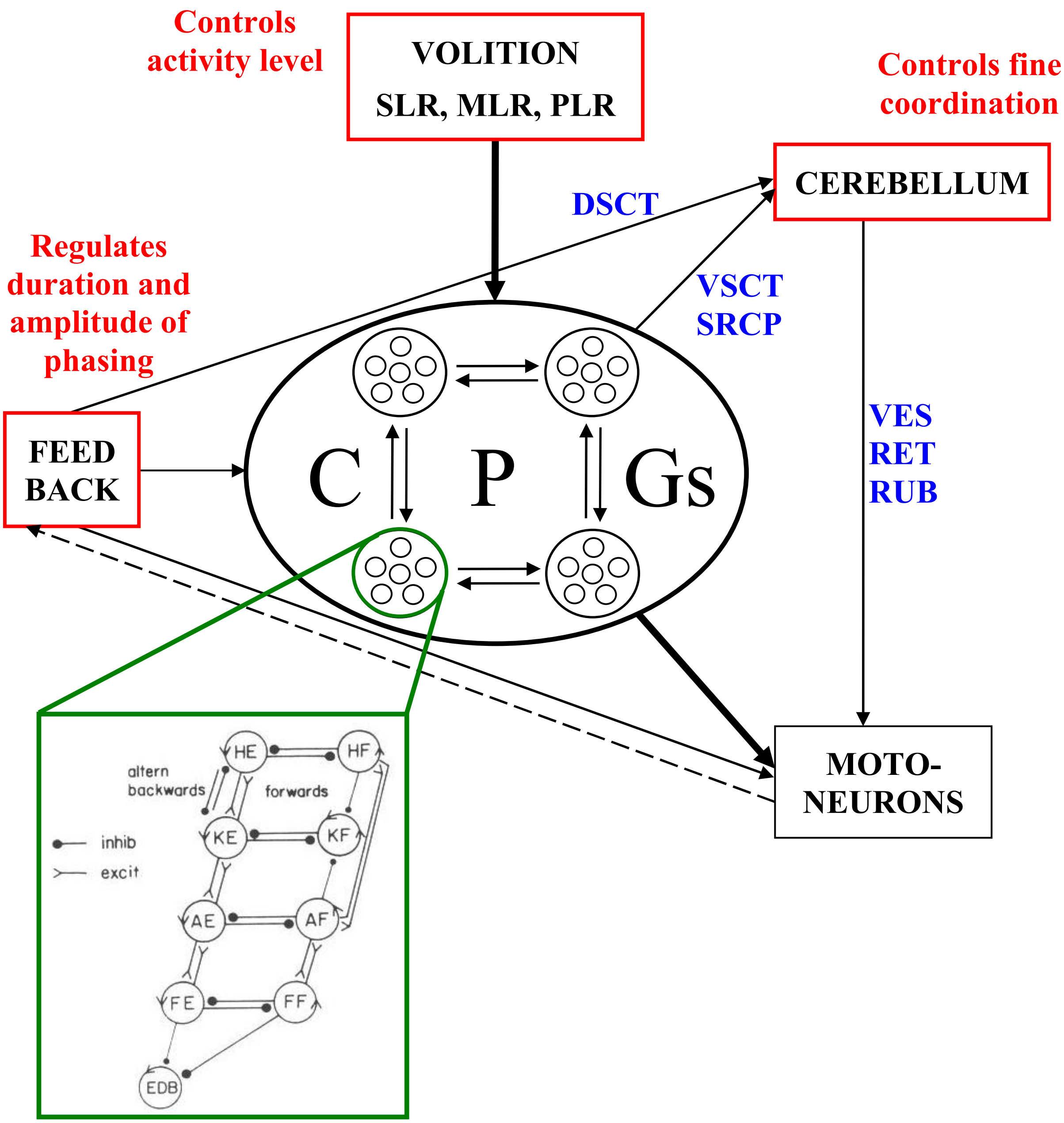

Figure 1. Proposed Locomotor Control System. Central pattern generators (CPGs) act on motoneurons to cause movement. CPG activity levels are modulated by volitional control from higher centers in the brain (subthalmic, mesencephalic and pontine locomotor regions (SLR, MLR, and PLR, respectively)), while peripheral feedback regulates the phasing of the signals. The cerebellum controls the fine coordination of movement via the vestibulospinal (VES), reticulospinal (RET), and rubrospinal (RUB) systems. Peripheral feedback (via the dorsal spinocerebellar tract (DSCT)) and CPG output (via the ventral spinocerebellar tract (VSCT) and spinoreticulocerebellar path (SRCP)) also affect the cerebellum. The expansion shows a possible limb CPG, consisting of several unit burst generators with excitatory and inhibitory connections. (E – extensor, F – flexor, H – hip, K – knee, A – ankle, F – foot, EDB – extensor digitorum brevis (toe flexor)). Adapted from Grillner (1981), Figures 31 and 39.

also proposed that the same spinal circuitry could be used to control different forms of locomotion, such as backward walking. This theory has received substantial attention in the quadruped (Buford and Smith, 1990, 1993; Buford et al., 1990; Perell et al., 1993; Pratt et al., 1996; Trank and Smith, 1996), as well as other animal species (Ashley-Ross and Lauder, 1997; Eilam and Shefer, 1992), and in humans (Grasso et al., 1998; Thorstensson, 1986; Ting et al., 1999; Van Deursen et al., 1998; Vilensky et al., 1987; Winter et al., 1989). Although common features of muscle coordination, kinematics, and kinetics have been observed during forward and backward walking in the quadruped, there is no consensus on whether the movements are controlled by different CPGs or the same CPG with the output modified by proprioceptive and supraspinal input (Perell et al., 1993). There is agreement that the motor program produced by the CPG, although sufficient to drive locomotion in a controlled environment, is dependent on other inputs to create a more robust locomotion pattern (Brown, 1911; Buford and Smith, 1990; Grillner, 1981).

CPGs in Humans

Although the idea of a CPG was originally based on animal studies, a similar control system is thought to exist in humans (Duysens and Van de Crommert, 1998). As evidence, in human subjects with incomplete spinal cord transections (but not in patients with complete transections) rhythmic muscle activity has been observed (Duysens and Van de Crommert, 1998; Leroux et al., 1999). With treadmill training spinal cord injury patients have been able to recover some locomotor activity, providing additional evidence for the presence of CPGs (Van de Crommert et al., 1998). However, the definitive existence of spinal cord CPGs in humans, and the extent of their interaction with afferent and supraspinal signals is still uncertain (Duysens and Van de Crommert, 1998). One problem is that researchers must rely on the SCI patient population to gain insight into this mechanism; the wide range of injury types and degrees, and the ability of the nervous system to reorganize after an injury complicate the search for proof of spinal CPGs (Van de Crommert et al., 1998).

Biomechanics and Neural Control

Setting aside the lingering question about the existence of human CPGs, many researchers have focused on using biomechanics to study the output of the intact control system during locomotor tasks. Although the link is not immediately apparent, the field of biomechanics is closely related to the study of neural control. The neuromuscular system uses metabolic energy to produce muscle tension that causes or resists movement by generating or absorbing mechanical energy. Several levels of signal summation occur to effect the desired movement and achieve the goal of the central nervous system (CNS) (Figure 2). First, multiple neurons converge on an alpha motoneuron, and the sum of their excitatory and inhibitory signals controls a single motor unit in a muscle. At the whole muscle level, forces from multiple motor units sum to cause a force at the associated joint(s). Multiple muscles act at each joint, and their interactions create a net joint moment. The net moments at all the joints work together to cause whole body motion. Therefore, studying the final movement provides a unique insight into the overall goal of the CNS (Winter, 1990). Winter (1995) argues that a complete

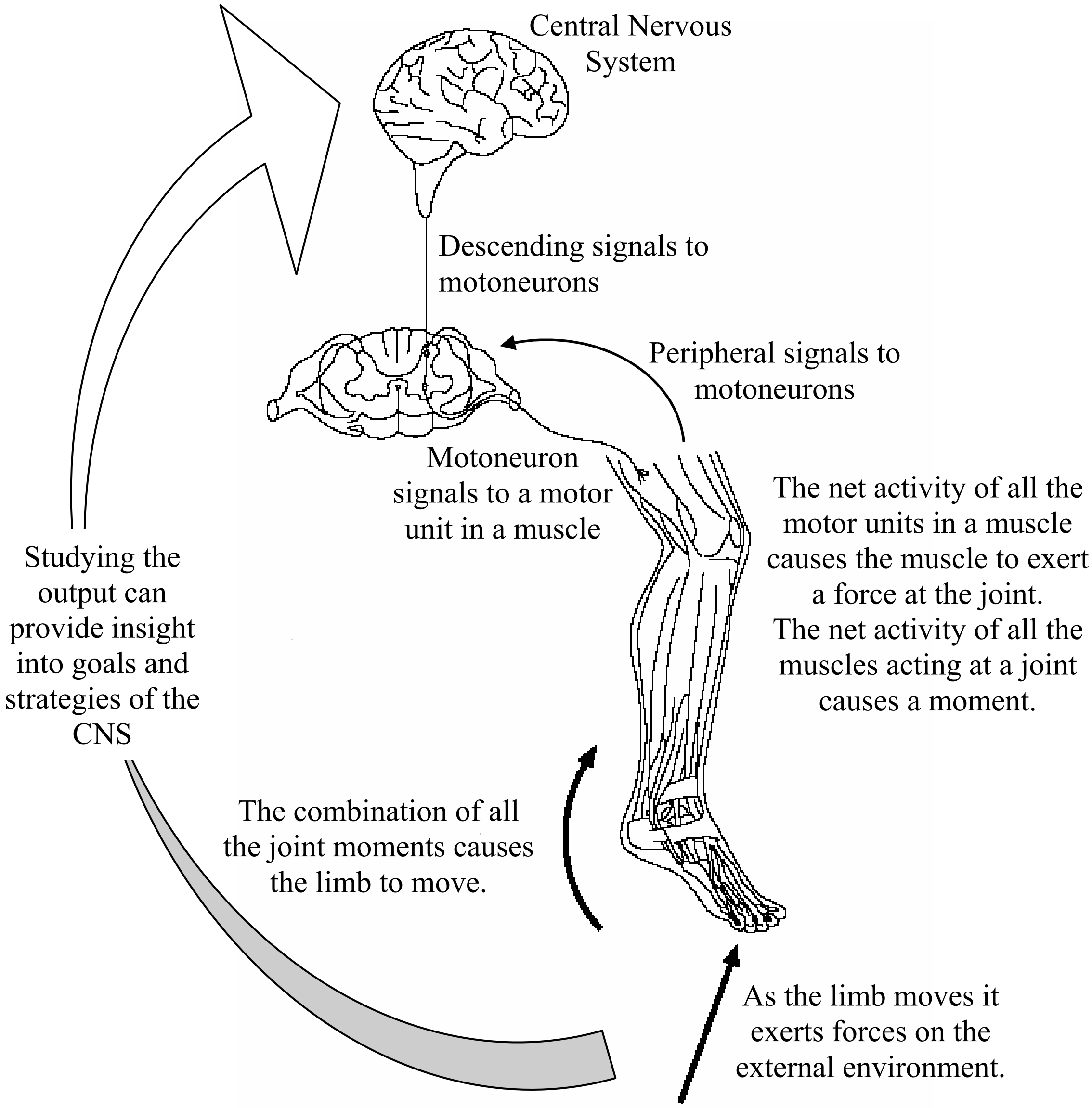

Figure 2. Neuro-musculo-skeletal Integration. Signal summation in the neruro-muscular system occurs at the motoneuron, muscle, joint, and limb during movement. Using biomechanics to study the final movement can reveal information about neural control. Adapted from Figure 1.2 of Vaughan (1999), also see Figure 1 of Winter (1995).

biomechanical analysis also reveals the mechanisms by which the CNS chooses to achieve the overall goal.

One aspect of a biomechanical analysis is kinematics, the study of motion independent of the forces causing or resulting from it. In the laboratory, markers placed on important anatomical landmarks are tracked with cameras to provide information about limb motion. Figure 3 shows the marker system used for the experiments in this project. The 3D coordinates of the markers are used to calculate the positions of the joint centers, which are then used to calculate the position, velocity, and acceleration of the segment centers of gravity (COG) (Vaughan et al., 1999). The orientation of the segments with respect to one another is analyzed to determine joint angles (Figure 4), as well as the angular velocity and acceleration of each segment. Anthropometric measurements (Figure 5) are input into regression equations (based on cadaver data) to yield body segment parameters (BSP) such as segment masses and moments of inertia. The forces that cause or result from the motion studied in kinematics are termed kinetics, and involve both internal and external forces. The external forces that occur during locomotion can be measured using a force platform. These forces can then be integrated with the kinematic data in a process called 'Inverse Dynamics' to calculate the internal forces at each joint (Figure 6). The inverse dynamics process assumes that at a finite time point the limb is not moving and can be divided into rigid segments (foot, shank and thigh). Starting with the isolated foot segment, the known variables are the external forces (measured), the segment mass (calculated), and the COG linear and angular velocity and acceleration (calculated). Using Newton's laws the two unknown variables, the force and moment at the ankle joint center, can therefore be calculated. Because the

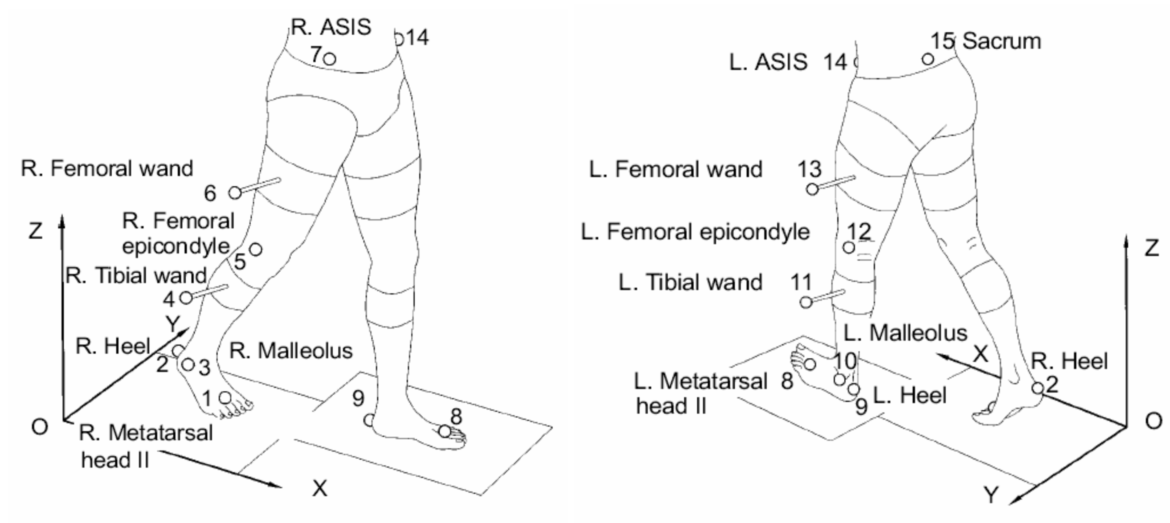

Figure 3. Helen Hayes Kinematic Marker System. Fifteen retroreflective markers are placed on anatomical landmarks to allow limb position data to be collected by cameras with infrared lights during an experiment. Adapted from Figure 3.4 of Vaughan (1999).

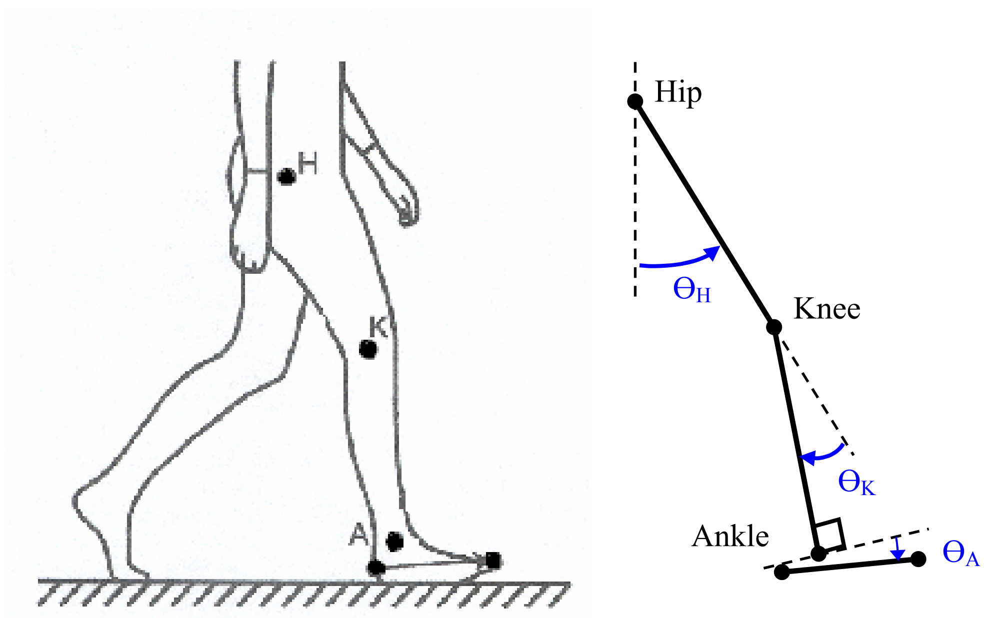

Figure 4. Joint Angles Convention, shown schematically. The dashed line at the hip represents the vertical, at the knee represents the continuation of the long axis of the thigh, and at the ankle represents a perpendicular to the long axis of the shank. Positive angles are indicated by the directions of the arrows, which correspond to flexion for the hip and knee, and plantarflexion for the ankle.

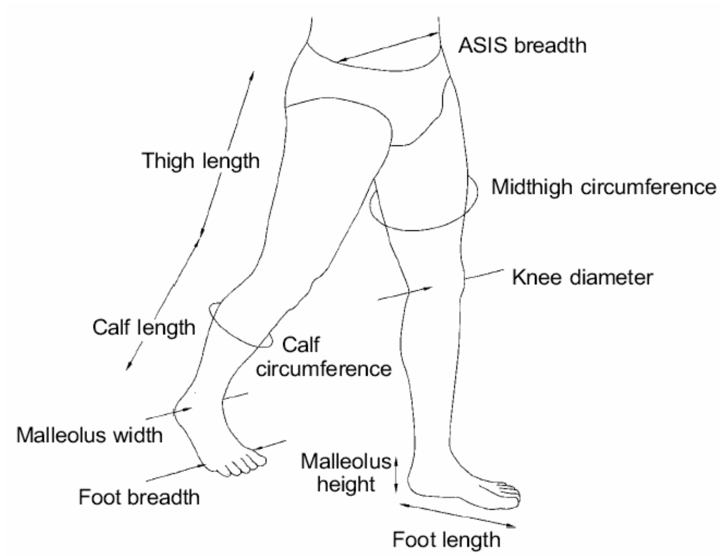

Figure 5. Anthropometric Measurements. These data are used to calculate body segment parameters (segment masses and moments of inertia), as well as joint center positions. The regression equations that are used for the calculations are based on average data obtained from cadavers. Figure from Vaughan (1999).

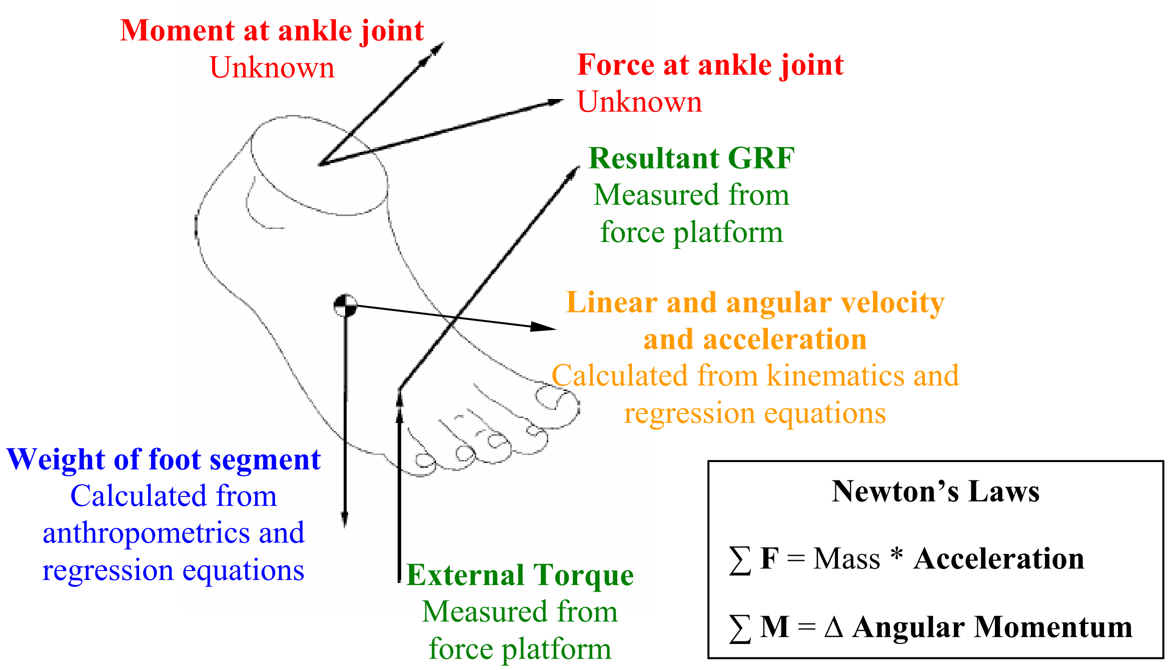

Figure 6. The Inverse Dynamics Process. Forces, torques, velocities and accelerations for each segment (foot shown here) are either measured or calculated, as indicated. Newton's laws are then used to calculate the unknown force and moment at the joint. This process is repeated for the shank and thigh segments to calculate the forces and moments acting at the knee and hip. Adapted from Figure 3.20, Vaughan (1999).

ankle joint center in the isolated foot and shank segments is the same, a force and moment equal (but opposite) to the newly calculated force and moment must be acting on the ankle joint center in the shank segment. With the force and moment at the ankle joint known, the number of unknowns for the shank segment is reduced to two (the force and moment acting at the knee joint center). Again, these variables can easily be calculated using Newton's laws. Following the same logic, the force and moment acting at the hip joint center are calculated. These three sets of calculations are performed for each time point during the trial to reveal a pattern of forces and moments at each joint.

Once the joint angles and moments are known the joint power can be calculated by multiplying the angular velocity (rate of change of the joint angle) and the joint moment. Positive power at a joint occurs when the joint moment and angular velocity are in the same 'direction.' For example, if there is a knee extensor moment and the knee is extending, power is generated. To be more specific, if is an extensor moment the extensor muscles must be active, and since the joint is extending, the extensor muscles must be producing power by actively shortening. Negative power occurs when the joint moves in the opposite direction to the joint moment generated by the active muscles, which requires the active lengthening of some muscles at the joint. Therefore negative power, or power absorption, is commonly associated with eccentric muscle activity (Winter, 1990). Unfortunately, joint power cannot completely describe the conditions of muscle contraction. For example, negative joint power can occur due to passive lengthening of the muscle-tendon unit rather than active lengthening of the muscle. In addition, even overall 'active' lengthening of the muscle-tendon complex can be achieved via a combination of localized shortening and lengthening contractions. Finally, even if the extensor muscles are actively shortening when joint power is positive, as in the example above, the activity of the flexor muscles is unknown; they could be lengthening actively or passively, or they could be in an isometric condition due to changes in kinematics. In spite of these limitations, however, joint power can at least give *some* indication of the conditions of muscular contraction at the joint.

One can think of the calculated moments as being caused by the external forces. However, at each time point the limb is not moving, so there must be some equal and opposite internal moment resisting the applied moment (and force resisting the applied force). In the body, the structures available to cause this resistance are the ligaments and tendons at each joint. Although ligaments can provide some resistance, it would not be enough to counter the large moments that occur when the foot is supporting the body weight. Muscle contractions (transmitted via tendons) are primarily responsible for the internal joint moments that occur during locomotion. Therefore, kinetics can also be considered to include the study of muscle mechanical power and energy changes due to muscle power (Winter, 1990). The internal joint moment, which is equal and opposite to the joint moment calculated by inverse dynamics, represents the net activity of all the muscles acting at a joint. The obvious question, then, is "how is this net activity distributed among the muscles?" For example, if the inverse dynamics solution was a 20 Nm flexor moment acting at the knee at a particular time point, then the muscles acting at the knee joint must be producing a 20 Nm internal extensor moment. But, there are several ways that moment could be produced: the vastus medialis (VM) could be producing the entire moment; or the VM and rectus femoris (RF) could each be producing a fraction of the moment; or the VM and RF could together be producing a 30

Nm extensor moment and the hamstrings could be producing a 10 Nm flexor moment. This problem is termed "indeterminacy" and is due to the large number of muscles acting at each joint. On the other hand, this redundancy gives the neuromuscular system the flexibility to adapt to compensate for an injury or deficiency.

One way to gain some insight into what muscles may be contributing to an internal joint moment is to use electromyography. Electrodes are placed either in (fine-wire) or on the skin over (surface) important muscles to record the electrical signal that occurs when the muscle contracts. This electrical signal is called an electromyogram (EMG), and it indicates when a muscle is contracting and also the magnitude of its activation. It is important to note that EMG data gives the activation level of the muscle, NOT the amount of force it is producing; these two quantities are not necessarily directly related. In other words, EMG activity signifies that a muscle is contributing to the joint moment, but the extent of its contribution is unknown. Nevertheless, EMG data are important in a complete biomechanical analysis of motion because they give insight into the ultimate control signal for each muscle (Winter, 1990).

The collection of kinematic, kinetic and electromyographic data in the laboratory allows for a very detailed analysis of human gait that is not possible when gait is observed in a natural environment. Such analyses can indicate adaptations that compensate for injuries, pathologies, or age (McGibbon, 2003; Winter and Eng, 1995), and can be used for research purposes or for treatment purposes. Clinical gait assessment can be used to design a rehabilitation regime, plan a surgical intervention, or evaluate the effectiveness of a treatment (pharmaceutical, surgical, or rehabilitation) (Kleissen et al., 1998; Whittle, 2001). In this study gait analysis was used to increase our understanding of the control strategies used by the nervous system during locomotion.

Perturbations and Control Strategies

As mentioned, biomechanical analyses can be used to illuminate adaptations in the nervous system to accommodate for injuries, disease, or deterioration due to aging. An extension of this idea is that challenging locomotor tasks or environments may require similar adaptations (changes in the pattern and magnitude of joint kinematics, kinetics, and EMG activity) and may therefore provide some insight into different control strategies used by the nervous system (Winter and Eng, 1995). For example, backward walking is a task that has been used to study neural control in humans (Grasso et al., 1998; Thorstensson, 1986; Van Deursen et al., 1998; Vilensky et al., 1987; Winter et al., 1989) and in a variety of animal species, such as cats, salamanders, and mole rats (Ashley-Ross and Lauder, 1997; Buford and Smith, 1990; Buford et al., 1990; Eilam and Shefer, 1992). Much of this research has been motivated by the theory that backward walking could be controlled by the same spinal circuitry that controls forward walking if the activity of the hip flexors and extensors was reversed, causing a mixed synergy (i.e. if there was an inhibitory connection between the knee and hip extensors rather than an excitatory connection in Figure 1) (Grillner, 1981; Grillner, 1985). Kinematic data from quadrupeds suggested that a common neural mechanism was used in both forward and backward walking (Buford et al., 1990). However, electromyographic (EMG) data did not support the proposed mixed synergy mechanism (Buford and Smith, 1990). Similarly, investigations of salamander locomotion discredited the idea of a common neural mechanism for forward and backward walking in tetrapods (Ashley-Ross and Lauder, 1997). Forward and time-reversed backward walking are kinematically similar tasks in humans, however the muscle firing patterns used to accomplish these tasks are different (Grasso et al., 1998; Thorstensson, 1986; Van Deursen et al., 1998; Winter et al., 1989). Although all of the backward walking studies in humans report similar data, there is no consensus among the authors on whether or not the same control strategy is used during these forms of locomotion.

Other locomotor tasks also seem to require motor programs that are different from the level walking pattern. For example, during stair walking at 24˚, 30˚, and 42˚ inclines (45%, 58%, and 90%, respectively) the ground reaction forces, joint kinematics, and joint kinetics differed considerably from those observed during level walking, suggesting there may be a motor program specific to stair climbing (Riener et al., 2002). Similarly, subjects taking a single step on a short incline segment (a wedge) utilized different motor programs for steeper 20˚ (36%) and 30˚ (58%) grades than for level and a 10˚ (18%) grade (Earhart and Bastian, 2000). The two motor programs were identified based on interjoint coordination patterns, as revealed by angle-angle plots, and muscle activation patterns. In contrast, other tasks do not seem to require a unique motor program. Brunt and colleagues (1999) studied two different forms of movement initiation, normal gait initiation and stepping over an obstacle. The observed similarities in peak ground reaction force values, muscle latencies, and muscle activity durations led the authors to conclude that a similar set of commands (a motor program) could be governing muscle activation patterns in both tasks (Brunt et al., 1999).

Negotiating sloped surfaces is a challenge in our daily environment that places unique demands on the neuromuscular system, and therefore provides a novel method for investigating the neural control of locomotion (Gregor et al., 2001; Leroux et al., 1999; Smith and Carlson-Kuhta, 1995). The findings from the previously discussed locomotor tasks suggest that slope walking could also require movement control strategies different from those for level walking that are not yet well understood. In spite of the potential insights that could be gained from studying slope walking, this task has not been thoroughly evaluated in humans.

Slope Walking in Quadrupeds

Slope walking has been studied in a range of animal species, such as horses (Dutto et al., 2004; Hoyt et al., 2005; Wickler et al., 2005), rats (Gillis and Biewener, 2002), lizards (Higham and Jayne, 2004), and cats (Gregor et al., 2001; Kaya et al., 2003; Smith and Carlson-Kuhta, 1995). Many of these studies have investigated basic properties of muscles during slope walking. In cats, slope walking has been used to investigate neural control, and several studies have reported patterns of neuromuscular response that are different from those observed during level walking (Carlson-Kuhta et al., 1998; Gregor et al., 2001; Kaya et al., 2003; Smith and Carlson-Kuhta, 1995; Smith et al., 1998). Smith and colleagues (1995, 1998) have used the adult cat model extensively in both treadmill and over-ground studies, and have reported unique muscle firing patterns. For example, hip flexor (iliopsoas) activity has been observed during the stance phase of downslope walking, when hip extensor activity normally occurs during level walking (Smith et al., 1998). In addition, unexpected hamstring activity occurred during stance in upslope walking (Carlson-Kuhta et al., 1998). The authors attempted to explain these results in light of kinetic data, but were limited because their testing system did not include a force plate. They were able to use their observed muscle firing patterns in conjunction with a previously proposed central pattern generator model (see Figure 7) to develop two task-specific movement control models (Figure 7) (Smith et al., 1998). The changes relevant to the hip, knee, and ankle joints are discussed here.

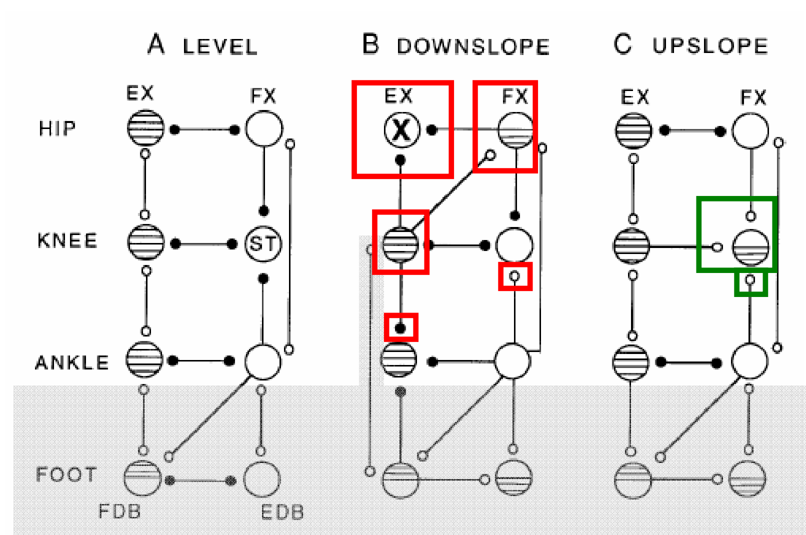

Figure 7. Models for Motor Control of the Feline Hindlimb. The level walking model (A) was adapted from Grillner (1981) by Smith *et al.* (1998) to account for muscle firing pattern changes observed during downslope (B) and upslope (C) walking. Small open circles are excitatory connections; small closed circles are inhibitory connections. Large striped circles are units that act together during stance; large white circles are units that act together during swing; half-striped circles represent "dual-phase units." Changes from level walking for the hip, knee, and ankle are boxed and discussed in the text.

During downslope walking stance the hip extensors were inactive, the hip flexors were active, knee extensor activity was constant, and ankle extensor activity decreased. Therefore in the downslope model (Figure 7B) the hip extensors are inhibited by all adjacent units and a "dual-phase unit" is created for the hip flexors. That is, the hip flexors are excited by both the knee extensors and the ankle flexors such that they fire during the main periods of both extensor and flexor activity (i.e. during both stance and swing). In addition, the knee extensors receive no excitatory input, and they inhibit the ankle extensors. This change restricts the ankle extensors to firing briefly only when a transition occurs between flexor and extensor activity. During downslope swing knee flexors activity was prolonged and similar to that of ankle flexors. To model this, the knee flexors are excited rather than inhibited by the ankle flexors.

During upslope walking knee flexor activity was observed during stance, and was also prolonged during swing, similar to the ankle flexors as in downslope walking. The corresponding model for upslope walking (Figure 7C) therefore includes an excitatory connection from the knee extensors to the knee flexors, creating another "dual-phase unit." In addition, the same knee/ankle flexor change that was made in the downslope model is also made for the upslope model. These changes allow the knee flexors (specifically the hamstrings) to be active during most of the stride for upslope walking.

The original CPG model (Figure 7A) was developed as a pure spinal cord pattern generator that could operate in the absence of both afferent feedback and descending signals (Grillner, 1981). However, the muscle firing patterns used by Smith *et al.* (1998) to modify this model were obtained in experiments on intact preparations, which included both of these additional inputs. It is therefore possible that the changes required for the original model to apply to slope walking (Figure 7B and 7C) reflect alterations in afferent feedback and/or supra-spinal control, rather than fundamental changes to the CPG output. It is also possible that spinalized, deafferented cats walking on a slope would exhibit the same muscle firing pattern changes observed by Smith and colleagues, supporting the idea of altered CPG output. Leroux *et al.* (1999) cite a Neuroscience abstract reporting that spinalized cats have difficulty altering their EMG and locomotor patters to walk on inclined surfaces. The authors also found that human subjects with a spinal cord injury use a variety of kinematic and motor strategies to adapt to sloped walking surfaces. This information seems to support the first conclusion about the modified CPG models. Because supraspinal and feedback control are present in the quadruped preparation, these new models will be referred to as *movement control* models rather than *CPG* models. This terminology is more fitting for extending these findings to humans, where the distinction between CPG, supraspinal, and feedback control would be nearly impossible to make. Regardless of terminology, these models afford us some insight into the chosen strategy of the nervous system for negotiating sloped surfaces.

The understanding of motor control can further be increased when EMG and kinetic data are combined. For example, consider the findings of Gregor *et al.* (2001) with regard to the soleus (Sol) force during quadruped slope walking. The soleus appears to be the primary muscle contributing to the plantarflexor moment during downslope walking. In contrast, for upslope walking the soleus force decreases in spite of an increased plantarflexor moment. These results have been confirmed (Kaya et al., 2003) and may be explained by analyzing the associated joint kinetics. The authors report increased ankle plantarflexor and knee flexor moments during the stance phase of upslope walking. These joint moment demands favor activation of the two-joint gastrocnemius over the single-joint soleus (Gregor et al., 2001). On the other hand, during downslope walking the joint moment demands (increased knee flexor and decreased plantarflexor moments) do not favor the activation of the gastrocnemius over the soleus. This may help to explain the authors' finding that the soleus is the primary contributor during downslope walking, but that its force decreases during upslope walking. This example illustrates the advantage of combining knowledge of muscle firing patterns and joint kinetics to elucidate control strategies during specific tasks.

Collectively, these results from quadruped slope-walking studies have not been consistent with the responses of the neuromuscular system reported for level walking. These findings have helped generate the notion that sloped surfaces require specialized responses from the nervous system and they support the need for more extensive investigations in this environment. The studies also highlight the importance of a complete biomechanical analysis in order to more thoroughly explain the observed results. Finally, the unique findings from these quadruped studies suggest that human slope walking studies might yield some equally interesting findings.

Slope Walking in Humans

The effects of slope walking on human locomotion have been studied in terms of kinetics, kinematics, and electromyography (EMG), on both treadmills and ramps, in both laboratory and natural environments. For the most part, these studies have been conducted on relatively young healthy subjects. The studies are outlined here and the relevant results are used as a basis for comparison in later sections.

Gait characteristics of pedestrians in natural settings are known to differ from those recorded in a gait analysis laboratory (Finley and Cody, 1970; Sun et al., 1996). In light of this, Finley and Cody (1970) submitted that it was necessary to use data obtained in a natural gait environment as a basis of comparison for gait data obtained in a lab. In general, the study of gait in a natural environment has important implications in ergonomics, environmental safety design, prosthetics, and assistive devices. The desired outcomes of designs or products in any of these areas should be based as much as possible on natural gait rather than the constrained gait that may be observed in a laboratory (Finley and Cody, 1970; Sun et al., 1996). Unfortunately, the amount of information that can be obtained without altering the natural gait environment is limited as subjects cannot be aware that they are being observed. In spite of these limitations, Finley and Cody (1970) and Sun *et al.* (1996) studied characteristics of pedestrians on inclines by simply observing pedestrians in a pre-characterized setting (i.e., a pre-defined walkway distance on a measured incline). Finley and Cody (1970) recorded the number of steps and time required for 1106 urban pedestrians to traverse a marked distance. The locations used in their study had inclines up to 3.25˚ (5.7%), and approximately equal numbers of pedestrians were observed walking uphill as walking downhill. Since all of their locations had an incline, the authors did not have level walking data to use for comparison. Sun *et al.* (1996) performed a more comprehensive investigation of pedestrian gait parameters during walking on sloped surfaces; the authors unobtrusively observed 2400 pedestrians walking up and down a concrete ramp whose incline ranged from 2˚ to 9˚ (3.5% to 15.8%). The authors recorded subject gender, the number of steps taken and time required to traverse a marked distance, but did not take measurements for level walking either.

Although gait observed in a natural environment can be useful, the limited amount of data that can be collected necessitates laboratory gait analysis, in spite of the fact that subjects might alter their gait. There have been numerous studies conducted on subjects walking on treadmills at different grades. With treadmill walking subjects can take a significant amount of time to adjust to the walking condition, speed can either be controlled or self-selected, and kinematic and EMG data can be collected. Leroux and colleagues studied subjects walking upslope at grades up to 15˚ (26.8%) (Leroux et al., 1999), and both upslope and downslope up to 5.71˚ (10%) (Leroux et al., 2002). The authors collected EMG data for upslope walking in addition to kinematic data, as did another group studying knee rehabilitation (Lange et al., 1996). A third group also studied kinematics and EMG of subjects walking upslope and downslope on a treadmill, but the information was published only in abstract form (Mitsui et al., 2001). Wall and coworkers studied the changes in kinematics and transverse plane pelvic oscillations, as well as spatial and temporal gait parameters for upslope and downslope treadmill walking at grades up to 11.31˚ (20%) (Wall et al., 1981). Two groups have studied the plantar pressure patterns of subjects during incline walking on a treadmill (Grampp et al., 2000; Simpson et al., 1993). Others have used graded treadmills to study postural changes (Leroux et al., 2002) and oxygen uptake and energetics during slope walking (Laursen et al., 2000; Minetti et al., 2002; Pivarnik and Sherman, 1990).

One limitation of using a treadmill is that kinetic data cannot typically be collected. Although force treadmills have been designed, tested, and used to study level walking and running (Gottschall and Kram, 2005; Kram et al., 1998), they have not yet been modified for incline studies. Ramps with embedded force plates solve this problem, and have been in use since the late 1960s (Morrison, 1969). Instrumented ramps allow less time for subject accommodation than treadmills, and walking speed is not readily controllable. In response, some groups have allowed subjects to walk at a comfortable self-selected pace (Cham and Redfern, 2002; McVay and Redfern, 1994; Redfern and DiPasquale, 1997) while others have controlled the step frequency of their subjects (Kuster et al., 1995). Instrumented ramps have been used to study the COP trajectory (Kawamura et al., 1991), hip joint contact forces (Stansfield and Nicol, 2002), friction requirements (Buczek et al., 1990; McVay and Redfern, 1994), and the effect of changing the floor conditions (slope and coefficient of friction) on subject behavior (Cham and Redfern, 2002) during slope walking. Such ramps have also been used to better understand the kinematics and kinetics of slope walking (Buczek, 1990; Cham and Redfern, 2002; Kuster et al., 1995; Redfern and DiPasquale, 1997; Schwameder et al., 1999) and running (Buczek and Cavanagh, 1990). These studies generally used grades of 10˚ (18%) or less, but some used grades as steep as 20˚ (36%).

Muscle activation during slope walking has also been studied independently of kinetics and kinematics. Tokuhiro and coworkers analyzed the activities of the tibialis anterior (TA), medial gastrocnemius (MG), RF, semitendinosus (ST) and gluteus maximus (GM) during uphill, level and downhill walking up to grades of 12° (21%) (Tokuhiro et al., 1985). The materials and methods section lacks most relevant experimental details, so the number of trials, walking speed, order of inclines walked, and data filtering/processing/averaging techniques, as well as whether the subjects walked on a ramp or treadmill are all unknown (Tokuhiro et al., 1985). Nevertheless, their data can be compared to those from similar slope walking EMG studies (Brandell, 1977; Patla, 1986; Powers et al., 1996). Patla (1986) investigated EMG patterns of the TA, Sol, MG, vastus lateralis (VL), RF, and biceps femoris (BF) during level and upslope treadmill walking up to 14° (25%) (Patla, 1986). Powers and coworkers (1996) studied the EMG patterns of the four vastus muscles in patients with and without patellofemoral pain walking upslope and downslope at 12˚ (21%). Brandell (1977) also studied the function of muscles at the knee joint during slope walking, although their data are complicated by their use of a fast walking speed. EMG data from these studies can also be compared to the EMG data and correlated to the kinematics data from the treadmill walking studies already discussed (Lange et al., 1996; Leroux et al., 1999).

There are currently no studies that integrate all the relevant gait data (kinematics, kinetics and EMG) for slope walking. Joint kinetics have been reported for downslope walking in only three studies (Cham and Redfern, 2002; Kuster et al., 1995; Redfern and DiPasquale, 1997). No study to date has reported kinetic data for upslope walking. EMG data from slope walking is also limited, especially for downslope walking and walking at steep grades. One purpose of this project was to address this gap in our fundamental knowledge about the mechanical demands of slope walking through a comprehensive biomechanical analysis of human slope walking. The information from such an analysis would lead to a more thorough understanding of how the nervous system chooses to adapt to the demands that occur during slope walking.

Backward Locomotor Tasks in Humans

Backward locomotor tasks have received limited attention in the literature, compared to forward locomotor tasks. This discrepancy makes sense when one considers that backward locomotor tasks are used much less frequently than the corresponding forward tasks. Injury rehabilitation, metabolic cost, and strength training are some common applications of backward locomotor tasks such as walking, running, and cycling. These applications provide the motivation for many of the published studies on backward locomotion (Chaloupka et al., 1997; Clarkson et al., 1997; DeVita and Stribling, 1991; Eisner et al., 1999; Flynn et al., 1994; Myatt et al., 1995; Neptune and Kautz, 2000; Terblanche et al., 2004; Threlkeld et al., 1989). Other studies have looked at the relationship between control strategies used during the forward and backward tasks. For example, in cycling the activity of some muscles is independent of pedaling direction while the activity of other muscles is not (Neptune et al., 2000; Ting et al., 1999), which suggests the possibility of "biomechanical functional pairs" that are used to control the muscles during pedaling and perhaps also during walking (Ting et al., 1999). As mentioned, backward level walking has also used for the purpose of studying the control system (Grasso et al., 1998; Thorstensson, 1986; Van Deursen et al., 1998; Vilensky et al., 1987; Winter et al., 1989). These studies all report that the tasks of forward and backward level walking are kinematically similar, and Winter (1989) even shows similarities in the joint kinetics. The differences in muscle activity that are observed between the tasks, however, makes it difficult to conclude that the same control system is regulating the tasks.

Backward upslope walking is used clinically for rehabilitation of the knee joint for patients with patellofemoral pain, anterior cruciate ligament injuries, and osteoarthritis, for example (Cipriani et al., 1995). In spite of the increasing popularity of this exercise, however, it has not been extensively studied. The majority of published data on backward upslope walking report the metabolic and cardiovascular demands that occur during the task (Chaloupka et al., 1997; Hooper et al., 2004; Minetti and Ardigo, 2001). There is currently only one study documenting the biomechanics of backward upslope walking, but it reports only kinematics and some EMG (Cipriani et al., 1995). There have been no studies on the reaction forces or joint kinetics of backward upslope walking, and there are no reports at all on backward downslope walking. As with backward walking on level ground, backward slope walking was expected to be kinematically very similar to forward slope walking; backward upslope walking was expected to be a kinematic reversal of forward downslope walking, and vice versa. Therefore the mechanical demands were also expected to change during backward slope walking. However, the demands were expected to be reversed from forward walking. For example, backward upslope walking should require power generation to lift the body, while its kinematic counterpart (forward downslope walking) should require power absorption to control the descent of the body. The changing demands were expected to require changes in the motor programs, as with forward slope walking. Again, such changes have the potential to provide further insight into the neural control of locomotion. Therefore, another purpose of this project was to examine the reaction forces, joint kinematics and kinetics, and muscle activity in the lower extremity during backward slope walking.

CHAPTER 3

THE INSTRUMENTED RAMPED WALKWAY[1]

Introduction

The majority of previous investigations on slope walking used inclined treadmills to create the walking surface. Kinematic and EMG data can be conveniently collected when using an inclined treadmill, but ground reaction force data is more difficult and expensive to obtain, which limits the scope of the biomechanical analyses. In order to evaluate the kinetics of sloped locomotion, several approaches have been taken to incorporate force plates within ramped walkways. One type of structure utilizes a ramp of fixed grade with an independent section of the surface coupled to a force plate in the floor beneath the ramp (Kuster et al., 1995; Stansfield and Nicol, 2002). This design presumes that the forces are directly transmitted through the rigid connections from the ramp surface to the force plate and that the “independent” section of the ramp surface has no interaction with the rest of the ramp surface. Another type of structure has the force plate embedded in the surface of the ramp, with slope intensity either fixed (Buczek and Cavanagh, 1990; Schwameder et al., 1999) or adjustable (Cham and Redfern, 2002; McVay and Redfern, 1994; Redfern and DiPasquale, 1997). This type of structure allows ground reaction forces to be captured directly by the force plate. Designs using a fixed grade offer the advantage of allowing the force plate to be fully supported, such as by

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[1]The text and figures in this chapter are adapted from the article "Characterization of a System for Studying Human Gait During Slope Walking" published in the Journal of Applied Biomechanics (Lay et al., 2005b). © 2005 Human Kinetics. Reprinted with permission from Human Kinetics Publishers (Champaign, IL).

mounting the force plate on a concrete block so it is flush with the ramp surface surrounding it (Buczek and Cavanagh, 1990). A disadvantage of these fixed-grade systems is the limited amount of data that can be collected, i.e., a separate ramp construction is required for each grade. Adjustable ramps allow different slope intensities to be studied but are potentially less stable. Although these systems have been used, the literature does not offer any discussion about the limitations of the designs, and the integrity of force plate recordings in such systems has not yet been presented.

Our goal for the first step of this project was to construct and characterize an adjustable ramp system for analysis of the kinetics of over-ground locomotion at a variety of slope intensities, i.e. up to ±39%. There were several constraints on the design of the system. First, the force platform had to be mounted securely and properly supported so the integrity of the data would not be compromised. The platform also needed to be positioned to allow forces to be captured during upslope and downslope walking from the same foot on each subject (either the right or the left foot, according to subject preference). The walkway had to be long enough to require a person of average height to take several strides to traverse it; this would help eliminate the effect of the transition on the recorded data. Finally, the system needed to be *easily* adjustable so that walking at several grades could be studied. The system that evolved from these requirements and its validation are described below.

The System

The ramp system (Figure 8) consists of a two-segment walkway (A and B) that is 1.07m wide and totals 4.3m in length (segment A 1.19m, segment B 3.11m). Segment A provides a transition from the floor to segment B, which contains a force plate (C, 0.4m x 0.6m x 0.083m, model FP4060-08, Bertec Corp., Columbus, OH). Segment B is hinged to a platform (D, 0.52m long) at one end and to the shorter walkway segment (A) at the opposite end, where it is also mounted on 0.1m hard rubber swivel locking casters. The casters allow the ramp to move freely as the platform is raised or lowered. Two adjustable nylon base swivel leveling mounts are affixed to the walkway frame adjacent to the casters. Once the ramp is in the desired position, the mounts are lowered, lifting the casters off the ground, and supporting the lower end of segment B. The platform is mounted on a scissor lift (shown schematically as F) that is powered by a hydraulic pump. The platform and scissor lift are constrained by a 2.34m high tower (E) that also acts as a safety railing when subjects are standing on the platform.

The tower (E) is made from 0.05m square steel tubing with a 0.003m wall thickness and rests on four adjustable nylon base swivel leveling mounts. Specially designed steel brackets pin to the tower at six different heights corresponding to ramp grades of 0, 5, 15, 25, 35, and 39%. The platform (D) rests on these brackets to effectively decouple the platform from the scissor lift.

The frame for walkway segment B is constructed from 0.1m aluminum c-channel with a 0.006m flange and is stabilized by pieces of 0.006m thick aluminum plate welded to the open face of the c-channel. The frame has four crossbeams, and the center two create the upper and lower boundaries of the force plate mount, which is centered in ramp segment B (Figure 8, lower panel). Two pieces of c-channel affixed between these crossbeams form the side boundaries for the force plate mount. Additional L- shaped pieces of aluminum are welded to the insides of the pieces of c-channel that form the

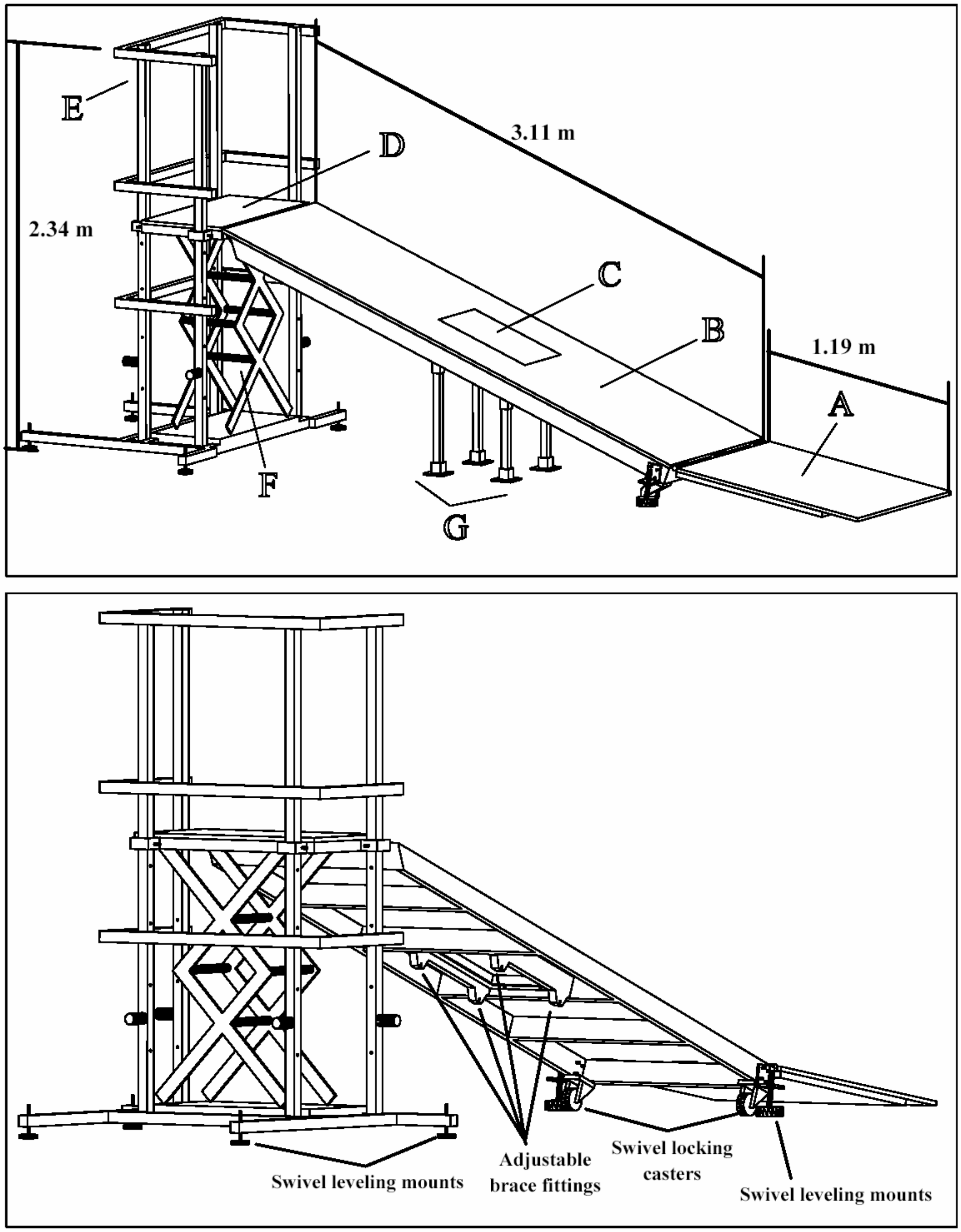


Figure 8. The Ramped Walkway System. Upper panel: anterior-oblique view of the ramped walkway system showing the two segment walkway (A, B), force plate (C), platform (D), tower (E), scissor lift (schematically, F), and vertical supports (schematically, G) as well as dimensions. Lower panel: posterior-oblique view of the ramp showing the force plate mount in the frame and select components.

borders of the force plate mount, creating a ledge that supports the force plate. The force plate is bolted directly to this ledge in the ramp frame. The force plate mount is slightly larger than the force plate, ensuring that the sides of the force plate do not contact the ramp frame or surface when the force plate is centered in the mount. Four adjustable Unistrut® brace fittings are bolted onto the underside of the ramp frame adjacent to each corner of the force plate. Incline-specific Unistrut® channel posts (shown schematically as G) are attached to these fittings once the ramp is positioned at the desired grade to provide additional stabilization for the force plate. The walkway frame is topped with 0.018m thick plywood that is covered with corrugated vinyl sheeting to improve traction. The force plate is flush with the surface of the vinyl sheeting.

Experimental Analyses

The following tests were conducted to determine the accuracy of the force plate output obtained using the ramped walkway system. Unless otherwise specified, ground reaction forces and moments were collected from the force plate at 300 Hz and processed in the Peak Motus system (Peak Performance Technologies, Inc.). Force platform data were subsequently used to calculate the instantaneous center of pressure. All human subjects gave their informed consent to participate in this study, which was approved by the Institutional Review Board of the Georgia Institute of Technology.

Natural Frequency. The natural frequency of the ramped walkway system was evaluated and compared to the natural frequency of the force plate alone to verify that ground reaction forces could be accurately measured during sloped locomotion (Kram et al., 1998). The force plate was first tested while mounted in the floor of the gait analysis laboratory. Force plate data were collected at 1200Hz for five separate trials of ten seconds, during which time a rubber mallet was used to strike the force plate. The force plate was then removed from the floor and mounted in the ramped walkway system at 0% grade. Evaluation of the force plate/ramp system included five trials where the mallet was used to strike the force plate and five more trials where the mallet was used to strike the walkway alone. Anterior-posterior (AP), medial-lateral (ML), and normal components of the force output were analyzed separately. The "ringing" in the force trace was concentrated in the first 100ms of data following the mallet strike. Therefore the natural frequency in each direction was calculated by performing a power spectrum analysis on these 120 data points using MATLAB (Mathworks, Inc.)

Center of Pressure. To test the accuracy of the center of pressure (COP) calculated from the force plate output, a 200N bar was supported such that it rested on a small nut (0.01m in diameter) centered over a known location on the force plate surface. For each ramp grade, force plate data were recorded for ten seconds during static loading (condition A). In addition, force plate data were collected for two ten-second trials where a representative healthy subject (Subject 1, male, age 32, 77kg body mass) walked up and down the ramp around the force plate (condition B). This condition was used to evaluate the influence of vibrations in the walkway (caused by walking on the ramp) on the force plate output. The average and standard deviation of the COP location (x and y coordinates) were calculated for each trial from conditions A and B. The radial distance between the calculated COP location and the actual COP location (assumed to be the center of the area of application of the force) was calculated for all trials.

Force. The accuracy and variability of the force plate output were quantified during loading with a known, static weight under conditions A and B (as above), because fluctuations in the force output may occur due to vibrations in the system caused by walking. Static loads were applied to the plate using a 327N dumbbell and various weight plates. The applied loads were representative of typical forces observed in each loading direction during normal locomotion. At 0% grade the dumbbell was placed on the force plate to apply a 327N load normal to the surface. In addition, a 110N load was applied in the negative AP shear direction, and a 44N load was applied in the negative ML shear direction. AP and ML shear forces were applied by attaching free weights to the dumbbell with rope and then hanging the weights off the top and sides of the ramp, respectively, using a simple pulley system. The lines of action of the ropes were maintained parallel to the ramp surface such that the majority of the hanging weight was contributing to a pure shear load. As the ramp was inclined, the 327N dumbbell on the force plate contributed force in the positive AP direction, opposite to the direction of the force applied by the hanging weights. The hanging load was reduced and eventually removed, so these forces did not cancel each other out. Force output was recorded for ten seconds under condition A and for two trials of ten seconds under condition B for all grades. The average, standard deviation, and coefficient of variation of the force output during each trial were calculated and compared between conditions A and B. In addition, the average normal force at each grade was compared to the actual applied force (327N * cosine(ramp angle)) to test the accuracy of the force plate output. The AP and ML loads were not compared to the actual load because the rope/pulley setup made it difficult to precisely quantify the 'actual' applied load.

Ground Reaction Forces. Ground reaction forces (GRF) were collected as an exemplary subject (Subject 2, male, age 30, 87kg body mass) walked at a comfortable, self-selected pace first on the force plate located in the floor of the gait analysis laboratory and then on the same force plate mounted in the ramp system at 0% grade (n=10 for each condition). Stance time was calculated for each trial from the component of the GRF normal to the plate. To limit the effect of speed on the GRF comparison, the eight trials from each condition with the most similar stance times were selected for analysis. The two groups of stance times were compared using a paired t-test. To eliminate any possible noise from ramp vibrations, all GRF data were filtered using a 4th order zero-lag Butterworth digital lowpass filter with a cutoff frequency set at 25 Hz (Kram et al., 1998). GRF data were normalized to 200 points and ensemble averaged across all trials for each condition. The repeatability of the GRF waveforms for the ramp and floor trials was evaluated using the "within test day" adjusted coefficient of multiple correlation (CMC) (Kadaba et al., 1989), with "day" corresponding to condition (floor or ramp) here. The similarity of the ramp and floor ground reaction forces was assessed using the CMC for the repeatability of a waveform "between test days" (Kadaba et al., 1989). These CMC values were compared to those reported by Kadaba et al. (1989).

As a final test, a healthy subject (Subject 3, male, age 24, 70kg body mass) walked up and down the ramp at a comfortable pace at each of the six ramp grades. The subject was asked to try to maintain a similar walking speed at all grades. 10 walking trials were recorded at each grade, with an upslope and a downslope footfall captured during each trial. Stance times for each grade were compared *post-hoc* to stance times for 0% grade using a paired t-test. GRF data were processed and filtered as above.

Marker Tracking. The ramped walkway system was surrounded by a six camera Peak Motus 3D Optical Capture system. All kinematic data were sampled at 60 Hz. An L-shaped calibration frame (4 control points, 1.50m x 1.00m x 0.09m) and wand (2 control points, 0.913m) were used for 3-D space reconstruction. The ramp surface served as the base for the coordinate system (the x-y plane). Within the calibrated volume, the motion capture system has been shown to be accurate to 0.002m. To analyze the possible effect of ramp motion on the calculated marker coordinates, six retroreflective markers were affixed to random positions on the surface of the ramp. At each grade marker position data (x, y, and z coordinates) were collected for ten seconds under conditions A and B (as above) and the average difference in position between the conditions was calculated.

Results

Natural Frequency. The natural frequency of the force plate while in the floor was approximately 350Hz for all three components. When the walkway was struck with the mallet, the natural frequencies of the force plate output were approximately 100Hz for the normal component, 120Hz for the ML component, and 230Hz for the AP component. When the force plate in the ramp was struck directly, the power spectra from the force outputs showed two peaks, typically around 100Hz and 350Hz. During a walking trial, the resolution of the force plate would be limited to the lower of these two frequencies. Since 99% of the integrated power content of the normal and AP components of the GRF of human walking has been shown to lie below 15Hz (Antonsson and Mann, 1985; Kram et al., 1998), the natural frequency of the ramp system is acceptable for accurate representation of the GRF during locomotion.

Center of Pressure. For the purpose of quantifying a difference between calculated and actual COP location, the actual COP was assumed to be at the center of the 0.01m diameter area of force application (x = 0.298m, y = 0.199m). The average and standard deviation of the calculated COP location over the 3000 points for all trials are given in Table 1. The radial distance between the average calculated COP and the actual COP was less than or equal to 0.005m for all trials, indicating that the calculated COP was always within the actual area of force application.

Force. Table 2 shows the average, standard deviation (SD), and coefficient of variation (CV = SD/average*100) of the force output over the 3000 data points from each trial. For condition A, the CV of the force output for the AP and ML components was always less than 1.0%, and for the normal component it was always less than 0.2%. The standard deviation of the force output always had a fairly consistent minimum magnitude regardless of the applied load (approximately 0.3 N). This constant noise, referred to as "broadband noise," results in a higher variation when lower forces are applied and therefore explains the larger CV value for the AP and ML components. For the dynamic

therefore explains the larger CV value for the AP and ML components. For the dynamic

Table 1. Average (SD) Position of the Calculated COP (in m)

	Condition A		Condition B			
	Trial 1		Trial 1		Trial 2	
	x	y	x	y	x	y
0%	0.295 (0.001)	0.198 (0.001)	0.298 (0.001)	0.198 (0.001)	0.297 (0.001)	0.196 (0.001)
5%	0.299 (0.001)	0.195 (0.001)	0.302 (0.001)	0.201 (0.001)	0.296 (0.001)	0.196 (0.001)
15%	0.301 (0.001)	0.198 (0.001)	0.301 (0.001)	0.198 (0.001)	0.300 (0.001)	0.197 (0.001)
25%	0.303 (0.001)	0.200 (0.001)	0.302 (0.001)	0.196 (0.001)	0.300 (0.001)	0.194 (0.001)
35%	0.297 (0.001)	0.202 (0.001)	0.298 (0.001)	0.200 (0.001)	0.298 (0.001)	0.199 (0.001)
39%	0.298 (0.001)	0.200 (0.001)	0.297 (0.001)	0.199 (0.001)	0.299 (0.001)	0.198 (0.001)

Note: Condition A was pure static loading of force plate; Condition B included a male, age 32, body mass 77kg, walking up and down the ramp while the force plate was statically loaded. Actual COP was assumed to be at the center of the 0.01m diameter area of force application (x = 0.298m, y = 0.199m); n = 3,000 for each trial.

trials at all inclines, the CV was less than 1.5% for the normal component, less than 4% for the AP component, and less than 8% for the ML component. The difference in the average force between conditions A and B was less than 0.5% for all trials at all inclines in the normal and AP components and was less than 1.5% for all trials at all inclines in the ML component. The normal force differed from the actual applied force by less than 0.8% (corresponding to an actual magnitude of less than 3N) for all trials at all grades for both conditions A and B.

Table 2. Average (SD) (in N) and Coefficient of Variability (%) of Force Output

	Anterior-Posterior Force			Medial-Lateral Force			Normal Force			
	Condition A	Condition B		Condition A	Condition B		Actual Load (N)	Condition A	Condition B	
		Trial 1	Trial 2		Trial 1	Trial 2			Trial 1	Trial 2
0%	-107.97 (0.33) 0.31%	-108.35 (1.24) 1.14%	-108.35 (1.06) 0.98%	-42.51 (0.32) 0.74%	-42.28 (1.37) 3.25%	-42.47 (1.56) 3.67%	327.33	327.72 (0.57) 0.18%	327.68 (1.62) 0.49%	327.56 (1.93) 0.59%
5%	-93.43 (0.33) 0.35%	-93.32 (1.13) 1.22%	-93.39 (1.26) 1.35%	-42.21 (0.32) 0.76%	-42.54 (1.80) 4.23%	-41.22 (1.78) 4.33%	327.02	327.28 (0.51) 0.15%	327.71 (3.41) 1.04%	327.87 (3.39) 1.03%
15%	-83.90 (0.38) 0.45%	-83.89 (1.53) 1.83%	-83.74 (1.70) 2.03%	-41.93 (0.34) 0.80%	-42.10 (1.82) 4.32%	-41.85 (1.99) 4.75%	323.74	323.23 (0.53) 0.17%	324.66 (1.29) 0.40%	324.73 (1.59) 0.49%
25%	77.23 (0.35) 0.45%	77.15 (2.17) 2.82%	77.13 (2.44) 3.17%	-40.78 (0.33) 0.81%	-40.34 (2.31) 5.73%	-40.17 (2.48) 6.18%	318.29	318.32 (0.54) 0.17%	318.66 (1.64) 0.51%	318.84 (1.67) 0.52%
35%	105.80 (0.35) 0.33%	105.83 (3.24) 3.06%	105.85 (3.77) 3.56%	-42.32 (0.35) 0.84%	-42.72 (2.50) 5.85%	-42.51 (2.28) 5.36%	309.50	308.67 (0.52) 0.17%	308.42 (2.01) 0.65%	307.36 (2.57) 0.83%
39%	117.98 (0.37) 0.31%	117.85 (3.76) 3.19%	117.87 (3.49) 2.97%	-40.63 (0.39) 0.96%	-40.59 (3.11) 7.65%	-41.00 (3.20) 7.80%	305.60	307.98 (0.49) 0.16%	307.66 (2.34) 0.76%	307.50 (2.18) 0.71%

Note: The applied AP and ML forces are not reported as the setup made them difficult to assess accurately. For these loading directions, the force output from Condition B is compared to that from Condition A to assess the effect of ramp vibrations on the force plate output. Actual load applied in the direction normal to the plate is shown for comparison to measured force; n = 3,000 for each trial.

Ground Reaction Forces. The average GRF components for the floor and ramp trials are shown in Figure 9. The t-test indicated that the stance times for the floor (avg ± sd: 0.69 ± 0.01 m/s) and ramp (0.69 ± 0.01 m/s) trials were not significantly different (n=8, p = 0.91). The CMC values are presented in Table 3 along with the reported values (Kadaba et al., 1989). The CMC approaches 1 with increasing similarity between waveforms. The repeatability of the waveforms for the ramp and floor trials (i.e. within-condition) was excellent and comparable to the reported data. Furthermore, the comparison between the floor and ramp GRF components indicates that these waveforms are nearly identical.

Upslope and downslope GRF data from all grades are shown in Figure 10. Each curve is the ensemble average of the 10 trials for that condition (11 conditions: 0, ±5, ±15, ±25, ±35, ±39%). The average stance times are shown in Table 4. Paired *t*-tests indicated that only the stance times from the steepest slopes, +35% and ±39%, were significantly different from those at level ($p < 0.05$). The normal and AP forces varied with ramp grade, while the ML force appeared similar across all ramp grades.

Table 3. Repeatability of Ground Reaction Force Data

	Within Condition CMC			Between Condition CMC	
	Reported	Floor	Ramp	Reported	Floor vs Ramp
Normal Force	0.997	0.996	0.993	0.995	0.994
AP Force	0.993	0.997	0.993	0.988	0.991
ML Force	0.953	0.951	0.944	0.942	0.948

Note: Reported CMC values are the average over 40 participants, with "within condition" assessing the repeatability of 3 trials on the same test day, and "between condition" assessing the repeatability of each participant over 3 test days (Kadaba et al., 1989). CMC value approaches 1 with increasing similarity of compared waveforms, and approaches 0 when the waveforms are very dissimilar. Average GRF curves shown in Figure 9.

Marker Tracking. The average differences in the positions of the markers between conditions A and B are shown in Table 5. The greatest difference in marker coordinates was 0.001m in any direction for all markers at all inclines and was 0.000m for the majority of the trials. Because the markers were affixed to the ramp, these differences can only be due to ramp motion, and they demonstrate that the effect of ramp motion on calculated marker coordinates is negligible.

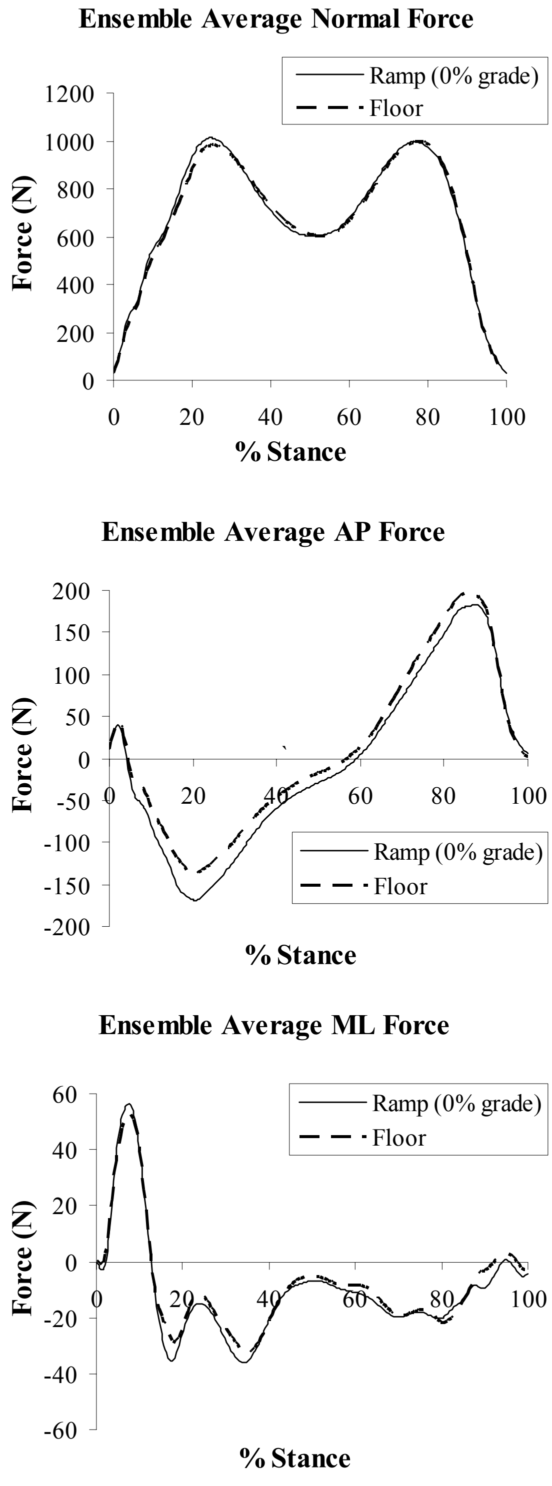
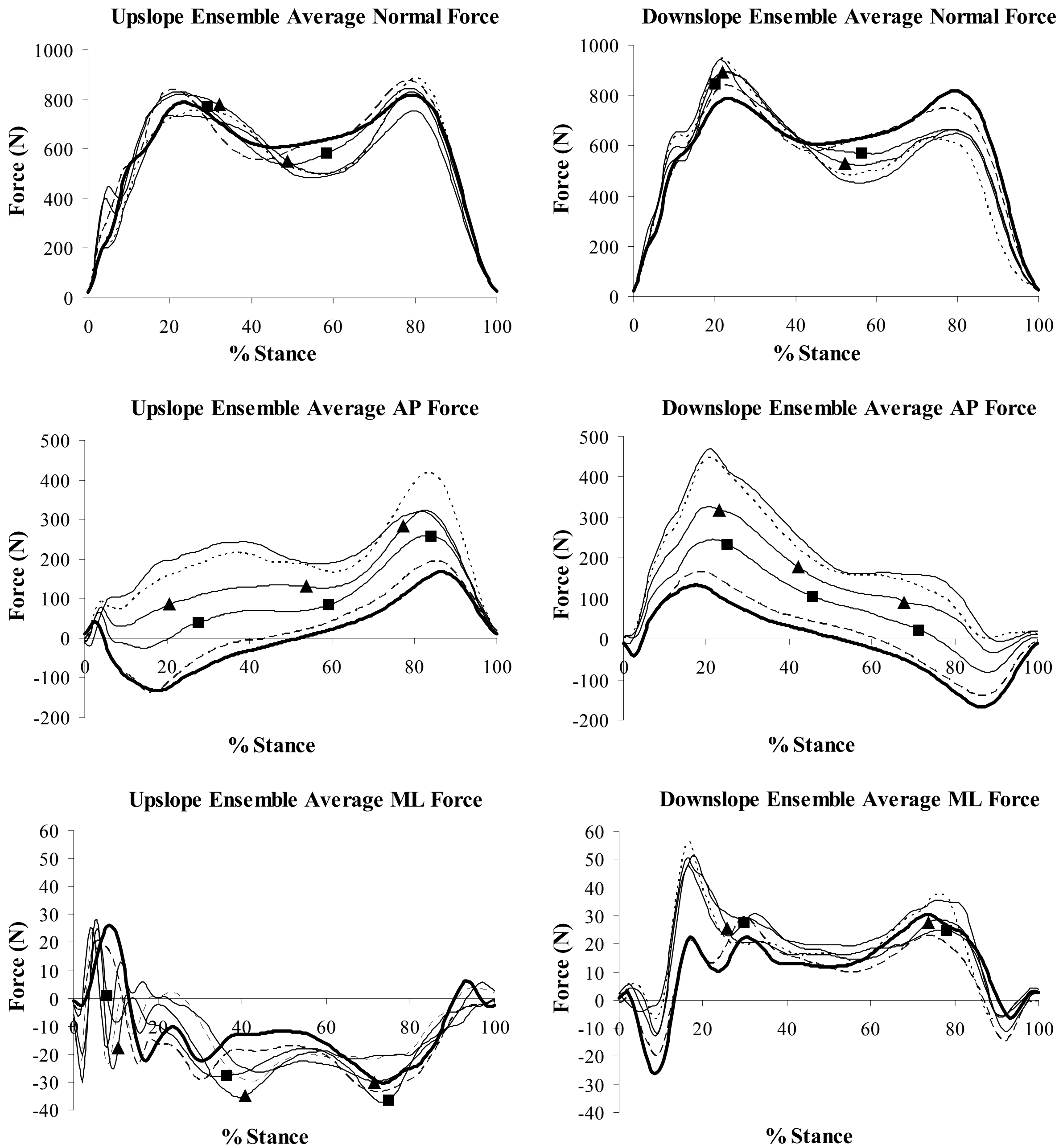


Figure 9. Floor vs. Ramp Ensemble Averaged GRF Components (normal, anterior-posterior, and medial-lateral) from right lower extremity of Participant 2 (male, age 30, body mass 87kg) walking on the force plate in the floor and in the ramp at 0% grade (n = 8 for each condition). Stance phase was normalized to 200 points for each trial before ensemble averaging.



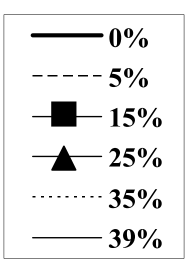


Figure 10. Upslope and Downslope Ensemble Averaged GRF Components from right lower extremity of Participant 3 (male, age 24, body mass 70kg) at all 6 ramp grades (n = 10 for each condition). Stance phase was normalized to 200 points for each trial before averaging. Because the walking direction changes, for upslope walking a positive AP force indicates a propulsive force, while for downslope walking a positive AP force indicates a braking force. Similarly, positive upslope ML forces are equivalent to negative downslope ML forces and indicate a laterally directed force from the foot.

Table 4. Average (SD) Stance Times (s)

	Upslope	Downslope
0%	0.799 (0.027)	0.799 (0.027)
5%	0.797 (0.028)	0.775 (0.023)
15%	0.792 (0.026)	0.786 (0.030)
25%	0.803 (0.015)	0.781 (0.039)
35%	0.864* (0.055)	0.782 (0.012)
39%	0.883* (0.042)	0.754* (0.033)

Note: Corresponding GRF curves are shown in Figure 10. * Significant differences in stance time from the 0% grade condition (as indicated by paired *t*-tests), $p < 0.05$.

Table 5. Difference in Average Marker Position between Conditions A and B (in m)

		Marker 1	Marker 2	Marker 3	Marker 4	Marker 5	Marker 6
0%	x	0.000	0.000	0.000	0.000	0.000	0.000
	y	0.000	0.000	0.001	0.000	0.000	0.000
	z	0.000	0.000	0.000	0.000	0.000	0.001
5%	x	0.000	0.000	0.000	0.000	0.000	0.000
	y	0.001	0.001	0.000	0.000	0.001	0.000
	z	0.000	0.000	0.001	0.000	0.000	0.000
15%	x	0.000	0.000	0.000	0.000	0.000	0.000
	y	0.001	0.000	0.000	0.000	0.000	0.000
	z	0.000	0.000	0.000	0.000	0.001	0.000
25%	x	0.000	0.000	0.000	0.000	0.000	0.000
	y	0.000	0.000	0.000	0.000	0.000	0.000
	z	0.000	0.000	0.000	0.000	0.000	0.000
35%	x	0.000	0.001	0.000	0.000	0.000	0.000
	y	0.001	0.000	0.001	0.001	0.000	0.001
	z	0.000	0.000	0.000	0.001	0.000	0.000
39%	x	0.000	0.001	0.000	0.001	0.000	0.000
	y	0.000	0.000	0.000	0.000	0.000	0.000
	z	0.000	0.001	0.000	0.001	0.001	0.001

Note: Average differences between Conditions A and B in the coordinates of the 6 retroreflective markers affixed to ramp surface.

Discussion

The results of these validation tests demonstrate that the ramped walkway system presented here provides an accurate means of testing the kinetics of over-ground slope walking in humans. The natural frequency of the ramp system is adequate for accurate GRF measurement, although it is lower than that of the isolated force plate. At all ramp grades the force and center of pressure outputs were accurate, and the variability of the force output was low, even with a subject walking up and down the ramp around the force plate. Level walking ground reaction forces for a subject were effectively the same when measured either by the force plate in the ramp or by the force plate in the gait laboratory floor. Ground reaction force data across the range of ramp grades show similar trends to those previously reported in the literature (Cham and Redfern, 2002; McVay and Redfern, 1994; Redfern and DiPasquale, 1997). Collectively, these data indicate that ramp movement and vibrations are minimal and that mounting the force plate in the ramp does not adversely affect its ability to accurately capture ground reaction force data. Finally, the optical capture system cannot detect ramp motion induced by a subject walking up and down the ramp, which implies that 3D kinematic marker coordinate data can be accurately collected.

The ramped walkway system presented here does have several limitations. Unlike graded treadmills, the ramp does not allow much time for subject accommodation to a walking grade. A second limitation is that the range of ramp grades is limited by the range of motion of the scissor lift used in the construction of this system. However, the steepest ramp grade (39%) presents a walking environment steeper than most slopes encountered in daily locomotion and undoubtedly presents a challenge to the neuromuscular system. Steeper slopes would result in a testing condition that some subjects might not be able to negotiate and would also greatly increase the risk of injury during testing. Lastly, the force plate is not as isolated from the external environment as it is when secured in the gait analysis laboratory floor. Hence foot-strikes on the walkway do cause small ramp vibrations that result in fluctuation in the force output. However, these fluctuations are minimal and, in the case of a GRF, can be effectively attenuated by applying a digital filter. We feel these disadvantages and limitations of the ramped walkway are strongly outweighed by the ability to collect kinetic data from this environment.

In conclusion, the validity of the system presented here was rigorously tested to ensure the unusual testing environment allowed for accurate data collection. The data presented here demonstrate the integrity of this system and justify its use in the investigation of incline walking.

CHAPTER 4

METHODS

Participant Preparation

Nine healthy adults, without any known impairments that might influence locomotion, volunteered for the study (See Table 6 for subject information). Each volunteer read and signed an informed consent form approved by the Institutional Review Board of the Georgia Institute of Technology before participating.

Table 6. Participant Information

Participant	Age	Height (in)	Height (m)	Weight (Kg)	Sex	Foot
1	24	68	1.73	80.28	F	R
2	27	71	1.80	71.19	F	R
3	24	72	1.83	75.79	M	R
4	20	71	1.80	74.55	M	R
5	30	74	1.88	88.12	M	L
6	23	72	1.83	77.13	M	R
7	23	67	1.70	65.24	F	R
8	21	64	1.63	58.61	F	R
9	20	72	1.83	69.32	M	R
Avg	**23.56**	**70.11**	**1.78**	**73.36**		
SD	3.28	3.14	0.08	8.60		

Participants were fitted with 15 retroreflective markers placed over bony landmarks according to the Helen Hayes system (Figure 3). Each marker was secured with tape to ensure it did not move during testing. Eight pairs of self-adhesive bipolar

Ag–AgCl EMG electrodes were applied unilaterally on each participant's preferred limb (8 R, 1 L). The skin was prepared by shaving (if necessary), abrading with fine grain sandpaper, and cleansing with alcohol swabs. The electrodes were positioned over the bellies of the GM, RF, VM, BF, semimembranosus (SM), MG, Sol, and TA as previously described (Basmajian and Blumenstein, 1989). Inter-electrode distance was maintained at 2 cm, and a ground electrode was placed on the anterior superior iliac spine (ASIS). All electrodes were connected to a transmitter box, and EMG signals were collected telemetrically (Konigsburg Instruments). Electrode positions and instrument function were verified by manual muscle testing and confirmation of the activity on an oscilloscope.

Experimental Protocol

Participants were asked to walk around the laboratory until the markers, wands, and EMG wires felt comfortable; adjustments were made if necessary. Participants performed at least eight forward and eight backward walking trials at each of the following five grades: -39% (-21˚), -15% (-8.5˚), 0% (0˚), +15% (+8.5˚), and +39% (+21˚). Each participant began with forward walking at the 0% condition, walking at a self-selected speed. Participants were told they would need to maintain a similar speed for all forward walking trials, and were asked to choose their level walking speed with this in mind. The stance time for each forward walking 0% trial was computed from the normal ground reaction force component and the average stance time for forward level walking was calculated. The remaining grades were presented randomly to each participant. At each grade the participants performed the forward walking trials (up and down) first and then performed the backward walking trials. Before any trials were collected the participants walked up and down the ramp several times to habituate to the task and so the starting position could be adjusted so the correct foot would strike the force platform. Each subject took at least two steps on the ramp before and after contacting the force platform. After all trials at a grade were completed the ramp grade was changed and the subject was asked to rest during this time (~10 minutes).

During testing a trial was repeated if foot contact with the force platform was incomplete or if the participant made visible stride alterations to target the platform. Eight acceptable trials were recorded for each condition. The stance time was computed for randomly selected forward walking trials during testing so some feedback could be given to the participant about their walking speed. Walking speed was not calculated or monitored during backward walking trials as that task was sufficiently difficult without asking subjects to concentrate on speed. During post-processing both forward and backward walking trials were discarded if the stance duration was greater than ±5% of that participant's average stance duration during the 0% condition. This final criterion was employed to minimize the effect of walking speed on the data (Hof et al., 2002; Kirtley et al., 1985; Winter, 1987).

Five participants were also asked to participate in an ongoing gait initiation (GI) study. If they agreed, they were asked to stand in front of the force platform, to step onto the platform on a verbal signal, to stand still on the platform, and to begin walking on a second verbal signal. Five trials were collected at each grade. The GI tasks varied by which direction the participant was walking (up or down) and by which foot contacted the platform (stance, swing, or both). If the trials were stance foot trials, the subject was asked to step onto the platform with their non-initiating foot (determined by watching the subject initiate gait several times on the laboratory floor), and to place their initiating (swing) foot next to the force platform. The opposite instructions were used for swing foot trials. For the third type of trial subjects were asked to step onto the force platform with both feet. The particular combination of trials was dictated by the data needs of the GI study. The processing and use of these data will not be included here.

Data Processing

The GRF and EMG data were sampled at 1200 Hz and kinematic data were captured at 60 Hz using a 6 camera 3D Optical Capture system (Peak Performance Technologies). All data were synchronized in the Peak Motus video analysis system. Marker position data were smoothed using a quintic spline interpolation technique (Woltring, 1986) (implemented by Peak Performance Technologies). Major gait events (forward walking: strike foot (SF) heel strike (HS), contralateral foot (CF) toe-off (TO), CFHS, SFTO, and SFHS; backward walking: SF toe strike (TS), CF heel off (HO), CFTS, SFHO, SFTS) were visually identified and marked. GRF data were filtered using a 4^{th} order zero-lag Butterworth filter (25 Hz). Based on the quality of the kinematic data and the stance duration criterion, at least five trials were acceptable for further processing for each subject at each grade (Tables 7 and 8 give the specific number of trials used).

Kinematic, GRF, and EMG data from all acceptable trials were exported to custom Labview (National Instruments) software for further processing. Joint angles were calculated with the neutral standing position (leg axis perpendicular to foot) representing 0˚ for each joint; flexion/plantarflexion beyond neutral were positive angles, and extension/dorsiflexion beyond neutral were negative angles (See Figure 4). Joint kinetics (moments and powers) were calculated using standard inverse dynamics (See Background). Body segment parameters (Vaughan et al., 1999) and the support moment (Winter, 1980) were calculated as previously described. All kinematic and kinetic data were time-normalized to 300 points over the stride (200 for stance, 100 for swing) and ensemble averaged across all participants.

Table 7. Number of Forward Walking Trials Used in Processing

Participant	- 39%	- 15%	0%	+ 15%	+ 39%
1	7	5	8	6	5
2	5	6	6	5	5
3	5	8	6	8	5
4	8	8	8	6	5
5	6	7	6	6	7
6	8	7	8	6	7
7	7	6	8	7	8
8	6	5	8	5	7
9	6	7	8	8	5
TOTAL	**58**	**59**	**66**	**57**	**54**

Table 8. Number of Backward Walking Trials Used in Processing

Participant	- 39%	- 15%	0%	+ 15%	+ 39%
1	6	8	8	7	5
2	5	6	7	6	6
3	7	5	7	7	5
4	5	8	6	6	5
5	6	5	7	6	6
6	6	4	6	5	6
7	5	5	7	6	6
8	7	7	5	8	7
9	5	7	5	8	8
TOTAL	**52**	**55**	**58**	**59**	**54**

Stride duration, stride length (normalized to leg length), and stance duration were calculated for each trial, averaged within subjects for each grade, and compared across grades using a repeated measures analysis of variance design (ANOVA). Points of interest (POI) were identified on the joint angle, GRF, and joint/support moment curves for statistical analysis (Forward Walking, Table 9; Backward Walking, Table 10). The values for each point of interest were taken from all trials and averaged within subjects and grades. These dependent variables were submitted to separate repeated measures ANOVAs to assess the effect of grade. When a significant main effect for grade was identified for any variable, Bonferroni confidence interval adjustments were used during follow-up analysis. Post-hoc comparisons were focused between all slope conditions and level, and within the upslope and downslope conditions, but not between upslope and downslope. All statistical tests were performed using SPSS 11.0 for Windows (unless otherwise noted, $\alpha=0.05$, a priori).

EMG data were bandpass filtered (cutoff frequencies 30 and 500 Hz), rectified, and wavelet filtered in the region of interest, from heel strike to toe off of the stride on the force plate for forward walking (toe strike to heel off for backward walking). An average quiet period activity level was calculated for each trial. EMG bursts were identified when the activity level exceeded the quiet period average by at least 2 standard deviations for at least 40ms. Each trial was visually inspected and the program parameters were adjusted to ensure accurate burst identification. Burst onset, duration, and offset (as % of stride) as well as mean activity (in volts) were calculated for each burst in every trial. For each subject, the average mean activity for the upslope and downslope grades was then expressed as a percentage of the average mean activity for the

Table 9. Forward Walking Points of Interest used for Statistical Analysis

Joint Angles	Ankle	AA1	Angle at HS
		AA2	Maximum stance plantarflexion
		AA3	Maximum swing plantarflexion
	Knee	KA1	Angle at HS
		KA2	Angle at 75% stance (150 on normalized scale)
		KA3	Maximum swing flexion
	Hip	HA1	Angle at HS
		HA2	Minimum stance flexion/Maximum stance extension
		HA3	Maximum swing flexion
GRF	AP	AP1	Maximum anterior (-) force in early stance
		AP2	Maximum posterior (+) force in late stance
	Normal	N1	Maximum force in early stance
		N2	Maximum force in late stance
	Resultant	FR1	Maximum force in early stance
		FR2	Maximum force in late stance
Joint Moments	Ankle	AM1	Minimum plantarflexor moment in early stance
		AM2	Maximum plantarflexor moment in late stance
	Knee	KM1	Maximum flexor moment in early stance
		KM2	Maximum extensor moment in early mid-stance
		KM3	Value at 50% stance (100 on normalized scale)
		KM4	Maximum flexor/extensor moment in late stance
	Hip	HM1	Value of second peak in early stance
		HM2	Point of zero crossing in mid-stance
		HM3	Peak extensor moment in late stance
	Support	SM1	Peak value in early stance
		SM2	Peak value in late stance

Note: These Points of Interest (POIs) were used to calculate statistical differences between the different walking grades for the forward walking trials.

Table 10. Backward Walking Points of Interest used for Statistical Analysis

Joint Angles	Ankle	AA1	Max stance plantarflexion
		AA2	Angle at HO
		AA3	Max swing plantarflexion
	Knee	KA1	Angle at 25% stance (50 on normalized scale)
		KA2	Angle at HO
		KA3	Max swing flexion
	Hip	HA1	Min stance flexion/Max stance extension
		HA2	Angle at HO
		HA3	Max swing flexion
GRF	AP	AP1	Maximum posterior (+) force in early stance
		AP2	Maximum anterior (-) force in late stance
	Normal	N1	Maximum force in early stance
		N2	Maximum force in late stance
	Resultant	FR1	Maximum force in early stance
		FR2	Maximum force in late stance
Joint Moments	Ankle	AM1	Maximum plantarflexor moment in early stance
		AM2	Minimum plantarflexor moment in late stance
	Knee	KM1	Peak value in early mid-stance (~25% stance)
		KM2	Value at 50% stance (100 on normalized scale)
		KM3	Maximum extensor moment in late mid-stance
		KM4	Maximum flexor moment in late stance
	Hip	HM1	Peak value in early stance (~10% stance)
		HM2	Point of zero crossing in mid-stance
		HM3	Peak extensor moment in late stance
	Support	SM1	Peak value in early stance
		SM2	Peak value in late stance

Note: These Points of Interest (POIs) were used to calculate statistical differences between the different walking grades for the backward walking trials.

0% condition. The dependent variables were submitted to repeated measures ANOVAs to assess the effect of grade (α=0.05, a priori). When a significant main effect for grade was identified dependent t-tests with an adjusted p-value ($p = 0.01667$) were used in follow-up analyses. Post-hoc comparisons were focused between all slope conditions and level, and within the upslope and downslope conditions. All statistical tests were performed using SPSS 11.0 for Windows.

These group mean EMG data are presented with actual filtered rectified EMG data from a representative trial for each muscle at each walking grade. The EMG data is plotted against '% Stride', with the toe-off time for that particular trial marked. The group mean data for each burst are presented as a horizontal line that begins at the group mean burst onset, ends at the group mean burst offset (%Stride), and whose magnitude represents the group mean activity level for that burst. The values for the group mean data are also given on each graph (See Figure 17 for an example). The limitation of this presentation method is that the horizontal lines show the group average values, but not the distribution of muscle activity during the bursts. The actual EMG data is intended to be "representative" of this distribution, but in reality is from a single trial from one subject. To provide the most complete picture of the muscle activity an ensemble average of the EMG activity during the stance phase was calculated as follows. For each subject, the mean activity level of the major stance phase burst of all eight muscles was calculated for each level walking trial, and the average mean activity level for each muscle was then calculated. The raw EMG data from each trial were bandpass filtered (30-500 Hz), wavelet filtered, rectified, and smoothed using a moving window average of 39 ms (Leroux et al., 1999). The stance phase of the smoothed data was then normalized to 1000 points, and the magnitude of the smoothed data was normalized to the previously calculated average mean activity level for that muscle during level walking. Once the smoothed normalized data had been calculated for all trials for each subject, the data were ensemble averaged across all subjects for each walking grade. The result was a single curve for each muscle at each walking grade representing the group average muscle activity. These curves are presented in Appendices A, B, and C.

Correlations were used to assess the relationship between joint moments and EMG data. The extensor moment duration and extensor muscle burst duration were plotted against one another for corresponding muscle-joint pairs: GM, BF, and SM – hip, RF and VM – knee, MG and Sol – ankle. The peak extensor moment and extensor muscle burst peak magnitude were also plotted against one another. For each subject, the peak burst magnitude for a slope walking trial was normalized to the value during the level walking trials so the data could averaged across subjects. The average peak moment for each walking grade was normalized to the average peak moment during level walking so the correlation was between two normalized values. Linear regressions were performed based on the data from all lower limb joints and also based on the joint-specific data sets.

Gait Cycle Parameters

The average gait cycle parameters are given in Tables 11 and 12. No significant grade effect was observed for the stance duration for either walking direction (indicated by gray shading in the table), which confirmed that walking speed was controlled during testing. This result was unexpected for backward walking considering that stance duration was not monitored during those trials. For forward walking, a significant grade effect was observed for the stride duration and normalized stride length (indicated by the lack of gray shading in the table). However, no significant differences were observed between level walking and any grade, or within the upslope or downslope conditions for these variables. In other words, the significant differences occurred between the upslope and downslope grades; because these differences were not the focus of the statistical analysis they are not indicated in the table. For backward walking a significant grade effect was observed only for the normalized stride length, and the results of the follow-up comparisons are indicated by the following symbols: **§** – denotes a significant difference from 0% grade; **†** – denotes a significant difference between 15% and 39% in the same walking direction (upslope or downslope). These conventions (gray shading, no shading, symbols indicating significant differences) will be used in all tables in this project.

Table 11. Forward Walking Gait Cycle Parameters

	- 39%	- 15%	0%	+ 15%	+ 39%
Stance Duration (s)	0.74 (0.05)	0.73 (0.05)	0.74 (0.05)	0.75 (0.05)	0.75 (0.05)
Stride Duration (s)	1.18 (0.07)	1.17 (0.08)	1.20 (0.08)	1.23 (0.08)	1.20 (0.07)
Normalized Stride Length (m/m)	1.26 (0.13)	1.42 (0.16)	1.49 (0.16)	1.61 (0.15)	1.53 (0.18)

Note: Average (SD) of the gait cycle parameters of all trials at each ramp grade. Stance duration was monitored during testing and used as a *post-hoc* measure to ensure speed consistency between ramp grades. Gray shading indicates no significant grade effect was observed (ANOVA).

Table 12. Backward Walking Gait Cycle Parameters

	- 39%	- 15%	0%	+ 15%	+ 39%
Stance Duration (s)	0.86 (0.09)	0.83 (0.09)	0.83 (0.08)	0.81 (0.08)	0.84 (0.09)
Stride Duration (s)	1.36 (0.13)	1.35 (0.13)	1.35 (0.11)	1.31 (0.12)	1.30 (0.12)
Normalized Stride Length (m/m)	1.32 (0.29)	1.32 § (0.31)	1.21 (0.29)	1.16 (0.28)	0.96 §† (0.22)

Note: Average (SD) of the gait cycle parameters. Gray shading indicates no significant grade effect was observed (ANOVA). When a significant grade effect was observed, Bonferonni confidence interval adjustments were used in the follow-up comparisons. The results are indicated as follows: § – a significant difference from 0% grade; † – a significant difference between 15% and 39% in the same walking direction (upslope or downslope).

CHAPTER 5

FORWARD DOWNSLOPE WALKING[2]

Introduction

This chapter will discuss the results from the **first aim** of this project, the investigation of forward downslope walking. From the limited data reported in the literature, it was expected that both the temporal pattern and peak magnitude of the joint moments would change during downslope walking (Redfern and DiPasquale, 1997). Because the joint moments arise from the muscle activity (as is discussed in Chapter 2), it was expected that the pattern and magnitude of muscle activity would also change. Specifically, it was *hypothesized that changes in the pattern and magnitude of the muscle activity would be directly related to the changes in the pattern and magnitude of the joint moments during downslope walking*. This would be tested using correlations of the moment and muscle activity duration and magnitude. Based on the joint angles and moments presented by Redfern et al. (1997), it was also expected that power absorption would increase at all joints during downslope walking (Redfern and DiPasquale, 1997), and therefore that the relative amount of eccentric muscle activity would increase. This finding would suggest that length-dependent sensory feedback from the muscle spindles would increase. Because Abelew et al. (2000) showed that such length feedback was necessary for quadrupeds to achieve normal kinematics during downslope walking, but not in level or upslope walking, it was thought that this increased feedback might play a

[2]Some text and figures in this chapter are adapted from the article "The effects of sloped surfaces on locomotion: A kinematic and kinetic analysis" published in the Journal of Biomechanics (Lay et al., 2005a), with permission from Elsevier, Inc.

role in the control strategy for human downslope walking. Although the increased length feedback and its role in changing the control strategy could not be implicitly tested in this project its potential impact was considered. In order to test the aforementioned hypothesis and expectations, joint moment magnitudes and patterns were quantified during level and downslope walking at -15% and -39%. The muscle firing patterns and muscle activity levels that occur during level and downslope walking were also characterized. The downslope walking data were compared to level walking data to define the changes that occur when a downslope walking grade is introduced. Once these data were collected and analyzed (using the methods outlined in Chapter 4) the correlation between joint moments and muscle activity was assessed.

Results

Joint Kinematics. Downslope joint kinematic data are presented in Figure 11 and Table 13. A significant grade effect was observed for all variables (ANOVA). Results from the follow-up comparisons are indicated with symbols in Table 13. The downslope walking ankle joint angles (Figure 11A) had a similar pattern to the level walking joint angles, but the peak magnitudes changed: the early stance peak plantarflexion angle (AA2) increased progressively and the swing phase peak plantarflexion angle (AA3) decreased progressively with the decreasing walking grade. At the steepest grade of downslope walking the knee flexion angle at heel strike increased from that during level walking. In general during downslope walking greater stance phase flexion (KA2) was necessary to lower the body down the slope, and continued into early swing (KA3) to

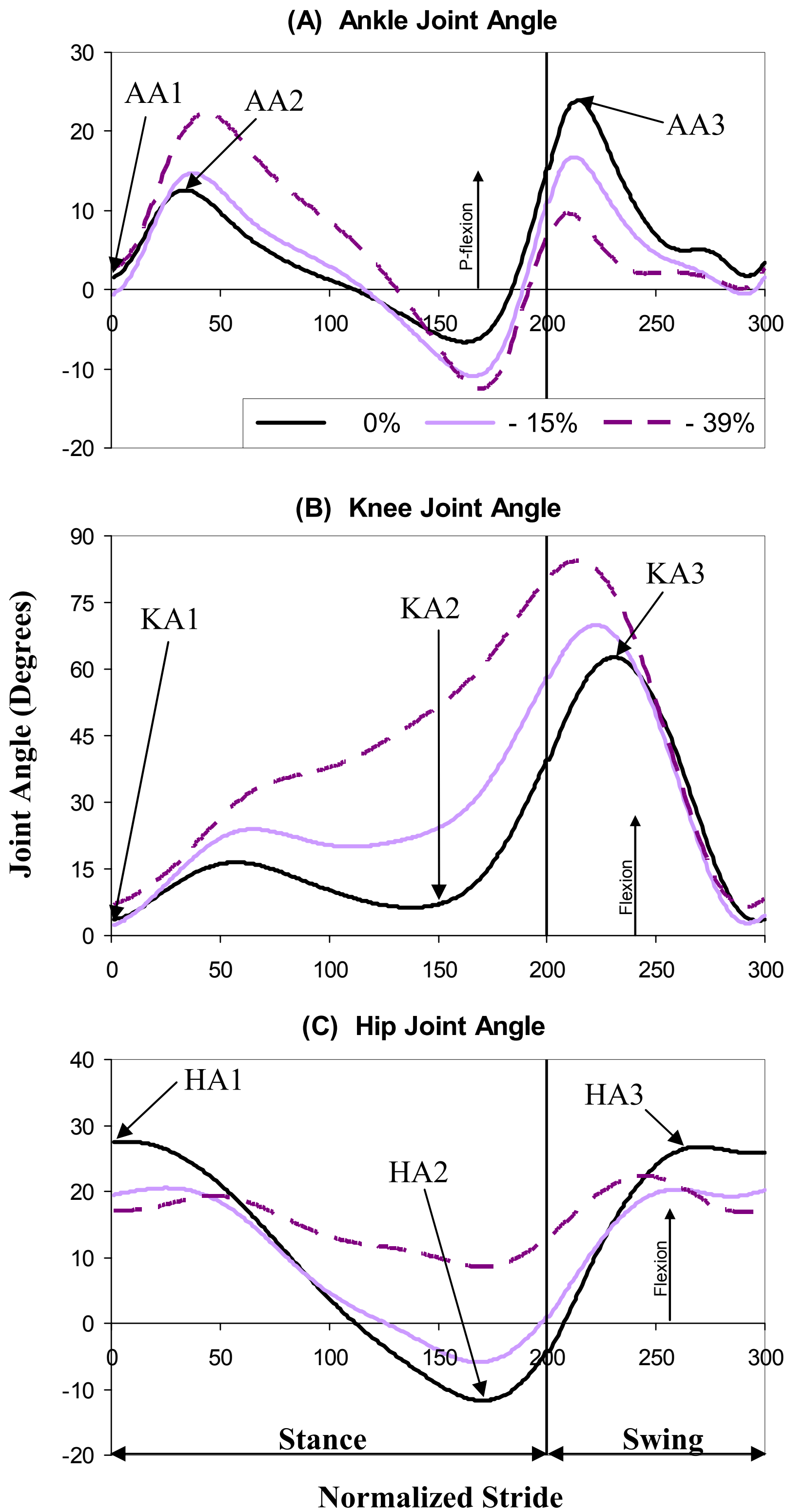


Figure 11. Average Joint Angles during Downslope Walking. Joint angle convention is shown in Figure 4. Stance phase was normalized to 200 points and swing phase to 100 points before ensemble averaging. Vertical line marks toe-off.

help ensure foot clearance. The knee joint angle for the second half of swing was identical to that during level walking. The hip flexion angle decreased uniformly from level in early stance and late swing (HA1, HA3), but increased progressively during midstance (HA2) as the walking grade decreased. At -15% the hip angle pattern was similar to the pattern observed during level walking, but at -39% these changes resulted in a relatively constant joint angle and therefore a much smaller hip joint range of motion.

Table 13. Downslope Walking Joint Angle POI Data

	- 39%	- 15%	0%
AA1	2.58 † (5.40)	-0.56 § (4.21)	1.51 (4.51)
AA2	22.68 §† (4.10)	15.02 § (4.18)	12.75 (4.31)
AA3	10.49 §† (5.15)	17.37 § (5.85)	24.23 (6.38)
KA1	7.05 §† (5.32)	2.32 (5.65)	3.56 (4.60)
KA2	51.54 §† (4.68)	24.22 § (6.64)	7.01 (4.70)
KA3	84.60 §† (3.40)	70.21 § (3.09)	62.82 (3.88)
HA1	16.94 § (5.92)	19.49 § (5.90)	27.56 (6.16)
HA2	8.01 §† (6.35)	-6.20 § (7.47)	-11.83 (6.83)
HA3	22.76 § (5.29)	21.54 § (5.46)	27.43 (5.40)

Note: Mean (SD) of the joint kinematic POI (all values in degrees), averaged across all subjects for each grade. § - indicates a significant difference from 0% grade ($p<0.05$), † - indicates a significant difference between 15% and 39% in the same walking direction (upslope or downslope) ($p<0.05$). For details about the POI see Table 9 or Figure 11.

Ground Reaction Forces. There was a significant grade effect for all downslope GRF points of interest (Figures 12 and 13, Table 14). The AP braking force (Figure 12A, AP1) increased dramatically, while the propulsive force (AP2) decreased markedly from level to -15% grade and did not exist at -39% grade. Because the ML component (Figure 12B) is such a small contributor to the sagittal plane joint moments, which are of the greatest interest here, no points of interest were analyzed from these data. It appears from the curves that the laterally directed foot force increased during downslope walking. The first peak of the normal component (Figure 12C, N1) increased at -15% walking grade, but showed no change from level at -39% walking grade. This may be because the steep slope of the ramp surface caused more of each subject's body weight to contribute to the shear (AP) component of the GRF, and therefore reduced the magnitude of the normal component. The second peak of the normal component (N2) decreased progressively as the walking grade decreased, again because of the decreased propulsive requirements during downslope walking. The sagittal plane resultant force (FR, Figure 13) is calculated as the vector sum of the AP and normal force components, and is an indication of the total force acting on the limb in the sagittal plane. FR was similar in shape to the normal force, but increased uniformly from level walking in early stance (FR1) and decreased uniformly in late stance (FR2).

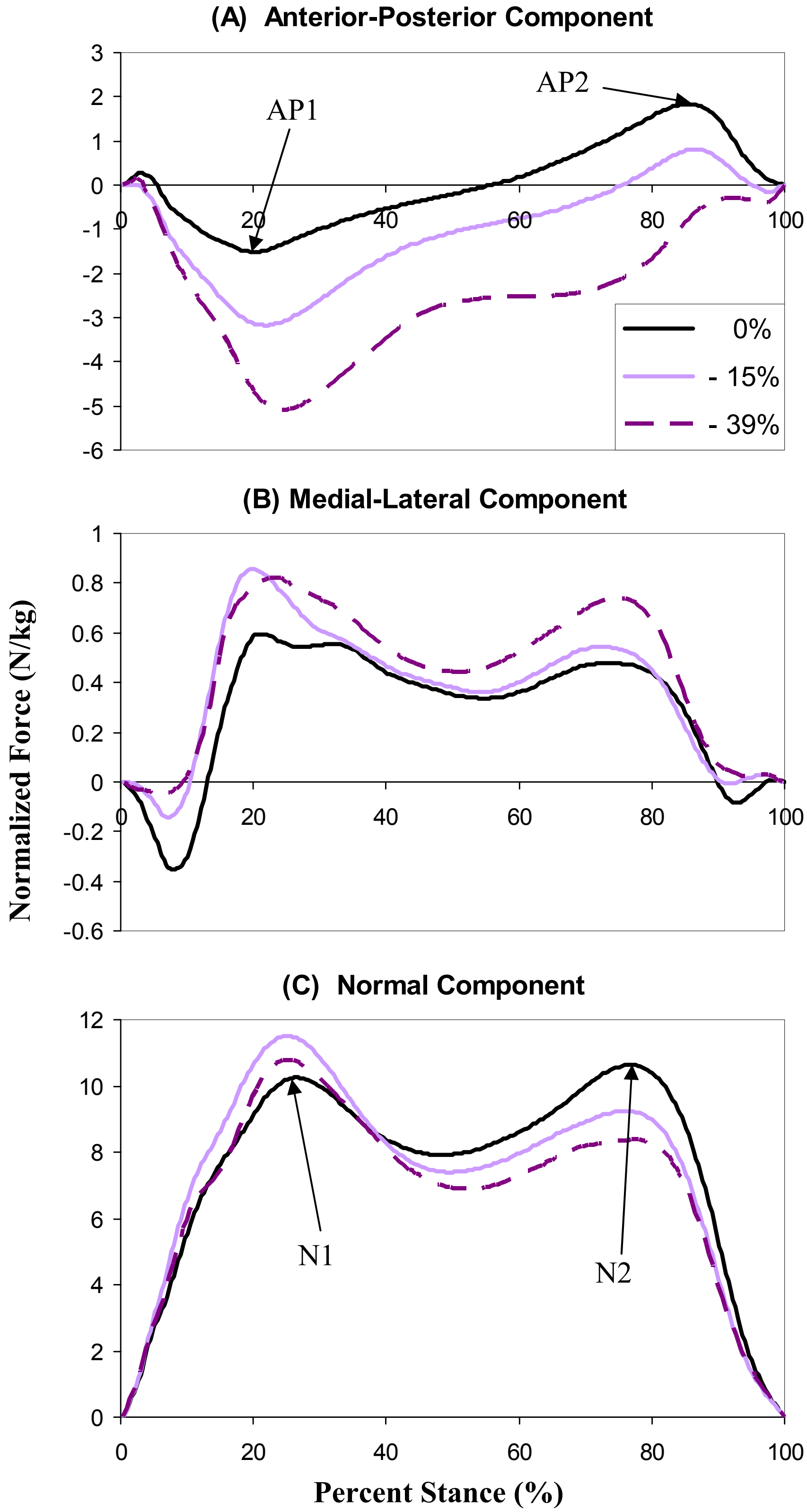


Figure 12. Average Ground Reaction Force Components during Downslope Walking. Positive forces indicate posteriorly and laterally directed foot forces.

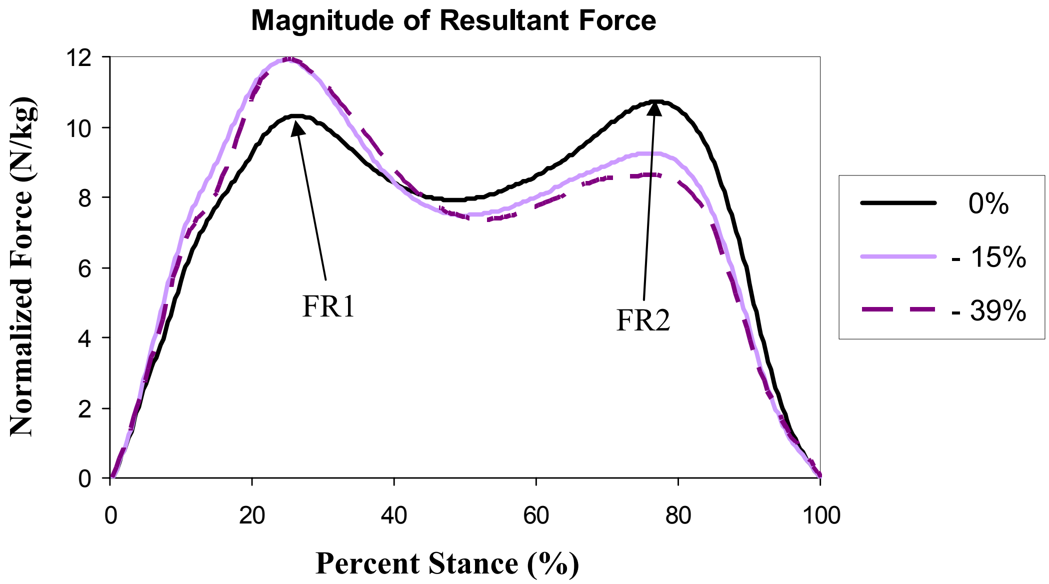


Figure 13. Average Resultant Force during Downslope Walking, calculated as the sum of the normal and AP component.

Table 14. Downslope Walking GRF Component POI Data

	- 39%	- 15%	0%
AP1	-5.35 §† (0.66)	-3.27 § (0.33)	-1.55 (0.14)
AP2	N/A	0.91 § (0.34)	1.85 (0.34)
N1	11.18 (0.90)	11.67 § (0.56)	10.37 (0.43)
N2	8.70 §† (0.56)	9.41 § (0.30)	10.71 (0.46)
FR1	12.39 § (1.09)	12.09 § (0.58)	10.45 (0.43)
FR2	8.95 § (0.58)	9.42 § (0.30)	10.79 (0.49)

Note: Mean (SD) of the GRF component POI (in N/kg). As defined, point AP2 did not exist for the -39% condition so no value is included in the table. § - a significant difference from 0% ($p<0.05$), † - a significant difference between 15% and 39% in the same walking direction (upslope or downslope) ($p<0.05$).

Joint Kinetics. During swing the joint moment patterns were generally similar for the three walking grades. In contrast, during stance phase the joint moments changed significantly with grade (Figures 14 and 15). Significant grade effects were observed for all variables (ANOVA). The mean and standard deviation of the POI are given and the results of follow-up comparisons are denoted with symbols in Table 15. The ankle joint (Figure 14A) peak dorsiflexor moment in early stance increased uniformly during downslope walking (AM1). The most noticeable change in the ankle moment was the progressive and significant decrease in the peak magnitude (AM2) as the ramp grade decreased from 0% to -39%. The knee moment pattern (Figure 14B) during downslope walking was similar to that of level walking in early stance, with only small changes in the peak flexor moment (KM1). However, for the remainder of stance the knee moment differed dramatically from the level walking pattern. The first peak extensor moment, KM2, increased progressively as the walking grade decreased, and then the moment remained extensor (KM3, KM4) rather than switching to a flexor moment. Finally, the peak hip extensor moments (Figure 14C) in early (HM1) and late (HM3) stance were similar to those during level walking. The time of the transition to a flexor moment in midstance (HM2) was significantly earlier during downslope walking, although there were no significant differences between the two downslope walking grades. The net extension moment of the lower extremity, the support moment (Figure 15), increased for most of stance with a large and progressive increase in late stance (SM2).

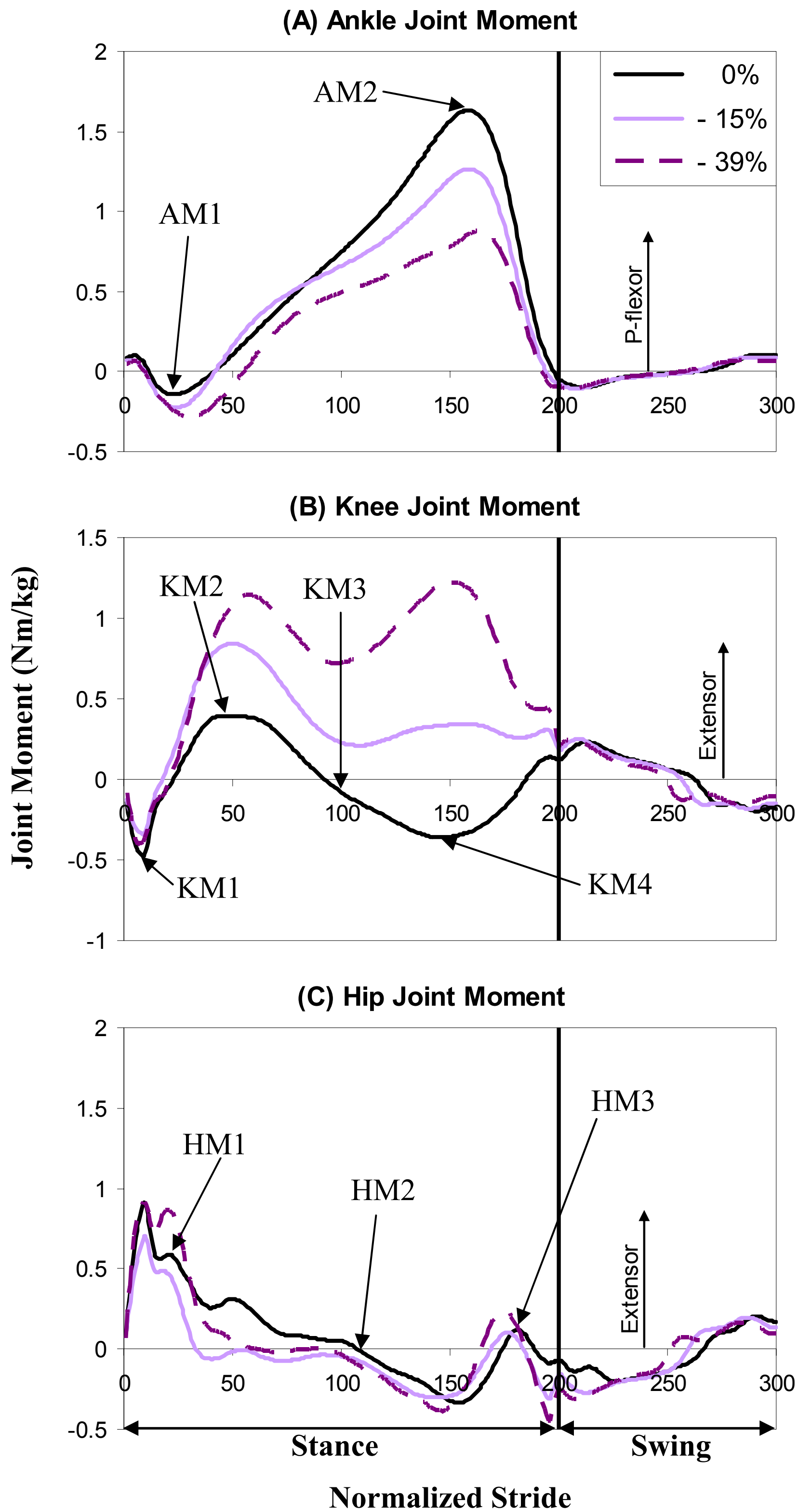


Figure 14. Average Joint Moments during Downslope Walking. Positive moments are plantarflexion and extension, as indicated on the figure. Stance phase for each trial was normalized before ensemble averaging. Vertical line marks toe off.

Table 15. Downslope Walking Joint Moment POI Data

	- 39%	- 15%	0%
AM1	-0.32 § (0.09)	-0.24 § (0.11)	-0.15 (0.06)
AM2	0.92 §† (0.10)	1.29 § (0.11)	1.64 (0.14)
KM1	-0.47 † (0.12)	-0.38 § (0.11)	-0.51 (0.09)
KM2	1.18 §† (0.27)	0.88 § (0.26)	0.44 (0.20)
KM3	0.72 §† (0.18)	0.23 § (0.20)	-0.09 (0.17)
KM4	1.25 §† (0.12)	0.36 § (0.26)	-0.38 (0.22)
HM1	0.75 † (0.35)	0.36 (0.32)	0.54 (0.22)
HM2	69.12 § (46.10)	60.63 § (44.95)	97.64 (42.76)
HM3	0.30 (0.23)	0.18 (0.14)	0.16 (0.15)
SM1	1.36 §† (0.31)	1.11 (0.33)	0.91 (0.23)
SM2	2.05 §† (0.25)	1.53 § (0.23)	1.14 (0.15)

Note: Mean (SD) of the joint moment POI (values N*m/kg, except HM2 is normalized stride). § - a significant difference from 0% grade ($p<0.05$), † - a significant difference between 15% and 39% in the same walking direction (upslope or downslope) ($p<0.05$).

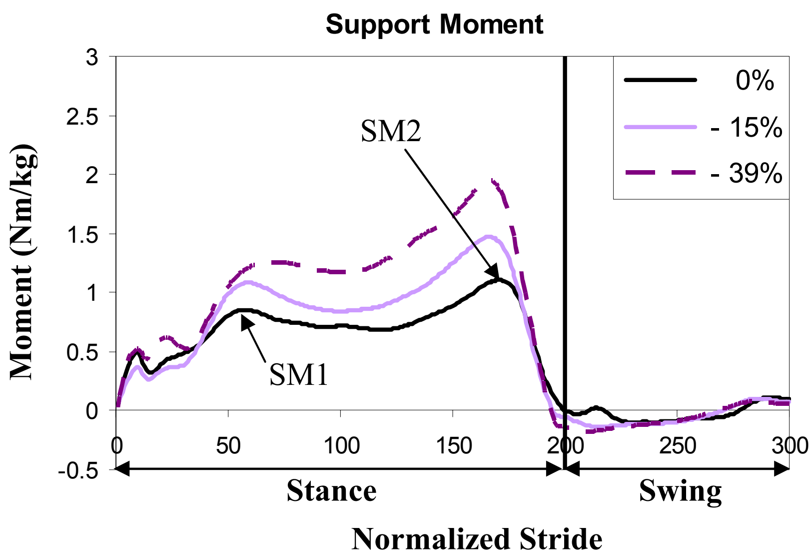


Figure 15. Average Support Moment during Downslope Walking (Winter, 1980).

Joint Power. The joint power curves (Figure 16) show that power generation decreased at all joints. In addition, the percent of stance phase during which power was absorbed increased progressively at all joints as the walking grade decreased, with the largest changes observed at the knee and hip (Table 16). At the knee power was absorbed for most of the stance phase, suggesting that eccentric muscle activity dominated during the lowering of the body over the stance foot during downslope walking.

Table 16. Average Percent of Stance Phase when Power is Absorbed

	- 39%	- 15%	0%
Ankle	82.44	79.51	74.63
Knee	87.32	68.29	45.85
Hip	73.66	80.49	40.49

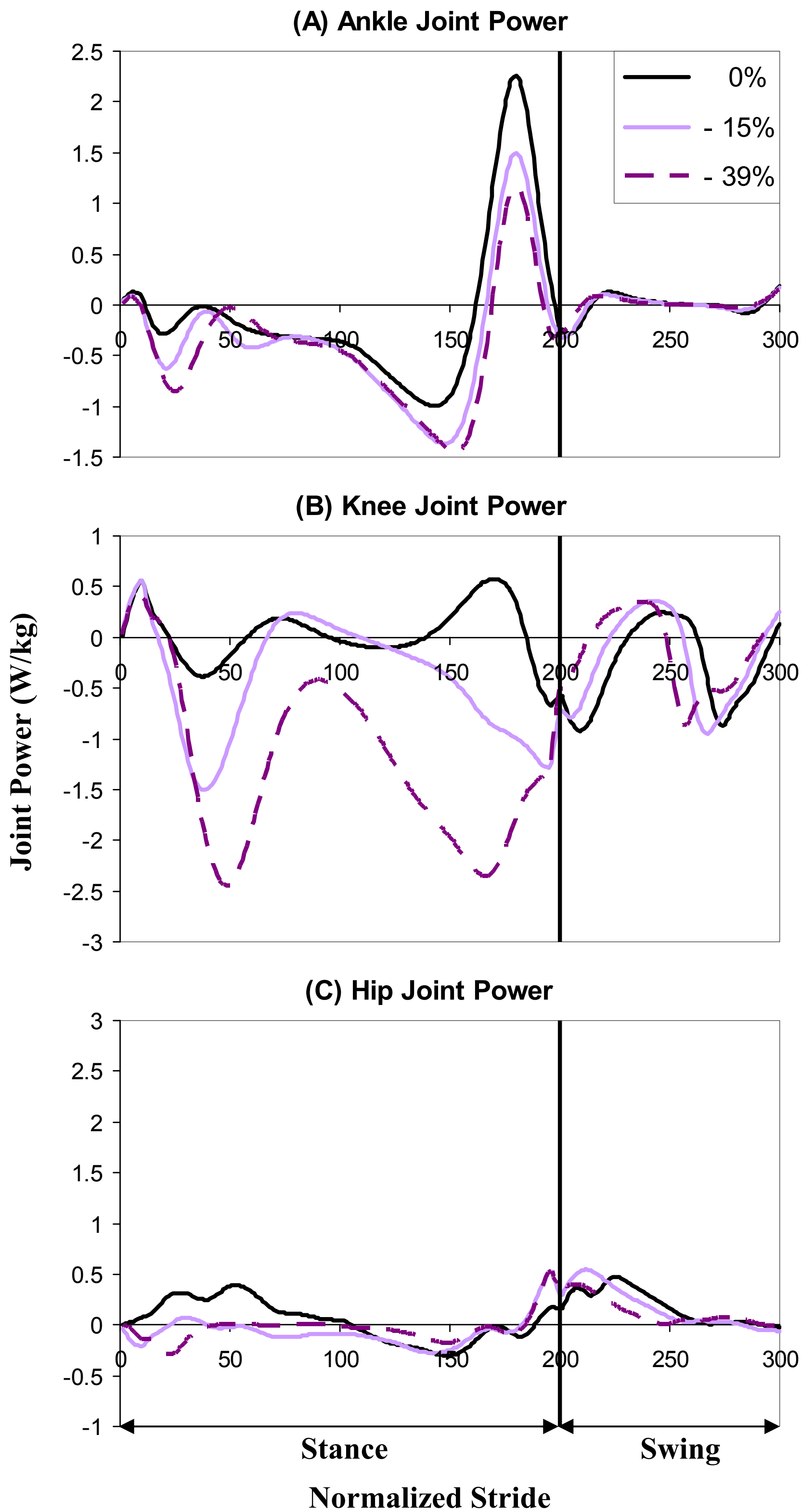


Figure 16. Average Joint Power during Downslope Walking. Positive power is generation, negative is absorption. Stance and swing phases of each trial were normalized before ensemble averaging. Vertical line marks toe off.

Muscle Activity. Exemplar and average EMG data are shown in Figures 17 (GM, BF, and SM), 18 (RF and VM), and 19 (Sol, MG, and TA). As in our discussion of the joint moments, the focus here will be on the muscle activity during stance. Ensemble average stance phase EMG data can be found in Appendix A. During downslope walking the mean activity of four of the eight muscles was not statistically different than that during level walking, and the MG showed a significant reduction in mean activity. Only the RF, VM, and TA showed increases in mean activity levels as the walking grade changed from 0% to -39%. In addition, only the RF, VM, Sol, and MG stance phase burst durations increased significantly as the walking grade changed; the burst durations of all other muscles were similar to those during level walking. In some trials the increased mean activity of the RF and VM was distributed uniformly across the longer stance phase bursts at -15% and -39%. In other trials the increased activity appeared as two approximately equal peaks in the longer bursts (see Figure 18). The group average curve showed the latter form; the increased mean activity was distributed as two approximately equal peaks in the longer stance phase burst (Appendix A, Figure 68). During level walking the activity of the Sol and MG peaked in late stance (Figure 19; Appendix A, Figure 69). During downslope walking the mean activity was distributed more uniformly across the increased stance phase burst durations such that there was no appreciable late stance phase peak in the Sol or MG activity (Appendix A, Figure 69).

There were some deviations from the normal bursting patterns of several muscles. For example, three subjects did not exhibit any distinguishable bursts of GM activity during downslope walking. All three subjects exhibited identifiable bursts during upslope walking (Chapter 7), which precludes the possibility of instrumentation errors. A majority of subjects exhibited a late stance phase BF burst during downslope walking that was not present during level walking (the mean activity levels could not be normalized to the mean activity at 0%, and so are reported in volts in Figure 17). In addition, a minority of subjects (n=3 at -15%, n=4 at -39%) exhibited a similar late stance burst in the SM during downslope; all of these subjects also exhibited the late stance BF burst. These hamstring bursts are likely needed to counteract the late stance knee extensor activity in downslope walking to prepare the leg for swing.

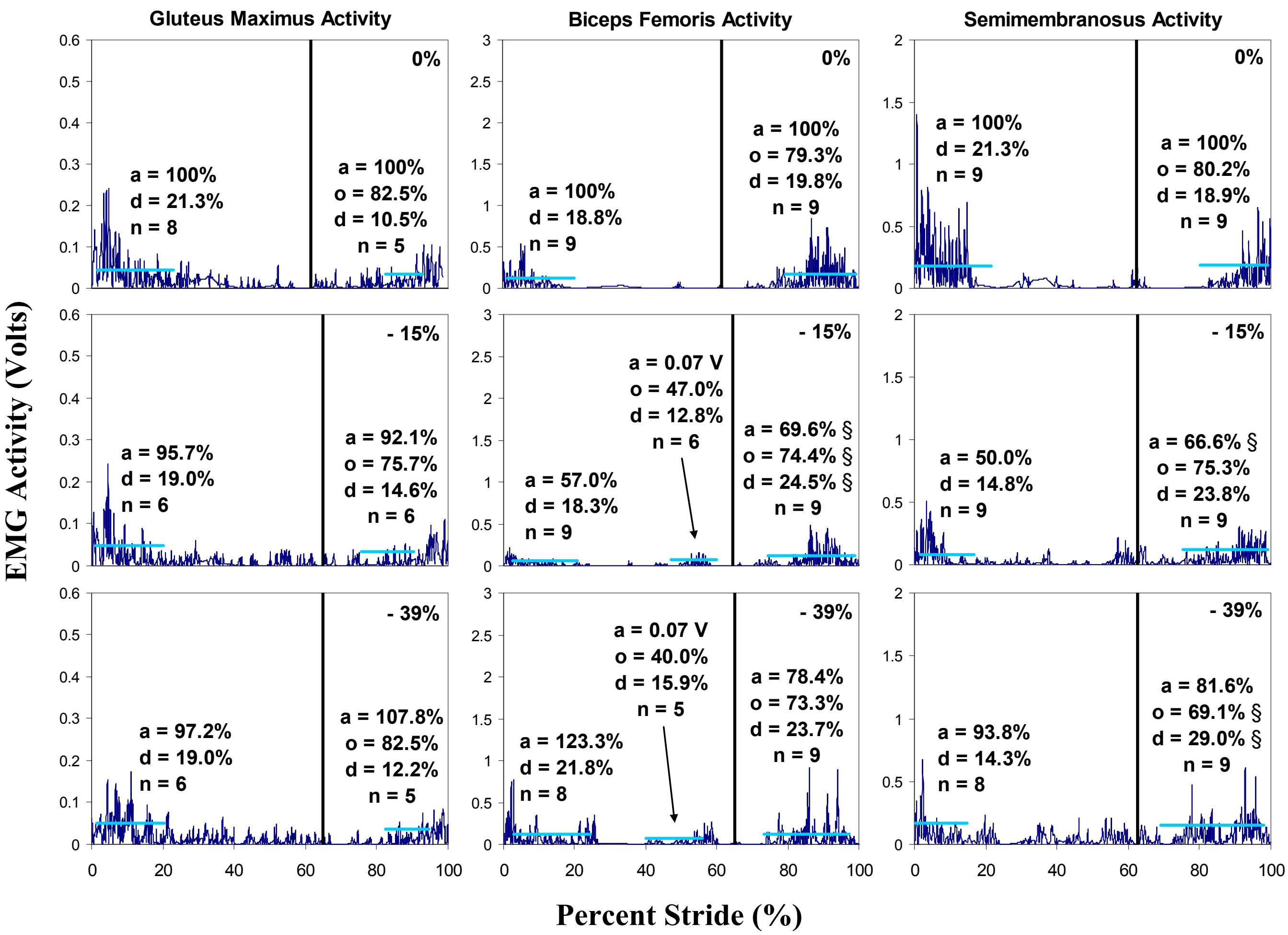


Figure 17. Representative GM, BF, and SM EMG Data during Downslope Walking. For each muscle a representative trial was chosen for each grade from a single participant's data to demonstrate the typical activity. Each trial was normalized to its stride time and plotted in %stride, with heel strike occurring at 0%. The toe-off time for each trial was also normalized to %stride and is indicated with a vertical line (average over all 24 representative trials for all eight muscles = 62.5%). Data from seven participants were used to represent the eight muscles. Horizontal lines representing the average burst data (mean for all subjects exhibiting the burst) are overlaid on each plot, and the values are also displayed on the plots: a = the group average mean activity (as a percentage of mean activity at 0% grade), d = the group average burst duration (in %stride), and n = the number of subjects exhibiting the burst. For bursts that do not start at heel strike o = the onset time (in %stride). These variables were submitted to repeated measures ANOVAs to assess the effect of grade. When a significant grade effect was observed, dependent t-tests were used for follow-up comparisons (adjusted p value of 0.01667). All significant differences are indicated on the figure by the following symbols: § - significantly different than the value at 0% grade ($p < 0.01667$) and † - significantly different than the value at -15% ($p < 0.01667$). See Appendix A for the group ensemble average curves.

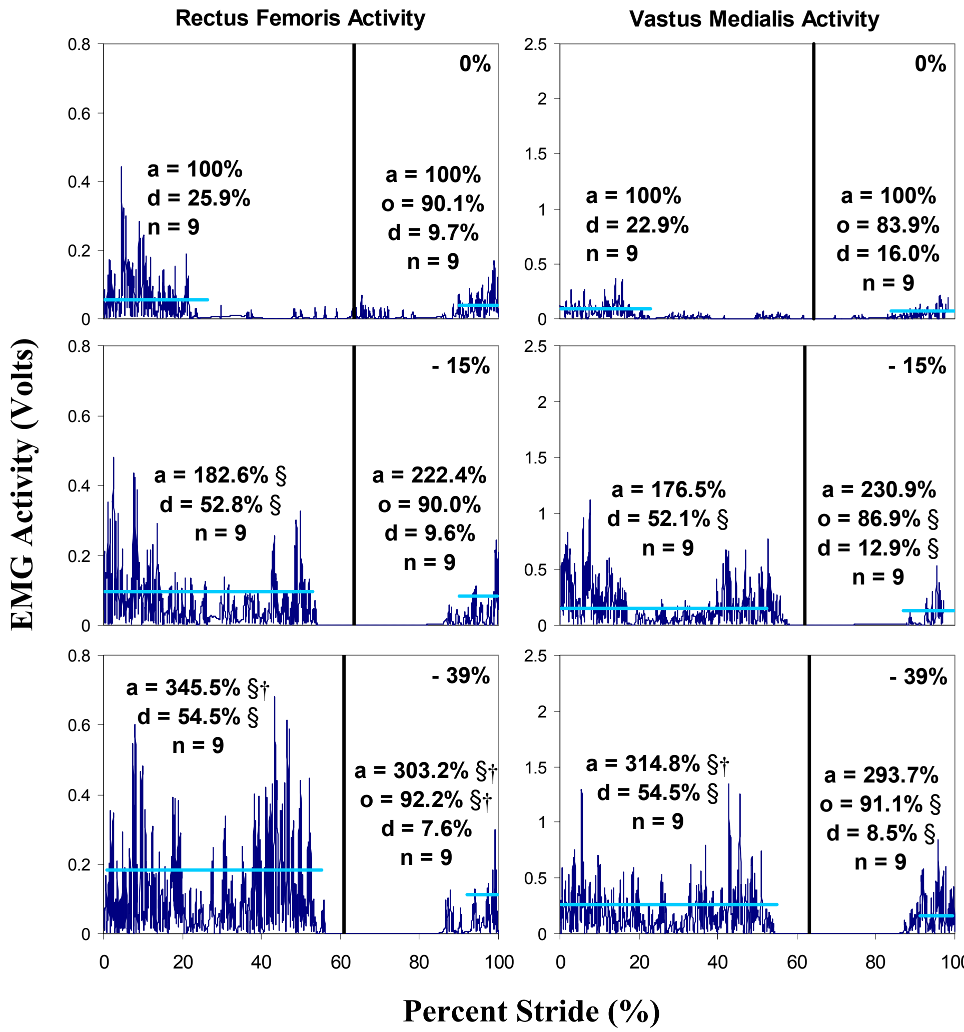


Figure 18. Representative RF and VM EMG Data during Downslope Walking. Details are the identical to Figure 17. All significant differences are indicated on the figure by the following symbols: § - significantly different than the value at 0% grade (p < 0.01667) and † - significantly different than the value at -15% (p < 0.01667).

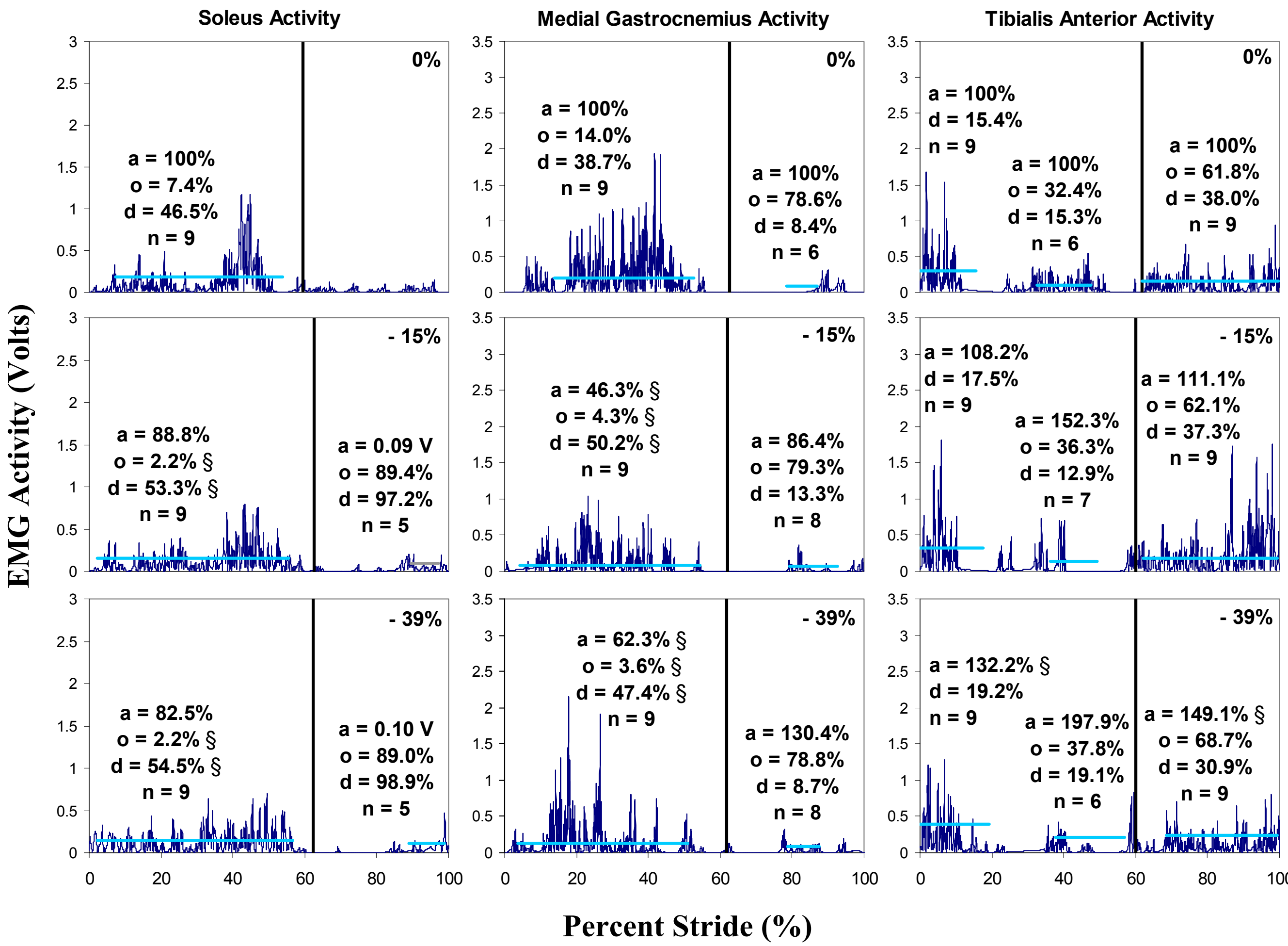


Figure 19. Representative Sol, MG, and TA EMG Data during Downslope Walking. Details are the identical to Figure 17. All significant differences are indicated on the figure by the following symbols: § - significantly different than the value at 0% grade ($p < 0.01667$) and † - significantly different than the value at -15% ($p < 0.01667$).

Correlations between Muscle Activity and Kinetics. In order to asses the relationship between joint moments and EMG, the durations and peak magnitudes of the extensor joint moments and the extensor muscle bursts were correlated (Figures 20 and 21). In both cases the linear regression showed the same trend: good overall correlation for the lower limb, with the knee showing the best joint-specific correlation. For the moment and burst durations the overall limb correlation (Figure 20A) had an R-value of 0.904, while the knee correlation (Figure 20B) value was 0.994 (r^2 values are shown in the figures). The R-values from the hip joint (0.509) and ankle joint (0.445) correlations (Figure 20B) were not as good because there was almost no change in the muscle burst duration and the extensor moment burst duration, respectively. For the moment and muscle burst peaks (Figure 21) the overall and knee joint correlations were also high, with R-values of 0.933 and 0.989, respectively. Again, the hip joint (R = 0.558) and ankle joint (R = 0.327) correlations were not as good, in this case because there was almost no change in the muscle burst peak magnitude at either joint.

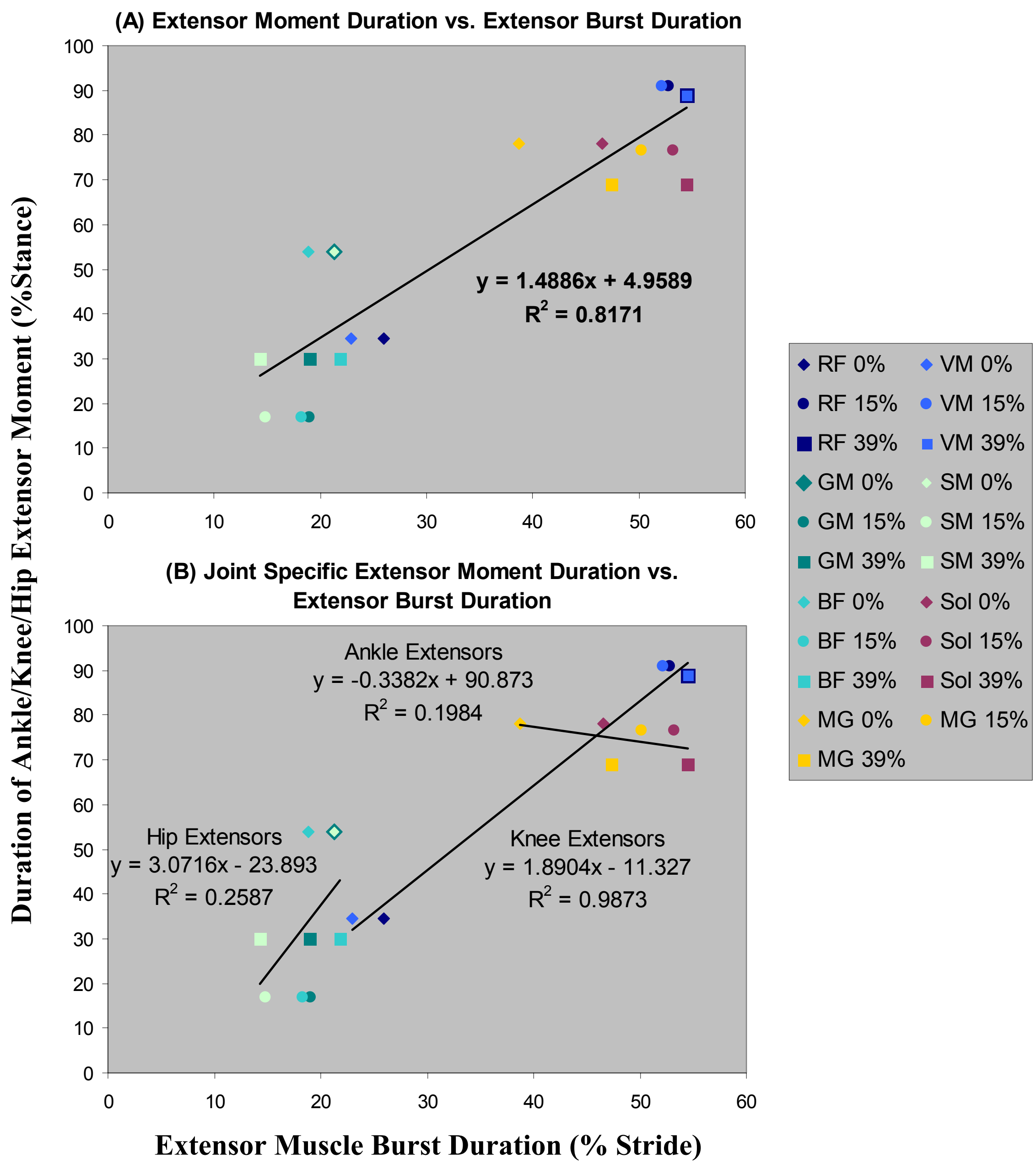


Figure 20. Correlations of Extensor Moment Durations and Extensor Muscle Burst Duration during Downslope Walking for lower limb extensor muscles and the joints at which they act. ♦ – data from level walking trials, ● – data from -15% trials, ■ – data from -39% trials. R^2 values are given for (A) overall lower limb correlation and (B) joint-specific correlations.

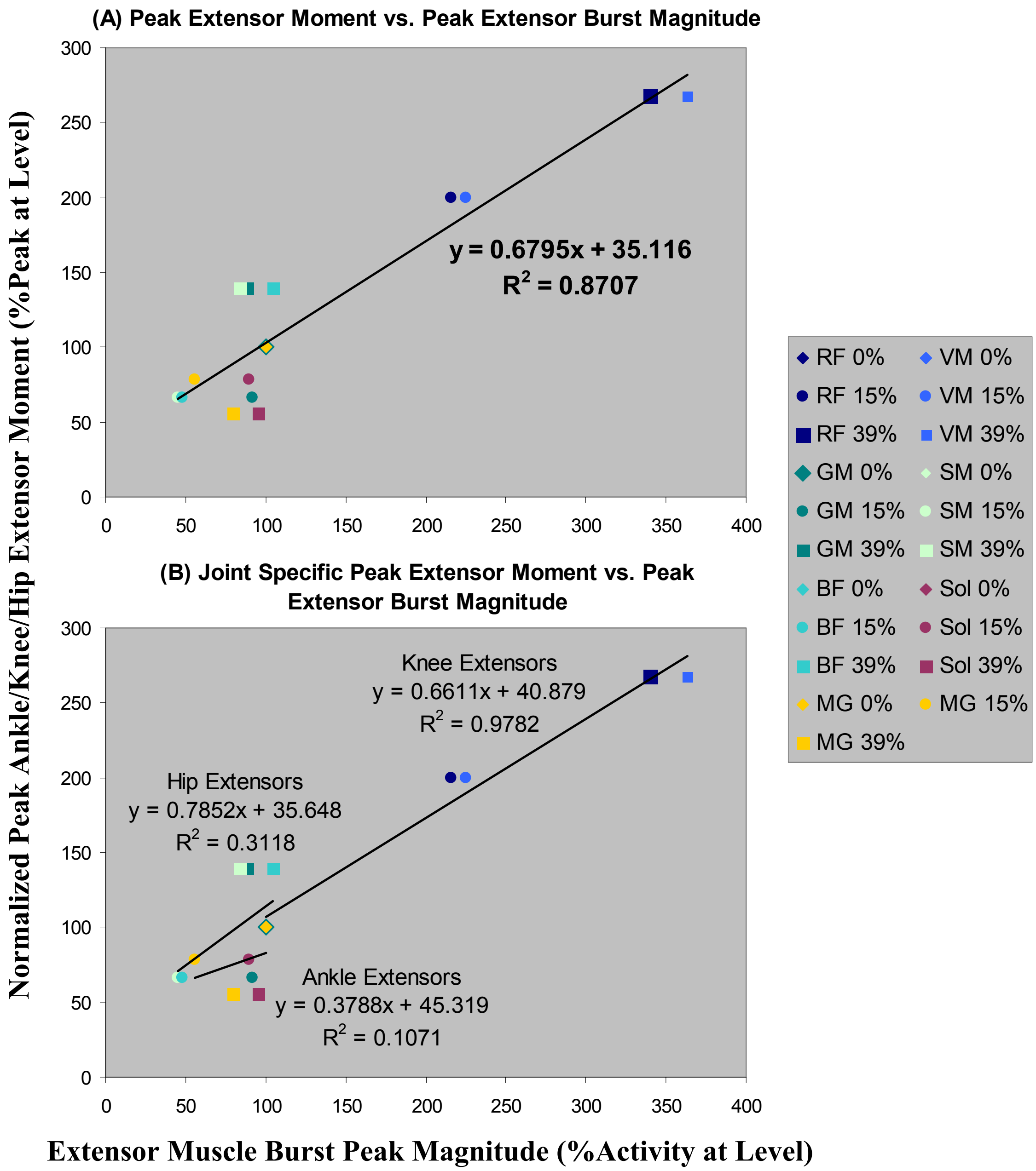


Figure 21. Correlations of Extensor Moment Peaks and Extensor Muscle Burst Peak Magnitudes during Downslope Walking for lower limb extensor muscles and the joints at which they act. The muscle burst peak magnitudes were normalized to the peak magnitude of the burst during level walking. The extensor moment peaks were also normalized to the peak extensor moment during level walking. Other details are as in Figure 20.

Discussion

Joint Kinematics. Joint kinematic patterns show good agreement with previous reports on human slope walking (Davis et al., 1995; Kuster et al., 1995; Redfern and DiPasquale, 1997; Wall et al., 1981). The changes in the angles from level walking correspond to the need for a controlled descent of the body and suggest that the single joint extensors may be lengthening during stance, absorbing energy to create a smooth descent. The relevance of these kinematic data therefore lies in their association with muscle length changes, and the corresponding proprioceptive feedback to the control system from the muscle spindle afferents. While muscle length changes are not investigated here, this idea warrants further study.

Ground Reaction Forces. The ground reaction forces are also generally similar to data reported in the literature (McVay and Redfern, 1994; Redfern and DiPasquale, 1997), particularly the increasing braking force of the AP component and the decreasing peak of the normal component in late stance. The FR peaks indicate that during downslope walking shock absorption increases in early stance and propulsion decreases in late stance compared to level walking. The uniform, rather than progressive changes in FR indicate that the overall change in the GRF is similar for both downslope walking grades; however, the force is distributed differently between the AP and normal force components due to the walking grade. Changes in the magnitudes of the ground reaction force components have implications for the output of cutaneous and muscle load sensors; output of these mechanoreceptors may increase in early stance due to the high loads and decrease in late stance of downslope walking when the loads are reduced.

Joint Kinetics. The downslope walking kinetics reported herein are consistent with the limited data in the literature for both humans (Cham and Redfern, 2002; Kuster et al., 1995; Redfern and DiPasquale, 1997) and quadrupeds (Gregor et al., 2001). These studies all report a decrease in the magnitude of the peak plantarflexor moment and a knee joint moment that is extensor for the majority of stance. Human data also show an earlier switch to a hip flexor moment in stance than during level walking (Redfern and DiPasquale, 1997), as was reported here.

The support moment is indicative of the integrated response of the lower extremity muscles to an environmental demand (Winter, 1980). Here, increases in the support moment are not shared equally among the joints: the knee extensor moment increases drastically during downslope walking while the ankle and hip extensor moments show small decreases. In addition, changes in the joint moment patterns indicate that the increased moment demands during downslope walking are not scalar multiples of the demands during level walking. For example, the hip joint moment exhibits an earlier transition to a flexor moment, and the knee joint moment is extensor rather than flexor during late stance in downslope walking.

Joint Power. A study of the energy cost during slope walking suggests that walking at downslope grades greater than 15% is achieved through “pure” eccentric muscle activity (Minetti et al., 2002). Although the joint power data presented here (Figure 16) show that muscle activity is not entirely eccentric during downslope walking at 15% or 39% (there is actually some power generation at the ankle), eccentric activity certainly predominates. In fact, the power absorption by the single joint knee extensors here is similar to that of the vastus lateralis in the rat model, which has been shown to lengthen for all of stance during downslope walking (Gillis and Biewener, 2002).

Knee Joint Biomechanics. The changes in the knee joint moment and power are explained in Figure 22. The stick figures indicate exemplar limb positions, GRF vectors (straight arrows), and a first approximation of the resultant applied moments at the knee joint (curved arrows, with the thickness representing the magnitude) at 75% of stance for level and walking at -39% grade. The knee is more flexed during downslope walking and the GRF vector is directed posterior to the joint center. As a first approximation, this would cause a large flexor moment to be applied at the knee joint, as shown requiring the lower limb muscles to produce a large knee extensor moment to counteract the effect. In contrast, during level walking the GRF passes near the knee joint center and should the associated joint moment should therefore be relatively small. The actual joint moment data confirm these first approximation estimates. Finally, because the knee is flexing to lower the body down the slope, power is absorbed at the knee joint during late stance of downslope walking, where it is typically generated during level walking.

A similar change in the knee joint moment patterns has also been observed during stair descent (McFadyen and Winter, 1988; Riener et al., 2002); the stance phase knee joint moment has the same double-hump shape. The knee is flexing during stair descent to lower the body down the steps, and therefore power is absorbed at the knee (McFadyen and Winter, 1988; Riener et al., 2002). Based on these data it seems that the knee performs similar roles in these tasks, controlling the lowering of the body over the support foot via lengthening contractions of the extensors (McFadyen and Winter, 1988).

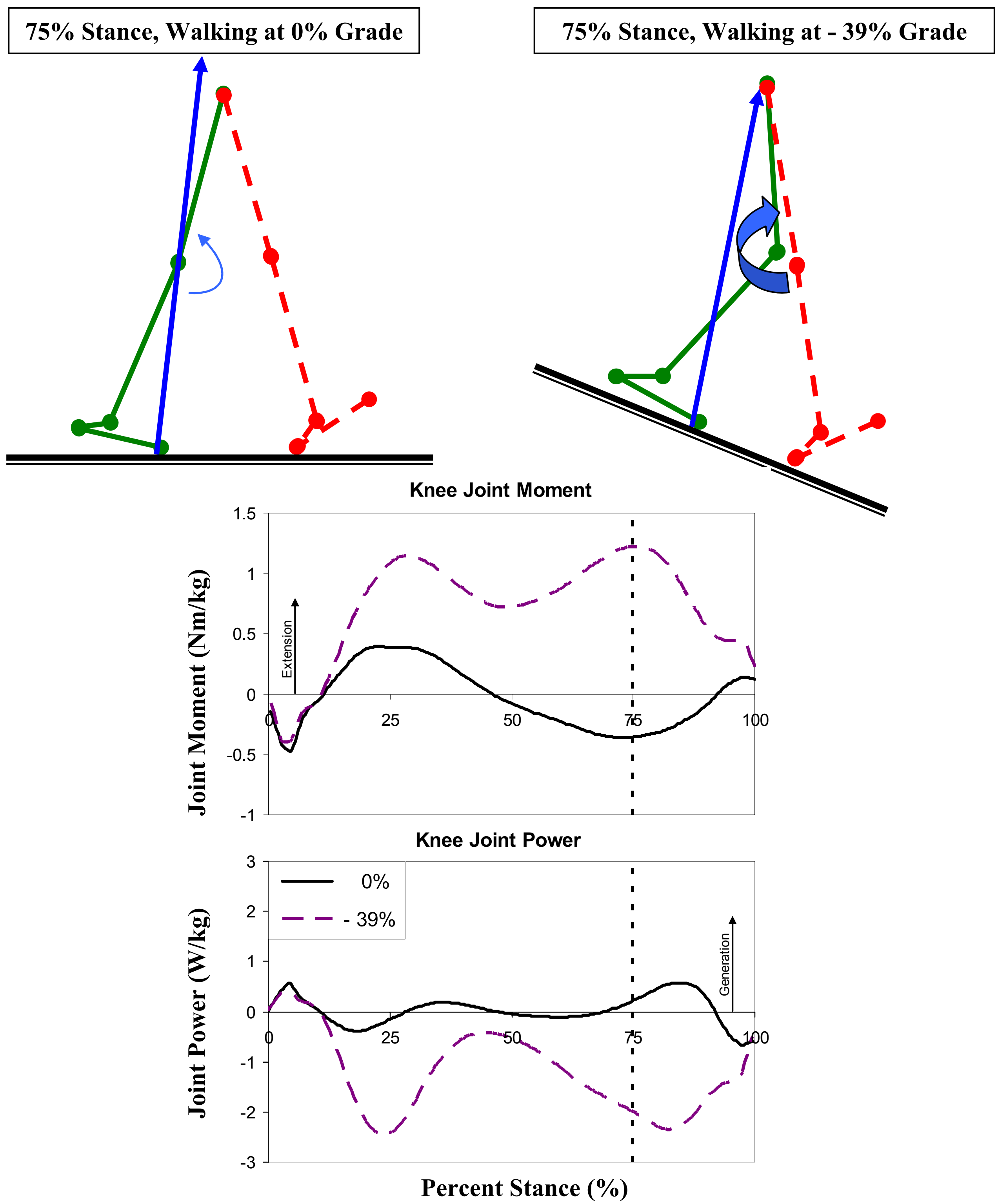


Figure 22. Schematic Model of the Lower Limbs at 75% Stance during level and downslope walking at -39%, where the circles, straight arrows, and curved arrows represent the joint centers, GRF, and first approximation external joint moments, respectively. Graphs show knee joint moment and power during stance phase of level and downslope walking, 75% of stance is marked with a vertical line.

Muscle Activity. The level walking EMG data are in agreement with previous reports (Shiavi, 1985; Winter and Yack, 1987). EMG activity during downslope walking has been reported in only two publications. The first study reports burst timing information for five lower limb muscles (GM, semitendinosus, RF, MG, and TA), but does not report amplitude data for the muscle bursts, and also does not report their methods clearly or completely (Tokuhiro et al., 1985). The second study is a conference abstract that reports burst timing and magnitude information for the lateral gastrocnemius, MG, Sol, and TA (Mitsui et al., 2001). The changes in burst duration found here are consistent with those reports: increased MG and RF burst durations (via earlier MG onset and later RF offset), a midstance hamstring burst, and no change in the GM or TA. The burst amplitudes observed here are also consistent with those reported by Mitsui et al. (2001).

Increased eccentric (and/or decreased concentric) activity during downslope walking has been observed in quadrupeds (Gillis and Biewener, 2002), and in humans (here). In addition, downslope walking in quadrupeds (rats and cats) is a task that has been shown to require reduced muscle activity (Gillis and Biewener, 2002; Smith et al., 1998). With the exception of the knee extensors, the muscles studied here generally showed either no change in activity or a decrease in activity from level walking. These activity and power data are consistent with each other, because when force is constant, eccentric muscle contractions require lower activation levels than concentric contractions (Enoka, 1996). Muscle activity during eccentric contractions still scales with force, but for a given force output is less than during a concentric contraction (Enoka, 1996).

In quadrupeds it has been found that the hindlimb hip extensors are inactive during downslope walking (Gillis and Biewener, 2002; Smith et al., 1998) and that the hip extension is largely passive and caused by gravity. Because humans do not have forelimbs to cause the necessary deceleration, the hip extensors must be active in early stance to absorb energy as the body is pulled down the slope. However, the magnitude of the power being absorbed is small, and therefore the intensity of the activity is not different than the relatively low-level activity seen during level walking. And in fact, some subjects demonstrated complete inactivity of the GM muscle (3 out of 9 subjects) during downslope walking. This is most likely because the activity of the biarticular muscles was sufficient in these subjects to cause the necessary deceleration. This example speaks to the variability of muscle activity in humans, and also to the redundancy in the neuromuscular system.

Joint Kinetics and Muscle Activity at the Hip and Knee. Based on the observed changes in the joint mechanics during downslope walking one would anticipate significant increases in the knee extensor activity magnitude and burst duration, but minimal change in the activity of knee flexor or hip extensor muscles, compared to level walking. The data from the knee extensors, RF and VM, were in agreement with these expectations: the burst durations and mean activity levels were significantly longer and higher than during level walking. These increases in activity from level walking correspond to the second knee extensor moment peak observed during downslope walking, which is evident from the high correlations for both the magnitude and duration of the moment and EMG (Figure 20B and 21B). In addition, the knee flexors and hip extensors (GM, BF, SM) showed little change from their activity during level walking. The short duration and low activity level of the hip extensors in early stance does little to counteract the increased activity of the biarticular RF, therefore resulting in the earlier onset of the hip flexor moment. One might expect the flexor moment onset to decrease progressively with walking grade (i.e. be earliest at -39% and latest at 0%), but that is not the case. Recall from Figure 14 and Table 15 that the onset of the hip flexor moment occurred earliest during walking at -15% grade, followed by -39% and finally by level walking. The EMG activity can be used to explain this unexpected finding. During walking at -15% grade the mean activity of the biarticular hip flexors decreases to approximately 50% of the activity at level, but then during -39% walking it returns to near 100% of the activity at level. Although the decrease at -15% grade was not statistically significant, it is relevant here because it means that the increased RF activity is consistent with an earlier flexor moment onset at -15% grade than at -39% grade. This slight mismatch in the EMG and moment for the hip joint results in only moderate hip joint-specific correlations.

During mid-stance the simultaneous knee extensor and hip flexor moments might have resulted in preferential activation of the biarticular RF over the uniarticular VM (Prilutsky, 2000). However, the burst durations and mean activity levels increased similarly in these two muscles. It is possible that the amount of hip flexor activity needed to generate the small flexor moment (almost neutral during midstance) was not enough to require increased activation of the RF over the VM. Alternatively, the knee extensor moment may be too large to be caused by RF activity alone, requiring VM activity as well. It could be that downslope grades steeper than -15% would elicit this preferential activation.

Joint Kinetics and Muscle Activity at the Knee and Ankle. The decreasing peak ankle plantarflexor moment suggests that mid-stance plantarflexor activity should decrease as the walking grade decreases from 0% to -39%. Indeed, the mean activity levels of the MG and Sol bursts decreased while the mean activity level of the early- and mid-stance TA bursts increased, which is in agreement with the joint moment patterns. In spite of this agreement, the peak moment and burst magnitudes at the ankle were not well correlated, because the changes in the peak bursts were small, and they did not decrease with grade for the MG. The onset of MG and Sol activity during downslope walking was also earlier than during level walking, such that there was a period of co-contraction with the TA during early stance. Because the plantarflexor moment duration did not change during downslope walking the joint-specific correlation between the EMG and moment durations was poor for the ankle (Figure 20B). The mean TA activity in early stance was also greater than during level walking. There was a dorsiflexor moment at that time and power was absorbed, indicating that the TA was acting eccentrically to control the lowering of the foot from heel strike to foot flat. The new period of co-contraction during early stance in downslope walking would therefore serve to stabilize the ankle joint as the foot was lowered to meet the ramp surface. A similar period of early stance co-contraction of the TA and soleus has been observed during stair descent in humans, and was also thought to stabilize the ankle during that task (McFadyen and Winter, 1988).

Conclusions

The goal of this aim was to characterize the task of forward downslope walking, and to determine whether the muscle activity patterns and joint kinetics are related in a task with increased eccentric contractile conditions where the joint moment patterns change. From the downslope walking data presented here it is ***evident that muscle activity and joint kinetics are indeed correlated in such a task***. The overall lower limb correlations for both magnitude and duration of joint moments and muscle activity were good (R values > 0.82). The hypothesis that changes in muscle firing patterns would be required was correct, and the most substantial changes occurred at the knee joint. The observed changes in both muscle burst durations and activity levels were exactly those that would be expected based on the joint moments: increased ankle dorsiflexor mean activity in early stance, and increased knee extensor mean activity and burst durations and decreased ankle plantarflexor mean activity during mid-stance. The other changes in muscle activity (mid-stance burst in BF, earlier onset of MG and Sol) are consistent with the need to stabilize the ankle and knee joints as the body is lowered over the stance foot. The knee joint showed the best correlations when joint-specific regressions were performed. This is because the changes in the hip and ankle moments and muscle activity were small compared to the changes in the knee joint moment and muscle activity. Generally speaking, the muscle activity patterns corresponded to the joint moment patterns for the task of downslope walking.

The question that then arises is: from a control standpoint how are these muscle activity and joint kinetic patterns related to those for level walking? If the same motor program was governing both level and downslope walking, then the tasks should have the same muscle firing pattern. That is, the onset times and/or burst durations for each muscle would be similar for both conditions, with changes only in the amplitude of the muscle activity. The activity patterns of the TA, GM, and SM seem to support this idea: the stance phase burst onsets and durations are independent of the walking grade, and the only changes observed are in the burst amplitudes. On the other hand, if the control strategy was different there would be non-uniform changes in the muscle firing patterns, similar to those that occur in the joint kinetics. The latter is more consistent with our findings from the other muscles: the burst durations (and activity levels) of the knee and ankle extensors increase during downslope walking. These data suggest that either different control systems are needed for level and downslope walking, or that the level walking pattern at the knee and ankle joints is altered to produce the muscle firing patterns needed for downslope walking. This change in control strategies is consistent with the finding that eccentric muscle activity requires “unique activation strategies” from the nervous system (Enoka, 1996). It is possible that increased muscle spindle feedback (because eccentric contractions increase) could cause this shift or modification in the control strategy. For example, the increased spindle feedback could act specifically at the knee and ankle extensor control centers to modify the motor program. Although it is not possible to determine whether there is a complete change in control systems or just a modification, the data reported here at least provide some insight into the goals of the nervous system during the task of downslope walking (the joint moments), and the mechanisms used to achieve those goals (the muscle firing patterns).

CHAPTER 6

FORWARD UPSLOPE WALKING[3]

Introduction

Results from the **second aim** of this project, the investigation of forward upslope walking, are discussed in this chapter. There were no prior data regarding the kinetics of upslope walking, but there was some information reported in the literature about upslope walking muscle activity (Leroux et al., 1999; Tokuhiro et al., 1985). The majority of the changes in the muscle activity during upslope walking occurred in the burst magnitudes, rather than in the burst durations. Using the muscle activity data as a first approximation for the joint moments, it was expected that the changes in the joint moments would be predominately changes in magnitude rather than in pattern. Specifically, it was *hypothesized that changes in the magnitude of the muscle activity would be directly related to the changes in the magnitude of the joint moments, but that the patterns of both the joint moments and muscle activity would remain unchanged*. Again, this relationship would be tested using correlations of the moment and muscle activity magnitudes and durations. As already discussed, it was expected that the relative amount of concentric muscle activity would increase. This finding would suggest that force-dependent feedback from the golgi tendon organs (GTO) would increase. Again, the potential impact of this increased feedback on the control strategy was considered although it could not be implicitly tested.

[3]Some text and figures in this chapter are adapted from the article "The effects of sloped surfaces on locomotion: A kinematic and kinetic analysis" published in the Journal of Biomechanics (Lay et al., 2005a), with permission from Elsevier, Inc.

Results

Joint Kinematics. Upslope walking joint kinematic data are presented in Figure 23 and Table 17. A significant grade effect was observed for all variables (ANOVA). Results from the follow-up comparisons are indicated with symbols in Table 17. The ankle joint (Figure 23A) was progressively more dorsiflexed at heel strike (AA1) as the walking grade increased. Except for a brief period of plantarflexion (AA2) during upslope walking at +15% the ankle joint was dorsiflexed until late stance. During walking at +39% grade the ankle was dorsiflexed from heel strike until late stance. The upslope and level walking ankle angles were similar only in late stance and early swing (AA3). The knee flexion angle at heel strike (Figure 23B, KA1) increased with the walking grade, but the knee angle during late stance (KA2) was similar for all walking grades; this resulted in increased knee extension during midstance to lift the body up the incline. In early swing the knee angle (KA3) was still similar across the walking grades, but flexion increased during late swing to prepare for heel strike. Unlike the other joints, the hip joint angle (Figure 23C) maintained the same temporal pattern during level and upslope walking, but progressively increased in amplitude (HA1, HA2, HA3) as walking grade increased.

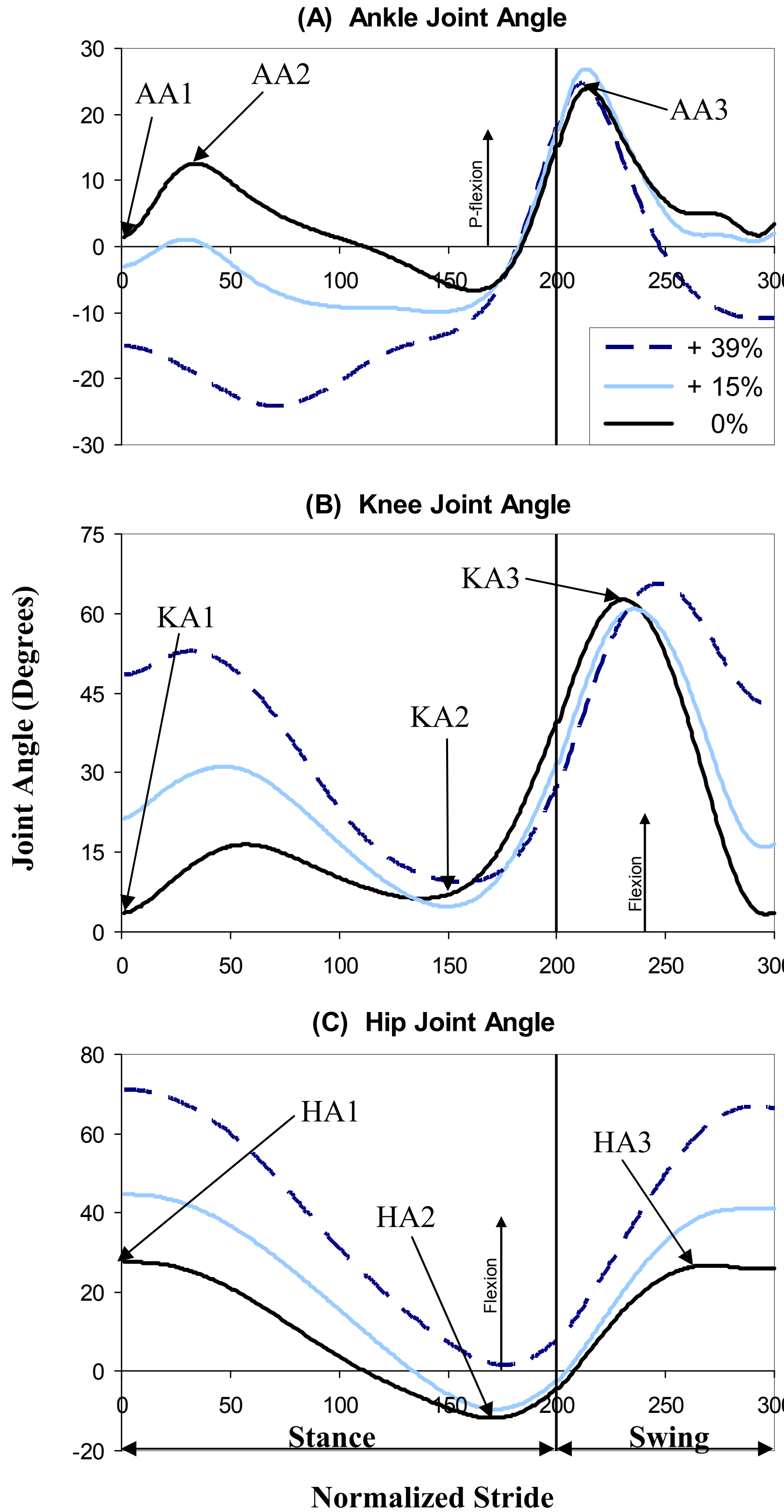


Figure 23. Average Joint Angles during Upslope Walking. Joint angle convention is shown in Figure 4. Stance phase was normalized to 200 points and swing phase to 100 points before ensemble averaging. Vertical line marks toe-off.

Table 17. Upslope Walking Joint Angle POI Data

	0%	+ 15%	+ 39%
AA1	1.51 (4.51)	-3.00 § (5.70)	-15.04 §† (5.91)
AA2	12.75 (4.31)	1.45 § (6.86)	-14.84 §† (5.87)
AA3	24.23 (6.38)	27.44 § (7.75)	25.17 (9.54)
KA1	3.56 (4.60)	21.32 § (8.06)	48.41 §† (5.83)
KA2	7.01 (4.70)	4.77 § (4.18)	9.49 (6.40)
KA3	62.82 (3.88)	61.12 (4.71)	66.24 † (5.59)
HA1	27.56 (6.16)	44.65 § (9.83)	71.05 §† (9.48)
HA2	-11.83 (6.83)	-9.64 (6.41)	0.53 §† (7.65)
HA3	27.43 (5.40)	41.92 § (7.60)	67.64 §† (7.88)

Note: Mean (SD) of the joint kinematic POI (all values in degrees), averaged across all subjects for each grade. § - indicates a significant difference from 0% grade ($p<0.05$), † - indicates a significant difference between 15% and 39% in the same walking direction (upslope or downslope) ($p<0.05$).

Ground Reaction Forces. There was a significant grade effect (ANOVA) for all GRF points of interest except the peak normal component during early stance (Figure 24, Table 18). The early stance AP force (Figure 24A, AP1) was negligible at +15% walking grade and nonexistent at +39% grade, while the propulsive force (AP2) increased significantly with walking grade. As explained in Chapter 5, the ML force (Figure 24B) was not analyzed statistically. The late stance normal component peak (Figure 24C, N2) and resultant force peak (FR, Figure 25) increased significantly at +15% but not at +39% walking grade. Both the normal force and the resultant force decreased during midstance.

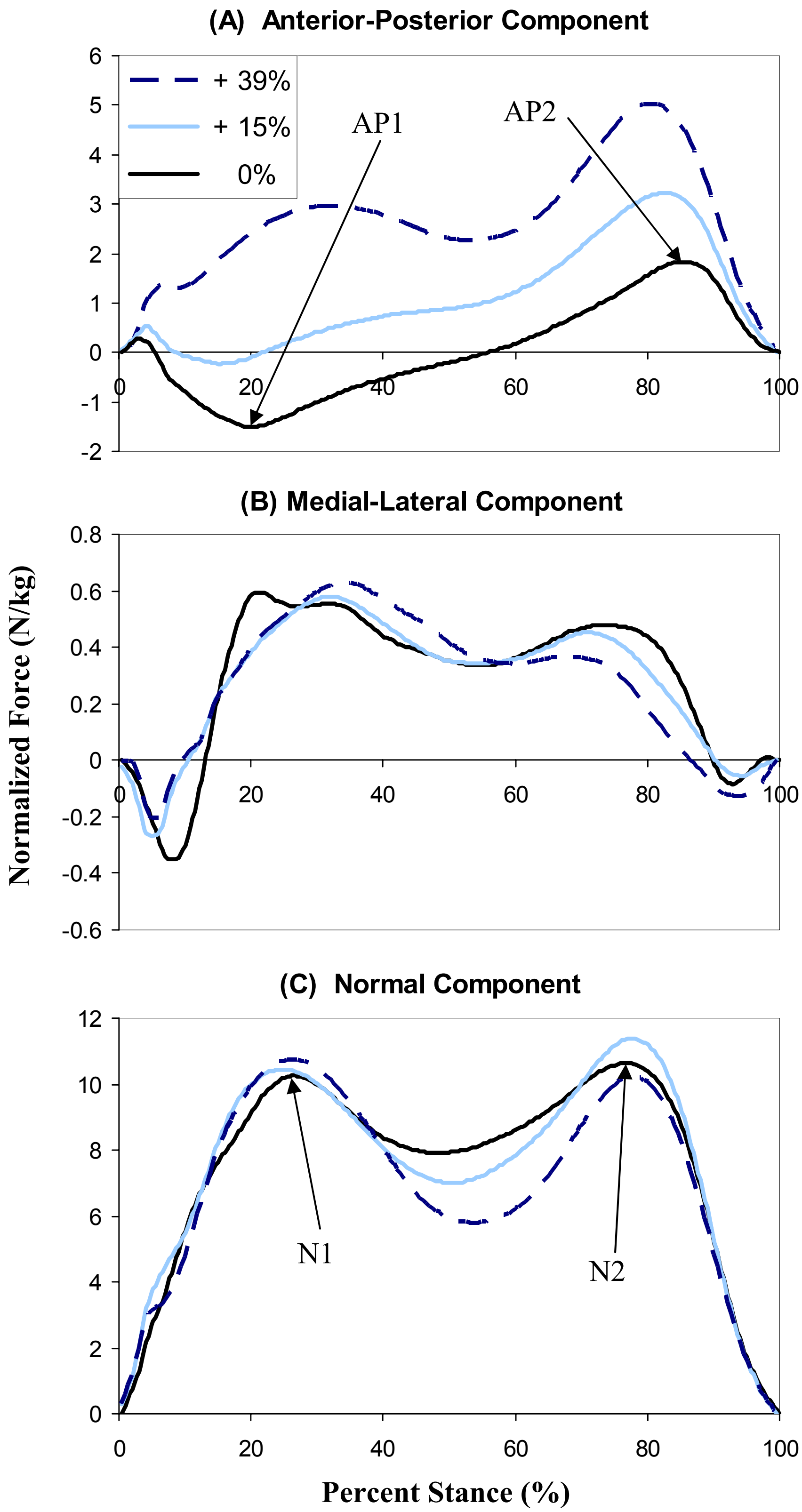


Figure 24. Average Ground Reaction Force Components during Upslope Walking. Positive forces indicate posteriorly and laterally directed foot forces.

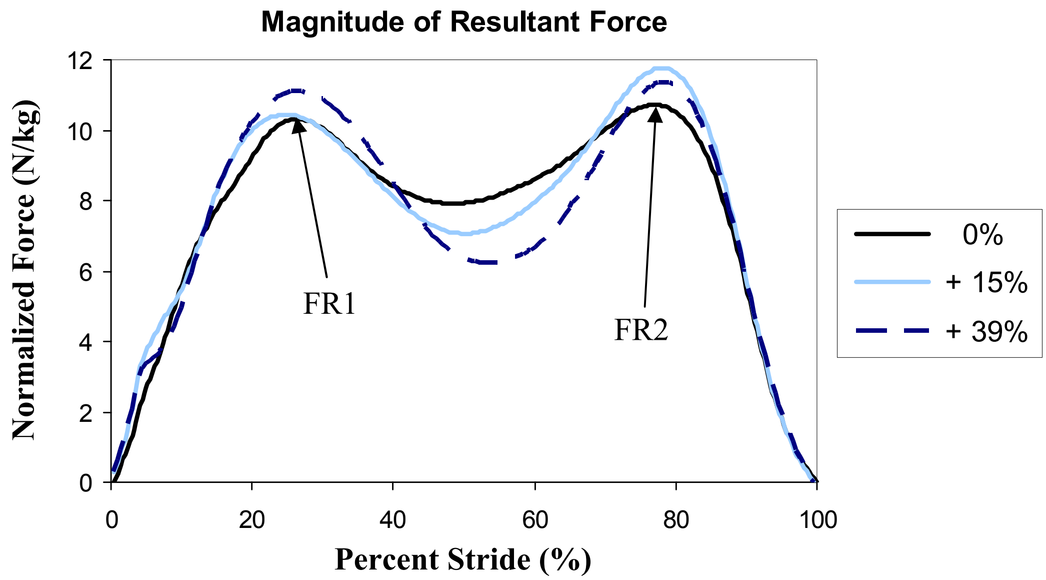


Figure 25. Average Resultant Force during Upslope Walking, calculated as the sum of the normal and AP component.

Table 18. Upslope Walking GRF Component POI Data

	0%	+ 15%	+ 39%
AP1	-1.55 (0.14)	-0.29 § (0.23)	N/A
AP2	1.85 (0.34)	3.27 § (0.48)	5.08 §† (0.63)
N1	10.37 (0.43)	10.60 (0.63)	10.88 (0.90)
N2	10.71 (0.46)	11.46 § (0.76)	10.33 † (0.88)
FR1	10.45 (0.43)	10.61 (0.63)	11.24 (0.91)
FR2	10.79 (0.49)	11.85 § (0.83)	11.46 (1.0)

Note: Mean (SD) of the GRF component POI (in N/kg). As defined, point AP1 did not exist for the +39% condition so no value is included in the table. Gray shading indicates no significant grade effect was observed (ANOVA). § - a significant difference from 0% ($p<0.05$), † - a significant difference between 15% and 39% in the same walking direction (upslope or downslope) ($p<0.05$).

Joint Kinetics. During swing the joint moment patterns were generally similar across slope conditions. In contrast, during stance the joint moments changed significantly with grade (Figure 26). Significant grade effects (ANOVA) were observed for all joint moment variables except KM1 (post-hoc comparisons are indicated in Table 19). The ankle moment (Figure 26A) in early stance (AM1) increased with the walking grade, such that the typical dorsiflexor moment present during level walking was either negligible or absent. The peak ankle plantarflexor moment (AM2) increased uniformly during upslope walking. The knee moment pattern (Figure 26B) was nearly identical to that during level walking, with mostly non-significant changes in the peak extensor and flexor moment values (KM2 and KM4). The knee flexor moment at 50% of stance (KM3) increased at the steepest walking grade. The peak hip extensor moment (Figure 26C) in early stance (HM1) and the time of the transition to a flexor moment in midstance (HM2) increased progressively with the upslope walking grade. The late stance hip extensor peak (HM3) showed no change with walking grade. Finally, the support moment (Figure 27) increased from level walking for the majority of stance, with the largest increases observed in early stance (SM1).

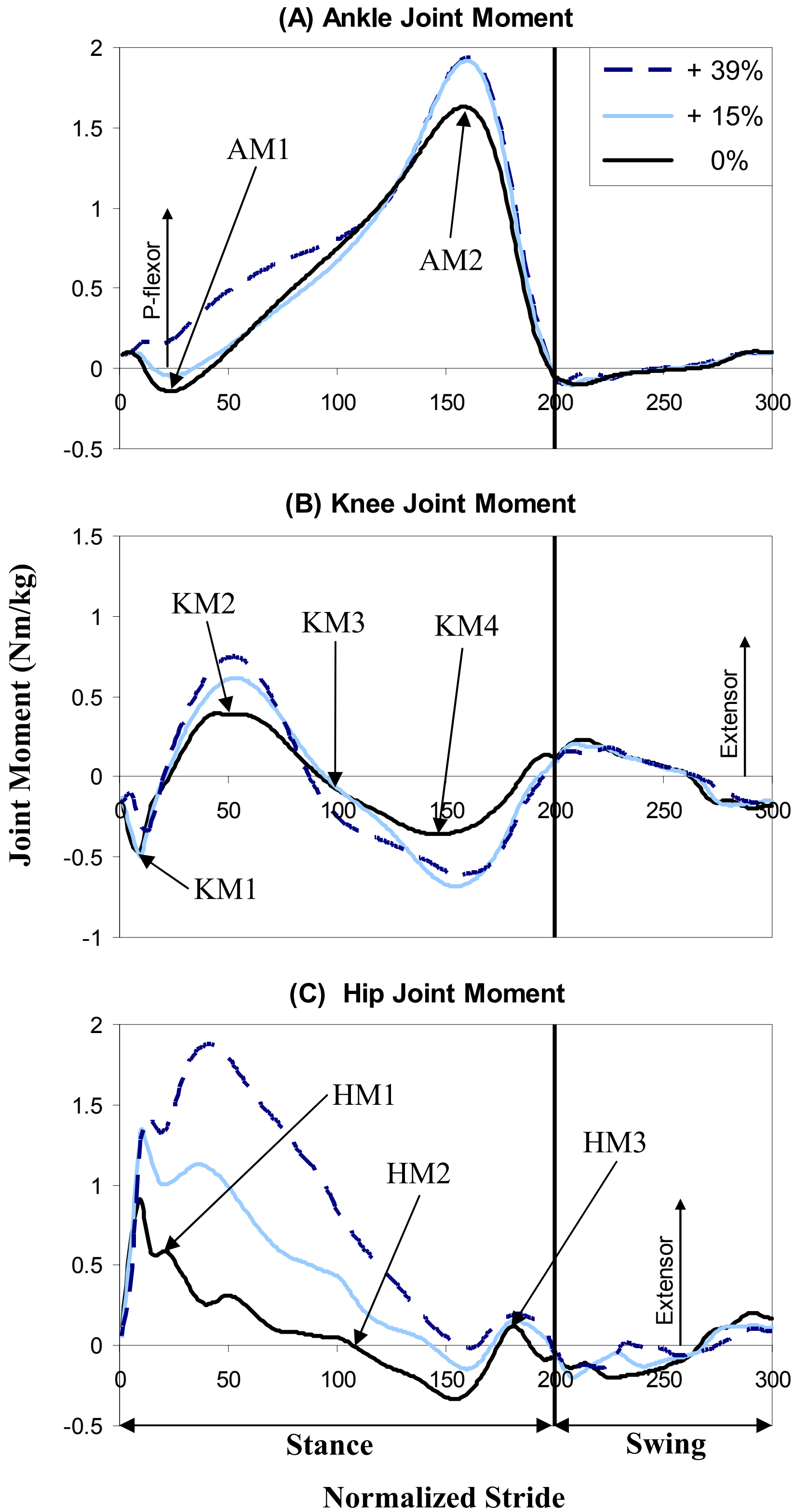


Figure 26. Average Joint Moments during Upslope Walking. Positive moments are plantarflexor and extensor, as indicated. Stance phase was normalized to 200 points and swing phase to 100 points before ensemble averaging. Vertical line marks toe-off.

Table 19. Upslope Walking Joint Moment POI Data

	0%	+ 15%	+ 39%
AM1	-0.15 (0.06)	-0.06 § (0.07)	0.08 §† (0.08)
AM2	1.64 (0.14)	1.94 § (0.22)	1.95 § (0.18)
KM1	-0.51 (0.09)	-0.56 (0.21)	-0.43 (0.21)
KM2	0.44 (0.20)	0.64 (0.38)	0.81 (0.41)
KM3	-0.09 (0.17)	-0.08 (0.22)	-0.27 §† (0.19)
KM4	-0.38 (0.22)	-0.70 § (0.31)	-0.68 (0.31)
HM1	0.54 (0.22)	1.10 § (0.38)	1.93 §† (0.34)
HM2	97.64 (42.76)	146.86 § (38.22)	168.19 §† (28.98)
HM3	0.16 (0.15)	0.21 (0.13)	0.32 (0.25)
SM1	0.91 (0.23)	1.79 § (0.25)	3.00 §† (0.44)
SM2	1.14 (0.15)	1.26 (0.26)	1.64 §† (0.34)

Note: Mean (SD) of the joint moment POI (values N*m/kg, except HM2 is normalized stride). § - different from 0% grade, † - different from 15% (same direction), ($p<0.05$).

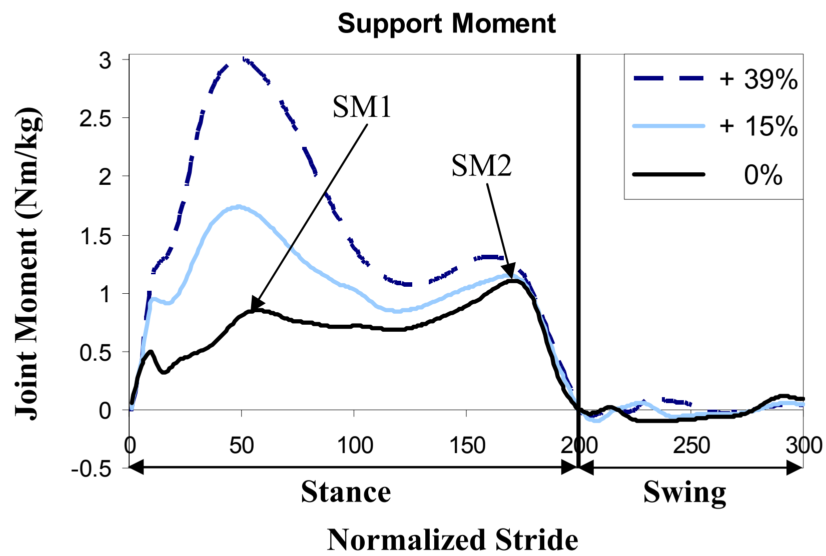


Figure 27. Average Support Moment during Upslope Walking. Calculated as the sum of the extensor moments at the three lower limb joints (Winter, 1980).

Joint Power. Figure 28 shows that power generation increased at all joints as the walking grade increased, with especially large increases at the hip joint. In addition, the percent of the stance phase during which power was generated increased progressively at all joints as the walking grade increased (Table 20). These data suggest that muscles were active during concentric contractions as the body was lifted over the stance foot up the ramp.

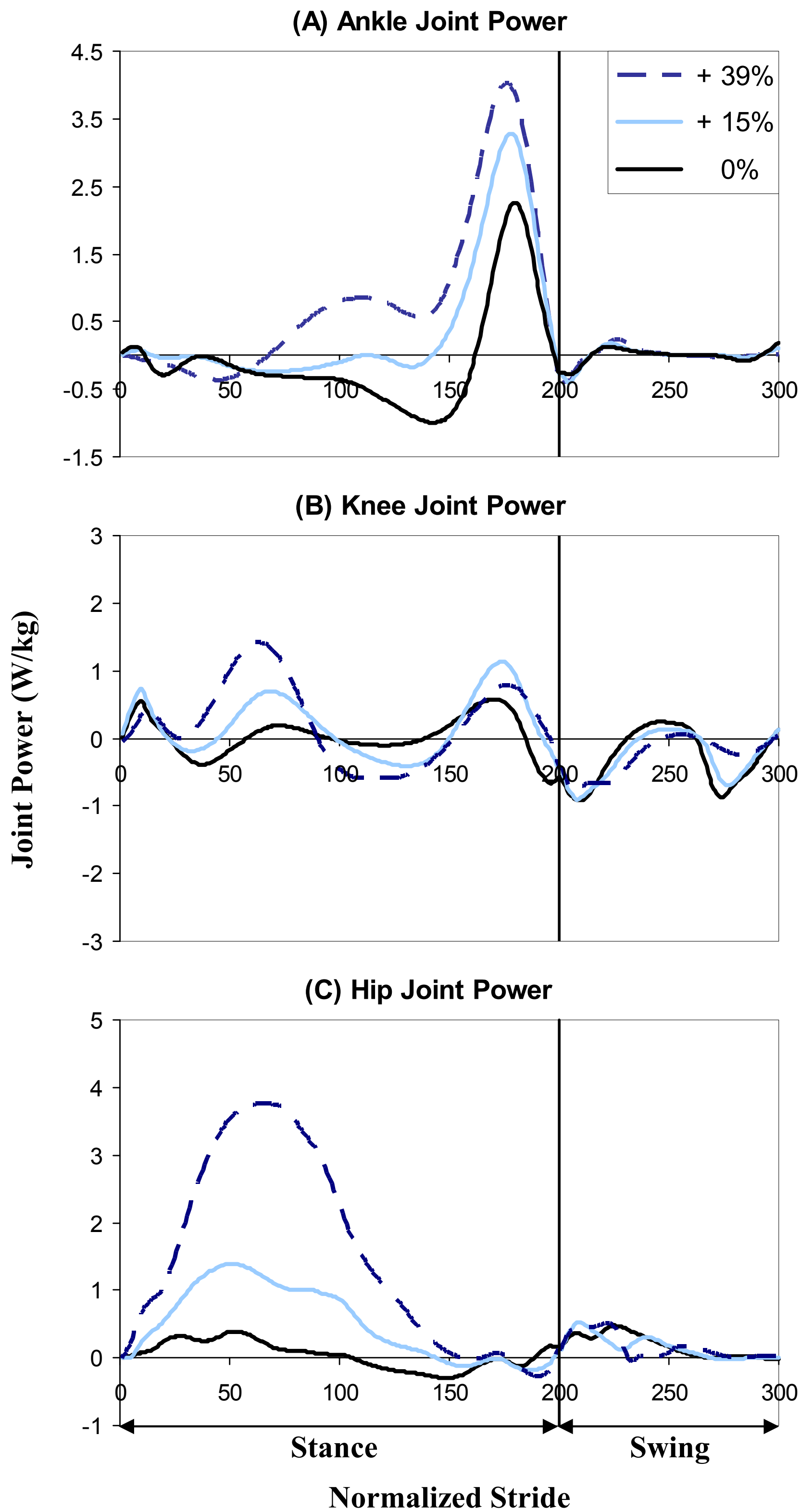


Figure 28. Average Joint Power during Upslope Walking. Positive power is generation, negative is absorption. Stance and swing phases of each trial were normalized before ensemble averaging. Vertical line marks toe off.

Table 20. Average Percent of Stance Phase when Power is Generated

	0%	+ 15%	+ 39%
Ankle	22.93	36.10	63.41
Knee	51.71	57.07	61.95
Hip	57.07	70.73	80.98

Muscle Activity. Representative and group average EMG data are presented in Figures 29 (GM, BF, and SM), 30 (RF and VM), and 31 (Sol, MG, and TA). The graphs show the muscle activity produced by the majority of participants, as described in the methods (Chapter 4); deviations from these patterns will be discussed below. For comparison, ensemble average stance phase EMG data can be found in Appendix A.

During stance phase the mean activity of the GM (at +39% only), BF, SM, RF, VM, Sol, and MG increased significantly as the walking grade increased. The activity of the GM at +15% and of the TA at +15% and +39% also increased, but the differences were not significant. The duration of the GM, BF, SM, RF, and VM stance phase bursts increased and significantly as the walking grade increased; the duration of the Sol, MG, and TA bursts did not change with increased walking grade. In the +39% condition, the increase in mean BF activity was uniform across the longer burst, while the increase in mean SM activity appeared as two approximately equal peaks in the longer burst (see Appendix A, Figure 70). The increased mean RF and VM activity in the +39% condition was concentrated in the first half of the longer stance phase burst, as suggested by the representative data in Figure 30, with much lower activity levels in the second half of the burst (Appendix A, Figure 71). The Sol and MG showed the opposite pattern; peak activity occurred late in the stance phase burst (Appendix A, Figure 72).

All nine subjects exhibited the same bursting pattern for six of the eight muscles; only the bursting patterns of the GM and TA showed variability between subjects. One subject did not exhibit an identifiable GM burst during level walking. Three subjects did not have the mid-stance TA burst at level walking, and one subject did not have this burst at +39%. The variability in the firing patterns here was much lower than during downslope walking; it could be that the increased power requirements during upslope walking result in increased recruitment and decreased variability in the EMG patterns.

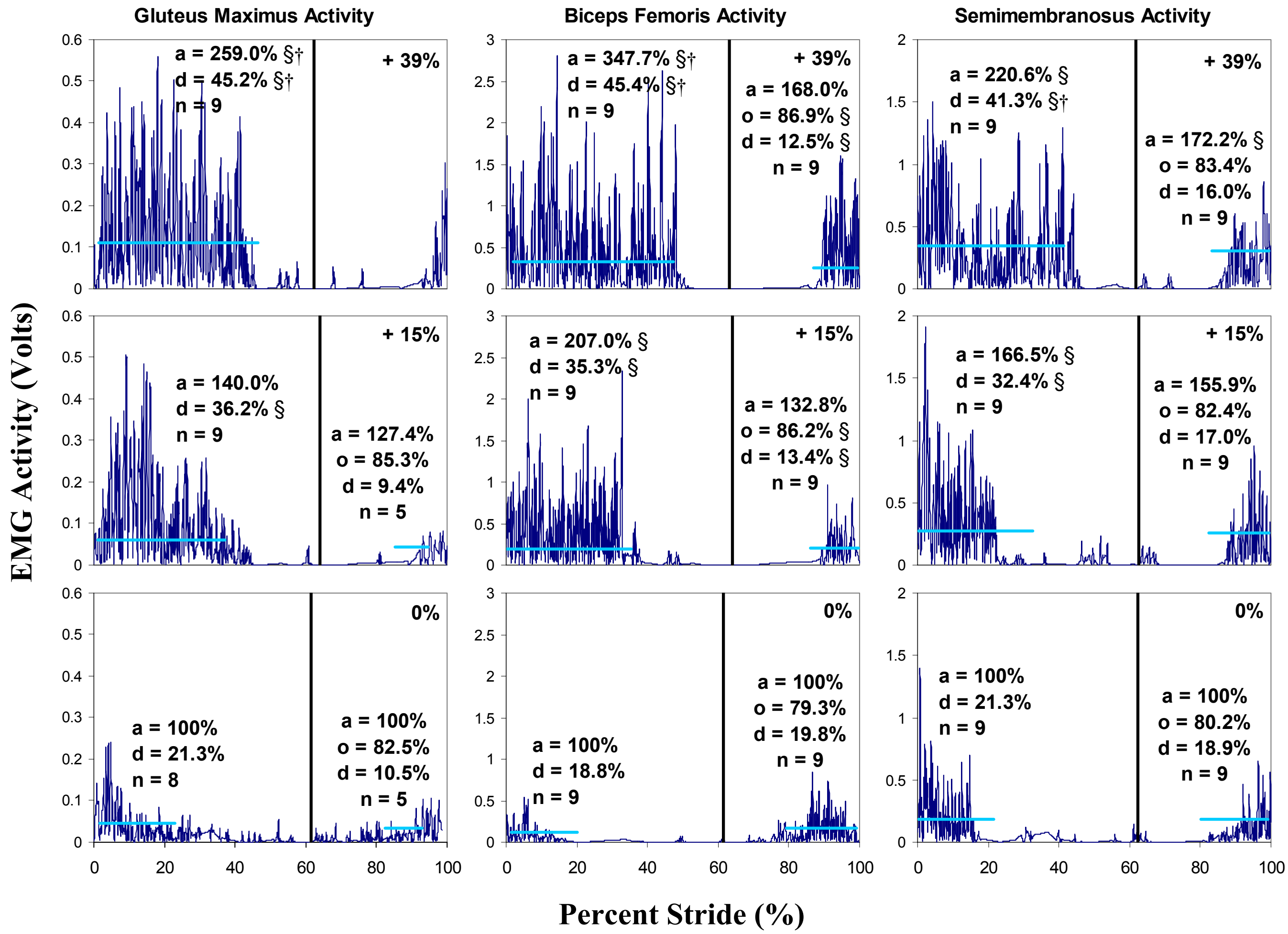


Figure 29. Representative GM, BF, and SM EMG Data during Upslope Walking. For each muscle a representative trial was chosen for each grade from a single participant's data to demonstrate the typical activity. Each trial was normalized to its stride time and plotted in %stride, with heel strike occurring at 0%. The toe-off time for each trial was also normalized to %stride and is indicated with a vertical line (average over all 24 representative trials for all eight muscles = 62.1%). Data from seven participants were used to represent the eight muscles. Horizontal lines representing the average burst data (mean for all subjects exhibiting the burst) are overlaid on each plot, and the values are also displayed on the plots: a = the group average mean activity (as a percentage of mean activity at 0% grade), d = the group average burst duration (in %stride), and n = the number of subjects exhibiting the burst. For bursts that do not start at heel strike o = the onset time (in %stride). These variables were submitted to repeated measures ANOVAs to assess the effect of grade. When a significant grade effect was observed, dependent t-tests were used for follow-up comparisons (adjusted p value of 0.01667). All significant differences are indicated on the figure by the following symbols: § - significantly different than the value at 0% grade ($p < 0.01667$) and † - significantly different than the value at +15% ($p < 0.01667$).

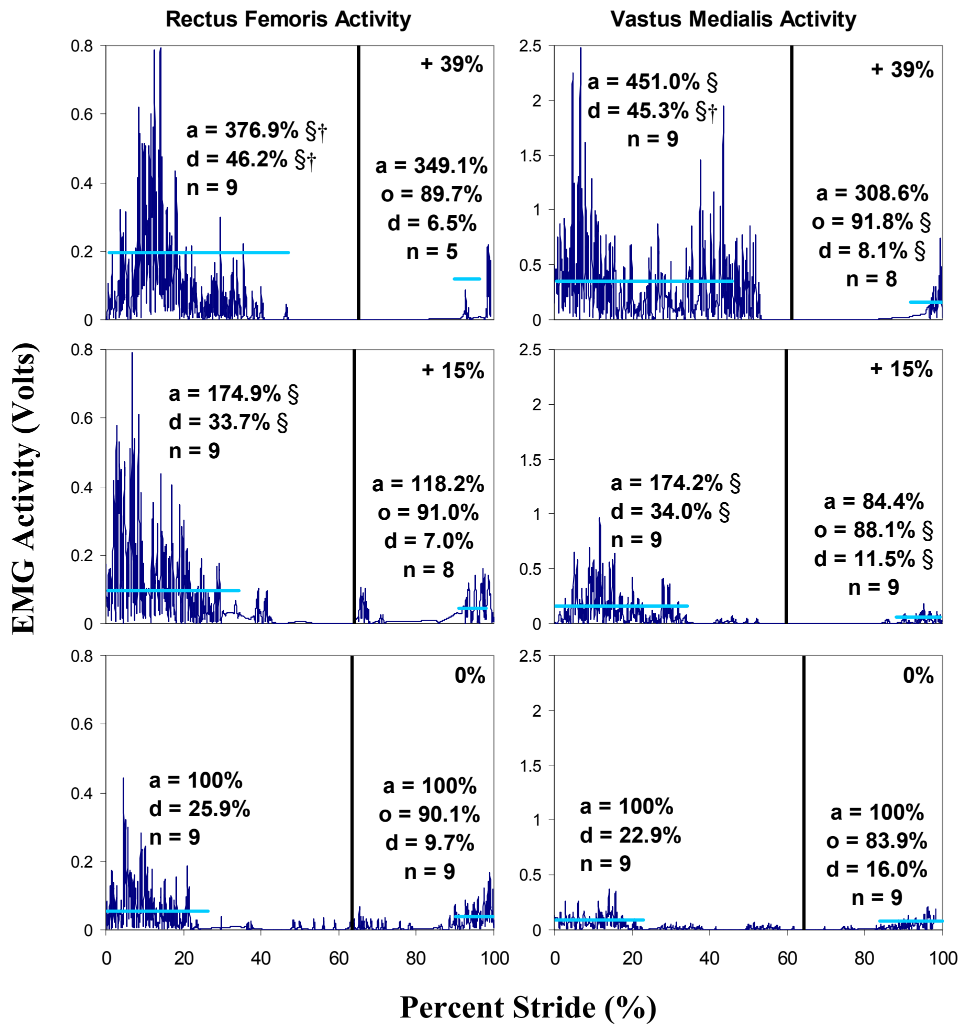


Figure 30. Representative RF and VM Data during Upslope Walking. Details are the identical to Figure 29. All significant differences are indicated on the figure by the following symbols: § - significantly different than the value at 0% grade ($p < 0.01667$) and † - significantly different than the value at +15% ($p < 0.01667$).

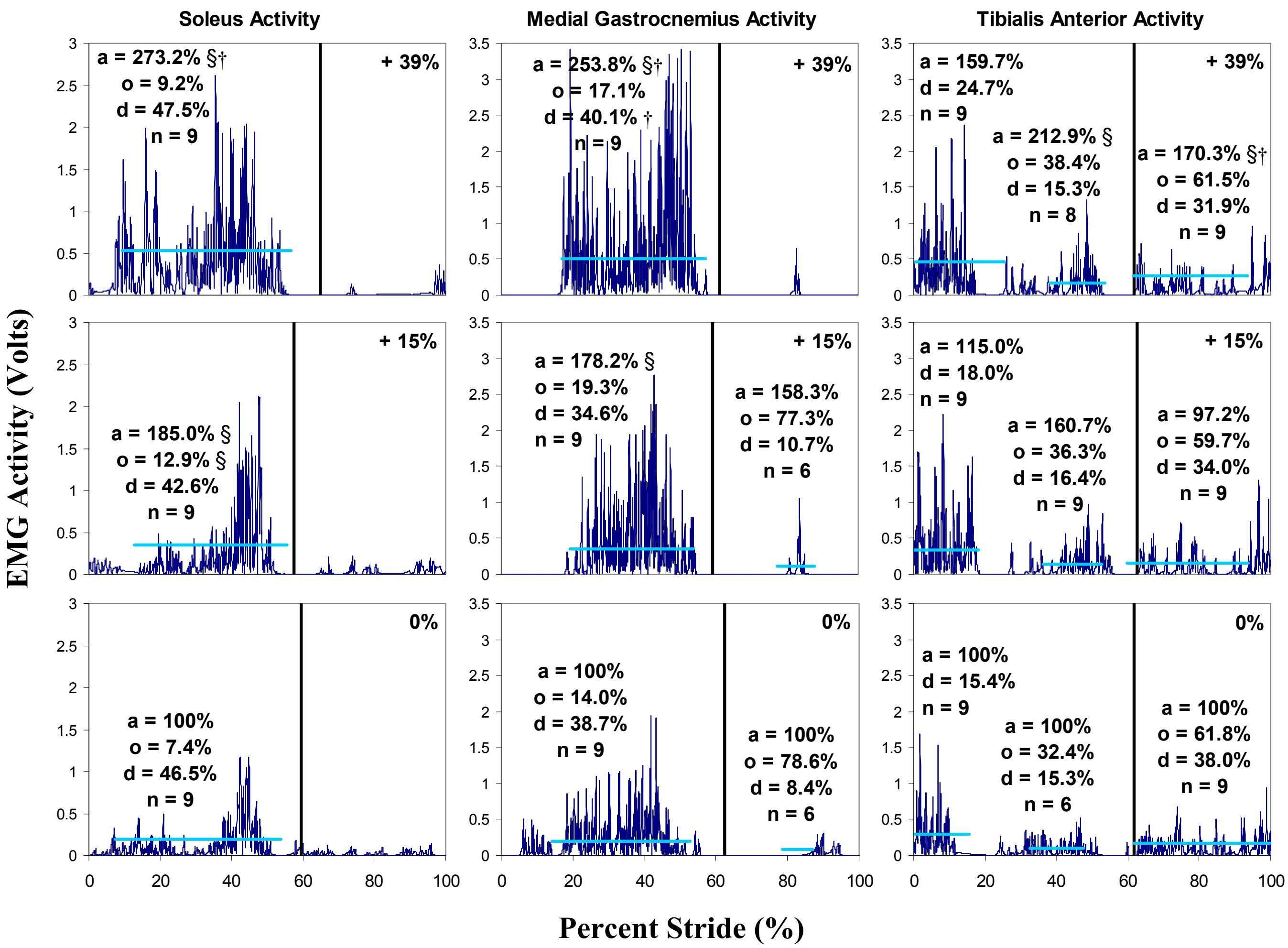


Figure 31. Representative Sol, MG, and TA EMG Data during Upslope Walking. Details are the identical to Figure 29. All significant differences are indicated on the figure by the following symbols: § - significantly different than the value at 0% grade ($p < 0.01667$) and † - significantly different than the value at +15% ($p < 0.01667$).

Correlations between Kinetics and EMG. In order to asses the relationship between joint moments and EMG, the durations and peak magnitudes of the extensor joint moments and the extensor muscle bursts were correlated (Figures 32 and 33). The duration and peak correlations had similar patterns: a low correlation for the overall limb, with a range of joint-specific correlations.

For the moment and muscle burst durations (Figure 32) the correlation for the overall lower limb (Figure 32A) was relatively low, with an R-value of 0.434 (r^2 values are shown in the figures). When joint-specific linear regressions were performed (Figure 32B), the results varied widely: the hip and knee joint correlations were better than the overall limb correlation (R-values of 0.976 and 0.555, respectively), but the ankle joint correlation was worse (R = 0.264). The hip joint correlation was high because the extensor muscle burst durations increased as the moment duration increased. At the knee joint, muscle burst durations increased although the moment duration did not change appreciably, resulting in a relatively low correlation. The reverse was true at the ankle joint, the moment duration increased at the steepest slope, but the muscle burst duration remained relatively constant; this also resulted in a poor correlation.

The overall limb correlation for the joint moment and muscle burst peak magnitudes (Figure 33A) had an R-value of just 0.502. The joint-specific correlation for the hip (Figure 33B) was similar to the overall correlation, with an R-value of 0.575. The knee and ankle joint-specific correlations (Figure 33B) were much better, with values of 0.898 and 0.999, respectively. Considering the moment and EMG data, the low hip joint correlation was unexpected. Review of Figure 33B revealed that the three data points for each extensor muscle formed lines of slightly different slopes, which made the overall

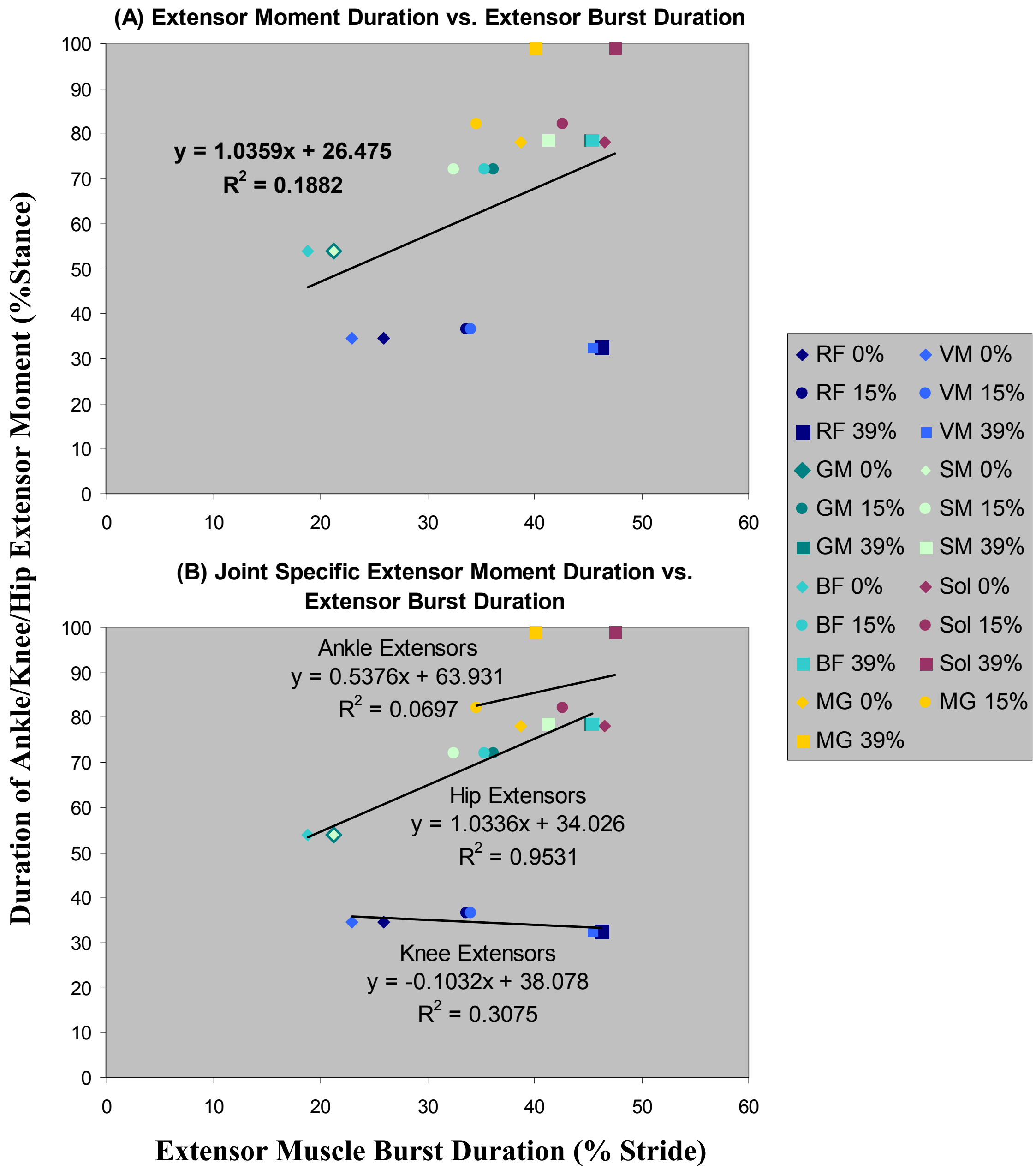


Figure 32. Correlations of Extensor Moment Durations and Extensor Muscle Burst Durations during Upslope Walking for lower limb extensor muscles and the joints at which they act. ♦ – data from level walking trials, ● – data from -15% trials, ■ – data from -39% trials. R^2 values are given for (A) overall lower limb correlation and (B) joint-specific correlations.

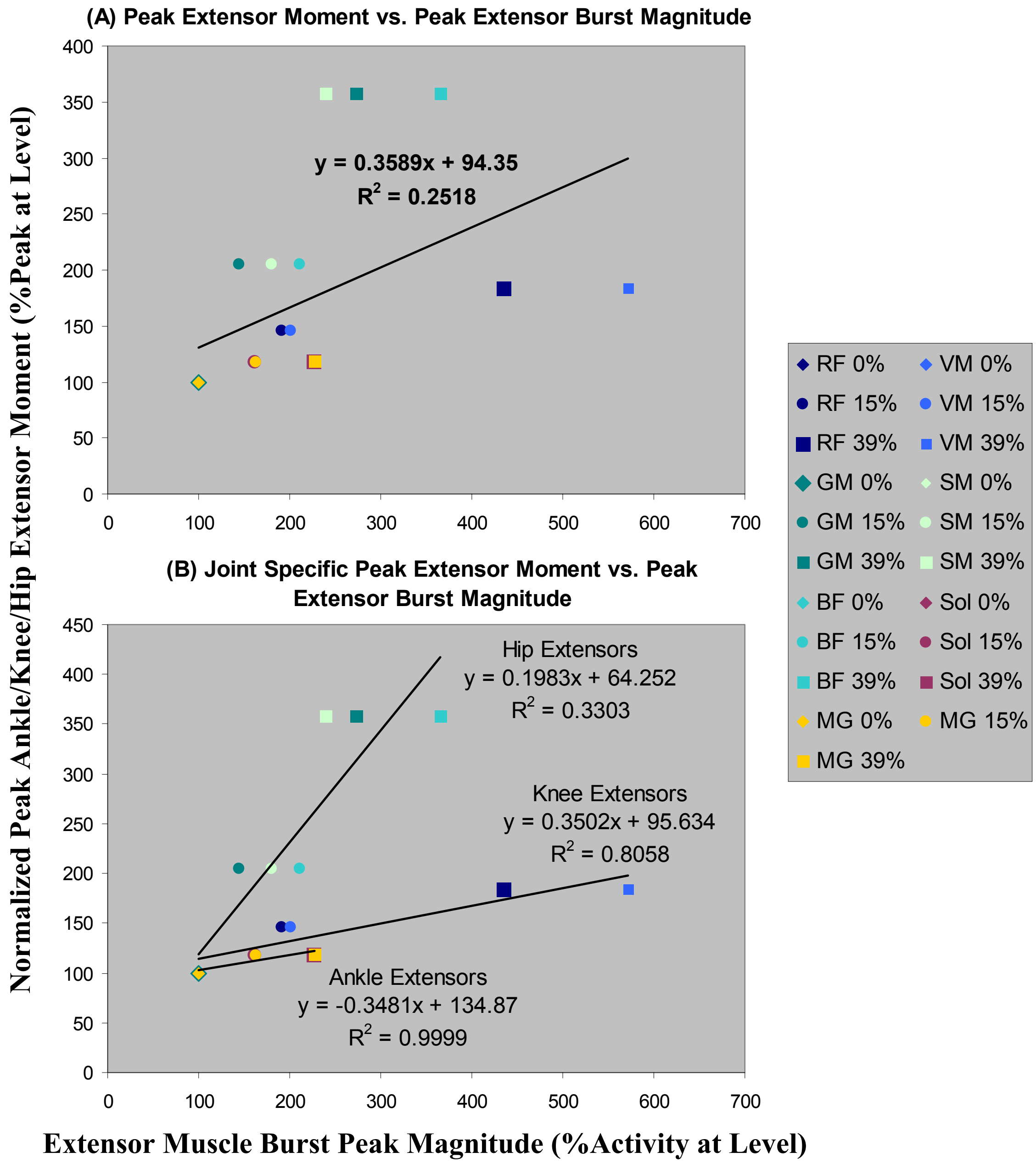


Figure 33. Correlations of Extensor Moment Peaks and Extensor Muscle Burst Peak Magnitudes during Upslope Walking for lower limb extensor muscles and the joints at which they act. The muscle burst peak magnitudes were normalized to the peak magnitude of the burst during level walking. The extensor moment peaks were also normalized to the peak extensor moment during level walking. Other details are as in Figure 32.

correlation lower than anticipated. Separate linear regressions were performed for each of the three hip extensor muscles (Figure 34), which resulted in strong correlations between the peak extensor moment and the peak EMG activity: SM (R = 0.982), BF (R = 0.999), GM (R = 0.987).

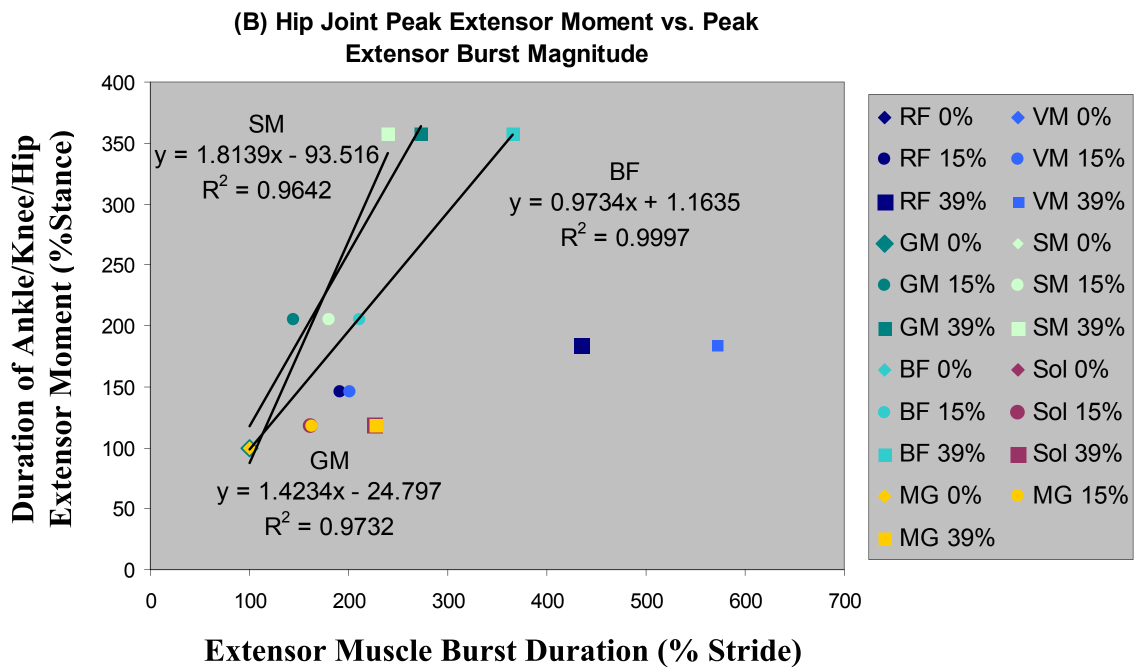


Figure 34. Correlations of Muscle Burst Durations during Upslope Walking. ♦ – data from level walking trials, ● – data from -15% trials, ■ – data from -39% trials. R^2 values are given for each hip extensor muscle.

Discussion

Joint Kinematics. The observed joint kinematic patterns (Figure 23) show good agreement with previous reports on human slope walking (Lange et al., 1996; Leroux et al., 1999, 2002; Wall et al., 1981). Collectively, these data indicate that for upslope walking all three joints are flexed more at heel strike and extend more during midstance, compared to level walking. These postural changes are consistent with the need to raise the limb for toe clearance and heel strike and then to propel the body up the incline, which would require muscle shortening during stance, particularly in the single joint knee and ankle extensors (Lange et al., 1996; Leroux et al., 1999). Similar to downslope walking, the relevance of these kinematic data lies in their association with muscle length changes, and the subsequent potential effect on force and proprioceptive feedback to the muscle pattern generator.

Ground Reaction Forces. The GRF data (Figure 24) are similar to the limited data available in the literature (McVay and Redfern, 1994); most noticeably, the propulsive force of the AP component increases. The normal force and resultant force are generally similar to those for level walking, with the exception of the decrease in force observed in midstance. Because the resultant force, which is the overall force acting on the foot, is similar during level and upslope walking, the output from cutaneous load sensors should also be similar in these two cases. In other words, feedback from these cutaneous sensors would be unlikely to contribute to the switch between control strategies if there were two different control strategies for these tasks. Instead, other sources of feedback, such as load receptors, may serve as signals for switching.

Joint Kinetics. The only reports on joint moments during upslope walking are for the ankle and knee moment for humans (in a thesis) (Buczek, 1990) and for quadrupeds (Gregor et al., 2001; Kaya et al., 2003). The joint moment data reported here again show some similarity to the quadruped model: in both cases the ankle plantarflexor moment increases with slope intensity and the knee moment is extensor during early stance. In humans the upslope walking knee and ankle joint moment patterns are similar to those during level walking, which is in agreement with the reported data (Buczek, 1990). The hip moment pattern, as reported here, is different from the level walking pattern. The peak hip extensor moment increases progressively with the upslope walking grade. The hip moment also has a progressively delayed transition to a flexor moment, from 48% of stance during level walking to 84% of stance at the steepest upslope grade. Again, this joint moment has not yet been reported in the literature, so no comparisons can be made. As with downslope walking, increases in the support moment are unequally distributed among the joints and the increased moment demands are not scaled linearly from those found during level walking.

Joint Power. The joint power data indicate that the muscle activity is increasingly concentric (active shortening) during upslope walking, as the muscles propel the body up and over the stance foot, against the force of gravity. This result has also been found in quadrupeds (Gillis and Biewener, 2002). The increases in power generation are large at the hip (> 900%) and knee (> 650%), and small at the ankle (< 100%). This increased shortening would be accompanied by an increase in force feedback from the golgi tendon organs, which could result in changes in the control strategy.

Hip Joint Biomechanics. The increases in the hip joint moment and power are explained in Figure 35. The stick figures indicate exemplar joint center and limb positions, GRF vectors (straight arrows), and a first approximation of the resultant applied moments at the knee joint (curved arrows, with the thickness representing the magnitude of the moment) at 25% of stance for level and upslope walking (+39% grade). During level walking the GRF passes near the hip joint center, which as a first approximation would cause a small applied flexor moment (thin curved arrow). The associated internal joint moment should therefore be a relatively small extensor moment; this is confirmed by the calculated joint moment. During upslope walking, however, the GRF vector is directed anterior to the hip joint center, resulting in a larger applied hip flexor moment (thick curved arrow). This, in turn, would require a larger internal hip extensor moment during upslope walking, which is again confirmed by the calculated joint moment. Finally, because the hip is extending in stance during level and upslope walking, power is generated by the hip extensors in both cases. The large hip extensor moment during upslope walking results in more power generation than during level walking.

Because of the similarities in the knee joint mechanics during walking downslope and down stairs, the hip joint mechanics observed here were compared to the mechanics from walking up stairs (McFadyen and Winter, 1988; Riener et al., 2002). . Unlike downslope walking, the mechanics for these two tasks were not found to be similar. The hip joint extensor moment did not increase during walking up stairs, but did during upslope walking (Riener et al., 2002). During upslope walking the hip extensors are the primary source of propulsion, while during walking up stairs the knee extensors performed this role (McFadyen and Winter, 1988).

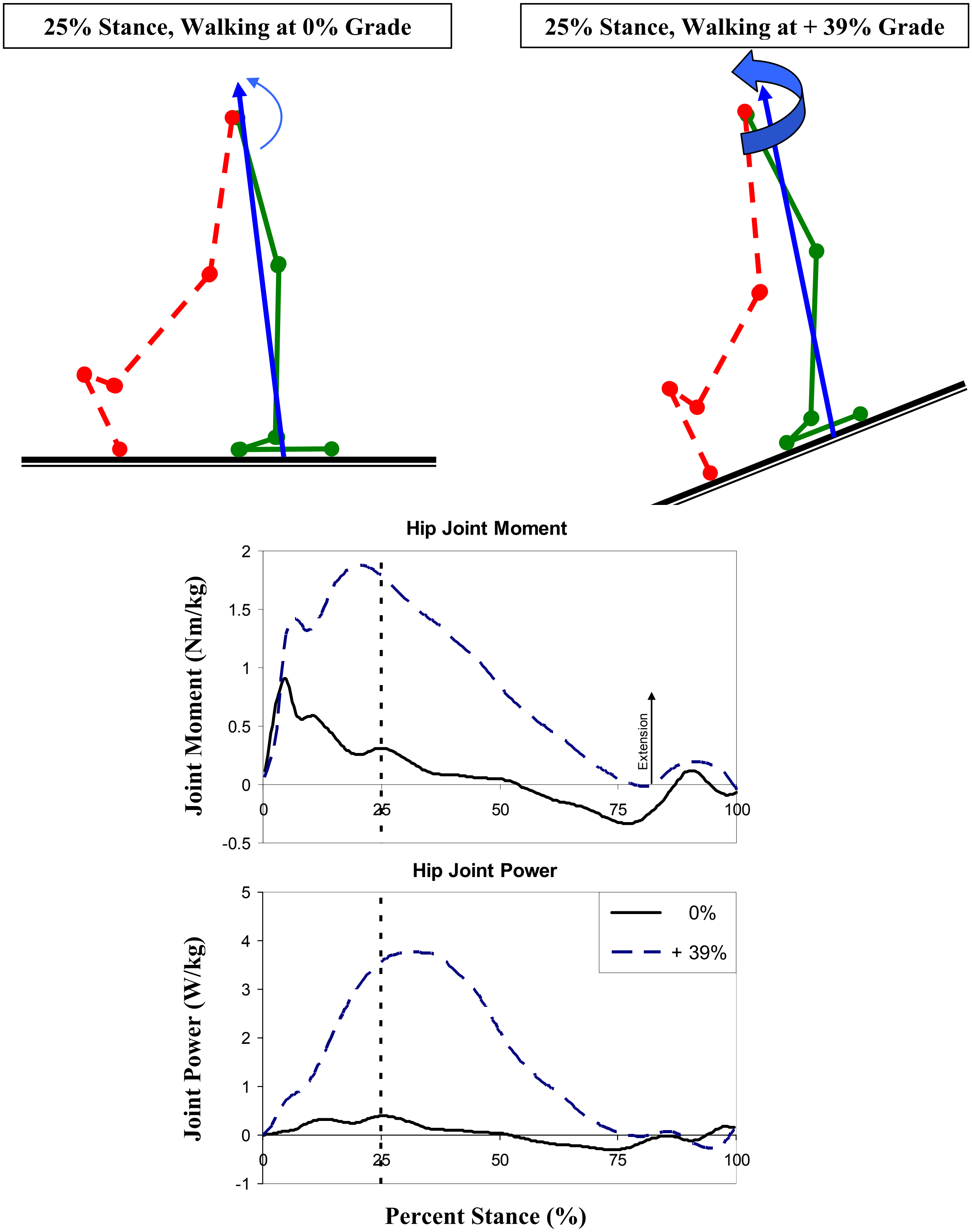


Figure 35. Schematic Model of the Lower Limbs at 25% Stance during level and upslope walking. The hip joint moment and power during stance phase of level and upslope walking are also shown, with 25% of stance marked by a vertical line.

Muscle Activity. The EMG activity observed during upslope walking is in general agreement with other reports in the literature. The increased GM and hamstring burst durations and constant MG burst durations are similar to previously reported burst timings (Tokuhiro et al., 1985). The observed increases in hamstring, RF, and VM activity and burst duration, and Sol and MG activity are in agreement with reported EMG data (Leroux et al., 1999). The VM and hamstring data are also consistent with a third study, which reports increases in early stance phase activity and burst durations for these muscles (Lange et al., 1996). The late stance TA burst observed here is different from both of the reported data sets: no late stance phase activity reported by Tokuhiro et al. (1985), and only very low level activity is reported by Leroux et al. (1999). These differences could be due to the high variability generally reported for the TA, or due to differences in preparation methods, electrode placement techniques, or data processing methods.

Joint Kinetics and Muscle Activity at the Hip and Knee. The large magnitude and delayed cessation of the hip extensor moment, compared to that at level walking, correspond to increases in both the magnitude and duration of hip extensor muscle activity, as demonstrated by the correlation in Figure 35. Figure 36 also illustrates this point. At +39% grade the mean GM activity is 260% of the mean activity during level walking, and the burst duration increases by 24% of the stride cycle. The duration and mean activity of the biarticular BF and SM also increased, further contributing to the large hip extensor moment, and in mid-stance also contributing to the knee flexor moment. The increased power generation at these joints suggests that these muscles are acting concentrically to move the body upslope.

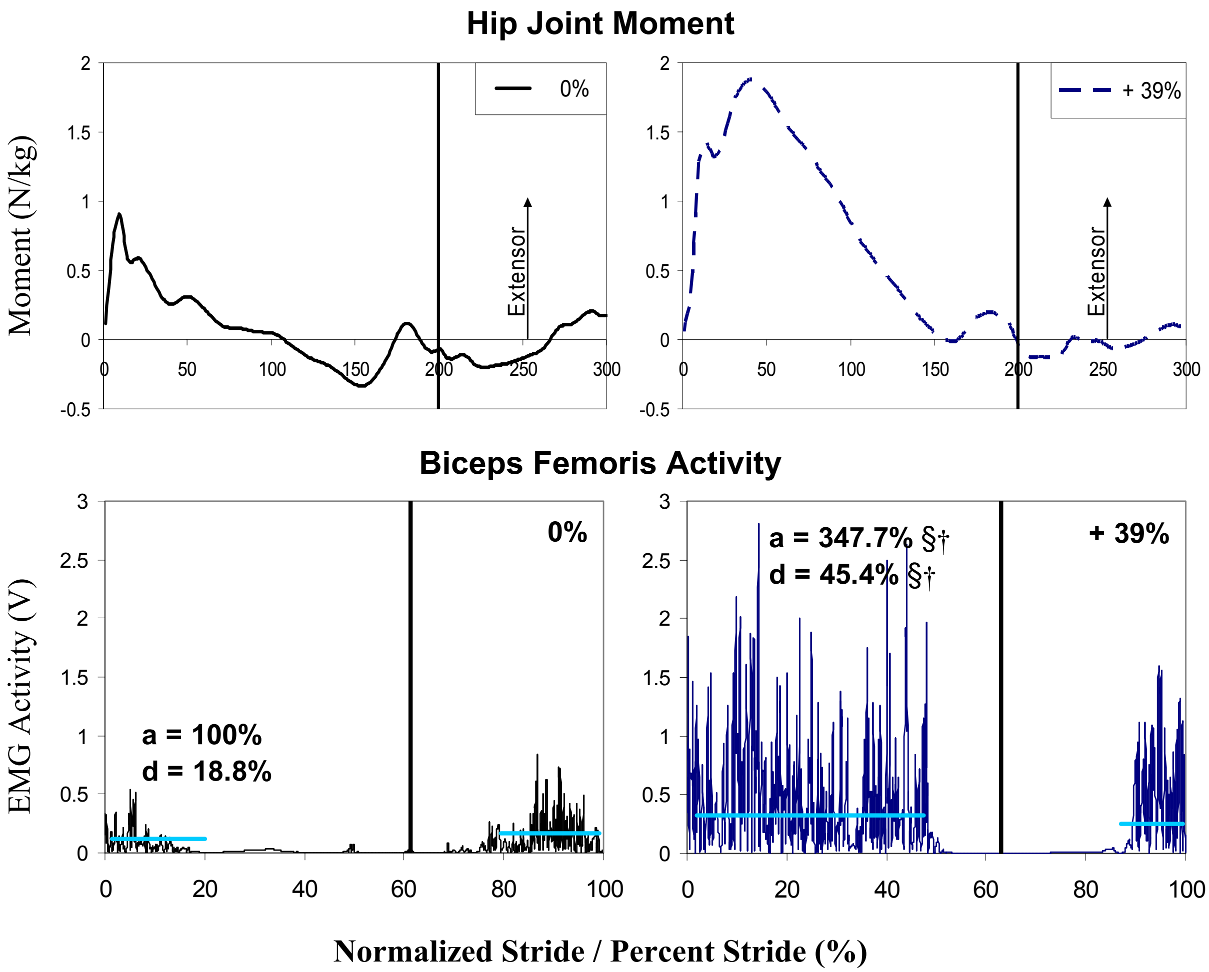


Figure 36. Hip Joint Moment and Muscle Activity during Upslope Walking, organized to more clearly show the relationship between the moment and muscle activity. Biceps Femoris activity was chosen as representative of the muscle activity at the hip joint for this figure. On the left are the hip joint moment and BF activity during level walking (0% grade); on the right are the data from +39% walking. The joint moment data are given in normalized stride (See Figure 26) and the EMG data are shown in percent stride (see Figure 29). It is clear from this figure that the increases in the mean BF burst activity (a) and BF burst duration (d) during stance correspond to the increased magnitude and duration of the hip extensor moment.

One unexpected finding involves the stance phase bursts of the knee extensors, VM and RF (also a hip flexor). During upslope walking the knee extensor moment in the first half of stance is not significantly different than that during level walking and there is no hip flexor moment (Figure 26). As a result, one might have expected RF and VM activity to be similar to that of level walking. This was not the case; both VM and RF burst durations and mean activity levels increased progressively with increases in the walking grade, with the largest increase in activity occurring during the first half of the stance phase burst. This mismatch between the joint moment and muscle activity is illustrated in Figure 37. One possible explanation for the unexpected increases in activity is that in order to maintain the knee extensor moment the RF and VM must increase their activity to also act as antagonists against the hamstrings, whose increased activity is needed to produce the large hip extensor moment during the first part of stance. During the second half of stance, when there is a knee flexor moment, the activity of the RF and VM is lower, resulting in a net flexor moment at the knee. The co-activation of the hamstrings and quadriceps throughout stance would also serve to increase the stability of both the knee and hip joints as the body is lifted up and over the stance foot.

Joint Kinetics and Muscle Activity at the Ankle and Knee. The mean activity levels of the MG and Sol bursts increase more during upslope walking than the activity level of the mid-stance TA burst, contributing to the larger peak plantarflexor moment. The joint moment data also indicate that there is no dorsiflexor moment in early stance of upslope walking as there is during level and downslope walking. However, the onsets of the MG, Sol, and TA bursts are not different during upslope and level walking; the MG and Sol

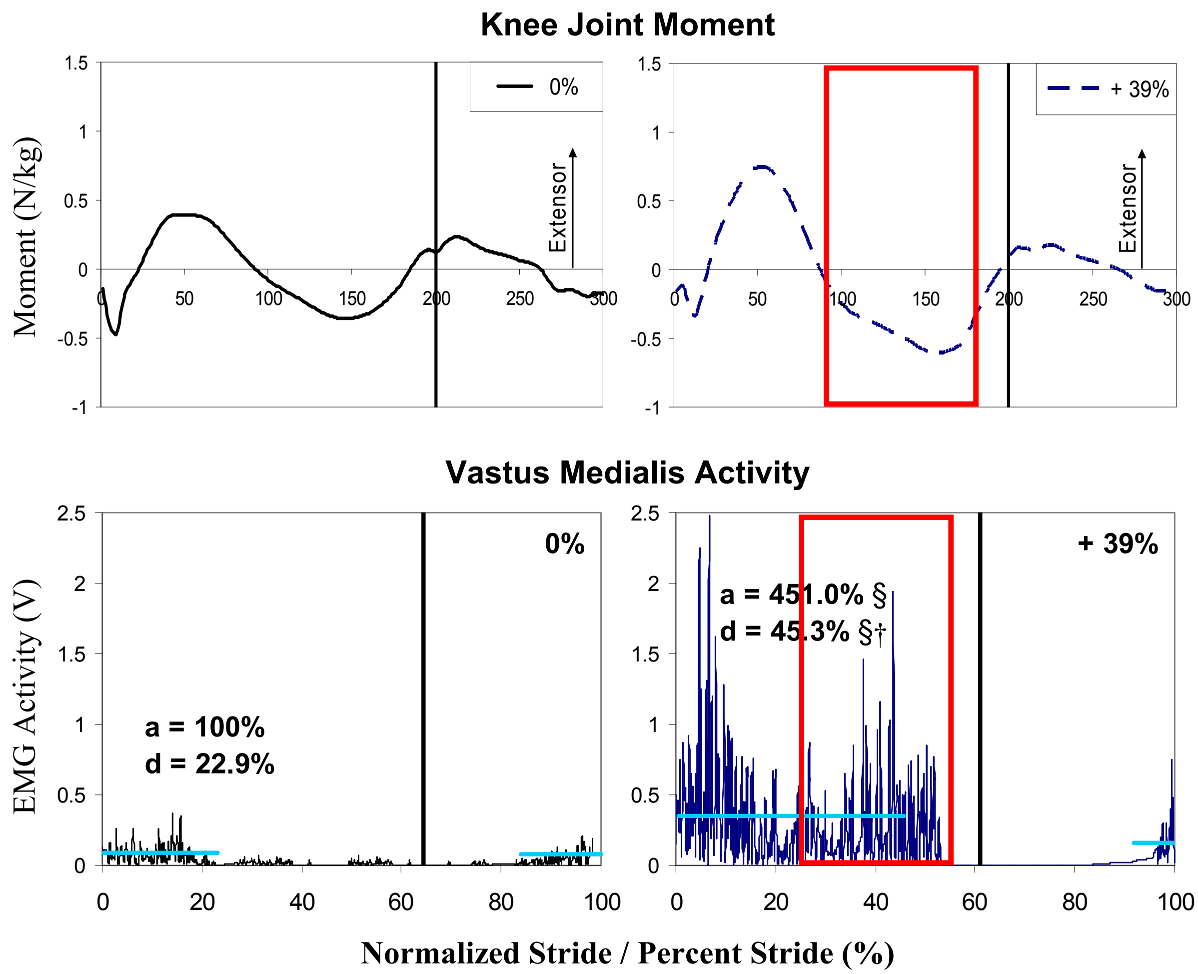


Figure 37. Knee Joint Moment and Muscle Activity during Upslope Walking, organized to more clearly show the relationship between the moment and muscle activity. Vastus Medialis activity was chosen as representative of the muscle activity at the knee joint for this figure. On the left are the knee joint moment and VM activity during level walking (0% grade); on the right are the data from +39% walking. The joint moment data are given in normalized stride (See Figure 26) and the EMG data are shown in percent stride (see Figure 30). The rectangles in the figures highlight the portion of stance phase where the knee extensor muscle activity does not correspond to the knee joint moment: the joint moment is flexor, but the mean VM burst activity (a) and VM burst duration (d) increase. This "paradoxical" activity of the knee extensor muscles is related to the joint power requirements and the activity of biarticular muscles.

activity does not occur earlier and the onset of TA activity is not delayed. This activity pattern is unexpected because some plantarflexor activity should be necessary to counteract the early stance TA activity in order to create the net plantarflexor moment. It is possible that the lateral gastrocnemius (not measured here) is active during this time and is the primary contributor to this moment. There is some evidence that the two heads of the gastrocnemius can be activated differentially in humans (Segal and Song, 2005).

The muscle burst patterns and joint moment patterns may be explained as follows: In early stance the ankle joint power was negative, indicating that the ankle plantarflexors were actively lengthening while the TA was actively shortening. Although it does not dominate the joint moment, this concentric TA activity represents the first phase of propulsion during upslope walking: the TA pulls the stance leg forward into an even more dorsiflexed position as the contralateral limb pushes off. Sol activity onset then occurs, which initiates ankle plantarflexion and therefore power generation and the primary phase of propulsion, where the body is 'pushed' forward over the stance limb by the plantarflexors and hip extensors. The onset of MG activity is later and it further contributes to the plantarflexor moment and propulsion, and to the knee flexor moment that occurs in the second half of stance. The biarticular action of the MG may also explain why its onset is later than that of the Sol; there is a knee extensor moment in the first half of stance, so activating the MG would counteract the knee extensor activity and require even greater extensor activation. By first using the Sol to plantarflex the ankle the need for this additional antagonist activity and Lombard's paradox are delayed (Gregor et al., 1985; Kuo, 2001). The final propulsive phase occurs during push off, when the co-activation of the TA and gastrocnemius muscles serves to stabilize the ankle joint.

Interestingly, the TA activity pattern provides an example of why it is necessary to study joint mechanics and muscle activity together. The TA is active during early stance in all walking conditions and only by looking at the joint mechanics is it evident that the contractions are concentric during upslope walking and eccentric during downslope walking.

Conclusions

The goal of this chapter was to discuss the data from forward upslope walking, and specifically the relationship between joint moments and EMG during this task. Based on the limited amount of data reported in the literature, it was expected that the changes in the joint moments would be largely magnitude changes rather than pattern changes. It was therefore hypothesized that changes in the magnitude of the muscle activity would be directly related to the changes in the magnitude of the joint moments, and that the patterns of both the joint moments and muscle activity would remain relatively unchanged. This task of upslope walking was also used because it was expected that power generation would increase significantly with the walking grade, which would distinguish the task from level and downslope walking.

The data reported here support the hypothesis that the magnitudes of the joint moments and muscle activity would be correlated. Although the overall lower limb correlation was moderate (R = 0.502), the knee and ankle joint-specific correlations and the hip muscle-specific correlations were very high (all $R > 0.89$). However, from the data presented here it is evident that the joint moment and muscle firing patterns were different during level and upslope walking: ***The hypothesis that the patterns of joint moments and muscle activity would remain unchanged is refuted by these data***. The ankle and hip joint moment patterns changed during upslope walking: the duration of the plantarflexor and extensor moments increased compared to level walking. In addition, it appears that more complex changes in muscle firing patterns occur than would be expected from the relatively simple changes in the joint moment patterns; the ankle plantarflexor burst durations did not change, and the knee extensor burst durations changed unexpectedly. As a result, the correlations were poor between the extensor moment durations and extensor muscle burst durations, with the exception of the joint-specific correlation for the hip. Hence, during upslope walking the joint moments alone are not enough to predict the observed muscle activity patterns. For example, the knee joint moment pattern during upslope walking is not significantly different than during level walking, so one might expect that the knee extensor muscle firing patterns during upslope and level walking would also be similar. However, this was not the case; the magnitude of the knee extensor muscle activity and burst duration increased significantly during upslope walking compared to level walking. A detailed examination of the joint moments and muscle activity showed that this increased knee extensor muscle activity was needed to counteract the increased activity of the biarticular hip extensors/knee flexors contributing to the increased hip extensor moment. Looking at the joint moments in isolation, however, this change could not be predicted. During downslope walking, the changes in the joint moment patterns could be used to predict the changes in muscle activity. This difference between upslope and downslope walking could be due to the differences in the power demands. In this case the muscles are actively shortening to propel the body up the slope against the force of gravity, rather than just controlling the motion as the body is pulled down the slope. This increased power output requires activation of biarticular muscles to meet the high joint moment demands. Activation of biarticular muscles, in turn, requires activation of antagonist muscles to counteract unwanted joint moments and explain Lombard's paradox (Gregor et al., 1985). It is this co-activation that makes predicting muscle activity from the joint moments difficult during upslope walking. The tasks of backward upslope and downslope walking will be used in Aim 3 to further explore the idea that joint power requirements are responsible for deciding the muscle activity patterns.

If the same motor program was governing upslope and level walking, the level walking muscle firing pattern should be conserved. As in the discussion of downslope walking, this could mean that the onset times and/or burst durations for each muscle would be similar across all conditions, with changes only in the amplitude of the muscle activity. It is also possible that the burst durations could increase (or decrease) uniformly for all the muscles, as was observed for the thigh muscles here; the burst durations of all five muscles (GM, BF, SM, RF, VM) increase by approximately the same percent of the stride cycle. However, the three shank muscles do not support this idea: the burst onsets and durations are independent of the walking grade, and the only changes observed are in the burst amplitudes. Collectively, these EMG data may indicate a feedback-mediated change to the control system. Power output increases at all the joints, but it increases most substantially at the hip and knee joints (Figure 28). The corresponding increase in

GTO output could inhibit the cessation of thigh muscle activity, resulting in relatively uniform increases in thigh muscle activity. Because the change in the peak joint moment and in power generation at the ankle was relatively small the GTO feedback from these muscles may not have been enough to alter the control strategy. Alternatively, the changes in feedback could cause a transition to a new, upslope walking control strategy. Distinguishing between these two possibilities would be difficult in humans, but at least the data here demonstrate that some change in the level walking control strategy is necessary to meet the unique challenges of upslope walking.

The data presented here and in Chapter 5 can also be used to address the question posed in Chapter 1; whether the change in surface grade would invoke a gradual change in the joint mechanics, or an all-or-nothing effect with a transition grade, as seen in the wedge-stepping. The upslope and downslope walking joint moment patterns are very similar at 15% and 39%, although the peak magnitude often increases from the 15% to the 39% grade. These data indicate that the same control strategy is likely governing slope walking at both 15% and 39%. The question that then arises is at what grade does the nervous system switch from the level walking strategy to the slope walking strategy? Earhart and Bastian (2000) studied control strategies for subjects taking a single step up a wedge and identified two distinct strategies, one for grades of 10˚ (18%) or less and another for grades of 20˚ (36%) or higher, with a transition grade at 15˚ (~27%). The similarities between our +15% and +39% kinematic data do not support this 27% transition grade. Differences in the tasks may explain this conflict: the level walking strategy may be sufficient for a single step on a 10˚ (18%) grade, but inadequate when sustained upslope walking is required. This idea is supported by another study that investigated the biomechanics of the first step on a slope as subjects walked up an incline (Prentice et al., 2004). The authors suggest that a transition grade exists between 6˚ and 9˚ (11% and 16%) of ramp inclination where the movement strategy changes to prepare the limb for an elevated heel strike with increased propulsive requirements. This adaptation would not be necessary if the subjects were simply stepping to a flat surface, as in Earhart and Bastian's (2000) study. In light of these reports and our data, we would expect that for grades of +10% or less the level walking control strategy will prevail, resulting in similar kinematic and EMG patterns to level walking. Exploration of this idea for both upslope and downslope walking warrants further investigation.

CHAPTER 7

BACKWARD SLOPE WALKING

Introduction

Results from the investigation of backward slope walking, which was used to further examine the robustness of the movement control strategies employed by the nervous system for forward slope walking, will be discussed in this chapter. Backward walking was chosen as the locomotor perturbation because in humans during level walking the kinematics are exactly reversed from the forward walking kinematics (Grasso et al., 1998; Thorstensson, 1986; Winter et al., 1989). A primary difference between forward and backward walking tasks is that backward walking is considered a novel form of locomotion. While this may be the case, it has been suggested that in quadrupeds the control parameters for these tasks are similar (Buford and Smith, 1990; Perell et al., 1993). The evidence that forward and backward walking in humans are driven by the same movement control strategy is not conclusive (Grasso et al., 1998; Lamb and Yang, 2000; Thorstensson, 1986; Winter et al., 1989). Another difference between these tasks is that the power requirements are reversed; power generation during forward level walking is replaced by power absorption during backward level walking (Winter et al., 1989). This reversal would also be expected to occur in the corresponding forward and backward slope walking conditions. Given the differences in the power requirements and task familiarity, backward upslope and downslope walking were used to test the robustness of the movement control strategies used for forward upslope and downslope walking. In order to present this information most clearly, Aim 3 was divided into three parts: i) to define the relationship between the forward and backward slope walking tasks, ii) to analyze the task of backward upslope walking, and iii) to analyze the task of backward downslope walking.

Part 1 of this aim was to establish the relationship between forward and backward slope walking. Because the biomechanics of backward slope walking have not been analyzed, it was unclear how they would compare to the biomechanics of forward slope walking. Based on comparisons between forward and backward level walking, *it was hypothesized that the joint kinematics and joint moments would be similar and the joint power would be reversed between backward upslope walking and forward downslope walking, and between backward downslope walking and forward upslope walking* (Grasso et al., 1998; Thorstensson, 1986; Winter et al., 1989).

After a relationship between forward and backward slope walking was established, an analysis of the biomechanics and muscle activity during backward slope walking was completed. This analysis relied on the same data processing and statistical analysis techniques outlined in Chapter 4 and used in the analysis of forward slope walking. Part 2 of this aim studied backward upslope walking, which was intended to parallel the investigation of forward downslope walking from Aim 1 (Chapter 5). Part 3 of this aim investigated backward downslope walking, as a complement to the study of forward downslope walking in Aim 2 (Chapter 6). It was expected these investigations of backward upslope and downslope walking would provide further insight into the robustness of the movement control strategies for slope walking.

One definition for the 'robustness' of a movement control strategy is its ability to govern multiple tasks, or perturbations of its primary task. The authors that studied backward walking were attempting to show that the forward walking movement control strategy was robust enough to also control backward walking (Grasso et al., 1998; Thorstensson, 1986; Winter et al., 1989), a concept that has been suggested to be true in infants (Lamb and Yang, 2000). However, their data show differences in the muscle firing patterns for forward and backward level walking, indicating it is unlikely that the same control strategy is valid for both walking directions. Assuming that there are different control strategies for forward and backward walking, it is then possible to test the 'robustness' of *each* control strategy by using the slope perturbation. In other words, if the grade is changed to +15% during forward and backward walking (in the same direction), requiring forward upslope walking and backward downslope walking, do the same changes occur in the initial control strategy? Is that also true when the grade is changed to -15%, requiring forward downslope walking and backward upslope walking? Because the changes in the joint kinetics were expected to be similar between these corresponding tasks, the changes in the control strategies with slope were also expected to be the same. As stated in Part 1 of this Aim, it was expected that the power requirements (and therefore the feedback signals) would be reversed between these corresponding tasks. For example, during forward downslope walking power absorption and length feedback would be emphasized; during the kinematically and kinetically similar task of backward upslope walking, power generation and force feedback were expected to be emphasized. It was expected that the movement control strategies would be robust in spite of the expected changes in power requirements, such that the relationship between muscle firing patterns and joint kinetics would be similar for locomotor tasks with similar patterns of joint kinetics. This line of reasoning and the hypotheses from forward downslope and forward upslope walking led to the following expectations for backward slope walking: For backward upslope walking, it *was hypothesized that changes in the pattern and magnitude of the muscle activity would be directly related to the changes in the pattern and magnitude of the joint moments.* For backward downslope walking it was *hypothesized that the changes in the magnitude of the muscle activity would be directly related to the changes in the magnitude of the joint moments*.

Part I: A Brief Comparison to Forward Walking

As mentioned, the joint angles and moments were expected to be similar during the tasks of forward and backward slope walking, but the joint powers were expected to be reversed. The joint angles are compared in Figure 38, the joint moments and the support moments are compared in Figures 39 and 40, respectively, and the joint powers are compared in Figure 41. For simplicity, only the data from level walking and the steepest slopes (+/- 39%) are shown. The backward walking data, as collected, began and ended at toe strike, with stance phase occurring first. In order to compare these data to forward walking data they are time-reversed, such that swing phase occurs first and stance phase begins with heel off (Winter et al., 1989). The data are then switched so stance phase is plotted first and the stride begins and ends with heel off (Vilensky et al., 1987). This manipulation makes the task kinematically similar to forward walking, but causes some discontinuities at the stance-swing transition. The swing-phase joint moments were similar across all the walking conditions, so only the stance phase moments are shown in Figures 39 and 40.

The results of the comparison of the level forward and backward walking joint angle curves are in agreement with similar comparisons in the literature (Thorstensson, 1986; Vilensky et al., 1987; Winter et al., 1989). Backward level walking shows decreased ankle plantarflexion during late stance and swing and increased knee flexion and hip flexion in midstance, compared to forward walking. The level walking joint and support moments are also in agreement with reported data (Winter et al., 1989), with a decrease in the ankle plantarflexor moment, no knee flexor moment in late stance, and a decreased peak hip extensor moment magnitude during backward walking, compared to forward walking.

Backward upslope walking joint angle data has been reported only once in the literature, and only select values of the joint angles were given (Cipriani et al., 1995). The authors did not make any comparisons between forward and backward walking in their study (Cipriani et al., 1995). Joint angle curves for backward downslope walking have not yet been reported, and therefore there is no existing comparison to forward slope walking. Figure 38 shows differences in the magnitudes of the joint angle curves similar to those observed in the level walking comparison: the backward slope walking conditions show decreased ankle plantarflexion during swing and increased knee flexion and hip flexion in midstance, compared to the corresponding forward slope walking conditions. The patterns of the joint angle curves were similar between conditions. In general, the forward upslope and backward downslope joint angles are similar, as are the forward downslope and backward upslope joint angles.

Backward slope walking joint moments have not been reported in the literature, so this is the first comparison of such data to forward slope walking. From Figure 39 it is clear there are some differences in the corresponding joint moment curves (forward downslope vs. backward upslope, and forward upslope vs. backward downslope). The backward walking ankle joint moments tend to be smaller than the corresponding forward walking moments. At the knee joint the backward walking extensor moment is always smaller than the corresponding forward walking extensor moment in the first half of stance and larger than it in the second half of stance. During late stance there is no knee flexor moment during backward level or backward downslope walking, as there is in the forward walking conditions; all of the backward walking conditions resemble forward downslope walking in that respect. The peak hip joint moments are smaller during backward walking than during the corresponding forward walking conditions. The transition from a hip extensor moment to a hip flexor moment is also more tightly clumped for the three backward walking conditions than for the forward walking conditions. The support moment (Figure 40) during backward walking is smaller than that during forward walking in early stance, but greater than during forward walking in late stance. In spite of these differences, the changes in the joint moment patterns are similar for the corresponding forward and backward walking conditions. As in the previous studies, therefore, the tasks of forward and backward walking can be considered kinematically and kinetically similar.

As expected, the joint powers were reversed during forward and backward level walking (Winter et al., 1989), and also during the slope walking conditions (Figure 41). The power curves for forward downslope walking and backward upslope walking were generally almost mirror images of each other, as were the curves for forward upslope walking and backward downslope walking. There were some differences in the curves for the knee joint power in late stance; there were regions of simultaneous generation and absorption for the forward and backward walking curves at 0% and 15% grades. There were also some differences in the peak magnitudes between the corresponding conditions; peak absorption at the ankle was higher during backward walking at -39% than during forward walking at +39%, generation at the knee was greater during backward walking at +39% than during forward walking at -39%, and peak absorption at the hip joint was lower during backward walking at -39% than during forward walking at +39%. In spite of these magnitude differences, the joint powers were generally reversed between the corresponding forward and backward slope walking conditions.

For the remainder of this chapter backward slope walking will be considered independently from forward slope walking, with the same type of analysis that has been performed for forward walking in the previous chapters. The data will no longer be time reversed or switched. The mechanics will be reported for all walking grades, and the EMG data will be analyzed in a similar fashion as for forward walking.

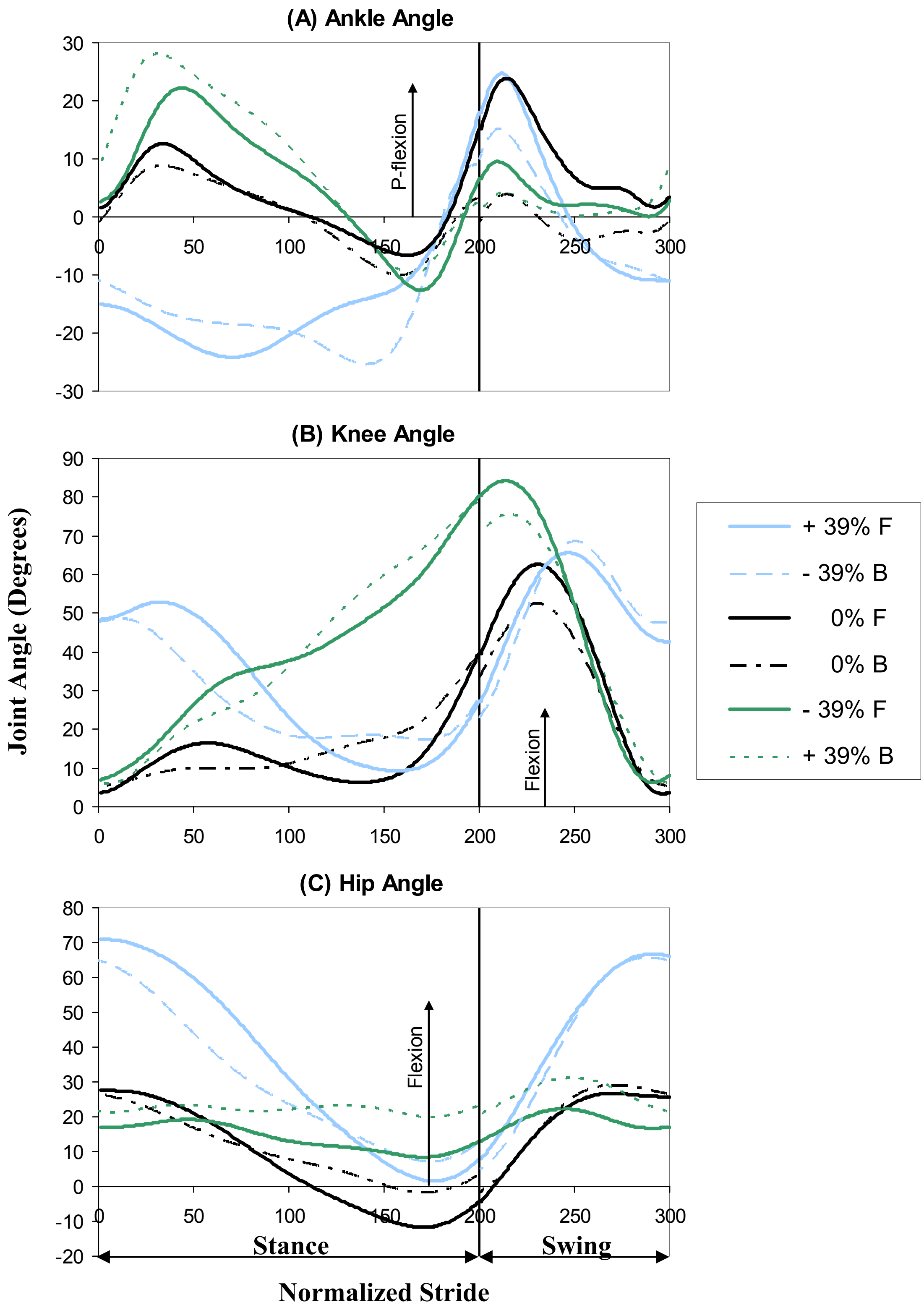


Figure 38. Comparison of Forward and Backward Walking Joint Angles. Backward walking angles (B) are time-reversed (HO is at 0, TS is at 200) and swing and stance are flipped to match the forward walking (F) angles (hence the discontinuity in the curves). All curves are ensemble averages, vertical line marks toe off for forward walking.

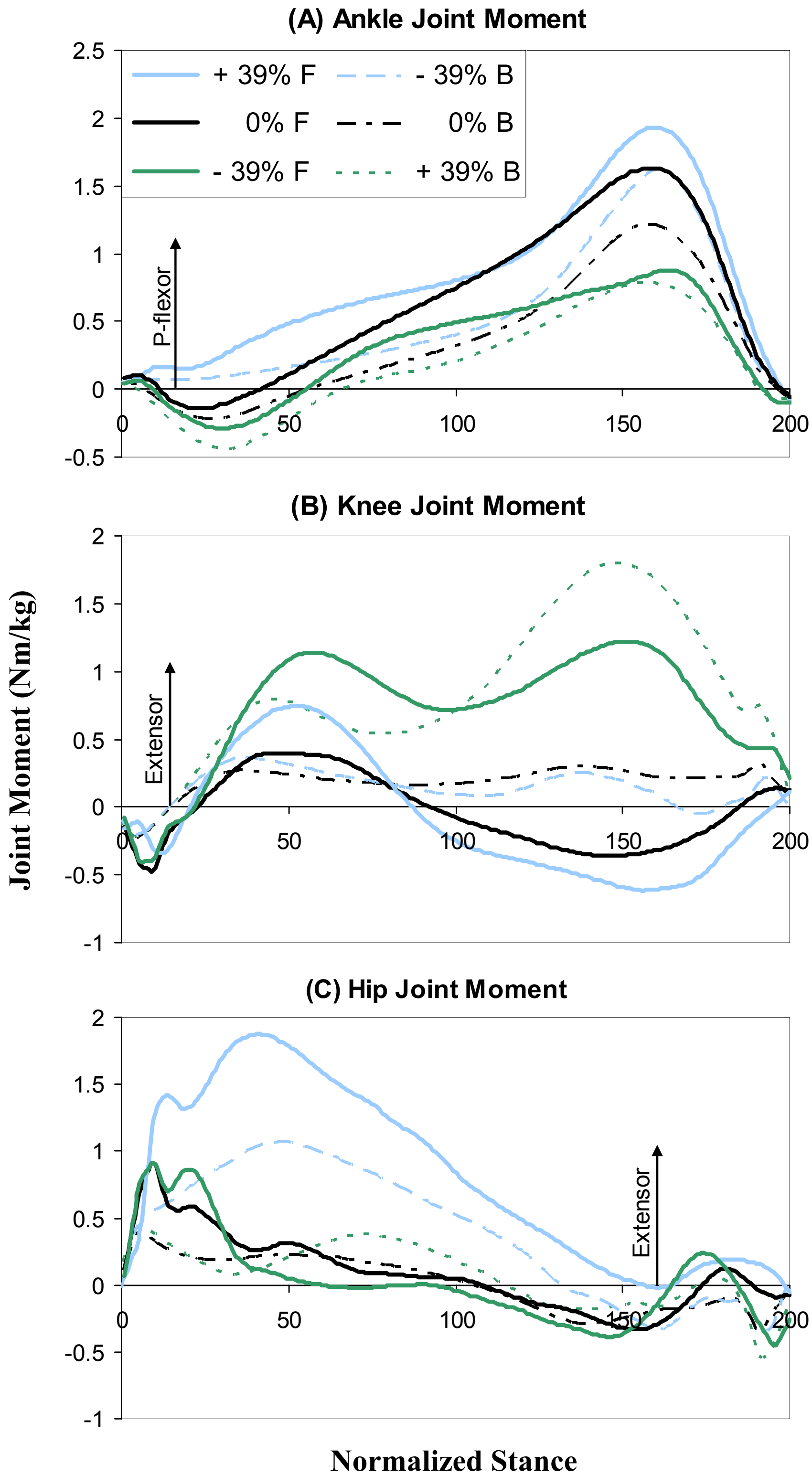


Figure 39. Comparison of Forward and Backward Walking Joint Moments. Stance was normalized to 200 points for each trial before averaging. Backward moments (B) are time-reversed (HO is at 0, TS is at 200); forward walking (F) begins with HS.

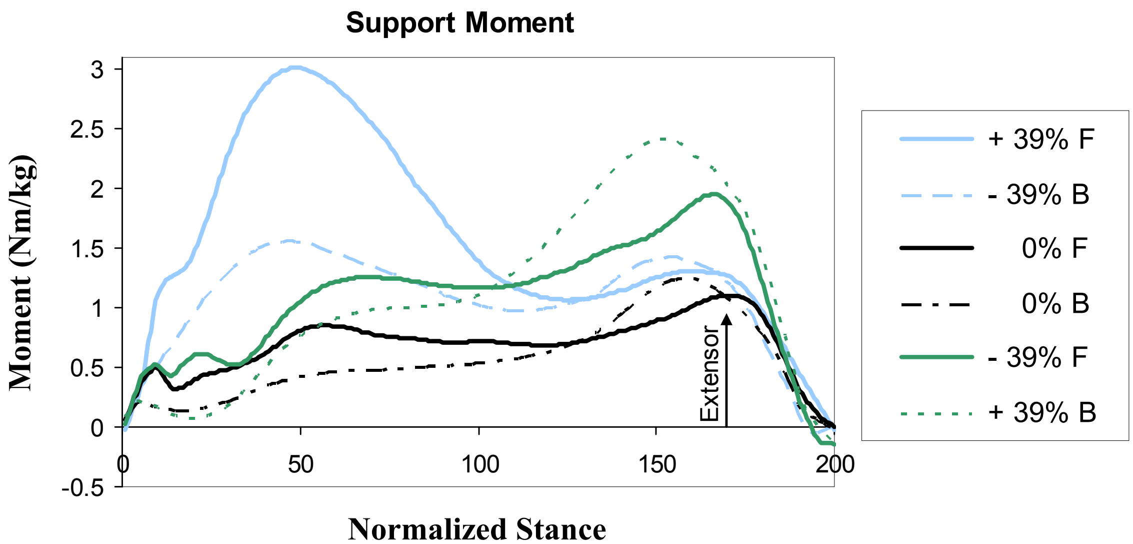


Figure 40. Comparison of Forward and Backward Walking Support Moments. Stance was normalized to 200 points for each trial before averaging. Backward moments (B) are time-reversed (HO is at 0, TS is at 200); forward moments (F) begin with HS.

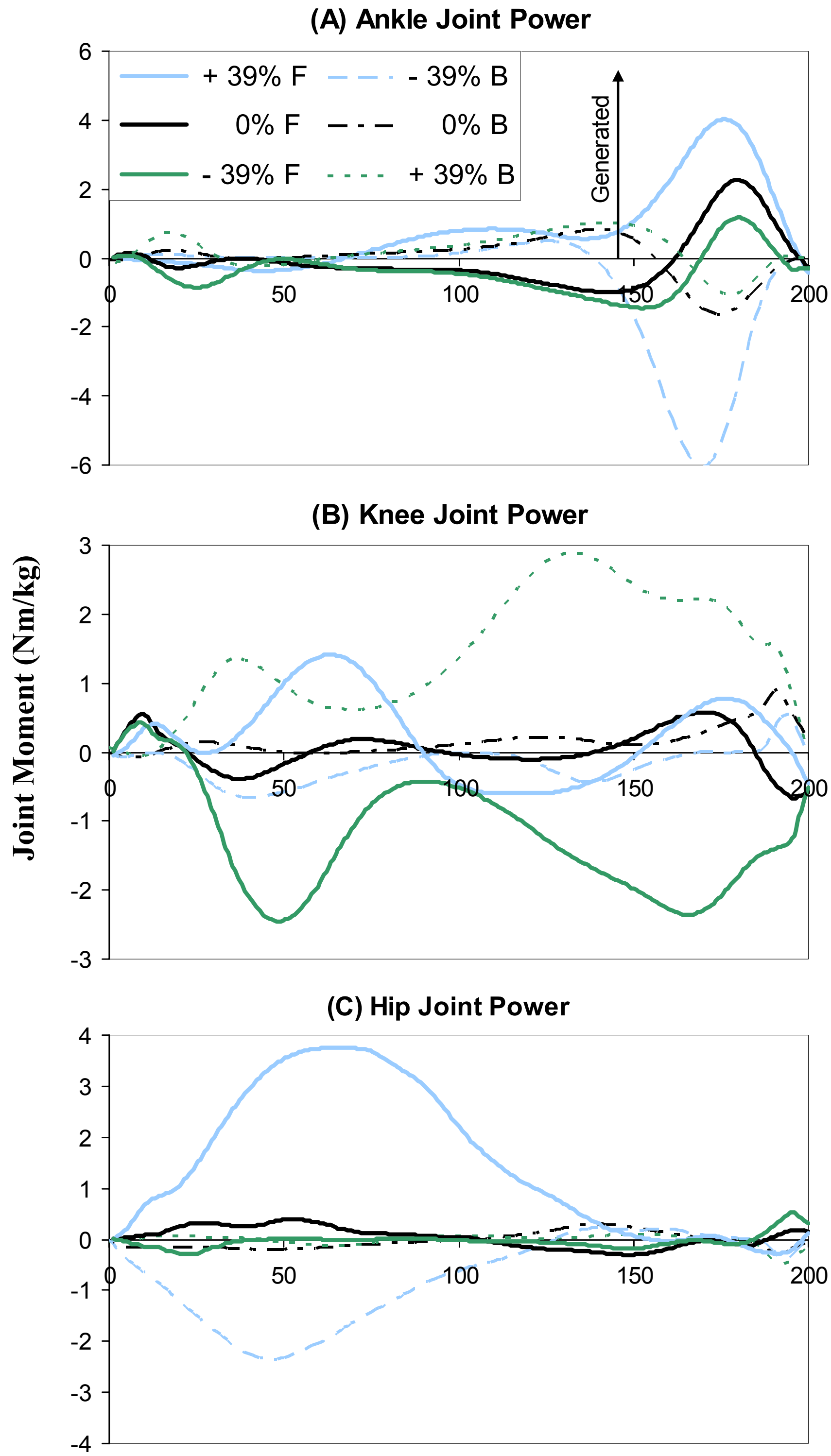


Figure 41. Comparison of Forward and Backward Walking Joint Powers. Stance was normalized to 200 points for each trial before averaging. Backward moments (B) are time-reversed (HO is at 0, TS is at 200); forward moments (F) begin with HS.

Part II: Backward Upslope Walking

Joint Kinematics and GRF. Although the kinematic and GRF data from backward upslope walking do not directly relate to the questions posed about this task in Aim 3, they should be included in this work so the biomechanical description of the task is complete. To accomplish this, but reduce the length of this chapter, those data are presented in Appendix B.

Joint Kinetics. The joint moments (Figure 42) were similar for all grades during swing, but showed marked changes during stance as the walking grade changed. Two points of interest, KM4 and HM1, were unaffected by the changing ramp grade (ANOVA); the results of the follow up comparisons for the other points are given in Table 21. Compared to the level walking joint moments, increasing the slope during backward walking resulted in a decreased peak ankle plantarflexor moment (Figure 42A, AM1) and increases in the ankle dorsiflexor moment (AM2) and the knee extensor moment (Figure 42B, KM1, KM2, KM3). The first peak of the knee extensor moment (KM1) increased progressively with the walking grade, while increases in points KM2 and KM3 were more uniform (i.e. there were no significant differences in the values between +15% and +39% walking grade). The hip moment (Figure 42C) during upslope walking was statistically similar to that for level walking at all three points of interest. The support moment (Figure 43) increased significantly in the first half of stance (SM1) as the grade increased, but not in the second half of stance (SM2).

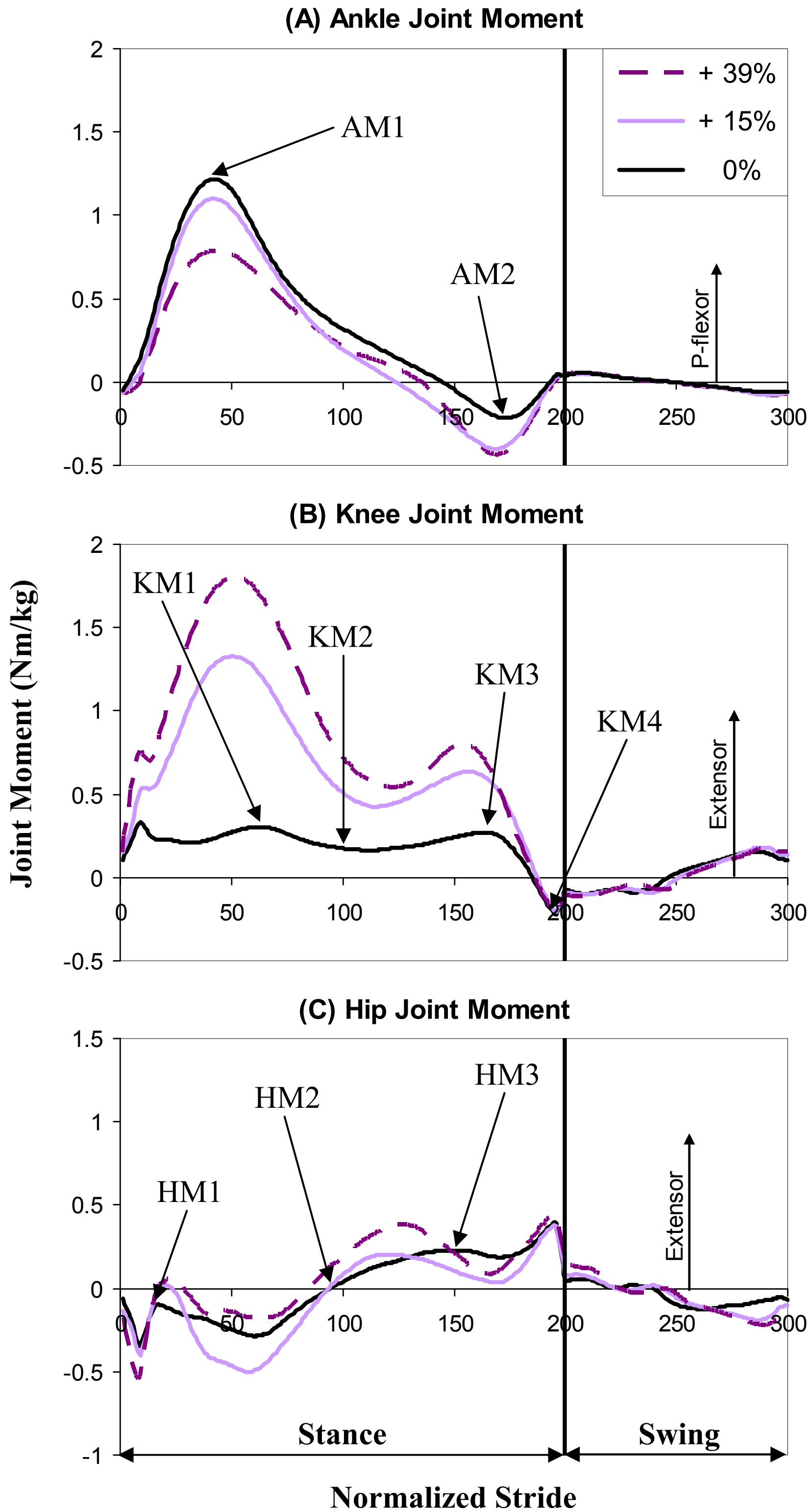


Figure 42. Average Joint Moments during Backward Upslope Walking. Positive moments are plantarflexor and extensor, as indicated. Stance was normalized to 200 points and swing to 100 points before ensemble averaging. Vertical line marks heel off.

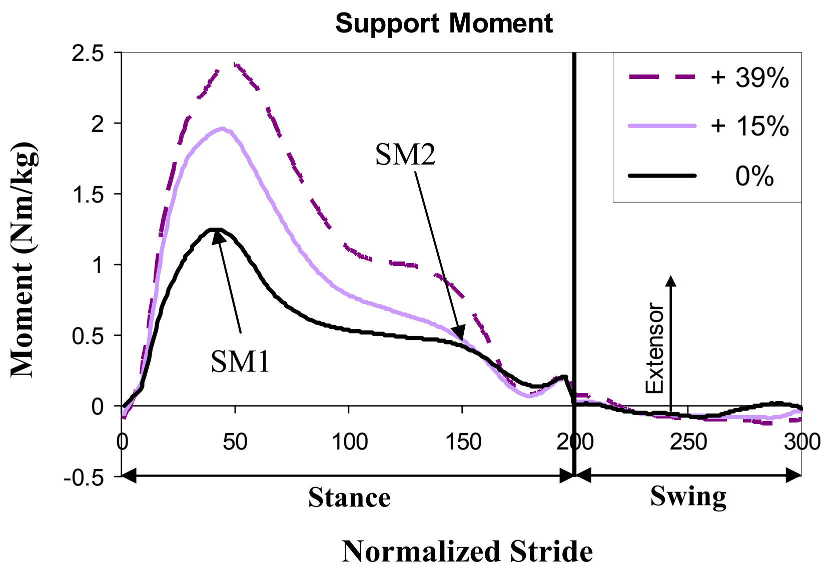


Figure 43. Average Support Moment during Backward Upslope Walking (Winter, 1980). Stance begins with toe strike, vertical line marks heel off.

Table 21. Backward Upslope Walking Joint Moment POI Data

	0%	+ 15%	+ 39%
AM1	1.26 (0.25)	1.13 (0.30)	0.84 §† (0.16)
AM2	-0.23 (0.08)	-0.42 § (0.11)	-0.47 § (0.10)
KM1	0.38 (0.30)	1.35 § (0.20)	1.85 §† (0.26)
KM2	0.17 (0.28)	0.50 § (0.19)	0.70 § (0.26)
KM3	0.31 (0.18)	0.66 § (0.24)	0.83 § (0.38)
KM4	-0.23 (0.10)	-0.19 (0.13)	-0.21 (0.10)
HM1	-0.06 (0.31)	0.09 (0.21)	0.20 (0.21)
HM2	96.89 (52.29)	97.78 (34.44)	79.56 (47.15)
HM3	0.26 (0.24)	0.21 (0.19)	0.37 (0.30)
SM1	1.31 (0.28)	2.08 § (0.27)	2.66 §† (0.40)
SM2	0.40 (0.28)	0.46 (0.25)	0.76 (0.47)

Note: Mean (SD) of the joint moment POI (values N*m/kg, except HM2 is normalized stride). Shading indicates no grade effect was detected by the ANOVA; § indicates a significant difference from 0% grade; † indicates a significant difference from 15% (in the same walking direction), ($p<0.05$).

Joint Power. Ensemble average joint power curves are shown in Figure 44. During backward upslope walking peak power generation increased slightly at the ankle and hip (Figure 44A and C), although for the majority of stance the power curves were similar for level and upslope walking. At the knee, however, power generation increased markedly (Figure 44B), suggesting that concentric muscle activity at this joint increased. The percent of stance phase during which power was generated was similar at the ankle during level and upslope walking, but increased at the knee and hip during upslope walking (Table 22).

Table 22. Power Generation during Backward Upslope Walking Stance

	0%	+ 15%	+ 39%
Ankle	67.32	62.44	67.32
Knee	81.95	91.71	93.17
Hip	34.63	41.46	63.41

Note: Values are percent of stance phase (%), and are the average across all subjects.

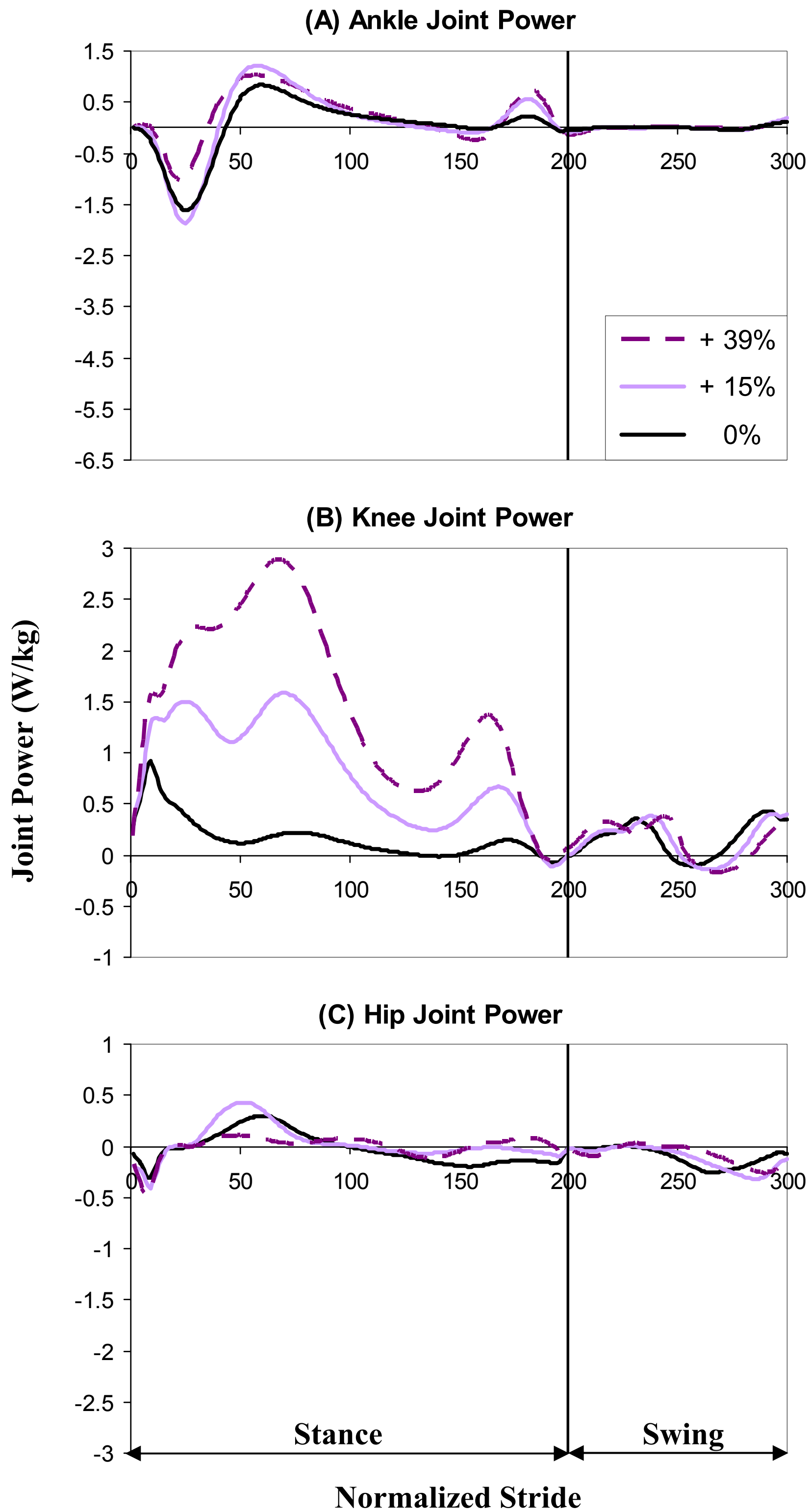


Figure 44. Average Joint Power during Backward Upslope Walking. Positive power is generation, negative is absorption. Stance and swing phases of each trial were normalized before ensemble averaging. Vertical line marks heel off.

Muscle Activity. Representative and group average EMG data are presented in Figures 45 through 47; the ensemble average smoothed data are given in Appendix B. In most cases the EMG burst patterns at +15% grade and +39% grade were the same, with only small differences in the burst durations. There were only two cases where the burst pattern at +39% was different from that at +15%: the TA at +39% exhibited a burst in early stance that was not present at +15%, and the MG did not exhibit a late swing burst at +39%.

During upslope walking the mean GM activity levels increased progressively, although the burst onset and duration did not change (Figure 45). The duration of the hamstrings activity (Figure 45) increased during upslope walking, such that the hamstrings were active for the majority of stance, although the two muscles achieved this increase differently. The onset of the BF stance phase burst was earlier during upslope walking, and the burst duration increased. In the SM a new burst occurred in early stance and the mid-stance burst remained statistically the same as during level walking. In both the BF and SM the mean activity levels during upslope walking were statistically the same as during level walking. The RF and VM were active for the majority of stance during level and upslope walking (Figure 46). The durations of the stance RF and VM bursts did not change during upslope walking, but the mean activity levels increased progressively with grade. The Sol, MG, and TA muscles (Figure 47) also showed some changes during upslope walking, compared to level walking. The single stance phase Sol burst that occurs during level walking split into two bursts during upslope walking. The activity levels of both bursts were compared to the level walking burst activity; the activity increased from level with walking grade for the first burst, but not for the second.

The duration of the first stance phase MG burst increased during upslope walking, and the onset of the second stance phase burst was delayed. The burst that begins at the stance-swing transition did not change with the walking grade. Finally, the TA exhibited a new early stance burst during +39% walking, as mentioned, and the onset of the mid-stance burst was delayed.

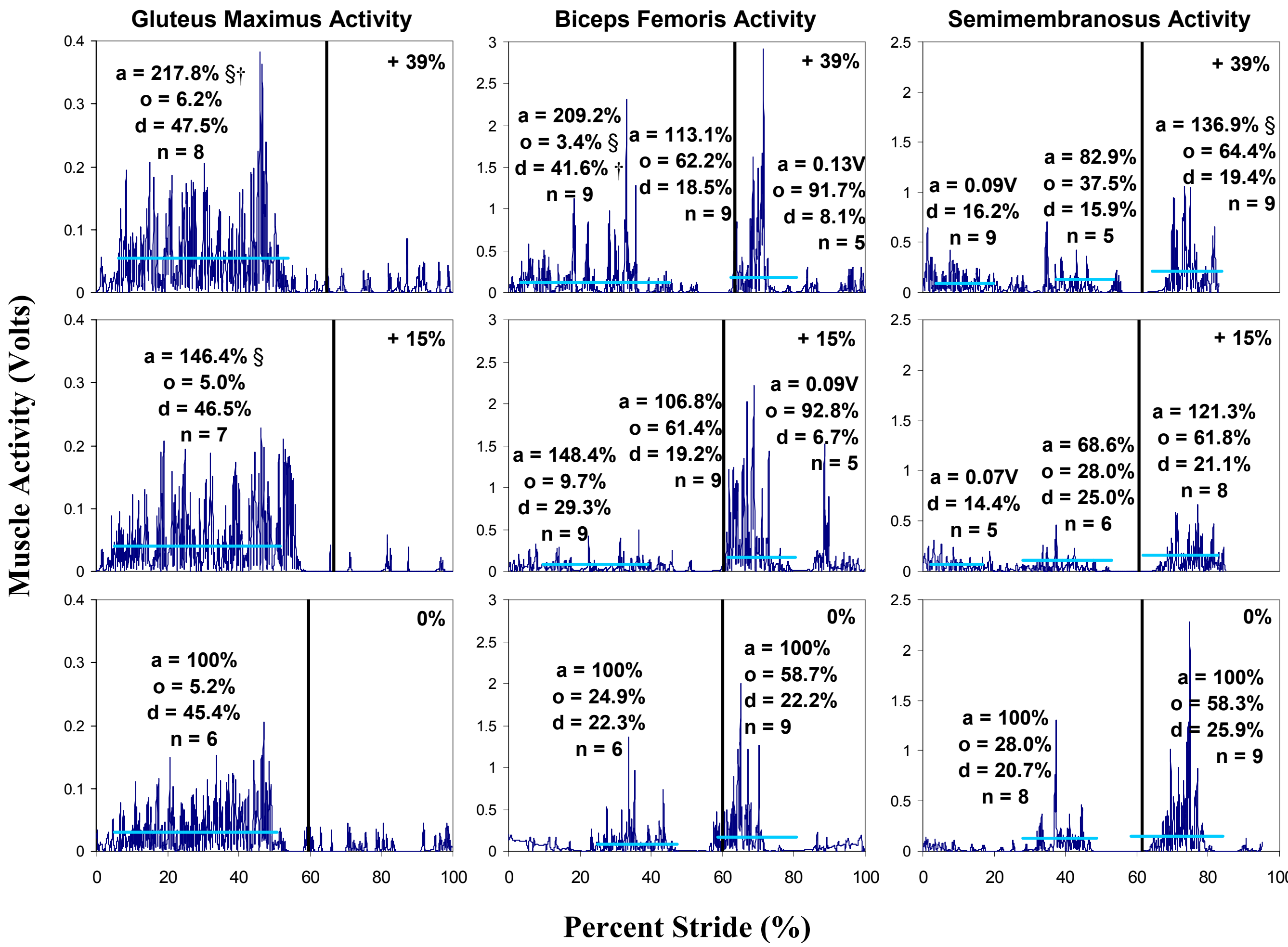


Figure 45. Representative GM, BF, and SM Data during Backward Upslope Walking. For each muscle a representative trial was chosen for each grade from a single participant's data to demonstrate the typical activity. Each trial was normalized to its stride time and plotted in %stride, with heel strike occurring at 0%. The toe-off time for each trial was also normalized to %stride and is indicated with a vertical line (average over all 24 representative trials for all eight muscles = 62.8%). Data from seven participants were used to represent the eight muscles. Horizontal lines representing the average burst data (mean for all subjects exhibiting the burst) are overlaid on each plot, and the values are also displayed on the plots: a = the group average mean activity (as a percentage of mean activity at 0% grade), d = the group average burst duration (in %stride), and n = the number of subjects exhibiting the burst. For bursts that do not start at heel strike o = the onset time (in %stride). These variables were submitted to repeated measures ANOVAs to assess the effect of grade. When a significant grade effect was observed, dependent t-tests were used for follow-up comparisons (adjusted p value of 0.01667). All significant differences are indicated on the figure by the following symbols: § - significantly different than the value at 0% grade ($p < 0.01667$) and † - significantly different than the value at -15% ($p < 0.01667$). See Appendix B for the group ensemble average curves.

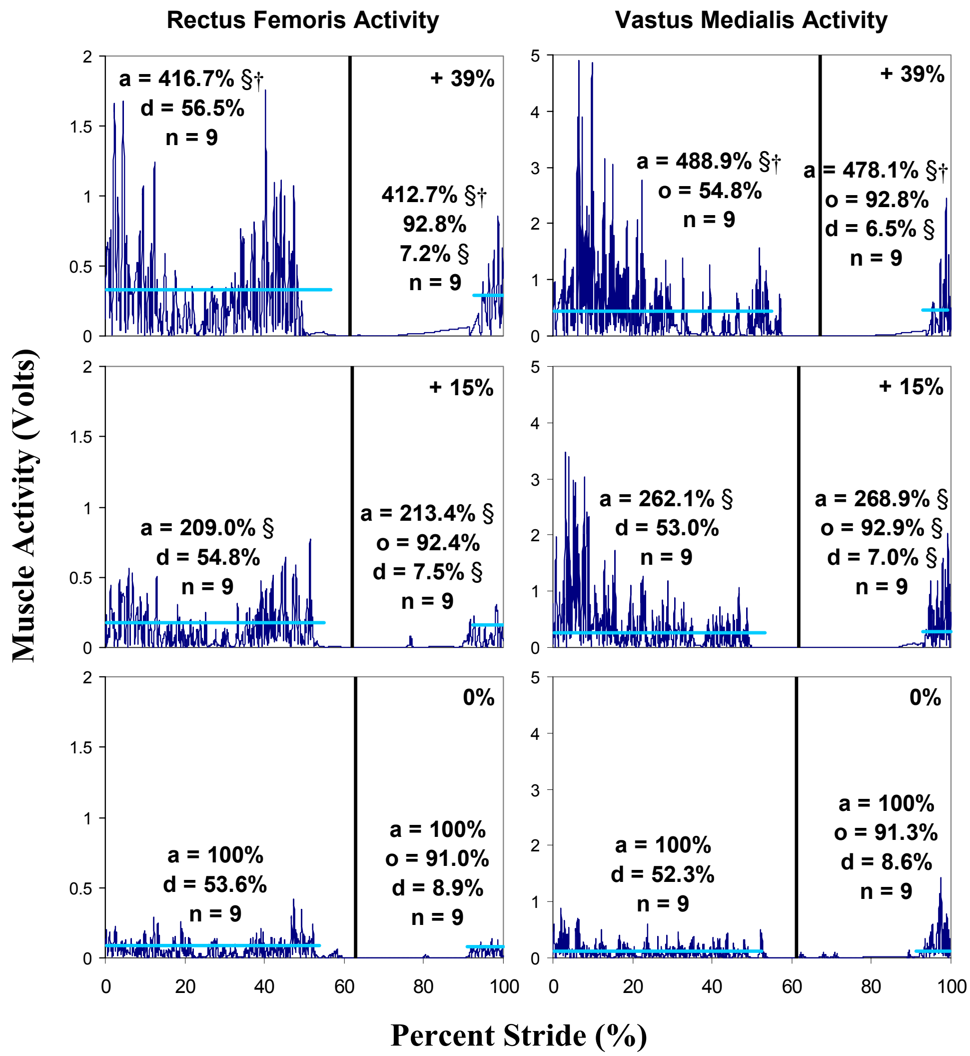


Figure 46. Representative RF and VM Data during Backward Upslope Walking. For details see Figure 45. All significance ($p < 0.01667$) is indicated on the figure: § - significantly different than 0% grade and † - significantly different than +/-15% (in the same walking direction).

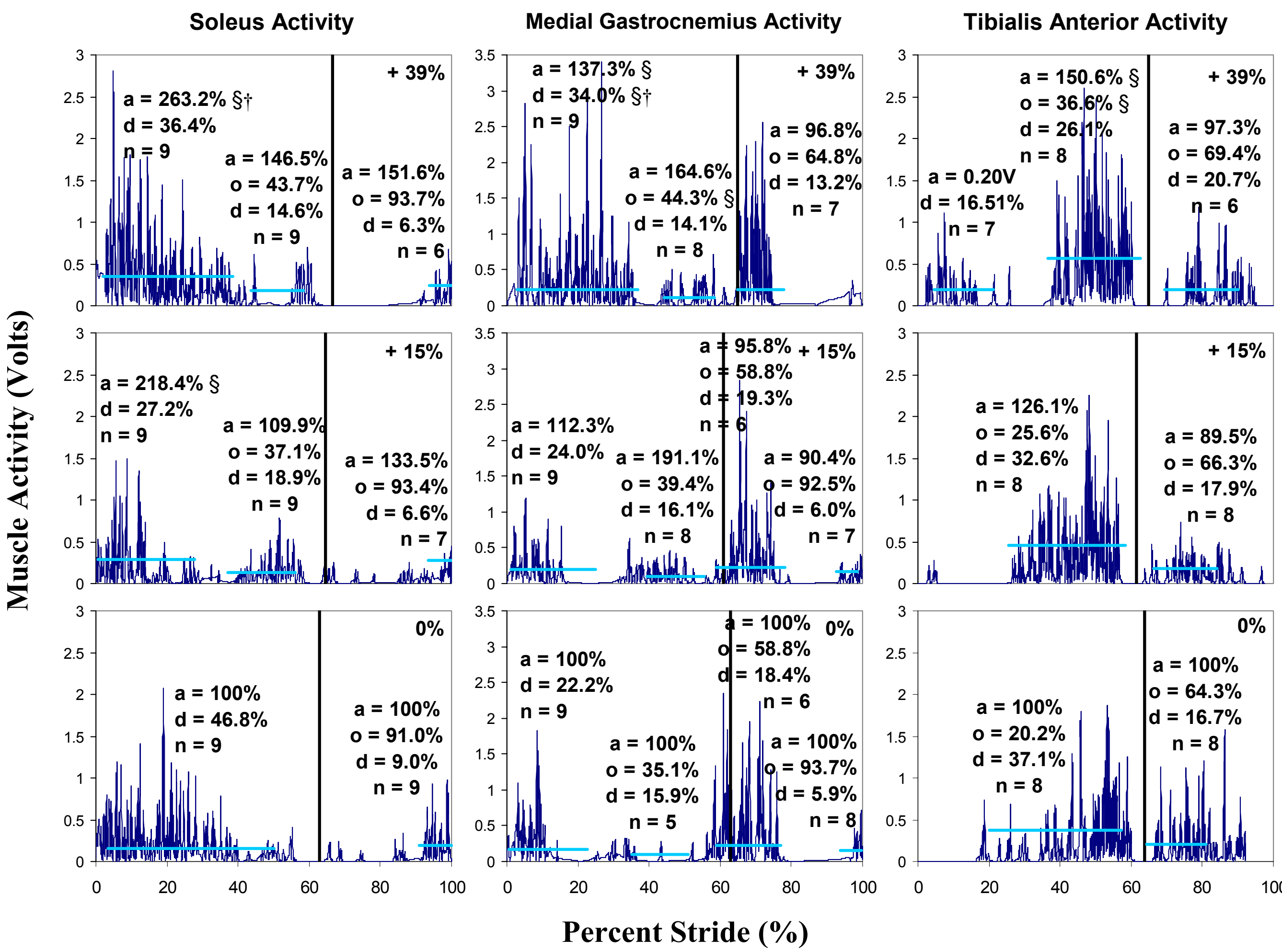


Figure 47. Representative Sol, MG, and TA Data during Backward Upslope Walking. For details see Figure 45. All significance ($p < 0.01667$) is indicated on the figure: § - significantly different than 0% grade and † - significantly different than +/- 15% (in the same walking direction).

Correlations between Kinetics and Muscle Activity. In order to quantitatively assess the relationship between joint moments and muscle activity the durations and peak magnitudes of the extensor joint moments and the extensor muscle bursts were correlated for upslope walking (Figures 48 through 51). The overall correlation for the extensor moments and extensor muscle burst durations (Figure 48A) was good ($R = 0.693$, r^2 values are shown in the figures). To understand how the extensor muscle activity at each joint was related to the resulting moment at that joint, joint-specific correlations were also performed. The joint-specific correlations (Figure 48B) for the knee ($R = 0.126$), and hip ($R = 0.164$) were poor because the extensor moment durations did not change much with grade, resulting in nearly horizontal regression lines. The ankle joint correlation ($R = 0.517$) was better than the correlations at the knee and hip. From Figure 48B the data points from the hip joint muscles seemed to be grouped separately, so muscle-specific correlations were done (Figure 49). In this case, the GM, BF, and SM correlations were high ($R = 0.949$, 0.969, and 0.776, respectively), indicating that each muscle was well-correlated to the joint moment data, but that the joint moment-muscle burst duration relationships were different for each muscle. These joint- and muscle-specific correlations provide additional insight into how the joint moments and muscle activity are related within the overall data set from the lower limb.

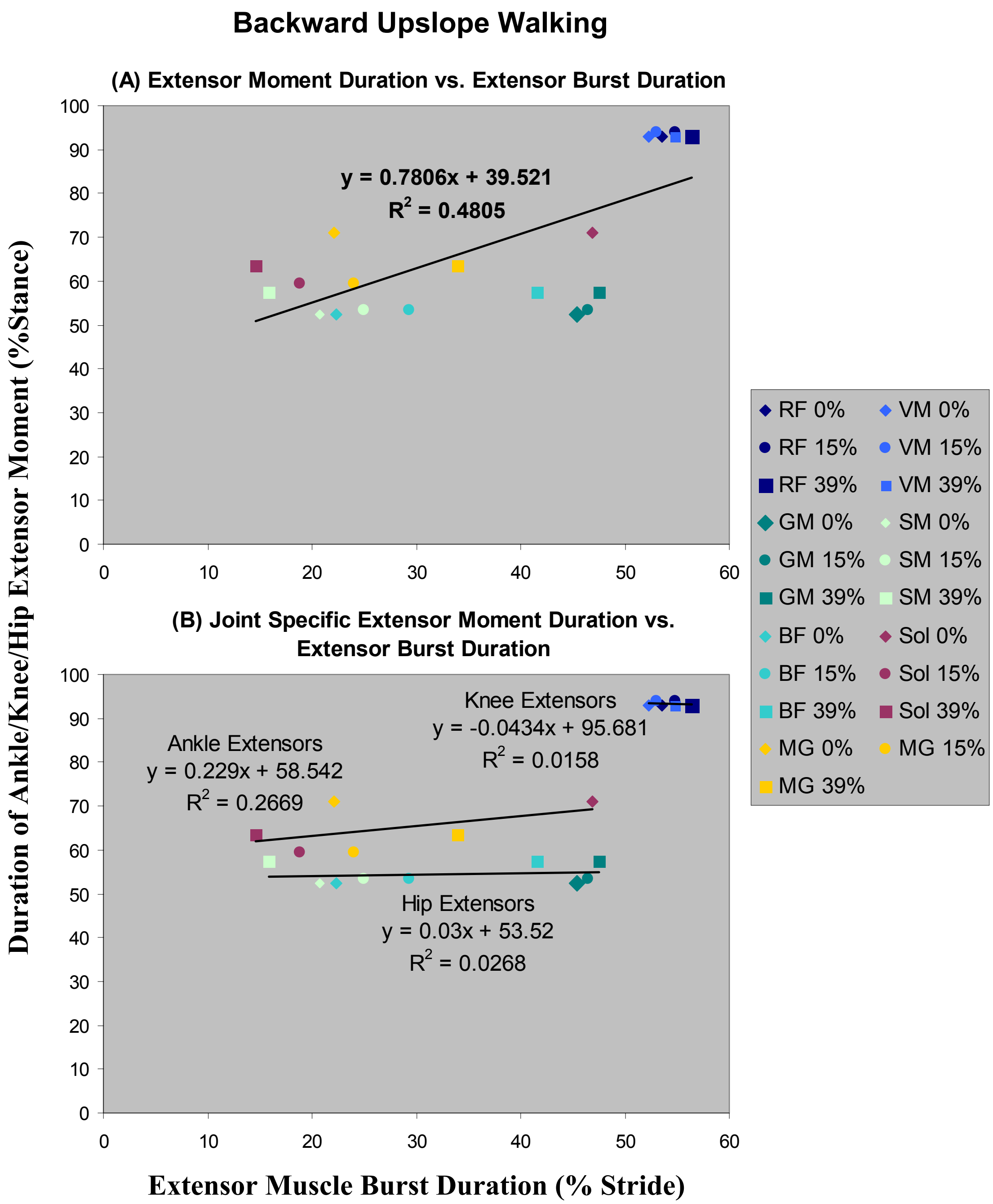


Figure 48. Correlation of Extensor Moment and Muscle Burst Durations during Backward Upslope Walking for lower limb extensor muscles and the joints at which they act. ♦ – data from level walking trials, ● – data from -15% trials, ■ – data from -39% trials. R^2 values are given for (A) overall lower limb correlation and (B) joint-specific correlations.

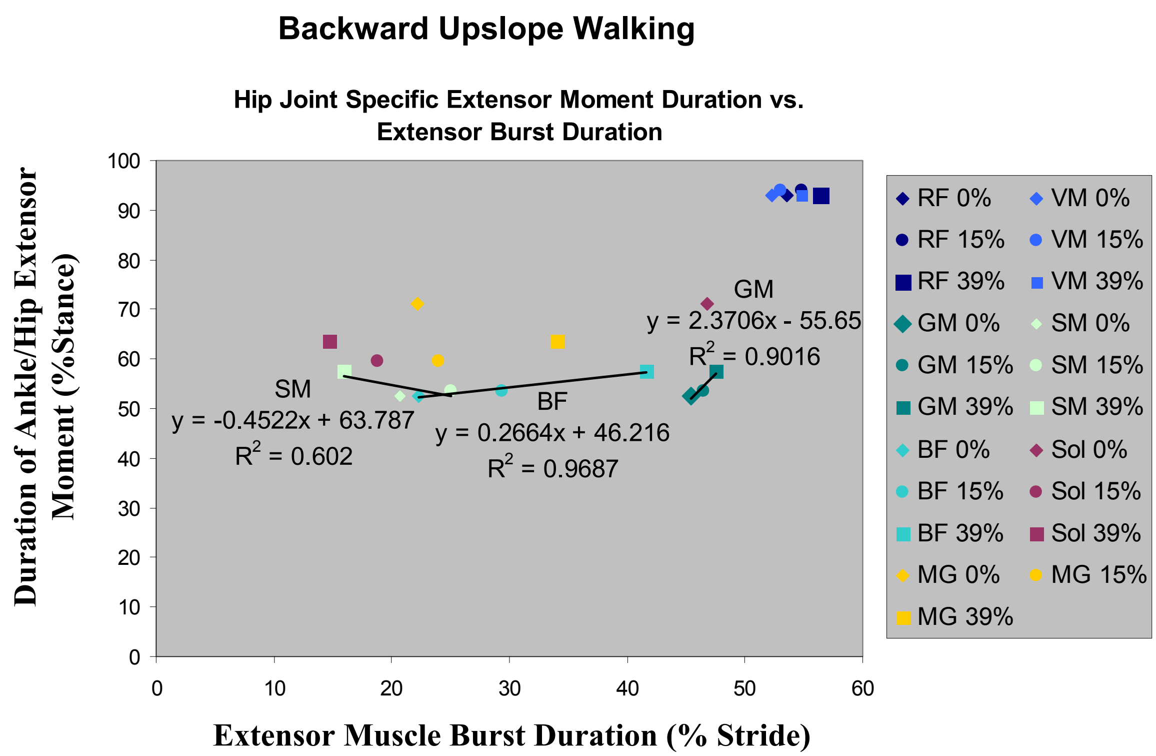


Figure 49. Correlation of Hip Extensor Moment and Muscle Burst Durations during Backward Upslope Walking for lower limb extensor muscles and the joints at which they act. ♦ – data from level walking trials, ● – data from -15% trials, ■ – data from -39% trials. R^2 values are given.

The overall peak extensor moment and peak extensor burst magnitude correlation (Figure 50A) was high for upslope walking (R = 0.884). The knee joint had the highest joint-specific correlation (Figure 50B, R = 0.918), while the hip and ankle correlations were much lower (R = 0.426 and R = 0.252, respectively). The knee joint moment and muscle peaks increased the most, overshadowing the changes at the hip and ankle and resulting in a higher correlation at the knee.

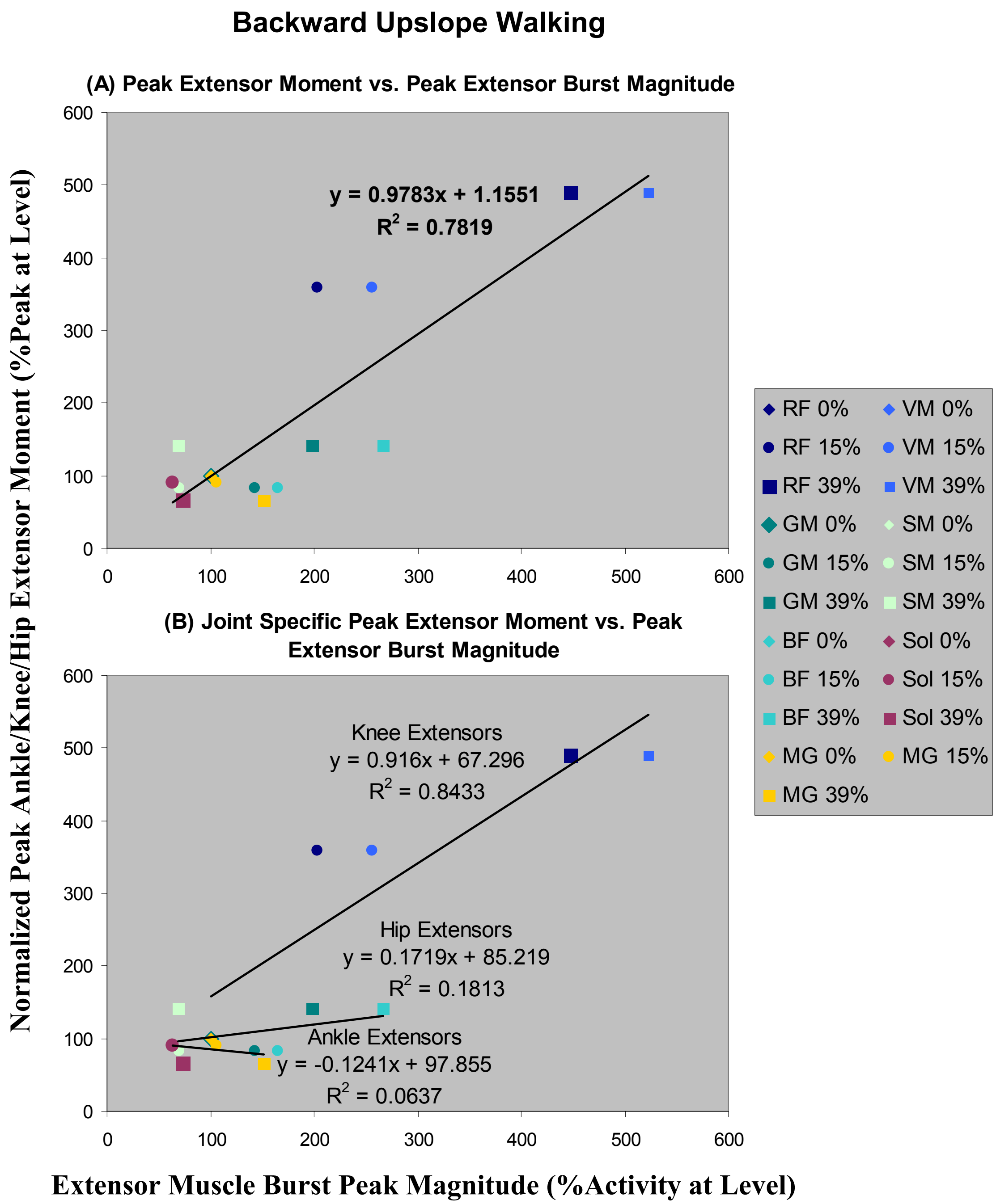


Figure 50. Correlations of Extensor Moment and Extensor Muscle Burst Peak Magnitudes during Backward Upslope Walking for lower limb extensor muscles and the joints at which they act. The muscle burst peak magnitudes were normalized to the peak magnitude of the burst during level walking. The extensor moment peaks were also normalized to the peak extensor moment during level walking. ♦ – data from level walking trials, ● – data from -15% trials, ■ – data from -39% trials. R^2 values are given.

Reviewing Figure 50B it seemed that a muscle-specific correlation at the knee might better represent the data. However, because the data point at level was the same for all muscles (100%, 100%; because each axis is normalized to the value at level) the individual RF and VM correlations were similar to the overall knee-joint correlation. Muscle-specific correlations did improve the representation of the hip extensor data joint (Figure 51). The GM (R = 0.741) and BF (R = 0.776) correlations were better than the overall hip joint correlation, but the SM correlation (R = 0.257) was worse.

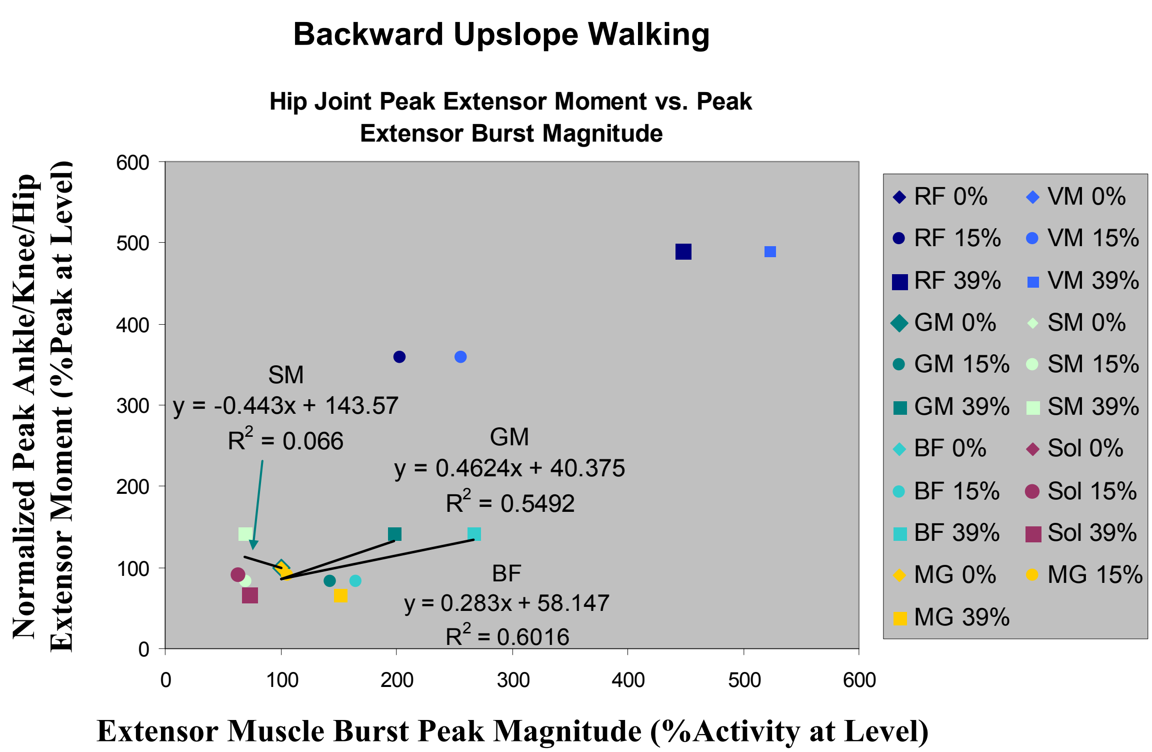


Figure 51. Correlations of the Hip Extensor Moment and Extensor Muscle Burst Peak Magnitudes during Backward Upslope Walking for lower limb extensor muscles and the joints at which they act. The muscle burst peak magnitudes were normalized to the peak magnitude of the burst during level walking. The extensor moment peaks were also normalized to the peak extensor moment during level walking. ♦ – data from level walking trials, ● – data from -15% trials, ■ – data from -39% trials. R^2 values are given.

Discussion. During early stance there was a plantarflexor moment and power was absorbed, indicating that the MG and Sol muscles were absorbing energy to control the lowering of the foot from toe-strike to foot flat. The *mean* activity levels of the MG and Sol increased during the first stance burst (Figure 47), but Figure 73 (Appendix B) shows that only the peak Sol EMG activity increased in early stance during upslope walking; the MG peak was similar during level and upslope walking. The increased activation of Sol compared to MG could be due to the biarticular function of the MG, because there is a large knee extensor moment in early stance and the MG functions as a knee flexor. Regardless of the reason, the increased plantarflexor activity does not seem to correspond to the decreasing peak plantarflexor moment. Figure 73 also shows increasing early stance TA activity, which may counteract the increased Sol and MG activity to cause the decreased moment, and most likely increases the stability of the ankle joint. For most of the remainder of stance the ankle joint plantarflexed due to the concentric activity of the MG and Sol. During terminal stance the ankle joint dorsiflexed due to concentric TA activity, and was stabilized by low level Sol and MG activity.

The knee joint was greatly flexed at toe strike, because the foot was positioned up the slope behind the body. During stance the knee joint extended with a large extensor moment, generating the large amount of power necessary to successfully lift the body upwards over the stance limb. The RF and VM, whose activity increased to more than 400% of the activity observed during level walking, contributed to this increased power generation. The BF and SM were also active during early stance of upslope walking (not during level walking) and the magnitude of the GM activity increased. This increased hip extensor activity may be used to counteract the hip flexor function of the RF, resulting in a hip joint moment similar to that for level walking. At the same time, the early stance BF and SM activity stabilizes and protects the knee joint as it undergoes the large extensor moment to lift the body.

The hip joint also started in a more flexed position than during level walking, due to the short stride length and the extreme flexion at the knee joint. Unlike the knee, however, the hip joint angle did not change appreciably during stance, and the hip joint moment was similar to that calculated for level walking. Although the moment did not change appreciably, the hip muscles were all active and the GM and BF had higher activity levels than during level walking. The power output at the hip joint was negligible during upslope walking, suggesting that the increased extensor activity was needed to counteract the increased RF activity (necessary to create the large knee extensor moment) and stabilize the hip joint as the stance limb supported the full body weight.

In general, the changes in the extensor moment durations and extensor muscle burst durations were small during backward upslope walking. As predicted by the hypothesis, the overall correlation and hip joint muscle-specific correlations were good in spite of the small changes. The peak extensor moments and peak muscle burst magnitudes were also well correlated for the overall lower limb, which also supported the hypothesis for this task. The knee joint data and the muscle-specific GM and BF data had high correlations, while the ankle joint data and the SM data were not well-correlated.

Comparison to Forward Slope Walking. The joint moment patterns were similar for level and the two backward upslope walking grades, but the magnitudes of the moments changed, which is similar to the changes in the joint moments during forward upslope walking. As described in Part 1 of this chapter, however, the large increase in the knee extensor moment is similar to that seen in forward downslope walking. The substantial power generation in backward upslope walking is also similar to that observed during forward upslope walking, except it was at the knee during backward walking rather than at the hip joint like forward walking. The behavior of the hip extensor/knee flexor muscles is different than that observed during forward slope walking; the two hamstrings exhibited different activity patterns and activation levels here. The increased GM output, and the pattern of the RF and VM outputs (Appendix B, Figure 72) are similar to those during forward upslope walking, as are the increases in Sol and MG mean activity levels. From these comparisons it is clear that the backward upslope walking data are similar to both forward upslope walking and forward downslope walking in different ways. Recall that backward upslope walking was expected to be most similar to that of forward downslope walking based on the similarities in joint kinematics and kinetics. However, the changes in joint kinetics that occurred as the slope increased were somewhat different from the changes observed during forward downslope walking: the joint moment patterns did not change substantially with grade, most noticeably because the knee joint moment was already extensor for most of stance during backward level walking. The activity patterns of the hip and ankle extensor muscles, on the other hand, did show changes with the walking grade. The changes in the joint moment magnitudes and peak muscle activity at the knee and hip joints generally had good correlations, but the correlation at the ankle was poor. Therefore, the data presented here both support and refute the hypothesis for this part of the project: ***the changes in the pattern and magnitude of the muscle activity were not directly related to changes in the pattern and magnitude of the joint moments at all joints***.

Part of the reason the hypothesis was not fully supported by the data relates to the similarities between backward upslope and forward upslope walking. In forward upslope walking, the RF and VM activity increased significantly in magnitude and duration, in spite of small changes in the peak knee extensor moment and no change in the knee moment pattern. This 'unexpected' change in muscle activity was attributed to the need to counteract the activity of the biarticular hamstrings, which were active to contribute to the large hip extensor moment demands but could also cause knee flexion. This situation was an example of Lombard's paradox, where an active biarticular muscle appears to cause an undesired effect at one joint if not counteracted (Gregor et al., 1985). A similar situation occurred here in backward upslope walking, where as the slope increased the magnitude and pattern of the hip joint moment did not increase, but the mean activities of the GM and BF did increase, and the duration of the BF and SM activity also increased during stance. As in forward upslope walking, this increased activity does not seem to correspond to the joint moment pattern until the role of biarticular muscles is considered. There is a large knee extensor moment during stance of backward upslope walking and power is generated, which requires increased concentric activity of the knee extensor muscles, RF and VM. However, the increased activity of RF would also tend to increase the flexor moment at the hip. The increased BF and GM activity and the SM burst that does not occur during level walking counteract the action of the RF at the hip, resulting in a joint moment similar to that during backward level walking. Relative to level walking, the activation of the BF increases more than that of the GM in early stance (Figure 45 and Appendix B, Figure 71), which may be because it is biarticular and therefore would also stabilize the knee joint. In the second half of stance, when there is a hip extensor moment and the knee extensor moment is reduced, the activity of the BF is lower than that of the GM, which may be because using the BF would also tend to cause unwanted knee flexion. Although the kinetic demands of this task are similar to those of forward downslope walking, these changes in muscle activity are similar to those observed during forward upslope walking, which also requires power generation. Based on these data, the similarity in the power requirements seems to be the primary cause of the changes in the muscle activity pattern. In other words, in spite of the similarity in the changes in kinetics in forward downslope and backward upslope walking, the changes in the muscle activity patterns are different. These differences may be attributed to the increased force feedback present during backward upslope walking, and may indicate a difference in control strategies for these tasks.

Part III: Backward Downslope Walking

Joint Kinematics and GRF. For completeness of the biomechanical description of backward downslope walking, the kinematic and GRF data from this task are presented in Appendix C.

Joint Kinetics. The sagittal plane lower extremity joint moments and the support moment (Figures 52 and 53) were similar for all grades during swing, but showed marked changes during stance as the walking grade decreased from level to -39%. No grade effect was observed for points KM4 or HM1 (ANOVA); the POI data and results of the follow up comparisons are given in Table 23. During backward downslope walking the peak ankle plantarflexor (AM1) and dorsiflexor (AM2) moments increased and decreased, respectively (Figure 52A). The knee joint moment (Figure 52B) showed no significant changes from its level walking pattern. The hip joint moment (Figure 52C) was similar to level in early stance, but the time of transition to a flexor moment decreased nonsignificantly with the walking grade. The peak extensor moment increased progressively with the walking grade during late stance (HM3). The early stance support moment peak (SM1) was similar during downslope and level walking, but the second peak (SM2) increased progressively as the walking grade decreased (Figure 53).

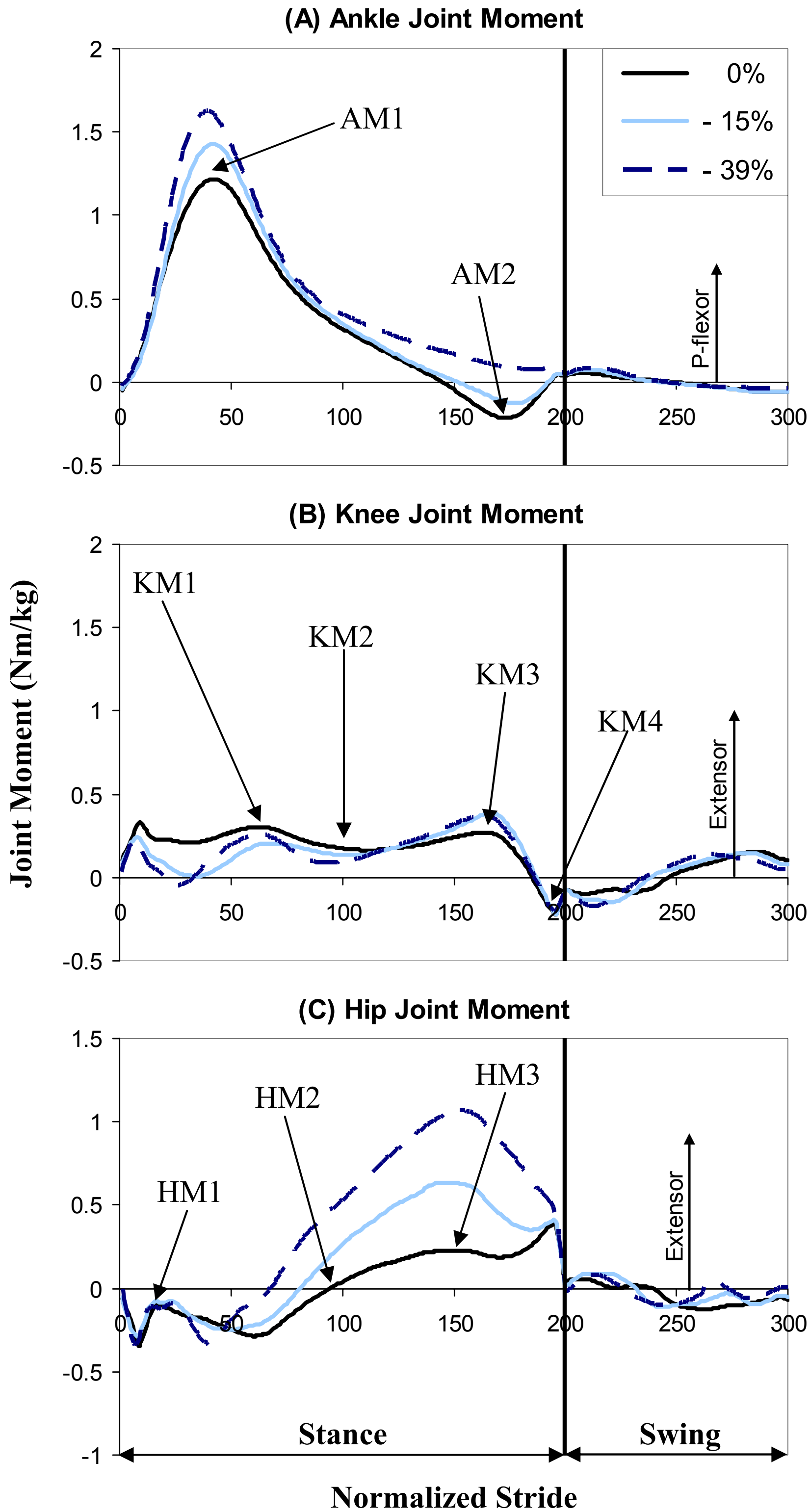


Figure 52. Average Joint Moments during Backward Downslope Walking. Positive moments are plantarflexor and extensor, as indicated. Stance was normalized to 200 points and swing to 100 points before ensemble averaging. Vertical line marks heel off.

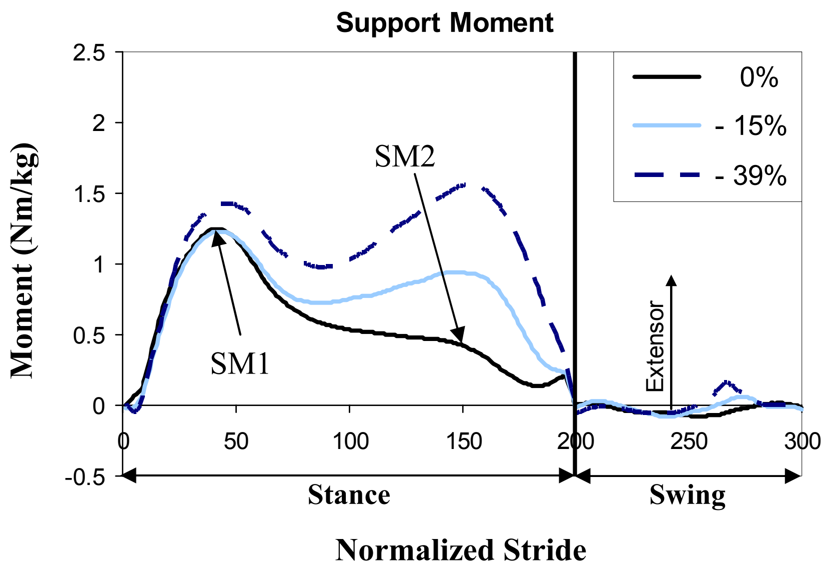


Figure 53. Average Support Moment during Backward Downslope Walking (Winter, 1980). Stance begins with toe strike, vertical line marks heel off.

Table 23. Backward Downslope Walking Joint Moment POI Data

	- 39%	- 15%	0%
AM1	1.69 (0.42)	1.49 (0.28)	1.26 (0.25)
AM2	0.02 §† (0.05)	-0.15 § (0.07)	-0.23 (0.08)
KM1	0.41 (0.52)	0.28 (0.46)	0.38 (0.30)
KM2	0.10 (0.39)	0.12 (0.32)	0.17 (0.28)
KM3	0.42 (0.21)	0.42 (0.22)	0.31 (0.18)
KM4	-0.22 (0.09)	-0.23 (0.09)	-0.23 (0.10)
HM1	0.05 (0.28)	0.04 (0.21)	-0.06 (0.31)
HM2	63.22 (23.26)	70.56 (33.89)	96.89 (52.29)
HM3	1.04 §† (0.20)	0.64 § (0.25)	0.26 (0.24)
SM1	1.63 † (0.51)	1.35 (0.34)	1.31 (0.28)
SM2	1.55 §† (0.26)	0.94 § (0.23)	0.40 (0.28)

Note: Mean (SD) of the joint moment POI (values N*m/kg, except HM2 is normalized stride). Shading indicates no grade effect was observed (ANOVA); § indicates a significant difference from 0% grade; † indicates a significant difference from 15% (in the same walking direction), ($p<0.05$).

Joint Power. Ensemble average joint power curves are shown in Figure 54. During backward downslope walking power absorption increased markedly at the ankle and hip (Figure 54A and C), and also somewhat at the knee joint (Figure 54B), indicating increased eccentric muscle activity. The percent of stance phase during which power was generated (Table 24) was similar during level and backward downslope walking at the ankle joint, but decreased at the knee and hip during downslope walking. In other words, the percent of stance where power absorption occurred increased at the knee and hip joints, but not at the ankle joint, as the walking grade decreased from level to -39%.

Table 24. Power Generation during Backward Downslope Walking Stance

	- 39%	**- 15%**	**0%**
Ankle	68.29	53.17	67.32
Knee	13.66	40.98	81.95
Hip	29.27	30.73	34.63

Note: Values are percent of stance phase (%), and are the average across all subjects.

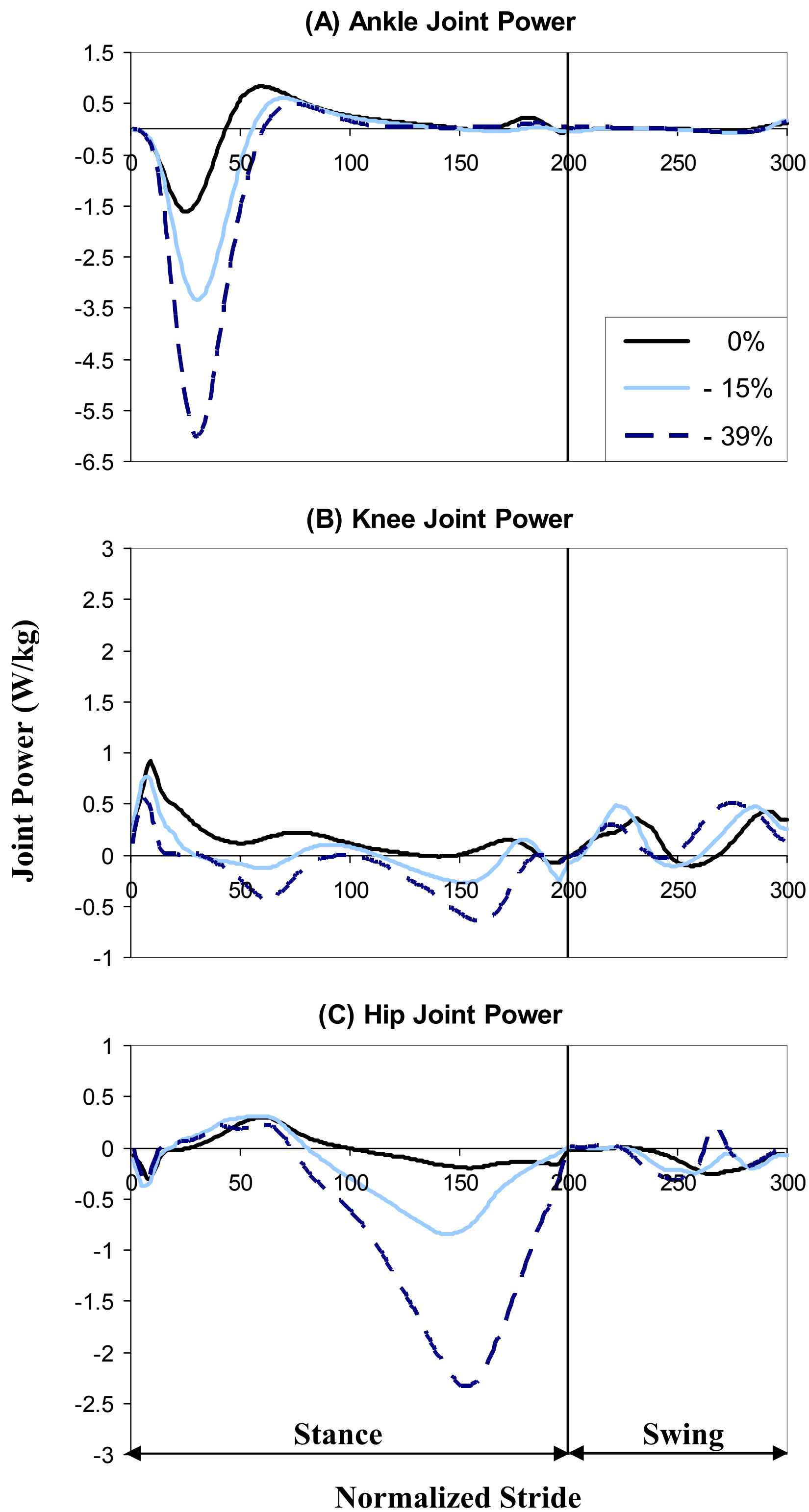


Figure 54. Average Joint Power during Backward Downslope Walking. Positive power is generation, negative is absorption. Stance and swing phases of each trial were normalized before ensemble averaging. Vertical line marks heel off.

Muscle Activity. Representative and group average EMG data are presented in Figures 55, 56, and 57; ensemble average EMG data are presented in Appendix C. With one exception, the EMG burst patterns at -15% and -39% were the same or had only small differences in the burst durations. The exception was the GM, which had a burst in late swing at -39% that was not present at -15%. Most muscles showed progressive changes in mean activity from as the walking grade decreased from 0% to -39%.

During backward downslope walking the mean activity level of the stance phase GM burst was statistically similar to that of level walking, and the burst timing did not change with walking grade (Figure 55). The onset of the stance phase hamstring muscle bursts (Figure 55) was earlier during downslope walking, and the duration increased such that the muscles were active for more of stance than during level walking. The mean activity levels of the BF and SM increased as the walking grade decreased from level to -39%, although only the SM showed statistical increases. Both muscles exhibited a burst at or immediately after heel off that did not change with the walking grade. The RF and VM (Figure 56) and the Sol (Figure 57) exhibited bursts for the majority of stance during downslope walking; the burst durations and mean activity values showed no statistical differences from level walking. The MG (Figure 57) typically had four bursts; two during stance, one at the stance-swing transition, and another in late swing. During downslope walking the mean activity of the first burst increased significantly but the second stance phase burst was no longer present, so the MG was not active during mid stance. The two TA bursts (Figure 57) typically seen during level walking merged into one burst during downslope walking. The mean activity of the new burst was reduced at -15%, but was similar to level at -39%.

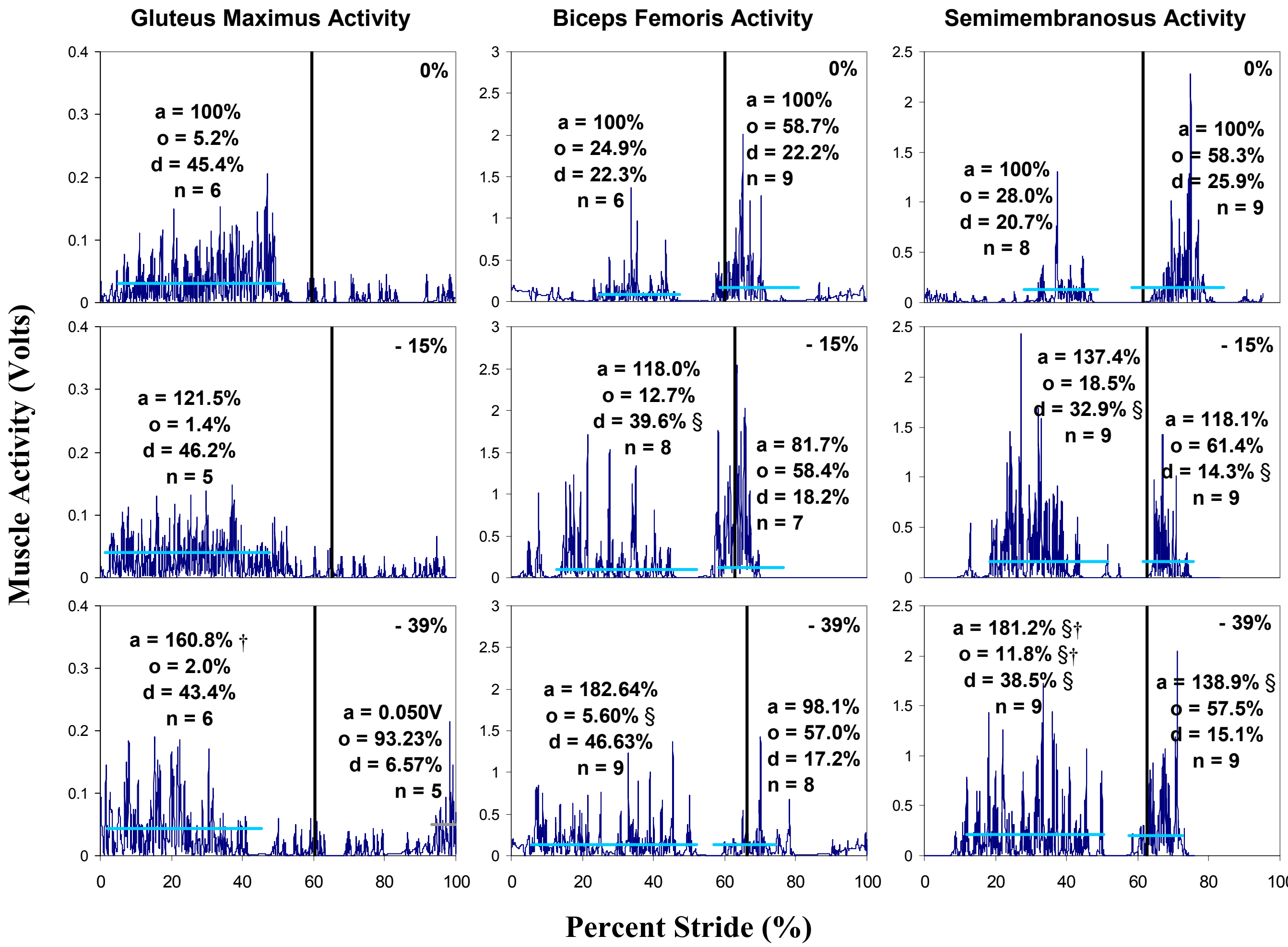


Figure 55. Representative GM, BF, and SM Data during Backward Downslope Walking. For each muscle a representative trial was chosen for each grade from a single participant's data to demonstrate the typical activity. Each trial was normalized to its stride time and plotted in %stride, with heel strike occurring at 0%. The toe-off time for each trial was also normalized to %stride and is indicated with a vertical line (average over all 24 representative trials for all eight muscles = 62.7%). Data from seven participants were used to represent the eight muscles. Horizontal lines representing the average burst data (mean for all subjects exhibiting the burst) are overlaid on each plot, and the values are also displayed on the plots: a = the group average mean activity (as a percentage of mean activity at 0% grade), d = the group average burst duration (in %stride), and n = the number of subjects exhibiting the burst. For bursts that do not start at heel strike o = the onset time (in %stride). These variables were submitted to repeated measures ANOVAs to assess the effect of grade. When a significant grade effect was observed, dependent t-tests were used for follow-up comparisons (adjusted p value of 0.01667). All significant differences are indicated on the figure by the following symbols: § - significantly different than the value at 0% grade ($p < 0.01667$) and † - significantly different than the value at -15% ($p < 0.01667$). See Appendix C for the group ensemble average curves.

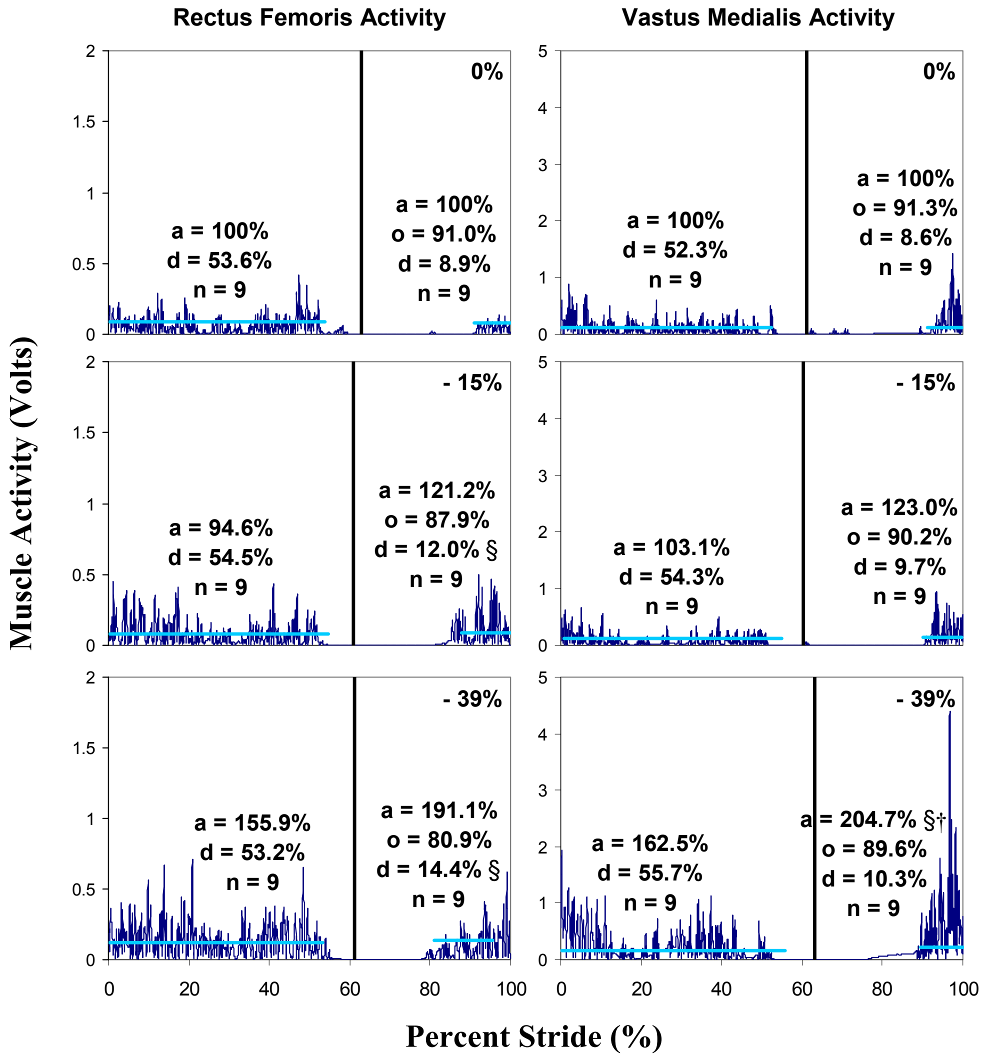


Figure 56. Representative RF and VM Data during Backward Downslope Walking. For details see Figure 55. All significance ($p < 0.01667$) is indicated on the figure: § - significantly different than 0% grade and † - significantly different than +/-15% (in the same walking direction).

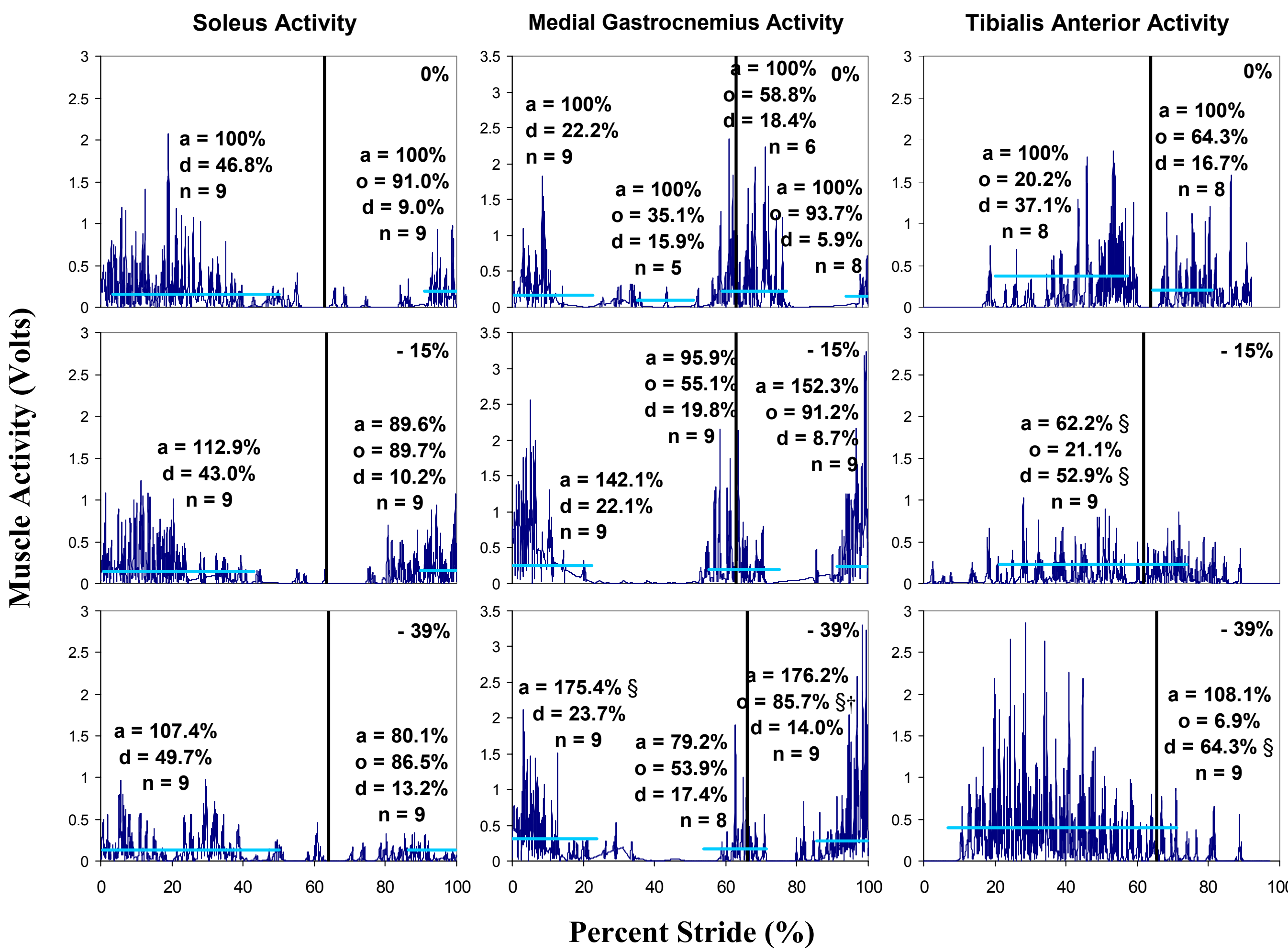


Figure 57. Representative Sol, MG, and TA Data during Backward Downslope Walking. For details see Figure 55. All significance ($p < 0.01667$) is indicated on the figure: § - significantly different than 0% grade and † - significantly different than +/- 15% (in the same walking direction).

Correlations between Kinetics and Muscle Activity. In order to asses the relationship between joint moments and EMG, the durations and peak magnitudes of the extensor joint moments and the extensor muscle EMG bursts were correlated (Figures 58 through 61). The overall correlation for the extensor moments and extensor muscle burst durations (Figure 58A) was low (R = 0.481, r^2 values shown in Figures). The joint-specific correlations (Figure 58B) showed low correlations for the ankle (R = 0.123), knee (R = 0.231), and hip (R = 0.590) also. The knee joint correlation was low because the data points were very tightly clumped, because neither the EMG burst duration nor the moment duration changed appreciably. The data from the ankle and hip moments and muscles seemed to fall into distinct groups for the different muscles, so a muscle-specific correlation was also performed (Figure 59). Separating the muscles drastically improved the MG correlation (R = 0.990) and the BF/SM correlation (R = 0.932). The Sol (R = 0.720) and GM (R = 0.654) correlations were also higher than the grouped ankle and hip joint correlations, but were not as good as the MG and BF/SM correlations. These muscle-specific correlations indicate that the agonist muscles at a joint have different activation patterns even when contributing to the same moment. As discussed in this and previous chapters, these differences may relate to the biarticular nature of some of the muscles included in this data set.

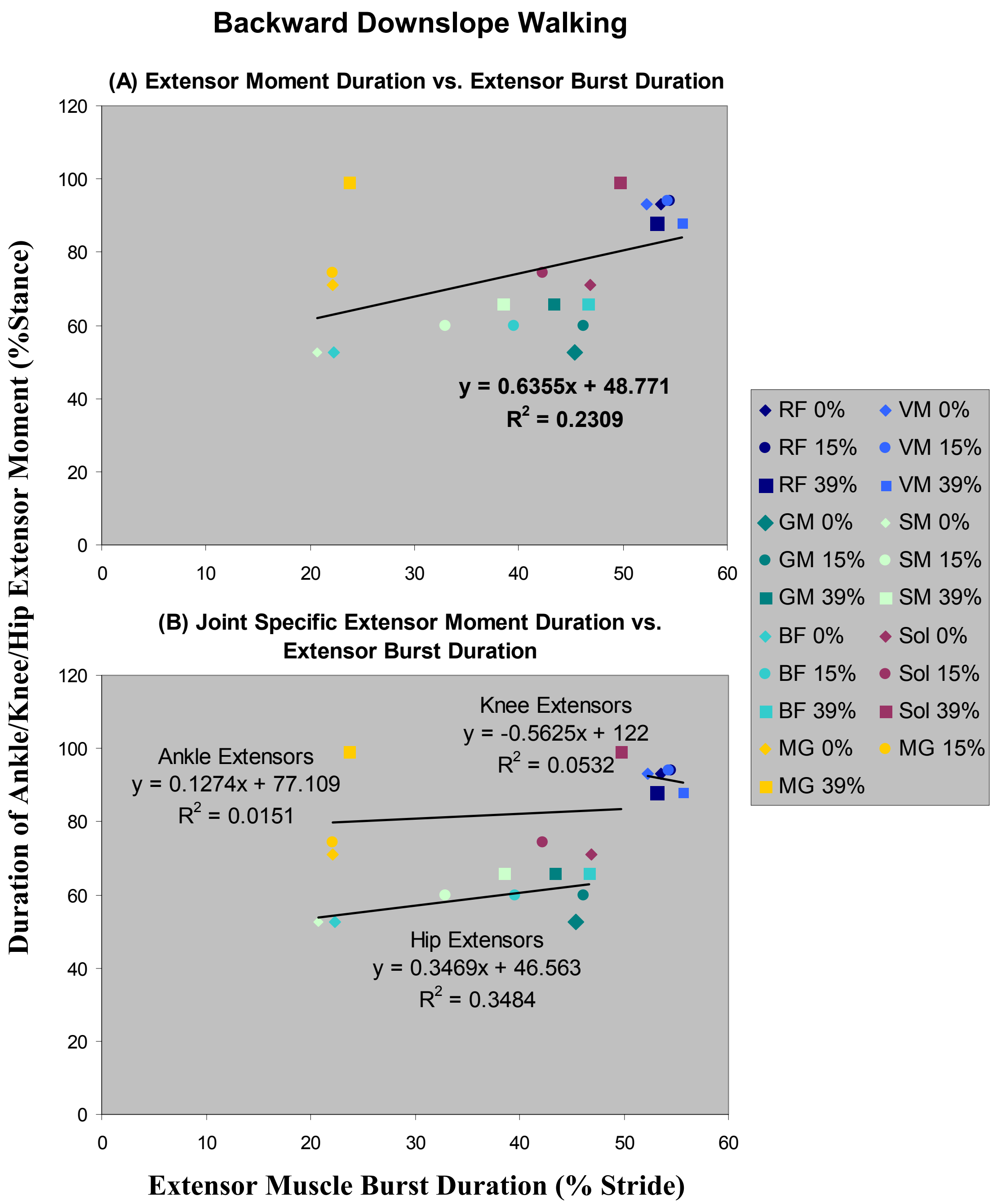


Figure 58. Correlation of Extensor Moment and Muscle Burst Durations during Backward Downslope Walking for lower limb extensor muscles and the joints at which they act. ♦ – data from level walking trials, ● – data from -15% trials, ■ – data from -39% trials. R^2 values are given for (A) overall lower limb correlation and (B) joint-specific correlations.

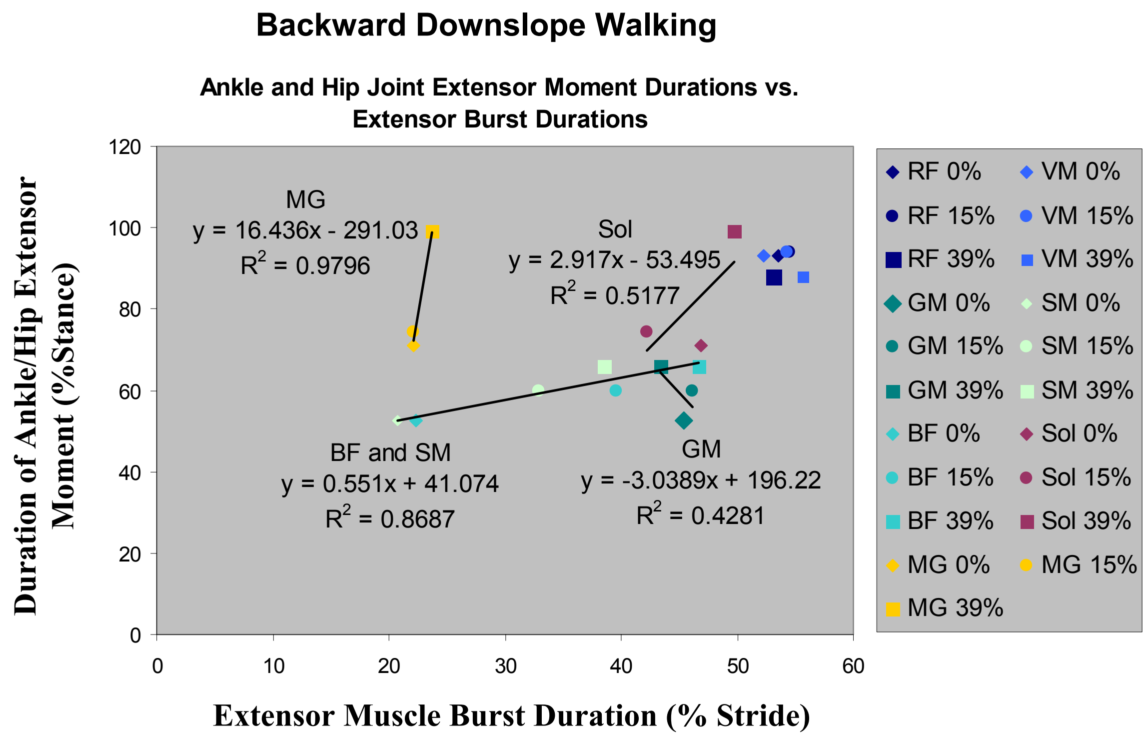


Figure 59. Correlation of Ankle and Hip Extensor Moment and Muscle Burst Durations during Backward Downslope Walking. ♦ – data from level walking trials, ● – data from -15% trials, ■ – data from -39% trials. R^2 values are given.

The overall correlation for the peak moments and peak EMG activities during downslope walking (Figure 60A) was good (R = 0.805), as were the joint specific correlations (Figure 60B, hip R = 0.948, knee R = 0.552, ankle R = 0.819). The ankle and knee joint moment peaks did not increase nearly as much as the hip joint moment peak, which resulted in data points that were much closer together for those two joints. Although the hip joint correlation was good, the data from each of the three hip extensor muscles seemed best fit by lines with different slopes (as in forward upslope walking, see Figure 34), so a muscle-specific regression was performed (Figure 61). As expected, the correlation coefficient for each muscle was even better than the already high hip joint correlation (GM R = 0.997, BF R = 0.982, SM R = 0.999).

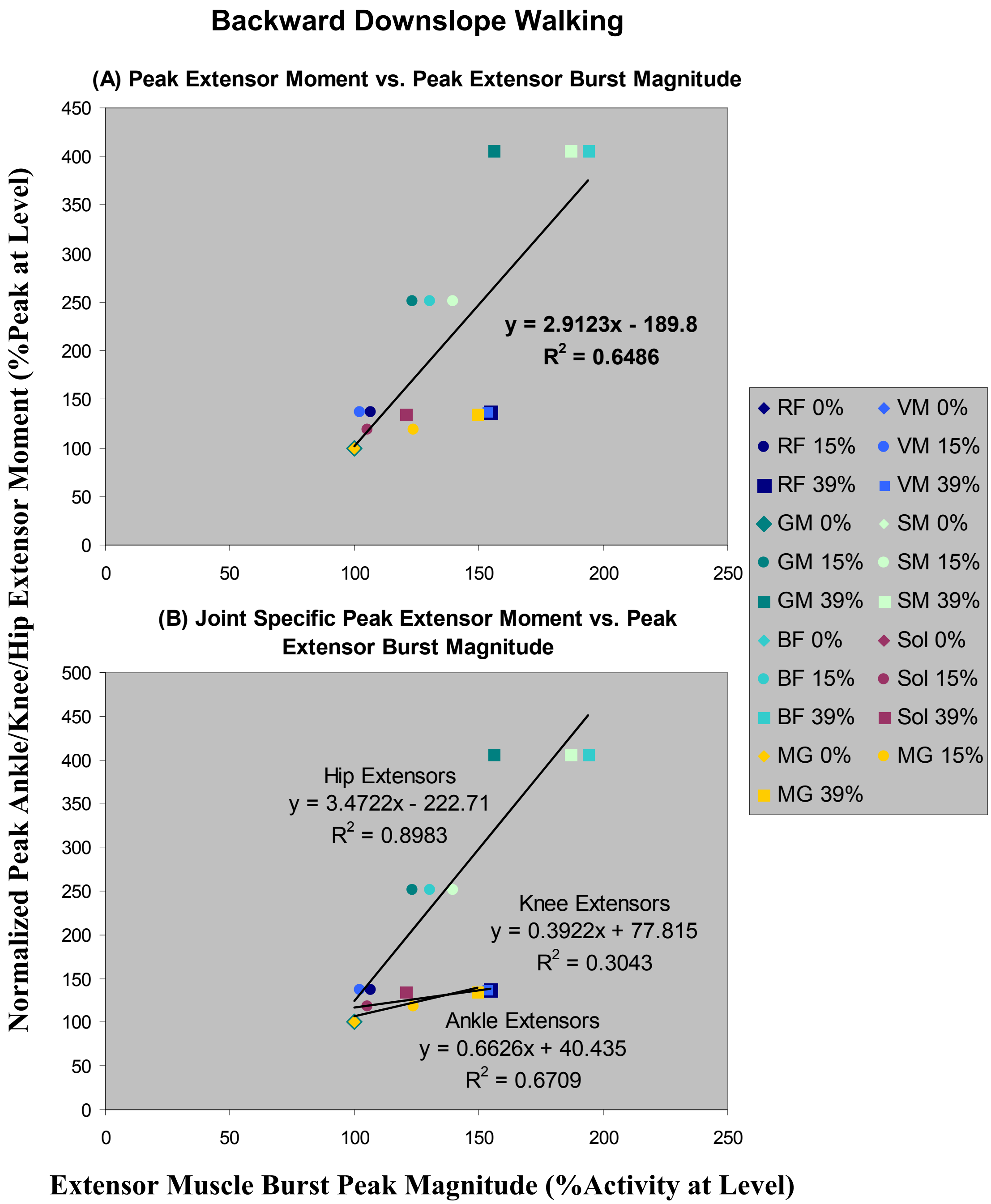


Figure 60. Correlations of Extensor Moment and Extensor Muscle Burst Peak Magnitudes during Backward Downslope Walking for lower limb extensor muscles and the joints at which they act. The muscle burst peak magnitudes were normalized to the peak magnitude of the burst during level walking. The extensor moment peaks were also normalized to the peak extensor moment during level walking. ♦ – data from level walking trials, ● – data from -15% trials, ■ – data from -39% trials. R^2 values are given.

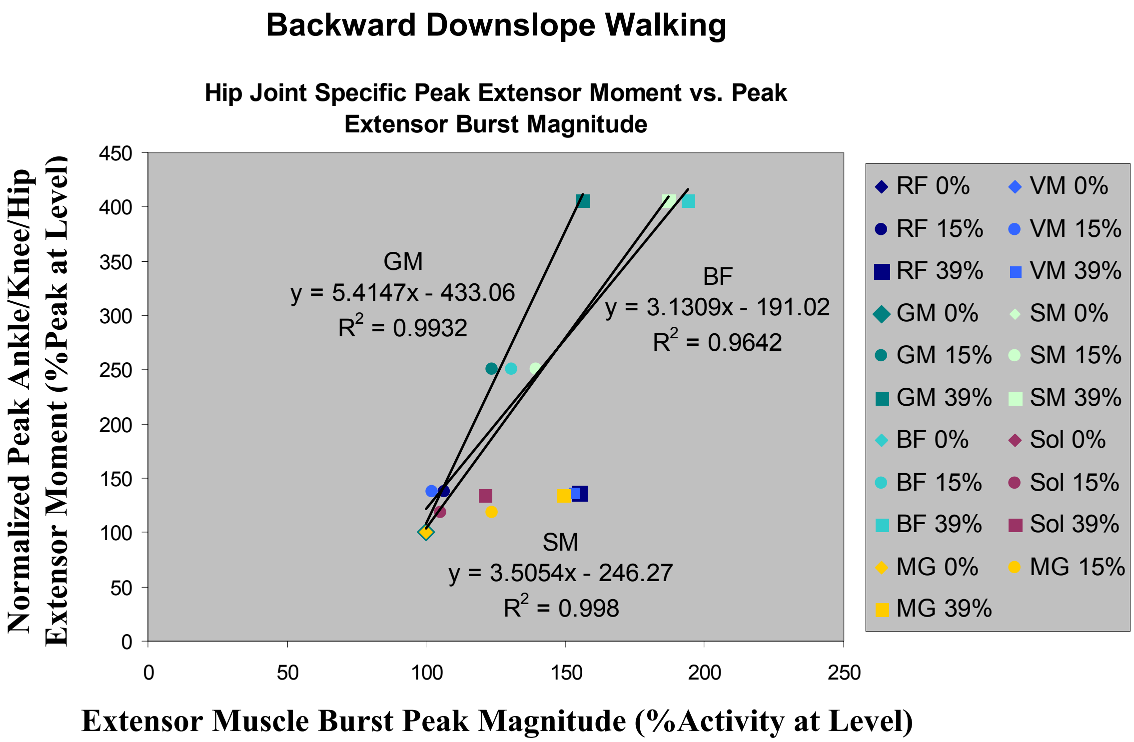


Figure 61. Correlations of the Hip Extensor Moment and Extensor Muscle Burst Peak Magnitudes during Backward Downslope Walking for lower limb extensor muscles and the joints at which they act. The muscle burst peak magnitudes were normalized to the peak magnitude of the burst during level walking. The extensor moment peaks were also normalized to the peak extensor moment during level walking. ♦ – data from level walking trials, ● – data from -15% trials, ■ – data from -39% trials.

Discussion. Due to the novelty of the data presented in this chapter, there are no data available in the literature for comparison. As presented here previously, during early stance the ankle joint was dorsiflexing, and because of the downward slope dorsiflexed even more when compared to backward upslope walking. There was also a large plantarflexor joint moment, so power was absorbed at the ankle during early stance. As a result, the Sol and MG were again acting eccentrically to control the progression of the foot from neutral to foot flat. The TA was not active in early stance, indicating that the ankle dorsiflexion was caused by gravity. After the foot was in full contact with the ramp the ankle plantarflexed for the remainder of stance and for most of swing. The role of the MG and Sol most likely was to produce the initial plantarflexion, initiating the backward movement of the body over the foot. Plantarflexion for the second half of stance occurred without an active contribution from the MG; the joint moment showed net plantarflexor activity, but power generation was negligible. It is likely, therefore, that plantarflexion during the second half of stance was a result of gravitational acceleration as the body was lowered down the slope over the stance foot. The TA and Sol were likely counteracting one another and stabilizing the ankle joint.

The position of the knee joint was similar to level walking at toe strike, but the joint extended less and flexed more in stance than during level walking. In early stance the knee extended briefly, due to the activity of RF and VM, to prepare the foot for weight acceptance. The knee flexors BF and SM were activated earlier than during level walking, which halted the extension of the knee joint earlier, reversed the knee extensor moment, and decreased the power being generated. From ~20% to 50% of stride, as the body weight was transferred to the stance limb and the contralateral limb entered the swing phase, the knee joint angle did not change much, there was a small extensor moment, and power absorption was negligible. The knee extensors and flexors were co-active at this time to stabilize the knee joint. In mid to late stance the knee flexed to lower the body and prepare to transfer the body weight to the contralateral limb. There was still a small knee extensor moment, so power was absorbed by the knee extensors to prevent collapse.

The hip joint extended dramatically during swing in backward downslope walking, and continued to extend in early stance. This extension was halted by contraction of the hip flexors, which generated some power in early stance to initiate the backward movement of the body over the stance foot. The hip joint continued to flex throughout stance, but the moment switched to an extensor moment in mid stance and large amounts of power were absorbed at the hip. Although the RF was active during stance, it is likely that the acceleration of the body over the stance foot down the ramp tended to naturally flex the hip joint, and required resistance by the hip extensors to stabilize the joint and control the movement of the body. The GM, BF, and SM were all active during stance, and were therefore most likely contracting eccentrically.

In general, the extensor moment and muscle burst durations were not well correlated, except for the individual muscles at the ankle and hip. On the other hand, the peak extensor moment and muscle burst magnitudes were well correlated, overall, at the joint level, and also at the individual muscle level. These data support the hypothesis that changes in the magnitude of muscle activity would be directly related to changes in the magnitude of the joint moments for backward downslope walking.

Comparison to Forward Slope Walking. The backward downslope walking joint moments showed some changes in both pattern and magnitude as the walking grade decreased from 0% to -39%. The changes in pattern were small, while the changes in magnitude were substantial, especially at the hip. These changes are similar to those observed during forward walking at the grade increases from 0% to +39%. Changes in the joint powers, on the other hand, resembled those from forward downslope walking, where power absorption increased. The muscle activity was similar to that from forward upslope walking, with activity increasing as the slope changed. However, the RF and VM activity pattern did not change with slope, which is more similar to the results from forward downslope walking. The overall moment duration – EMG burst duration correlation and the peak moment – peak EMG burst magnitude correlations were also similar to those for forward upslope walking.

The data presented here show that the task of backward downslope walking is similar to the task of forward upslope walking, with the exception of the joint power requirements and some specific muscle activity patterns. The data also ***support the hypothesis that the changes in the magnitude of the muscle activity would be directly related to the changes in the magnitude of the joint moments***. As discussed, the moment and peak burst magnitudes were well-correlated during this task. It was also thought that, similar to forward upslope walking, the changes in the muscle firing patterns would not relate well to the changes in the joint moment patterns. During forward upslope walking, for example, the RF and VM activity duration increased during upslope walking although there was no change in the knee extensor moment. This same discontinuity was not observed in backward downslope walking, where ***instead the muscle activity patterns corresponded directly to the joint moment patterns***: during backward downslope walking the hip extensor moment onset was earlier in the stance phase, and the hamstring muscle EMG burst durations increased to account for this, resulting in a high correlation for the BF and SM. At the knee neither the moment pattern nor the muscle durations changed, resulting in a clump of data points and a low correlation. At the ankle, the moment stayed plantarflexor at the end of stance at -39%

walking; the TA activity decreased in conjunction with this change, and the MG and Sol burst durations were highly correlated as well. As previously mentioned, the primary difference between the tasks of forward upslope and backward downslope walking is the power requirements. In Chapter 6 the discontinuity between the joint moment pattern and the muscle activity was attributed to the need to activate the biarticular hamstrings to contribute to the large hip extensor moment demands, which in turn required increased activation of the knee extensors to counteract the knee flexion caused by the hamstrings, an example of Lombard's paradox (Gregor et al., 1985). It is clear from the EMG data (Figure 55 and Appendix C, Figure 78) that the hamstrings are actively contributing to the increased hip extensor moment during backward downslope walking as well, but the knee extensor activity does not increase as it does during forward upslope walking (Figure 30 and Appendix A, Figure 66). From this discussion and the data presented in this chapter it seems that under similar kinetic demands, replacing power generation with power absorption reduces the need to activate antagonists to the biarticular muscles, which makes the muscle activity patterns more closely related to the joint moment patterns.

Conclusions

Recall that the purpose of this aim was to use the tasks of backward upslope and downslope walking as further tests of the robustness of the movement control strategies employed by the nervous system for forward slope walking. It was expected that the tasks of backward upslope walking and forward downslope walking would be kinematically and kinetically similar, as would the tasks of backward downslope walking and forward upslope walking. The kinematics and kinetics presented here show generally support this expectation. The primary difference is that the knee joint moment is never a flexor moment during backward walking, unlike forward walking where it is flexor for the second half of stance in level and upslope walking. Given the anticipated similarities in the kinetics, it was also expected that the changes in muscle firing patterns would be similar to the changes observed in the corresponding forward slope walking tasks. More specifically, it was expected that the magnitude and duration of the joint moments and muscle activity would be correlated in backward upslope walking, and that the magnitudes of the joint moments and muscle activity would be correlated in backward downslope walking.

During backward upslope walking the correlations of moment and muscle burst durations and peak EMG were variable across the joints. In other words, the changes in the pattern and magnitude of the muscle activity were not directly related to changes in the pattern and magnitude of the joint moments at all joints. Although the task was kinematically and kinetically similar to forward downslope walking, the changes in the muscle activity patterns showed some similarities to forward upslope walking. In both forward upslope and backward upslope walking power generation (at the hip and knee, respectively) increases as the slope changes, and increased activity of biarticular muscles is required at these ‘primary’ joints. This, in turn, requires recruitment of muscles at the adjacent joint(s) to counteract undesired activity of the biarticular muscles. The increased activity of the antagonists at the adjacent joint(s) does not directly correspond to the joint moment pattern in either case. This similarity between forward upslope and backward upslope walking is likely due to the similarity in the power requirements, which seems to supersede the similarity in the kinetic patterns between backward upslope and forward downslope walking as an influence on control strategies. Data from the task of descending stairs also supports this idea (McFadyen and Winter, 1988). The knee joint moment during stair descent is similar to the knee joint moment during forward downslope walking and backward upslope walking; but power is absorbed in the first two tasks, and generated in the third. In both stair descent (McFadyen and Winter, 1988) and forward downslope walking (Chapter 5) the knee extensor activity increases dramatically, but the GM and hamstring activity is similar to that during level walking. In backward upslope walking, however, the GM and hamstring activity increases substantially from level walking. Again, these data indicate that power requirements during a task outweigh the similarities in joint moment patterns in determining the muscle activity patterns.

In backward downslope walking the peak magnitudes of the muscle bursts and joint moments were well-correlated, as expected. The hip extensor moment and EMG burst peaks increased the most, while the ankle and knee moments and muscle peaks increased slightly as the walking grade decreased. As in forward upslope walking, the hip joint moment was the major contributor to the increases in the support moment during backward downslope walking. The difference was that the hip extensor muscles were absorbing power during backward walking. Recall that in forward upslope walking there were increases in the knee extensor EMG burst magnitudes and durations that did not correspond to the knee extensor moment. During backward downslope walking the knee extensor burst durations did not change because the muscles were already active for most of stance due to the continuous knee extensor moment, but the magnitude of the knee extensor EMG activity did not change either. This lack of change in the knee extensor activity meant the muscle activity patterns corresponded more closely to the joint moment patterns than during forward upslope walking; in fact, this relationship was more like that observed in forward downslope walking. Again, the similarity in the power requirements between forward and backward downslope walking seems to have a major influence on the changes in the level walking activity pattern that are needed to achieve the locomotor task.

To summarize, it was expected that tasks with similar changes in kinetic patterns would also have similar changes in muscle activity patterns regardless of the joint power profiles. Although the correlation data presented here do support some of the hypotheses developed from this idea, a more detailed look at the data refutes this concept. It appears that the power requirements of a task dictate the muscle activity pattern needed to accomplish that movement. This idea was suggested by Grasso et al. (1998) for level walking, where it appears that kinematic patterns are conserved across a gait reversal via a reorganization of muscle activity patterns. The data presented here suggest that this reorganization is due to changes in the power requirements of the task: increases in power generation at a joint require compensatory muscle activity at adjacent joints that results in a mismatch between the moments and muscle activity at those joints. Increases in power absorption during a task do not appear to have this effect on the muscle activity patterns; the muscle activity patterns correspond closely to the joint moment patterns. This apparent importance of the joint power requirements in determining the neuromuscular patterns during a task is an interesting finding and raises the question of what the actual mechanism is that causes the changes in the muscle activity.

CHAPTER 8

GENERAL SUMMARY AND FUTURE WORK

Summary of Findings

The goal of this project was to gain insight into human neural control strategies during upslope and downslope walking by synthesizing an analysis of lower limb biomechanics and muscle activity patterns. Backward upslope and downslope walking were used as additional perturbations to test the robustness of the control strategies observed during forward slope walking. The main findings from this study were that (1) forward upslope and downslope walking appear to be controlled by different motor programs, (2) the changes in the muscle activity pattern depend on the power requirements during the task rather than on the joint moments, and (3) the changes in joint moments, power requirements, and muscle activity that occur during slope walking are distributed unequally among the three major lower limb joints. These findings, as well as their relevance, will be discussed here.

The results from forward upslope and downslope walking indicate that these tasks are controlled by different motor programs. During forward downslope walking the joint moment pattern at the knee changed significantly and power absorption increased, compared to level walking. Under these conditions, the changes in the muscle activity patterns corresponded directly to the changes in the joint mechanics. During upslope walking, compared to level walking, the hip joint moment pattern was most affected and power generation increased at all three major joints, but most significantly at the hip. In this case, the changes in the muscle activity patterns were indirectly related to the changes in the joint moments because the increased activation of biarticular muscles required increased activation of antagonists that did not correspond to the joint moment pattern. These data support the idea that different control strategies, or modifications of the same control strategy, were used to accomplish these tasks. The differences in the mechanics of the tasks and the corresponding muscle patterns suggest that length and force feedback may be important in modifying the control strategy.

Backward upslope and downslope walking were used to further explore the different control strategies observed in forward upslope and downslope walking. When the data from all four tasks are viewed collectively, it appears that the muscle activity pattern that occurs during a locomotor task depends on the power requirements of the task rather than on the joint kinematics or kinetics. Similar to the previous reports for level forward and backward walking (Grasso et al., 1998; Thorstensson, 1986; Winter et al., 1989), it was thought that the 'reversed' slope walking tasks would be similar: forward upslope and backward downslope walking, and forward downslope and backward upslope walking. Kinematically and kinetically, these pairs of tasks were indeed the most similar; the backward walking tasks were almost identical time-reversals of the corresponding forward walking tasks. The power demands of these corresponding tasks were exact opposites, however, with power absorption increasing during backward downslope walking, and production increasing during backward upslope walking. As expected based on the level walking data, these pairs of tasks did not have the same muscle activation patterns. More importantly, however, the different muscle firing patterns for forward and backward level walking did not respond the same way to changes in the slope of the walking surface, as it was hypothesized they would. The changes in the muscle firing patterns were more similar between the tasks with similar joint power demands. During tasks where power absorption increased (forward downslope and backward downslope walking) the changes in the muscle activity patterns were directly related to the changes in the joint moment patterns. In contrast, during tasks where power generation increased (forward upslope and backward upslope walking) the changes in the muscle activity were related to the changes in the joint moments *only* at the ‘primary’ joint; at the adjacent joints the changes in the muscle activity were unrelated to the joint moment pattern. The ‘paradoxical’ changes in the muscle activity at the adjacent joints were related to the activation of biarticular muscles required by the increasing power generation at the primary joint. In total, these data indicate that increased power generation requirements at a joint impact the control of muscle activity at adjacent joints, most likely via force-feedback from biarticular muscles.

Finally, these data illustrate that the changes in joint moments, power requirements, and muscle activity that occur during slope walking are distributed unequally among the contributing lower limb joints. In all slope walking conditions the support moment, a measure of the net demand on the lower limb, increased from level walking as the slope changed. The increases were distributed unequally among the ankle, knee, and hip joint moments: the knee joint moment increased the most during forward downslope and backward upslope walking and the hip joint moment increased the most during forward upslope and backward downslope walking. As one might expect, the muscles at these joints showed the greatest changes in activity. During forward and backward downslope walking the joint moments and muscles at the ankle and hip joints were less affected than those at the knee joint, but even the small changes in muscle activity remained directly related to the changes in the associated joint moments. During forward and backward upslope walking, however, the changes in the activation of the muscles at the knee and hip joints, respectively, did not correspond to the joint moments. As already discussed, this finding has been attributed to the increased power generation requirements during these tasks and the increased activation of biarticular muscles that span the knee and hip joints. One question that follows is how the ankle joint is affected during these tasks. During forward upslope walking the changes in the muscle activity at the ankle joint corresponded to the changes in the ankle joint moment, but during backward upslope walking they do not. In both cases mean plantarflexor muscle activity levels increased; during forward upslope walking the peak plantarflexor moment increases, but during backward upslope walking the peak moment decreases. So, what determines how the ankle joint is affected by the changing moment and power demands in the limb during a task? One explanation is that the proximity to the 'primary' joint during the task determines the effect of the changing demands. During forward upslope walking the ankle was farther from the influence of the primary contributor to the propulsion of the body (the hip joint) than during backward upslope walking (the knee joint). Therefore during backward upslope walking the increased activity of the knee extensors had an effect on the biarticular muscles at both of the adjacent joints, the hip and ankle. During forward upslope walking, the changes in the hip joint muscle activity only had an impact on the knee joint muscles. A second explanation is the degree of power generation required at the ankle joint itself. During forward upslope walking the peak and duration of power generation increased markedly, while during backward upslope walking the increases were small. Increased power generation at the ankle may dominate control of the muscles at that joint, where little change in the power generation may instead lead to a greater influence by the activity of the adjacent knee joint muscles. The non-uniform distribution of changes in mechanics and muscle activity at the joints suggest that control system has the ability to modulate each joint separately.

Relevance of Findings and Future Work

Understanding the mechanics of slope walking may help those in the medical field evaluate if such tasks, which are activities of daily living, pose a problem for patient populations. Such knowledge could lead to the development or improvement of physical therapy and rehabilitation exercise programs. For example, increased eccentric knee extensor activity is often associated with knee pain in patients with osteoarthritis and patellofemoral joint pain (Cipriani et al., 1995). The large power absorption requirements at the knee joint during forward downslope walking could therefore make this task difficult for such patients, which may limit their mobility. During backward downslope walking, however, the increased power absorption occurs at the hip instead. A strategy for these patients could be to walk down hills backwards to avoid knee pain.

The increased reliance on biarticular muscles during upslope walking may be problematic for other patient populations. For example, when the lower limbs are mechanically decoupled contralateral sensorimotor feedback influences the activity of ipsilateral biarticular muscles (Kautz et al., 2002; Ting et al., 1999; Ting et al., 1998). Related to these findings, one may question how the activity of biarticular muscles changes in those who have a sensory deficit, or in lower-limb amputees, who are often missing all or some of their biarticular muscles. If such patients are able to successfully locomote in such environments, does that discount the importance of sensorimotor feedback from the contralateral limb, or does it indicate an adaptation that those patients have developed in response to their deficit? An understanding of the strategies used by patients in these groups could lend further insight into the components of the human locomotor control system, and could lead to advances in treatments or prosthetic device design.

Finally, the changes in control strategy required for slope walking may impact certain patient populations. For example, it has been shown that during gait initiation older adults do not always express the correct motor program, and that the motor program, when expressed, does not function as efficiently as in younger adults (Polcyn et al., 1998). This phenomenon may also occur during slope walking, especially because the slope walking task is much more demanding than gait initiation. Older adults may not be able to consistently use the correct control strategy to produce the necessary muscle activation patterns. A related problem has also been expressed in a different patient group: it has been shown that spinal cord injury patients use different strategies than normal subjects when walking up slopes (Leroux et al., 1999). It could be that descending signals from the central nervous system are very important in activating the correct control strategy for slope walking. Because this pathway is often either interrupted or totally missing in SCI patients, they may not be receiving the correct signals to regulate the motor programs. Investigations of slope walking in infants have the potential to explain the role of descending inputs to the control of slope walking, because infant stepping is thought to be free from descending control and driven predominately by the pattern generators (Lamb and Yang, 2000; Yang et al., 2004). If infants are able to step on sloped surfaces with a consistent motor pattern, that would suggest the existence of a basic control program for slope walking. Data from infants could be compared to that from spinal cord injury patients to determine the extent to which the basic control pattern exists in adults. This knowledge may help improve treatment programs for SCI patients.

Conclusions

In conclusion, this project has detailed the biomechanics of both forward and backward upslope and downslope walking; these data were for the most part unavailable before this report. The forward walking tasks are activities of daily living, and so a thorough understanding of these tasks can benefit several different patient populations, as discussed above. Backward slope walking is a novel task where the mechanics were completely undefined prior to this investigation. More importantly, the backward slope walking tasks were useful perturbations for gaining more information about locomotor control strategies. Using these data we confirmed that the power requirements during a task provide an important signal to the control system, and therefore appear to influence the muscle activation patterns. The actual mechanism of this signal (force output, force feedback, etc.) cannot be determined from these findings, but it would be interesting to explore this idea. Although this study has increased the knowledge base on the tasks of forward and backward slope walking, and on the control of such locomotor tasks, further research is warranted. The examples given above demonstrate that further studies on slope walking including different populations, such as amputees and older adults, would be beneficial in understanding the abilities and/or deficits in sensorimotor integration in these groups. Understanding the manifestations of the deficit, i.e. what tasks or components of the task do these patients have the most difficulty accomplishing, may lead to an increased understanding of the deficit itself. This increased understanding could, in turn, lead to better treatment programs, rehabilitation training, or device design.

APPENDIX A

FORWARD WALKING ENSEMBLE AVERAGE EMG SIGNALS

The data in this Appendix are intended to supplement the EMG data presented in Chapters 5 and 6 (Figures 17-19 and 29-31). In those figures, the horizontal lines show the group average burst onset, offset, and mean activity level, but not the distribution of muscle activity during the burst. The actual EMG data is intended to be “representative” of this distribution, but in reality is from a single trial from one subject. An ensemble average of the EMG activity would help fill in this gap in information, but also requires a smoothing function that distorts the magnitude of the EMG activity. To provide the most complete picture of the muscle activity, therefore, the data is presented in all three ways.

The ensemble average EMG activity was calculated as follows. For each subject, the mean activity level of the major stance phase burst of all eight muscles was calculated for each level walking trial, and the average mean activity level for each muscle was then calculated. The raw EMG data from each trial were bandpass filtered (30-500 Hz), wavelet filtered, rectified, and smoothed using a moving window average of 39 ms (Leroux et al., 1999). The stance phase of the smoothed data was then normalized to 1000 points, and the magnitude of the smoothed data was normalized to the previously calculated average mean activity level for that muscle during level walking. Once the smoothed normalized data had been calculated for all trials for each subject, the data were ensemble averaged across all subjects for each walking grade. The result is a single curve for each muscle at each walking grade representing the group average muscle activity. Those curves are presented in the following figures.

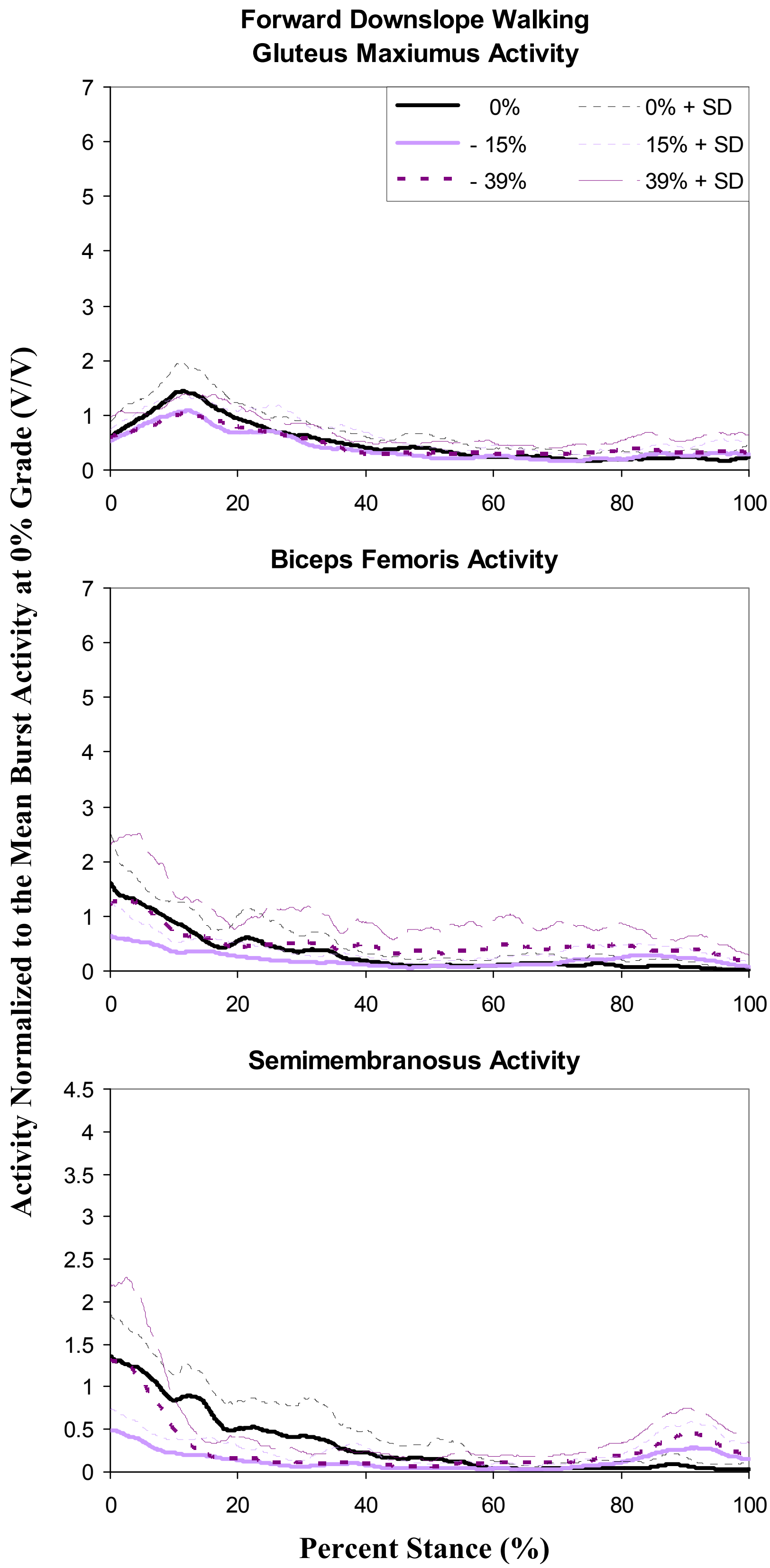


Figure 62. Ensemble Average GM, BF, and SM Activity during Forward Downslope Walking for comparison to Figure 17.

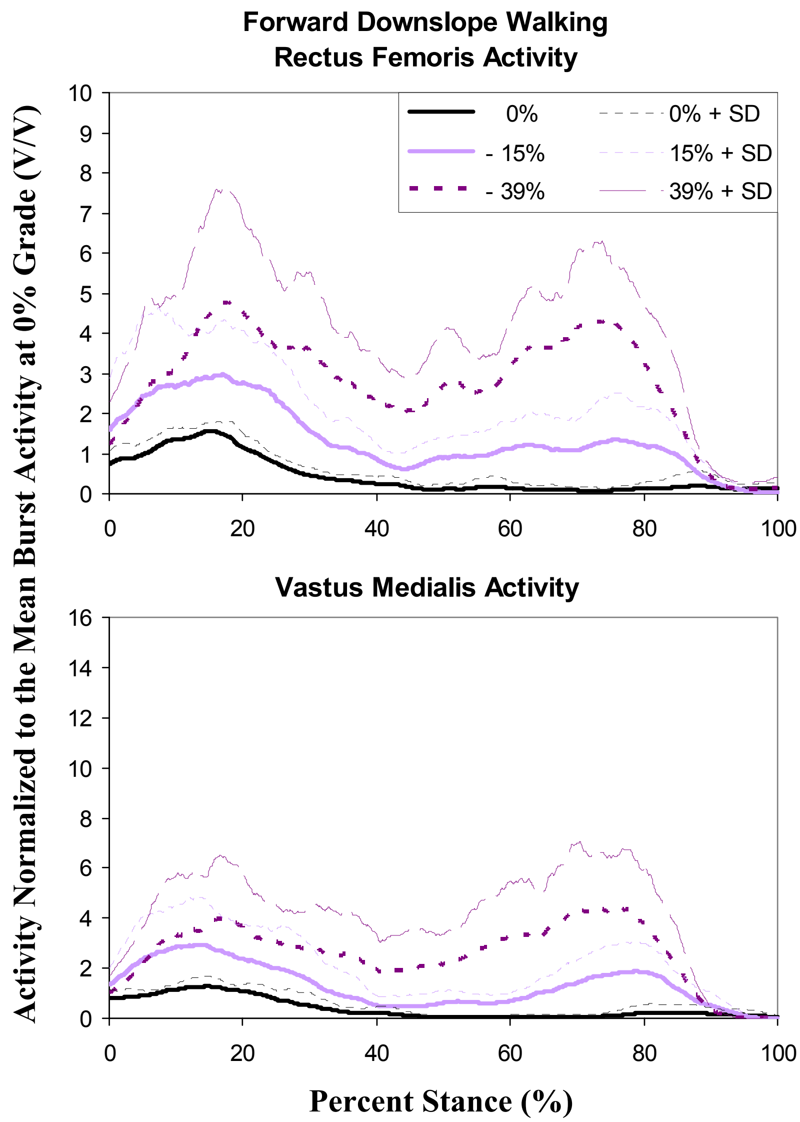


Figure 63. Ensemble Average RF and VM Activity during Forward Downslope Walking for comparison to Figure 18.

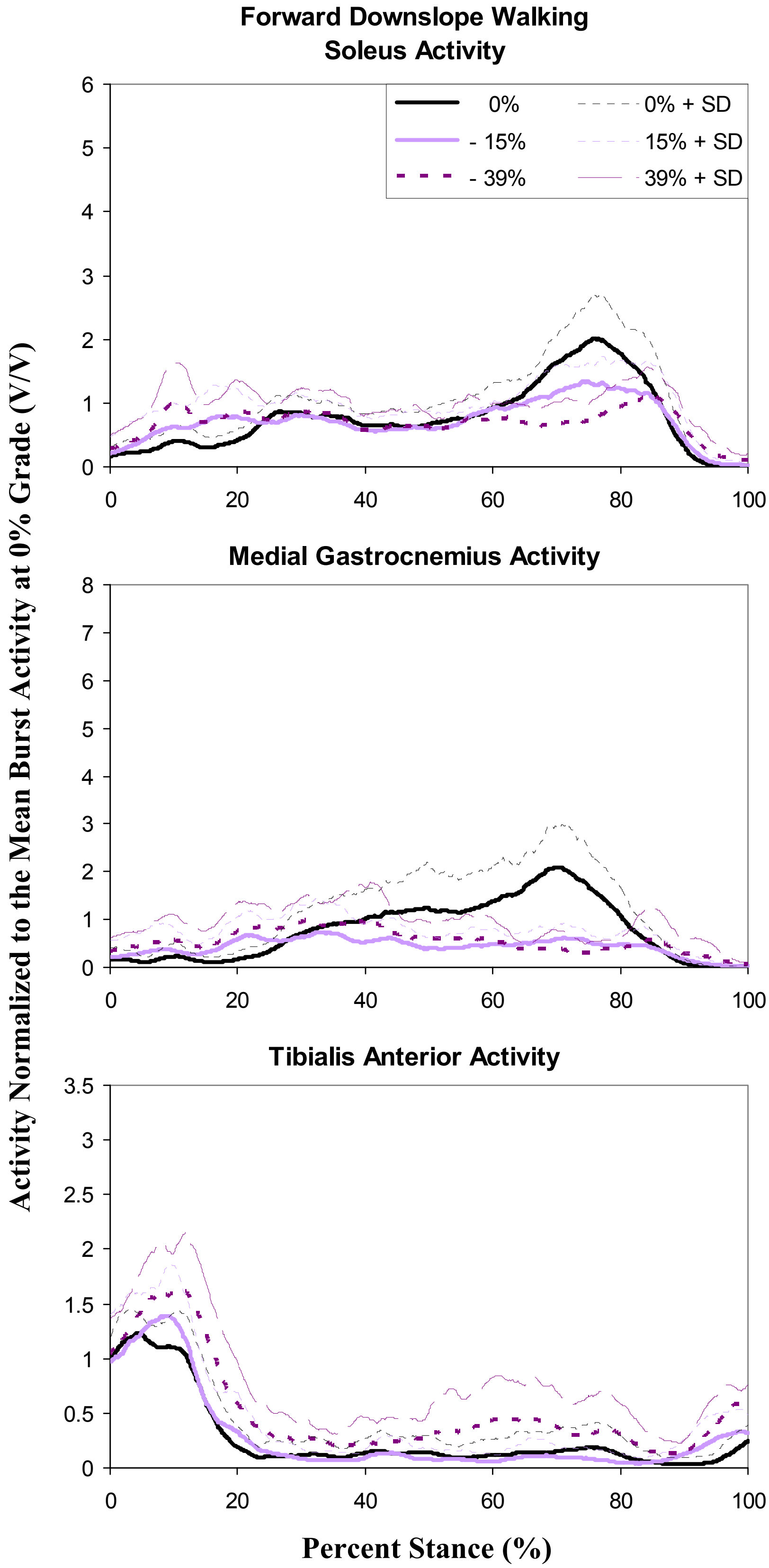


Figure 64. Ensemble Average Sol, MG, and TA Activity during Forward Downslope Walking for comparison to Figure 19.

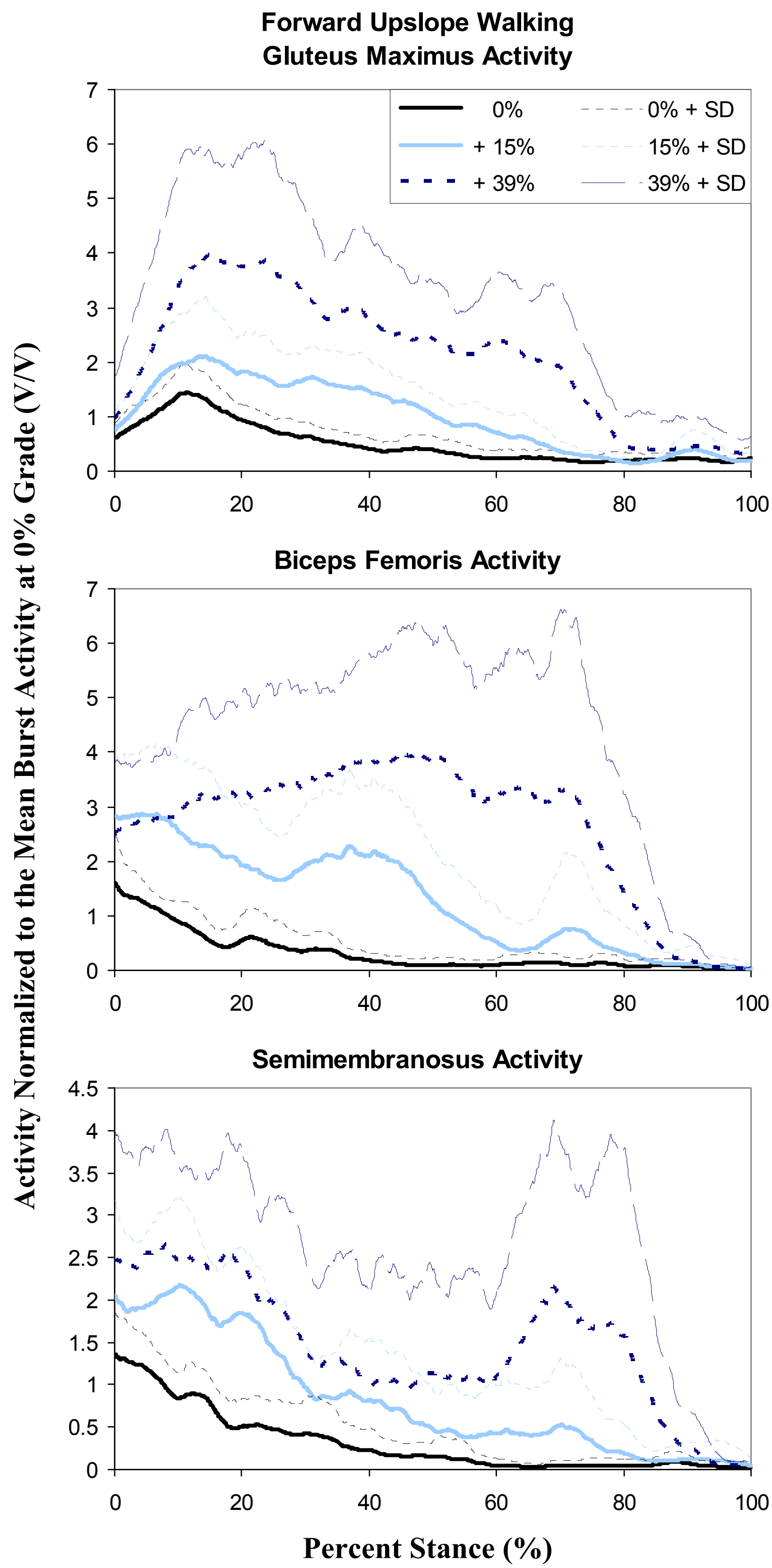


Figure 65. Ensemble Average GM, BF, and SM Activity during Forward Upslope Walking for comparison to Figure 29.

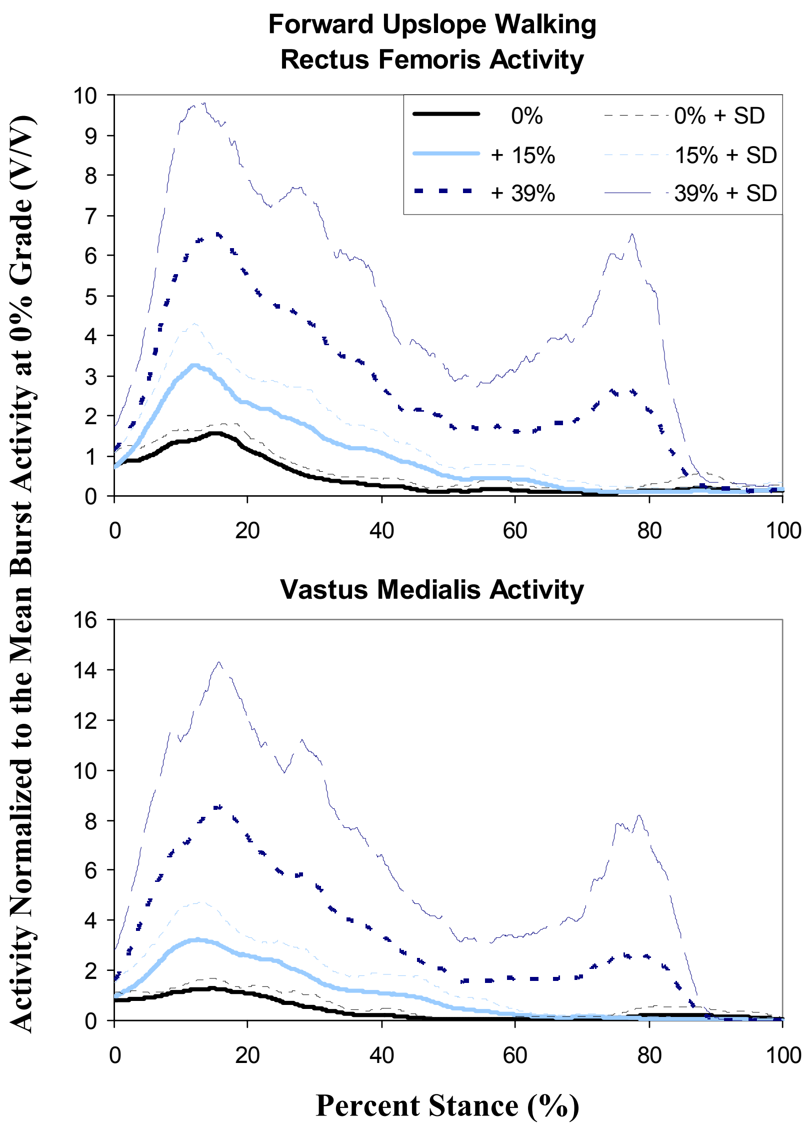


Figure 66. Ensemble Average RF and VM Activity during Forward Upslope Walking for comparison to Figure 30.

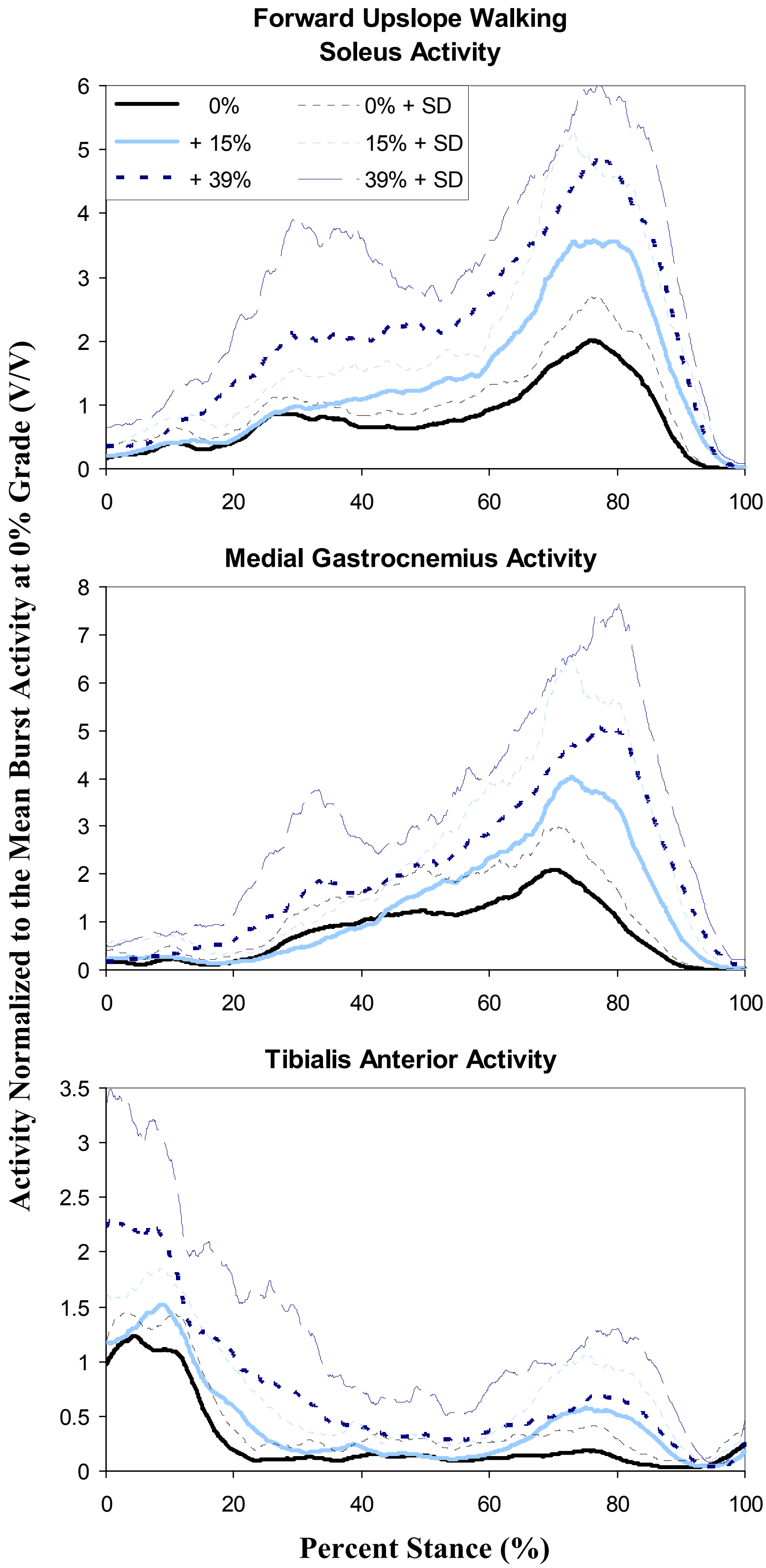


Figure 67. Ensemble Average Sol, MG, and TA Activity during Forward Upslope Walking for comparison to Figure 31.

APPENDIX B

ADDITIONAL BACKWARD UPSLOPE WALKING DATA

Joint Kinematics

Ensemble average joint angle curves are presented in Figure 68. There was a significant grade effect for all variables (ANOVA); POI data and the results of follow-up comparisons are given in Table 25. During backward level walking the ankle joint (Figure 68A) is almost neutral at toe strike, dorsiflexes in early stance as the body weight is transferred to the stance limb, plantarflexes in midstance (AA1) as the body moves backward over the foot, and then dorsiflexes again for heel off (AA2) and swing. The knee joint (Figure 68B) is flexed at toe strike, extends throughout stance (KA1, KA2), and then flexes during swing (KA3) to ensure toe clearance. The hip joint (Figure 68C) is also almost neutral at toe strike, extends slightly in early stance (HA1), and then flexes for the majority of the stride (HA2) as the body moves backward over the foot, until mid-swing (HA3) when it begins to extend to prepare for toe strike.

During upslope walking the ankle joint angle in late swing (AA3) and early stance is almost identical to that during level walking, but then the ankle plantarflexes more during upslope stance as the foot remains flat on the ramp surface and the body moves up and back over it (AA1, AA2). The knee and hip joint angles are similar to those for level walking at the stance-swing transition (heel off, KA2 and HA2), but the knee joint is more flexed during swing (KA3) and both joints are more flexed for the majority of stance (KA1, HA1). The knee and hip must be more flexed at toe strike for contact to occur up the ramp behind the contralateral foot. Because the knee and hip angles at heel off do not change, this increased flexion at toe strike results in more knee extension and less hip flexion during the stance phase.

One group has investigated the kinematics of backward upslope walking in humans at grades up to 10% (Cipriani et al., 1995). The joint angle curves are not reported, but joint angle values at select points in the stride cycle are presented. Despite the difference in the walking grades, the reported data should show a similar trend to what is observed here. For the knee angle, the data is comparable, with the authors reporting increased flexion at TS and during mid-stance, but not at HO (Cipriani et al., 1995). However, the authors report no change in the hip joint angle and changes in the ankle joint angle that are not consistent with the findings presented here. These discrepancies could be due to the smaller walking grade, the use of a treadmill rather than a ramp, or a different joint angle convention (the authors do not clearly report the joint angle convention) in the published report compared to the protocol used here.

Table 25. Backward Upslope Walking Joint Angle POI Data

	0%	+ 15%	+ 39%
AA1	9.07 (6.86)	15.89 § (6.61)	28.74 §† (6.78)
AA2	-0.70 (5.47)	1.599 (7.09)	8.03 §† (5.79)
AA3	5.73 (5.41)	6.720 (6.56)	9.635 (5.56)
KA1	17.84 (7.18)	41.40 § (4.27)	59.71 §† (5.34)
KA2	5.45 (5.78)	4.713 (5.12)	6.171 (4.67)
KA3	53.13 (7.69)	63.36 § (5.77)	76.11 §† (6.40)
HA1	-2.13 (8.60)	7.61 § (10.03)	19.61 §† (10.60)
HA2	26.61 (8.79)	23.23 § (9.55)	21.71 § (9.68)
HA3	29.49 (8.55)	27.532 (9.35)	31.436 (9.86)

Note: Mean (SD) of the joint kinematic POI (all values in degrees), averaged across all subjects for each grade. § - indicates a significant difference from 0% grade ($p<0.05$), † - indicates a significant difference between 15% and 39% in the same walking direction (upslope or downslope) ($p<0.05$). As defined, point AA1 did not exist at -39% walking grade, and so is not included in the table.

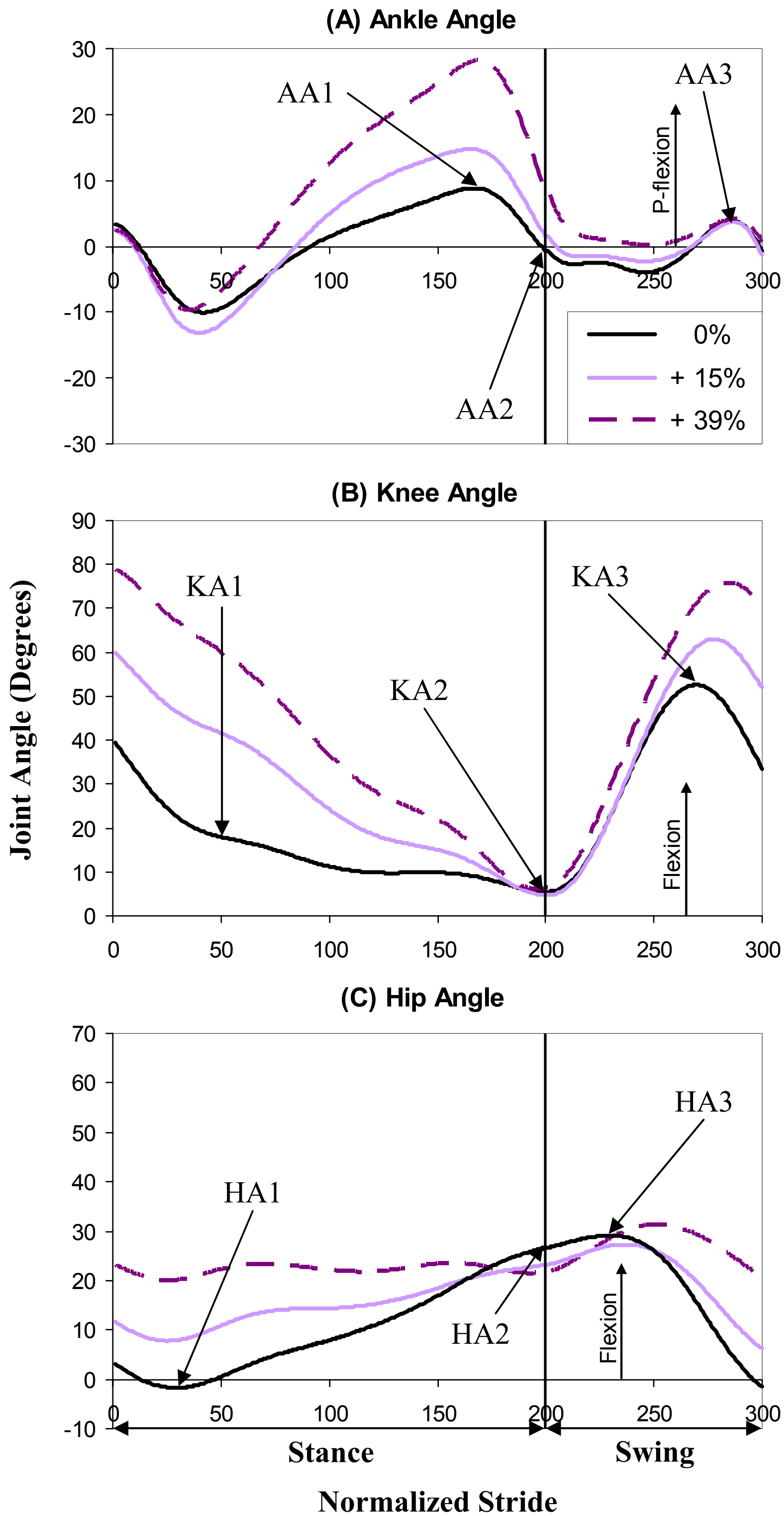


Figure 68. Average Joint Angles during Backward Upslope Walking. For joint angle convention see Figure 4. Stance begins with toe strike, vertical line marks heel off.

Ground Reaction Forces

The GRF curves are shown in Figures 69 and 70. There was a significant grade effect for all variables except N1 (ANOVA). Values of the POI and results of the follow-up comparisons are given in Table 26. Backward upslope walking required little or no braking force (AP1), but large propulsive forces (AP2) (Figure 69A). The ML force was laterally directed for all conditions (Figure 69B). The first peaks of the normal (N1, Figure 69C) and resultant force (FR1, Figure 70) curves showed no significant changes with walking grade, but the second peaks (N2, FR2) increased as the walking grade increased, which is consistent with the higher propulsive requirements during backward upslope walking.

Table 26. Backward Upslope Walking GRF Component POI Data

	0%	+ 15%	+ 39%
AP1	1.22 (0.25)	0.32 § (0.34)	N/A
AP2	-1.25 (0.30)	-2.67 § (0.23)	-4.69 §† (0.50)
N1	11.33 (1.18)	11.61 (0.92)	10.82 (0.89)
N2	8.86 (0.26)	9.46 § (0.48)	9.25 (0.77)
FR1	11.76 (0.90)	11.65 (0.90)	11.29 (0.90)
FR2	8.92 (0.24)	9.76 § (0.50)	10.26 § (0.81)

Note: Mean (SD) of the GRF component POI (in N/kg). As defined points AP1 and AP2 did not exist for +39% and -39%, respectively. Shading indicates no grade effect was detected by the ANOVA. § - significant difference from 0% ($p<0.05$), † - significant difference from 15% in the same direction (up/down) ($p<0.05$).

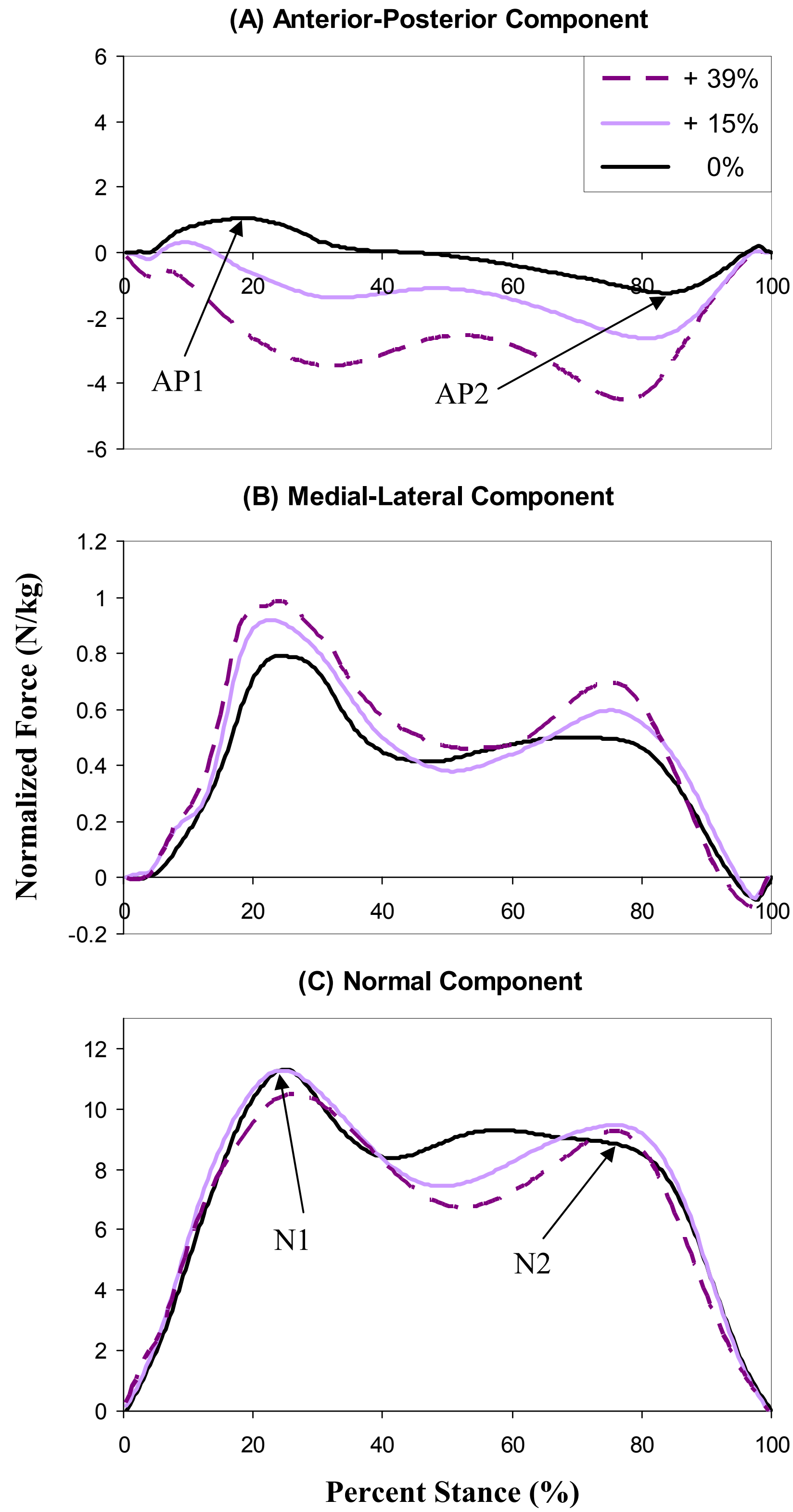


Figure 69. Average GRF Components during Backward Upslope Walking. Positive forces indicate posteriorly and laterally directed foot forces.

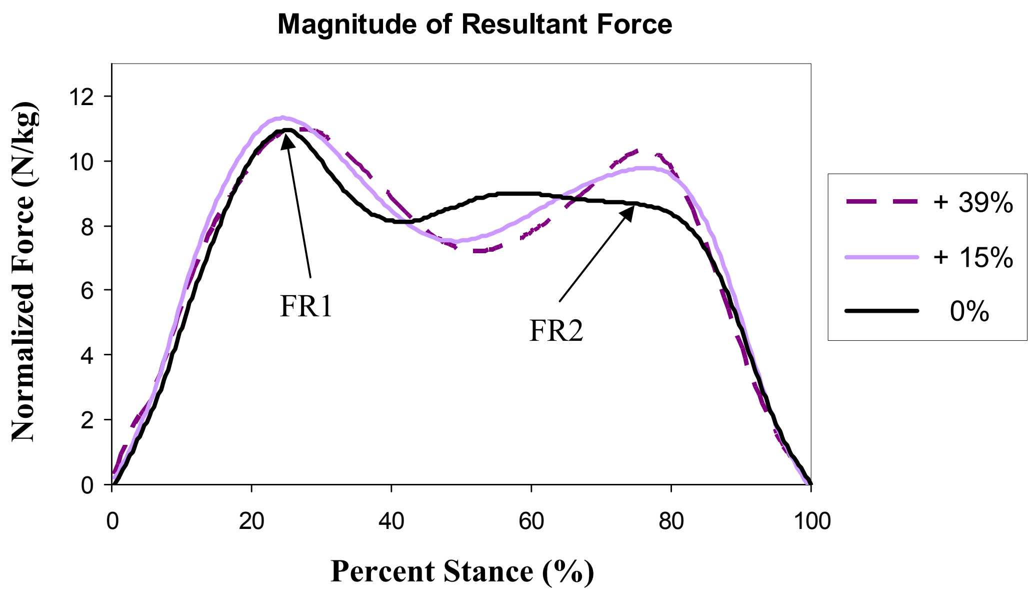


Figure 70. Average Resultant Force during Backward Upslope Walking, calculated as the sum of the normal and AP component.

Muscle Activity

The data presented here are intended to supplement the EMG data presented in Chapter 7 (Figures 45-47). In those figures, the horizontal lines show the group average burst onset, offset, and mean activity level, but not the distribution of muscle activity during the burst. The raw data is intended to be "representative" of this distribution, but in reality is from a single trial from one subject. An ensemble average of the EMG activity would help fill in this gap in information, but also requires a smoothing function that distorts the magnitude of the EMG activity. To provide the most complete picture of the muscle activity, therefore, the data is presented in all three ways.

The ensemble average EMG activity was calculated as follows. First, for each subject the mean activity level of the major stance phase burst of all eight muscles was calculated for each level walking trial, and the average mean activity level for each muscle was then calculated. This is the same number that the average mean activity of the bursts from the slope walking trials was normalized to, which resulted in the average mean activity level for each muscle as a percent of the activity at level walking (reported as “a” on the EMG figures in Chapter 7). Next, the raw EMG data from each trial were bandpass filtered (30-500 Hz), wavelet filtered, rectified, and smoothed using a moving window average of 39 ms (Leroux et al., 1999). The stance phase of the smoothed data was then normalized to 1000 points, and the magnitude of the smoothed data was normalized to the previously calculated average mean activity level for that muscle during level walking. Once the smoothed normalized data had been calculated for all trials for each subject, the data were ensemble averaged across all subjects for each walking grade. The result is a single curve for each muscle at each walking grade representing the group average muscle activity pattern. These curves are presented as bold lines in the following figures; the thin dashed lines represent the average EMG plus one standard deviation. Note that the standard deviations are large for some of the muscles, but the variability is predominately in the magnitude of the EMG activity; the patterns of muscle activity are consistent across subjects.

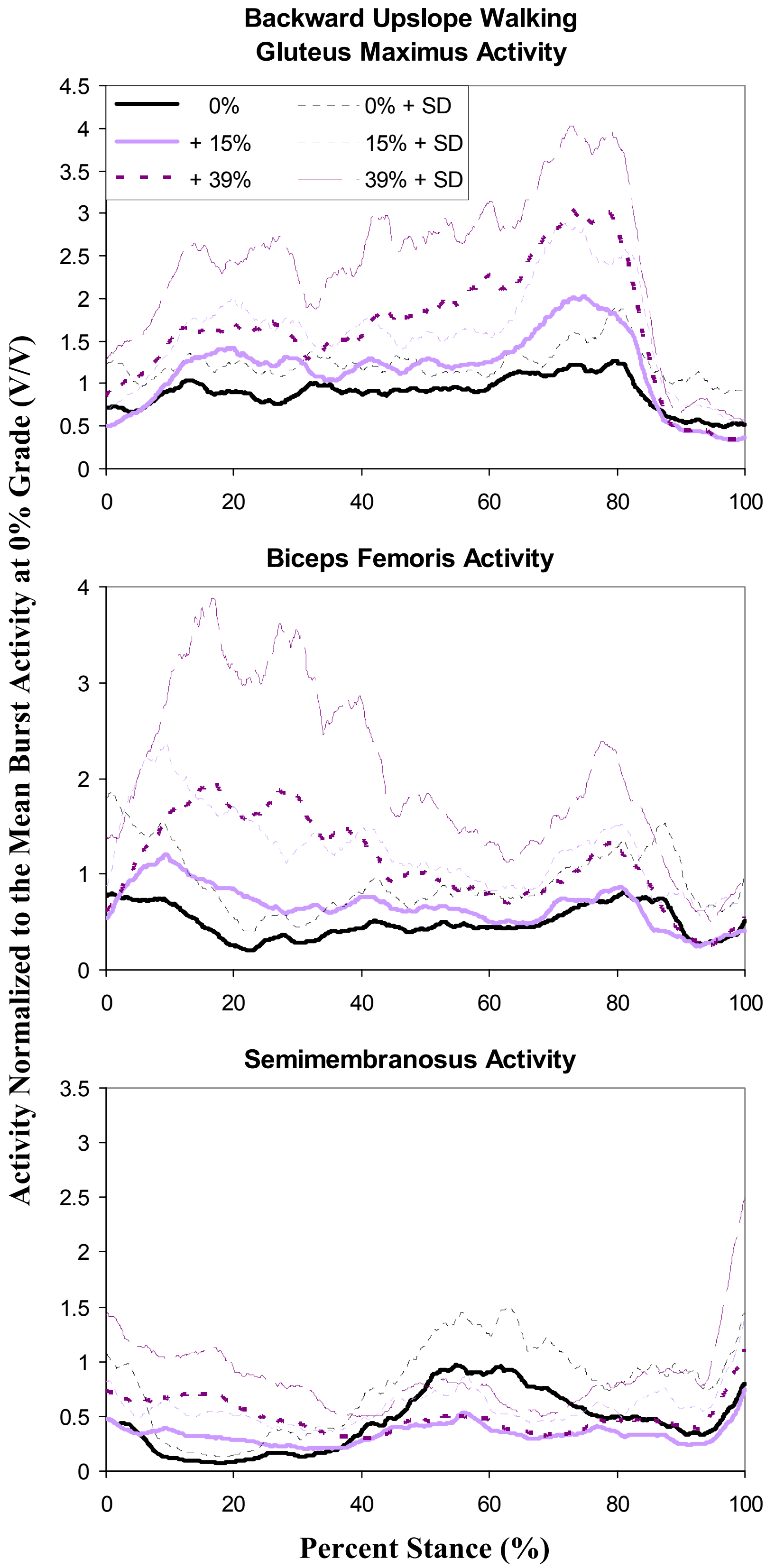


Figure 71. Ensemble Average GM, BF, and SM Activity during Backward Upslope Walking for comparison to Figure 43.

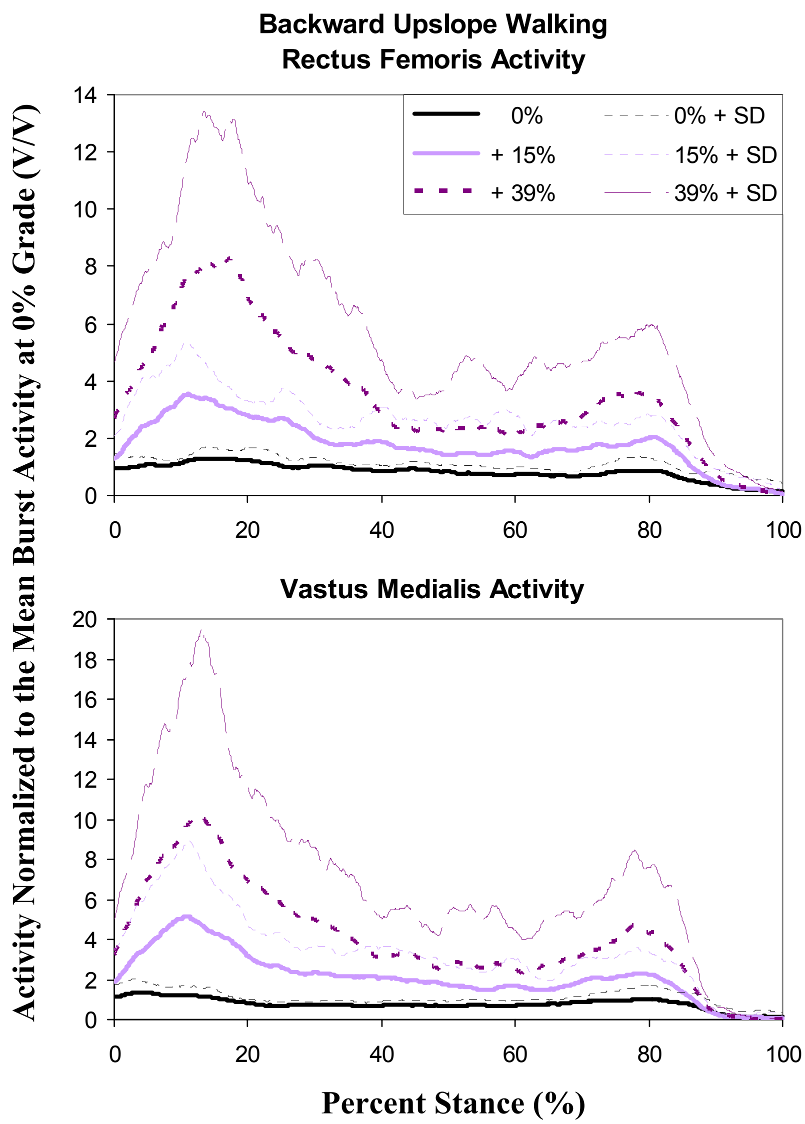


Figure 72. Ensemble Average RF and VM Activity during Backward Upslope Walking for comparison to Figure 46.

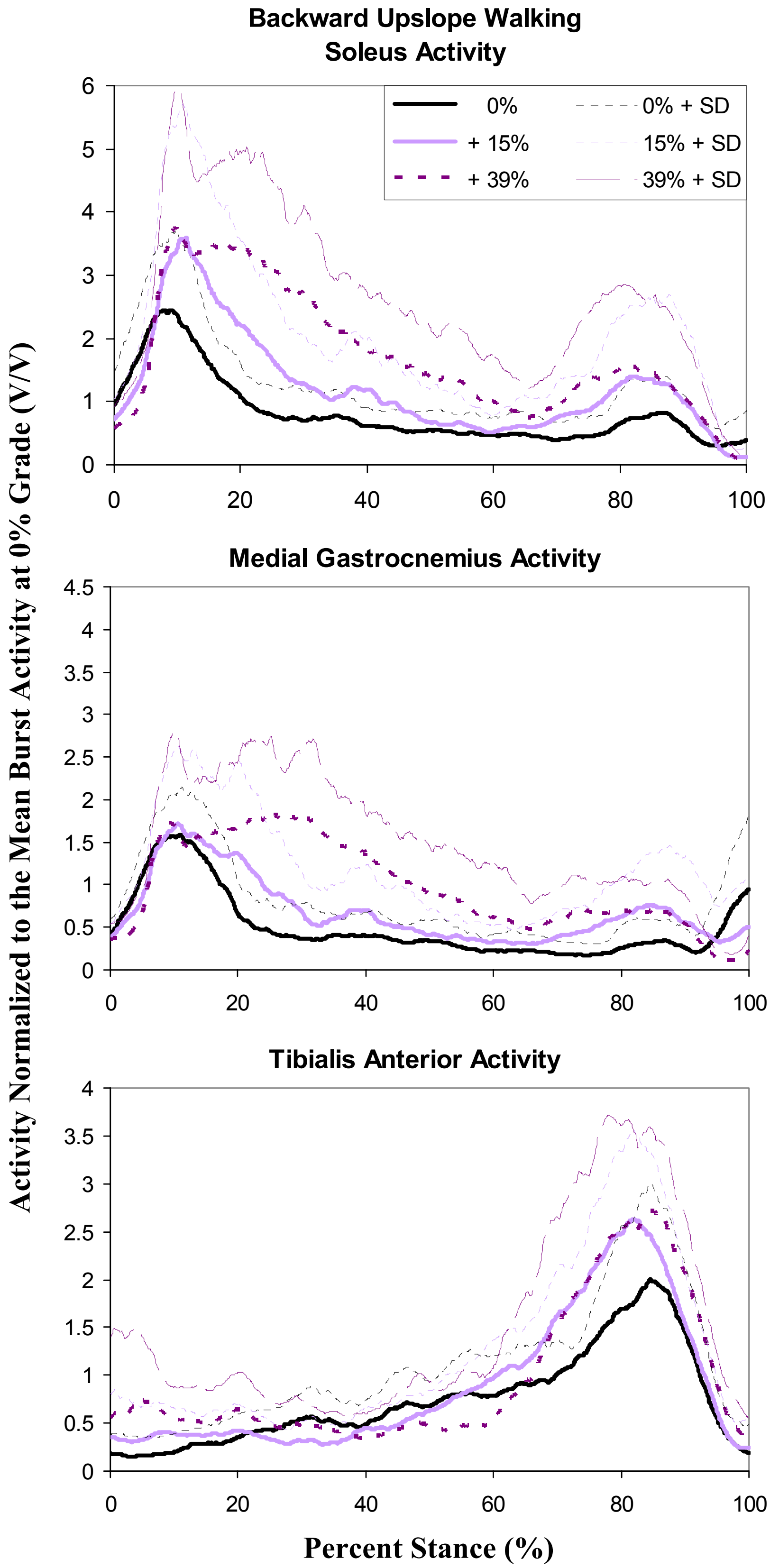


Figure 73. Ensemble Average Sol, MG, and TA Activity during Backward Upslope Walking for comparison to Figure 47.

APPENDIX C

ADDITIONAL BACKWARD DOWNSLOPE WALKING DATA

Joint Kinematics

Ensemble average joint angle curves are presented in Figure 74. There was a significant grade effect for all variables (ANOVA); POI data and the results of follow-up comparisons are given in Table 27. Compared to the level walking joint angles, the ankle and knee joint angles during backward downslope walking are slightly more extended (plantarflexed) at initial contact as the foot reaches out behind and below the contralateral foot to make contact. The ankle joint (Figure 74A) dorsiflexes more in early stance as the entire foot makes contact with the ramp surface; the increased flexion is maintained for all of stance (AA2) and part of swing, until the ankle extends for toe contact. The knee joint (Figure 74B) also flexes more from mid-stance to mid-swing (KA2) as the body is lowered down the ramp. The hip joint (Figure 74C) is more flexed during downslope walking than during level walking for the entire stride (HA1, HA2, HA3), with the largest increases at heel off. This increased flexion occurs because the body is being lowered down the slope, resulting in the stance limb being in front and above the body at the end of stance.

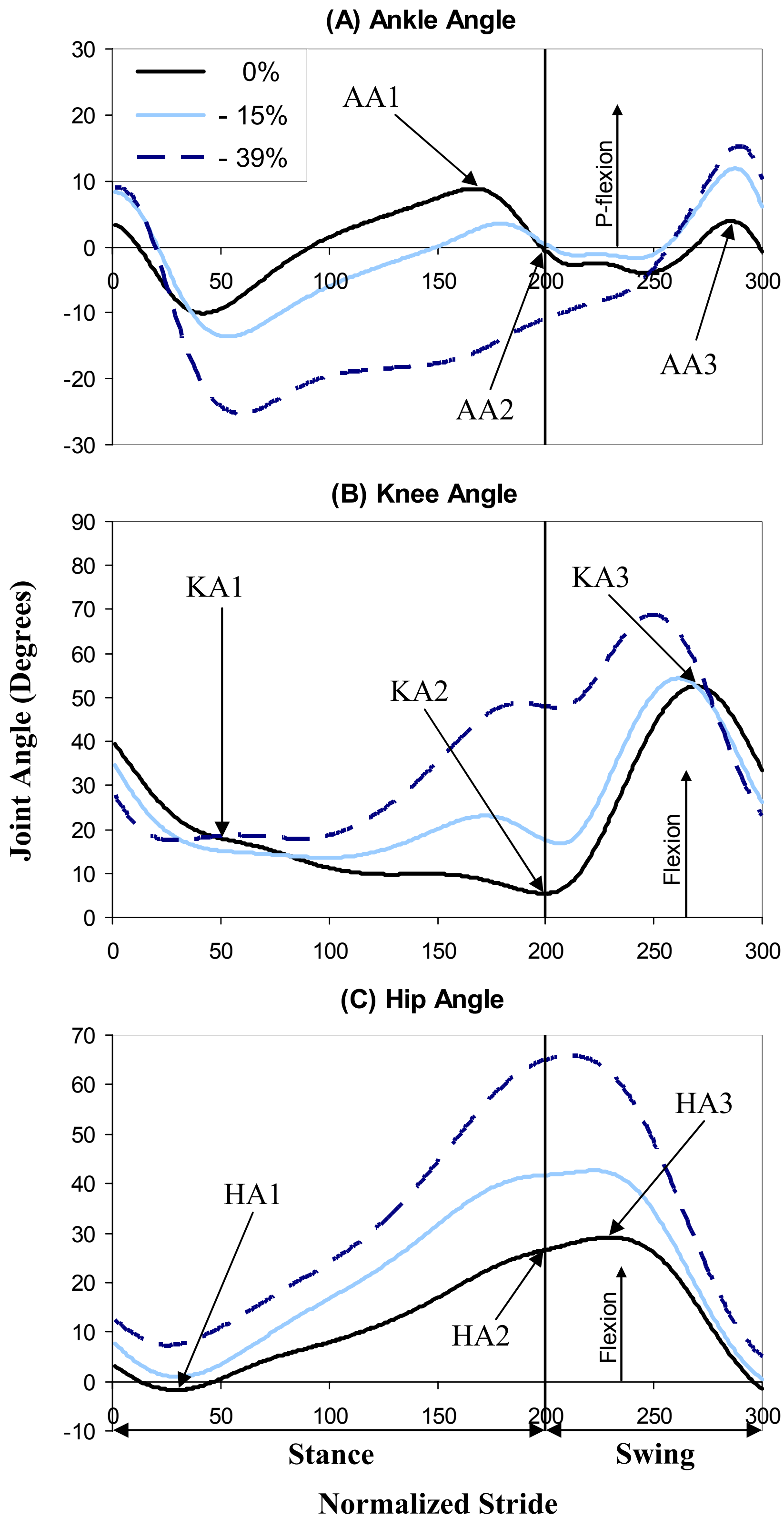


Figure 74. Average Joint Angles during Backward Downslope Walking. For joint angle convention see Figure 4. Stance begins with toe strike, vertical line marks heel off.

Table 27. Backward Downslope Walking Joint Angle POI Data

	- 39%	- 15%	0%
AA1	N/A	4.05 § (6.21)	9.07 (6.86)
AA2	-10.96 §† (6.49)	0.346 (5.63)	-0.70 (5.47)
AA3	16.38 § (7.79)	12.76 § (5.95)	5.73 (5.41)
KA1	18.433 (6.95)	15.089 (6.55)	17.84 (7.18)
KA2	47.77 §† (6.02)	17.47 § (7.09)	5.45 (5.78)
KA3	69.52 §† (6.20)	55.025 (6.11)	53.13 (7.69)
HA1	6.39 §† (6.72)	0.099 (7.08)	-2.13 (8.60)
HA2	64.95 §† (10.69)	41.74 § (10.33)	26.61 (8.79)
HA3	66.38 §† (10.01)	43.66 § (8.96)	29.49 (8.55)

Note: Mean (SD) of the joint kinematic POI (all values in degrees), averaged across all subjects for each grade. § - indicates a significant difference from 0% grade ($p<0.05$), † - indicates a significant difference between 15% and 39% in the same walking direction (upslope or downslope) ($p<0.05$). As defined, point AA1 did not exist at -39% walking grade, and so is not included in the table.

Ground Reaction Forces

The GRF curves are shown in Figures 75 and 76. There was a significant grade effect for all variables except N1 (ANOVA). Values of the POI and results of the follow-up comparisons are given in Table 28. Backward downslope walking required large braking forces and small propulsive forces. In fact, the AP force (Figure 75A) was entirely a braking force at the steepest grade. The ML force was laterally directed for all grades (Figure 75B). The first peak of the normal force (N1, Figure 75C) showed no grade effect, but the corresponding peak of the resultant force (FR1, Figure 76) increased slightly at the steepest grade, indicating the increased braking force requirement. The second peaks of the normal and resultant force curves (N2, FR2) decreased as the walking grade decreased, which is consistent with the lower propulsive requirements.

Table 28. Backward Downslope Walking GRF Component POI Data

	- 39%	- 15%	0%
AP1	5.40 §† (0.62)	2.87 § (0.42)	1.22 (0.25)
AP2	N/A	-0.40 § (0.28)	-1.25 (0.30)
N1	12.64 (1.02)	12.52 (1.22)	11.33 (1.18)
N2	6.69 §† (0.44)	7.87 § (0.38)	8.86 (0.26)
FR1	13.73 § (1.17)	12.83 (1.28)	11.76 (0.90)
FR2	6.93 §† (0.46)	7.88 § (0.38)	8.92 (0.24)

Note: Mean (SD) of the GRF component POI (in N/kg). As defined points AP1 and AP2 did not exist for +39% and -39%, respectively. Shading indicates no grade effect was detected by the ANOVA. § - significant difference from 0% ($p<0.05$), † - significant difference from 15% in the same direction (up/down) ($p<0.05$).

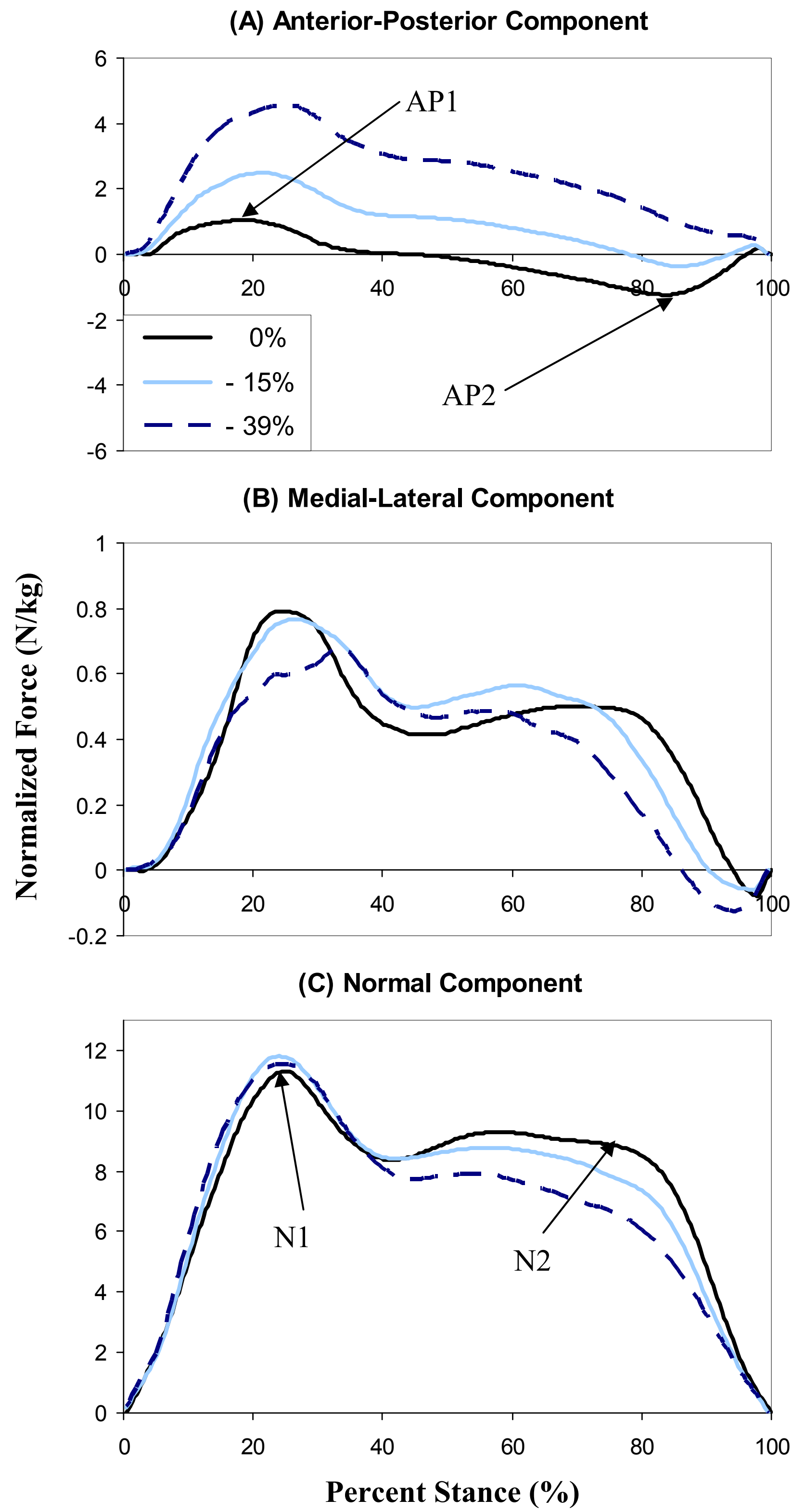


Figure 75. Average GRF Components during Backward Downslope Walking. Positive forces indicate posteriorly and laterally directed foot forces.

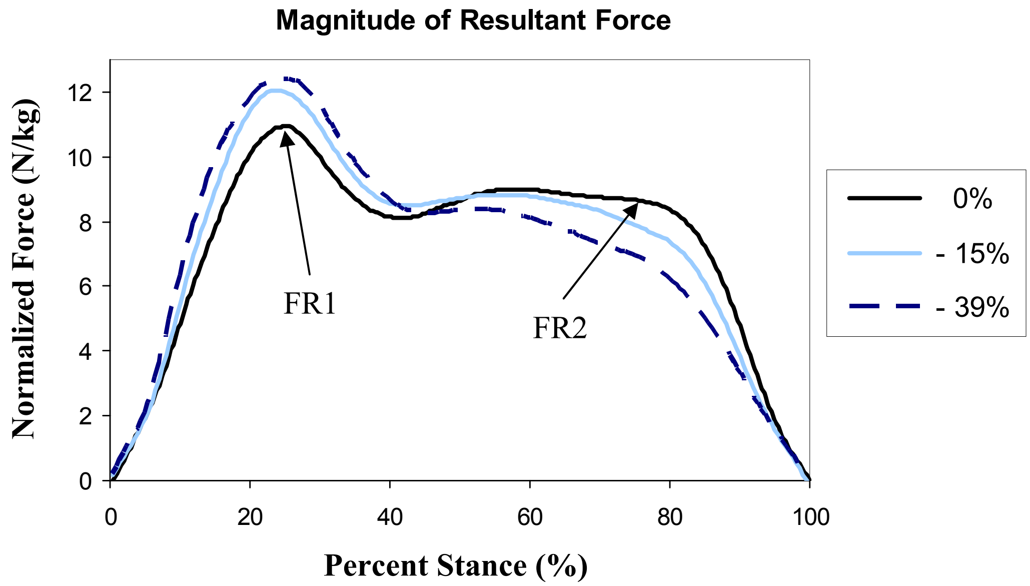


Figure 76. Average Resultant Force during Backward Downslope Walking, calculated as the sum of the normal and AP component.

Muscle Activity

The data presented here are intended to supplement the EMG data presented in Chapter 7 (Figures 55-57). In those figures, the horizontal lines show the group average burst onset, offset, and mean activity level, but not the distribution of muscle activity during the burst. The raw data is intended to be "representative" of this distribution, but in reality is from a single trial from one subject. An ensemble average of the EMG activity would help fill in this gap in information, but also requires a smoothing function that distorts the magnitude of the EMG activity. To provide the most complete picture of the muscle activity, therefore, the data is presented in all three ways.

The ensemble average EMG activity was calculated as follows. First, for each subject the mean activity level of the major stance phase burst of all eight muscles was calculated for each level walking trial, and the average mean activity level for each muscle was then calculated. This is the same number that the average mean activity of the bursts from the slope walking trials was normalized to, which resulted in the average mean activity level for each muscle as a percent of the activity at level walking (reported as "a" on the EMG figures in Chapter 7). Next, the raw EMG data from each trial were bandpass filtered (30-500 Hz), wavelet filtered, rectified, and smoothed using a moving window average of 39 ms (Leroux et al., 1999). The stance phase of the smoothed data was then normalized to 1000 points, and the magnitude of the smoothed data was normalized to the previously calculated average mean activity level for that muscle during level walking. Once the smoothed normalized data had been calculated for all trials for each subject, the data were ensemble averaged across all subjects for each walking grade. The result is a single curve for each muscle at each walking grade representing the group average muscle activity pattern. These curves are presented as bold lines in the following figures; the thin dashed lines represent the average EMG plus one standard deviation. Note that the standard deviations are large for some of the muscles, but the variability is predominately in the magnitude of the EMG activity; the patterns of muscle activity are consistent across subjects.

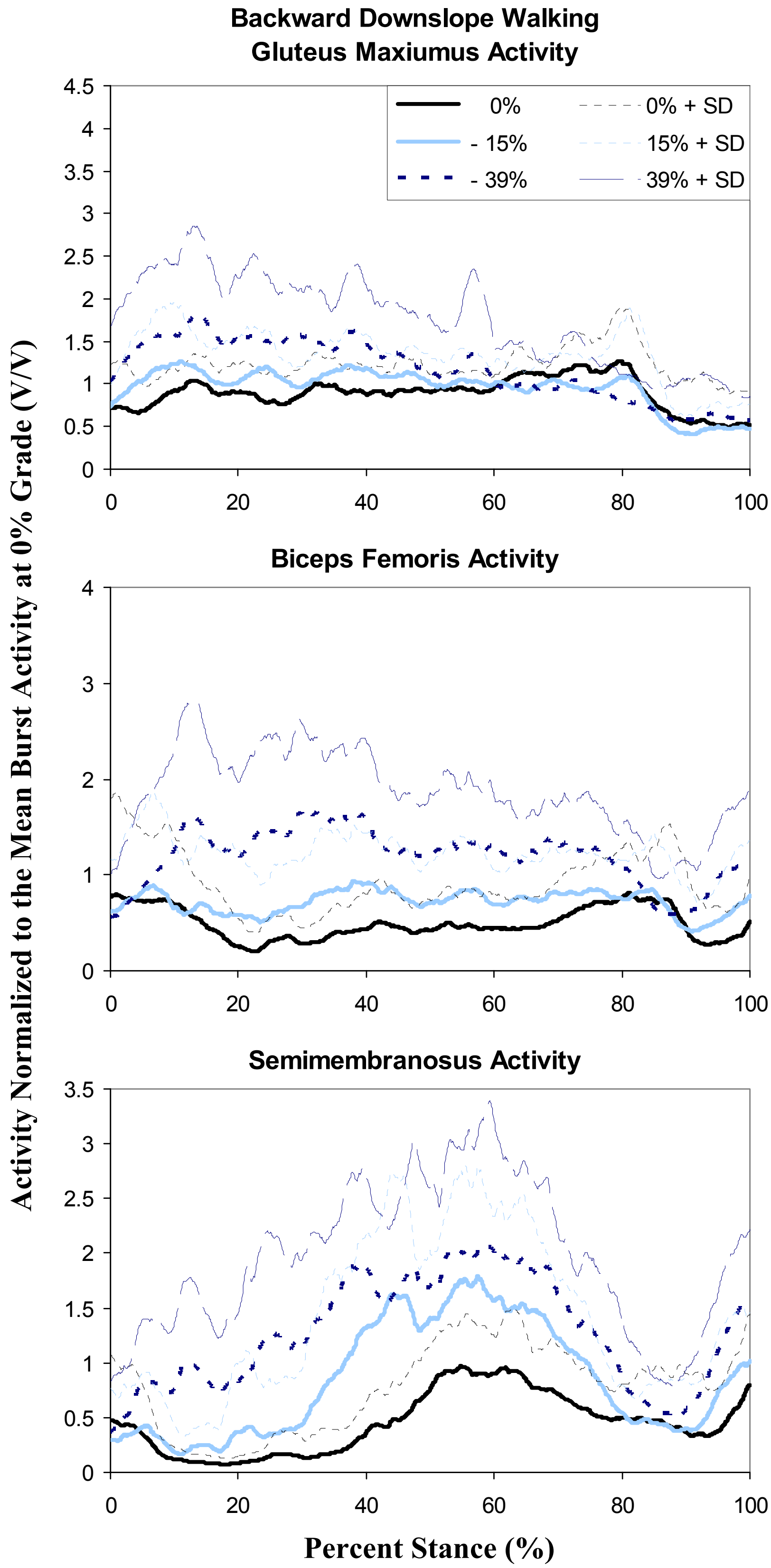


Figure 77. Ensemble Average GM, BF, and Sol Activity during Backward Downslope Walking for comparison to Figure 55.

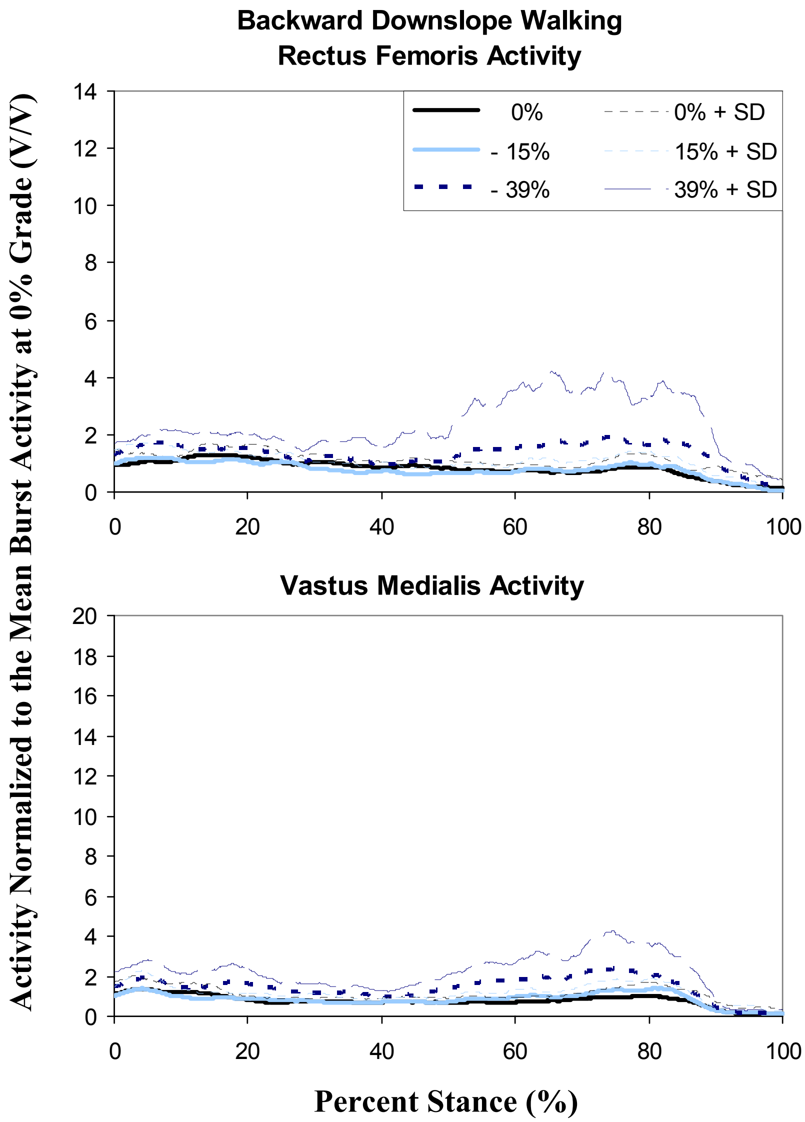


Figure 78. Ensemble Average RF and VM Activity during Backward Downslope Walking for comparison to Figure 56.

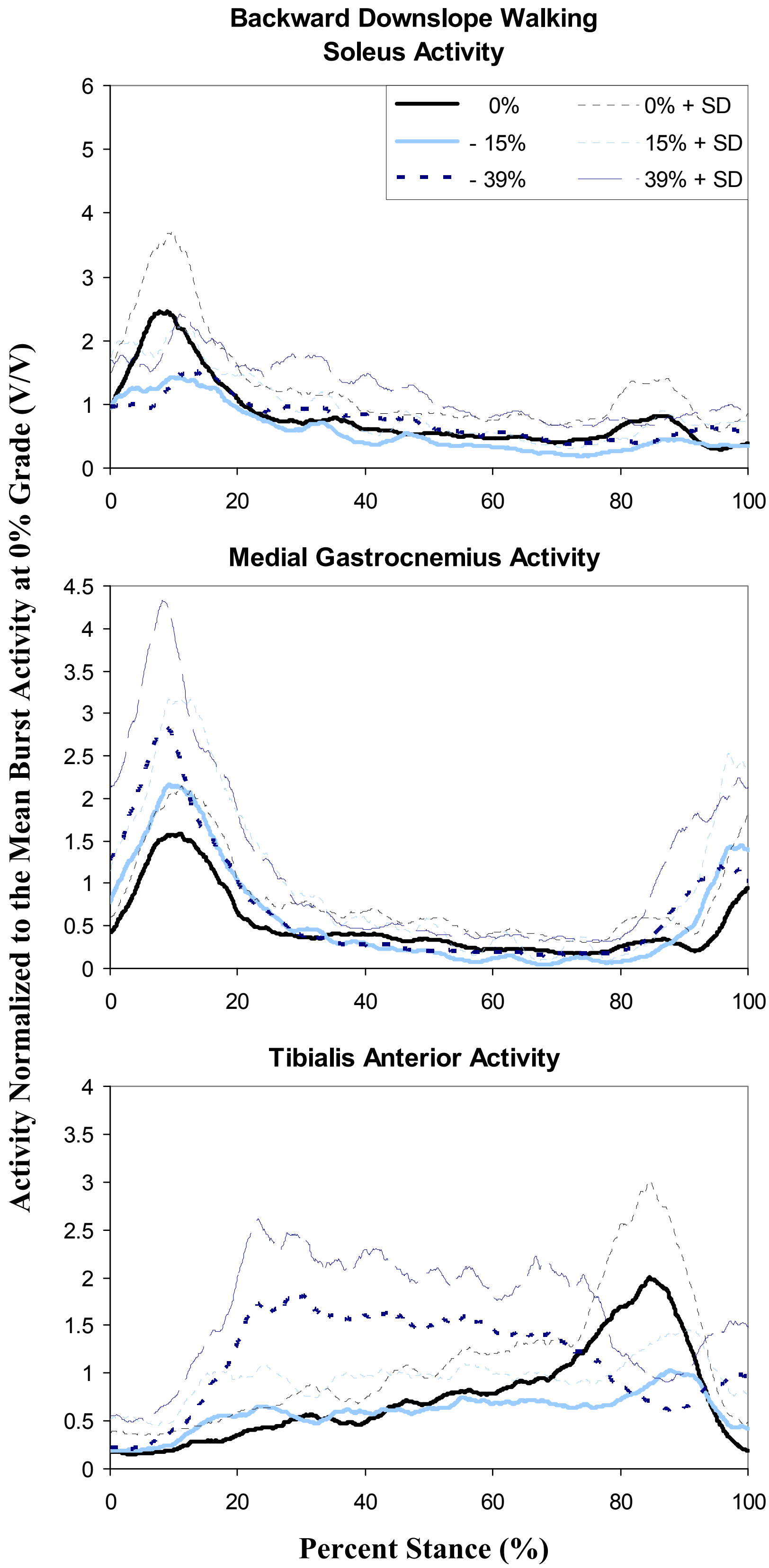


Figure 79. Ensemble Average Sol, MG, and TA Activity during Backward Downslope Walking for comparison to Figure 57.

WORKS CITED

Abelew, TA, Miller MD, Cope TC, and Nichols TR (2000). Local loss of proprioception results in disruption of interjoint coordination during locomotion in the cat. *Journal of Neurophysiology*, **84**(5): 2709-14.

Antonsson, EK, and Mann RW (1985). The frequency content of gait. *Journal of Biomechanics*, **18**(1): 39-47.

Ashley-Ross, MA, and Lauder GV (1997). Motor patterns and kinematics during backward walking in the pacific giant salamander: evidence for novel motor output. *Journal of Neurophysiology*, **78**(6): 3047-60.

Basmajian, JV, and Blumenstein R (1989). Electrode Placement in Electromyographic Biofeedback. In: Biofeedback, Principles and Practice for Clinicians. JV Basmajian, Ed. Baltimore, Williams & Wilkins: 369-82.

Brandell, BR (1977). Functional roles of the calf and vastus muscles in locomotion. *American Journal of Physical Medicine*, **56**(2): 59-74.

Brown, TG (1911). The intrinsic factors in the act of progression in the mammal. *Proceedings of the Royal Society of London*, **84**(572): 308-19.

Brown, TG (1914). On the nature of the fundamental activity of the nervous centres; together with an analysis of the conditioning of rhythmic activity in progression, and a theory of the evolution in function in the nervous system. *J Physiology*, **48**: 18-46.

Brunt, D, Liu SM, Trimble M, Bauer J, and Short M (1999). Principles underlying the organization of movement initiation from quiet stance. *Gait and Posture*, **10**(2): 121-8.

Buczek, FL (1990). Three-dimensional kinematics and kinetics of the ankle and knee joints during uphill, level, and downhill walking. Dissertation, The Pennsylvania State University. Available from University Microfilms International, UMI No. 9032253.

Buczek, FL, and Cavanagh PR (1990). Stance phase knee and ankle kinematics and kinetics during level and downhill running. *Medicine and Science in Sports and Exercise*, **22**(5): 669-77.

Buczek, FL, Cavanagh PR, Kulakowski BT, and Pradhan P (1990). Slip resistance needs of the mobility disabled during level and grade walking. In: Slips, Stumbles, and

Falls: Pedestrian Footwear and Surfaces, ASTM STP 1103. BE Gray, Ed. Philadelphia, ASTM: 39-54.

Buford, JA, and Smith JL (1990). Adaptive control for backward quadrupedal walking. II. Hindlimb muscle synergies. *Journal of Neurophysiology*, **64**(3): 756-66.

Buford, JA, and Smith JL (1993). Adaptive control for backward quadrupedal walking. III. Stumbling corrective reactions and cutaneous reflex sensitivity. *Journal of Neurophysiology*, **70**(3): 1102-14.

Buford, JA, Zernicke RF, and Smith JL (1990). Adaptive control for backward quadrupedal walking. I. Posture and hindlimb kinematics. *Journal of Neurophysiology*, **64**(3): 745-55.

Carlson-Kuhta, P, Trank TV, and Smith JL (1998). Forms of forward quadrupedal locomotion. II. A comparison of posture, hindlimb kinematics, and motor patterns for upslope and level walking. *Journal of Neurophysiology*, **79**(4): 1687-701.

Chaloupka, EC, Kang J, Mastrangelo MA, and Donnelly MS (1997). Cardiorespiratory and metabolic responses during forward and backward walking. *Journal of Orthopaedic and Sports Physical Therapy*, **25**(5): 302-6.

Cham, R, and Redfern MS (2002). Changes in gait when anticipating slippery floors. *Gait and Posture*, **15**(2): 159-71.

Cipriani, DJ, Armstrong CW, and Gaul S (1995). Backward walking at three levels of treadmill inclination: An electromyographic and kinematic analysis. *Journal of Orthopaedic and Sports Physical Therapy*, **22**(3): 95-102.

Clarkson, E, Cameron S, Osmon P, McGraw C, Smutok M, Stetts D, and Underwood F (1997). Oxygen consumption, heart rate, and rating of perceived exertion in young adult women during backward walking at different speeds. *Journal of Orthopaedic and Sports Physical Therapy*, **25**(2): 113-8.

Crenna, P, and Frigo C (1991). A motor programme for the initiation of forward-oriented movements in humans. *J Physiol*, **437**: 635-53.

Davis, M, Newsam CJ, and Perry J (1995). Electromyograph analysis of the popliteus muscle in level and downhill walking. *Clinical Orthopaedics and Related Research*, **310**: 211-7.

DeVita, P, and Stribling J (1991). Lower extremity joint kinetics and energetics during backward running. *Medicine and Science in Sports and Exercise*, **23**(5): 602-10.

Dutto, DJ, Hoyt DF, Cogger EA, and Wickler SJ (2004). Ground reaction forces in horses trotting up an incline and on the level over a range of speeds. *Journal of Experimental Biology*, **207**(20): 3507-14.

Duysens, J, and Van de Crommert HW (1998). Neural control of locomotion; The central pattern generator from cats to humans. *Gait and Posture*, **7**(2): 131-41.

Earhart, GM, and Bastian AJ (2000). Form switching during human locomotion: traversing wedges in a single step. *Journal of Neurophysiology*, **84**(2): 605-15.

Eilam, D, and Shefer G (1992). Reversal of interleg coupling in backward locomotion implies a prime role of the direction of locomotion. *Journal of Experimental Biology*, **173**(1): 155-63.

Eisner, WD, Bode SD, Nyland J, and Caborn DN (1999). Electromyographic timing analysis of forward and backward cycling. *Medicine and Science in Sports and Exercise*, **31**(3): 449-55.

Enoka, RM (1996). Eccentric contractions require unique activation strategies by the nervous system. *Journal of Applied Physiology*, **81**(6): 2339-46.

Finley, FR, and Cody KA (1970). Locomotive characteristics of urban pedestrians. *Archives of Physical Medicine and Rehabilitation*, **51**(7): 423-6.

Flynn, TW, Connery SM, Smutok MA, Zeballos RJ, and Weisman IM (1994). Comparison of cardiopulmonary responses to forward and backward walking and running. *Medicine and Science in Sports and Exercise*, **26**(1): 89-94.

Gailey, RS, and Clark CR (2004). Physical Therapy. In: Atlas of Amputations and Limb Deficiencies: Surgical, Prosthetic, and Rehabilitation Principles. DG Smith, JW Michael and JH Bowker, Eds. Rosemont, American Academy of Orthopedic Surgeons: 589-620.

Gillis, GB, and Biewener AA (2002). Effects of surface grade on proximal hindlimb muscle strain and activation during rat locomotion. *Journal of Applied Physiology*, **93**(5): 1731-43.

Gottschall, JS, and Kram R (2005). Ground reaction forces during downhill and uphill running. *Journal of Biomechanics*, **38**(3): 445-52.

Grampp, J, Willson J, and Kernozek T (2000). The plantar loading variations to uphill and downhill gradients during treadmill walking. *Foot and Ankle International*, **21**(3): 227-31.

Grasso, R, Bianchi L, and Lacquaniti F (1998). Motor patterns for human gait: backward versus forward locomotion. *Journal of Neurophysiology*, **80**(4): 1868-85.

Gregor, RJ, Cavanagh PR, and LaFortune M (1985). Knee flexor moments during propulsion in cycling--a creative solution to Lombard's Paradox. *Journal of Biomechanics*, **18**(5): 307-16.

Gregor, RJ, Smith DW, and Prilutsky BI (2005). Mechanics of Slope Walking in the Cat: Quantification of Muscle Load, Length Change, and Ankle Extensor EMG Patterns. *J Neurophysiology*, **In Press**.

Gregor, RJ, Smith JL, Smith DW, Oliver A, and Prilutsky BI (2001). Hindlimb Kinetics and Neural Control During Slope Walking in the Cat: Unexpected Findings. *Journal of Applied Biomechanics*, **17**: 277-86.

Grillner, S (1981). Control of Locomotion in bipeds, tetrapods, and fish. In: Handbook of Physiology. The nervous system. Motor Control. Bethesda, Am Physiol Soc. **II**: 1179-236.

Grillner, S (1985). Neurobiological bases of rhythmic motor acts in vertebrates. *Science*, **228**(4696): 143-9.

Higham, TE, and Jayne BC (2004). In vivo muscle activity in the hindlimb of the arboreal lizard, Chamaeleo calyptratus: general patterns and the effects of incline. *Journal of Experimental Biology*, **207**(Pt 2): 249-61.

Hof, AL, Elzinga H, Grimmius W, and Halbertsma JP (2002). Speed dependence of averaged EMG profiles in walking. *Gait and Posture*, **16**(1): 78-86.

Hooper, TL, Dunn DM, Props JE, Bruce BA, Sawyer SF, and Daniel JA (2004). The effects of graded forward and backward walking on heart rate and oxygen consumption. *Journal of Orthopaedic and Sports Physical Therapy*, **34**(2): 65-71.

Hoyt, DF, Wickler SJ, Biewener AA, Cogger EA, and De La Paz KL (2005). In vivo muscle function vs speed. I. Muscle strain in relation to length change of the muscle-tendon unit. *Journal of Experimental Biology*, **208**(6): 1175-90.

Kadaba, MP, Ramakrishnan HK, Wootten ME, Gainey J, Gorton G, and Cochran GV (1989). Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait. *Journal of Orthopaedic Research*, **7**(6): 849-60.

Kautz, SA, Brown DA, Van der Loos HF, and Zajac FE (2002). Mutability of bifunctional thigh muscle activity in pedaling due to contralateral leg force generation. *Journal of Neurophysiology*, **88**(3): 1308-17.

Kawamura, K, Tokuhiro A, and Takechi H (1991). Gait analysis of slope walking: a study on step length, stride width, time factors and deviation in the center of pressure. *Acta Medica Okayama*, **45**(3): 179-84.

Kaya, M, Leonard T, and Herzog W (2003). Coordination of medial gastrocnemius and soleus forces during cat locomotion. *Journal of Experimental Biology*, **206**(20): 3645-55.

Kirtley, C, Whittle MW, and Jefferson RJ (1985). Influence of walking speed on gait parameters. *Journal of Biomedical Engineering*, **7**(4): 282-8.

Kleissen, RF, Buurke JH, Harlaar J, and Zilvold G (1998). Electromyography in the biomechanical analysis of human movement and its clinical application. *Gait and Posture*, **8**(2): 143-58.

Kram, R, Griffin TM, Donelan JM, and Chang YH (1998). Force treadmill for measuring vertical and horizontal ground reaction forces. *Journal of Applied Physiology*, **85**(2): 764-9.

Kramers-de Quervain, IA, Lauffer JM, Kach K, Trentz O, and Stussi E (2001). Functional donor-site morbidity during level and uphill gait after a gastrocnemius or soleus muscle-flap procedure. *Journal of Bone and Joint Surgery*, **83-A**(2): 239-46.

Kuo, AD (2001). The action of two-joint muscles: The legacy of W. P. Lombard. In: Classics in Movement Science. ML Latash and VM Zatsiorsky, Eds, Human Kinetics: 289-315.

Kuster, M, Sakurai S, and Wood GA (1995). Kinematic and kinetic comparison of downhill and level walking. *Clinical Biomechanics (Bristol, Avon)*, **10**(2): 79-84.

Lamb, T, and Yang JF (2000). Could different directions of infant stepping be controlled by the same locomotor central pattern generator? *Journal of Neurophysiology*, **83**(5): 2814-24.

Lange, GW, Hintermeister RA, Schlegel T, Dillman CJ, and Steadman JR (1996). Electromyographic and kinematic analysis of graded treadmill walking and the implications for knee rehabilitation. *Journal of Orthopaedic and Sports Physical Therapy*, **23**(5): 294-301.

Laursen, B, Ekner D, Simonsen EB, Voigt M, and Sjogaard G (2000). Kinetics and energetics during uphill and downhill carrying of different weights. *Applied Ergonomics*, **31**(2): 159-66.

Lay, AN, Hass CJ, and Gregor RJ (2005a). The Effects of Sloped Surfaces on Locomotion: A Kinematic and Kinetic Analysis. *Journal of Biomechanics*, **In Press**.

Lay, AN, Hass CJ, Smith DW, and Gregor RJ (2005b). Characterization of a system for studying human gait during slope walking. *Journal of Applied Biomechanics*, **21**(2): 153-66.

Leroux, A, Fung J, and Barbeau H (1999). Adaptation of the walking pattern to uphill walking in normal and spinal-cord injured subjects. *Experimental Brain Research*, **126**(3): 359-68.

Leroux, A, Fung J, and Barbeau H (2002). Postural adaptation to walking on inclined surfaces: I. Normal strategies. *Gait and Posture*, **15**(1): 64-74.

Lindle, RS, Metter EJ, Lynch NA, Fleg JL, Fozard JL, Tobin J, Roy TA, and Hurley BF (1997). Age and gender comparisons of muscle strength in 654 women and men aged 20-93 yr. *Journal of Applied Physiology*, **83**(5): 1581-7.

McFadyen, BJ, and Winter DA (1988). An integrated biomechanical analysis of normal stair ascent and descent. *Journal of Biomechanics*, **21**(9): 733-44.

McGibbon, CA (2003). Toward a better understanding of gait changes with age and disablement: neuromuscular adaptation. *Exercise and Sport Sciences Reviews*, **31**(2): 102-8.

McVay, EJ, and Redfern MS (1994). Rampway safety: Foot forces as a function of rampway angle. *American Industrial Hygiene Association Journal*, **55**(7): 626-34.

Minetti, AE, and Ardigo LP (2001). The transmission efficiency of backward walking at different gradients. *Pflugers Archiv. European Journal of Physiology*, **442**(4): 542-6.

Minetti, AE, Moia C, Roi GS, Susta D, and Ferretti G (2002). Energy cost of walking and running at extreme uphill and downhill slopes. *Journal of Applied Physiology*, **93**(3): 1039-46.

Mitsui, T, Gregor RJ, and Matsuo A (2001). Electromyographic activity of triceps surae muscles during human treadmill locomotion at various slope intensities. XVIIIth International Society of Biomechanics Congress, Zurich, Switzerland.

Morrison, JB (1969). Function of the knee joint in various activities. *Bio-Medical Engineering*, **4**(12): 573-80.

Myatt, G, Baxter R, Dougherty R, Williams G, Halle J, Stetts D, and Underwood F (1995). The cardiopulmonary cost of backward walking at selected speeds. *Journal of Orthopaedic and Sports Physical Therapy*, **21**(3): 132-8.

Neptune, RR, and Kautz SA (2000). Knee joint loading in forward versus backward pedaling: implications for rehabilitation strategies. *Clinical Biomechanics (Bristol, Avon)*, **15**(7): 528-35.

Neptune, RR, Kautz SA, and Zajac FE (2000). Muscle contributions to specific biomechanical functions do not change in forward versus backward pedaling. *Journal of Biomechanics*, **33**(2): 155-64.

Patla, A (1986). Effects of salking on various inclines on EMG patterns of lower limb muscles in humans. *Human Movement Science*, **5**(4): 345-57.

Perell, KL, Gregor RJ, Buford JA, and Smith JL (1993). Adaptive control for backward quadrupedal walking. IV. Hindlimb kinetics during stance and swing. *Journal of Neurophysiology*, **70**(6): 2226-40.

Pijnappels, M, Bobbert MF, and van Dieen JH (2004). Contribution of the support limb in control of angular momentum after tripping. *Journal of Biomechanics*, **37**(12): 1811-8.

Pijnappels, M, Bobbert MF, and van Dieen JH (2005). How early reactions in the support limb contribute to balance recovery after tripping. *Journal of Biomechanics*, **38**(3): 627-34.

Pivarnik, JM, and Sherman NW (1990). Responses of aerobically fit men and women to uphill/downhill walking and slow jogging. *Medicine and Science in Sports and Exercise*, **22**(1): 127-30.

Polcyn, AF, Lipsitz LA, Kerrigan DC, and Collins JJ (1998). Age-related changes in the initiation of gait: degradation of central mechanisms for momentum generation. *Archives of Physical Medicine and Rehabilitation*, **79**(12): 1582-9.

Powers, CM, Landel R, and Perry J (1996). Timing and intensity of vastus muscle activity during functional activities in subjects with and without patellofemoral pain. *Physical Therapy*, **76**(9): 946-55.

Pratt, CA, Buford JA, and Smith JL (1996). Adaptive control for backward quadrupedal walking V. Mutable activation of bifunctional thigh muscles. *Journal of Neurophysiology*, **75**(2): 832-42.

Prentice, SD, Hasler EN, Groves JJ, and Frank JS (2004). Locomotor adaptations for changes in the slope of the walking surface. *Gait and Posture*, **20**(3): 255-65.

Prilutsky, BI (2000). Coordination of two- and one-joint muscles: functional consequences and implications for motor control. *Motor Control*, **4**(1): 1-44.

Redfern, MS, and DiPasquale J (1997). Biomechanics of descending ramps. *Gait and Posture*, **6**(2): 119-25.

Riener, R, Rabuffetti M, and Frigo C (2002). Stair ascent and descent at different inclinations. *Gait and Posture*, **15**(1): 32-44.

Schwameder, H, Roithner R, Muller E, Niessen W, and Raschner C (1999). Knee joint forces during downhill walking with hiking poles. *Journal of Sports Sciences*, **17**(12): 969-78.

Segal, RL, and Song AW (2005). Non-uniform activity of human calf muscles during an exercise task. *Archives of Physical Medicine and Rehabilitation*, **86**(10): 2013-17.

Shiavi, R (1985). Electromyographic patterns in adult locomotion: a comprehensive review. *Journal of Rehabilitation Research and Development*, **22**(3): 85-98.

Simpson, KJ, Jiang P, Shewokis PA, Odum S, and Reeves KT (1993). Kinematic and plantar pressure adjustments to downhill gradients during gait. *Gait and Posture*, **1**(3): 172-9.

Smith, JL, and Carlson-Kuhta P (1995). Unexpected motor patterns for hindlimb muscles during slope walking in the cat. *Journal of Neurophysiology*, **74**(5): 2211-5.

Smith, JL, Carlson-Kuhta P, and Trank TV (1998). Forms of forward quadrupedal locomotion. III. A comparison of posture, hindlimb kinematics, and motor patterns for downslope and level walking. *Journal of Neurophysiology*, **79**(4): 1702-16.

Stansfield, BW, and Nicol AC (2002). Hip joint contact forces in normal subjects and subjects with total hip prostheses: walking and stair and ramp negotiation. *Clinical Biomechanics (Bristol, Avon)*, **17**(2): 130-9.

Sun, J, Walters M, Svensson N, and Lloyd D (1996). The influence of surface slope on human gait characteristics: a study of urban pedestrians walking on an inclined surface. *Ergonomics*, **39**(4): 677-92.

Terblanche, E, Page C, Kroff J, and Venter RE (2004). The effect of backward locomotion training on the body composition and cardiorespiratory fitness of young women. *Int J Sports Medicine*, **25**: 1-6.

Thorstensson, A (1986). How is the normal locomotor program modified to produce backward walking? *Experimental Brain Research*, **61**(3): 664-8.

Threlkeld, AJ, Horn TS, Wojtowicz GM, Rooney JG, and Shapiro R (1989). Kinematics, ground reaction force, and muscle balance produced by backward running. *Journal of Orthopaedic and Sports Physical Therapy*, **11**(2): 56-63.

Ting, LH, Kautz SA, Brown DA, and Zajac FE (1999). Phase reversal of biomechanical functions and muscle activity in backward pedaling. *Journal of Neurophysiology*, **81**(2): 544-51.

Ting, LH, Raasch CC, Brown DA, Kautz SA, and Zajac FE (1998). Sensorimotor state of the contralateral leg affects ipsilateral muscle coordination of pedaling. *Journal of Neurophysiology*, **80**(3): 1341-51.

Tokuhiro, A, Nagashima H, and Takechi H (1985). Electromyographic kinesiology of lower extremity muscles during slope walking. *Archives of Physical Medicine and Rehabilitation*, **66**(9): 610-3.

Trank, TV, and Smith JL (1996). Adaptive control for backward quadrupedal walking VI. metatarsophalangeal joint dynamics and motor patterns of digit muscles. *Journal of Neurophysiology*, **75**(2): 678-9.

Van de Crommert, HW, Mulder T, and Duysens J (1998). Neural control of locomotion: sensory control of the central pattern generator and its relation to treadmill training. *Gait and Posture*, **7**(3): 251-63.

Van Deursen, RW, Flynn TW, McCrory JL, and Morag E (1998). Does a single control mechanism exist for both forward and backward walking? *Gait and Posture*, **7**(3): 214-24.

Vandervoort, AA (2002). Aging of the human neuromuscular system. *Muscle and Nerve*, **25**(1): 17-25.

Vaughan, CL, Davis BL, and O'Connor JC (1999). Dynamics of Human Gait. Cape Town, Kiboho Publishers.

Vilensky, J, Gankiewicz E, and Gehlsen G (1987). A kinematic comparison of backward and forward walking. *Journal of Human Movement Studies*, **13**(1): 29-50.

Wall, JC, Nottrodt JW, and Charteris J (1981). The effects of uphill and downhill walking on pelvic oscillations in the transverse plane. *Ergonomics*, **24**(10): 807-16.

Whittle, M (2001). Gait Analysis: An Introduction. Oxford, Butterworth Heinemann.

Wickler, SJ, Hoyt DF, Biewener AA, Cogger EA, and De La Paz KL (2005). In vivo muscle function vs speed. II. Muscle function trotting up an incline. *Journal of Experimental Biology*, **208**(Pt 6): 1191-200.

Winter, DA (1980). Overall principle of lower limb support during stance phase of gait. *Journal of Biomechanics*, **13**(11): 923-7.

Winter, DA (1987). The Biomechanics and Motor Control of Human Gait. Waterloo, University of Waterloo Press.

Winter, DA (1990). Biomechanics and Motor Control of Human Movement. New York, John Wiley & Sons, Inc.

Winter, DA, and Eng P (1995). Kinetics: our window into the goals and strategies of the central nervous system. *Behavioural Brain Research*, **67**(2): 111-20.

Winter, DA, Pluck N, and Yang JF (1989). Backward walking: A simple reversal of forward walking? *Journal of Motor Behavior*, **21**(3): 291-305.

Winter, DA, and Yack HJ (1987). EMG profiles during normal human walking: stride-to-stride and inter-subject variability. *Electroencephalography and Clinical Neurophysiology*, **67**(5): 402-11.

Woltring, HJ (1986). A FORTRAN package for generalized, cross-validatory spline smoothing and differentiation. *Advances in Engineering Software*, **8**(2): 104-13.

Yang, JF, Lam T, Pang MY, Lamont E, Musselman K, and Seinen E (2004). Infant stepping: a window to the behaviour of the human pattern generator for walking. *Canadian Journal of Physiology and Pharmacology*, **82**(8-9): 662-74.

Zernicke, RF, and Smith JL (1996). Biomechanical Insights into Neural Control of Movement. In: Handbook of Physiology: Integration of Motor, Circulatory, Respiratory, and Metabolic Control During Exercise.