The effects of walking speed on obstacle crossing in healthy young and healthy older adults

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Abstract

The effects of walking speed and age on the peak external moments generated about the joints of the trailing limb during stance just prior to stepping over an obstacle and on the kinematics of the trailing limb when crossing the obstacle were investigated in 10 healthy young adults (YA) and 10 healthy older adults (OA). The peak hip and knee adduction moments in OA were 21–43% greater than those in YA ($p < 0.046$). The angular velocity of hip flexion from toe-off to when the toe was over the obstacle in OA was 20% less than that in YA ($p = 0.048$). Eight external peak moments about the hip, knee, or ankle increased significantly with speed in both groups ($p < 0.01$); the largest in the sagittal plane was 62% for knee flexion ($p < 0.0001$). Toe–obstacle clearance was not affected. Trailing foot placement was affected by speed, but not by age. Speed produced a small change ($\approx 1^{\circ}$) in knee abduction–adduction, the only joint angle affected ($p = 0.022$). The greater magnitudes of hip and ankle adduction moments occurring in the trailing limb of OA place larger demands on the hip and ankle abductors of the trailing limb to maintain dynamic balance when stepping over the obstacle with the leading limb.

1. Introduction

Falls represent an important health problem for the elderly, affecting an estimated 30% of community-dwelling persons over the age of 65 each year (Blake et al., 1988; O’Laughlin et al., 1993; Tinetti et al., 1988). In 1977, Overstall reported that 47% of falls in the elderly are caused by tripping over obstacles, and tripping has since been cited as the most frequent cause of falling in the elderly (Blake et al., 1988; Campbell et al., 1990; Overstall et al., 1977; Prudham and Evans, 1981). The danger posed by tripping calls for a better understanding of how lower limb actions change when negotiating obstacles under different conditions.

While investigating the movements of the lower limbs during obstacle crossing, previous researchers have studied both the leading and trailing limbs (Chen et al., 1991; Patla et al., 1996). Our interest lies in the trailing limb because it presents several unique risks for tripping. For instance, during obstacle crossing, neither the trailing limb nor the obstacle is within the subject’s field of vision, increasing the chance that an unexpected foot–obstacle contact may occur. In addition, at toe-off just prior to crossing, the toe of the trailing limb is one step-length closer to the obstacle than the toe of the leading limb. In order to clear the obstacle, the trailing limb must achieve an adequate toe elevation in a shorter period of time for a given crossing speed. Researchers have found that the trailing limb did indeed exhibit higher vertical toe velocities than the leading limb and that toe–obstacle clearance of the trailing limb was consistently lower than that of the leading limb (Patla et al., 1996). The foregoing suggests that the risk of toe–obstacle contact is greater for the trailing limb.

Both faster gait (Kirtley et al., 1985) and obstacle crossing (Chou and Draganich, 1997; Murray et al.,...
1966, 1984) have been shown to increase flexion of the swing limb and the magnitudes of the three-dimensional moments on the stance limb in healthy YA. The increased motions and moments during faster obstacle crossing may require greater balance, muscular strength and range of motion than those encountered during normal walking, especially in OA.

Studies of the ability to regain balance following a release from a forward lean demonstrated decreased abilities of older adults to regain balance that were attributed to their inabilities to move their swing feet forward fast enough (Thelen et al., 1997; Wojcik et al., 1999). This inability to move their swing feet forward fast enough implies a limitation in their ability to flex one or more of their hip, knee or ankle joints rapidly enough, which may result in a reduction in toe–obstacle clearance during faster crossing speeds. Furthermore, Patla et al. (1991) observed that young adults who attempted to step over a suddenly appearing obstacle used two different strategies for increasing ground clearance of the leading limb, including increased angular velocity of the knee or ankle. Thus, differences may exist between the young and old in angular velocity of knee or ankle flexion when stepping over an obstacle at increased speeds. Although studies of obstacle crossing have been performed on subjects walking at self-selected speeds (Chen et al., 1991, 1994; Chou and Draganich, 1997, 1998a, b; Chou et al., 2001; McFadyen et al., 1993; McFadyen and Winter, 1991; Patla et al., 1996, 1991; Patla and Rieddy, 1993; Pavol et al., 2001), only one of these studies found that faster walking speed was associated with falling when trying to recover from a trip (Pavol et al., 2001). Consequently, combining an obstacle with increased walking speed may result in an even greater demand on the lower limb than either factor alone to increase the risk of tripping in OA.

The current study reports on the peak moments generated about the joints of the trailing limb during stance just prior to stepping over the obstacle with the trailing limb and on the kinematics of the trailing limb when crossing the obstacle. The hypotheses tested were that increasing walking speed would increase trailing limb toe–obstacle clearance, increase flexion of the hip, knee and ankle when the trailing toe is over the obstacle, increase the three-dimensional moments about the hip, knee and ankle, and increase the angular velocities of hip, knee and ankle flexion during stance of the trailing limb in healthy YA and healthy OA, and that the magnitudes of these variables in OA would be different than those in YA.

2. Methods

Gait analysis was performed on 10 healthy YA (five females, five males) having a mean age of 25.9 yr (23–34 yr), an average height of 172.3 cm (165.1–187.9 cm) and a mean weight of 712.5 N (587.0–919.2 N), and 10 healthy OA (three females, seven males) having a mean age of 71.6 yr (65–84 yr), height 171.4 cm (157.5–182.9 cm) and weight 759.5 N (624.2–969.2 N). Medical histories were obtained and older subjects had a physical (including neurologic) examination to exclude the presence of poor uncorrected vision or chronic conditions that might affect gait, including a history of heart disease, hip fracture, active cancer, or stroke. The Institutional Review Board of The University of Chicago approved the protocol for the study.

Ground reaction forces were measured with a multi-component force platform (Advanced Mechanical Technology Inc., Newton, MA) in the center of walkway. Clusters of six or eight infrared light-emitting diodes (IREDs) were strapped to the foot, shank and thigh of each subject. Kinematic parameters were collected with the Optotrak (Northern Digital Inc., Waterloo, Ontario, Canada) optoelectronic, three-dimensional digitizing system. Kinematic and force parameters were synchronized and sampled at a rate of 100 Hz. The overall accuracy of the system was better than 0.5 mm.

DAP software (Northern Digital Inc., Waterloo, Ontario, Canada) was used to determine the spatial orientation of each rigid segment in the inertial coordinate system. Euler angles were used to compute the three-dimensional angles. The external three-dimensional moments about the hip, knee, and ankle of the trailing limb during stance of the crossing stride (from heel contact just before the obstacle to toe-off just before the obstacle) were computed using inverse dynamics. The joint moments were normalized to body weight times lower limb length. Custom software, developed in our laboratory, was used to identify and extract the data of interest. The kinetics of the hip, knee and ankle were analyzed at the instant the toe was over the obstacle. Angular velocities of hip, knee and ankle flexion from toe-off to when the toe was over the obstacle were computed by dividing the angular range traversed by the time required.

Subjects all wore their own low-heeled shoes. Subjects were naive to the specific aims of the study. The obstacle was a white wooden dowel 94 cm long and 0.5 cm in diameter, held 20 cm high by grooves in the two vertical arms of an aluminum frame. Each subject first crossed the obstacle at a “normal speed” (0.95–1.10 m/s). The subjects were then asked to cross the obstacle at “slow speed” (<0.85 m/s) and then at “fast speed” (>1.20 m/s) in a random order. Trials outside of these ranges were repeated. The subject had several practice trials on the walkway without the obstacle present to ensure that the subject was able to walk consistently at the speed to be tested. We then measured the subject’s step length. In a prior study, it was discovered that
subjects stepped over a 20 cm high obstacle with a mean toe–obstacle distance of 42–44% of step length found for unobstructed level walking (Chou and Draganich, 1998a). Thus, the obstacle was placed at a distance of 43% of step length from the center of the forceplate. Next, the subject’s starting position was adjusted near one end of the walkway. Subjects were asked to walk along the 9.5 m walkway, step over the obstacle with their right limbs first and their left limbs (trailing limbs) second and continue walking to the end of the walkway while maintaining the same speed. With repeated trials, we discovered a starting position that allowed the subject to step onto the force platform and over the obstacle consistently, in a self selected manner and without the subject having to make adjustments in his/her stride. This procedure was repeated for each of the three speeds.

SPSS (SPSS Science, Chicago, IL) was used to perform the statistical analysis. The mean of three trials for each walking speed was used in formulating the results. Two-way analysis of variance with repeated measures was used to determine the effects of walking speed and age on toe–obstacle clearance and toe–obstacle distance, the three-dimensional joint angles and moments of the trailing limb, crossing speed and the angular velocities of flexion of the joints of the trailing limb. The Greenhouse-Geiser adjustment to the degrees of freedom was used in assessing significance levels. Values of \( p \leq 0.05 \) were considered significant.

3. Results

Four peak out-of-plane moments were significantly affected by age (Table 1, Fig. 1). In early stance, these were the maximum hip and knee adduction moments. In late stance, these were the maximum hip and ankle adduction moments. For all of these parameters, the magnitudes of the moments in the OA were from 21% (peak hip adduction in early stance) to 43% (peak ankle adduction in late stance) greater than those in the YA. In addition, there was a significant interaction between speed and age in the maximum hip internal rotation moment (\( p = 0.024 \)).

Three peak out-of-plane moments increased significantly with increased speed (Table 1, Fig. 2). In early

Table 1

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Group</th>
<th>Slow gait speed</th>
<th>Normal gait speed</th>
<th>Fast gait speed</th>
<th>% Change from slow to fast ( p )-value for speed</th>
<th>Age ( p )-value for age</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum hip adduction moment in early stance</td>
<td>YA</td>
<td>8.57 ± 2.76</td>
<td>9.06 ± 3.13</td>
<td>10.28 ± 3.41</td>
<td>16.6( % ) 0.271</td>
<td>9.30 ± 3.09 0.004</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>12.09 ± 4.02</td>
<td>10.75 ± 2.73</td>
<td>11.03 ± 3.01</td>
<td>8.8( % ) 0.439</td>
<td>11.29 ± 3.24</td>
</tr>
<tr>
<td>Maximum hip adduction moment in late stance</td>
<td>YA</td>
<td>8.62 ± 2.79</td>
<td>8.59 ± 2.80</td>
<td>8.27 ± 2.93</td>
<td>−4.1( % ) 0.439</td>
<td>8.49 ± 2.76 0.013</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>10.82 ± 6.70</td>
<td>11.55 ± 5.56</td>
<td>8.97 ± 4.84</td>
<td>−17.1( % ) 0.439</td>
<td>10.45 ± 5.66</td>
</tr>
<tr>
<td>Maximum hip external rotation moment in early stance</td>
<td>YA</td>
<td>0.39 ± 0.40</td>
<td>1.32 ± 0.46</td>
<td>1.83 ± 0.74</td>
<td>105.6( % ) 0.003</td>
<td>1.35 ± 0.65 0.778</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>1.29 ± 1.00</td>
<td>1.36 ± 0.67</td>
<td>1.64 ± 0.70</td>
<td>27.1( % ) 0.010</td>
<td>1.43 ± 0.79</td>
</tr>
<tr>
<td>Maximum hip internal rotation moment in late stance</td>
<td>YA</td>
<td>0.81 ± 0.40</td>
<td>0.92 ± 0.40</td>
<td>1.31 ± 0.48</td>
<td>61.7( % ) 0.010</td>
<td>1.01 ± 0.47 0.831</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>0.92 ± 0.40</td>
<td>1.01 ± 0.40</td>
<td>1.16 ± 0.51</td>
<td>26.1( % ) 0.013</td>
<td>1.03 ± 0.43</td>
</tr>
<tr>
<td>Maximum knee adduction moment in early stance</td>
<td>YA</td>
<td>4.79 ± 1.55</td>
<td>5.19 ± 1.74</td>
<td>6.46 ± 1.76</td>
<td>34.9( % ) 0.392</td>
<td>5.48 ± 1.78 0.046</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>6.04 ± 1.56</td>
<td>6.01 ± 1.94</td>
<td>6.97 ± 2.37</td>
<td>15.4( % ) 0.392</td>
<td>7.04 ± 3.10</td>
</tr>
<tr>
<td>Maximum knee internal rotation moment in late stance</td>
<td>YA</td>
<td>1.84 ± 0.79</td>
<td>2.08 ± 0.89</td>
<td>2.30 ± 0.87</td>
<td>11.5( % ) 0.004</td>
<td>2.07 ± 0.85 0.513</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>2.29 ± 0.64</td>
<td>2.42 ± 0.82</td>
<td>2.42 ± 0.79</td>
<td>5.7( % ) 0.004</td>
<td>2.37 ± 0.73</td>
</tr>
<tr>
<td>Maximum ankle adduction moment in late stance</td>
<td>YA</td>
<td>0.39 ± 0.51</td>
<td>1.01 ± 0.54</td>
<td>1.25 ± 0.51</td>
<td>40.4( % ) 0.634</td>
<td>1.05 ± 0.52 0.040</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>1.79 ± 0.99</td>
<td>1.40 ± 0.84</td>
<td>1.31 ± 0.60</td>
<td>26.8( % ) 0.634</td>
<td>1.50 ± 0.82</td>
</tr>
</tbody>
</table>

\( a \) In this and the following tables, data are given as mean ± 1 sd. \( p \)-values are for two-way ANOVA.

\( b \) In this and the following tables, positive values indicate increased speed and negative values indicate decreased speed.

\( c \) In this and the following tables, values were computed using the data for all three speeds.
stance, the maximum hip external rotation moment increased 27.1% in OA and 105.1% in YA. In late stance, the maximum knee internal rotation and hip internal rotation moments increased, with the corresponding increases ranging from 5.7% to 61.7%.

Five in-plane moment peaks increased significantly with increased speed (Table 2, Fig. 3). In early stance, these were the maximum hip flexion, knee flexion and ankle plantarflexion moments, with the increases ranging from 48% for ankle plantarflexion to 62.3% for knee flexion, which was the largest moment generated in the sagittal plane. In late stance, these were the maximum hip extension and ankle dorsiflexion moments, with the increases ranging from 10.2% for ankle dorsiflexion to 35.1% for hip extension. All of the increases for a given parameter were similar for the YA and OA. Age did not significantly affect these measures.

The angular velocities of hip, knee and ankle flexion increased significantly with increased speed (Table 3). The increases ranged from 32.1% for the knee to 67.9% for the ankle. Age significantly affected angular velocity of hip flexion; the velocity for OA was 20% less than that of YA. In addition, age had a marginally significant effect on angular velocity of knee flexion ($p = 0.076$).

When the toe of the trailing limb was over the obstacle, the knee adduction angle increased significantly with walking speed. However, the magnitude of
the change was small (~1°). Walking speed was not found to significantly affect any other of the three-dimension al hip, knee or ankle joint angles in YA or in OA.

Although the clearances in OA were 2–4 cm greater than those in YA, vertical toe-obstacle clearance of the trailing foot was not found to significantly change with obstacle speed or age (Table 4). Similarly, when the clearances were normalized to lower limb length for each subject, clearance was not found to significantly change with obstacle speed or age. At toe-off, the mean horizontal toe-obstacle distances significantly increased with increased walking speed for both groups, but they were not affected by age. In both YA and OA, increased walking speed resulted in increased crossing stride velocity, demonstrating that the crossing speeds were significantly different.

The subjects crossed the obstacle with the trailing limb at 75.7% ± 2.1% of the gait cycle, where heel strike of the trailing limb marked the beginning of the gait cycle.

### 4. Discussion

In this study, we wanted to elucidate the effects walking speed and age would have on the three-dimensional angles and moments of the trailing limb of healthy YA and healthy OA when crossing an obstacle. Various studies have demonstrated that comfortable walking speeds decrease with age (Bohannon, 1997). In studying the differences in gait between YA and OA, earlier studies have shown that many, but not all, of the significant kinetic and kinematic differences between groups when walking at self-selected “comfortable” speeds corrected or disappeared when the OA walked slightly faster, thus matching YA gait speed (Kerrigan et al., 1998; Riley, 2001). In the current study, subjects were asked to walk within prescribed ranges of speed in order to eliminate differences between groups that could be due to speed alone, allowing us to determine the effects of age alone.

Studies have shown that muscle strength begins to diminish from the fifth decade until the ninth decade at a rate of approximately 12–15% per decade, and over the age of 60 years, muscle mass reduces by 1–5% per year (Hurley, 1995; Wolfson et al., 1995). In the current study, the largest in-plane moment increase from slow to fast speeds in OA was 62% for knee flexion. Furthermore, the magnitudes of external knee flexion moments generated when crossing a 20 cm high obstacle were 49% greater than those generated during unobstructed gait (Chou and Draganich, 1997, 1998). This greater demand on the knee extensors may be particularly important when considering that the body’s center of mass is in front of the base of support (i.e., in front of the trailing limb stance foot) when stepping over an obstacle and that a cross-sectional study found knee extension strength to be a significant determinant of performance on dynamic balance tests in 65- to 75-year-old women (Carter et al., 2002). Thus, without adequate

### Table 2

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Group</th>
<th>Slow gait speed</th>
<th>Normal gait speed</th>
<th>Fast gait speed</th>
<th>% Change from slow to fast</th>
<th>p-value for speed</th>
<th>Age</th>
<th>p-value for age</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum hip flexion moment in early stance</td>
<td>YA</td>
<td>4.69 ± 1.62</td>
<td>6.49 ± 1.78</td>
<td>10.05 ± 2.81</td>
<td>53.3</td>
<td>&lt; 0.0001</td>
<td>6.46 ± 3.13</td>
<td>0.732</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>5.47 ± 1.14</td>
<td>8.12 ± 2.76</td>
<td>11.72 ± 2.82</td>
<td>53.3</td>
<td>6.04 ± 2.88</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Maximum hip extension moment in late stance</td>
<td>YA</td>
<td>5.15 ± 2.64</td>
<td>6.19 ± 2.92</td>
<td>8.03 ± 3.33</td>
<td>35.1</td>
<td>&lt; 0.0001</td>
<td>7.08 ± 3.06</td>
<td>0.122</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>4.82 ± 2.91</td>
<td>5.74 ± 2.48</td>
<td>7.57 ± 2.78</td>
<td>31.0</td>
<td>8.44 ± 3.47</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Maximum knee flexion moment in early stance</td>
<td>YA</td>
<td>3.99 ± 2.02</td>
<td>5.79 ± 2.77</td>
<td>8.88 ± 3.58</td>
<td>55.1</td>
<td>&lt; 0.0001</td>
<td>6.22 ± 3.45</td>
<td>0.143</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>2.71 ± 2.18</td>
<td>4.22 ± 2.26</td>
<td>7.19 ± 2.14</td>
<td>62.3</td>
<td>4.71 ± 2.84</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Maximum ankle plantarflexion moment in early stance</td>
<td>YA</td>
<td>2.05 ± 0.67</td>
<td>2.89 ± 0.74</td>
<td>3.91 ± 1.05</td>
<td>48.0</td>
<td>&lt; 0.0001</td>
<td>2.95 ± 1.12</td>
<td>0.660</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>2.13 ± 0.69</td>
<td>2.86 ± 0.88</td>
<td>4.24 ± 1.25</td>
<td>50.0</td>
<td>3.13 ± 1.25</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Maximum ankle dorsiflexion moment in late stance</td>
<td>YA</td>
<td>16.20 ± 1.58</td>
<td>17.45 ± 1.83</td>
<td>18.33 ± 1.90</td>
<td>11.6</td>
<td>&lt; 0.0001</td>
<td>17.32 ± 1.93</td>
<td>0.607</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>16.00 ± 1.40</td>
<td>17.22 ± 1.61</td>
<td>17.81 ± 1.61</td>
<td>10.2</td>
<td>17.01 ± 1.68</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
strength, the trailing limb is much more likely to come in contact with the obstacle at increased walking speeds to increase the risk of tripping in frail OA.

The incidence of hip fractures in the elderly is increasing, with 650,000 fractures expected in the United States by the year 2050 (Ellis, 2003). Furthermore, falling to the side not only greatly increases the risk of hip fracture (Hayes et al., 1993; Hayes and Myers, 1997), but is also an independent predictor of hip fracture (Greenspan, 1998; Wei et al., 2001). During single limb support in gait, the body’s center of mass trajectory is medial, towards the centerline of the plane of progression of gait (MacKinnon and Winter, 1993; Lyon and Day, 1997). Thus, the body is unstable as it tends to fall sideways under gravity. Similarly, this instability occurs when stepping over an obstacle, although the duration of this instability is likely longer since stride time is increased as compared with that for unobstructed gait (Chou et al., 2001). Balance in the frontal plane during gait is controlled by the precise interaction between active muscle moments, passive joint acceleration moments and destabilizing gravitational moments about the ankle and hip to control the mediolateral trajectory of the body’s center of mass (MacKinnon and Winter, 1993). In the current study, age was found to significantly affect the peak hip and ankle adduction moments during stance. In OA these were 21–43% greater, respectively, than those of the YA. The greater magnitudes of hip and ankle adduction moments occurring in the trailing limb of OA place larger demands on the hip and ankle abductors of the trailing limb to maintain dynamic balance when stepping over the obstacle with the leading limb.

Furthermore, hip and ankle adduction moments generated during obstacle crossing were found to be 27–77% greater, respectively, than those generated during unobstructed gait (Chou and Draganich, 1997, 1998a, b). In addition, as noted above, muscle strength diminishes with age. Also, in a review of all pertinent literature sources published in the English language between 1966 and 2002 concerning hip fracture epidemiology, hip fracture injury mechanisms, and hip fracture management strategies, muscle weakness was one of three factors most likely explaining the rising incidence of hip fracture injuries (Marks et al., 2003).

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**Table 3**

Angular velocities (deg/s) of flexion for the hip, knee and ankle

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Group</th>
<th>Slow gait speed</th>
<th>Normal gait speed</th>
<th>Fast gait speed</th>
<th>% Change from slow to fast</th>
<th>p-value for speed</th>
<th>Age</th>
<th>p-value for age</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td>YA</td>
<td>127.96±34.07</td>
<td>146.84±27.18</td>
<td>185.83±29.80</td>
<td>45.2</td>
<td>≤0.0001</td>
<td>153.54±38.29</td>
<td>122.85±47.61</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>104.35±32.19</td>
<td>114.47±44.40</td>
<td>149.72±55.09</td>
<td>43.5</td>
<td>≤0.0001</td>
<td>153.54±38.29</td>
<td>122.85±47.61</td>
</tr>
<tr>
<td>Knee</td>
<td>YA</td>
<td>298.08±57.56</td>
<td>333.29±55.60</td>
<td>393.65±59.48</td>
<td>32.1</td>
<td>≤0.0001</td>
<td>341.68±68.53</td>
<td>295.12±69.13</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>253.11±39.62</td>
<td>292.61±63.20</td>
<td>339.64±75.26</td>
<td>34.2</td>
<td>≤0.0001</td>
<td>295.12±69.13</td>
<td>295.12±69.13</td>
</tr>
<tr>
<td>Ankle</td>
<td>YA</td>
<td>52.47±19.47</td>
<td>56.78±21.21</td>
<td>88.11±37.96</td>
<td>67.9</td>
<td>0.005</td>
<td>65.78±30.99</td>
<td>55.68±50.33</td>
</tr>
<tr>
<td></td>
<td>OA</td>
<td>42.56±51.65</td>
<td>59.33±47.65</td>
<td>65.13±54.02</td>
<td>65.1</td>
<td>0.005</td>
<td>55.68±50.33</td>
<td>55.68±50.33</td>
</tr>
</tbody>
</table>

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*aFrom plantar- to dorsi-flexion.*

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![Graph](image-url)
These increased demands on the hip and ankle abductors during obstacle crossing at increased speeds are expected to increase the risk of falling to the side in frail OA.

All of the angular velocities of joint flexion increased with crossing speed. Although the angular velocities of joint flexion for the OA were less than those for the YA, only angular velocity of hip flexion was significantly affected by age. Hip angular velocity for the OA was 20% less than that for the YA. However, there was also a trend towards significance for the effects of age on angular velocity of the knee (p = 0.076). Thus, the decreased abilities of OA to regain balance following a release from a forward lean, which were attributed to their inabilities to move their swing feet forward fast enough (Thelen et al., 1997; Wojcik et al., 1999), may in turn be due to the inability to flex the hip and possibly also the knee fast enough.

Although the OA tended to clear the obstacle by 3–4 cm more than the YA for each gait speed and the clearances for the fast and slow speeds were less than that for the normal speed, neither speed nor age were found to significantly affect clearance. These results in light of prior studies noting no change in clearance with obstacle height within YA (Chou and Draganich, 1997) or between YA and OA (Chen et al., 1991) for self-selected walking speeds suggest that obstacle crossing with the trailing limb operates under a preset locomotor pattern that calibrates the necessary flexion of the hip, knee and ankle with the end goal of adequate clearance. Patla et al. (1996) have suggested that the locomotor pattern for crossing an obstacle with the trailing limb is different from that of the leading limb. The distinct locomotor pattern for the trailing limb may be especially important because both the obstacle and limb are behind the subject’s field of view during obstacle crossing, so subjects are unable to utilize visual inputs during this time to help prevent toe–obstacle contact.

On the other hand, the reductions in toe–obstacle clearance in the OA group are consistent with the reduced angular velocities of joint flexion observed in the OA group. Thus, it is possible that the reduced toe–obstacle clearances in the OA group would have become significant had a larger population of OA and YA been tested.

A limitation of this study is that obstacle height was not scaled to body height or lower limb length. Thus, it is possible that any subtle effects of age or speed on clearance were missed. However, this appears unlikely since neither speed nor age were found to significantly affect clearance when clearance was normalized to lower limb length for each subject. Similarly, Chen et al. (1991) found no effects of age on clearance in 24 young and 24 old adults when clearance was normalized by anthropometric variables, including stature.

When the toe was over the obstacle, only one joint angle, knee adduction–abduction angle, was affected significantly by speed in young adults. However, the change was small, approximately 1°, and therefore is likely unimportant. Age did not affect any of the joint angles. Thus, the idea that range of motion in the lower extremity does not have a profound effect on gait in OA (Kerrigan et al., 1998) can be extended to crossing an obstacle of 20 cm height over the range of speeds tested in this study.
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References


