FINAL REPORT

Muscle strength and age effects in balance recovery

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ABSTRACT

The purpose of this project was to investigate a potential mechanism by which age-related reductions in muscle strength contribute to falls in older adults. Tripping and falling accidents are a major problem in both occupational and non-occupational settings. Older adults in particular have a higher incidence of falling and fall related injuries. Despite numerous studies that have implicated age-related strength reductions in the higher incidence of older adult falls, the mechanism by which strength reductions contribute to falls remains unclear. Determining this mechanism will help toward understanding why people fall, and contribute to the development of intervention strategies aimed at fall prevention.

Age-related strength reductions may contribute to the higher incidence of older adult falls by limiting muscle forces during balance recovery. Insufficient muscle forces during balance recovery would not adequately decelerate trunk angular momentum and/or would not prevent the stepping leg from collapsing. As a result, the individual would fall.

An existing experimental model of balance recovery was used to evaluate insufficient muscle forces during balance recovery as a mechanism contributing to tripping and falling accidents in older adults. The experiment involved repeatedly releasing subjects from progressively larger stationary forward leaning postures to simulate a tripping accident. Lower extremity muscle forces during balance recovery (quantified using joint torques) were estimated using an inverse dynamics analysis, and compared to experimentally-collected strength measurements.

Muscle forces increased as the forward lean from which subjects were released increased. Ankle muscle force during balance recovery approached its maximum during recovery from the largest achieved lean angle, and the ankle muscle force that was predicted to be required for the next largest lean angle exceeded 100% of subject strength. These results imply that ankle strength limited the largest achieved lean angle (i.e. balance recovery ability), and suggest that increased ankle strength would increase the largest achieved lean angle. Further research is needed to determine if these results can be generalized to falls during walking (i.e. trips and slips) and falls outside of the laboratory.

HIGHLIGHTS/SIGNIFICANT FINDINGS

Hip, knee, and ankle muscle forces increased as subjects were released from larger stationary forward leans (*Hypothesis 1*). Ankle muscle forces during balance recovery approached maximum ankle strength during recovery from the largest achieved forward lean, and the ankle muscle force that was predicted to be required for the next largest forward lean exceeded 100% of the ankle strength (*Hypothesis 2*). These results imply that ankle strength limited the largest lean (i.e. balance recovery ability), and suggest that increasing ankle strength would increase the largest achieved forward lean. Somewhat similar trends were found at the knee, but these results did not reach statistical significance.

TRANSLATION OF FINDINGS

Results from this study imply that ankle strength limits the ability to recover from a forward fall with a single step, and suggest that increasing ankle strength will increase balance recovery ability in the laboratory. These findings could have implications on fall prevention interventions by emphasizing strength exercises at the ankle. However, further work is first needed to determine if these results can be generalized to falls during walking (i.e. trips and slips) and falls outside of the laboratory.

OUTCOMES/RELEVANCE/IMPACT

This study provides preliminary evidence that ankle strength may play an important role in limiting the ability of an older adult to recover his/her balance when falling. As such, emphasizing ankle strength exercises in later life may help to prevent falls. Further work is needed before these results can be generalized to falls outside of the research laboratory.

SCIENTIFIC REPORT

Background

Falls are a leading cause of injuries and fatalities, both in occupational and non-occupational settings. Almost 18% of all accidents involving absence from work for more than three days were found to be related to slipping, tripping, or falls on the same level (Leamon and Murphy, 1995). In 1992, falls resulted in 12,400 fatalities, representing the single most common cause of death due to accidents in the home and public sector (National Safety Council, 1993).

Falls are a particular problem in older adults due to higher fall rates and a greater likelihood of injury and death from a fall. An estimated 30-40% of community-dwelling adults over 65 years old fall each year (Blake et al., 1988,Horak et al., 1989). Eleven percent of these falls result in serious injury (National Safety Council, 1996). The high rate of falls among the older adults coupled with the growth in the older adult population and workforce (DHHS (NIOSH), 1996,Schultz, 1992) warn of an onerous public health problem.

In an effort to determine the causes behind the elevated fall rates in older adults, researchers have studied balance recovery from large-scale balance perturbations (Wojcik et al., 1999,Thelen et al., 1997,Luchies et al., 1994,Wojcik et al., 2001,McIlroy and Maki, 1996,Schultz et al., 1997,Do et al., 1982). These studies have isolated specific biomechanical factors and attempted to determine their role in balance recovery. For example, the reaction time between a balance perturbation and the initiation of a step is longer in older adults compared to young adults. This increase (on the order of 10 msec) appears to have little influence on balance recovery capability, though, since it is less than 5 % of the total time required to complete a single-step response (approximately 500 msec) (Thelen et al., 1997,Luchies et al., 1994,Wojcik et al., 1999). This suggests that the sensory processes or motor control involved in response initiation plays a relatively minor role in balance recovery (Schultz et al., 1997). In contrast, a decrease in the step velocity of older adults during balance recovery seems to be largely responsible for the decreased ability of older adults to recover balance (Thelen et al., 1997,Wojcik et al., 1999). This decrease in step velocity may be due to a decline in the ability of older adults to rapidly develop joint torques, resulting from a decrease in type-2 muscle fibers (Thelen et al., 1996,Clarkson et al., 1981,Hakkinen and Hakkinen, 1991).

Reductions in muscle strength may also contribute to elevated fall rates in older adults. One study reported that nursing home residents with a history of falls exhibited less than half of the knee and ankle strength compared to non-falling residents (Wolfson et al., 1995). Another study reported that healthy community-dwelling older adults with a history of falls exhibited a reduction in muscle strength compared to non-fallers (Gehlsen and Whaley, 1990). The aging process itself causes a 12-15% reduction in muscle strength each decade starting at about 50 years of age (Hurley, 1995). Quadriceps muscles strength, for example, is 20-40% less in older adults aged 60-80 years old compared to young adults (Porter et al., 1995). Similarly, ankle plantar flexor strength in older adults is 28-38% less than young adults (Davies et al., 1986). The main reasons for these age-related reductions in strength are a loss of muscle fibers, and a decrease in muscle fiber size (Porter et al., 1995).

Despite extensive epidemiological evidence linking age-related strength reductions to elevated falls rates in older adults, the mechanism by which strength reductions contribute to falls remains unclear. Few studies have examined joint torques during recovery from large-scale balance perturbations to investigate this mechanism. A recent study reported poor correlation between selected muscle strengths and the ability of older adults to recover balance from an induced forward fall (Wojcik et al., 2001). The selected muscle strengths (ankle plantarflexion of the non-stepping leg and hip flexion of the stepping leg) were involved in the step initiation phase of balance recovery (Figure 1). The critical relationship between muscle strength and balance recovery may not exist during the step initiation phase of balance recovery, but instead when the body is being supported after taking a step. Pavol et al. (2001) reported that at least 44% of older adult fallers fell after an induced trip despite completing a balance recovery step. These falls were attributed to the inability of subjects to simultaneously re-stabilize the trunk and resist buckling of the stepping leg during the support phase of balance recovery (SPBR). This inability may be due to insufficient joint torques in the stepping leg. Therefore, age-related strength reductions may contribute to the higher incidence of older adult falls by limiting joint torques during the support phase of balance recovery. Joint torques during the step initiation phase of balance recovery have been quantified (Wojcik et al., 2001), but no studies have quantified joint torques during the support phase of balance recovery.

An age-related redistribution of lower extremity joint torques during balance recovery may also contribute to falls in older adults. A recent study indicated that age causes a redistribution of joint torques and powers during the stance phase of gait (DeVita and Hortobagyi, 2000). Other studies have identified several lower extremity joint torque differences during gait between older adults subjects with a history of unexplained falls and older adults subjects with no history of falls (Kerrigan et al., 2000,Lee and Kerrigan, 1999). Although the experimental protocols of these studies involved gait, the results demonstrate the potential for age-related differences in joint torques during the support phase of balance recovery. Any joint torque redistribution during balance recovery could place increased torque demands on a joint already weakened by aging, and potentially exacerbate any limiting effects that muscle strength already has on balance recovery capability.

In summary, the mechanism(s) by which age-related reductions in muscle strength contribute to falls in older adults remains unknown. An existing experimental model of balance recovery (Thelen et al., 1997) will be used to evaluate insufficient joint torques in the stepping leg during the support phase of balance recovery as a mechanism contributing to tripping and falling accidents. Baseline measurements of joint torques during balance recovery will be collected from a young adult population. These measurements will be compared with similar measurements from older adult subjects to investigate age-related differences in joint torques during balance recovery. The proposed research will also determine if healthy older adults exhibit a redistribution of joint torques during the support phase of balance recovery. These results are needed to better understand the biomechanical factors that contribute to falls in older adults, and to facilitate more effective intervention strategies aimed at fall prevention.

Specific Aims

An existing experimental model of balance recovery will be used to evaluate insufficient joint torques during the support phase of balance recovery as a mechanism contributing to tripping and falling accidents in older adults. The experimental model involves releasing subjects from a forward-leaning posture to simulate a tripping accident where the body center of mass translates forward beyond the base of support. Stepping leg joint torques will be estimated during the support phase of balance recovery using an inverse dynamics analysis.

Specific Aim 1: To establish baseline measurements of stepping leg joint torques during the support phase of balance recovery in young, healthy adults aged 18-25. These joint torques will be used with muscle strength measurements to 1) evaluate if joint torques during balance recovery increase as the magnitude of a balance perturbation increases, and 2) determine if muscle strength limits joint torques at the largest lean from which balance can successfully be recovered upon release.

Hypothesis 1: Peak joint torques in the stepping leg during the support phase of balance recovery will increase with increasing lean magnitude from which subjects are released.

Hypothesis 2: At least one joint torque in the stepping leg during the support phase of balance recovery will approach its maximum when subjects are released from the largest achieved lean magnitude.

Specific Aim 2: To quantify stepping leg joint torques during the support phase of balance recovery in healthy, older adults aged 55-65. In addition to repeating the analysis from specific aim 1 for older adult subjects, these data will be used to investigate an age-related redistribution of peak joint torques during the support phase of balance recovery. Such a redistribution has been demonstrated during gait, but has not been studied for balance recovery. This redistribution could place increased torque demands on a joint already weakened by aging, and potentially exacerbate any limiting effects that muscle strength already has on balance recovery capability.

Hypothesis 3: Older adults will exhibit an age-related redistribution of peak extensor joint torques in the stepping leg during the support phase of balance recovery.

<u>Methods</u>

Fourteen young adults (aged ±) and fourteen older adults (aged ±) participated. Both age groups included an equal number of males and females. Participants had no self-reported musculoskeletal, neurological, cardiovascular, or cognitive disorders, participated in physical activity 2-4 days per week, were right-footed, and reported no episodes of musculoskeletal injury in the previous 6 months. In addition, older participants were required to pass a medical screening performed by a physician (detailed history and physical examination). The experimental protocol was approved through the Virginia Tech Institutional Review Board, and all subjects provided informed consent prior to participation.

The experimental protocol was adapted from Wojcik et al. (8), and has been reported (7). Trips were simulated by releasing participants from a forward-leaning posture. After release, participants attempted to recover their balance using a single step of the right foot. Successful recoveries were followed by another trial at a larger lean, and failed recoveries were followed by a second trial at the same lean. This process was repeated until participants failed to recover their balance with a single step for two consecutive trials at the same lean. The ability to recover from a fall was quantified by the maximum lean from which participants could recover their balance with a single step after being released ($Lean_{MAX}$).

To start each trial, participants stood with their feet shoulders-width apart at a toe line and were leaned forward. Participants were held in this forward-leaning posture using a lean support rope spanning from the back of a belt worn by the participants to a releasable clasp affixed to a stable wooden structure. In this position, participants were asked to equally distribute their weight across both feet while maintaining heel contact with the ground. Participants were asked to keep their arms folded across their chest throughout each trial. Once the participants were in position at the correct lean, they were verbally reminded to take a single step with their right foot for recovery. Participants were released without warning 0-10 seconds after this verbal reminder. The initial lean corresponded to 10 degrees measured between vertical and a line connecting the ankle and greater trochanter, and lean was increased by 5 degrees after each successful recovery. In the event of an unsuccessful recovery, falls to the ground were prevented using a full-torso harness tethered to a ceiling-mounted support track with a fall-prevention lanyard. Three criteria were used to define a failed recovery: 1) when more than one step was taken with the right foot, 2) when more than 30% BW force was applied to the harness at any point during trip recovery, and 3) when the left foot took a step longer than 30% of the participant's body height. All participants practiced the single step balance recovery prior to the start of the experiment.

A battery of muscle strength tests were performed to determine the maximum torque generating capacity in the right ankle, knee, and hip. These data were used to express joint torques during balance recovery as a function of their maximum joint generating capacity. These tests included isometric and isokinetic (concentric and eccentric) maximum voluntary contractions (MVCs), each performed in three exertion directions: hip extension, knee extension, and ankle plantar flexion. These joint/direction combinations were selected based on evidence that only extensor torques are employed during SPBR. All testing was done on the right leg with a Biodex System 3 dynamometer (Biodex Medical Systems, Inc., Shirley, New York, USA).

Body segment positions during balance recovery trials were sampled at 200 Hz using a Vicon 460 Motion Analysis System (Lake Forest, CA). Infrared markers were placed on the right side of the body at the fifth metatarsal head, heel, lateral malleolus, lateral femoral epicondyle, greater trochanter, and acromion. Marker data were filtered with a 4^{th} order, 7 Hz low-pass, zero-phase-shift Butterworth filter. In the initial forward-leaning posture, a forceplate was under each foot (AMTI, Watertown, MA). Subjects stepped onto a third forceplate (90cm $\times$ 90cm, Bertec Corp., Columbus, OH) during recoveries to provide ground reaction forces during SPBR. Forceplate and harness load cell data were sampled at 1000 Hz, and load cell data were subsequently filtered with a 4^{th} order, 10 Hz low-pass, zero-phase-shift Butterworth filter. To estimate joint torques in the stepping leg during SPBR, the body was modeled as a 2D system of 4 rigid body segments, connected by frictionless pin joints, that included the right foot, right shank, right thigh, and a head/arms/trunk (HAT) segment. The mass and inertial characteristics of the body segments were defined using an anthropometric model (15). Sagittal plane joint torques were estimated for the right ankle, knee, and hip using the governing Newton-Euler equations as described by Winter (16). Peak extensor torques at the right ankle, right knee, and right hip were determined for SPBR, and expressed as a percentage of the maximum joint torque measured during strength testing. This provided a estimate of relative effort during SPBR. The

SPBR was operationally defined to be the time interval between foot contact and the instant when both knee flexion velocity and HAT forward angular velocity had reached zero (11).

Results

Older adults exhibited a smaller maximum lean angle compared to young adults (older = 16.8±3.2 degrees; young = 23.6±2.3 degrees; $p<0.001$).

Hip, knee, and ankle torques in the stepping leg were predominantly extensor (or plantar flexor) dominant during SPBR. As such, our analysis was limited to peak extensor torques. Peak extensor torques during SPBR (Figure 1) increased with increasing lean angle at the hip ($p=0.013$), knee ($p<0.001$), and ankle ($p<0.001$). Peak extensor torques at the hip, knee, and ankle exhibited no age effect or age × lean angle interactions. Based on the increase in peak extensor torque with increasing lean angle, we will accept our first hypothesis. Based on the lack of an age effect or an age × lean angle interaction, we will reject our third hypothesis.

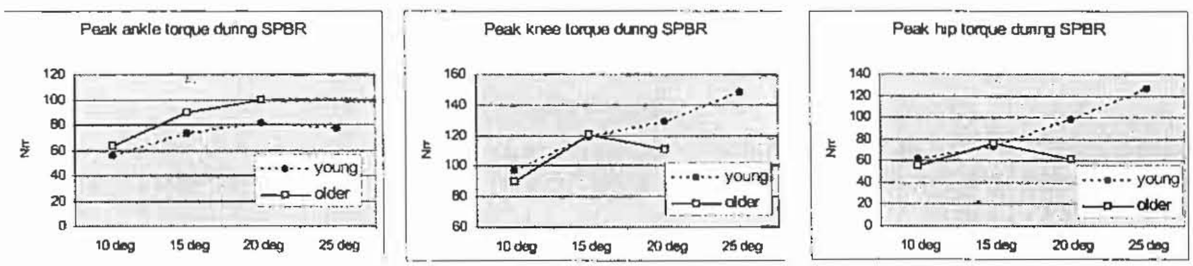

Figure 1 – Peak extensor (plantar flexor) torques for young and older adults during SPBR. These data illustrate the increase in peak extensor torques as lean angle increases. Error bars omitted for clarity.

Hip, knee, and ankle torques expressed as a percentage of maximum torques measured during strength testing are shown in Figure 2. The peak ankle plantar flexor torque that would be required for subjects to recover from a lean angle 5 degrees larger than their maximum achieved lean angle was greater than 100% of their strength ($p=0.011$). This was not the case for the knee ($p=0.377$) or the hip ($p>0.900$). Based on these results, we will accept our second hypothesis.

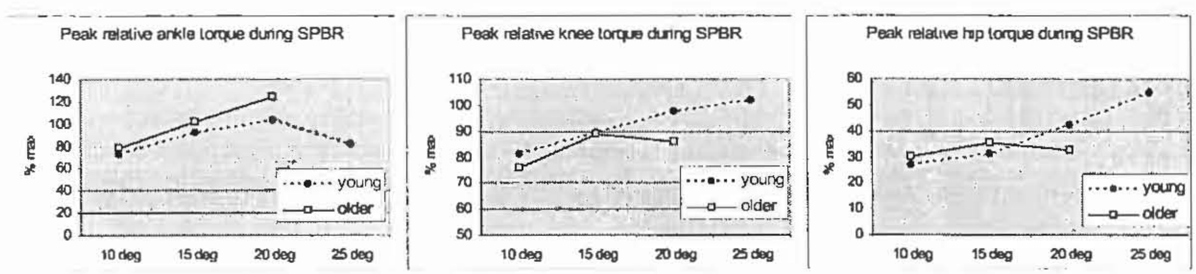

Figure 2 – Peak extensor torques for young and older adults during SPBR expressed as a percentage of the maximum torque measured during strength testing. These data illustrate the relation between these torques and the maximum available torque (100% of max). Note that some measurements exceeded 100% max, which is discussed below. Error bars omitted for clarity.

Discussion

The purpose of this study was to investigate a mechanism by which age-related reductions in muscle strength contribute to falls in older adults. Determining this mechanism will help toward understanding why people fall, and contribute to the development of intervention strategies aimed at fall prevention. We used an existing experimental model of balance recovery that involved repeatedly releasing subjects from a progressively larger forward-leaning postures to simulate a tripping accident where the body center of mass translates forward beyond the base of support. The maximum achieved lean angle was a surrogate measure for balance recovery ability. During balance recovery, peak joint torques increased as the angle from which subjects were released increased. This suggests an increased strength requirements as the angle (i.e. task difficulty) was increased. At the largest achieved lean angle, the peak ankle torque approached the its maximum value measured during strength testing. We then showed that peak ankle torque that would have used during recovery from the next highest lean angle was above the subjects' strength capability. This implies that increased strength would be required to increase the maximum achieved lean angle. In other words, this suggested that ankle strength limited the balance recovery ability. We also found no evidence of a redistribution of peak extensor torques in older adults that previous research hypothesizes may occur as a compensatory mechanism used to mitigate the deleterious effects of muscle strength on balance recovery ability.

Comparing these results to our earlier work reveals some similarities and differences. The increase in peak extensor torques is consistent with findings from an earlier study in our lab (Madigan and Lloyd; 2005) that included adults older than those used in the current study (aged 65 and older). The previous study did not include strength measurements, so the findings in the current study regarding the relation between joint torques during SPBR and strength measurements are novel. Unlike the current study, this earlier study did find a redistribution of extensor torques in older adults. This redistribution consisted of a trend toward increased ankle and hip torques in older adults compared to young adults. One possible reason for not finding this redistribution in the current study is that it included adults 55-65 years old whereas the earlier study included adults over 65 years old. It is notable that similar trends exist in the data (Figure 2) in that older adults showed larger mean values of peak ankle plantar flexor torque and hip extensor torque. The fact that these differences were not significant may be due to the fact that these differences increase as age increases, and they had not yet increased to the extent necessary to make it statistically significant.

One difficulty we experienced in the current study was expressing joint torques during SPBR as a percentage of maximum joint torques collected during strength measurements. Approximately 30% of peak extensor torque values during SPBR exceeded 100% of strength measurements. This should not be the case if the strength measurements accurately reflect the theoretical maximum values for joint torques during SPBR. This is likely due to two issues. First, this approach is dependent upon subjects providing maximal effort during strength testing. Although we provided verbal encouragement during strength testing, subjects may have not provided their maximal effort. Second, the model we used to represent maximum joint torques has limitations. This model is essentially used to interpolate between a relatively small number of experimental strength measurements over a wide range of experimental conditions (varying joint angle and joint angular velocity). It is based on a numerical optimization approach that minimizes the sum-of-the-squared error terms. As such, the numerical optimization often fits the model such that the model predicts a maximum torque that is lower than an experimental strength measurement. As a result, if a subject generates their maximum torque during SPBR, it will appear to be over 100% of the maximum torque. We are investigating more sophisticated optimization techniques to minimize this effect in the future.

Conclusion

In conclusion, peak extensor torques increased as the lean from which subjects were released increased. The peak ankle torque approached its maximum during recovery from the largest achieved lean, and peak ankle torque that was predicted to be required for the next largest lean exceeded 100% of the maximum ankle torque. These results imply that ankle strength limited the largest lean (i.e. balance recovery ability), and suggest that increased strength would increase the largest achieved lean. Further research is needed to determine if these results can be generalized to falls during walking (i.e. trips and slips).

PUBLICATIONS

Madigan ML. Age-related differences in muscle power during single step balance recovery. *Journal of Applied Biomechanics* 22, 2006: 185-92.

This paper describes age-related differences in muscle power (which is related to joint torques) which are investigated in Specific Aims 1 and 3.

Anderson DE, Madigan ML, Nussbaum MA. Maximum voluntary joint torque as a function of joint angle and angular velocity: model development and application to the lower limb *Journal of Biomechanics* 40, 2007: 3105-13.

This paper describes the development of a model of joint strength that was used to express joint torques during balance recovery as a percentage of their maximum. This was used to address hypothesis 2 in Specific Aim 1.

Bieryla KA, Anderson DA, Madigan ML. Estimations of relative effort during sit-to-stand increase when accounting for variations in maximum voluntary torque with joint angle and angular velocity *Journal of Electromyography and Kinesiology* (in press).

This paper demonstrates the application of the model of joint strength on a different task. This model was used to address hypothesis 2 in Specific Aim 1.

We anticipate at least one more publication that focuses on ankle strength limiting balance recovery ability.

INCLUSION OF GENDER AND MINORITY STUDY SUBJECTS

See the Inclusion Enrollment Report attached.

INCLUSION OF CHILDREN

Eleven children, defined by NIH as individuals under the age of 21, were included in this study as a part of the young adult group. No children under the age of 18 were used.

MATERIALS AVAILABLE TO OTHER INVESTIGATORS

A mathematical model of maximum voluntary joint torque was developed as a part of this work and is available to other investigators. It can be accessed via its dissemination in a peer-reviewed article published in the *Journal of Biomechanics*.

Inclusion Enrollment Report

This report format should NOT be used for data collection from study participants.

Study Title: Muscle strength and age effects in balance recovery

Total Enrollment: 34 **Protocol Number:**

Grant Number: 1 R03 OH007821-02

PART A. TOTAL ENROLLMENT REPORT: Number of Subjects Enrolled to Date (Cumulative) by Ethnicity and Race

Ethnic Category	Females	Males	Unknown or Not Reported	Total
Hispanic or Latino	0	1		1 **
Not Hispanic or Latino	17	16		33
Unknown (individuals not reporting ethnicity)	0	0		0
Ethnic Category: Total of All Subjects*	17	17	0	34 *
Racial Categories				
American Indian/Alaska Native	0	0	0	0
Asian	1	2	0	3
Native Hawaiian or Other Pacific Islander	0	0	0	0
Black or African American	3	2	0	5
White	13	13	0	26
More Than One Race	0	0	0	0
Unknown or Not Reported	0	0	0	0
Racial Categories: Total of All Subjects*	17	17	0	34 *

PART B. HISPANIC ENROLLMENT REPORT: Number of Hispanics or Latinos Enrolled to Date (Cumulative)

Racial Categories	Females	Males	Unknown or Not Reported	Total
American Indian or Alaska Native	0	0	0	0
Asian	0	0	0	0
Native Hawaiian or Other Pacific Islander	0	0	0	0
Black or African American	0	0	0	0
White	0	1	0	1
More Than One Race	0	0	0	0
Unknown or Not Reported	0	0	0	0
Racial Categories: Total of Hispanics or Latinos**	0	1	0	1 **

* These totals must agree.

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