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Effect of contralateral cane use on hip moment impulse in the frontal plane during the stance phase

Takuma Inai*, Tomoya Takabayashi, Mutsuaki Edama, Masayoshi Kubo

Institute for Human Movement and Medical Sciences, Niigata University of Health and Welfare, 1398 Shimami-cho, Kita-ku, Niigata City, Niigata, 950-3198, Japan

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ABSTRACT

Background: Recent reports have shown that the daily cumulative moment in the frontal plane (i.e., product of hip moment impulse in the frontal plane during the stance phase and mean steps per day) is a risk factor for hip osteoarthritis. This study aimed to clarify the effect of contralateral cane use on hip moment impulse in the frontal plane of the stance limb.

Methods: This study included 15 healthy subjects who walked under four experimental conditions: (1) without a cane and (2–4) contralateral cane use with 10%, 15%, and 20% body weight support (BWS), respectively. To maintain the same walking speed in all conditions, the cadence was set to 80 steps/min, and the step length was fixed. The hip moment impulses in the frontal plane (i.e., area under the hip ab-adduction moment waveform) and peak hip adduction moments in all conditions were calculated.

Results: Contralateral cane use significantly decreased the hip moment impulse in the frontal plane and peak hip adduction moment compared to non-cane use. Moreover, the hip moment impulse in the frontal plane and peak hip adduction moment decreased significantly with increased cane BWS. There were no significant differences in walking speed, cadence, and step length between the four conditions.

Conclusion: Contralateral cane use decreases the hip moment impulse in the frontal plane and peak hip adduction moment in the stance limb. These findings may help clarify how to delay the progression of hip osteoarthritis.

1. Introduction

Hip osteoarthritis is a representative disease of all orthopedic diseases. Reportedly, hip osteoarthritis causes hip joint pain [1], with decreased muscle strength [2], range of motion [3], and functional activity (e.g., gait and stair climbing [1]). Moreover, a previous study [4] reported poor health-related quality of life in patients with hip osteoarthritis. Thus, it is important to prevent and delay the progression of hip osteoarthritis.

Although several previous studies have reported biomechanical risk factors of knee osteoarthritis (e.g., peak knee adduction moment [5,6] and knee adduction moment impulse [5]), no biomechanical risk factor for hip osteoarthritis has been found until recently. Tateuchi et al. [7] found a biomechanical risk factor for hip osteoarthritis called "daily cumulative hip moment," and this index is the product of hip moment impulse during the stance phase and mean steps per day [7]. They found that higher daily cumulative hip moment, particularly in the frontal plane, was a predictor of the radiographic progression of hip osteoarthritis over 12 months. Thus, this finding suggests that decreases in hip moment impulse in the frontal plane (i.e., area under the hip ab-adduction moment waveform) and mean steps per day (only heavy physical activity [8]) may delay the progression of hip osteoarthritis.

Contralateral cane use for patients with hip osteoarthritis has been recommended by Osteoarthritis Research Society International [9]. However, the level of evidence is IV (expert consensus) because of insufficient rationale. Previous studies on gait with a cane reported that contralateral cane use generates external hip abduction moment and decreases hip abductor muscle activity in the stance limb [10,11]. Therefore, contralateral cane use may change hip moment impulse in the frontal plane during the stance phase. Other studies have reported the effects of contralateral cane use on peak hip adduction moment [12,13], peak knee adduction moment [14,15], and knee adduction

* Corresponding author.
E-mail address: hwd17001@nuhw.ac.jp (T. Inai).

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moment impulse [14]. However, the effect of contralateral cane use on hip moment impulse in the frontal plane during the stance phase is unknown [16]. If contralateral cane use decreases hip moment impulse in the frontal plane, this knowledge may enhance our of understanding gait patterns with low hip moment impulse in the frontal plane and may provide a rationale of delaying the progression of hip osteoarthritis.

This study aimed to examine the effect of contralateral cane use on hip moment impulse in the frontal plane during the stance phase. We hypothesized that contralateral cane use would decrease hip moment impulse in the frontal plane during walking.

2. Methods

2.1. Participants

Fifteen subjects [five men and ten women, mean (SD) age: 19.5 (1.1) years, height: 1.62 (0.07) m, and body mass: 53.6 (4.8) kg] participated in this study. Inclusion criteria for participation were the absence of (1) body pain, (2) orthopedic disease (e.g., anterior cruciate ligament injury), and (3) neurological disorders. The study was approved by the Niigata University of Health and Welfare ethics committee. Informed consent was obtained from all participants prior to participation.

2.2. Experiment

Four experimental conditions were set. The first was to walk without a cane. The second, third, and fourth conditions were to walk with a cane on the right hand at 10%, 15%, and 20% body weight support (BWS), respectively [14]. More than 10 trials were conducted for each condition.

Previous studies have reported a significant decrease in walking speed [12,17,18] and cadence [12,17–19] with contralateral cane use compared with cane non-use. Moreover, a change in walking speed affects the hip ab-adduction moment waveform [20,21]. Therefore, the cadence was set to 80 steps/min based on a previous study [18], using a metronome in the present study. The target tape was placed on the force place (Fig. 1) (i.e., the location of the target tape was the same for all subjects). All subjects were instructed to walk naturally with the 80 steps/min cadence from the temporary start tape as a practice trial. The experimenter confirmed the subject's natural step length with the 80 steps/min and replaced the temporary start tape for landing the second step of the left foot on the target tape (i.e., the start tape locations were different for each subject). The start and target tapes were used to maintain the same walking speed for each subject in all conditions. The hip moment impulse in the frontal plane of the left limb was analyzed, and all subjects held the cane in their right hand. After each condition, the subjects were asked to identify which body parts were uncomfortable and to evaluate the discomfort level on a numerical rating scale (0–10 points).

The cane height was set to the distal wrist crease from the floor [22]. Subjects were instructed regarding the two-point gait pattern and to place the cane on the ground simultaneously, or just preceding, heel contact of the stance limb [14]. Moreover, subjects were to place the cane in line with the stance limb and at a lateral distance roughly equal to shoulder width [14]. Before each condition with the cane, subjects practiced putting the cane with the specified target BWS using a real-time visual monitor (Fig. 1).

A three-dimensional motion capture system (Vicon, Oxford, UK) with 12 cameras was used to capture marker trajectories. To obtain pressure centers and ground reaction forces of the stance limb and cane, four force plates were used. The sampling rates of the motion capture system and force plates were 100 Hz and 1000 Hz, respectively. Fourteen reflective markers were attached to the subject's body as follows: both acromia, midpoint of the posterior superior iliac spine, both anterior superior iliac spines, lateral and medial epicondyles of the left femur, lateral and medial malleoli of the left tibia, both first metatarsal heads, left fifth metatarsal head, and both heels.

2.3. Gait analysis

In the first condition, two successful trials closest to the cadence of 80 steps/min were used for analysis. In the three cane conditions, when the peak cane vertical force was not within ± 2.5% of the target BWS (i.e., 7.5–12.5%BWS, 12.5–17.5%BWS, and 17.5–22.5%BWS, respectively), the trial was excluded. Then, two successful trials that were closest to the target cane BWS for each condition were obtained for analysis.

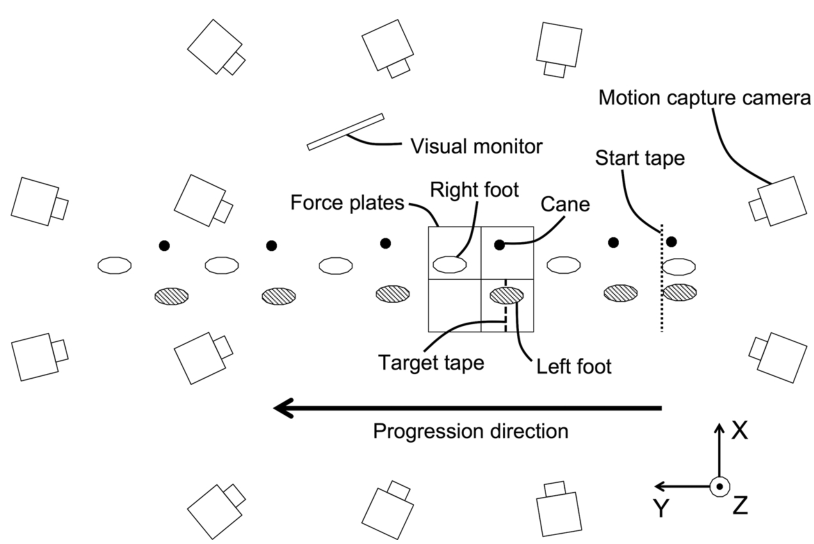

Fig. 1. Study schematic. The start tape was placed on the ground for each subject. The target tape was placed at the point where the second step of the left foot naturally landed.

All marker trajectories, ground reaction forces, and pressure centers were filtered using a fourth-order zero-lag Butterworth low-pass filter with a 6-Hz cut-off frequency. Next, hip joint moments during the stance phase of the left limb for all subjects were calculated using inverse dynamics. Subsequently, the hip joint moment impulse in the frontal planes was calculated by integration of the hip ab-adduction moment. The hip joint moment impulse in the frontal plane during the stance phase (i.e., from left heel contact to left toe-off) was analyzed. First peak hip adduction moment was also calculated. Hip moment impulse in the frontal plane (Nm·s) and peak hip adduction moment (Nm) were normalized with body weight (Nm·s/BW and Nm/BW, respectively). Masses, mass positions, and inertia parameters of segments reported in previous studies were used [23,24].

We calculated gait characteristics (stance time, walking speed [20,21], cadence [21], step length {instead of stride length [21]}, and step width [25]), kinematics (peak contralateral and ipsilateral trunk lean angles [26]), and cane lateral position [14] because these variables could affect hip moment impulse in the frontal plane during the stance phase. The stance time was defined as the time from left heel contact to left heel toe-off. The walking speed during the stance phase was calculated using the reflective marker at the midpoint of the posterior superior iliac spines and was normalized with height [m/(s·HT)]. The actual cadence (steps/min) was calculated using the time from left heel contact to right heel contact. Step lengths at the left and right heel contacts were determined as distances between the left heel to the right heel, and the step lengths were averaged. The step widths at the left and right heel contacts were determined as distances between the left and right first metatarsal heads, and the step widths were averaged. The averaged step length and step width were normalized with height (m/HT), respectively. The peak contralateral and ipsilateral trunk lean angles (°) were calculated using reflective markers at the midpoint of the posterior superior iliac spines and both acromia. The cane lateral position was calculated as the distance from the left ankle joint center to the cane's pressure center at 50% of the stance phase [14] and was normalized by height (m/HT). Moreover, the peak vertical cane force and cane impulse were calculated to confirm whether the target magnitude for each cane BWS was met. The peak vertical cane force (N) and cane impulse (N·s) were normalized with body weight (N/BW and N·s/BW, respectively). Gait analyses were conducted using MATLAB (MathWorks, USA) and Scilab (Scilab Enterprises, France).

2.4. Statistical process

We compared gait characteristics (stance time, walking speed, cadence, step length, step width), kinetics (hip moment impulse in the frontal plane and peak hip adduction moment), kinematics (peak contralateral and ipsilateral trunk lean angles), and cane characteristics (peak cane vertical force, cane impulse, and cane lateral position) between the four conditions. Normality of data was evaluated using the Shapiro-Wilk test. The Bonferroni method for multiple comparisons was used for normally distributed variables. The Wilcoxon signed-rank test with Bonferroni correction was used for non-normally distributed variables. The significance level was set at $p < 0.05$. Statistical analysis was performed using the R language (R Development Core Team).

3. Results

Fig. 2a shows the hip moment impulses in the frontal plane during the stance phase. The hip moment impulse in the frontal plane decreased significantly with an increase in cane BWS. The mean (SD) hip moment impulse in the frontal plane in unaided, 10%BWS, 15%BWS, and 20%BWS was 0.057 (0.005), 0.042 (0.005), 0.034 (0.005), and 0.027 (0.006) Nm·s/BW, respectively. Fig. 2b shows the peak hip adduction moments, which also decreased significantly with an increase in cane BWS. The mean (SD) peak hip adduction moment in unaided, 10%BWS, 15%BWS, and 20%BWS was 0.091 (0.008), 0.072 (0.008), 0.061 (0.009), and 0.052 (0.012) Nm/BW, respectively.

Fig. 3a shows the hip ab-adduction moment waveforms during the stance phase in the four conditions. An increase in cane BWS caused a decrease in peak hip adduction moment. Fig. 3b shows the vertical cane force in 10%BWS, 15%BWS, and 20%BWS conditions. The peak vertical cane forces in the waveforms of the three conditions almost reached their respective target magnitudes for each cane BWS.

Table 1 shows the gait characteristics, kinematics, and cane characteristics in the four conditions. There were no significant differences in stance time, walking speed, cadence, step length, step width, and lateral cane position between the four conditions ($p > 0.05$). There were significant differences in the peak contralateral trunk lean angle between unaided and 15%BWS ($p = 0.013$), unaided and 20%BWS ($p = 0.011$), and 10%BWS and 20%BWS ($p = 0.046$). There was a significant difference in the peak ipsilateral trunk lean angle between unaided and 20%BWS ($p = 0.019$). There were significant differences in the peak vertical cane force and cane impulse between the three cane conditions ($p < 0.001$).

Table 2 shows the numerical rating scale of discomfort during walking with contralateral cane use. In the 10%BWS, 15%BWS, and 20%BWS conditions, 1, 8, and 12 subjects, respectively, felt discomfort. Body parts with discomfort were in the right upper extremity (i.e., shoulder, upper arm, forearm, wrist, and finger).

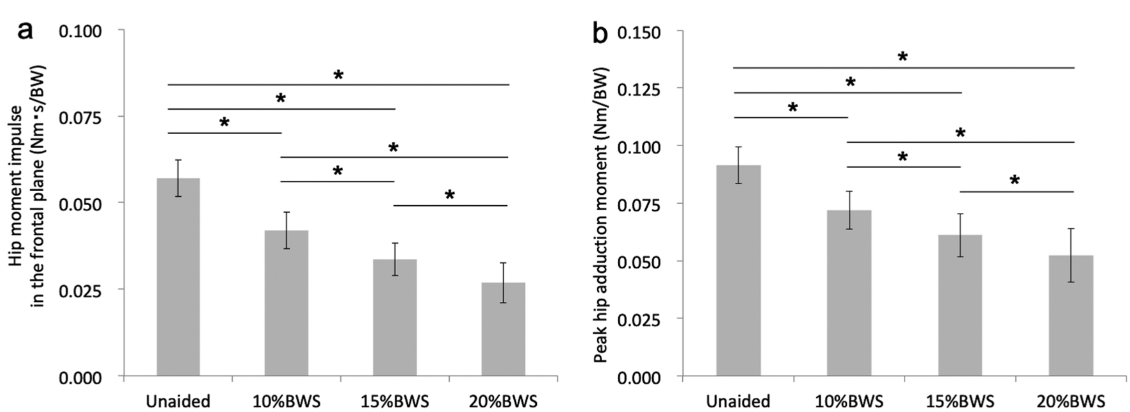

Fig. 2. (a) Hip moment impulse in the frontal plane and (b) peak hip adduction moment in the four conditions. The hip moment impulse in the frontal plane and peak hip adduction moment decreased significantly with an increase in cane body weight support (BWS).
* $p < 0.01$.

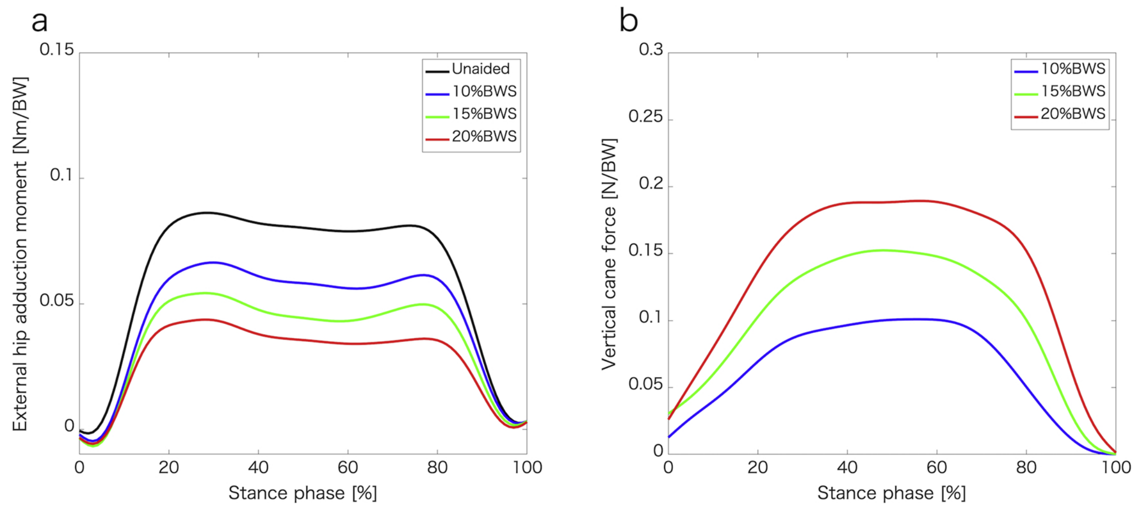

Fig. 3. (a) Hip ab-adduction moment and (b) vertical cane force during the stance phase.
(a) The bimodal waveform and (b) peak vertical cane forces in the three cane condition waveforms almost reached their respective target cane BWS magnitudes.

Table 1
Gait parameters with and without contralateral cane use.

	Unaided (mean (SD))		Attempted 10%BWS		Attempted 15%BWS		Attempted 20%BWS	
Gait characteristics								
Stance time, s	0.91	(0.04)	0.90	(0.05)	0.89	(0.04)	0.90	(0.04)
Walking speed, m/(s·HT)	0.50	(0.04)	0.50	(0.03)	0.50	(0.04)	0.50	(0.04)
Cadence, steps/min	82.4	(2.6)	82.6	(4.1)	81.8	(3.0)	80.3	(3.3)
Step length, m/HT	0.34	(0.02)	0.33	(0.02)	0.33	(0.02)	0.33	(0.02)
Step width, m/HT	0.06	(0.02)	0.05	(0.01)	0.05	(0.02)	0.06	(0.01)
Kinematics								
Peak contralateral trunk lean angle, °	0.4	(1.6)	1.3	(1.7)	1.7	(1.6)*	2.0	(1.7)* †
Peak ipsilateral trunk lean angle, °	2.9	(1.6)	2.1	(1.6)	1.9	(1.7)	1.5	(1.3)*
Cane								
Peak vertical force, N/BW	N/A		0.11	(0.01)	0.16	(0.01)†	0.21	(0.01)† ‡
Impulse, N·s/BW	N/A		0.06	(0.01)	0.09	(0.01)†	0.12	(0.01)† ‡
Lateral position, m/HT	N/A		0.19	(0.02)	0.19	(0.02)	0.19	(0.03)

HT: Height. BW: Body weight. N/A: Not applicable.
* Significantly difference when compared with unaided.
† Significantly difference when compared with 10%BWS.
‡ Significantly difference when compared with 15%BWS.

Table 2
Numerical rating scale of discomfort during walking with contralateral cane use.

Subject	Contralateral cane use		
	10%BWS	15%BWS	20%BWS
1	0	0	0
2	0	2 (right upper arm)	6 (right upper arm) 3 (right forearm)
3	0	0	2 (right wrist)
4	2 (right wrist)	3 (right wrist)	3 (right wrist) 1 (right finger)
5	0	0	0
6	0	0	2 (right wrist)
7	0	1 (right upper arm)	4 (right upper arm)
8	0	5 (right elbow)	6 (right upper arm) 5 (right elbow)
9	0	0	3 (right upper arm)
10	0	2 (right wrist)	2 (right wrist)
11	0	0	0
12	0	2 (right finger)	2 (right upper arm) 2 (right finger)
13	0	2 (right wrist)	3 (right wrist)
14	0	0	3 (right wrist)
15	0	3 (right upper arm)	3 (right upper arm)

Range: 0–10, 11 points.

4. Discussion

The novel findings of this study are as follows: (1) compared with unaided walking, contralateral cane use decreased the hip moment impulse in the frontal plane during the stance phase (Fig. 2a), and (2) the hip moment impulse in the frontal plane during the stance phase decreased with an increase in cane BWS (Fig. 2a).

In this study, stance time, walking speed, cadence, step length and step width, peak contralateral and ipsilateral trunk lean angles, and cane lateral position were calculated and compared between conditions because changes in these variables affect hip ab-adduction moment waveforms [14,20,21,25,26] and may potentially affect hip moment impulse in the frontal plane. However, no significant differences were found in these variables except for the peak contralateral and ipsilateral trunk lean angles between the conditions (Table 1). Therefore, we consider that the effects of these variables, except for the peak contralateral and ipsilateral trunk lean angles, on hip moment impulse in the frontal plane are small.

The peak contralateral trunk lean angle increased significantly with an increase in cane BWS, and the peak ipsilateral trunk lean angle with cane non-use was greater than that with 20%BWS (Table 1). Hence, an increase in cane BWS caused an increase in contralateral trunk lean (right side) during the stance phase of the left limb. Hunt et al. [26]

examined the effect of lateral (i.e., ipsilateral side) trunk lean on peak hip adduction moment during the stance phase and found that early stance hip adduction moment in small (4°) lateral trunk lean (i.e., lateral trunk lean toward the ipsilateral side) was significantly lower than that in normal walking. This previous finding [26] revealed that contralateral trunk lean during the stance phase of the left limb may cause increased peak hip adduction moment and hip moment impulse in the frontal plane. However, our results (Fig. 2a and b) showed that the hip moment impulse in the frontal plane and peak hip adduction moment during the stance phase of the left limb decreased although contralateral trunk lean occurred in the cane conditions. In the present study, the peak contralateral trunk lean angle was 2.0° even with 20% BWS (Table 1). Therefore, we considered that the effect of contralateral trunk lean on hip moment impulse in the frontal plane and peak hip adduction moment was minimal.

Contralateral cane use generates vertical ground reaction force and may cause external hip abduction moment, as shown in Fig. 4. Therefore, contralateral cane use may cause a decrease in peak hip adduction moment in the stance limb, and a previous study [13] supports this hypothesis. However, another previous study [12] reported no significant difference in peak hip adduction moment between contralateral cane use and non-use. Although walking speed [20,21], cadence [21], step length {instead of stride length [21]}, and cane BWS magnitude may affect peak hip adduction moment in the stance limb, Chan et al. [12] did not match these variables between contralateral cane use and non-use. We believe that they could not find a significant difference in the peak hip adduction moment between contralateral cane use and non-use because these variables were not matched. Therefore, we matched these variables in all conditions. Consequently, contralateral cane use significantly decreased the peak hip adduction moment in the stance limb (Fig. 2b).

In the present study, the peak hip adduction moment without cane use and walking speed were 0.90 (0.08) Nm/kg [$= 9.8\,m/s^2$ x 0.091 (0.008) Nm/BW] and 0.50 (0.04) m/(s·HT), respectively. For the Chan et al. study [12], these values were 0.97 (0.30) Nm/kg and 0.60 (0.07) m/(s·HT), respectively, and for Tateuchi et al. [7], the values were 1.05 (0.29) Nm/kg and 0.70 (0.13) m/(s·HT), respectively. Although the participants in their studies were patients with knee osteoarthritis [12] or hip osteoarthritis [7], we observed that an increase in walking speed increased the first peak hip adduction moment, which agrees with a previous study [20]. Hence, we consider that the value of the peak hip adduction moment without cane use in the present study is quantitatively reasonable.

In the present study, the hip moment impulse in the frontal plane with cane non-use was 0.56 (0.05) Nm·s/kg [$= 9.8\,m/s^2$ x 0.057 (0.005) Nm·s/BW]; Tateuchi et al. [7] reported this value as 0.41 (0.13) Nm·s/kg. Therefore, the value in the present study is quantitatively reasonable. As mentioned in the previous paragraph, walking speed without cane use in the present study was lower than that in Tateuchi et al.'s study. Another previous study [27] revealed that an increase in walking speed causes a decrease in knee adduction moment impulse during walking. If this relationship could also be applied to the hip joint, the hip moment impulse in the frontal plane in the present study may be reasonable. However, the effect of walking speed on the hip moment impulse in the frontal plane has not been reported. Further studies are needed to examine the effect of walking speed on the hip moment impulse in the frontal plane to identify the difference in hip moment impulse values between the present study and that in Tateuchi et al. [7].

This study has some limitations. First, the participants were young and without hip osteoarthritis. A previous study [28] reported that an increase in pelvis motion was even at an early stage of hip osteoarthritis. Moreover, most patients with hip osteoarthritis have hip joint pain [7] and some patients have other affected joints (e.g., knee joint [3]). A change in kinematics due to hip joint pain and other affected joints may affect our main findings. Therefore, the effect of contralateral cane use on hip moment impulse in the frontal plane during gait in patients with hip osteoarthritis should be further examined. Second, most subjects felt discomfort in the upper limb in the 15%BWS and 20%BWS conditions. Contralateral cane is asymmetrical, and such movement may cause increases in load on the shoulder joint and lumber spine and may lead to their degenerative changes due to

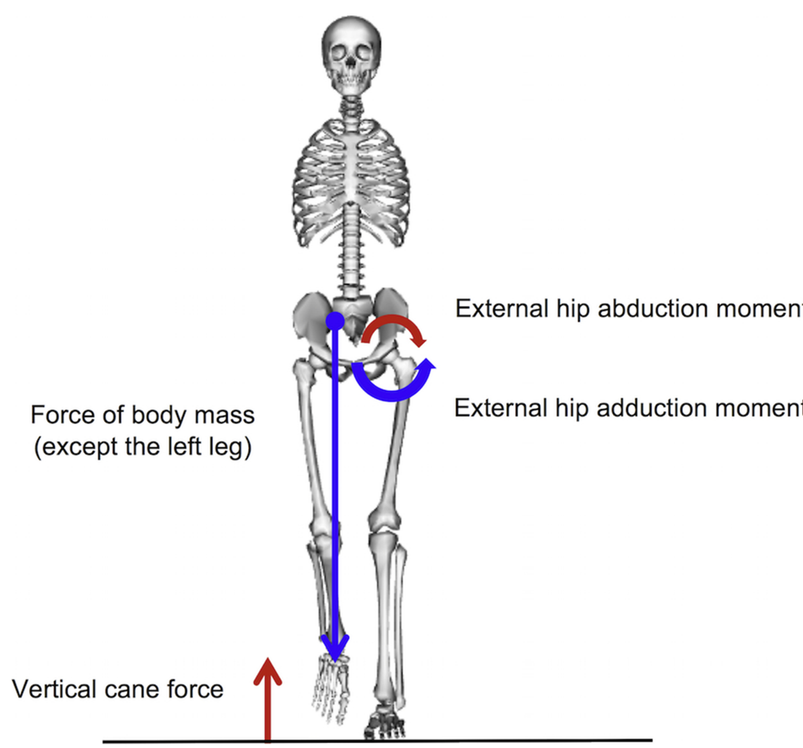

Fig. 4. Illustration of the hip abduction and adduction moments with contralateral cane use.
The vertical cane force generates external hip abduction moment in the left hip joint. This figure was created using OpenSim 3.3 [30].

overload for a long duration. In vivo measurement indicated that walking aids (forearm crutches and axillary crutches) exert shoulder joint force during walking [29]. Therefore, the effect of long-term contralateral cane use on overloading and pain (e.g., upper limb and low back) should be further evaluated.

5. Conclusion

This study aimed to examine the effect of contralateral cane use on hip moment impulse in the frontal plane and found that (1) compared with non-cane use, contralateral cane use decreased the hip moment impulse in the frontal plane during the stance phase, and (2) hip moment impulse in the frontal plane during the stance phase decreased with an increase in cane BWS. These findings may improve our understanding of gait patterns with low hip moment impulse in the frontal plane and of how to delay the progression of hip osteoarthritis.

Conflict of interest statement

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References

[1] Y.H. Pua, T.V. Wrigley, S.M. Cowan, K.L. Bennell, Hip flexion range of motion and physical function in hip osteoarthritis: mediating effects of hip extensor strength and pain, Arthritis Care Res. (Hoboken) 61 (2009) 633–640.
[2] A. Loureiro, P.M. Mills, R.S. Barrett, Muscle weakness in hip osteoarthritis: a systematic review, Arthritis Care Res. (Hoboken) 65 (2013) 340–352.
[3] M.P. Steultjens, J. Dekker, M.E. van Baar, R.A. Oostendorp, J.W. Bijlsma, Range of joint motion and disability in patients with osteoarthritis of the knee or hip, Rheumatology (Oxford) 39 (2000) 955–961.
[4] F. Salaffi, M. Carotti, A. Stancati, W. Grassi, Health-related quality of life in older adults with symptomatic hip and knee osteoarthritis: a comparison with matched healthy controls, Aging Clin. Exp. Res. 17 (2005) 255–263.
[5] A.H. Chang, K.C. Moisio, J.S. Chmiel, F. Eckstein, A. Guermazi, P.V. Prasad, Y. Zhang, O. Almagor, L. Belisle, K. Hayes, L. Sharma, External knee adduction and flexion moments during gait and medial tibiofemoral disease progression in knee osteoarthritis, Osteoarthr. Cartil. 23 (2015) 1099–1106.
[6] E.F. Chehab, J. Favre, J.C. Erhart-Hledik, T.P. Andriacchi, Baseline knee adduction and flexion moments during walking are both associated with 5 year cartilage changes in patients with medial knee osteoarthritis, Osteoarthr. Cartil. 22 (2014) 1833–1839.
[7] H. Tateuchi, Y. Koyama, H. Akiyama, K. Goto, K. So, Y. Kuroda, N. Ichihashi, Daily cumulative hip moment is associated with radiographic progression of secondary hip osteoarthritis, Osteoarthr. Cartil. 25 (2017) 1291–1298.
[8] T.W. O'Neill, P.S. McCabe, J. McBeth, Update on the epidemiology, risk factors and disease outcomes of osteoarthritis, Best Pract. Res. Clin. Rheumatol. 32 (2018) 312–326.
[9] W. Zhang, R.W. Moskowitz, G. Nuki, S. Abramson, R.D. Altman, N. Arden, S. Bierma-Zeinstra, K.D. Brandt, P. Croft, M. Doherty, M. Dougados, M. Hochberg, D.J. Hunter, K. Kwoh, L.S. Lohmander, P. Tugwell, OARSI recommendations for the management of hip and knee osteoarthritis, Part II: OARSI evidence-based, expert consensus guidelines, Osteoarthr. Cartil. 16 (2008) 137–162.
[10] D.A. Neumann, An electromyographic study of the hip abductor muscles as subjects with a hip prosthesis walked with different methods of using a cane and carrying a load, Phys. Ther. 79 (1999) 1163–1176.
[11] D.A. Neumann, Hip abductor muscle activity as subjects with hip prostheses walk with different methods of using a cane, Phys. Ther. 78 (1998) 490–501.
[12] G.N.Y. Chan, A.W. Smith, C. Kirtley, W.W.N. Tsang, Changes in knee moments with contralateral versus ipsilateral cane usage in females with knee osteoarthritis, Clin. Biomech. 20 (2005) 396–404.
[13] S. Ajemian, D. Thon, P. Clare, L. Kaul, R.F. Zernicke, B. Loitz-Ramage, Cane-assisted gait biomechanics and electromyography after total hip arthroplasty, Arch. Phys. Med. Rehabil. 85 (2004) 1966–1971.
[14] M. Simic, K.L. Bennell, M.A. Hunt, T.V. Wrigley, R.S. Hinman, Contralateral cane use and knee joint load in people with medial knee osteoarthritis: the effect of varying body weight support, Osteoarthr. Cartil. 19 (2011) 1330–1337.
[15] G. Kemp, K.M. Crossley, T.V. Wrigley, B.R. Metcalf, R.S. Hinman, Reducing joint loading in medial knee osteoarthritis: shoes and canes, Arthritis Rheum. 59 (2008) 609–614.
[16] T. Inai, T. Takabayashi, M. Edama, M. Kubo, Evaluation of factors that affect hip moment impulse during gait: a systematic review, Gait Posture 61 (2018) 488–492.
[17] M.A. Fang, C. Heiney, J.M. Yentes, N.D. Harada, S. Masih, K.L. Perell-Gerson, Effects of contralateral versus ipsilateral cane use on gait in people with knee osteoarthritis, PM&R 7 (2015) 400–406.
[18] M.A. Fang, C. Heiney, J.M. Yentes, N.D. Harada, S. Masih, K.L. Perell-Gerson, Clinical and spatiotemporal gait effects of canes in hip osteoarthritis, PM&R 4 (2012) 30–36.
[19] D.R. Aragaki, M.C. Nasmyth, S.C. Schultz, G.M. Nguyen, J.M. Yentes, K. Kao, K. Perell, M.A. Fang, Immediate effects of contralateral and ipsilateral cane use on normal adult gait, PM&R 1 (2009) 208–213.
[20] E.F. Chehab, T.P. Andriacchi, J. Favre, Speed, age, sex, and body mass index provide a rigorous basis for comparing the kinematic and kinetic profiles of the lower extremity during walking, J. Biomech. 58 (2017) 11–20.
[21] M.M. Ardestani, C. Ferrigno, M. Moazen, M.A. Wimmer, From normal to fast walking: impact of cadence and stride length on lower extremity joint moments, Gait Posture 46 (2016) 118–125.
[22] R. Kumar, M.C. Roe, O.U. Scremin, Methods for estimating the proper length of a cane, Arch. Phys. Med. Rehabil. 76 (1995) 1173–1175.
[23] P. de Leva, Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters, J. Biomech. 29 (1996) 1223–1230.
[24] D.G.E. Robertson, G.E. Caldwell, J. Hamill, G. Kamen, S.N. Whittlesey, Research Methods in Biomechanics, second ed., Human Kinetics, Champaign, 2004.
[25] M. Wesseling, F. de Groote, C. Meyer, K. Corten, J.P. Simon, K. Desloovere, I. Jonkers, Gait alterations to effectively reduce hip contact forces, J. Orthop. Res. 33 (2015) 1094–1102.
[26] M.A. Hunt, M. Simic, R.S. Hinman, K.L. Bennell, T.V. Wrigley, Feasibility of a gait retraining strategy for reducing knee joint loading: increased trunk lean guided by real-time biofeedback, J. Biomech. 44 (2011) 943–947.
[27] S.M.K. Robbins, M.R. Maly, The effect of gait speed on the knee adduction moment depends on waveform summary measures, Gait Posture 30 (2009) 543–546.
[28] E. Watelain, F. Dujardin, F. Babier, D. Dubois, P. Allard, Pelvic and lower limb compensatory actions of subjects in an early stage of hip osteoarthritis, Arch. Phys. Med. Rehabil. 82 (2001) 1705–1711.
[29] P. Westerhoff, F. Graichen, A. Bender, A. Halder, A. Beier, A. Rohlmann, G. Bergmann, In vivo measurement of shoulder joint loads during walking with crutches, Clin. Biomech. 27 (2012) 711–718.
[30] S.L. Delp, F.C. Anderson, A.S. Arnold, P. Loan, A. Habib, C.T. John, E. Guendelman, D.G. Thelen, OpenSim: open-source software to create and analyze dynamic simulations of movement, IEEE, Trans. Biomed. Eng. 54 (2007) 1940–1950.