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**Kinetic Analysis of Forwards, Step-by-Step and Backwards Stair Descent**

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# Kinetic Analysis of Forwards, Step-by-Step and Backwards Stair Descent

Tyler Cluff

This thesis is submitted to the  
Faculty of Graduate and Postdoctoral Studies  
in partial fulfillment of the requirements  
for the degree M.Sc. in Human Kinetics

Faculty of Health Sciences  
School of Human Kinetics  
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## ABSTRACT

The purpose of this research was to evaluate biomechanical differences in alternate stair descent strategies to determine the mechanisms by which these strategies modulate task mechanics. Seventeen healthy subjects (9 males, 8 females) performed ten trials in four experimental conditions; forwards (FD), backwards (BD), step-by-step lead (SBSL) and trail (SBST) limbs.

Results showed that peak ankle plantiflexor powers and work in BD and SBSL were similar in magnitude ( $p>0.05$ ) and reduced relative to FD at initial contact ( $p<0.001$ ). Similarly, the midstance peak plantiflexor moments, powers and work in FD and SBST were greater than both BD and SBSL ( $p<0.001$ ). The plantiflexor moment and work in BD were larger than those at the SBSL limb ( $p<0.01$ ), whereas power magnitudes were not significantly different ( $p>0.05$ ). Lastly, peak moments, powers and work at push-off were greater in FD compared to all other conditions ( $p<0.001$ ) and further, were larger in SBST relative to BD and SBSL ( $p<0.001$ ). Peak moments, powers and work were similar in BD and SBSL conditions ( $p>0.05$ ).

Peak knee moments, powers and work were reduced during the eccentric midstance burst. Knee moments, powers and work were similar in FD and SBST ( $p>0.05$ ), but were greater than BD and SBSL ( $p<0.001$ ). Peak moments, powers and work were greater in BD compared to the SBSL limb ( $p<0.001$ ). Thus, the step-by-step descent strategy is more appropriate for individuals with unilateral knee problems, whereas backwards descent is more suitable for bilateral quadriceps weakness.

Peak hip powers and work were larger in BD relative to FD ( $p<0.001$ ) during initial contact. Similarly, the extensor moment in BD was significantly larger than the

flexor moment demonstrated in FD. No consistent stair or condition effects were observed for the concentric hip flexor burst at push-off in FD, SBSL and SBST conditions.

Finally, this study considered foot clearance in stair descent. Clearance was greater in BD relative to FD, SBSL and SBST in the mid staircase region, but was similar at step 1, which represented the transition from standing to stair descent. No foot contacts, trips or stumbles were recorded, demonstrating the likelihood of contacting a stair edge in all patterns was nonexistent for these healthy participants.

The present results have implications to improve functionality in stair descent for individuals that find the task difficult or painful. Future studies will likely consider the strategies employed during the initiation of stair descent by elderly and clinical populations to gain insight on the strategies used to minimize mechanical burden on both the joints and overlying musculature. Lastly, research should consider the effects of handrail use on alternate stair descent strategy mechanics, such as backwards and step-by-step descent.

To Michelle,

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*T.C.*

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## TABLE OF CONTENTS

<b>ABSTRACT</b> .....	<b>i</b>
<b>ACKNOWLEDGEMENTS</b> .....	<b>iv</b>
<b>TABLE OF CONTENTS</b> .....	<b>v</b>
<b>LIST OF TABLES</b> .....	<b>vii</b>
<b>LIST OF FIGURES</b> .....	<b>viii</b>
<b>GLOSSARY</b> .....	<b>xi</b>
<b>REVIEW OF LITERATURE</b> .....	<b>3</b>
KINETICS OF STAIR NEGOTIATION .....	3
MECHANICAL WORK IN BIOMECHANICAL ANALYSES .....	12
INVERSE DYNAMICS AND THE LINK-SEGMENT MODEL.....	15
MINIMAL FOOT CLEARANCE AND THE RISK OF TRIPPING IN STAIR DESCENT .....	19
RATIONALE.....	21
PURPOSE .....	22
RESEARCH HYPOTHESES.....	22
<i>Kinetics</i> .....	22
<i>Minimal Foot Clearance</i> .....	23
<b>METHODS</b> .....	<b>24</b>
PARTICIPANTS.....	24
INSTRUMENTATION AND PROTOCOL .....	25
DATA ANALYSIS .....	27

STATISTICAL ANALYSES .....	30
<b>RESULTS .....</b>	<b>32</b>
CORRELATION ANALYSES.....	64
MINIMAL FOOT CLEARANCE.....	66
KINETICS .....	71
PROGRESSION TO STEADY-STATE STAIR DESCENT .....	79
TRANSITION EFFECT IN STAIR DESCENT .....	85
<b>SUMMARY AND CONCLUSIONS.....</b>	<b>90</b>
SUMMARY.....	90
CONCLUSIONS.....	92
<b>REFERENCES.....</b>	<b>95</b>
<b>APPENDIX A .....</b>	<b>105</b>
<b>INFORMED CONSENT DOCUMENTATION .....</b>	<b>105</b>
<b>APPENDIX B .....</b>	<b>114</b>
<b>SCREENING QUESTIONNAIRE.....</b>	<b>114</b>

## LIST OF TABLES

Table 3-1: Subject characteristics.....	24
Table 4-1: Summary of statistical results for the ankle peak moment, power and work done by the plantiflexor moment at initial-contact (A2).....	44
Table 4-2: Summary of statistical results for ankle peak plantiflexor moment and concentric power at push-off (A3) .....	44
Table 4-3: Summary of statistical results for ankle plantiflexor concentric power and total work at push-off (A3) .....	45
Table 4-4: Summary of statistical results for the peak plantiflexor moment, power and work done by the ankle plantiflexors in midstance (A2) .....	47
Table 4-5: Summary of statistical results for peak knee moments, powers and work performed about the flexion/extension axis ( $x$ -axis) in mid- through terminal stance.....	52
Table 4-6: Summary of statistical results for peak hip moments, powers and work performed about the flexion/extension axis ( $x$ -axis) in early stance (through terminal stance in backwards descent).....	55
Table 4-7: Summary of statistical results for peak hip moments, powers and work performed by the flexor moment during the generative burst, H4, in terminal stance.....	59
Table 4-8: Summary of statistical results for maximum progression velocity computed tangential to staircase inclination.....	61
Table 4-9: Summary of statistical results for minimal foot clearance in forwards and backwards stair descent.....	62

## LIST OF FIGURES

Figure 3-1: Laboratory set-up for motion capture.....	26
Figure 3-2: The uOttawa marker set used for motion capture.....	28
Figure 3-3: Schematic representing the three-dimensional distance between a point and a line.....	29
Figure 4-1: Grand ensemble averaged kinetics of forwards descent.....	33
Figure 4-2: Grand ensemble averaged, body mass normalized kinetics of backwards descent.....	35
Figure 4-3: Grand ensemble averaged, body-mass normalized kinetics of the lead limb in step-by-step stair descent.....	36
Figure 4-4: Grand ensemble averaged, body-mass normalized kinetics of the trail limb in step-by-step stair descent.....	38
Figure 4-5: Grand ensemble averaged ( $n = 17$ ), body-mass normalized moments of force (top row) and moment powers (bottom row) for the ankle (left column), knee (middle column) and hip (right column) joints at stair 2, across all conditions. FD: Blue; BD: Black; SBSL: Green; SBST: Red.....	41
Figure 4-6: Grand ensemble averaged ( $n = 17$ ), body mass normalized moments of force (top row) and moment powers (bottom row) for the ankle (left column), knee (middle column) and hip (right column) joints at stair 3, across all conditions. FD: Blue; BD: Black; SBSL: Green; SBST: Red.....	42

- Figure 4-7: Grand ensemble averaged ( $n = 17$ ), body mass normalized moments of force (top row) and powers (bottom row) for the ankle (left column), knee (middle column) and hip (right column) joints at stair 4, across all conditions. FD: Blue; BD: Black; SBSL: Green; SBST: Red.....43
- Figure 4-8: Grand ensemble averaged ( $n = 17$ ), body mass normalized moments of force (top row) and powers (bottom row) for the ankle (left column), knee (middle column) and hip (right column) joints at stair 5, across all conditions. FD: Blue; BD: Black; SBSL: Green; SBST: Red.....44
- Figure 4-9: Ensemble-averaged, body mass normalized peak moments, powers and work done at weight-acceptance, A1.....50
- Figure 4-10: Ensemble-averaged, body mass normalized moments, powers and work during the push-off burst, A3.....51
- Figure 4-11: Ensemble-averaged, body mass normalized moments, powers and work done by the ankle plantiflexors during the midstance burst, A2.....52
- Figure 4-12: Ensemble-averaged, body mass normalized moments, powers and work done at the dissipative stance phase knee burst, K3.....57
- Figure 4-13: Ensemble-averaged, body mass normalized moments, powers and work done at the hip over four consecutive gait cycles for forwards and backwards gait strategies.....58
- Figure 4-14: Ensemble-averaged, body mass normalized moments, powers and work done at the hip for concentric burst, H3 at push-off.....59

- Figure 4-15: Maximum velocity parallel to progression over four consecutive gait cycles in FD, BD and SBSD patterns.....62
- Figure 4-16. Minimal foot clearance over five consecutive stairs for forwards, backwards and the lead and trail limbs in step-by-step descent. Error bars represent the 95% confidence interval.....63
- Figure 4-17: Correlation plots between stance phase ankle (A1), knee (K3) and hip (H2) powers and progression velocity in forwards FD.....68
- Figure 4-18: Correlation plots for stance phase ankle, knee and hip powers and progression velocity in BD.....69

**GLOSSARY**

A1	Eccentric plantiflexor moment burst at initial-contact.
A2	Eccentric burst of the plantiflexor moment in midstance.
A3	Concentric burst of the ankle plantiflexor moment at push-off.
K1	Eccentric knee extensor burst at initial-contact.
K2	Concentric knee extensor burst at weight-acceptance.
K3	Eccentric knee extensor burst in mid through terminal stance (controlled lowering).
K4	Eccentric knee flexor burst in terminal swing.
H1	Concentric hip burst at initial-contact.
H2	Eccentric burst of the stance phase hip moment (flexor: forwards descent; extensor: backwards descent).
H3	Concentric burst of the hip flexor moment at push-off.
H4	Concentric burst of the swing phase hip flexor moment.
H5	Eccentric burst of the hip extensor moment at terminal swing/early stance (backwards descent).
BD	Backward (stair) descent
FD	Forward (stair) descent
SBSD	Step-by-step (stair) descent
SBSL	Step-by-step (stair descent) lead leg
SBST	Step-by-step (stair descent) trail leg
CL	Controlled lowering
FCL	Foot clearance

FCN	Forward continuance
FP	Foot-placement
IC	Initial contact
PL	Pull-through
PU	Push-off
WA	Weight acceptance
<i>df</i>	Degrees of freedom (statistical)
MEE	Mechanical energy expenditure
GRF	Ground reaction force

## INTRODUCTION

The ability to ambulate on stairs is important since stairs are often encountered in daily life. According to the Canadian Institute for Health Information (2003-2006), the number of injuries associated with falls on or from stairs or steps requires further attention. In 2002 through 2004, the most commonly specified fall that resulted in injury was a fall on or from stairs or steps; which accounted for approximately 20% of fall injuries (22%,  $n = 574$ ; CIHI, 2003; 20%:  $n = 573$ , CIHI, 2004; 20%:  $n = 631$ ; CIHI, 2005). When fall type was grouped by age, falls on or from stairs was the most commonly reported fall for those aged 35-64, from 2001 through 2005. This was also the case for those above 65 in 2001 through 2003 (CIHI, 2003; CIHI, 2004), but was surpassed by falls on the same level by slipping, tripping and stumbling in 2004-2005 (CIHI, 2006). According to CANSIM Table 102-0540 (StatCan, 2007), for the years 2000 to 2004, between 12 and 15% of fatal falls were falls on or from stairs. Consequently, this problem warrants an investigation of stair negotiation to prevent the injuries, losses of mobility, disabilities and fatalities that result from stair falls.

The biomechanics of stair negotiation are well documented in the literature (Joseph & Watson, 1967; Shinno, 1971; Lyons, Perry, Gronley, Barnes & Antonelli, 1983; McFadyen & Winter, 1988; Reiner, Rabuffetti & Frigo, 2002; Stacoff, Diezi, Luder, Stussi & Kramers-de Quervain, 2003). These studies performed kinematic, kinetic and/or electromyographic (EMG) analyses to quantify joint angles and ranges of motion, moments of force and moment powers and muscle activity patterns at the ankle, knee and hip joints that characterize stair ambulation.

This thesis focuses on lower extremity mechanics during stair descent. Accordingly, the literature review will be limited to data specific to the moments of force and moment powers at the ankle, knee and hip joints in stair descent. The literature will be examined to demonstrate that, while considerable data exists concerning the mechanics of stair negotiation, there are certain implications for descent that have not been adequately addressed. For example, while much is known about the peak moments and powers in stair descent, mechanical work has yet to be examined. Analyzing peaks contributes to our understanding of the instant in the gait cycle where the underlying musculature must generate the greatest force (in the case of moments) or energy to be generated or dissipated (in the case of moment powers). Unfortunately, these analyses offer no explanation about the overall mechanical demand imposed on the musculature; the sum total of energy generated or dissipated during locomotion. This understanding can only be rendered through an analysis of mechanical work, the time integral of the moment power curve. As a result, there is a discontinuity in the understanding of stair descent kinetics, that when addressed, will lead to a more comprehensive understanding of this locomotion. The review will also consider literature related to foot clearance in stair descent. The prevailing motive for studying foot clearance in descent has been to understand factors that lead to tripping (Hamel, Okita, Higginson & Cavanagh, 2005).

## REVIEW OF LITERATURE

### Kinetics of Stair Negotiation

McFadyen and Winter (1988) produced seminal research on the biomechanics of stair negotiation. A sagittal plane analysis was performed that examined the contribution of the flexors/extensors of the ankle, knee and hip joints during stair ascent and descent. The gait cycle was partitioned into five phases. The stance phase of ascent was divided into weight acceptance (WA), pull-up (PU) and forward continuance (FCN). Swing was divided into foot clearance (FCL) and positioning for foot placement (FP). Descent was divided into weight acceptance (WA), forward continuance (FCN) and controlled lowering (CL) for stance phase, whereas swing was divided into pull-through (PL) and foot placement (FP).

The main vertical displacement of the body in stair ascent occurred from the beginning of single limb support to mid-stance, when the contralateral limb was in swing. This corresponded to PU, which was attributed to an eccentric burst of knee extensor activity that controlled knee flexion (K1) and a concurrent concentric peak in the ankle plantiflexor moment (A2). During this phase, body weight was supported by the stance limb. Thus, the knee extensors and ankle plantiflexors were responsible for maintaining the support moment, preventing collapse and providing vertical lift to the body. The FCN phase corresponded to the greatest forward displacement. A second peak in ankle plantiflexor power occurred during this phase where the plantiflexors generated energy (A3) to translate the body over the contralateral support limb and lift the body during push-off. During FCL, the authors reported a positive power burst of the hip (H3) and knee flexors (K2), which served to flex the leg at both the hip and knee, thereby

providing clearance between the foot and stair to swing the limb through to the next step. Knee flexor activity then became extensor, performing negative work to control flexion and position the lower limb for foot placement (K3). Late swing was characterized by an eccentric bout of knee flexor activity (K4) that limited extension prior to initial-contact. Final placement of the foot was controlled by eccentric activity of the hip flexors (H4) and ankle dorsiflexors (A1).

In stair descent, the researchers found WA began with initial-contact in mid-stride. Energy absorption at the ankle and knee (A1 and K1) characterized this phase. Thus, the ankle plantiflexors and knee extensors maintained support during weight acceptance. Following this absorption, there was energy generation at the stance knee, which corresponded to K2 and contributed to the support moment and leg extension. During this phase the body progressed both horizontally and upward, initiating the FCN phase. Following FCN, the body entered the downward displacement phase of descent; the CL phase. During CL, there was energy absorption at both the ankle and knee represented by power bursts A2 and K3, which together controlled forward and downward progression of the body. At the end of CL (mid to late-stance), there was a concentric power burst where the hip flexors (H1) and ankle plantiflexors (A3) performed positive work that contributed to the limb's energy for swing. This small amount of push-off energy, compared to level walking, is attributable to reduced stride length in stair descent. Further, the minimal push-off energy contributes only to clearance between the foot and stair and not to providing sufficient energy to overcome gravity, as in stair ascent. Simultaneously, the knee extensors performed negative work during K3 that controlled flexion. In early swing, the researchers observed a positive power burst at the hip (H2).

The authors speculated that hip musculature contributed little to the work necessary to lower the body, since hip moments and powers were variable both within and between subjects (McFadyen & Winter, 1988). During mid-swing all three joints flexed, preparing the body for contact with the stairs and WA. The authors referred to this phase as FP.

Reiner, Rabuffetti and Frigo (2002) documented the lower extremity mechanics of stair negotiation. The focus of this research was to understand the influence of staircase inclination on joint moments and powers. Sagittal plane mechanics were reported for the ankle, knee and hip joints at staircase inclinations of 24°, 30° and 42°. The researchers reported no significant inclination effect for joint moments or powers in swing. However, the stance phase kinetics showed marked inclination dependencies for both ascent and descent. The maximum moment values at the knee and ankle increased as a function of inclination for both ascent and descent. The hip joint produced a flexion moment during the stance phase of descent, whereas an extension moment was produced in ascent. At the knee joint, moments were similar in early stance. However, there were differences late in stance; the authors documented a second peak moment that further extended the knee, whereas during ascent the net knee moment became a flexor moment of small magnitude. The magnitude of the ankle plantiflexor moment increased with inclination at initial-contact in both ascent and descent.

The authors discovered that joint powers were more dependent on inclination than the moments (Reiner *et al.*, 2002). Similar to McFadyen and Winter (1988), Reiner and colleagues argued that the primary task of the lower extremity musculature was to transfer muscle energy to the potential energy of the body during stair ascent. On the other hand, the descent task requires the muscles of the lower limb dissipate gravitational

(potential) energy via eccentric contractions. For both stair ascent and descent, the absolute magnitudes of the knee and ankle joint powers increased as a function of inclination. In stair descent, the greatest increase was observed for ankle dynamics at the loading response phase (weight acceptance), which occurred immediately following initial-contact. Thus, the primary task of the ankle plantiflexors during this phase was to dissipate energy. For stair ascent, the greatest increase occurred at the hip joint, where the magnitude of hip extensor power paralleled increased inclination.

In stair descent, the lower extremities were activated in a sequence aimed at shared energy absorption. The ankle and knee demonstrated peaks in absorption at initial-contact. Thus, the ankle and knee extensor musculature were primarily responsible for dissipating energy and preventing collapse while the stance limb supported body weight. The authors reported that increased inclination resulted in a greater tendency to plantiflex the ankle for foot placement. Accordingly, greater moments and powers at the ankle were associated with initial-contact in descent. As a result, the plantiflexors absorbed potential energy and kept the ankle plantiflexed since reduced tread did not permit heel contact. The knee demonstrated peak absorption late in stance. Thus, the knee extensor moment acted eccentrically to control knee flexion and support the body, while the contralateral limb was in swing.

Peak knee moments and powers were significantly larger in descent than level walking, reaching magnitudes 3.0 and 3.8 times greater, respectively. The knee extensor moments and powers were also significantly greater in descent relative to ascent. Other authors have documented similar results (McFadyen & Winter, 1988; Kowalk, Duncan & Vaughan, 1993). Descent imposes considerable demand on the extensor muscles of the

lower limb, which must generate great magnitudes of eccentric force to dissipate gravitational energy and thereby perform negative work.

Several studies have examined stair negotiation in clinical populations. Hooper and colleagues (2002) examined gait adaptations of participants with chronic posterior knee instability. Whereas the instability group demonstrated reduced knee extensor moments and powers in both ascent and descent, the difference in the extensor moment failed to reach significance in descent. The authors concluded that posterior knee instability patients adopt gait patterns that reduce strain on the posterior stabilizers during stair descent. Using the Flandry-Hughston Clinic questionnaire, the authors observed a significant relationship between self-assessed knee function and the peak extensor moment within the patient group: individuals with greater laxity of the posterior cruciate ligament tended toward greater reductions in the stance phase peak extensor moment.

Research also considered the effects of anterior cruciate ligament (ACL) deficiencies on knee mechanics in stair negotiation. Berchuck, Andriacchi, Bach and Reider (1990) discovered reduced or absent net external flexion moments in level gait, signifying a “quadriceps avoidance” strategy when the knee was flexed less than 40°. However, the ACL-deficient group demonstrated external flexion moments in ascent and descent and consequently did not avoid contracting the quadriceps muscles in either condition. These moments were resisted by internal extension moments at greater angles of knee flexion (62.4 and 66.0°). The authors argued that greater knee flexion caused the hamstrings to stabilize the tibia more effectively, which thereby prevented anterior tibial displacement at the knee and moderates anteroposterior shear forces stabilized by the ACL. In support of this hypothesis, Arms, Pope, Johnson, Fischer, Arvidsson and

Erikkson (1984) demonstrated that when knee flexion angles exceed  $60^\circ$ , contraction of the quadriceps decreased strain on the ACL. Thus, it appeared these individuals were capable of normal stair ascent and descent by altering typical task mechanics.

Alternatively, reduced peak external knee flexor moments were observed in ACL deficient patients ascending stairs, where no significant differences were observed in peak flexion angles (Thambyah, Thiagarajan, & Goh Cho Yong, 2002). The authors discovered reduced ground reaction forces at initial-contact, which they argued reduced the moment arm of the ground reaction force vector to the knee centre, thereby reducing external flexion moments (Thambyah *et al.*, 2002). Thambyah and colleagues speculated that ACL-deficient subjects controlled strain on the ACL by some mechanism of energy absorption and less forceful foot-to-ground contact. The magnitude of the knee flexion angles reported by Berchuck *et al.* (1990) were attributed to greater stair inclination used in their study, rather than a direct result of ACL deficiency (Thambyah *et al.*, 2002). Regardless of the discrepancy; these studies demonstrated that ACL deficient individuals modify the typical mechanics of stair gait.

Researchers examined the implications of patellofemoral pain (PFP) on knee mechanics in stair negotiation. Salsich, Heino Brechter and Powers (2001) hypothesized that PFP subjects would demonstrate compensatory stair gait patterns compared to healthy participants to reduce knee extensor moments in both ascent and descent, since quadriceps force decreases the patellofemoral joint space. In essence, contraction of the knee extensors increases the patellofemoral joint reaction force. In response to this compensation, the authors speculated that secondary compensation would occur at either the ankle or hip so the overall support moment was not compromised. In confirmation,

the stance-phase knee extensor moments were reduced in both ascent and descent. While the PFP group demonstrated trends towards cadence reductions in both ascent and descent, the trend was significant in stair descent only. As a result, the authors argued that PFP subjects modified their gait pattern to minimize symptom aggravation, a primary compensatory strategy indicative of “quadriceps avoidance”. However, differences in the hip, ankle, or overall support moments were not documented. Thus, the authors concluded that secondary compensation was not exclusive to the hip or ankle joints in PFP subjects, which suggested a distributed compensation between the two joints. By maintaining the support moment and not placing exceedingly high demands at either the ankle or hip, the PFP group modified gait to ascend and descend stairs safely.

Subsequent research compared the peak extensor moments, patellofemoral joint reaction force (PFJRF) and patellofemoral joint stress (PFJS) between healthy and PFP subjects (Heino Brechter & Powers, 2002). The authors hypothesized that PFP subjects would demonstrate increased PFJS in ascent and descent, thereby explaining PFP. The researchers discovered that group differences in PFJS were not significant, although there were significant differences in peak extensor moments, cadence and PFJRF in stair ascent. The authors suggested that PFP subjects adopt a “quadriceps avoidance” strategy to reduce the magnitude of forces acting across the patellofemoral joint and minimize pain. There were statistically insignificant trends toward reduced knee extensor moments and PFJRF during stair descent, despite the significant reduction in cadence demonstrated by the PFP group. The knee extensor moments and PFJRF were larger in descent compared to ascent, posing particular difficulty for these individuals. Moreover, the PFJS-time integral was significantly larger in stair descent as compared to ascent,

indicating that overall joint pressure was larger in descent. This finding was consistent with clinical complaints expressed by the PFP subjects: these individuals experienced more pain and expressed more difficulty with stair descent relative to ascent. In terms of the peak PFJS, no significant differences existed between the PFP and control groups, for either ascent or descent. As a result, the experimental hypothesis was invalidated. The authors concluded that PFP subjects develop compensatory strategies to minimize PFJS.

Subsequent research revealed that retropatellar pain was reduced when PFP subjects descended stairs wearing a brace compared to a non-brace condition (Powers, Ward, Chen, Chan & Terk, 2004). In this research, patellofemoral joint kinetics were computed from a biomechanical model with input variables that included knee joint angle, magnitude of the knee extensor moment and patellofemoral joint contact area (PFJCA), determined from MRI. Whereas knee kinematics were not influenced by the bracing condition, the peak knee extensor moment, PFJRF and PFJCA were significantly reduced in the non-braced condition. As such, bracing did not have a significant influence on PFJS, since forces acting across the patellofemoral joint were significantly increased in the braced condition. Reductions in the peak knee extensor moment suggest that PFP subjects compensated to reduce the forces acting across the patellofemoral joint to minimize pain and keep PFJS at a tolerable level (Powers *et al.*, 2004). From the previously outlined research, it is evident that stair negotiation is difficult for clinical populations to the extent that compensatory strategies are often used to maintain functionality and minimize pain.

The previously outlined studies revealed two important considerations regarding stair negotiation mechanics. First, in comparison to ascent, few studies considered the

mechanics of descent. Secondly, studies that examined both ascent and descent mechanics revealed that descent is more demanding for the musculature of the lower extremity, particularly at the knee. The moments and powers necessary for ambulation are inherently larger in stair descent. These demands exacerbate the difficulty encountered by elderly and patient populations. As a result, these populations may encounter difficulty to the extent that they develop alternate gait patterns as compensatory mechanisms.

One such alternate pattern is the step-by-step gait pattern, whereby the individual ascends or descends stairs, one stair at a time, placing both feet on each stair (Musselman, 2003; Reid, Lynn, Musselman & Costigan, 2007). These studies established that step-by-step gait significantly lowered the sagittal plane contact forces, moments and powers of the lead knee in descent and the trail knee in ascent, as compared to the contralateral limb. As such, the authors labelled the lead limb in descent and the trail limb in ascent as the 'resting limb' since sagittal plane contact forces, moments and powers were reduced relative to the contralateral limb (Musselman, 2003; Reid *et al.*, 2007). However, there were no significant differences in the frontal plane forces, moments and powers between the step-over-step and modified step-by-step patterns. As such, step-by-step stair gait decreased the forces, moments and moment powers in the plane of progression, but did not influence the musculature responsible for maintaining balance and support.

An alternative to the step-by-step pattern was offered by Beaulieu (2004). Beaulieu (2004; 2007) proposed backwards descent because it significantly reduced eccentric knee extensor powers relative to forward descent. This remained the case when progression velocity in forwards descent was controlled to that demonstrated in

backwards descent. Thus, the noted difference is attributable to the backwards descent pattern and not velocity artefact. However, Beaulieu's study was limited in that it performed only a two-dimensional inverse dynamics analysis, which may induce errors in joint moment and power computations when the segmental sagittal planes are not aligned with that of the laboratory (Winter, 2005). Further, mechanical work performed during backwards descent is not known. The following section outlines the rationale for a mechanical work analysis of stair descent.

### **Mechanical Work in Biomechanical Analyses**

Mechanical work represents the change in energy of a body (Robertson, Caldwell, Hamill, Kamen & Whittlesey, 2004). There are two methods of determining the mechanical work performed by joint moments during locomotion. The first cited method combined external ground reaction forces and the kinematics of the lower extremity to calculate joint forces and moments of force using inverse dynamics equations (Elftman, 1939). The joint moments were used to determine moment powers and subsequently, mechanical work. This was referred to as the "absolute power" method (Purkiss & Robertson, 2003). Conversely, Winter (1979) and later, Pierrynowski, Winter and Norman (1980) derived a method to compute the mechanical work of the body from changes in the mechanical energies (potential and kinetic) of segments. With this method, the absolute changes in potential and kinetic energies of body segments are summed to determine total body mechanical work, while accounting for energy transfers both within segments and between adjacent segments. This technique was referred to as the "absolute work" method (Purkiss & Robertson, 2003).

Since the inception of the absolute work method, authors have disputed its ability to perform accurate estimates of the mechanical energy expenditure (MEE) of movement (Aleshinsky, 1986a; 1986b, 1986c; Caldwell & Forrester, 1992). Aleshinsky demonstrated that the absolute work method conveys certain assumptions that may disrupt the validity of MEE estimates. First, the absolute work method considers all energy sources intercompensated and recuperative; energy lost by the system is assumed to be recovered through inter- and intrasegmental energy transfers and elastic potentiation. These circumstances rarely occur in human motion. In fact, Aleshinsky argued that these assumptions lend themselves to conclusions that do not make physical sense (Aleshinsky, 1986a; 1986b; 1986c). In summary, Aleshinsky argued the absolute work method produces severe underestimates of MEE and is “not even useful for the determination of muscular energy expenditure limits” (1986d, p.309).

Caldwell and Forrester (1992) demonstrated that the absolute power method more effectively determined the MEE and mechanical efficiency of walking and running. In particular, the authors discovered that errors in absolute work magnitudes occurred when muscle powers offset one another. Therefore, the authors concluded the absolute power method provided better estimates of the mechanical cost of movement. Purkiss and Robertson (2003) also contested the validity of absolute work estimates when they showed work estimates demonstrated greater variability relative to the absolute power method. The researchers interpreted this variability as an indicator of compromised reliability. The absolute power method was argued to be superior because it was more sensitive to gait inefficiencies, considered both positive and negative work and explained

energy generation, absorption and transfer in detail. These findings reiterate the analytical superiority of the absolute power method (Purkiss & Robertson, 2003).

Despite its superiority, the absolute power method is not without limitation. The absolute power method assumes all energy sources are compensated; energy lost by a source cannot be returned to the system by intercompensation (energy transfers between segments) nor by the recuperation of stored muscle elastic energy (Aleshinsky, 1986a). Aleshinsky (1986a) admitted that intercompensation and recuperation do occur in human movement via the contraction of two-joint muscles and muscles shortening immediately after lengthening. In fact, 22-26% of the work performed in an isolated knee bend experiment was attributed to the recovery of elastic energy (Thys *et al.*, 1972; Assmussen & Bonde-Peterson, 1974). Thus, the absolute power method may overestimate MEE. Nevertheless, Wells (1988) demonstrated that these assumptions were acceptable and the associated errors were minimal in walking analyses. A final source of error is the inability to distinguish between mechanical costs of positive and negative work, where previous analyses demonstrated negative work to be of less metabolic cost to the musculature (Abbott & Bigland, 1953; Nagle *et al.*, 1965). Despite shortcomings, the absolute power method is superior for estimating the mechanical costs of human motion.

The importance of a mechanical work analysis is to analyze the effect of alternate gait patterns on lower extremity mechanics – does one method reduce the total energy dissipated by the net joint moment in stair descent? According to Ingen Schenau and Cavanagh (1990), the only sources capable of supplying power to or absorbing it from the body's segments are the net joint forces and moments. The only active sources are the joint moments; joint forces only redistribute energy between segments. Thus, it follows

from this argument that both energy generation and absorption are achieved predominantly through skeletal muscle and to a lesser extent, connective tissue. Similar arguments were brought forth by Chapman and Caldwell (1983), Winter (1983), Williams and Cavanagh (1983) and Robertson and Winter (1980). Winter (1983) argued that an understanding of the net muscle functions, in terms of the positive and negative work performed can only be accomplished by analyzing the patterns of mechanical power generation and absorption at each joint. Therefore, mechanical work analysis quantifies the overall energy generated or dissipated by the net moment and will permit comparison between gait conditions as to the overall mechanical cost of each strategy.

### **Inverse Dynamics and the Link-Segment Model**

The process by which muscle moments and reaction forces are computed from kinematic and kinetic data was first outlined by Elftman (1939) and later refined by Bresler and Frankel (1950). The process is referred to as link-segment modeling. In link segment modeling, the segments of the lower extremity are considered rigid bodies (geometric solids) defined by segment masses, joint centers and moments of inertia.

An inverse dynamics solution combines kinematic and anthropometric data, as well as external ground reaction forces to compute joint reaction forces and internal muscle forces (Robertson *et al.*, 2004; Winter, 2005). This method relies on Newton's laws of motion to partition the resultant force acting across a joint into known and unknown forces. Since there are several unknown forces but only one equation to solve for the translational motion of a body, the problem is indeterminant. However, the unknown forces can be combined into a single net force that can be solved. The vector equation for the translational dynamics of the ankle joint is as follows:

$$\vec{F}_{\text{Ankle}} = m\vec{a}_{\text{CM}} - m\vec{g} - \vec{F}_{\text{GRF}} \quad (1)$$

where  $\vec{F}_{\text{Ankle}}$  is the vector describing the ankle joint force,  $m$  is the segment mass,  $\vec{a}_{\text{CM}}$  is the linear acceleration vector of the center of mass,  $\vec{g}$  is the gravitational force vector, and  $\vec{F}_{\text{GRF}}$  is the GRF vector. The next step of the inverse dynamics procedure is to calculate the rotational dynamics, the ankle joint moment from Euler's equation:

$$\vec{M}_{\text{Ankle}} = \vec{I}\alpha - (\vec{r}_1 \times \vec{F}_{\text{Ankle}}) - (\vec{r}_2 \times \vec{F}_{\text{GRF}}) - \vec{\tau} \quad (2)$$

where  $\vec{M}_{\text{Ankle}}$  is the ankle joint moment;  $\vec{I}$  is the moment of inertia tensor,  $\alpha$  is the angular acceleration matrix;  $(\vec{r}_1 \times \vec{F}_{\text{Ankle}})$  is the moment that results from the ankle joint reaction force;  $(\vec{r}_2 \times \vec{F}_{\text{GRF}})$  is the moment resulting from the ground reaction force; and  $\vec{\tau}$  is the ground reaction moment vector. To expand the equation, it is necessary to compute the vector that describes the moment about the x-, y- and z-axes of the proximal joint center:

$$\vec{M}_{\text{JRF}} = \vec{r}_1 \times \vec{F}_{\text{Ankle}} \quad (3)$$

where  $\vec{M}_{\text{JRF}}$  is the moment arising from the JRF,  $\vec{r}_1$  is the vector describing the xyz distance between the center of mass and the proximal joint center and  $\vec{F}_{\text{Ankle}}$  is the vector of the xyz components of the proximal joint reaction force. The moment vector that results from the GRF,  $\vec{M}_{\text{GRF}}$ , is the cross-product of the distance between the COM and COP vectors and the GRF vector:

$$\vec{M}_{\text{GRF}} = \vec{r}_2 \times \vec{F}_{\text{GRF}} \quad (4)$$

where  $\vec{r}_2$  is the vector of the xyz distance between the COM and the COP and  $\vec{F}_{\text{GRF}}$  is the vector of the xyz components of the GRF.

The moment at the ankle joint, taken about the center of mass of the foot is then calculated from:

$$M_{\text{Ankle}_X} = I_{XX}\alpha_X + (I_{ZZ} - I_{YY})\omega_{ZZ}\omega_{YY} - M_{\text{GRF}_X} - M_{\text{jrf}_X} - \tau_X \quad (5)$$

$$M_{\text{Ankle}_Y} = I_{YY}\alpha_Y + (I_{XX} - I_{ZZ})\omega_{XX}\omega_{ZZ} - M_{\text{GRF}_Y} - M_{\text{jrf}_Y} - \tau_Y \quad (6)$$

$$M_{\text{Ankle}_Z} = I_{ZZ}\alpha_Z + (I_{YY} - I_{XX})\omega_{YY}\omega_{XX} - M_{\text{GRF}_Z} - M_{\text{jrf}_Z} - \tau_Z, \quad (7)$$

where  $I_{XX}$ ,  $I_{YY}$ ,  $I_{ZZ}$  are the principal moments of inertia about the x, y and z axes;  $\alpha_X$ ,  $\alpha_Y$  and  $\alpha_Z$  are the x, y and z components of ankle angular acceleration;  $\omega_{XX}$ ,  $\omega_{YY}$  and  $\omega_{ZZ}$  are the components of the segmental angular velocity,  $M_{\text{GRF}_X}$ ,  $M_{\text{GRF}_Y}$ ,  $M_{\text{GRF}_Z}$  are the moments arising from the GRF;  $M_{\text{jrf}_X}$ ,  $M_{\text{jrf}_Y}$  and  $M_{\text{jrf}_Z}$  are the components of the moment resulting from the joint reaction force and  $\tau_X$ ,  $\tau_Y$  and  $\tau_Z$  are the ground reaction force moments. The joint reaction forces and moments are first computed at the ankle, followed by the knee and finally, the hip joint.

Inverse dynamics, as outlined here, computes net measures and therefore, it is not possible to determine the force or moment within a specific structure. As such, the net agonist/antagonist muscle effort or single equivalent moment is computed (Robertson *et al.*, 2004; Winter, 2005). Similarly, in this method friction is considered negligible. Thus, inverse dynamics determines the minimal moment necessary to produce movement (Robertson *et al.*, 2004). According to Winter (2005), friction would reduce the effective muscle moment. In this case, muscle contractile elements would be required to compensate by producing moments higher than those analyzed at the tendon. However, this assumption remains valid for the present study since friction is negligible through the midrange of motion to be encountered in this study (Winter, 2005).

Joint power analysis is a method used by biomechanists to examine the flow of energy across a joint that results from the net joint forces and moments of force (Ingen Schenau & Cavanagh, 1990; Robertson *et al.*, 2004). Computing the moment power requires the relative joint angular velocity:

$$P_M = M_j(\omega_p - \omega_d), \quad (8)$$

where  $P_M$  is the moment power;  $M_j$  is the magnitude of the moment of force acting at the joint; and  $(\omega_p - \omega_d)$  is the relative angular velocity of the two segments that form the joint. Thus, the direction of the moment and angular velocity determine whether energy is generated or dissipated at a particular joint. The total body mechanical work  $W_{tb}$ , which is the change in energy of the body over the duration of movement, is computed from the numerical time integral of equation (8):

$$W_{tb} = \sum_{n=1}^N \sum_{j=1}^J |M_{n,j} \omega_{n,j}| \Delta t, \quad (9)$$

where  $J$  is the number of joints in the body and  $N$  is the number of time intervals for the trial. The external work performed by the body for a given movement is:

$$W_E = \sum_{n=1}^N \sum_{j=1}^J M_{n,j} \omega_{n,j} \Delta t. \quad (10)$$

The internal work  $W_I$  is the work performed by contractile tissue during movement and is computed as:

$$W_I = W_{tb} - W_E, \quad (11)$$

where  $W_{tb}$  is the total body mechanical work and  $W_E$  is the external work performed by or on the body during movement.

## **Minimal Foot Clearance and the Risk of Tripping in Stair Descent**

Debilitating stair falls motivate researchers to understand the factors that predispose both young and elderly populations to falling. Of late, foot clearance has received considerable attention as such a factor. Foot clearance is a descriptive kinematic variable that quantifies the distance between the stair edge and the sole of the shoe during the swing phase of gait. Therefore, foot clearance is a method to measure the likelihood of the foot contacting the stair edge and subsequently, measures the risk of tripping during stair accommodation (Cavanagh & Higginson, 2002; Startzell, 1998; Hamel, Okita, Higginson & Cavanagh, 2005). The majority of these studies focused on clearance in stair descent since three-quarters of falls in stair ambulation are sustained in descent (Cohen, Templer & Archea, 1985; Svanstrom, 1974; Cohen, 2000).

To date, the literature has demonstrated that foot clearance varies between steps within a staircase. Simoneau and colleagues (1991) discovered that elderly women demonstrated greater clearance at the onset of descent but decreased clearance through the mid-stair region. More recently, Cavanagh and Higginson (2002) further supported this phenomenon when they discovered that clearance progressively decreased among healthy young subjects during descent. Therefore, it appears that clearance is refined to progressively smaller values as the individual descends the staircase and accommodates stair dimensions. In a subsequent research, ambient luminance was decreased to 3 lux, to

represent the lower limit of civil twilight. At this intensity, the individuals could see the stairs but had difficulties locating the edge (Hamel *et al.*, 2005). When the intensity of staircase illumination was manipulated by occlusion, the young subjects maintained higher clearance over all stairs compared to the normal vision (300 lux) condition (Cavanagh & Higginson, 2002). However, Hamel *et al.* (2005) demonstrated that elderly individuals failed to increase clearance when ambient lighting was reduced and showed greater variability in clearance than young participants. Not surprisingly, there were three falls among the elderly participants, whereas no falls were suffered by the younger group. The authors concluded that failure to increase clearance with decreased luminance and reduced, more variable clearance likely accounts for the greater incidence of stair falls in the elderly (Hamel *et al.*, 2005).

## Rationale

The review of literature demonstrated that lower extremity moments and powers are larger for descent than both level walking and stair ascent, particularly at the knee. As a result, the injured and elderly often develop compensatory or alternate gait patterns to deal with the demand. Although previous studies examined the three-dimensional knee biomechanics of the step-by-step pattern (Musselman, 2003; Reid *et al.*, 2007) and the sagittal plane joint mechanics of backwards descent (Beaulieu, 2004), we know little about these alternate stair descent strategies. In fact, we know little about how alternate strategies influence foot clearance during the swing phase of stair descent. As revealed by Hamel and colleagues, this consideration is likely of great importance for the elderly who demonstrate reduced and more variable clearance in stair descent. In short, the following summary presents the rationale for this study:

1. The ability to descend stairs is an important aspect of functionality, since we often encounter stairs in daily life.
2. Stair descent has larger moments and powers than both level walking and ascending stairs. This is especially true at the knee.
3. For both the elderly and certain clinical populations, the demands imposed by the descent task can be problematic. These populations often develop compensatory or alternate stair descent patterns to minimize the incidence of pain or reduce the demand placed on the lower extremity musculature.
4. Literature on these alternate strategies is scarce and a direct comparison of these alternate strategies is lacking. As a result, several important questions remain unanswered: Does one strategy result in greater foot clearance, thereby reducing

the likelihood of the swing leg contacting the stair edge while negotiating the stairs? Is one strategy less demanding for the musculature of the lower extremity?

## **Purpose**

The purpose of this research was to understand biomechanical differences among traditional step-over-step forwards stair descent and two alternate descent patterns; forwards step-by-step and backwards descent.

## **Research Objectives**

The objectives of this study are two-fold:

1. Determine whether one of the strategies is less demanding at the joint level, based on inverse dynamics and mechanical work analyses.
2. Determine whether foot clearance is greater in one of the strategies, thereby reducing the likelihood of inadvertently contacting the stair edge during the swing phase of stair descent.

## **Research Hypotheses**

### **Kinetics**

1. The peak knee extensor moment will be smaller for the step-by-step lead limb as compared to the step-by-step trail limb, forwards and backwards descent.
2. There will be no difference in the peak knee extensor moment between forwards step-over-step and backwards descent.
3. The total mechanical work will be lower for the step-by-step lead limb and backwards descent, as compared to the step-by-step trail limb and forwards step-over-step patterns.

4. The total mechanical work will not be significantly different between the alternate descent strategies; forwards step-by-step and backwards descent.

### **Minimal Foot Clearance**

1. Foot clearance will be larger in backwards as compared to all other descent strategies.
2. Minimal foot clearance will be greatest at the onset of forwards step-over-step descent and decrease progressively through to the midstair region.
3. Minimal foot clearance will not vary between steps in either the lead or trail limbs of step-by-step descent.

## METHODS

### Participants

Healthy males ( $n = 9$ ) and females ( $n = 8$ ) voluntarily participated in this study. Table 1 summarizes participant characteristics. All participants were free of neurological and musculoskeletal conditions of the lower extremity, as identified by an extensive screening questionnaire. Exclusionary criteria were: musculoskeletal conditions of the lower limbs including surgical intervention, visual disorders not corrected with lenses, vestibular disorders or epilepsy. Provided inclusionary criteria were met, candidates were considered eligible for the study. Minimum sample size was determined by *a priori* power computations based on effect sizes determined in small pilot sample ( $n = 5$ ). Power computations were performed with the G\*Power software (Erdfelder, Faul & Buchner, 1996) so that the likelihood of committing a type II error,  $\beta < 0.20$  (Cohen, 2005).

Table 3-1. Subject characteristics.

Subject group	$n$	Age (years)	Height <sup>†</sup> (cm)	Mass (kg)
Male	9	22.9 (2.7)	175 (5.5)	78.3 (7.0)
Female	9	22.7 (1.7)	168 (5.1)	58.9 (3.8)
Combined	18	22.8 (2.3)	173 (6.3)	70.3 (11.4)

The data are presented as mean (S.D.).

<sup>†</sup>Significant difference ( $p < 0.05$ ).

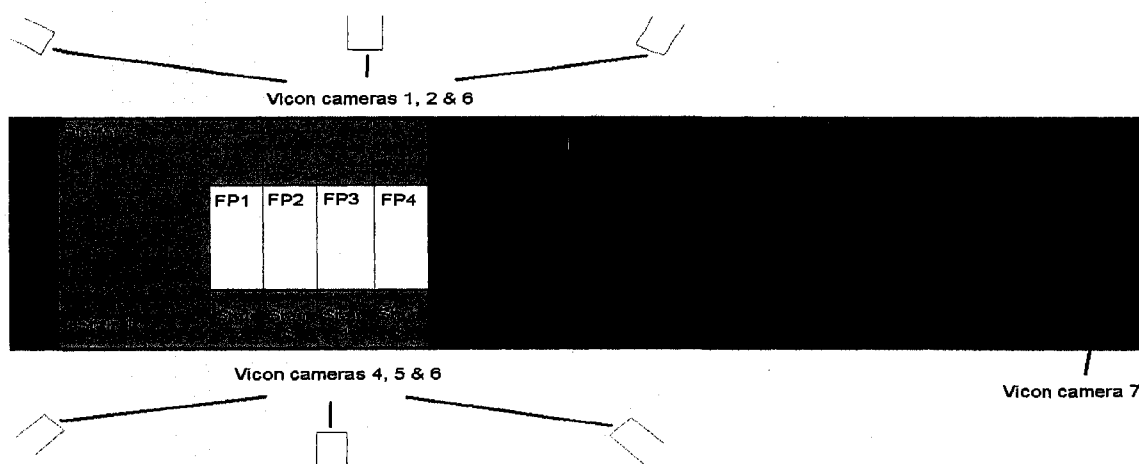
## Instrumentation and Protocol

Participants read and signed informed consent forms approved by the University of Ottawa Research Ethics Board. Following a brief familiarization period, ten trials were performed in step-over-step forwards (FD) and backwards descent (BD). Trials were initiated with the subject's preferred limb and performed at a self-selected pace, where preferred limb was defined as the limb with which the participant would habitually begin descending stairs. In addition, twenty step-by-step descent (SBSD) trials were performed- 10 to analyze lead limb kinetics (SBSL) and 10 for trail limb kinetics (SBST). Since participants were healthy control subjects, SBSD kinetics represent those derived from the right limb. Condition presentation was counterbalanced across subjects. Participant height and mass were recorded at the outset for inverse dynamics analysis.

The staircase was instrumented with four force plates embedded in steps 2 through 5 (step 2 & 5: 9286A, Kistler Instrumente AG, Winterthur, Switzerland; step 3 & 4: OR6-7-1000 & OR6-7-2000, AMTI, Watertown, MA, USA). The stair dimensions were 20 cm rise and 30 cm run (Ontario Building Code, Section 9.8.3, 2006). Force platform data were sampled at 200 Hz. Marker displacement data were collected at 200 Hz with a seven-camera Vicon MX-13 (Lake Forest, CA, USA) motion analysis system and processed offline with the Vicon Workstation software. Figure 3-1 outlines the experimental apparatus. The setup allowed for simultaneous capture of kinetic and kinematic data for two complete stride cycles of both the lead and trail limbs. In SBSD descent, the apparatus permitted capture of kinetic and kinematic data for four consecutive cycles of the lead and trail limbs. Analog data were low-pass filtered with a 4<sup>th</sup>-order, dual-pass Butterworth digital filter (Robertson and Dowling, 2003) ( $f_c=10$  Hz

force; 6 Hz for displacement data). No hand railings were necessary because only normal subjects were investigated who showed no risk of falling.

Fourteen-millimetre reflective spherical markers were positioned bilaterally on the participants' limbs at the hallux (TOE), 1<sup>st</sup> and 5<sup>th</sup> metatarsophalangeal joints (MT1 and MT5), calcaneus (CAL), medial and lateral malleoli (MAN, LAN), shank (TIB), medial and lateral femoral condyles (KNE, MKN), thigh (THI), greater trochanter (HIP), and the anterior (ASI) and posterior superior iliac spines (PSI), according to the uOttawa marker set represented in Figure 3-2. The uOttawa marker set was derived from the Vicon Plug-in-gait set, but was modified by incorporating hip and medial ankle and knee markers.



**Figure 3-1.** Laboratory set-up for motion capture. The staircase was instrumented with four force platforms to capture kinetic data. Vicon cameras captured kinematic data during stair descent.

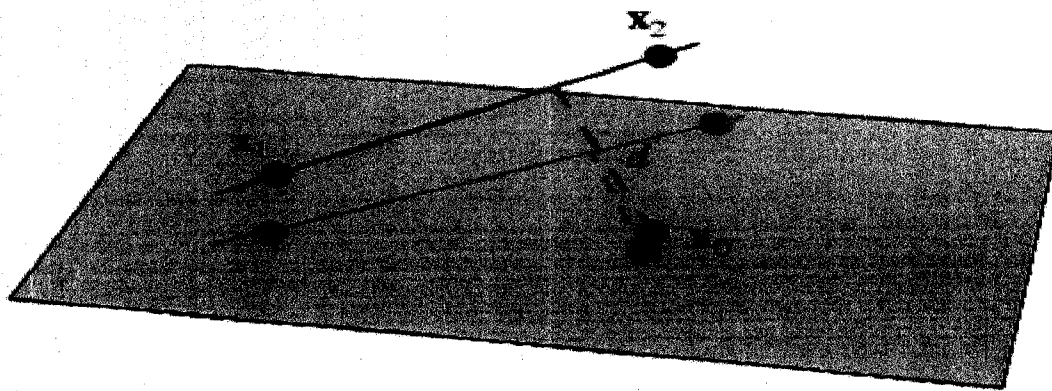
The marker set permits six degrees of freedom and the requisite three independent, non-collinear surface markers necessary to track motion in 3-dimensional space (Winter, 2004). Further, under the uOttawa marker set, markers are affixed at well-defined bony landmarks of the lower extremity, which improves both the within- and between-day reproducibility of joint kinematics and kinetics in stair negotiation (Yu, Queen & Schrod, 2003).

## Data Analysis

The joint and segment kinematics and force platform signals were incorporated into a standard link-segment model to compute joint angular velocities (rad/s), moments of force (N.m) and moment powers (W). Inverse dynamics computations were performed for the ankle, knee and hip joints of both lower extremities using Visual3D version 3.90.21. Joint angular velocities, moments and moment powers were projected onto local coordinate systems embedded at proximal joint centres of each segment. The local coordinate systems used a  $z$ -axis (*axial rotation*) that projected from the proximal joint centre to the distal joint centre of the segment. The  $y$ -axis (*ab/adduction*) was formed by projecting an axis orthogonal to the plane formed by the markers that defined proximal and distal joint centres (i.e., for the shank, this plane is defined by the medial and lateral condylar and malleoli markers). Subsequently, the  $x$ -axis (*flexion/extension*) was the cross-product of  $z$ - and  $y$ -axes, producing a right-handed orthogonal axis system.



Where  $d$  was the minimal three-dimensional distance between a marker (foot markers: TOE, MT1, MT5 or CAL) and the line defining the stair edge;  $\underline{x}_2$  and  $\underline{x}_1$  were the coordinates of markers that defined the stair edge corners; and  $\underline{x}_0$  represented the coordinates of any one of the foot markers. Minimal clearance was the absolute minimum clearance of foot markers. Figure 3-3 outlines the process for computing the three-dimensional distance between a point and a line.



**Figure 3-3.** Schematic representing the three-dimensional distance between a point and a line. The variables  $x_1$  and  $x_2$  correspond to three-dimensional position vectors that specify the locations of the corners of the stair edge;  $x_0$  represents the time-varying three-dimensional coordinates of any one of the foot markers. Therefore,  $d$  represents the three-dimensional distance between the foot markers and the stair edge, which corresponds to foot clearance.

This study followed a repeated-measure, within-subjects quantitative design. Data processing and inverse dynamics analysis were performed with Visual3D software version 3.90.21 (C-Motion Inc., Rockville, MD, USA). Moment and power profiles were normalized to percent stride cycle, where one stride was defined as toe-off to toe-off of the same foot. Therefore, ensemble-averaged figures (Figures 4-1 to 4-8) present the swing followed by stance phase dynamics. Moment and power magnitudes were body-mass normalized and reported as N.m/kg and W/kg. Body-mass normalized mechanical work (J/kg) was computed by the absolute power method with BioProc2 software. Peak moments, powers and mechanical work at the ankle, knee and hip joints were extracted with the BioProc2 software (Robertson, 2008) for each trial and averaged to create unique within-subject means.

Lastly, maximum velocity parallel to the inclination of the staircase was computed for each gait cycle and condition. Maximum velocity was computed iteratively from velocity of the total body COM as:

$$V_t = \sqrt{V_x^2 + V_z^2} , \quad (13)$$

parallel to the inclination of the staircase and defined by the plane of progression.

## Statistical Analyses

Separate 4×4 factorial ANOVAs with repeated-measures were performed for ensemble-averaged moments, powers and work done about the flexion/extension joint axis with stair (within-subjects factor with 4 steps) and condition (within-subjects factor with 4 levels: FD, BD, SBSL and SBST) factors. For all statistical tests, the probability of Type I error,  $\alpha$ , was set to 0.05. The data for males ( $n = 9$ ) and females ( $n = 8$ ) were pooled

since mixed – model ANOVA analyses with a gender between-subjects factor showed no significant gender effects, or gender  $\times$  condition, gender  $\times$  stair, or gender  $\times$  condition  $\times$  stair interactions ( $p > 0.05$ ). Accordingly, only the results of the repeated-measures ANOVA analyses will be presented here.

Repeated-measures ANOVA assumes equality of the differences in variance between treatment levels, or sphericity. This study employed Mauchly's test to determine whether the sphericity assumption was obeyed. Where Mauchly's test showed the sphericity assumption was violated and the Greenhouse-Geisser  $df$  correction factor,  $\epsilon < 0.7$ , the multivariate Pillai's Trace statistic was employed. Pillai's Trace is recommended when sphericity is seriously violated (i.e.,  $\epsilon < 0.7$  and  $p < 0.05$  for Mauchly's test) and the number of subjects,  $n > a + 10$ ; where  $a$  is the number of levels of the within-subjects factor (Maxwell & Delaney, 1990). In short, the multivariate test employs a specific error term for contrasts with 1  $df$  so that each contrast is associated only with its specific error term, rather than the pooled error variance term used in omnibus ANOVA with repeated-measures (O'Brien & Kaiser, 1985). Unless indicated, results present summary statistics from univariate 4 $\times$ 4 factorial ANOVA. Post-hoc analyses were performed using Tukey HSD tests with Bonferroni corrections.

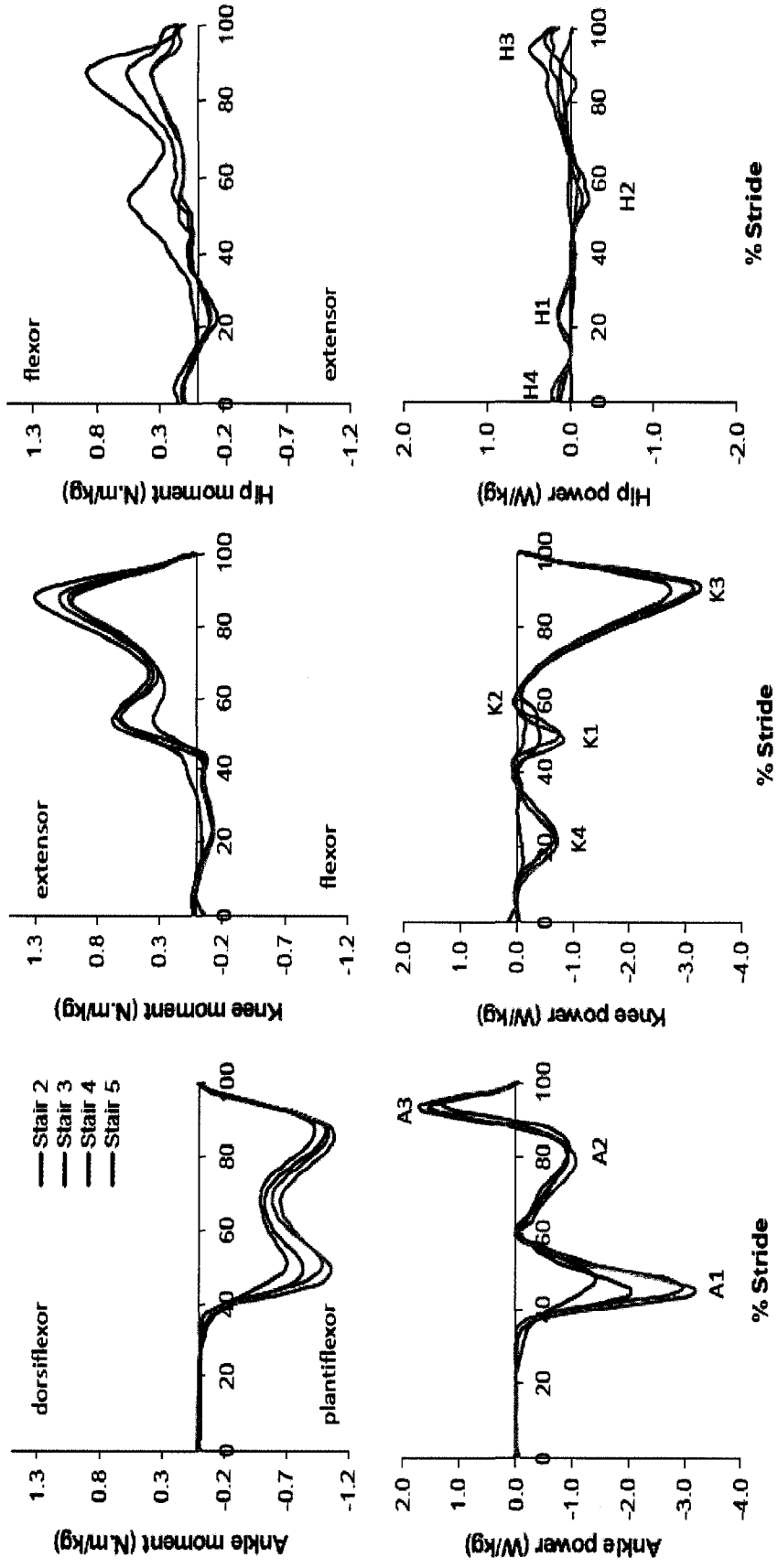
Maximum velocity was the maximum velocity in the plane of progression for a particular gait cycle. Correlation analyses were performed to determine whether a deterministic relationship existed between maximum progression velocity and stance phase power magnitudes using the Pearson Product Moment Correlation statistic (Pearson's  $r$ ). The coefficient of determination,  $r^2$ , was then computed to determine percentage variance in peak power explained by maximum progression velocity.

## RESULTS

Figure 4-1 shows the ensemble-averaged body mass normalized moments and powers in forwards stair descent. The abscissas are normalized to 100% stride cycle, where one stride represents toe-off to toe-off of the same limb. As such, moment and power profiles at stairs 1 and 3 represent those of the lead limb (limb used to initiate descent), while those at stairs 2 and 4 represent moments and powers of the contralateral limb. The ankle moment was plantiflexor for the entire stance phase, with definitive peaks at initial-contact and midstance. The peak plantiflexor moment at weight-acceptance was associated with eccentric power burst A1, when the plantiflexor moment limited dorsiflexion and controlled lowering of the body at foot-contact. In midstance, the plantiflexor moment acted eccentrically (burst A2) to control dorsiflexion while the contralateral limb was in swing. Finally, the plantiflexor moment performed positive work in terminal stance that contributed to push-off energy (burst A3).

A net knee flexor moment in mid swing was associated with the burst K4, when the knee flexors dissipated energy to control extension prior to foot-contact. At weight-acceptance, the knee extensor moment acted eccentrically to support the body and prevent collapse. The maximum knee extensor moment occurred in terminal stance, when the contralateral limb was in swing. At this point, the lower extremity was in single-support; body weight was supported solely by the stance limb and the knee extensors controlled lowering of the body to the next step (K3).

In stance, the net hip flexor moment alternated between bouts of negative (H2) and positive work (H3). In early to mid stance (50-65% stride), the hip flexors acted



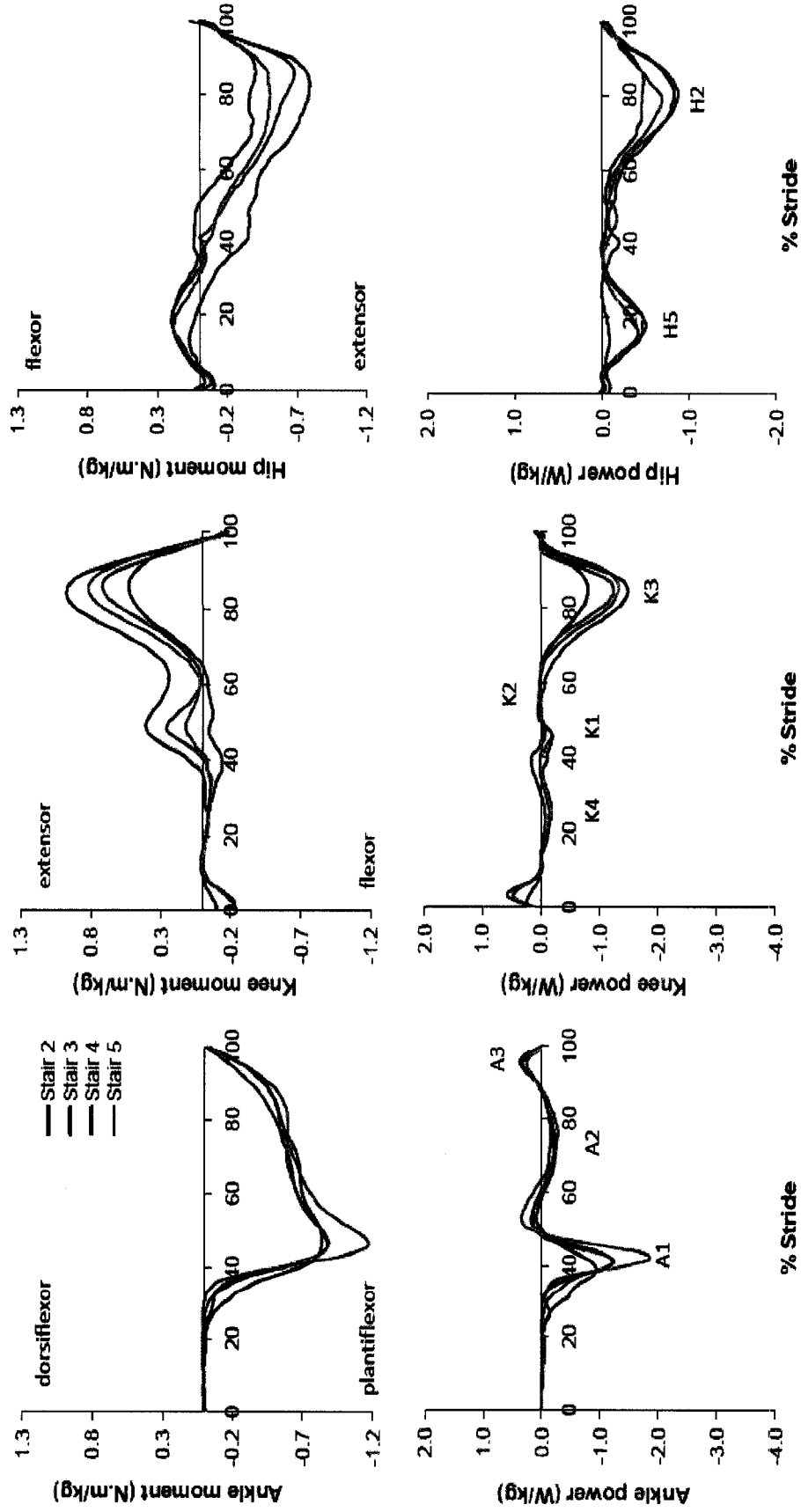
**Figure 4-1.** Grand ensemble averaged ( $n = 17$ ) kinetics of forwards descent. Body mass normalized moments of force (top row) and powers (bottom row) are presented for the ankle (left column), knee (middle column) and hip (right column) joints in forwards stair descent.

eccentrically to limit extension while the contralateral limb was in swing (H2).

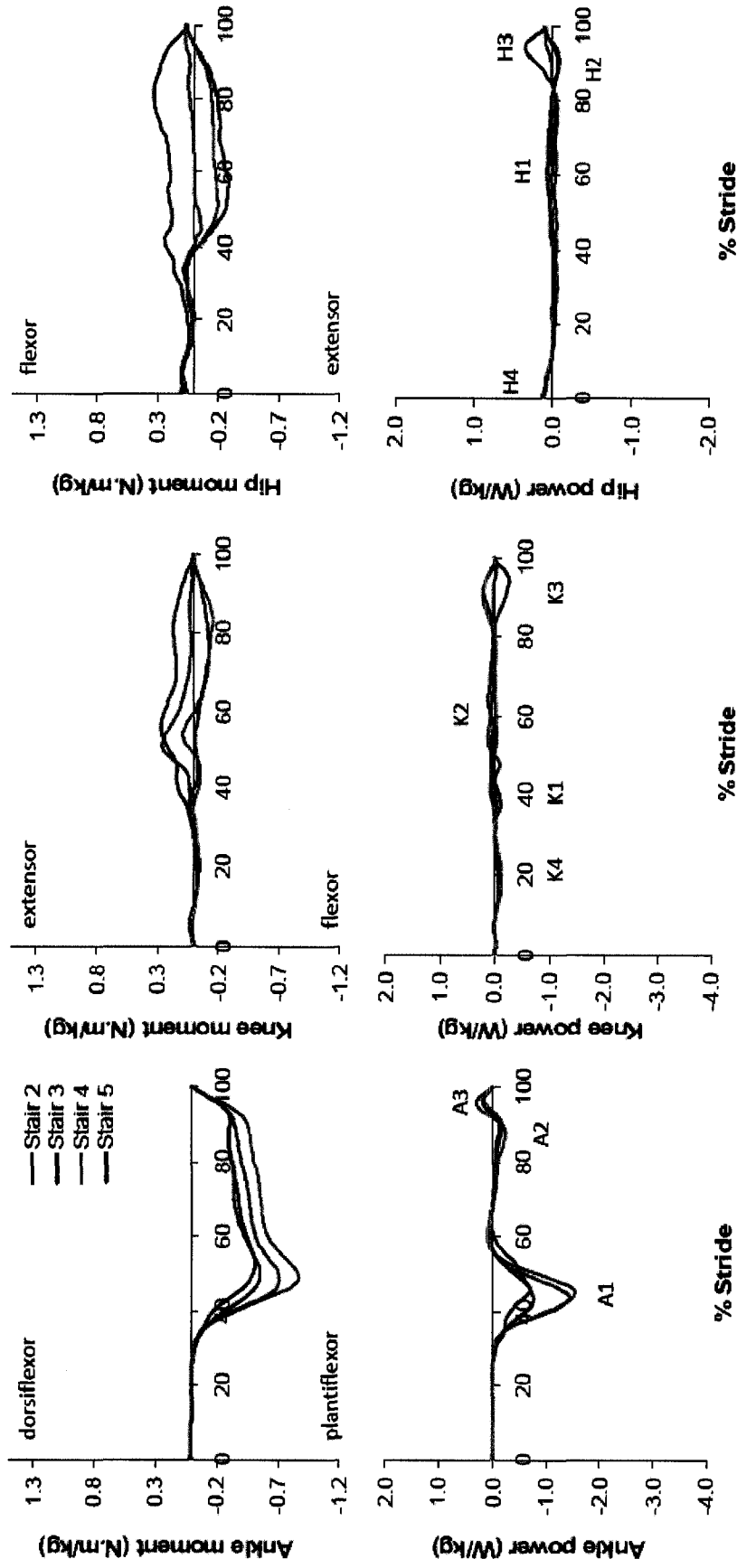
Conversely, in mid through terminal stance (65-100% stride) the hip flexors did positive work to pull the leg up in preparation for swing (H3).

Figure 4-2 plots the ensemble-averaged body mass normalized moments and powers for backwards descent. At the ankle joint, the net stance phase moment was plantiflexor. Similar to forwards descent, the plantiflexor moment did negative work at initial contact (A1) to absorb gravitational energy and in mid stance (A2) to control lowering to the next step. In terminal stance there was a generative burst, where the plantiflexor moment contributed to the limb's energy for swing. At the knee joint, the net moment in backwards descent alternated between a flexor and extensor moment that did positive and negative work. At initial contact, the net moment was extensor and did negative work. In terminal stance, there was an eccentric power burst of the knee extensor moment that controlled lowering of the body while the contralateral limb swung through to the next step. Swing phase dynamics in BD were markedly different relative to FD. In early swing, the knee flexors performed positive work to flex the knee and consequently, provided clearance between the foot and stair. Backwards descent hip kinetics were different than FD. In stance, a net hip extensor moment did negative work to prevent collapse. In mid swing through early stance, there was a flexor moment that performed negative work to limit the rate of extension prior to foot-contact (H5).

Figure 4-3 plots the ensemble averaged kinetics for the lead limb in step-by-step stair descent (SBSL) over four consecutive gait cycles. Ankle kinetics followed a pattern similar to those observed for the FD and BD conditions.



**Figure 4-2.** Grand ensemble averaged ( $n = 17$ ), body mass normalized kinetics of backwards descent. Moments of force (top row) and powers (bottom row) are plotted for the ankle (left column), knee (middle column) and hip (right column) joints in backwards stair descent.

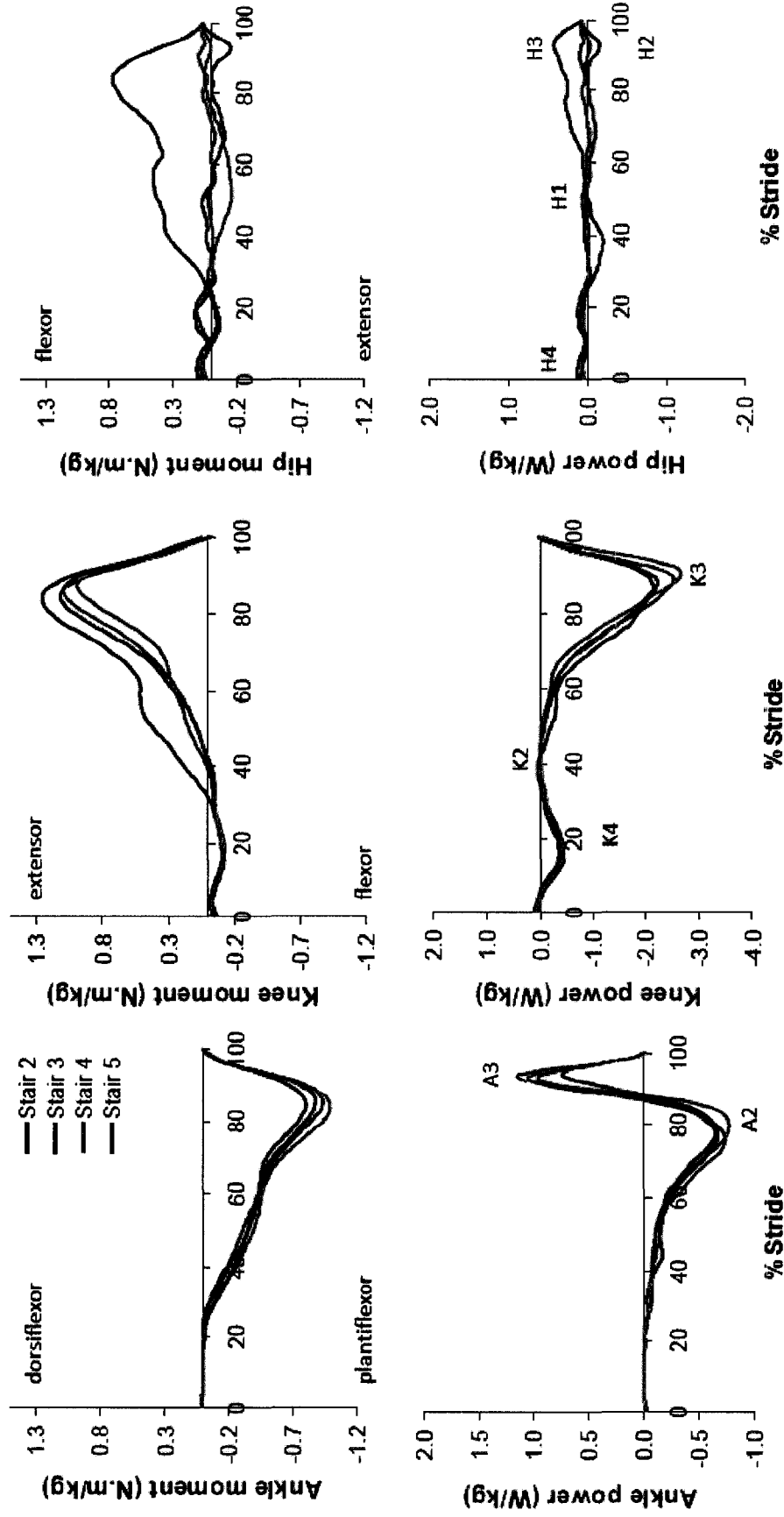


**Figure 4-3.** Grand ensemble averaged ( $n = 17$ ), body-mass normalized kinetics of the lead limb in step-by-step stair descent. Moments of force (top row) and powers (bottom row) are reported for the ankle (left column), knee (middle column) and hip (right column) joints of the lead limb in step-by-step stair descent.

Only the generative hip power burst, H4, when the hip flexor moment contributed to the energy of the limb at push-off was consistent across all gait cycles.

Figure 4-4 shows the ensemble averaged moments and powers performed about the joints of the trail limb in step-by-step stair descent (SBST). Stance phase ankle and knee kinetics at initial contact were different than those observed in FD, BD and SBSL. The dissipative plantiflexor power burst A1 and dissipative knee burst K1 were absent in SBST, since initial-contact in SBST is characterized by tandem stance; body weight is supported by the lead limb at initial-contact. In terminal stance, knee kinetics were similar to those observed in FD and BD conditions, characterized by an eccentric burst of the knee extensor moment (K3). Similar to the SBSL condition, the stance phase hip moment and power magnitudes were near zero so that the hip contributed little to progression in SBST. However, at terminal stance the net hip flexor moment did positive work to swing the limb through to the next step (H3), comparable to hip kinetics observed in FD and SBSL conditions.

To facilitate comparison across conditions, Figures 4-5 to 4-8 show the ensemble averaged lower extremity kinetics demonstrated in FD, BD, SBSL and SBST conditions across stairs 2, 3, 4 and 5. Statistical analyses were performed on peak ankle moment and power magnitudes at bursts A1, A2 and A3. For moment, power and work comparisons at initial contact (A1), SBST kinetics were omitted due to the absence of the burst.



**Figure 4-4.** Grand ensemble averaged ( $n = 17$ ), body-mass normalized kinetics of the trail limb in step-by-step stair descent. Moments of force (top row) and powers (bottom row) for the ankle (left column), knee (middle column) and hip (right column) joints of the trail limb in step-by-step stair descent.

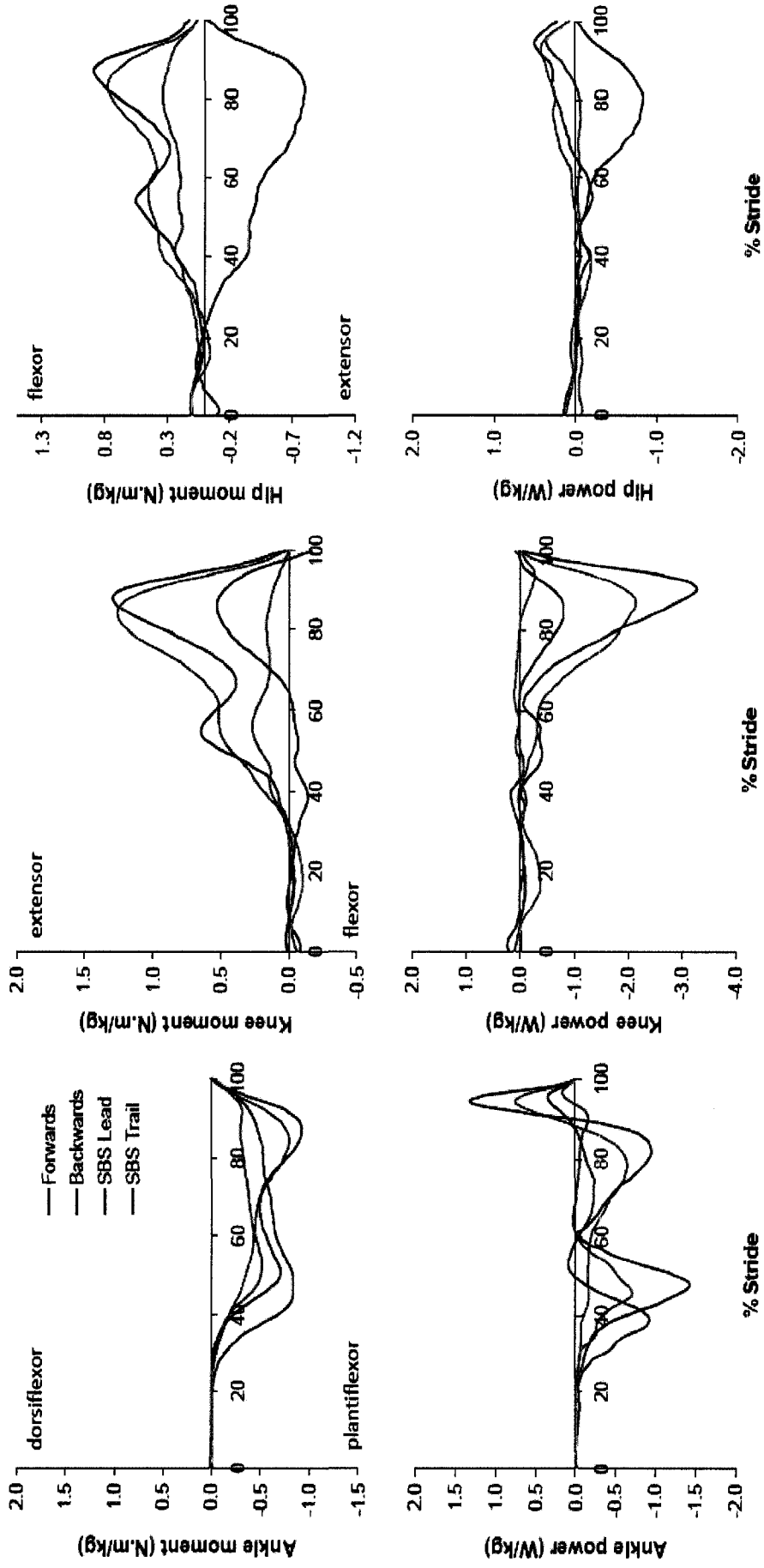
Statistical analyses for knee parameters were performed for peak moments, powers and the work done by the stance phase knee extensor moment. Figures 4-5 to 4-8 show that the net knee moment in terminal stance was extensor in FD, BD and SBST conditions; however, in SBSL the net knee moment was reduced in magnitude and changed from an extensor moment that performed little negative work to a flexor moment that did almost no work at stairs 3 and 4. The peak knee moments appeared similar in FD and SBST, but larger than those observed over all steps in BD. Similarly, burst K3 appeared larger in FD and SBST than BD and SBSL (at stairs 2 and 4). The knee extensor moment was similar in magnitude between FD and SBST, but reduced in SBSL relative to BD.

Table 4-1 revealed significant stair and condition main effects and a stair  $\times$  condition interaction for the plantiflexor moment at initial-contact. However, Figure 4-9 demonstrated both stair and condition main effects were overridden by the stair  $\times$  condition interaction. As such, the peak ankle plantiflexor moment patterns showed no clear pattern across conditions, so that in FD the plantiflexor moment increased in magnitude through the midstair region. Conversely, in BD and the SBSL limb, the plantiflexor moment at initial contact was similar in magnitude across all stairs.

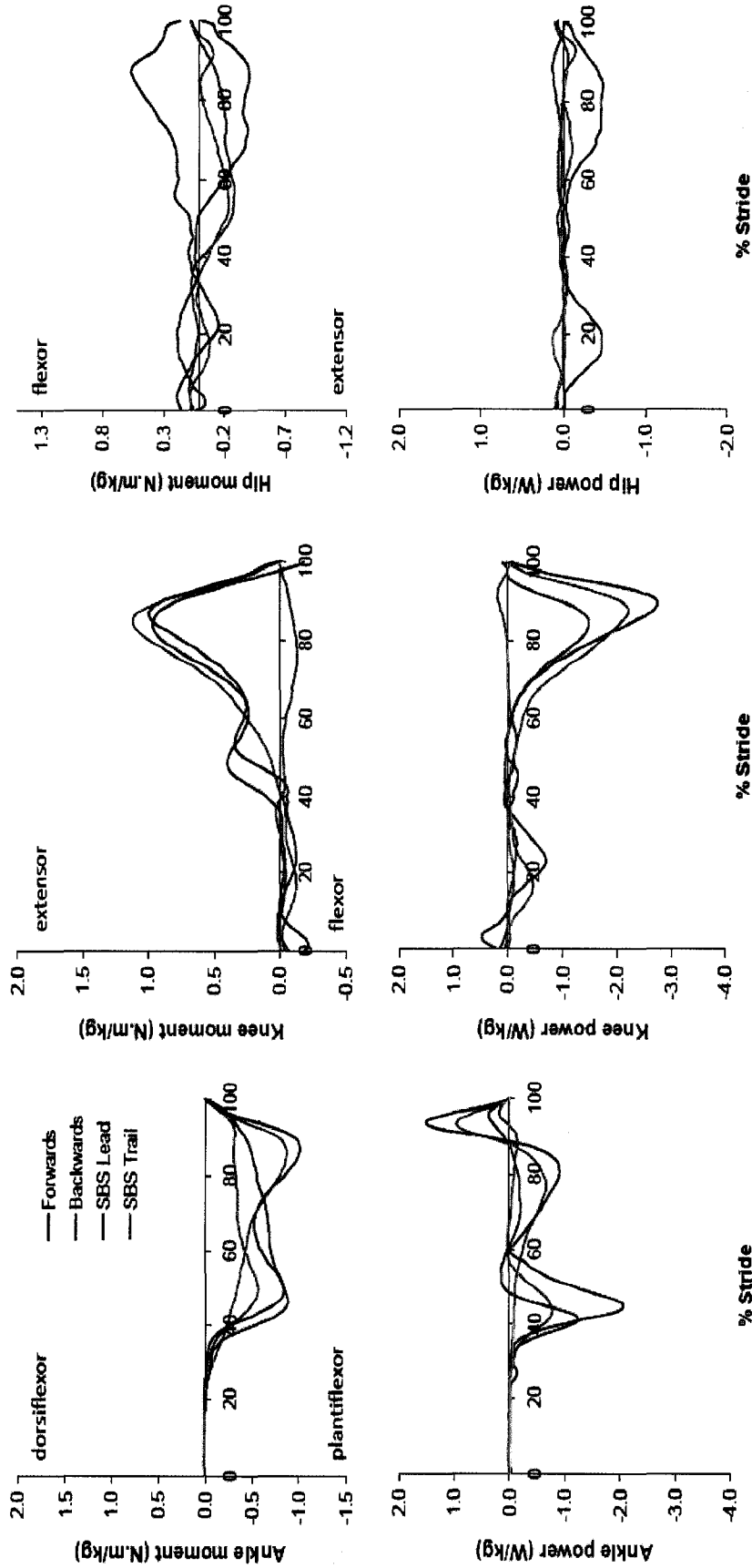
Further, Table 4-1 demonstrates significant stair and condition main effects and a stair  $\times$  condition interaction for peak plantiflexor power burst at initial contact (A1). For the stair effect, ankle plantiflexor powers were not different between stairs 2 and 3 ( $p > 0.05$ ), but were smaller than those at stairs 4 ( $p < 0.01$ ) and 5 ( $p < 0.001$ ). Similarly, there were no discernible differences between stairs 4 and 5 ( $p > 0.05$ ). There was no

difference in the ankle plantiflexor burst A1 between SBSL and BD ( $p>0.05$ ), which were smaller than those demonstrated in FD ( $p<0.001$ ).

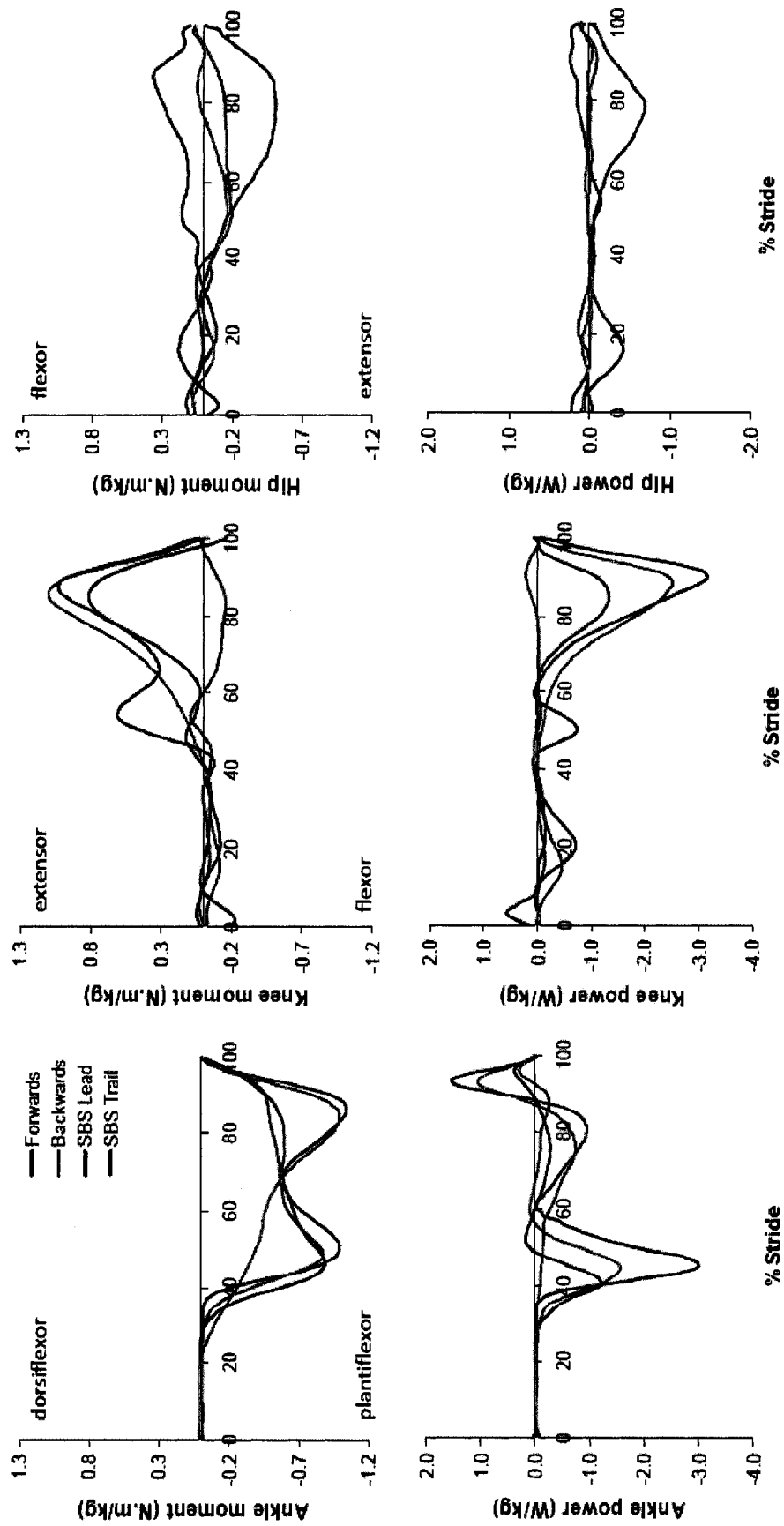
Mechanical work magnitudes at burst A1 followed a similar trend. There were significant stair and condition main effects and a stair  $\times$  condition interaction. Whereas the stair effect was overridden by the interaction, work done by the plantiflexor moment was similar for BD and SBSL conditions ( $p>0.05$ ), but greater for FD than both BD ( $p<0.01$ ) and SBSL ( $p<0.001$ ). Figure 4-9 shows the stair  $\times$  condition interaction. The interaction can be explained as follows. Work magnitudes remained constant across stairs in the BD and SBSL conditions, but increased monotonically as the individual progressed through the staircase in FD.



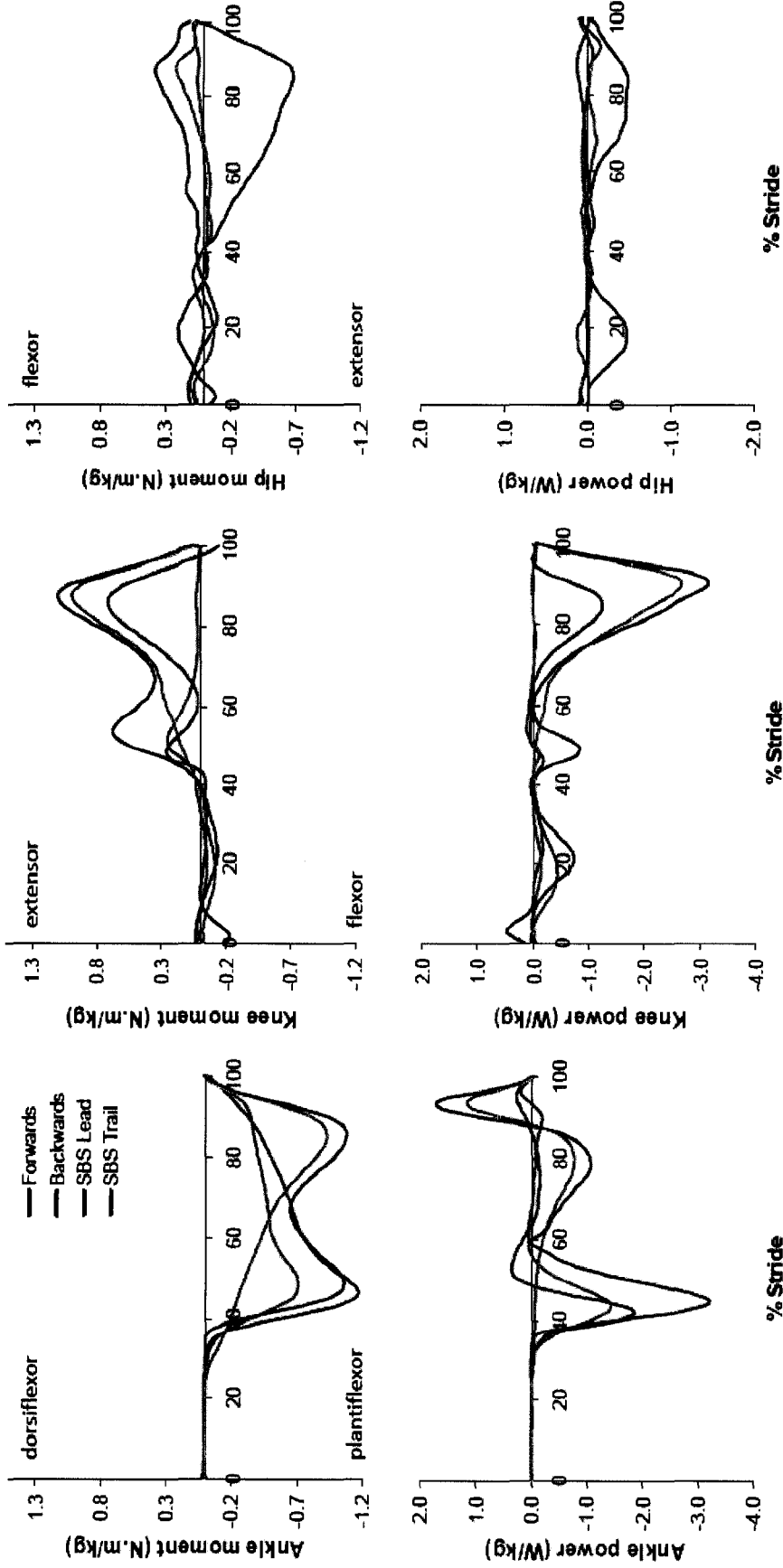
**Figure 4-5.** Grand ensemble averaged ( $n = 17$ ), body-mass normalized moments of force (top row) and moment powers (bottom row) for the ankle (left column), knee (middle column) and hip (right column) joints at stair 2, across all conditions. FD: Blue; BD: Black; SBSL: Green; SBSST: Red



**Figure 4-6.** Grand ensemble averaged ( $n = 17$ ), body mass normalized moments of force (top row) and moment powers (bottom row) for the ankle (left column), knee (middle column) and hip (right column) joints at stair 3, across all conditions. FD: Blue; BD: Black; SBSL: Green; SBST: Red.



**Figure 4-7.** Grand ensemble averaged ( $n = 17$ ), body mass normalized moments of force (top row) and powers (bottom row) for the ankle (left column), knee (middle column) and hip (right column) joints at stair 4, across all conditions. FD: Blue; BD: Black; SBSL: Green; SBST: Red.



**Figure 4-8.** Grand ensemble averaged ( $n = 17$ ), body mass normalized moments of force (top row) and powers (bottom row) for the ankle (left column), knee (middle column) and hip (right column) joints at stair 5, across all conditions. FD: Blue; BD: Black; SBSL: Green; SBST: Red.

**Table 4-1.** Summary of statistical results for the ankle peak moment (N.m/kg), power (W/kg) and work done by the plantiflexor moment (J/kg) at initial-contact (A1).

	Univariate tests of within-subjects effects ( 4×3 RM-ANOVA)		
	Stair ( $F_{(3, 48)}$ )	Condition ( $F_{(2, 32)}$ )	Stair×Condition ( $F_{(6, 96)}$ )
<i>Plantiflexor moment</i>	22.36***	6.53**	3.80**
<i>Plantiflexor power</i>	31.49***	19.29***	7.43***
<i>Work</i>	27.30***	29.48***	9.88***

\* $p < 0.05$ , \*\* $p < 0.01$ , \*\*\* $p < 0.001$

**Table 4-2.** Summary of statistical results for ankle peak plantiflexor moment (N.m/kg) and concentric power (W/kg) at push-off (A3).

	Univariate tests of within-subjects effects ( 4×4 RM-ANOVA)		
	Stair ( $F_{(3, 48)}$ )	Condition ( $F_{(3, 48)}$ )	Stair×Condition ( $F_{(9, 144)}$ )
<i>Plantiflexor moment</i>	16.53***	79.62***	3.37**

\* $p < 0.05$ , \*\* $p < 0.01$ , \*\*\* $p < 0.001$

Table 4-2 demonstrates that significant stair and condition effects and a stair × condition interaction were found for the ankle plantiflexor moment at push-off (A3). Figure 4-10 shows the ankle plantiflexor moment varied across stairs and conditions, culminating in the stair effect being overridden by the interaction. However, there was a condition effect so that the plantiflexor moment at push-off was larger in FD than all other conditions ( $p < 0.05$ ). Moment magnitudes were similar between BD and SBSL ( $p > 0.05$ ). Lastly, the peak moment at toe-off was significantly larger in SBST than BD

and SBSL ( $p < 0.001$ ). The stair  $\times$  condition interaction was mitigated by differences in patterning at stair 3 in the SBSL condition, where the plantiflexor moment decreased in magnitude relative to FD, BD and SBST conditions.

**Table 4-3.** Summary of statistical results for ankle plantiflexion concentric power (W/kg) and total work (J/kg) at push-off (A3).

	Multivariate tests: Pillai's Trace <sup>†</sup>		
	Stair ( $F_{(3, 14)}$ )	Condition ( $F_{(3, 14)}$ )	Stair $\times$ Condition ( $F_{(9, 8)}$ )
<i>Plantiflexor power</i>	15.58***	102.40***	1.83*
<i>Total work</i>	21.93***	92.48***	7.09**

\* $p < 0.05$ , \*\* $p < 0.01$ , \*\*\* $p < 0.001$

<sup>†</sup>Mauchly's test showed sphericity was violated, Greenhouse-Geisser correction,  $\epsilon < .7$ , and  $n > 10$ ; multivariate test statistics are reported that are not dependent on the sphericity assumption.

Table 4-3 reports summary statistics for the peak power and work done at push-off by the plantiflexors, across all stairs and conditions. Significant stair and condition effects and a stair  $\times$  condition interaction were observed for the magnitude of plantiflexor powers at push-off. In all conditions but FD, plantiflexor powers were significantly smaller at stair 2 than other stairs, but similar thereafter. As such, the stair  $\times$  condition interaction dominated the reported stair effect. The condition effect remained significant, despite the interaction. Figure 4-10 shows the peak power at push-off was larger in FD compared to all other conditions ( $p < 0.001$ ) and further, was significantly larger in SBST compared to BD and SBSL ( $p < 0.001$ ). No discernible differences in peak plantiflexor powers were found between SBSL and BD conditions ( $p > 0.05$ ).

Finally, there were stair and condition main effects and a stair  $\times$  condition interaction effect for work done by the plantiflexors at push-off. Figure 4-10 demonstrates there were no systematic stair effects across conditions. Similarly, there was no consistent condition effect. As a result, the stair  $\times$  condition interaction precipitated from differences in work magnitudes across both stairs and conditions. Work done by the ankle plantiflexors in the FD, BD and SBSL conditions remained constant across stairs. However, work done in the SBST condition was reduced during the first gait cycle, but was not different thereafter.

Table 4-4 shows summary statistics for the dependent variables peak ankle moment, power and work done by the plantiflexors during midstance, when the plantiflexors controlled body lowering while progressing to the next step. Peak ankle moment magnitudes were influenced by stair and condition main effects. Further, there was a significant stair  $\times$  condition interaction. Idiosyncrasies in ankle moment magnitudes rendered the stair effect inconsequential, since no stair effect was consistent across conditions. In BD, the plantiflexor moment was near constant through the midstair region, in contrast to all other conditions, where larger increases in the ankle moments were observed between steps. The condition effect can be summarized as follows. FD plantiflexor moments were not significantly different from SBST ( $p > 0.05$ ), whereas those observed in FD and SBST were significantly greater than SBSL and BD ( $p < 0.001$ ). Additionally, the BD plantiflexor moment was larger than the SBSL moment ( $p < 0.01$ ). The interaction effect resulted from differences at stair 4 in the SBSL condition (increased in absolute magnitude) and stair 5 in BD (decreased in absolute magnitude).

**Table 4-4.** Summary of statistical results for the peak plantiflexor moment (N.m/kg), power (W/kg) and work (J/kg) done by the ankle plantiflexors in midstance (A2).

	Multivariate tests: Pillai's Trace <sup>†</sup>		
	Stair ( $F_{(3, 14)}$ )	Condition ( $F_{(3, 14)}$ )	Stair×Condition ( $F_{(9, 8)}$ )
<i>Plantiflexor moment</i>	18.47***	64.52***	3.58*
<i>Plantiflexor power</i>	2.60	58.66***	3.68*
<i>Total work</i>	1.13	42.01***	3.40*

\* $p < 0.05$ , \*\* $p < 0.01$ , \*\*\* $p < 0.001$

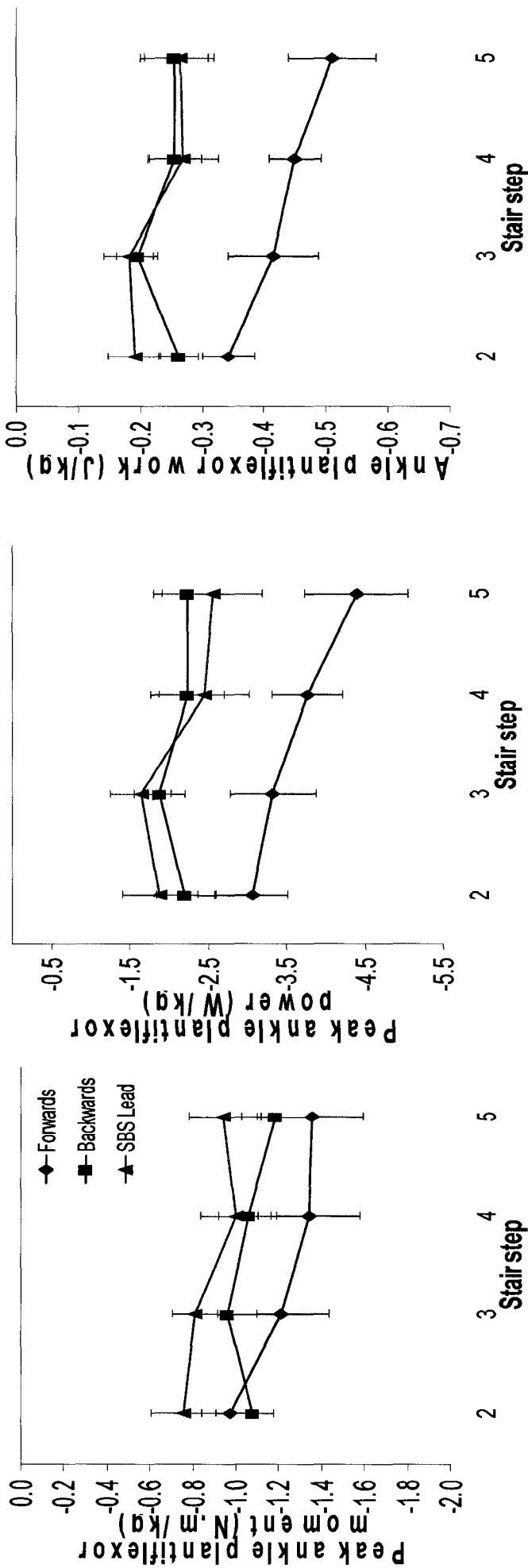
<sup>†</sup>Mauchly's test showed sphericity was violated, Greenhouse-Geisser correction,  $\epsilon < 0.7$ , and  $n > 10$ ; multivariate test statistics are reported that are not dependent on the sphericity assumption.

There was a condition effect and a significant stair × condition interaction for the dependent variable, peak plantiflexor power in midstance. Plantiflexor powers were greater in FD than SBST ( $p < 0.05$ ), while magnitudes for FD and SBST were larger than those demonstrated in SBSL and BD ( $p < 0.001$ ). The magnitude of the plantiflexor power was not significantly different between BD and SBSL ( $p > 0.05$ ). The stair × condition interaction arose from differences in the magnitude of the ankle power among BD and FD, SBSL, and SBST. The peak power decreased in magnitude at stair 3 (second complete gait cycle) in FD, SBSL and SBST and remained relatively constant thereafter. In BD the patterning was opposite, the peak plantiflexor power increased in absolute magnitude at stair 3, but decreased thereafter.

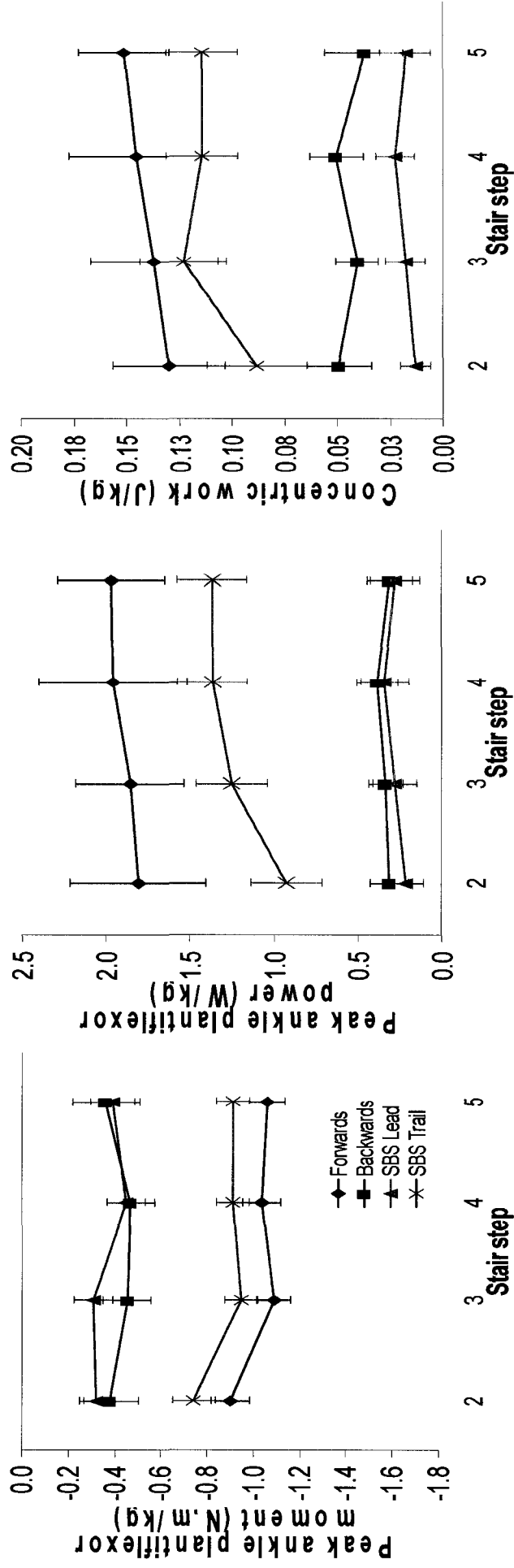
Finally, there was a significant condition effect and a stair × condition interaction for work performed by the plantiflexors during the eccentric midstance burst (A2). The

stair effect was not significant. Work magnitudes were not significantly different between FD and SBST ( $p>0.05$ ), while work done by the plantiflexor moment was significantly larger for FD and SBST than BD ( $p<0.001$ ) and SBSL ( $p<0.001$ ). Lastly, work done by the plantiflexor moment was significantly larger in BD relative to SBSL ( $p<0.01$ ). The interaction effect can be summarized as follows. Work was close to constant across all four stairs in the FD and SBST conditions. Conversely, less work was done at stair 2 for the SBSL condition, increased in magnitude at stair 3 and stayed constant thereafter. For the BD condition, work was constant across stairs, but decreased markedly in the transition to level walking at stair 5.

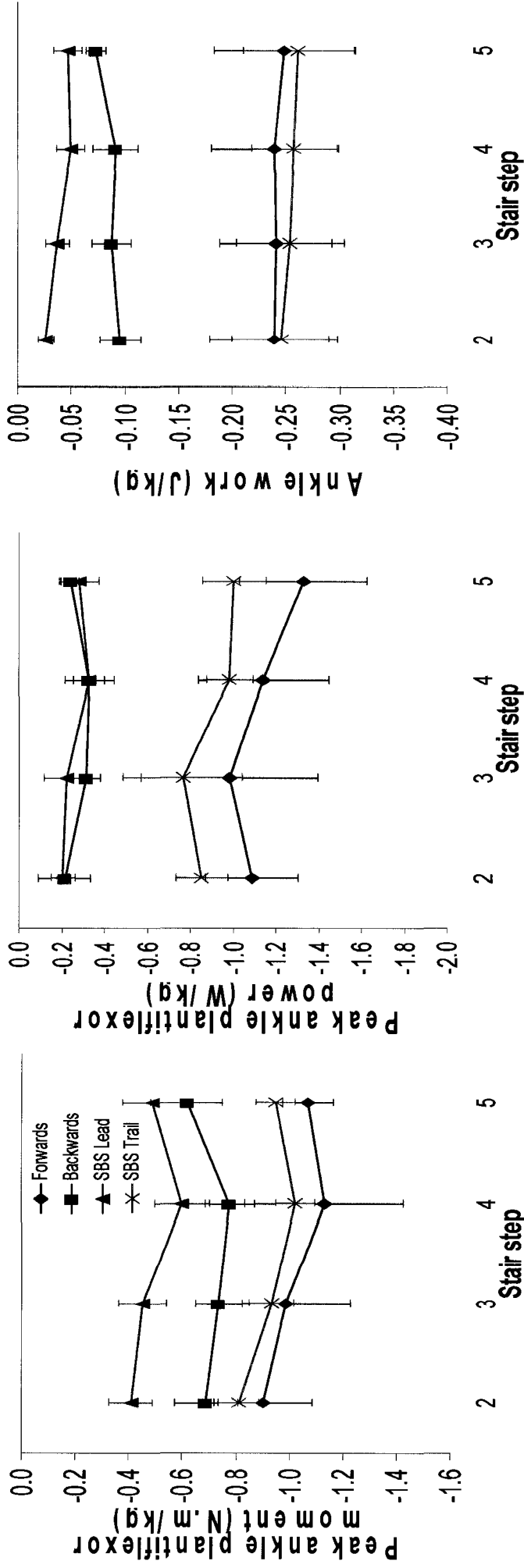
Table 4-5 presents summary statistics for the peak knee moment, power and work from mid- through terminal stance. There was a significant stair  $\times$  condition interaction, such that the knee moment depended on both stair and condition. In the SBSL condition, the knee moment was flexor at stairs 3 and 4. In FD, BD and SBST there was a knee extensor moment across all stairs with knee moments largest at stair 2 for FD and SBST. In BD the trend was opposite; the knee moment was smallest in transition from quiet standing and increased from stair 3. Therefore, no stair main effect was consistent across conditions. However, there was a significant condition effect, such that peak knee moments in BD and SBSL were smaller than those demonstrated in FD and SBST ( $p<0.001$ ). Knee moments were not significantly different between FD and SBST ( $p>0.05$ ). Peak knee moments in BD were larger relative to those demonstrated in SBSL ( $p<0.001$ ).



**Figure 4-9.** Ensemble-averaged, body mass normalized peak ankle plantiflexor moments (N.m/kg) powers (W/kg), and work plotted over four consecutive gait cycles for forwards, backwards and the step-by-step lead gait patterns. **A1 is period when plantiflexor moment dissipates gravitational energy after initial-contact.** Error bars represent 95% confidence intervals.



**Figure 4-10.** Ensemble-averaged, body mass normalized moments (N.m/kg), powers (W/kg) and work (J/kg) done by the ankle plantiflexors plotted over four consecutive gait cycles for forwards, backwards and step-by-step gait patterns. **A3 is period when ankle plantiflexor moment provides energy at push-off.** Error bars denote 95% confidence intervals.



**Figure 4-11. . Ensemble-averaged, body mass normalized moments (N.m/kg), powers (W/kg) and work (J/kg) done by the ankle**

plantiflexors plotted over four consecutive gait cycles for forwards, backwards and step-by-step gait patterns. **A2 is period when**

**ankle plantiflexor moment controls lowering during midstance and when the contralateral limb is in swing.** Error bars represent

95% confidence intervals.

**Table 4-5.** Summary of statistical results for peak knee moment (N.m/kg) power (W/kg) and work (J/kg) performed about the flexion/extension axis ( $x$ -axis) in late stance.

	Univariate tests of within-subjects effects ( 4×4 RM-ANOVA)		
	Stair ( $F_{(3, 14)}$ )	Condition ( $F_{(3, 14)}$ )	Stair×Condition ( $F_{(9, 8)}$ )
<i>Knee moment</i>	12.50***	104.231***	25.27***
<i>Knee power</i>	14.95***	172.00***	17.88***
<i>Total work</i>	3.143	171.84***	3.58**

\* $p < 0.05$ , \*\* $p < 0.01$ , \*\*\* $p < 0.001$

†Mauchly's test showed sphericity was violated ( $p < 0.05$ ), Greenhouse-Geisser correction,  $\epsilon < .7$ , and  $n > a + 10$ ; multivariate test statistics are reported that are not dependent on the sphericity assumption.

Table 4-5 also shows significant stair and condition main effects and a stair × condition interaction for the stance phase knee power burst K3. As shown in Figure 4-12, knee power tended to be largest at stair 2 in FD, SBST and SBSL and remained relatively constant in the midstair region (gait cycles 2, 3 and 4). In BD the trend was opposite, the knee power magnitude was smallest at stair 2, but increased from stair 3 so that the overall stair effect was overridden. Nevertheless, there was a parsimonious condition effect, such that knee power was larger in FD than all other conditions ( $p < 0.001$ ). Further, the eccentric knee powers were larger in SBST than BD and SBSL ( $p < 0.001$ ). Knee powers in BD were larger than those demonstrated in SBSL ( $p < 0.001$ ).

Lastly, Table 4-5 shows a significant condition effect and a stair × condition interaction for work performed during the eccentric burst K3. Stair did not elicit a significant effect. In terms of the condition effect, mechanical work associated with the

burst K3 was substantially larger in SBST than other conditions ( $p < 0.001$ ). Work done in FD was larger than that in BD ( $p < 0.001$ ), which in turn was greater than that performed in SBSL ( $p < 0.001$ ). Mechanical work was largest at stair 2 for FD, SBSL and SBST, decreased at stair 3 and remained relatively constant at stairs 4 and 5. In BD, knee kinetics were different, since work done by the knee extensor moment increased from stair 2 to stair 3 and gradually decreased at stairs 4 and 5. However, as indicated by the 95% confidence interval, work magnitudes were variable so that the observed trends were minimized.

Table 4-6 summarizes statistical tests for the peak hip moments, powers and work done at the hip during the dissipative burst H2 in FD and BD. Hip moment, power and work magnitudes were reduced to near zero in SBSL so that the lead and trail limb hips contributed little to progression and there was no consistent pattern across gait cycles. As such, statistics are presented for only FD and BD, where consistent kinetic patterning was observed across gait cycles. In FD, summary statistics represent tests performed for the dissipative hip flexor burst that occurred at weight acceptance, where the flexor moment acted eccentrically to limit extension. Given the nature of BD, summary statistics for this condition represent dissipative activity of the hip extensor moment, which prevented collapse in mid through late stance.

Table 4-6 reveals significant stair and condition main effects and a stair  $\times$  condition interaction for the peak hip moment. Post-hoc testing with Bonferroni correction revealed no significant differences in the hip moment across stairs ( $p > 0.05$ ). Conversely, the condition effect emanated from kinetic differences related to the forwards and backwards descent tasks. The hip moment was flexor in FD, and thus, opposite in sign

relative to BD. Not surprisingly, the hip moment was larger in FD than BD ( $p < 0.001$ ). The interaction was such that the absolute magnitude of the hip moment was greatest in transition from upright stance at stair 2, decreased (in absolute magnitude) into the midstair region and remained relatively constant thereafter.

Table 4-6 also demonstrates stair and condition effects and a stair  $\times$  condition interaction for the dissipative hip powers. As previously mentioned, this represents the hip flexor burst, H2, in FD and the dissipative extensor burst, H2, in BD. Figure 4-13 shows the interaction was such that in FD, the absolute magnitude of the peak hip power was largest at stair 2, decreased at stair 3 and remained close to constant for stairs 4 and 5. In BD, the peak hip power was largest in transition from quiet stance (first gait cycle; stair 2), decreased over the second gait cycle and increased progressively over the third and fourth gait cycles. Since peak hip powers varied idiosyncratically in FD and BD, the reported stair effect was inconsistent with the observed interaction. Nevertheless, the magnitude of the eccentric hip power was markedly larger in BD relative to FD ( $p < 0.001$ ).

Finally, Table 4-6 demonstrates that eccentric work done at the hip was influenced by stair and condition effects and a stair  $\times$  condition interaction. Figure 4-13 demonstrates that eccentric work done by the flexor moment in FD was constant across gait cycles. Similarly, work magnitudes in BD were relatively constant through the staircase despite a marginal decrease from stair 2 to stair 3, increased from stair 3 to 4, and decreased from stair 4 to stair 5. Thus, the reported stair effect was overridden by the interaction for work done at the hip, since the observed work magnitude trends were

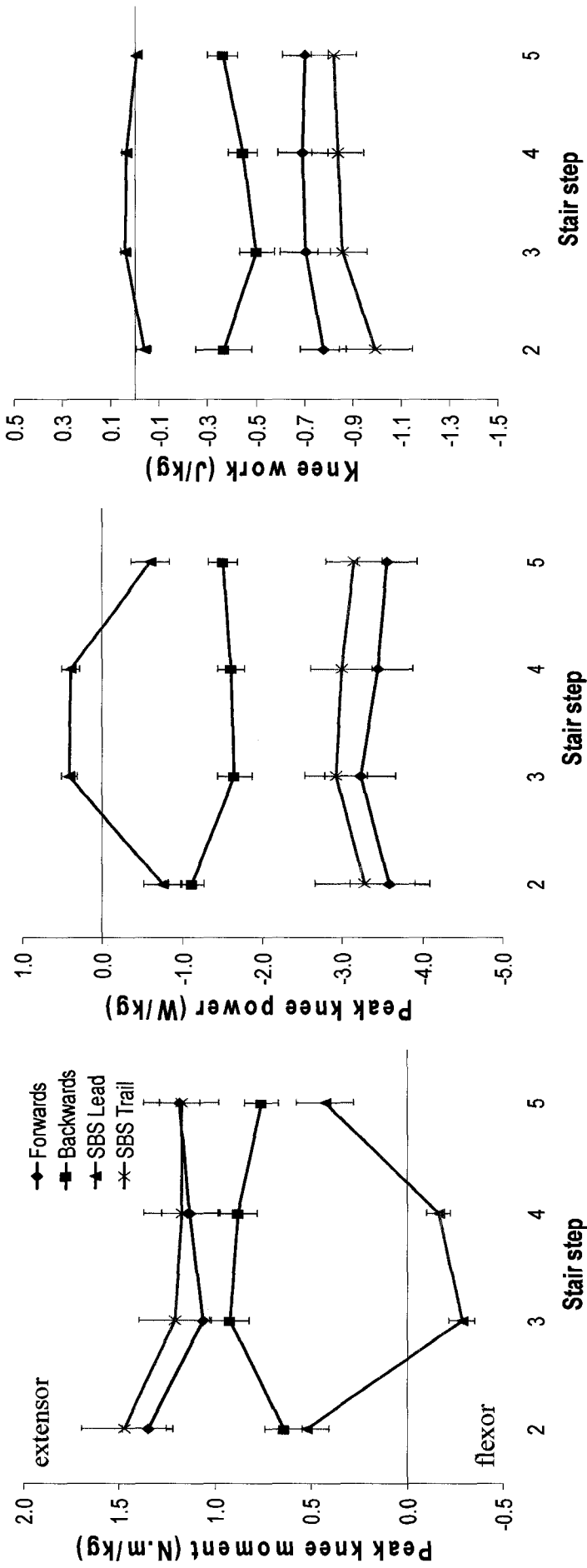
different in FD and BD. Nevertheless, there was a condition effect with the magnitude of eccentric work done at the hip markedly larger in BD relative to FD ( $p < 0.001$ ).

**Table 4-6.** Summary of statistical results for peak hip moment (N.m/kg), power (W/kg) and work (J/kg) performed about the flexion/extension axis ( $x$ -axis) in early stance (through terminal stance in backwards descent).

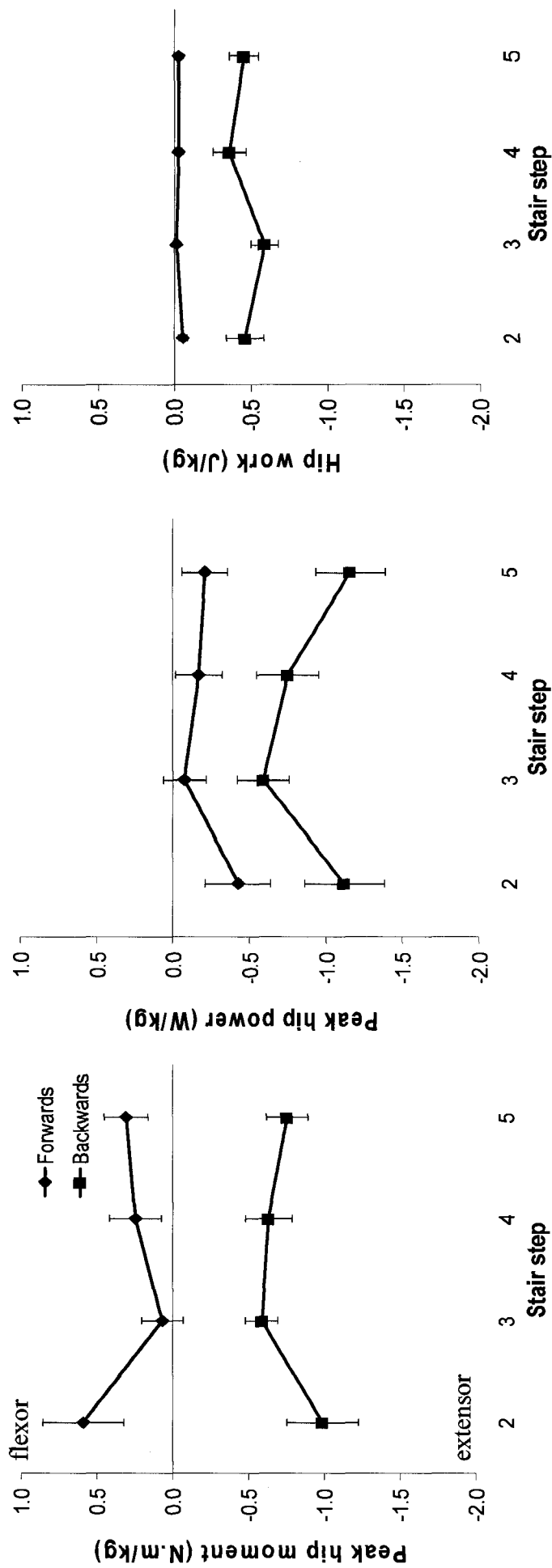
	Multivariate test: Pillai's Trace <sup>†</sup>		
	Stair ( $F_{(3, 14)}$ )	Condition ( $F_{(1, 16)}$ )	Stair×Condition ( $F_{(3, 14)}$ )
<i>Hip moment</i>	3.42*	94.92***	26.41***
<i>Hip power</i>	16.22***	56.65***	3.94*
<i>Hip work</i>	20.61***	77.02***	10.34***

\* $p < 0.05$ , \*\* $p < 0.01$ , \*\*\* $p < 0.001$

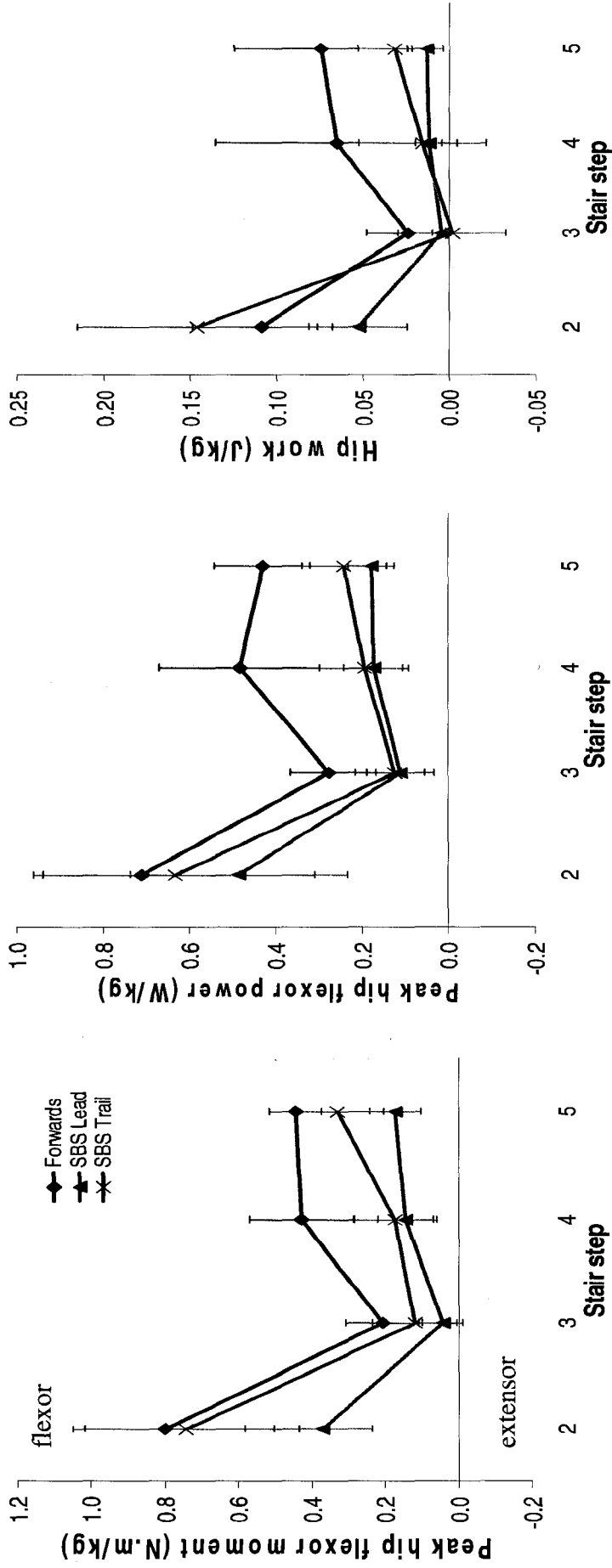
<sup>†</sup>Mauchly's test showed sphericity was violated, Greenhouse-Geisser correction,  $\epsilon < .7$ , and  $n > 10$ ; multivariate test statistics are reported that are not dependent on the sphericity assumption.



**Figure 4-12.** . Ensemble-averaged, body mass normalized moments (N.m/kg), powers (W/kg) and work (J/kg) done at the knee over four consecutive gait cycles for forwards, backwards and step-by-step gait patterns. **K3 is period when knee extensor moment controls lowering during midstance.** Error bars denote 95% confidence intervals.



**Figure 4-13.** Ensemble-averaged, body mass normalized peak hip moments (N.m/kg) and powers (W/kg) plotted over four consecutive gait cycles for forwards and backwards descent strategies. **H2 is period when hip flexor moment performs negative work during the stance phase of forwards stair descent or when hip extensor moment controls flexion during stance phase of backwards descent.** Error bars denote 95% confidence intervals.



**Figure 4-14.** Ensemble-averaged, body mass normalized moments (N.m/kg), powers (W/kg) and work (J/kg) done at the hip for four consecutive gait cycles in forwards, SBSL and SBST descent strategies. **H3 is period when hip flexor moment performs positive work during the stance phase to swing the limb to the next step.** Error bars represent the 95% confidence intervals.

**Table 4-7.** Summary of statistical results for peak hip moment (N.m/kg), power (W/kg) and work (J/kg) performed by the flexor moment during the generative burst, H4, in terminal stance.

	Multivariate test: Pillai's Trace <sup>†</sup>		
	Stair ( $F_{(3, 14)}$ )	Condition ( $F_{(2, 15)}$ )	Stair×Condition ( $F_{(6, 11)}$ )
<i>Hip moment</i>	42.38***	28.07***	4.57*
<i>Hip power</i>	10.06***	13.00**	1.29
<i>Hip work</i>	15.23***	6.77***	3.16*

\* $p < 0.05$ , \*\* $p < 0.01$ , \*\*\* $p < 0.001$

<sup>†</sup>Mauchly's test showed sphericity was violated ( $p < 0.05$ ), Greenhouse-Geisser correction,  $\epsilon < .7$ , and  $n > a + 10$  (where  $a$  is the number of the levels of the independent variables, stair and condition); multivariate test statistics are reported that are not dependent on the sphericity assumption.

Table 4-7 reports summary statistics for the peak hip flexor moments and powers, and work done by the flexor moment in terminal stance. In this phase, the hip flexor moment generates energy to prepare the limb for swing phase. Statistics are reported for FD, SBSL and SBST conditions. No comparable event existed in the BD condition. Table 4-7 shows that peak hip flexor moments were influenced by stair and condition main effects and a stair × condition interaction. Figure 4-14 elucidates the interaction. As demonstrated, hip moments were greatest in the transition from quiet stance, but decreased over the second gait cycle (stair 3). Immediately following, the peak hip moments increased through the staircase, over the third and fourth gait cycles. Further, the moment magnitude was most variable during the transition from quiet stance, as shown by the magnitude of the confidence intervals for all conditions in Figure 4-14.

Whereas hip moments stabilized over the second to fourth gait cycles in FD and SBSL conditions, they tended to increase for the SBST limb. As such, there were no stair or condition effects, since the magnitude of the hip flexor moment was dominated by the interaction.

Further, Table 4-7 presents summary statistics for concentric hip powers in terminal stance. As demonstrated, both stair and condition factors influenced the peak hip power. While hip powers at push-off appeared larger in FD than SBSL and SBST conditions, variability rendered the effect inconsequential. Similarly, hip powers were stable across the mid- to terminal staircase in SBSL and SBST conditions, but there were marked differences in power across stairs 3 and 4 in FD. Despite the seemingly overbearing interaction, the stair  $\times$  condition interaction failed to reach significance.

Lastly, work done by the hip flexor moment at push-off was influenced by stair and condition effects and a stair  $\times$  condition interaction. Similar to the trends observed for both hip moments and powers, work contributed to push-off by the hip flexor moment was greatest at transition and decreased thereafter. Despite variability, work done by the hip flexors was smaller in SBSL than FD. Work done by the hip flexors was not different between SBSL and SBST conditions in the midstair region, but was larger over the first gait cycle in SBST.

Table 4-8 shows stair and condition effects and a stair  $\times$  condition interaction for maximum velocity of the total body COM in the direction of progression. As demonstrated in Figure 4-15, the interaction was due to different velocity trends across conditions. Velocity increased monotonically in FD. Similar trends were observed in SBSL, SBST and BD, but with increased variability of velocity magnitudes so that they

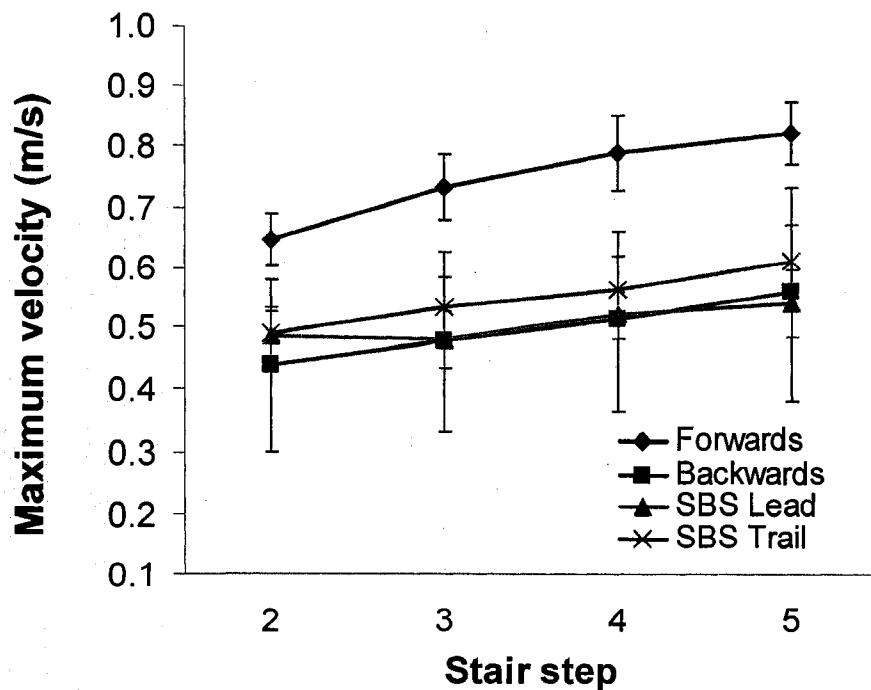
were not significantly different across stairs. Moreover, there was a condition effect, with maximum velocity being significantly larger in FD than BD ( $p < 0.05$ ), SBSL and SBST ( $p < 0.001$ ). Maximum velocity was not significantly different between BD, SBSL and SBST ( $p > 0.05$ ) conditions.

**Table 4-8.** Summary of statistical results for maximum progression velocity (m/s) computed tangential to staircase inclination.

	Multivariate tests: Pillai's Trace <sup>†</sup>		
	Stair ( $F_{(3, 14)}$ )	Condition ( $F_{(3, 14)}$ )	Stair $\times$ Condition ( $F_{(9, 8)}$ )
Maximum progression velocity	41.48***	45.92***	4.47*

\* $p < 0.05$ , \*\* $p < 0.01$ , \*\*\* $p < 0.001$

<sup>†</sup>Mauchly's test showed sphericity was violated, Greenhouse-Geisser correction,  $\epsilon < .7$ , and  $n > 10$ ; multivariate test statistics are reported that are not dependent on the sphericity assumption.



**Figure 4-15.** Maximum velocity parallel to progression over four consecutive gait cycles in FD, BD and SBS- descent patterns. Error bars represent 95% confidence intervals.

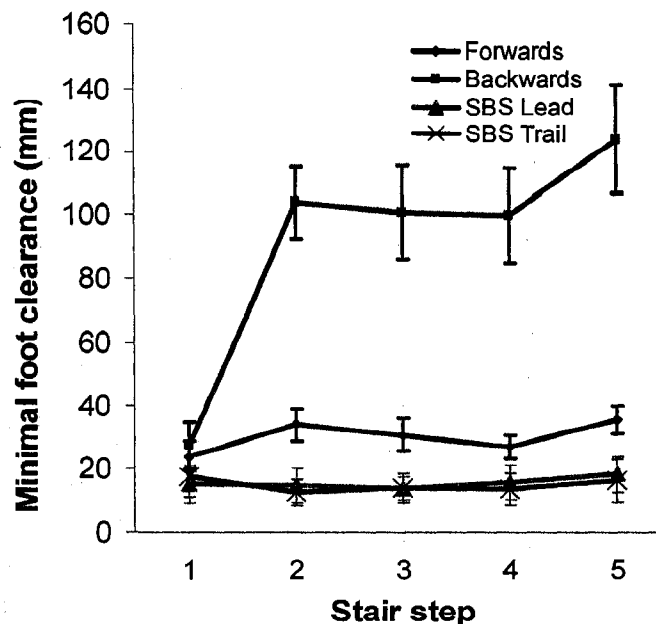
Table 4-9 reveals significant stair and condition main effects and a stair  $\times$  condition interaction for the dependent variable, minimal foot clearance. Figure 4-15 shows that both main effects were overridden by the interaction since foot clearance was similar

**Table 4-9.** Summary of statistical results for minimal foot clearance in forwards, backwards and step-by-step stair descent.

	Univariate tests of within-subjects effects (4 $\times$ 4 RM-ANOVA)		
	Stair ( $F_{(4, 14)}$ )	Condition ( $F_{(3, 15)}$ )	Stair $\times$ Condition ( $F_{(12, 6)}$ )
<i>Minimal foot clearance</i>	52.74***	113.10***	26.66***

\*\*\* $p < 0.001$ , \*\* $p < 0.01$ , \* $p < 0.05$

across all conditions at stair 1. Foot clearance increased in the midstair region (stair 2) in both FD and BD and decreased slightly thereafter, whereas foot clearance



**Figure 4-16.** Minimal foot clearance over five consecutive stairs for forwards, backwards and the lead and trail limbs in step-by-step descent. Error bars represent the 95% confidence interval.

remained constant in the SBSB conditions. The transition to level walking saw a further increase in foot clearance in both FD and BD conditions, though that demonstrated in BD was more drastic. Conversely, foot clearance for both limbs in SBSB was significantly lower than that demonstrated in either FD or BD and was constant across stairs. As such, the interaction rendered it such that clearance varied non-systematically between stairs and conditions so that the elicited stair and condition effects were not meaningful.

### **Correlation Analyses**

Correlation analyses were performed between stance phase eccentric power magnitudes at the ankle (A1), knee (K3) and hip (H2) and maximum sagittal plane progression velocity. There was a moderate inverse relationship between progression velocity and the magnitude of the dissipative ankle burst A1 in FD ( $r(66) = -0.487, p < 0.001$ ), with a significant proportion of peak ankle power magnitude variance explained by progression velocity ( $r^2 = 0.24, F(1, 66) = 21.74, p < 0.001$ ). There were no significant correlations between knee power at burst K3 ( $r^2 = 0.0021, r(66) = .148, p > 0.05$ ) or hip power at burst H1 ( $r^2 = 0.00005, r(66) = -.047, p > 0.05$ ). Figure 4-17 shows correlation plots between progression velocity and peak joint powers in FD.

Similarly, correlation analyses were carried out between maximum progression velocity (tangential to staircase) and ankle, knee and hip power magnitudes in BD. These analyses are presented in Figure 4-18. There were no significant correlations between the magnitude of the dissipative ankle burst, A1 ( $r(66) = .122, p > 0.05, r^2 = 0.0148$ ), the dissipative hip burst, H5 ( $r(66) = -.035, p > 0.05, r^2 = 0.0012$ ) or the dissipative knee burst

K3 ( $r(66) = -.091, p > 0.05, r^2 = 0.0082$ ) and maximum progression velocity across all stairs.

Correlation analyses were not performed for the SBSL and SBST conditions. The magnitude of hip moments and powers were drastically reduced in SBSD so that very little work was done by the lead and trail limb hip moments. As a result, the hip power bursts were not reproducible between subjects. As previously mentioned, the terminal stance concentric hip flexor burst H3 was the only burst consistent across subjects.

## DISCUSSION

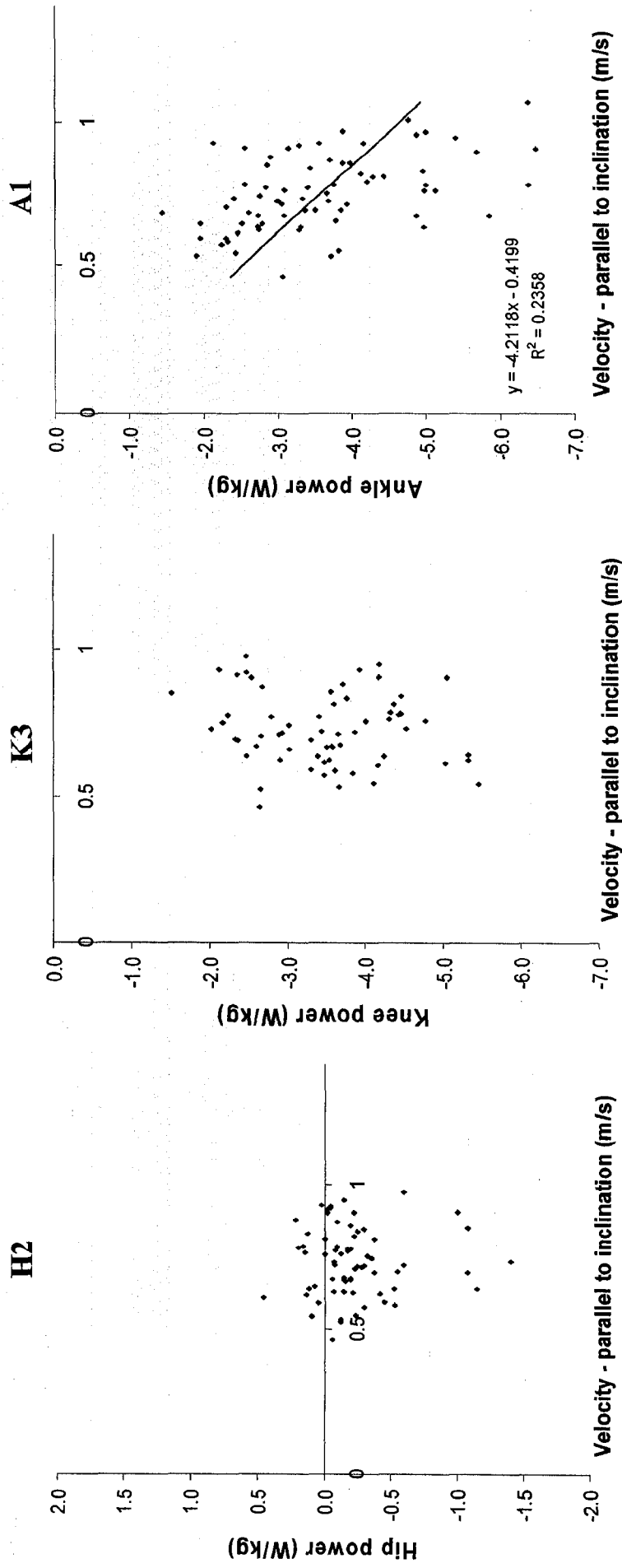
The purpose of this study was to understand mechanical differences between forwards, backwards and step-by-step descent patterns. Peak moments, powers and mechanical work about the flexion/extension axes of the ankle, knee and hip joints were computed and compared across conditions. The aforementioned parameters were extracted for ankle bursts A1, A2 and A3, and the knee burst, K3. For the hip, dissipative bursts H2 in FD and BD were contrasted. The magnitude of peak moments and powers, as well as work done during the concentric hip flexor burst H3 were compared across FD, SBSL and SBST conditions. The ensuing text will highlight and discuss results to determine if one pattern effectively moderated the demand of the descent task. Secondly, this research was concerned with outlining transition effects from quiet standing to stair descent. Whereas previous research concluded transition to steady-state descent was completed by the end of the first gait cycle in forwards descent (Andriacchi *et al.*, 1980), the data presented here suggest researchers misrepresented the transition to steady-state progression. Lastly, this study considered foot clearance in FD, BD and SBSL to determine whether foot clearance was greater in any descent strategy and thereby reduced the likelihood of contacting a stair edge during the swing phase of stair descent.

### Minimal Foot Clearance

Foot clearance has received considerable attention as a factor underlying falls in young and elderly individuals (Cavanagh and Higginson, 2002; Hamel *et al.*, 2005). Foot clearance is the distance between the traversed stair edge and the foot during the swing phase. To date, the majority of research (Cavanagh and Higginson, 2002; Hamel *et al.*, 2005), including the present study, measured clearance in descent since falls are three

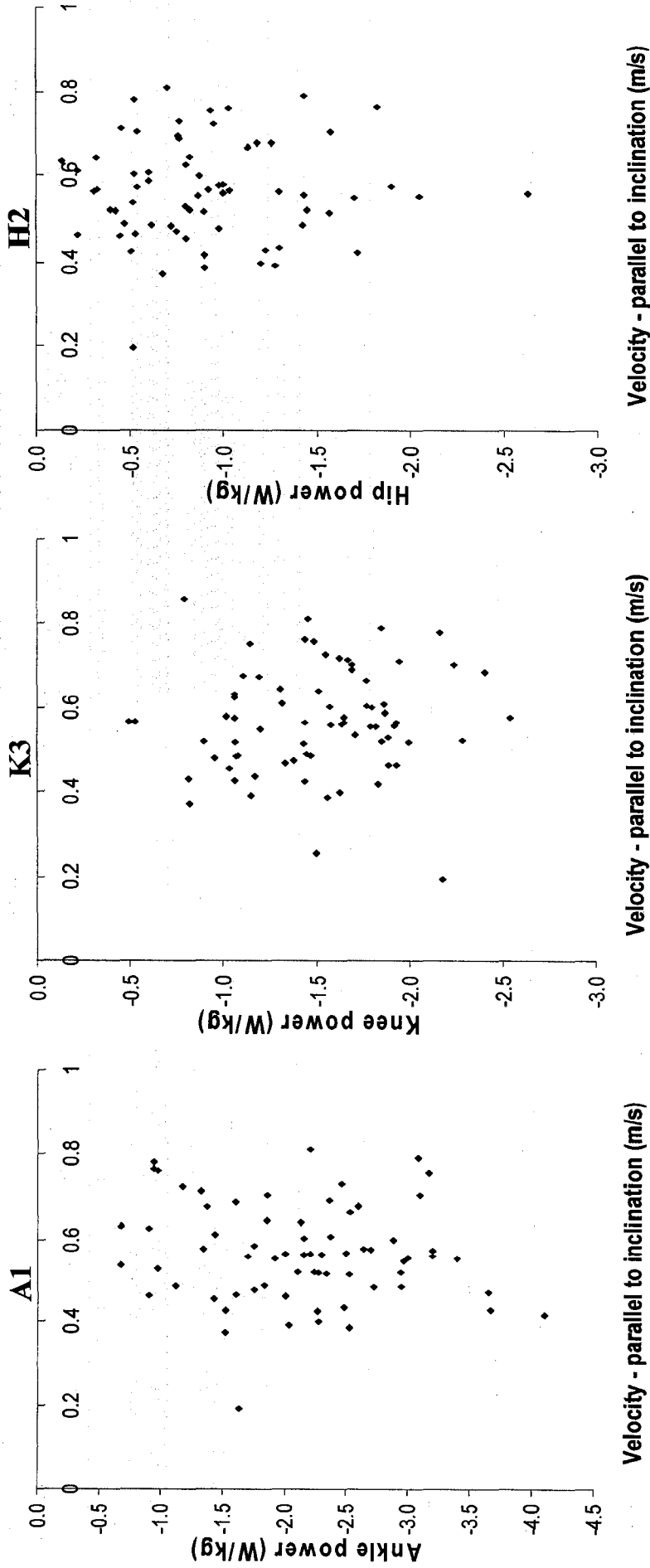
times more numerous than those incurred in ascent (Cohen, 2000). The results of this research replicated those published by Hamel and colleagues (2005) for the FD and BD strategies, in that clearance decreased progressively through the midstair region.

However, reduced clearance was reported in transition from quiet stance to stair descent at stair 1, relative to all other stairs in both FD and BD. This discrepancy likely reflects methodological differences between these results and previous research. In the Hamel and colleagues study (2005), participants were instructed to begin walking 2m before the first stair edge on the upper landing of the staircase apparatus. They were instructed to clear the first stair edge with the left foot, and continue down the stairs at a fixed speed of 0.65m/s. In this research, participants stood at the top of the staircase with instructions to begin descending the stairs with their preferred limb in FD and BD, the right limb in SBSL and the left limb in SBST. As such, Hamel and colleagues reported that clearance was greatest at the top of the staircase and decreased progressively thereafter. Hamel and colleagues (2005) argued that this trend suggests subjects were accommodating stair dimensions as they continued down the stairs and that this strategy likely represents the most energy efficient way to negotiate stairs.



**Figure 4-17.** Correlation plots between stance phase ankle (A1), knee (K3) and hip (H2) powers and progression velocity in forwards

FD. Ankle powers represent the magnitude of the dissipative plantiflexor power, A1 at initial-contact. Knee powers depict the magnitude of the dissipative knee extensor burst, K3. Hip powers represent the magnitude of the hip flexor or extensor burst in early stance, H2.



**Figure 4-18.** Correlation plots for stance phase ankle, knee and hip powers and progression velocity in BD. Ankle powers represent the magnitude of the dissipative plantiflexor power, A1 at initial-contact. Knee powers depict the magnitude of the dissipative knee extensor burst, K3. Hip powers represent the magnitude of the dissipative hip extensor burst, H2, in mid- to terminal stance.

The transition from standing to stair descent represents the initiation of descent, where progression down the staircase requires the foot clear only a single stair edge. Therefore, it seems that reduced clearance over the first step relative to all other steps in the staircase might result from this constraint. The trend toward reduced clearance at step 1 was especially pronounced in backwards descent, where clearance over stair 2 was five times higher relative to clearance over step 1. Similarly, clearance saw a second increase at step 5, which represented the transition from stair descent to level walking. In contrast to BD, this effect was more pronounced in FD, which saw a 24% increase in clearance, whereas that observed in BD increased approximately 20%. Increasing clearance would facilitate the transition to level walking, since the individual is not constrained to set stair dimensions that limit stride length. Similar findings were reported by Hamel and colleagues (2005). Aside from similar clearance over step 1 in all conditions, foot clearance was markedly lower in SBSD than the two step-over-step conditions; FD and BD. In SBSD, clearance was consistently low across all stairs, whereas it increased in the midstair region for FD and BD.

The foremost concerns of Hamel and colleagues (2005) were to delineate how age and luminance influenced foot clearance in healthy young and community-dwelling elderly individuals. In contrast, the work outlined here quantified clearance to substantiate BD as an alternate descent pattern, whereby the likelihood of foot contact with the traversed stair edge is minimized in swing. In confirmation of this premise, clearance in the midstair region was strikingly greater in BD than all other conditions. Therefore, aside from the apparent transition effect and despite greater inter-subject variability, clearance in BD was markedly larger than that observed in FD and the SBSD

conditions. While the observed transition effect reflected the constraints of the phase, the finding should not warrant significant clinical concern. As demonstrated, foot clearance was sufficiently large so that variability considered; in healthy participants, the likelihood of contacting the traversed stair was non-existent for both FD and BD conditions. Similarly, the likelihood of contacting a stair edge in the SBSD condition was non-existent, despite an overall reduction in clearance relative to other conditions. Subsequent research should examine whether this phenomenon merits further attention in older adults. Population trends demonstrated that older adults are prone to reduced, more variable clearance relative to their younger counterparts and are more likely to trip from contacting a stair edge in stair descent (Hamel *et al.*, 2005).

## **Kinetics**

Figures 4-1 to 4-4 plotted the grand-ensemble averaged moments and powers over stairs 2 to 5 in forwards and backwards descent, and the lead and trail limb kinetics in step-by-step stair descent. These figures confirmed previously reported results in that the three joints of both limbs were activated in a sequence directed toward shared energy absorption (McFadyen & Winter, 1988; Reiner *et al.*, 2002). Stair descent is a conservative task. During the swing phase, gravitational or potential energy is transferred to kinetic energy of the body that must be absorbed at initial-contact. As such, the ankle burst A1 was present in all conditions except the SBST limb. The absence of burst A1 in SBST is attributable to the asymmetric functions fulfilled by the lead and trail limbs in the step-by-step pattern. The role of the SBST limb is to control lowering of the body from one step to the next. In contrast, the SBSL dissipates energy at initial contact and maintains support as the trail limb is swung through to the next step. Therefore, burst A1

is absent for the SBST limb because initial-contact is characterized by tandem stance - gravitational energy was already dissipated by the lead limb plantiflexor moment.

In FD, the contralateral knee began dissipating gravitational energy prior to initial-contact (~60-100% stride of the contralateral limb). At this point, the knee extensor moment acted eccentrically to control body lowering while the ipsilateral limb progressed to the next step. In FD, the contralateral hip responded with energy absorption in early stance (~40-60% stride), when the hip flexors dissipated energy during the progression phase of stair descent. This phase was followed by a positive burst of the hip flexors, which generated energy to swing the limb through.

In BD, activity at the hips was opposite, but was nevertheless aimed at shared energy absorption. Given the kinetics of the BD task, there was a hip extensor moment in stance (~40-100% stride) that performed negative work to limit flexion and contributed to controlled lowering of the body. In comparison to FD and BD, hip moment and power magnitudes were significantly reduced in SBS, so that the hip contributed very little to supporting the body in this strategy. Only burst H4, where the hip flexor moment does positive work to swing the limb through, was reproduced across trials and subjects.

**Ankle kinetics.** Figure 4-1 plotted the grand-ensemble averaged ankle moments and powers in FD, which were similar to power profiles that existed in other studies (McFadyen & Winter, 1988; Reiner *et al.*, 2003; Beaulieu, Pelland, & Robertson, 2007). Figure 4-2 plotted the grand ensemble averaged kinetics for BD. With regards to BD kinetics, this study reported power bursts that were similar to those reported by Beaulieu and colleagues (2007). Figures 4-3 and 4-4 reported grand ensemble averaged kinetics for the lead and trail limbs in SBS. Unfortunately, studies that considered SBS kinetics

used an external moment convention (Musselman, 2003; Reid *et al.*, 2007), and thus, comparing the magnitude of peak moments and power bursts is not appropriate.

Discrepancies between previously reported peak moments and powers can be attributed to differences in stair dimensions, staircase design (number of steps) and instrumentation. McFadyen and Winter (1988) used stair dimensions of 22.0 cm rise and 28.5 cm tread and reported ensemble averaged stance phase moments and powers for a single subject over the fourth stair. In contrast, Beaulieu and colleagues reported stance phase moments and powers recorded at the second stair of a three step staircase with dimensions identical to those used in this research. As such, the values reported here more closely reflect those reported by Beaulieu and colleagues (2007) and Reiner *et al.* (2002).

Finally, discrepancies between the aforementioned results and this study can be attributed to the inverse dynamics model used to compute joint moments and subsequently, powers. This study reported results derived from a three-dimensional model, thereby reducing error in flexor/extensor moment and power calculations when outside of the sagittal plane. Alkjaer, Simonsen and Dyrhe-Poulsen (2001) demonstrated joint moment profiles were similar between 2D and 3D models but magnitudes differed between computational approaches. Some previous studies (Beaulieu *et al.*, 2008; McFadyen & Winter, 1988) resolved joint moments and powers onto a sagittal projection and therefore ignored out-of-plane contributions from the flexor/extensor moments and powers (Winter, 2005).

At the ankle joint, BD and SBSL kinetics were less demanding than those of their FD counterpart. While peak plantiflexor moments were not significantly different across

conditions, there were pronounced differences in power and work done by the plantiflexor moment at burst A1. Burst A1 was larger in FD and SBST than both BD and SBSL, where the power and work magnitudes were not significantly different. For instance, the magnitude of burst A1 in FD (marginal mean collapsed across stairs) was approximately 1.7 times greater than both BD and SBSL. Peak power and work done by the plantiflexor moment at burst A1 were not significantly different between BD and SBSL. Similarly, work done by the ankle plantiflexors was larger in FD than either BD or SBSL, where work magnitudes were not different from one another. Since peak moments were not different between conditions, angular velocities were greater in FD and resulted in increased power and work reported for the FD condition. Power is a scalar value derived from the dot product of joint moment and angular velocity, and thus, these parameters are not independent entities. As such, statistical analyses performed on any two of the three parameters are sufficient to draw conclusions regarding the third, without error rate inflation. Thus, given that powers were significantly different between conditions whilst moments were not, then angular velocities were necessarily different.

Moreover, there were condition effects for the peak ankle moments, powers and work done by the ankle plantiflexors at push-off. Peak plantiflexor moments, powers and work magnitudes that were substantially larger in FD and SBST than BD and SBSL conditions. Whereas the magnitude of the moments and powers were not significantly different in SBSL and BD, work done by the plantiflexors was smaller in SBSL. Figures 4-2 and 4-3 show that the difference in work observed in Figure 4-10 was related to timing of the burst A3, which was more prolonged in BD compared to the short burst observed in SBSL. Similarly, peak moments at push-off were not significantly different

between SBST and FD, while plantiflexor power at burst A3 was smaller in SBST. Therefore, angular velocity was larger in FD. Subsequently, there was a difference in work done by the ankle plantiflexor moment, which rendered work done by the plantiflexor moment in FD larger than that done by the plantiflexor moment of the SBST limb.

Finally, the current study considered differences in the peak moment, power and work done by the ankle plantiflexors in midstance, when the ankle acted in conjunction with the ipsilateral knee to control body lowering to the next step. Figure 4-11 demonstrated that peak ankle plantiflexor moments, powers and work done by the plantiflexor moment at burst A2 for FD were not different from those observed in SBST, but were significantly greater than SBSL and BD. The peak plantiflexor moments and powers were not significantly different in the BD and SBSL conditions. The pronounced differences in work done by the plantiflexor moment in midstance were therefore related to differences in the length of burst A2: as demonstrated in Figures 4-2 and 4-3, burst A2 was substantially longer in BD relative to SBSL and consequently, led to more work being done by the plantiflexor moment in BD (Figure 4-11). Figure 4-11 revealed peak powers were not different between BD and SBSL conditions. However, the magnitude of burst A1 was elevated for a prolonged duration in BD relative to the short burst demonstrated in SBSL.

The results showed conclusively that BD and SBSL kinetics successfully reduced the demand of stair descent at the ankle. Reductions in peak power and work done by the plantiflexor moment at burst A1, peak moments, powers and work done by the plantiflexors in midstance (A2) and at push-off (A3) demonstrated that substantially

lower forces were recruited from the plantiflexors at reduced rates in BD and the SBSL limb. Furthermore, the mechanical work or overall demand imposed by the descent task was considerably lower in BD and the SBSL limb relative to FD and SBST. These effects were witnessed for the lead limb in the step-by-step descent pattern only, but were bilateral in the BD condition.

**Knee kinetics.** Table 4-5 presented summary statistics for the magnitudes of peak knee moments, powers and mechanical work done at the knee during the burst K3. In the FD and SBST conditions, the peak extensor moment was largest at step 2, decreased at step 3 and remained constant thereafter. Conversely, in BD the trend was opposite: the peak knee extensor moment, power and work done by the knee extensor moment were smallest at step 2, increased at step 3 and remained close to constant at steps 4 and 5. In SBSL, the peak knee moment reversed polarity at steps 2 and 4, resulting in a net flexor moment that did positive work. The overall reduction in knee moment and power magnitudes reiterates the role of the lead limb as a 'resting limb' in SBSD (Musselman, 2003; Reid *et al.*, 2007). These studies considered the knee forces, moments and powers in stair descent and appropriately labelled the lead limb the resting limb in SBSD since moment and power magnitudes were significantly reduced relative to the SBST limb and both limbs in FD. As such, the SBSL knee extensor moment supports the body but does not lower the body from one step to the next. Controlled lowering is performed by the SBST limb knee extensor moment, which led the authors to label the trail limb the 'working limb' in step-by-step stair descent (Musselman, 2003; Reid *et al.*, 2007).

While peak knee moments were smallest in the SBSL condition, there were no significant differences between SBST and FD knee extensor moments. The condition

main effect was preserved for peak knee extensor power and work done at the knee during the eccentric burst, K3. Peak knee extensor powers were greatest in FD and smaller in SBSL than other conditions. The mean knee extensor powers (across stairs; marginal mean for condition) were 2.3 and 2.1 times greater in FD and SBST than those demonstrated in BD, respectively. Similarly, work done by the knee extensors was 1.7 times greater in FD and 2.1 times greater in SBST than that done in BD. These data show the overall energy dissipated by the knee extensors was significantly reduced in BD compared to FD and the SBST limb. In contrast, the absolute magnitude of the peak powers and work done at the knee were substantially lower at SBSL limb relative to BD.

In terms of the statistical interaction, in FD and SBST peak knee powers and work done at the knee were largest at step 2, which represented the transition from quiet stance to stair descent and decreased in magnitude thereafter. In BD, the trend was reversed: peak moments, powers and work done by the knee extensor moment were smaller at step 2, increased at step 3 and remained constant at steps 4 and 5. Figure 4-12 demonstrated that peak knee extensor moments and powers stabilized following one complete gait cycle. Further, peak knee moments and powers were larger in FD than alternate gait patterns. As such, compensatory strategies reduced the demand at the knee during stair descent. However, the overall patterns of net moments and powers were idiosyncratic; they depended on both condition and stair. This was illustrated with BD, where peak knee moments and powers increased at step 2 but decreased thereafter. Since the peak knee extensor moment, power and work done by the extensor moment were significantly reduced in BD relative to FD, BD is recommended as an alternative for individuals with

knee extensor weakness. While similar results were observed in SBSL, the reduction only occurred for the lead limb.

In summary, the knee moment, power and work were significantly lower for the SBSL compared to all other conditions. Therefore, BD might represent a more feasible strategy in individuals with bilateral knee extensor weakness, whereas SBSL with the affected limb serving as the lead or resting limb would be more appropriate for individuals with unilateral knee problems.

**Hip kinetics.** Differences in peak power and work magnitudes were drastic between FD and BD, with power and work done by the hip extensor moment in BD reaching 5.6 and 22.5 times greater than those of the flexor moment in FD, respectively. BD reduces loading at the ankle and knee joints but displaces loading to the hip. As such, BD might not be a feasible strategy when hip extensor strength is compromised, as is common with degenerative neural pathologies that include motor neuron deficits (Perry & Clarke, 1997). Hip extensor strength is a key determinant of walking ability (Salsich & Mueller, 1997; Burnfield, Josephson, Powers & Rubenstein, 2000) and the ability to maintain balance in upright stance (Duncan, Weiner, Chandler & Studenski, 1990).

In level walking, compensation for hip extensor weakness occurs early in the stance phase, where the hip extensors of normal subjects counteract the external flexion moment that arises from the GRF falling anterior to the hip joint. With weak or paralyzed hip extensors, the individual may compensate by moving the trunk backwards so that the GRF's line of action is projected posterior to the hip joint flexion/ extension axis (Whittle, 2007). Mechanically, the compensatory strategy either reduces the moment arm of the GRF (if the line of action is anterior to the hip joint centre) and reduces loading of

the extensors, or reverses the polarity of the external moment (should the GRF fall posterior to the hip joint axis) (Perry & Clarke, 1997). As previously mentioned, BD requires eccentric hip extensor activity to control flexion that arises from the GRF in stance. Therefore, BD might not be a feasible strategy for individuals with pronounced hip extensor weakness since the compensatory trunk strategy, if enacted, would impart posterior instability. Subsequent research should examine the implications of BD in clinical populations.

### **Progression to Steady-state Stair Descent**

This study examined joint moments, powers and work over four consecutive gait cycles to determine whether there was a transition effect when initiating stair descent from upright stance. Further, the experimental design permitted contrasts across conditions including forwards, backwards and step-by-step stair gait to determine whether the transition effect was similar across strategies. Whereas peak powers and work done by the ankle plantiflexors at burst A1 remained relatively stable between stairs in the BD and SBSL conditions, the moment, power and mechanical work magnitudes increased systematically across stairs in FD. The plantiflexor power at burst A1 saw a 30% increase in magnitude over the first to fourth gait cycles, whereas the corresponding work done by the plantiflexor moment increased 33%. Conversely, Figure 4-9 demonstrated ankle moments, powers and mechanical work were not different across gait cycles in BD and SBSL. As such, ankle moment, power and work magnitudes were steady for the entire descent task.

Similarly, Figures 4-10 and 4-11 demonstrated that ankle moments, powers and work done in midstance (A2) and push-off (A3) were consistent across steps. There was

an interaction for the peak ankle moments at burst A2 with moments being smaller over the first gait cycle in FD and SBST and stabilized thereafter. Differences in angular velocity rendered it such that power and work were constant across gait cycles and conditions.

Table 4-5 and Figure 4-12 revealed a significant stair  $\times$  condition interaction for the peak moments, powers and work done at the knee during mid- to terminal stance (burst K3). As such, knee extensor moments were observed to be greater at step 2 relative to other stairs in the FD and SBST conditions, decreased in magnitude at step 3 and remained constant. The magnitude of burst K3 and work done by the knee extensors was not different across gait cycles. In BD, the trend was opposite. In BD, peak power at burst K3 and work done by the knee extensors was smaller in transition, increased into the midstair region and stabilized thereafter. In the SBSL condition, the kinetics involved a dissipative knee extensor moment that did negative work in transition (stairs 2 and 5). Conversely, in the midstair region, the net moment was a flexor moment that did positive work to flex the knee in terminal stance. As such, the results suggest participants employed a strategy whereby the demand at the knee was minimized.

Powers and work done by the hip flexor moment at burst H2 were near constant across gait cycles in FD. In SBSL, there were no reproducible patterns observed at the hip, except the generative hip flexor burst in terminal stance (H3). Mechanically, the SBSL pattern isolates both the lead limb knee and hip joints by landing near full extension. As such, the GRF force passes close to the joint centres so that corresponding moment arms are reduced. Accordingly, motion about the SBSL knee and hip joints is minimized and these joints impart a passive stability. Relatively minor changes in how

the GRF passes with respect to the joint center could reverse polarity of the external moment and consequently, internal joint moments that prevent collapse of the lower extremity in stance. In short, task mechanics in SBSL rendered it such that the hip contributed very little to controlled lowering in FD. Inconsistency in the moment and power profiles may be further attributed to task unfamiliarity among participants. Despite partaking in a familiarization period, SBSL was a novel and unnatural task for healthy individuals that were absent of musculoskeletal disorders.

Figure 4-17 showed maximum velocity parallel to the plane of progression across gait cycles for all experimental conditions. Velocities were markedly higher in FD compared to other conditions. There were no obvious differences in velocity between BD, SBST and SBSL. In terms of the interaction, velocity increased approximately monotonically in FD, suggesting that individuals approached but did not reach a steady-state progression velocity in FD. Similarly, there was a trend towards increased velocity in BD, SBSL and SBST conditions but these trend failed to reach significance. Increased velocity over successive gait cycles might seem counterintuitive for the step-by-step gait pattern given the individual appears stationary during tandem stance. However, this study was intended to be a control research to investigate the mechanics of alternate descent patterns in healthy participants. As such, extraneous variables including pathology and age were eliminated so that observed effects would precipitate solely from task mechanics rather than secondary compensatory mechanisms. Given the movement was inherently slow and unnatural for these healthy participants; they tended to increase velocity through the staircase. As such, the participants tended to carry forward momentum through to the next step, which served to increase progression velocity.

However, inter-subject variability was greater in these conditions and consequently, this trend failed to reach significance. Consequently, participants effectively attained steady-state progression in these alternate descent patterns.

Correlation analyses were performed on the peak ankle (burst A1), knee, and hip powers (burst K3) observed in FD since vertical GRF in stair descent (Protopapadaki, Dreschler, Cramp, Coutts & Scott, 2007) and joint moments and powers in level walking (Winter, 1983) are proportional to progression velocity. The magnitude of the dissipative ankle burst A1 and work done by the ankle plantiflexors at initial-contact decreased linearly (increased in absolute magnitude) as the individual progressed toward a steady-state velocity in FD. However, a similar trend was absent at the knee and hip joints since correlation analyses revealed no discernible relationships between maximum progression velocity and the magnitudes of the knee burst K3 or hip burst H2. Therefore, the ankle plantiflexors appeared to act as a 'sink' for energy dissipation in FD. The outlined strategy effectively minimized loading at the knee and hip joints with increased velocity in FD. Given the concatenation of the lower extremity, the joints of both limbs were activated in a sequence directed toward shared energy absorption; dissipation was relegated to the ankle plantiflexors rather than the knee extensor or hip flexor moments.

The outlined strategy is consistent with a postulate by Reiner *et al* (2002). These researchers examined the influence of staircase inclination on joint moments and powers in stair ascent and descent. In descent, the authors reported a 67.3% increase in plantiflexor power at initial-contact (burst A1) with increased inclination. In contrast, absolute magnitudes of knee extensor and hip flexor powers increased only 26.7% and 24.3%, respectively. Consequently, the authors reasoned that eccentric plantiflexor

activity is directed toward distributed energy dissipation between segments. Landing with the ankle plantiflexed affords a considerable range for dorsiflexion, while the plantiflexors act eccentrically (Reiner *et al.*, 2002). In this study, the magnitude of burst A1 was observed to increase across gait cycles, as participants progressed down the stairs with increasing pace.

Conclusive evidence regarding neuromuscular control mechanisms in gait is scant (Pedotti, 1977; Dietz, 1997). The results presented here suggest that a realistic control strategy for stair descent might be to minimize knee and hip powers, and subsequently, energy dissipated by the knee extensor and hip flexor moments. The minimization of knee and hip powers might be consistent with an optimal control approach to biomechanical modeling. Optimization research assumes that the central nervous system attempts to minimize loading on key parameters underlying movement (Todorov, 2004). Subsequently, task goals are quantified with respect to a cost function, where deviation from the optimality criterion is penalized accordingly (Todorov, 2004). Models that incorporate optimization have a longstanding history in the literature. Motor control research has contemplated control laws for the formation of limb trajectories in reaching tasks (Hogan, 1984; Flash and Hogan, 1985; Todorov and Jordan, 1998; Smeets and Brenner, 1999; Uno, Kawato and Suzuki, 1989). In the biomechanical literature and locomotion studies in particular, the majority of models strive to minimize muscle energy expenditure (Hatze and Buys, 1977; Davy and Audu, 1987; Collins, 1995; Anderson and Pandy, 2001). As previously outlined, the cost function or optimality criterion for stair descent might be to minimize energy dissipated by eccentric activity of the knee and hip musculature. Forwards dynamical simulations that employ the load minimization strategy

might replicate the current results and further consolidate the role of the ankle as a compensatory sink for energy dissipation. Subsequent researches should determine whether similar strategies are employed by the elderly and clinical populations in stair descent.

Figure 4-12 demonstrated that BD involved a different strategy compared to FD. In BD, the magnitude of knee burst K3 and consequently, energy dissipated at the knee joint increased as the individual descended the stairs. This trend was also evident in Figure 4-12, where the magnitude of eccentric knee power increased from step 2 to step 3. Figure 4-18 demonstrated that the magnitude of the eccentric ankle plantiflexor power burst A1 was not correlated to progression velocity, nor was eccentric activity of the hip extensors at burst H2. As a result, the outlined strategy might be moderated by two factors. First, BD reduced power and work done by the knee extensor moment during the dissipative stance phase burst K3, relative to FD. As such, the magnitude of the knee extensor power might not be minimized. That is, given the power of the knee extensor moment was reduced in BD, the system or underlying musculature is more tolerant to increased knee power. Conversely, a secondary explanation is contingent upon task mechanics. In BD, the trunk is projected forward over the staircase throughout the gait cycle, presumably to minimize momentum in the direction of progression down the staircase and stabilize the body. The result of projecting the trunk over the staircase is to displace the CoP away from the stair edge and over the staircase (Beaulieu, 2004). Further research should determine whether the trunk angle and subsequently, horizontal displacement of the CoM (and thus, CoP) are moderated by a stair effect so that they decrease as the individual progresses down the staircase. If a tangible stair effect existed and the trunk flexion angle

was inversely proportional to progression velocity, then the external flexor moment caused by the ground reaction force (GRF) would increase. In BD, the ground reaction force vector passes posterior to the knee joint centre during the stance phase and thus, tends to flex the knee. Support in this case requires increased involvement from the internal knee extensor moment. As demonstrated in Figure 4-12, the magnitude of the knee extensor moments, eccentric power burst K3, and work done by the knee extensor moment increased into the midstair region (at stair 3) and remained relatively stable thereafter. Two factors may explain the observed effect: the magnitude of the ground reaction force likely paralleled increased velocity. Alternately, the forward lean of the trunk may have decreased with increased velocity, which would consequently increase the perpendicular distance between the knee joint centre and GRF and thus, the external knee flexor moment caused by the GRF. Further research should consider the legitimacy of both explanations. Regardless, the strategy employed in BD did not result in minimized knee power magnitudes, and as such, represented a different strategy with respect to FD.

### **Transition Effect in Stair Descent**

The only research to consider the transition from initiation to steady-state progression in stair descent was performed by Andriacchi *et al.* (1980). With only a three-step staircase, the authors attempted to differentiate kinetic parameters in stair descent from the transition to level walking. In short, their protocol may have misrepresented the transition from standing to stair descent: participants transitioned from standing to stair-descent to level-walking within two gait cycles, with each of these tasks reflecting different mechanical demands. Further, peak stance phase moments were reported for gait cycles

that involved descending from stair 3 (top) to stair 1 (bottom) and stair 2 (middle) to the floor. By definition, transition represents the “passage from one state [...] to another” or alternately, “a movement, development, or evolution from one form, stage, or style to another” (Gove, 1986). As such, the transition from standing to stair descent would necessarily involve discerning kinetic parameters in the first gait cycle from standing at stair 3 to stair descent at stair 2, but was not considered by the researchers. No further research contemplated the transition phase in stair descent. Therefore, an objective of this research was to determine whether a transition effect occurred from standing to stair descent that was consistent across subjects.

The peak dissipative knee moment in mid- through terminal stance showed apparent transition effects in FD, SBST and BD. In the FD and SBST conditions, the eccentric knee extensor moment tended to be greater in transition, at stair 2, whereas moment powers and work done by the knee extensor moment to dissipate gravitational energy remained relatively constant across gait cycles, reflecting reduced angular velocity in the transition from standing to stair descent for FD. Similarly, the hip flexor moment, eccentric power burst H2 and work done by the hip flexor moment in FD tended to be greatest at step 2, which represented the first gait cycle. As such, the forces required from the knee extensor and hip flexor moment were greatest at the top of the staircase. Moreover, the magnitude of the eccentric power burst H2 implies that force was required at a greater rate relative to other gait cycles, culminating in an overall increase in energy dissipated by the hip flexor moment.

The kinetics of the knee and hip joints in FD, and the knee joints of the SBSL and SBST limbs stabilized following one complete gait cycle, which represented the

transition from quiet stance to stair descent. Except in the SBSL condition, there were no apparent transition effects into level-walking. In the SBSL condition, the net knee moment reversed polarity, changing from a flexor moment to an extensor moment at the transition to level-walking. The outlined effect likely precipitated from the mechanics of SBSL. In SBSL, the limb contacts the stair with the knee joint near full extension, with that GRF passing very near to the joint centre and the moment arm of the GRF being reduced. Therefore, only a small deviation from full extension could cause the net moment to become flexor and subsequently, to do positive work to flex the limb in the pull-up phase.

In BD, there was an apparent transition effect for the stance phase knee kinetics, though the observed effect was opposite relative to that described for FD. As previously outlined, magnitudes of the knee extensor moment, power and work at burst K3 were reduced at stair 2, increased at stair 3 and decreased thereafter (stairs 4 and 5). This effect likely resulted from the overall reduction in knee extensor moment, power and work magnitudes with respect to FD and SBST conditions. However, similar to FD, the absolute magnitude of the hip moment at burst H2 was greatest at stair 2, decreased over the second gait cycle and increased thereafter. Similar trends were observed for the magnitude of hip extensor power. However, work done by the hip extensor moment was variable, but remained relatively constant across conditions.

The magnitude of the hip extensor moment and power were greatest in transition to standing to stair descent at stair 2 and the transition from stair descent to level-walking at stair 5. Again, these results imply that transition involves different mechanics with respect to steady-state descent. that imposes greater force demands on the knee extensors

(knee extensors in FD, SBSL, SBST) and hip musculature (flexors in FD, extensors in BD).

Previous research examined the initiation of level walking and reported that steady-state progression velocity is attained at the end of the first step (Breniere and Do, 1986), whereas others concluded, on the basis of kinetic and mechanical energy analyses, that steady-state level walking is not attained until the end of the third step (Mann, Hagy, White, and Liddell, 1979; Miller and Verstraete, 1996; Miller and Verstraete, 1999). This study revealed that ankle, knee and hip parameters reached a steady-state following the first gait cycle in BD. Similar results were observed for ankle and knee parameters in the lead and trail limbs of SBSL and knee and hip parameters in FD. However, there were pronounced differences in the magnitudes of ankle powers in FD, suggesting that four gait cycles was not sufficient to attain steady-state progression in this mode, since ankle power increased with progression velocity down the staircase.

Therefore, the present results refute the results of Andriacchi and colleagues (1980) by demonstrating that peak ankle plantiflexor power and work at initial-contact increased with increased progression velocity in step-over FD. Power and work magnitudes reported here were not steady-state following a single gait cycle in FD; they increased through the staircase such that they paralleled increases in progression velocity. Unfortunately, laboratory constraints prohibited a larger stair apparatus. Despite quantifying kinetic parameters for an ample number of gait cycles to improve the current understanding of stair descent, the staircase was not sufficient to reach a steady-state progression in FD. Conversely, individuals converged to steady-state progression in alternate stair gait patterns, including BD, SBSL and SBST.

Future research should consider the transition strategies employed by the elderly and individuals with lower extremity musculoskeletal conditions that compromise the force-generating capacities of the knee and hip joint musculature. To employ a strategy similar to that demonstrated here by healthy participants would further jeopardize safety in stair descent by having these individuals work near maximal capacity. Previous research speculated that the descent task is difficult for the elderly due to strength reductions in the quadriceps musculature, which led to them to work at an increased relative proportion of their maximal knee extensor moment capacity (Hortobagyi, Mizelle, Beam & Devita, 2003; Reeves, Spanjaard, Mohagheghi, Baltzopoulos & Magnaris, 2008). As such, the reported transition effects and descent strategies outlined here might not represent a feasible alternative and is probably not used by elderly individuals. The transition from upright stance to stair descent appears to impart some increased difficulty for the elderly, since these individuals employ more cautious behaviour including handrail usage and hesitation (Hamel & Cavanagh, 2004). Further research should examine whether the cautious behaviour demonstrated by the elderly in transition reflects mechanical or psychological factors, or both.

## SUMMARY AND CONCLUSIONS

### Summary

The purpose of this thesis was to determine the mechanisms by which alternate descent strategies (backwards; forwards step-by-step) modulate task difficulty. Secondly, this study sought to determine whether foot clearance was greater in alternate descent patterns and thereby minimized the likelihood of inadvertently contacting a stair edge in the swing phase of descent. Seventeen healthy males ( $n = 9$ ) and females ( $n = 8$ ) participated in the study.

The results demonstrated that alternate descent patterns mitigate task demand at the ankle and knee joints since joint power and work magnitudes were reduced relative to forwards step-over-step descent. Stair effects were overridden by stair  $\times$  condition interaction effects. Whereas these effects were prominent for only the lead leg in step-by-step descent, they were bilateral in the backwards condition. Thus, backwards descent reduced loading bilaterally at the ankle and knee joints but displaced loading to the hips.

Joint mechanics were also contrasted across gait cycles. There were obvious transition effects at the ankle, knee and hip joints, since moments tended to be greatest when initiating stair descent. Similar results were observed for knee and hip kinetics in FD. However, in FD there were pronounced increases in ankle power magnitudes that paralleled progression velocity, suggesting the ankle acts as a mechanism for shared energy absorption. Correlation analyses reiterated these results by demonstrating there was a relationship between ankle power magnitudes at burst A1 and progression velocity

in FD, but no significant relationships between hip and knee power magnitudes and progression velocity. There were no significant correlations between ankle (A1), hip (H2) or knee (K3) power magnitudes when descending backwards, which suggests the strategy was different in BD.

Finally, this study examined foot clearance in stair descent. While foot clearance was significantly greater in BD at stairs 2 through 5 relative to the SBSL and SBST limbs and FD, our results show the likelihood of contacting a stair edge in either pattern was nonexistent for these healthy participants since no trips, stumbles or falls were recorded. There were apparent transition effects in forwards and backwards descent, with clearance being smallest in transition to descent, whereas clearance was consistent across stairs for both the SBSL and SBST limbs.

The present results have implications in the rehabilitation sciences to improve functionality for elderly or patient groups that experience difficulty or pain during stair descent. Future research should consider the effects of and the feasibility of backwards descent for these individuals. Moreover, studies will likely consider initiation mechanisms employed by elderly and clinical populations when transitioning to stair descent from quiet stance. These studies will offer insight regarding strategies used to minimize mechanical burden experienced by the joints and overlying musculature.

## Conclusions

The following conclusions are supported by the results:

1. BD and SBSL kinetics successfully mitigate the demand of the descent task at the ankle. These effects were witnessed for the lead limb in the step-by-step descent pattern, but were bilateral in the BD condition.
2. Peak knee moments were smallest in the SBSL condition. There were no apparent differences between those associated with SBST and FD.
3. Peak knee extensor powers were greatest in FD and smaller in SBSL than other conditions. In BD, peak powers were smaller than those observed at the SBST knee.
4. Work done by the knee extensor moment at burst K3 was greatest at the SBST limb. Work done in FD was also larger than that done in BD. In contrast, the peak powers and work done at the knee were substantially lower in SBSL relative to BD.
5. The hip moment was larger in FD than BD. Differences in peak power and work magnitudes were drastic, with power and work done by the hip extensor moment was larger in BD than those of the flexor moment in FD.
6. The magnitude of the plantiflexor burst A1 increased monotonically over gait cycles in FD. Similarly, work done by the plantiflexor moment at initial-contact increased through the staircase. Ankle moments, powers and work done were not discernibly different across gait cycles in BD and SBSL. As such, there was no apparent transition effect so that kinetic parameters at initial contact were steady for the entire descent task in the alternate patterns.

7. Ankle moments, powers and work done in midstance (A2) and push-off (A3) saw no apparent transition effect.
8. Knee extensor moments were greatest in transition in the FD and SBSL conditions, decreased in magnitude through the midstair region and stabilized. The magnitude of burst K3 and work done by the knee extensors was not different across gait cycles. In BD, the trend was opposite. In BD, peak power at burst K3 and work done by the knee extensors was slightly smaller in transition, increased into the midstair region and stabilized thereafter.
9. Peak hip moments, powers and work done during the burst H2 were stable across gait cycles and showed no obvious transition effect.
10. Velocities were markedly higher in FD compared to other conditions. There were no obvious differences in velocity between BD, SBST and SBSL. Velocity increased approximately linearly through the staircase in FD, suggesting that individuals did not reach a steady-state progression velocity on the laboratory stair apparatus.
11. Correlation analyses demonstrated that the ankle plantiflexors appeared to act as a 'sink' for energy dissipation in FD since plantiflexor power magnitudes paralleled increases in progression velocity. Similar relationships were not present for dissipative knee (K3) or hip power bursts (H2). The outlined strategy effectively minimized loading at the knee and hip joints with increased velocity in FD. A similar strategy was not present in BD since dissipative ankle (A1), knee (K3) and hip (H2) bursts were not correlated to progression velocity.

12. The present results suggest Andriacchi and colleagues (1980) misrepresent the transition to steady-state descent by demonstrating the peak ankle plantiflexor power and work done by the plantiflexors at initial-contact encounter increases that parallel increases in progression velocity. In effect, power and work magnitudes are not steady-state following a single gait cycle in FD, whereas they approach steady-state following a single cycle in the alternate descent patterns.
13. Foot clearance was markedly smaller in transition at stair 1 and increased at stair 2 in FD and BD. Foot clearance decreased through the midstair region and increased in the transition to level walking. These results reflect methodological differences between this study and Hamel and colleagues (2005), where clearance was reported to be largest at the top of the staircase. However, similar trend were observed in the midstair region where values were observed to decrease progressively.
14. Clearance was markedly lower for both lead and trail limbs in SBSB relative to FD and BD. Clearance was similar between lead and trail limbs in the SBSB pattern.

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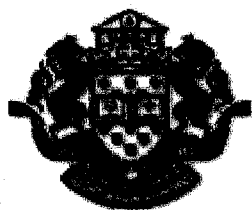
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## **Appendix A**

### **Informed Consent Documentation**



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## **INFORMATION AND CONSENT DOCUMENT**

### **BIOMECHANICAL ANALYSIS OF STAIR ASCENT AND DESCENT**

#### **GAIT LABORATORY, UNIVERSITY OF OTTAWA**

**INVESTIGATORS :** D. Gordon E. Robertson PhD FCSB  
Faculty of Health Sciences, University of Ottawa  
125 University, Ottawa, Ontario, K1N 6N5  
Tel.: 613-562-5800 extension 4246  
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François G. D. Beaulieu MA  
Faculty of Health Sciences, University of Ottawa  
125 University, Ottawa, Ontario, K1N 6N5  
Tel.: 613-562-5800 extension 2923  
E-mail: fdbeauli@uottawa.ca

#### **STATEMENT OF INVITATION**

You are invited to participate in a research project which is a collaborative effort between researchers in the School of Human Kinetics at the University of Ottawa. This research project is conducted by the two investigators listed above. We greatly appreciate your interest in our work.

#### **PURPOSE OF THE STUDY**

The goal of this study is to gain a better understanding of the three-dimensional mechanics underlying several variants of stair negotiation, including backwards stair descent. The ability to maintain balance during stair descent is a prerequisite to safe and independent mobility necessary for independent living at home and for community participation. This mobility activity is often challenged in adults with physical and neurological impairments as well as under various functional constraints, including weight carriage. Our interest is founded by previous work conducted in our laboratory which suggests that this form of locomotion may decrease the risk of falling while descending stairs. We are also interested in determining the mechanisms that predispose

individuals to falls in stair negotiation, particularly within the first step of initiating stair gait.

#### **YOUR PARTICIPATION IN THIS STUDY INVOLVES:**

1. Attend a detailed information session prior to the experiment.
2. Providing informed consent prior to the experimental session.
3. Reflective markers will be affixed to several body sites using a hypoallergenic two-way adhesive tape. These markers enable segmental displacement data to be recorded during stair ascent/descent. The data will be captured using several infrared cameras and analyzed with the VICON system.
4. The experimental trials will be performed on a five step wooden staircase constructed in the Biomechanics Laboratory at the School of Human Kinetics (Montpetit Hall, Room 319). The staircase will be instrumented with 3 to 5 force platforms, which will permit the researchers to quantify the forces produced about the joints of the lower extremity during the experimental trials.
5. You will receive instructions detailing the manner in which you will ascend or descend the staircase and a metronome will ensure that the pacing of ascent/descent remains similar between trials. A familiarization period will be allotted to ensure that you are comfortable with the experimental tasks. A research assistant will be within close proximity at all times during the experiment to provide assistance in the event that you lose your balance.
6. Performing the following tasks for data acquisition:
  - You will be asked to perform 5 trials of stair negotiation under the following experimental conditions: a) forwards ascent, b) forwards descent, and c) backwards descent. You will also be asked to perform all conditions while wearing a weighted bag, for a total of 30 experimental trials.

Note: Each session will last approximately 90 minutes. You will perform a total of 10 stair ascents and 20 stair descents. Rest periods will be provided as necessary.

#### **RISKS AND DISCOMFORTS**

The repetitive nature of the experimental tasks may evoke discomfort of the muscles or hip, knee, and ankle joints of the lower extremities. However, since rest periods will be allotted as needed, this discomfort will be minimized. Further, there is remote chance of a fall due to an error in stepping. In the event that a fall occurs, a handrail attached to the staircase apparatus and two spotters will ensure that the magnitude of the fall is not serious. Rest assured that several studies have considered stair negotiation within children and adults without any injuries. During the protocol you may feel winded from repeated stair ascent and descent trials. Be sure to inform the researcher of any discomfort that you experience so that sufficient rest periods are allotted.

## **BENEFITS**

There are no personal benefits to be derived from participating in this study. The information that we obtain will help us further our understanding of the mechanics of stair negotiation as well as outline the mechanisms that increase the risk of incurring a fall while negotiating stairs. It is hoped that through such studies, we will be able to effectively identify the factors underlying stair-related falls to improve the safety for a client.

## **EMERGENCY SITUATIONS**

In the event that a fall or cardiorespiratory problem occurs during the experimental session, the researchers will ensure that you remain safe and comfortable while briefly waiting for additional help. The University of Ottawa Emergency Services will be contacted from the laboratory telephone immediately, and if necessary, the Ottawa emergency services personnel (911) will be contacted.

## **CONFIDENTIALITY**

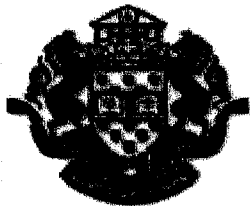
Confidentiality of your records will be maintained at all times, by using an alpha-numeric identification code rather than your initials during data analysis and in reporting the results of this study. **No personal identification will be used in publications of any form pertaining to this study.**

## **INQUIRIES CONCERNING THIS STUDY**

This research project has been approved by the Research Ethics Board at the University of Ottawa. For questions related to the ethical approval of this study, you may contact *Ms. Lise Frigault at (613) 562-5800, extension 1787 (e-mail: lfrigault@uottawa.ca)*, Protocol Officer for Ethics in Research at the University of Ottawa.

If you require further information concerning the study (experimental procedures or other details), please do not hesitate to contact *Dr. Gordon Robertson and M. Francois Beaulieu* at the number or address listed at the beginning of this document.

A copy of this form will be given to you before the end of the experimental session.



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## CONSENT

I, \_\_\_\_\_, AGREE TO VOLUNTARILY PARTICIPATE IN THE STUDY DESCRIBED ABOVE ABOUT THE THREE-DIMENSIONAL MECHANICS OF STAIR NEGOTIATION.

I HAVE RECEIVED AND READ A DETAILED DESCRIPTION OF THE EXPERIMENTAL PROTOCOL. I AM COMPLETELY SATISFIED WITH THE EXPLANATION GIVEN TO ME REGARDING THE NATURE OF THIS RESEARCH PROJECT, INCLUDING THE POTENTIAL RISKS AND DISCOMFORTS RELATED TO MY PARTICIPATION IN THIS STUDY.

I AM AWARE THAT I HAVE THE RIGHT TO WITHDRAW MY CONSENT AND DISCONTINUE MY PARTICIPATION AT ANY TIME WITHOUT ANY PREJUDICES.

## SIGNATURES

### SUBJECT

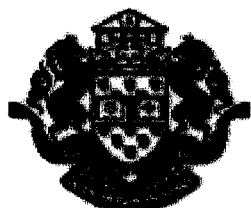
\_\_\_\_\_ (signature) \_\_\_\_\_ (print name)

Date: \_\_\_\_\_

### WITNESS

\_\_\_\_\_ (signature) \_\_\_\_\_ (print name)

Date: \_\_\_\_\_



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## INFORMATION ET FORMULE DE CONSENTEMENT

### ANALYSE BIOMÉCANIQUE DE LA MONTÉ ET DESCENTE D'ESCALIER

#### UNIVERSITÉ D'OTTAWA, "GAIT LABORATORY"

#### CHERCHEURS:

D. Gordon E. Robertson PhD FCSB

Faculty of Health Sciences, University of Ottawa  
125 University, Ottawa, Ontario, K1N 6N5  
Tel.: 613-562-5800 extension 4246  
E-mail: dger@uottawa.ca

François G. D. Beaulieu MA  
Faculty of Health Sciences, University of Ottawa  
125 University, Ottawa, Ontario, K1N 6N5  
Tel.: 613-562-5800 extension 2923  
E-mail: fdbeauli@uottawa.ca

#### INVITATION

Vous êtes invité à participer à un projet de recherche au département d'activité physique de l'Université d'Ottawa. Ce projet est sous la supervision des deux chercheurs ci-haut mentionnés. Nous vous remercions pour l'intérêt que vous portez à ce projet.

#### PROPOSITION DE LA RECHERCHE

Le but de cette recherche est d'obtenir une meilleure compréhension de l'étude mécanique à 3 dimensions des variations de l'utilisation de descente et monté des escaliers. La capacité de maintenir son équilibre lors d'utilisation d'escaliers prédestine l'indépendance d'une personne à rester autonome. Cette mobilité individuelle est souvent mise en contexte lors de difficulté physique ou neurologique de même que lorsque certaines contraintes fonctionnelles sont appliquées tel que le transport d'objets et blessures. Poussé par les précédentes recherches effectuées dans notre laboratoire, nous nous intéressons dans la réduction des risques d'accident lors de la descente d'escalier. Nous nous intéressons aussi dans la détermination des mécanismes qui prédisposent les chutes plus spécifiquement lors de l'amorce de la descente d'escalier. L'information

acquise lors de cette recherche nous fournira une meilleure compréhension des facteurs qui contribuent aux risques de chutes lors de la descente d'escaliers.

#### **CE QUE L'ON ATTEND DE VOUS POUR CETTE ÉTUDE**

7. Obtenir un consentement informé avant la session expérimentale
8. Signer le formulaire de consentement avant l'expérimentation.
9. Des marqueurs réfléchissants autocollants seront fixés sur votre corps à de multiples endroits. Ceci nous permettra d'enregistrer les données de position durant votre descente. Les données seront captées par plusieurs caméras et analysées avec le système «VICON».
10. Pour l'expérience, nous utiliserons un module d'escalier de 5 marches fabriqué dans le laboratoire de biomécanique à l'École des sciences de l'activité physique (Pavillon Monpetit, Salle 319). De trois à cinq plates-formes de force sont intégrées sur les marches. Ces plates-formes de forces nous permettront de mesurer la force que vous appliquerez sur vos membres inférieurs lors de la monte ou descente d'escalier.
11. Vous recevrez des directives sur la façon de descendre l'escalier par-devant et par derrière et un métronome assurera le maintien du rythme des descentes. Vous pourrez pratiquer jusqu'à ce que vous vous sentiez confortable avec l'exécution de la tâche expérimentale. Un assistant de recherche sera près de vous en tout temps pour vous venir en aide si vous perdez l'équilibre.
12. Effectuer les tâches suivantes pour l'acquisition de données:  
 Vous aurez à déambuler une série de marche d'escalier (longueur 20cm, hauteur 30cm) un minimum de 5 fois dans les conditions suivantes: **a)** monté de face, **b)** descente de face et **c)** descente à reculons. Par la suite, vous aurez à effectuer de nouveau ces mêmes tâches avec une charge supplémentaire (poids).

Note: Chaque session sera de 90 minutes environ. Des périodes de repos et d'adaptation seront fournies si nécessaire.

#### **RISQUES ET INCONFORTS**

La nature répétitive de la tâche peut occasionner un endolorissement musculaire ainsi que de la douleur à la hanche, au genou ou à la cheville. Par contre, comme des périodes de repos seront permises au besoin, les effets secondaires devraient être minimes. Il existe aussi la possibilité d'entorse à la cheville ou de chute causée par un faux pas durant la descente d'escalier. Un assistant de recherche sera présent en tout temps pour vous assister lors de chaque descente. Veuillez noter que plusieurs autres études sur la descente d'escalier furent réalisées chez les enfants ainsi que chez les adultes sans aucune blessure. Vous allez peut être vous sentir un peu essoufflé lors de la montée et de la descente d'escalier. N'hésitez pas de communiquer tout inconfort au chercheur; vous aurez la possibilité de vous arrêter pour vous reposer si vous ressentez certains malaises ou inconforts.

## AVANTAGES

Il n'y a aucun gain personnel qui découle de votre participation à cette étude. Cependant, l'information que nous obtiendrons nous servira à comprendre le maintien de l'équilibre et ce qui prédestine les chutes lorsqu'une personne descend un escalier. Nous espérons ainsi identifier plus efficacement les limites de sécurité chez une personne.

## ANONYMAT ET CONFIDENTIALITÉ

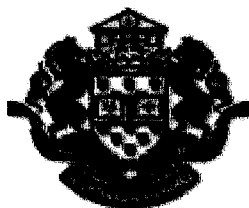
Votre anonymat comme participant sera maintenu en tout temps en utilisant un numéro d'identification au lieu de vos initiales lors de l'analyse des données et de la publication des résultats de cette étude. Aucune identification personnelle ne sera utilisée dans les publications de cette étude. Toutes les informations enregistrées seront gardées à l'Université d'Ottawa pour la durée de l'étude et pour une période de 10 ans après la fin de l'étude.

## EN CAS D'URGENCE

En cas de chute ou de troubles cardio-respiratoires durant la session d'enregistrement, les chercheurs s'assureront que vous êtes en sécurité et confortable. Le service d'urgence de l'Université d'Ottawa sera appelé immédiatement à partir du laboratoire et vous portera assistance dans un bref délai, ainsi que de communiquer avec le service 911 de la région d'Ottawa si nécessaire.

## ENQUÊTE AU SUJET DE CETTE ÉTUDE

Ce projet de recherche a été approuvé par le Comité d'éthique en recherche à l'Université d'Ottawa. Pour des questions concernant l'approbation d'éthique de cette étude, vous pouvez contacter *Mme Lise Frigault* au (613) 562-5800, poste 1787 (courriel: [lfrigault@uottawa.ca](mailto:lfrigault@uottawa.ca)), officier du protocole pour l'éthique en recherche à l'Université d'Ottawa. Si vous avez besoin de plus amples renseignements au sujet de l'étude (procédures expérimentales ou autres détails), n'hésitez pas à contacter le *Dr. Gordon Robertson* et *M. François D. Beaulieu* au numéro ou à l'adresse mentionnée au début de ce document. Une copie de ce formulaire vous sera fournie avant la fin de la session.



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CONSENTEMENT

JE, \_\_\_\_\_, ACCEPTE VOLONTAIREMENT DE PARTICIPER À L'ÉTUDE MENTIONNÉE CI-HAUT AU SUJET DU CONTRÔLE BIOMÉCANIQUE DE LA DESCENTE D'ESCALIER.

J'AI REÇU ET LU UNE DESCRIPTION DÉTAILLÉE DU PROTOCOLE EXPÉRIMENTAL. JE SUIS COMPLÈTEMENT SATISFAIT AVEC LES EXPLICATIONS QUI M'ONT ÉTÉ FOURNIES AU SUJET DE LA NATURE DE CE PROJET DE RECHERCHE, INCLUANT LES RISQUES ET INCONFORTS POSSIBLES RELIÉS À MA PARTICIPATION À CETTE ÉTUDE.

JE SUIS AU COURANT QUE JE POSSÈDE LE DROIT DE RETIRER MON CONSENTEMENT ET MA PARTICIPATION EN TOUT TEMPS SANS AUCUN PRÉJUDICE.

SIGNATURES

PARTICIPANT/E

\_\_\_\_\_  
(signature)

\_\_\_\_\_  
(Nom imprimé)

TÉMOIN

\_\_\_\_\_  
(signature)

\_\_\_\_\_  
(Nom imprimé)

Date: \_\_\_\_\_

**Appendix B**  
**Screening Questionnaire**



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## SUBJECT INFORMATION AND MEDICAL HISTORY QUESTIONNAIRE

### SUBJECT IDENTIFICATION

Name: \_\_\_\_\_ ID code: \_\_\_\_\_

Age: \_\_\_\_\_ years Sex: M or F Telephone number: \_\_\_\_\_

Emergency telephone number: \_\_\_\_\_ Weight: \_\_\_\_\_ Height: \_\_\_\_\_

### MEDICAL HISTORY

1. Have you recently complained of pain in the lower limbs? Yes or No

If yes, specify

\_\_\_\_\_

2. Have you ever sustained severe muscle or bone related lower extremity injuries? Yes

or No

If yes, specify

\_\_\_\_\_

3. Have you ever been affected by the following disorders?

a) Joint disorders Yes or No

If yes, specify

\_\_\_\_\_

b) Visual disorders Yes or No

If yes, specify

---

c) Vestibular disorders                      Yes or No

If yes, specify

---

d) Epileptic disorders                      Yes or No

If yes, specify

---

4. Have you ever undergone surgical interventions, particularly to the lower extremities?

Yes or No

If yes, specify

---

5. Are you currently taking any medication?                      Yes or No

If yes, specify

---

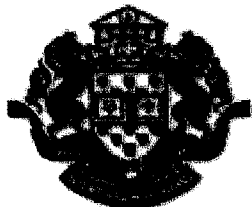
6. Do you have other medical conditions which should be mentioned?      Yes or No

If yes, specify

---

\_\_\_\_\_  
(signature)

\_\_\_\_\_  
(date)



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## QUESTIONNAIRE POUR INFORMATION SUR LE SUJET ET ANTÉCÉDENTS MÉDICAUX

### IDENTIFICATION DU SUJET

Nom: \_\_\_\_\_ Code d'identification: \_\_\_\_\_

Âge: \_\_\_\_\_ ans Sexe: M or F Numéro de téléphone: \_\_\_\_\_

### ANTÉCÉDENTS MÉDICAUX

7. Vous êtes-vous récemment plaint de douleur aux membres inférieurs?

Oui ou Non

Si oui, spécifiez

\_\_\_\_\_

a) Avez-vous déjà subi des blessures sévères? Oui ou Non

Si oui, spécifiez

\_\_\_\_\_

8. Avez-vous déjà été affecté(e) par les troubles suivants?

a) Troubles articulaires

Oui ou Non

Si oui, spécifiez

\_\_\_\_\_

b) Troubles visuels

Oui ou Non

Si oui, spécifiez

\_\_\_\_\_

c) Troubles vestibulaires

Oui ou Non

Si oui, spécifiez

\_\_\_\_\_

d) Troubles épileptiques

Oui ou Non

Si oui, spécifiez

---

e) Avez-vous déjà subi une ou des interventions chirurgicales? Oui ou Non

Si oui, spécifiez

---

9. Prenez-vous des médicaments?

Oui ou Non

Si oui, spécifiez

---

10. Êtes-vous déjà tombé?

Oui ou Non

Si oui, spécifiez

---

11. Souffrez-vous d'autres conditions médicales qui devraient être mentionnées?

Oui ou Non

Si oui, spécifiez

---