STRENGTH REQUIREMENTS AND ENERGY EFFICIENCY OF DIFFERENT STAIR-STEPPING STRATEGIES IN PERSONS WITH CHRONIC STROKE AND HEALTHY ADULTS

by

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the degree of Master’s of Science

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Kingston, Ontario, Canada
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Abstract

The majority of stroke survivors return to living in the community; however, muscle weakness and cardiovascular deconditioning can restrict mobility, limit community access and independence, particularly when challenging activities like stair negotiation are involved. A “step-by-step” (SBS) strategy (both feet per step) may be adopted in lieu of a “step-over-step” (SOS) method (one foot per step) to increase stability and off-load the paretic limb though the physical demands of the two methods are unknown. The main objective of this thesis was to investigate the strength and energy demands of the two stair-stepping strategies in chronic stroke compared to healthy adults. The first study identified the relative strength and aerobic demands of both strategies. The results showed that the stroke group produced similar peak joint moments compared to controls, despite their slower cadence suggesting that the stroke group exerts comparable ‘effort’ to move more slowly. The SBS method was associated with lower strength costs (relative to individuals’ maximum strength output) than SOS, however aerobic cost was significantly higher. The second study identified the mechanical energy expenditures (MEEs) and transfers related to both strategies. The MEEs were found to be lower when the SBS strategy was used. Though expenditures were similar between groups, the stroke group had higher expenditures associated with the work of the less affected knee extensors (lead limb) during ascent and descent and controls exhibited higher expenditures for the plantarflexors during ascent. The reduced output of the trail (affected) limb plantarflexors likely resulted in the increased workload of the knee extensors. Overall, the aerobic cost per step was higher in stroke, particularly during descent, suggesting that in addition to reducing cadence, persons with stroke may be co-contracting to increase stabilization during descent, thus increasing oxygen demands.

This thesis provides novel information on the physical demands associated with two methods of stair negotiation demonstrating that the SBS strategy might be better suited to persons with chronic stroke by minimizing the strength demands on the paretic side, but the benefit comes
at an elevated aerobic cost. This information is valuable to rehabilitation professionals engaged in retraining mobility to facilitate community reintegration.
Acknowledgements

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<th>Description</th>
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<tbody>
<tr>
<td>AFF</td>
<td>affected</td>
</tr>
<tr>
<td>CBM</td>
<td>community balance and mobility scale</td>
</tr>
<tr>
<td>COM</td>
<td>centre of mass</td>
</tr>
<tr>
<td>DF</td>
<td>dorsiflexors</td>
</tr>
<tr>
<td>DOM</td>
<td>dominant</td>
</tr>
<tr>
<td>EMG</td>
<td>electromyography</td>
</tr>
<tr>
<td>Ext</td>
<td>extensors</td>
</tr>
<tr>
<td>Flex</td>
<td>flexors</td>
</tr>
<tr>
<td>HR</td>
<td>heart rate</td>
</tr>
<tr>
<td>HR(_{\text{max}})</td>
<td>maximal heart rate</td>
</tr>
<tr>
<td>IREDs</td>
<td>infrared emitting diodes</td>
</tr>
<tr>
<td>LAFF</td>
<td>less affected</td>
</tr>
<tr>
<td>MEE</td>
<td>mechanical energy expenditure</td>
</tr>
<tr>
<td>NDOM</td>
<td>non-dominant</td>
</tr>
<tr>
<td>PF</td>
<td>plantarflexors</td>
</tr>
<tr>
<td>RER</td>
<td>respiratory exchange ratio</td>
</tr>
<tr>
<td>ROM</td>
<td>range of motion</td>
</tr>
<tr>
<td>SBS</td>
<td>step-by-step</td>
</tr>
<tr>
<td>SD</td>
<td>standard deviation</td>
</tr>
<tr>
<td>SOS</td>
<td>step-over-step</td>
</tr>
<tr>
<td>TUG</td>
<td>timed up and go</td>
</tr>
<tr>
<td>VO(_{2})</td>
<td>oxygen consumption</td>
</tr>
<tr>
<td>VO(<em>{2})(</em>{\text{max}})</td>
<td>maximal oxygen consumption</td>
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Chapter 1

General Introduction

Stair ambulation is an important determinant for community living and functional independence. More than 85% of stroke survivors return to living in the community and prioritize community participation (Stineman & Granger, 1998, Lord et al., 2004), however, mobility impairments post-stroke, reduced levels of physical activity and an inability to negotiate stairs can restrict access (Pound et al., 1998, Hamel & Cavanagh, 2004).

Compared to level-ground walking, stair negotiation is associated with greater lower limb joint range of motion in the sagittal plane (Andriacchi et al., 1980, Costigan et al., 2002, Riener et al., 2002, Nadeau et al., 2003, Olney, 2005), increased joint moments (Andriacchi et al., 1980, McFadyen & Winter, 1988, Costigan et al., 2002, Riener et al., 2002, Nadeau et al., 2003, Protopapadaki et al., 2007) and an elevated aerobic cost (Basset et al., 1997, Startzell et al., 2000 (review), Teh & Aziz, 2001). The increased physical demands of stair walking compared to overground walking could bode poorly for those with mobility limiting physical impairments.

The extent of physical deficits post-stroke including muscle weakness, spasticity, sensory loss, and gait or balance impairments generally relates to the location and the severity of the stroke (Gordon et al., 2004, Michael et al., 2005, Kluding & Gajewski, 2009). Superimposed on normal losses attributed to aging, marked physical decline experienced by stroke survivors can result in a loss of functional independence and/or lead to significant adaptations in how the activity is performed. Though it is important to understand how these physical limitations manifest during mobility tasks like stair walking, it is equally important to study compensatory mechanisms, and how stroke survivors accomplish the task. Such knowledge would inform the safest method for stair ambulation and the implementation of rehabilitation to enhance safety and function.
It is important to study the biomechanics of stair walking and the movement strategies associated with healthy aging alongside the patterns observed in stroke to determine stroke-specific compensations i.e. those not observed as part of normal aging. Healthy older adults consider stair-walking to be the most difficult mobility task; they attribute this to aging (Williamson & Fried, 1996, Startzell et al., 2000 (review)). Compared to young adults, older adults demonstrate unique kinematic and kinetic patterns, which have been described as compensatory (Nadeau et al., 2003 Reeves et al., 2009, Novak & Brouwer, 2011). One of the earliest mobility compensations to manifest in older adults is reduced speed or cadence secondary to age-related decline in muscle strength; though when stroke is superimposed additional adaptations would be reasonably expected. Despite slower cadence, a re-distribution of joint torques has been observed in older adults essentially shifting the work from weaker muscles to those with better preserved strength; likely a strategy to keep muscular output within optimal limits (Reeves et al., 2009). At slower speeds, joint moments are correspondingly lower which allows the muscle work relative to the maximum force generating capacity to be comparable to that observed in young adults (Reeves et al., 2009, Novak & Brouwer, 2011).

Studies investigating the biomechanics of stair walking in stroke are limited (Novak & Brouwer, 2012a). In comparison to older adults, persons with stroke have a reduced capacity to generate the moments needed to accomplish the task as quickly, therefore, this group exhibits markedly reduced cadence (Novak & Brouwer, 2012a). Despite the slower speed, the relative strength costs are exceedingly high (Novak & Brouwer, 2012b) which may have implications for safety and community independence.

Another compensatory strategy adopted by some older adults is an altered stepping pattern. That is, both feet land on the same step (step-by-step, SBS) to reduce the muscular strength requirements (Shiomi, 1994). This strategy compensates for instability and weakness, and most notably reduces the knee extensor moment in the trail leg compared to the lead limb and
yields lower muscular demands than step-over-step (SOS, i.e. one foot lands on each successive step) (Shiomi, 1994). This is observed as an adaptation in hemispheric stroke to off-load the more affected limb. The impact on other physical systems such as the cardiovascular demands has not been adequately explored.

Approximately 75% of stroke survivors have diagnosed cardiovascular disease, which when paired with low levels of physical activity often results in significant deconditioning (Roth, 1993). In healthy individuals, the metabolic cost of stair ambulation is three times higher than overground walking (Teh & Aziz, 2001). As such, the fitness level required to negotiate stairs could pose a serious challenge for individuals with stroke. Indeed the aerobic cost for those with hemiparesis due to stroke is reportedly higher than that of similarly aged individuals when both groups use the usual SOS gait pattern (Novak & Brouwer, 2012b). Further, the mechanical inefficiencies of hemiparetic gait are associated with energy costs up to twice that of healthy counterparts (Potempo et al., 1995, Roth et al., 2000, Macko et al., 2001). The mechanical efficiency and oxygen costs of the SBS method, however, have not been described in stroke.

In order to characterize the demands of stair ascent and descent in stroke, it is important to evaluate the strength and aerobic costs, and determine the lower limb kinetics and mechanical efficiency. It is also relevant to investigate how these demands might differ as a function of compensatory behaviours; specifically, handrail use or altered gait pattern. A comprehensive analysis of the physical demands of stair negotiation using the SOS and SBS stair-stepping strategies will provide new knowledge applicable to the rehabilitation field to inform the development of new approaches and targeted interventions to enhance safety and mobility independence in stroke survivors.

1.1 Thesis outline

The focus of this thesis is to describe/evaluate the physical demands of stair negotiation (muscular and energy efficiency, both mechanical and aerobic) in persons with chronic stroke
compared to their healthy counterparts during two commonly used stepping strategies (SOS and SBS).

Chapter 2 provides a review of the relevant literature and provides the rationale for the work undertaken. Chapter 3 presents the first study which evaluates the relative strength and aerobic costs of stair ascent and descent using different stair negotiation strategies in stroke and healthy older adults. Chapter 4 presents the results of a secondary analysis of the energy efficiency (mechanical energy expenditure and oxygen consumption) during SOS and SBS stair negotiation strategies in the two groups (stroke and control).

A general discussion is presented in Chapter 5. Here, a summary of the findings, the study limitations and suggestions for future directions for research are described.
1.2 References


Chapter 2

Literature Review

Stroke is the leading cause of adult neurological disability in Canadian adults, disproportionately affecting the aging population. Over 50,000 strokes occur each year and over 300,000 are living with the effects of stroke (Statistics Canada, 2008, Tracking Heart Disease and Stroke in Canada, 2009). Nearly all survivors of mild stroke, and 85% of survivors of moderate to severe stroke return to living in the community (Stineman & Granger, 1998), although only a small proportion 30-50% are functionally independent in activities of daily living (Wade & Turnbull, 1987, McDowell, 1990).

The extent of physical deficits including muscle weakness, spasticity, sensory loss, and gait or balance impairments generally relates to the location and the severity of the stroke (Gordon et al., 2004, Michael et al., 2005, Kluding & Gajewski, 2009). The impact is often significant in terms of mobility restriction, loss of independence, social isolation, and reduced community participation (Duncan et al., 1994, Jorgensen et al., 1995, Pound et al., 1998, Mayo et al., 2002, Hankey et al., 2002, Hamel & Cavanagh, 2004,) and may be exacerbated by comorbidities and aging.

Seventy-five percent of individuals discharged post-stroke prioritize being active in the community (Lord et al., 2004), though mobility limitations, lower levels of physical activity post-stroke, and an inability to negotiate stairs can restrict access (Pound et al., 1998, Hamel & Cavanagh, 2004). It follows that the primary focus of rehabilitation is the restoration of independent ambulation; walking being the principle goal (Jette et al., 2005, Belda-Lois et al., 2011). While 71% of stroke survivors report incomplete recovery (Gadidi et al., 2011), 60-70% are able to walk at discharge from hospital (Wade & Turnbull, 1987, Thorngren et al., 1990, Lord
et al., 2004). This is tempered by accounts that only 7-22% can walk independently outside of their homes (Hill et al., 1997).

Arguably, independent living requires physical mobility beyond walking. Stair negotiation is an important determinant of discharge destination and independence, surpassing walking speed as the single best predictor of community ambulation (Stineman & Granger, 1998, Alzahrani et al., 2009) though we know little about the physical demands of stair ascent and descent. The movement patterns, strength requirements and cardiovascular demands of walking have been studied extensively in stroke (Olney et al., 1991, Olney et al., 1994, Nadeau et al., 1999, Kim & Eng, 2004, Parvataneni et al., 2007, Brouwer et al., 2009). A similar depth of understanding of the physical ‘cost’ of stair negotiation is essential to establish physical rehabilitation goals, retrain safe stair ambulation and understand the impact of alternate movement strategies adopted to accomplish the task.

2.1 Physical requirements of stair ambulation

2.1.1 Kinetic and kinematic requirements of stair ambulation

Stair walking requires both concentric and eccentric muscle activation to lift (or lower) the body vertically and translate it horizontally. Stair ascent primarily involves positive (concentric) work as the stance limb accepts body weight, pulls the body up to full support, and maintains progression while the swing limb clears the intermediate step and makes contact with the next step (forward continuance). During descent, the stance limb accepts body weight and controls the lowering of the body’s centre of mass (eccentric muscle work) as the swing limb is pulled forward to contact the lower step. A combination of adequate joint mobility, strength and cardiovascular function is required to successfully negotiate a flight of stairs.

Compared to walking, stair negotiation requires ~30% to 300% greater range of motion (ROM) in the sagittal plane at lower limb joints (Andriacchi et al., 1980, Costigan et al., 2002, Riener et al., 2002, Nadeau et al., 2003, Olney, 2005). The greatest range of motion (stairs: 90°,
walking: 70°) occurs at the knee (Nadeau et al., 2003); though the ankle ROM required for stair ambulation can be double that of walking (30-40° versus 10-20°, respectively) (Nadeau et al., 2003). It follows that restrictions in joint ROM attributable to weakness, elevated stiffness and/or joint disorders can adversely affect stair walking, but may have negligible or considerably lesser impact on walking.

In terms of strength, muscles must be able to generate the forces required to produce the necessary movement and to control the excursion of the centre of mass relative to the base of support. As with walking, the plantarflexors are important in generating an extensor moment that assists in maintaining upright support, reaching a peak during late stance to raise the body upward and progress it forward during ascent (McFadyen & Winter, 1988, Riener et al., 2002, Nadeau et al., 2003, Protopapadaki et al., 2007). In descent, the plantarflexors work eccentrically to accept body weight and provide stability (Riener et al., 2002, McFadyen & Winter, 1988, Novak & Brouwer, 2011). The magnitudes of the moments generated are comparable to those reported for walking.

In contrast, the knee extensors generate net moments of 12% to 25% greater peak magnitude during stair climbing than in walking (Andriacchi, et al., 1980, McFadyen & Winter, 1988, Costigan et al., 2002, Nadeau et al., 2003, Protopapadaki et al., 2007). Their main contribution, with the plantarflexors, is to the total support moment in keeping the body upright during weight acceptance and forward continuance (Novak & Brouwer, 2012a). During descent, the knee extensor moments are generally lower in magnitude than those associated with ascent (Novak & Brouwer, 2011) and the muscles work eccentrically to control lowering of the body with gravity. They also actively contribute to joint stiffness to stabilize the stance limb (McFadyen & Winter, 1988, Novak & Brouwer, 2011).

During walking, the hip extensors play a major role in energy generation (with the plantarflexors) for forward progression. In stair ascent more so than descent, the hip extensor
moment in early stance contributes to the support moment (McFadyen & Winter, 1988, Costigan et al., 2002, Rieker et al., 2002, Nadeau et al., 2003, Protopapadaki et al., 2007). Throughout the remainder of the stance phase, the hip moment shifts from extensor to flexor (higher magnitude in descent) to aid in stabilizing the mass of the trunk and upper body over the base of support (Winter, 1995, Novak & Brouwer, 2011). Figure 2-1 compares the lower limb joint moment profiles in the sagittal plane associated with walking, stair ascent and descent.

Figure 2-1 Net joint moment values for the ankle (a.), knee (b.) and hip (c.) during stance phase of overground walking (smooth line) (Eng & Winter, 1995), stair ascent (dotted line) (Novak & Brouwer, 2011) and stair descent (dashed line) (Novak & Brouwer, 2011) in healthy young adults; net internal flexor moments are negative.
Investigation of the frontal plane kinetic patterns in stair walking is limited (Nadeau et al., 2003, Novak & Brouwer, 2011). Hip abductor moments are comparable in magnitude during stair and overground walking, though there is a marked distinction in terms of function. In stair negotiation medio-lateral stability is controlled largely by the hip abductors and adductors (Novak & Brouwer, 2011) and during ascent, the abductor moment facilitates step clearance of the contralateral limb (Nadeau et al., 2003), unlike level ground walking where the hip remains in an adducted position during stance (Nadeau et al., 2003).

2.1.2 Energy requirements of stair ambulation

The muscle moments that generate the movements required for stair negotiation produce mechanical energy that can contribute to movement efficiency. When adjoining segments about a joint rotate in the same direction, energy may be transferred from one segment to another in a way that assists propulsion, for example, and reduces or compensates the magnitude of active force generation. When adjoining segments rotate in opposite directions, muscle energy must be entirely absorbed (eccentric muscle activity) or generated (concentric muscle activity), in other words, no segmental energy transfer occurs (see Appendix A for details).

Indeed level ground walking at self-selected speeds is quite energy efficient due largely to intersegmental energy transfer that compensates muscle work (Robertson & Winter, 1980, Olney et al., 1991, Zajac et al., 2002). Furthermore, the relationship between gait speed and oxygen consumption is U-shaped, where metabolic efficiency is optimum at moderate (natural) speed (Frederickson et al., 2007). In contrast, mechanical energy compensation during stair negotiation is markedly lower (Novak et al., 2011). That is, the muscles are not compensated to any significant degree and therefore must perform the positive (concentric) work of ascent and negative (eccentric) work of descent with only limited exploitation of the energy of motion. The fact that significant vertical displacement in addition to horizontal movement is required poses particular challenges. The mechanical energy expended by the ankle plantarflexors is particularly
high during late stance, and the knee extensors generate 1.5 to 10 times the energy during the pull up phase than that associated with walking; the requirements of ascent being higher than descent (Novak et al., 2011). The hip flexors primarily generate energy during descent, likely to control the lowering of the upper body (head, arms and trunk) during forward continuance.

The higher mechanical energy expenditures of stair ambulation are accompanied by higher metabolic requirements compared to overground ambulation (Waters et al., 1983, Reddy et al., 1989, Teh & Aziz, 2001). In healthy individuals, the aerobic cost of stair climbing at ~70-95 steps/min can exceed 3 times that of walking at ~1 m/sec (Basset et al., 1997, Startzell et al., 2000, Teh & Aziz, 2001). Stair descent is less aerobically demanding given the predominantly negative work, but nonetheless still yields higher oxygen demands than walking (Basset et al., 1997, Teh & Aziz, 2001). Individuals with compromised physical conditioning may find that the incremental challenge of meeting the demands of stair negotiation imposes restrictions in community ambulation.

Table 2-1 summarizes the relative demands of walking, stair ascent and descent in terms of joint mobility, muscle output and oxygen consumption in healthy, young adults illustrating the greater physical requirements associated with negotiating stairs. For older adults experiencing natural age-related changes in physical function, the higher demands may pose particular challenges and may explain why many describe stair ambulation as one of the most difficult tasks attributable to aging (Williamson & Fried, 1996).
Table 2-1 Joint angular displacement, peak joint moments and oxygen consumption associated with overground walking, stair ascent and stair descent in healthy adults

<table>
<thead>
<tr>
<th>ROM (°)</th>
<th>Overground Walking</th>
<th>Stair Ascent</th>
<th>Stair Descent</th>
</tr>
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<tbody>
<tr>
<td>Ankle plantarflexion</td>
<td>20</td>
<td>25-35</td>
<td>25-45</td>
</tr>
<tr>
<td>Ankle dorsiflexion</td>
<td>20</td>
<td>11-25</td>
<td>21-33</td>
</tr>
<tr>
<td>Knee flexion</td>
<td>70</td>
<td>90-95</td>
<td>90-105</td>
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<tr>
<td>Knee extension</td>
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<td>~0</td>
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<tr>
<td>Hip flexion</td>
<td>30</td>
<td>65-70</td>
<td>40</td>
</tr>
<tr>
<td>Hip extension</td>
<td>20</td>
<td>~0</td>
<td>~0</td>
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</tbody>
</table>

Peak Moments (Nm/kg)

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<th></th>
<th>Overground Walking</th>
<th>Stair Ascent</th>
<th>Stair Descent</th>
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</thead>
<tbody>
<tr>
<td>Ankle plantarflexor</td>
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<td>1.2-1.5</td>
<td>0.9-1.4</td>
</tr>
<tr>
<td>Ankle dorsiflexor</td>
<td>0.25</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
</tr>
<tr>
<td>Knee flexor</td>
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<td>0.4-1.0</td>
<td>0.46-1.3</td>
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<tr>
<td>Knee extensor</td>
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<td>0.58-1</td>
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<tr>
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<td>0.52</td>
</tr>
<tr>
<td>Hip extensor</td>
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<td>0.5-0.7</td>
<td>0.13-0.6</td>
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</table>

VO₂ (mL/kg/min)*

<table>
<thead>
<tr>
<th></th>
<th>Overground Walking</th>
<th>Stair Ascent</th>
<th>Stair Descent</th>
</tr>
</thead>
<tbody>
<tr>
<td>(Steady-state)</td>
<td>12-15</td>
<td>35</td>
<td>17</td>
</tr>
</tbody>
</table>

*values reported for steady state walking, ranging from self-selected slow to natural speed (estimated 1.0m/s and 1.1m/s, respectively), and stair ascent and descent at pace of 95 steps/min


2.2 Age-related changes in stair ambulation

Characteristics of the aging musculoskeletal system include increased joint stiffness attributable to losses in distensibility and elasticity of ligaments and muscles and degradation of articular cartilage (Bailey, 2001) and loss in strength due to reductions in muscle mass and motor unit numbers (Bailey, 2001, Doherty, 2003). These changes can restrict joint movement and be associated with joint pain, particularly under high load conditions as occurs at the knee during stair climbing (Andriacchi et al., 1980, Nadeau et al., 2003, McFadyen & Winter, 1988, Reeves et al., 2008). To compensate, individuals may alter their movement patterns and/or reduce cadence to limit the loads and muscle forces required.
By 70-80 years of age, men and women have about 50% of the strength of young adults (Doherty, 2001) and exhibit markedly reduced joint ROM (Hortobagyi & DeVita, 1999). It follows that older adults require greater effort to perform a specific physical task than their younger counterparts and therefore operate closer to their maximum capacity. Fortunately most activities of daily living require low to moderate physical effort keeping the strength requirements well within healthy individuals’ force generating capacity (Hortobagyi, 2003). For this reason it is not uncommon for age-related declines in strength to go largely undetected until capacity falls below some threshold level or the challenge of the task is increased.

Slowing or reducing the speed of the task is typically the earliest indicator of compensation for losses in strength. More demanding tasks can expose reductions in muscular system output sooner and may require adaptations in performance to adjust the motor demands of the task to keep the force requirements within optimal limits. Reeves et al. (2008, 2009) reported a redistribution of ankle and knee moments associated with stair negotiation in older adults. By ‘transferring’ the work from one muscle group to another with greater force generating capacity, these individuals were able to keep the muscle workloads within comfortable limits. There is ample evidence demonstrating that the relative contribution of the ankle, knee and hip flexors and extensors vary considerably in walking, however their summative contribution to maintaining an upright posture remains consistent in magnitude and profile (Winter et al., 1984). The same adaptability has been shown for stair ambulation providing options for compensating some degree of limitation (Novak & Brouwer, 2011).

During stair ascent and descent, in conditions where older and younger adults walked at the same cadence, older adults generated lower plantarflexor moments compared to young adults (Reeves et al., 2008, Reeves et al., 2009), but when these values were expressed relative to their maximum strength, the effort was similar. The same was observed for knee extensor moments (Reeves et al., 2009). Since both groups performed at similar speeds, it follows that older adults
rely on generating higher hip moments than their counterparts, though these values were not reported.

In our laboratory older adults typically ascend and descend stairs more slowly than young adults producing lower sagittal plane joint moments (Novak & Brouwer, 2011). Decreasing the speed of ascent or descent however, prolongs the duration over which the task is performed which may challenge endurance and elevate the metabolic cost. In the community, where distances and flights of stairs to be negotiated to arrive at a given destination may be fixed, physical conditioning could be a key factor of functional independence.

Novak et al. (2011) investigated the mechanical energy transfers across lower limb segments during stair negotiation in young and older adults. They found that the energy expenditures at the ankle, knee and hip for both stair ascent and descent were comparable, despite older adults having a slower cadence than young adults. This is somewhat surprising given the lower joint moments associated with the slower speed and suggests that perhaps inefficiencies in energy transfer may be more prevalent in older adults. The lesser compensation of concentric muscle activity in ascent and eccentric activity in descent in older adults than demonstrated by young adults supports this view (Novak et al., 2011). It is the case that both young and older adults adopt self-selected speeds of gait that optimize energy requirements, yet in walking, older adults use 15-25% more energy (J/kg) than younger people, at all speeds (Martin et al., 1992, Malatesta et al., 2003, Mian et al., 2006, Ortega & Farley, 2007, Ortega et al., 2008) suggesting a general upregulation of energy consumption with aging.

It is plausible that the reduced ability of older adults to optimize efficiency through energy transfer in stair walking is secondary to age-related physiological changes. If so, the relative mechanical inefficiency may correspond to a higher aerobic cost. This has not been directly addressed in the literature and few reports have examined the metabolic demands of stair negotiation in aging.
Healthy adults use approximately 44% of their maximum aerobic capacity (max VO\textsubscript{2}) to climb three flights of stairs (Reddy et al., 1989, Teh & Aziz, 2001). If one considers that the average healthy, untrained 70-year-old adult has an aerobic capacity equivalent to approximately half that of a 20 year old (Robinson, 1938, Astrand, 1960, Waters et al., 1983) and that energy expenditures of stair ascent and descent are comparable in young and old, then the metabolic demand of stair negotiation is likely significantly higher in older adults. This is of particular concern since the elevated mechanical inefficiency of the task in older adults could exacerbate the impact of reduced aerobic capacity. Additionally, when physical disability is superimposed on aging, mobility may be seriously compromised.

2.3 Changes in stair ambulation in people with stroke

There is limited research describing the biomechanics and physical demands of stair negotiation in the stroke population. Comparing task performance to age and sex matched healthy counterparts would enable the characterization of those adaptations and compensations attributable to stroke by controlling for the otherwise confounding age-related effects. From a rehabilitation perspective this is important.

Hartman-Maeir and colleagues (2007) reported that among stroke survivors of one year who had returned to living at home, 57% required mild-to-moderate assistance with managing stairs and 4% required full assistance. Others have found that the mobility status of 20% of those with chronic stroke deteriorates significantly between 1 and 3 years post-stroke (van de Port et al., 2006). Considering the physical demands of stair ambulation, the rate of decline in the ability to negotiate stairs may be particularly steep in individuals living with stroke. Furthermore if there is little reserve capacity in one or more of the neuromuscular, postural or cardiorespiratory systems then in addition to mobility limitation, safety may be at risk (Young, 1986, Skelton et al., 1994, Lord & Menz, 2002, Tiedemann et al., 2007).
Joint mobility is often limited following stroke due to increased joint stiffness (Levin & Hui-Chan, 1994, Rydahl & Brouwer, 2004, Den Otter et al., 2007) and muscle weakness which is most pronounced on the paretic side with distal muscles most affected (Teixeria-Salmela et al., 1999, Eng et al., 2002, Kim & Eng, 2003). Notably, the hemiparetic ankle lacks range in dorsiflexion secondary to plantarflexor stiffness and paresis of the dorsiflexors, which in combination may be associated with ‘foot drop’ (Intiso et al., 1994). This presents a tripping risk in walking and reduces the ability to achieve adequate step clearance during stair negotiation (Nadeau et al., 2003). To compensate, individuals may adopt extraneous movements, such as hip circumduction to elevate the lower limb and swing it laterally to clear the rise of the stair, but risking medio-lateral instability.

Recently published work from our laboratory has revealed that the peak ankle and knee extensor moments generated during stair ascent and descent in people with stroke represent a 15-50% higher percentage of maximum strength than age and sex matched healthy adults (Novak & Brouwer, 2012b). The abnormally elevated strength ‘cost’ of stair ambulation bodes poorly for maintaining mobility independence with increasing age. Such high demands on the muscular system leaves little reserve force generating capacity to increase speed or control/compensate for a loss in balance. This is a significant concern.

The kinetic profiles associated with stair negotiation in stroke reveal a general reduction in lower limb joint moment magnitudes, particularly on the affected side compared to healthy adults (Novak & Brouwer, 2012b). This is partly explained by the slower cadence, which is compensatory for limitations in speed control associated with paresis and other physical impairments. The maximum muscle forces generated on the affected side reflect 20-88% of those on the sound side, which are also abnormally weak (Hsu et al., 2002, Kim & Eng, 2003, Barbic & Brouwer, 2008, Novak & Brouwer, 2012). Compensating the weakest (distal most) muscles by
substituting with those having better preserved strength (proximal muscles) can be an effective strategy.

In hemiparetic walking, the hip flexors generate the power to bring the paretic leg through swing offsetting the loss in plantarflexor power production (Parvataneni et al., 2007). Strategies as described by Reeves et al. (2008, 2009) to redistribute muscle work during stair negotiation to accommodate the relative capacity of the ankle and knee flexors and extensors may also be useful in stroke, though it may come at a high metabolic cost. Additional research is warranted to explore this further considering that 75% of those with stroke have cardiovascular disease (Roth, 1993). Poor cardiovascular health and associated low aerobic capacity (Mackay-Lyons & Howlett, 2005, Pang et al., 2005) compromise the ability to meet the oxygen demands of routine activities like walking and managing stairs.

As a result of motor inefficiencies, hemiparetic gait yields a higher mechanical energy cost than normal gait accompanied by an elevated metabolic cost; 66%-76% of maximum VO$_2$ in stroke compared to 27% in healthy counterparts (Michael et al., 2005, Ivey et al., 2005). It is reasonable to infer that similar differences exist for stair negotiation. The elevated strength and aerobic demands associated with ascent and descent (Basset et al., 1997) could render the activity unmanageable and unsafe for those with stroke. Stair ambulation is rated as the most difficult activity to perform following stroke rehabilitation often leading to avoidance of stairs (Tsuji et al., 1995) or the adoption of alternate strategies to accomplish the task (Startzell et al., 2000).

A compensatory strategy adopted by people with mobility or balance deficits (including stroke) is to climb and descend stairs using the “step-by-step” method (both feet contacting each step). This permits most work to be accomplished by the stronger muscles on the less affected side (Startzell et al., 2000, Kim & Eng, 2003) and prolongs periods of double support for (re)stabilization (Winter, 1995). Ascending stairs in a step-by-step fashion however, elevates both heart rate and oxygen consumption by about 10% in healthy individuals compared to the step-
over-step method (Shiomi, 1994). The longer time to task completion resulted in higher cardiovascular cost. Our own preliminary data suggest that the oxygen consumption associated with step-by-step ascent is 15%-20% higher compared to step-over-step for a single flight of stairs (16 steps). If the estimated maximum VO\(_2\) is nearly 25% lower in stroke (Novak & Brouwer, 2012b), this presents substantial cardiovascular challenge, but may be a necessary compromise if the strength deficits would otherwise preclude performance of the less energy-demanding step-over-step strategy.

Stroke survivors are one of the largest consumer groups of rehabilitation services (Heart Disease and Stroke Statistics, 2002). After completion of conventional physical therapy in the (sub)acute phase, there are no recommended protocols and few resources promoting regular exercise for those with chronic stroke (Ivey et al., 2005). In fact, more than half of stroke survivors are inactive and sedentary, which affects both their physical and social functioning (Patel et al., 2006).

The extensive body of research describing hemiparetic gait and the factors that determine gait speed in level-ground walking has contributed to the implementation of targeted interventions for gait retraining even years beyond the stroke event (Olney et al., 1991, Olney et al., 1994, Nadeau et al., 1999, Kim & Eng, 2004, Parvataneni et al., 2007, Novak et al., 2009, Brouwer et al., 2009). In a similar vein, if we are to enhance stair mobility we require a sound understanding of the factors that limit or restrict stair mobility and their inter-relationships. It is equally important to investigate the movement strategies that individuals with stroke adopt to compensate for physical limitations in order to determine the impact on movement efficiency. Only in this way will we gain insight into the relative demands placed on the muscular and cardiovascular systems in order to successfully ascend and descend stairs.

This study examined the kinematic, kinetic and cardiovascular demands of stair ascent and descent using two stepping strategies (step-by-step and step-over-step) in stroke and healthy,
age-matched control subjects (Chapter 3). A secondary analysis of the data to explore the mechanical and metabolic efficiencies of the two strategies was carried out to determine whether both indicators of energy efficiency support similar conclusions (Chapter 4).
2.4 References


Heart Disease and Stroke Statistics-2003 Update. Heart and Stroke Facts. Dallas, TX: American Heart Association; 2002


Tracking Heart Disease and Stroke in Canada. Released June 2009.


Chapter 3

Strength and aerobic demands of different stair-stepping strategies in healthy aging and stroke

3.1 Introduction

Stroke is the leading cause of adult neurological disability in Canada. Eighty-five percent of stroke survivors return to living in the community (Stineman & Granger, 1998), and independent mobility is a major goal to enhance independence and community participation (Mayo et al., 2002, Lord et al., 2004, Flansjber et al., 2006). A main focus of rehabilitation has been on gait re-training (Lord et al., 2004, Ditunno et al., 2005, Jette et al., 2005, Flansjber et al., 2006, Belda-Lois et al., 2011 (review)), yet even though 60-85% of stroke patients are able to walk by discharge (Wade & Turnbull, 1987, Dean & Mackey, 1992, Jorgensen et al., 1995), only 7-22% report they are able to ambulate independently outside of their homes (Hill et al., 1997, Jorgensen et al., 1995). Others indicate they do not achieve a walking level that enables them to perform their daily activities unassisted (Flansjber et al., 2005).

Stroke-related gait impairments attributed to physical deficits including muscle weakness (Nakamura et al., 1985, Bohannon & Andrews, 1990, Nadeau et al., 1999, Kim & Eng, 2003), balance deficits (Bohannon, 1987, Dettmann et al., 1987, Bohannon, 1989), and de-conditioning (Michael et al., 2005) are well described. The slow walking speed, usually between 0.4 and 0.8 m/s in stroke may provide a means for compensating for physical deficits or reflect a direct consequence of them (Hill et al., 1997, Duncan et al., 1998, Eng et al., 2002, Green et al., 2002, Pohl et al., 2002). However, it is slower than what is considered adequate for safe community ambulation (for example, crossing an intersection) (Hill et al., 1997). Furthermore, inefficiencies in motor patterns are associated with an elevated metabolic cost of up to 1.5 to 2 times that of normal overground walking (Corcoran et al., 1970, Gersten & Orr, 1971, Ivey et al., 2005, Platts
et al., 2006). It is important to understand the physical requirements, mobility challenges and compensatory strategies of ambulation activity post-stroke in order to restore function and independence, but to date, certain task requirements remain ill-defined. Stair negotiation, for example, is an essential mobility task in most community settings and has been identified as the single best predictor of active community living (Alzahrani et al., 2009), yet few studies have explored the physical requirements needed to perform the task post-stroke (Novak & Brouwer, 2012b).

Compared to walking, stair ascent and descent require greater lower limb joint range of motion (Andriacchi et al. 1980, Andriacchi et al., 1982, Nadeau et al., 2003, Riener et al., 2002), strength (Andriacchi et al., 1980, Andriacchi et al., 1982, Nadeau et al., 2003, Novak & Brouwer, 2011) and cardiovascular fitness (Reddy et al., 1989, Teh & Aziz, 2001). The biomechanical challenge of moving the body’s centre of mass (COM) vertically as well as horizontally increases the physical demands considerably. Older adults report stair negotiation as the most challenging physical task which they attribute to aging (Williamson & Fried, 1996). Following physical rehabilitation stroke survivors rate stair ambulation among the most difficult tasks often leading to avoidance of stairs (Tsuji et al., 1995).

Considering that community-dwelling stroke survivors are able to generate about 70% of the maximum muscle power and have approximately 50% of the maximum aerobic capacity of their gender and age-matched counterparts (Gordon et al., 2004), they are likely working closer to the limits of their physical capabilities (Macko et al., 2001, Ivey et al., 2005, Novak & Brouwer, 2012b). When task demands are high and physical capacity is reduced, compensatory movements may be adopted in order to complete the task. Older adults have been shown to re-distribute the ankle and knee extensor output compared to young, healthy adults as a means of compensating ankle weakness and keeping the muscle workloads within comfortable limits (Reeves et al., 2008, 2009). Similarly, an exaggerated hip abductor moment to aid in step clearance has been observed.
to compensate for contralateral plantarflexor weakness associated with hemiparesis which limits the vertical excursion of the centre of mass (Nadeau et al., 2003). When weakness compromises stability, older adults may modify the stepping pattern from the traditional step-over-step (SOS) to a step-by-step (SBS) (both feet on each step) strategy (Reid et al., 2007). In hemiparesis secondary to stroke this is more common due to the asymmetrical lower limb weakness and balance deficits (Startzell et al., 2000 (review), Kim & Eng, 2003). With the SBS strategy, twice as many steps are required to cover the same distance which has been shown to increase the metabolic cost (Shiomi, 1994). This may pose a challenge for individuals with cardiovascular disease such as stroke. The purpose of this study was to quantify the relative strength and aerobic demands of step-over-step and step-by-step strategies of stair ascent and descent in people with stroke in comparison with similarly aged healthy counterparts.

3.2 Methodology

3.2.1 Subject characteristics

Fifteen individuals with chronic hemispheric stroke (> 6 months) were recruited through outpatient stroke clinics, newspaper advertisements, and stroke support groups. Age- (within 3 years) and sex-matched older adults were recruited from the community and served as a control group. All participants were able to walk independently (though some in the stroke group used a walking aid outdoors), ascend and descend a flight of stairs with or without the use of a handrail, and all self-reported that they were in good health. Individuals were excluded from the study if they reported a non-stroke related mobility restriction (e.g. severe arthritis, joint replacement) or heart disease (e.g. peripheral vascular disease, uncontrolled hypertension, congestive heart failure, unstable angina) as these comorbidities could influence the outcome measures of the study. All procedures were approved by the university ethics board and participants provided their written, informed consent prior to testing.
Demographic information including age, sex, height, weight, time since stroke, and relevant medications (i.e. beta-blockers) was collected. Performance measures including the Community Balance and Mobility scale (CBM), Timed Up and Go (TUG) and 10 meter walk velocity were measured to provide an indication of overall physical ability.

### 3.2.2 Biomechanical assessment of stair negotiation

Participants were asked to ascend and descend a custom four-step staircase (rise: 15cm, run: 26cm, width: 56cm) with a moveable handrail positioned on one side (the less affected side in stroke and the dominant side in healthy controls). A forceplate (AMTI, Newton, MA) mounted on concrete blocks formed the centre of the second step. The handrail was instrumented with two uniaxial load cells positioned at the top and bottom of the stairs to measure compressive forces transmitted through the hand during stair ascent and descent. An optoelectric camera (Northern Digital Inc., Waterloo, ON) was positioned on each side of the staircase to track the position of infrared emitting diodes (IREDs) placed on the lateral side of participants’ lower limbs as they negotiated the stairs. The IRED clusters (consisting of 3 or 4 diodes) were secured at the midfoot, midshank, midthigh and on either side of a fin projecting outward from the sacrum region (S2). Bilateral kinematic data were acquired at a sampling rate of 50Hz and kinetic (ground reaction force) data were acquired at a sampling rate of 100Hz.

Lower limb segment lengths, joint centres and rotational axes were defined using a digitizing probe with 4 IREDs in fixed positions relative to the tip. A static standing reference trial in which bilateral virtual landmarks (first and fifth metatarsal heads, medial and lateral malleoli, medial and lateral epicondyles, greater trochanters, and anterior superior iliac spines) were located for each subject.

Data are reported for the sagittal plane on the affected (AFF) and less affected (LAFF) sides in stroke, and the dominant (DOM) and non-dominant (NDOM) side in healthy controls. In order to determine the dominant side, participants were asked which side they write with and
which foot they would use to kick a ball. The peak joint moments at the ankle, knee and hip in the sagittal plane generated during stair ascent and descent were computed for each trial, using commercial software (Visual 3D, C-Motion, Inc., Germantown, MD) and an inverse dynamics approach incorporating a seven segment link-segment model (see Novak & Brouwer, 2011 for details).

Typically five trials of each stair walking condition (SBS ascent/descent, SOS ascent/descent) were collected from each participant. Successful trials, defined as those in which all IREDs were visible and the stance limb contacted the force plate were selected for analysis.

3.2.3 Maximum strength assessment

Isometric joint torques at the ankle, knee and hip were obtained using a dynamometer (Biodex, System 3, Shirley, NY). For testing at the ankle and knee, participants were seated in a chair tilted back 5° from vertical. When testing the ankle, the foot was secured in a footplate and strapped tightly with a padded Velcro strap so that the ankle remained in a neutral position (~90°) with the knee positioned in approximately 60° (0° full extension) of flexion and the shank supported by an adjustable padded fixture such that the lower leg was parallel to the floor. An additional padded strap was secured around the thigh to stabilize the proximal segments. When testing at the knee, the joint was positioned in 90° of flexion with the thigh securely strapped to the chair and the Velcro pad placed 2cm above the lateral malleolus at the distal end of the dynamometer arm.

For testing at the hip, the backrest of the chair was flattened to allow participants to lie supine with their lower limbs hanging freely over the edge of the seat. The thigh of the test limb was raised slightly forming a hip angle of 15° from the horizontal and supported in this position using a padded strap attached to the end of the dynamometer and secured proximal to the knee.

For all testing, participants were allowed a practice trial. They were instructed to cross their arms over their chest and to breathe normally while they pushed or pulled as hard as
possible against the immovable arm of the dynamometer (velocity set to 0°/sec); encouragement was given by the researchers throughout the testing. Each maximal contraction was held for five seconds (flexion or extension). After twenty seconds of rest, subjects contracted the opposing muscle group for five seconds. Two trials of maximal flexor and extensor contractions were performed at each joint. The peak torque generated was operationally defined as maximum strength.

3.2.4 Estimation of relative strength cost

Peak flexor and extensor joint moments generated during the stance phase of stair ascent and descent were expressed as a ratio of the corresponding maximum isometric torque. This provided an estimate of the relative strength cost associated with each of the stair-walking strategies.

3.2.5 Aerobic assessment of stair negotiation

Oxygen consumption (VO₂) and heart rate (HR) data were acquired using a K4b² metabolic unit (Cosmed, Italy) while subjects ascended or descended a full flight of stairs (16 steps). Subjects wore a mask covering their mouth and nose which was equipped with a flow meter (turbine) that transmitted exhaled air to the portable unit for breath-by-breath analysis. The portable unit was supported in a harness worn over the chest and back of the participant. A Polar chest belt monitored heart rate through a receiver connected to the portable unit. Prior to data collection, standard calibration of the metabolic unit was performed according to the manufacturer’s specifications (K4b² User Manual, 2006) to ensure accurate sensor and airflow volume readings.

Resting metabolic values were recorded during 7-8 minutes of quiet sitting. The participants then performed one stair trial for each strategy (SOS descent, SOS ascent, SBS descent, SBS ascent) at their self-selected speed. All participants used the handrail for safety. After completing each trial, participants were asked to sit until their heart rate and oxygen
consumption returned to resting values (usually 2-3mins). Peak heart rate and oxygen consumption were determined for each of the four stair-walking trials.

3.2.6 Maximum aerobic assessment

The YMCA submaximal cycle ergometer test was administered to estimate each subject’s maximum oxygen consumption (VO$_{2\text{max}}$). A cycle test is preferred for older adults and those with stroke because it is minimally impacted by balance impairments. Furthermore, with this test, the timing of increments in workload is based on the participant’s heart rate during the exercise test rather than occurring at fixed or standard intervals. To enhance stability, a back rest with optional shoulder straps was added to the bike seat and straps were used to secure the feet to the pedals.

Estimated maximal heart rate (HR$_{\text{max}}$) was calculated for each participant using the equation (208 - 0.7 x age), as recommended for adults older than 60 years of age (Tanaka et al., 2001), or (162 - 0.7 x age) for individuals on beta-blocker medication (Brawner et al., 2002).

The test protocol was explained to the participant and the pedals and seat were adjusted as needed. The day prior to testing, subjects were familiarized with the test and given the opportunity to practice pedaling at a rate of 50 rpm which was assisted by a metronome. During the test protocol, blood pressure was recorded at rest and at the 2 minute mark of each workload; the test was discontinued if the systolic pressure rose above 250mmHg and/or the diastolic pressure rose above 120mmHg. A minimum of 6 minutes (2 workloads of 3 minutes each) were required to estimate VO$_{2\text{max}}$.

All participants started at a workload of 25 watts which was increased by 25 watts after 3 minutes provided that the HR at the 2 and 3 minute marks differed by less than or equal to 5 beats per minute (bpm), otherwise the workload was maintained for an additional minute until the HR stabilized. The workload was increased by an increment greater than 25 watts if the HR remained below a defined threshold based upon the estimated maximum HR (see Appendix C).
The test was terminated when any of the following criteria was met: HR within 10 beats of 85% of the estimated HR$_{\text{max}}$, respiratory exchange ratio (RER) > 1.15, or the participant reached exhaustion (score of 18-20 on the Borg scale of perceived exertion, Borg, 1982).

3.2.7 **Estimation of relative aerobic cost**

The oxygen consumed during stair ascent and descent was expressed as a ratio of the estimated VO$_{2\text{max}}$ to provide an estimation of relative aerobic cost for each of the stair-walking strategies. Since maximum heart rate can be estimated from an equation based on age, and we know that rises and falls in VO$_2$ are coupled with fluctuations in heart rate, the slope of the VO$_2$ increase during activity can be used to predict VO$_{2\text{max}}$ (Astrand & Rodahl, 1986). See Appendix D for detailed calculations.

3.2.8 **Procedure**

Data collection took place over two test days (2-3 hours each day) approximately one week apart. The biomechanical evaluation of stair negotiation, isometric strength testing and demographic information were completed on the first day. The performance measures (CBM, TUG, 10m walk), aerobic requirements of stair ascent and descent and the submaximal cycle ergometer test were completed on the second test day.

3.2.9 **Statistical analyses**

Descriptive statistics (means and standard deviations, (SDs)) were calculated for all outcome measures (SPSS version 21). Independent t-tests were used to identify demographic and performance differences between groups (stroke vs. control). A one-way ANOVA was performed on average peak joint moments, and relative strength ratios to identify differences across limbs (LAFF, AFF, DOM, NDOM). Post-hoc pairwise comparisons were conducted as appropriate. Two by two ANOVAs were performed to examine the effect of strategy (SOS and SBS) and group (stroke, control) as well as interactions between these factors on peak joint moments and
strength costs. The ANOVAs were carried out accounting for the lead (LAFF, DOM) and trail (AFF, NDOM) limbs for the SBS strategy. A two by two ANOVA was used to determine between-group (stroke, control) and within-group (strategy; SBS, SOS) differences in aerobic costs and the presence of an interaction between factors. For all analyses in which significant group or interaction effects were found, a covariate relating to speed was incorporated into the model since speed can impact peak joint moment values. Analyses were carried out in relation to ascent and descent separately and a significance level of p<.05 was used for all analyses.

3.3 Results

3.3.1 Subject characteristics

Thirty subjects (fifteen with chronic stroke) participated in the study. Thirteen stroke and thirteen controls were included in the analysis (8 males, 5 females) as the data for 2 subjects from each group were discarded due to incorrigible artefacts in the kinematic data. All participants were able to ascend and descend the stairs using the handrail, and two participants in the stroke group were able to only perform SBS stair ambulation including one who could ascend but not descend.

Stroke and healthy controls were similar in height and weight (p>0.06). In the stroke group (1-16 years post-stroke), eight presented with right hemiparesis; all control participants were right side dominant. As expected, the control group scored higher (better balance) on the Community Balance and Mobility Scale (p<0.001) and demonstrated faster Timed Up and Go times (p=0.01) than the stroke group. The self-selected walking speed was also faster (p<0.001) for the control group. Group characteristics are summarized in Table 3-1.
Table 3-1 Characteristics of stroke (n=13) and healthy control (n=13) groups.

<table>
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<th>Stroke</th>
<th>Control</th>
<th>p-value</th>
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<td>Age (years)</td>
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<td>66.5 ± 10.7</td>
<td>.743</td>
</tr>
<tr>
<td>Sex (M/F)</td>
<td>8/5</td>
<td>8/5</td>
<td></td>
</tr>
<tr>
<td>Height (m)</td>
<td>1.65 ± 0.1</td>
<td>1.72 ± 0.1</td>
<td>.064</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>71.8 ± 13.1</td>
<td>74.8 ± 15.8</td>
<td>.603</td>
</tr>
<tr>
<td>Right side affected or dominant (n)</td>
<td>8</td>
<td>13</td>
<td></td>
</tr>
<tr>
<td>Time since stroke (months)</td>
<td>50 ± 47</td>
<td>n/a</td>
<td></td>
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<tr>
<td>Beta-blocker medications</td>
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<td></td>
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<tr>
<td>CBM (maximum score = 96)</td>
<td>46 ± 20</td>
<td>82 ± 14</td>
<td>&lt;.001</td>
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<td>TUG (s)</td>
<td>13.8 ± 9.3</td>
<td>6.6 ± 1.2</td>
<td>.010</td>
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<tr>
<td>10m walk speed (m/s)</td>
<td>0.9 ± 0.3</td>
<td>1.4 ± 0.2</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>SBS Ascent cadence (steps/min)</td>
<td>31.6 ± 9.6</td>
<td>57.6 ± 16.1</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>SBS Descent cadence (steps/min)</td>
<td>32.8 ± 6.9</td>
<td>63.0 ± 19.5</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>SOS Ascent cadence (steps/min)</td>
<td>68.7 ± 11.1</td>
<td>113.3 ± 16.8</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>SOS Descent cadence (steps/min)</td>
<td>63.5 ± 18.2</td>
<td>120.1 ± 28.7</td>
<td>&lt;.001</td>
</tr>
</tbody>
</table>

CBM=Community Balance and Mobility Scale; TUG=Timed Up and Go; SBS=step-by-step; SOS=step over-step; n/a=not applicable
p-values reflect the between group comparison

3.3.2 Biomechanical assessment of stair negotiation

On average, individuals with stroke ascended and descended stairs at approximately half the cadence of their healthy counterparts (p<.001), though cadence was influenced by the strategy used (SOS or SBS), p<.001 (Table 3-1). There was an interaction effect (p<.001) between group and strategy suggesting that the impact of strategy was more pronounced in stroke. Both groups had a markedly higher cadence using the SOS strategy relative to the SBS method (p<.001).

Ascent

In general, joint moment profiles are similar in shape for both groups (Figure 3-1). The SOS and SBS lead limbs perform a similar role, which is to lift the body mass vertically. The major work being accomplished by the knee extensors. The SBS strategy effectively off-loads the
AFF (trail limb) knee extensors; as the trail limb need not perform the work to raise the body’s centre of mass upwards against gravity, but rather need only provide stability and elevate the limb to the next step. The trail limb plantarflexors produce a greater torque than the lead limb plantarflexors particularly in controls (p<.001) likely reflecting its role in pushing off to accelerate the limb upward to the next step. No group or limb effects were detected at the hip. These findings are summarized in Table 3-2.

![Figure 3-1](image-url) Figure 3-1 Average peak joint moment profiles of the stance phase of stair ascent for the (a.) step-over-step lead limb, (b.) step-by-step lead limb, and (c.) step-by-step trail limb, in the stroke affected, less affected and control dominant and non-dominant ankle, knee and hip.
With the SOS strategy, peak plantarflexor moments were higher for the control group than in stroke for either limb (p=.024), though accounting for speed eliminated the plantarflexor difference between groups. No differences were detected in the peak moments associated with other lower limb joints (p≥0.131). The absence of differences in moment magnitudes across limbs relative to stroke and controls at the knee or hip despite the difference in cadence between groups in combination with the speed related differences at the ankle could reflect the relative importance of the plantarflexors in controlling speed.

The joint moment profiles for the SOS strategy appear to have greater peak magnitudes compared to the SBS (lead limb) strategy. Contrasting the two strategies along with group (less affected stroke, control dominant), a main effect of strategy was observed for the plantarflexors (p<.001) as well as a group by strategy interaction effect (p=.019). Control subjects produced a mean peak plantarflexor moment during SOS ascent that is more than twice the magnitude associated with SBS (1.30 vs .62 Nm/kg) whereas the difference between strategies in stroke is less pronounced (.71 vs .46 Nm/kg), Table 3-2. The interaction effect is retained even after considering speed as a covariate (p=.037).

A main effect of strategy is also apparent for the knee flexors (p=.001) and hip flexors (p=.031) with higher moments associated with SOS than SBS. There was no main effect of group or interactions between factors (p≥.053).
Table 3-2 Average peak joint moments (Nm/kg) associated with the step-by-step (SBS) and step-over-step (SOS) strategies of stair ascent for the affected and less affected side in stroke and dominant and non-dominant control limbs.

<table>
<thead>
<tr>
<th>SBS</th>
<th>Stroke Less Affected</th>
<th>Stroke Affected</th>
<th>Control Dominant</th>
<th>Control Non-dominant</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>(lead limb)</td>
<td>(trail limb)</td>
<td>(lead limb)</td>
<td>(trail limb)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Plantarflexor</td>
<td>.46 ± .29†</td>
<td>.65 ± .43†</td>
<td>.62 ± .22*</td>
<td>1.18 ± .25*†‡</td>
<td>.001</td>
</tr>
<tr>
<td>Dorsiflexor</td>
<td>.11 ± .18</td>
<td>.01 ± .00</td>
<td>-</td>
<td>.01 ± .00</td>
<td>.514</td>
</tr>
<tr>
<td>Knee Extensor</td>
<td>.94 ± .76†‡</td>
<td>.25 ± .44‡^</td>
<td>1.08 ± 0.37‡^</td>
<td>.32 ± .14*‡</td>
<td>.001</td>
</tr>
<tr>
<td>Knee Flexor</td>
<td>.17 ± .07†</td>
<td>.25 ± .17</td>
<td>.15 ± .07*</td>
<td>.38 ± .24*†</td>
<td>.001</td>
</tr>
<tr>
<td>Hip Extensor</td>
<td>.33 ± .44</td>
<td>.29 ± .24</td>
<td>.22 ± .19</td>
<td>.43 ± .31</td>
<td>.415</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>SOS</th>
<th>Stroke Less Affected</th>
<th>Stroke Affected</th>
<th>Control Dominant</th>
<th>Control Non-dominant</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>(lead limb)</td>
<td>(lead limb)</td>
<td>(lead limb)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Plantarflexor</td>
<td>.71 ± .79†</td>
<td>.65 ± .72*</td>
<td>1.30 ± .18*†</td>
<td>Na</td>
<td>.024</td>
</tr>
<tr>
<td>Dorsiflexor</td>
<td>.02 ± .02</td>
<td>.01 ± .01</td>
<td>-</td>
<td>Na</td>
<td>.131</td>
</tr>
<tr>
<td>Knee Extensor</td>
<td>.75 ± .96</td>
<td>.47 ± .75</td>
<td>1.0 ± .36</td>
<td>Na</td>
<td>.208</td>
</tr>
<tr>
<td>Knee Flexor</td>
<td>.28 ± .15</td>
<td>.21 ± .18</td>
<td>.21 ± .12</td>
<td>Na</td>
<td>.464</td>
</tr>
<tr>
<td>Hip Extensor</td>
<td>.28 ± .54</td>
<td>.20 ± .44</td>
<td>.24 ± .34</td>
<td>Na</td>
<td>.887</td>
</tr>
<tr>
<td>Hip Flexor</td>
<td>.18 ± .06</td>
<td>.30 ± .27</td>
<td>.23 ± .13</td>
<td>Na</td>
<td>.242</td>
</tr>
</tbody>
</table>

p-value is for one-way ANOVA, *, †, ‡, ^ signify where the difference exists, Na = not analyzed

When contrasting SOS and SBS strategies with respect to the trail limb for SBS, it is important to note that for control participants the limb (NDOM, DOM) was not matched across strategies as was the case in stroke. In the absence of inter-limb asymmetries in controls using SOS ascent (descent), it was not considered necessary (Novak & Brouwer, 2011). A main effect of group was observed for the plantarflexors (p=.001), associated with higher moments generated in the control group than the stroke group. The strategy used did not impact the moment magnitudes for the plantarflexors (p=.554), but did for the knee flexors (p=.05) and extensors (p<.001). The flexor moments, which tend to be relatively low during ascent, were lower in association with SOS than SBS; whereas, the opposite pattern was noted for the extensors and especially so for control subjects (interaction effect, p =.045). No differences in peak hip moments were observed between ascent strategies (p>.171).
Descent

In general, the joint moment profiles are similar in shape for stroke and control groups (Figure 3-2). During SOS and SBS stair descent, the lead limbs have similar roles to prepare to support the weight of the body as it is transferred to the step below. The SBS trail limb has the role of controlling the lowering of the body’s centre of mass. In SBS, there was a main effect of limb for the mean peak plantarflexor moment (p=.036) and knee extensor moments (p=.038), demonstrating markedly higher magnitudes in the trail limb of control subjects than the lead and trail limbs in stroke, respectively (Table 3-3).

Although a significant effect of limb was detected for the dorsiflexors (p=.032), the relevance of this finding given the moment magnitudes is considered negligible. No differences across limbs were observed for the SOS strategy despite the appearance of greater plantarflexor and knee extensor moments in the control group relative to the stroke group (Table 3-3).

Figure 3-2 Average peak joint moment profiles of the stance phase of stair descent for the (a.) step-over-step lead limb, (b.) step-by-step lead limb, and (c.) step-by-step trail limb, in the stroke affected, less affected and control dominant and non-dominant ankle, knee and hip.
Table 3-3 Average peak joint moments (Nm/kg) associated with the step-by-step (SBS) and step-over-step (SOS) strategies of stair descent for the affected and less affected side in stroke and dominant and non-dominant control limbs.

<table>
<thead>
<tr>
<th></th>
<th>Less Affected (lead limb)</th>
<th>Affected (trail limb)</th>
<th>Dominant (lead limb)</th>
<th>Non-dominant (trail limb)</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>SBS</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Plantarflexor</td>
<td>.50 ± 0.34*</td>
<td>.53 ± .69</td>
<td>.70 ± .26</td>
<td>.95 ± .20*</td>
<td>.036</td>
</tr>
<tr>
<td>Dorsiflexor</td>
<td>.09 ± 0.10*</td>
<td>.01 ± .00*</td>
<td>.02 ± .02</td>
<td>.01 ± .01</td>
<td>.032</td>
</tr>
<tr>
<td>Knee Extensor</td>
<td>.65 ± .48</td>
<td>.56 ± .71*</td>
<td>.63 ± .36</td>
<td>1.10 ± .15*</td>
<td>.038</td>
</tr>
<tr>
<td>Knee Flexor</td>
<td>.10 ± .06</td>
<td>.07 ± .07</td>
<td>.09 ± .07</td>
<td>.08 ± .05</td>
<td>.773</td>
</tr>
<tr>
<td>Hip Extensor</td>
<td>.02 ± .29</td>
<td>.25 ± .55</td>
<td>.02 ± .18</td>
<td>.12 ± .08</td>
<td>.195</td>
</tr>
<tr>
<td>Hip Flexor</td>
<td>.50 ± .41</td>
<td>.33 ± .39</td>
<td>.52 ± .39</td>
<td>.35 ± .22</td>
<td>.424</td>
</tr>
<tr>
<td><strong>SOS</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Plantarflexor</td>
<td>.75 ± .80</td>
<td>.82 ± .60</td>
<td>1.13 ± .25</td>
<td>Na</td>
<td>.267</td>
</tr>
<tr>
<td>Dorsiflexor</td>
<td>.03 ± .03</td>
<td>.01 ± .01</td>
<td>.03 ± .02</td>
<td>Na</td>
<td>.447</td>
</tr>
<tr>
<td>Knee Extensor</td>
<td>.79 ± .97</td>
<td>.65 ± .79</td>
<td>1.14 ± .44</td>
<td>Na</td>
<td>.284</td>
</tr>
<tr>
<td>Knee Flexor</td>
<td>.13 ± .11</td>
<td>.09 ± .08</td>
<td>.18 ± .24</td>
<td>Na</td>
<td>.540</td>
</tr>
<tr>
<td>Hip Extensor</td>
<td>.04 ± .32</td>
<td>.20 ± .44</td>
<td>.37 ± .74</td>
<td>Na</td>
<td>.183</td>
</tr>
<tr>
<td>Hip Flexor</td>
<td>.45 ± .23</td>
<td>.52 ± .42</td>
<td>.89 ± .87</td>
<td>Na</td>
<td>.140</td>
</tr>
</tbody>
</table>

p-value is for one-way ANOVA, * signify where the difference exists, Na = not analyzed

Contrasting the strategies of descent, the joint moment profiles for SOS appear to be greater in magnitude than those associated with the SBS strategy. Examination of the two strategies (lead limb for SBS) indicated a main effect of strategy for the plantarflexors (p=.001), knee extensors (p=.026), and hip flexors (p=.006) which in all cases were larger during SOS. The impact of strategy on the hip moments generated is unique to descent as a similar observation was not present during ascent. This may be a reflection of the greater demand placed on the hip muscles to stabilize the mass of the upper body (head, arms, and trunk) over a small base of support against gravity. Of note is that these differences disappeared when accounting for speed (p>.296). No group or interaction effects were observed.

A similar analysis but involving the SBS trail limb (affected side stroke and non-dominant control) revealed a main effect of strategy only for the peak plantarflexor moment
(p=.006), which disappeared when speed was considered as a covariate (p=.481). No other differences were identified.

The amount of compressive loading transmitted through the handrail was examined to determine if it could explain any of the findings described above. In general, the average compressive forces through the handrail were higher during descent (~26N) than ascent (~18N), with all values ≤38N. The applied force in the stroke group ranged from 19-38N compared to 6-14N in controls, though the difference between groups did not prove to be statistically significant (p>.381). As such, handrail use was not considered a factor in explaining any group effects observed.

3.3.3 Maximum strength assessment

As expected, the control group generally produced higher isometric torques than stroke, significantly so in comparison to the plantarflexors (p=.04), dorsiflexors (p<.01) and knee flexors (p=.03) on the affected side and the dorsiflexors (p=.02) on the less-affected side. Given the heterogeneity in the groups the variances were high limiting the ability to detect differences. The less affected limb produced greater force output that the affected limb at all lower limb joint, but none were statistically significant (p>.229). Data are summarized in Figure 3-3.
3.3.4 Estimation of relative strength cost

The ratio of the peak moment produced during stair ascent (descent) to the corresponding maximum isometric torque provided an estimate of the strength demand of stair negotiation. It is the case that often the relative strength costs exceeded 100% maximum isometric strength likely attributable to the mechanics of the task requirements and the relative orientation of the lower-limb segments. The isometric testing was done in seated or supine positions and open-chain, whereas stair negotiation is dynamic, upright and closed-chain thus it is not surprising that greater torques may be produced during an upright position considering the need for an overall support moment as well as force production (Porter et al., 1996, Porter & Vandervoot, 1997).

In general, the highest strength costs for both groups were associated with the plantarflexors, and these tended to be greatest when subjects used the SOS strategy compared to the SBS during ascent and descent. Considering the cadence during SBS stair negotiation is
approximately half that of SOS paired with the influence of speed on plantarflexor moments (described above), this may have been anticipated.

**Ascent**

For the SBS strategy only knee extension strength ratios differed between limbs (p=.001). Pairwise comparisons showed a higher strength cost associated with the lead limb in stroke (LAFF) compared to the trail limb in control subjects (p=.001). This finding is in part an artifact of the different mechanical roles associated with the lead and trail limbs. A similar explanation applies to the difference detected between the relative strength cost associated with the lead (LAFF) and trail (AFF) knee extensors in stroke (p=.005). Data are illustrated in Figure 3-4.

Contrasting the SOS and SBS lead limbs (LAFF, DOM), a greater relative strength cost of the plantarflexors is observed in SOS (p<.001). There was no evidence of differences between groups (p=.307) or interaction effects (p=.196). No differences were found with respect to the knee or hip (p> .119).

Contrasting SOS and SBS strategies (trail limbs) by group revealed main effects of strategy for the plantarflexors (p=.003) and knee extensors (p<.001) indicating the higher strength costs associated with the SOS strategy. There were however, no group effects (p> .191), or interactions (p> .295). The strength costs associated with the hip flexors and extensors were similar between strategies and groups (p> .490).

**Descent**

There were no differences in the relative strength ratios across limbs during either SBS or SOS stair descent (Figure 3-4). However, when contrasting the SOS and SBS with respect to the lead limbs by group, a main effect of strategy emerged for the plantarflexors and knee extensors (p<.001) reflecting the higher costs associated with SOS. At the hip, there was a trend towards greater strength costs with SOS in the flexors (p=.076) and extensors (p=.081). No group effects
or interactions were observed (p > .124). Repeating the analysis with respect to the SBS trail limb showed higher strength costs associated with the plantarflexors (p < .001) and hip flexors (p = .029) when the SOS strategy was used. No group effects or interactions were observed at any joint (p > .095).

Figure 3-4 Mean relative strength cost of the plantarflexors (PF), knee and hip extensors (Ext) and hip flexors (Flex) associated with stair ascent (top) and descent (bottom) for step-by-step (SBS) and step-over-step (SOS) strategies. SBS lead limbs are represented by control dominant (white) and stroke less affected (light gray) limbs; SBS trail limbs are represented by control non-dominant (striped) and stroke affected (dark gray) limbs. Error bars reflect one SD.

*p < .05 between-limb differences for each strategy.

### 3.3.5 Aerobic assessment of stair negotiation

Stroke and control groups had similar resting metabolic VO$_2$ and HR (p > .349). The peak rate of oxygen consumption (Peak VO$_2$) and heart rate (Peak HR) were comparable between groups (p > .313), see Table 3-4. However, there was a main effect of strategy for peak VO$_2$ in
stair ascent (p<.001) reflecting a greater peak value using the SBS strategy, and for peak HR in stair descent (p=.014) reflecting a greater peak value using the SOS strategy. Oxygen (O$_2$) consumption for the duration of the task performance was greater using the SBS strategy for both ascent and descent (p<.001). The lack of a group effect during ascent (p=.174) indicated that overall, the groups consumed similar amounts of oxygen. The lack of between-group difference could be the result of the high variability in the oxygen consumption during SBS ascent within the stroke group. A different pattern was observed for descent. A main effect of group and strategy (p<.001) and an interaction effect (p=.017) indicated that the stroke group was more strongly influenced by the strategy used.
Table 3-4 Average resting and average peak heart rate (beats per minute, bpm) and oxygen uptake (mL/kg/min) and oxygen consumption (mL/kg); for step-by-step (SBS) and step-over-step (SOS) strategies in stroke and control groups

<table>
<thead>
<tr>
<th></th>
<th>Stroke</th>
<th>Control</th>
<th>Independent T-test</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Resting</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>HR</td>
<td>67.7 ± 10.9</td>
<td>66.4 ± 7.4</td>
<td>.745</td>
</tr>
<tr>
<td>VO₂</td>
<td>2.5 ± 0.8</td>
<td>2.8 ± 0.4</td>
<td>.349</td>
</tr>
<tr>
<td><strong>Est. Max</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>VO₂</td>
<td>25.6 ± 4.1</td>
<td>30.2 ± 5.7</td>
<td>.093</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th>Group</th>
<th>Strategy</th>
<th>Interaction</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Peak HR</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ascent</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SBS</td>
<td>93.1 ± 11.6</td>
<td>92.8 ± 12.6</td>
<td>.884</td>
<td>.684</td>
<td>.504</td>
</tr>
<tr>
<td>SOS</td>
<td>93.1 ± 12.9</td>
<td>93.2 ± 14.7</td>
<td>.884</td>
<td>.684</td>
<td>.504</td>
</tr>
<tr>
<td>Descent</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SBS</td>
<td>86.1 ± 8.8</td>
<td>81.5 ± 9.0</td>
<td>.313</td>
<td>.014</td>
<td>.991</td>
</tr>
<tr>
<td>SOS</td>
<td>90.6 ± 12.9</td>
<td>86.1 ± 13.8</td>
<td>.313</td>
<td>.014</td>
<td>.991</td>
</tr>
<tr>
<td><strong>Peak VO₂</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ascent</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SBS</td>
<td>11.6 ± 2.6</td>
<td>10.9 ± 2.5</td>
<td>.818</td>
<td>&lt;.001</td>
<td>.575</td>
</tr>
<tr>
<td>SOS</td>
<td>8.6 ± 2.0</td>
<td>8.6 ± 1.5</td>
<td>.818</td>
<td>&lt;.001</td>
<td>.575</td>
</tr>
<tr>
<td>Descent</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SBS</td>
<td>6.7 ± 1.4</td>
<td>7.8 ± 2.1</td>
<td>.402</td>
<td>.717</td>
<td>.223</td>
</tr>
<tr>
<td>SOS</td>
<td>7.4 ± 2.7</td>
<td>7.5 ± 1.3</td>
<td>.402</td>
<td>.717</td>
<td>.223</td>
</tr>
<tr>
<td><strong>O₂ consumption</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ascent</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SBS</td>
<td>3.1 ± 1.77</td>
<td>2.2 ± 0.6</td>
<td>.174</td>
<td>&lt;.001</td>
<td>.685</td>
</tr>
<tr>
<td>SOS</td>
<td>1.4 ± 0.4</td>
<td>1.1 ± 0.6</td>
<td>.174</td>
<td>&lt;.001</td>
<td>.685</td>
</tr>
<tr>
<td>Descent</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SBS</td>
<td>2.7 ± 0.6</td>
<td>1.7 ± 0.4</td>
<td>.001</td>
<td>&lt;.001</td>
<td>.017</td>
</tr>
<tr>
<td>SOS</td>
<td>1.4 ± 0.6</td>
<td>0.9 ± 0.3</td>
<td>.001</td>
<td>&lt;.001</td>
<td>.017</td>
</tr>
</tbody>
</table>

HR=heart rate, beats per minute; peak VO₂=peak oxygen consumption, mL/kg/min; Est. Max=estimated maximum; O₂ consumption=mean oxygen consumed over the trial multiplied by the time to complete the trial (mL/kg). P-value is significant at <.05

3.3.6 Estimated maximum aerobic capacity

All control subjects completed the submaximal ergometer test, whereas only 6 individuals with stroke were able to complete the test; one of whom was on beta-blocker medication. The stroke participants who were unable to complete the test could not maintain the pedaling rate (n=2), could not pedal at a higher workload (n=1), experienced lower limb pain (n=1), or terminated the test prior to reaching the target HR (n=3). The estimation of VO₂max in these seven subjects could not be determined. Comparing the estimated maximal oxygen uptake in those who completed the submax test (control group, n=13 and stroke group, n=6) revealed a
trend toward higher VO$_{2\text{max}}$ in control subjects (p=.093), recognizing that the power to detect group differences was low.

3.3.7 Estimated relative aerobic cost

The ratio of oxygen consumption over time during stair negotiation to the estimated maximum oxygen consumption reflected the aerobic challenge of the two stepping strategies for stroke and control participants (see Figure 3-5). For ascent there was a main effect of strategy (p<.001) with the cost of SBS almost twice that of SOS, however no group effect was observed (p=.123) suggesting the aerobic cost is similar between groups. For descent however, there was a main effect for strategy (p<.001), group (p=.002) as well as an interaction effect (p=.001) suggesting a higher cost for SBS, and in stroke, and a disproportionately high cost for the stroke group using the SBS strategy. The group effect was in part driven by the lower estimated maximum aerobic capacity in stroke compared to controls.

![Figure 3-5 Mean relative aerobic cost associated with stair ascent (A) and descent (D) for step-by-step (SBS) and step-over-step (SOS) strategies in stroke (n=6) and control (n=13) participants. Error bars reflect one SD. * reflects significance at p<.05.](image)
3.4 Discussion

The main findings of this study are that the SBS stair-walking strategy is associated with lower joint moments and strength costs than SOS, but cadence is much lower. This was true for both groups although cadence in stroke for any strategy used was nearly half of that associated with control subjects, explaining the lower magnitudes of joint moments of the major muscle groups (i.e. plantarflexors and knee extensors). The consequence of taking twice as many steps to cover the same distance or perform the same amount work using the SBS strategy over the SOS approach however, imposes considerably greater demands on the cardiovascular system as reflected by the substantially higher oxygen consumption, particularly in stroke for descent compared to SOS. Of note is that despite the speed differential between stroke and control subjects, there was no relative economy in terms of either strength or aerobic costs in stroke; as such those with stroke exerted greater effort to move more slowly. Considering that the stroke group in this study was relatively high functioning based on walking speed (0.88 m/s) and the TUG test (14 s) (Hill et al., 1997, Duncan et al., 1998, Eng et al., 2002, Green et al., 2002, Pohl et al., 2002, Ada et al., 2003, Jonsdottir et al., 2007, Hill et al., 2012), these findings suggest that those more severely affected might have much more serious limitations to their mobility.

3.4.1 Biomechanical assessment of stair negotiation

Ascent

In walking, individuals with stroke typically reduce gait speed to compensate for muscle weakness. Similarly, during stair negotiation stroke participants adopt a slower cadence compared to healthy control participants, which effectively reduces the muscle forces required as demonstrated by others (Novak & Brouwer, 2012a). If this were exclusively the case, then controlling for speed would eliminate any between-group differences in peak joint moments. This was the case for the plantarflexors demonstrating their major role in power production (McFadyen & Winter 1988, Protopapadaki et al., 2007); however, the knee extensor output, a
major contributor to the work of stair ascent (Costigan et al., 2002; Novak & Brouwer, 2011) was comparable to that of control subjects despite the fact that they are moving at about half the speed. The same is true for the hip extensors. These findings indicate that individuals with stroke produced comparable joint moments as their healthy counterparts, but would need to do so for extended periods of time to accomplish the same task which could limit their overall capacity for mobility. This is particularly salient when one considers the general muscle weakness, particularly on the affected side.

The knee extensors contribute in a significant way to the total support moment in stair ascent (Novak & Brouwer, 2011). Using a SBS strategy effectively offloads the knee extensors associated with the trail limb reducing the moment magnitude to less than 33% of that generated by the lead limb. The different biomechanical requirements of lead and trail limbs explains this observation since the trail limb knee extensors are not involved in elevating the body’s centre of mass. In the case of hemiparetic stroke, the restricted functional requirement of the trail limb, which would be the affected side, has obvious benefits. In contrast, the plantarflexors of the trail limb produce approximately twice the output as the lead limb in the SBS strategy, but is comparable to that associated with the end of the SOS stance phase. Considering their role in propelling the limb upward to the next step in both strategies this is not unexpected. Overall, the findings of this study indicate that the relative strength cost is substantially lower with SBS than SOS, which leaves reserve capacity in the event of an unexpected perturbation or instability (Rantanen, 2002 (review)).

The hip extensor moments were similar for stroke and control groups and no differences were detected as a function of the strategy used. Novak and Brouwer (2012a) showed that of the lower limb joints, the hip extensor contribution to the total support moment (i.e. the extensor moment maintaining an upright position (Winter, 1980) was comparatively small. This study supports this view. The hip is however, important in controlling the relative position of the upper
body (head, arms and trunk) to the base of support to maintain stability (McFadyen & Winter, 1988, Nadeau et al., 2003). The similarity across strategies may reflect that in terms of stability, the two techniques are comparable. Whether or not this would be the case in the presence of moderate to severe stroke however, is unknown.

Although the absolute extensor output associated with generating the work associated with stair ascent is important, the relative cost (i.e. as a percentage of maximum capacity) is informative in terms of establishing the extent of the residual capacity (Novak & Brouwer, 2012b). It is the latter that enables increases in speed as well as recovery from unexpected perturbations (Pavol et al., 2002). Previous work has shown that the greatest relative strength cost during SOS stair ascent is attributed to the plantarflexors for both stroke and control groups followed by the knee extensors (Novak & Brouwer, 2012b) This study extends these finding to demonstrate substantial cost reduction when the SBS strategy is adopted rather than SOS. For the lead limb the cost for the plantarflexors is reduced during SBS compared to SOS; from 121% to 71% in stroke and from 178% to 87% in controls; however for the knee extensors the cost remains similar between strategies for both groups (90% to 91% in stroke, 61% to 67% in controls) reflecting the significant role of the knee extensors for performing the work of stair climbing. Reductions are seen for the trail limb in both groups with respect to the knee extensors (61% to 23% in controls and 76% to 31% in stroke), however the plantarflexors exert comparable output to that observed during SOS ascent.

It warrants explanation as to why the relative strength costs sometimes exceeded 100%. Maximum strength was determined isometrically in a seated or supine position and open chain. Dynamically, in a closed chain configuration in which the individual is standing upright, higher torques are generated (Porter et al., 1996, Porter & Vandervoot, 1997, Novak & Brouwer, 2012b). As such, estimated costs of a dynamic task can exceed 100%. The validity of comparing relative costs across groups and conditions is not compromised since the method of cost
estimation was standardized across all subjects and there is no reason to suspect that one groups would be differentially affected than another.

**Descent**

Stair descent is an inherently unstable activity. Healthy adults have enough eccentric muscle strength to control the downward movement of the COM, even with a high cadence; however in the presence of weakness, as in stroke, individuals slow down to promote stability. The lead limb accepts the weight of the body while the trail limb provides stability during lead limb swing and loading. In descent, no group differences were observed in peak joint moments with respect to the SBS lead limb despite differences in cadence across groups. For the trail limb, the control group produced significantly greater peak joint moments than the stroke group, though it warrants mention that the affected limb served as the trail. For example, the control trail limb knee extensors produced a peak moment double that of the affected side; however speed was also a factor. Considering the important role of the trail limb in stabilization during descent, people with hemiparetic stroke would normally lead with their affected limb such that the stronger limb accomplishes the higher amount of negative work; this was not the case in the current study.

During SBS and SOS descent, the relative strength costs for the plantarflexors are exceedingly high. However, in SBS, the difference in relative strength cost between the trail and lead limbs appears to be greater in the stroke group compared to healthy controls (133% to 69% and 127% to 100% respectively), likely a reflection of weakness in affected plantarflexors (trail limb). Groups did not differ significantly in terms of relative strength cost for either strategy, indicating that the slower cadence adopted by stroke may serve to effectively keep the strength costs within a normal range.

The higher strength costs observed for the plantarflexors and knee extensors using the SOS strategy relative to the SBS strategy offers evidence as to why individuals with stroke may voluntarily adopt the SBS strategy. When paresis is asymmetrical, as is typically the case in
hemispheric stroke, the uneven distribution of work between lead and trail limbs allows the opportunity for compensation by leading with the stronger limb in ascent, and the weaker limb in descent. In the current study, this was not an option as the non-paretic limb led in both ascent and descent.

3.4.2 Aerobic assessment of stair negotiation

In overground walking, an elevated energy cost of 1.5 to 2 times that of healthy adults has been reported for hemiparetic gait (Macko et al., 1997, Macko et al., 2001, Ivey et al., 2005, Pang et al., 2008). Similarly, a recent study on stair negotiation (SOS pattern only) demonstrated that the aerobic cost (ratio of peak VO\(_2\) to VO\(_{2\text{max}}\)) was significantly higher in individuals with chronic stroke (Novak & Brouwer, 2012b). Data from the current study confirm these findings for descent only and in addition, demonstrate an increased aerobic cost associated with the SBS method for both ascent and descent. The increased oxygen consumption (mL/kg) and relative aerobic cost observed in stroke compared to healthy controls during descent, may be explained in part by their slower cadence which necessarily means that they are performing the task over a longer period of time; however a slower cadence was also adopted for ascent. Therefore, it may be the case that stroke adopt an altered movement strategy during descent contributing to an increased O\(_2\) consumption. It could be a consequence of co-activation of agonist and antagonist muscle groups, which could increase the oxygen demand without increasing the net moment. The effect would be increased stability, which is limited during descent, particularly in the presence of weakness. Further testing including the recording of muscle activation using electromyography would be required to confirm this hypothesis.

Approximately 75% of stroke survivors have cardiovascular disease (Gordon et al., 2004, Roth, 1993) which translates to reduced cardiorespiratory fitness. Though the estimated VO\(_{2\text{max}}\) of our stroke participants (25.5 mL/kg/min) is higher in comparison to previous reports in the literature (14.0 to 23.2 ml/kg/min) (Teh & Aziz, 2001, Kelly et al., 2003, Eng, 2004, Tang et al.,
2006, Pang et al., 2008), only 6 participants were capable of completing the submaximal testing to calculate this value, so it is likely the more fit individuals made up this group. Typically, persons with chronic stroke are physically inactive (Michael et al., 2005) and de-conditioned, which has been demonstrated through reductions in maximum oxygen uptake (Kelly et al., 2003, Eng, 2004, Tang et al., 2006, Pang et al., 2008). Difficulty with conducting maximal and submaximal testing has also been reported in the literature with stroke participants not meeting the minimum criteria used to establish that VO$_{2\text{max}}$ has been achieved (increase in oxygen consumption < 150 mL in the final minute of exercise, RER > 1.15, systolic blood pressure > 200 mm Hg, peak HR within 15 beats per minute of predicted maximal heart rate) (Howley et al., 1995, Mackay-Lyons & Howlett, 2005). Submaximal testing was selected for the current study, as previous reports indicate that maximal tests are terminated prematurely by individuals with stroke (Mackay-Lyons & Howlett, 2005). Our participants who did not complete the submaximal test (n=7) either reached their perceived maximum exertion or stopped prior to the end of the test (6 minutes); therefore too few measures were collected or inadequate workload achieved to make an appropriate estimation of VO$_{2\text{max}}$. In addition, submaximal exercise testing generally overestimates maximal oxygen consumption (Hartung et al., 1995, Kelly et al., 2003), though they have demonstrated good reliability and concurrent validity with those determined from maximal testing in persons with chronic stroke (Eng, 2004).

The reduced cardiovascular fitness in stroke evidenced by the trend of diminished VO$_{2\text{max}}$ paired with the elevated O$_2$ requirements of the task result in a relatively high aerobic cost compared to healthy control subjects. This was particularly evident during descent, likely for the reasons described above.

The SBS strategy was associated with higher O$_2$ costs than SOS. In stroke, the SBS strategy enables conservation of strength as the muscle output is lower, and for some it may be the only way that they can accomplish the task given their muscle weakness. The elevated aerobic
cost that accompanies the SBS strategy however, is a concern in stroke considering their reduced cardiovascular fitness (Potempa et al., 1995, Macko et al., 1997, Kelly et al., 2003). Improving the functional capacity in stroke survivors can reduce the need for compensatory strategies (e.g. SBS strategy), improve performance and lower strength and aerobic costs (Potempa et al., 1995, Rimmer & Wang, 2005), ultimately leading to increased independence and community participation. The results of this study not only support the need for continued exercise and aerobic training in chronic stroke (van de Port et al., 2006), but also underscore the importance of investigating multiple aspects of movement and compensations adopted by people with stroke. Adopting an alternate movement strategy can reduce the challenge in one system but tax another. These findings reinforce the need to monitor multiple systems when evaluating movement.
3.5 References


Chapter 4

Mechanical energy expenditure and efficiency of different stair-stepping strategies in healthy aging and stroke

4.1 Introduction

As muscles perform work to produce movement, they generate (via concentric contraction) or absorb (via eccentric contraction) mechanical energy which, in turn, can be transferred across the joint(s) that they span. This active, inter-segmental energy transfer between adjoining segments can compensate the work done by muscles when the segments rotate in the same direction (Robertson & Winter, 1980, Aleshinsky, 1986a-e, Zajac et al., 2002). For example, the terminal stance phase of walking on level ground involves a great deal of inter-segmental energy transfer as the energy or momentum is transferred distally assisting the ankle extensors in propelling the leg through the swing phase. This power flow of energy across segments serves to economize on the work required by the muscles by harnessing the energy of motion and consequently optimizing energy efficiency during ambulation (i.e. mechanical efficiency). Examining the energy and efficiency of movement can be important in appreciating the challenge of mobility, particularly when mobility is compromised due to physical limitations.

One way to investigate the efficiency of a particular mobility task or different movement strategies is through measuring mechanical energy expenditure (MEE) derived from analyzing the net joint moments. Essentially, the MEE at a given joint is the net amount of energy produced by the muscles to control movement.

Healthy aging is associated with a gradual decline in muscle function due to loss in muscle mass and reduced numbers of muscle fibres (Doherty, 2001 (review)). As a result, the mechanical characteristics of muscle are affected in terms of contraction time and peak force production. In stroke, a disease disproportionately affecting adults over the age of 65 years, the
superimposition of muscle weakness secondary to stroke on age-related loss of strength is a significant concern for mobility, independence and safety. Any associated loss in movement coordination, altered movement patterns and/or slowing could reasonably be expected to impact the mechanical efficiency in the mobility of stroke survivors.

The degree to which muscle weakness is manifest in physical performance depends on the relative demands of the task or activity. One of the earliest mobility compensations observed in older adults and stroke survivors alike secondary to weakness is reduced speed or cadence. Moving more slowly reduces the magnitudes of the joint moments and therefore the muscle work required which in turn lowers the energy expended providing, for example, inter-segmental coordination is not compromised (McGibbon et al., 2001a, Novak et al., 2011). Whether the energy can compensate muscle work to the same degree is less certain. In terms of metabolic efficiency, the U-shaped relationship between gait speed and oxygen consumption clearly indicates that there exists an optimum speed (usually natural speed) at which the oxygen demands are lowest (Frederickson et al., 2007). In stroke, the mechanical inefficiencies of hemiparetic gait are associated with energy costs up to twice that of healthy counterparts (Potempo et al., 1995, Macko et al., 1997, Roth et al., 2000). Speed is a likely contributor, though co-contraction of agonist and antagonist muscle groups as a means of enhancing stability when balance is compromised could also serve to elevate the oxygen demand whilst also reducing the net moments produced.

Stair negotiation is more physically demanding than overground walking, but is similarly considered essential for community ambulation and independence. The higher joint moments, aerobic requirements, and elevated instability (particularly during descent) (Andriacchi, et al., 1980, McFadyen & Winter, 1988, Basset et al., 1997, Startzell et al., 2000, Teh & Aziz, 2001, Costigan et al., 2002, Riener et al., 2002, Nadeau et al., 2003, Protopapadaki et al., 2007) contribute to it being described as one of the most challenging mobility tasks post-stroke (Tsuji et
Indeed many stroke survivors cannot meet the demands of step-over-step (SOS) stair ascent and descent (i.e. reciprocal stepping such that one foot lands on each step) and instead, must adopt a step-by-step (SBS) pattern where both feet land on each step in order to accomplish the task. This adaptation compensates for unilateral weakness and instability (Chapter 3, Reid et al., 2007), however the impact on energy expenditure (mechanical and metabolic) is unknown.

The primary purpose of this paper is to examine the mechanical energy expenditures associated with different stair walking strategies (SBS & SOS) in persons with chronic stroke compared to healthy older adults. Secondarily, the relationship between mechanical and metabolic costs is explored.

### 4.2 Methodology

A secondary analysis was performed on the data obtained from a group of 13 chronic stroke and 13 age- and sex-matched control participants. Participant demographics and the experimental protocol are described in the previous Chapter. Briefly, all participants reported themselves to be in good health, able to ascend and descend a flight of stairs independently and had no non-stroke related mobility impairments or other serious conditions. The study protocol was approved by the university ethics board and all participants provided informed consent.

Three-dimensional kinematic and kinetic data were acquired using optoelectic cameras (Optotrak, Northern Digital Inc., Waterloo, ON) tracking infra-red emitting diodes secured to the lower limbs of the participants as they ascended and descended a custom 4-step staircase with a force plate (AMTI) replacing the centre of the second step. Participants adopted a self-selected pace while using the step-by-step (SBS) and step-over-step (SOS) methods as they were able (see Chapter 3). For the SBS condition, participants always led with their dominant limb (less-affected in stroke). Based on a seven-segment model, joint kinematics and net internal moments were computed using an inverse dynamics approach and commercial software (Visual 3D, C-Motion Inc., Germantown, MD).
For the purposes of this study, 'mechanical efficiency' refers to the efficiency of the task performance based on movement patterns and aerobic and mechanical energy costs. To determine mechanical energy transfer between segments it is necessary to consider the kinematic and kinetic conditions and power flow. When segments rotate in the same direction, power generated (concentric contraction) or absorbed (eccentric contraction) flows from one segment to another. When segments rotate in the opposite direction, no transfer occurs, and energy must be entirely absorbed (eccentric) or generated (concentric) by muscles; that is, muscles are not compensated by segmental energy of motion.

A joint power greater than zero (P_j>0) indicates concentric muscle activity, while a negative joint power represents eccentric muscle activity (P_j<0). The net joint power (W/kg) is calculated by summing the power at the distal (d) and proximal (p) endpoints of the adjoining segments (Equation 1).

\[ P_j = P_d + P_p \] (W/kg)

The power is the product of the joint moment (M_j), where \( M_j = M_p - M_d \) (Nm/kg) and the segment’s angular velocity (\( \omega \)) in rads/s (Equation 2). The net joint power was calculated for each participant during each stair-stepping condition.

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

To evaluate the energy expenditures of the system, mechanical energy expenditures (MEE) were calculated using the method described by McGibbon and colleagues (McGibbon et al., 2001a, McGibbon et al., 2001b, McGibbon & Krebs, 2001). The joint MEE is calculated as the definite integral on the x-axis of the function between two points (time in concentric transfer, for example), essentially, the area under the curve (McGibbon et al., 2001b). Six conditions reflecting all combinations of the direction of power flow (distal, proximal, and no transfer) and energy absorption or generation are identified (see Appendix E). Subsequently, MEE is quantified.
for concentric transfer conditions (MEEc), eccentric transfer conditions (MEEe), and no transfer conditions (MEEn).

4.2.1 Statistical analyses

ANOVAs were used to determine differences between control and stroke for a given strategy. One-way ANOVAs were used to determine differences between limbs for SOS (stroke affected, less affected and control limbs) and SBS (stroke affected, less affected, and control dominant and non-dominant limbs). Post-hoc pairwise comparisons were implemented as appropriate. Within-group differences (to contrast SBS and SOS strategies) were then identified using 2 x 2 ANOVAs (control/stroke and SOS/SBS) for both the lead and trail limbs. For all analyses in which significant group or interaction effects were found, a covariate relating to speed was incorporated into the model since speed can impact the mechanical efficiency of movement. These analyses were carried out in relation to ascent and descent separately and a significance level of p<0.05 was used for all analyses.

4.3 Results

As reported previously (Chapter 3), people with stroke (regardless of strategy) manage stairs more slowly than their healthy counterparts and for both groups, the SOS strategy is associated with about twice the cadence as the SBS strategy. This is relevant when considering energy requirements.

4.3.1 Comparison of group and strategy differences in mechanical energy transfer

Ascent

The stance phase of stair ascent begins with weight acceptance (~0-26% stance), the initial movement of the body’s centre of mass (CoM) forward over the lead limb’s base of support. The pull-up phase (~26-57% stance) is the main progression of elevating the CoM as the body weight is transferred to the lead limb on the subsequent step. The final stage, forward
continuance (~57-100% stance), the phase in which completion of step ascent occurs and the trail limb overtakes the support limb to progress to the next step (SOS) or meets the support limb on the same step (SBS) (McFadyen & Winter, 1988, Zachazewski et al., 1993).

Using the SBS strategy, the lead limb power profiles (less affected side in stroke, (LAFF), dominant side in controls, (DOM)) and trail limb power profiles (affected side in stroke, (AFF), non-dominant side in control participants, (NDOM)) are similar in shape (see Figure 4-1). The lead limb knee extensors are the major power generators throughout the pull-up phase.

Forward continuance is primarily achieved by power generated by the plantarflexors of the trail limb as they propel the trail limb upward to meet the lead limb. The unique biomechanical roles of the lead and trail limbs are reflected by the significant interlimb differences in mechanical energy generated by the knee extensors and plantarflexors (MEE_N & MEE_C, p≤0.021; see Table 4-1). In the case of the plantarflexors; however, the MEE_C generated by the trail limb in stroke is less than 40% of the magnitude produced by the control group.

During pull-up, the lead limb’s plantarflexors transfer energy proximally through concentric activity and the hip extensors transfer energy distally combining to assist the knee extensors in raising the weight of the body against gravity. The power profiles (Figure 4-1) suggest that the less affected hip may have a greater role in stroke, although post-hoc analysis revealed no difference compared to the lead limb in controls (Table 4-1).
Figure 4-1 Step-by-step ascent power profiles at the ankle (top), knee (middle) and hip (bottom) for the lead limb (left panels) and trail limb (right panels) associated with each group. The shaded bars represent the type of mechanical energy transfer; black=no transfer, dark gray=concentric, and light gray=eccentric. The arrows indicate the direction of energy transfer, upwards=proximal, downwards=distal, outward arrows=concentric no transfer, inward arrows=eccentric no transfer.
Table 4-1 Mechanical energy expenditures at the ankle, knee and hip for lead and trail limbs associated with control (dominant (DOM); non-dominant (NDOM) limbs) and stroke groups (less affected (LAFF); affected (AFF) limbs) during step-by-step and the lead limbs for step-over-step ascent

<table>
<thead>
<tr>
<th>100*J/kg</th>
<th>LAFF (lead)</th>
<th>AFF (trail)</th>
<th>DOM (lead)</th>
<th>NDOM (trail)</th>
<th>P-value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>SBS</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Ankle</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>3.79 ± 2.28†</td>
<td>4.65 ± 4.90*</td>
<td>5.74 ± 3.72‡</td>
<td>13.20 ± 7.77*†‡</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>2.29 ± 0.79</td>
<td>3.51 ± 1.52*</td>
<td>1.90 ± 0.94*†</td>
<td>3.44 ± 1.88†</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>6.05 ± 4.00</td>
<td>6.85 ± 4.98</td>
<td>5.12 ± 3.93*</td>
<td>11.89 ± 8.72*</td>
</tr>
<tr>
<td><strong>Knee</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>3.37 ± 2.62*†</td>
<td>1.04 ± 0.71*</td>
<td>2.57 ± 1.20†</td>
<td>0.88 ± 0.39†‡</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>0.34 ± 0.19†</td>
<td>0.73 ± 0.44*†</td>
<td>0.30 ± 0.37*</td>
<td>0.54 ± 0.28</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>53.91 ± 21.67*‡</td>
<td>10.84 ± 17.20*†</td>
<td>51.44 ± 19.12†^</td>
<td>6.70 ± 3.03‡^</td>
</tr>
<tr>
<td><strong>Hip</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>7.82 ± 8.70*†</td>
<td>1.55 ± 2.66*</td>
<td>4.51 ± 4.21</td>
<td>0.92 ± 0.83†</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>0.76 ± 0.73</td>
<td>0.81 ± 0.53</td>
<td>1.35 ± 2.29</td>
<td>0.75 ± 0.76</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>9.02 ± 7.73</td>
<td>6.22 ± 4.53</td>
<td>5.19 ± 2.83</td>
<td>5.07 ± 2.68</td>
</tr>
<tr>
<td><strong>SOS</strong></td>
<td></td>
<td></td>
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<td></td>
</tr>
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<td><strong>Ankle</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>16.53 ± 7.39</td>
<td>9.67 ± 7.29*</td>
<td>22.12 ± 10.54*</td>
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<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>3.88 ± 2.48</td>
<td>5.28 ± 2.66</td>
<td>3.68 ± 5.73</td>
<td>na</td>
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<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>17.84 ± 10.82</td>
<td>10.15 ± 15.17</td>
<td>19.27 ± 15.17</td>
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<tr>
<td><strong>Knee</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>5.49 ± 1.71</td>
<td>4.69 ± 2.83</td>
<td>4.40 ± 3.25</td>
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<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>1.18 ± 0.80</td>
<td>1.26 ± 1.18</td>
<td>1.17 ± 1.72</td>
<td>na</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>54.57 ± 11.11*</td>
<td>37.28 ± 16.44*</td>
<td>38.53 ± 14.62</td>
<td>na</td>
</tr>
<tr>
<td><strong>Hip</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>11.42 ± 8.70</td>
<td>8.95 ± 5.91</td>
<td>5.74 ± 6.16</td>
<td>na</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>1.01 ± 0.81</td>
<td>1.07 ± 1.17</td>
<td>1.97 ± .279</td>
<td>na</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>13.55 ± 5.67</td>
<td>14.87 ± 14.13</td>
<td>9.75 ± 5.37</td>
<td>na</td>
</tr>
</tbody>
</table>

P-value significant at p<.05; *,†,‡,^ denotes where the differences occur; na=not analyzed

Using the SOS strategy, the reciprocal nature ensures that there is no distinctive functional role of one limb versus the other and in the case of controls, the activity is symmetrical (Novak & Brouwer, 2011); whereas the same is not necessarily the case in hemiparetic stroke.

Weight acceptance (in controls) typically involves energy absorption at the ankle via eccentric plantarflexor activity and generation at the hip; both actions appearing attenuated in stroke (Figure 4-2). As weight is accepted and in the early pull-up phase, substantial energy is
generated at the knee to raise the CoM vertically; the counter rotations of the shank and thigh segments achieve knee extension but preclude energy transfer (see MEE_N, Table 4-1). In contrasting the magnitude of MEE_N as a function of limbs, a significant effect was detected (p=.009) which was attributed to the greater reliance on the knee extensors when the LAFF limb leads than when the AFF limb leads (Table 4-1); no differences from control subjects were noted.

During forward continuance, the ankle plantarflexors transfer energy distally through the foot to assist propulsion of the body upwards. Control subjects expended the greatest amount of energy and the power flow associated with the paretic plantarflexors was lowest, about 50% lower than observed for control subjects (p=.006). Inter-segmental transfers and power generation at the knee are of low amplitude. At the hip, the direction of energy transfer is variable and of relatively low amplitude though important to controlling stability and the position of the upper body (head, arms and trunk) relative to the base of support of the support limb (Figure 4-2).
Figure 4-2 Step-over-step ascent power profiles at the ankle (top), knee (middle), hip (bottom), for the control (left), stroke less affected (centre) and stroke affected (right) limbs. The shaded bars represent the type of mechanical energy transfer, black=no transfer, dark gray=concentric, and light gray=eccentric. The arrows indicated the direction of energy transfer, upwards=proximal, downwards=distal, outward arrows=concentric no transfer, inward arrows=eccentric no transfer.
Comparing the SOS and the SBS (lead limb) strategies of stair ascent revealed a general pattern of greater mechanical energy expenditure associated with the SOS strategy in relation to concentric extensor activity (all joints) (p ≤ .007) and no-transfer conditions at the ankle and hip for both control and stroke groups (p < .001). The slower SBS cadence relative to SOS was a primary factor. Controlling for speed as a covariate eliminated the significance of all main effects of strategy (p ≥ .155), with the exception of the MEEc associated with the plantarflexors (p = .002). The only main effect of group observed was associated with the knee (no-transfer condition) (p = .005), but none for other joints, and there were no interactions between group and strategy (p ≥ .105).

Substituting the trail limb associated with the SBS strategy into the model revealed a similar trend. That is, the MEEs attributed to concentric activity and no-transfer conditions were in all cases of greater magnitude for the SOS strategy than SBS at all joints (p ≤ .017). The main effects disappeared when speed was entered as a covariate. The only group effect observed was for the plantarflexor MEEc condition (p = .001) reflecting the reduced power output in the stroke affected limb compared to controls. The absence of group effects elsewhere (p ≥ .072) or interaction effects (p ≥ .282) suggests that the mechanical energy expenditures were comparable despite the significantly slower speed of the stroke group and the impact of strategy yielded similar effects on both groups.

**Descent**

The stance phase of stair descent begins with the initial foot contact on the lower step and weight acceptance (0-22%) followed by forward continuance which marks the commencement of single leg support (22-52%) and lastly, controlled lowering accounting for the major portion of stance (52-100%) as full weight is borne on the support limb on the lower step (Zachazewski et al, 1993).
In SBS descent, the lead limb (LAFF, DOM) and the trail limb have very different biomechanical roles. The lead limb accepts body weight in early stance while the hip muscles work to stabilize the body’s COM during the weight transfer as evidenced by the fluctuations in power flow of relatively low magnitude. The trail limb (AFF, NDOM) plantarflexors and knee extensors eccentrically performs the majority of the work during forward continuance and controlled lowering the COM before moving into the swing phase and meeting the contralateral limb on the next step (see Figure 4-3). These distinctive roles account for the limb differences observed at all joints (Table 4-2; \( p < .045 \)). The power absorption at the ankle and knee of the trail limb being critical throughout most of stance and in the case of plantarflexors, the power related energy transfer was \( \sim 32-40\% \) higher for the control group than stroke (Table 4-2). At terminal stance, the concentric power generation by the plantarflexors and associated MEE\(_C\) directed distally serves to propel the limb through swing; the control subjects performed this more effectively than those with stroke (\( p < .010 \)). This is likely attributable in part to weakness on the AFF side. The MEEs are similar between control and stroke lead limbs as revealed from the post-hoc analysis (\( p > .422 \)).

During weight acceptance using the SOS strategy, the plantarflexors absorb energy (eccentric, no transfer). A difference across limbs (\( p = .014 \)) indicated this was less pronounced on the AFF side relative to the LAFF, though neither were different from the DOM limb of the control group (Table 4-2). The knee extensors play a major role during forward continuance and controlled lowering (Figure 4-4) absorbing considerable energy in performing negative work. A difference between limbs (\( p = .019 \)) reflected higher than normal energy expenditure associated with the LAFF limb despite the slower cadence. As observed with the SBS strategy, the power profiles show substantial fluctuation at the hip as the muscles work to stabilize the upper body relative to the base of support. During weight acceptance and at the start of single support (early forward continuance) the hip action is assisted by the proximally directed transfer of energy from
Figure 4-3 Step-by-step descent power profiles at the ankle (top), knee (middle) and hip (bottom) for the lead limb (left panels), and for the trail limb (right panels). The shaded bars represent the type of mechanical energy transfer, black=no transfer, dark gray=concentric, and light gray=eccentric. The arrows indicated the direction of energy transfer, upwards=proximal, downwards=distal, outward arrows=concentric no transfer, inward arrows=eccentric no transfer.
<table>
<thead>
<tr>
<th>100*J/kg</th>
<th>LAFF (lead)</th>
<th>AFF (trail)</th>
<th>DOM (lead)</th>
<th>NDOM (trail)</th>
<th>P-value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>SBS</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Ankle</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>3.21 ± 1.67*</td>
<td>6.80 ± 4.10‡</td>
<td>3.58 ± 2.82*</td>
<td>11.51 ± 4.25*‡</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>.500 ± 5.76*</td>
<td>14.67 ± 5.90*‡</td>
<td>4.10 ± 1.27*</td>
<td>21.15 ± 7.22*‡</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>11.39 ± 7.23†</td>
<td>2.89 ± 2.20*‡</td>
<td>11.66 ± 5.61*</td>
<td>6.64 ± 6.46</td>
</tr>
<tr>
<td><strong>Knee</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>1.26 ± 0.74</td>
<td>0.76 ± 1.31</td>
<td>1.28 ± 0.67*</td>
<td>0.25 ± 0.22*</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>2.06 ± 2.25</td>
<td>0.97 ± 1.09</td>
<td>1.44 ± 1.35</td>
<td>3.10 ± 7.48</td>
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<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>22.62 ± 23.65*‡</td>
<td>49.00 ± 26.57†‡</td>
<td>13.61 ± 10.34†^</td>
<td>66.33 ± 15.26*^</td>
</tr>
<tr>
<td><strong>Hip</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>3.89 ± 3.37*</td>
<td>1.43 ± 1.43*</td>
<td>2.10 ± 1.26</td>
<td>3.38 ± 2.09</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>1.92 ± 2.02</td>
<td>1.43 ± 1.80</td>
<td>1.72 ± 1.30</td>
<td>0.73 ± 0.75</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>7.36 ± 7.35</td>
<td>4.30 ± 3.23</td>
<td>5.91 ± 4.70</td>
<td>4.29 ± 2.62</td>
</tr>
<tr>
<td><strong>SOS</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Ankle</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>12.49 ± 4.27</td>
<td>8.42 ± 4.05</td>
<td>9.56 ± 5.01</td>
<td>Na</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>25.48 ± 6.66*</td>
<td>17.20 ± 4.90*</td>
<td>19.13 ± 8.09</td>
<td>Na</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>17.03 ± 9.20</td>
<td>7.64 ± 6.93*</td>
<td>23.09 ± 9.08*</td>
<td>Na</td>
</tr>
<tr>
<td><strong>Knee</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>2.14 ± 1.37</td>
<td>1.91 ± 1.36</td>
<td>1.59 ± 1.34</td>
<td>Na</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>5.46 ± 6.12</td>
<td>7.60 ± 7.68</td>
<td>8.69 ± 9.95</td>
<td>Na</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>88.62 ± 24.31*</td>
<td>58.51 ± 25.46</td>
<td>56.79 ± 35.70*</td>
<td>Na</td>
</tr>
<tr>
<td><strong>Hip</strong></td>
<td>MEE&lt;sub&gt;C&lt;/sub&gt;</td>
<td>3.66 ± 2.93</td>
<td>1.90 ± 0.76</td>
<td>3.89 ± 1.77</td>
<td>Na</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;E&lt;/sub&gt;</td>
<td>1.48 ± 0.63</td>
<td>1.76 ± 1.37</td>
<td>1.07 ± 1.04</td>
<td>Na</td>
</tr>
<tr>
<td></td>
<td>MEE&lt;sub&gt;N&lt;/sub&gt;</td>
<td>8.71 ± 5.55</td>
<td>3.92 ± 2.66</td>
<td>8.62 ± 7.59</td>
<td>Na</td>
</tr>
</tbody>
</table>

P-value significant at p < .05; *,†,‡,^ denotes where the differences occur, Na=not analyzed
Figure 4-4 Step-over-step descent power profiles at the ankle (top), knee (middle) and hip (bottom) for control (left), stroke less affected (centre) and stroke affected (right) limbs. The shaded bars represent the type of mechanical energy transfer; black=no transfer, dark gray=concentric, and light gray=eccentric. The arrows indicate the direction of energy transfer, upwards=proximal, downwards=distal, outward arrows=concentric no transfer, inward arrows=eccentric no transfer.
the knee (Figure 4-4).

Comparing the SBS (lead limb) and the SOS (LAFF and dominant side) strategies during descent resulted in similar findings as reported for ascent. Specifically, the energy expenditures associated with ankle plantarflexors (concentric and eccentric activation) and knee extensors working eccentrically were higher when using the SOS than the SBS strategy ($p \leq 0.005$). For the plantarflexors, these findings were not altered by controlling for speed ($p \leq 0.032$), however the difference disappeared at the knee ($p \geq 1.16$). No group effects ($p \geq 0.077$) nor interactions of group by strategy were observed ($p \geq 0.167$) with the exception of the energy absorption associated with the knee extensors (no transfer condition), ($p=0.003$ and $p=0.028$, respectively). This reflects the much greater difference in $\text{MEE}_N$ between SOS and SBS in the stroke group than the control group, possibly the result of attempting to stabilize and control descent with the LAFF limb, resulting in enhanced asymmetry between the less affected and affected sides. There were no significant findings associated with the $\text{MEEs}$ calculated at the hip ($p \geq 0.092$) implying that their role was consistent regardless of strategy used (mainly stabilization) and both groups expended similar amounts of energy even in the presence of marked differences in cadence (between group and strategy).

Examining the trail limb associated with the SBS strategy (AFF, NDOM) and the AFF/DOM limbs related to the SOS strategy revealed a different pattern than that described above. Considering the different functions of the lead and trail limbs in SBS descent, this may not be surprising. No differences related to strategy were detected in the distally directed energy transfer via the eccentric plantarflexor activity ($\text{MEE}_E$) while controlling the forward and downward displacement of the COM ($p=0.891$) although the eccentric (no transfer) energy expenditure was higher in relation to the SOS strategy ($p<0.001$). This was attributable to speed, although an interaction effect ($p=0.010$) indicates that the distinction between strategies is less
pronounced in stroke, likely a reflection of plantarflexor weakness (see Chapter 3) and limited capacity to produce greater force output.

The other major power producers, the knee extensors, showed comparable eccentric energy absorption (\(MEE_N\)) across strategies (p=.998), but not so for eccentric inter-segmental transfer (\(MEE_E\)) which was nearly three times higher with the SOS than the SBS strategy (p<.001). This was the case for stroke and control groups alike and was associated with differences in speed across the two strategies. A similar observation was noted for concentric knee extensor activity (p=.003); however, the extent of concentric activity and the magnitude of power generated are extremely small and considered negligible. Indeed, introducing speed as a covariate had no impact on this finding.

No differences in energy expenditure were evident at the hip in association with stair stepping strategy implying a similar role of the hip flexors and extensors in both SOS and SBS and for both groups (p>.055).

4.3.2 Comparison of group and strategy differences in metabolic energy

The metabolic energy requirements (i.e. oxygen consumption) associated with the two stepping strategies are summarized by group and strategy in Table 4-3. With respect to both stair ascent and descent there were main effects of group (p<.033) and strategy (p<0.001) indicating higher metabolic energy demands for those with stroke than control subjects and that the SBS strategy required more energy than the SOS strategy. For stair descent only, there was an interaction effect (p=.014) reflecting that the impact of strategy was more pronounced in the stroke group than the control group.
Table 4-3 Mean oxygen consumption per step during ascent and descent of one flight of stairs (16 steps) using step-by-step and step-over-step strategies in stroke and control participants

<table>
<thead>
<tr>
<th>Strategy</th>
<th>Stroke</th>
<th>Control</th>
<th>Group effect</th>
<th>Strategy effect</th>
<th>Interaction</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ascent</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SBS</td>
<td>0.20 ± 0.10</td>
<td>0.14 ± 0.04</td>
<td>.033</td>
<td>&lt;.001</td>
<td>.911</td>
</tr>
<tr>
<td>SOS</td>
<td>0.09 ± 0.02</td>
<td>0.06 ± 0.02</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Descent</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SBS</td>
<td>0.17 ± 0.03</td>
<td>0.11 ± 0.02</td>
<td>&lt;.001</td>
<td>&lt;.001</td>
<td>.014</td>
</tr>
<tr>
<td>SOS</td>
<td>0.09 ± 0.04</td>
<td>0.06 ± 0.01</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

VO₂ = oxygen consumption; P-value significant at p < .05

4.4 Discussion

This is the first study to investigate mechanical energy expenditures and oxygen consumption during two stair negotiation strategies in persons with chronic stroke compared to healthy older adults. The main finding of the study is that mechanical expenditures are generally reduced by adopting the SBS strategy, however aerobic cost is increased. Stroke and control groups usually had similar mechanical expenditures for a given strategy even though those with stroke negotiated the stairs at about half the cadence compared to controls. Exceptions to this were expenditures for the lead limb knee extensors during no transfer conditions (MEEₙ) which were higher in the stroke group during both ascent and descent, and trail limb plantarflexors during concentric transfer conditions (MEEₖ) during ascent, where expenditures were greater in the control group. The aerobic cost; however, was in all cases higher for the stroke group. This is an important consideration for rehabilitation professionals.

The ability to manage stairs is one of the single best predictors of physical activity levels in the community (Alzahrani et al., 2009). To date, only few studies have investigated this aspect of mobility in persons with chronic stroke (Alzahrani et al., 2009, Novak & Brouwer, 2012a,
Novak & Brouwer, 2012b, Pinheiro et al., 2013). In level ground walking, mechanical energy expenditures during transfer conditions (MEE<sub>c</sub> & MEE<sub>e</sub>) at the ankle, knee and hip of healthy adults range between 0.02 – 0.38 J/kg (McGibbon et al., 2001a). In the current study stair ascent expenditures range between 0.01-0.22 J/kg and during descent expenditures range between 0.01-0.19 J/kg in healthy adults using the SOS strategy. These findings support work done by Novak and colleagues (2011) and demonstrate the limited opportunity to transfer energy via the contracting muscles during stair ambulation. This is best explained by the constraints of the task, where one must move their COM upwards against gravity (ascent) or control vertical displacement of the COM downwards (descent) in addition to progressing horizontally in order to advance to the next step. The task becomes significantly more challenging in the presence of disability, as indicated by the current results. The mechanical energy expenditures associated with stair ascent and descent were frequently higher or similar in stroke compared to controls with respect to the knee and hip indicating that the slowing of speed does not translate into energy conservation. The elevated oxygen cost compared to control subjects suggests that in stroke, inefficient motor patterns increase the physical demands of the task.

**Ascent**

In the presence of lower limb impairment, stair ascent presents as a challenge for persons with stroke. Data from the current study indicate that mechanical expenditures for stair ascent are 3 to 10 times greater at the ankle and knee compared to overground walking in healthy older adults (McGibbon et al., 2001a). Many stroke survivors rely on handrail use or adopt an alternative strategy, such as step-by-step stepping to accomplish the task. The results of the present study show a reduction in the energy expenditure associated with the concentric work of the trail limb or affected plantarflexors during push-off when using the SBS strategy compared to the typical SOS. In persons with stroke, this demonstrates that the SBS strategy effectively reduced the demands of the trail limb plantarflexors, thereby capitalizing on the relative strength...
of the LAFF limb. The need for plantarflexor push-off is minimized further when a slower cadence is adopted. By contrast, in the control trail limb, the plantarflexor expenditure ($\text{MEE}_c$) was higher than the corresponding energy demand in stroke, which may be attributed to their increased cadence.

The elevated work through concentric activation of the knee extensors of the LAFF limb compensates for the reduced propulsive power generated by the plantarflexors on the AFF side to raise the trail limb as it prepares for the swing phase. During SBS stepping, this compensatory strategy allows the LAFF limb to take on a greater proportion of work compared to the trail (AFF) limb, producing a similar expenditure to that of controls, who negotiated the stairs twice as fast. A similar pattern (low plantarflexor output paired with exaggerated knee extensor output on the contralateral side) was also observed for the stroke group using the SOS pattern. During level ground walking, it has been observed that muscle weakness following stroke can result in compensatory gait patterns, where the stronger muscle groups counter deficiencies by generating more than normal work (Olney & Richards, 1996, Olney et al., 1998). In a similar vein, the adoption of the step-by-step strategy forces a redistribution of work to the lead limb which is particularly advantageous in the presence of interlimb strength asymmetry as occurs in hemiparetic stroke. In controls, there is no apparent functional benefit.

The concentric energy expenditure at the hip for the lead limb was greater in stroke compared to the lead limb in controls during SBS. This strategy may have served to better distribute the extensor work across both the knee and the hip; whereas control subjects generated most power from the knee extensors. A similar pattern was also detected during SOS, though at the higher cadence, no differences were detected between LAFF and the DOM limbs. The most salient feature of the MEEs associated with the SBS strategy, was the asymmetry between lead and trail limbs. The disproportionate workload on the lead limb effectively minimizes the output required on AFF trail limb as noted above, but may also provide an elevated sense of safety.
resulting from enhanced stability. Control subjects, on the other hand, would not derive any
physical benefit. In fact, the elevated per step oxygen consumption compared to SOS ascent
underscores the undesirability of the SBS strategy for this group. In stroke, the higher O$_2$ cost
may be the necessary consequence of adopting a strategy that permits greater stability (Reid et
al., 2007) and enables stair mobility.

When contrasting stair ascent strategies, the SBS strategy yields greater aerobic cost and
lower mechanical energy expenditures than ascending using a SOS pattern of stepping. The
elevated oxygen consumption relates to the fact that each limb touches each step thus doubling
the steps taken and increasing the metabolic requirement. Mechanically, there is limited capacity
for intersegmental energy transfer due to the discontinuous SBS gait pattern and the relatively
slow cadence minimizes opportunities to exploit the energy of motion. During level ground
walking, the conservation of energy through intersegmental energy transfer is quite high at
preferred walking speeds (Olney & Richards, 1996), however in stroke, impaired gait patterns
and relative slowness result in marked loss in energy efficiency (Waters et al., 1982, Waters et
al., 1988, Cunha et al., 2002). The reciprocal nature of gait facilitates energy exchange and in
particular, temporal-spatial symmetry between the two limbs is essential for optimal efficiency
(Ellis et al., 2013). Departure from symmetry has been shown to elevate the mechanical costs
upwards of 50% (Ellis et al., 2013). Unlike overground walking, in stair negotiation, the
relatively large vertical displacement compared to horizontal motion requires considerable energy
to be expended in moving against gravity. Superimpose an asymmetrical pattern, i.e. SBS, the
ability to transfer energy is minimized and the energy cost elevated. The reciprocal nature of the
SOS pattern has been shown in the current study to more effectively transfer energy, though a
more detailed analysis would be required to determine the extent to which muscle work is
compensated by such transfer.
Descent

The task of stair descent is dominated by periods of energy absorption in the absence of transfer, with the majority of work done at the knee during the controlled lowering phase. In late stance, energy is transferred distally via concentric contraction of the plantarflexors to assist with push-off. Powers at the hip vary considerably in both groups during descent as the muscles are involved in controlling the mass of the upper body (trunk, head and arms) relative to the base of support during controlled lowering.

In terms of mechanical expenditure, the lead limbs associated with the SBS strategy (LAFF and DOM) did not differ during descent. On the other hand, when contrasting the trail limbs (AFF& NDOM), significant differences in energy expenditure were observed during transfer conditions at the ankle (MEEc & MEEe). Greater expenditures observed in the control group indicate their ability to achieve the needed control through their plantarflexors of the trail limb ankle compared to stroke participants who are less able due to plantarflexor weakness on the affected side.

Similarly using the SOS method, the plantarflexor expenditure in the stroke affected limb was reduced compared to controls. The less affected knee extensors, however, produced the greatest expenditure (energy absorption, MEEN), even when compared to controls who descended the stairs at double the cadence. This disproportionate expenditure attributed to the LAFF knee extensors in stroke demonstrated the relative importance of their role in providing support and enhancing balance during stair descent, as has been shown in previous research (Novak et al., 2011).

When contrasting SOS and SBS lead limbs during descent, MEEs associated with plantarflexor activity (all conditions) were of greater magnitude using the SOS strategy, however, speed was only a factor for eccentric transfer conditions (MEEe). The greater speed may have in fact contributed to enhanced efficiency provided muscle work is compensated. With respect to the
knee extensors, expenditures (MEE_E & MEE_N) were greater during SOS with speed accounting for the difference. An interaction effect during the no transfer condition (MEE_N) suggests that the SOS strategy yielded a greater expenditure compared to SBS, however the difference is more pronounced in the stroke limb. Therefore, during SOS, the LAFF knee extensors are required to perform more work in terms of energy absorption in compensation for abnormally low plantarflexor power from the AFF side.

When contrasting SOS with the SBS trail limb, a main effect of strategy was observed for the plantarflexors (MEE_N), however speed was a factor. Additionally, the presence of an interaction with group, which remained when speed was accounted for, indicated that the difference in expenditure between SBS and SOS in controls is greater than the difference in stroke, perhaps a reflection of a limited range of capacity in stroke.

Regardless of the strategy used during descent, the stroke group consumed more oxygen per step compared to controls. Similarly to ascent, the SBS strategy for descent yielded an aerobic cost nearly double that of SOS. The inability to transfer energy coupled with a reduced cadence is one factor that might explain energy costs in stroke. However, for stair descent only, an interaction effect reflects that the difference in oxygen cost between strategies was more pronounced in the stroke group.

It is evident that speed plays an important role in the mechanical and metabolic efficiency of movement. In healthy adults, the relationship between gait speed and metabolic efficiency is U-shaped, where minimum oxygen costs occur at moderate, comfortable walking speeds (Fredrickson et al., 2007). When speed is reduced, the oxygen cost is increased. In terms of mechanical efficiency, there is evidence that efficiency was maximized at a higher than preferred stride rate (Umberger & Martin, 2007). This indicates also that at slower speeds mechanical efficiency, or the opportunity to transfer energy to compensate muscles, is reduced. It is this fact paired with the longer time it would take to complete the task that most reasonably explains the
increased metabolic demand with slower gait. This was evident in the current study with respect to stair negotiation. In stroke, however, other factors likely also contributed to the higher than normal aerobic costs.

In stroke, the elevated aerobic cost on stairs could be the result of co-contraction of agonist and antagonist muscle groups. Particularly during stair descent, an inherently unstable task (Startzell et al., 2000 (review), Reid et al., 2007, Novak & Brouwer, 2011, Reid et al., 2011), this strategy can be used to increase joint stiffness and provide additional mechanical stability (Levin & Hui-Chan, 1994, Lamontagne et al., 2000, Den Otter et al., 2007). In a study comparing joint stiffness during a stepping down task, older adults demonstrated 50% greater stiffness compared to young adults (Hortobagyi & DeVita, 1999), attributed in part to a stiff- or straight-legged posture, and co-activation of muscles around the knee and ankle. In stroke, where muscle strength and postural balance are limited (Bohannon, 1986, Kim & Eng, 2003), increased stiffness could certainly enhance stability during stair descent, resulting in reduced power absorption but at a greater aerobic cost.

There are several limitations of the current study. Inverse dynamics does not parcel out individual muscle contributions, but rather quantifies the net activity of the muscles spanning a particular joint. As such, the occurrence of co-contraction is unknown, and simply hypothesized. The lower power from the stroke group on the affected side may be due to co-contraction of the musculature (which would be expected to increase the metabolic costs) although the extent to which this occurs in our subjects remains unknown. Future studies should utilize electromyography to investigate specific muscle usage and co-contraction during stair negotiation, and the relationship with metabolic cost. The segmental energy approach used in the current study defines only the conditions under which transfer occurs, and the outcome is energy expenditure. As such, the study does not provide any indication of the proportion of muscle energy compensated through intersegmental transfer, and it is possible that this compensation
may contribute to differences in metabolic cost. Future work should investigate this aspect of the mechanical work related to stair ambulation. Despite the limitations, however, the results from the current research supports the need to increase strength and aerobic reserve in stroke to facilitate the adoption of more energy efficient movement strategies.
4.5 References

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energy expenditure problem – II. Movement of the multi-link chain model. J Biomech 19: 
295-300.

energy expenditure problem – III. Mechanical energy expenditure reduction during one 

energy expenditure problem – IV. Criticism of the concept of „energy transfers within 

energy expenditure problem – V. The mechanical energy expenditure reduction during 

physical activity in community-dwelling people with stroke: an observational study. Aust 
J Physiother 55, 277-281.


Energy cost of stair climbing and descending on the college alumnus questionnaire. Med 


Chapter 5

General Discussion

5.1 Summary of findings

Following stroke, the majority of survivors return to living in the community (Stineman & Granger, 1998) and the ability to ascend and descend stairs is crucial for functional independence and community ambulation (Alzahrani et al., 2009). In the presence of stroke-related impairment, the high physical demands of stair negotiation can pose difficulties in accomplishing the task, leading to the adoption of altered movement patterns or compensatory mechanisms. This study examined the demands associated with the step-by-step (SBS) and the step-over-step (SOS) methods in hemiparetic stroke and healthy age and sex-matched controls.

In Chapter 3 the relative strength and aerobic demands of SBS and SOS strategies were examined. In general, the SBS strategy was associated with lower joint moments and relative costs (expressed as a ratio of maximum isometric strength) compared to the SOS strategy in both groups. During ascent, it was revealed that the SBS strategy effectively offloads the trail limb knee extensors, as the majority of the work to lift the centre of mass (COM) is accomplished by the lead limb. In stroke, the trail limb was their affected (paretic) side, which effectively diminished the relative strength costs compared to SOS. For the affected side plantarflexors though, the strength cost was similar to that associated with SOS given the similar functional role of the plantarflexors in the two conditions; the result is a challenge in stroke since the plantarflexors tend to be amongst the weakest muscle groups post-stroke (Teixeira-Salmela et al., 1999, Eng et al., 2002, Kim & Eng, 2003).

During descent the relative strength costs were lower using the SBS strategy, however the plantarflexor cost is abnormally high for both groups using both strategies. This is attributable to their role in stabilization during controlled lowering of the COM, but considering this was the
affected limb in stroke, this posed a significant challenge. During overground walking, the plantarflexors have a crucial role in push-off, and the power produced is related to gait speed. During stair negotiation, the plantarflexors also have a role in push-off, therefore the high plantarflexor strength cost observed in controls is likely attributable to speed.

The stroke group negotiated the stairs at approximately half the cadence of control participants in all conditions (SBS and SOS, ascent and descent). In stroke gait, slowing down the movement may come as a result of muscle weakness, however it might also be a strategy to reduce net joint moments (i.e. muscular demand) and enhance stability (Novak & Brouwer, 2012a). The same adaptation has been observed in the current study, however, the reduced speed did not result in higher strength economy, suggesting that stroke exert greater effort to move more slowly. In terms of aerobic demands, prolonging the time to complete the movement increases the amount of oxygen required, thereby increasing aerobic cost.

Though the SBS strategy appears to lower the strength requirements for stair negotiation, the result of doubling the steps (and time) to cover the same distance comes at a greater aerobic cost; this was true for both ascent and descent in both groups. Although the stroke group tended to consume more oxygen than controls during all strategies, it was only during descent that a significant interaction between group and strategy was observed, suggesting that likely an alternate movement strategy (beyond their reduced cadence) was adopted. Co-contraction of agonist and antagonist muscle groups for example, would result in increased oxygen requirements without increasing the net moment, however specific muscular contributions were not measured in this study.

The reduced muscle strength and reduced cardiovascular system capacity in stroke resulted in a high physical cost that may be considered a limiting factor for successful and safe stair ambulation. The oxygen requirements associated with stair negotiation in stroke relative to control subjects supports the characteristic inefficiencies of ambulation reported by others.
(Potempa et al., 1995, Macko et al., 1997, Ivey et al., 2005, Michael et al., 2005 Cunha et al., 2002). Chapter 4 explored the mechanical energy expenditures (MEEs) associated with stair negotiation as well as metabolic efficiency of the two stepping strategies.

In walking, stroke-related impairments reduced the intersegmental mechanical transfer (potential and kinetic energy) (Olney & Richards, 1996). The findings of this study illustrated a limited opportunity for energy transfer in stroke and controls based on the biomechanics of stair negotiation. In healthy individuals, the current study suggests that MEEs were 3 to 10 times greater during stair climbing than has been reported for overground walking (McGibbons, 2001a); the highest in association with the SBS strategy, likely due to its discontinuous nature. Consequently the transfer of energy generated or absorbed was limited, yielding little opportunity for compensating muscle activity.

During ascent, the important contributions of the trail limb plantarflexors at push-off, and lead limb knee extensors in pull-up were evident. Mechanical expenditures were high and revealed limited opportunity for energy transfer to compensate muscle activity, particularly for the SBS strategy. Though in both cases, expenditures were lower than in SOS, which parallels the finding of lower joint moments in Chapter 3, and provides support as to why individuals with stroke may adopt this compensatory pattern. During descent, the trail limb knee extensors and plantarflexors had a major role in energy absorption and transfer through eccentric activation respectively, producing high expenditures. In accordance with the study protocol, this meant that in stroke, this high demand was placed on the affected side, making stair descent particularly challenging.

In general, the SBS strategy had lower mechanical expenditures compared to SOS, however the per step aerobic cost was higher in SBS. In stroke, the result of reduced cadence on stairs lowered joint moments of major muscle groups (plantarflexors & knee extensors, Chapter 3), however, mechanical expenditures remained similar to controls and aerobic cost was
increased. This could be a result of co-contraction. Considering the level of muscle weakness and reduced postural balance in the stroke population, increased joint stiffness would be an effective strategy to enhance stability, particularly during descent where the affected limb served as the trail limb, resulting in a reduction in power, but a greater aerobic cost.

5.2 Limitations

There were three main limitations to the studies undertaken. The stroke group was high functioning which was indicated by their self-selected walking speeds, low TUG (timed-up-and-go) scores and average-to-high CBM (Community Balance and Mobility) scores in reference to what has been previously reported in the literature and may not be representative of the general stroke population. The findings, therefore, cannot be generalized to all people who have had a stroke. Despite this, differences in physical demands and energy efficiency (mechanical and metabolic) were observed between stroke and controls indicating the impact of even mild impairment on mobility. It is reasonable to hypothesize that these difficulties would be exaggerated in the presence of more severe impairment.

Another limitation of this work is the small sample size. The current study recruited fifteen participants with stroke, with datasets obtained for thirteen. In many cases, the high variability in the data reduced statistical power. A larger sample size would result in decreased variability and thus, greater statistical power.

The current study selected the lead limb as the less affected side in stroke and to the dominant side in control subjects. For ascent this is reasonable, though considering the high demand placed on the trail limb during descent, individuals with stroke would typically lead with their affected limb. The impact is that the findings for descent may not truly reflect the mechanics associated with natural task performance.
5.3 Future Directions

Future work should continue to study how the pattern of mobility relates to the output of the muscular and cardiovascular systems. Particularly in a population with multiple physical deficiencies, as in stroke, it is imperative to monitor how making adaptations in one system might affect another.

Future studies can build on the current work by recruiting larger sample sizes, and including a greater range in stroke severity. This would increase generalizability and better capture the impact of severe impairment on mobility in the chronic stroke population.

In terms of mechanical energy transfer, the methodology employed in the current study was unable to determine individual contributions from specific muscle groups, and only provided the net joint contribution. Future work should utilize electromyography to demonstrate the specific muscle activation patterns in individuals with stroke during stair negotiation in order to determine if they do in fact co-contract agonist and antagonist muscles to enhance stability, as was hypothesized in relation to the findings of the current work. Additionally, energy transfer was limited to the lower limbs, however energy transfer through the trunk during stair negotiation should be incorporated in future studies, particularly in the context of balance.

Although the stroke group in the current studies had better physical performance compared to what has been previously reported in the literature, there were still marked differences compared to healthy control participants. In addition, the relative strength and aerobic costs of stair ambulation remain high compared to overground walking. The findings demonstrated the appropriateness of improving the reserve capacity in individuals with stroke in order to reduce physical costs of mobility. An increase in reserve capacity could also result in a reduction of risks associated with working close to physical capacity (such as fatigue, falls etc.). Future work should continue to investigate the physical requirements of mobility tasks in the
context of relative cost. This information will assist in the development of specific strength and fitness targets for intervention programs.
5.4 References


References


Heart Disease and Stroke Statistics-2003 Update. Heart and Stroke Facts. Dallas, TX: American Heart Association; 2002


Tracking Heart Disease and Stroke in Canada. Released June 2009.


Appendix A

Research Ethics Board Approval Letter

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. R. Brison, Professor, Department of Emergency Medicine, Queen's University
Dr. M. Evans, Community Member
Dr. S. Horgan, Manager, Program Evaluation & Health Services Development, Geriatric Psychiatry Service,
Providence Care, Mental Health Services Assistant Professor, Department of Psychiatry
Ms. J. Hudacin, Community Member
Mr. D. McNaughton, Community Member
Ms. P. Newman, Pharmacist, Clinical Care Specialist and Clinical Lead, Quality and Safety, Pharmacy
Services, Kingston General Hospital
Dr. W. Racz, Emeritus Professor, Department of Pharmacology & Toxicology, Queen's University
Ms. S. Rohland, Privacy Officer, ICES-Queen's Health Services Research Facility, Research Associate,
Division of Cancer Care and Epidemiology, Queen's Cancer Research Institute
Dr. B. Simchison, Assistant Professor, Department of Anaesthesiology and Perioperative Medicine, 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 Bioethics, Queen's University
Dr. E. van DenKerkhof, Professor, School of Nursing and Department of Anaesthesiology and Perioperative
Medicine, Queen's University

has reviewed the request for renewal of Research Ethics Board approval for the project Biomechanical,
Neuromuscular and Cardiorespiratory Costs of Stair Negotiation in Healthy Individuals and Persons
With Stroke (new title May 5/11) Determining Physical Requirements for Every Day Activities (original
title) as proposed by Dr. Brenda Brouwer of the School of Rehabilitation Therapy, at Queen's University
The approval is renewed for one year, effective May 05, 2012. If there are any further amendments or changes
to the protocol affecting the participants 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.

Date: May 14, 2012

Chair, Research Ethics Board

Renewal 1 [ ] Renewal 2 [ X ] Extension [ X ] Code# REH-435-08 Romeo file# 6004264
Appendix B

Joint Power Calculations

For motion about a joint axis (i.e. rotational motion), the joint power \( P_j \) may be calculated by adding the powers at the distal \( P_d \) and proximal \( P_p \) ends of the articulating segments,

\[
P_j = P_d + P_p
\]

Where, \( P \) = muscle power, Watts/kg

The distal and proximal powers, in turn, can be calculated by multiplying the muscle moment \( (M_j) \) and the segment’s angular velocity \( (\omega) \), therefore, \( P_j = P_d + P_p \) becomes:

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

Where, \( M \) = muscle moment, Nm/kg, and \( \omega \) = angular velocity, rads/sec.

The sign of the power depicts whether the muscles generate \( (P_j > 0) \) or absorb energy \( (P_j < 0) \), and in cases when transfer occurs, the energy may be directed proximally \( (P_d < 0 \) and \( P_p > 0) \) or distally \( (P_d > 0 \) and \( P_p < 0) \).
### Appendix C

Heart rate and loading sequence for different stages of the YMCA cycle ergometer test

<table>
<thead>
<tr>
<th>1st Workload</th>
<th>150 kpm/min⁻¹ (0.5 kp or 25W)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subsequent Workloads</td>
<td>HR during Last Minute of Previous Workload</td>
</tr>
<tr>
<td></td>
<td>&lt;80bpm</td>
</tr>
<tr>
<td>2nd</td>
<td>750 kpm/min⁻¹ (2.5kp or 125W)</td>
</tr>
<tr>
<td>3rd</td>
<td>900 kpm/min⁻¹ (3.0kp or 150W)</td>
</tr>
<tr>
<td>4th</td>
<td>1050 kpm/min⁻¹ (3.5kp or 175W)</td>
</tr>
<tr>
<td>Etc...</td>
<td>If additional workloads are required to achieve within 10 beats of 85% HRmax, add 150kpm/min⁻¹ (0.5kp or 25W) to the previous workload.</td>
</tr>
</tbody>
</table>
Appendix D
YMCA VO\textsubscript{2max} calculation

For each of the last two workloads, calculate the oxygen cost (VO\textsubscript{2}) in mL•kg\textsuperscript{-1}•min\textsuperscript{-1} using the following equation:

\[ \text{VO}_2 = (\text{Workload (W)} \times 10.8) + 3.5 + 3.5 \]

Body mass (kg)

\[ \text{SM}_1 = \text{VO}_2 \text{ at Second-Last Workload} \]
\[ \text{SM}_2 = \text{VO}_2 \text{ at Last Workload} \]

From these two oxygen cost (VO\textsubscript{2}) values estimate the VO\textsubscript{2max} in mL•kg\textsuperscript{-1}•min\textsuperscript{-1} using the equations for the multistage model to calculate the slope of the line based on the HR response to the last two workloads

\[ \text{Slope (b)} = \frac{\text{SM}_2 - \text{SM}_1}{\text{HR}_2 - \text{HR}_1} \]

\[ \text{VO}_2\text{max} = \text{SM}_2 + [b \times (\text{HR}_\text{max} - \text{HR}_2)] \]
## Appendix E

### Mechanical energy transfer conditions

<table>
<thead>
<tr>
<th>Condition</th>
<th>Muscle Moment</th>
<th>Angular Velocity (Proximal)</th>
<th>Angular Velocity (Distal)</th>
<th>Power flow</th>
<th>Type of contraction</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<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>
<td>All power is generated by the muscles; no transfer occurs</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>
<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>
<td>All power is absorbed by the muscles; no transfer occurs</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>
<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>
<td>Power generated (concentric) or absorbed (eccentric) and proximally directed (transferred from distal to proximal)</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>
<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>
<td>Power generated (concentric) or absorbed (eccentric) and distally directed (transferred from proximal to distal)</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>
<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>
<td>Power generated (concentric) or absorbed (eccentric) and distally directed (transferred from proximal to distal)</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>
<td></td>
</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>
<td>Power generated (concentric) or absorbed (eccentric) and distally directed (transferred from proximal to distal)</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>
<td></td>
</tr>
</tbody>
</table>

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