

Mobility in Older Adults

by

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ABSTRACT

Mobility plays an important role in determining quality of life in older adults as it is closely tied to health, participation, and independence in later years. Although much of the literature to date has focused on walking and stair climbing, mobility also encompasses driving and the use of public transportation to access the community.

Comprehensive definitions of mobility and techniques for objectively measuring community mobility are generally lacking. This thesis describes a new theoretical framework for mobility that illustrates how impairments can lead to limitations in accessing different life-spaces, and stresses the associations among determinants that influence mobility. The feasibility of using global positioning system (GPS) watches and accelerometers to monitor community mobility in older adults was also examined. Data acquired from the equipment were quite variable. While the technology offers promise for capturing detailed information (e.g., the timing, distances covered, and speeds reached on foot and in-vehicle), new GPS solutions are required to allow for data collection over an extended period of time.

In addition, projects were conducted to examine ankle strength and power in older women because these muscle groups are important for physical function and mobility. Test-retest reliability on the dynamometer was found to be generally good for isotonic and isokinetic tests, but relatively poor for isometric rate of torque development. Measures of dorsiflexion (DF) and plantar flexion (PF) strength and power-related variables were significantly correlated with functional performance (gait speed, stair climb power, and foot movement time). An intervention study was conducted to determine the effects of ankle DF and PF resistance training performed concentrically “as

fast as possible” on movement time in a brake response task. Power training with elastic bands resulted in the greatest reduction in movement time, which suggests that this low-cost, practical form of exercise may benefit older adults in circumstances when rapid generation of ankle torque is required.

This thesis adds to the literature by examining mobility from a number of perspectives. Mobility determinants are comprehensively defined, community and laboratory-based measures are examined, and the effects of an intervention are evaluated to improve mobility assessment and treatment techniques in older adults.

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DEDICATION

I would like to dedicate this thesis to my parents, Hilton and Esther Wright. They instilled in me an interest and a love for learning. My dad passed away around the time that I was seriously considering embarking on this path, and unfortunately, he was unable to see me through it. Throughout my life, my parents have provided a positive stable base and an encouraging influence, and I owe much of my success to them.

LIST OF ABBREVIATIONS

CV	coefficient of variation
CV _{TE}	coefficient of variation of typical error
DF	dorsiflexion
GPS	global positioning system
ICC	intraclass correlation coefficient
LOA	limits of agreement
MDC	minimal detectable change
PASE	Physical Activity Scale for the Elderly
PF	plantar flexion
PP	peak power
PT	peak torque
RLOA	ratio limits of agreement
RM	repetition maximum
RTD	rate of torque development
SD	standard deviation
SEM	standard error of the measurement
SPPB	Short Physical Performance Battery
TE	typical error

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CHAPTER 1: GENERAL INTRODUCTION

Mobility is an important topic, relevant to older adults because it has far-reaching implications in terms of health and quality of life (Groessl *et al.* 2007; Yeom *et al.* 2008; Metz, 2000). The maintenance of mobility, which in a broad sense may refer to movement through one's environment (e.g., walking, wheeling a wheelchair, driving a car, taking a bus, etc.), is fundamental to active aging, allowing older adults to continue to lead independent lives and participate in society (World Health Organization, 2007). The maintenance of independence is largely dependent on one's ability to access required services and social contacts, and access is principally related to mobility status. Physical impairments and/or difficulties with transportation may result in immobility and isolation that can contribute to other social and physical problems in the older individual. Ultimately, mobility limitations have been shown to be associated with falling, loss of independence, institutionalization, and death in the older population (von Bonsdorff *et al.* 2006; Hirvensalo *et al.* 2000; Rubenstein *et al.* 2001).

This thesis examines mobility from a broad perspective as it is relevant to older adults. The overall objectives focus on defining and measuring mobility comprehensively, along with determining the effects of an exercise intervention on specific neuromuscular system factors that impact mobility. In addition, measures such as strength, power, and velocity are examined in older women to determine the relationships among these variables and to establish which parameters best predict functional outcomes relevant to mobility. In particular, the introduction will describe the following: 1) the need for a comprehensive definition of mobility relevant to older adults, 2) the prevalence of mobility problems, 3) factors that influence mobility, 4) methods of measuring and

monitoring mobility in older adults, 5) resistance exercise interventions to improve properties related to mobility, and 6) hypotheses and rationale for the individual thesis projects. The introduction is followed by a series of chapters, each consisting of a manuscript (either published, under review, or in preparation) that has been introduced in this first chapter. The concluding chapter of the thesis contains a general discussion of the findings and the relevance of the results.

This thesis incorporates six papers which will be or have been peer reviewed (four have been accepted for publication, one has been accepted pending minor revisions, and one is yet to be submitted). The thesis was successfully defended on February 8, 2010. The published papers may differ slightly from the format in this thesis. Readers are encouraged to refer to the final published manuscripts.

Section 1: The need for a definition of mobility relevant to older adults

Although there is widespread acceptance regarding the importance that mobility plays in the lives of older adults, there is no standardized definition or framework that encompasses the concept of mobility. From a practical standpoint, in research studies, older adults' mobility capabilities have frequently been determined based on objective, laboratory-based walking or stair climbing tests (Simonsick *et al.* 2008; Alexander *et al.* 2000; Bean *et al.* 2002b; Chang *et al.* 2004; Patel *et al.* 2006) and/or self-report of walking capabilities (Shumway-Cook *et al.* 2005; Simonsick *et al.* 2005; Guralnik *et al.* 1995; Fried *et al.* 2000; Hamalainen *et al.* 2006).

However, researchers have suggested that the definition of mobility should not be restricted to walking and/or stair climbing, rather that it should be extended to include movement to more distant locations, requiring the use of some form of transportation

(Myers *et al.* 2005). In fact, there is a large body of literature that deals specifically with mobility issues related to driving, driving cessation, and options for public transportation in older adults (Dickerson *et al.* 2007; O'Neill, 2000). Yet this work, and other research related to the built environment and its influence on mobility have been generally disconnected from more traditional studies of gait, walking tolerance, and balance.

The broader concept of community mobility has been traditionally measured with questionnaires in terms of life-space (May *et al.* 1985; Peel *et al.* 2005). Mobility capabilities in the community have been theoretically portrayed as a mobility continuum and as a mobility wheel (Patla & Shumway-Cook, 1999). These models have focused primarily on the physical and environmental influences on mobility. Despite the fact that many other types of factors are recognized as important in influencing mobility in older adults (e.g., psychosocial factors, financial capabilities, cognitive function, gender and cultural issues), there are no models for mobility that account for the complexity of the concept. In 2000, Metz challenged readers to develop an operationally useful and multidisciplinary definition for the concept of mobility. He suggested five fundamental elements that should be encompassed by a mobility model: 1) mobility includes travel to access desired people and places; 2) there are psychological benefits associated with “getting out and about”; 3) exercise benefits are associated with mobility; 4) mobility allows for social involvement in the local community; and 5) the potential for travel (i.e., knowing a trip could be made, e.g., in an emergency) is also important. To date there is still a need for such a framework to raise awareness of the complexity of factors that influence mobility, and stimulate new integrative, comprehensive, and interdisciplinary research ideas.

Section 2: The prevalence of mobility problems in older adults

In Canada, approximately 33% of individuals 65 years of age or older report having difficulty with walking, climbing stairs, and carrying objects short distances (Statistics Canada, 2007). This figure jumps to 76% for those people with disabilities aged 65 and older. According to the Participation and Activity Limitation Survey (2006), women are more likely than men to report have difficulties related to mobility for all age groups (Statistics Canada, 2007). The rate of mobility limitations tends to increase with age and the gender gap also becomes more apparent.

If we adopt an inclusive definition of mobility, one that incorporates the use of various modes of transportation, then other interesting statistics emerge. It has been suggested that the older adult in today's society requires the use of a private vehicle, as well as other forms of transportation, in order to access essential services, activities and people in the community (Oxley & Whelan, 2008). Analyses from the 2005 General Social Services Survey in Canada demonstrated that the majority (94%) of adults 65 years and older do have access to these transportation options (Turcotte, 2006). However, numbers declined with advancing age, especially when use of a private vehicle was solely considered. Approximately 83% of older men (75-84 years of age) reported that they held a valid driver's license and had access to a vehicle compared to only 45% of women in the same age bracket. Generally, women, seniors with lower incomes, and people 85 years of age and older living in rural areas were found to be most at risk for mobility limitation.

Overall, these statistics suggest that the prevalence of mobility impairments (including both walking-related movement and transportation difficulties) is relatively

high in the older Canadian population. Given the significant role that mobility plays in determining independence, health, and general quality of life, it is an important topic that warrants further study.

Section 3: Factors that influence mobility

Mobility, considered in a broad sense, involves the complicated inter-play of a multitude of factors. Physical, psychological, socioeconomic, environmental, and cultural influences may all impact mobility status (Myers *et al.* 2005; Turcotte, 2006; Stalvey *et al.* 1999). Personal factors such as age, health, education level, sex, and gender are also important (Finlayson & Kaufert, 2002; Stalvey *et al.* 1999). For some of these factors, the effect on mobility is quite straight-forward and direct (e.g., it is clear that environmental factors such as the location of one's neighborhood, and the presence or absence of acceptable sidewalks, lighting, weather, traffic, and overall urban design may be important). For other factors, influences on mobility are less obvious. For example, older women are less likely to drive a car (Turcotte, 2006), more likely to suffer from fear of falling (Filiatrault *et al.* 2009), and more likely to be diagnosed with disability-related health conditions (Murtagh & Hubert, 2004). All of these factors put older women at greater risk for mobility limitations. As another example, it is not always clear how culture may play a role in determining mobility. However, by influencing social relationships, educational opportunities, occupational experiences, and physical activity habits, cultural norms may impact mobility options (Mollenkopf *et al.* 1997; Golant, 1984).

Examples of many of the above-mentioned mobility-related factors are provided in Chapter 2 which explains the rationale for and the development of a theoretical

framework for mobility in older adults. Because much of the work described in this thesis deals specifically with measuring and attempting to influence neuromuscular factors that impact mobility (e.g., strength, power, and movement time), they will be discussed in more detail in this section.

Strength is defined as the ability to produce force (in a linear system) or torque (in an angular system). The capacity to produce force or torque is dependent on many neuromuscular factors which undergo age-related changes. It is generally accepted that strength declines with age as a result of loss of muscle fibre numbers (Lexell *et al.* 1983; Lexell *et al.* 1988), muscle fibre atrophy which primarily affects the stronger and faster type II muscle fibres (Vandervoort, 2002; Porter *et al.* 1995; Coggan *et al.* 1992), reduction in contractile speed (Roos *et al.* 1997), and possible reductions in specific tension (force per cross-sectional area of muscle) (Frontera *et al.* 2000b; Larsson *et al.* 1997).

Other age-related changes in the neuromuscular system also have the potential to influence strength production. For example, although human data are not available, experiments in rats suggest that motoneuron gain and current threshold are reduced with age (Kalmar *et al.* 2009). This means that motoneurons in an older animal may require lower levels of input current in order to fire at similar frequencies as motoneurons in younger animals. As a result, force production may be higher at lower levels of current input in older animals. However, as part of the age-related motor unit remodeling process, axons from low-threshold motoneurons may re-innervate muscle fibres that were previously type II, rendering them slow-twitch (Kanda & Hashizume, 1989; Stalberg *et al.* 1989). This change likely contributes to the lower maximal force levels seen in older

individuals. For a more complete discussion of the neuromuscular changes that occur with age and the effects of these changes on recruitment, rate-coding, and resultant force output, readers are referred to the paper in Appendix A entitled “Modeling age-related neuromuscular changes in humans.” This paper, published in *Applied Physiology, Nutrition, and Metabolism* [34:(4), August 2009] was written in partial fulfillment of my candidacy requirements. In this paper, the Heckman-Binder model (Heckman & Binder, 1991) originally developed to portray a pool of 100 simulated motor units in the cat, was used for studying the relationships among current input, firing frequency, and force output in simulated younger and older humans.

Cross-sectional studies suggest that strength levels decline at a rate of approximately 1.0% – 1.5 % per year after age 50 (Vandervoort & McComas, 1986; Larsson *et al.* 1979) and longitudinal data support a 1.0 % – 2.0 % decline per year (Frontera *et al.* 2000a; Metter *et al.* 1997; Hughes *et al.* 2001). The neuromuscular changes which result in reductions in strength also impact on mobility performance. For example, older adults with reduced lower extremity strength levels also demonstrate slower walking speeds (Bendall *et al.* 1989; Tiedemann *et al.* 2005), slower sit-to-stand performance (Lord *et al.* 2002), greater levels of fall risk (Liu-Ambrose *et al.* 2004), and faster rates of mobility decline measured through timed walking and turning tests (Buchman *et al.* 2007).

Power is another parameter highly dependent on neuromuscular factors. Power is defined as the amount of work performed in a given period of time and, as such, can be calculated as the product of torque and velocity (American College of Sports Medicine, 2009). Data concerned with the rate of decline of muscle power seem to suggest that

power declines at a faster rate than does strength (approximately 3.5 % per year) (Skelton *et al.* 1994; Metter *et al.* 1997). Metter and colleagues (1997) found that reductions in power generating capabilities also occurred at earlier ages compared to reductions in strength in women. It is assumed that the greater rate of decline in power with age may be related to selective atrophy of type II muscle fibres, decreases in maximal shortening velocity and specific tension (Larsson *et al.* 1997), and/or reductions in tendon stiffness (Macaluso & DeVito, 2004).

Although both strength and power have been shown to decline with age, the results of some studies have suggested that it is the loss of lower extremity muscle power that plays a larger role in influencing functional mobility (Basseley *et al.* 1992; Bean *et al.* 2003; Marsh *et al.* 2006; Suzuki *et al.* 2001; Bean *et al.* 2002a) and fall risk in older adults (Skelton *et al.* 2002; Perry *et al.* 2007). Metter and colleagues (1997) summarized their results by stating that “power changes to a greater extent with age than strength” and “the independent effect of age on power argues for the importance of movement speed, coordination, and other factors in power generation” (page B275). Indeed, variables related to response time are important in situations that demand fast movements, for example, when recovering balance after a trip and when applying the brakes in a vehicle (Mackey & Robinovitch, 2006; Margolis *et al.* 2002). It has been shown that older adults at increased risk for falling demonstrate slower reaction times and slower movement times compared to those in lower risk categories (St George *et al.* 2007). In these types of situations, in order to maintain safe mobility, the speed of torque generation and the velocity of the resultant movement may be more important than the absolute magnitude of the torque generated (Mackey & Robinovitch, 2006).

Section 4: Methods of measuring and monitoring mobility in older adults

Although researchers have developed a number of laboratory-based evaluations designed to measure specific aspects of mobility (e.g., gait speed over a specified course, time to climb a set of stairs, balance assessments), tools capable of objectively and comprehensively measuring and monitoring mobility in the broader community setting are lacking. Many interventions aimed at the older population focus on improving mobility and resultant quality of life (e.g., through exercise trials, provision of walking aids, and driver training courses), however, in actual fact, capabilities for measuring changes in mobility in the “real world” are limited. Because it is impossible to replicate the environmental demands of specific community settings, valid assessment of community mobility patterns must be measured within the life-spaces that are relevant to older adults.

Questionnaires have been used to attempt to measure physical activity levels and travel patterns in subjects of different ages. Various life-space instruments have been designed to measure the spatial extent of older adults’ travel within home and community environments (Stalvey *et al.* 1999; Baker *et al.* 2003; May *et al.* 1985; Tinetti & Ginter, 1990). These tools typically provide information about locations visited (in the home and community), the degree of assistance required to accomplish travel, and the frequency of travel. However, specific information about individual trips within the community is not attained with these surveys, and many diary-type tools have been shown to be inaccurate, lack detail, and place a high degree of burden on subjects (Stopher & Greaves, 2007).

In order to avoid some of the short-comings associated with indirect methods of measuring and monitoring physical activity levels, two types of motion sensors have been

developed to measure physical activity more directly in a practical manner (see review by Tudor-Locke & Myers, 2001). These devices, namely pedometers and accelerometers, have been used to quantify physical activity in a number of different age groups. Both of these small instruments are usually worn clipped to a belt at the waist. A pedometer contains a horizontal, spring-suspended lever arm which deflects with vertical acceleration, causing an electrical circuit to open and close with each deflection. In this way, steps associated with foot fall during gait can be measured. An accelerometer, on the other hand, contains piezoelectric crystals which generate charges that are detected when they are accelerated during motion. Accelerometers can detect movement in one plane (uniaxial) or three planes (triaxial) depending on the number and orientation of the piezoelectric crystals. Accelerometers generally record activity counts which can be used to categorize the intensity of physical activity and estimate energy expenditure. It is also possible to get a reading of the number of steps recorded when an individual wears an accelerometer on the hip.

Because accelerometers are more sensitive to step detection at slower walking speeds than pedometers, they have been recommended for use in older adults (Culhane *et al.* 2005; Le Masurier & Tudor-Locke, 2003). Accelerometers also have greater data recording and storage capabilities compared to pedometers and can provide step and activity counts within user-defined time frames (e.g. every minute), storing data for many consecutive days without requiring interaction between the wearer and the device (Trost *et al.* 2005). Despite these benefits, accelerometer data are still limited to step or activity counts and do not provide direct information about gait speed or mobility habits in different environments. For example, without additional information, it is not possible to

know whether a recorded activity occurred indoors or outdoors, in the home, or in the community. As well, pedometers and accelerometers provide no useful information about vehicle travel. In fact, when either of these devices is worn inside a vehicle, they may mistakenly count steps if enough vertical movement (jarring) occurs.

Satellite-based global positioning system (GPS) technology, however, offers potential advantages for studying movement in the community which may involve outdoor walking and using various forms of transportation (Elgethun *et al.* 2007; Le Faucheur A. *et al.* 2007; Phillips *et al.* 2001; Rodriguez *et al.* 2005). GPS devices provide rich information regarding geographical and temporal aspects of travel, as well as detailed speed and route descriptions that can not be accurately obtained using surveys and/or physical activity monitors (Stopher *et al.* 2008). Until recently, relatively large GPS devices were primarily used in vehicles to capture transportation-related data (Porter & Whitton, 2002; Wolf *et al.* 2001). However, the development of smaller GPS receivers that can more comfortably be carried or worn by individuals has made their use more practical for studying all types of outdoor movement (e.g., walking, bicycling, driving) (Stopher *et al.* 2008). In the future it may be possible to use these devices to monitor changes in mobility parameters and patterns in older adults in response to interventions that are designed with the aim of improving mobility in the “real world.” However, initial studies must first investigate the feasibility of using such devices to monitor community mobility in this population.

Section 5: Resistance exercise interventions to improve properties related to mobility

In 2004 Latham and colleagues published a systematic review of randomized controlled resistance training trials in older adults (Latham *et al.* 2004). The review confirmed that traditional resistance training in older adults is effective in causing muscle hypertrophy and improving strength, with high intensity programs eliciting greater effects than lower intensity training regimes. In terms of functional mobility-related tasks, the review determined that traditional progressive resistance training has a moderate effect on improving gait speed, reducing the time required for sit-to-stand tests, and improving walking endurance. A more recent Cochrane review supported these findings, suggesting that progressive resistance training has a large effect on muscle strength, a moderate to large effect on improving the ability to get up from a chair and climb stairs, and a modest effect on gait speed (Liu & Latham, 2009).

Another systematic review recently examined the role of progressive resistance training on balance performance in older adults (Orr *et al.* 2008). The review found that only 22% of the static, dynamic, functional, and computerized balance test results examined in 29 studies demonstrated improvements following the implementation of resistance training. The authors of the review suggested that the inconsistent effect of resistance training on balance may have resulted from variability in the subject groups and balance tests chosen; inadequate sample sizes, resistance training doses, or exercise compliance rates; and/or lack of statistical power.

Because research suggests that power, as opposed to strength, is more closely associated with functional mobility (Bassey *et al.* 1992; Bean *et al.* 2003; Marsh *et al.*

2006; Suzuki *et al.* 2001; Bean *et al.* 2002a) and demonstrates greater decline with age (Skelton *et al.* 1994; Metter *et al.* 1997), there has been recent interest in examining the effects of power training interventions on mobility-related outcomes. In older adults, the term “power training” has most commonly been used to describe resistance training which is conducted with the concentric portions of the training movements performed “as fast as possible” and the eccentric phases of the movements completed over 2-3 seconds (Marsh *et al.* 2009; Orr *et al.* 2006; Bean *et al.* 2009; de Vos *et al.* 2005; Earles *et al.* 2001; Fielding *et al.* 2002; Henwood & Taaffe, 2005).

The number of power training intervention studies conducted to date is relatively small and results are not definitive. As well, the muscle groups studied and the modes used for power training and testing have not been consistent. One study found significantly better results on the repeated chair rise test and the 8-foot-up-and-go test in those older adults who performed power training as opposed to traditional strength training (Bottaro *et al.* 2007). Other studies have demonstrated improvements in chair rise, gait speed performance (Hruda *et al.* 2003; Henwood & Taaffe, 2005), and balance (Orr *et al.* 2006) in response to power training, however strength training groups were not included for comparison. Another study found increases in power in response to power training, but no change in other physical function test results (balance, chair rise, gait speed, stair climb) (Sayers *et al.* 2003). Henwood and colleagues found no differences in improvements in functional performance between older adults who trained with high velocity movements and those who performed traditional strength training protocols (Henwood *et al.* 2008). This finding has been demonstrated in additional studies (Earles *et al.* 2001; Bean *et al.* 2009; Marsh *et al.* 2009). Reasons cited for the lack of differences

in functional outcomes between strength and power training groups include: inadequate resistance training doses to induce specific strength and power improvements, differences in specificity between training modes and testing techniques, and lack of statistical power.

Therefore, despite the apparent association between power generation capabilities and function/mobility, it is not clear whether power training is more effective than traditional strength training in improving function/mobility in older adults. It has been suggested that there is a need to explore optimal resistance training protocols that focus on the muscle groups most relevant to specific mobility-related tasks and target the types of neuromuscular adaptations pertinent to maintaining balance and generally improving movement control in older adults (Orr *et al.* 2008). It may be important to match specific power training protocols to particular muscle groups and certain functional or mobility-related tasks in order to truly determine the relative importance of power training compared to traditional strength training in older adults.

Summary of the issues

Mobility, considered in a broad sense, contributes significantly to older adults' physical and psychological health. Despite its importance, and the prevalence of mobility impairments, the concept of mobility is not consistently considered in an inclusive manner. For this reason, research and social programs targeting issues related to mobility in later life have been largely discipline-specific. There is a need to develop a comprehensive new framework for mobility by examining the concept along with its primary predictors and modifiers, from a variety of points of view. There is also a need for developing tools capable of monitoring different types of mobility (e.g., walking, driving, taking public transportation) in a variety of life-spaces. Successfully meeting

these goals may increase awareness of the issues and strengthen the associations among different mobility contexts in the hopes of developing more extensive and effective intervention programs and having the capability to measure mobility change in a variety of environments in the future.

While there are a multitude of factors that have the potential to influence mobility, this thesis deals more specifically with measuring and attempting to influence neuromuscular factors (e.g., strength, power, velocity, and movement time). The reliability of these different measures and the relationships among them are important factors to study. Although some research suggests that resistance training programs should be tailored to enhance power generation capabilities in older adults, it is not yet clear how this is best accomplished (i.e., the specifics of the resistance training protocol and equipment have not been determined).

Section 6: Rationale, objectives and hypotheses for projects

Overall, this thesis deals with mobility in different contexts as it is relevant to older adults. The paper in Chapter 2 describes the development of a new theoretical framework for mobility that attempts to comprehensively explain the complexities of mobility determinants. Chapter 3 describes a feasibility study that was undertaken to examine the combined use of GPS devices and accelerometers in monitoring community mobility in older adults. The work described in Chapter 4 was undertaken to determine the reliability of a number of strength and power-related variables that can be measured on an isokinetic dynamometer. It is important to understand the reliability of these types of tests that are increasingly being used to attempt to measure the effects of resistance training interventions. Chapter 5 presents the results of an ankle power training

intervention, designed to investigate the effects of training with weights and elastic bands on movement time in self-identified mobility-impaired older women. Chapter 6 presents a discussion of some of the different ways that strength, power, maximum voluntary velocity, and acceleration can be measured on an isokinetic dynamometer. Developing a greater understanding of the variety of tests that can be conducted using the different modes on a dynamometer may result in a better appreciation for the strengths and limitations of individual tests. Relationships among different isokinetic and isotonic strength, power, and power-related variables are presented. In addition, relationships among different ankle measures and specific functional outcomes are evaluated.

What follows is an introduction of each body of work included in this thesis. The title, rationale, and hypotheses are provided (where appropriate) for each project.

CHAPTER 2 - Mobility in Older Adults: A Comprehensive Framework

Rationale for the project: Although it is known that mobility capabilities are vitally important to the health and well-being of older adults, few theoretical frameworks exist to describe this multi-dimensional, inter-disciplinary concept. It is necessary to have a clear definition of mobility in order to determine the most effective ways to measure and influence mobility (at the societal and individual level) as it pertains to older adults.

Objective: The objective of this project was to develop and present a new theoretical framework for mobility that built upon previous models. The new theoretical framework was designed to include mobility factors relevant to older adults living independently in the community and those requiring specialized assistance. The framework explains how determinants that influence mobility are inter-connected and demonstrates how mobility impairments can lead to limitations in accessing different life-spaces.

CHAPTER 3 - Monitoring mobility in older adults using global positioning system (GPS) watches and accelerometers: a feasibility study

Rationale for the project: GPS devices have not previously been used in combination with accelerometers to monitor community mobility in older adults. While studies have demonstrated that GPS devices are capable of gathering detailed information about the location, timing, and speed of movement during discrete bouts of outdoor activity, the same success has not been realized in studies that have attempted to monitor functional everyday human movement over an extended period of time.

Objective: The purpose of this study was to determine the feasibility of using GPS watches along with hip-worn accelerometers to gather information about walking and other forms of transportation travel in older adults over the course of one day.

CHAPTER 4 – Reliability of ankle isometric, isotonic, and isokinetic strength and power testing in older women

Rationale for the project: The muscles around the ankle play an important role in influencing mobility in older adults. These distal leg muscles exhibit reductions in strength and power with aging (Lanza *et al.* 2003; Thelen *et al.* 1996). In order to measure changes in strength and power with age, reliable testing techniques are required. Although strength and power parameters are frequently measured with dynamometers, few studies have specifically examined the reliability associated with these measures about the ankle in older adults.

Objective: The purpose of this study was to examine the relative and absolute intra-rater reliability of isometric, isotonic, and isokinetic ankle strength, power, and velocity measures in older women.

CHAPTER 5 – Effects of ankle power training on movement time in mobility-impaired older women

Rationale for the project: The muscles around the lower leg are important for function in older adults. Specifically, ankle strength and power have been shown to be significantly related to chair rise time, stair climb time, gait velocity, balance, and the occurrence of falls (Suzuki *et al.* 2001; Whipple *et al.* 1987; Wolfson *et al.* 1995; Mackey & Robinovitch, 2006; Skelton *et al.* 2002; Bendall *et al.* 1989). Cross-sectional studies have demonstrated that physical performance may be more closely tied to power generation capabilities than strength potential in older adults (Suzuki *et al.* 2001; Bean *et al.* 2002a). Being able to generate torque quickly may be especially important in situations that demand fast movement responses (e.g., to avoid a fall or apply the brakes quickly in a vehicle).

Objective: The objective of this study was to determine the effects of ankle power training (with weights and with elastic bands) on foot movement time in a group of mobility-impaired older women. We were specifically interested in determining whether movement time changed within each training group. Secondly, we were interested in measuring changes in plantar flexion (PF) and dorsiflexion (DF) strength and power in response to training.

Hypotheses: We hypothesized that movement time would become faster and PF strength and power would improve in the weight-trained group. As well, we anticipated that the smaller DF muscles would achieve a training effect with lower loads, therefore DF strength and power measures would improve in both the weight and elastic band training groups.

CHAPTER 6 – Relationships among ankle strength, power, velocity and function in older women

Rationale for the project: Isokinetic dynamometry is used to measure power in older adults. Most commonly, power generation has been studied during constant velocity movements (i.e., using the isokinetic mode on the dynamometer) (Suzuki *et al.* 2001; Hartmann *et al.* 2009; Symons *et al.* 2005), however there has been some interest in using the isotonic mode to measure power and velocity in this population (Stauber *et al.* 2000; McNeil *et al.* 2007). To date, the relationships among different measures of ankle strength, power, and velocity are unclear. As well, it is not known whether certain measures predict functional mobility-related outcomes better than others.

Objectives: The objective of this study was to determine the relationships among ankle isokinetic and isotonic measures and functional performance (usual gait speed, fast gait speed, stair climb power, and movement time).

Hypotheses: While it was anticipated that many of the measures would be related to function, we hypothesized that plantar flexion (PF) would be more closely related than dorsiflexion (DF) to the chosen functional measures. Of the four functional outcomes, we hypothesized that power-related PF variables would be most highly correlated with fast gait speed and stair climb power.

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CHAPTER 2: MOBILITY IN OLDER ADULTS: A COMPREHENSIVE FRAMEWORK

Running head: Mobility in Older Adults

Key Words: Community, Driving, Life-space, Transportation, Walking

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MY CONTRIBUTION TO THE PUBLICATION

The topic for this paper was provided to me by my committee as part of the requirements for my candidacy exam. I conducted all of the literature reviews required for the development of this theoretical framework on mobility. I wrote the initial paper and presented it at my candidacy exam. I then made changes to the manuscript based on the feedback received from my committee. I presented the material in poster format at the International Association on Gerontology and Geriatrics World Congress in Paris, France in July 2009. I selected the journal to which we submitted the manuscript and, based on the reviewers' comments that were received, I revised the manuscript in consultation with Dr. Porter and Dr. Menec. I then resubmitted the manuscript along with the responses to the reviewers. When the manuscript was accepted I made any further required corrections such as those included in the galley proofs.

ABSTRACT

Mobility is fundamental to active aging and is intimately linked to health status and quality of life. Although there is widespread acceptance regarding the importance of mobility in older adults, there have been few attempts to comprehensively portray mobility and research has to a large extent been discipline-specific. In this paper a new theoretical framework for mobility is presented with the goals of raising awareness of the complexity of factors that influence mobility and stimulating new integrative and interdisciplinary research ideas. Mobility is broadly defined as the ability to move oneself (e.g., by walking, by using assistive devices, or by using transportation) within community environments that expand from one's home, to the neighborhood and to regions beyond. The concept of mobility is portrayed through 5 fundamental categories of determinants (cognitive, psychosocial, physical, environmental, and financial), with gender, culture, and biography (personal life history) conceptualized as critical cross-cutting influences. Each category of determinants consists of an increasing number of factors, demonstrating greater complexity, as the mobility environment expands farther from the home. The framework illustrates how mobility impairments can lead to limitations in accessing different life-spaces and stresses the associations among determinants that influence mobility. By bridging disciplines and representing mobility in an inclusive manner, the model suggests that research needs to be more interdisciplinary, current mobility findings should be interpreted more comprehensively, and new, more complex strategies should be developed to address mobility concerns.

INTRODUCTION

The maintenance of mobility is thought to be fundamental to active aging, allowing older adults to continue to lead dynamic and independent lives (World Health Organization [WHO], 2007). Loss of mobility may occur, for example, when an older adult is no longer able to safely drive a vehicle and/or when physical disabilities interfere with walking. Because activity restriction is associated with numerous consequences related to physical de-conditioning and reduced levels of social participation, mobility is intimately linked to health status and quality of life (Groessl et al., 2007; Metz, 2000; Yeom, Fleury, & Keller, 2008). Impaired mobility has been shown to be an early predictor of physical disability (Hirvensalo, Rantanen, & Heikkinen, 2000) and ultimately, to be associated with falling, loss of independence, institutionalization, and death (Hirvensalo et al., 2000; Rubenstein, Powers, & MacLean, 2001; von Bonsdorff, Rantanen, Laukkanen, Suutama, & Heikkinen, 2006). For these reasons, researchers and older adults themselves are keenly interested in determining factors that influence mobility and ways to maximize mobility as people age.

Despite the importance of mobility in older adults, a comprehensive theoretical framework that describes mobility in an inclusive manner is lacking and research and social programs targeting issues related to mobility in later life have been largely discipline-specific. For example, rehabilitation-related studies concerned with the biomechanical and physiological aspects of gait typically have not been linked to driving-related research and efforts to promote the use of public transportation systems have not been coordinated with programs designed to improve functional walking tolerance. Models developed to date have focused on specific components of mobility (e.g., life-

spaces relevant to older adults and environmental factors that influence physical aspects of mobility), but have not defined mobility in a broad or inclusive way (e.g., incorporating movement on foot and in vehicle; recognizing biographical, psychosocial, cognitive, and financial influences). Our purpose was to develop a comprehensive new framework by examining the concept of mobility, along with its primary predictors and modifiers, from a variety of points of view. In doing so, it was not our intention to review all aspects of the pertinent literature in detail, but rather to highlight the key determinants of mobility and demonstrate inter-related factors that influence mobility in different environments. We propose that this interdisciplinary model for mobility will challenge researchers, clinicians, and policy-makers to consider mobility from a number of different perspectives which will increase awareness of the different contexts of mobility issues relevant to older adults. Demonstrating linkages among the various areas of mobility may guide future research to be more holistic and contribute to the development of more extensive and effective intervention programs.

In this paper, we briefly discuss some of the perspectives taken by different disciplines and bodies of research in defining and measuring mobility. Conceptual frameworks that have contributed to our understanding of mobility are then described and components are used to generate a new, more comprehensive model.

Defining and Measuring Mobility

Self-reported mobility limitation is common in older people. The prevalence of mobility impairment depends on the definition used to identify restrictions.

Approximately one third to one half of individuals 65 years of age or older report difficulties related to walking or climbing stairs (Shumway-Cook, Ciol, Yorkston,

Hoffman, & Chan, 2005; Statistics Canada, 2007) which are common tests used to quantify mobility capabilities (Guralnik, Ferrucci, Simonsick, Salive, & Wallace, 1995; Shumway-Cook et al., 2005; Simonsick et al., 2008). However, when mobility is defined more broadly, the concept also includes movement outdoors and beyond the home, which frequently entails the use of some form of transportation (Myers, Cyarto, & Blanchard, 2005; Stalvey, Owsley, Sloane, & Ball, 1999). Stalvey et al. referred to mobility as “the spatial extent of one’s travel within the environment”, encompassing “travel in, around, and outside the home as one conducts the business and social aspects of everyday life” (p. 461). The World Health Organization’s *International Classification of Functioning, Disability and Health* also recognizes a broad description of mobility, including both indoor and outdoor movement as well as the use of assistive devices and transportation (WHO, 2001).

In this paper mobility has been defined as the ability to move oneself (either independently or by using assistive devices or transportation) within environments that expand from one’s home to the neighborhood and to regions beyond. For research purposes, mobility within and beyond one’s home has frequently been defined and measured in terms of life-space (Baker, Bodner, & Allman, 2003; May, Nayak, & Isaacs, 1985; Peel et al., 2005; Stalvey et al., 1999). Life-space “estimates the magnitude or extent of travel into the environment, regardless of how one gets there” (Stalvey et al., 1999, p. 472).

Research has also focused on the effects of the built environment on community mobility (Clarke, Ailshire, Bader, Morenoff, & House, 2008; Nagel, Carlson, Bosworth, & Michael, 2008). Vehicles and other forms of transportation are required in order for

older adults to maintain access to essential services, activities and people (Oxley & Whelan, 2008). There is a large body of literature that deals specifically with mobility issues related to transportation options for older adults. For example, the majority of older adults still rely on the personal automobile as their primary form of transportation. Much research is therefore focused on maintaining and improving safety in older drivers to keep them behind the wheel as long as possible (Dickerson et al., 2007; O'Neill, 2000) because driving cessation is associated with numerous negative consequences including increased dependency, social isolation, and depression (Edwards, Perkins, Ross, & Reynolds, 2009; Fonda, Wallace, & Herzog, 2001; Marottoli, 2000; Marottoli et al., 1997).

Conceptual Frameworks Relevant to Mobility in Older Adults

Over the past 40 years, a number of conceptual frameworks have been developed that relate directly or indirectly to mobility issues relevant to older adults. Early work related to mobility in older adults was largely focused on the environment and the role of the person-environment fit (Lawton & Nahemow, 1973; Rowles, 1983). Rowles (1983) explored locations that had meaning and were frequently visited by older adults. Lawton and Nahemow (1973) developed an ecological model of adaptation in older age that laid the theoretical foundation for examining an individual's ability to interact successfully with the demands of the environment. According to their model, matching levels of competence with the demands of the environment engenders well-being and fosters independence. Carp (1988) further conceptualized mobility itself as being central to well-being and independence in later life. In this model, mobility is portrayed as being fundamental to determining whether life-maintenance needs (e.g., food, clothing,

healthcare) are met independently, and whether higher-order needs (e.g., social relationships, recreational activities) are fulfilled to promote well-being.

Patla and Shumway-Cook (1999) introduced a mobility continuum which portrayed the positive relationship between independent walking tolerance and ability to access the community. They also presented a framework centered around physical and environmental factors and their influence on walking. Minimum walking distance, time constraints, ambient conditions, terrain characteristics, external physical loads, demands on attention, postural transitions, and traffic levels were conceptualized as eight spokes on a mobility wheel. Although this framework focused on factors relevant to accessing the community on foot, many factors would apply equally well to using wheelchairs or other forms of transportation.

Rose (2005) also proposed a theoretical framework for balance and mobility relevant to older adults. This framework included consideration of the inter-relationships among individual capabilities, environmental constraints, and task demands. Similar factors were included in the theory of driver behaviour proposed by Fuller (2005) which involves driving task demands and driver capabilities (e.g., physical, cognitive, and psychological characteristics) and recognizes that environmental factors, compensatory strategies, and driver perceptions influence driver action.

In order to conceptualize mobility more coherently, a framework that links factors relevant to walking, wheeling, driving, and taking alternate forms of transportation within different life-spaces is required. Much of the research to date has been relatively compartmentalized, focusing heavily on physical aspects related to the person and/or the environment. Existing theoretical frameworks have successfully

portrayed specific aspects but a new model capable of illustrating more comprehensively inter-related factors relevant to mobility in different contexts is necessary to bridge disciplines, drive new research questions, and inform at the societal and political level.

A NEW THEORETICAL FRAMEWORK FOR MOBILITY

In order to conceptualize mobility in a more holistic fashion, it is necessary to consider multiple determinants that influence mobility, for those living independently and for those requiring care, and extending from the home into the larger community. For the new theoretical framework, it is assumed that mobility may take many forms including, but not limited to, walking, using a wheelchair, driving, and using alternate forms of transportation.

As portrayed in the life-space literature (May et al., 1985; Peel et al., 2005) and the mobility continuum (Patla & Shumway-Cook, 1999), the framework includes concentric areas of expanding locations from home with increasing requirements for independent mobility. Included as mobility zones are the room where one sleeps, the home (e.g., house, apartment, institution), the outdoor area surrounding the home (e.g., yard, parking lot), the neighborhood (e.g., nearby streets or parks), the service community (e.g., shops, banks, health care facilities), the surrounding area (e.g., within one's country), and the world (see Figure 1). Each life-space portrayed in the vertical order may be represented by a cross-section made up of five categories of determinants that influence mobility at all life-space levels. The total cross-sectional area increases with expanding life-spaces, suggesting that a greater number of factors contribute to each determinant category as one moves farther from home.

The model recognizes that gender, culture and biography (personal life history) each fundamentally shape individuals' experiences, opportunities, and behaviors, and therefore act as cross-cutting influences on mobility (Commission on the Social Determinants of Health, 2008; WHO, 2002). For example, mobility limitations in older adults are not equally distributed, with women demonstrating greater limitations and greater risk of mobility disability compared to men (Leveille, Penninx, Melzer, Izmirlian, & Guralnik, 2000; Murtagh & Hubert, 2004; Shumway-Cook et al., 2005; Statistics Canada, 2007). Culture also influences mobility through its effect on social relationships, educational and occupational opportunities, and physical activity habits (Golant, 1984; Mollenkopf et al., 1997). Because gender, culture, and biography indirectly impact mobility through their influence on the five key determinants, they are depicted as encircling the entire mobility cone (Figure 1).

The key determinants of mobility illustrated in Figure 1 include cognitive, psychosocial, physical, environmental, and financial influences. Cognitive determinants include a broad range of factors such as mental status, memory, speed of processing and executive functioning, whereas psychosocial determinants include factors like self-efficacy, coping behaviours, affect/depression, fear, and relationships with others that affect interest and/or motivation to be mobile. The relative importance of different factors depends on the specific mobility context for an individual. For example, speed of information processing and visual attention are important for safe driving (Owsley et al., 1998). However, an individual with low self-efficacy beliefs may not even attempt to be mobile beyond the home, despite his/her actual driving or walking capabilities (McAuley et al., 2006; Perkins, Multhaup, Perkins, & Barton, 2008). Older individuals also

sometimes self-restrict mobility due to depression (Gayman, Turner, & Cui, 2008) or in response to opinions voiced by friends, family, and physicians (Rudman, Friedland, Chipman, & Sciortino, 2006). Older adults who have had a previous fall (Tinetti, Richman, & Powell, 1990), as well as those who have never fallen (Friedman, Munoz, West, Rubin, & Fried, 2002; Howland et al., 1993; Maki, Holliday, & Topper, 1991), may demonstrate fear of falling to the degree that it may interfere with mobility choices. Diagnoses such as mild cognitive impairment and dementia also have the potential to seriously challenge mobility outside of the home. Problems related to driving safety, wandering, and getting lost (in-vehicle and on foot) are common (Adler & Silverstein, 2008; Cotter, 2007).

Inherent to the mobility framework is the notion that mobility determinants are inter-related. For example, age-related changes such as impairments in vision and/or reaction time may be incongruent with specific challenges presented by environmental conditions, making mobility hazardous in certain circumstances (e.g., driving at night or in poor weather conditions, walking on a slippery surface with limited lighting). Because dimensions are linked (e.g., self-efficacy affects physical function), a change in one dimension (e.g., a fractured ankle that affects physical capabilities) may also result in an altered ability to meet specific factor requirements in other determinant categories (e.g., stairs and outdoor terrain in the environment may become obstacles to someone using a walker). This is illustrated in Figure 2 in which specific environmental and physical mobility requirements can only be met in the two lowermost life-spaces. Although financial, cognitive and psychosocial factors may not be affected by this change in health status (and are therefore portrayed as extending throughout all life-spaces in Figure 2),

mobility is limited to the level at which all factors in the five determinant categories can be met (e.g., the home). In fact, in this case, mobility may be limited to the first floor of the home if the individual is unable to climb the stairs to the second storey. One would hypothesize that mobility could be improved in this situation by providing further aids to deal with environmental challenges (e.g., crutches to use on the stairs, a scooter for short-distance outdoor travel, handi-transit or friends to provide rides) and/or by enhancing physical status by improving stamina for using walking aids and maximizing bone healing to minimize the need for walking aids.

Financial factors also directly influence mobility and interact with other key determinants to impact overall mobility status. Research indicates that people with lower incomes are at greater risk for mobility disability (Shumway-Cook et al., 2005). Economic resources dictate activity options away from home and accessible modes of transportation. In this way, financial factors have the potential to influence psychosocial factors (e.g., relationships that are maintained), physical factors (e.g., access to fitness classes) and environmental factors (e.g., income may be a primary factor in determining the location of one's home).

The inclusion of factors within each of the determinant categories is meant to be broad and inclusive. It is recognized that factors will have different levels of relevance depending on individual living situations. The mobility cone used to illustrate this new framework would be "built up" as one moves from childhood to adulthood and acquires greater physical capabilities and financial resources, as well as knowledge, experience, and confidence in being active beyond the home. However, community mobility generally declines at older ages (Cannon Hendrickson & Mann, 2005) and the cone may

erode from the top down, with different determinant wedge stacks remaining (e.g., Figure 2). Of course, it should not be assumed that all individuals will, at some time during their lifetime, have the ability to deal with all factors in all mobility dimensions to allow travel to distant parts of the world or even to areas surrounding the community. For some, financial constraints, cognitive limitations and/or psychosocial factors may preclude this type of travel. As well, it should be noted that mobility portrayed on this three-dimensional cone is meant to represent one's capacity to be mobile. What is most important is having the capability to handle increasingly complex mobility factors (i.e., the "ability to be mobile") for it is these capabilities that define mobility potential (Metz, 2000). In this way, the mobility model may be used to predict impairment and guide the design and interpretation of research studies. Consider, for example, an older woman who restricts her driving to the neighborhood during the winter because she does not feel comfortable driving in cold and snowy weather. The framework would predict that her mobility would be more impaired in the winter because of environmental and psychosocial (self-efficacy) factors, but only in terms of accessing larger life-spaces; no differences would be expected within her home. These kinds of predictions have a number of implications for research, for example, the need to measure all important contextual factors relating to mobility (i.e., potentially at different times of the year) and all types of mobility in order to gain a true picture of overall mobility status.

Although changes in mobility may occur gradually over time, some life events (e.g., fracturing a hip, losing a spouse who was the driver for the couple, changing the location of one's residence) have the potential to change one's mobility profile suddenly and substantially. Usually, mobility is limited to the life-space for which all factors in the

determinant categories are met. However this may not always be the case. An older person with dementia who continues to drive may lack the cognitive ability to be safely mobile beyond walking in the neighborhood, yet by continuing to drive, this person could access more distant locations.

Also intrinsic in the model is the concept that deficits affecting mobility at a particular life-space may be compensated for by altering other determinants affecting mobility at that level. This is in accordance with the competence-press model (Lawton & Nahemow, 1973) which suggests that people will adapt within a range of environmental demands in attempts to maintain control and independence. Backman and Dixon (1992) have suggested that people compensate for real or perceived mismatches in skills and environmental demands by investing more time or effort to improve abilities, by drawing on latent skills, and/or by acquiring new skills. Certain adaptations to driving which may be implemented voluntarily or imposed by license restrictions may also require learning new skills (e.g., to make use of vehicle adaptations or implement driver training recommendations) and/or improving upon existing capabilities (e.g., driving only at night or on roads with lower speed limits; Eberhard et al., 2006; Marshall, Man-Son-Hing, Molnar, Wilson, & Blair, 2007).

Figure 3A depicts a cross-section of the mobility cone at the service community level. The older adult with vision problems who no longer feels safe driving at night is represented with deficits relating to the physical determinants of mobility. However, by modifying the environment (e.g., limiting driving to daylight hours) the individual is able to continue independent community travel (Figure 3B). Advances in technology also continue to provide options which allow older adults to compensate for various mobility-

related deficits (e.g., robotic walkers, driving assistance and vehicle control devices, vision aids, global positioning system (GPS) devices, “kneeling” buses).

Previous research has substantiated that older individuals restrict their walking and/or driving behaviors because of cognitive, psychosocial, physical, environmental and financial influences, which supports the inclusion of these determinants in the mobility model (Rudman et al., 2006; Shumway-Cook et al., 2003; Shumway-Cook et al., 2005). This theoretical framework is presented in the hopes that it will encourage researchers and clinicians to consider broadly the determinants of mobility and the interactions among determinants in order to truly appreciate reasons for mobility impairments in particular life-space contexts and possible compensatory strategies. The model may drive new research to establish the relative importance of mobility determinants in different mobility contexts and lead to the development of further integrative instruments similar to the mobility questionnaire, a tool which includes questions about accessing different levels of life-space and driving habits (Edwards et al., 2009). Further to this, in order to better understand associations among determinants and their influences on indoor and outdoor mobility, new relationships need to be forged among disciplines (e.g., engineers, psychologists, and researchers who use physical activity monitors and GPS devices) to determine more effective ways to deal with challenges in the built environment and psychosocial factors that impact walking and driving habits.

From an applied perspective, the conceptual model can suggest a more holistic approach to mobility assessment and treatment, as well as more broadly to service program delivery and policy. As portrayed in the framework, mobility becomes more complex as one moves away from the home and neighborhood region, yet healthcare

professionals and society in general often do not consider the resources required to enable older adults to maintain independence in the larger life-spaces. Efforts may target discharge from hospital, yet older adults may then be left relatively isolated in their homes. The development of comprehensive mobility teams including, for example, geriatricians, rehabilitation professionals, ophthalmologists, licensing authorities, and social workers could provide more integrative assessment and treatment recommendations to target issues related to transfers, indoor and outdoor walking, driving, and options for alternate forms of transportation.

CONCLUSION

Given that mobility is fundamentally important to older adults being able to maintain their physical and psychological health (Groessler et al., 2007; Yeom et al., 2008), it is important that we are able to conceptualize mobility in a broad fashion that applies to all older adults, regardless of living situation or functional ability. The new theoretical framework recognizes that all forms of movement are important and that many inter-related determinants influence mobility. This conceptualization of mobility may drive new interdisciplinary research that recognizes the complex interactions between determinants of mobility in different mobility contexts. From a clinical perspective, it provides a more holistic view of mobility and may, thus, promote more effective assessment and treatment practices. Lastly, the framework may help shape program delivery and policy in order to enable older adults to live independently by maintaining their mobility for as long as possible.

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FIGURES

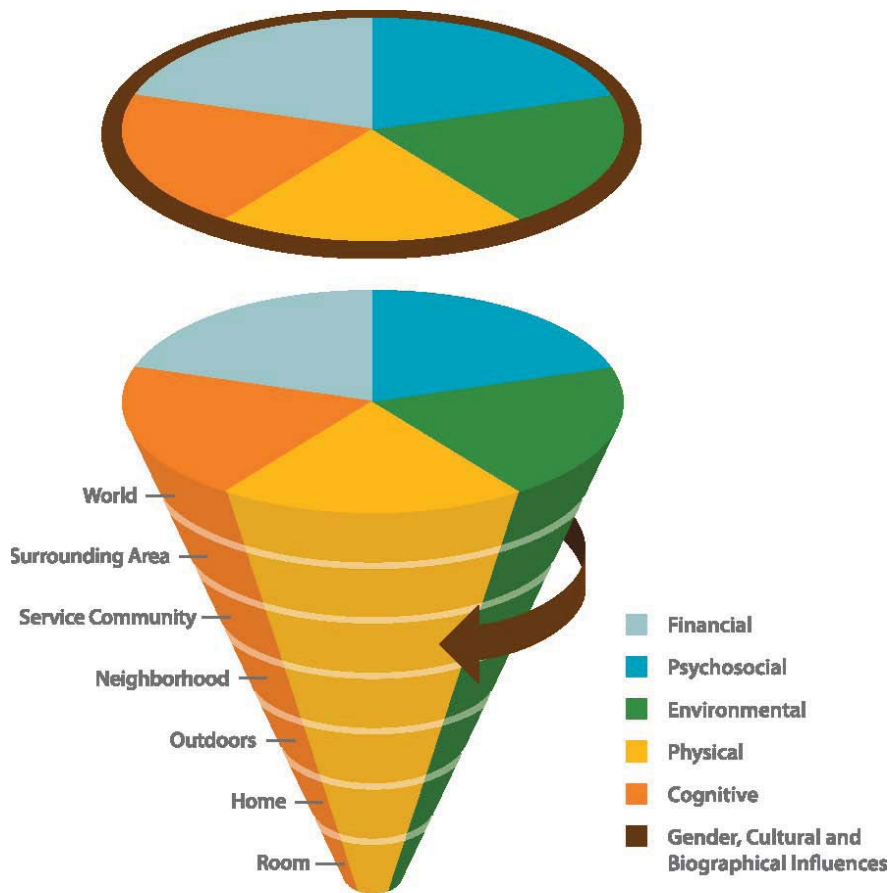


Figure 1. Conical model of the theoretical framework for mobility in older adults

illustrating seven life-space locations (ascending in order of increasing distance from the room where one sleeps), each of which is composed of mobility determinants related to cognitive, psychosocial, physical, environmental, and financial factors. A ring representing gender, culture, and biographical influences surrounds the entire cone exerting influence on all of the mobility determinants. A cross-section of the cone is illustrated above.

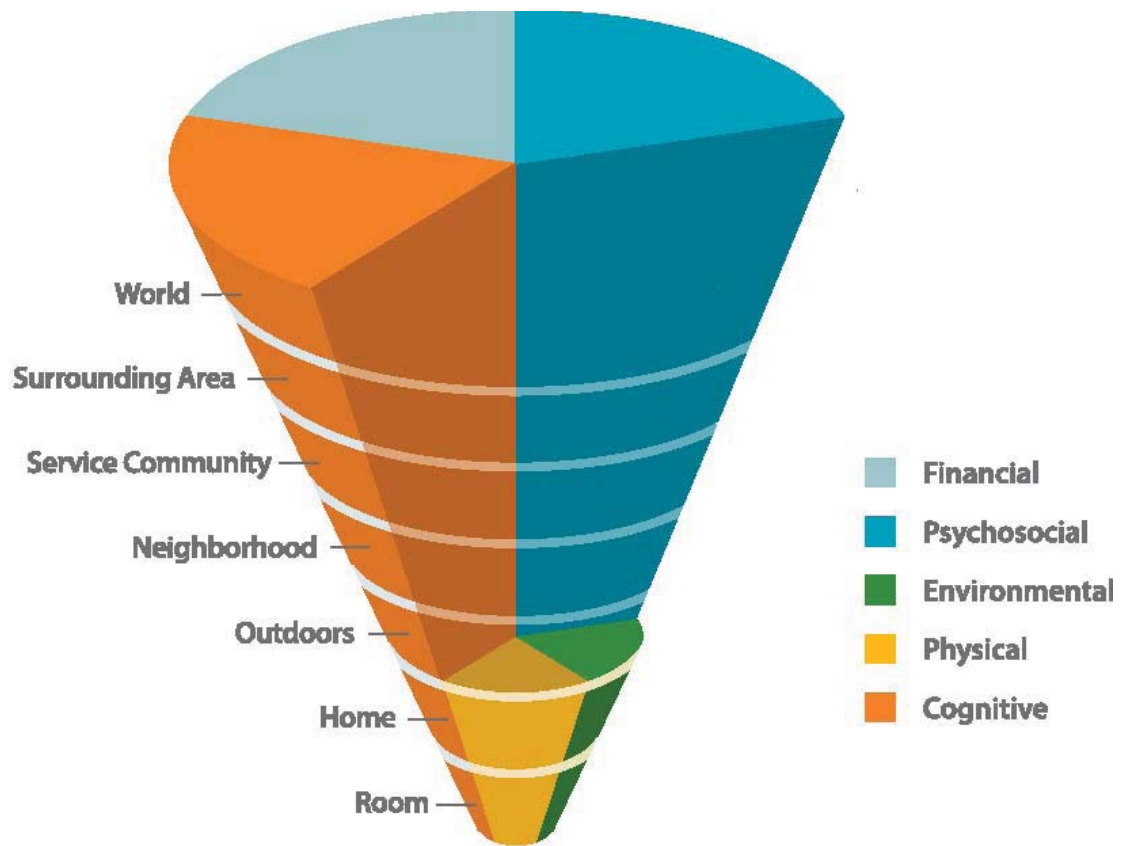
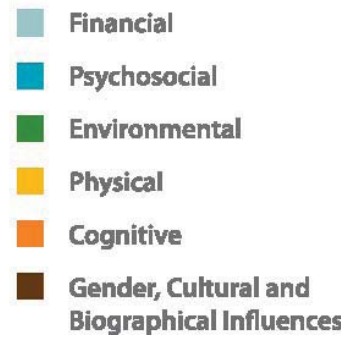
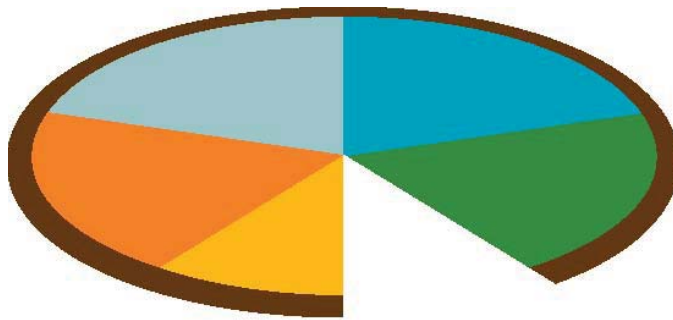


Figure 2. Mobility cone demonstrating impairments in environmental and physical determinant categories (mobility limited to the home).

A



B

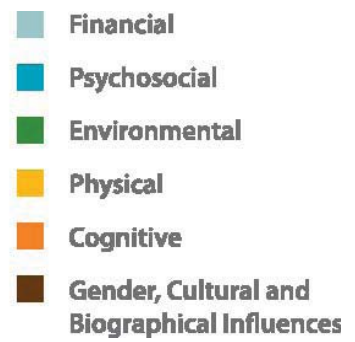
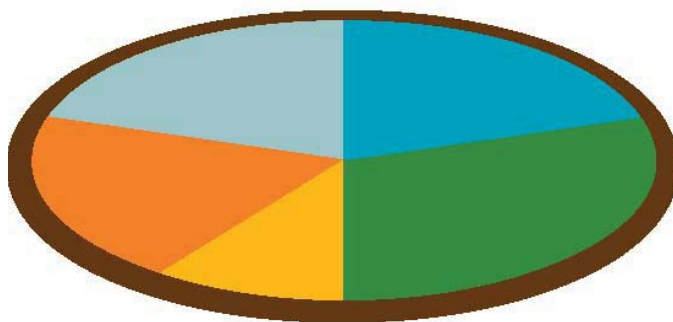


Figure 3. Cross-section of mobility cone at the service community level. (A) portrays deficits in physical determinants and (B) portrays compensation for these deficits with modifications in environmental factors.

**CHAPTER 3: MONITORING MOBILITY IN OLDER ADULTS USING
GLOBAL POSITIONING SYSTEM (GPS) WATCHES AND
ACCELEROMETERS: A FEASIBILITY STUDY**

Running head: Monitoring mobility with GPS

Key Words: active living, physical activity, aging, walking, driving, community mobility

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Webber, S.C., & Porter, M.M. (2009). Monitoring mobility in older adults using global positioning system (GPS) watches and accelerometers: A feasibility study. *Journal of Aging and Physical Activity, 17*(4), 455-467.

MY CONTRIBUTION TO THE PUBLICATION

I participated in determining the design for this study in consultation with my supervisor and conducted trials with the equipment in order to determine the protocol. I participated in data collection along with two other students working in the lab. I determined the methods used for data analysis in consultation with my supervisor and one other student. I conducted all of the data analysis for this study. With my supervisor's input, I wrote and submitted abstracts based on preliminary data analysis to an international annual conference (2007) and two local conferences (2007 and 2008). I presented the work at all of these conferences. When all data were collected and analyzed, I selected the journal to which we submitted the manuscript, and wrote and revised the manuscript in consultation with my supervisor. When the reviewers' comments were received, I made all the required revisions in consultation with my supervisor and resubmitted the manuscript along with the responses to reviewers. When the manuscript was accepted I made any further required corrections such as those included in the galley proofs.

ABSTRACT

This exploratory study examined the feasibility of using Garmin global positioning system (GPS) watches and ActiGraph accelerometers to monitor walking and other aspects of community mobility in older adults. After accuracy at slow walking speeds was initially determined, 20 older adults (74.4 ± 4.2 yr) wore the devices for 1 day. Steps, distances, and speeds (on foot and in vehicle) were determined. GPS data acquisition varied from 43 minutes to over 12 hr, with 55% of participants having more than 8 hr between initial and final data-collection points. When GPS data were acquired without interruptions, detailed mobility information was obtained regarding the timing, distances covered, and speeds reached during trips away from home. Although GPS and accelerometry technology offer promise for monitoring community mobility patterns, new GPS solutions are required that allow for data collection over an extended period of time between indoor and outdoor environments.

INTRODUCTION

Maintenance of mobility, defined as “the extent of movement within a person’s environment” (Stalvey, Owsley, Sloane, & Ball, 1999, p.460), is fundamental to active aging, allowing older adults to continue to lead dynamic and independent lives (World Health Organization, 2007). Mobility can be conceptualized as a continuum from being confined to bed (immobile) to being capable of traveling to distant locations (Patla & Shumway-Cook, 1998; Stalvey et al.). Research has demonstrated that the ability to move about effectively in one’s home and community surroundings is a strong and early predictor of physical disability in older adults (Fried, Bandeen-Roche, Chaves, & Johnson, 2000; Hirvensalo, Rantanen, & Heikkinen, 2000). The ability to walk at least short distances reduces the risk of mortality, cardiovascular disease, and mobility limitation (Chang et al., 2004; Newman et al., 2006).

Maintaining or improving walking ability is a consistent aim of rehabilitation for all older adults. In addition to being able to get around the neighborhood on foot, older adults rely heavily on being able to drive and/or access public transportation to continue to be mobile within the community (Dickerson et al., 2007). For those capable of functioning on the higher end of the mobility spectrum, it is impossible in the clinical situation to replicate the environmental demands of specific community settings (Patla et al., 1999). Interventions designed to enhance mobility success can only be evaluated through assessments that occur in community environments. Questionnaires have been developed to attempt to measure mobility in community-dwelling older adults. Various instruments have been designed to attempt to measure life space, or the spatial extent of travel within one’s environment (Baker, Bodner, & Allman, 2003; May, Nayak, & Isaacs,

1985; Stalvey et al., 1999; Tinetti & Ginter, 1990). In addition, the Environmental Analysis of Mobility Questionnaire (EAMQ) was developed by Shumway-Cook et al. (2003, 2005) as a self-report mechanism to measure community mobility disability. Both that questionnaire and the life-space tools attempt to consider the many environmental factors (e.g. walking speed and distance requirements, terrain characteristics, weather conditions) and intrapersonal aspects (e.g., functional motor ability, cognitive status) that influence mobility. Although these questionnaires cover many aspects important to mobility, they do not obtain specific information regarding individual trips in the community. For this reason, as well as the fact that questionnaires place a high degree of burden on participants, we were interested in determining whether existing physical activity monitors and global positioning system (GPS) technology might be useful in describing community mobility in older adults.

Accelerometers and pedometers have typically been used to quantify physical activity (see the review by Tudor-Locke & Myers, 2001). They provide objective measures of steps taken per day and estimates of energy expended during walking. Accelerometer activity counts have been used to categorize physical activity intensity, estimate caloric expenditure, and estimate walking speed (Barnett & Cerin, 2006; Hendelman, Miller, Baggett, Debold, & Freedson, 2000; Pober, Staudenmayer, Raphael, & Freedson, 2006). Although accelerometers have also been used at the wrist and ankle, placement at the hip ensures that they are relatively insensitive to upper body movement and primarily sensitive to capturing information related to walking (Hendelman et al.). Because accelerometers are more sensitive to step detection at slower walking speeds than pedometers, their use has been recommended for quantifying walking in older adults

(Culhane, O'Connor, Lyons, & Lyons, 2005; Le Masurier & Tudor-Locke, 2003).

Accelerometers can provide step counts within user-defined time frames (e.g. every minute) and are capable of storing data for many consecutive days without requiring interaction between the wearer and the device (Troost, McIver, & Pate, 2005). However, data gathered with an accelerometer are limited to step or activity counts and do not provide information about mobility in different environments. For example, without additional information it is impossible to determine where the steps were taken (e.g., indoors or outdoors, in the home, or in the community).

Satellite-based GPS technology, however, offers potential advantages for studying movement in the community, which may involve outdoor walking, as well as the use of various forms of transportation (Elgethun, Yost, Fitzpatrick, Nyerges, & Fenske, 2007; Le Faucheur et al., 2007; Phillips, Hall, Esmen, Lynch, & Johnson, 2001; Rodriguez, Brown, & Troped, 2005). GPS receivers provide information about location, speed of movement, and elevations encountered and may be used to validate physical activity diaries (Rodriguez et al.) and estimate free-living walking capacity (Le Faucheur et al., 2007, 2008). However, GPS reception is frequently interrupted in “urban canyons,” under heavy tree canopy, and inside or proximal to large buildings (Elgethun, Fenske, Yost, & Palcisko, 2003; Phillips et al.; Rodriguez et al.). Receiver interference also occurs in proximity to power substations and microwave ovens (Elgethun et al., 2003). For these reasons another monitor should be used in conjunction with GPS to ensure detection of indoor mobility. Although a few studies have utilized GPS devices and accelerometers in combination to identify types and locations of outdoor physical activities (Rodriguez et al.; Troped et al., 2008) they have not been used together in older adults. The purpose of

this exploratory study was to determine the feasibility of using GPS watches and accelerometers in combination to gather information about walking and other forms of transportation over the course of a day to see whether these technologies may ultimately be useful to measure community mobility patterns in older adults.

METHODS

Participants

A convenience sample of 20 older adults (68-88 years of age) was recruited to participate in this study. Based on previous studies (Elgethun et al., 2003; Elgethun et al., 2007 ; Phillips et al., 2001; Rodriguez et al., 2005), this sample size was thought to be adequate to provide information regarding a variety of mobility levels and patterns of movement in different locations. All participants lived independently in the community (5 in houses and 15 in apartments); 4 used walking aids; and on average they reported about two current health problems. All participants provided written consent. Ethical approval was granted by the Education/Nursing Research Ethics Board at the University of Manitoba, Winnipeg, MB, Canada.

Equipment

Garmin Forerunner 205 GPS watches ($53.3 \times 17.8 \times 68.6$ mm, 77 g, Garmin International Inc., Olathe, KS) were set to “smart recording” mode, resulting in variable recording frequencies and up to 10 hr of data collection. ActiGraph model 7164 dual-mode uniaxial accelerometers ($50.8 \times 38.1 \times 15.2$ mm, 37.5 g, ActiGraph, LLC, Pensacola, FL) were initialized to record in 1-min epochs and were worn centered over the right thigh in a fitted pouch attached to a belt around the waist. All accelerometers were calibrated by the manufacturer before being used in the study.

Accuracy Tests

Accuracy of the GPS watches' average speed determination and accelerometer step counts were assessed simultaneously with paced outdoor walking tests conducted by research staff (set distances 200 or 400m at 0.45 m/s, 0.67 m/s, 0.89 m/s, 1.33 m/s, 1.67 m/s and 2.08 m/s). The error associated with GPS watches' average walking speed and calculated walking speed (known distance/time) was very low (coefficients of variation [CV] of the measurement errors were 0.8%, 1.5%, 2.0%, 0.7%, 1.4% and 1.3% for the respective speeds). These results are consistent with those previously reported for other Garmin units (Barnett et al., 2006; Le Faucheur et al., 2007). Although ActiGraph accelerometer step counts have been shown to be accurate at speeds as low as 0.9 m/s under controlled conditions (Le Masurier, Lee, & Tudor-Locke, 2004; Le Masurier & Tudor-Locke, 2003), studies have also found that they significantly underestimate steps at speeds only slightly lower (Esliger et al., 2007; Storti et al., 2008). Our results demonstrated that ActiGraph step counts were associated with very low error rates at speeds above 0.90 m/s (CV = 1.1%, 0.3%, and 0.8% for speeds ranging from 0.90–1.33 m/s, 1.34–1.67 m/s, and 1.68–2.09 m/s, respectively), but they performed very poorly at lower speeds (CV = 32.7%, 34.7%, and 7.4% for speeds <0.45 m/s, 0.46–0.67 m/s, and 0.68–0.89 m/s).

Protocol

At a mutually convenient time, study staff delivered a Garmin GPS watch (battery fully charged) and an ActiGraph accelerometer to each participant and provided very basic information about how to wear the equipment the following day as the participant went about his or her usual routine. Participants were shown how to position the accelerometer

on the belt around their waist, how to power on the watch, and how to start the stopwatch to enable GPS data collection. They were given no further instruction regarding the operation of the equipment and were told to disregard any messages that might appear on the face of the watch during the course of the day. The accelerometers were preprogrammed to start recording at 4:30 a.m. After wearing the devices for 1 day, participants completed a short questionnaire regarding their experiences with the equipment and listed trips taken away from home (by time of day and type of transportation).

Data Processing and Analysis

GPS data were uploaded to MotionBased software (Sausalito, CA) and saved in formats appropriate for Google Earth (Google, Inc., Mountain View, CA) and SigmaStat (version 3.10, Systat Software Inc., San Jose, CA). Google Earth's mapping service was used to define activity that took place at home or away from home. When time gaps in GPS data collection were identified in the spreadsheet, corresponding locations (pre- and post gap) were determined on the maps.

ActiGraph analysis software was used to determine steps per minute from the cycle counts recorded on the accelerometer. Steps per day were determined in SigmaStat. Accelerometer data were time-matched with GPS data. Outdoor walking distances were identified in the GPS data for time periods that coincided with accelerometer activity. GPS data were also scanned to determine periods of vehicle travel (speed >3 m/s for > 1 min) and distances traveled by vehicle. Data were checked for overlap between periods of accelerometer activity and vehicle travel, and steps detected during these periods were subtracted from the daily totals.

RESULTS

GPS Data

Overall, some GPS information was successfully recorded for 18 of the 20 participants (Table 1). The average total duration of GPS data capture was 7.7 ± 4.1 hr, with 55% of participants successfully acquiring some data for more than 8 hr (range from 43 min to 12 hr and 19 min). Google Earth maps provided detailed information about participants' walking and driving routes. Representative GPS data have been included in Figure 1. The Google Earth map depicts 1 participant driving to a golf course, walking the nine-hole course, and driving away. Speed of movement (in vehicle and on foot) is displayed in the lower part of the figure. Letters *a* and *c* correspond to speeds associated with periods of driving, and letter *b* corresponds to speeds associated with time spent walking on the golf course. Figure 2 depicts a magnified view of movement on the southeast hole on the golf course. The graph displays speed of movement and steps per minute for each minute spent on the hole. Locations coinciding with brief periods of reduced movement speed between the tee box and the green are denoted as 1-4, which may correspond with separate golf shots and the resultant reduced speed that accompanied preparing to hit the golf ball. The steps accumulated over each minute are plotted at the end of the minute (e.g., step number plotted at 13:36 represents the sum of steps taken between 13:35 and 13:36).

Although it was possible for the GPS unit to acquire a signal when some participants were inside some buildings (which presumably depended on the participant's location, type of building construction, and the number of windows), the signal was frequently lost when participants went indoors. Delays in GPS data acquisition were

encountered at start-up (turning the GPS watch on indoors resulted in “cold,” indoor starts) and when participants left buildings.

Data from the 8 older adults who acquired more than 10 hr of GPS data (GPS watch start time to stop time) were analyzed to investigate the variable data-recording rate that resulted from using the “smart recording” setting on the Garmin watch. For these 8 participants, 97% of successive data points were recorded within 1 min of each other. Analysis of these data points demonstrated that the average duration between GPS data points was 10.35 ± 1.58 s. Median times between data points varied between 6 and 8 s, a frequency similar to that used in other GPS data-acquisition studies (Phillips et al., 2001; Rodriguez et al., 2005).

Accelerometer Data

Accelerometer data indicated that participants’ step counts ranged from 3,564 to 23,888 steps per day ($10,012 \pm 5,026$) with the longest walk for each participant (occurring indoors or outdoors) lasting from 8 to 53 min (26.2 ± 13.9 min).

Time-matched GPS and Accelerometer Data

Average walking speed was determined for participants who walked for more than 10 min consecutively outdoors when accelerometer and GPS data were successfully captured simultaneously. Average walking speed for the 11 participants who met these criteria was 1.27 ± 0.32 m/s (range 0.76-1.72 m/s). Analysis of the periods of vehicle travel (determined by velocity threshold) indicated that accelerometers recorded $793 \pm 1,205$ steps erroneously while participants traveled in vehicles.

Questionnaires

All participants rated the GPS watch and accelerometer easy to use and reported that they wore the equipment for the full day. In addition, all reported that the accelerometer was comfortable to wear on a belt. Nineteen of 20 rated the GPS watch comfortable to wear. On the questionnaires, no participants reported any concerns with using the equipment or with having their mobility tracked. Although there are many issues emerging relating to the ethics of using GPS to track human movement, concerns largely focus on situations in which the monitor is in a position of power over the person being monitored (e.g., parents monitoring children, law-enforcement officials monitoring criminals, and caregivers monitoring people with dementia; Michael, McNamee, & Michael, 2006). In describing the methods of data collection to our participants, we made it clear that we would potentially be able to monitor their location and speed throughout the entire day.

DISCUSSION

GPS receivers, which are capable of providing information about how often people leave their homes, when they leave their homes, where they travel, how they travel (on foot or in vehicle), and how quickly they move, offer promise for ultimately being able to objectively monitor community mobility in older adults. They collect detailed information about individual trips away from home that cannot be attained with existing questionnaires. However, results from this study indicate that advances in technology are required before it will be possible to gather complete data over the course of many days, which may then be used to describe older adults' mobility patterns.

Previous studies have demonstrated that GPS devices function well when used during discrete (20- to 45-min) outdoor bouts such as walking (Le Faucheur et al., 2007,

2008; Troped et al., 2008), cycling, in-line skating or in a vehicle (Troped et al.). For example, GPS receivers have successfully been used to measure outdoor walking capacity (in terms of distances walked and duration of walk and rest periods) in older adults with peripheral arterial disease (Le Faucheur et al., 2008). In addition, it would be anticipated that GPS devices would function well and return detailed information about extended bouts of outdoor activity (e.g., golfing, orienteering).

However, difficulties have been encountered in GPS studies that attempted to monitor people's movements and locations over much longer periods of time and between indoor and outdoor positions. Rodriguez et al. (2005) attempted to use spatial data from GPS units to complement physical activity measures detected by accelerometers. Bouts of physical activity that were easily identified in the accelerometer data were time-matched to the GPS data to determine and classify the location of physical activity bouts (e.g. indoors, outdoors, and location relative to street and land use). In that 3-day study of 35 young adults, 11 participants had incomplete GPS data. In another study, conducted by Phillips et al. (2001), GPS units were used to track 25 participants' locations to assess potential environmental exposure over a 24-hr period (Phillips et al.). Data were captured for only 30% of the total monitoring time and 14 participants had less than 8 hr of data. Many of their waist-worn GPS units suffered apparent battery failure. In our study, with a newer device, GPS information was collected for over 8 hr in 55% of participants. However, within this time frame, a number of participants experienced gaps in GPS signal acquisition that may represent a lost signal related to prolonged indoor time or could represent failure of the GPS unit to collect information related to outdoor movement. It is known that some participants experienced significant delays in satellite

lock-in on initially starting the watches if they presumably remained indoors for a long period of time after waking in the morning (see Table 1, Participants 2, 10, and 18), and this may also have occurred midday after they exited buildings. Other problems including premature battery failure and accidental bumping of the buttons on the face of the watch may also have contributed to lost data.

Despite the problems with GPS data acquisition, the use of GPS technology provides the opportunity to collect much more detailed and relevant information about mobility in community environments (e.g. where an individual walks, how quickly he or she must cross a street, what degree of incline is encountered) than step or activity counts monitored by accelerometers or pedometers. As is evident in Figure 2, GPS information gathered as an individual walks provides much more detailed information about the experience than step counts measured by an accelerometer. When using this equipment with older adults, it is also important to consider the fact that accelerometer accuracy may be compromised by age-related changes in mechanical gait parameters (e.g., slower, shuffling gait or changes related using a walking aid) that influence acceleration in the vertical plane. GPS measurements of speed and displacement would not be sensitive to these changes.

Using both GPS devices and accelerometers together provides the most complete information about mobility in community environments. Because there is a potential for incomplete GPS data acquisition, especially when mobility is monitored over longer periods of time, using accelerometers in combination with GPS devices provides the assurance that at least some form of mobility information will be acquired even when the satellite signal is not available. For example, because GPS units do not function well

indoors, the addition of accelerometers provides the opportunity to acquire information about the number and timing of steps taken in the home and other buildings. From this, it is at least possible to know when an individual was mobile or stationary during their time indoors. This same functionality is also provided if the individual is outdoors and the GPS signal is lost (e.g., in an urban canyon).

Although accelerometers are more sensitive than mechanical pedometers (Le Masurier et al., 2004; Le Masurier & Tudor-Locke, 2003) and their use has been recommended in the older population (Marsh, Vance, Frederick, Hesselmann, & Rejeski, 2007), the greater sensitivity of the device can result in false step recordings during travel in a vehicle. When GPS receivers are used in conjunction with accelerometers, periods of vehicle travel can be identified and erroneous step or activity counts can be removed from the data. In addition, using a GPS device in combination with an accelerometer allows for detection of outdoor gait velocities below the threshold for accelerometer use (0.86 m/s), which occurs frequently in frail older people (Lindemann et al., 2007).

There are many different handheld and watch- or phone-based GPS receivers available on the market today. Performance characteristics vary depending on sampling frequency, lock-in time to acquire or reacquire the GPS signal, battery life, size, weight, method of wearing the device, and cost. Because most of these devices have not been developed for research purposes, no systematic comparison of performance features among GPS units has been done. Many of the published studies to date that have used GPS to track human movement required participants to wear relatively large, cumbersome devices. GPS data recorders have been attached on the belt along with external battery packs and antennae fixed on the shoulder (Phillips et al., 2001).

Alternatively, GPS data recorders, batteries, and antennae have been sewn into vests and bib overalls for children to wear (Elgethun, et al., 2003). Other studies have positioned the GPS devices in a small backpack with the antenna attached to a shoulder strap (Le Faucheur et al., 2008; Troped et al., 2008). We chose GPS watches for use in this study because of their relatively small size, light weight, and unobtrusive nature. Garmin GPS watches were designed to be used by athletes to track distances and speeds during outdoor workouts. Despite the 10-hr battery life, they are not intended to be used for extended periods of time with wearers moving between indoor and outdoor locations without stopping to allow the watch to lock in to satellites each time the user goes outdoors. To minimize influences on normal everyday mobility, we decided that participants should not be instructed to allow time for satellite lock-in upon going outdoors during the day. This procedure likely resulted in lost data in some instances and explains the relatively frequent occurrence of gaps in GPS data collection. In addition, battery life could have been maximized by discontinuing GPS data collection when indoors and restarting on going outdoors. However, we felt that this would have added unreasonable burden to the participants, and they might not have remembered to restart the watch when returning outdoors, which also would have resulted in missing data.

In future studies participants could be given more detailed instructions about using the GPS watch in order to gain more complete GPS data. For example, participants could be told to stand stationary for 2-3 min to ensure satellite lock-in every time they exit a building or emerge from an area consisting of tall buildings or substantial tree cover. GPS devices could be turned off when participants go indoors to try to prolong battery life and maximize data collection. In this case, of course, it would be essential to

use another mobility monitoring device such as an accelerometer to be able to gain some information about relative movement inside buildings. However, researchers must be aware that putting these requirements on individuals may in fact influence participants' mobility patterns (in terms of the number and duration of trips taken from home), or result in missing data (e.g., if people do not remember to turn the devices back on after they have been turned off).

In conclusion, although combined GPS and accelerometer technologies have been used successfully to gather detailed information about discrete bouts of outdoor activity (physical endeavors as well as driving), the same success has not been realized in studies that have attempted to monitor functional everyday human movement over an extended period of time. Although these technologies offer the potential to accurately monitor community mobility patterns in older adults, some issues still need to be addressed before this is realized.

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TABLES**TABLE 1.** Details of GPS Data Capture for 20 Older Adults Over 1 Day

Participant	Delay in GPS capture after watch turned on (min)	Duration between initial and final GPS data acquisition (hr)
1	79.88	10.07
2	319.68	1.12
3	9.93	2.02
4	5.42	11.65
5	No GPS data	No GPS data
6	67.33	10.55
7	3.17	12.32
8	11.87	11.50
9	0	0.72
10	307.88	2.88
11	54.67	3.38
12	18.98	11.65
13	12.98	7.63
14	No GPS data	No GPS data
15	Unknown	9.93
16	18.10	4.46
17	37.28	8.58
18	182.18	8.22
19	64.46	10.53
20	67.00	10.67

FIGURES

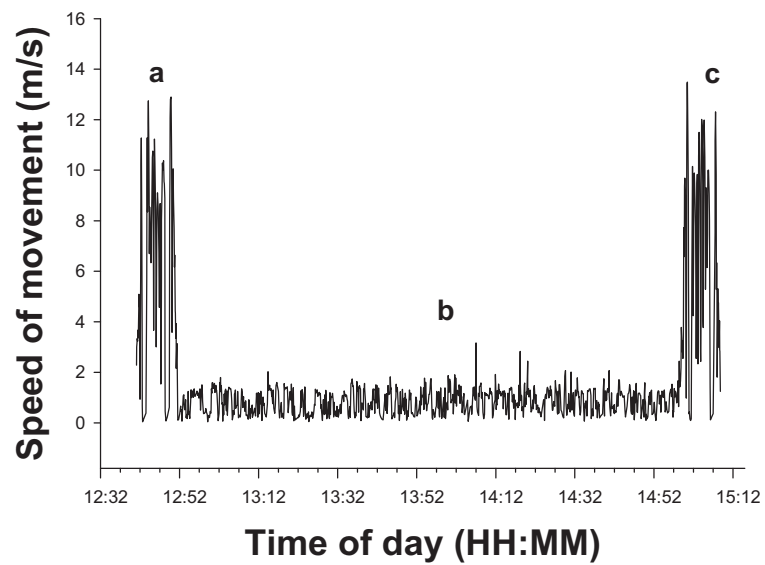
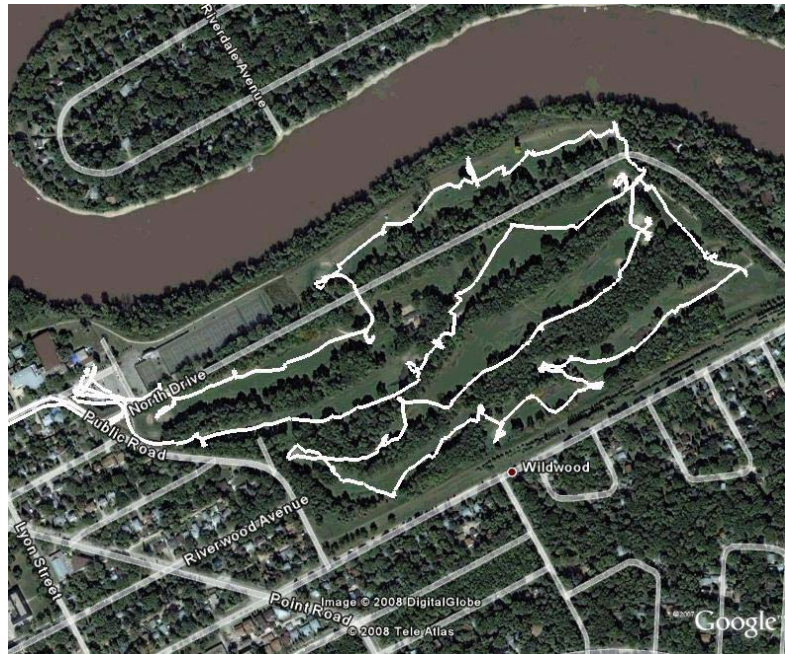


Figure 1 - Google Earth map (top) and speed data (bottom) obtained from part of the GPS data for one participant. Letters *a* and *c* correspond to driving to and from the golf course, and letter *b*, to walking the nine-hole course. Satellite image courtesy of DigitalGlobe.

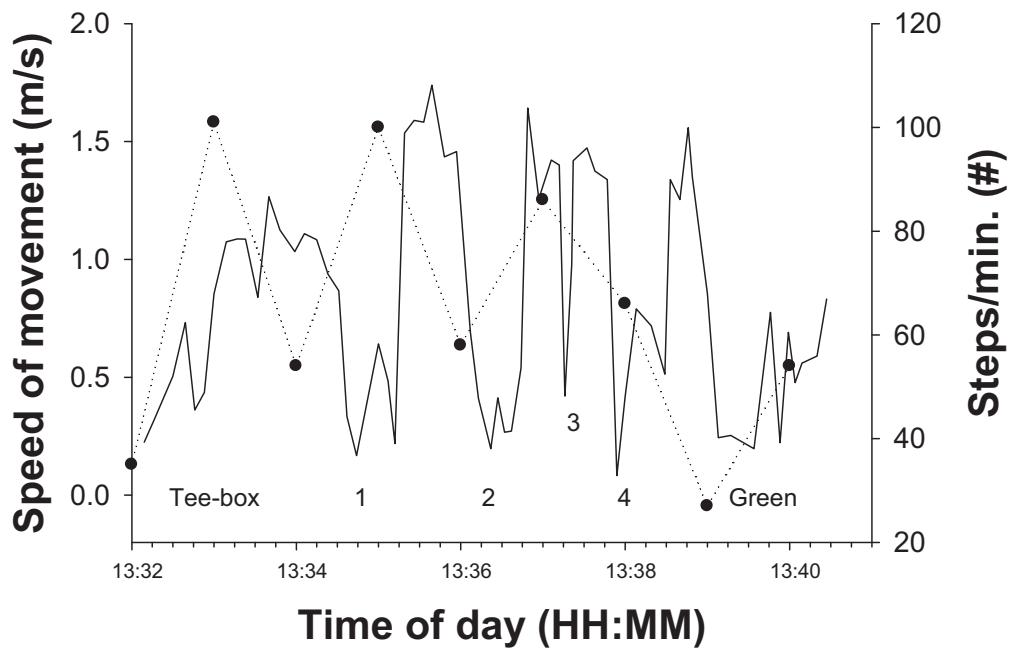


Figure 2 - Google Earth map (top) and speed/steps data (bottom) data obtained from the southeast hole on the golf course. Numbers 1-4 represent brief periods of reduced speed that may coincide with individual shots on the hole. Steps per minute are plotted for each minute (connected by the dotted line). Satellite image courtesy of DigitalGlobe.

CHAPTER 4: RELIABILITY OF ANKLE ISOMETRIC, ISOTONIC, AND ISOKINETIC STRENGTH AND POWER TESTING IN OLDER WOMEN

The manuscript in Chapter 4 has been accepted for publication pending minor revisions.
Webber, S.C., & Porter, M.M. Reliability of ankle isometric, isotonic, and isokinetic strength and power testing in older women.

MY CONTRIBUTION TO THE MANUSCRIPT

I participated in determining the design for this study in consultation with my supervisor and conducted trials with the equipment in order to determine the protocol. I also developed the documents for submission to the ethics committee, the intake questionnaire, and scripts/forms for testing in consultation with my supervisor. I recruited subjects and spoke with the majority of potential subjects to determine their eligibility to participate in the study. I was responsible for all of the data collection in this study. I determined the methods used for data analysis in consultation with my supervisor and conducted the majority of data analysis for this study. I trained a summer student to assist with some of the initial data reduction. With my supervisor's input, I wrote and submitted two abstracts based on preliminary data analyses to two national conferences (2009). I presented the work at the conferences. When all of the data were collected and analyzed, I selected the journal for submission of the manuscript, wrote and revised the manuscript in consultation with my supervisor, and submitted the manuscript for peer review.

ABSTRACT

Background: Ankle strength and power capabilities influence physical function (e.g., walking and balance) in older adults. Although strength and power parameters are frequently measured with dynamometers, few studies have examined the reliability of different types of contractions.

Objective: The purpose of this study was to examine relative and absolute intrarater reliability of isometric, isotonic, and isokinetic ankle measures in older women.

Design: This was a prospective, descriptive methodological study.

Methods: The following dorsiflexion (DF) and plantar flexion (PF) measures were assessed twice (7 days apart) by the same examiner in 30 older women (mean 73.3 ± 4.7 years of age): isometric peak torque and rate of torque development (RTD); isotonic peak velocity, average acceleration, and peak power; and isokinetic peak torque and peak power ($30^\circ/\text{s}$ and $90^\circ/\text{s}$). Several statistical methods were used to examine relative and absolute reliability.

Results: ICC values for DF tests (0.76 to 0.97) were generally higher than ICC values for matched PF tests (0.58 to 0.93, $p < 0.001$). Measures of absolute reliability (e.g., coefficient of variation [CV_{TE}]) also demonstrated more reliable values for DF tests (5-18%) compared to PF tests (7-37%, $p < 0.001$). Isotonic peak velocity tests at minimal loads were associated with the lowest CV_{TE} and RLOA values for both DF (5% and 14% respectively) and PF (7% and 18% respectively). Isometric RTD variables were the least reliable ($CV_{TE} = 16\text{-}37\%$).

Conclusions: Test-retest reliability was adequate for determining changes at the group level for all strength and power variables except isometric RTD. However, better

reliability would provide more confidence in interpreting clinically meaningful change within individuals.

INTRODUCTION

Loss of neuromuscular mass, strength and power are closely associated with functional decline, loss of independence and mortality in older adults.¹ Some studies suggest that the rate of loss of neuromuscular power exceeds the rate of loss of strength with age.^{2,3} As well, cross-sectional studies have demonstrated that functional capabilities such as the ability to get up and down from a chair, climb stairs, and walk quickly may be more closely associated with power than strength⁴⁻⁶ and that loss of power may be more related to the etiology of falls.⁷ In order to determine the true nature of the relationships among changes in strength and power with age, and changes in function, reliable testing techniques are required.

Isokinetic dynamometers are frequently used to assess neuromuscular function because they provide detailed torque, velocity and position data with high mechanical reliability.⁸ Although researchers have investigated the reliability of dynamometer strength assessment in older adults, the focus has largely been on the knee.⁹⁻¹⁴ The distal leg muscles exhibit reductions in strength and power with aging^{15,16} and are important for walking,¹⁷ maintaining balance, avoiding falls,^{7,18} and braking a vehicle. However, there have been only a few evaluations of the reliability of ankle strength protocols,^{11,12,19} and only one assessment concerned with reliability associated with measures of ankle power in this population.¹¹

Because power is defined as work (force \times distance) divided by time, it is influenced by both strength and speed. Peak or average power (Watts = Nm \times radians/s) can be

measured using either the isokinetic or isotonic modes on a dynamometer. Other indirect measures that may be associated with the ability of the neuromuscular system to generate force or torque rapidly can be evaluated using either the isometric mode (rate of torque development [RTD])^{16, 21} or the isotonic mode (velocity, acceleration).^{2, 20} Hartmann et al.¹¹ have reported reliability scores for ankle average isokinetic power tests, but reliability of isotonic and isometric measures related to power have not previously been investigated in older adults. Preliminary findings suggest that isokinetic evaluations of power may not be as reliable as strength measures,^{9, 11} and that RTD averaged over a specified range (e.g., from 30% to 60% of peak torque) may yield more consistent results compared to peak RTD.²¹ Further research is required to compare reliability of strength and power measures and to determine which measures are associated with lower levels of measurement error for use in research and clinical situations.

Physical therapists and other healthcare providers need to be able to properly interpret measurement change to determine the relative effectiveness of different interventions. Test-retest studies provide information about relative reliability, the degree to which repeated measurements reveal consistent ranking of individuals' scores within a group. Measures of absolute reliability describe individual variability and measurement error, and are therefore important in determining levels for clinically significant change.

Although the literature suggests that power may be more important than strength in terms of function in older adults, very little is known about the reliability of different measures of power or power-related variables (e.g., velocity during isotonic movements). The objective of this study was to determine the relative and absolute intrarater reliability for

ankle strength and power measures obtained using isometric, isotonic, and isokinetic tests on a dynamometer in older women.

METHODS

Subjects

Thirty older women (mean (SD) = 73.3 (4.7) years) were recruited to take part in this study. Participant characteristics are presented in Table 1. None of the participants had previously been tested on an isokinetic dynamometer. Exclusion criteria included acute or unstable chronic disease, and neurological or musculoskeletal impairment that would interfere with testing. A physician's signed Physical Activity Readiness Medical Examination form (PARmed-X) was required when potential contraindications to exercise were identified (e.g., history of cardiac disease, uncontrolled hypertension, hernia, detached retina). At the initial evaluation session, participants provided their written informed consent. Ethical approval for this study was granted by the Education / Nursing Research Ethics Board of the University of Manitoba.

The median number of co-morbidities reported by participants was 2 (range = 0 to 3). Arthritis (18), hypertension (14), previous cancer diagnosis (5), and diabetes (4) were most commonly reported. Seven of the participants reported that they had fallen in the past year and one regularly used a cane. The median number of medications prescribed was 1 (range = 0 to 3). None of the participants changed their medication(s) during their testing week.

Procedure

Upon admission to the study, participants completed a health/demographic questionnaire. All other tests were conducted twice, by the same examiner, exactly one week (7 days) apart, at the same time of day. Resting blood pressure, heart rate, body mass, height, and active range of motion for the ankle were measured using standard procedures.

Dynamometer Tests

DF and PF torque, position, and velocity were measured using a Biodex System 3 Pro* dynamometer. The mechanical reliability of this dynamometer has been shown to be excellent.⁸ Calibration of the dynamometer was verified each day prior to testing. Participants warmed-up by walking for 4-minutes on a treadmill before being seated on the Biodex (right lateral malleolus aligned with the axis of rotation, right knee flexed 45-55°, trunk reclined 5° from vertical). Only the dominant leg (defined as the leg that would be used to kick a ball) was tested. All subjects reported right leg dominance. Each participant kept her arms folded across her chest during testing and belts provided stabilization around the waist and over the right thigh. The end-limits of range of motion were set at 10° DF and 30° PF for all tests. Subjects performed isometric tests, concentric isotonic tests, and concentric isokinetic tests, always in that order, with DF contractions preceding PF contractions. Standardized, consistent verbal encouragement was provided for all tests.

* Biodex Medical Systems Inc., 20 Ramsey Rd, Shirley, NY 11967

Isometric Tests

Following 3 practice trials, 3 maximal voluntary isometric contractions were performed for DF (at 25° PF) and then for PF (at 0°). Test angles were chosen based on previous literature^{2, 12, 16, 22} to correspond approximately to the angles at which maximum isometric torques can be produced. Subjects were strongly encouraged to contract “fast” and to hold each contraction for 3-5 seconds. They were given 90s of rest between trials.

Isotonic Tests

The dynamometer was then switched to the isotonic mode. DF and PF movements were each performed against two set resistance levels: first, a minimal resistance level (DF = 1 Nm, PF = 15 Nm) and then a load equal to 50% isometric peak torque. These resistance levels represent the boundaries of those previously published for the dorsiflexors.²

Isotonic DF trials were initiated from 30° PF, and PF trials were initiated from 10° DF. Once again, participants were strongly encouraged to move “fast.” Two practice trials preceded 5 test trials for each of the 4 conditions (DF and PF, 2 loads each). Thirty seconds of rest was provided between all repetitions.

Isokinetic Tests

Maximal effort isokinetic concentric DF and PF tests were then performed at 30°/s and 90°/s (in that order). The passive mode on the dynamometer was used for these constant velocity tests because it provided a passive return to the start position after each concentric contraction (therefore all concentric DF contractions were completed before PF testing began, i.e., DF and PF contractions were not performed immediately back-to-back). As well, many subjects would not be able to generate enough DF torque to overcome the torque related to the combined mass of the foot and footplate to initiate DF

movement if the isokinetic mode was used. Matching concentric contractions with the onset of passive movement avoided this difficulty. Participants were given 3-5 sub-maximal practice trials for familiarization before 5 test trials were conducted for each movement, at each velocity. A 2-minute rest period was provided between velocities. One subject was unable to generate torque in the DF direction at the higher velocity, therefore DF peak torque and peak power values were recorded at 90°/s in 29 subjects.

Data Analysis

Biodex data were collected at a frequency of 100 Hz and exported for analyses in SigmaPlot[†] (version 11.0). All variables used in the analyses were means of the repetitions performed (isometric measures = mean of 3 repetitions, isotonic and isokinetic measures = mean of 5 repetitions). Because all scores inherently include some random error (which either adds to or subtracts from the true score), using mean scores may reduce the magnitude of the error component contributing to the total score.²³

For each isometric contraction, peak torque (Nm) was identified and RTD (Nm/s) was calculated by 2 methods. Change in Nm/time was first determined from 0-50% peak torque and then from 40-80% peak torque (Figure 1). Calculating RTD over a specified range has been shown to be more reliable than determining peak RTD.²¹ These specific ranges were chosen to allow comparison of RTD reliability between relatively steep sections of the isometric torque curve (0-50% peak torque) and less steep sections (40-80% peak torque).

[†] Systat Software Inc., 1735 Technology Drive, Ste. 430, San Jose, CA 95110

For each isotonic contraction, peak velocity ($^{\circ}/s$), average acceleration (peak velocity/time to reach peak velocity, $^{\circ}/s^2$) and peak power (Watts = Nm \times radians/s) were determined (Figure 2). Although the dynamometer was set to the isotonic mode for these tests, torque is not held absolutely constant throughout the range of motion on this setting (Figure 2). As has been noted previously^{24, 25}, the sampling rate (100 Hz) of the dynamometer does not permit adjustments in speed to occur fast enough to result in a continuous torque level.

For each isokinetic contraction, peak torque (Nm), and peak power (Watts = Nm \times radians/s) were analyzed.

Statistical analyses were conducted using SPSS[‡] (Version 15.0) and SigmaPlot[†]. Means and standard deviations were calculated for each variable tested at time 1 and time 2. Paired t-tests were conducted to look for significant bias between test sessions ($P < 0.05$). The intraclass correlation coefficient ($ICC_{(2,k)}$) was used to evaluate both systematic and random errors that may affect relative test-retest reliability.²⁶ Specifically, $ICC_{(2,3)}$ was used for all isometric measures because they were based on the mean of 3 repetitions and $ICC_{(2,5)}$ was used for all isokinetic and isotonic measures that were scored as the mean of 5 repetitions. Normality of the difference scores was assessed using the Shapiro-Wilk test. Data were checked visually with Bland-Altman plots for the presence of heteroscedasticity, and Pearson correlation coefficients were calculated between absolute

[‡] SPSS Inc., 233 S. Wacker Drive, Chicago, IL 60606

[†] Systat Software Inc., 1735 Technology Drive, Ste. 430, San Jose, CA 95110

differences and the means of the two tests. The standard error of the measurement (SEM) and limits of agreement (LOA) were calculated along with the coefficients of variation of the typical error (CV_{TE}), the ratio limits of agreement (RLOA), and the minimal detectable change (MDC). SEM was determined as the square root of the residual mean square error term from the analysis of variance table.²⁷ The LOA were calculated as the systematic bias (mean difference between two test sessions) \pm the random error component ($1.96 \times$ SD of the difference between the two test sessions). This random error component was equal to the MDC_{95} value (calculated as $1.96 \times \sqrt{2} \times SEM^{28, 29}$). Typical error (TE) was calculated as the standard deviation of the differences scores between sessions divided by $\sqrt{2}$.³⁰ CV_{TE} is defined as typical error divided by the mean of all trials from both sessions, multiplied by 100.³¹ RLOA were calculated as a percentage of the mean ($[MDC_{95}/\text{mean of all observations}] \times 100$).

RESULTS

Means and standard deviations (SD) for the isometric, isotonic and isokinetic strength and power variables are presented in Table 2. There were no significant differences between session 1 and session 2 for almost all of the variables, however, PF isometric torque and RTD increased ($P < 0.05$). In addition, changes in DF isotonic average acceleration (1 Nm load) and PF isokinetic peak power at $30^\circ/s$ were very close to being statistically significant ($P = 0.05$ and $P = 0.06$ respectively).

Table 3 reports the reliability data for all DF and PF tests. ICC values for DF tests (0.76 to 0.97) were higher (signed rank test, $p < 0.001$) than ICC values for matched PF tests

(0.58 to 0.93), with the exception of 2 isotonic values (peak power against minimal load and peak velocity against 50% maximum isometric load). Measures of absolute reliability (CV_{TE}) also demonstrated more reliable values for all DF tests (5-18%) compared to PF tests (7-37%) except for the same 2 isotonic measures (signed rank test, $p < 0.001$).

Isotonic peak velocity tests at minimal loads were associated with the lowest CV_{TE} and RLOA values for both DF (5% and 14% respectively) and PF (7% and 18% respectively). Isometric $RTD_{0-50\%}$ and $RTD_{40-80\%}$ demonstrated the highest levels of variability between test sessions for both DF ($CV_{TE} = 16\%$ and 18% respectively and $RLOA = 45\%$ and 50% respectively) and PF ($CV_{TE} = 35\%$ and 37% respectively and $RLOA = 97\%$ and 104% respectively). MDC values, considered to be the minimal amount of change in an outcome measure that can be measured for an individual that is not due to systematic or chance variation in measurement^{28, 32} are included for all variables. Specifically, MDC_{95} values indicate that one can be 95% confident in the true nature of changes that exceed these levels. The LOA were equal to the systematic bias \pm the MDC_{95} value.

Bland-Altman plots (individual subject differences plotted against the mean for both test sessions) were created for all outcome variables to look for systematic bias, outliers, and the presence of heteroscedasticity. Pearson correlation coefficients were not significant for 23 of the 26 tests for heteroscedasticity, but isotonic PF peak velocity (against 50% isometric peak torque), PF $RTD_{0-50\%}$, and PF $RTD_{40-80\%}$ did demonstrate significant positive correlations ($r = 0.45 - 0.65$, $p \leq 0.01$). Reliability statistics associated with the RTD and peak isotonic velocity (against 50% isometric peak torque) variables were

relatively low, therefore other strength and/or power variables should be chosen in test-retest situations. No data transformation was conducted.

DISCUSSION

This study was conducted to establish relative and absolute reliability scores for isometric, isotonic, and isokinetic strength and power-related measures about the ankle in older women. Although the reliability of some of these measures has been previously investigated (e.g., isokinetic tests), other parameters (e.g., isotonic values) have been infrequently reported in the literature with no associated reliability information provided. This study found that isometric, isotonic and isokinetic measures of strength and power were associated with good relative reliability (all ICC's >0.75, with the exception of PF RTD).²³ This indicates that these measures generally exhibit consistency for repeated measurements at the group level.

Isometric peak torque and isokinetic results are most commonly reported in dynamometer studies, whereas RTD and isotonic variables have been less cited. In terms of strength, women in this study obtained DF and PF isometric and isokinetic peak torque values similar to those previously reported.^{11, 12, 16, 33, 34} However, direct comparisons could not be made with the isokinetic peak power data. As well, very little information exists regarding RTD capabilities in older adults. Plantar flexion RTD_{0-50%} and RTD_{40-80%} values recorded in this study were lower than PF RTD_{30-60%} findings reported for younger adults²¹, and also lower than maximum absolute RTD results recorded about the ankle in older adults.^{2, 16}

Few researchers have utilized isotonic dynamometer settings for evaluating velocity, acceleration, and power. In one study, McNeil and Rice²² reported that older men (mean age 84 years) reached peak isotonic DF velocities of 120 °/s, whereas slightly younger men (mean age 64 years) peaked at 154 °/s. DF velocities achieved by women in our study (161 °/s with 1 Nm load) were slightly higher than both of these values, however, the load used in this study (1 Nm) was less than the McNeil and Rice study (20% isometric peak torque) and the starting ranges of motion differed by 5° (30° PF in our study compared to 25° PF in the study by McNeil and Rice).

In our study, virtually all isometric, isotonic and isokinetic DF and PF measures demonstrated good relative reliability. With the exception of PF RTD_{0-50%} ($ICC_{(2,k)} = 0.63$) and PF RTD_{40-80%} ($ICC_{(2,k)} = 0.58$), all ICC values exceeded 0.75 and more than half of the values reached 0.90 or greater. However, it is important to also examine the 95% confidence intervals associated with the ICC's (Table 3) to gain a more thorough understanding of the reliability of these measurements. In the majority of cases (18/26), the lower confidence interval did not fall below 0.70, however, at the worst extreme, PF RTD measurements demonstrated lower confidence limits of 0.20 and 0.12, indicating very poor test-retest reliability. ICC values associated with isokinetic DF and PF peak torque and peak power (0.85-0.97) were very similar to those reported by Hartmann and colleagues¹¹ (ICC range 0.92 – 0.98) when they tested older women and men at 60°/s. Isokinetic DF peak torque ICC values in our study (0.95 at 30°/s and 0.96 at 90°/s) were slightly higher than those previously reported for younger men and women³¹ but virtually

identical to those demonstrated in a group of older men and women (70-85 years of age)³⁵ when testing was conducted at the same velocities (30°/s and 90°/s). This study represents the first to report reliability statistics associated with isometric and isotonic tests about the ankle in older women, therefore no comparisons of these variables could be made.

Measures of absolute reliability describe within-subject variation and the degree to which observed scores will vary with repeated measurements. Random variation responsible for measurement error during testing arises primarily from biological variability (e.g., changes in the subject's mental or physical condition), mechanical variability related to testing equipment, and variability introduced by the tester.³⁰ SEM values describe (in the units of the actual measure) the limits for change required to indicate a real increase or decrease for a group of subjects following some sort of intervention.³⁵ In this study, SEM results took on a wide range of values, related to the magnitude of the mean score for each evaluation. SEM values were similar to those reported in other ankle strength/power studies for isokinetic parameters^{11, 31, 35} and isometric RTD.²¹ Whereas SEM values are expressed in original units, CV_{TE} expresses typical error as a percentage, making it useful for comparing reliability between different measures and across different studies. Values of CV_{TE} for DF and PF isometric peak torque were relatively low in this study (6% and 12% respectively), whereas CV_{TE} results were slightly higher for isokinetic peak torque and peak power results, ranging from 9-14%. These results were similar to those reported by Holmback and colleagues for isokinetic concentric DF peak torque tests in young adults³¹ and older adults.³⁵ In addition, CV_{TE} values associated with isokinetic peak

power tests (9-13%) were similar to those previously reported for DF and PF average power at 60°/s (11-15%).¹¹

LOA, RLOA and MDC₉₅ provide additional, more conservative measures of absolute reliability. Generation of these statistics provides information about the confidence limits associated with measurement error so that, for example, one can state with 95% confidence that an individual's change score that exceeds the LOA represents a true change. As has been noted previously, LOA statistics are identical to systematic bias \pm MDC₉₅ calculations.³⁶ For this reason, only the latter are listed in Table 3. MDC₉₅ values in our study were 4.2 Nm (DF 30°/s), 3.2 Nm (DF 90°/s), 24.2 Nm (PF 30°/s) and 24.7 Nm (PF 90°/s) for isokinetic peak torque measurements. In comparison, Hartmann et al.¹¹ reported slightly lower values of 2.3 Nm and 15.9 Nm respectively, for average isokinetic DF and PF peak torques at 60°/s. The LOA determined in our study (systematic bias \pm MDC₉₅) were virtually identical to those reported by Holmback and Lexell³⁵ (listed as mean difference \pm smallest real difference) for DF peak torque at 30°/s and 90°/s, tested in older men and women. Random error results (equivalent to MDC₉₅ values) reported by Hartmann et al.¹¹ for average isokinetic power were also slightly lower than those reported for peak power in our study. Overall differences among study results were small, and may be related to differences in the subjects (e.g., sex, age), the raters and/or the test protocol itself (e.g., test velocities, measurement of peak versus average power, and subject positioning). Ordway et al.¹² and Hartmann et al.¹¹ reported RLOA values ranging from 24-43% for isokinetic DF and PF peak torque in older adults. In our study, RLOA

for these variables similarly ranged from 30-40%, indicating that individuals must demonstrate relatively large changes over time to be assured of real change.

Using the isotonic mode on a dynamometer may allow for evaluation of functionally relevant contractions in which velocity is not constrained or held constant. However, these settings have been infrequently utilized in clinical practice and research situations.²⁰ For this reason, very little is known about the reliability of these types of tests. This study adds to the literature by providing detailed information about relative and absolute reliability associated with different isotonic parameters. In the present study, isotonic peak velocity and average acceleration were associated with low CV_{TE} values when the load was minimal (CV_{TE} for DF = 5% and 6% for peak velocity and average acceleration respectively, CV_{TE} for PF = 7% and 10% respectively) and slightly higher CV_{TE} values (DF = 13% and 11% and PF = 12% and 15%) when the load was equal to 50% of isometric peak torque. Isotonic peak velocity was associated with less variation compared to average acceleration and peak power variables. Although other researchers have studied peak velocities achieved with different levels of load,^{2, 20} additional reliability data is not available. Further research involving other joint movements and different populations is required to determine whether peak velocity is consistently more reliable than acceleration or isotonic power. In addition, further research may determine whether isotonic variables are generally more reliable when lower loads are used compared to higher loads.

In all but two instances (isotonic peak power against minimal load and isotonic peak velocity against 50% maximum isometric load), DF scores demonstrated better reliability compared to PF scores. This is in agreement with the findings reported by Hartmann et al.¹¹ In both studies, participants were positioned with the knee flexed for PF tests. Although the upper body and thigh were well stabilized with straps, it is conceivable that attempts to extend the knee and/or hip may have also occurred during PF movements, adding variability to these PF measurements that did not occur with DF movements.

It has been suggested that from a functional perspective, increases in RTD may represent one of the most important adaptations that occurs in response to resistance training in older adults.³⁷ That is, the ability to generate moderate forces quickly may be more important than being able to generate high forces, especially when quick action is required (e.g., to regain balance and avoid a fall). Although RTD may be a functionally important variable, our study demonstrated that it had the lowest absolute reliability of all the power-related variables studied about the ankle. Sleivert et al.²¹ reported that average RTD (RTD_{30-60%}) was more reliable than peak RTD, but the ICC for average RTD still only reached 0.63. Given this information, we recommend that RTD as measured on a dynamometer should not be used to reliably detect change over time in older individuals. Instead, measures of peak velocity and/or average acceleration during isotonic tests (against minimal resistance levels) should be considered to provide reliable, relevant information. These measures demonstrated good relative reliability and CV_{TE} values less than or equal to 10%. For both DF and PF, isotonic peak velocity and average

acceleration (against minimal loads) were associated with lower levels of within-subject variation compared to isokinetic and isotonic peak power measures.

In this study, a familiarization session was not provided on the dynamometer before the two test sessions. This may represent a limitation of the study if learning had an effect on the scores during the second testing session. However, familiarization sessions are rarely provided in clinical situations and may not always be feasible in research circumstances because of time constraints, associated costs, and availability of equipment. For these reasons, we elected to omit a familiarization session. Measured levels of systematic bias were minimal for most variables, indicating that there was no substantial learning effect.

CONCLUSIONS

Interpreting and setting threshold levels for acceptable reliability results depends on the particular testing circumstance.²³ In this study, many variables demonstrated good ICC results and CV_{TE} values in the range of 6-13% which are comparable to previous strength/power assessments in younger and older people.^{9, 10, 31, 35} These levels are likely adequate to determine gross changes in strength and power-related parameters among groups over the course of a training study, but ideally, more reliable measures would provide greater confidence in interpreting clinically meaningful change within individuals. Further research is needed to examine the reliability of isotonic variables that have been infrequently studied using dynamometers in the past. These measures may prove to be more reliable and relevant to function in older adults than isometric and isokinetic strength and power parameters that have been more commonly reported. In the

meantime, MDC_{95} scores have been presented for all DF and PF isometric, isotonic and isokinetic variables to provide meaningful thresholds for clinicians and researchers to identify changes beyond those expected by measurement error.

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TABLES

Table 1.
Participant Characteristics

Characteristic	Mean (SD)
Age (years)	73.3 (4.7)
Body mass (kg)	73.8 (11.9)
Height (cm)	159.9 (4.8)
Body mass index (kg/m ²)	28.8 (4.1)
Active DF ROM (degrees)	11 (5)
Active PF ROM (degrees)	53 (6)

Table 2.

Means and Standard Deviations (SD) for Isometric, Isotonic and Isokinetic Tests

Dorsiflexion	Test 1	Test 2	P-value^a
<i>Isometric Results</i>			
Peak torque (Nm)	21.6 (5.1)	21.2 (5.5)	0.28
RTD ^b (to 50% peak torque, Nm/s)	97.8 (28.6)	95.3 (34.9)	0.55
RTD ^b (40-80% peak torque, Nm/s)	62.3 (18.6)	59.3 (23.1)	0.30
<i>Isotonic Results</i>			
Peak velocity (1 Nm load, °/s)	160.9 (31.0)	158.5 (28.9)	0.24
Average acceleration (1 Nm load, °/s ²)	685.5 (183.0)	662.0 (176.7)	0.05
Peak power (1 Nm load, W)	14.7 (6.1)	13.8 (5.6)	0.18
Peak velocity (50% max isometric load, °/s)	78.2 (18.5)	80.0 (15.0)	0.53
Average acceleration (50% max isometric load, °/s ²)	345.2 (95.3)	350.9 (81.9)	0.57
Peak power (50% max isometric load, W)	15.7 (6.3)	15.5 (6.2)	0.64
<i>Isokinetic Results</i>			
Peak torque (30°/s, Nm)	14.0 (4.6)	13.9 (4.8)	0.68
Peak torque (90°/s, Nm)	10.5 (4.2)	10.6 (4.1)	0.68
Peak power (30°/s, Nm)	7.2 (2.3)	7.1 (2.5)	0.76
Peak power (90°/s, Nm)	11.2 (4.5)	10.9 (4.4)	0.35
Plantar Flexion	Test 1	Test 2	P-value^a
<i>Isometric Results</i>			
Peak torque (Nm)	71.0 (21.5)	77.5 (24.0)	0.03
RTD ^b (to 50% peak torque, Nm/s)	113.5 (60.1)	142.0 (65.3)	0.02
RTD ^b (40-80% peak torque, Nm/s)	68.9 (30.1)	90.3 (48.8)	0.02
<i>Isotonic Results</i>			
Peak velocity (15 Nm load, °/s)	275.2 (47.8)	274.8 (50.1)	0.93
Average acceleration (15 Nm load, °/s ²)	1686.4 (477.2)	1698.3 (460.5)	0.79
Peak power (15 Nm load, W)	171.3 (73.0)	180.0 (71.2)	0.24
Peak velocity (50% max isometric load, °/s)	224.6 (44.2)	217.9 (43.6)	0.34
Average acceleration (50% max isometric load, °/s ²)	1304.7 (334.8)	1235.6 (318.5)	0.17
Peak power (50% max isometric load, W)	158.9 (59.1)	162.7 (57.0)	0.51
<i>Isokinetic Results</i>			
Peak torque (30°/s, Nm)	66.7 (20.0)	69.7 (20.2)	0.11
Peak torque (90°/s, Nm)	61.4 (15.8)	62.0 (18.5)	0.82
Peak power (30°/s, Nm)	35.0 (10.3)	37.1 (10.6)	0.06
Peak power (90°/s, Nm)	76.2 (18.0)	77.8 (21.2)	0.54

^aP- values from paired t-tests, except DF isotonic peak power (1Nm load) and PF isometric peak torque which were analyzed with Signed Rank Test

^bRTD = rate of torque development

Table 3.
Relative and Absolute Reliability Scores for Isometric, Isotonic, and Isokinetic Tests

Dorsiflexion	ICC^a	95% CI^b for ICC	SEM^c (units)	95% CI^b for SEM (units)	CV_{TE}^d (%)	Systematic Bias^e ± MDC₉₅^f (units)	Ratio LOA (% of mean)
<i>Isometric Results</i>							
Peak torque (Nm)	0.97	0.94-0.99	1.3	± 2.4	6	0.4 ± 3.5	16
RTD (0-50%, Nm/s)	0.86	0.71-0.94	15.8	± 31.0	16	2.5 ± 43.8	45
RTD (40-80%, Nm/s)	0.84	0.67-0.92	11.0	± 21.5	18	3.0 ± 30.4	50
<i>Isotonic Results</i>							
Peak velocity (1 Nm load, °/s)	0.96	0.92-0.98	8.0	± 15.8	5	2.5 ± 22.3	14
Average acceleration (1 Nm load, °/s ²)	0.97	0.93-0.98	43.7	± 85.6	6	23.5 ± 121.1	18
Peak power (1 Nm load, W)	0.90	0.79-0.95	2.5	± 4.9	17	0.9 ± 6.9	48
Peak velocity (50%, °/s)	0.76	0.50-0.89	10.6	± 20.7	13	-1.7 ± 29.2	37
Average acceleration (50%, °/s ²)	0.90	0.79-0.95	37.9	± 74.3	11	-5.7 ± 105.1	30
Peak power (50%, W)	0.95	0.90-0.98	1.9	± 3.7	12	0.2 ± 5.3	34
<i>Isokinetic Results</i>							
Peak torque (30°/s, Nm)	0.95	0.89-0.98	1.5	± 3.0	11	0.2 ± 4.2	30
Peak torque (90°/s, Nm)	0.96	0.92-0.98	1.2	± 2.3	11	-0.1 ± 3.2	31
Peak power (30°/s, Nm)	0.94	0.88-0.97	0.8	± 1.6	11	0.1 ± 2.3	32
Peak power (90°/s, Nm)	0.97	0.94-0.99	1.0	± 2.0	9	0.3 ± 2.9	26
Plantar Flexion							
	ICC^a	95% CI^b for ICC	SEM^c (units)	95% CI^b for SEM (units)	CV_{TE}^d (%)	Systematic Bias^e ± MDC₉₅^f (units)	Ratio LOA (% of mean)
<i>Isometric Results</i>							
Peak torque (Nm)	0.90	0.74-0.95	9.2	± 18.0	12	-6.5 ± 25.4	34
RTD (0-50%, Nm/s)	0.63	0.20-0.82	44.7	± 87.6	35	-28.4 ± 123.8	97
RTD (40-80%, Nm/s)	0.58	0.12-0.80	29.9	± 58.6	37	-19.5 ± 82.9	104
<i>Isotonic Results</i>							
Peak velocity (15 Nm load, °/s)	0.93	0.85-0.97	18.2	± 35.6	7	0.4 ± 50.4	18
Average acceleration (15 Nm load, °/s ²)	0.93	0.86-0.97	168.2	± 329.6	10	-11.9 ± 466.1	28
Peak power (15 Nm load, W)	0.92	0.83-0.96	27.6	± 54.1	16	-8.6 ± 76.5	44
Peak velocity (50%, °/s)	0.77	0.51-0.89	27.0	± 53.0	12	6.7 ± 74.9	34
Average acceleration (50%, °/s ²)	0.79	0.56-0.90	192.0	± 376.2	15	69.1 ± 532.1	42
Peak power (50%, W)	0.92	0.68-0.93	22.2	± 43.5	14	-13.8 ± 61.5	40
<i>Isokinetic Results</i>							
Peak torque (30°/s, Nm)	0.89	0.77-0.95	8.7	± 17.1	13	-3.8 ± 24.2	36
Peak torque (90°/s, Nm)	0.85	0.68-0.93	8.9	± 17.5	14	-0.5 ± 24.7	40
Peak power (30°/s, Nm)	0.88	0.75-0.95	4.6	± 9.0	13	-2.4 ± 12.8	35
Peak power (90°/s, Nm)	0.86	0.71-0.93	9.8	± 19.2	13	-1.6 ± 27.1	35

^aICC_(2,3) for isometric results and ICC_(2,5) for isotonic and isokinetic results

^bCI = confidence interval

^cSEM = standard error of measurement

^dCV_{TE} = coefficient of variation of typical error

^eSystematic bias = average difference between the two tests (time 1-time 2)

^fMDC₉₅ = minimal detectable change with 95% confidence

FIGURES

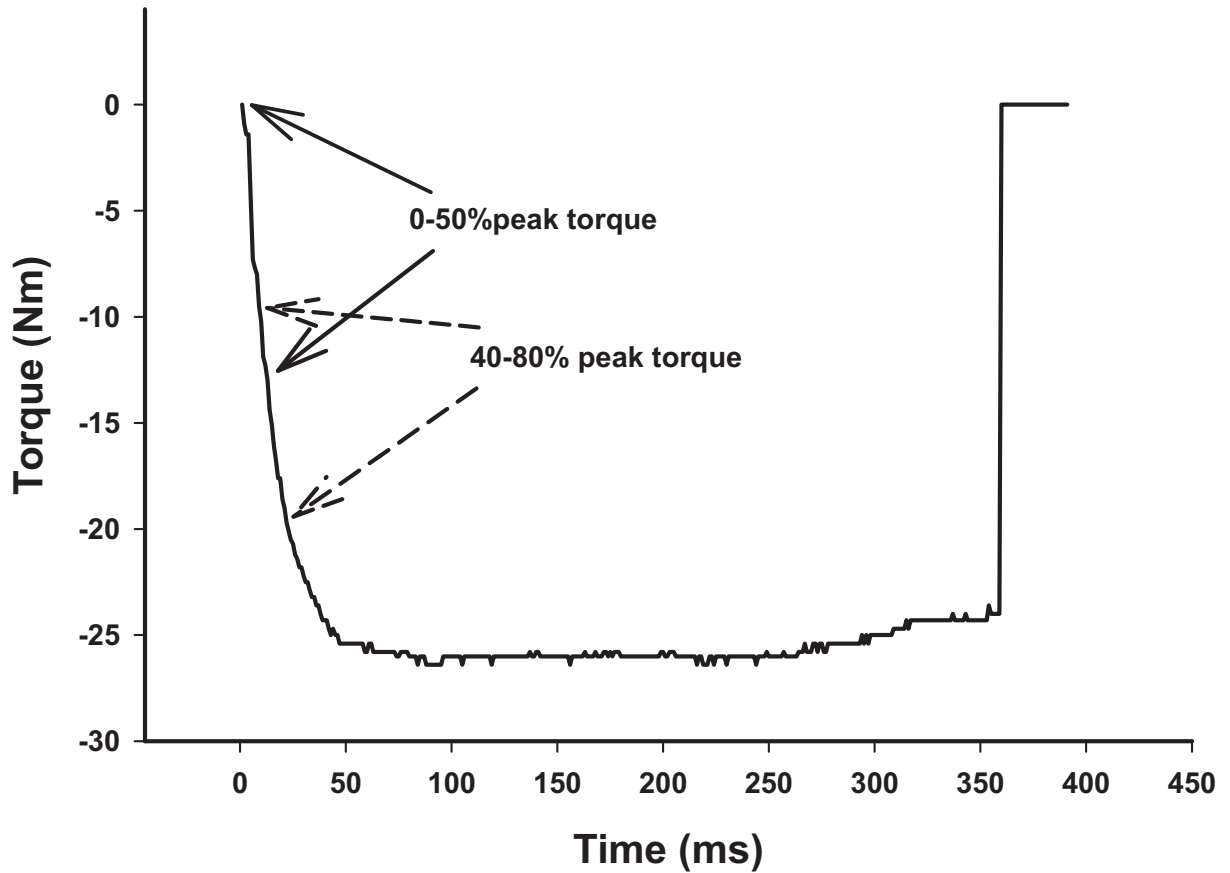


Figure 1. Torque recorded during one repetition of isometric dorsiflexion for one representative subject. Rate of torque development was calculated as the change in Nm/change in time from 0-50% of peak torque and also from 40-80% of peak torque. Torque is designated as negative in the dorsiflexion direction.

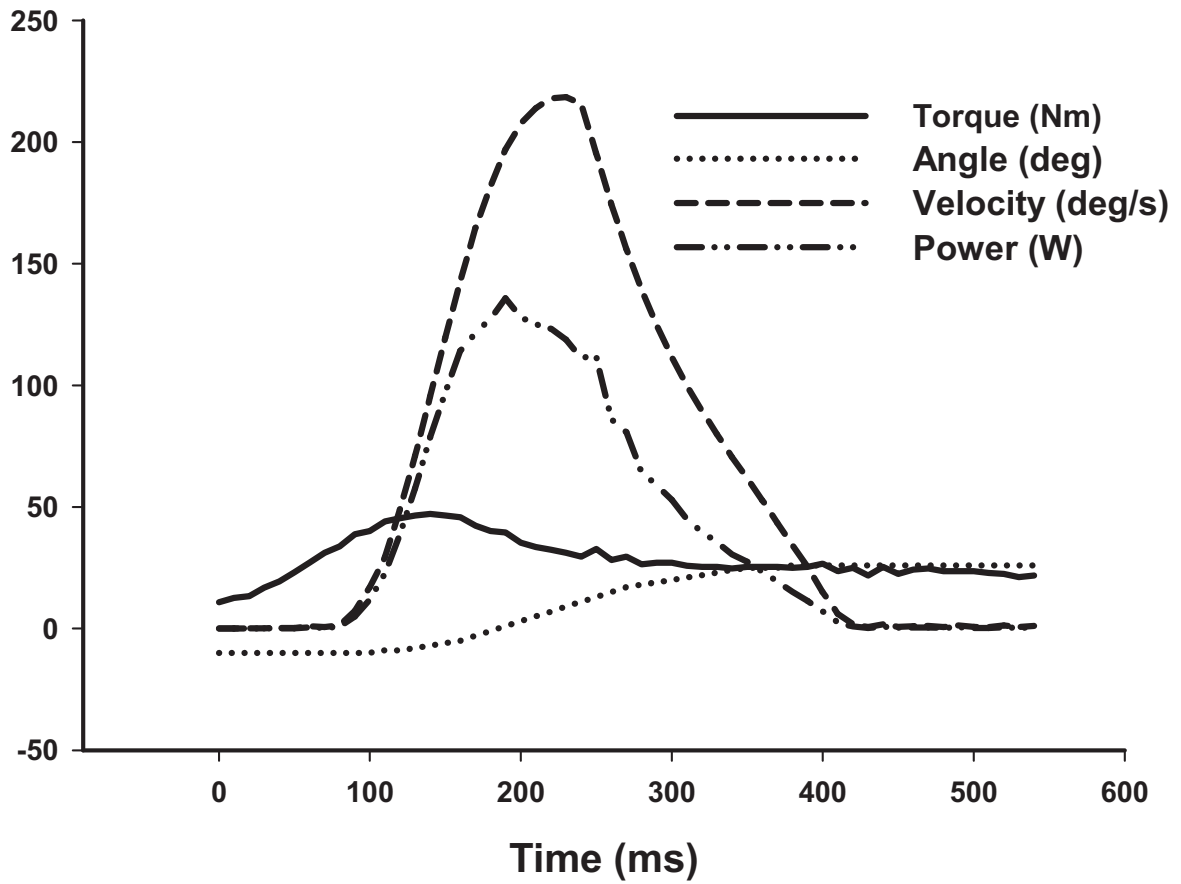


Figure 2. Torque, angle, velocity, and power measurements during one repetition of isotonic plantar flexion (against 50% isometric peak torque) for one representative subject.

CHAPTER 5: EFFECTS OF ANKLE POWER TRAINING ON MOVEMENT TIME IN MOBILITY-IMPAIRED OLDER WOMEN

Running head: Power training in mobility-impaired women

Key Words: response time, aging, resistance training, high velocity, concentric

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MY CONTRIBUTION TO THE PUBLICATION

I participated in determining the design for this study in consultation with my supervisor and conducted trials with the equipment in order to determine the protocol. I also developed the documents for submission to the ethics committee, the intake questionnaire, and scripts/forms for testing in consultation with my supervisor. I assisted other students working in the lab with recruiting subjects and spoke with the majority of potential subjects to determine their eligibility to participate in the study. With input from my supervisor, I trained the assessors in the tests that were included in the protocol and the exercise leaders in developing and implementing the exercise intervention. I assisted with the exercise sessions when help was required. I determined the methods used for data analysis in consultation with my supervisor and conducted the majority of data analysis for this study. I trained summer students to assist with some of the initial data reduction. With my supervisor's input, I wrote and submitted four abstracts based on preliminary data analyses to three national conferences (2007 and 2008) and one local conference (2008). I presented the work at all of the conferences (three presentations and one poster). When all data were collected and analyzed, I selected the journal to which we submitted the manuscript, and wrote and revised the manuscript in consultation with my supervisor. When the reviewers' comments were received, I made all the required revisions in consultation with my supervisor and resubmitted the manuscript along with the responses to reviewers. When the manuscript was accepted I made any further required corrections such as those included in the galley proofs.

ABSTRACT

Purpose: Reduced abilities to generate power put older adults at risk in situations that demand rapid movements. Slower movement times are associated with greater risk of falling and of being involved in a motor vehicle crash. The purpose of this study was to determine the effects of power training on foot movement time, and secondarily, on ankle strength and power in mobility-impaired older women.

Methods: Fifty mobility-impaired women (70-88 years) trained twice per week for 12 weeks in one of three groups (Weights, Elastic Bands, or placebo Control). All groups performed seated warm-up exercises, followed by either concentric dorsiflexion (DF) and plantar flexion (PF) resistance exercises (Weights and Bands) performed “as fast as possible,” or upper body flexibility exercises (Control). Foot reaction/movement time, and ankle DF and PF peak torque (30°/s) and peak power (90°/s) were measured before and after training.

Results: Participants who trained with elastic bands demonstrated improvements in movement time (decreased by 24 ms or 12%, $P = 0.003$). All groups demonstrated improvements in DF and PF strength and power which were not statistically different.

Conclusions: High velocity/low load (elastic bands) training improved movement time, which may have important implications in circumstances when rapid generation of torque is required (e.g., to avoid a fall or prevent a vehicle crash). Elastic bands are relatively inexpensive and provide a practical form of training that could be considered in programs designed for older adults with mobility limitations.

INTRODUCTION

Loss of muscle mass and muscle weakness are closely associated with functional decline, loss of independence, and mortality in older adults (27), with women typically living longer than men with more disability and functional limitations (11). Preferential atrophy of the faster type II muscle fibres, reduced tendon stiffness, along with slowing of muscle twitch properties and a decrease in maximal motor unit discharge frequency (24, 26), may contribute to more substantial reductions in power compared to reductions in strength or endurance with aging (9). Power is defined as the amount of work performed in a given period of time and, as such, is the product of torque and velocity (1). Cross-sectional studies have demonstrated that functional capabilities such as the ability to get up and down from a chair, climb stairs, and walk quickly are more closely associated with power than strength (2, 31) and that limitations in power may be more related to the etiology of falls (28).

Prolongation of response initiation times and reduced rapid torque generation capabilities (9, 14) put older adults at risk in situations that demand fast movement responses. In some situations (e.g., to avoid a fall or to brake a vehicle), being able to generate torque quickly may take equal or greater importance to being able to generate large magnitudes of torque (14). Cross-sectional data have demonstrated that older adults at high risk for falling demonstrate slower response times and slower movement times than those at low risk for falling (29), with women exhibiting slower maximum velocities compared to men when recovering balance (35). Slower reaction/movement times have also been shown to be associated with a greater risk of motor vehicle crashes (15).

Measures of ankle strength and power have been shown to be significantly related to chair rise time, stair climb time, gait velocity, balance, and the occurrence of falls in older adults (3, 14, 28, 31, 34, 36). In an isokinetic study, Whipple et al. (34) found that nursing home residents with a history of falls demonstrated lower strength and power measures about the knee and ankle compared to non-fallers. Specifically, ankle dorsiflexion (DF) power was the most limited in the fallers. Similarly, Skelton et al. (28) also demonstrated reductions in DF strength and knee and ankle power measures in self-reported fallers. Concentric exercise performed with the intention to move quickly (power training) has been shown to result in improvements in strength, endurance, power, balance, and walking speed in older adults (7, 17, 19). While most resistance training studies have utilized large, facility-based resistance training machines, strength gains have been reported in older adults using elastic bands which are much less expensive and more appropriate for home programs (10). The majority of power training studies to date have focused on the larger lower extremity muscles around the hip and knee, with very little emphasis on the distal leg muscles. In addition, the effectiveness of ankle plantar flexion (PF) and DF training on improving movement time has not been previously examined.

The objective of this study was to determine the effects of ankle power training on movement time in a group of mobility-impaired older women. Specifically, we were interested in determining whether movement time changed within each training group. Secondarily, we were interested in measuring changes in PF and DF strength and power in response to training. We hypothesized that movement time would become faster and PF strength and power would improve in the weight-trained group. As well, we

anticipated that the smaller DF muscles would achieve a training effect with lower loads, therefore DF strength and power measures would improve in both the weight and band training groups.

METHODS

Participants

Participants in this controlled trial were randomly assigned in blocks, by a person external to the study, to one of three groups (two resistance training groups and one exercise control group). This meant that once three subjects had agreed to participate in the study, they were each randomly assigned to one of the three groups. This was done in order to keep the size of the intervention groups similar and to control for the effects of season (the study spanned spring, summer and fall and activity levels could potentially differ based on weather conditions etc.). Before randomization and after 12 weeks of training, testing was conducted by an examiner blinded to the study objectives, number of groups, and types of interventions.

The study was advertised in newspapers, newsletters, and on posters, television and radio shows. Women, 70 years of age or older, with mobility limitation (self-reported inability to walk one mile at a moderate pace) who were willing to be randomized and able to attend for 12 weeks, were eligible to participate. Exclusion criteria included unstable acute or chronic disease, participation in an exercise program more than once/week in the past 6 months, and neurological or musculoskeletal impairment that would interfere with the ability to participate. Potential participants who responded to advertisements completed a telephone interview that included the Physical Activity Readiness Questionnaire (PAR-Q). A physician's signed Physical Activity Readiness

Medical Examination form (PARmed-X) was required when potential contraindications to exercise were identified (e.g., history of cardiac disease, uncontrolled hypertension, hernia, detached retina).

One hundred ninety-two women responded to study advertisements and underwent telephone screening. Of these initial responders most were deemed ineligible because they were too young or too active. Seventy-five eligible participants were sent further information after which 13 participants reported they were no longer interested, leaving 62 participants (32% of initial respondents) who underwent initial testing (Figure 1). One woman was further excluded because she did not meet all study criteria, leaving 61 participants to be randomized to the three study groups. Fifty of these women completed 12 weeks of training. At the initial evaluation session, participants provided their written informed consent. Ethical approval for this study was granted by the Education / Nursing Research Ethics Board of the University of Manitoba.

Descriptive Measures

Participants completed the Physical Activity Scale for the Elderly (PASE)(33) as well as a health/demographic questionnaire. Resting blood pressure, heart rate, body mass and active range of motion for the ankle were measured using standard procedures. The Short Physical Performance Battery (SPPB), a test of standing balance, timed-walking and repetitive chair stands, was also performed (8).

Movement Time

Foot reaction and movement time was assessed with the Lafayette Timer, Model 63017 (Lafayette Instruments Co., Lafayette Indiana). Participants were instructed to move their right foot as quickly as possible from the right switch (“the gas”) to the left

switch (“the brake”) when a red light was presented. Participants were cued to upcoming trials with a white light. A standardized 1, 2, or 3 second delay was provided after the white light to discourage anticipation of the red light. Ten familiarization trials were followed by ten test trials. Average reaction times and movement times (of the 10 test trials) were used in analyses. Reaction time was the time between onset of the red light and onset of movement from the “gas” switch, and movement time was the time between onset of movement from the “gas” switch to depression of the “brake” switch. The reliability of this test was assessed in a separate test-retest study at a one week interval in our laboratory (n = 30 women, mean age = 73.3 ± 4.7 years). The intra-class correlation coefficient was 0.90 for movement time and the standard error of the measurement (SEM) was 18.3 ms.

Ankle strength and power measures

DF and PF torque, position and velocity were measured using a Biodex System 3 Pro dynamometer (Biodex Medical Systems Inc., Shirley, NY). Calibration of the dynamometer was verified each day prior to testing. A 4-minute warm-up was provided on a treadmill before participants were positioned on the Biodex with the right lateral malleolus aligned with the axis of rotation of the dynamometer, the right knee flexed 45-55° and the trunk reclined 5° from vertical. Participants were instructed to keep their arms folded across their chest during test trials and belts were used for stabilization. The end-limits of range of motion were set at 10° DF and 30° PF for all tests.

To obtain a measure representative of passive resistive torque about the ankle, the difference between end range DF and PF torque measures were determined from passive

trials (5 repetitions, 5°/s). There were no changes in passive resistive torque between pre and post tests ($P = 0.59$).

Maximal effort isokinetic concentric ankle DF and PF tests were performed at 30°/s and 90°/s. DF tests preceded PF tests and testing at 30°/s preceded that at 90°/s. Participants were given 3-5 sub-maximal practice trials to familiarize themselves with the concentric movement before 5 test trials were conducted. A 2-minute rest period was provided between test velocities.

Biodex data were collected at a frequency of 100 Hz and exported for analyses in SigmaStat (Version 3.10, Systat Software Inc., San Jose, CA). Power (Watts) was calculated as the product of torque (Nm) and velocity (radians/s). In our lab, test-retest reliability of DF and PF peak torque and peak power measures in older women yielded intra-class correlation coefficients in the range of 0.86 to 0.97. SEM results were as follows: peak torque DF 30°/s = 1.2 Nm, PF 30°/s = 6.6 Nm; peak power DF 90°/s = 0.8 Watts, PF 90°/s = 8.1 Watts.

Torque, velocity, and power data were analyzed for the concentric DF and PF tests. At 30°/s, peak torque and peak power frequently occurred at the same joint angle (and were therefore redundant), so only peak torques are reported for the slower speed of testing. At 90°/s, peak torque frequently occurred before a constant velocity was achieved, whereas peak power usually occurred at or very near 90°/s. For this reason, peak power (and not peak torque) values are reported for the faster speed of testing.

Training Program

Participants were randomly assigned to one of three groups that met for 45 minute training sessions, twice per week for 12 weeks. All exercise sessions were supervised by

kinesiology graduates and/or a licensed physiotherapist. Sessions began and finished with a standardized warm-up and cool-down routine that consisted of a total of 25-30 minutes of general lower body exercises performed in sitting for all groups (e.g., marching, knee extensions, ankle circles, toe-tapping).

Weight Training Group (Weights)

The first three training sessions for participants in the Weights group were designed to promote familiarization with the Hammer Strength Tibia Dorsi Flexion and Super Horizontal Calf weight training machines (Life Fitness, Schiller Park, IL). Following these sessions, each participant's 1 RM (repetition maximum) load was estimated from a submaximal test (8-10 RM test). Thereafter, participants completed the standardized warm-up routine and then performed 3 sets of 8-10 repetitions of concentric ankle DF and PF against 80% of a 1 RM load. DF was performed unilaterally and PF bilaterally with two minutes of rest between sets. Loads were progressed every 2-3 weeks to present a consistent challenge while still maintaining the speed of contraction as assessed visually.

Elastic Resistance Training Group (Bands)

The Bands group performed 3 sets of 8 repetitions of concentric ankle DF and PF movements against Thera-Band (The Hygenic Corporation, Akron, OH) elastic resistance bands after warming-up. Movements were performed unilaterally with 2 minutes of rest between sets. Participants were progressed to more difficult resistance bands every 2-3 weeks.

Participants in both the Weights and Bands groups were encouraged to perform the concentric portions of the training movements "as fast as possible", whereas the

eccentric phases of the movements were completed in a slow and controlled manner over 2-3 seconds (17).

Control Group

The placebo control group participants performed static neck stretches and shoulder range of motion exercises, and received education about proper head and neck posture between their warm-up and cool-down exercises.

Statistical Analyses

Based on preliminary movement time data (the primary outcome variable), sample size calculations were conducted (minimally detectable difference = 10 ms, expected standard deviation of residuals = 15 ms, 3 groups with $\alpha = 0.05$ and desired power = 0.8). It was determined that 45 participants were required for the study.

Analyses were conducted using SigmaStat (Version 3.10, Systat Software Inc., San Jose, CA) and SPSS (SPSS 15.0, SPSS Inc., Chicago, IL) software packages. Descriptive statistics were calculated as mean \pm standard deviation (SD) for normally distributed variables, and as medians and ranges for those that were not normally distributed. Mean differences between variables at baseline were assessed using one-way analysis of variance (ANOVA) tests ($P < 0.05$). When normality failed, a one-way Kruskal-Wallis ANOVA on ranks was performed. A series of two-way ANOVA tests with repeated measures were used to investigate between (group) and within (time) factors with movement time, peak torque, and peak power values as dependent variables ($P < 0.05$). Pre-planned within-group comparisons were conducted to investigate changes in movement time using the Bonferroni adjustment. Because three within-group

comparisons were planned, the significance level was set at 0.017 (0.05/3) for these analyses so that the overall probability of making a type I error was still 0.05 (25).

RESULTS

Participants

Eleven of the 61 women who were randomized to the three study groups did not complete 12 weeks of training. As well, one woman completed all tests except the movement time test because she could not see the indicator lights due to poor vision. Reasons for not completing training included: unexpected move from the province (1), joining a fitness facility (1), sustaining injuries in a fall unrelated to the study (1), increasing time commitments elsewhere (4), and illness related to persistent upper respiratory tract infection (1). In addition, three women discontinued training because they felt it was exacerbating pre-existing musculoskeletal conditions. One woman in the Weights group reported worsening knee pain in response to the initial testing procedures and two participants in the Control group complained of worsening hip/back pain related to exercise. Of the 50 participants who trained, 36 attended 90 to 100% of scheduled sessions, 12 attended 75 to 89% of sessions, and 2 attended 63 to 67% of sessions. There were no differences in median attendance among the groups ($P = 0.88$).

Participant characteristics are listed in Table 1. Participants ranged in age from 70-88 years. On average, participants had three chronic medical conditions and had performance scores consistent with moderate mobility limitations. There were no significant differences ($P \geq 0.19$) between the training groups at baseline in their descriptive characteristics (Table 1).

Movement Time and Reaction Time

There were no significant differences in reaction time ($P = 0.15$) or movement time ($P = 0.80$) among groups at baseline. Results of the two-way repeated measures ANOVA are presented in Table 2. Movement time demonstrated a significant main effect for time. A priori within-group comparisons demonstrated a significant decrease in movement time in the Bands group ($P = 0.003$, significance level set at $P < 0.017$ with Bonferroni adjustment), whereas changes in the Weights group ($P = 0.03$) and Control group ($P = 0.38$) were not significant. Reaction time demonstrated no main effects for group or time. In terms of percent change, movement time decreased 12% in the Bands group, 8% in the Weights group, and 2% in the Control group.

The effect size was calculated using the means and standard deviations from the pre and post intervention measures (25). The effect size for change in movement time was 0.60 in the Bands group, 0.28 in the Weights group and 0.17 in the Control group (Table 2). According to Cohen, effect sizes of 0.80 represent large changes, effect sizes of 0.50 represent moderate changes, and effect sizes of 0.20 represent small changes (6).

Strength and Power

No significant differences in DF or PF strength or power were noted among groups at baseline (DF peak torque [$P = 0.84$], PF peak torque [$P = 0.98$], DF peak power [$P = 0.89$], PF peak power [$P = 0.84$]). Main effects for time were significant for all strength and power variables but group by time interactions were not significant (Table 3). Percent change and effect size results were similar among groups for DF strength and power, however, changes in PF strength and power were greater in the Weights group.

DISCUSSION

The primary objective of this study was to determine the effects of power training on movement time in mobility-impaired older women. In this study, ankle training with elastic bands resulted in significant improvements in movement time as assessed with the braking task. This is an important finding because slower reaction/movement times have been shown to be associated with increased risk of falling (13), reduced sit-to-stand performance (12), slower speeds ascending and descending stairs (32), and greater risk of motor vehicle crashes (15).

Although many brake response studies have not differentiated between reaction time and movement time in evaluating speed of response, a recent study by Marques et al. (16) demonstrated that patients at 10 and 30 days post total knee arthroplasty had longer movement times (and similar reaction times) compared to their pre-operative results. Thirty days post surgery, brake movement times for women were 20 ms slower and movement times for men were 25 ms slower than pre-operative values. In this study, training with elastic bands resulted in an average improvement in movement time of 24ms, which, although small, exceeded the measurement error for the test and was approximately equal to the decrement in movement time that occurred post surgery in knee arthroplasty patients. This suggests that concentric training with elastic bands performed with the intention to move quickly may play an important role in the rehabilitation of older adults to improve movement time response. Falls and motor vehicle crashes represent the leading injury causes of death in the older population (4, 30). Safety may be improved if older adults are able to move and respond more quickly to balance threats and/or dangerous driving situations. Because training with elastic bands is

relatively inexpensive and more practical than most other modes of resistance training, these results may have important public health implications.

Although participants in both resistance training groups demonstrated improvements in movement time in this study, only the change in the Bands group reached statistical significance. Changes in movement time in the Weights group verged on being significant ($P = 0.03$) but did not meet the modified level for multiple comparisons ($P < 0.017$ with Bonferroni adjustment). This statistical finding was supported by the effect size data (i.e., effect sizes for change in movement time were small in the Weights group [0.28] and moderate in the Bands group [0.60]). This was contrary to our original hypothesis which stated that participants in the Weights group would demonstrate faster movement times after the exercise intervention. It was not possible to measure the speeds attained during training in this study, however, participants who performed the DF and PF “as fast as possible” against elastic bands conceivably trained at higher velocities and against lower loads compared to those using 80% 1 RM loads with the Hammer Strength equipment. It seems likely that training at higher velocities resulted in greater improvement in the brake movement time task in the Bands group and that the higher loads experienced by the Weights group may have limited the velocities achieved in training and reduced the carryover of effect to the braking task.

Orr et al. (19) found that older adults who performed rapid concentric (and slower eccentric) lower extremity resistance exercises at lower loads (20% 1 RM) demonstrated greater improvements in balance (measured using a computerized force platform) compared to those who trained at higher loads (50% and 80% 1 RM). Changes in peak

power capabilities were similar in all three training groups in their study. Additionally, attainment of lower peak velocities at 20% and 40% of 1 RM during baseline testing independently predicted improvements in balance with training. Increased average velocity at post-test for 40% of 1RM demonstrated a trend ($P = 0.065$) toward significantly predicting balance improvement. These results seem to suggest that for some aspects of balance, movement velocity may be more important than torque or power generation. In situations where the load is relatively small, being able to move quickly may be the most important factor determining balance.

Similarly, velocities attained during training may have had the greatest influence on movement time in our study. However, factors responsible for improved movement time in this cohort were not directly determined. Analyses did demonstrate that changes in movement time were not correlated with changes in DF or PF strength or power as measured on the dynamometer. Because the brake response task involved coordinated activation of both DF and PF and may have also included a small degree of hip and/or knee movement, it is not surprising that changes in movement time were not closely related to changes in single-joint, velocity-constrained strength and power tests on the dynamometer.

Orr et al. (19) have suggested that training with fast concentric contractions against low loads may result in enhanced neural function (reduced response latency, increased recruitment of postural muscles, and improved interpretation of sensory information) and improved force control (decreased co-contraction). As well, in the upper extremity, rapid concentric contractions against relatively light loads (45 % 1 RM) have been associated with greater average muscle activity throughout the concentric action and

a shorter deceleration phase at the end of the movement (18). These types of neuromuscular adaptations may have occurred in response to the concentric training with elastic bands and contributed to improvements in movement time which required coordinated activation of the DF muscles, quickly followed by deceleration of DF and activation of PF.

The PF and DF peak torques measured in this study were, as expected, lower than those previously reported for similarly aged, more active women (23). Contrary to our hypothesis, changes in strength and power were similar among the groups. The absolute and percentage change values for all groups for DF strength and power increased similarly. All participants (including controls) attended up to 24 exercise sessions away from home which may have resulted in a form of “functional” training (increased overall activity level) for these mobility-impaired individuals. As well, motor skill training has been shown to result in increased excitability of the motor cortex, especially when the training requires coordination of visual input and motor performance (20, 21). Many of the warm-up exercises performed by all participants required a high degree of concentration and coordination (e.g., marching and toe-tapping exercises with alternating legs). Regular participation in the warm-up protocol may have resulted in greater excitability of the motor cortex, and, given that there are more pronounced corticospinal projections to the tibialis anterior motoneurons than to any other lower extremity muscle group in humans (5, 22), this may partially explain why DF strength and power improved in all participants.

For PF strength and power, absolute and percentage increases in the Weights group were 2 to 4 times greater than changes seen in the Control group but these

differences among groups did not reach statistical significance. Similarly, effect sizes for changes in PF strength and power were greater in the Weights group (0.35 and 0.38 respectively) compared to the Control group (0.15 and 0.07 respectively). Because movement time was the primary variable of interest in this study, sample size calculations were conducted based on pilot movement time data. A much larger sample size would be required to detect significant changes in ankle strength or power.

Strengths of this study include the fact that participants were randomized, group allocation was concealed, a placebo control group was included, and strength and power were tested by a blinded assessor using different testing equipment (an isokinetic dynamometer) compared to that used in the intervention (weights and elastic bands). It is recommended that further investigations with larger sample sizes be conducted to determine changes in strength and power with similar training protocols. Future research may also examine the actual velocities attained during training using bands and weight machines. For example, further research is needed to determine if weight training performed with lower loads (and therefore at higher velocities) is equal to or better than band training.

In conclusion, our results demonstrated that concentric ankle movements performed as quickly as possible against elastic bands improved foot movement time, which may have important implications in circumstances when rapid generation of ankle torque is required (e.g., to avoid a fall or prevent a vehicle crash). Training with bands represents a low-cost, practical form of exercise that could be considered as an addition to programs designed to address mobility limitations in older adults.

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TABLES

TABLE 1. Participant characteristics.

Characteristic	Total	Weights	Bands	Control
	N = 50	N = 17	N = 17	N = 16
Age (years)	75.0 (70 – 88)	77.0 (70 – 88)	74.0 (70 – 88)	75.0 (70 – 81)
Body weight (kg)	77.9 ± 18.5	74.2 ± 14.5	80.7 ± 22.6	78.8 ± 17.9
Height (cm)	159.4 ± 5.8	158.6 ± 6.9	159.1 ± 6.4	160.6 ± 3.7
Body mass index (kg/m ²)	30.5 ± 6.2	29.3 ± 4.3	31.6 ± 7.7	30.5 ± 6.3
Active DF ROM (degrees)	10.0 ± 8.1	9.2 ± 9.3	12.1 ± 8.3	8.8 ± 6.4
Active PF ROM (degrees)	46.8 ± 9.5	45.9 ± 7.9	49.9 ± 10.1	44.4 ± 10.3
Physical Activity (PASE)	75.4 (25.0 - 229.8)	83.0 (52.2 – 222.0)	95.7 (33.0 - 229.8)	62.5 (25.0 - 185.0)
SPPB	9 (5 – 11)	9 (6 – 11)	9 (6 – 11)	9 (5 – 11)
Medical diagnoses (no.)	3 (0 – 6)	3 (0 - 6)	3 (1 – 5)	3 (1 – 6)
Prescription medications/day	3.8 ± 2.5	4.7 ± 2.5	3.2 ± 1.8	3.5 ± 3.0

(no.)				
Fallers, <i>n (%)</i>	24.0	23.5	23.5	25.0
No walking aid (no.)	42			
Cane (no.)	7	3	3	1
Walker (no.)	1			1

There were no significant differences ($P \geq 0.19$) between the training groups at baseline.

Values of normally distributed data are presented as mean \pm standard deviation (SD).

Skewed data are presented as median (range).

P values were determined by one-way ANOVA on ranks for ages, PASE and SPPB scores, and number of medical diagnoses and by one-way ANOVA for all other variables.

“Fallers” represents the number of participants who reported 1 or more falls in the past 12 months.

PASE = Physical Activity Scale for the Elderly

SPPB = Short Physical Performance Battery

TABLE 2. Pre and post-training movement time and reaction time values (mean \pm SD, % change, effect size) with repeated measures analysis of variance results.

					<i>P Value</i>		
		Weights	Bands	Control	Group	Time	Group X Time
Movement	Pre	191 \pm 48	188 \pm 41	181 \pm 43	0.84	<0.001	0.20
Time (ms)	Post	177 \pm 52	164 \pm 39	175 \pm 36			
	Change	8%	12%	2%			
	Effect Size	-0.28	-0.60	-0.17			
Reaction	Pre	366 \pm 41	360 \pm 34	347 \pm 43	0.52	0.91	0.51
Time (ms)	Post	359 \pm 39	363 \pm 39	351 \pm 46			
	Change	2%	1%	1%			
	Effect Size	-0.17	0.09	0.09			

TABLE 3. Pre and post-training dorsiflexion (DF) and plantar flexion (PF) peak torque (PT) and peak power (PP) values (mean \pm SD, % change, effect size) with repeated measures analysis of variance results.

					<i>P</i> Value		
		Weights	Bands	Control	Group	Time	Group X Time
DF PT 30°/s (Nm)	Pre	11.9 \pm 6.0	11.9 \pm 4.9	12.9 \pm 6.0	0.85	<0.001	0.98
	Post	13.8 \pm 6.4	13.5 \pm 6.3	14.6 \pm 5.2			
	Change	25%	16%	30%			
	Effect Size	0.32	0.34	0.29			
PF PT 30°/s (Nm)	Pre	50.3 \pm 15.8	51.1 \pm 18.0	49.7 \pm 19.6	0.91	0.004	0.75
	Post	55.9 \pm 12.6	56.0 \pm 20.2	52.6 \pm 19.1			
	Change	17%	11%	8%			
	Effect Size	0.35	0.27	0.15			
DF PP 90°/s (Watts)	Pre	9.3 \pm 5.9	9.5 \pm 4.5	10.1 \pm 5.2	0.94	<0.001	0.88
	Post	10.8 \pm 6.3	11.0 \pm 5.6	11.2 \pm 4.7			
	Change	27%	29%	29%			
	Effect Size	0.27	0.34	0.21			
PF PP 90°/s (Watts)	Pre	58.5 \pm 17.8	62.8 \pm 19.5	60.4 \pm 24.8	0.76	0.005	0.38
	Post	65.3 \pm 15.8	69.2 \pm 21.9	62.0 \pm 15.8			
	Change	17%	12%	4%			
	Effect Size	0.38	0.33	0.07			

FIGURES

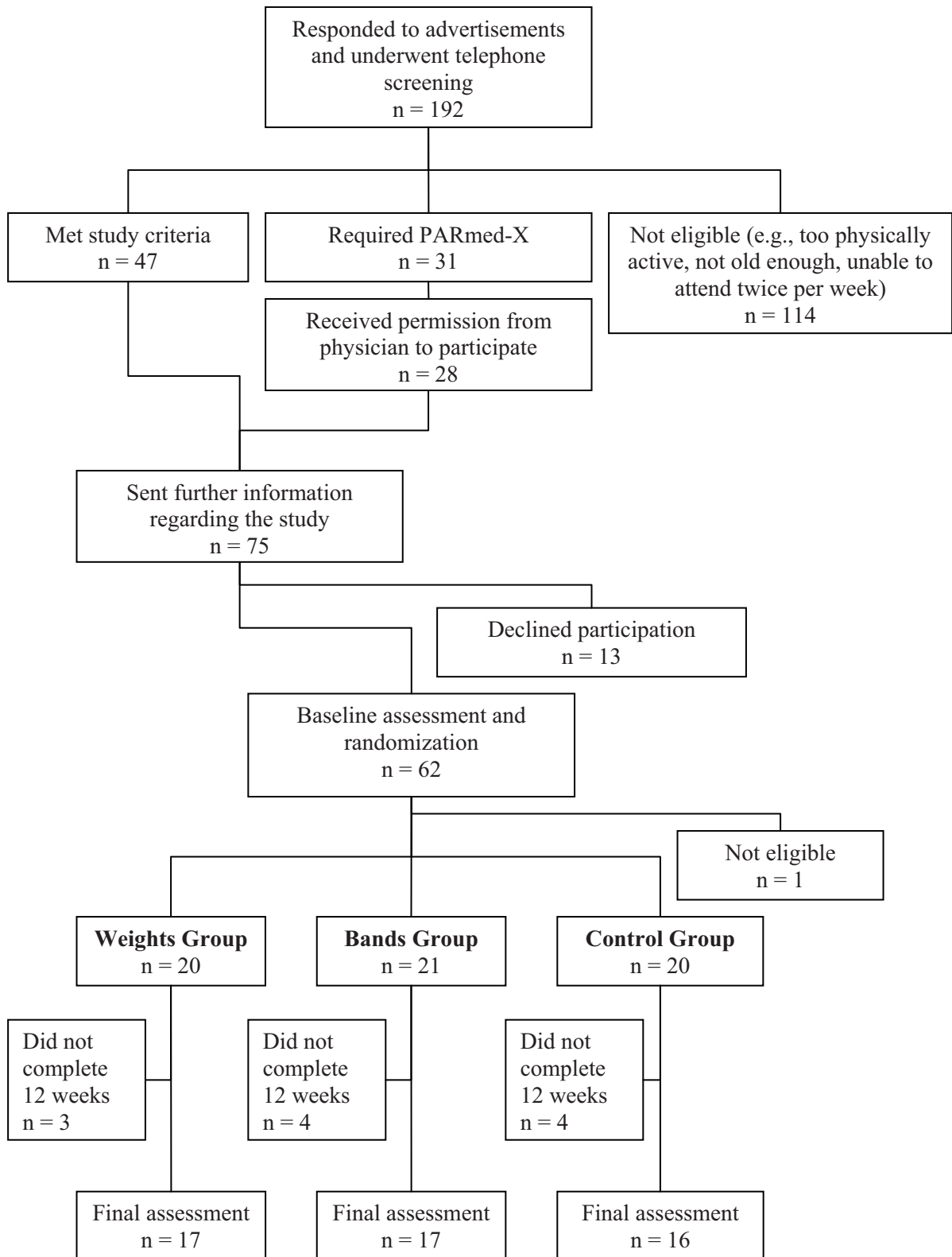


Figure 1. Flow chart outlining numbers of participants at various stages of the study.

**CHAPTER 6: RELATIONSHIPS AMONG ANKLE STRENGTH, POWER,
VELOCITY, AND FUNCTION IN OLDER WOMEN**

Running head: Ankle measures and function

Key Words: Dynamometer, Gait, Isotonic contraction, Neuromuscular

The manuscript in Chapter 6 is being prepared for submission for peer review.

MY CONTRIBUTION TO THE MANUSCRIPT

The data analyzed in this study were gathered from the projects described in Chapters 4 and 5 of this thesis. With my supervisor's input, I wrote and submitted two abstracts based on preliminary data analyses to two national conferences (2009). I presented the work at the conferences. When all of the data were collected and analyzed, I selected the journal that we intend to submit the manuscript to, and wrote and revised the manuscript in consultation with my supervisor.

ABSTRACT

Objective: To examine the relationships among ankle isokinetic and isotonic measures and functional performance, in order to determine which parameters best predict mobility-related function in older women.

Design: Cross-sectional.

Setting: University laboratory.

Participants: Volunteer sample of community-dwelling older women (N = 92; mean age, 74.9y) free of unstable acute or chronic disease.

Intervention: Not applicable.

Main Outcome Measures: Isokinetic (30°/s, 90°/s) dorsiflexion (DF) and plantar flexion (PF) peak torque and peak power, and isotonic DF and PF peak power, peak velocity, and average acceleration measurements were made on a dynamometer. Functional evaluations included usual gait speed, fast gait speed, and stair climb power. In addition, movement time was assessed using a brake response test.

Results: Correlations for isokinetic and isotonic measures were moderate to excellent (DF: $r = 0.77$ to 1.00 , PF: $r = 0.55$ to 0.99 , $p < 0.001$). DF and PF measures were significantly correlated with usual gait speed, stair climb power, and foot movement time, whereas only PF measures were correlated with fast gait speed. Multiple regression analyses revealed that isokinetic PF measures explained the most variance in usual gait speed ($R^2 = 0.42$; PF peak power 30°/s, with the co-variates of weight, age, and co-morbidities) and fast gait speed ($R^2 = 0.37$; PF peak torque 30°/s) while isotonic PF measures contributed the most to stair climb power ($R^2 = 0.53$; PF peak velocity, age, co-morbidities) and movement time ($R^2 = 0.27$; PF acceleration, weight).

Conclusions: Both isokinetic and isotonic strength and power-related variables measured about the ankle were associated with lower extremity functional performance in older women. Parameters that best predicted mobility-related function differed depending on the specific task evaluated.

INTRODUCTION

Power is defined as work (force \times distance) divided by time. It can be calculated as the product of torque and velocity, and as such is a function of both strength and speed. In the older adult, the assessment of power generation capabilities and the evaluation of power training intervention programs have received recent attention. Researchers are interested in studying power because findings suggest that power declines at a faster rate compared to strength (approximately 3.5 % compared to 1.0% – 1.5 % decline per year after age 50) (15, 24, 38, 45). Also, the loss of lower extremity muscle power seems to play an important role in influencing functional mobility (1, 2, 4, 19, 42) and fall risk (30, 39).

Lower extremity neuromuscular power has been assessed with a variety of assessment procedures including the 30-second cycle-ergometer Wingate test and stair climb tests (28, 46), vertical jumping tests (46), sprinting tests (28), and pneumatic resistance and isokinetic dynamometer tests (6, 28). In older adults, power is commonly evaluated using either pneumatic resistance equipment (3, 4, 11, 18, 27) or dynamometers (13, 41-43). With pneumatic resistance machines, the external load is set (relative to an individual's one repetition maximum result) and power is displayed during specific exercise movements. Isokinetic dynamometers have passive, isometric, isotonic, and isokinetic modes that can be utilized to measure torque, angle, and velocity during different types of contractions. Using the isokinetic mode, the actuator arm velocity is constrained to a preset maximum (this does not infer that muscle contractions occur at a constant velocity). During isokinetic tests, peak power (26, 42) and average power measured over a specified range of motion (7, 37, 43) have been reported. Peak power

has also been measured using the isotonic mode in which the load is set and the velocity of movement is under voluntary control (22, 23, 26). Other parameters that can be measured on a dynamometer that are inherently related to power (i.e., the fast generation of torque) include rate of torque development during isometric contractions (23, 40, 44) and maximum voluntary velocity or acceleration achieved during testing with the isotonic mode (23, 41).

Although the majority of studies involving older adults have focused on strength and power about the hip and knee, research has demonstrated that the muscles around the ankle are important for walking (5, 21, 48), and maintaining balance as well as avoiding falls (17, 39, 47). Findings suggest that older adults with limited abilities to generate strength and power about the ankle generally adopt slower walking speeds and are at greater risk of falling (5, 39, 47, 48). To date, the relationships among different measures of ankle strength, power, and velocity are unclear; and isotonic variables have been infrequently studied despite the suggestion that isotonic measures may be more functionally relevant than isokinetic measures (23). As well, it is not known whether certain measures of strength or power predict functional performance better than others. The objective of this study was to determine the relationships among ankle isokinetic and isotonic measures and functional performance (usual gait speed, fast gait speed, stair climb power, movement time). It was hypothesized that many of the measures would be related to function; in particular, that plantar flexion (PF) would be more closely related than dorsiflexion (DF) to the chosen functional measures. Of the four functional outcomes, we hypothesized that power-related PF variables would be most highly correlated with fast gait speed and stair climb power.

METHODS

Participants

Subject data from two studies were combined for these analyses (Chapters 4 and 5). Ninety-two women 65 years of age and older were recruited. Exclusion criteria included unstable acute or chronic disease, and neurological or musculoskeletal impairment that could interfere with testing. Each participant provided written informed consent prior to data collection. Ethical approval for the studies was granted by the Education / Nursing Research Ethics Board of the University of Manitoba. Details of the participants are provided in Table 1.

Ankle strength and power measures

DF and PF torque, position, and velocity were measured using a Biodex System 3 Pro dynamometer (Biodex Medical Systems Inc., Shirley, NY). Calibration of the dynamometer was verified each day prior to testing. Participants initially warmed-up by walking for 4 minutes on a treadmill and then were positioned with the right lateral malleolus aligned with the axis of rotation of the dynamometer, the right knee flexed 45-55°, and the trunk reclined 5° from vertical. They were instructed to keep their arms folded across their chest during test trials, and belts were used for stabilization around the waist and thigh. The end-limits of range of motion were set at 10° DF and 30° PF for all tests.

Isotonic testing preceded the isokinetic tests. DF and PF movements were each performed against a set minimal resistance level (DF = 1 Nm, PF = 15 Nm), as these tests were found to be more reliable than using higher loads in a previous test-retest reliability study conducted in our laboratory. Intra-class correlation coefficients (ICC's) ranged from 0.90 to 0.97 and coefficient of variation [CV] values ranged from 5 to 17% for the

lower load conditions (Chapter 4). Isotonic DF trials were initiated from 30° PF, and PF trials were initiated from 10° DF. Two practice trials preceded 5 test trials for both movements. With each repetition, participants were strongly encouraged to move “fast”. Thirty seconds of rest was provided between all repetitions.

Maximal effort isokinetic concentric ankle DF and PF tests were then performed at 30°/s and 90°/s. Previous test-retest reliability results yielded ICC's in the range of 0.86 to 0.97, with CV's ranging from 9 to 13% for isokinetic peak torque and peak power variables (Chapter 4). The passive mode on the dynamometer was used for these constant velocity tests because it provides a passive return to the start position after each concentric contraction, enabling participants to focus on one concentric movement (e.g., DF) at a time. As well, many participants would not be able to generate enough DF torque to overcome the torque created by the combined mass of the foot and footplate to initiate DF movement if the isokinetic mode was used. Matching concentric contractions with the onset of passive movement avoided this difficulty. Participants were given 3-5 sub-maximal practice trials to familiarize themselves with the concentric movement before 5 test trials were conducted. A 2-minute rest period was provided between test velocities. One subject was unable to generate torque in the DF direction at the higher velocity, therefore DF peak torque and peak power values were recorded at 90°/s in 91 participants.

Functional Performance Variables

Usual gait speed over a 4 m distance was measured in all participants as the average of two trials (N=92). No participants required the use of a walking aid to

complete this test. Fast gait speed (4 m distance, average of two trials) was also measured in 30 women.

A standard flight of 10 stairs (18.7 cm step height) was used for the stair climb test (2, 3). Participants were instructed to ascend the stairs once as quickly as possible, using the handrail only if absolutely necessary. Timing started when the first foot left the ground and ended when both feet reached the top landing. A total of 73 women completed this test (data from 5 women who used the handrail were included in the analyses). One participant was unable to climb stairs and did not complete this test.

Foot reaction and movement time was assessed with the Lafayette Timer, Model 63017 (Lafayette Instruments Co., Lafayette Indiana). Participants were instructed to move their right foot as quickly as possible from the right switch (“the gas”) to the left switch (“the brake”) when the red light was presented. Ten familiarization trials were followed by ten test trials. Averages of the 10 test trials were used in analyses. Movement time was defined as the time between onset of movement from the “gas” switch to depression of the “brake” switch. Data from 91 women are included in the analyses. One woman was unable to perform the test due to vision problems.

Data Analyses

Biodex data were collected at a frequency of 100 Hz and exported for analyses in SigmaPlot (version 11.0, Systat Software Inc., San Jose, CA). All isotonic and isokinetic variables used in the analyses were means of the 5 repetitions performed. Peak torque (Nm) and peak power (Watts = Nm \times radians/s) were calculated for each isokinetic repetition. For each isotonic repetition, peak velocity ($^{\circ}$ /s), average acceleration (peak velocity/time to reach peak velocity, $^{\circ}$ /s²) and peak power (Watts = Nm \times radians/s) were

determined (Figure 1). Although the dynamometer was set to the isotonic mode for these tests, torque is not held absolutely constant throughout the range of motion on this setting (Figure 1). As has been noted previously (16, 26), the sampling rate (100 Hz) of the dynamometer does not permit adjustments in speed to occur fast enough to result in a continuous torque level.

Stair climb power was calculated using the following formula: $\text{power} = \text{force} \times \text{velocity}$, where $\text{force} = \text{body mass} \times \text{acceleration } 9.8\text{m/s}^2$, and where $\text{velocity} = \text{total stair climb height} / \text{stair climb time}$ (3).

Statistical Analyses

Statistical analyses were conducted using SigmaPlot. All data were examined visually and statistically for normality of distribution. The ranges, means, and standard deviations were calculated for each variable. Pearson correlation coefficients were calculated (for DF and PF variables separately) to determine the relationships among isokinetic and isotonic parameters. Because isotonic average acceleration calculations included the peak velocity term ($\text{average acceleration} = \text{peak velocity} / \text{time to reach peak velocity}$), these variables were not independent and correlations were not determined for these pairs. Previous studies have found that relationships between leg strength or power and physical performance variables may be either curvilinear or linear (2-4, 10, 33), therefore, scatter plots were visually inspected to look at the relationships between dynamometer measures and functional measures. These scatter plots demonstrated linear relationships, therefore, best subsets multiple linear regression analyses were conducted (using R^2 as the best criterion and $p < 0.05$ for inclusion in the equation). Each of the three most highly correlated DF and PF variables were used as independent variables. Height,

weight, age, and number of co-morbidities were included as independent variables when they were significantly correlated ($p < 0.10$) with the dependent variables.

Because participants demonstrated a range of mobility levels (62 women had self-reported mobility limitations and 30 women did not), data were also analyzed for the separate groups. Correlation coefficients for the isokinetic and isotonic tests were similar for both groups, and strength, power, and velocity parameters also explained similar amounts of variance in the functional performance regression equations.

RESULTS

The results of the functional tests are presented in Table 1. Correlation coefficients (r) among isokinetic and isotonic variables averaged 0.87 for DF tests and 0.76 for PF tests, ranging from 0.77 to 1.00 and from 0.55 to 0.99 for DF and PF respectively.

Correlation coefficients were determined for each functional performance variable (usual gait speed, fast gait speed, stair climb power, and movement time) with the various isokinetic and isotonic measures. All DF and PF measures were significantly correlated with usual gait speed (DF: $r = 0.24$ to 0.34 , $p < 0.05$ and PF: $r = 0.33$ to 0.49 , $p < 0.01$) and stair climb power (DF: $r = 0.43$ to 0.50 , $p < 0.001$ and PF: $r = 0.50$ to 0.60 , $p < 0.001$). All DF measures were significantly related to foot movement time ($r = -0.24$ to -0.35 , $p < 0.05$) and 5 of the 7 PF measures also demonstrated significant correlations ($r = -0.28$ to -0.37 , $p < 0.01$). No DF measures were correlated with fast gait speed but all PF measures reached significance ($r = 0.50$ to 0.64 , $p < 0.01$). The three strongest correlations for each functional performance variable are presented in Table 2. Six multiple regression equations were then generated (using each of the most highly correlated DF and PF

measures) to determine the combination of independent variables that best predicted each of the functional outcomes. Models that explained the most variance for the functional performance variables are presented in Table 3. Scatter plots demonstrating the relationships between functional performance and the PF variables from the models described above are portrayed in Figure 2.

DISCUSSION

This is the first study to examine the relationships among the more commonly reported isokinetic measures and the less commonly used isotonic measures in older women. Results demonstrated that ankle DF and PF strength, power, and velocity/acceleration results were highly correlated with one another. This suggests that isotonic measures such as peak velocity and average acceleration which do not strictly measure power, do reflect common factors of neuromuscular performance. In addition, regression analyses indicated that for some outcomes such as stair climb power and foot movement time, isotonic velocity and acceleration measures explained more variance in function than isokinetic strength and power measures.

Using the isotonic mode on “isokinetic” dynamometers to measure velocity, acceleration, and power has only recently been recommended and studied. A few researchers have examined age-related changes in velocity and power during velocity dependent concentric contractions (16, 22, 23, 31), but this represents the first study to explore relationships among isokinetic and isotonic measures about the ankle and function in older women. We hypothesized that PF measures would be more closely related to functional outcomes compared to DF measures. Results demonstrated that PF

isokinetic and isotonic measures were indeed more highly correlated with usual gait speed, fast gait speed, and stair climb power compared to DF parameters. PF isotonic peak power was the most highly correlated with movement time, however, DF peak torque at 90°/s was the second most highly correlated variable, suggesting that DF capabilities had a substantial influence on this variable. As hypothesized, PF measures were more highly correlated with fast gait speed and stair climbing compared to usual gait speed and movement time (Table 2). These relationships did not apply strictly to measures of PF power. Power (isokinetic peak power), velocity (isotonic peak velocity), and strength (isokinetic peak torque) variables all demonstrated moderately strong correlations with fast gait speed and stair climbing. Similarly, power, velocity, acceleration, and strength each contributed to one of the models generated to explain the greatest proportion of variance for the functional performance measures (Table 3).

Generally, power is considered to be more closely related to function in older adults as opposed to strength capabilities. However, in reporting these findings, measurement techniques for determining strength and power have varied. Isometric tests and/or composite measures of strength have frequently been compared with isotonic power results obtained on pneumatic resistance equipment (2, 4, 19). As well, although a variety of functional tests have been examined (e.g., usual and fast paced walking over different distances, stair climbing, repetitive chair stands, standing balance), strength and power measurements have most commonly been limited to hip and/or knee extension movements (2, 3, 8, 9, 14, 20, 33, 34, 36). Presumably, different functional performance tasks require different strength and power capabilities, and the lower extremity muscle groups that contribute the most also vary.

Studies that have specifically examined the relationships among ankle indices and function in older adults have reported mixed results. In separate bivariate linear regression models, Bean et al. (2002) determined that isokinetic PF peak power (90°/s) independently explained 37% of the variance in the 6 Minute Walk Test in 45 older adults with mobility limitations, compared to isometric PF strength which explained only 24% of the variance. However, these authors also reported that virtually the same amount of variance in the 6 Minute Walk Test was explained by PF power and PF strength when these variables were entered into multiple regression equations along with age, gender, chronic conditions, body mass index, and falls efficacy ($R^2 = 0.60$ and 0.59 , respectively). Suzuki et al. (2001) also measured DF and PF isometric strength and isokinetic peak torque and peak power in 34 older women with self-reported functional limitations. PF isometric strength was more highly correlated with usual and fast walking speeds ($r = 0.53$ and 0.47 , respectively) compared to PF power measures (tested at 120°/s). These correlations were very similar to those attained in our study describing relationships between isokinetic peak torque (30°/s) with usual and fast gait speeds ($r = 0.49$ and 0.64 , respectively). Suzuki and colleagues also reported that isokinetic DF power (120°/s) was an independent predictor of stair climb time along with physical functioning and general health scores ($R^2 = 0.54$). Similarly, in our study, DF peak power at 30°/s and 90°/s explained 45% and 46% of the variance, respectively, in stair climb time when entered into multiple linear regression equations along with weight, age, and co-morbidity values. However, greater proportions of variance in stair climb time and stair climb power were explained by PF parameters in our study. PF peak power and isometric strength were both correlated with stair climb time in the study by Suzuki and colleagues but their

effects were not significant when entered into multivariate equations along with 7 other variables (DF strength, power, general health, mental health etc.). In our study we analyzed the contributions of DF and PF strength and power in separate regression equations because the variables were highly correlated with one another, and this avoided multi-collinearity issues (32).

Peak power was measured during constant velocity contractions (isokinetic tests) and against minimal loads during isotonic conditions in this study. Velocity and acceleration measurements were also determined from the isotonic trials. The strongest relationships among measures of strength, power, velocity, and function were found when the muscle performance tests and the functional situations were similar in terms of the velocities and loads encountered. For example, the velocities and loads encountered during walking likely corresponded more closely with those experienced during the isokinetic PF tests, compared to the minimal loads and relatively high velocities achieved during isotonic tests. It has been shown that ankle joint peak velocities approach 150°/s in the PF direction and approximately 60°/s in the DF direction during walking in older men (25). This may explain why isokinetic PF measures independently predicted usual and fast walking speeds, yet isotonic PF velocity and acceleration contributed more substantially to measures of stair climb power and movement time when the speed of movement may have been the more critical determining factor.

In our study, all correlations for measures of DF strength, power, and velocity/acceleration were greater than 0.73, indicating good to excellent relationships (32). PF results demonstrated moderate to excellent relationships that were slightly lower ($r \geq 0.55$). One previous study examined relationships between PF peak torque (isokinetic

30°/s) and isotonic peak velocities and peak power in fifteen younger people (26).

Although the specific testing techniques and parameters used for analyses differed from those in our study, similar correlation coefficients in the range of 0.73 to 0.83 were obtained. In our study, isokinetic measures were generally more closely correlated with other isokinetic measures at the same velocity and isotonic measures were more closely related to other isotonic parameters. As has been previously suggested, strength and power measures may be specific to particular assessment techniques because of mechanical and neural activation differences associated with specific types of contractions (12, 29, 35).

In this study, isokinetic and isotonic strength, power, and velocity relationships were evaluated about the ankle. Further research is required to determine the strength of similar relationships measured about different joints, involving more proximal muscle groups. As well, it is recognized that evaluations of gait speed conducted over a relatively short distance (4 m) in the laboratory may not reflect actual community walking behaviours. Puthoff et al. have determined that the relationships between walking speed evaluated over 4 m in the laboratory and average speeds attained during community walking are significant but only moderately strong ($r = 0.56$) and, in their study, measures of hip and knee extension strength and power were more closely related to everyday walking behaviours than laboratory tests of gait (33, 34). This suggests that relationships determined in this study among strength, power, and velocity measures and gait speeds may have been greater if a different walking test that was more representative of walking behaviours in the community had been used.

Further research is required to determine which functional assessments are most relevant and yet still practical to be used to evaluate performance in older adults. In addition, more comprehensive evaluations of strength, power, and velocity capabilities about different joints in the lower extremity are required in order to determine the relative importance of strength and power capabilities in different muscle groups to functional mobility-related tasks.

CONCLUSIONS

This study demonstrated that different measures of strength, power, and velocity attained during isokinetic and isotonic ankle tests on a dynamometer were strongly related in a group of older women. Measures involving similar modes and velocities of testing were the mostly closely related. Isotonic evaluations, which do not constrain velocity and therefore may be more relevant to function, were moderately to strongly related to isokinetic results and explained the most variance in the stair climb power and movement time outcomes. This suggests that the isotonic mode on dynamometers could be used more frequently for evaluation of baseline status and change in neuromuscular properties over time or with interventions.

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TABLES

Table 1.
Participant Characteristics and Functional Test Results

Characteristic	Mean \pm SD	Range
Age (years) N=92	74.9 \pm 4.7	65 - 88
Body mass (kg) N=92	76.0 \pm 16.2	46.5 – 129.4
Height (cm) N=92	159.8 \pm 5.4	144.3 – 175.8
Co-morbidities (#) N=92	2.6 \pm 1.4	0 – 6
Usual gait speed (m/s) N=92	1.0 \pm 0.2	0.4 – 1.5
Fast gait speed (m/s) *N=30	1.5 \pm 0.2	1.1 – 2.0
Stair climb time (s) *N =73	6.9 \pm 3.2	3.4 – 17.0
Stair climb power (Watts) *N =73	228.3 \pm 80.5	81.4 – 445.3
Foot movement time (ms) *N =91	192 \pm 46	104 – 335

*N varied from 92 based on the number of subjects tested for specific variables

Table 2. Strongest dorsiflexion and plantar flexion correlations with functional performance variables.

Variables	Strongest DF correlations	Strongest PF correlations
Usual gait speed (N=92)	r = 0.34, IsoDF PV r = 0.30, DF90 PT r = 0.30, IsoDF Acc	r = 0.49, PF30 PP r = 0.49, PF30 PT r = 0.46, PF90 PP
Fast gait speed (N =30)	No significant correlations	r = 0.64, PF30 PT r = 0.61, PF90 PT r = 0.59, PF30 PP
Stair climb power (N =73)	r = 0.50, IsoDF PV r = 0.49, IsoDF acc r = 0.48, DF30 PT	r = 0.60, PF90 PP r = 0.58, PF90 PT r = 0.57, IsoPF PV
Movement time (N =91)	r = -0.35, DF90 PT r = -0.31, DF30 PP r = -0.30, DF30 PT	r = -0.37, IsoPF PP r = -0.35, PF90 PT r = -0.35, IsoPF Acc

P<0.05 for all correlations

Abbreviations: DF, dorsiflexion; PF, plantar flexion; PT, peak torque; PP, peak power; 30 and 90 refer to isokinetic testing velocities (30°/s or 90°/s); Iso, isotonic; PV, peak velocity; Acc, average acceleration.

Table 3. Best subset multiple regression models for functional performance variables.

Dependent Functional Variable	Independent Variables in Best Subset Equation	Coefficient	Coefficient p-value	Model R² value	Model p-value
Usual gait speed (N =92)	Weight	-0.00365	0.002	0.42	<0.001
	Age	-0.0143	0.001		
	Co-morbidities	-0.0283	0.032		
	PF30 PP	0.00719	<0.001		
Fast gait speed (N =30)	PF30 PT	0.00672	<0.001	0.37	<0.001
Stair climb power (N =73)	Age	-6.280	<0.001	0.53	<0.001
	Co-morbidities	-17.388	<0.001		
	IsoPF PV	0.552	<0.001		
Movement time (N =91)	Weight	-1.114	<0.001	0.27	<0.001
	IsoPF Acc	-0.037	<0.001		

Abbreviations: PF, plantar flexion; PT, peak torque; PP, peak power; 30 and 90 refer to isokinetic testing velocities (30°/s or 90°/s); Iso, isotonic; PV, peak velocity; Acc, average acceleration.

FIGURES

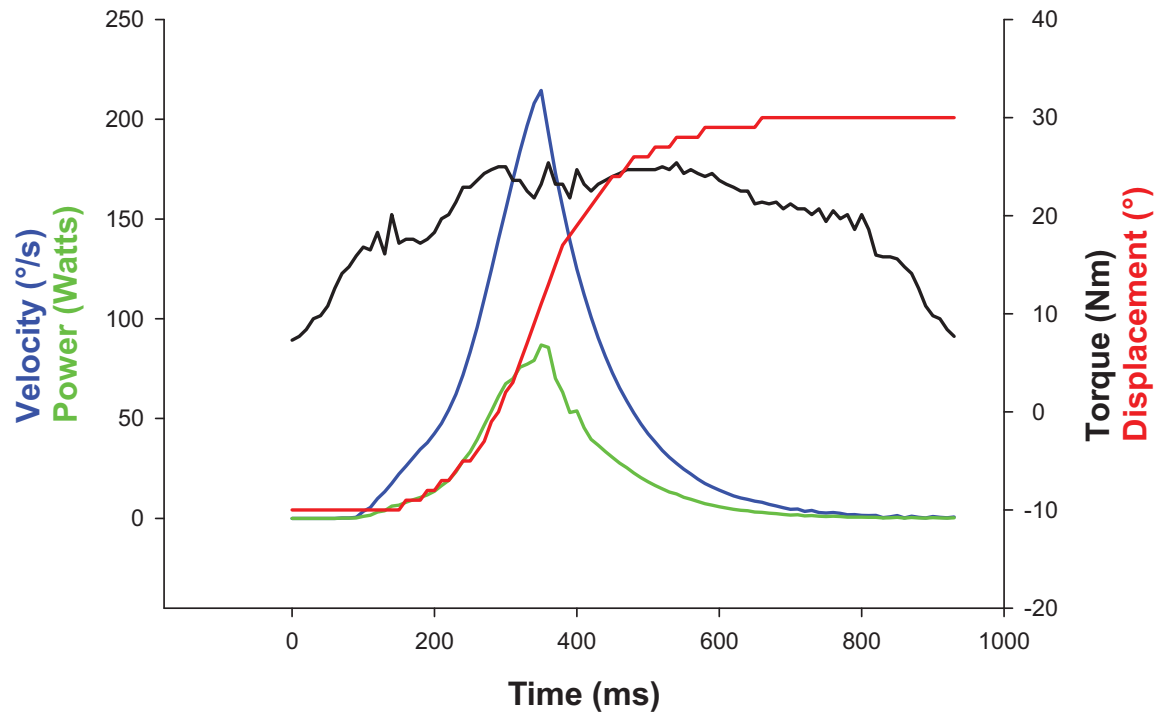


Figure 1. Torque, displacement, velocity, and power plotted during one repetition of isotonic plantar flexion for one participant. Initial load was set at 15 Nm. Power was calculated as the product of torque (Nm) and velocity (°/s).

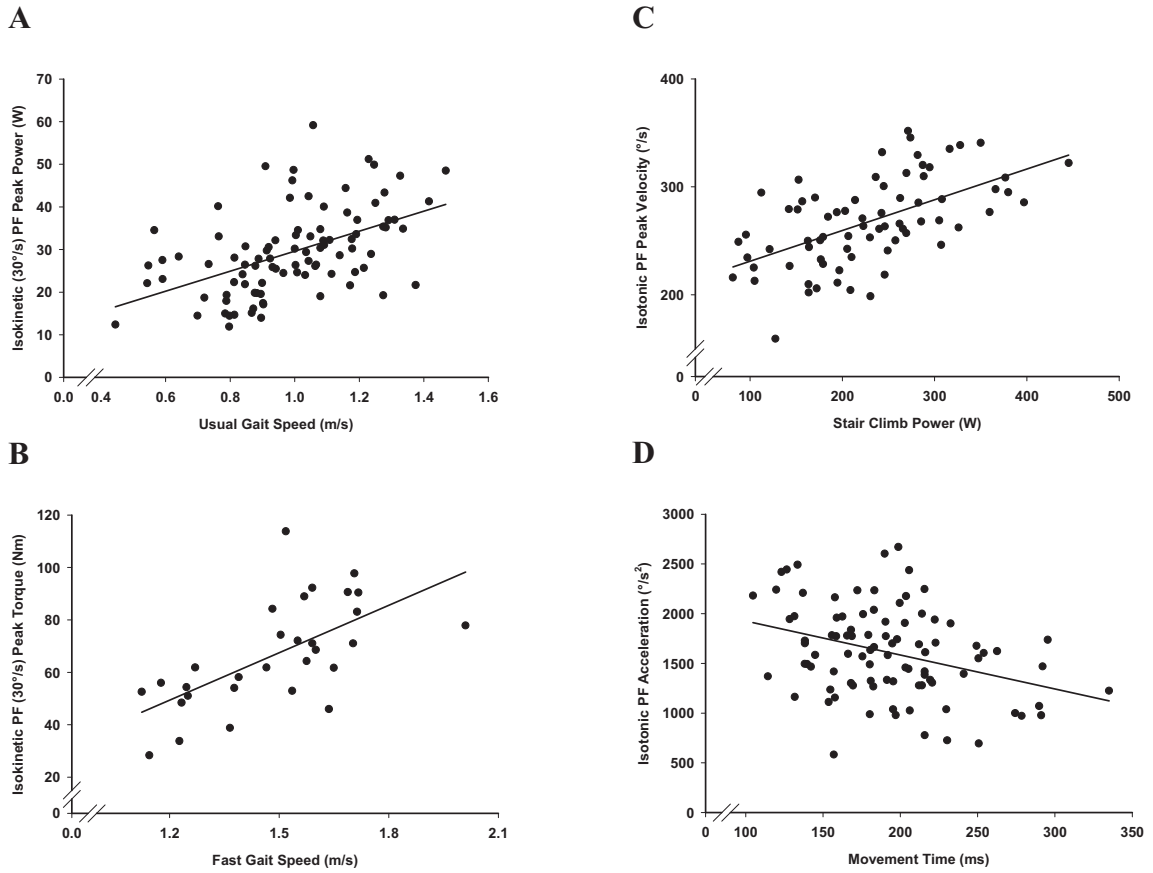


Figure 2. Scatter plots of usual gait speed (A), fast gait speed (B), stair climb power (C), and movement time (D) plotted with the plantar flexion (PF) variable found to be most significant by multiple regression (A = PF30 peak power; B = PF30 peak torque; C = isotonic PF peak velocity and D = isotonic PF average acceleration).

CHAPTER 7: DISCUSSION

This thesis describes aspects of mobility, studied from a number of different perspectives, as it is relevant to older adults. The discussion will focus on: 1) the bases for studying mobility in older women, 2) the main findings and the implications of results from each of the projects, and 3) the limitations of the studies and possible future directions. For additional discussion, the reader is referred to each individual paper.

Section 1: Bases for studying mobility-related parameters in older women

Mobility considered in a broad sense is extremely important because it is so intimately linked with overall health and quality of life in older adults (Groessl *et al.* 2007; Metz, 2000; Yeom *et al.* 2008). Included in this thesis are projects that dealt with investigating the effects of specific high-velocity resistance training programs on mobility-related parameters such as movement time, strength, and power. Intra-rater reliability of dynamometer measures was determined. As well, relationships among these variables and between these variables and functional outcomes were also examined. These studies (Chapters 4, 5, and 6) focused on movements about the ankle in older women. Ankle DF and PF movements were chosen for study because, in terms of mobility, the ankle muscles are important for driving a vehicle, walking and climbing stairs (Bendall *et al.* 1989; McGibbon & Krebs, 1999; Suzuki *et al.* 2001; Wolfson *et al.* 1995), maintaining balance, and avoiding falls (Mackey & Robinovitch, 2006; Skelton *et al.* 2002; Whipple *et al.* 1987). As well, although there is much interest in investigating the relationships between lower limb power generation capabilities and physical function in older adults, the majority of studies to date have focused on the more proximal lower extremity muscle groups (Bean *et al.* 2002a; Bean *et al.* 2002b; Bean *et al.* 2003; Marsh

et al. 2009; Marshall & Gilbert, 1999; Sayers *et al.* 2003; Sayers *et al.* 2005), despite the important role of ankle function in mobility. Women were specifically recruited for the studies because they are at particular risk for mobility impairment. Older women demonstrate greater limitations in terms of walking tolerance and stair climbing ability compared to older men (Leveille *et al.* 2000; Statistics Canada, 2007). They are also more likely to suffer from fear of falling (Filiatrault *et al.* 2009) and to be diagnosed with disability-related health conditions (Murtagh & Hubert, 2004). In terms of accessing the community, older women are less likely to drive a car compared to older men (Turcotte, 2006), especially beyond 75 years of age.

Section 2: Main findings and implications

In Chapter 2 a new theoretical framework for mobility was developed. This novel work was undertaken to attempt to define mobility more comprehensively than has typically been done in the past. A small number of mobility frameworks have previously been developed (Carp, 1988; Patla & Shumway-Cook, 1999; Rose, 2005), but each of these models is heavily focused on environmental and physical factors without regard for other important mobility determinants. As well, much of the research relating to mobility to date has been discipline-specific with very few attempts to consider the determinants of mobility from a variety of perspectives. The new mobility framework incorporates concepts from the existing life-space literature relevant to both walking and driving (Baker *et al.* 2003; May *et al.* 1985; Stalvey *et al.* 1999), previous mobility models (Patla & Shumway-Cook, 1999; Rose, 2005), and other psychological theories dealing with the importance of the environment and the person-environment fit (Carp, 1988; Lawton & Nahemow, 1973; Rowles, 1981). It describes five categories of determinants of mobility

(cognitive, psychosocial, physical, environmental, and financial), with gender, culture, and biography (life history, mobility preferences and experiences) conceptualized as critical cross-cutting influences that impact indirectly on mobility. Each category of determinants consists of an increasing number of factors, to represent the increasing complexity of factors that impact on mobility as one travels farther from home.

This model was developed to raise awareness regarding the topic of mobility as it is relevant to older adults in different life situations, to stimulate new integrative and interdisciplinary research ideas, to encourage a broader interpretation of mobility findings, and to contribute to the development of more extensive and effective intervention programs. In our lab, there has been a consistent interest in considering mobility in older adults from a broad perspective, at least in terms of walking, driving, and accessing public transportation. It was this interest, combined with the need to explore practical methods for measuring different types of mobility that led to the development of the project described in Chapter 3.

Chapter 3 examined the feasibility of using GPS watches and accelerometers to monitor community mobility in older adults. Although many interventions aimed at the older population focus on improving mobility and resultant quality of life, in actual fact, capabilities for measuring changes in mobility in the “real world” are limited. Questionnaires have been used to measure physical activity and travel patterns, but many have been shown to be inaccurate, lack detail, and place a relatively high degree of burden on subjects (Stopher & Greaves, 2007; Tudor-Locke & Myers, 2001). Because it is impossible to replicate the environmental demands of specific community settings, valid assessment of community walking and general mobility patterns (e.g., driving

habits, use of public transportation) should occur within the older adults' communities as they participate in usual routines. Previous studies have demonstrated that GPS devices (used with or without accelerometers) function well and provide valuable mobility data during discrete bouts of outdoor activities such as walking (Le Faucheur A. *et al.* 2007; Le Faucheur A. *et al.* 2008; Troped *et al.* 2008), cycling, in-line skating or riding/driving in a vehicle (Troped *et al.* 2008).

To our knowledge, this study was the first to report on the use of GPS watches and accelerometers to measure community mobility in older adults. Twenty older adults wore the devices during the waking hours for one day. Detailed mobility information was obtained regarding the timing, distances covered, and speeds reached during trips away from home (on-foot and in-vehicle). As well, step counts were acquired for each minute of the day. However, our results demonstrated that there are a number of issues with the technology that still need to be addressed before these devices can be used for data collection over an extended period of time. Difficulties with interruptions in GPS signal acquisition resulted in gaps in the GPS data, and battery life did not allow for more than 12.3 hours of data collection. Similar problems have been reported in previous studies that have attempted to track movement in younger adults over 1 to 3 days (Phillips *et al.* 2001; Rodriguez *et al.* 2005).

This study adds to the existing literature because it presents a new application (i.e., community mobility assessment in older adults) for using GPS devices and accelerometers in combination. As well, it is the first study examining the use of GPS watches for data collection over an extended period of time. GPS watches are small and relatively unobtrusive to wear compared to other GPS devices and may be more

appropriate for use in studies that span multiple days. Although this study determined that the technology is not yet sufficient for monitoring indoor and outdoor mobility over extended periods of time, the data that were obtained contained rich mobility-related information that could not be gained using other methods (e.g., questionnaires). For this reason, GPS and accelerometry technology continue to offer promise for ultimately being able to objectively monitor community mobility in older adults.

Chapters 4, 5, and 6 describe studies that involved ankle strength and power assessment in older women on the Biodex dynamometer. Reliability statistics associated with isometric, isotonic, and isokinetic ankle tests on the dynamometer were reported in Chapter 4. In general, all measures of ankle strength, power, velocity, and acceleration demonstrated high relative reliability values (ICC's > 0.75) except for PF RTD results which were associated with very poor reliability scores. Relative and absolute reliability for DF tests were greater than PF results in 24 of the 26 tests conducted. Isotonic peak velocity tests against minimal loads were associated with the lowest CV_{TE} and RLOA values for both DF (5% and 14% respectively) and PF (7% and 18% respectively). Overall, the following parameters were found to be reliable ankle measures in older women: isometric peak torque, isokinetic peak torque, isokinetic peak power, isotonic peak velocity, and isotonic average acceleration (against low loads for both isotonic parameters). These variables were associated with relatively high ICC, low CV_{TE} , and low SEM results, suggesting that they can reliably measure change in group test-retest situations (Holmback & Lexell, 2007).

This was the first study to determine test-retest reliability for isotonic tests about the ankle in older women (Chapter 4). As well, the study reported in Chapter 6 was the

first to examine relationships among functional mobility-related outcomes and isotonic measures. Correlations between the more commonly reported isokinetic measures (peak torque and peak power) and the less frequently utilized isotonic measures (velocity, acceleration, peak power) were moderately strong to excellent (DF: $r = 0.77$ to 1.00 , PF: $r = 0.55$ to 0.99 , $p < 0.001$), which suggests that the parameters represent related factors of neuromuscular performance. Although few studies have utilized isotonic testing techniques to date, it is possible to gain detailed information about neuromuscular function during functionally relevant, maximum velocity movements using this mode. Indeed, our results demonstrated that isotonic measures of PF peak velocity and average acceleration explained the most variance in the stair climb power and movement time outcomes tested in older women. Because these isotonic parameters have been shown to be reliably measured and relevant to mobility function, the isotonic mode on dynamometers should be considered more frequently for evaluation of baseline status and change in neuromuscular properties.

The project described in Chapter 5 of this thesis studied the effects of ankle resistance training with weights and elastic bands on foot movement time in mobility-impaired older women. A placebo control group was included for comparison. The primary outcome variable (movement time) was chosen because of its relevance to mobility in terms of ability to recover from a possible fall and to operate the pedals on a vehicle safety. It has been shown that older adults at high risk for falling demonstrate slower response times and slower movement times than those at low risk for falling (St George *et al.* 2007). Slower reaction/movement times have also been shown to be associated with a greater risk of motor vehicle crashes (Margolis *et al.* 2002).

Our results demonstrated that older women who trained twice/week for 12 weeks performing concentric contractions of the ankle PF and DF muscles “as fast as possible” against elastic bands obtained the most benefit in terms of reducing foot movement time (24 ms reduction in movement time, $p=0.003$). Changes in the weight-trained group (12 ms reduction in movement time) and the control group (6 ms) were not significant. Ankle DF and PF strength and power measures (peak torque at $30^\circ/s$ and peak power at $90^\circ/s$) improved in all groups.

This was the first study to compare ankle training with bands and weights when subjects were encouraged to perform the concentric contractions “as fast as possible.” The fact that the PF and DF band training resulted in reduced foot movement times suggests that this type of training may improve older adult safety in challenging situations that demand quick movements (e.g., to maintain balance or apply the brakes in a vehicle). Because training with elastic bands is relatively inexpensive and can practically be implemented in a home exercise program, these results may have important public health implications.

Section 3: Limitations of the studies and future directions

One limitation associated with the development of the theoretical framework on mobility for older adults relates to the fact that the model has not yet been tested and it lacks empirical support. Attempting to gather empirical data to support the framework will be a difficult task to undertake, given that the concept is so complex and multi-factorial. However, development of a tool (or a set of inter-related tools) capable of measuring barriers and compensation strategies that influence mobility in different contexts and in different life-space locations is required. Focus groups of older adults

may provide more information about restrictions in life-space mobility and the reasons for changes that lead to those limitations. Computer-adaptive testing, which has been used to measure functional mobility in children (Haley *et al.* 2005; Haley *et al.* 2006), may be explored to provide a method for comprehensive mobility assessment without unreasonable administration times.

In terms of future directions for the mobility framework, it is our hope that a more comprehensive conceptualization of mobility contexts will be adopted which will result in changes that benefit older adults. One example would be the development of mobility intervention teams that are capable of approaching mobility challenges in a clinical environment from a variety of points of view. In terms of research, new relationships may arise among disciplines interested in studying mobility-related issues. Finally, policies could be established so that older adults will be able to maintain their mobility, their independence, and their health for as long as possible.

The GPS watch and accelerometer mobility monitoring study reported in this thesis represents one of our first attempts to bridge mobility disciplines, by attempting to measure walking, driving, and the use of public transportation in older adults. Limitations associated with this study were largely related to inadequacies in the GPS technological capabilities (e.g., limited battery life, difficulty locking-in to satellite signals, accidentally shutting off the watch by bumping the buttons), even though there were some cases of very rich, day-long data. Future studies should investigate the latest equipment capabilities (e.g., experiment with devices that do not require 2-3 minutes to acquire satellite signals upon going outdoors) and consider adjusting the protocol to maximize GPS watch battery life (e.g., turning watches off when indoors, charging watches

overnight). As the technology continues to evolve to address these difficulties, it will be possible to design larger, more involved trials. For example, researchers may be able to examine changes in community mobility patterns with seasonal transitions, and determine differences between older men and women in terms of driving, walking habits, and use of public transportation.

In this thesis the studies that involved assessment/training of ankle DF and PF strength and power (Chapters 4, 5, and 6) included only older women. Although similar reliability results, training responses, and relationships among strength, power, and functional outcome variables would be expected in other segments of the population, they were not tested in these projects. In both the reliability study and the power training intervention study no familiarization session was provided prior to collection of the baseline data. Although this could represent a limitation of the study protocols, no substantial learning effect was seen in the test-retest situation over 7 days (measured levels of systematic bias were minimal). It would be expected that testing conducted 3 months later (as occurred in the power training intervention study) would be associated with even less of a learning effect. Indeed, many of the women in the intervention trial reported during the final testing session that they did not remember ever doing some of the tests in the baseline evaluation. In addition, possible effects related to learning were controlled in the intervention trial with the inclusion of the placebo control group.

It should be noted that screening of the older women for inclusion in the resistance training intervention study was based on the subjects' responses to the intake questionnaire. Of particular importance were the subjects' responses to the questions about their ability to walk. For this study, subjects were categorized as being mobility

impaired if they reported they were unable to walk 1 mile (8-12 blocks) at a moderate pace. Although the majority of women admitted to the study demonstrated obvious limitations in walking tolerance, actual walking abilities varied. Objective evaluation of walking tolerance to determine eligibility for the study may have resulted in a more homogeneous sample of mobility-impaired older women.

Another limitation associated with the ankle training intervention study relates to the fact that it was not possible to measure the velocities achieved during training with the equipment that was used (weights and elastic bands). It is possible (and even likely) that subjects in the weight training group trained at lower velocities than those in the elastic band training group (because the weight training subjects had 80% 1RM loads to work against). For this reason it was difficult to determine the likely mechanism(s) for the changes in movement time that occurred (e.g., changes may have occurred as a result of differences in the resistance equipment, resistance loads, unilateral versus bilateral training, and/or training velocities). Further research is needed to determine whether weight training performed “as fast as possible” against lower loads (and presumably at higher velocities) is equal to or better than similar training with bands in terms of reducing movement time in older adults.

As mentioned in Chapter 5, changes in movement time were not significantly correlated with changes in any of the DF or PF strength and power variables. However, baseline movement time values were significantly related to all baseline strength and power measures evaluated on the dynamometer. It is likely that the correlations demonstrated among strength, power, movement time, and functional outcomes such as gait speed and stair climbing (reported in Chapter 6) reflected the common influence of

central nervous system changes associated with aging (Seidler *et al.* in Press) and not cause-effect relationships among these variables.

The sample size determined for the ankle power training intervention study was based upon pilot movement time data because movement time was the primary variable of interest. However, if we had been able to recruit greater numbers of older women for the study, it is possible that: 1) the weight training group might have had significant, albeit smaller, changes in movement time than the band training group; and 2) the changes seen for PF strength and power variables (which were 2 to 4 times greater than changes in the control group) may have reached statistical significance. As well, it would have been more ideal to have had a larger sample of women who completed the fast gait speed test ($n = 30$ versus $n \geq 73$ for the other functional outcomes).

In summary, the projects included in this thesis focused on describing and measuring mobility, along with determining the effects of an exercise intervention on specific neuromuscular factors important for safe mobility. Measures of ankle strength, power, and velocity were studied in older women to determine the reliability of the measures, the relationships among these variables, and to establish which parameters best predict functional outcomes relevant to mobility. A comprehensive conceptual framework that describes mobility as it is relevant in many different contexts and life-space locations has been developed. Experimental data were collected using a relatively new combination of measurement tools (GPS watches and accelerometers) which hold promise for monitoring community mobility in older adults. Finally, the results of an exercise intervention study have been presented, demonstrating that concentric ankle training performed with the intention to move quickly using relatively simple equipment

(elastic bands) resulted in improvements in movement time which may have implications in terms of exercise prescription for fall prevention and safe driving in older adults.

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APPENDIX A: MODELING AGE-RELATED NEUROMUSCULAR CHANGES IN HUMANS

Key Words: aging, strength, muscle, motoneuron, force, quadriceps, effort.

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MY CONTRIBUTION TO THE PUBLICATION

The topic for this paper was provided to me by my committee as part of the requirements for my candidacy exam. Dr. Gardiner worked closely with me to assist in the interpretation of the Heckman-Binder model and the rat model program that he had previously developed. I conducted all of the literature reviews required for determining the background literature for the paper and experimented with the model in consultation with Dr. Gardiner. I wrote the initial paper and presented it at my candidacy exam. I then made changes to the manuscript based on the feedback received from my committee. I presented the material in poster format at a national conference in 2009. In consultation with Dr. Gardiner, I selected the journal to which we submitted the manuscript and, based on the reviewers' comments, I revised the manuscript with assistance from Dr. Gardiner and Dr. Porter. I resubmitted the manuscript along with the responses to reviewers. When the manuscript was accepted I made any further required corrections such as those included in the galley proofs.

Modeling age-related neuromuscular changes in humans

Sandra C. Webber, Michelle M. Porter, and Phillip F. Gardiner

Abstract: With aging, motoneurons and muscle tissue undergo significant changes, which influence function in terms of strength, mobility, and overall independence. Mathematical modeling provides a practical method of studying the relationships among recruitment, rate-coding, and force output in motor units, and may be used to predict functional neuromuscular changes related to aging. For this study, the Heckman-Binder model was used to examine changes in human quadriceps motor units. Relationships among current input, firing frequency, and force output were defined for both a younger and an older individual. Included in the model were age-related effects associated with reduced muscle contractile speed; reduced muscle-fibre number, size, and specific tension; reduced gain of the frequency-current relationship; decreased size of motoneurons; and altered motor unit remodeling. Adjustment of this model to reflect age-related changes resulted in a leftward shift of the force-frequency function, lower firing frequency for any given current injected into the motoneuron, and a reduction in maximal force output. The model suggests that older individuals are capable of reaching force levels up to approximately 50% of those attained by younger individuals, with relatively similar or even slightly lower levels of current input. This could mean that the sense of effort and the contribution of factors other than degree of effort from afferent inputs to the pool, including conscious supraspinal centres, might be different in the older adult.

Key words: aging, strength, muscle, motoneuron, force, quadriceps, effort.

Résumé : Des modifications significatives au niveau des motoneurones et du tissu musculaire sont observées au cours du vieillissement; ces changements ont un effet significatif sur la force musculaire, la mobilité et l'autonomie globale. Les modèles mathématiques sont pratiques quand il s'agit d'analyser la relation entre la sommation spatiale, la sommation temporelle et la production de tension des unités motrices; on peut les utiliser comme prédicteurs des modifications neuromusculaires associées au vieillissement. Pour cette étude, le modèle de Heckman et Binder a été utilisé pour examiner des changements des unités motrices du quadriceps fémoral humain. On détermine tant chez un jeune individu que chez un plus vieux la relation entre l'arrivée du courant, la fréquence des potentiels d'action et la tension produite. On intègre dans le modèle les variables suivantes : les effets dus à l'âge concernant la diminution de la vitesse de contraction musculaire, la diminution du nombre de fibres musculaires, de leur calibre et de leur tension spécifique, la diminution du gain dans la relation fréquence-courant, la diminution de la taille des motoneurones et la restructuration des unités motrices altérées. Quand on teste ce modèle en fonction des modifications observées avec l'âge, on observe un déplacement vers la gauche de la fonction tension-fréquence, une diminution de la fréquence des potentiels d'action pour tout courant donné aboutissant au motoneurone et une diminution de la tension maximale produite. D'après ce modèle, les personnes âgées peuvent produire jusqu'à 50 % de la tension maximale produite par les jeunes en envoyant un courant de même intensité ou légèrement affaibli. Cela signifie que la sensation de l'effort et que la contribution des facteurs autres que le courant aboutissant au pool, y compris ceux des centres supramédullaires, pourraient différer chez les personnes âgées.

Mots-clés : vieillissement, tension musculaire, muscle, motoneurone, force, quadriceps, effort.

[Traduit par la Rédaction]

Introduction

As our knowledge of the neuromuscular system continues to grow with information gained from basic science and clinical studies, researchers have also developed mathematical models and computer simulations in attempts to further understand aspects of this physiological system. More than

50 years ago, the first theories proposing a sliding filament structure in skeletal muscle were developed (Huxley and Niedergerke 1954; Huxley 1953); since that time, numerous models have been developed to study a variety of different aspects related to muscle and its neurological connection. For example, models have been developed to investigate re-

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relationships among force output, recruitment, and rate-coding (Fuglevand et al. 1993; Heckman and Binder 1991; Moritz et al. 2005; Taylor et al. 2002); muscle activation, fatigue, and recovery (Liu et al. 2002); and the effects of electrical stimulation (Davidson et al. 2006; Perumal et al. 2006; Wexler et al. 1997).

A primary purpose of developing simulations is to quantitatively analyze the effects of changing parameters and assumptions of the system being modeled. In studying muscle and motoneuron properties, it is difficult to record information from more than the few motor units in close proximity to electrodes in human and animal studies. Therefore, motor unit models provide a practical method for studying the relative influence of recruitment and rate-coding, and for predicting force output from an entire motor unit pool. Although many studies have identified changes that affect skeletal muscle and its activation with aging (e.g., reduction in size and current threshold of motoneurons, reductions in the number of muscle fibres, atrophy of remaining fast-twitch fibres, and slowing of contractile speed), the relative influences of each of these changes on force production have not been simulated in a comprehensive model.

In 2003, Thelen (2003) developed a model to investigate age-related effects of reduced contraction velocity and increased noncontractile tissue on the force-velocity and force-length properties of muscle during concentric contractions. While this model was found to generate simulated ankle torque results consistent with those measured in healthy older adults, the focus was strictly on muscular changes with aging, and age-related alterations associated with neural activation were not included. In comparison, Barry et al. (2007) incorporated motor unit data gathered from experiments on the first dorsal interosseus muscle in older adults to investigate the relationship between variability in discharge rate and force steadiness with aging. Although this model included some parameters related to muscle activation, few motoneuron properties were represented. Heckman and Binder (1991) developed a model based on motoneuron activation properties and frequency-current relationships for low- and high-threshold motoneurons. The model consisted of a pool of 100 simulated motor units for which steady-state neural and mechanical properties were matched to experimental data for cat medial gastrocnemius motor units. It was designed to predict whole system motor input and force output functions based on single unit data for cats, and has since been modified to describe a similar application in rats (Gardiner 2001). However, it has not been previously used to describe motor unit characteristics in humans. As well, it has not been used to predict age-related changes to motor unit output.

The objectives of this study were to modify the model developed by Heckman and Binder (1991) to make it representative of human neuromuscular properties, based on available literature, and to incorporate typical age-related neuromuscular changes to determine the effects on force output for the aged condition. The quadriceps muscle was used in the model because it is one of the most thoroughly studied in terms of age-related strength and morphological changes (see review by Roos et al. 1997), and it is functionally important in terms of contributing to the ability to sit, stand, walk, and climb stairs, which are fundamental to maintaining mobility.

Materials and methods

The model originally developed by Heckman and Binder (1991) was used as the foundation for this study. It was developed to quantitatively study the relationship between synaptic input and force output for an entire pool of motor units during steady-state conditions. The Heckman-Binder model incorporated motoneuron excitability, tetanic force, twitch/tetanic properties, and contractile speed factors, along with rate-coding and recruitment properties required to generate isometric contractions. Simulations were based on experimental data gained from studying the medial gastrocnemius muscle of the cat. The motoneuron input to force output relationship was modeled for a pool of 100 motor units, based on motoneuronal frequency-current ($f-i$) and muscle unit force-frequency ($F-f$) relationships (for examples, see Figs. 1B and 1A, respectively). Motoneuronal $f-i$ functions were modeled using the following parameters: current threshold for repetitive firing at steady state (I_{thres}), firing frequency at current threshold (f_{thres}), gain (slope) of the $f-i$ relationship (G_1 in the primary range of firing and G_2 in the secondary range of firing), and the firing frequency at the transition from primary to secondary firing ranges (f_{trans}). Isometric $F-f$ functions of the muscle unit were described by the following parameters: maximal tetanic force (F_{max}), a frequency constant related to the speed of contraction and the left-right position of the $F-f$ curve (T_f), and an exponent (P) used to transform the exponential $F-f$ function to a sigmoidal function to better describe the relationship between force and frequency (Erim et al. 1996; Kernell 1983; Kernell et al. 1983).

$F-f$ curves were generated for each motor unit using the following formula (Heckman and Binder 1991):

$$F = F_{max} \times \left(1 - e^{(-\text{frequency} \times T_f) \times P} \right)$$

Readers are directed to review the original work of Heckman and Binder (1991) for a more detailed explanation of the model.

The model applied to a human motoneuron pool

Existing experimental human data were used (when available) to generate a human application representative of the quadriceps muscle in a younger person. The model was run in Microsoft Excel 2003. Justifications for adjustments to model parameters are explained below.

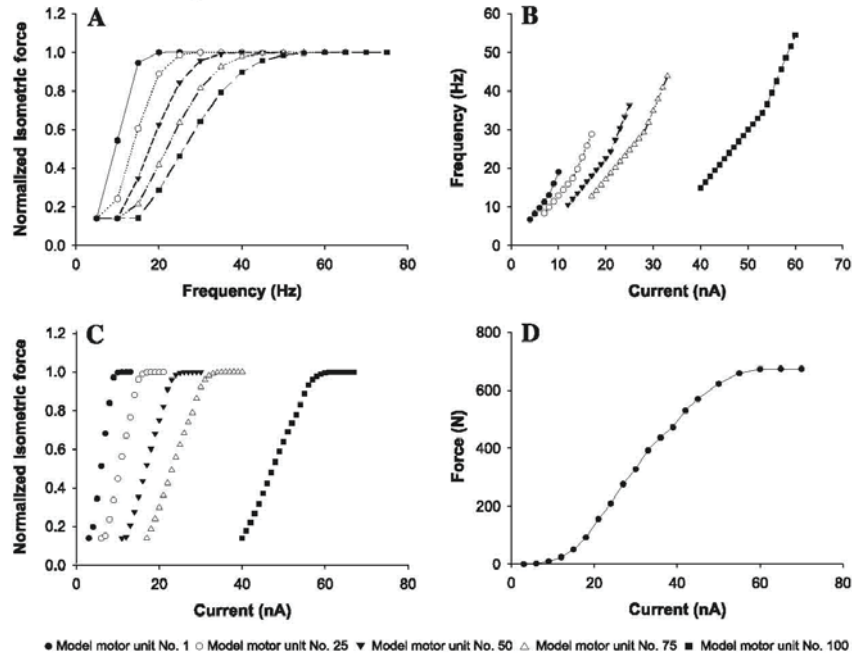
I_{thres}

Because there are no human data available describing rheobase or rhythmic firing thresholds of motoneurons, current threshold values in the younger human model were not modified from those used in the cat motoneuron model. The current threshold was set to vary from 3.5–40 nA, in agreement with experimental results from the cat and rat (Bakels and Kernell 1994; Baldissera et al. 1987; Munson et al. 1997).

f_{thres}

In the human model, the frequency threshold was set to increase linearly with recruitment threshold, but values were set slightly lower (approximately 6–15 Hz.) than those found in the cat model. This is consistent with the firing rates ob-

Fig. 1. Representative force–frequency relationships (A), frequency–current relationships (B), and force–current relationships (C) for model motor units. Force was normalized to maximum tetanic force for each motor unit (A and C). (D) Whole muscle force (N) vs. current (nA) injected into the entire motoneuron pool.



served in tibialis anterior motor units in younger people (Erim et al. 1996; Van Cutsem et al. 1997). To our knowledge, no data are available describing frequency thresholds for low- and high-threshold motor units in the quadriceps.

G_1 and G_2

Because there is no evidence to suggest that the gain on the excitatory drive–firing rate relationship differs in humans, G_1 and G_2 values from the cat model were used.

f_{trans}

In the cat motoneuron model, the firing frequency at the transition between primary and secondary firing ranges occurred at 80% of maximum force levels for individual motor units (Kernell 1979). For the human model, values for f_{trans} were set to 50% of the cat values (12.5–35.3 Hz.) to account for the lower mean firing rates in humans than in cats (Hornby et al. 2002).

F_{max}

Absolute tetanic force levels were scaled in increasing proportion exponentially to create a 100-fold range between the tetanic force produced by the lowest threshold motor units and force produced by the highest threshold motor units, as has been supported by work in humans (Milner-Brown et al. 1973; Stephens and Usherwood 1977; Van Cut-

sem et al. 1997). Values were adjusted so maximal stimulation of the entire motoneuron pool would result in a force output close to 650 N, which has been measured for the quadriceps muscle in younger people (Allman and Rice 2004; Roos et al. 1999). For simplicity, the model was generated for a pool consisting of 100 motor units, but in actual fact, the 4 muscles that constitute the quadriceps group likely each consist of a few thousand motor units (Lexell et al. 1988; Ling et al. 2007).

T_f and P

The T_f and P formulae include a value for the twitch/tetanus ratio. This constant was set at 0.08 in the cat model, but was changed to 0.14 for the human model, based on data from human quadriceps muscle (Scott et al. 2006).

Specific age-related changes incorporated into the model

Based on physiological changes associated with aging, model parameters were adjusted in simulations of the motoneuron pool to represent properties that would be found in an older adult (70–80 years of age).

Reduced muscle contractile speed

For the entire pool of motor units, values for the frequency constant related to the speed of contraction (T_f) were reduced to account for slowing in contractile speed

and changes in the F - f relationship that have been demonstrated to occur with aging (Allman and Rice 2004; Roos et al. 1999). The majority of researchers have reported longer contraction and half-relaxation times in older adults (see review by Roos et al. 1997), although not all researchers have found this (Trappe et al. 2003). Results vary substantially, with contraction duration reportedly being 3%–72% slower in older adults. One study that specifically examined properties of the quadriceps muscle from older men demonstrated that time to peak tension lengthened by 11% in men 73–91 years of age, compared with the younger group (19–35 years of age) (Roos et al. 1999).

Reduced muscle unit strength

Maximal force generation capabilities for individual motor units were reduced in a linear fashion; there was a 2% reduction in maximal force potential for each successive group of 5 motor units to represent changes related to loss of numbers of muscle fibres, reductions in size of fibres (preferentially affecting fibres in the high threshold motor units), and reductions in specific tension levels.

Studies have demonstrated that muscle-fibre numbers are reduced with age (Lexell et al. 1983, 1988). In comparing quadriceps vastus lateralis total muscle-fibre number between individuals in their 20s and 70s, Lexell et al. (1988) found a 40% reduction with age, which increased to 50% when younger individuals were compared with those in their 80s. Because preferential loss of a specific type of muscle fibre would result in altered fibre-type proportions with age, studies have examined relative changes in numbers of slow- and fast-twitch muscle fibres. However, while a number of biopsy studies have examined fibre-type proportions in the vastus lateralis muscle in younger and older subjects, the results are conflicting (Grimby et al. 1984; Larsson et al. 1979; Lexell et al. 1988; Trappe et al. 2003). Fibre-type proportions can be influenced by changes in fibre size, histochemical profile, and fibre number. As summarized in review papers by Porter et al. (1995) and Roos et al. (1997), at least part of the reason for the reportedly variable effect of age on muscle-fibre numbers may be related to the fact that biopsy results are not always representative of changes across the entire muscle. However, data from cross-sectional analyses of whole muscle suggest that the loss in number of muscle fibres does not differ between fibre types (Lexell et al. 1988; Lexell and Downham 1992).

Along with declining muscle-fibre numbers, muscle-fibre sizes also decrease with age, with type II fibres demonstrating the most significant reductions. As summarized in reviews by Vandervoort (2002) and Porter et al. (1995), a large range of quadriceps muscle-fibre size reductions has been noted in the literature. While type I fibres have demonstrated size reductions in the range of 1%–25%, type II fibres have shown greater atrophy, in the range of 10%–60%.

The majority of studies, although not all (Trappe et al. 2003), have found that specific tension (force expressed relative to muscle physiological cross-sectional area) also decreases with age. Studies have demonstrated that for the quadriceps muscle, specific tension is 14%–34% lower for

people aged 65–85 years than for those aged 20–43 years (Frontera et al. 2000; Larsson et al. 1997; Yu et al. 2007). Contrary to these findings, Trappe et al. (2003) found similar force per cross-sectional area results among younger and older women and men, and suggested that changes in specific tension with age may be limited to those with a particular activity profile. Despite possible debate about changes in specific tension with age, reductions in overall strength with age are well established. Although changes in absolute total muscle strength seem to be relatively minor until approximately 50 years of age, strength decreases at a rate of 1.0%–1.5% per year as people continue to age (Larsson et al. 1979; Vandervoort and McComas 1986). Peak isometric force levels for the quadriceps have been reported to be 31%–48% lower in older men (aged 70 years and older) than in younger men (Allman and Rice 2004; Roos et al. 1999; Young et al. 1985).

Reduced motoneuron gain

For the older model, G_1 and G_2 were reduced by 36% and motoneuron current threshold was reduced by 25%, as has been suggested by rat experimental data. No data are currently available to describe age-related changes in gain of the f - i relationship in humans because of the invasive nature of this measure. However, studies on rat motoneurons suggest that values for G_1 and G_2 are 36% lower in older animals (Kalmar et al. 2008). As well, older motoneurons are smaller and exhibit approximately 25% lower current thresholds in experiments on rats (Kalmar et al. 2008) and cats (Morales et al. 1987); however, human data are not available.

Loss and sprouting of motor units

To represent changes related to loss of motor units and re-innervation of some fast-type muscle fibres with low-threshold motoneurons, the model was adjusted so that 30% of the motoneurons were lost. As well, based on the work by Pettigrew and Gardiner (1987), low-threshold motor units were made larger (i.e., capable of producing greater amounts of force) and high-threshold motor unit size was not changed. Although denervation, axonal sprouting, and re-innervation of muscle occur throughout life as part of a regular motor unit remodeling process at neuromuscular junctions, with age this process is modified. While some motor units are simply lost, it is believed that some type II muscle fibres may be re-innervated by sprouting axons from low-threshold motoneurons, with subsequent conversion of muscle-fibre types from fast to slow (Howard et al. 1988; Kanda and Hashizume 1989; Stalberg et al. 1989). The overall result is a reduction in total motor unit number, with the possibility that remaining motor units consist of a greater number of muscle fibres in older people than in younger people (Doherty et al. 1993). Human studies have estimated the motor unit number loss to be about 50% in subjects over 60 years of age (Brown et al. 1988; Tomlinson and Irving 1977). As well, it has been demonstrated in the quadriceps muscles of older men that there are unusually high incidences of muscle fibres containing more than 1 myosin heavy chain isoform, which is consistent with the process of denervation and re-innervation (Klitgaard et al. 1990).

Table 1. Key model parameters representative of younger human quadriceps motoneuron pool.

Motor unit No.	I_{thres} (nA)	G_1	G_2	Frequency threshold (Hz)	Frequency transition (Hz)	Force max (N)	T_f	P
1	3.50	1.5	3.0	6.00	12.50	0.28	10.79	3.23
5	3.96	1.5	3.0	6.36	13.42	0.34	11.55	3.17
10	4.54	1.5	3.0	6.81	14.58	0.43	12.51	3.11
15	5.17	1.5	3.0	7.26	15.73	0.54	13.46	3.06
20	6.00	1.5	3.0	7.71	16.88	0.68	14.42	3.02
25	6.83	1.5	3.0	8.16	18.03	0.86	15.37	2.99
30	7.89	1.5	3.0	8.61	19.18	1.09	16.33	2.96
35	9.00	1.5	3.0	9.06	20.33	1.37	17.28	2.93
40	9.91	1.5	3.0	9.51	21.48	1.73	18.24	2.91
45	10.82	1.5	3.0	9.96	22.64	2.19	19.19	2.89
50	11.89	1.5	3.0	10.41	23.79	2.76	20.15	2.87
55	13.00	1.5	3.0	10.86	24.94	3.48	21.10	2.85
60	13.91	1.5	3.0	11.32	26.09	4.40	22.06	2.83
65	14.82	1.5	3.0	11.77	27.24	5.55	23.01	2.82
70	15.89	1.5	3.0	12.22	28.39	7.00	23.96	2.81
75	17.00	1.5	3.0	12.67	29.54	8.83	24.92	2.80
80	18.43	1.5	3.0	13.12	30.70	11.14	25.87	2.78
85	20.50	1.5	3.0	13.57	31.85	14.06	26.83	2.77
90	25.00	1.5	3.0	14.02	33.00	17.74	27.78	2.77
95	31.00	1.5	3.0	14.47	34.15	22.38	28.74	2.76
100	40.00	1.5	3.0	14.92	35.30	28.24	29.69	2.75

Note: I_{thres} , current threshold (nA); G_1 and G_2 , gain on the frequency-current relationship in the primary and secondary firing ranges, respectively; force max, maximum tetanic force; T_f , frequency constant related to the speed of contraction and the left-right position of the force-frequency curve; P , exponent used to transform the exponential force-frequency function to a sigmoidal function. See text for formulae used to determine motor unit forces from currents and the resultant motoneuronal firing frequencies.

Results

Model of younger human quadriceps motoneuron pool

A model of 100 motor units representative of motoneuron and muscle properties from the quadriceps muscle of a younger person was generated using the parameters in the Heckman-Binder model (Table 1). As outlined in Table 1, current thresholds ranged from 3.5–40 nA and frequency thresholds ranged from 6–14.92 Hz. All motor units in this model were recruited when 40 nA of current was delivered. At this current level, when all motor units reached firing threshold, the force output was equal to approximately 75% of maximum isometric force generation, similar to published data for the tibialis anterior muscle (Van Cutsem et al. 1997).

Simulated F - f relationships were generated for representative motor units, ranging from No. 1 to No. 100 (Fig. 1A). The f - i relationship is depicted in Fig. 1B, and the force-current (F - i) relationship is shown in Fig. 1C. It is evident that higher-threshold motoneurons had higher frequency thresholds and reached higher maximal firing rates. Variable levels of input current were simulated in the model to measure the resultant whole muscle force output (Fig. 1D). Maximal isometric force for the young quadriceps muscles rose steadily, with increasing current reaching peak values of approximately 675 N, which is consistent with previous studies of knee extension strength in humans (Allman and Rice 2004; Roos et al. 1999).

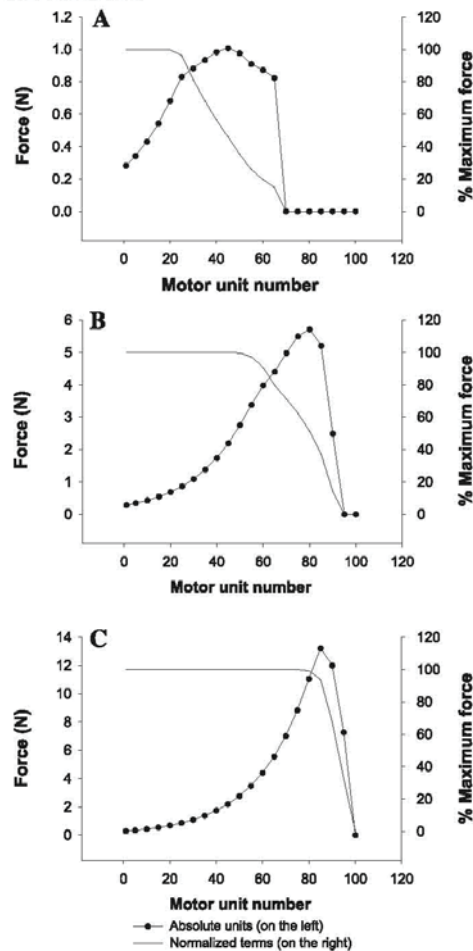
Figure 2 illustrates the relationship between the number of motor units activated and the resultant force per motor unit

(in absolute and normalized terms), when different levels of input current were injected into all motoneurons in the pool. Figures 2A, 2B, and 2C represent increasing levels of current (15, 25, and 35 nA, respectively). When 15 nA of current was injected into the motoneuron pool, approximately 25% of the motor units were fully activated (i.e., 100% maximum force was reached by 25 motor units in this diagram). The overall resultant force was 52.5 N (approximately 8% of total maximum isometric force). Approximately 65% of the motor units were recruited, with the lower-threshold motor units contributing relatively little absolute force (in the range of 0.25–0.65 N each), despite being maximally activated. Many of the slightly larger motoneurons were not fully rate-coded; yet, because of their size and the types of muscle fibres they innervate, they contributed relatively more force toward the total. In Fig. 2B, when 25 nA was injected into the model, a greater number of motor units were activated (approximately 90% were activated to some extent, with the total force output of 239.5 N, or 35% of maximum). Although motor unit Nos. 60–90 were not fully rate-coded, they generated much larger forces than the lower-threshold smaller motor units. Figure 2C demonstrates the system when the current was set to 35 nA and the total force output was 425.2 N, or 63% of maximum. This current was sufficient to recruit 95% of motor units in the pool, with the majority (80%) being fully rate-coded.

Model of older human quadriceps motor unit

The model for the younger human motoneuron pool was

Fig. 2. Generated sample relationships between the number of motor units activated and the resultant force (in absolute units on the left and normalized terms on the right) when different levels of current were injected into the model. (A, B, and C) Increasing levels of current (15, 25, and 35 nA, respectively). Each dot represents 5 motor units.



systematically modified to reflect changes in the neuromuscular system with aging. That is, specific changes in motoneuron and (or) muscle properties related to aging were first modeled individually to determine their effects on frequency and force output from the system. A composite model that incorporated all individually modeled changes was then generated to determine the overall effect of aging on force production.

The first modification made to the model involved reducing the frequency constant related to the speed of contraction (T_f) to account for slowing in contractile speed with aging. The resultant F - f plot and F - i plots are portrayed in Figs. 3A and 3B, respectively. F - f data are displayed for representative motor units (Nos. 1, 50, and 100). Plots for the representative motor units from the older individual are steeper and shifted to the left, relative to the younger motor units (Fig. 3A), and the overall force/current ratio is slightly elevated in the older model (Fig. 3B). Figure 3C depicts the change in F - i that occurred when the older motoneuron model was adjusted to simulate changes related to decreased specific tension, loss of muscle fibres, and decreased size of muscle fibres within individual motor units (and the resultant decrease in maximum force capacity for individual motor units). Maximal force generation capabilities of individual motor units were reduced by a factor that increased in a linear fashion, so that force production was reduced to a greater extent in the higher-threshold motor units than in the lower-threshold motor units (to reflect changes in numbers of muscle fibres and specific tension, which affect all motor units, and atrophy of type II muscle fibres that preferentially affect higher-threshold motor units). The younger motoneuron pool was capable of generating greater total force with lower input current, compared with the older motoneuron pool. Whereas maximal total muscle force reached approximately 675 N in the younger model, maximal force in the older simulation was 440 N.

Data obtained from experiments on rats suggest that G_1 and G_2 (the gain of the f - i function in the primary and secondary ranges, which reflects the excitatory drive-firing rate relationship) may decrease with aging (Kalmar et al. 2008). Reducing the values for G_1 and G_2 resulted in a significant reduction in the slope of the f - i relationship (Fig. 4A). To simulate age-related changes in motoneuron size that affect current threshold, I_{thres} was reduced by 25% for all motoneurons. This resulted in a greater slope in the F - i relationship and larger forces per given current for older motor units until a similar maximum force was reached (Fig. 4B).

The model was then adjusted to represent loss of motor units (low- and high-threshold units), with some re-innervation of type II muscle fibres by low-threshold motoneurons (Fig. 5). In this model, the total number of motor units represented was reduced by 30%, incorporating relatively equal loss of both low- and high-threshold motor units. Without adjustment for sprouting from low-threshold motoneurons (Fig. 5A), total force output was reduced to nearly half of what was available in the younger model. However, when motor units associated with the smallest motoneurons were given greater force potential (to represent a greater number of innervated muscle fibres), the total force generation capability for the quadriceps increased by approximately 100 N (Fig. 5A). A motoneuron model was then generated, incorporating all age-related factors, to determine the combined effect of reducing contractile speed (T_f); reducing muscle-fibre number, size, and specific tension; reducing the gain on the frequency-current (F - i) relationship; decreasing the size of motoneurons; and altering motor unit remodeling (Fig. 5B). Maximum isometric force levels and their associated current requirements are listed in Table 2 for both the younger and composite older moto-

Fig. 3. (A) Changes in the force–frequency relationship associated with slowing in contractile speed with aging in motor units Nos. 1, 50, and 100. The force–current relationship is depicted in (B). (C) Whole muscle force vs. current when the older model accounted for changes in the size of muscle fibres, the loss of muscle fibres, and the reduction in specific tension.

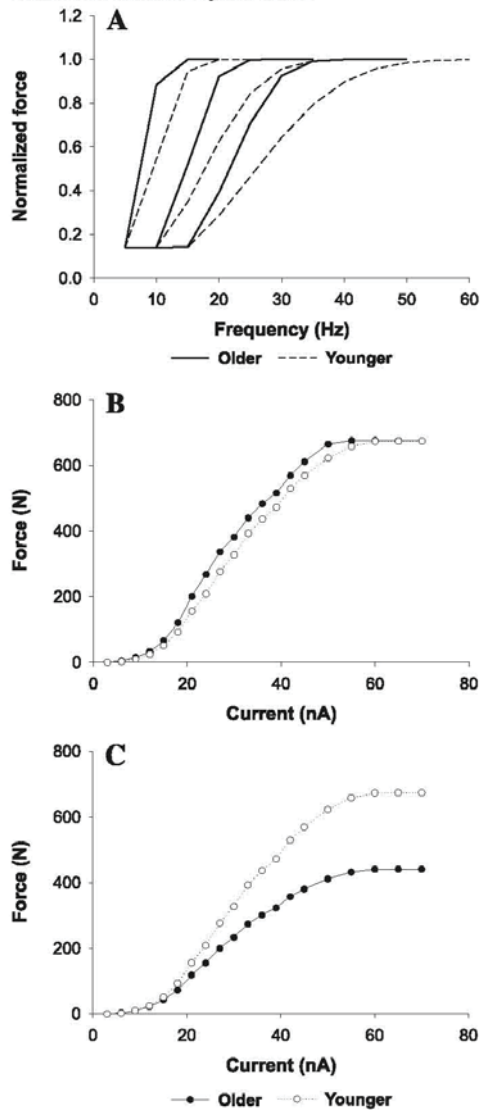
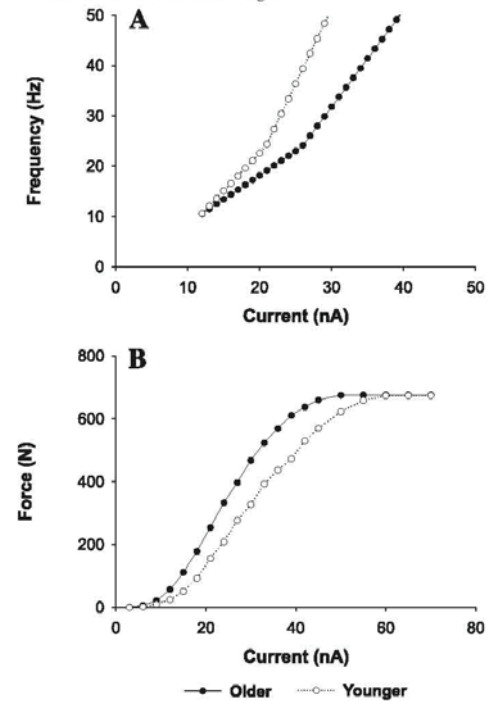


Fig. 4. (A) Frequency–current relationship for motor unit No. 50 in the model for younger and older motor units when G_1 and G_2 were reduced for the older simulation. (B) Force–current relationship for younger and older motor units when the older model was adjusted for lower current thresholds with age.

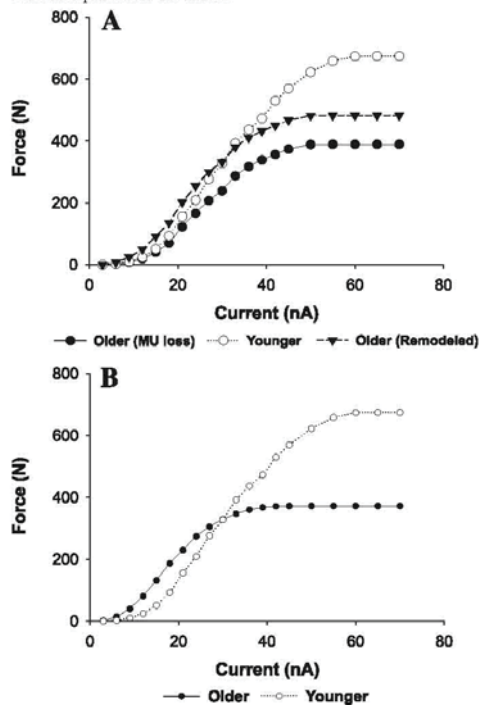


neuron pool models. Comparisons of current levels required to reach similar absolute force levels (10%, 25%, 50% maximum younger isometric force, 50 N, 300 N) are also listed for both models.

Discussion

Morphological changes in skeletal muscle with age result in loss of muscle mass caused by a combined reduction in total muscle-fibre number and fibre sizes (Porter et al. 1995). Motoneurons are not immune to the effects of aging, and also demonstrate age-related altered functional properties (Kalmar et al. 2008). Taken together, these result in significant age-related changes in the neuromuscular system, which may threaten an older individual's mobility and independence. Because aging affects both the motoneuron and the muscle fibres it innervates, models attempting to simulate changes in strength with aging need to be capable of manipulating both motoneuron and muscle properties. For this study, the simulation originally described by Heckman and Binder (1991) was utilized in a novel way to model changes in neuromuscular function with aging in human

Fig. 5. (A) Force–current relationship for younger motor units (MU), older simulation when motor units were lost, and older motor units when lower-threshold motor units were given greater force potential related to innervating a greater number of fibres with re-modeling. (B) Force–current relationship for the entire motoneuron pool for younger and older motor units when all age-related factors were incorporated in the model.



quadriceps motor units. The Heckman–Binder model was chosen for this application because it was designed to be used to determine mechanical force output in response to different levels of current input.

Both human models (the younger and older motoneuron pools) developed in this study were built on the foundations described by Heckman and Binder (1991) and Gardiner (2001). In comparing the model for the younger human motoneuron pool with that previously published for the cat and rat, it is evident that a number of differences and similarities exist. Because the twitch/tetanus ratio is greater in humans (0.14) than in cats (0.08), minimum normalized isometric forces were slightly higher in the human model. Otherwise, the shape of the normalized F - i and F - f relationships were very similar between the human and cat models. Although the F - i functions were not truly sigmoidal in shape in the human models, cat data from the Heckman–Binder model demonstrated a similar finding. Heckman and Binder (1991) stated that because the frequency threshold for each motor

unit was relatively high, the lower parts of the curves were missed (i.e., the sigmoidal shape would be more apparent if lower currents were capable of activating the muscle at lower forces). As well, they proposed that the existence of secondary firing ranges (influenced by G_2) resulted in relatively steep approaches to maximum force. It is expected that this reasoning also applies to the newly developed human models.

Although the frequency threshold and range of firing frequencies were set slightly lower in the human model, and values for T_f and P were adjusted for human data, these modifications did not appreciably change the isometric F - f relationship from that previously demonstrated for the cat and rat. Although the magnitude of maximal isometric force generation varies substantially among the species, depending on the amount of contractile protein available to be activated, the shape of the F - f relationship was similar.

Adjusting the human model parameters for changes expected to occur with aging resulted in some predictable findings. For example, accounting for slowing in muscle contractile speed with aging (and, therefore, summation of twitches at lower frequencies), the F - f plots for representative motor units demonstrated steeper slopes and were left-shifted, relative to those generated in the younger model (Fig. 3A). This has been found in numerous human studies involving the quadriceps muscle (Allman and Rice 2004; Roos et al. 1999), tibialis anterior (Connelly et al. 1999), and muscles of the hand (Barry et al. 2007). Because firing frequency is directly dependent on input current, the F - i relationship for the older model demonstrated a similar leftward shift (Fig. 3B). Taken together, these data suggest that an older individual can attain higher percentages of maximum force for a given current (and, therefore, for a given frequency) than a younger person.

Loss of muscle mass and strength are closely linked to declining function, dependence, and mortality in older adults (Roubenoff 2003). The fact that older individuals are capable of recruiting motor units as effectively as younger people (verified using twitch interpolation by Vandervoort and McComas 1986) suggests that reductions in strength result from changes in the muscle and its nerve supply, rather than from centrally generated motor commands. By the seventh decade, isometric maximum voluntary knee extension torque has been reported to be reduced by 25%–49% (Larsson and Karlsson 1978; Roos et al. 1999; Stalberg et al. 1989; Young et al. 1985). Although loss of muscle mass may be related to loss of muscle-fibre numbers and to atrophy of remaining fibres (largely type II muscle fibres), it is believed that changes in cross-sectional area are more greatly affected by changes to the total fibre number (Lexell and Downham 1992). Figure 3C demonstrates the change in total muscle force demonstrated when the motor unit model was adjusted to account for changes in muscle-fibre number and atrophy of type II muscle fibres with aging. Whereas maximal total muscle force reached approximately 675 N in the younger model, maximal force in the older simulation was 440 N. These results, which represent a 35% reduction in maximal isometric torque, closely agree with previously published findings from experiments involving younger and older individuals (Allman and Rice 2004; Roos et al. 1999; Young et al. 1985). Figure 3C depicts little difference be-

Table 2. Comparison of current requirements in the younger and composite older motoneuron pool models.

Current requirements	Younger model (nA)	Older model (nA)
Current associated with maximum isometric force	64	45
Current associated with 337 N (50% maximum force in younger model)	31	31
Current associated with 168 N (25% maximum force in younger model)	22	17
Current associated with 68 N (10% maximum force in younger model)	16	11
Current associated with 50 N force	15	10
Current associated with 300 N force	28	26

Note: The maximum isometric force is 676 N for the younger model and 376 N for the older model. nA, current.

tween force output for a given current at low levels of current (or during tasks requiring low-moderate effort), but as task difficulty and required current input levels increase, the F - i relationship diverges, with older individuals being unable to produce the higher forces reached by younger individuals.

Kalmar et al. (2008), in their studies on rats, demonstrated that the slope of the f - i relationship decreases with age, effectively resulting in a lower firing frequency for the same given current in an older animal. They also determined that motoneuron size decreases with age, resulting in lower current thresholds in older rats (i.e., lower rheobase and higher input resistance). Similar findings have been reported in the cat (Morales et al. 1987). These 2 age-related changes have opposite effects on frequency and current (i.e., reducing G_1/G_2 reduces frequency for a given current, and reducing current threshold acts to increase frequency for a given current) (Table 3).

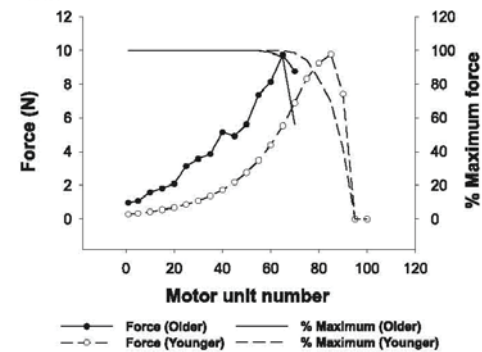
A primary goal of this modeling exercise was to determine the combined effects of age-related changes in motoneuron and muscle properties in terms of input current and force output. Effective synaptic current input may be related to the degree of voluntary effort required to exert force in these human models. Data demonstrated in Fig. 5B and listed in Table 2 show total muscle force levels relative to current input for younger subjects and older subjects (with the older model accounting for reduced muscle contractile speed, reduced muscle-fibre number and size, reduced gain on the f - i relationship, decreased size of motoneurons, and altered motor unit remodeling). Although the models suggest that younger individuals are capable of reaching much greater absolute maximal muscle forces (676 N vs. 376 N), the current level required to reach 50% of a younger person's maximal isometric force is the same for both younger and older individuals (31 nA input results in a force output of 337 N in both models). As is demonstrated in Fig. 6, 31 nA of current input results in maximal force output for most motor units (55 of 70) in the older model, whereas in the younger model, virtually the same current input results in maximal force output from 70 of 100 existing motor units. In the older model, more force is provided by all activated motor units than in the younger simulation.

Reducing the gain on the f - i relationship (G_1 and G_2) in

Table 3. Firing frequencies for given currents in the younger model, older model adjusted for changes in G_1/G_2 , and older model modified for changes in I_{thmax} .

Input current (nA)	Motor unit No.	Younger model (Hz)	Older model; G_1 and G_2 reduced (Hz)	Older model; I_{thmax} reduced (Hz)
12	1	25.0	15.8	27.6
	25	15.9	13.1	18.9
	50	10.6	10.5	15.0
30	1	79.0	50.4	81.6
	25	67.8	42.8	72.9
	50	51.4	31.8	60.3

Note: G_1 and G_2 , gain on the frequency-current relationship in the primary and secondary firing ranges respectively; I_{thmax} , current threshold (nA).

Fig. 6. Relationship between the number of motor units activated and the resultant force (in absolute and normalized terms) when 31 nA of current was injected into the older model and younger model.

the older model effectively resulted in lower firing frequencies for any given current. Therefore, as explained in the example above, when virtually the same current (31 nA) was injected into both the younger and older simulations, although the total force output was identical, generation of the force was accomplished at lower frequencies in the older model. This result is in agreement with the study by Barry et al. (2007), who found that although minimal firing rates were similar in the first dorsal interosseus muscle in younger and older people, peak discharge rates were lower in the older group, which resulted in a lower range of rate-coding potential for each motor unit.

The fact that equivalent input current resulted in the generation of 337 N of force in both models suggests that, despite the fact that this represents 90% of maximal force production in the older person, the degree of effort required to exert this force level might not be appreciably different between the 2 age groups. In fact, the models suggest that for resistance tasks requiring <50% maximal voluntary isometric force (relative to the younger individual's capabilities), the degree of input current required by older subjects is actually slightly less than that required by younger individuals (Fig. 5B). This finding is in agreement with the fa-

Table 4. Input current required to generate isometric force (when specific age-related motoneuron and motor unit properties were tested with different values to assess the sensitivity of the model to these specific changes).

Model	Current requirements (nA) for 68 N (10% maximum force in younger model)	Current requirements (nA) for 168 N (25% maximum force in younger model)	Current requirements (nA) for 337 N (50% maximum force in younger model)
Younger model	16	22	31
Older composite model	11	17	31
Older model with G_1 and G_2 unchanged from younger values	10	15	28
Older model with motoneuron current thresholds unchanged from younger values	15	25	48
Older model with 50% of motor units lost	12	21	337 N not possible, maximum force is 280 N

tigue-related literature that has shown that for submaximal repetitive tasks, older individuals may demonstrate greater fatigue resistance than younger subjects on some standardized tasks (Kent-Braun et al. 2002; Lanza et al. 2004). Preferential atrophy of type II muscle fibres, combined with the fibre-type transition from type II to type I, which is thought to occur when fast-twitch fibres are re-innervated by surviving lower-threshold motoneurons, results in an increased percentage of type I muscle fibres (Trappe et al. 2003), which are inherently fatigue resistant. In addition, other factors, such as a lower rate of decline in motor unit firing rate with fatigue (Rubinstein and Kamen 2005), may contribute to fatigue resistance in older adults.

As recognized by Heckman and Binder (1991) and Gardiner (2001), this model is not without its limitations. One of the major shortcomings relates to the fact that the model does not put any upper limits on firing frequencies for motoneurons, allowing the lower-threshold units to reach supratetanic frequencies before the largest motoneurons are activated. In actual fact, studies of voluntary isometric conditions in humans have demonstrated firing frequencies in the range of 15–25 Hz for the triceps brachii (Del Valle and Thomas 2004) and 10–50 Hz for the quadriceps muscle (Knight and Kamen 2008; Pucci et al. 2006) across contractions of different intensities from 25% to 100% maximal voluntary contraction. As well, studies of older adults suggest that maximal firing rates may be reduced with age (Kamen et al. 1995; Kamen and Knight 2004; Nelson et al. 1984); therefore, this model may specifically overestimate older adults' potential to generate force, because maximal firing rates are not restricted.

This model simulates slow isometric contractions, and more extensive modeling would be required to represent rapid contractions where force thresholds of motor units differ (Desmedt and Godaux 1977). It is assumed that remodeled motor units in this model will produce forces scaled relative to the extent of remodeling (i.e., increased innervation ratios). For various reasons (neuromuscular junction failure, distal axonal block, less biomechanical efficiency), this may not be the case during sustained contractions or during rhythmic contraction during exercise. In addition, our model does not account for late adaptation, which can result in reduced motor unit firing rates during sustained constant current injection (Kernell 1979). Results of animal (Kalmar et al. 2008) and human (Erim et al. 1999) studies suggest that motor units in older individuals demonstrate

less adaptation, which may actually benefit older subjects if this results in lower effort requirements to maintain force. Animal research also indicates that the incidence of persistent inward current is greater in older rats (Kalmar et al. 2008). This also could contribute to greater force production with less effort if motoneurons in older adults are more depolarized and require less current to reach firing threshold.

Our model deals with the well-studied isometric condition; however, dynamic concentric and eccentric contractions are often more relevant to human function. Substantial modifications would be required to use the model to predict force in a dynamic task. It is known that relationships among current, frequency, and force vary with different types of contractions. Recruitment thresholds are lower during concentric, as opposed to isometric, contractions, even when the velocity is very low, and initial firing frequencies are dependent on the type of contraction being executed (Tax et al. 1989). Although the simulation sums forces from individual motor units to determine total maximum force, 1 group of researchers has demonstrated in studies on cats that there is actually not a linear relationship between summed tensions and total force (Troiani et al. 1999). As well, our model included only 100 motor units, a small percentage of the actual number of motor units that would be found in the quadriceps. Spreading the same ranges of values for different model parameters over a much larger pool of motor units would result in smoother, steadier force generation. Overall, using this model to estimate force production in humans is also limited by the lack of human data available to represent some parameters in the model.

Although a comprehensive sensitivity analysis was not conducted to illustrate the relative influence of all age-related parameter changes incorporated in the model, analyses were conducted to demonstrate force generation capabilities predicted by the model if certain parameters had not been altered from the younger model. Specifically, the model was adjusted to determine current input levels and subsequent force production when G_1 and G_2 were maintained at younger values, when input current thresholds were unchanged from the younger model, and when a greater percentage (50%) of motor units were lost (Table 4). Preserving the gain on the $f-i$ relationship in the older model predictably had relatively little influence on input current requirements to generate low-, moderate-, and high-level forces (10%, 25% and 50% maximal force levels obtained in the younger model, respectively). Similarly,

in the alternate model, in which 50% of motor units were lost, current requirements did not change substantially to generate low and moderate force levels, but this model indicated that an older person would not be able to reach force levels equal to 50% of maximal values in the younger individual. As illustrated in Table 4, when current thresholds were maintained at levels equal to those used in the younger model, substantially greater amounts of input current were required to generate equal amounts of force. This would be expected, given the reductions in individual motor unit maximum force potential that were incorporated into the model. These analyses demonstrate that the changes in motoneuron current threshold that were incorporated in the composite older model were largely responsible for the changes in the F - i relationship with age. Since the parameters tested are those for which comprehensive information is currently lacking, more emphasis should be placed, in future, on these changes with age to better understand alterations in voluntary neuromuscular function in older adults.

In summary, modifying the Heckman-Binder motoneuron model for human data (where available) resulted in the development of f - i , F - f , and F - i functions for the human quadriceps muscle. Adjusting this model to reflect age-related changes in motoneurons and muscle tissue resulted in a leftward shift of the F - f function, lower firing frequencies for any given current, and reductions in maximal force output. Interestingly, results generated from the composite model for the older neuromuscular system (which included the combined effects related to reduced muscle contractile speed; reduced muscle fibre number, size, and specific tension; reduced gain on the f - i relationship; decreased size of motoneurons; and altered motor unit remodeling) suggest that older individuals may be capable of reaching force levels up to approximately 50% of those attained by younger individuals with relatively similar or even slightly lower levels of current input. This suggests that the degree of effort associated with functional tasks that require lower levels of force output may not be significantly different between younger and older individuals. Because muscle weakness is associated with decreased mobility and morbidity with aging, simulating the effects of these changes and their influence on potential force production may provide further insight into how age-related changes in the neuromuscular system affect functional strength levels, which are important to overall health and well-being in older adults.

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