FACULTÉ DES ÉTUDES SUPÉRIEURES ET POSTOCTORALES

uOttawa

L'Université canadienne
Canada's university

FACULTY OF GRADUATE AND POSDOCTORAL STUDIES

Andrew Post

AUTEUR DE LA THÈSE / AUTHOR OF THESIS

M.Sc. (Human Kinetics)

GRADE / DEGREE

School of Human Kinetics

FACULTÉ, ÉCOLE, DÉPARTEMENT / FACULTY, SCHOOL, DEPARTMENT

Biomechanical Comparison of Stair and Ramp Descent

TITRE DE LA THÈSE / TITLE OF THESIS

Gordon Roberston

DIRECTEUR (DIRECTRICE) DE LA THÈSE / THESIS SUPERVISOR

CO-DIRECTEUR (CO-DIRECTRICE) DE LA THÈSE / THESIS CO-SUPERVISOR

EXAMINATEURS (EXAMINATRICES) DE LA THÈSE / THESIS EXAMINERS

Edward Lemaire

Heidi Sveistrup

Gary W. Slater

LE DOYEN DE LA FACULTÉ DES ÉTUDES SUPÉRIEURES ET POSTDOCTORALES /
DEAN OF THE FACULTY OF GRADUATE AND POSTDOCORAL STUDIES

Biomechanical Comparison of Stair and Ramp Descent

by

Andrew Post, B.Sc. (Hons. H.K.)

Submitted in partial fulfillment of the requirements for the degree
Master of Science (Human Kinetics)
in
Faculty of Graduate and Postdoctoral Studies,
University of Ottawa

Your file Votre référence
ISBN: 0-494-14940-X
Our file Notre référence
ISBN: 0-494-14940-X

ABSTRACT

Purpose: Although many different forms of human gait have been researched, a lack of knowledge exists when comparing the moments and powers felt by the lower extremity during the descent of stairs and ramps. The purpose of this project was to quantify and analyze the changes that occur to the moments of force and moment powers at the joints of the lower extremity during stair and ramp descent. **Methods:** A sample population of five male and five female volunteers were asked to walk five times down a 10-degree ramp at normal gait speed, followed by five stair descent trials. Force platforms mounted on the ramp and stairs measured ground reaction forces while a VHS camera collected the sagittal view trajectories of markers placed on the left side of the body. An inverse dynamics approach was used to compute the moments and powers at each joint. These data were then ensemble averaged and normalized to body mass. **Results and Discussion:** Stair descent has a larger eccentric plantar flexor peak at the ankle joint during weight acceptance. The knee exhibits slightly larger eccentric knee extensor peaks during ramp descent at push - off and there was higher loading of the hip during ramp descent as compared with stair descent. The higher ankle powers at FS during stair descent reveal a concern for those people suffering from ankle pathology and the larger hip and knee peaks during ramp descent is of concern to those with hip and knee problems. These results could be used in the rehabilitative considerations of patients with hip, knee and ankle replacements and as a baseline for future research.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

Acknowledgements

I would like to thank the following people for the help they gave me in the completion of this thesis. I would not have been able to complete this without them:
D.G.E. Robertson,
Shaunna Taylor,
and my family.

PART ONE: THEORETICAL AND METHODOLOGICAL CONSIDERATIONS

CHAPTER 1

INTRODUCTION

Stairs are used daily to ascend and descend to different levels and are also used as a means to promote active living (Blamey *et al.*, 1995). Healthy benefits of stair use include improvements to cardiovascular and cardiopulmonary fitness as well as strengthening of the active and passive structures of the locomotor system (Schwameder *et al.*, 1999) that lead to a philosophy of active living in a society where obesity has become a topic of concern.

Recent research regarding stairs focused on muscle action and reaction forces during descent. For example, higher loads on the knee and the hip exist while descending stairs (Yu *et al.*, 1997) as compared with level walking. These higher joint reaction forces are one reason why many people who ascend or descend stairs find it painful (Kaufman *et al.*, 2001). From this research, a need exists to find a method of vertical movement that eases the joint moments and powers to a point where less discomfort is likely to occur. Elevators are one solution to this problem; although they are costly. A more feasible solution lies in the use of ramps. People with certain physical disabilities use ramps daily, for example, people in wheelchairs or walkers and even uninjured people may also use ramps. Unfortunately, few biomechanical ramp studies that compare stair descent to ramp descent exist.

Ramp research has been a relatively new phenomenon with the bulk of the research by Kuster and Schwameder. Kuster's work (1997) examined the effects of ramp descent on knee moments and found that they were larger than those incurred in level walking. These results caused speculation that steep downhill walking might lead to chronic muscular pain (Schwameder *et al*, 1999). Schwameder *et al*'s (1999) ramp study, which studied injuries that affect hikers, indicated that moving downhill was significantly more intensive physically than walking over level ground. Their results demonstrated that joint reaction forces were much higher than those found during level walking but possibly lower than those felt while descending stairs. These researchers attributed the higher joint reaction forces to the increase in lower extremity injuries found in those who frequently walk downhill.

To date, no research has been done comparing walking down stairs versus down ramps using the same subjects. This study quantified the lower extremity's moments and powers as each subject walked down both ramps and stairs. Such research makes an important contribution to the present understanding of how gait is adapted for stair and ramp descent.

Objective Statement

To discover how the patterns of the ankle, knee and hip moments differ between descending stairs versus ramps. This will be done by quantifying and analyzing the patterns of the moments of force and moment powers at the joints of the lower extremity during stair and ramp descent.

Hypotheses

Ramp descent reduces the peak knee moment of force when compared with descent down stairs. The peak ankle moment of force will show no difference between the stair and ramp descent, however the peak hip moment will be larger for stair descent.

Delimitations

Only a 10-degree ramp and only one stair dimension (20 x 30 cm) was tested. This ramp angle was selected since this is the maximum permissible ramp inclination for moveable ramps.

Only a 4-step stairway was used to examine the characteristics of stair descent because of the need to instrument the stairs and due to height restrictions within the laboratory. These restrictions fall within acceptable building standards.

Assumptions and Limitations

One limitation of this research project was that subjects will be tested in the controlled environment of a laboratory as opposed to real-life scenarios. This is required because of the need to inlay force platforms on the walking surfaces and for suitable lines of view for the recording cameras.

For the biomechanical research of human motion, the identification of certain assumptions necessary to quantify the activity that is being analyzed are important. This study assumes that the body is a system of linked rigid bodies connected by frictionless pin-connected joints. The paths of the joint centers can be traced by placing markers at the approximate centers of the joints to discover their motion. Further errors involve the relative movement of skin, joint centers and camera perspective errors. The deformable nature of the segments of the body make the moments of force difficult to quantify validly. These errors have been found to be within acceptable limits and this method has become the standard for gait analysis (Winter,1990). For these reasons, the rigid body model will be used to calculate the moments around the hip, knee and ankle.

Rationale

The results from this study showed whether a difference exists in the mechanical demands of the lower extremity during stair and ramp descent. The hip, knee and ankle were studied and a major concern was which structure incurred the highest moments and powers to help establish norms for descending ramps or stairs. This research may also determine more completely the differences between ramp and stair gait to identify possible areas of future research. Ramps could possibly indicate a method to lessen the peak moments of force in some joint structures in comparison with stair descent. Finally, this research could be applied to the Activities of Daily Living (ADL) where the impact of stair and ramp use by the infirm or rehabilitating patients could be given more consideration by their medical and rehabilitation practitioners.

CHAPTER 2

REVIEW OF LITERATURE

Biomechanical analysis has improved significantly since the first experiments performed by Fenn in 1930 on investigating the instantaneous powers of sprinters. Since then, a great deal of research has been devoted to analyzing the many aspects of human gait (Ralston, 1976, Robertson & Winter, 1980, Winter, 1991, Whittle, 1995 & 2001, Andriacchi & Alexander, 2000). This literature review focuses on normal gait and the biomechanical properties of stair and ramp descent. It covers the knowledge concerning level gait and stair descent and will show a lack of knowledge exists in our understanding of the biomechanics of ramp descent, especially in comparison with other modes of descent (i.e., stair descent).

Analysis of Gait

In walking, the joints of the ankle, knee and hip have been the major concern because the movements of the head, trunk and arms (H.A.T.) have been found to have a negligible effect on the powers and energy exchanges that occur during gait (Winter, 1978). In the knee, several actions occur in normal gait. During swing phase, initially the inertial load is positive, which indicates the extension of the leg (Winter & Robertson, 1978). This leg extension is mostly the result of gravitational forces that cause a moment to accelerate the leg forward. During the last half of the swing, the inertial load reverses and so do the components contributing to the moment decelerating the leg. At this point, the hamstring muscle contributes approximately 80% of the decelerating moment.

Energy requirements of the legs were much higher during gait than those of the rest of the body; furthermore, the H.A.T. did very little other than move in the direction the legs are moving. The primary demand for new energy occurred at the push-off phase of gait, so that the leg could start moving forward and upward. This energy was then stored in the swinging leg and subsequently removed or dissipated during the second half of the swing. This loss of energy in the swinging leg was the result of negative work by the hip extensors and knee flexors that decelerated the thigh and the leg.

The power analysis done by Winter and Robertson (1978) demonstrated the complex generation, absorption and transfer of energy that occurs among segments during normal gait. At push-off, the energy of the foot, leg and thigh increased by 65 J, 86 J and 97 J, respectively by a powerful push from the ankle plantar flexors (Winter & Robertson, 1978). The energy flow out of the common plantar flexor tendon was 533 W; 65 W was contributed to the foot and the remaining 469 W indicated energy flow up from the foot through the ankle joint. Of this power, 108 W continued to flow up through the knee joint and had little contribution from the muscles that acted upon the knee. Only 23 W continued into the hip joint and then into the H.A.T. This analysis showed that most of the energy was being produced by the muscles at the ankle, which was added to the energy of the four body segments with little involvement from the muscles that acted upon the knee or hip (Winter & Robertson, 1978). During toe-off, a similar increase in energy occurred in the distal segments of the leg; however, this energy source was from the hip flexors which acted across the hip joint. The acceleration phase, which occurred after the foot leaves the ground, showed some muscle activity and small energy exchanges occurred from the thigh to the foot through the joints. The deceleration phase of the thigh and leg was performed by the hamstrings and only slightly by the hip extensors (6 J) characterized by a decrease in energy. At the end of the swing phase, kinetic energy leaves the lower leg (foot 50 W and leg 47 W) through the joints with minimal muscle involvement at the thigh (gained 23 W) and trunk (gained 58 W) (Winter and Robertson, 1978). The final stage of energy absorption is weight acceptance where 244 W flowed out of the trunk and across the knee joint; 164 W was dissipated by the knee extensors and the rest in the dorsiflexors (33 W) which was indicated by an energy increase of 89 W in the foot

To summarize, in normal gait the major muscle group that acted at push-off is the ankle plantar flexors (Winter & Robertson, 1978). At a faster walk, a more forceful contraction by this group occurred and therefore, a higher power was measured. At toe-off, the hip flexors acted across the hip joint to increase energy. Notably, variations in cadence did not result in energy changes across the hip. As the leg accelerated forward, energy flowed from the trunk into the leg through the joints with little muscle activity. The deceleration of the leg towards initial contact (IC) occurred as a result of energy absorption by hip extensors and knee flexors with the final deceleration indicated by a flow of energy from the foot to the thigh and trunk via the leg. At weight acceptance the energy stored in the trunk traveled across the hip and knee and was dissipated by the knee extensors and foot dorsiflexors.

From similar within segment power and energy analyses, Winter *et al.* (1976) derived that the torso acted as a conservative system with about half of its kinetic and potential energy interchanging. Winter *et al.* (1976) also noticed that the thigh had a high level of conservation, saving approximately one third of its energy and the lower leg had the largest increase in total energy, but almost no exchange of energy. All the segments except the lower leg had rotational kinetic components so small that they could be ignored. The lower legs rotational component was found to contribute approximately 10% of the total energy but was thought to contribute more when walking faster or running, which could add to error in 2 dimensional camera analyses (Winter *et al.*, 1976). In a later study,

Winter's (1991) kinetic analyses of normal gait at fast, slow and normal cadence showed that ankle moments differ between early and mid stance phase (5% to 40% of cycle) and push-off (40% to 60%, Figure 1). During the first part of stance phase at IC there is a small ankle dorsiflexor moment and, as the leg rotates over the foot, the ankle moment becomes plantar flexor towards push-off. Late in stance phase, those same muscles plantar flex and generate an explosive burst of power. Fast cadence produced a lower moment than slow and normal cadence during the energy absorption phase, while the opposite was observed during push-off.

The magnitude of the moments in the knee during walking increased proportionally with cadence. At initial contact (IC) the knee underwent a flexor moment followed by the activation of the knee extensors to dissipate energy and control knee flexion until 15% of stride, then assisted in extending the knee and add potential energy to the body. At 30% to 50% of stance, the knee moment becomes flexor. As the knee starts to collapse following push-off the knee extensors activated eccentrically to control the degree of flexion and to decelerate the backward swinging leg. Near the end of swing phase an eccentric knee flexor moment occurred that results from the hamstrings decelerating the forward swinging leg.

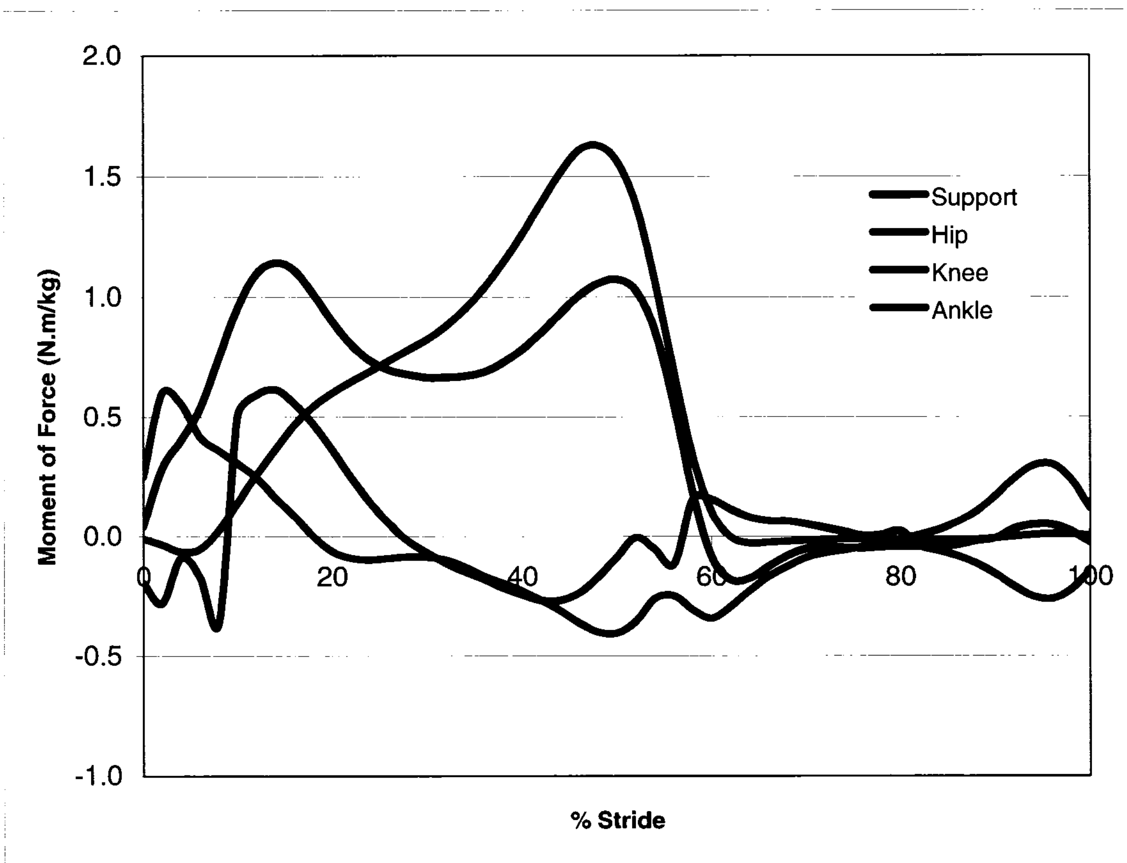

Figure 1. Hip, knee ankle and support moments for normal cadence walking (from Winter, 1991, included with permission)

The hip moments were dissimilar throughout the three cadence speeds, mostly due to timing. At IC the hip extensors activated to control hip flexion followed by the extensors to control the trunk and stabilize forward rotation of the pelvis. At midstance, the hip flexors activated to control the backward rotating thigh and to stop hip rotation. At 50% of stride; a reversal occurred and the hip flexors initiated a pull up of the lower limb that continued into mid swing.

Winter's analysis of the powers (figure 2) involved in normal gait showed that between IC and 5% of stride the foot plantar flexes under the control of a dorsiflexor moment. From 5% to 40% of stride, the rotation of the leg, due to an eccentric plantar flexor moment, resulted in an energy absorption phase (A1). At 35% and 40% of stride, the plantar flexor moment caused heel off and rapid plantar flexion of the ankle that resulted in power burst A2, which was regarded as the most important energy generation phase during the gait cycle: 80-85% of the energy is generated during A2. At 60% of stride, toe-off (TO) occurred and the plantar flexion moment ends. A subsequent dorsiflexor moment caused the foot to dorsiflex until 75% of stride, at which point the foot can safely clear the ground.

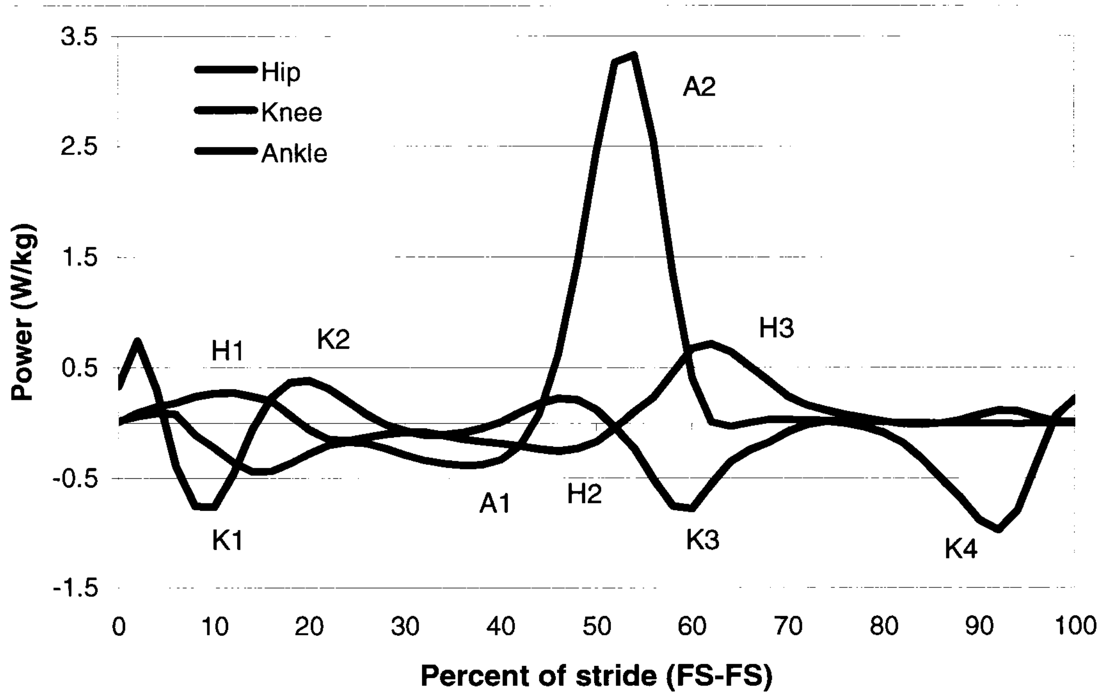

Figure 2. Hip, knee and ankle powers for normal cadence walking (from Winter, 1991, included with permission)

Analysis of the powers at the knee revealed four phases of energy absorption and generation. From IC to 15% of stride, the knee flexed under the control of the knee extensors, which resulted in the first absorption power burst (K1). From 15% to 40% of stride, the knee extended by a concentric knee extensor contraction that produced a positive burst of power from the knee extensors (K2), which represented only 10% to15% of the energy generated in the gait cycle. At 40% of stride, the knee began to flex, which continued into early swing phase (70% of stride). The knee extensor moment at 70% of stride was associated with another power burst (K3) which served two functions: (a) prior to TO, the knee extensors absorbed energy and control the collapsing knee joint and (b) after TO the extensors continued to absorb energy by decelerating the backward swing of the leg and prepared it for extension during swing phase. The knee extensors did not propel the leg forward during swing phase; this was accomplished primarily by pendulum action. During the last half of swing, the knee flexors absorbed much of the energy from the swinging leg and foot, (K4).

The powers at the hip for most of stance (H1 and H2) were low and not well defined. These hip powers were attributed to trunk balance, which can be highly variable between subjects. The hip powers only became substantial with faster cadences. Energy absorption by the hip flexors (H2) occurred during deceleration of the thigh as it rotates in a backwards direction. The pull-off burst (H3) increased under the inertial load of the swinging limb.

Although gait has been researched extensively, establishment of a normal pattern of gait that can be applied to everyone remains difficult (Simonsen *et al.*, 1997). For example, the energy and power patterns of gait among individuals differ as a result of cadence and stride length variations (Robertson & Winter, 1980). These variations can sometimes, but not always, be correlated to age and subject height. Segmental analyses, like those conducted by Winter and Robertson (1978) and Winter (1991), were not ideal as errors exist in the method they used to calculate their values. In complex natural movements such as gait, muscle activity at a single joint was rarely responsible for all the energy changes and exchanges in adjacent segments (Ingen Schenau & Cavanagh, 1990). As well, the lack of muscular activity in one segment does not mean that energy changes were not occurring in another segment (Winter & Robertson, 1978). Finally, Aleshinsky (1986) reported that the segmental model does not account for muscles that cross more than one joint. This could lead to transfers of energy between two non-adjacent segments.

Stair Descent Analyses

The majority of stair studies were focused around the various anomalies of stair gait based on subjects with a disability, disease or injury (Yu *et al.*, 1997). Stair descent has been linked to injuries such as trauma from falls and chronic conditions. Falls alone have occurred three times more often descending stairs than when ascending stairs (Christina & Cavanagh, 2002), which causes higher risk to cement in hip replacements. The following section will focus on stair descent kinetics and some pathologies that are affected by stair descent.

Research focused on muscle activity has shown that, during stair descent, the body is lowered by an eccentric contraction by the soleus and quadriceps femoris (Joseph and Watson, 1967), with smaller levels of stance phase muscle activity produced in the other muscles except for the hamstrings. The gluteus maximus contracted at this point to prevent the trunk from flexing at the hips. The contraction of the gluteus medius is important during the support phase to keep the unsupported side of the body from falling. At toe-off the tensor fasciae latae and adductor magnus muscles had very little activation as the leg left the step. During swing phase, the hamstrings were responsible for flexing the leg. The tibialis anterior activated during the swing phase to help the foot clear the step upon step-down and inverted the foot slightly; concurrently the soleus contracted to overcome the dorsiflexion moment of the tibialis anterior. Finally, the erector spinae contracted to help prevent forward bending of the trunk at the vertebral column. During late swing phase and through to the loading phase, the gluteus medius and maximus had EMG levels of 20% and 15% of maximum (Lyons *et al.*, 1983). In comparison with stair ascent, Lyons *et al.* (1983) found that stair descent used less muscular activity.

In analysis of stance and swing phase, Joseph and Watson (1967) found that, similar to walking on level ground and up stairs, both limbs had a swing and support phase. During descent, the support phase was about three fifths of the step while the swing phase was two fifths. Double support by both limbs occurred for only one fifth of the step down. Of the major muscles involved in stair descent they found that the tibialis anterior was active during swing phase to help the foot clear the step and prepare for heel-strike, the soleus was active during the support phase and the hip flexors were active at toe-off to drive the thigh forward, in order to descend to the next step.

In research on the kinematics of valgus and varus motion of the knee during level walking and stair climbing, Yu *et al.* (1997) discovered that stair gait could be detrimental to patients with joint anomalies such as osteoarthritis. They discovered that ascending stairs was more stressful than level walking because the body was lifted with each step and then decelerated with each descending step, which would cause more weight to be supported by the leg. Their research showed that the varus angle of the knee and the valgus moment at the knee increased during stair ascent in comparison to level gait. As well, the vertical reaction force was a major factor in the valgus-varus moment, implying a greater reaction force when ascending and descending stairs than when walking on level ground (Kirkwood *et al.,* 1999). These increased reaction forces during step down were of particular concern with respect to possible injury, especially if the muscles were not used to dampen the force upon contact with the lower step in stair descent (Freedman *et al.*, 1976). These results indicated that the forces developed during ordinary step-down could contribute to joint injury.

Further research on pathologies linked with step down led to Brechter and Powers (2002) study that analyzed the causes of patellofemoral pain (PFP) and showed that ascending and descending stairs was one of the most painful activities of daily living for people with PFP. Patellofemoral joint reaction force (PFJRF) during stair descent was more than three times that of level walking (Brechter & Powers, 2002) which can add to articular cartilage degeneration and wear. The peak patellofemoral joint (PFJ) stress occurred earlier in stance phase for PFP patients and may be explained by a reduction in utilized contact area at lesser knee flexion angles. This smaller contact area resulted in greater peak PFJ stress during the first 60% of the stair descent's stance phase. As the knee flexion angle increased (last 40% of stance), the PFJ contact area in the PFP group increased, resulting in decreased PFJ stress late in stance phase. During descent, the PFP group showed a higher peak knee moment and peak PFJRF. The PFJ stress-time integral was 1.5 times greater during descent when compared with ascent and could explain why patients frequently report greater pain during descent over ascent.

The analysis of joint forces was expanded upon in a later study by Christina and Cavanagh (2002) which revealed the ground reaction forces in stair descent was characterized by two unequal peaks; the first higher than the second. The first peak was higher by 0.45 and 0.56 of body weight (BW) for stair one and two, respectively, with the mediolateral ground reaction forces were directed more medially. Between these peaks was a valley that was more flat than normally expected during level walking. The first peak vertical force for descending stairs was 0.35 times BW higher than for level walking. The second vertical peak force was 0.15 times BW lower than that for level walking. Bergmann *et al.* (2001) also indicated that when climbing stairs the average torque about the hip was 23% higher than during level gait, which confirms the possibility of higher joint forces proposed by Christina and Cavanagh (2002) and Yu *et al.* (1997).

In a pilot study furthering the research of Christina and Cavanagh (2002), Bergmann *et al.* (2001) and Yu *et al.* (1997), Beaulieu *et al.* (2002), analyzed the effects of walking forward and backward down stairs on joint moments. Beaulieu *et al.* (2002) showed that the knee moments for descending stairs were higher than those for level walking by 40% to 60%, showing a slightly lower moment than Andriacchi *et al.*'s (1997) conclusions that stair descent could have up to 400% higher knee joint moments when descending stairs. The centers of pressure for the ground reaction forces were closer to the edge of the force plate for forward descent down stairs, implying a possible hazard for tripping which confirmed the fall data by Christina and Cavanagh (2002). The results found were somewhat inconsistent with the knee and hip moments being significantly different in many of the subjects which indicated a need for a larger study. The ankle moments and powers were not greater than those for level walking.

Joint reaction forces and moments have been shown in the literature to be higher in stair ascent and descent. This leads to a possible predisposition of injury to people who frequently use stairs or could make stair negotiation dangerous for people with joint injury or bone pathologies (Freedman *et al.*, 1976). Supporting these conclusions, higher kinetic values associated with stair descent has led to the consideration of ramps being researched as an alternative.

Ramp Descent Analyses.

Ramps are currently used as an everyday tool by people for wheelchair access to buildings and facilities. Ramps may be a solution to the dilemma of higher joint stresses and falls that may arise from the use of stairs which threaten knee and hip replacement wear due to the more horizontal movement as opposed to the vertical lowering of stair descent; however, little research exists regarding this topic.

One significant study concerning the aspects of downhill gait was conducted by Schwameder *et al.* (1999) who focused on knee joint forces during downhill walking with hiking poles at 25 degrees. This study showed that during downhill walking, injuries affect hikers because of the higher loads on the joints of the lower extremities with the knee joint being loaded more during downhill walking than the other joints. This supports the findings of Kuster *et al.* (1994) who discovered that peak ground forces at an 11 degree decline increased by 38% over level walking, the peak knee flexion moments increased by 117% and the peak knee muscle power increased by 490%, concluding that downhill walking significantly stressed the different knee joint structures (Kuster *et al.*, 1995). Schwameder *et al.* (1999) showed that, as the grade increased, the peak ground reaction forces increased by 7 N/kg over level walking at a 25-degree decline. This increase was the result of the increased potential energy dissipated when walking downhill. However, the external peak forces were reduced by 15% when hiking poles were used, which indicated that some sort of supporting structure could reduce the impact forces on ramps by transferring some of the ground reaction forces to that structure. The peak knee joint moments during downhill walking were found to be five times higher during level walking and an increase of 100% for peak flexion moments occurred when walking at an 11-degree decline (Kuster *et al.*, 1997). Schwameder *et al.* (1999) noticed much higher values than Kuster, a result attributed to the higher degree of decline. Research has indicated that for the duration of downhill walking, many of the gait adjustments occurred primarily at the hip joint during heel-strike rather than at the knee throughout the stance phase (Kuster *et al.,* 1995). At toe-off, as swing phase began, both the hip and ankle joint make angular changes to compensate for the slope. As well, muscle work changed from a concentric plantar flexor effort at toe-off to an eccentric knee extensor action at the beginning of stance phase. In addition, knee extensor power data on a decline of 11 degrees approach those normally found for running on a level surface (Kuster *et al.,* 1995).

In a kinematic and kinetic analysis of gait patterns on ramps of different angles, Redfern and DiPasquale (1997) examined subjects descending ramps. They found that the biggest change occurred at the knee extension moments while descending, accompanied by large eccentric dorsiflexion moments at the ankle. They also discovered that the shear ground reaction force increased proportionally with ramp angle, and that step length and period decreased as ramp angle increased but the speed of gait did not increase significantly. Redfern and DiPasquale (1997) concluded that little adjustment in gait patterns was required for ramp descent; however, ramps did require large increases in knee moments as the ramp angle is increased.

Summary

There was a considerable body of knowledge concerning the mechanics of level human gait. On the other hand, an apparent gap exists in the literature regarding the comparison of the kinetics of stair and ramp descent. Although several studies have established that downhill gait can be strenuous in the joints of the lower extremity (Kuster *et al.*, 1994, Kuster *et al.*, 1995, Redfern & DiPasquale, 1997 and Schwameder *et al.*, 1999) none has directly compared stair descent to downhill walking kinetics. Further analysis of ramp and stair descent could also lead to a better understanding of the differences between the two methods of descent.

CHAPTER 3

METHODS

Population

The sample population was taken from a group of university students who volunteered for the study. Five males and five females, between the ages of 18 and 30, were selected randomly from volunteers. The subjects were free of any lower extremity pathology that may have interfered with normal gait. All subjects provided their written consent prior to the trials.

Protocol

For the trials, the subjects wore slippers and dark clothing (shorts and Lycra top) that had non-reflective surfaces. It was important to have tight fitting clothing so that the reflective digitizing markers did not move on the clothing while the subject was in motion. The reflective markers were placed on the following locations on the left side of the body: hallux, 5^{th} metatarsal (distal), calcaneus, lateral malleolus, lateral femoral condyle (knee), the greater trochanter and head of the humerus. The subjects were required to wear slippers supplied to them by the gait lab to maintain a consistent footwear type.

A grid board of control points (1×1.5 m) was filmed in a horizontal position prior to subject testing to establish a frame of reference for digitizing. Half the subjects walked at a self-selected pace down the stairs and then down the ramp for five trials, while a VHS camera recorded their motion.. The other half of the subjects walked at a self-selected pace down the ramp and then the stairs, to control for any testing order that may have affected their gait patterns. The subjects completed the ramp and stair gait trials in the same recording session.

The ramp used in this study was a solid structure composed of iron and was set at a fixed angle of 10 degrees with a length of 2.30 m. The surface of the ramp was composed of $1\frac{3}{8}^{th}$-inch plywood that allowed for a section to be taken out so that the force plate was flush with the walkway. Before data collection, the subjects descended the ramp several times to become comfortable with stepping on the force plate and with the inclination of the ramp. The subjects started walking down the ramp from an elevated box at the top of the ramp. When walking down the ramp the subjects completed the minimum of one full stride while making contact with the force plate. The ramp itself was covered with a non-slip surface to ensure the safety and to prevent slipping.

The stairs consisted of 3 steps of 20.0 cm rise and 30.0 cm tread each of which were wide enough to hold a force platform that could be placed on any step using a special bracket. Before any trials were recorded, the subjects were permitted to practice the step down procedure so that they became comfortable with the stairway. The stair trials were conducted with the subject stepping from the top step and walking down the stairs as they would normally. This meant the subject took a full step down to the force plate that was fixed to the last step (step 3) before stepping to the ground. This ensured that the subject engaged in normal stair descent gait for the force recording.

Techniques

A Kistler Slimline (9286AA) force platform was placed on the ramp allowing for the reaction force of one cycle to be captured per trial. The force plate used for the stair trials was the Kistler 9281C, which was placed on the stairs in such a way that allowed the subject to contact the force plate two steps down. The force plate was sampled at 240 Hz. A computer using the BioWare software for force plate collection recorded the reaction forces from the plate. The motion was recorded using a VHS camera sampling at 60 Hz. The camera was placed to record the left side sagittal view of the subjects while the subjects descended the ramp and the stairs.

Data Acquisition

All force data were collected by a computer connected to the force plate. The data analysis of the force recordings was acquired using the BioWare program. The VHS data was captured using Ariel Performance Analysis Software (APAS) software for the marker trajectories. The Biomech Motion Analysis System and BioProc2 performed the inverse dynamics analyses and computed the powers produced by the moments of force.

Data Analysis

To determine whether significant differences occurred among the patterns of the moments of forces and their associated powers, each subject's moments of force and powers were ensemble averaged and normalized to body mass. Differences which would be selected for statistical testing were said to have occurred between level walking (Winter, 1991) and stair or ramp descent whenever the ensemble averages differed by +/- one standard deviation. Significant differences between the peak powers in the hip, knee and ankle for ramp and descent were tested using a t-test with alpha set at 0.05.

REFERENCES

Aleshinsky, S.Y. (1986). An energy 'sources' and 'fractions' approach to the mechanical energy expenditure problem-IV. Criticism of the concept of 'energy transfers within and between links'. *Journal of Biomechanics, 19(4),* 307-310.

Andriacchi, T.P., Hurwitz, D., Bush-Joseph, C., & Bach, B. (1997). Clinical implications of functional adaptations in patients with ACL deficient knees. *Sportorthopadie-sporttraumatologie, 13(3)*, 153-160.

Andriacchi, T.P. & Alexander, J.A. (2000). Studies of human locomotion: past, present and future. *Journal of Biomechanics, 33*, 1217-1224.

Beaulieu, F.D., Pelland, L., & Robertson, D.G.E. (2003). Descending stairs, forwards and backwards. *Proceedings International Society of Biomechanics Congress XIX.* Dunedin, New Zealand.

Bergmann, G., Graichen, F., & Rohlmann, A. (2001). Hip contact force and gait patterns from routine activities. *Journal of Biomechanics, 26*, 859-871.

Blamey, A., Mutrie, N., & Aitchison, T. (1995). Health promotion by encouraged use of stairs. *British Medical Journal, 311*, 289-290.

Brechter, J.H. & Powers, C.M. (2002). Patellofemoral joint stress during stair ascent and descent in persons with and without patellofemoral pain. *Gait and Posture, 16*, 115-123.

Christina, K.A. & Cavanagh, P.R. (2002). Ground reaction forces and frictional demands during stair descent: effects of age and illumination. *Gait and Posture, 15*, 153-158.

Fenn, W.O. (1930). Frictional and kinetic factors in the work of sprint running. *American Journal of Physiology, 92*, 583-611.

Freedman, W., Wannstedt, G., & Herman, R. (1976). EMG patterns and forces developed during step-down. *American Journal of Physical Medicine, 55*, 275-290.

Ingen Schenau, G.J. van, & Cavanagh, P.R. (1990). Power equations in endurance sports. *Journal of Biomechanics, 23*, 865-881.

Joseph, J. & Watson, R. (1967). Telemetering electromyography of muscles used in walking up and down stairs. *The Journal of Bone and Joint Surgery, 49*, 774-780.

Kaufman, K.K., Hughes, C., Morrey, B.F., Morrey, M., & An, K. (2001). Gait characteristics of patients with knee osteoarthritis. *Journal of Biomechanics, 34*, 907-915.

Kirkwood, R.N., Culham, E.G., & Costigan, P. (1999). Hip moments during level walking, stair climbing and exercise in individuals aged 55 years or older. *Physical Therapy, 79(4),* 360-370.

Kuster, M., Wood, G.A., Sakurai, S., & Blatter, G. (1994). Downhill walking: A stressful task for the anterior cruciate ligament? A biomechanical study with clinical applications. *Knee Surgery, Sports Traumatology and Arthroscopy, 2*, 2-7.

Kuster, M., Sakurai, S., & Wood, G.A. (1995). Kinematic and kinetic comparison of downhill and level walking. *Clinical Biomechanics, 10*, 79-84.

Kuster, M.S., Wood, G.A., Stachowiak, G.W., & Gachter, A. (1997). Joint Load Considerations in total knee replacements. *Journal of Bone and Joint Surgery, 79,* 109-113.

Lyons, K., Perry, J., Gronley, J.K., Barnes, L., & Antonelli, D. (1983). Timing and relative intensity of hip extensor and abductor muscle action during level and stair ambulation. *Physical Therapy, 63,* 1597-1605.

Ralston, H.J. (1976). Energetics of human walking. *Neural Control of Locomotion.* New York: Plenum Press.

Redfern, M.S. & DiPasquale, J. (1997). Biomechanics of descending ramps. *Gait and Posture, 6(2)*, 119-125.

Roberston, D.G.E. & Winter, D.A. (1980). Mechanical energy generation, absorption and transfer amongst segments during walking. *Journal of Biomechanics, 13*, 845-854.

Simonsen, E.B., Dyhre-Poulsen, P., Voigt, M., Aagaard, P., & Fallentin, N. (1997). Mechanisms contributing to different joint moments observed during human walking. *Scandinavian Journal of Medicine & Sports Science, 7*, 1-13.

Schwameder, H., Roithner, R., Muller, E., Niessen, W., & Raschner, C. (1999). Knee joint forces during downhill walking with hiking poles. *Journal of Sports Sciences, 17*, 969-978.

Whittle, M.W. (1995). Clinical gait analysis: A review. *Human Movement Science*, *15,* 369-387.

Whittle, M.W. (2001). *Gait Analysis: An Introduction* (3rd ed.). Oxford: Butterworth-Heinemann.

Winter, D.A., Quanbury, A.O., & Reimer, G.D. (1976). Analysis of instantaneous energy of normal gait. *Journal of Biomechanics, 9*, 253-257.

Winter, D.A. (1978). Calculation and interpretation of mechanical energy of movement. *Exercise and Sport Science Reviews, 6*, 183-201.

Winter, D.A. & Robertson, D.G.E. (1978). Joint torque and energy patterns in normal gait. *Biological Cybernetics, 29*, 137-142.

Winter, D.A. (1990). Biomechanics and Motor Control of Human Movement (2nd ed.). John Wiley & Sons, Inc.

Winter, D.A. (1991). *The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological* (2nd ed.). Waterloo, ON: University of Waterloo Press.

Yu, B., Stuart, M.J., Kienbacher, T., Growney, E.S., & An K. (1997). Valgus-varus motion of the knee in normal level walking and stair climbing. *Journal of Clinical Biomechanics, 12*, 286-293.

PART 2

THESIS IN ARTICLE FORMAT

INTRODUCTION

Remarkably, Macleans magazine reported that 236 Canadians died in 2000 from stair falls compared with 404 pedestrian deaths. No similar data were reported for falls caused by ramps but Randall Atlas (1997) estimates "Ramp fall accidents represent only about 10% of all slip and fall accidents, but that number is increasing as older buildings are being equipped with ramps…." These statistics point to the danger that ramps and stairs have, yet little research has compared the differences between these two means of moving from one level to another.

Stairs and ramps have been a topic of research for many years, with the majority of the studies concerning the ascent or descent by individuals with various muscular and joint pathologies. Recent research on stair descent has focused primarily on muscle action and reaction forces with application to the aging geriatric population (Stacoff *et al.*, 2005). Stair descent, due to its complex nature and higher relative moments of force and powers about the lower extremity, has also become a concern for the design and wear of joint replacements, arthritis interventions and lower extremity injuries (Catani *et al.*, 2003, Saari *et al.*, 2004, Kaufman *et al.*, 2001, Thambyah *et al.*, 2002). Unlike stairs, ramps are used less frequently; however, ramp research has had the same focus as stair descent. The objective of the most recent ramp research has been to identify the difference between ramp ascent or descent when compared with walking (Kuster *et al.*, 1997; Cham and Redfern, 2002). To date, there have been some discoveries that indicate that descent down steep inclines produce higher moments of force in the leg (Kuster *et al.*, 1997; Schwameder *et al.*, 1999), which have been implicated as a reason for acute injuries in hikers when they descend steep terrain (Blake and Ferguson, 1993; De Loes, 1995). These acute injuries, however, have not been linked to everyday ramp structures such as access ramps to buildings, perhaps these inclines are too shallow to produce the extreme moments experienced by hikers.

Research concerning stair gait has been examining injuries and chronic conditions that make stair ascent or descent difficult or painful (Kaufman *et al.*, 2001, Salsich *et al.*, 2001, Saari *et al.*, 2003,). Stairs have also been a popular topic of fall related research, due to a great number of people being injured or killed by falling down stairs. These injuries were more likely to occur on descent, which may be a result of the body centre of gravity leaning over the foot or an improper foot placement (Beaulieu *et al.*, 2002).

During stair descent, the weight acceptance phase was evidenced by an energy absorption peak in both the ankle and knee (A1 and K1) by the triceps surae, the rectus femoris and vastus lateralis (McFadyen and Winter, 1988). However, during the large K1 absorption burst at initial contact, most of the energy of the descending body was absorbed by the ankle plantar flexors. By toe-off of the contralateral limb, the body has dropped to the lower step and the tibialis anterior and the soleus activate to support the ankle and maintained the body weight on the lateral part of the foot (Joseph & Watson, 1967; McFadyen & Winter, 1988). At the beginning of single support, the power absorption at the knee turns into a brief power generation (K2), as the body begins to move forward in a period termed "forward continuance" by McFadyen and Winter (1988). From midstance to the start of swing phase, the body is lowered by the quadriceps to the next step. There is also a small period of negative power at the ankle by the soleus (A2), which helps the quadriceps contribute to forward movement and controls the lowering of the body (McFadyen & Winter, 1988). For the majority (85%) of stride, the hip powers were small. From 85% of stride to the end of the cycle, there was a positive power peak (H1) in which the hip flexors activated to pull the leg to the next position as well as pull it off the upper step, as the knee extensors controlled knee flexion (McFadyen & Winter, 1988). Ankle plantar flexor activity as the foot leaves the upper step was to relieve the dorsiflexion of the foot as well as push the swing leg through. Early in swing, the hip flexor moment continued to pull the leg through to initial contact with a positive power burst (H2). However, due to the relatively small amount of power generated from the hip during decent, the hip musculature did little work to lower the body. This lowering was accomplished primarily by the knee and a little by the ankle moments of force (McFadyen & Winter, 1988). There was little knee flexion during swing, as foot clearance was not as important in stair descent as in level gait. During swing phase, flexion about the knee and hip was reversed while the ankle begins to plantar flex. As all three joints were extended, the leg was preparing to accept body weight in a phase called "foot placement" by McFadyen and Winter (1988) as the semitendinosus activated in late swing to decelerate the extending leg (K5). The gluteus medius activated late in swing phase at the beginning of double support in an action which was hypothesized to be necessary to shift the weight of the torso to the support limb (McFadyen & Winter, 1988). During mid-swing phase, the foot underwent very little sagittal rotation with little muscular activity, leading to the possibility that the foot plantar flexion was a result of a pendulum effect with the foot as a passive mass pivoting about its point of attachment (McFadyen & Winter, 1988). Finally, during foot placement, the foot was found to be inverted slightly laterally, which appeared to be necessary for impact preparation (Greenwood & Hopkins, 1976, Freedman *et al.*, 1976, Craik *et al.*, 1982).

Stair descent has been shown to be an important area of study for rehabilitation and medical research. Some of this research has been kinetic in nature, focusing on moments and powers in an attempt to quantify the risks to joint replacements, prosthetic devices and chronic conditions such as patellofemoral pain (PFP) (Salsich *et al.*, 2001). Recent research has shown that stair descent incurred higher ground reaction forces (GRF) than level walking (up to 0.35 times body weight) and the moments can be up to 4 times higher than those incurred while walking (Stacoff *et al.*, 2005). Stair descent has also been linked to patellofemoral pain due to the (PFJ) stress occurring earlier in stance phase for PFP patients during stair descent, which may be explained by a reduction in utilized patellar contact area at lesser knee flexion angles. The smaller contact area resulted in greater peak PFJ stress during the first 60% of stair descent stance phase (Salsich *et al.*, 2001). This research outlines the problems faced by individuals descending stairs, which may not be as severe during ramp descent.

There is relatively little research currently available on the mechanics of ramp descent and of those studies, the knee was the major concern of research since the knee extensors have been identified as the primary muscles used to control the lowering of the body down the ramp (Redfern & DiPasquale, 1997). During the first 20% of stance the ankle underwent an increased dorsiflexor moment that changed to a plantar flexor moment, which increased throughout stance (Lay *et al.* 2005). The peak knee extensor moment resisted the lowering of the torso during stance and the hip moment increased in comparison with level walking and had a delayed transition to a flexor moment (Lay *et* *al.* 2005). The hip moment towards the end of stance was reported to be less than that of level walking, with a maximum hip flexion moment at 20 degrees of 0.5 N.m/kg (Redfern & DiPasquale, 1997). Of the studies examining the knee, the inclines studied were as steep as 25-degrees, which was not suitable for everyday use (Schwameder *et al.*, 1999). From this research it has been agreed that the steeper the angle the larger the joint forces and moments of force (Lay *et al.,* 2005; Schwameder *et al*, 1999; Kuster *et al.*, 1997; Redfern & DiPasquale, 1997) . At an everyday decline of 11 degrees, research showed that peak ground forces increased by 38%; peak knee flexion moments by 117%; and peak knee power by 490% (Kuster *et al.*, 1997). These large moments found by Kuster were not reported by Redfern and DiPasquale (1997), who observed maximum knee extensor moments of 1.5 N.m/kg whereas Kuster recorded 2.5 N.m/kg. There were also differences in the shape of the ankle moment. The tibiofemoral bone-on-bone forces for level walking has been found to be 4 times body weight (BW) on average, however, these forces exceed 8 times BW for downhill walking (Kuster *et al.*, 1997). Schwameder *et al.* (1999) also discovered higher knee flexion moments, at a ramp angle of 25 degrees, observing 4 times higher knee moments when compared with level walking, whereas the hip and ankle moments remained roughly the same. At toe-off, Kuster *et al.* (1995) noticed that, during swing phase, both the hip and ankle joints made angular changes which compensated for the slope. As well, the powers changed from a concentric plantar flexor effort at toe-off to an eccentric knee extensor action at the beginning of stance phase. The shear forces in downhill walking were also greater than during level walking due to the deceleration of the centre of mass in the walking direction (Schwameder *et al.*, 1999). These changes in shear GRFs as the ramp angle increased have been linked with increased potential to fall and the prolonged higher moments may be a problem for populations with strength or musculoskeletal limitations, especially in the elderly (Redfern & DiPasquale, 1997).

Larger moments and bone-on-bone forces have been of a particular concern with respect to knee replacements (Kuster *et al.*, 1997). Knee replacements are designed to deal with considerations such as contact area, load, material properties and length of time it has been implanted (Kuster *et al.*, 1997). However, the most destructive process was fatigue through repeated high loads and repetitive stressing (Kuster *et al.*, 1997). Considering that the moments of force can exceed 8 times BW during downhill walking at 11 degrees and downstairs walking can exceed 6 times BW, it may be considered potentially detrimental to knee replacements. Kuster pointed out that not only are the forces and moments higher, but the area of contact at particular joint angles of flexion was less. Therefore, higher forces were being distributed over a smaller area in the knee replacements when compared with healthy knees.

The findings in biomechanical research involving stair and ramp descent have been employed to examine and better understand the effects that descent can have on weakened joint structures and muscles. These studies have helped develop rehabilitation options for injured individuals and to establish a baseline for the analysis of pathological gait.

The research accomplished so far in ramp or stair descent has been compared exclusively to normal walking as a baseline for the subjects involved. When comparisons have been drawn between ramps and stairs, they have been comparing different populations, which would add error to the value of those conclusions. The purpose of this study was to examine the differences in the moments of force and moment powers at the joints of the lower extremity for the same subjects during ramp and stair decent and to compare stair and ramp descent with each other and with Winter's (1991) benchmark level walking data.

Methods

The sample population was taken from a group of university students who volunteered for the study. Five males and five females, between the ages of 18 and 30, were selected from volunteers. The subjects were free of any lower extremity pathology that may have interfered with normal gait. All subjects provided their written consent prior to the trials.

Table 1. Subject parameters

Subject	Sex	Age	Height (cm)	Weight (kg)	Stair Cadence (steps/min)	Ramp Cadence (steps/min)
1	F	22	163.5	60	122	107
2	M	20	184	79.5	109	124
3	M	25	165	64.1	81.8	94.8
4	M	23	173	58.2	91.1	111
5	F	21	165	61	116	129
6	M	18	170	77.3	97.3	113
7	F	21	184	79.5	109	116
8	F	21	177	54.5	113	100
9	F	22	168	58.2	103	101
10	M	22	180	77.3	109	94.7
Mean	-	21.5	172.95	66.9	105.1	109.1
Std. Dev.	-	1.86	7.5	9.6	11.5	11.5

For the trials, the subjects wore laceless shoes and dark clothing (shorts and Lycra top) that had non-reflective surfaces. It was important to have tight fitting clothing to minimize the movement of the reflective digitizing markers while the subject was in motion. The reflective markers were placed on the following locations on the left side of the body: hallux, 5th metatarsal, calcaneus, lateral malleolus, lateral femoral condyle (knee), the greater trochanter and head of the humerus. The subjects were required to wear slippers supplied to them by the gait lab to maintain a consistent footwear type.

A grid board of control points (1×1.5 m) was filmed prior to subject testing to establish a frame of reference for digitizing. Half the subjects walked at a self – selected pace down the stairs and then down the ramp for five trials, while recorded by the camera. The other half of the subjects walked at a self – selected pace down the ramp and then the stairs, to minimize order effects that may have affected their gait patterns. The subjects completed the ramp and stair gait trials in the same recording session.

The ramp (figure 3) used in this study was a solid structure composed of iron and was set at a fixed angle of 10 degrees. The surface of the ramp was composed of $1\frac{3}{8}^{th}$ inch plywood, which allowed for a section to be taken out so that the force plate could lie flush with the walkway. Before data collection, the subjects descended the ramp several times to become comfortable with stepping on the force plate and with the inclination of the ramp. The subjects started walking down the ramp from an elevated box at the top of the ramp. When walking down the ramp the subjects completed the minimum of one full stride cycle. The ramp was covered with a non-slip surface to ensure the safety and to prevent slipping.

The stairs (figure 4) consisted of 3 steps of 20.0 cm rise and 30.0 cm tread each of which were wide enough to hold a force platform that could be placed on any step using a special bracket. Before any trials were recorded, the subjects were permitted to practice the step down procedure so that they became comfortable with the stairway. The stair trials were conducted with the subject stepping from the top step and walking down the stairs as they would normally. This meant the subject took a full step down to the force plate that was fixed to the last step (step 3) before stepping to the ground. This ensured that the subject engaged in normal stair descent gait for the force recording.

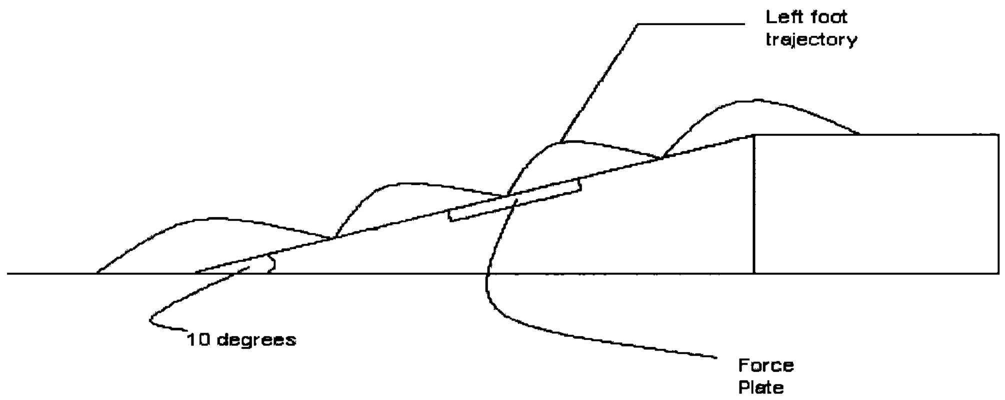

Figure 3. Ramp design

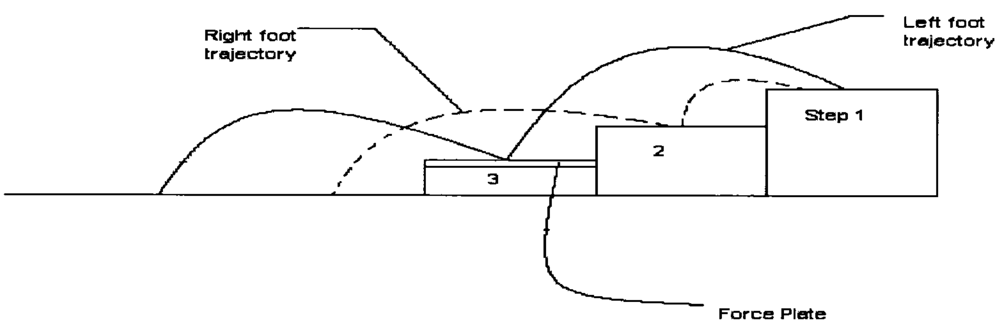

Figure 4. Stair Design

A Kistler Slimline (9286AA) force platform was placed on the ramp allowing for the ground reaction forces of one step to be captured per trial. The force plate used for the stair trials was the Kistler Inc. 9281C, which was placed on the stairs in such a way that allowed for the subject to contact the force plate two steps down. The force plate was sampled at 240 Hz. A computer using the BioWare software for force plate collection recorded the reaction forces from the plate. The video was recorded using a VHS camera sampling at 60 Hz. The camera recorded the left side sagittal view of the subjects while they descended the ramp and the stairs.

All force data were collected by a computer connected to the force plate. The data analysis of the force recordings was acquired using the BioWare and BioProc2 programs. The VHS data was captured using APAS (Ariel Performance Analysis Software) software for the marker trajectories. The Biomech Motion Analysis System combined the force recordings to the APAS digitized marker trajectories for kinematic and kinetics analysis. From the computer analysis, the moments and powers produced by the ankle knee and hip for stair and ramp descent were calculated.

To determine whether significant differences occurred among the patterns of the moments of forces and their associated powers, each subject's data was ensemble averaged and normalized to body mass. Differences between stair or ramp descent were statistically tested whenever the ensemble averages differed by more than +/- one standard deviations. Significant differences between the peak powers in the hip, knee and ankle for ramp and descent were tested using a t-test with alpha set to 0.05.

Results

The moments and powers histories are presented in figures 5 and 6. Figure 5 shows the comparisons of the grand ensemble averages of the moments and powers for the hip, knee and ankle joints for level walking, ramp descent and stair descent in stride time. It should be noted that the ordinal scaling is the same for the moments of force but different for the powers. Figure 6 shows the comparisons of the grand ensemble averages of the moments and powers for the hip, knee and ankle joints for ramp and stair descent with +/-1 SD to show variability in percentage of stride.

Stair Descent

Ankle. The ankle plantar flexor moment was characterized by two roughly equal peaks during stance (Figure 5). The first occurred early in stance (0.05s) and the second late in stance (0.5s). The peak ankle plantar flexor moment of 1.159 N.m/kg occurred at 0.08s, followed by a smaller second peak of 1.017 N.m/kg occurred at 0.5s (figure 5). The stair descent peak ankle moment was 118% larger than the ramp ankle moment peak, which occurred at 0.5s (figure 5). The ankle powers during stair descent showed a large eccentric plantar flexor power of –5.3 W/kg (A0) just after FC (0.07s, Figure 5). A second concentric plantar flexor power peak (A2) was present at 56% at push-off (0.8 W/kg). The peak ankle power (0.07s) was 320% larger than the peak power during ramp descent (0.5s). The A2 power burst in ramp and stair descent was not significantly different ($p = 0.074$).

Knee. The knee moment was extensor throughout most of stance, with a brief (0 - 0.1s) period of flexor activity (Figure 5). The peak knee extensor moment was at 0.5s with a value of 0.7 N.m/kg (figure 5). The knee moments had two other peaks, one at 0.05s (–0.4 N.m/kg) and one at 0.15s (0.5 N.m/kg). The knee powers during stair descent had three major peaks, two eccentric extensor peaks at 0.1s (K1, –0.7 W/kg) and 0.5s (K3, –2.3 W/kg) and one concentric extensor peak at 0.1s (K2, 0.7 W/kg) (Figure 5). There was also a small concentric flexor peak after FC. The peak knee power occurs just before TO at 0.5s (–2.3 W/kg). The knee powers were not statistically different (K1, $p = 0.217$ and K3, $p = 0.983$).

Hip. The hip moment had one brief concentric extensor peak (0.5 N.m/kg) early in stance (0.05s) and a concentric extensor peak at push-off (figure 5). The hip powers for stair descent showed a small concentric extensor peak (H1) of 0.2 W/kg at FC (0.05s). The peak hip power (H3) occurred at 0.6s, just before TO with a value of 0.4 W/kg (figure 5).

Ramp Descent

Ankle. Excepting for a brief period of dorsiflexion (0 - 0.1s), the ankle moment was plantar flexor throughout stance phase, characterized by a single peak (0.9 N.m/kg) near TO (0.5s) (Figure 5). The ankle powers during ramp descent followed an eccentric plantar flexor trend until push-off where there was a large concentric peak of 1.7 W/kg at 0.5 (A2, Figure 5).

Knee. The knee moment was extensor with two roughly equal peaks at 0.15s and 0.5s (0.8 N.m/kg and 0.75 N.m/kg). The peak moment for ramp descent (0.8 N.m/kg) was 117% larger than the peak knee moment in stair descent. The knee powers followed a similar eccentric and concentric extensor trend as stair descent (Figure 5). There were two eccentric extensor peaks at 0.1s (K1, –1.2 W/kg) and 0.5s (K3, –2.3 W/kg), with a small concentric extensor peak at 0.2s (K2). The peak knee power occurred just before TO and was 105% of the peak knee power in stair descent (at 0.5s, –2.3 W/kg).

Hip. The hip moment was mostly flexor throughout stance, except for the first 0.1s, which was extensor (Figure 5). The flexor moment increased until near TO where it peaked at 0.8 N.m/kg (0.52s) which was 176% larger in ramp descent than in stair descent. The hip powers during ramp descent followed a similar trend to walking, except for an initial concentric extensor powers burst just after foot - strike (H1, figure 5). There were two peaks, the first was an eccentric flexor peak (H2) at 0.5s of –0.8 W/kg. The second concentric flexor peak (H3) occurred just before push - off at 0.6s with a value of

1.0 W/kg (figure 5). The peak hip power for ramp descent was 231% of the peak hip power for stair descent (0.6s, at 0.4 W/kg). The H3 concentric power bursts were significantly larger in ramp descent ($p = 0.043$) than stair descent.

Variability. The results in figure 6 show a high degree of variability in both ramp and stair descent. These results differ from Winter's (1991) walking data, which showed a very small variability, due to his large number of subjects and a controlled speed of gait. In the present study of stair and ramp descent, the variability was much larger due to the small sample size and the self-selected pace of the subjects. It was also possible that the methods of descent differ between individuals more than walking.

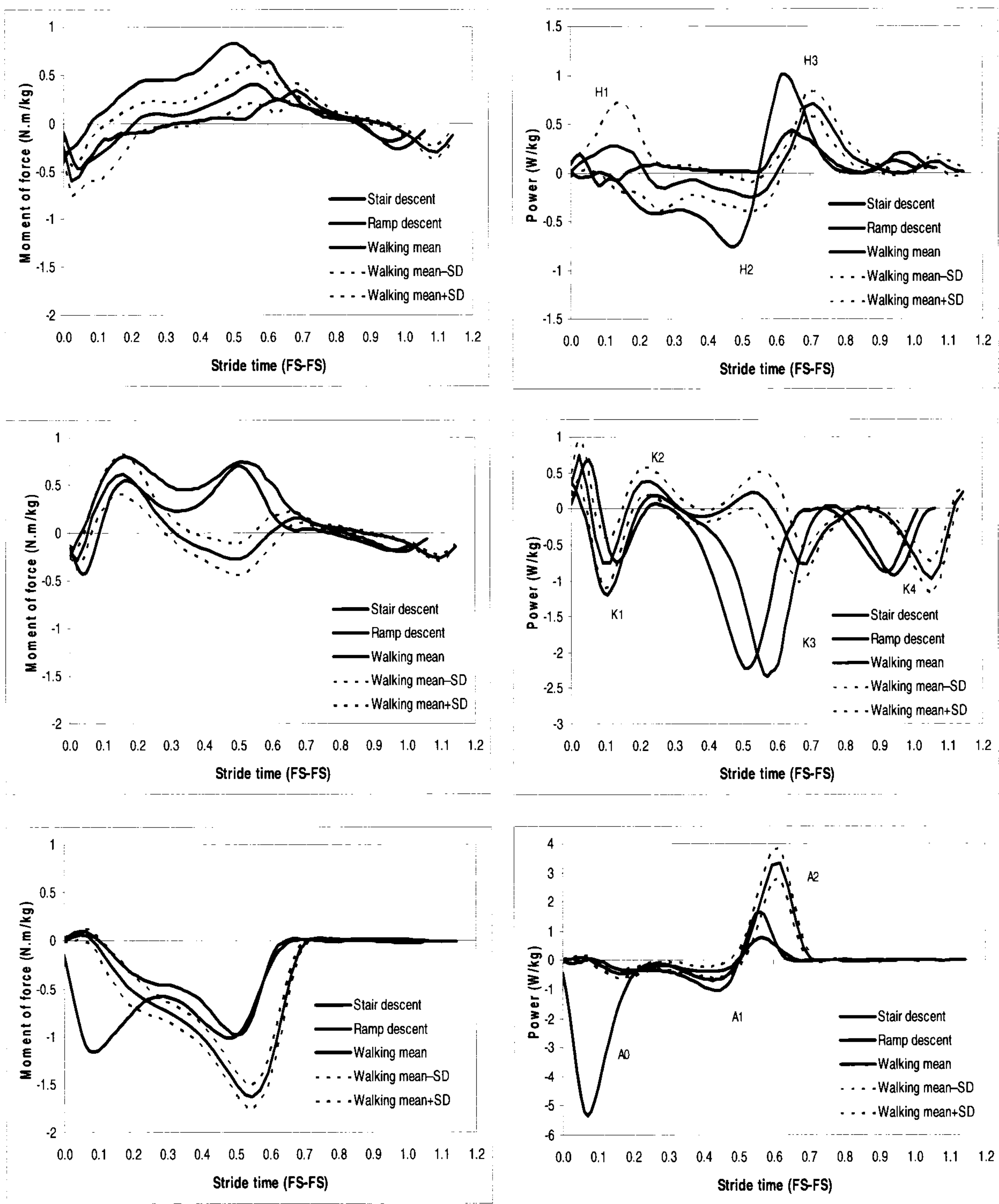

Figure 5. Comparisons of the grand ensemble averages of the moments (left) and powers (right) of the hip (top), knee (middle) and ankle (bottom) joints during level walking, stair descent and ramp descent. Note ordinal scaling is the same for the moments of force but different for the powers. Data were normalized to body mass and average stride time. Dotted lines are the mean +/– 1 SD of the level walking data from Winter (1991).

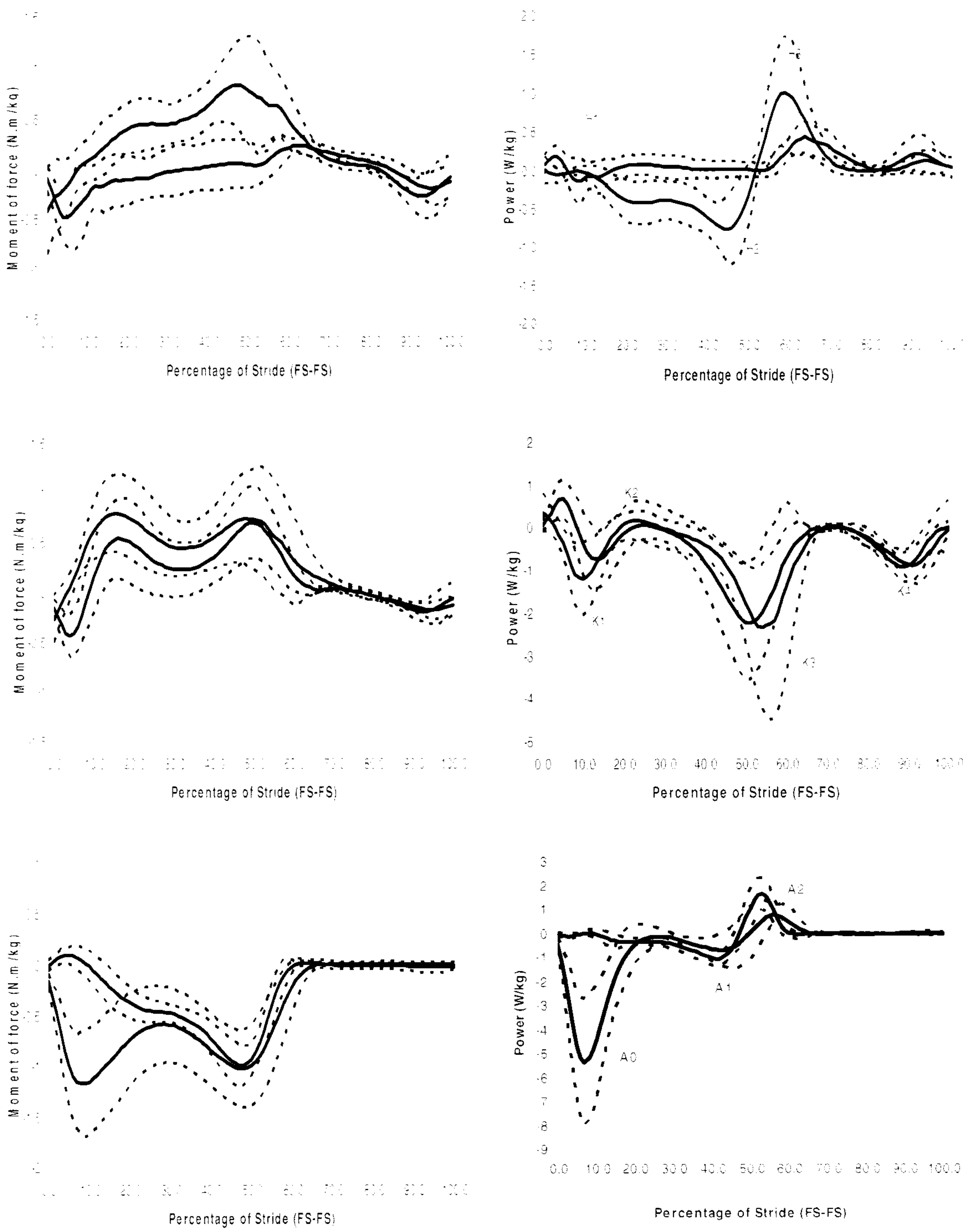

Figure 6. Comparisons of the grand ensemble averages of the moments (left) and powers (right) of the hip (top), knee (middle) and ankle (bottom) joints during stair descent (blue line) and ramp descent (green line). Data were normalized to body mass and average percentage of stride. Dotted lines are the means +/– 1 SD; solid lines are the means.

Discussion

The purpose of this study was to identify the typical patterns of the moments of force and powers at the hip, knee and ankle for ramp and stair descent and compare the two. The moments of force and powers found in stair descent in this study are consistent with those found by McFadyen and Winter (1988). The moment at the ankle was similar to their results in that the moment showed two plantar flexor peaks, one at IC and one just before push-off. The moment of force curves at the knee joint showed a typical double hump, indicating the activation of the knee extensors working eccentrically to support the body as it descends to the next step. The moment at the hip was small but is consistent with McFadyen and Winter's (1988) data, confirming that the hip contributes fairly little to the work of lowering the body.

The powers at the hip show similar trends to McFadyen and Winter's (1988) study, with a similar concentric extensor power burst at H1, to help extend the leg to position for IC and a concentric flexor burst at H3 to help lift the leg off the upper step. The knee powers showed two major bursts of power absorption during stance at K1 and K3. This is consistent with the findings by McFadyen and Winter (1988), indicating K1 was due primarily to absorption of energy as the foot makes contact with the lower step followed by a brief period of generation (K2). The last peak of energy absorption (K3) was interpreted as a controlled lowering of the torso over the step by the knee extensors. The power activity at the ankle was also consistent with previous studies (McFadyen & Winter, 1988), showing absorption by the plantar flexors, at A0, which aids the knee (K1) to slow the descent of the torso at IC (McFadyen & Winter, 1988). The second important power peak comes at TO (A2), showing the activation of the ankle plantar flexors to push-off, however, this peak was small and is felt to assist progression of the step, but not the main source of propulsion. This ankle activity has also been postulated to be used mainly to alleviate the aggressive dorsiflexion of the foot as the torso descends to the next step (McFadyen and Winter, 1988).

There are few studies which have been conducted with which to compare the ramp descent data. The most pertinent of those would be work done by Kuster *et al.* (1997); Redfern & DiPasquale, (1997); Schwameder *et al.* (1999) and Lay *et al.* (2005). All researchers noted a larger moment and power for the knee during ramp descent, however, Schwameder (1999) used an angle above 10 degrees and showed that the moments and powers increased proportionately with the angle of ramp (Redfern & DiPasquale, 1997; Schwameder *et al.*, 1999). The moment of force about the hip exhibit a similarity with that of Redfern & DiPasquale (1997) and Lay *et al.* (2005), indicating that the hip flexors were used eccentrically throughout stance to slow the descent of the torso and during push-off to provide the energy for the swing phase. The knee and ankle moments were both similar to those of Redfern & DiPasquale (1997) and Lay *et al.* (2005), showing a similar curve of two peaks for the knee as the extensors acted eccentrically to lower the torso down the ramp. However, it should be noted that Redfern & DiPasquale (1997) did not have a well-defined 2nd "peak", which may be explained by the slightly larger decline of 15 degrees. The ankle moment was primarily extensor, with a concentric peak during push-off that was similar to Redfern & DiPasquale, (1997) and Lay *et al.*'s (2005) data. This indicates that the ankle absorbed energy as the body descended the ramp, then at push-off reversed to aid the foot to begin to drive the lower leg forward as in level gait (Winter, 1991).

The powers generated at the hip during ramp descent differ from those found in walking and stair descent due to the hip flexors activating to control hip extension. The large concentric hip flexor power burst (H3) is larger than Winter's (1991) walking data, showing that the hip flexors activated to help lift the leg and propel it forwards, more than what would normally be required for level gait, most likely due to a need to lift the foot from a higher position and swing it through to a lower IC. The knee powers for ramp descent are much larger at K3 than level walking, similar to Kuster *et al.*'s (1997) data. The knee goes through similar phases of absorption and generation of power as walking, except for K3 where the knee extensors are required to activate to control the rate of descent of the torso just before TO, similar to stair descent. The ankle powers in ramp descent are similar to those in level gait showing a similar, but smaller, power burst at A2 indicating that the ankle plantar flexors could be acting passively to alleviate the extreme dorsiflexion much like during stair descent (McFadyen & Winter, 1988).

There are large differences between ramp and stair descent for the hip, knee and ankle moments. For example, the moments of force were larger for the hip and knee in ramp descent while the eccentric extensor ankle power at IC was larger for stair descent (A0), which shows larger eccentric plantar flexor activity. This plantar flexor activity is required in stair descent to help lower the torso down onto the next step (McFadyen & Winter, 1988), an activity that is not necessary in ramp descent because there is more forward momentum and not as much of a vertical component to the motion. The reverse is true at push-off where ramp descent has a slightly larger concentric ankle plantar flexor peak, while during stair descent the ankle plantar flexors are acting mostly to relieve some dorsiflexion. Note that while many of the powers were similar, they occurred for different reasons. The stair descent absorptions were due to vertical lowering of the body while the ramp-descent, power absorptions were for a combination of vertical lowering and deceleration of the forward momentum of the body down the ramp.

These results are significant to those with hip, knee or ankle injuries or pathologies. This research has shown that ramp descent is more strenuous on the knee and hip but stair descent would be more strenuous at the ankle at foot-strike. These results may be used to advise patients to avoid stairs or ramps until their injuries heal.

Conclusions

Based on the results obtained, the following conclusions are warranted. This research demonstrated that significant differences exist with the moments and powers of the lower extremity during level gait, stair descent and ramp descent. The patterns of the moments and powers were clearly different between stair and ramp descent. In particular, the ankle moment at push-off (A2) is larger in level gait than during stair and ramp descent. The knee moment and power at push-off and early swing phase are significantly smaller in direction and peak value (K3) in level gait. The hip moment was larger in ramp descent during stance phase (H2, H3), and less so for stair descent.

When comparing ramps and stair descent, significant differences were found. The ankle moment powers during stair descent (A0) were larger as the ankle plantar flexors performed a large burst of negative work at weight acceptance. No significant differences occurred for eccentric or concentric powers at push-off (A1 or A2); furthermore, there were no significant differences between the knee moments or powers, although the ramp had a slightly higher eccentric extensor power burst (K3). Finally and most significantly, the hip flexor moment for ramp descent was significantly larger than in stair descent for the eccentric power burst during mid-stance (H2) and concentric power at push-off (H3).

REFERENCES

Atlas, R. (1987). Ramps, Railings, and Rounded Edges. *Florida Architect Journal*, Nov/Dec.

Beaulieu, F.D., Pelland, L., & Robertson, D.G.E. (2003). Descending stairs, forwards and backwards. *Proceedings International Society of Biomechanics Congress XIX.* Dunedin, New Zealand.

Blake, R.L. & Ferguson, H.L. (1993). Walking and hiking injuries: A one-year follow-up study. *Journal of the American Podiatric Medicine Association, 83*, 499-503.

Catani, F., Benedetti, M.G., De Felice, R., Buzzi, R., Gianni, S., & Aglietti, P. (2003). Mobile and fixed bearing total knee prosthesis functional comparison during stair climbing. *Clinical Biomechanics, 18*, 410-418.

Cham, R. & Redfern, M.S. (2002). Changes in gait when anticipating slippery floors. *Gait and Posture, 15*, 159-171.

Craik, R., Cozzens, B., & Freedom, W., (1982). The role of sensory conflict on stair descent performance in humans. *Experimental Brain Research, 45*, 399-409.

De Loes, M. (1995). Epidemiology of sports injuries in the Swiss organization 'Youth and Sports' 1987-1989. *International Journal of Sports Medicine, 16*, 134-138.

Freedman, W., Wannstedt, G., & Herman, R. (1976). EMG patterns and forces developed during step-down. *American Journal of Physical Medicine, 55*, 275-290.

Joseph, J. & Watson, R. (1967). Telemetering electromyography of muscles used in walking up and down stairs. *The Journal of Bone and Joint Surgery, 49*, 774-780.

Kaufman, K.K., Hughes, C., Morrey, B.F., Morrey, M., & An, K. (2001). Gait characteristics of patients with knee osteoarthritis. *Journal of Biomechanics, 34*, 907-915.

Kuster, M., Sakurai, S., & Wood, G.A. (1995). Kinematic and kinetic comparison of downhill and level walking. *Clinical Biomechanics, 10*, 79-84.

Kuster, M.S., Wood, G.A., Stachowiak, G.W., & Gachter, A. (1997). Joint Load Considerations in total knee replacements. *Journal of Bone and Joint Surgery, 79,* 109-113.

Lay, A.N., Haas, C.J., & Gregor, R.J., (2005, in press). The effects of sloped surfaces on locomotion: a kinematic and kinetic analysis. *Journal of Biomechanics*.

McFadyen, B.J. & Winter, D.A., (1988). An integrated biomechanical analysis of normal stair ascent and descent. *Journal of Biomechanics, 21 (9)*, 733-744.

Greenwood, R. & Hopkins, A., (1976). Landing from an unexpected fall and a voluntary step. *Brain, 99*, 375-386.

Redfern, M.S. & DiPasquale, J. (1997). Biomechanics of descending ramps. *Gait and Posture, 6(2)*, 119-125.

Saari, T., Tranberg, R., Ziigner, R., Uvehammer, J., & Karrholm, J. (2004). Total knee replacement influences both knee and hip joint kinematics during stair climbing. *International Orthopaedics, 28*, 82-86.

Salsich, G.B., Brechter, J.H., & Powers, C.M. (2001). Lower extremity kinetics during stair ambulation in patients with and without patellofemoral pain. *Clinical Biomechanics, 16*, 906-912.

Schwameder, H., Roithner, R., Muller, E., Niessen, W., & Raschner, C. (1999). Knee joint forces during downhill walking with hiking poles. *Journal of Sports Sciences, 17*, 969-978.

Stacoff, A., Diezi, C., Luder, G., Stussi, E., & Kramers-de Quervain I.A. (2005). Ground reaction forces on stairs: effects of stair inclination and age. *Gait and Posture, 21*, 24-38.

Thambyah, A., Palaniappan, T., & Goh Cho Hong, J. (2004). Knee joint moments during stair climbing of patients with anterior cruciate ligament deficiency. *Clinical Biomechanics, 19*, 489-496.

Winter, D.A. (1991). *The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological* (2nd ed.). Waterloo, ON: University of Waterloo Press.