BIOMECHANICAL AND PHYSICAL REQUIREMENTS OF STAIR NEGOTIATION WITH RESPECT TO AGING AND STROKE

by

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Abstract

The ability to safely and efficiently negotiate stairs is an essential skill for independent ambulation. To date, basic research to identify biomechanical and physical costs is limited in older adults. In persons with stroke, this aspect of mobility is virtually unexplored. The main objective of this thesis was to investigate biomechanical alterations during stair negotiation and to evaluate the physical costs of the task in older adults and persons with stroke. This was approached by conducting four studies. The first study identified age-related alterations in joint kinetics during stair negotiation. The results showed age-related differences in moment magnitudes, an exaggerated net support moment and sustained abductor moments through stance. To gain insight into these adaptive changes with respect to mechanical efficiency, the second study evaluated age-related changes in mechanical energy transfers during stair negotiation. During ascent, older adults achieve similar efficiencies as young adults by slowing their cadence. During descent, age-related differences in mechanical energy expenditures and related variances in mechanical energy compensation coefficients reflect a loss in mechanical efficiency. The impact was likely the provision of enhanced extensor support and stability. The results also highlight a functional role for concentric energy expenditures during descent. The third study provided a detailed biomechanical description of stair negotiation in people with stroke, revealing important differences in how stroke survivors manage stairs and how handrail use modifies the magnitudes of lower limb joint moments. The fourth study evaluated the strength and aerobic requirement of stair ambulation in persons with stroke. The findings reveal increased costs of the task, primarily due to reduced neuromuscular and aerobic capacities and serve to identify factors that may be limiting during stair negotiation.
This thesis provides new information regarding movement control in older adults during stair negotiation, providing a normative benchmark of age-related alterations in movement patterns. In persons with stroke, this work is the first to quantify the biomechanical patterns and physical requirements of stair negotiation. Future work may extend these findings to explore mobility challenges in persons with greater levels of impairment as well as guide the development of targeted and task oriented rehabilitation programs.
Co-Authorship

This dissertation contains material from several published works (chapters 2 & 3), a manuscript accepted for publication (chapter 4) and submitted manuscripts (chapters 5 & 6). The intended authorship is as follows:


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Abbreviations

Abd  Abduction
Add  Adduction
Aff  Affected side
CBM  Community Balance and Mobility Scale
D    Dominant side
DF   Dorsiflexion
Ev   Eversion
Ext  Extension
Flex Flexion
H    Handrail
Inv  Inversion
LAff Less affected side
MEE  Mechanical energy expenditure
MEC  Mechanical energy compensation
mVO$_2$ Maximal VO$_2$
ND   Non-dominant side
NH   No handrail
PF   Plantarflexion
RER  Respiratory exchange ratio
ROM  Range of motion
Chapter 1

Introduction

Stair negotiation is an essential skill for independent ambulation and community accessibility that is described by older adults as one of the more challenging activities of daily living [1,2] with associated high risk for falls and injury particularly during descent [1]. This is not surprising as previous research has shown significantly greater biomechanical and neuromuscular demands of stair ascent and descent when compared to level ground walking [3-5]. Consequently, adaptations during stair negotiation frequently occur in association with normal aging as strength and joint mobility decline. It follows that if physical impairment is superimposed on normal aging, the ability to negotiate stairs safely could be seriously compromised, which in turn could be a critical factor in the loss of independence in older adults [6]. Understanding the normal, age-related alterations in movement control during functional tasks such as stair negotiation is important to provide a benchmark against which rehabilitation professionals working with a wide range of people with physical limitations can gauge mobility.

Making use of gait analysis tools, the biomechanical demands of stair ascent and descent have been described to varying extents in healthy, young adults [5,7-12]. Fewer studies have reported the mechanics of the task in older adults. Of those, most are limited to the sagittal plane [13-15], or one or two lower limb joints [13,15-17]. Indeed, in a recent review of stair negotiation in older adults, Startzell et al. [1] indicated that more research is needed to identify key determinants of difficulty and safety on stairs. There is a need for a comprehensive biomechanical analysis of stair negotiation in older adults. Such analysis would aid in identifying movement alterations or compensations in older adults which may put them at risk of falling and injury, or
may contribute to inefficiency and ultimately increased energy costs of the task. Further, for rehabilitation professionals, it is important to appreciate the nature and extent of adaptations during stair negotiation as a natural progression with aging in order to be able to identify unique alterations in kinematic or kinetic patterns attributable to impairment rather than aging.

In persons with stroke, in particular, loss of independent ambulation is one of the most disabling aspects [18] and stair negotiation is considered a primary factor contributing to the attainment of community ambulation. However, to date, no published studies have characterized the biomechanics of stair negotiation in persons with stroke. Further to this, it has been previously highlighted that the ability to negotiate stairs is a function of multiple factors, including strength, joint mobility, physical conditioning and coordination between lower limb joints [1,4]. In persons with stroke, reductions in strength and joint mobility are common and contribute to significant limitations in overground walking [1,19-23]. In addition, up to 75% of stroke survivors have cardiovascular disease [24] contributing to reduced physical activity and deconditioning [25]. The metabolic cost of climbing stairs is approximately three times as high as walking over level ground in healthy individuals [1,26], which when paired with the higher metabolic demand of level walking in stroke [27], suggests that the higher energy costs of ambulating stairs may be limiting. Further, the challenge of stair negotiation can present as a barrier to active community living in persons with stroke, potentially leading to declines in physical function, independence and ultimately health-related quality of life.

In older adults the literature is limited and in stroke survivors we know even less about the physical requirements of stair ambulation. Knowledge of the lower limb mechanics, strength and metabolic costs is essential to characterize the requirements of stair ascent and descent and understand compensatory behaviors related to healthy aging, and is an important contribution to
understanding the mobility challenges following stroke.

1.1 Dissertation outline and intent

The overall focus of this thesis was to evaluate the biomechanical and physical (including neuromuscular and aerobic) requirements of stair ascent and descent related to normal, healthy aging and impairment associated with stroke. To address this, a comprehensive literature review was undertaken to portray the current state of knowledge (Chapter 2). This provided the rationale for the approach of conducting four studies (Chapters 3-6). The first two studies (Chapter 3 & 4) focus on biomechanical alterations and mechanical efficiency related to healthy aging. Chapter 5 provides an in-depth analysis of the kinematic and kinetic profiles of stair ascent and descent in persons with chronic stroke. The final study (Chapter 6) determines the strength and aerobic requirements of stair negotiation in healthy older adults and stroke survivors relative to a standard evaluation of maximum capacity. The dissertation concludes with a summary and general discussion in Chapter 7, including future directions for rehabilitation research and clinical practice.
1.2 References


Chapter 2

Literature Review

Portions of the literature review have been previously published:


Independent mobility involves navigating over changing terrain, obstacle avoidance, frequent modulation of speed and direction and stair negotiation. Each task imposes different biomechanical and neuromuscular system demands and the ability to meet them is related to functional capacity or peak system performance. For example, in the stance phase of level walking, one of the demands is to support the body against gravity. To meet this demand the hip, knee and ankle work together to generate an adequate lower limb extension moment. Healthy individuals have the muscular strength to generate the required extension moment and they have ample reserve strength to accommodate higher demands should they be required (for example, walking at a faster pace [1]). As individuals age, mobility and strength decline, reducing biomechanical and neuromuscular system capacities, respectively. This results in altered or adapted movement patterns during ambulatory tasks [2]. For more functionally demanding tasks, the implications of decreased capacity can be significant, particularly if community access and independence are affected.

Older adults have identified stair negotiation as one of the most difficult tasks attributable to aging and is one of the leading causes of fall related injuries [3-5]. In a sample of 310
nondisabled older adults, more than 45% reported difficulties in climbing stairs, and about 30% reported difficulties in stair descent [6]. Further, stair descent is often associated with increased risk of falls and self-reported falls has been shown to be higher during stair descent compared to stair ascent in able-bodied individuals [7]. If physical impairment is superimposed on normal aging, the ability to negotiate stairs safely could be seriously compromised, which in turn could be a critical factor in the loss of independence in older adults [8]. Understanding the normal, age-related alterations in movement control during functional tasks such as stair negotiation is important to provide a benchmark against which rehabilitation professionals working with a wide range of people with physical limitations can gauge mobility.

2.1 Gait cycle of stair ascent and descent

During stair negotiation, as during walking, the legs move in a cyclical pattern. The cycle for both ascent and descent is divided into two distinct phases: the stance phase and the swing phase (Figure 2-1) [9-11]. In ascent, the stance phase has three sub-phases: weight acceptance (shifting the body into an optimal position to be pulled up); pull-up (progression to full support on the next step); and forward continuance (ascent of a step has been completed and progression continues). The swing phase is subdivided into two sub-phases: foot clearance (the leg is raised to clear of the intermediate step); and foot placement (the swing leg is positioned for foot placement on the next step) (Figure 2-1a).
Figure 2-1 A schematic of the gait cycle of step over step stair ascent (a) and stair descent (b). (from Novak et al. [12], with permission)
The stance phase of descent is divided into three sub-phases: weight acceptance; forward continuance (the start of single limb support and forward body movement); and controlled lowering (the body’s mass is lowered onto the support limb) [9-11]. The swing phase has two sub-phases: leg pull through (the swing limb is pulled forward); and foot placement [9-11] (Figure 2-1b). During step over step ascent and descent, brief periods of double support occur at the transition points as one limb moves into swing and the other into stance.

2.2 The kinematics of stair negotiation

As with walking, the greatest range of motion (ROM) required at lower limb joints occurs in the plane of progression (sagittal) [9,11,13-15]. The ROM associated with stair negotiation, however, is greater than that associated with level walking and particularly so during ascent compared to descent. In general, upwards of 10° to 20° of additional mobility is needed at each lower limb joint in healthy adults (Table 2-1) when compared to walking [14-16]. It follows that limitations in joint ROM may impact the ability to go up and down stairs, even if they do not manifest any detrimental effect on level ground walking.

Table 2-1  Comparison of maximum sagittal plane angular displacements (in degrees) required for walking and stair negotiation* over the entire cycle (stance and swing phases)

<table>
<thead>
<tr>
<th>Joint</th>
<th>Walking</th>
<th>Stair ascent</th>
<th>Stair descent</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Flexion</td>
<td>Extension</td>
<td>Flexion</td>
</tr>
</tbody>
</table>

* Standard stair dimensions (rise/run) apply to stair ascent and descent values

Compared to motion in the sagittal plane, frontal plane movement is rarely examined; only a few studies report the kinematics during stair ascent and descent [13,14,18,19]. The angular motion required in the frontal plane (abduction/adduction) is relatively small, only about...
15° ROM at the ankle and less than 10° ROM at the knee and hip joints [13,14,18] which may explain why it is frequently overlooked. However, it is important to note that when joint mobility is compromised limiting the amount of flexion and extension, that step clearance is often achieved by exaggerated hip abduction allowing the leg to swing out to the side [19].

Researchers [19] have reported that healthy, older adults show reduced sagittal plane movement at the ankle and knee compared to young adults when descending stairs, which is accompanied by increased frontal plane motion at the hip and pelvis. When climbing stairs, age-related reductions in ankle dorsiflexion could result in a trip if the toes fail to clear the step [20], although individuals may take advantage of compensatory movements such as hip flexion or abduction to successfully negotiate the step. There is ample evidence that reduced joint mobility poses risks for stair negotiation [4], but the compensatory strategies adopted to enable stair ascent and descent also can introduce safety concerns. Excessive motion in the frontal plane can cause significant shifts in the location of the body’s centre of mass relative to the small base of support. This requires adequate control and muscle strength to maintain stability, which are known to decline with increasing age [21] and likely contribute to a reduction in the ability to safely ascend and descend stairs [19].

2.3 The kinetics of stair negotiation

2.3.1 Joint moments

Internal moments of force are generated to counter the external forces acting on the body throughout locomotion and represent the net effect of agonist and antagonist muscle activity acting at a specific joint. The algebraic sum of the lower limb joint moments is termed the net support moment, which must be extensor to prevent lower limb collapse during stance. The support moment is shown to be consistent between subjects even when individual joint moment
profiles vary [10,22]. The investigation of the individual joint moments and the relative support moment provides insight into the motor control strategies used during stair negotiation that cause the observed movement patterns.

Stair descent requires largely eccentric muscle contractions to control the lowering of the body, whereas concentric contractions of lower limb muscles predominate during stair ascent to lift the body’s mass against gravity [9,10,13,17]. During the stance phases of ascent and descent, much of the work is done by the knee and ankle extensors (in the sagittal plane), while the hip abductors (frontal plane) serve to control lateral movement of the trunk and pelvis [14,17]. It is important to note, however, that the staircase dimensions (rise and run) will affect the task demands including the degree of muscle activation required [15]. During stair ascent, studies in healthy young and middle aged adults have indicated that knee extensor moments are approximately 12% to 25% greater than those generated during level walking, corresponding to peak values ranging from 0.76 Nm/kg to 1.50 Nm/kg during ascent [9,10,13-15]. The higher demands of stair negotiation therefore requires greater knee extensor strength than walking, which can have negative consequences such as fatigue and instability if the additional muscle force required begins to approach the maximum strength available [23,24].

At the ankle, a plantar flexion moment predominates reaching a maximum magnitude during late stance as the body is raised upward and forward to complete the rise onto the step [10,17]. During stair descent, a plantarflexion moment of similar magnitude (1.1 Nm/kg to 1.4 Nm/kg) is also generated, but eccentrically during weight acceptance [10,15,17]. In stair climbing as with walking, the plantarflexors are main contributors to the work produced to complete the task [9,10,17], including the maintenance of upright support (the net support moment) [10]. This contrasts with the knee extensors, which do not play a dominant role in walking but do during
both stair ascent and stair descent [14,15,18]. A hip extensor moment is also evident in early stance during ascent [10,13,17], and to a lesser degree during stair descent (maximum hip extensor moments of 0.76 ± 0.19 Nm/kg during ascent compared to 0.52 ± 0.19 Nm/kg during descent [17]). It is noteworthy that the demands on the hip muscles can be quite variable in response to even small alterations in trunk position [10,17]. For example, antero-posterior shifts in trunk position during ascent or descent require compensating hip extensor or flexor moments to maintain dynamic stability of the combined mass of the head, arms and trunk over the base of support [25]; a challenging prospect in the presence of instability and muscle weakness.

Beyond 50 years of age, muscle strength declines at a rate of about 10% per decade [21]. In terms of mobility, the impact would likely be negligible for many years with respect to walking because the strength requirements to accomplish the task are quite low relative to the force-generating capacity of the muscles. However, the same may not be the case with stair negotiation. Reeves et al. [23,24] reported that healthy, older adults adopted alternate kinetic strategies at the ankle and knee relative to their younger counterparts as a means of recalibrating muscle workloads to within their comfortable limits while still meeting the demands of stair ascent and descent.

During stair ascent, older adults showed reduced ankle plantarflexor moments compared to young adults (1.24±.21 Nm/kg and 1.48±.27 Nm/kg, respectively) [24]. Similarly, older adults demonstrated lower magnitude knee extensor moments than young adults (0.89±22 Nm/kg and 1.19±.24 Nm/kg, respectively) [24]. This is notable considering that there was no significant difference in cadence between older and young adults, suggesting that the hip extensors likely contribute to performing the work of the task to a greater extent in older than younger adults. The hip moments, however were not measured.
Relative to the maximum strength capacity, both young and older adults worked at an estimated 90% of their plantarflexor limit, whereas older adults operated a higher level of knee extensor output (75%) than young adults (53%) during stair ascent [24]. These data suggest that older adults require a greater intensity of effort, leaving less reserve muscle strength capacity, which may compromise their comfort and perceived stability while climbing stairs.

The pattern was similar for stair descent. Older adults generated lower peak ankle plantarflexor moments than young adults (1.03 Nm/kg and 1.32 Nm/kg, respectively); a strategy enabling older adults to operate at a similar relative proportion of their maximum capacity compared to young adults (about 75%) [23]. In contrast, older adults generated knee extensor moments of a similar magnitude as those of young adults (0.83 Nm/kg and 0.91 Nm/kg, respectively) such that they performed at a higher proportion of their maximal capacity (42%) than their younger counterparts (30%) [23]. The authors concluded that older adults redistribute the relative extensor moment outputs at the knee and ankle joints (they did not study the hip) as a strategy to keep the costs within their physical capabilities and safe limits [23] during stair descent.

2.3.2 Joint powers (mechanical energies)

For rotational motion, that is, motion about a joint axis, the power of the proximal ($P_p$) and distal ($P_d$) ends of the distal and proximal articulating segments, respectively is calculated as the product of the muscle moment ($M_j$, where $M_j^p = -M_j^d$) and the segment’s angular velocity ($\omega$). At a given joint (j), the net muscle power ($P_j$, Watts/kg) is the summation of the power at the endpoints of adjoining segments:
\[ P_j = P_p + P_d = M_j^p \omega_p + M_j^d \omega_d \]

Where,  
- \( P = \) muscle power, Watts/kg  
- \( M = \) muscle moment, N·m/kg  
- \( \omega = \) angular velocity, rads/sec

The sign and relative magnitudes of the power terms indicate whether the muscles generate or absorb energy and whether the energy is transferred across segments [26-30]. When segments on either side of the joint rotate in opposite directions, muscles either generate mechanical energy (via concentric contractions) or absorb mechanical energy (via eccentric contractions) entirely. If the adjoining segments rotate in the same direction, energy transfers can be directed proximally \( (P_d < 0 \text{ and } P_p > 0) \) or distally \( (P_d > 0 \text{ and } P_p < 0) \) to facilitate muscular effort [26-30].

During stair negotiation only a few studies have reported joint powers. Of these, the data reported are limited to the net power at the joint which does not distinguish between active or passive sources, but does provide valuable information regarding the function of the muscle contraction, i.e. whether it is to absorb energy to decelerate or to generate energy to do external work [31]. In general, stair ascent requires considerable positive power to lift the body against gravity in addition to generating movements which advance the body in a forward progression [14]. In healthy young adults, most of this positive work is performed by the knee extensor muscles, which contribute the greatest power during the pull-up phase [10]. In late stance or during forward continuance, the ankle plantarflexors also produce a significant amount of positive power during stair ascent [10,14,15].

Unlike stair ascent, the net muscle powers are predominantly negative during stair descent, representing energy absorption to control the lowering of the body when descending.
from one step to the next. In healthy young adults, the ankle plantarflexors absorb a significant amount of energy in addition to the moderate absorption occurring at the knee during weight acceptance. Through late stance or controlled lowering, the knee extensors provide significant energy absorption [10], reflecting the gravity assisted nature of stair descent. Only the hip flexors generate some energy in late stance during stair descent [10,15].

It is clear that stair negotiation poses high physical and biomechanical demands even in healthy older adults. When physical impairments due to disability are superimposed on age-related decline in physical function, it follows that the ability to climb and descend stairs can be severely compromised and consequently, independence and community participation may be limited.

2.4 Stroke

2.4.1 Overview

Stroke is a leading cause of adult neurological disability in Canada and the number affected is increasing as the population ages and survival rates post-event improve. Over 75% of strokes occur in people over 65 years of age [32,33]. Depending on the severity and location of the stroke, survivors present with varying degrees of neurologic and functional deficits, including motor, sensory, cognitive, perceptual and language impairments [34,35]. With more than 85% of those who experience moderate to severe strokes and almost all who survive mild strokes living in the community [36], the risk of social isolation and dependency secondary to disability is high [37-40]. Indeed the majority of community-dwelling stroke survivors report restrictions in physical capacity and mobility that impact their reintegration into the community [40].

Mobility is fundamental to maintaining independence and regaining the ability to walk is the most frequently stated objective of stroke survivors. While most (approximately 60-70%) [41-
recover the ability to walk by the time they are discharged from hospital, it is estimated that only about 7-22% regain the capacity to walk independently outside of their homes according to an audit of mobility outcomes post-stroke [44] and a one week follow-up of patients discharged home [41]. The reasons are many-fold, but of primary importance are the residual impairments in strength, coordination, balance and deconditioning that manifest in functional limitations. Deficits in one or more of these physical performance measures reflective of neuromuscular, postural and cardiorespiratory system capabilities can impact mobility and if reserve capacity to generate greater ‘output’ from any of these systems is limited, then compensatory strategies will be correspondingly constrained as will be the ability to perform more demanding tasks. Understanding the relationships among system impairments and functional ability is critical for the development and implementation of rehabilitation strategies that are optimally restorative.

2.4.2 Stroke related physical impairments and their association with mobility

Muscle weakness, elevated muscle tone, poor motor coordination and postural instability characterize a large proportion of individuals with hemispheric stroke [45-50]. In addition, up to 75% of stroke survivors have cardiovascular disease [51]. In combination these deficits contribute to the high energy cost of mobility, estimated to be at least twice that of age-matched healthy individuals [52,53].

Muscle weakness is among the most commonly reported and studied deficit following stroke. The force producing capacity of a muscle is compromised secondary to disuse atrophy [54] and as a direct consequence of the stroke’s disruption of descending motor pathways thus limiting the capacity to bring spinal motor neurons to threshold and activate muscles (recruitment) [55,56]. Furthermore, damaged inhibitory circuits or neuromodulatory controls can adversely affect the ability to keep muscles silent or relax those previously activated.
(derecruitment) [56]. Co-activation between agonist and antagonist muscle groups is common in stroke resulting in increased joint stiffness, reduced net torque production and incoordination [57,58]. It is well established that muscles contralateral to the side of the lesion in the brain are significantly weaker than ipsilateral muscles though the extent of the paresis can vary widely. Isokinetic testing of concentric peak torque production following mild to moderate stroke has revealed that the paretic limb generates between 20% and 88% of the torque produced on the less-affected side with the greatest asymmetry occurring in the more distal muscles (Table 2-2). The adoption of the term “less-affected” is deliberate as there are frequently strength deficits on the side ipsilateral to the lesion [59,60] that may reflect damage to the uncrossed corticospinal pathways and/or activity related trophic changes [61].

Table 2-2 Relative isokinetic strength of the affected side to the less-affected side. The shaded area reflects the relative strength of the less-affected side to age and gender matched healthy subjects.

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip extensors</td>
<td>.88 (60°/s)</td>
<td></td>
<td></td>
<td>.64 (60°/s)</td>
</tr>
<tr>
<td>Hip flexors</td>
<td>.63 (60°/s)</td>
<td>.76 (30°/s)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee extensors</td>
<td>.47 (30-60°/s)</td>
<td>.63 (60°/s)</td>
<td>.72 (60°/s)</td>
<td></td>
</tr>
<tr>
<td>Knee flexors</td>
<td>.24 (30-60°/s)</td>
<td>.51 (90°/s)</td>
<td>.62 (60°/s)</td>
<td>.68 (60°/s)</td>
</tr>
<tr>
<td>Plantarflexors</td>
<td>.20 (30-60°/s)</td>
<td>.51 (30°/s)</td>
<td>.55 (30°/s)</td>
<td>.79 (30°/s)</td>
</tr>
<tr>
<td>Dorsiflexors</td>
<td>.31 (30-60°/s)</td>
<td>.74 (30°/s)</td>
<td>.51 (30°/s)</td>
<td>.72 (30°/s)</td>
</tr>
</tbody>
</table>

aUnpublished data from 22 stroke survivors and age- and gender-matched healthy controls.

The impact of muscle weakness is determined by the demands of the activities that individuals need to or wish to perform. Bohannon [60] illustrated this point in the form of sigmoidal curves reflecting the theoretical relationship between strength and functional performance. He described that a certain amount of strength is required to perform a given activity and until that strength level is attained, the individual will be unable to perform the task.
Once the minimum strength requirement is met, further gains translate into better performance, though at some point performance plateaus regardless of continued improvements in strength. The absolute amount of strength required and the degree to which gains in strength are accompanied by improvements in performance differ in accordance with the task demands. In practical terms, these thresholds remain ill-defined but are compelling in the abstract and explain why individuals may be able walk on level ground, but not ascend stairs.

Healthy adults are able to walk at a range of gait speeds and increase their walking cadence by generating progressively higher plantarflexor, hip flexor and hip extensor moments, which they are able to do because they have ample reserve strength [1]. In contrast, people with stroke tend to walk quite slowly and have limited capacity to increase their speed, perhaps due to insufficient residual strength. Nadeau and colleagues [64] evaluated the extent to which the plantarflexors were used during comfortable and fast walking by calculating the muscle utilization ratio (MUR). The MUR is an index of the amount of muscle force used during a task (inverse dynamics) relative to the maximum force output produced under standardized conditions (dynamometry). They found in their sample of 17 subjects with hemiparesis, that the plantarflexors worked at an average of 76.4% of their maximum at a walking speed of 0.8m/s and 85.9% at 1.09 m/s. This compares to 65.6% and 58.8% for healthy controls walking at their comfortable speed or at a comparable speed to those with stroke, respectively. The researchers concluded that weakness was a factor limiting gait speed, however this assumes that other muscle groups could not compensate for the limited plantarflexor output. Several studies have reported strong associations between hip flexor strength and gait speed in stroke [46,62,64-66] suggesting that the hip flexors could effectively pull the stance limb off the ground to compensate for limited plantarflexor push-off [66,67]. Considering that hip flexor strength tends to be better preserved
post stroke than distal muscles (see Table 2-2) it is conceivable that they compensate for the weaker plantarflexors. Similar data are not currently available in association with stair negotiation, but are important to understanding mobility.

Correlational analyses have revealed significant associations between gait speed and the affected knee extensor [62,68], knee flexor [69,70] and plantarflexor [46,62] torques (r values ranging from 0.57 to 0.85). Although only limited studies have investigated factors associated with stair ambulation, the ability to climb stairs has been most strongly linked to the isokinetic strength of the affected and less affected knee extensors and flexors, and ankle plantarflexors (r values ranging from .45 to .73) [46,71]. The speed of stair descent is correlated with isokinetic knee flexor and extensor strength of the affected limb [69]. In combination these findings serve as a reminder that strength on both the affected and less-affected sides should be evaluated, but offer little insight to the clinician about their relative importance or how the muscle groups interact (including compensatory strategies) to maximize function.

There is also growing recognition that poor cardiovascular and cardiorespiratory health can limit recovery following stroke because of limited capacity to meet and respond to the demands of activity and mobility [see reviews: [72,73]]. Among people with stroke, up to 75% have cardiovascular disease [51] contributing to reduced physical activity and deconditioning [74]. Further, research has consistently shown that the energy cost associated with gait post-stroke is at least twice that of age matched, healthy individuals [75-77]. The abnormally low peak aerobic capacity in stroke, estimated to be between 20-50% lower than healthy counterparts [75,78-80], paired with the elevated energy cost of walking on level ground can leave little reserve to handle higher demand activities or conditions such as curb and stair negotiation.
Healthy older adults (average of 78 years) consume oxygen at a rate between 13.3 and 23.7 ml/kg/min while climbing stairs at a rate of 24 steps/min (44% of their maximal VO$_2$ (mVO$_2$)) [81] compared to 11.0 - 12.0 ml/kg/min during comfortable walking [82-84]. Young adults climb nearly 4 times faster, but at a higher O$_2$ cost of about 33.5 ml/kg/min or 83% of their mVO$_2$ [85]. Stair descent requires about half the energy consumption [85]. It follows that deficits in cardiorespiratory function could pose a major limitation to stair negotiation in stroke. A single study from over 40 years ago reported that following stroke, stair ascent required twice the normal O$_2$ cost when compared to healthy adults [86]. However, in the absence of accompanying kinematic and kinetic data, the differential cost cannot be explained. A recent systematic review of the effects of exercise training on walking competency following stroke provided evidence to support the utility of aerobic training in improving stair climbing ability [87]. Following 8 weeks of training using a leg cycle ergometer, the exercise group (n=46) demonstrated a significant increase in the ability to climb one flight of stairs compared to a control group (n=44) [88]. Logistic regression modeling confirmed the effect was due to the training, although it could be argued that corresponding training-related increases in strength (not measured) could have contributed to the outcome. Because weakness and poor cardiorespiratory function generally coexist in stroke it is important to determine the extent to which one or both factors contribute to mobility including stair negotiation.

Regression analyses have included measures related to impairments of multiple systems (muscular, balance, and cardiorespiratory) to identify possible predictors of mobility. A recent study identified the ability to negotiate stairs surpassed walking speed as the single best predictor of community ambulation activity in stroke survivors [89]. The additional strength, coordination and physical conditioning required for stairs compared to walking provides a plausible
explanation of why this might be the case. Unpublished data from our lab (Table 2-3) derived from 72 chronic stroke survivors identified plantarflexor and knee flexor strength as determinants of stair climbing ability after entering flexor and extensor strength for ankle, knee and hip joints into the model. Furthermore, gains in muscle strength was found as a significant independent factor associated with improvements in stair climbing capacity emphasizing the value of improving muscle function to mobility. Although only changes in dorsiflexor strength on the affected side was found to be significant, isometric strength rather than isokinetic strength was measured; the latter providing an indication of force generating capacity during movement. Unpublished data have also identified the physiological cost index (a proxy measure for oxygen cost [90]) after 2 minutes of walking as the second most important determinant of stair climbing capacity (R² = .113) (see Table 2-3).

Table 2-3  Determinants of stair climbing capacity in chronic stroke (n=72) and those impairment measures that when improved translate into increased mobility.

<table>
<thead>
<tr>
<th>Dependent Variable</th>
<th>Independent Variable</th>
<th>Standardized β coefficient</th>
<th>R²</th>
<th>Change in R²</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stairs climbed</td>
<td>Affected knee flexor strength</td>
<td>.234</td>
<td>.293</td>
<td></td>
<td>&lt;.001</td>
</tr>
<tr>
<td></td>
<td>Physiological cost index</td>
<td>-.333</td>
<td>.406</td>
<td>.113</td>
<td>.001</td>
</tr>
<tr>
<td></td>
<td>Unaffected plantarflexor strength</td>
<td>.264</td>
<td>.455</td>
<td>.048</td>
<td>.017</td>
</tr>
<tr>
<td></td>
<td>Ankle tone (modified Ashworth)</td>
<td>-.234</td>
<td>.499</td>
<td>.045</td>
<td>.017</td>
</tr>
<tr>
<td>Δ in number of</td>
<td>Δ affected dorsiflexor strength</td>
<td>.429</td>
<td>.184</td>
<td></td>
<td>.002</td>
</tr>
<tr>
<td>stairs climbed</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Unpublished data; Motor Performance Lab, Queen’s University

To date there have been no published studies characterizing the biomechanics of stair climbing in stroke, yet the ability to negotiate stairs is key to determining discharge destination.
[36]. Seventy five percent of those discharged home consider independent mobility to be of very high importance and stair negotiation is a factor contributing to the attainment of community ambulation [41].

Stair use has been rated as the most difficult task following stroke rehabilitation often leading to avoidance of stairs [91]. It is clear from the reported literature that the ability to negotiate stairs is a function of several factors including strength, joint mobility, physical conditioning and coordination between lower limb joint actions. However, there is little known regarding the biomechanical, strength and aerobic costs of the task, particularly in stroke survivors but in healthy older adults as well. Knowledge of lower limb kinematics and kinetics and the metabolic costs is essential to characterize the requirements of stair ascent and descent and understand compensatory behavior and associated costs in stroke relative to nondisabled controls. Firstly, a comparison between healthy older adults and young adults will identify normal, age-related changes. With this information, the superimposition of impairment related to stroke permits identification of stroke-specific alterations in the requirements of the task as distinct from the changes related to healthy aging. Understanding the physical requirements to negotiate stairs in absolute terms and relative to maximal capacity would provide valuable insight for rehabilitation professionals to assist patients in attaining mobility goals and optimize movement strategies to enhance function and safety.
2.5 References


90. Fredrickson E, Ruff RL, and Daly JJ. Physiological cost index as a proxy measure for the oxygen cost of gait in stroke patients. Neurorehabilitation and Neural Repair, 2009.

Chapter 3
Sagittal and frontal lower limb joint moments during stair ascent and descent in young and older adults

Manuscript published in Gait and Posture 2011; 33:54-60.

3.1 Abstract

Stair negotiation is an essential skill required for independent mobility, and is described by older adults as a challenging task that is associated with high fall risk. Little is known about the age-related changes in joint kinetics and the relative contribution of lower limb joint moments during stair negotiation. This study characterized lower extremity joint kinetics and their variability associated with stair ascent and descent in young and older adults. Twenty three young and 32 older adults (>55 years) participated. Three dimensional, bilateral gait analysis provided ankle, knee, and hip moment profiles, which in the sagittal plane were summed to provide the support moment. In addition, intra- and inter-subject coefficients of variation were calculated for ensemble averaged curves. Age-related differences were found in the magnitudes of the moment contributions during event transitions for stair ascent and descent. Within groups, the moment profiles were generally consistent. Ankle and knee moments predominantly contributed to extensor support in the sagittal plane. In the frontal plane, proximal joint abductor moments maintained lateral stability and were larger at the hip in older adults. Understanding age-related alterations in movement control during functional tasks can help inform the rehabilitation management and assessment of patient populations.
3.2 Introduction

Stair negotiation comprised of stair ascent and descent is required to carry out many activities of daily living and is important for independence in the community. Adults 60 years of age and older describe stair negotiation as one of the more challenging tasks [1]. In community dwelling seniors stair walking is associated with high risk for serious injury [2] particularly during stair descent [3]. In comparison to walking on level ground, the range of motion [4-7] and the muscle moments [4,5,8,9] required at the ankle, knee and hip are significantly higher. The task demands may prove challenging for adults experiencing age-related decline in physical capacity, which becomes significant beyond 50 years of age [10]. Reduced function can increase the variability in movement patterns which if excessive, could be unstable [11,12].

McFadyen and Winter [9] reported consistent sagittal ankle and knee joint moment profiles during stair ascent and descent in three young, healthy men. However, high intra- and inter-individual variability at the hip was attributed to trial to trial modulation in how individuals carried their upper bodies. A larger study of 33 young adults [5] confirmed that sagittal joint moments were stereotypical and added that variability was less during positive work (ascent) than during negative work (descent). The same may not be true in older adults who use proportionately more of their muscle capacity to walk stairs and consequently redistribute the joint moments to maintain muscle output within comfortable limits [6,7]. These strategies could alter the variability in movement patterns, a parameter linked to fall risk [11,12] which we suggest may be particularly relevant if the overall (net) support is compromised. In level walking the support moment reflects overall support against lower limb collapse during stance. The support moment in walking is consistent between subjects even though the magnitude of individual joint moment profiles vary, thus providing an indication of the relative contribution of ankle, knee and hip
moments [13]. Whether the same is true in stair negotiation across age groups is unknown. Few studies report the kinetics from all lower limb joints [6,7,14] or for more than one age group [5,9].

Most studies describing the kinetics of stair negotiation are limited to the sagittal plane [5-7,9,14,15]. There is a need for a comprehensive biomechanical analysis of lower limb kinetics in both sagittal and frontal planes [16]. Nadeau et al. [4] emphasized the importance of the hip abductors in controlling the pelvis during stair ascent and others report the significance of an internal knee abduction moment throughout stance for stabilization [8]. If stability is of concern, as is often the case with increasing age, then movement patterns may be altered to compensate [17].

The purpose of this study was to conduct a comprehensive evaluation of the kinetic requirements of stair negotiation and identify age-related alterations in moment contributions in sagittal and frontal planes. We hypothesize that older adults over the age of 55 will display different moment distributions (ankle, knee, hip and support moments) than their younger counterparts and will demonstrate greater variability in their moment profiles. This information serves to characterize the biomechanical demands of stair negotiation and establish a benchmark of performance in normal healthy aging to allow rehabilitation professionals working with older adults with mobility disorders to identify limitations attributable to impairment rather than aging.

### 3.3 Methods

#### 3.3.1 Subjects

Twenty four adults less than 30 years old (17 female) and thirty three older adults over 55 years (19 female) participated. Subjects were recruited from the university campus and local community through newspaper advertisements. Inclusion required that subjects reported
themselves healthy and could independently manage a flight of stairs without the use of a handrail. Exclusions were neurological or orthopaedic disorders affecting mobility or a history (self report) of an unexplained fall. The protocol was approved by the university’s research ethics board and all subjects provided informed consent.

3.3.2 Procedure

Motion was tracked as subjects ascended and descended a specially constructed 4-step flight of stairs of standard dimensions (rise = 15cm; tread = 26cm). A force platform (AMTI, Newton, MA) mounted on concrete blocks formed the second step. Two optoelectric cameras (Optotrak 3020, Northern Digital Inc., Waterloo, ON) positioned on either side of the staircase allowed bilateral tracking of infrared emitting diodes (IREDs). Clusters of three or four IREDs mounted on moulded plastic plates were strapped over the midthigh, midshank, and midfoot bilaterally and a fin positioned over the sacrum projected outward (see Figure 3-1).

Subjects stood in front of the staircase and following a “go” command, ascended at their self-selected pace using a step over step pattern. Similarly, on a “go” command subjects descended the staircase in the same manner. Kinematic and kinetic data were collected specifying the leading leg; the order (dominant/non-dominant as per the leg used to kick a ball) was randomized. Three successful trials (full contact on the forceplate and all IREDs in camera view) were collected for each lead limb and condition (ascent, descent).

Finally, virtual landmarks were defined using a probe embedded with 4 IREDs with fixed orientation to the tip to enable the calculation of joint centres and segment lengths and to define the rotational axes (see [18,19]). The first and fifth metatarsal heads, medial and lateral malleoli, medial and lateral epicondyles, greater trochanters, and a point aligned vertically with each greater trochanter at the level of the anterior superior iliac spine were probed.
Figure 3-1 Standard configuration of lower limb and pelvis marker clusters during stair negotiation trials. Marker clusters were secured over the midthigh, midshank and midfoot bilaterally and a fin positioned over the sacrum projected outward.
3.3.3 Data Processing and Analysis

Coordinate and force platform data were sampled at 100 Hz, filtered (second order, low pass 6 Hz Butterworth) and synchronized (Visual 3D, C-Motion, Inc., Germantown, MD). Sagittal and frontal plane joint kinetics were calculated using a seven segment, link-segment model and an inverse dynamics approach [20]. Internal joint moments of the ankle, knee and hip were resolved in the local coordinate system of the foot, shank and thigh, respectively. Segment lengths were defined using the appropriate proximal and distal landmarks and joint centres were calculated as the midpoint between the malleoli and epicondyles for the ankle and knee, respectively [21]. The hip joint centre was located at one quarter the distance between the greater trochanters from the left or right trochanter [21]. Initial pilot testing of the experimental set-up compared with previous published literature in healthy young and older adults confirmed face validity of the model used.

Initial foot contact and toe off were identified for each trial from the force platform data. The subsequent foot contact was determined using an algorithm that matched the motion of the foot cluster to that associated with initial contact and foot contact of the contralateral limb was established in a similar manner allowing for the identification of double support [16]. Cadence (steps/min) during ascent and descent was averaged over trials for each subject. Internal ankle, knee and hip joint moments were normalized to body mass and 100% of the stance phase and in the sagittal plane were summed to produce the support moment [13]. Peak flexion, extension, abduction and adduction moments were determined for each trial as well as distinct peaks associated with specific phases when present.

The moment curves for each subject were ensemble averaged and visually inspected for qualitative differences in shape (i.e. the number of peaks or valleys). In the frontal plane, two
distinct patterns were observed, therefore, curves that were qualitatively similar (by group) were ensemble averaged for ascent and descent. Intra- and inter-individual coefficients of variation (CV) were determined for the ankle, knee, hip, and support moments.

3.3.4 Statistical Analysis

Descriptive statistics (means and standard deviations) were calculated for all outcome variables (SPSS 17.0, San Rafael, CA). Mixed factor analyses of variance with one between subject factor (group) and one (condition: ascent and descent) or two within subject factors (condition and side: dominant and non-dominant) were conducted for temporal-distance and moment data. A significance level of $p < 0.05$ was adopted and where an interaction was observed post-hoc comparisons controlling for each factor were performed (Tukey).

3.4 Results

Complete datasets were obtained from 23 young ($23.7 \pm 3.0$ years; range = 20-30 years) and 32 older adults ($67.0 \pm 8.2$ years; range = 55-83 years). One subject per group was excluded due to problems with data acquisition. Eighteen young and 31 older adults were right leg dominant.

3.4.1 Temporal distal parameters of stair ascent and descent

Cadence was influenced by group ($p=0.031$) and condition ($p<0.001$)) reflecting higher cadences among young adults compared to older adults and faster descent than ascent. Correspondingly, stance time was shorter in young versus older adults ($p=0.037$) and for descent as compared to ascent ($p<0.001$) Side was not a significant factor ($p=0.484$). Data are summarized in Table 3-1.
Table 3-1 Temporal distance parameters of stair ascent and descent

<table>
<thead>
<tr>
<th></th>
<th>Cadence (steps/min)</th>
<th>Dominant side Stance time (seconds)</th>
<th>Non-dominant side Stance time (seconds)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean ± SD</td>
<td>Range</td>
<td>Mean ± SD</td>
</tr>
<tr>
<td>Ascent†</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Young</td>
<td>102.50 ± 8.86</td>
<td>87.16-119.42</td>
<td>.74 ± .08</td>
</tr>
<tr>
<td>Older</td>
<td>94.76 ± 13.03</td>
<td>72.53-122.27</td>
<td>.82 ± .13</td>
</tr>
<tr>
<td>Descent†</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Young</td>
<td>110.64 ± 10.24</td>
<td>87.48-134.90</td>
<td>.70 ± .08</td>
</tr>
<tr>
<td>Older</td>
<td>103.68 ± 15.64</td>
<td>74.01-130.51</td>
<td>.74 ± .14</td>
</tr>
</tbody>
</table>

* A significant difference between groups for cadence and stance time ($p < 0.05$)
† A significant difference between condition for cadence and stance time ($p < 0.05$)

3.4.2 Joint Moment Profiles

Because groups differed in cadence and joint moments are influenced by speed [22], independent factorial analyses (group, side) for ascent and descent were performed with cadence as a covariate. Peak sagittal and frontal plane joint moments generated during ascent and descent showed notable differences between groups (Tables 3-2 and 3-3), but no main effect of or interaction with side (p>0.117).

During weight acceptance through pull-up a higher plantarflexor moment was generated by young compared to older subjects ($p<0.02$) for ascent and descent, contributing to a higher support moment during this phase ($p<0.04$). Although the peak plantarflexor moment was also higher for young adults during forward continuance (later stance) ($p=0.008$), the associated support moment peak was greater in older adults ($p=0.03$) likely because they generated smaller knee flexion moments than their young counterparts ($p<0.001$). During controlled lowering (descent), the peak support moment was higher for older adults ($p=0.001$) attributable to a larger knee extensor moment ($p=0.02$) and smaller hip flexor moment ($p<0.001$) than observed in young adults (Table 3-2).

Within group contrasts of maximum peak moments generated during ascent and descent
revealed greater ankle and hip extensor moments during ascent in young adults contributing to a higher support moment (p<0.005). Older adults showed larger peak hip extensor moments during ascent accompanied by a greater support moment (p<0.001). Both groups showed higher knee extensor moments during descent than ascent (p<0.007) and older subjects also had higher hip flexor moments when descending stairs (p<0.001). The moment profiles were in all cases similar in pattern within each group (Figure 3-2).

Table 3-2  Peak joint moments (Nm/kg) for the sagittal plane during the stance phase of stair ascent and descent (mean ± SD)

<table>
<thead>
<tr>
<th></th>
<th>Young adults (n=23)</th>
<th></th>
<th>Older adults (n=32)</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>D</td>
<td>ND</td>
<td>D</td>
<td>ND</td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak PF</td>
<td>1.31 ± .16†</td>
<td>1.27 ± .14</td>
<td>1.19 ± .11</td>
<td>1.18 ± .12</td>
</tr>
<tr>
<td>PF 1†</td>
<td>.94 ± .22</td>
<td>.87 ± .23</td>
<td>.78 ± .24</td>
<td>.71 ± .18</td>
</tr>
<tr>
<td>PF 2†</td>
<td>1.31 ± .17</td>
<td>1.26 ± .14</td>
<td>1.18 ± .11</td>
<td>1.18 ± .12</td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexor</td>
<td>.30 ± .14</td>
<td>.35 ± .19</td>
<td>.21 ± .06</td>
<td>.20 ± .06</td>
</tr>
<tr>
<td>Extensor</td>
<td>1.02 ± .20†</td>
<td>1.06 ± .20</td>
<td>.99 ± .21†</td>
<td>.99 ± .19</td>
</tr>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexor</td>
<td>.11 ± .08</td>
<td>.11 ± .77</td>
<td>.15 ± .10</td>
<td>.14 ± .06</td>
</tr>
<tr>
<td>Extensor</td>
<td>.56 ± .19†</td>
<td>.50 ± .17</td>
<td>.55 ± .18†</td>
<td>.52 ± .16</td>
</tr>
<tr>
<td>Support</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak Ext*</td>
<td>2.41 ± .32†</td>
<td>2.31 ± .32</td>
<td>2.19 ± .33†</td>
<td>2.13 ± .22</td>
</tr>
<tr>
<td>Ext 1†</td>
<td>2.41 ± .32</td>
<td>2.31 ± .32</td>
<td>2.19 ± .33</td>
<td>2.13 ± .22</td>
</tr>
<tr>
<td>Ext 2†</td>
<td>1.19 ± .36</td>
<td>1.03 ± .32</td>
<td>1.24 ± .26</td>
<td>1.23 ± .32</td>
</tr>
</tbody>
</table>

D = Dominant side; ND = Non-dominant side; PF = plantarflexion; ext = extension; abd = abduction

* p < 0.05; indicates a significant main effect of group (young and older adults)
† p<0.05; indicates a significant difference between corresponding moments generated during ascent and descent for the dominant limb within a group (young or older)
Figure 3-2 Ensemble averages (± 1 SD) of the sagittal plane joint and support moment curves (dominant limb) during stair ascent (left) and descent (right) for young adults and older adults. Positive values represent internal flexion moments. All curves are normalized to 100% of the stance phase.
In the frontal plane, peak abductor moments were similar for young and older adults at all joints (p>0.59) and were generally larger during descent than ascent (p<0.029). At the knee and hip, visual inspection of subjects’ mean moment profiles identified in most cases two abductor peaks during ascent and descent (Figure 3-3); the second hip abductor peak during forward continuance and controlled lowering being larger in older adults (p<0.04, Table 3-3). This double peak pattern was the norm in young adults; however a subset of older adults demonstrated a single peak. The single peak pattern was observed at the knee in 10 subjects and at the hip in four subjects during ascent. During descent, this secondary pattern was limited to the knee (7 subjects).

Table 3-3  Peak joint moments (Nm/kg) for the frontal plane during the stance phase of stair ascent and descent (mean ± SD)

<table>
<thead>
<tr>
<th>Frontal plane</th>
<th>Ascent</th>
<th>Young adults</th>
<th>Older adults</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>D ND</td>
<td>D ND</td>
</tr>
<tr>
<td>Ankle</td>
<td>Inverter</td>
<td>.09 ± .04 .11 ± .04</td>
<td>.08 ± .07 .13 ± .11</td>
</tr>
<tr>
<td>Knee</td>
<td>Peak abductor</td>
<td>.29 ± .15† .30 ± .14</td>
<td>.33 ± .12† .34 ± .14</td>
</tr>
<tr>
<td></td>
<td>Abd 1</td>
<td>.32 ± .14 .29 ± .12</td>
<td>.35 ± .10 .35 ± .12</td>
</tr>
<tr>
<td></td>
<td>Abd 2</td>
<td>.19 ± .07 .21 ± .08</td>
<td>.23 ± .06 .26 ± .09</td>
</tr>
<tr>
<td>Hip</td>
<td>Peak abductor</td>
<td>.61 ± .16† .62 ± .12</td>
<td>.68 ± .14† .72 ± .18</td>
</tr>
<tr>
<td></td>
<td>Abd 1</td>
<td>.61 ± .16 .62 ± .12</td>
<td>.68 ± .14 .71 ± .16</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Descend</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle</td>
<td>Inverter</td>
<td>.12 ± .04 .10 ± .05</td>
<td>.12 ± .05 .11 ± .09</td>
</tr>
<tr>
<td>Knee</td>
<td>Peak abductor</td>
<td>.37 ± .13† .36 ± .16</td>
<td>.36 ± .13† .36 ± .13</td>
</tr>
<tr>
<td></td>
<td>Abd 1</td>
<td>.37 ± .10 .37 ± .14</td>
<td>.36 ± .10 .39 ± .09</td>
</tr>
<tr>
<td></td>
<td>Abd 2</td>
<td>.19 ± .07 .18 ± .12</td>
<td>.23 ± .06 .24 ± .11</td>
</tr>
<tr>
<td>Hip</td>
<td>Peak abductor</td>
<td>.76 ± .14† .75 ± .16</td>
<td>.74 ± .13† .76 ± .17</td>
</tr>
<tr>
<td></td>
<td>Abd 1</td>
<td>.74 ± .15 .75 ± .16</td>
<td>.73 ± .17 .76 ± .16</td>
</tr>
<tr>
<td></td>
<td>Abd 2</td>
<td>.53 ± .13 .50 ± .17</td>
<td>.60 ± .15 .64 ± .16</td>
</tr>
</tbody>
</table>

D = Dominant side; ND = Non-dominant side; PF = plantarflexion; ext = extension; abd = abduction
Note: At the knee and hip, two distinct patterns were observed. In these cases, Abd 1 and Abd 2 represent the mean values for the older adults who presented with the “main pattern” only (see text for details)
†p < 0.05; indicates a significant main effect of group (young and older adults)
†p<0.05; indicates a significant difference between corresponding moments generated during ascent and descent for the dominant limb within a group (young or older)
Figure 3-3 Ensemble averages (± 1 SD) of the frontal plane joint moment curves (dominant limb) during stair ascent (left) and descent (right) for young and older adults. An exemplar profile illustrating the alternate pattern displayed at the knee (ascent and descent) and the hip (ascent) by a subgroup of older adults is also shown. Positive values represent internal abduction moments. All curves are normalized to 100% of the stance phase.

3.4.3 Variability

As expected, intra-subject variability was lower than the inter-subject variability for all variables (Table 3-4). In the sagittal plane, both groups showed a distal to proximal pattern of
increasing variability with the ankle moments being the least and the hip the most variable (see Figure 3-2). Only the support moment was more consistent than the ankle moment profile.

Factorial analysis (group and condition) of the intra-subject variability associated with the dominant limb revealed only main effects. For the support moment, variability was lower in older than young adults (p=0.009), but comparable at lower limb joints. The intra-subject CVs of the ankle, hip and support moments were lower in stair ascent than descent (p<0.019), Table 3-4.

**Table 3-4** Mean (± SD) coefficient of variation (%CV) for the sagittal plane observed across the three trials recorded for individual subjects (intra-subject variability) and the %CV across all subjects by group during stair ascent and descent. Data are reported for the dominant side.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Moment</th>
<th>Young adults</th>
<th>Older adults</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intra-subject variability</td>
<td>Ankle†</td>
<td>12.0 ± 4.4</td>
<td>13.3 ± 6.2</td>
</tr>
<tr>
<td></td>
<td>Knee</td>
<td>19.4 ± 9.5</td>
<td>19.2 ± 7.1</td>
</tr>
<tr>
<td></td>
<td>Hip†</td>
<td>24.4 ± 15.1</td>
<td>22.3 ± 10.6</td>
</tr>
<tr>
<td></td>
<td>Support*†</td>
<td>10.5 ± 3.9</td>
<td>9.5 ± 2.7</td>
</tr>
<tr>
<td>Ascent</td>
<td>Ankle</td>
<td>24.0</td>
<td>26.0</td>
</tr>
<tr>
<td></td>
<td>Knee</td>
<td>51</td>
<td>46.3</td>
</tr>
<tr>
<td></td>
<td>Hip</td>
<td>73.5</td>
<td>63.6</td>
</tr>
<tr>
<td></td>
<td>Support</td>
<td>24.3</td>
<td>23.1</td>
</tr>
<tr>
<td>Inter-subject variability</td>
<td>Ankle</td>
<td>28.3</td>
<td>30.3</td>
</tr>
<tr>
<td></td>
<td>Knee</td>
<td>39.6</td>
<td>33.0</td>
</tr>
<tr>
<td></td>
<td>Hip</td>
<td>88.2</td>
<td>140.8</td>
</tr>
<tr>
<td></td>
<td>Support</td>
<td>25.7</td>
<td>23.3</td>
</tr>
</tbody>
</table>

*p<0.05; Indicates a significant main effect of group
†p<0.05; Indicates a significant main effect of condition (ascent/descent)

In the frontal plane (Table 3-5), the opposite variance pattern emerged. The hip moments showed the lowest and the ankle moments the highest variability (see Figure 3-3). For intra-
subject variability, no differences were found between groups or between ascent and descent. The inter-subject variability was extreme at the ankle relative to the knee or hip, even though all subjects showed similar moment profiles. At the knee and hip however, two distinct profile patterns were observed in older adults as described above.

Table 3-5 Mean (± SD) coefficient of variation (%CV) for the frontal plane observed across the three trials recorded for individual subjects (intra-subject variability) and the %CV across all subjects by group during stair ascent and descent. Data are reported for the dominant limb.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Moment</th>
<th>Young adults</th>
<th>Older adults</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ascent</td>
<td>Intra-subject variability</td>
<td>Ankle</td>
<td>84.1 ± 49.1</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Knee</td>
<td>29.6 ± 15.9</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Hip</td>
<td>15.2 ± 4.7</td>
</tr>
<tr>
<td></td>
<td>Inter-subject variability</td>
<td>Ankle</td>
<td>84.0</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Knee</td>
<td>87.1</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Hip</td>
<td>31.5</td>
</tr>
<tr>
<td>Descent</td>
<td>Intra-subject variability</td>
<td>Ankle</td>
<td>82.97 ± 59.8</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Knee</td>
<td>26.1 ± 10.7</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Hip</td>
<td>15.3 ± 3.6</td>
</tr>
<tr>
<td></td>
<td>Inter-subject variability</td>
<td>Ankle</td>
<td>185.6</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Knee</td>
<td>106.5</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Hip</td>
<td>36.3</td>
</tr>
</tbody>
</table>

N.B. Frontal plane values of intra and inter-subject variability are reported for the primary, “two-peak” profiles.

* *p<0.05; Indicates a significant main effect of group
† †p<0.05; Indicates a significant main effect of condition (ascent/descent)

3.5 Discussion

This is the first study to report sagittal and frontal plane ankle, knee, and hip joint moments during stair ascent and descent in young and older adults (≥55 years). The main findings were that groups differed in the magnitude of the moment contributions during transition from double to single support (early stance) or single to double support (late stance) of stair
ascent and descent. Intra-subject variability was comparable for joint moment profiles between groups, although older adults showed greater consistency in the net support moment than their young counterparts.

The sagittal moment profiles for both groups were similar to those reported for young adults [5,9]. Reeves et al. [6,7] compared young and old subjects reporting lower ankle and knee moments in older adults during ascent and descent. Our findings are compatible and add that while the ankle and knee extensors are primary contributors to the support moment, the hip also contributes during weight acceptance and controls trunk position. Toward the end of stance during ascent, the plantarflexors are primarily responsible for the support moment but their output is comparatively less in older adults than in young adults. In overground walking, reducing the piston-like push-off at the ankle minimizes instability [23]; the same could apply here. Alternatively, less plantarflexor output may be needed since unlike younger subjects, older adults generate lower knee flexor moments; a strategy that could limit loading at the knee.

Older adults produce higher late stance support moments in ascent and descent which may serve to enhance perceived stability [23] during transition between single and double support. Compatible with this view is the greater reliance on hip abductors by older adults. The importance of these muscles in allowing the swing leg to clear the intermediate step during ascent has been previously described [4] and our older adults may augment abductor output to maximize safety. Paired with low intra-subject variability of the frontal plane hip moments, the implication is that the tight motor control could be a strategy to enhance balance [12]. A review of stair negotiation suggests that older adults often perceive difficulty due to concern about falling which is intensified during descent [17]. In the current study, peak hip abductor moments were higher during descent than ascent in older subjects. If it is the case that the hip abductors are important to
safely negotiate stairs, this may have implications for patients with physical impairments. Indeed, during level walking, weakness of the abductors can result in frontal plane instability thus increasing the risk of falls [24].

During stair descent, the eccentric plantarflexor moment was markedly smaller in older adults at weight acceptance contributing to the smaller support moment at this point. Reeves et al. [6] also reported lower plantarflexor moments in older adults which they attributed to weakness.

We cannot determine whether that was the case in our subjects in the absence of measuring strength. Combined with a strong knee extensor moment and a small hip flexor moment however, the overall extensor support was larger than in young adults, perhaps to augment confidence. The consistency of the pattern within and across subjects suggests this to be a strategy characteristic of older adults.

The extensor moments associated with the positive work of stair ascent were generally of greater magnitude than during descent as expected from other reports [5,9]. The exception was the higher knee extensor moment generated during descent in older adults which would effectively provide greater control during body lowering. Noteworthy is the correspondence between intra-subject variability and task demand such that variability is lower for the condition (ascent or descent) in which the moment magnitudes are highest [25].

In the sagittal plane, young and older adults demonstrated reproducible moment profiles across trials. The lowest CVs were associated with the ankle and highest at the hip attributable to trial to trial adjustments in trunk position [5,9] to meet the challenge of balancing the head, arms and trunk [26,27]. Of note is that despite variances in the magnitude of individual joint moments, the net extensor support moment is very consistent and particularly so in older adults which may keep the risk of falling in check.
In the frontal plane, the moments generated reflected predominantly knee and hip abductor moments which were greater during descent compared to ascent. Abductor moments are important in maintaining the body’s centre of mass within the narrow base of support during controlled lowering while countering the destabilization associated with the upper body and mass of the swing leg producing a gravitational adductor moment [28]. The higher magnitude hip abductor moments observed in older adults may be secondary to the generalized concern about safety when negotiating stairs [17]. The observation of a more uniform abductor moment profile in a subset of older adults may be a strategy to provide additional stability.

Unlike the proximal-distal pattern of decreasing variability seen in the sagittal plane, the opposite was true in the frontal plane. The high intra-subject variability in ankle joint moment profiles is likely a consequence of the small moment magnitudes (see Figure 3-3) such that minor fluctuations to control dynamic posture and balance would result in large CVs [28]. The least variable joint moment patterns occurred at the hip. From a motor control perspective, when demands are high the number of possible solutions to accomplish the task are fewer and, as was found in the sagittal plane, the variance is lowest in those muscle groups (and conditions) that generate the greatest output. The general lack of between group differences in variability could be interpreted to mean that motor control aspects are similar in healthy young and older adults. Alternatively, variability in kinetic patterns may be a less valid indicator of movement control [29] or dynamic stability [12] than variability in temporal-distance or ground-reaction force data, which are by nature, more consistent [13,20]. Further study is warranted to explore the usefulness of measuring variability of kinetic variables as in indicator of neuromuscular stability.

In summary, we have characterized sagittal and frontal plane lower extremity joint kinetics during stair negotiation and identified age-related differences in peak moments generated...
reflecting a strategy to promote greater support and lateral balance in older adults. The variability of the joint moment profiles were comparable between groups, although the lower support moment CV in older adults could reflect a reduction in fall risk. The data provide a benchmark of normal age-related moment patterns against which performance in the presence of impairment or disability can be compared. Future studies should incorporate the kinematic data to assist in interpretation of the kinetic profiles, particularly when evaluating pathological conditions.
3.6 References


Chapter 4

Mechanical energy transfers across lower limb segments during stair ascent and descent in young and healthy older adults

Manuscript In Press, Gait and Posture; doi:10.1016/j.gaitpost.2011.06.007, June 2011

4.1 Abstract

Older adults present with altered movement patterns during stair negotiation although the extent to which modifications in pattern and speed influence mechanical efficiency is unknown. This study evaluated mechanical energy transfers attributed to active force production during stair negotiation in young and older adults to provide insight into age-related changes in mechanical efficiency. Secondary analysis on data obtained from 23 young (23.7 ± 3.0 years) and 32 older adults (67.0 ± 8.2 years) during self-paced stair ascent and descent was conducted. Mechanical energy expenditures (MEE) during concentric transfer, eccentric transfer and no-transfer phases were determined for the ankle, knee and hip power profiles in the sagittal plane. Mechanical energy compensations (MEC) were also determined at each joint. During ascent, MEEs were similar for young and older adults although older adults compensated ankle muscles to a lesser extent during concentric muscle action. Controlling for cadence eliminated this difference. During descent, older adults demonstrated lower energy expenditures at the ankle and hip and similar expenditures at the knee compared to young adults. Changes in joint MEE in the older group resulted in reduced energy compensation at the ankle during concentric and eccentric activity and at the knee during eccentric activity. These age-related differences in mechanical energy transfers and related adjustments in MEC were not a function of the slower cadence in
older adults and suggest a loss in mechanical efficiency. These results provide a benchmark against which physical impairments in older adults may be explored.
4.2 Introduction

Muscles generate or absorb mechanical energy (power) via concentric or eccentric contractions, respectively. Mechanical energy can also be transferred between adjoining segments via muscle activation (i.e. active energy transfer) if the two segments are rotating in the same direction [1,2]. For example, during walking, the foot and shank rotate in the same direction during push-off resulting in energy being transferred to the foot segment which partially compensates the work done by the plantarflexor muscles. In this case, inter-segmental energy transfer assists with propulsion thereby improving efficiency. One way to evaluate inter-segmental energy transfers and the efficiency of mobility is to consider mechanical energy expenditure, which is the net amount of energy produced by the muscles controlling the movements (the net result of agonist, synergist and antagonist activity at a joint), and the mechanical energy compensations [3-8]. The latter reflects the proportion of active contractile muscle energy compensated by inter-segmental energy transfer through the muscles [3-8]. By partitioning mechanical energy expenditure as a function of concentric/eccentric sources for each joint and transfer/no-transfer compensations, McGibbon and colleagues have quantified altered motor control strategies attributable to age and pathology [8-10]. Such insight into the efficiency of movement has provided valuable information about movement control and compensatory strategies associated with level ground walking [8-10] but other types of gait including stair negotiation remain unexplored.

Beyond 50 years of age measurable declines in joint mobility and strength result in altered or adapted movement patterns, particularly during higher demand physical activities [11-16]. Studies have reported a redistribution of the moments generated at lower limb joints during stair negotiation in older adults compared to young adults, reflecting an attempt to maintain
muscle activity within safe, comfortable limits [15-17]. We have shown that older adults produce higher net support moments in late stance than their younger counterparts; possibly to enhance balance when transitioning from single to double support [17]. Further, older adults generally ascend and descend stairs more slowly [18] than young adults. It seems reasonable to speculate that such modifications in pattern and speed would influence mechanical efficiency by altering the energy transferred across segments and/or the amount of muscle energy generated or absorbed when no inter-segmental transfer takes place. However, the extent to which this occurs or if localized deficits in energy flow can be compensated elsewhere is unknown. Determining mechanical energy expenditures and the proportion of muscle energy compensated by inter-segmental energy transfer is an important step toward understanding the energy costs associated with stair negotiation in older adults and is a key consideration for mobility independence.

The purpose of this study was to evaluate mechanical energy transfers attributed to the net effect of active muscular force production (active transfers) during stair negotiation in young and older adults. Based on the evidence cited above, we hypothesize that healthy older adults will demonstrate altered inter-segment energy transfer during stance compared to young adults.

4.3 Methodology

A secondary analysis of biomechanical data obtained from 23 young (23.7 ± 3.0 years) and 32 older adults (67.0 ± 8.2 years) was performed. Details about the subjects and study protocol are described elsewhere [17, Chapter 3]. Briefly, self-reported healthy subjects were recruited from the university and local communities and all were able to negotiate a flight of stairs independently without the use of a handrail. The protocol was approved by the university’s research ethics board and all subjects provided informed consent. Subjects completed three ascent and three descent self-paced trials without a handrail on a specially constructed 4-step staircase of
standard dimensions (rise = 15cm; run = 26cm) with a forceplate forming the centre of the second step. Optoelectric cameras tracked clusters of infrared emitted diodes (Northern Digital Inc., Waterloo, ON) secured to the subjects’ lower limbs to provide the three dimensional spatial coordinates of lower limb joints and segments. The data were synchronized with the force plate data using commercial software (Visual 3D, C-Motion, Inc., Germantown, MD), and segment kinematics and joint moments were computed using an inverse dynamics approach based on a seven segment model. All data were resolved in the global coordinate system and kinetic variables were normalized to body mass. Data are reported for the dominant limb only (18 young and 31 older adults were right leg dominant) and normalized to 100% of the stance phase.

4.3.1 Mechanical energy calculations

For rotational motion, that is, motion about a joint axis, the power of the proximal (Pp) and distal (Pd) ends of the distal and proximal articulating segments, respectively was calculated as the product of the muscle moment (Mj, where Mjp = -Mjd) and the segment’s angular velocity (ω). At a given joint (j), the net muscle power (Pj, Watts/kg) is the summation of the power at the endpoints of adjoining segments:

\[ P_j = P_p + P_d = M_{jp} \omega_p + M_{jd} \omega_d \]

Where, P = muscle power, Watts/kg
M = muscle moment, N·m/kg
ω = angular velocity, rads/s

The sign and relative magnitudes of the power terms indicate whether the muscles generate or absorb energy and whether the energy is transferred across segments [1,3-5]. When segments on either side of the joint rotate in opposite directions, muscles either generate or absorb power entirely. If the adjoining segments rotate in the same direction, energy transfers can be directed proximally (Pd < 0 and Pp > 0) or distally (Pd > 0 and Pp < 0) to facilitate muscular effort.
Table 4-1  Summary of the power flow conditions that determine the energy transfer between adjoining segments (modified from McGibbon et al. [8], Robertson and Winter [1]). The relative velocities of the segments, power flow conditions and net joint power determine transfer of energy, direction and the type of muscle contraction, respectively

<table>
<thead>
<tr>
<th>Condition</th>
<th>Muscle moment</th>
<th>Angular velocity</th>
<th>Power flow</th>
<th>Type of contraction</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Proximal</td>
<td>Distal</td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>(M_p &gt; 0)</td>
<td>(\omega_p &gt; 0)</td>
<td>(\omega_d &lt; 0)</td>
<td>(P_p &gt; 0, P_d &gt; 0, P_j &gt; 0)</td>
<td>Concentric</td>
</tr>
<tr>
<td></td>
<td>(M_p &lt; 0)</td>
<td>(\omega_p &lt; 0)</td>
<td>(\omega_d &gt; 0)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>(M_p &gt; 0)</td>
<td>(\omega_p &lt; 0)</td>
<td>(\omega_d &gt; 0)</td>
<td>(P_p &lt; 0, P_d &lt; 0, P_j &lt; 0)</td>
<td>Eccentric</td>
</tr>
<tr>
<td></td>
<td>(M_p &lt; 0)</td>
<td>(\omega_p &gt; 0)</td>
<td>(\omega_d &lt; 0)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>(M_p &gt; 0)</td>
<td>(\omega_p &gt; 0)</td>
<td>(\omega_d &gt; 0)</td>
<td>(P_p &gt; 0, P_d &lt; 0, P_j &gt; 0)</td>
<td>Concentric</td>
</tr>
<tr>
<td></td>
<td>(M_p &lt; 0)</td>
<td>(\omega_p &lt; 0)</td>
<td>(\omega_d &lt; 0)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>(M_p &gt; 0)</td>
<td>(\omega_p &gt; 0)</td>
<td>(\omega_d &gt; 0)</td>
<td>(P_p &gt; 0, P_d &lt; 0, P_j &lt; 0)</td>
<td>Eccentric</td>
</tr>
<tr>
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<td>(M_p &lt; 0)</td>
<td>(\omega_p &lt; 0)</td>
<td>(\omega_d &lt; 0)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>5</td>
<td>(M_p &gt; 0)</td>
<td>(\omega_p &lt; 0)</td>
<td>(\omega_d &lt; 0)</td>
<td>(P_p &lt; 0, P_d &gt; 0, P_j &gt; 0)</td>
<td>Concentric</td>
</tr>
<tr>
<td></td>
<td>(M_p &lt; 0)</td>
<td>(\omega_p &gt; 0)</td>
<td>(\omega_d &gt; 0)</td>
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</tr>
<tr>
<td>6</td>
<td>(M_p &gt; 0)</td>
<td>(\omega_p &lt; 0)</td>
<td>(\omega_d &lt; 0)</td>
<td>(P_p &lt; 0, P_d &gt; 0, P_j &lt; 0)</td>
<td>Eccentric</td>
</tr>
<tr>
<td></td>
<td>(M_p &lt; 0)</td>
<td>(\omega_p &gt; 0)</td>
<td>(\omega_d &gt; 0)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

\(M\), muscle moment; \(\omega\), angular velocity; \(P\), power; \(j\), joint; \(p\), proximal segment; \(d\), distal segment

To determine the net amount of energy produced by the system, the mechanical energy expenditure (MEE) was calculated using the method discussed by Aleshinsky [3-7] and later described by McGibbon and colleagues [8-10]. The net joint MEE was calculated separately for concentric transfer conditions (\(\text{MEE}_{\text{C}} = \text{MEE}_{\text{condition3}} + \text{MEE}_{\text{condition5}}\)), eccentric transfer conditions
\( \text{MEE}_E = \text{MEE}_{\text{condition4}} + \text{MEE}_{\text{condition6}} \) and no transfer conditions \( \text{MEE}_N = \text{MEE}_{\text{condition1}} + \text{MEE}_{\text{condition2}} \); where conditions are those described in Table 4-1 and MEE is the integral of the net joint power curve segmented by power condition. In addition to the net joint MEE, mechanical energy compensation (MEC), defined as the proportion of muscle energy compensated by inter-segmental energy transfer [8], was determined for the periods of concentric and eccentric activity at each joint. The MEC is the ratio of the net joint MEE for each contraction type (concentric, eccentric) and the total absolute MEE. MEC is always zero when no segmental transfer occurs and therefore is not reported for those conditions.

4.3.2 Statistical Analysis

Descriptive statistics (means and standard deviations) were calculated for all outcome measures (SPSS version 17.0, San Rafael, CA). Independent samples t-tests comparing groups were conducted for the MEE and MEC measures at each joint (ankle, knee and hip) and for stair ascent and descent. A significance level of \( p < 0.05 \) was adopted for all analyses.

4.4 Results

Cadence differed significantly by group \( p<0.031 \) during stair ascent (young adults: 102.50 ± 8.86 steps/min; older adults: 94.76 ± 13.03 steps/min) and descent (young adults: 110.64 ± 10.24 steps/min; older adults: 103.68 ± 15.64 steps/min).

4.4.1 Stair ascent

The power profiles were similar in shape between young and older adults (see Figure 4-1), although visual inspection indicates the peak magnitudes are lower in the older group. In general, during early stance or weight acceptance (~0% to 10% of stance), energy was generated at the hip and absorbed at the ankle. It was transferred concentrically and directed proximally across the
knee joint at weight acceptance. As individuals progressed to full support on the next step (pull-up or loading phase; ~10% to 50% of stance), energy was mostly transferred distally across the hip due to concentric contraction (active energy generation) while energy was generated at the knee during loading as the counter rotations of the shank and thigh preclude transfer across segments. At the ankle, energy was directed proximally (concentric activity) during early loading and generated through midstance. In late stance, little energy was produced at the hip and transferred across the knee. Similarly, energy generated at the knee was transferred distally across the ankle contributing (or adding power) to the ankle power during unloading, prior to the transition from stance to swing.

Ankle and hip MEEs were similar for young and older adults (p>.451) although older adults demonstrated increased eccentric energy expenditure at the ankle (p=.03), which from Figure 4-1 appears to be of limited relevance. The compensation of ankle muscles through concentric energy transfer (MEC\textsubscript{Ankle}) was lower in older than young adults (p=.009); a function of their faster cadence since the difference disappeared when cadence was considered as a covariate. The other difference between groups was the significantly lower eccentric MEE (MEE\textsubscript{Knee}) at the knee in older adults (p=.001), although the knee muscles were compensated to a similar degree in both groups (MEC\textsubscript{Knee}, p=.795). That the knee MEE differences remained after controlling for cadence is surprising given the limited power flow via eccentric activation (Figure 4-1); the concentric nature of the work done during stair ascent suggests this finding may be of limited importance. In contrast, the tendency of lower compensation of hip muscles in older adults through concentric inter-segmental transfer effectively limited the power of the thigh and energy generated (early stance, Figure 4-1) though the difference from young adults was not significant. Data are summarized in Table 4-2.
Figure 4-1 Mean power profiles of the proximal and distal endpoints of the adjoining segments about the hip (top), knee (middle), and ankle (bottom) joints for the young adults (left) and older adults (right) identifying concentric and eccentric efforts, and the energy flow during stair ascent. Shaded bars represent the energy transfer conditions: black=no transfer, dark gray=concentric transfer, light gray=eccentric transfer. The arrows indicate the direction of energy transfer: up=proximal, down=distal, outward=concentric no transfer, inward=eccentric no transfer. To avoid artifacts associated with averaging across trials and subjects and due to temporal dispersion, a five point moving average was applied to the ensemble average power curves for illustration.
4.4.2 Stair descent

As with stair ascent, the power profiles during descent were comparable between the two groups (Figure 4-2), though the peak magnitudes appeared somewhat lower in older adults compared to young adults, particularly at the hip (late stance). In general, during early stance or weight acceptance, hip energy was near zero indicating little to no energy generation, absorption, or transfer until the controlled lowering phase of stance (mid to late stance; ~60% to 100% of stance). At the knee, energy absorption occurs, followed by eccentric muscle activity resulting in energy being transferred proximally as individuals move through weight acceptance to forward continuance (midstance). During midstance, young adults transferred limited energy from concentric knee activity proximally, whereas older adults exploited eccentric muscle activity to a greater extent to transfer energy proximally through the forward continuance phase. The ankle muscles absorbed considerable energy in early stance then in mid to late stance, energy was transferred distally through eccentric action and concentric action (during controlled lowering prior to swing).

No significant group differences were found for ankle MEE during eccentric transfer (p=.158) yet young adults compensated their ankle muscles to a greater extent (p=.020). In late stance, older adults expended greater ankle energy through concentric activation transferring the energy distally (p=.018; Figure 4-2) though compensation was less than was observed for young adults (MEC_{Ankle}^{c} < .02). In the absence of energy transfer, energy expenditure was significantly lower in older adults compared to young adults (MEE_{Ankle}^{N} p=.014).

Concentric knee activity was limited in stair descent (see Figure 4-2). Eccentric knee activity, however resulted in higher energy expenditure in older adults directed proximally, though this was not borne out statistically (p=.084). Young adults more effectively compensated
Figure 4-2 Mean power profiles of the proximal and distal endpoints of the adjoining segments about the hip (top), knee (middle), and ankle (bottom) joints for the young adults (left) and older adults (right) identifying concentric and eccentric efforts, and the energy flow during stair descent. Shaded bars represent the energy transfer conditions: black=no transfer, dark gray=concentric transfer, light gray=eccentric transfer. The arrows indicate the direction of energy transfer: up=proximal, down=distal, outward=concentric no transfer, inward=eccentric no transfer. To avoid artifacts associated with averaging across trials and subjects and due to temporal dispersion, a five point moving average was applied to the ensemble average power curves for illustration.
their knee muscles through eccentric transfer (MEC\textsubscript{E}\textsuperscript{Knee}; p=.007). At the hip, concentric MEE was approximately twice as high in young adults than in older adults (p<.001; late stance in Figure 4-2) largely due to higher power of the thigh segment though compensations were similar (p>.102). The significance of these findings did not change when cadence was introduced as a covariate. Data are summarized in Table 4-2.

Table 4-2 Mechanical energy expenditures (MEEs) (J/kg) and compensation coefficients (MECs) of the ankle, knee and hip during the stance phase of stair ascent and descent (mean ± SD) for each transfer condition (concentric transfer, eccentric transfer, and no transfer)

<table>
<thead>
<tr>
<th></th>
<th>Mechanical energy expenditures (J/kg)</th>
<th>Ankle</th>
<th>Older adults</th>
<th>p-value</th>
<th>Young adults</th>
<th>Older adults</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle</td>
<td>MEE\textsubscript{C}</td>
<td>28.80 ± 9.99</td>
<td>26.56 ± 11.75</td>
<td>.462</td>
<td>14.08 ± 2.75</td>
<td>16.20 ± 3.46</td>
<td>.018</td>
</tr>
<tr>
<td></td>
<td>MEE\textsubscript{E}</td>
<td>1.58 ± 0.93</td>
<td>2.37 ± 1.64</td>
<td>.027</td>
<td>5.22 ± 1.09</td>
<td>6.58 ± 1.16</td>
<td>.158</td>
</tr>
<tr>
<td></td>
<td>MEE\textsubscript{N}</td>
<td>16.54 ± 10.07</td>
<td>14.07 ± 13.05</td>
<td>.452</td>
<td>28.60 ± 12.24</td>
<td>21.26 ± 9.19</td>
<td>.014</td>
</tr>
<tr>
<td>Knee</td>
<td>MEE\textsubscript{C}</td>
<td>7.18 ± 3.25</td>
<td>6.42 ± 4.03</td>
<td>.462</td>
<td>2.57 ± 1.80</td>
<td>0.98 ± 1.20</td>
<td>&lt;.001</td>
</tr>
<tr>
<td></td>
<td>MEE\textsubscript{E}</td>
<td>1.09 ± 0.71</td>
<td>0.54 ± 0.49</td>
<td>&lt;.001</td>
<td>5.58 ± 6.19</td>
<td>9.4 ± 9.0</td>
<td>.084</td>
</tr>
<tr>
<td></td>
<td>MEE\textsubscript{N}</td>
<td>43.55 ± 14.97</td>
<td>41.78 ± 10.29</td>
<td>.604</td>
<td>70.83 ± 16.55</td>
<td>71.34 ± 14.79</td>
<td>.904</td>
</tr>
<tr>
<td>Hip</td>
<td>MEE\textsubscript{C}</td>
<td>12.25 ± 6.86</td>
<td>11.79 ± 6.43</td>
<td>.798</td>
<td>3.60 ± 1.46</td>
<td>1.76 ± 1.19</td>
<td>&lt;.001</td>
</tr>
<tr>
<td></td>
<td>MEE\textsubscript{E}</td>
<td>0.36 ± 0.62</td>
<td>0.54 ± 1.02</td>
<td>.466</td>
<td>1.11 ± 0.73</td>
<td>0.98 ± 0.87</td>
<td>.555</td>
</tr>
<tr>
<td></td>
<td>MEE\textsubscript{N}</td>
<td>13.20 ± 6.25</td>
<td>12.64 ± 6.64</td>
<td>.751</td>
<td>2.44 ± 0.51</td>
<td>2.09 ± 0.52</td>
<td>.475</td>
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<table>
<thead>
<tr>
<th></th>
<th>Mechanical energy compensation coefficients</th>
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<th>p-value</th>
<th>Young adults</th>
<th>Older adults</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle</td>
<td>MEC\textsubscript{C}</td>
<td>0.44 ± 0.11</td>
<td>0.37 ± 0.12</td>
<td>.009</td>
<td>0.70 ± 0.04</td>
<td>0.67 ± 0.04</td>
<td>.007</td>
</tr>
<tr>
<td></td>
<td>MEC\textsubscript{E}</td>
<td>0.60 ± 0.12</td>
<td>0.53 ± 0.16</td>
<td>.065</td>
<td>0.50 ± 0.08</td>
<td>0.43 ± 0.11</td>
<td>.020</td>
</tr>
<tr>
<td>Knee</td>
<td>MEC\textsubscript{C}</td>
<td>0.36 ± 0.08</td>
<td>0.34 ± 0.12</td>
<td>.596</td>
<td>0.63 ± 0.10</td>
<td>0.65 ± 0.18</td>
<td>.588</td>
</tr>
<tr>
<td></td>
<td>MEC\textsubscript{E}</td>
<td>0.56 ± 0.11</td>
<td>0.55 ± 0.16</td>
<td>.795</td>
<td>0.52 ± 0.08</td>
<td>0.44 ± 0.12</td>
<td>.007</td>
</tr>
<tr>
<td>Hip</td>
<td>MEC\textsubscript{C}</td>
<td>0.27 ± 0.14</td>
<td>0.21 ± 0.08</td>
<td>.060</td>
<td>0.44 ± 0.12</td>
<td>0.49 ± 0.11</td>
<td>.102</td>
</tr>
<tr>
<td></td>
<td>MEC\textsubscript{E}</td>
<td>0.21 ± 0.19</td>
<td>0.25 ± 0.17</td>
<td>.437</td>
<td>0.57 ± 0.11</td>
<td>0.56 ± 0.11</td>
<td>.830</td>
</tr>
</tbody>
</table>

MEE\textsubscript{C}/MEC\textsubscript{C}, Concentric energy transfer condition; MEE\textsubscript{E}/MEC\textsubscript{E}, Eccentric energy transfer condition; MEE\textsubscript{N}, No transfer condition

Note: MEC always zero for no-transfer conditions.
4.5 Discussion

This is the first study to analyze mechanical energy transfers and compensations during stair negotiation as a means of identifying age-related compensations in healthy older adults relative to their younger counterparts. The major findings are that even though older adults ascend and descend stairs at a slower cadence than young adults, during ascent the energy generated and transferred through concentric muscle activation was generally comparable between groups at all joints. However, the proportion of muscle energy compensated by inter-segmental energy transfer was less for the ankle muscles and hip muscles in older adults suggesting that older adults require more muscle work to produce similar power output. During stair descent, group differences were more evident. Energy expenditures associated with ankle and knee muscles were higher in older adults, but contributed more to compensating the muscles in young adults, the opposite pattern occurred for the hip muscles acting concentrically. Controlling for the difference in cadence between the two groups failed to eliminate the significance of the age-related differences observed. This suggests that perhaps physiological changes such as declines in strength[11] and dynamic balance[19] may have limited the mechanical efficiency of stair negotiation in our healthy, older adults.

Studies reporting active power exchanges and compensations related to inter-segmental energy transfers through the muscles during movement are limited and are specific to level ground walking [1, 8-10]. Stair ascent is more demanding as it requires vertical progression against gravity as well as forward, horizontal progression [14, 20]. Compared to reported data for level walking, the amount of energy expended and transferred across segments in our older (and young) adults ascending stairs is similar in magnitude to older adults (mean age 73.7 years old) walking over ground [8]. Dissimilarities relate to the degree of mechanical energy compensation,
which in level walking yields MEC coefficients of approximately twice the magnitude of those we observed during positive ankle and knee work. Although direct comparison of walking and stair negotiation would be required to confirm these inferences, it seems quite evident that stair climbing is less energy efficient although older adults appear to achieve similar efficiencies as young adults by slowing their cadence.

The demands of stair climbing are further elevated by the amount of energy expended in the absence of energy transfer. That is, the muscles are not compensated by segmental energy transfer through the muscles (the no transfer condition). In level walking, reported values range from 1.9 J/kg at the ankle to 13.2 J/kg at the knee (11.8 J/kg at the hip) [8]. The energy expenditures we report during stair ascent are approximately 4 to 10 times higher at the knee and ankle, which contribute to the higher energy cost of stair ascent as reflected by the higher metabolic cost [21]. In stair descent, the net amount of energy absorbed by muscles without transfer across segments is 1.5 or 2 times as high at the ankle and knee as occurs during ascent, likely a strategy to augment balance.

Age-related differences in inter-segmental energy flow through the muscles were more pronounced during descent. In early stance, healthy young adults transfer energy through concentric knee muscle activity, a strategy that is not seen in older adults. This would assist in achieving efficient energy redistribution through the stance limb without contributing to the muscular compensation. In contrast, older adults transfer less energy which would serve to add balance through weight acceptance [22], and may compensate the lower energy expended by the ankle muscles at this point compared to young adults. The reduced mechanical energy compensation for the total eccentric power generated at the knee ($\text{MEC}_E$) during descent in older adults is noteworthy. Previous work investigating stair descent in older adults has shown larger
knee extensor moments compared to young adults [17] to compensate for weak ankle muscles and to augment balance. The tendency toward higher energy expenditure attributable to the knee extensors (p=.084) paired with the reduced MEC would be compatible with the exaggerated knee extension moment providing enhanced extensor support and balance [17].

In late stance, or controlled lowering, older adults produced higher energy at the ankle and lower energy at the hip through concentric muscles activity than young adults that were not attributable to differences in cadence. The ankle muscles were less compensated than occurred in young adults suggesting more energy expended by the muscles to assist in advancing the limb into swing [23]. At the hip, limiting the distally directed energy flow provides an effective strategy to augment balance through the control of the trunk and upper body [17,24-25]. Future work should investigate in greater detail the energetics associated with the hip and its impact on the trunk in light of the elevated fall risk during stair descent in older adults [18].

There are several limitations to our analysis. First, we restricted our analysis to active energy transfers. McGibbon and Krebs [26] have investigated passive transfers in level ground gait and found significant errors due to the joint gap that exists when unconstrained segment tracking is used, as was used in the current study. The active transfers we report in relation to stair negotiation are novel and provide a foundation for future work to explore the metabolic demands of this high demand activity. Second, mechanical energy analysis based on inverse dynamics does not allow for identification of the individual muscle sources responsible for the movement, but rather it is a reflection of the net effect attributable to the activity of all muscles that span the joint. This approach, however, yielded important information about the interactions of body segments and net muscle energy that identified age-related compensatory strategies adopted during stair negotiation. Future work should build on our findings associated with the sagittal
plane and consider active power transfers in the frontal plane, particularly at the hip considering
the importance of the hip abductors in step clearance and providing lateral balance [17,27].

In summary, we have demonstrated age-related alterations in mechanical energy
efficiency during stair negotiation. Older adults reduced their speed of stair ascent but, in general,
appeared to transfer energy to the same degree as young adults indicating that the adaptation did
not impact mechanical efficiency. During descent however, age-related differences in mechanical
energy transfers and related adjustments in MEC suggest a loss in mechanical efficiency since the
differences were preserved after controlling for cadence. Factors such as declines in strength,
deficits in dynamic balance control and elevated fall risk may have contributed to these findings.
It follows that the superimposition of physical impairments and disability on age-related changes
could have serious impact on mobility independence.
4.6 References


Chapter 5

Kinematic and kinetic evaluation of the stance phase of stair ambulation in persons with chronic stroke and healthy adults


5.1 Abstract

This study describes and contrasts the kinematics and kinetics of stair ambulation in people with chronic stroke and healthy control subjects. Three dimensional motion data were collected from 10 persons with stroke (7 males) and 10 sex and age-matched older adults as they ascended and descended an instrumented staircase at self-selected speed with and without a handrail. Ankle, knee and hip joint angle and moment profiles were generated during stance and range of motion and peak moments were contrasted between groups, sides (stroke only) and condition. Cadence was lower in stroke than controls, although the kinematics profiles were generally similar during ascent and decent. Notable differences in joint kinetics were evident as the extensor moments were typically lower on the affected side in stroke compared to controls and the less affected side accounting for the lower magnitude net extensor support moment. Lower affected side hip abductor moments limited lateral balance. Handrail use tended to reduce the peak moments on the affected side only leading to more side-to-side differences than occurred without the handrail. The findings reveal differences in task performance between stroke and healthy groups that help inform rehabilitation practice.
5.2 Introduction

Stair negotiation is a challenging and demanding locomotor task that is essential for independent ambulation. Biomechanical analyses have shown that compared to level walking, greater lower limb joint ranges of motion (ROM) and muscle moments are generally required to ascend and descend stairs, with the knee extensors having a dominant role [1-3]. Following stroke, reduced neuromuscular capabilities and limitations in joint mobility can increase the challenge of stair negotiation which can present as a barrier to active community living [4].

There is ample evidence that reduced joint mobility poses risks for stair negotiation [5]. Researchers have reported that healthy, older adults show reduced ankle and knee motion compared to young adults when descending stairs, accompanied by increased hip and pelvic motion in the frontal plane [6]. During stair ascent, age-related reductions in ankle dorsiflexion could result in a trip if the toes fail to clear the step [7]. Recent evidence has shown that older adults adopt alternative strategies to meet the high demands of the task [8-11]. A redistribution of joint moments [8-10], an exaggerated net support moment, and sustained abductor moments throughout stance are evident in older adults likely to compensate for declining muscle strength and/or to enhance balance [8]. In clinical populations such as stroke, where impairment is typically superimposed on aging, the ability to redistribute loads in a way that accomplishes the task of stair ambulation may be compromised.

Level ground gait of persons with stroke is characterized by a decrease in self-selected speed and altered sagittal and frontal plane kinematic and kinetic profiles in both magnitude and pattern [12-16]. These deviations from the normal gait pattern result from residual deficits in strength, poor coordination, loss of balance control and deconditioning (see [17] for review). It is the case that many individuals regain the ability to walk following stroke [18,19], however it is
stair negotiation that is the single best predictor of community living activity [4]. The higher physical demands associated with stair ambulation compared to walking likely explains why stair use has been rated as the most difficult task following stroke rehabilitation [20]. Unlike walking, the biomechanics of stair negotiation in stroke survivors are not well described, which presents challenges for rehabilitation specialists to develop effective retraining interventions. Knowledge of lower limb kinematics and kinetics is essential to characterize the requirements of stair ascent and descent and is important to understand stroke related compensations compared to nondisabled controls. This information can guide intervention strategies aimed at improving stair mobility and community access.

This study characterizes the sagittal and frontal plane lower limb kinematics and kinetics of stair ascent and descent in people with stroke in comparison with similarly aged healthy controls.

5.3 Methodology

5.3.1 Subjects

Community-dwelling chronic stroke survivors (at least 6 months post-stroke) were recruited through newspaper advertisements and from monthly clinics held at the rehabilitation hospital. Age and sex-matched controls were drawn from those individuals who responded to newspaper advertisements and postings at the local senior’s centre. All subjects self-reported good health and were able to ascend and descend at least 4 steps with or without the use of a handrail. In addition, subjects with stroke were independently ambulatory with or without an assistive device, and presented with mobility deficit. All individuals were screened to exclude those with conditions (other than stroke) that affected ambulation (e.g. mobility limiting arthritis) or who were unable to follow instructions. All procedures were approved by the university’s
research ethics board and subjects provided their informed consent prior to participation.

5.3.2 Gait assessment

Three-dimensional motion data were obtained as subjects ascended and descended an instrumented staircase consisting of four steps (rise: 15cm; run: 26 cm; width: 56cm). Bilateral kinematic data were acquired at a sampling rate of 50 Hz using two optoelectric cameras (Northern Digital Inc, Waterloo, ON) positioned on either side of the staircase to track the position of infrared emitting diodes (IREDs). Clusters of three to four IREDs were mounted non-collinearly in moulded plastic and secured on each segment of the lower body (midfoot, midshank and midthigh) bilaterally and over the sacrum at the level of S2 using a fin which projected outwards. A force platform (AMTI, Newton, MA) mounted on concrete blocks formed the middle of the second step and recorded ground reaction forces at a sampling rate of 100 Hz.

Limb segment lengths, joint centres and rotational axes were defined using an instrumented probe embedded with four IREDs fixed relative to the tip [8,21]. In addition, bilateral virtual landmarks were identified from a static standing reference trial representing the approximate locations of the first and fifth metatarsals, lateral and medial malleoli, lateral and medial epicondyles, greater trochanters, and points aligned vertically with each greater trochanter at the level of the anterior superior iliac spine. Details can be found in Chapter 3.

5.3.3 Procedure

Once instrumented, subjects performed several practice trials to ensure lead wires were non-restricting and adequately secured. Subjects were instructed to ambulate at a self-selected pace and to place one foot on each step (step-over-step). Subjects were asked to perform trials with (H) and without (NH) the use of a handrail as they were able. Because of constraints on the experimental set-up, the handrail was always placed opposite the limb contacting the forceplate.
Three successful trials were acquired for each condition (H, NH) for ascent and descent. A clinical indicator of functional mobility and balance (Community Balance and Mobility Scale, CBM [22]) was acquired for the stroke group only.

5.3.4 Data processing and analysis

All kinematic and force platform data were filtered (second-order, low pass, Butterworth, cutoff frequency 6 Hz), and synchronized using post-processing software (Visual 3D, C-Motion, Inc., Rockville, MD). Kinematic (joint angles) and kinetic data (joint moments) were computed using a seven segment, link-segment model and an inverse dynamics approach [23]. Internal moments were calculated at the ankle, knee and hip in the local coordinate system of the foot, shank and thigh, respectively. Spatial coordinates were transformed into Cardan angles by determining the orientation of the distal segment with respect to the reference proximal segment using the x, y, z ordered sequence of rotations (representing flexion/extension, abduction/adduction and axial rotation, respectively). Kinetic data were normalized to body mass (kg) and all data were normalized to 100% of the stance phase. Most affected (Aff) and less affected (LAff) sides were analyzed in stroke subjects and the dominant side only in healthy controls since no side-to-side differences have been reported in this population [8].

Cadence (steps/min) and stance time (s) were computed from the kinematic data and joint ranges of motion (ROM) were determined. Peak values for joint moments were obtained from individual curve profiles and where appropriate, distinct peaks associated with specific phases of stance were also identified. Outcome measures for individual trials were averaged for each condition to avoid attenuation due to minor temporal shifts. Kinematic and kinetic data are reported for the ankle, knee and hip joints in the sagittal and frontal planes of movement.

Descriptive statistics (means and standard deviations) were calculated for all outcome
measures (SPSS version 19.0, San Rafael, CA). A mixed ANOVA with 1 within-subject factor (condition: NH, H) and 1 between-subject factor (group) was carried out for all kinematic and kinetic variables of interest. Interaction effects, if present, were explored by extracting factors from the larger factorial design. Paired t-tests were conducted to examine the effect of side (Aff vs. LAff) in the stroke group. A significance level of $p < .05$ was adopted for all analyses.

5.4 Results

Ten people with stroke (7 male, 3 female) with a mean age of 60.1 ($\pm$ 10.3) years who experienced a stroke an average of 28.1 ($\pm$ 16.3) months prior participated. Six presented with right hemiparesis and four had left hemiparesis, and all exhibited some degree of mobility deficit (mean CBM = 53.9 ($\pm$ 20.3)). Eight subjects completed all tasks and conditions (ascent and descent, H and NH) and two performed trials with the handrail only. Ten healthy age (mean age $\pm$ SD = 59.4 $\pm$ 8.7 years) and sex-matched controls completed all trials. T-tests were substituted for ANOVAs to analyze between group effects for each condition because the two missing datasets for the NH condition in stroke would adversely impact the cell counts in a factorial analysis.

5.4.1 Temporal gait parameters

Cadence was in all cases (ascent, descent and H, NH) lower in stroke subjects than controls ($p<.001$). This was accompanied by longer stance times (Aff and LAff sides) during ascent ($p<.04$), but not descent ($p>.10$). Data are summarized in Table 5-1.
Table 5-1  Mean (± SD) values for cadence (steps/min) and stance time (seconds) during stair ascent and descent

<table>
<thead>
<tr>
<th></th>
<th>Cadence (steps/min)</th>
<th>Stance time (seconds)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Stroke</td>
<td>Control</td>
</tr>
<tr>
<td>Ascent NH</td>
<td>71.11 ± 10.30*</td>
<td>94.72 ± 10.32*</td>
</tr>
<tr>
<td>H</td>
<td>65.60 ± 10.82*</td>
<td>95.48 ± 10.84*</td>
</tr>
<tr>
<td>Descent NH</td>
<td>75.56 ± 13.43*</td>
<td>106.82 ± 16.98*</td>
</tr>
<tr>
<td>H</td>
<td>68.13 ± 10.58*</td>
<td>102.77 ± 21.35*</td>
</tr>
</tbody>
</table>

Aff, Affected side (stroke); LAff, Less affected side (stroke); NH, No handrail; H, handrail
* Indicates a significant difference between groups (p < .05)
† Indicates a significant difference between affected and control (p<.05)
α Indicates a significant difference between less affected and control (p<.05)

5.4.2 Stair ascent

Joint angles

In general, the mean joint angle profiles were similar between groups (Figure 5-1), although the ROM at the ankle and knee associated with the LAff side were greater in the stroke patients than in controls (p<.029) in both handrail conditions. At the ankle, the LAff ROM also exceeded the Aff ROM (p=.026). For the NH condition, the affected side hip ROM was about 5° greater than observed in control subjects (p=.014).

No differences in ROM in the frontal plane were detected between groups, conditions, or sides (stroke) (p>.072) even though the illustrated mean profiles suggest disparity. The magnitudes of the joint ROM, however, are small. Therefore in the presence of variability, differences are difficult to detect.

Net joint moments

The magnitude of the extensor moments were generally smaller on the affected side compared to the LAff side and controls (Figure 5-2). This was particularly evident at the ankle.
Figure 5-1 Mean sagittal plane (left) and frontal plane (right) angles of the ankle, knee and hip during stair ascent without handrail use (left) and with handrail use (right). Positive values represent eversion/abduction and flexion angles. All curves are normalized to 100% of the stance phase. Thin line=control group; Solid thick line=Affected side, stroke group; Dashed line=Less affected side, stroke group
Figure 5-2  Mean sagittal plane (left) and frontal plane (right) moments of the ankle, knee and hip and support during stair ascent without handrail use (left) and with handrail use (right). Positive values represent internal eversion/abduction and flexion moments. All curves are normalized to 100% of the stance phase. Thin line=control group; Solid thick line=Affected side, stroke group; Dashed line=Less affected side, stroke group
during late stance (p<.001) and at the knee in early stance or weight acceptance (p<.002). The latter paired with a lower hip extensor moment compared to the LAff side (p=.018) contributed to a markedly lower support moment in early stance on the Aff side compared to the LAff side (p<.001) and controls (p<.024). The support moment in late stance (forward continuance) was lower in stroke than controls (p<.003). Handrail use effectively diminished the support moment generated in both groups during weight acceptance (p<.001) corresponding to lower plantarflexor moments (p<.005).

In the frontal plane, differences between stroke and control groups were limited to the ankle, which exhibited much lower invertor moments in stroke (both sides, p<.033). Side-to-side differences were noted at the knee (p<.047) and hip (p<.026) during handrail use only as the magnitudes of the abductor moments on the Aff side appeared to be somewhat less than observed during the NH condition (see Figure 5-2). Handrail use, however, was not associated with any significant change in moment output at any joint in the frontal plane. Moment data are summarized in Table 5-2.

5.4.3 Stair descent

Joint angles

As with stair ascent, the patterns of angular displacement were similar in shape (Figure 5-3) although there was markedly smaller ROM at the ankle on the Aff side compared to the LAff side and controls (p<.021) and greater hip ROM on the Aff side compared to controls (p<.025). On the LAff side, the mean ROM at all joints tended to be greater than in controls, significantly so at the ankle (p=.029) and knee (p=.001); the latter for the NH condition only. The use of a handrail had no effect on ROM for either group.

In the frontal plane, the joint ROMs were relatively small (<14°). There were no
differences detected between groups, conditions, or sides (stroke) (p>.138).

Table 5-2  Sagittal and frontal plane peak moments (Nm/kg) during the stance phase of stair ascent (mean ± SD)

<table>
<thead>
<tr>
<th></th>
<th>No rail, Ascent</th>
<th></th>
<th>Control</th>
<th>Rail, Ascent</th>
<th></th>
<th>Control</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Aff</td>
<td>LAff</td>
<td>Control</td>
<td>Aff</td>
<td>LAff</td>
<td>Control</td>
</tr>
<tr>
<td><strong>Sagittal plane</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle PF 1</td>
<td>-.62 ± .10*</td>
<td>-.60 ± .13*</td>
<td>-.69 ± .24*</td>
<td>-.48 ± .14*</td>
<td>-.48 ± .18*</td>
<td>-.61 ± .28*</td>
</tr>
<tr>
<td>Ankle PF 2</td>
<td>-.85 ± .17†</td>
<td>-.96 ± .14†</td>
<td>-1.17 ± .09†</td>
<td>-.83 ± .14*†</td>
<td>-.99 ± .12*†</td>
<td>-1.17 ± .11†</td>
</tr>
<tr>
<td>Knee Flexor</td>
<td>.15 ± .10</td>
<td>.25 ± .10</td>
<td>.22 ± .07</td>
<td>.12 ± .08*†</td>
<td>.21 ± .09*</td>
<td>.22 ± .12†</td>
</tr>
<tr>
<td>Knee Extensor</td>
<td>-.80 ± .15*†</td>
<td>-1.09 ± .20*</td>
<td>-1.14 ± .21†</td>
<td>-.78 ± .15*</td>
<td>-1.17 ± .19*</td>
<td>-1.12 ± .15</td>
</tr>
<tr>
<td>Hip Flexor</td>
<td>.23 ± .10*†</td>
<td>.09 ± .09*</td>
<td>.12 ± .07†</td>
<td>.21 ± .07*†</td>
<td>.11 ± .08*</td>
<td>.11 ± .10†</td>
</tr>
<tr>
<td>Hip Extensor</td>
<td>-.45 ± .27*</td>
<td>-.59 ± .32*</td>
<td>-.41 ± .11</td>
<td>-.41 ± .21</td>
<td>-.46 ± .22</td>
<td>-.40 ± .14</td>
</tr>
<tr>
<td>Support Ext 1</td>
<td>-1.70 ± .35*</td>
<td>-2.13 ± .41*</td>
<td>-2.08 ± .29*</td>
<td>-1.56 ± .27*†</td>
<td>-1.92 ± .34*†</td>
<td>-2.00 ± .25*†</td>
</tr>
<tr>
<td>Support Ext 2</td>
<td>-.83 ± .22†</td>
<td>-.82 ± .14†</td>
<td>-1.16 ± .19†</td>
<td>-.85 ± .20†</td>
<td>-.82 ± .15†</td>
<td>-1.16 ± .16†</td>
</tr>
</tbody>
</table>

| **Frontal plane** |      |      |         |      |      |         |
| Ankle Inverter   | -.15 ± .06†    | -.16 ± .07† | -.29 ± .13† | -.12 ± .06     | -.14 ± .07   | -.36 ± .39    |
| Knee Peak abd    | .29 ± .13      | .35 ± .11   | .29 ± .09   | .27 ± .10*     | .36 ± .11*   | .28 ± .14    |
| Hip Peak abd     | .64 ± .17      | .69 ± .13   | .74 ± .14   | .54 ± .20*     | .66 ± .21*   | .71 ± .27    |

Aff, Affected side (stroke); LAff, Less affected side (stroke); PF, Plantarflexion; Ext, Extension; Abd, Abduction
Positive values indicate dorsiflexion, flexion, eversion or abduction; Negative values indicate plantarflexion, extension, inversion or adduction
*indicates a significant difference between sides (stroke only)
† indicates a significant difference between groups (stroke vs. control)
α indicates a significant difference between conditions (handrail vs. no handrail)

**Net joint moments**

During descent, the mean moment magnitudes were typically smaller on the Aff side as compared to control subjects (Figure 5-4), though statistical significance was limited to the knee extensor moment during early stance (p<.030) and the support moment during controlled lowering (p<.039). Side-to-side differences were evident in the extensor moments generated at the knee (p<.012) and the total support moment (p<.013), both in late stance reflecting lower values on the Aff side.
With handrail use, inter-limb differences were also evident at the ankle with a lower plantarflexor moment on the Aff side during controlled lowering (p=.037). The impact of handrail use was to lower the magnitudes of the moments generated, though the effect was significant only with respect to the support moment during weight acceptance (p<.012).

In the frontal plane, the ankle invertor and hip abductor moments on the Aff side were smaller than those of control subjects (p<.014), but only for the NH condition. With handrail use, a side-to-side difference in the hip abductor moment was evident (p<.013) and attributed to the reduction in muscle output on the Aff side relative to the NH condition. The data associated with stair descent are presented in Table 5-3.

<table>
<thead>
<tr>
<th>Sagittal plane</th>
<th>No rail, Descent</th>
<th>Rail, Descent</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle</td>
<td>Aff</td>
<td>LAff</td>
</tr>
<tr>
<td>PF 1</td>
<td>-.61 ± .18</td>
<td>-.65 ± .25</td>
</tr>
<tr>
<td>PF 2</td>
<td>-.91 ± .15</td>
<td>-.102 ± .11</td>
</tr>
<tr>
<td>Knee</td>
<td>Ext 1</td>
<td>-.58 ± .19†</td>
</tr>
<tr>
<td></td>
<td>Ext 2</td>
<td>-.99 ± .09*</td>
</tr>
<tr>
<td>Hip</td>
<td>Flexor</td>
<td>.33 ± .08</td>
</tr>
<tr>
<td></td>
<td>Extensor</td>
<td>-.15 ± .25</td>
</tr>
<tr>
<td>Support</td>
<td>Ext 1</td>
<td>-1.23 ± .53</td>
</tr>
<tr>
<td></td>
<td>Ext 2</td>
<td>-1.59 ± .15*†</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Frontal plane</th>
<th>No rail, Descent</th>
<th>Rail, Descent</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle</td>
<td>Inverter</td>
<td>-.19 ± .08†</td>
</tr>
<tr>
<td>Knee</td>
<td>Peak abd</td>
<td>.30 ± .12</td>
</tr>
<tr>
<td>Hip</td>
<td>Peak abd</td>
<td>.69 ± .11†</td>
</tr>
</tbody>
</table>

Aff, Affected side (stroke); LAff, Less affected side (stroke); PF, Plantarflexion; Ext, Extension; Abd, Abduction
Positive values indicate dorsiflexion, flexion, eversion or abduction; Negative values indicate plantarflexion, extension, inversion or adduction
*indicates a significant difference between sides (stroke only)
† indicates a significant difference between groups (stroke vs. control)
* indicates a significant difference between conditions (handrail vs. no handrail)
Figure 5-3  Mean sagittal plane (left) and frontal plane (right) angles of the ankle, knee and hip during stair descent without handrail use (left) and with handrail use (right). Positive values represent eversion/abduction and flexion angles. All curves are normalized to 100% of the stance phase. Thin line=control group; Solid thick line=Affected side, stroke group; Dashed line=Less affected side, stroke group
Figure 5-4 Mean sagittal plane (left) and frontal plane (right) moments of the ankle, knee and hip during stair descent without handrail use (left) and with handrail use (right). Positive values represent internal eversion/abduction and flexion moments. All curves are normalized to 100% of the stance phase. Thin line=control group; Solid thick line=Affected side, stroke group; Dashed line=Less affected side, stroke group
5.5 Discussion

This paper provides a detailed kinematic and kinetic description of stair ambulation comparing stroke and healthy adults. The major findings are that while people with stroke present with kinematic and kinetic profiles that are similar in shape, there are marked differences in ROM and the magnitudes of the moments produced between groups and between sides in stroke that are important in understanding deficiencies associated with stroke. Further, the impact of handrail use in stroke, provides insight into its use in compensating physical limitations.

In overground walking, the positive relationship between speed and both range of motion and muscle output is well established [12,16,24-26]. In stroke, walking speed is markedly reduced compared to similarly aged healthy adults as a means of compensating stroke-related impairments including paresis [16,25]. The lower cadence observed in our stroke subjects during stair ascent and descent may reflect a similar compensation strategy. Although the presence of the side-to-side kinematic and kinetic differences in stroke and joint-specific alterations compared to control subjects rather than generalized adaptations in outcome amplitudes suggest that following stroke, stair negotiation poses challenges that cannot be addressed by adjusting speed alone.

5.5.1 Stair ascent

Except for reduced ankle ROM on the affected side, lower limb joint excursions were similar to the less affected side. Despite the slower cadence, the less affected ankle and knee showed a greater degree of extension than was observed in controls during midstance (see Figure 5-1). The net effect would be to raise the vertical position of the centre of mass and facilitate clearance of the intermediate step by the affected leg as it progresses through swing [12]. The corresponding extensor moments were of similar magnitude to those generated by control
subjects suggesting that any anticipated reduction due to slower speed may have been offset by the exaggerated vertical displacement.

More so than the kinematic profiles, examination of how the work of stair ascent was accomplished revealed notable disparities in pattern between sides (in stroke) and groups. The moments produced on the affected side were mostly lower than on the less affected side and at all joints were lower than those of control subjects. In healthy adults, stair climbing places considerable demand on the knee extensors [2] and together with the ankle plantarflexors, contribute significantly to generating positive work [2,8,27] as we have shown in the present study. In stroke, the slower speed reduced the ankle and knee moment requirements, however, while the plantarflexor moments were lower bilaterally, the net knee extensor output was reduced on the affected side only. Furthermore, the mean extensor moments at the hip were higher in stroke (both sides) though the trend was not borne out statistically. Nonetheless, the contribution served to generate a net extensor support moment on the less affected side during weight acceptance comparable to that of controls; a strategy that would enhance balance during transition as the affected limb is off-loaded [8]. Consistent with this interpretation, when balance was enhanced by the use of a handrail, the magnitude of the net support moment was reduced. Though this was primarily associated with a decrease in the plantarflexor moment, a generalized pattern of somewhat lower extensor moments was observed at all joints and the side-to-side difference in hip extensor moment magnitudes disappeared. In control subjects, only the plantarflexor moment showed a reduction in magnitude during weight acceptance (and a corresponding decrease in the net support moment) but no trends were detected elsewhere when a handrail was used.

In the frontal plane, only the ankle inverter moments were lower in stroke (both sides) than controls, secondary to the group’s slower cadence during stair ascent. The moment
magnitudes were greatest at the hip in both groups as the abductors help stabilize the upper body over the base of support [2,8]. That a side-to-side difference was detected in subjects with stroke during handrail use only suggests that while the hip abductors on the affected side may be able to provide balance, the challenge is lessened if an alternate solution is presented. In the case of controls, the use of a handrail yielded a negligible impact. Considering that those with stroke showed deficits in balance and mobility, it may not be surprising that the availability of a handrail influenced joint kinetics. Certainly in the case of two individuals with stroke, it allowed them to perform the task.

5.5.2 Stair descent

In contrast to stair ascent, the ROM required at all lower limb joints is larger during stance than during swing when descending stairs [3,9]; consequently limited or impaired joint mobility is more likely to alter the pattern of movement during descent. Our stroke group exhibited reduced dorsiflexion on the affected side during late stance or controlled lowering. Reeves and colleagues [9] identified this period of stance as one of high risk for falls among the elderly because they are operating near their maximum available ROM in order to lower the body to allow the opposite limb to contact the next step. By compensating with exaggerated hip flexion (Aff side), subjects with stroke were able to adequately shorten the limb to lower the swing limb onto the next step [28]. The correspondingly low extensor support moment however, may compromise balance. Interestingly, with handrail use, the net extensor moment was sustained at a comparable magnitude which may be a strategy to offset the greater instability associated with descent [5] compared to ascent and at this point in stance in particular [9]. Alternatively, the unchanged support moment may suggest that handrail use is not for mechanical support but serves another purpose such as haptic aid,. Note that during weight acceptance, the
net extensor moment was lower with handrail use, but only on the LAff side and for control subjects which further supports the notion that on the Aff side, people with stroke are not inclined to reduce the net lower limb extensor support.

Stair descent is accomplished primarily through negative work as muscles contract eccentrically to control the lowering of the body with gravity and the knee extensors are a major contributor [8,27]. In stroke, markedly lower knee extensor moments were produced on the affected side compared to the LAff side and control subjects. The lower cadence could explain this finding although the fact that the LAff side compared well to controls suggests other factors also played a role. The knee exhibits the greatest ROM of all lower limb joints during descent, which necessarily requires considerable dynamic control [29,30]. To promote stability it is possible that people with stroke increase their knee stiffness by co-contracting the knee flexors, which would reduce the net extensor moment, but increase joint stability [31,32]. Electromyographic recordings would have been useful in determining if this was the case.

Similar to stair ascent, the frontal plane moments revealed the highest output at the hip as the abductors stabilize the trunk over the support limb. With handrail use the Aff side abductor moment was lower than without (though not statistically significant), whereas the LAff abductor moment was similar in magnitude resulting in a significant side-to-side difference. We suggest that the handrail serves to provide lateral stability enabling less reliance on the paretic abductors. Future studies warrant the instrumentation of the handrails to quantify the extent of any compensation occurring through the upper extremities. This would certainly permit confirmation of our interpretation, although it would not alter our conclusion regarding the impact on lower extremity moments given the inverse dynamics approach used.
Several limitations exist with the current study. First, the absence of an instrumented handrail precluded measurement of the forces applied through the hand. As such, we were unable to determine if any effects noted were due to actual mechanical aid (ie. upper limb loading), haptic aid (such as increased confidence secondary to sensory information from cutaneous receptors), altered orientation of the upper body (ie. postural effect), or some combination thereof. Future investigation of the loads placed through the handrail is warranted to facilitate interpretation of the data. It also warrants stating that this comprehensive biomechanical analysis of stair ambulation in stroke is limited to individuals with mild to moderate mobility impairment (as indicated by scores on the CBM) and are not generalizable to the entire stroke population. Even so, it is clear that compared to age-matched healthy adults, significant alterations in movement patterns and movement control were evident thus highlighting the challenge of stair ambulation in this group.

There has been a paucity of information describing how people with stroke manage stairs and the strategies they employ. The findings presented describe the mechanics and highlight stroke-related differences in lower limb mobility and motor output that are not attributable to aging. Slower cadence reduces the muscle work required in stroke however lower extremity extensor weakness limits the overall net support compared to healthy adults. Furthermore lower hip abductor moments reduce trunk stability in stroke. Future work is required to determine whether strengthening can augment balance during stair negotiation to promote safe mobility.
5.6 References


Chapter 6

Strength and aerobic requirements increase during stair ambulation in persons with chronic stroke

*Manuscript Under Review, Archives of Physical Medicine and Rehabilitation, August 2011*

6.1 Abstract

Information about the relative cost of stair ambulation in terms of strength and oxygen consumption can enhance our understanding of the challenges associated with mobility function post-stroke although to date these requirements remain ill-defined. The purpose of this study was to estimate the cost of stair ascent and descent in relation to a measured standard of neuromuscular (strength) and metabolic (aerobic) capacities in persons with chronic stroke in comparison to age-matched healthy adults. Ten persons with stroke (7 males, 3 females) and 10 sex and age-matched older adults participated in the study. To provide an estimate of the relative strength cost of stair ascent and descent, the maximum moment generated during stair walking was expressed as a ratio of the maximum strength assessed using a dynamometer. The relative aerobic cost of stair negotiation was determined as the ratio of the oxygen consumption measured during stair walking to the maximal oxygen consumption estimated from a submaximal cycle ergometer test. Self-selected cadence was lower in stroke than controls during both stair ascent and descent. During ascent, the strength cost was higher for the affected ankle (stroke) compared to the less affected ankle and controls and the knee (both sides) compared to controls. During stair descent, the relative strength cost was higher for the plantarflexors and knee extensors in stroke (both sides) than controls although the affected and less affected sides were similar in the
stroke group. In terms of aerobic requirements, oxygen consumption was comparable between
groups when ascending and descending one flight of stairs. However, the estimated maximum
aerobic capacity was significantly lower in stroke than control subjects, which translated into a
much higher relative aerobic cost of stair ascent and descent in this group. This is the first study
to examine and compare the relative costs of stair negotiation in people with stroke and healthy
controls. The results indicate strength and aerobic costs are higher during stair negotiation in
stroke primarily due to reduced neuromuscular and aerobic capacities, respectively. These results
serve to identify factors that may limit stair negotiation.
6.2 Introduction

Regaining independent ambulation is a major goal of rehabilitation post-stroke [1]. Deficits in muscle strength, coordination, and balance [2-5] all contribute to the reduced capacity for independent mobility following a stroke. In addition, up to 75% of stroke survivors have cardiovascular disease [6] which may contribute to reduced physical activity, deconditioning and decline in mobility [7]. In combination, these deficits contribute to the high energy cost of mobility, which in the case of level walking is estimated to be at least twice that of age-matched individuals [8,9]. The demands associated with physical activities such as stair negotiation, which requires greater lower limb strength [10-12], range of motion [10-13] and metabolic requirements compared to level ground walking [14-17] may challenge stroke survivors such that they approach the limits of their physical abilities. Information about the relative ‘cost’ of stair ambulation in terms of strength and oxygen consumption in relation to a standard measure of capacity can enhance our understanding of the challenges associated with mobility function post-stroke.

To estimate the relative strength cost of gait, the moments of force generated during level walking have been expressed as a proportion of peak torque generated isokinetically at the corresponding joint [18]. This approach has served to illustrate that even at the lower self-selected gait speeds typical of stroke survivors compared to healthy adults, hemiparetic individuals had to produce a greater level of muscular effort than the able-bodied group. That is, the joint moments generated (i.e. the task demands) expressed as a percentage of the peak muscular output determined through strength testing were larger for stroke survivors compared to healthy controls, although relative strength costs were similar between the paretic and non-paretic limbs in stroke [19,20]. Applying a similar paradigm to stair negotiation, Reeves and
colleagues recently showed that older adults utilize a higher proportion of their peak strength capacity than young adults when ascending and descending stairs [21,22]. The implication is that older adults have less reserve strength than their young counterparts, which may contribute to their higher fall risk [23] as their capacity to generate higher forces to, for example, maintain balance is limited.

In the presence of weakness secondary to stroke, it is not known whether the adoption of lower speed or cadence corresponds to maintaining the task demands within safe or normal limits or if the cost of stair ambulation is exceedingly high in this group. An additional concern is the aerobic challenge. Poor cardiovascular health in stroke [24,25] and associated abnormally low aerobic capacity [26] can compromise the ability to meet the oxygen demands of climbing stairs, which requires greater consumption than walking [15]. This has not been explored.

The ability to manage stairs is a significant determinant of independence and active community living [27], however the physical cost associated with stair negotiation remains ill-defined. Understanding the strength and metabolic requirements are important for assessing the capacity for community mobility. The purpose of this study was to estimate the cost of stair ascent and descent in relation to standard measures of peak neuromuscular (strength) and metabolic (aerobic) capacities in persons with chronic stroke in comparison to age-matched, healthy adults.

6.3 Methodology

6.3.1 Subjects

Individuals with hemispheric stroke of at least 6 months duration were invited to participate. Subjects were recruited through newspaper advertisements, stroke support groups and outpatient stroke clinics. Age and sex-matched control subjects were drawn from those
individuals who responded to newspaper advertisements and postings at the local senior’s centre. All participants met the following criteria: independent ambulators (cane or assistive device acceptable in stroke), able to ascend and descend at least 4 steps (with or without the use of a handrail) and were (self-reported) in good health. Those reporting a non-stroke related mobility restriction (e.g. arthritis), a history of serious cardiac disease (unstable angina, peripheral vascular disease, congestive heart failure), uncontrolled hypertension, or were unable to understand and follow instructions were not eligible. Demographic characteristics including age, sex, time post-stroke were documented and relevant medications such as beta-blockers were noted. All subjects provided their informed consent and the study protocol was approved by the university and affiliated hospitals research ethics board.

6.3.2 Gait assessment

Subjects ascended and descended a customized four step staircase in a step-over-step manner with (H) and without (NH) the use of a handrail as they were able. A forceplate (AMTI, Newton, MA) was mounted on concrete blocks, forming the middle of the second step. Two optoelectric camera banks were placed on either side of the staircase for bilateral acquisition of infrared emitting diodes (IREDs) secured on the subject’s lower limbs. Using commercial software (Visual 3D, C-Motion, Inc., Germantown, MD), segment kinematics and joint moments were computed using an inverse dynamics approach and a seven segment link-segment model (see [28] for details). In order to determine the relative strength cost associated with the each of the strategies of stair ascent and descent (see Relative strength cost below), the peak sagittal plane moments were identified during stance and the joint angle and velocity at which they occurred documented. These data were extracted for the affected (Aff) and less affected (LAff) sides in stroke and for the dominant limb only in the control group for the purpose setting the parameters
associated with testing strength in order to estimate the relative strength cost (see below).

6.3.3 Estimation of relative strength cost

Peak concentric and eccentric torques of the ankle, knee and hip flexors and extensors were measured bilaterally in stroke and on the dominant side only in control subjects using an isokinetic dynamometer (Biodex, System 3, Shirley, NY). When testing ankle and knee strength subjects were seated on the dynamometer chair tilted back 5º from vertical. For ankle testing, the foot was secured in a footplate in a neutral position and the hip and knee angle set such that the shank was horizontal, and the knee positioned in approximately 60º of flexion (0º = full extension). For testing at the knee, a padded strap at the distal end of the dynamometer’s lever arm was secured around the subject’s lower leg approximately 2 cm above the lateral malleolus. Hip flexors/extensors were tested with the subjects in supine with their lower limbs able to swing freely over the edge of the seat pan (the non-test limb was supported by a stool). For all testing, the relevant joint centre was aligned with the mechanical axis of the dynamometer and large straps were applied to stabilize the trunk, pelvis and the segment proximal to the joint of interest (see Figure 6-1).

Peak concentric and eccentric torques were measured within subjects’ active range of motion and at speeds corresponding to the velocity at which the respective peak joint moment was produced during stair ascent and descent for the NH and H conditions [21,22]. Four trials of five repetitions each were completed for every joint with each individual trial set at the appropriate velocity representing the four task conditions (NH and H ascent (concentric); NH and H descent (eccentric)) with ample rest provided between trials. The peak torque generated in a trial was operationally defined as the maximum strength. The relative strength requirement or cost of stair ascent (descent) was calculated as the ratio of the peak moment generated during
ascent (descent) to the maximum strength and was expressed in percent.

Figure 6-1 Dynamometric position when testing the ankle (left), knee (middle) and hip (right)

6.3.4 Estimation of relative metabolic cost

Metabolic demands during stair ascent and descent were determined using a telemetered metabolic unit (Cosmed K4b², Chicago, IL) and heart rate (HR) monitor which provided breath-by-breath gas analysis and HR, respectively. The metabolic unit was calibrated prior to each test session using a standard mixture of known oxygen (O₂) and carbon dioxide (CO₂) gas concentrations to ensure accurate sensor operation (as per manufacturer’s specifications). Flow sensor calibration was also performed using a 3000 ml syringe.

Subjects were asked to breathe normally as they wore a mask securely covering their nose and mouth. The mask was equipped with a turbine that allowed room air to enter and exhaled air to be transmitted to a gas exchange analyzer in the portable unit worn in a chest harness. A polar HR monitor was strapped around the subject’s chest and connected to the portable unit of the metabolic system (Figure 6-2).

Once instrumented, resting metabolic rate was recorded as subjects sat quietly for approximately 7-10 minutes. Metabolic data were then collected as the subjects descended a full flight of stairs in a step-over-step manner with handrail use (16 steps; rise=15 cm, run=27 cm)
and rested at the bottom. Once the oxygen consumption (VO$_2$) and HR returned to resting values, the subjects ascended the stairs using the same strategy.

HR, average and peak VO$_2$ (ml/kg/min), and the respiratory exchange ratio (RER) were recorded during both ascent and descent.

**Figure 6-2 Metabolic instrumentation**

To estimate the relative oxygen utilization associated with stair ascent and descent required knowledge about the maximal VO$_2$ (mVO$_2$). A cycle ergometer was fitted with straps on the pedals to hold the feet in place and an office chair seat pan with back support replaced the regular saddle in order to provide added support while subjects performed the Astrand-Ryhming submaximal cycle ergometer test [29,30]. This is a single-stage submaximal test requiring subjects to pedal for 6 minutes at a rate of 50 RPM. Following a 2 to 3 minute warm-up with zero resistance, the initial workload (50-150 W) was set based on the individual’s gender and fitness.
status. Workload increments (25-50 W) were applied after 2 minutes if the HR was below 125 beats/min (if higher than 170 beats/min, the workload was decreased). The mean HR for the last 10 sec of each of the last two minutes of the test was used to estimate the mVO$_2$ from a nomogram [30]. Note that the target HRs were adjusted for those participants who were on beta-blocker medications [31]. Testing was terminated in accordance with the American College of Sports Medicine guidelines [32] or if a subject was unable to maintain the required pedaling rate. Submaximal exercise testing has been previously shown to be a reliable and valid predictor of maximal oxygen consumption in chronic stroke [33].

The ratio of the VO$_2$ measured during stair walking to the estimated mVO$_2$ provided an estimate of the relative aerobic cost of stair ascent and descent.

### 6.3.5 Procedure

Data collection took place over two sessions approximately 1 week apart. Gait assessment and the submaximal bicycle ergometer test were completed in the first session. The dynamometric testing and metabolic assessment of stair ambulation were completed in the second session. It was necessary to conduct the gait assessment prior to dynamometric testing since the isokinetic test velocities were based on those achieved during stair ascent and descent.

To provide a clinical indication of mobility function, each subject was assessed using the Community Balance and Mobility Scale [34].

### 6.3.6 Statistical analysis

Descriptive statistics (means and standard deviations) were calculated for all outcome measures (SPSS version 19.0, San Rafael, CA). Independent samples t-tests were carried out to identify differences between groups for all metabolic measures and dynamometric measures. Paired t-tests were also conducted to examine the effect of side (affected vs. less affected) in the
stroke group only for dynamometric measures. To explore the effect of handrail condition (NH, H) and task (ascent, descent), a mixed ANOVA with 1 between-subject factor (group) and 2 within-subject factors (condition: NH, H; task: ascent, descent) was carried out for measures of relative strength costs. Interaction effects, if present, were explored by extracting factors from the larger factorial design. A significance level of p<.05 was adopted for all analyses.

6.4 Results

6.4.1 Subject characteristics

Fourteen people with stroke indicated a willingness to participate. Of these ten met the study criteria and were matched by sex and age with 10 healthy controls. All subjects completed the testing without incident, however, two subjects with stroke were only able to ascend and descend stairs with the use of the handrail. Subject characteristics are presented in Table 6-1.

Table 6-1 Subject characteristics

<table>
<thead>
<tr>
<th></th>
<th>Stroke group (n=10)</th>
<th>Control group (n=10)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>60.1 ± 10.3</td>
<td>59.4 ± 8.7</td>
</tr>
<tr>
<td>Sex (male/female)</td>
<td>7/3</td>
<td>7/3</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>72.4 ± 15.9</td>
<td>79.2 ± 13.0</td>
</tr>
<tr>
<td>Height (m)</td>
<td>1.7 ± .1</td>
<td>1.8 ± .1</td>
</tr>
<tr>
<td>β-blocker medication*</td>
<td>4/10 (40%)</td>
<td>0/10 (0%)</td>
</tr>
<tr>
<td>Side of stroke (right/left)</td>
<td>6/4</td>
<td>---</td>
</tr>
<tr>
<td>Time post-stroke (months)</td>
<td>28.1 ± 16.3</td>
<td>---</td>
</tr>
<tr>
<td>CBM score*</td>
<td>53.9 ± 20.3</td>
<td>89 ± 6.3</td>
</tr>
<tr>
<td>Stair ascent cadence</td>
<td>NH: 71.1 ± 10.3</td>
<td>NH: 94.7 ± 10.3</td>
</tr>
<tr>
<td>(steps/min)*</td>
<td>H: 65.1 ± 10.8</td>
<td>H: 95.5 ± 10.8</td>
</tr>
<tr>
<td>Stair descent cadence</td>
<td>NH: 75.6 ± 13.4</td>
<td>NH: 106.8 ± 17.9</td>
</tr>
<tr>
<td>(steps/min)*</td>
<td>H: 68.1 ± 10.6</td>
<td>H: 102.8 ± 21.4</td>
</tr>
</tbody>
</table>

CBM, Community Balance and Mobility Scale; NH, No handrail; H, Handrail
Note: Values are mean ± SD or n (%)  
* Indicates a significant difference between groups (p < .05)
### 6.4.2 Strength requirements

Kinematic and kinetic descriptions of stair ascent and descent are reported elsewhere (see Chapter 5), however a summary of the data used to set the parameters for isokinetic strength testing are presented in Table 6-2. Note that muscles performed positive work (concentric) during stair ascent and primarily negative work (eccentric) during descent except at the hip joint where both concentric and eccentric activity was evident during descent.

**Table 6-2** Gait variables (mean ± SD) during stair ascent and descent

<table>
<thead>
<tr>
<th></th>
<th>No handrail</th>
<th>Handrail</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Peak moment (Nm/kg)</td>
<td>Velocity at peak moment (%/s)</td>
</tr>
<tr>
<td><strong>Ascent</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Aff</td>
<td>-.85 ± .17*</td>
<td>22.36 ± 23.93*†</td>
</tr>
<tr>
<td>LAff</td>
<td>-.96 ± .14*</td>
<td>76.13 ± 34.21†</td>
</tr>
<tr>
<td>Control</td>
<td>-1.17 ± .09*</td>
<td>60.94 ± 29.62*</td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Aff</td>
<td>-.80 ± .15*†</td>
<td>100.02 ± 12.17</td>
</tr>
<tr>
<td>LAff</td>
<td>-1.09 ± .20†</td>
<td>93.63 ± 12.31</td>
</tr>
<tr>
<td>Control</td>
<td>-1.14 ± .21*</td>
<td>102.90 ± 11.81</td>
</tr>
<tr>
<td>Hip</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Aff</td>
<td>-.45 ± .27†</td>
<td>71.24 ± 22.43</td>
</tr>
<tr>
<td>LAff</td>
<td>-.59 ± .32†</td>
<td>55.09 ± 13.07*</td>
</tr>
<tr>
<td>Control</td>
<td>-.41 ± .11</td>
<td>87.11 ± 18.62*</td>
</tr>
<tr>
<td><strong>Descent</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Aff</td>
<td>-.98 ± .07†</td>
<td>33.66 ± 13.57</td>
</tr>
<tr>
<td>LAff</td>
<td>-1.06 ± .10†</td>
<td>53.62 ± 48.42</td>
</tr>
<tr>
<td>Control</td>
<td>-1.02 ± .14</td>
<td>44.08 ± 34.85</td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Aff</td>
<td>-.95 ± .15*†</td>
<td>135.63 ± 34.96</td>
</tr>
<tr>
<td>LAff</td>
<td>-1.19 ± .17†</td>
<td>119.29 ± 36.04</td>
</tr>
<tr>
<td>Control</td>
<td>-1.19 ± .22*</td>
<td>129.41 ± 36.52</td>
</tr>
<tr>
<td>Hip (Ext moment)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Aff</td>
<td>-.42 ± .41</td>
<td>7.28 ± 10.13</td>
</tr>
<tr>
<td>LAff</td>
<td>-.51 ± .30</td>
<td>9.69 ± 1.65</td>
</tr>
<tr>
<td>Control</td>
<td>-.43 ± .16</td>
<td>16.8 ± 13.58</td>
</tr>
<tr>
<td>Hip (Flex moment)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Aff</td>
<td>.35 ± .08</td>
<td>23.71 ± 16.42</td>
</tr>
<tr>
<td>LAff</td>
<td>.43 ± .21</td>
<td>16.88 ± 11.53</td>
</tr>
<tr>
<td>Control</td>
<td>.47 ± .22</td>
<td>21.19 ± 11.12</td>
</tr>
</tbody>
</table>

Aff, Affected side (stroke); LAff, Less affected side (stroke); Ext, extension; Flex, flexion
Positive values indicate dorsiflexion or flexion

Note: During stair descent, mean values for hip extension or hip flexion are taken from only the subjects whose peak moment generated represents the direction of interest and do not reflect the entire group

* indicates a significant difference between groups (stroke vs. control)
† indicates a significant difference between sides (stroke only)
In stroke, several subjects were unable to produce concentric and eccentric contractions at the required velocities. Therefore, the protocol was adjusted for all subjects to reflect a reduction in speed to 30°/s at the ankle and 60°/s at the knee in order to ensure standardized testing across all subjects. Furthermore, if the velocity required for testing at the hip was less than 30°/s, then isometric testing was substituted. All healthy older adults were able to complete isokinetic testing at angular velocities that matched those associated with peak moments generated during ascent or descent as appropriate; however, they were also tested at the fixed speeds to enable comparison with the stroke group. One individual in the stroke group was unable to perform the ankle testing on the affected side due to limited active range of motion (ROM). Although ROM was consistently smaller on the Aff side compared to the LAff side in stroke (p<0.05, Table 6-3), it rarely limited the ability to reach the desired test velocities.

Dynamometric measures confirmed affected side extensor weakness of the ankle and knee relative to the LAff side (p<0.003) and for all Aff side lower limb joint extensors compared to controls during concentric testing (p<.003). Similar results were found for eccentric strength, although both sides in stroke were weaker than control subjects (p<.032). Data are summarized in Table 6-3.

In terms of the relative strength cost of stair negotiation, the cost was higher for the affected plantarflexors than the LAff and controls muscles (p<.033) during stair ascent. The mean strength cost at the knee was also significantly higher than observed in the control group (both sides) (p<.041), Figure 6-3a. During stair descent, the relative strength cost was higher for the plantarflexors and knee extensors in stroke (both sides) than controls (p<0.041) but the Aff and LAff sides were similar. Handrail use did not alter the findings. As was the case with ascent, the relative strength requirements at the hip were comparable for both sides (stroke) and control
subjects (p>.108) (Figure 6-3b). For both conditions (NH and H) the relative strength costs were also generally lower during descent than ascent (p<.001) for stroke and control groups, except at the hip where they are comparable between tasks (p>.087) (see Figure 6-3a and b).

**Table 6-3** Dynamometric outcomes (mean ± SD)

<table>
<thead>
<tr>
<th></th>
<th>Stroke</th>
<th>Control</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Aff side</td>
<td>LAff side</td>
</tr>
<tr>
<td>Ankle ROM (degrees)</td>
<td>46 ± 12.09†*</td>
<td>58.7 ± 8.78†</td>
</tr>
<tr>
<td>Knee ROM (degrees)</td>
<td>101.2 ± 13.23*</td>
<td>107.3 ± 11.37</td>
</tr>
<tr>
<td>Hip ROM (degrees)</td>
<td>92.7 ± 20.31†*</td>
<td>99 ± 16.81†</td>
</tr>
</tbody>
</table>

Concentric

- Peak plantarflexor torque (Nm/kg), 30°/sec: -.45 ± .14†* - .77 ± .17† - .89 ± .29*
- Peak knee extensor torque (Nm/kg), 60°/sec: -.90 ± .25†* - 1.26 ± .32† - 1.65 ± .54*
- Peak hip extensor torque (Nm/kg), matched velocity, no-handrail: -1.14 ± .15* - 1.27 ± .32 - 1.76 ± .64*
- Peak hip extensor torque (Nm/kg), matched velocity, handrail: -1.13 ± .17* - 1.23 ± .33* - 1.82 ± .71*

Eccentric

- Peak plantarflexor torque (Nm/kg), 30°/sec: -.99 ± .28†* - 1.21 ± .37†* - 1.84 ± .41*
- Peak knee extensor torque (Nm/kg), 60°/sec: -1.45 ± .38†* - 1.80 ± .43†* - 2.65 ± .57*

Isometric

- Peak hip extensor torque (Nm/kg): -1.16 ± .22 - 1.19 ± .27 - 1.40 ± .34
- Peak hip flexor torque (Nm/kg): .70 ± .17 .73 ± .32 1.08 ± .56

Aff, Affected side; LAff, Less affected side; ROM, Range of motion
† indicates a significant difference between sides (stroke only)
* indicates a significant difference between groups (stroke vs. control)
Figure 6-3  Estimated relative strength cost (%) to ascend (a) and descend (b) stairs without (NH) and with (H) the use of a handrail. Values are means plus 1 standard deviation. 
* indicates significant difference between groups (p<.05).
† indicates significant difference between sides (p<.05).
6.4.3 Aerobic requirements

Resting oxygen consumption and HR were similar between groups (p>.108) and increased markedly during ascent and descent for both stroke and control subjects. The control group managed the stairs more quickly than those with stroke (p<.001), but despite this, the intensity of the task as reflected by the respiratory exchange ratio was comparable between groups for all conditions (p>.382).

Submaximal cycle ergometer testing was terminated for three individuals with stroke as they were unable to adequately pedal; all control subjects completed the test. The estimated maximum aerobic capacity was significantly lower in stroke than control subjects (p<0.05), which translated into a much higher relative aerobic cost of stair ascent and descent in stroke (p<0.005, Figure 6-4). The metabolic data are summarized in Table 6-4.

Table 6-4 Metabolic outcomes (mean ± SD) associated with different conditions.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Stroke</th>
<th>Control</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rest</td>
<td>Avg HR (beats/min)</td>
<td>64.8 ± 9.2</td>
<td>64.6 ± 11.8</td>
</tr>
<tr>
<td></td>
<td>Avg VO$_2$ (ml/kg/min)</td>
<td>3.17 ± .42</td>
<td>2.62 ± .72</td>
</tr>
<tr>
<td></td>
<td>RER</td>
<td>.81±.16</td>
<td>.85 ± .15</td>
</tr>
<tr>
<td>Ascent</td>
<td>Time of task (s)</td>
<td>15.1 ± 2.7*</td>
<td>9.1 ± 1.7*</td>
</tr>
<tr>
<td></td>
<td>Avg HR (beats/min)</td>
<td>75.7 ± 9.5</td>
<td>74.5 ± 13.8</td>
</tr>
<tr>
<td></td>
<td>VO$_2$ peak (ml/kg/min)</td>
<td>8.13 ± 2.83</td>
<td>6.54 ± 1.48</td>
</tr>
<tr>
<td></td>
<td>Avg VO$_2$ (ml/kg/min)</td>
<td>6.21 ± 2.22</td>
<td>5.19 ± 1.43</td>
</tr>
<tr>
<td></td>
<td>RER</td>
<td>.82 ± .17</td>
<td>.81 ± .09</td>
</tr>
<tr>
<td>Descent</td>
<td>Time of task (s)</td>
<td>16.3 ± 4.8*</td>
<td>8.6 ± 1.4*</td>
</tr>
<tr>
<td></td>
<td>Avg HR (beats/min)</td>
<td>73.9 ± 8.1</td>
<td>74.9 ± 15.4</td>
</tr>
<tr>
<td></td>
<td>VO$_2$ peak (ml/kg/min)</td>
<td>7.19 ± 1.78</td>
<td>6.23 ± 1.11</td>
</tr>
<tr>
<td></td>
<td>Avg VO$_2$ (ml/kg/min)</td>
<td>5.90 ± 1.41</td>
<td>5.36 ± .88</td>
</tr>
<tr>
<td></td>
<td>RER</td>
<td>.82 ± .17</td>
<td>.78 ± .08</td>
</tr>
<tr>
<td>Submaximal Cycle Test</td>
<td>mVO$_2$ (ml/kg/min)</td>
<td>27.91 ± 6.53*</td>
<td>36.67 ± 8.49*</td>
</tr>
</tbody>
</table>

HR, Heart rate; RER, Respiratory exchange ratio; mVO$_2$, Maximal VO$_2$ estimated from submaximal exercise test

Note: for measures of rest, ascent and descent n=10. For the submaximal cycle test, n=7.
6.5 Discussion

This is the first study to examine and compare the relative strength and aerobic costs of stair negotiation in people with stroke and healthy controls. The main findings are that the strength and aerobic costs are higher during stair negotiation in stroke even though they move more slowly. This is primarily due to reduced neuromuscular and aerobic capacities, respectively and serves to highlight the elevated task demands in stroke compared to similarly aged healthy adults.

6.5.1 Strength requirements during stair ascent and descent

The determination of relative strength cost at a given joint is dependent on both the moment produced during task performance and the peak torque generated under test conditions. Theoretically, the relative cost should not exceed 100%; however when one considers the different dynamics of stair negotiation and dynamometric testing (e.g. closed versus open chain; the relative orientation of segments) and any neural or mechanical effects of body positioning, this is not likely to be the case [35-37]. Porter et al. [36,37] reported consistently higher plantar-
and dorsiflexor torques when young and older healthy subjects were in an upright position compared to supine. They attributed this to different mechanics as the support moment would impact the standing condition only, as would be the situation in the current study. It is also the case that when managing stairs, motion occurs at multiple joints simultaneously, such that bi-articular muscles impact the net moments generated at those joints that they cross and energy can be transferred across segments [22,38]. These factors impact the calculated value of the relative strength cost, although there is no reason to suspect that people with stroke or older healthy adults would be differentially impacted. Consequently comparisons between groups are relevant and the calculated relative costs are scalable, though the actual values are difficult to interpret.

Muscle weakness is among the most commonly reported and studied deficit following stroke [2,19,20,39-45]. Our data confirm the findings of others, demonstrating the greatest weakness in the more distal muscles contralateral to the lesion (affected side) [5,42,46], but also to a lesser extent on the less affected side. Consequently, the relative strength cost to ascend stairs is abnormally high in stroke in order to produce adequate ankle and knee extensor support to perform the work necessary to accomplish the task. The slower cadence adopted by those with stroke served in part to reduce the magnitude of the net plantarflexor and knee extensor moments generated compared to healthy adults (with the exception of the affected ankle, the group differences disappeared with cadence as a covariate) however this strategy did not translate into a lower relative strength cost. Furthermore, unlike overground walking during which the hip flexors effectively compensate weak plantarflexors by pulling the stance limb into swing [45,47] a similar strategy is not possible in stair ascent since extensor activation is essential to progress both vertically and horizontally [12,48].
During stair descent, forces are primarily produced eccentrically to lower the body to the next step [48] and the efficiency associated with negative work is apparent by the lower cost or relative strength requirements compared to ascent. However, the relative strength requirements for both the affected and less affected plantarflexors and knee extensors exceeded those of healthy subjects. As reported elsewhere [49], in stroke as in healthy adults, muscles produce greater output eccentrically than concentrically, but even so eccentric paresis is evident in stroke. Similar to stair ascent, the lower cadence of individuals with stroke influenced moment magnitudes during task performance but the relative cost remained abnormally high. There was however, a tendency to scale the relative cost such that the costs were comparable for both Aff and LAff limbs.

The relative strength requirements at a joint during level ground walking have been quantified by others to reveal that lower self-selected speeds resulted in comparable muscular costs or levels of effort between affected and less affected sides in stroke [20]. The authors hypothesize that the sense of effort is involved in the regulation of motor output during functional activities such as gait. The same could apply to stair negotiation, as evidenced by the similarity in relative cost even though maximum strength differed between sides in stroke; the exception occurred at the ankle during ascent. In this case, the excessive asymmetry in concentric strength paired with the important extensor contribution at the ankle during ascent may have limited the ability to equilibrate “effort” between limbs.

Our results did not reveal any significant effect of handrail use in terms of modulating strength costs during either stair ascent or descent. However, during descent (higher instability than ascent [50]), the control group appears to demonstrate a tendency toward lower strength costs with handrail use than without (see Figure 6-3b), suggesting that the handrail may
compensate to some degree the strength requirements of the task. This is not so in stroke, likely a reflection of the greater perceived instability and risk associated with descent in the stroke group which is not offset by access to a handrail. The greater demands of the task coupled with coordination and balance deficits [51-53] evidenced by lower scores on the Community Balance and Mobility scale could limit the utility of a handrail as a means of compensating lower limb weakness.

6.5.2 Aerobic requirements during stair ascent and descent

Reduced cardiorespiratory fitness in chronic stroke contributes to reductions in gait performance and the elevated energy cost of walking [54-56]. Although there is a paucity of literature examining the impact on stair negotiation, it follows that the higher physical task demands could pose considerable challenge for a compromised cardiovascular system.

As expected, the estimated maximum aerobic capacity in our stroke group was markedly lower than in their healthy counterparts [57], although the values are higher than those derived from maximal rather than submaximal cycle ergometer tests (14.0 to 23.2 ml/kg/min; [16,33,56,58]. Previous reports have shown that predictions based on submaximal exercise testing generally overestimate maximal oxygen consumption [57,59]; in persons with stroke this is likely because maximal oxygen consumption is based on a symptom-limited VO2 peak measurement rather than a “true” VO2 max score [57]. Despite this, the VO2 measures from submaximal exercise testing have excellent reliability and good concurrent validity with those determined from maximal testing in persons with chronic stroke [33].

The mean peak and average oxygen consumption during ascent and descent were comparable between stroke and controls. In level ground walking, deviation from the preferred speed (i.e. faster or slower) results in greater oxygen consumption [60-62]. Since both groups
self-selected their speed of ascent and descent it is reasonable then that the absolute oxygen requirements were comparable, but when expressed relative to mVO\(_2\) it was evident that the stroke group accomplished the task at greater aerobic cost. The similar RER values between groups imply that the higher cost in stroke was still being met through aerobic metabolism.

We are unable to contrast our findings with other reports in the literature since our subjects ascended and descended only one flight of stairs. Usually subjects complete approximately 3 to 4 flights of stairs to reach steady-state [15,16,63]; a protocol not well-suited to people with stroke. As a consequence, the peak VO\(_2\) is likely lower than what would be measured in steady state, but this would be the case for both groups and therefore should not alter the interpretation of our findings.

The higher relative aerobic cost paired with higher strength costs of stair negotiation underscore the high demands of mobility in stroke; even in those mildly to moderately impaired. Post-stroke impairments in strength and coordination can result from a reduction in the number of motor units [64]. If a larger proportion of the available pool is required to generate the force required then aerobic demands can be elevated unless compensated by reducing moment magnitudes by slowing the cadence. Physical inactivity post-stroke can contribute to further deconditioning affecting cardiorespiratory fitness levels [7] which would impact maximum aerobic capacity as demonstrated in our stroke group. Recognizing the importance of stair mobility in maintaining independence and being active in the community [65], the elevated cost of the task can be very limiting. Improving functional capacity in stroke survivors can reduce the physiologic burden of performing mobility activities, thus increasing the likelihood that persons with stroke will be able to perform physical activities at a lower strength and aerobic cost [66,67]. Indeed exercise and aerobic training can improve stair climbing ability in stroke [68], which
according to our findings is clearly warranted.

Stair mobility is key to successful return to community activity [67]. Our findings in individuals with mild to moderate mobility impairment provide the first evidence of the abnormally high costs associated with stair negotiation in terms of strength and aerobic requirements. These results reinforce the need to improve physical capacity following stroke and serve to identify factors that may limit stair negotiation.
6.6 References


Chapter 7

General discussion

Older adults have identified stair negotiation as one of the most difficult mobility tasks attributable to aging and is one of the leading causes of fall related injuries [1-3]. In the presence of impairment secondary to stroke, the ability to ascend and descend stairs can be a determining factor of whether a person can return home following inpatient care and recent research has shown it to be the single best predictor of community ambulation activity [4]. Undoubtedly, the ability to safely and efficiently negotiate stairs is an essential skill for independent ambulation and community accessibility.

To date, extensive research has described movement patterns in older adults and stroke survivors during level ground walking. Community mobility, however, is not only a function of walking capacity, but the ability to climb and descend stairs safely [2,4]. In older adults, the literature describing the task demands of stair negotiation is limited and in stroke this aspect of mobility has been virtually unexplored. However, for rehabilitation professionals, it is important to appreciate the nature and extent of adaptations made during stair negotiation as a natural progression of aging in order to be able to identify unique alterations in movement patterns due to the superimposition of physical impairments or physical disability. This dissertation is a first step in addressing the knowledge gap.

7.1 Recapitulation of findings and their relevance

The compilation of the four studies that formed the basis of this dissertation advance our understanding of the physical demands and costs associated with stair negotiation related to
normal, healthy aging and impairment related to stroke. Specifically, this dissertation has: 1) identified age-related alterations in the mechanics of stair negotiation (Studies 1 & 2); 2) described the biomechanical patterns of movement and adaptations made during stair negotiation in persons with stroke (Study 3); and 3) determined the strength and aerobic requirements to negotiate stairs relative to a standard evaluation of maximum capacity (Study 4). A summary of the findings from this work and their relevance are discussed below.

7.1.1 Biomechanical alterations of stair negotiation related to healthy aging

Internal moments of force are generated to counter the external forces acting on the body throughout locomotion and represent the net effect of all agonist and antagonist muscle activity acting across a particular joint. As such, the investigation of muscle moments provides insight into the motor control strategies used during stair negotiation that cause the observed movement patterns. Few studies have reported the kinetics of stair negotiation and of those, most are limited to the sagittal plane of movement [5-10], one or two lower limb joints [7-9] or a single age group [5,6,11]. The first study (Chapter 3) provides a comprehensive evaluation of the lower extremity joint kinetics and their variability during stair ascent and descent in young and healthy older adults. The results showed age-related differences in peak moments generated reflecting a strategy to reduce reliance on the distal musculature, promote greater support in the sagittal plane and enhance lateral stability in older adults. Specifically, older adults demonstrated lower plantarflexor moments, a higher support moment in late stance and increased hip abductor moments. The literature suggests that older adults often perceive difficulty due to concern about falling, which is intensified during descent [2]. Adaptations that enhance balance in both the sagittal and frontal planes of movement are thus not surprising, even in healthy older adults. The
consistency of the moment patterns within and across subjects further suggests that these alterations are a strategy characteristic of older adults.

Previous literature has described a redistribution of lower limb moments during functional tasks [8,12,13]. Reeves et al. [8,13] reported that compared to young adults, healthy older adults adopted alternate kinetic strategies at the ankle and knee as a means of recalibrating muscle workloads to remain within their comfortable limits while still meeting the demands of the task. The results of the present study support this notion, as the older adults reduced their reliance on the ankle musculature by redistributing the work to more proximal muscle groups. The findings add to the literature by highlighting the age-related adaptations which act to promote stability. In the frontal plane, the sustained hip abductor moment throughout stance evident in the older group would effectively restrict lateral trunk mobility. Excessive motion in the frontal plane can cause significant shifts in the location of the body’s centre of mass relative to the small base of support [14] requiring adequate control and muscle strength to maintain balance; augmented hip abductor output would achieve this. The importance of the hip abductors to safely negotiate stairs [15], raises concern for individuals with physical impairments and may contribute to a reduction in the ability to safely ascend and descend stairs. Indeed, during level walking, weakness of the hip abductors results in frontal plane instability, leading to an increased risk of falls [16].

It was also important to determine the impact of the age-related modifications in moments and speed on the inter-segmental energetics to address whether they were associated with any compromise in efficiency. Study 2 (Chapter 4) evaluated the active energy transfer across segments and the degree to which it compensated muscle energy by conducting a secondary analysis of the biomechanical data from Study 1. Studies reporting active power
exchanges and compensations related to inter-segmental transfer during movement are few and specific to level ground walking [17-21]. Stair ascent is more demanding as it requires vertical progression against gravity as well as forward, horizontal progression [5,15]. Indirect comparison to data reported for walking [18], suggests that stair climbing is less energy efficient. In the present study, the amount of energy expended in the absence of transfer was approximately 4 to 10 times higher at the knee and ankle and the mechanical energy compensations were nearly twice the magnitude of level walking [18]. The findings suggest, however, that during ascent, older adults achieve similar efficiencies as young adults by slowing their cadence.

During descent, age-related differences in mechanical energy expenditures and related variances in mechanical energy compensation coefficients reflect a loss in mechanical efficiency since the differences between young and older subjects were preserved after controlling for cadence. Specifically, older adults demonstrated a tendency toward higher energy expenditures attributable to the eccentric knee extensor activity paired with reduced compensations consistent with the exaggerated knee extensor moments reported in the previous study (Study 1, Chapter 3). The impact was likely the provision of enhanced extensor support and balance. The results also highlight a functional role for concentric energy expenditures during descent. In late stance, older adults produced higher energy at the ankle and lower energy at the hip (concentrically) than young adults. At the ankle, the work done was compensated to a greater degree in young adults via inter-segmental transfer suggesting more energy expended by the muscles in the older group to assist in advancing the limb into swing [22]. Older adults limited the distally directed energy at the hip, which may be a useful strategy adopted to enhance dynamic stability by reducing destabilization of the trunk and upper body.

During descent, simply slowing down the cadence did not afford the older adults similar
energy expenditures as young adults as was the case with ascent. It follows that other factors play a role; perhaps physiological changes such as declines in strength [23] and dynamic balance [24] may have limited the mechanical efficiency of stair descent in this group of healthy older adults. The compromised ability of older adults to exploit energy transfer as a means of compensating active muscle activity may have implications for the energy cost of stair negotiation, particularly if impairment is superimposed on aging. The information gained from Studies 1 & 2 contribute to the understanding of the control of movement in young and older adults during stair negotiation and provides a normative benchmark of age-related alterations in movement patterns against which the superimposition of physical impairment can be evaluated.

7.1.2 Biomechanical and physical requirements of stair negotiation in persons with chronic stroke

Kinematic and kinetic profiles are well-described in stroke survivors during level ground walking [25-31], however, the mechanics of stair negotiation are unknown. Considering the importance of stair ambulation in terms of mobility independence, knowledge of the lower limb kinematics and kinetics is essential to understanding stroke related compensations, which can guide intervention strategies aimed at improving stair mobility and community accessibility. Following the identification of normal, age-related changes in movement patterns during stair negotiation, the second aspect of this thesis was to investigate biomechanical alterations in stair negotiation in persons with stroke and to explore the physical requirements of the task in relation to standard measures of peak neuromuscular and aerobic capacities.

The third study (Chapter 5) provided a detailed biomechanical description of stair negotiation in people with chronic stroke in comparison to similarly aged healthy control subjects. Although lower cadences were observed in the stroke group, several significant

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differences between groups highlight deficiencies in those with stroke that cannot be attributed to speed alone. Specifically, during ascent stroke survivors show reduced moments on the affected side. In addition, the stroke group generally modified their movement patterns to provide greater extension on the less affected side, which has the net effect of raising the vertical position of the centre of mass and facilitating clearance of the intermediate step by the affected leg [26]. Consistent with this approach, the stroke group also generated a net extensor support moment on the less affected side comparable to that of controls as a strategy to enhance balance as the affected limb is off-loaded [32]. Handrail use clearly afforded the stroke group greater balance and support as the reliance on the hip abductors was lessened and the magnitude of the net support moment was reduced compared to no handrail use. Considering that those with stroke demonstrate deficits in balance and mobility [30,33,34], it is not surprising that handrail use influenced joint kinetics. Certainly in the case of two individuals with stroke, the absence of handrails precluded them from performing the task.

During descent, the ROM required at all lower limb joints is reportedly larger during stance than during swing [6,13]; consequently limited or impaired joint mobility is more likely to alter the pattern of movement during the stance phase of stair descent. Indeed, as a likely strategy to compensate for reduced dorsiflexion on the affected side, the stroke group exhibited exaggerated hip flexion, which acts to adequately shorten the limb and lower the swing limb onto the next step (as previously shown in the work of others [35]), although possibly at the expense of dynamic balance. Markedly lower extensor moments were also evident on the affected side. The slower cadence may account for the alterations from the norm. However, co-contraction of the agonist and antagonist muscles would also promote stiffness of the joint, effectively reducing the net moment, but increasing joint stability [36]. If this is the case, the energy cost associated with the
adaptation could be significant. Finally, similar to ascent, the less affected hip abductors provided lateral stability through stance to an extent not evident in healthy controls and in conjunction with handrail use, resulted in less reliance on the paretic abductors through stance.

This study highlighted modifications in mobility and motor output among stroke survivors that were not attributable to aging. It is clear that deficits in joint mobility and lower limb extensor and abductor moment production affect stair climbing ability. The slower cadence is likely a compensation for stroke related impairments [30,34] but does not account fully for smaller ROM and lower moment magnitudes. This raises the question of whether gains in strength would translate into increases in cadence, although the cost in terms of metabolic demand may be prohibitive.

The identification of stroke-related biomechanical adaptations during stair negotiation provides direction to rehabilitation specialists about where they might target intervention, but it provides little insight about the challenge of the task in terms of relative physical costs. Study 4 (Chapter 6) evaluated the relative strength requirement (strength cost) and aerobic cost of ascending and descending stairs in stroke and healthy adults. Muscle weakness is among the most commonly reported deficit following stroke, with the greatest weakness occurring in the more distal muscles contralateral to the lesion, but also to a lesser extent on the ipsilateral side [37-43]. Even though the stroke group exhibited a slower cadence of stair ascent and descent compared to healthy adults and lower net plantarflexor and knee extensor moments, the relative strength cost (peak moment during ascent (descent) as a percentage of maximum moment generated isokinetically) was higher in stroke than healthy adults. Of note, was that the efficiency associated with the primarily negative work of stair descent was evident from the lower relative strength requirements compared to ascent in both groups. Also of interest was that the relative costs were
comparable for both the Aff and LAff sides despite the difference in maximum strength between the limbs.

In level ground walking, Milot and colleagues [42] demonstrated that reduced self-selected speed resulted in comparable muscular costs or levels of “effort” between the sides in stroke survivors, hypothesizing that the sense of effort is involved in the regulation of motor output during gait. The present study suggests the same could apply to stair negotiation. From the metabolic data, the oxygen demands associated with negotiating stairs were higher (though not significantly so) in stroke. In level ground walking, a U-shaped relationship exists whereby deviation from the preferred speed results in greater oxygen consumption [25,44,45]. Since both older adults and stroke survivors self-selected their speed of ascent and descent, it is reasonable to expect that the absolute oxygen requirements were comparable. However, in relative terms, the aerobic cost was markedly elevated in stroke compared to healthy adults reflecting the greater physiologic challenge in stroke.

The higher than normal strength and aerobic costs of stair negotiation underscores the high demands of mobility in stroke. On the basis of the work presented in this dissertation, the higher challenge of stair negotiation in stroke cannot be explained by age-related alterations, but rather biomechanical compensations and stroke-related limitation in strength and cardiovascular health. Recognizing the importance of stair mobility in maintaining independence and community accessibility [46], the demands of the task can be very restrictive. Improving the functional capacity in stroke survivors (including lower limb strength and cardiorespiratory fitness) may reduce the physiologic burden of managing stairs although the costs may still differ from age-matched controls if the mechanics of walking up and down stairs remain different.
7.2 Limitations

This thesis provides insight into the demands of stair negotiation in young and older adults and also persons with stroke, however there are several limitations of the work completed.

Firstly, the sample of healthy individuals may not be generalizable to the entire population of young and older adults. The young adults were sampled from the university community and the older group represented a relatively healthy, active sample of older adults > 55 years of age (Chapter 3, 4), as most indicated they participated in moderate physical activity approximately 3-4 or > 4 times per week. As a result, biomechanical alterations were perhaps not as marked as they might have been in less active adults or if the age criterion was > 65 years old. The choice of including adults > 55 years old was deliberate and necessary as the group served as a comparison for stroke. Further, it has been previously reported that declines in strength up to 10% per decade occurs beyond the age of 50 years as well as other indicators of physical function [23,47]. Therefore, it was important to determine whether declines in stair performance were evident in a group believed to be undergoing physical decline. The data presented in Chapters 3 & 4 provide a normative benchmark against which movement patterns produced by similarly aged older individuals with physical impairments or disability can be compared.

The sample of stroke survivors included in Chapters 5 & 6 was limited to those individuals with mild to moderate mobility impairment (as indicated by the scores on the CBM) as it was important that they could complete the physical task required. The findings therefore are not generalizable to the wider stroke population, particularly those with more severe mobility limitations. It is clear, however, that this group presented with significant alterations in movement patterns and movement control from their age-matched healthy counterparts indicating the impact of even mild/moderate impairment on mobility.
Another limitation of this work relates to the type of analysis utilized. The calculation of net joint moments and powers was completed using inverse dynamics analysis; a commonly used analytical approach to quantify normal and abnormal gait patterns [5-7,9,11,13,15,26,28,31,48]. This type of analysis, however, does not partition individual muscle contributions, but rather is a reflection of the net effect attributable to the activity of all muscles that span the joint. While reference to particular muscle groups (ie. ankle plantarflexors) is common, the net joint moments do not identify the role of individual muscles in coordinating the body segments and producing the emergent movement. This may be particularly important in older adults and stroke survivors increase antagonist muscle co-activation during level ground walking to assist with lower limb stability [36,49-51] and likely also during stair negotiation. Despite this, the results yielded important and novel information about adaptive movement patterns and compensatory strategies related to aging and impairment that is relevant in developing approaches to enhance mobility.

The current work examined the stance phase only of stair ascent and descent. Design and space limitations precluded the capture of both the swing and stance phases, therefore given the focus on examining the biomechanical (muscle output and load redistribution) and physical demands of the work associated with stair negotiation it was most relevant to evaluate the stance phase. During swing, limitations in ROM are of primary importance; for example, reductions in ankle dorsiflexion or limited hip flexion can increase the risk of a fall due to failure to clear the intermediate step [49]. This is an important consideration in aging and aging with stroke and should be addressed in future work.

Finally, our experimental set-up required that the handrail be placed on the side opposite the stance limb contacting the forceplate. As such, persons with stroke were not always able to use their preferred hand, which was typically the LAff side. Consequently, the lower limb
moments generated on the Aff side would be more likely to be reduced than those on the LAff side during handrail use due to the unequal capability of the arms to bear some of the load. This may explain why side-to-side kinetic differences were more prevalent during handrail use in stroke (Chapter 5). The lack of handrail instrumentation precluded measurement of the forces applied through the hand. Such information would be useful to facilitate interpretation and confirm that any lowering of lower limb moments was directly attributable to upper limb compensation. It is likely that in stroke survivors with greater degrees of impairment, the redistribution of loads from lower to upper limbs may be more significant and the incorporation of an instrumented handrail should be considered for future studies.

7.3 Future directions

This dissertation comprises original work that explored the biomechanical demands and physical requirements of stair negotiation related to both aging and stroke impairment and presented novel results that form the foundation for future studies.

Our experimental set-up precluded measurement of the loads through the handrail as well as analysis of the swing phase of gait. Nadeau et al. [15] hypothesized that restrictions in range of motion would result in reductions in functional performance for climbing stairs. In our sample of stroke subjects, ankle ROM was not significantly reduced on the affected side compared to controls (Chapter 5). This finding is somewhat surprising given the restricted range typically reported in stroke subjects, particularly of the affected ankle [26,27,30]. However, previous research has reported the most significant angular excursions at the ankle and knee are seen in the swing phase of ascent in order to clear the intermediate step [15] whereas our data reports values for the stance phase of gait only. Future investigations of the entire gait cycle including the swing phase are warranted and necessary to fully understand the risk of falls during stair ambulation in
this population where in the cycle fall risk is particularly high. Ideally, future work will also include a broader range of age and impairment to expand our understanding of the impact of mobility limitations.

Another important area of focus relates to understanding the individual muscle contributions during stair ascent and descent. Forward dynamics analyses derived from musculoskeletal models is an intriguing analytical approach that provides insight into muscle coordination and contributions during physical tasks such as stair negotiation. Muscle-actuated forward dynamics analysis pairs experimentally measured data with simulated data. By using an optimization algorithm, the analysis fine-tunes the estimated muscle excitation patterns for individual muscles to produce a well-coordinated walking pattern that emulates the experimental data, i.e. the simulation-generated kinematic and kinetic patterns closely match the actual patterns measured [52,53]. As such, the contributions of individual muscles to specific walking sub-tasks, such as forward propulsion, can be determined. In overground walking, forward dynamics-based simulations have quantified muscle powers during walking subtasks in healthy and hemiparetic individuals providing insight into segmental power exchanges performed by individual muscles in healthy adults [54] and muscle function in the post-stroke population [55]. Application of this approach to stair negotiation could provide interesting information partitioning segmental power into individual sources (e.g. muscle forces) that would enable rehabilitation specialists to target interventions to particular muscles and specific actions.

The findings relating to the strength and aerobic costs of stair ascent and descent, suggest that there could be benefit to increasing either or both of these factors (strength, aerobic capacity) to improve stair mobility. It has been previously shown that strength and aerobic capacity explain approximately 50% of the variance associated with stair performance [56], but it remains
unknown whether improvements can effect change in performance. Development of function-oriented interventions is warranted based on the findings of this work to determine their effectiveness in improving the mechanics and reducing the energy costs of stair ambulation.

Ultimately, the global goal of rehabilitation is to maximize function. In terms of stair negotiation this translates into safer and more efficient movement. The work from this dissertation provides a benchmark of the normal biomechanical profiles of stair negotiation in young and older adults characterizing age-related alterations. The superimposition of impairment on aging and the impact on stair negotiation was described as well as the relative strength and aerobic costs compared to healthy, age-matched controls. The knowledge gained has laid a foundation upon which targeted and task oriented rehabilitation programs may be developed, or more in-depth analysis of mobility function in disabled populations can be explored. As a whole this work has made a significant contribution to our understanding of mobility and exposed age-related and impairment-related compensations that could impact independence and safety.
7.4 References


Appendix A: Ethics

QUEEN’S UNIVERSITY HEALTH SCIENCES AND AFFILIATED TEACHING HOSPITALS
ANNUAL RENEWAL

Queen’s University, in accordance with the “Tri-Council Policy Statement, 1998” prepared by the Medical Research Council, Natural Sciences and Engineering Research Council of Canada and Social Sciences and Humanities Research Council of Canada requires that research projects involving human subjects be reviewed annually to determine their acceptability on ethical grounds.

A Research Ethics Board composed of:

Dr. A.F. Clark
Emeritus Professor, Department of Biochemistry, Faculty of Health Sciences, Queen’s University (Chair)

Dr. H. Abdollah
Professor, Department of Medicine, Queen’s University

Dr. C. Cline
Assistant Professor, Department of Medicine
Director, Office of Biethics, Queen’s University
Clinical Ethicists, Kingston General Hospital

Rev. T. Deline
Community Member

Dr. M. Evans
Community Member

Dr. S. Irving
Psychologist, Providence Care, St. Mary’s of the Lake Hospital Site

Prof. L. Keeping-Burke
Assistant Professor, School of Nursing, Queen’s University

Mrs. J. Kotecha
Research & Programs Manager, Centre for Studies in Primary Care, Department of Family Medicine, Queen’s University

Dr. J. Low
Emeritus Professor, Department of Obstetrics and Gynaecology, Queen’s University and Kingston General Hospital

Dr. W. Racz
Emeritus Professor, Department of Pharmacology & Toxicology, Queen’s University

Dr. R. Simchison
Assistant Professor, Department of Anaesthesiology, Queen’s University

Dr. A.N. Singh
WHO Professor in Psychosomatic Medicine and Psychopharmacology
Professor of Psychiatry and Pharmacology
Chair and Head, Division of Psychopharmacology, Queen’s University
Director & Chief of Psychiatry, Academic Unit, Quinte Health Care, Belleville General Hospital

Dr. E. Tsai
Associate Professor, Department of Paediatrics and Office of Biethics, Queen’s University

Rev. J. Warren
Community Member

Ms. K. Weisbaum
L.L.B. and Adjunct Instructor, Department of Family Medicine (Bioethics)

Dr. S. Wood
Director, Office of Research Services (Ex Officio)

has reviewed the request for renewal of Research Ethics Board approval for the project “Determining Physical Requirements for Every Day Activities” as proposed by Dr. Brenda Brouwer and Ms. Alison Novak of the School of Rehabilitation Therapy, at Queen’s University. The approval is renewed for one year, effective May 5, 2009. If there are any further amendments or changes to the protocol affecting the subjects in this study, it is the responsibility of the principal investigator to notify the Research Ethics Board. Any unexpected serious adverse event occurring locally must be reported within 2 working days or earlier if required by the study sponsor. All other adverse events must be reported within 15 days after becoming aware of the information.

Chair, Research Ethics Board

Date: April 24, 2009

ORIGINAL TO INVESTIGATOR - COPY TO DEPARTMENT HEAD - COPY TO HOSPITAL(S) - FILE COPY
Renewal 1 [X] Renewal 2 [ ] Extension [ ]

REB# REH-435-08
Appendix B: Letter of Information & Consent Form

CONSENT FORM

TITLE OF PROJECT: Exploring the physical demands associated with going up and down stairs.

INVESTIGATORS: Dr. Brenda Brouwer, School of Rehabilitation Therapy & Human Mobility Research Centre
Alison Novak (Ph.D. candidate), School of Rehabilitation Therapy & Human Mobility Research Centre

BACKGROUND INFORMATION
Access to buildings and homes often requires that individuals are able to manage stairs. It is known that the physical demands of stair climbing and descent are higher than those required to walk on level ground, but little is known about how much higher. This is important because it can help establish targets or goals for strength training and physical conditioning that if achieved can impact on individuals’ independence.

You are being invited to participate in a research study designed to determine the strength and metabolic (energy) requirements associated with going up and down stairs and how this compares to your maximum strength and metabolic capacity. One of the investigators will read through this consent form with you, describe the procedures in detail, and answer any questions you may have.

DETAILS OF THE STUDY
The purpose of this study is to determine the physical demands of stair climbing and how these demands compare to your maximum performance capacity. We are interested in exploring this in three groups of people: young, healthy people (aged 20-35 years); healthy seniors (aged 65+ years) and in people who have had a previous stroke affecting one side of the body, but are otherwise healthy. You will be considered for the study if you fall into one of these groups and are able to manage stairs.

Description of visits and tests to be performed as part of the study
The testing will be completed over two laboratory visits, the first in the Human Mobility Research Centre and the second in the School of Rehabilitation Therapy each lasting about 1.5 hours. During the first visit light emitting diodes (little disks) will be placed on your feet, lower legs, thighs and trunk using straps so we can measure how your legs and body move as you go up and down a set of 4 stairs. You will also be wearing a heart rate monitor around your chest. You will be asked to go up and down the stairs several times, but can rest as needed as we do not want you to become fatigued. After, you will be seated and the disks will be removed. At this time you will be asked to wear a mask that will cover your mouth and nose. This allows us to monitor your breathing and measure how much oxygen you use. We will measure how much oxygen
you use and your heart rate while you pedal a recumbent bicycle for 6 minutes. After every two minutes, you will find the pedaling becomes a little more difficult. At the end of six minutes, you will be instructed to slow your pedaling gradually and come to rest. When your heart rate and breathing return to rest values then the testing is complete.

You will be asked to return for your second visit two to five days later. During this visit you will be seated comfortably so that we can measure the strength of the flexor and extensor muscles of your ankles, knees and hips (you will lie down when we test hip strength) using a dynamometer. You will be asked to push and pull as hard and as fast as you can against a machine with your foot, leg or thigh. This machine measures how much force your muscles produce. After a rest period, you will be asked to wear the heart rate monitor and the mask over your mouth and nose as you did on the previous test day. This time we will measure how much oxygen you use and your heart rate while going up and down a flight of stairs. You will then sit down and rest. When your heart rate and breathing return to their resting values, the test is complete.

**Risks/Side-Effects:**

We have experience conducting all these tests in healthy people and in people who have had strokes. In the presence of cardiovascular disease, physical exertion can pose a risk. To minimize this risk, you will stretch and perform practice trials at low exertion to warm up prior muscle strength testing and stretch again afterwards. Other testing (stair climbing and cycling) do not require maximal exertion and can be terminated at your request. The investigator will carefully supervise you at all times. You may feel somewhat fatigued after testing depending on how active you normally are and your muscles may be a bit sore after strength testing. If this is the case, the soreness should disappear within a couple of days. The discomfort should not be excessive and is the result of physical activity.

If you have any concerns, please ask the investigators immediately or contact Alison Novak or Dr. Brouwer at a later time (contact information is provided below).

**Benefits**

You will not experience any direct benefit from participating in this study. However, the results will contribute to our understanding of the physical demands associated with performing an everyday physical activity. This will assist in developing an exercise/physical conditioning program aimed at improving strength and metabolic capacity to enable people to meet the demands of stair climbing.

**Exclusions**

The testing you will undergo requires physical exertion. You will not be considered for this study if you have any of the following: a pre-existing mobility restriction, unstable angina, peripheral
vascular disease or congestive heart failure. If you are not sure if you have any of these things please consult your doctor first.

Confidentiality
All information obtained during the course of this study is strictly confidential and your anonymity will be protected at all times. You will be identified by a code reflecting the project identifier, the group you are in (young, senior or stroke) and your subject number (each subject will be numbered in sequence when they enter the study). Data summary sheets and computer datafiles will identify your data by the code only. The list matching codes to the peoples’ names will be kept in the locked office of the principal investigator. You will not be identified in any publication, presentation or report.

Voluntary nature of study/Freedom to withdraw or participate
Your participation in this study is voluntary. You may withdraw from this study at any time for any reason and your withdrawal will not affect any future care you may require.

Withdrawal of subject by principal investigator or co-investigators
A study investigator may decide to withdraw you from this study if you exhibit signs of cardiovascular stress (dizziness, sudden change in heart rate) or an inability to follow the protocol.

Liability
In the event that you are injured as a result of the study procedures, health care will be provided to you until resolution of the problem.

By signing this consent form, you do not waive your legal rights nor release the investigators from their legal and professional responsibilities.

Payment
We appreciate your involvement in this study. After completion of both test sessions you will be given $25 to compensate you for travel costs. You will be asked to sign a receipt for this amount for our own internal accounting purposes. We can provide you with free parking at the rear of the Louise D. Acton building if needed.

SUBJECT STATEMENT AND SIGNATURE SECTION:
I have read and understand the consent form for this study. I have had the purposes, procedures and technical language of this study explained to me. I have been given sufficient time to consider the above information and to seek advice if I chose to do so. I have had the opportunity to ask questions which have been answered to my satisfaction. I am voluntarily signing this form. I will retain a copy of this consent form for my own records.
If at any time I have further questions, problems or adverse events, I can contact:

Dr. Brenda Brouwer, Principal Investigator at 613-533-6000 ext. 77557

OR

Alison Novak, Student Investigator at 613-533-6000 ext. 77850

OR

Dr. Elsie Culham, Director of the School of Rehabilitation Therapy at 613-533-6727

If I have questions regarding my rights as a research subject I can contact

Dr. Albert Clark, Chair, Research Ethics Board at 613-533-6081

By signing this consent form, I am indicating that I agree to participate in this study.

_______________________  _____________________
Signature of Subject     Date

_______________________  _____________________
Signature of Witness    Date

STATEMENT OF INVESTIGATOR:

I, or one of my colleagues, have carefully explained to the subject the nature of the above research study. I certify that, to the best of my knowledge, the subject understands clearly the nature of the study and demands, benefits, and risks involved to participants in this study.

_______________________
Signature of Principal Investigator Date
Appendix C: Community Balance and Mobility Scale*

Test Set-Up:

1. Unilateral Stance
**Instructions to patient:** Stand on your right/left leg and hold for as long as you can up to 45 seconds. Look straight ahead.
**Instructions to therapist:** Begin timing as soon as the patient’s foot leaves the ground. Do not allow for the patient to brace the elevated leg against the supporting leg. Stop timing if stance foot moves from starting position or opposite foot touches ground.

<table>
<thead>
<tr>
<th>Right limb</th>
<th>Left limb</th>
</tr>
</thead>
<tbody>
<tr>
<td>□ 5 45 seconds, steady and coordinated</td>
<td>□ 5 45 seconds, steady and coordinated</td>
</tr>
<tr>
<td>□ 4 ≥ 20 seconds</td>
<td>□ 4 ≥ 20 seconds</td>
</tr>
<tr>
<td>□ 3 10 to 20 seconds</td>
<td>□ 3 10 to 20 seconds</td>
</tr>
<tr>
<td>□ 2 4.5 to 9.99 seconds</td>
<td>□ 2 4.5 to 9.99 seconds</td>
</tr>
<tr>
<td>□ 1 2 to 4.49 seconds</td>
<td>□ 1 2 to 4.49 seconds</td>
</tr>
<tr>
<td>□ 0 Unable to sustain</td>
<td>□ 0 Unable to sustain</td>
</tr>
</tbody>
</table>

2. Tandem Walking
**Instructions to patient:** Walk forward on the line, heel touching toes. Keep your feet pointing straight ahead. Look ahead down the track, not at your feet. I will tell you when to stop.
**Instructions to therapist:** Position the patient with one foot positioned on the 8m track. If able, allow the patient to take a maximum of 7 steps for which the heel is on the line and the heel-toe distance is ≤ 8cm (3 inches).
0 Unable to complete 1 step on the line independently, ie. requires assistance, upper extremity support, or takes a protective step

1 Able to complete 1 step independently, acceptable to toe out

2 Able to complete 2 to 3 consecutive step on the line, acceptable to toe out

3 Able to complete >3 consecutive steps, acceptable to toe out

4 Able to complete 3 consecutive steps, in good alignment (heel-toe contact, feet straight on the line, no toeing out), but demonstrates excessive use of equilibrium reactions

5 Able to complete 7 consecutive steps, in good alignment (heel-toe contact, feet straight on the line, no toeing out), and in a steady and coordinated manner. Excessive use of equilibrium reactions or looking at feet is not acceptable.

3. 180 degree Tandem Pivot
Starting position: Tandem stance on bare spot in track – aligned heel to toe, no toeing out, arms at sides, head in neutral position and eyes forward. Patient allowed to choose either foot in front and may use assistance or upper extremity support to achieve, but not sustain, tandem stance
Instructions to patient: Lifting your heels just a little, pivot all the way around to face the opposite direction without stopping and maintain your balance in this position
Instructions to therapist: When right foot is in front in tandem position, patient to turn towards left. When left foot is in front in tandem position, patient to turn towards right. Therapist may assist patient to assume starting position. Test is over when patient puts heels down or steps out of position

0 Unable to sustain tandem stance independently ie. requires assistance or upper extremity support

1 Sustains tandem stance but unable to unweight heels and/or initiate pivot

2 Initiates pivot but unable to complete 180 degree turn

3 Completes 180 degree turn but discontinuous pivot (ie. pauses on toes)

4 Completes 180 degree turn in a continuous motion but unable to sustain reversed position (Not acceptable: heel-toe distance >8 cm)

5 Completes 180 degree turn in a continuous motion and sustains reversed position (acceptable to have feet slightly angled out in reversed position). Not acceptable: heel-toe distance >8 cm; excessive use of equilibrium reactions.
4. **Lateral foot scooting**  
**Starting position:** Patient to stand on the line beside the bare spot in the track in unilateral stance on right/left foot, arms at sides. Foot is perpendicular to the track.  
**Instructions to patient:** Stand on your right/left leg and move sideways by alternatingly pivoting on your heel and toe. Keep pivoting straight across until you touch the line and maintain your balance in this position.  
**Instructions to therapist:** The patient moves laterally along the length of the bare spot (40 cm). For the grading, one lateral pivot is defined as either pivoting on heel, moving toes laterally OR pivoting on toes, moving heel laterally.

<table>
<thead>
<tr>
<th>Right limb</th>
<th>Left limb</th>
</tr>
</thead>
<tbody>
<tr>
<td>□ 0 Unable to sustain unilateral stance independently</td>
<td>□ 0 Unable to sustain unilateral stance independently</td>
</tr>
<tr>
<td>□ 1 Able to perform 1 lateral pivot in any fashion</td>
<td>□ 1 Able to perform 1 lateral pivot in any fashion</td>
</tr>
<tr>
<td>□ 2 Able to perform 2 lateral pivots in any fashion</td>
<td>□ 2 Able to perform 2 lateral pivots in any fashion</td>
</tr>
<tr>
<td>□ 3 Able to perform ≥3 pivots but &lt; 40 cm</td>
<td>□ 3 Able to perform ≥3 pivots but &lt; 40 cm</td>
</tr>
<tr>
<td>□ 4 Able to complete 40 cm in any fashion. Acceptable to be unable to control final position</td>
<td>□ 4 Able to complete 40 cm in any fashion. Acceptable to be unable to control final position</td>
</tr>
<tr>
<td>□ 5 Able to complete 40 cm continuous rhythmical motion demonstrating a controlled stop briefly maintaining unilateral stance. Not acceptable to pause while pivoting to regain balance, veer from a straight line, excessive use of equilibrium reactions, or excessive trunk rotation while pivoting</td>
<td>□ 5 Able to complete 40 cm continuous rhythmical motion demonstrating a controlled stop briefly maintaining unilateral stance. Not acceptable to pause while pivoting to regain balance, veer from a straight line, excessive use of equilibrium reactions, or excessive trunk rotation while pivoting</td>
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5. **Hopping forward**  
**Starting position:** Unilateral stance on right/left with entire foot on the track. Heel placed on inside edge of starting line.  
**Instructions to patient:** Stand on your right/left foot. Hop twice straight along this line to pass the 1 m mark with your heel. Maintain your balance on your right/left leg at the finish.
Instructions to therapist: It is recommended that the therapist assess safety prior to commencing task by having the patient hop in one spot. Patient is successful in completing 1 m when the heel of the foot is touching or beyond the 1m line.

Right limb

☐ 0  Unable

☐ 1  1 to 2 hops, uncontrolled

☐ 2  2 hops controlled but unable to complete 1 meter

☐ 3  1 m in 2 hops but unable to sustain landing

☐ 4  1 m in 2 hops but difficulty controlling landing (hops or pivots)

☐ 5  1 m in 2 hops, coordinated with stable landing

Left limb

☐ 0  Unable

☐ 1  1 to 2 hops, uncontrolled

☐ 2  2 hops controlled but unable to complete 1 meter

☐ 3  1 m in 2 hops but unable to sustain landing

☐ 4  1 m in 2 hops but difficulty controlling landing (hops or pivots)

☐ 5  1 m in 2 hops, coordinated with stable landing

6. Crouch and walk

Starting position: Bean bag is placed to right or left side of 2m mark on the track according to which hand the patient will use to pick it up.

Instructions to patient: Walk forward and, without stopping, bend to pick up the bean bag and then continue walking down the line.

Instructions to therapist: This task is performed using only half of the track. Start timing when the patient’s foot leaves the ground. Stop timing when both feet cross the 4m line.

☐ 0  Unable to crouch (descend) to pick up bean bag independently

☐ 1  Able to descend but unable to maintain crouch to pick up bean bag or rise to stand independently

☐ 2  Descends and rises but hesitates, unable to maintain forward momentum

☐ 3  Crouches and walks in continuous motion, time ≤ 8 seconds and demonstrates protective step at any time

☐ 4  Crouches and walks in continuous motion, time ≤ 8 seconds and/or uses excessive equilibrium reaction to maintain balance. Not acceptable to veer off course.

☐ 5  Crouches and walks in continuous motion, time ≤ 4 seconds. Not acceptable to veer off course or to use excessive equilibrium reactions.
7. Lateral dodging

**Starting position:** Starting at the 2m mark with feet perpendicular to the track. The toes of both feet should cover the track.

**Instructions to patient:** Move sideways along the line by repeatedly crossing one foot in front of and over the other. Place part of your foot on the line with every step. Reverse direction whenever I call “Change”. Do this as fast as you can, yet at a speed that you feel safe.

**Instructions to therapist:** Patient moves laterally back and forth along the line, between the 2m and 4m marks by repeatedly crossing one foot over and in front of the other. It is acceptable for the patient to look at the line to monitor foot placement. Begin timing as soon as the patient’s foot leaves the ground. To cue the patient to change direction, call out “Change” when one foot passes the 2 and 4 m marks. The patient should believe direction changes are random. One cross-over = crossing one leg over to land beside the other and returning the back leg to an uncrossed position. One cycle = patient completes cross-overs for a 2m distance and return. The test requires that the patient perform 2 of these cycles (a total of 8m).

- □ 0 Unable to perform 1 cross-over in both directions without loss of balance or use of support
- □ 1 1 cross-over in both directions without use of support, but unable to contact the line with part of the foot
- □ 2 1 or more cycles to and from the 2m mark, but unable to contact line with every step
- □ 3 2 cycles in any fashion (to the 2m line and back twice) and one part of the foot contacts line during every step
- □ 4 2 cycles in any fashion as described in the response above (3) in 12 to 15 seconds
- □ 5 2 cycles in < 12 seconds in a continuous, rhythmic fashion with coordinated direction changes immediately after verbal cue.

8. Walking and looking

**Instructions to patient:** Walk at your usual pace to the end of the line. I will tell you when to look at the circle. Keep looking at the circle while you walk past it. I will then tell you when to look straight ahead again. Try not to veer off course while you walk.

**Instructions to therapist:** Start timing when the patient’s foot leaves the ground. Stop timing when both feet cross the 8m finish line. At the 2m mark, ask the patient to “Look at the circle”. Cue the patient to “Keep looking at the circle” as they look back over their shoulder until they reach the 6m mark. At the 6m mark, ask the patient to “Look straight ahead and continue walking until the end of the line”. Stand beside the target so that you can assess the patient’s ability to maintain fixation. It may be necessary to have another person present to walk along side the patient to ensure safety. It is acceptable to continue to remind the patient of where they should be looking at each segment. To score in the opposite direction, repeat the task starting from the opposite end of the line.
Target to the **Right**

- □ 0  Unable to walk and look eg. stops
- □ 1  Performs but loses visual fixation at or before 4m mark
- □ 2  Performs but loses visual fixation after 4m mark
- □ 3  Performs and maintains visual fixation between 2 to 6m mark but protective step
- □ 4  Performs and maintains visual fixation between 2 to 6m mark but veers
- □ 5  Performs, straight path, steady and coordinated ≤ 7 seconds

Target to the **Left**

- □ 0  Unable to walk and look eg. stops
- □ 1  Performs but loses visual fixation at or before 4m mark
- □ 2  Performs but loses visual fixation after 4m mark
- □ 3  Performs and maintains visual fixation between 2 to 6m mark but protective step
- □ 4  Performs and maintains visual fixation between 2 to 6m mark but veers
- □ 5  Performs, straight path, steady and coordinated ≤ 7 seconds

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9. **Running with controlled stop**

**Instructions to patient:** Run as fast as you can to the end of the track. Stop abruptly with both feet on the finish line and hold this position.

**Instructions to therapist:** Begin timing when initial foot leaves the ground. Stop timing when both feet reach the finish line. It does matter whether the feet land consecutively or simultaneously on the finish line.

- □ 0  Unable to run (with both feet off the ground for a brief instant), demonstrates fast walking or leaping from foot to foot
- □ 1  Runs in any fashion, time > 5 seconds
- □ 2  Runs in any fashion, time > 3 seconds but ≤ 5 seconds, unable to control stop (uses protective step or excessive equilibrium reactions)
- □ 3  Runs, time > seconds, but ≤ 5 seconds, with controlled stop, both feet on the line. Not acceptable to use excessive equilibrium reactions
- □ 4  Runs in any fashion, time ≤ 3 seconds, unable to control stop with both feet on the line (uses protective step(s) or excessive equilibrium reactions)
- □ 5  Runs in a coordinated, rhythmical manner and performs a controlled stop with both feet on the line, ≤ 3 seconds. Not acceptable to use excessive equilibrium reactions.
10. Forward to backward walking

**Instructions to patient:** Walk forward to the halfway mark, turn around and continue to walk backward until I say, “Stop”. Try not to veer off course. Walk as quickly as you can, yet at a speed that you feel safe.

**Instructions to therapist:** Start timing when the patient’s foot leaves the ground. Stop timing when both feet cross the 8m finish line. The patient is to turn at the 4m mark. It is acceptable for the patient to turn in any direction. Count the number of steps required to turn 180 degrees. Note: the first step in the turn is angled away from the forward trajectory. The last step in the turn completes the 180 degree turn and is oriented towards the starting line, initiating backwards walking. It is also acceptable to pivot on one foot rather than stepping around.

- □ 0 Unable to complete task (requires assistance or upper extremity support)
- □ 1 Performs but must stop to regain balance at any time during the task
- □ 2 Performs with reduced speed, time > 11 seconds and/or requires 4 or more steps to turn
- □ 3 Performs in ≤ 11 seconds and/or veers from straight path during backwards walking
- □ 4 Able to complete task in a continuous motion, ≤ 9 seconds and/or uses protective step(s) during or just after turn
- □ 5 Able to complete task in a continuous motion with brisk speed, ≤ 7 seconds, maintains straight path throughout

11. Walk, look and carry

**Starting position:** Start patient at the end of the track carrying a plastic grocery bag in each hand by the handle, with a 7.5 pound weight inside each bag.

**Instructions to patient:** Walk at your usual pace to the end of the line carrying the grocery bags. I will tell you when to look at the circle. Keep looking at it while you walk past it. I will then tell you to look straight ahead again. Try not to veer off course while you walk.

**Instructions to therapist:** Start timing when the patient’s foot leaves the ground. Stop timing when both feet cross the 8m finish line. At the 2m mark, ask the patient to “Look at the circle”. Cue the patient to “Keep looking at the circle” as they look back over their shoulder until they reach the 6m mark. At the 6m mark, ask the patient to “Look straight ahead and continue walking until the end of the line”. Stand beside the target so that you can assess the patient’s ability to maintain fixation. It may be necessary to have another person present to walk along side the patient to ensure safety. It is acceptable to continue to remind the patient of where they should be looking at each segment. To score in the opposite direction, repeat the task starting from the opposite end of the line.

**Note:** Patient to only carry one grocery bag if unable to perform bilaterally due to motor control problems of the upper extremity. Indicate that the patient was only able to carry one bag.
Target to the **Right**

- **□ 0** Unable to walk and look eg. Has to stop to look, or requires assistance or upper extremity support at any point during the test
- **□ 1** Able to continuously walk and initiate looking, but loses visual fixation on circle at or before 4m mark
- **□ 2** Able to continuously walk and look, but loses visual fixation on circle after 4m mark ie. while looking back over the shoulder
- **□ 3** Able to continuously walk and fixate upon the circle between the 2m and 6m mark, but demonstrates a protective step. Acceptable for patient to demonstrate inconsistent or reduced speed
- **□ 4** Able to continuously walk and fixate upon the circle between the 2m and 6m mark, but veers off course. Acceptable for patient to demonstrate inconsistent or reduced speed
- **□ 5** Able to continuously walk and fixate upon the circle between the 2m and 6m mark, maintains a straight path, in a steady and coordinated manner; ≤ 7 seconds. Not acceptable to walk at an inconsistent speed or reduced speed or to be looking down at feet

Target to the **Left**

- **□ 0** Unable to walk and look eg. Has to stop to look, or requires assistance or upper extremity support at any point during the test
- **□ 1** Able to continuously walk and initiate looking, but loses visual fixation on circle at or before 4m mark
- **□ 2** Able to continuously walk and look, but loses visual fixation on circle after 4m mark ie. while looking back over the shoulder
- **□ 3** Able to continuously walk and fixate upon the circle between the 2m and 6m mark, but demonstrates a protective step. Acceptable for patient to demonstrate inconsistent or reduced speed
- **□ 4** Able to continuously walk and fixate upon the circle between the 2m and 6m mark, but veers off course. Acceptable for patient to demonstrate inconsistent or reduced speed
- **□ 5** Able to continuously walk and fixate upon the circle between the 2m and 6m mark, maintains a straight path, in a steady and coordinated manner; ≤ 7 seconds. Not acceptable to walk at an inconsistent speed or reduced speed or to be looking down at feet

**12. Descending stairs**

**Starting position:** Quiet standing at top of staircase (minimum 8 steps). Depending on patient’s skill on the stairs, may begin by descending from the first or third step at the bottom of the flight.
**Instructions to patient:** Walk down the stairs. Try not to use the rail.

**Instructions to therapist:** Depending on the patient’s skill on the stairs, may use a cane as in level 1 or level 2

*Bonus: If the patient achieves a score of 4 or 5, and if deemed safe by the rating therapist, the patient is asked to repeat the task and descend the stairs while carrying a weighted basket (laundry basket with 2 pound weight in it). It is acceptable for the patient to intermittently look at the steps. Add one bonus point to the score of 4 or 5 if the patient can descend the stairs safely while carrying the basket without the need for continuous monitoring of their foot placement. If the patient is unable to hold the basket with one or both arms, they are not eligible for the bonus point.

**Instructions to patient:** Hold this basket, keeping it in front of you at waist level. Walk down the stairs and try not to look at your feet. You may look at the steps once in a while for safety

- □ 0 Unable to step down 1 step, or requires railing or assistance
- □ 1 Able to step down 1 step with/without cane
- □ 2 Able to step down 3 steps with/without cane, any pattern
- □ 3 3 steps reciprocal or full flight in step-to-pattern
- □ 4 Full flight reciprocal, awkward, uncoordinated*
- □ 5 Full flight, reciprocal, rhythmical and coordinated*
- □,+1 *BONUS for carrying basket

**13. Step ups X 1 step**

**Starting position:** In front of step at bottom of stairs

**Instructions to patient:** i) Step up and down on this step as quickly as you can until I say “Stop”. The pattern is right-left up and right-left down. Try not to look at your feet. ii) Step up and down on this step as quickly as you can until I say “Stop”. The pattern is left-right up and left-right down. Try not to look at your feet.

**Instructions to therapist:** Start timing when the patient’s foot leaves the ground. Stop timing after the completion of 5 cycles. A cycle is one complete step up and one complete step down
**Right**

- □ 0  Unable to step up, requires railing or assistance
- □ 1  Steps up, requires assistance or railing to descend
- □ 2  Steps up and down (1 cycle)
- □ 3  Completes 5 cycles, acceptable to demonstrate incoordination or inconsistent speed/rhythm
- □ 4  Complete 5 cycles in > 6 seconds but < 10 seconds, acceptable to demonstrate incoordination or inconsistent speed/rhythm
- □ 5  Completes 5 cycles in ≤ 6 seconds, rhythmical and coordinated manner

**Left**

- □ 0  Unable to step up, requires railing or assistance
- □ 1  Steps up, requires assistance or railing to descend
- □ 2  Steps up and down (1 cycle)
- □ 3  Completes 5 cycles, acceptable to demonstrate incoordination or inconsistent speed/rhythm
- □ 4  Complete 5 cycles in > 6 seconds but < 10 seconds, acceptable to demonstrate incoordination or inconsistent speed/rhythm
- □ 5  Completes 5 cycles in ≤ 6 seconds, rhythmical and coordinated manner

Appendix D: Astrand-Rhyming protocol*

1. Adjust seat height (legs nearly straight when extended - 5° bend)
2. Measure pre-exercise BP and HR with subject seated on bike
3. Pedal at 50 rpm (if using a metronome - 100x/minute)
4. Warm-up, zero resistance for 2-3 minute
5. Pedal rate is 50 rpm
6. Determine Workload

   unconditioned males - 300 or 600 kgm/min
   conditioned males - 600 or 900 kgm/min
   unconditioned females – 300 or 450 kgm/min
   conditioned females - 450 or 600 kgm/min

7. 6 minute test
8. At end of 2nd minute of pedaling take HR (BP at 1.25-1.5 min)
    want the HR to be between 125-170bpm
    if less than 125 increase resistance by 1 kp for men and 1/2 kp for women
    if greater than 170 bpm decrease resistance by 1 kp
    continue to monitor HR every minute until HR exceeds 125
    → adjust HR target range for subjects on β blockers
9. At the end of the 5th and 6th minute take HR and average the two values (make sure values are within ±6 bpm to assure a steady state HR was obtained)
10. BP at 4:30 and 5:30
11. Reduce resistance and cool-down for 4 minutes
    Continue pedaling at a work rate equivalent to the first stage of the exercise protocol or lower
12. Determine VO₂ from nomogram
13. Age-correction factor
11. Convert to relative value
