Frontal plane kinematics in walking with moderate hip osteoarthritis: Stability and fall risk

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Article history:
Received 7 August 2014
Accepted 18 May 2015

Keywords:
Hip osteoarthritis
Fall risk
Gait analysis
Frontal plane balance
Trunk movements

Abstract

Background: Hip abductor weakness and unilateral pain in patients with moderate hip osteoarthritis may induce changes in frontal plane kinematics during walking that could affect stability and fall risk. Methods: In 12 fall-prone patients with moderate hip osteoarthritis, 12 healthy peers, and 12 young controls, we assessed the number of falls in the preceding year, hip abductor strength, fear of falling, Harris Hip Score, and pain. Subjects walked on a treadmill with increasing speeds, and kinematics were measured optoelectronically. Parameters reflecting gait stability and regressions of frontal plane center of mass movements on foot placement were calculated. We analyzed the effects of, and interactions with group, and regression of all variables on number of falls.

Findings: Patients walked with quicker and wider steps, stood shorter on their affected leg, and had larger peak speeds of frontal plane movements of the center of mass, especially toward their unaffected side. Patients’ static margins of stability were larger, but the unaffected dynamic margin of stability was similar between groups. Frontal plane position and acceleration of the center of mass, especially toward their unaffected side, which could contribute to fall risk.

Interpretation: Quickening and widening steps probably increase stability. Shorter affected side stance time to avoid pain, and/or weakened affected side hip abductors, may lead to faster frontal plane trunk movements toward the unaffected side, which could contribute to fall risk.

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1. Introduction

Hip osteoarthritis (HOA) has a worldwide prevalence of around 1%, and is a major contributor to global disability (Cross et al., 2014). Patients suffer from pain, mobility limitations, and stability problems (Edwards et al., 2014). Stability problems may induce falls (Ambrose et al., 2013), which often lead to further disability, and even serious morbidity (Stel et al., 2004). Patients with mild or moderate HOA have an increased risk of falling — a relative risk ratio of 1.4 was reported (Arnold and Gyurscik, 2012). Still, the underlying mechanisms have remained insufficiently clear, and need to be better understood (Arnold and Gyurscik, 2012). The present study aims at contributing to the understanding of mechanisms that underlie stability and fall risk in HOA.

Most falls occur during walking (Robinovitch et al., 2013). Research on a dynamical gait model and on healthy subjects suggests that in walking, frontal plane stability requires more active control than sagittal plane stability (Bauby and Kuo, 2000). A major hip abductor, i.e., the m. gluteus medius, was shown to play an important role in the control of the frontal plane movements of the center of mass (CoM) (Pandy et al., 2010). The trunk, arms, and head constitute an unwieldy segment with frontal plane movements that need to remain within certain limits to ensure stability (Hof et al., 2005). But in HOA, the abductors are often weak (Arnold and Gyurscik, 2012), particularly at the affected side (Arokoski et al., 2002). For the present study, we decided to focus on the impact of hip abductor weakness and of frontal plane trunk movements on stability and fall risk during walking in patients with HOA.
coM projection and BoS borders is a static "margin of stability", dCoM (Hof et al., 2005). Hof added a linear function of CoM speed for a dynamic margin of stability, dXCoM (Hof, 2008). In a review of walking with HOA, Constantinou et al. (2014) reported that patients tend to have larger step width than controls, which may increase margins of stability (Hak et al., 2012). Still, these margins co-depend on amplitude and speed of frontal plane trunk movements. Some HOA patients walk with lateral trunk inclination toward the affected side (Reininga et al., 2012). However, patients with more severe pain tend to walk with lateral inclination toward their unaffected side (Thurston, 1985). The impact of frontal plane trunk movements on stability and fall risk in walking with HOA has been insufficiently studied.

In healthy subjects, step width appears to be adapted on-line to frontal plane trunk kinematics in preceding mid-stance (Hurt et al., 2010). Step width may be adapted to changes in frontal plane CoM movements to maintain relatively large margins of stability (cf. Hak et al., 2012). To the best of our knowledge, the relationship between frontal plane kinematics and step width has not been studied yet in walking with HOA.

In our study of frontal plane kinematics, stability and fall risk during walking with HOA, we included two other valid estimators of gait stability (Brujin et al., 2013), i.e., the variability, and the short term Lyapunov exponent of frontal plane CoM movements. The Lyapunov exponent, or "local divergence exponent", assesses local dynamic stability, i.e., the sensitivity of the system to small kinematic variations, assumed to result from small internal or external perturbations. Most gait parameters are speed dependent, and HOA patients are known to prefer lower gait speeds (Constantinou et al., 2014). We studied walking at a range of gait speeds, taking into account that gait stability in HOA could be speed dependent, and HOA patients are known to prefer lower gait speeds (Constantinou et al., 2014). We included two rather general measures of health, the Falls Efficacy Scale (FES) and the Harris Hip Score (HHS). The FES is a questionnaire which assesses confidence to be able to perform activities of daily living without falling (Delbaere et al., 2010; Chinese version, Kwan et al., 2013), 6–64 points, with higher scores representing less confidence. A Chinese version of the Harris Hip Score (HHS; Harris, 1969) was used. The HHS combines surgeon-observed ranges of motion with self-reported pain and problems with activities of daily living (e.g., distance walked, problems with stair climbing, problems with public transport), 0–100 points, with higher points being better. Subjects filled in a 100 mm VAS scale for current pain, from "no pain", 0 mm, to "maximum pain", 100 mm.

Maximum isometric hip abduction force was measured with a dynamometer (Commander PowerTrack II muscle tester, JTECH Medical, Salt Lake City, UT, USA). With the subjects lying on their side (Bohannon, 1999; Leetun et al., 2004), the trunk was stabilized with a strap around the bench, pillows supported the leg in 10° abduction, and the dynamometer was secured to the bench, 30 cm distal to the trochanter major. The subject had to push the leg upwards, against the dynamometer, with maximal effort during 5 s, and the maximum was registered. This was repeated three times per leg, the average was calculated per leg, then multiplied by the moment arm (0.3 m), and divided by the subject's weight, the resulting dimension being Nm/kg.

All the above measurements were performed before kinematic testing on a treadmill. Moreover, after the walking session, subjects filled in a second VAS for pain.

2. Methods

2.1. Participants

We recruited a convenience sample of 12 patients (4 females) from two different hospitals (Table 1). We selected patients with unilateral HOA, who reported to have fallen at least once during the preceding year. To reduce variance between participants, we selected patients with "moderate" HOA only (KL grade 2 or 3; Kellgren and Lawrence, 1957). Patients had to be without any other self-reported pathology that would affect walking. We also recruited 12 age and BMI matched healthy subjects (2 females), and 12 healthy young controls (4 females). This allowed us to differentiate between the effects of hip OA (patients versus both control groups) and age (both elderly groups versus the young). The local Medical Ethics Committee approved the protocol, and participants signed an informed consent.

2.2. Subject characteristics

An orthopedic surgeon and a radiologist determined the Kellgren–Lawrence scores. Subjects reported how many times they had fallen during the last year. In view of the multidimensional nature of fall risk (Fabre et al., 2010), we included two rather general measures of health, the Falls Efficacy Scale (FES) and the Harris Hip Score (HHS). The FES is a questionnaire which assesses confidence to be able to perform activities of daily living without falling (Delbaere et al., 2010; Chinese version, Kwan et al., 2013), 6–64 points, with higher scores representing less confidence. A Chinese version of the Harris Hip Score (HHS; Harris, 1969) was used. The HHS combines surgeon-observed ranges of motion with self-reported pain and problems with activities of daily living (e.g., distance walked, problems with stair climbing, problems with public transport), 0–100 points, with higher points being better. Subjects filled in a 100 mm VAS scale for current pain, from "no pain", 0 mm, to "maximum pain", 100 mm.

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2.3. Kinematic data acquisition

Clusters of 3 markers each (infrared light emitting diodes), fixed on light metal plates, were attached, with neoprene bands, to the thorax (Th 7), the pelvis (between the posterior superior iliac spines), thighs, shanks, heels, and forearms. Movements were recorded with two 3-camera arrays of OptoTrak™ (Northern Digital, Waterloo, Ontario, Canada). With the subjects in the anatomical position, a pointer with six infrared light emitting diodes was used to locate the anatomical landmarks required to estimate segmental CoM positions (Zatsiorsky, 2002).

Participants walked on a treadmill (Bonte, Culemborg, The Netherlands) at incremental speeds, from 1 km/h to 5 km/h (increments of 1 km/h). After 1 min of warming up, data were recorded at 100 samples/s during 3 min. Subjects had 2 min rest between each two subsequent speed conditions, and were encouraged to indicate if speed was too high, after which the experiment would be stopped.

2.4. Data analysis

Data analysis was performed with custom made software in Matlab 7.13 (The Mathworks, Natick, MA, USA). Head contacts were estimated from maximum heel marker forward positions, and toe offs from the maxima in their vertical velocity (Pijnappels et al., 2001). Step width was derived from the mediolateral distance between the heel markers
at heel contact, and stride time from the time between subsequent heel contacts on the same side. Stance time, from heel contact to toe off, was calculated for each leg as percentage of stride time. The left leg in controls was, arbitrarily, chosen to compare to the unaffected leg in patients, and the right leg to compare to the affected leg.

A 12-segment 3D model was used to derive total body CoM position from segmental CoM positions, which were calculated from pointered (virtual) anatomical landmarks. We determined the frontal plane range of motion of the total body CoM. For its variability (cf. Toebes et al., 2012), we used the velocity time series (Kiss, 2010), normalized strides to 101 samples (0%–100%), and calculated the between-stride standard-deviations per time point. These 101 values were averaged over the stride, per subject, and per condition. Also frontal plane peak speed toward either side was derived from the velocity time series. Margins of stability (Hof, 2008; Hof et al., 2005) were calculated at heel strike from the frontal plane distance between CoM position, or extrapolated CoM position, and the heel of the stance foot.

To test whether foot placement was adjusted to preceding frontal plane kinematics, we performed bivariate regression analyses of within-subject mediolateral CoM position and acceleration at midstance on subsequent step width. The resulting R-value was taken as an indicator of predictive control of foot placement (cf. Hurt et al., 2010). Note that, originally, this analysis was performed with the trunk CoM (Hurt et al., 2010). However, for balance control total body CoM appears more important, and trunk CoM and body CoM movements are likely to be closely correlated, given the high relative mass of the trunk (Zatsiorsky, 2002).

Finally, we calculated the Lyapunov exponent (Bruijn et al., 2013; Dingwell and Cusamano, 2000) of the mediolateral CoM velocity time series (England and Granata, 2007) for the minimum number of strides available across subjects and conditions (i.e., 75). The CoM velocity time series was first resampled, so that it contained 75 × 100 samples for each subject and condition. A state space was created with the resampled time series plus four copies with time delays of 10, 20, 30, and 40 samples. Note that, within limits, the time delay and dimensionality are rather arbitrary (Rosenstein et al., 1993), and our choices are well within the range used in the literature. Moreover, Van Schooten et al. (2013) showed that it is best to have fixed rather than varying state-space reconstruction parameters. Maximum Lyapunov exponents were calculated as the slope of the mean natural log of divergence in these state spaces, from 0 to 1 step (Stenum et al., 2014).

2.5. Statistical analysis

SPSS 20 (IBM, Somers, NY, USA) was used throughout, with the critical alpha set at 0.05 to determine statistical significance. We analyzed the effects of group for variables that were not speed dependent (number of falls, HHS, FES, VAS before and after walking, maximum hip abduction force), with ANOVAs (3 groups) or T-tests (independent samples for 2 groups, paired for side comparisons in the patients, or one sample against the mean of a control group with 0 variance); numbers of falls were compared between the elderly groups with a Mann–Whitney U-test.

To deal with missing speeds in speed-dependent variables, we used generalized estimating equations (GEEs), on Group, Speed, and their interaction. If data were not normally distributed (Kolmogorov–Smirnov), a log transform was performed. After the initial GEE, non-significant interactions were removed. In the Results section, significant effects of Group and Speed, and significant interactions are given, in that order. In case of a significant effect of Group, a post hoc LSD was performed, whereas effects of Speed or significant interactions were interpreted on the basis of visual inspection. Where relevant, Side × Speed GEEs were performed within the patient group only. Group × Speed GEEs on the strength of the relationship between foot placement and preceding trunk kinematics were performed after the R-values were Fisher Z transformed.

To identify potential determinants of falling, we first calculated, in both elderly groups, the Pearson correlation matrix between the individually reported numbers of falls last year, and all other variables measured. For speed-dependent variables, we entered averages over speeds. From the correlation matrix, we selected all variables with a significant (P < 0.01), and high (≥0.7) correlation with number of falls. Then, we performed, in the patient group, backward multivariate linear regression of these variables on the individually reported numbers of falls.

3. Results

3.1. Subject characteristics

General group characteristics are given in Table 1. Mean (standard-deviation) KL grade of the patients was 2.5 (0.5). The patients had fallen most often, followed by the healthy peers, whereas no young subject reported any falls. Of the patients, 4 reported to have fallen once in the preceding year, 3 two times, 3 three times, and 2 four times. Of the healthy peers, 7 reported no falls, 3 one fall, and 2 two falls.

Scores on the FES and the HHS were worse in the patients, followed by the healthy peers, and then the young, who had the maximum scores throughout. A similar pattern was found for the affected (controls: right) side maximum hip abduction moment, but at the unaffected (controls: left) side, the difference between the two elderly groups was not significant. The abduction moment at the patient’s affected side was smaller than at their unaffected side (P = 0.03).

In the control groups, the pain before the experiment was always 0 mm, whereas the patients reported some pain. After the experiment, both elderly groups reported pain, the patients significantly more so than the peers (Table 1).

3.2. General kinematic variables

Maximum walking speed was 5 km/h in all controls, but lower in most patients (Table 1). In 1 patient, it was 1 km/h, in 5 patients 3 km/h, and in 4 patients 4 km/h.

All speed-dependent variables had non-normal distributions. Step width (Fig. 1A) differed between groups (P = 0.001), and was larger in the patients and the peers than in the young (LSD, P-values ≤ 0.002). Stride time decreased with Speed (P < 0.001). Also in stride time (Fig. 1B), there was an effect of Group (P = 0.002), it being shortest in the patients, followed by the peers, and then the young (LSD, P-values ≤ 0.03). Stride time decreased with Speed (P < 0.001).

AFFECTED (controls: right) side relative stance time was also affected by Group (P = 0.001), and was shorter in the patients, 63% (4%), than in either control group, 66%–67% (standard-deviations 4% in both; LSD, P-values < 0.001). It decreased with Speed (P < 0.001), with a significant interaction (P = 0.02), decreasing most over speed in the young. At the unaffected (controls: left) side, the only significant effect was a decrease with Speed (P < 0.001). Patients’ relative stance time was different between sides (P < 0.001), being shorter at the affected than at the unaffected side (LSD, P < 0.001). There was a significant decrease with speed (P < 0.001), and a significant Side × Speed interaction (P = 0.02), with more decrease over speeds at the unaffected side.

3.3. Frontal plane stability

The range of frontal plane CoM movements (Table 2) was non-significantly larger in the patients, 0.07 (0.03) m, than in the young, 0.04 (0.02) m, and the peers, 0.05 (0.03) m. In the variability of frontal plane CoM movements no effect of, or interaction with Group was found. Both these variables increased with Speed.

The mediolateral peak speeds of the CoM toward either side (Fig. 1C–D, Table 2) were larger in the patients than in the young, who were not significantly different from the healthy peers. Frontal plane...
peak speeds decreased with Speed. A Side × Speed GEE in the patients revealed a significant effect of Side on peak speed ($P = 0.03$), which was larger toward the unaffected than toward the affected side (LSD, $P = 0.03$). Again, a significant decrease with Speed was found ($P < 0.001$).

There was an effect of Group on the affected (right) side static margin of stability (Table 2), dCoM, being larger in the elderly groups—patients 0.15 (0.06) m and peers 0.12 (0.05) m—than in the young, 0.08 (0.03) m. At the unaffected (left) side, the same pattern was found. A Side × Speed GEE in the patients revealed an effect of Side ($P = 0.001$), with smaller dCoM values at the unaffected than at the affected side ($P < 0.001$). In all analyses, dCoM decreased with Speed ($P < 0.001$).

Also the affected (right) side dynamic margin of stability, dXCoM (Table 2), revealed an effect of Group, with larger values in the elderly groups—patients 0.10 (0.05) m and peers 0.08 (0.04) m—than in the young, 0.05 (0.02) m. However, in unaffected (left) dXCoM, no effect of Group was found. At both sides, values decreased with Speed. In a Side × Speed GEE with the patients only, there was a significant effect of Side on dXCoM ($P = 0.001$), it being smaller at the unaffected than at the affected side (LSD, $P = 0.001$). Values decreased with Speed ($P < 0.001$), particularly at the affected (right) side (interaction, $P = 0.02$).

At affected (right) side mid-stance, regressions of frontal plane CoM position and acceleration on subsequent step width were significant—in the young 0.70 (0.13), peers 0.67 (0.15), and patients 0.67 (0.15). No effect of Group was found, but the regressions were stronger at higher speeds ($P < 0.001$). At the unaffected (left) side, the pattern was similar, with 0.68 (0.15) in the young, 0.67 (0.15) in the peers, and 0.60 (0.17) in the patients.

The maximum short-term Lyapunov exponent of frontal plane CoM movements was non-significantly larger (i.e., more instability) in the patients, 2.6 (0.5), than in the young, 2.2 (0.5) and the healthy peers, 2.1 (0.5). The Lyapunov decreased with Speed ($P < 0.001$). This decrease over speed was found in all groups, but less so in HOA ($P$-value of the interaction 0.003).

### 3.4. Fall prediction

The number of falls reported by the elderly for the last year was correlated with a large number of measured variables, which also had many mutual correlations (Table 3). In the patients, backward multivariate linear regression resulted in a model with peak speed of the CoM toward the unaffected (left) side plus HHS predicting the number of falls with an adjusted $R^2$ of 0.83 ($P$-value of the model < 0.001). In and

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**Table 2**

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| Note | a increase with speed, decrease with speed.
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| b     | When group was significant, least significant differences were used, comparing young controls; h, healthy peers; p, patients; $\approx$, not significantly different from; $>$, significantly larger than.
| c     | The patients having higher values than both control groups at the highest speeds. |
Table 3
Correlation matrix for variables significantly ($P < 0.01$) correlated with number of falls in the three groups, taken together. Negative correlations are given in bold italic, and non-significant correlations ($P \geq 0.01$) were left out.

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A: number of falls in the preceding year.
B: Falls Efficacy Scale.
C: Harris Hip Score.
D: affected side maximum hip abduction moment/kg body weight.
E: pain before the measurements.
F: pain after the measurements.
G: maximum walking speed.
H: step width.
I: range of frontal plane CoM movements.
J: variability of frontal plane CoM movements.
K: peak speed of frontal plane CoM movements toward the affected side.
L: peak speed of frontal plane CoM movements toward the unaffected side.
M: affected side frontal plane margin of stability.
N: affected side extrapolated frontal plane margin of stability.
O: maximum Lyapunov exponent of frontal plane CoM movement.
of itself, the peak speed toward unaffected had 55% common variance ($r_p = 0.74$) with number of falls reported.

4. Discussion

Fall-prone patients with moderate hip osteoarthritis walked with different speeds on a treadmill, and were compared to healthy peers and young controls.

4.1. General kinematics

Maximum walking speed was lower in the patients than in both control groups. Similar was reported for knee osteoarthritis (Fallah-Yakhdani et al., 2011). In the present study, both elderly groups walked with wider steps than the young, which has been interpreted as a strategy to improve mediolateral stability (Hak et al., 2012; Hof et al., 2007). In their review of walking with HOA, Constantinou et al. (2014) reported a large between-study variance in step width difference between hip osteoarthritis patients and healthy controls, with, however, in the studies that controlled for walking speed on average significantly wider steps in the patients.

Patients had shorter stride times than controls. Similar was reported for studies controlling for walking speed in Constantinoiu et al.’s (2014) review. Also walking with shorter stride times may be a strategy to increase mediolateral stability (Hak et al., 2012; Hof et al., 2007).

Patients’ stance time on the affected leg was shorter than stance time in controls, and shorter than stance time at their unaffected limb. Such asymmetry in patients agrees with the literature (Constantinoiu et al., 2014). Perhaps, having shorter stance time is related to avoiding the pain due to supporting body weight while standing on the affected leg; or it may be caused by affected side abductor weakness.

4.2. Frontal plane stability

Frontal plane range of movement of the center of mass (CoM) was non-significantly larger in the patients than in young controls, as was reported for trunk range of movement in elderly subjects by Hurt et al. (2010). We found no group effect on the variability of frontal plane movements. Toebes et al. (2012) studied this variability in 134 elderly subjects with a history of falling, and reported it to be positively related to the number of falls in the last year. Lack of power in our study may explain why we found no variability effect.

The peak speed of frontal CoM movements was higher in the patients than in the young, toward unaffected even more so than toward affected. This asymmetry may have resulted from a tendency to quickly unload the affected leg, and/or an inability to brake movements toward unaffected because of affected abductor weakness. Yang et al. (2009) argued that frontal plane trunk speed should remain within limits for gait to be stable. Accordingly, high peak speed of frontal plane CoM movements, particularly toward unaffected, may be a risk factor for falling.

The static margins of stability were larger in both elderly groups than in the young, because the elderly groups walked with wider steps (cf. Aberg et al., 2010; Hak et al., 2012; McAndrew Young and Dingwell, 2012). In the patient group, the static margin of stability was largest at the affected side, which suggests that patients walked with more lateral trunk inclination toward unaffected than toward affected, as Thurston (1985) reported for patients with more serious pain. Note, however that Reininga et al. (2012) observed more inclination toward affected; however, in that study inclination toward affected was used as a preselection criterion.

In the patients, also the dynamic margin of stability was larger at the affected than at the unaffected side. Moreover, the affected side dynamic margin of stability was larger in both elderly groups than in the young, but at the unaffected side it was not different between groups. Here, the stabilizing effect of widening the step appeared to be canceled by the destabilizing effect of having a larger CoM speed. Thus, as to our first hypothesis, results were ambiguous in that a larger speed of the CoM toward unaffected appeared to reduce the unaffected dynamic margin of stability, whereas, in fact, the unaffected side dynamic margin of stability was not different between groups. Note, however, that patients may require relatively large unaffected side dynamic margins of stability, since normal correction mechanisms may be failing, when affected side abductor weakness renders it difficult to quickly decelerate CoM movement toward the unaffected side. Hence, the “normal” dynamic margin of stability at the unaffected side may hide a fall risk.

Regression analysis of CoM position and acceleration on subsequent step width revealed $R^2$-values from 0.6 to 0.7. Hurt et al. (2010) reported an $R^2$ of 0.54, that is, an $R$ of 0.73, which is similar to the values we found. Contrary to our study, Hurt found a stronger regression in elderly than in young subjects, but Hurt did not control for speed. Anyhow, our study confirmed that frontal plane kinematics in mid-stance predict subsequent step width. Large, strongly accelerating frontal plane movements, and thus fast movements, imply wider steps.

Different from some studies (e.g., Bruijn et al., 2014), we found no main effect of group on the mediolateral Lyapunov exponent, but in HOA, values decreased less with speed (i.e., stability increased less). Note that this is not necessarily the same as the observation that elderly subjects walk with less safety at higher speeds (Mademli and Arampatzis, 2014). Nor is our hypothesis that the effects of HOA on gait stability would be more pronounced at higher gait speeds, unambiguously confirmed. Still, in our study, HOA patients appeared to be less able to produce a stable gait pattern at higher speeds than healthy controls.

4.3. Fall prediction

In and of itself, peak speed toward unaffected had 55% variance in common with the number of falls reported. Our second hypothesis, that frontal plane kinematic abnormalities would predict the number of falls reported, was thus confirmed. We assume that in daily life our patients sometimes had even faster trunk movements toward unaffected, which then could not be stopped and reversed in time, because of affected side abductor weakness.

Falling is multifactorial. The HHS gives a more global assessment of the patient’s functional status, and adding the HHS to the regression on the number of falls reported, led to 83% of the variance being accounted for. We were surprised to see such a high value (cf., Toebes et al., 2012; Van Schooten et al., 2015; Weiss et al., 2014). Unfortunately, however, our number of subjects was too small for HHS item-analysis, and we do not know which aspects of the HHS were most indicative of fall risk.

4.4. Limitations

We studied a small group of patients, but found a large number of significant results, and lack of power was probably no major problem. During our experiments, we observed no actual falls, but used the retrospectively reported number of falls in the preceding year. Note that inaccuracies of our subjects’ memories could have biased our results, but the fact that our model gave a high $R$-square on number of falls reported, suggests that random inaccuracies of memory played no major role.

Our study was largely observational, which precludes any causal conclusions. Nevertheless, the pattern of results made sense. Treadmill walking and overground walking are different in relevant respects (e.g., Dingwell et al., 2001), but this was the same for all groups. For correlations and regressions, we used averages over speeds, which may have biased our data toward the lower speeds in patients. We expect that this was a conservative bias. Finally, we used linear methods to study the prediction of number of falls reported, and our final model certainly requires further study.
5. Conclusion

Fall-prone patients with moderate hip osteoarthritis and pain in one hip walked with wider and quicker steps. They spent less time in stance while on the affected leg. Affected side hip abductors were weak, and peak speeds of frontal plane movements of the center of mass were faster toward the unaffected than toward the affected side. This appeared to cancel the stabilizing effect of increasing step width, and the unaffected side dynamic margin of stability was not different from that in controls. Still, peak speed toward unaffected had 55% common variance with the number of falls reported for the preceding year, probably because affected side abductor strength was insufficient to brake faster movements toward unaffected. Adding the Harris Hip Score led to 83% variance in the number of falls accounted for.

In sum, the peak speed of frontal plane trunk movements toward the unaffected side during walking was found to be a major risk factor for falling in patients with moderate hip osteoarthritis. It remains to be established if strength training of affected side hip abductors would reduce fall risk, and if frontal plane kinematics are affected by hip replacement.

Acknowledgments

W.H.W. was supported by a grant for Quanzhou Leading Scientific Talents in 2013, the Fund for Quanzhou Excellent Talents (#08A17, #11A02), the National Natural Science Foundation of China (#10272161), the Natural Science Foundation of Fujian Province, China (#2013J01125), and the Fund of Professorship for Academic Development of Fujian Medical University (JS11001). S.M.B was funded by a grant from the Netherlands Organization for Scientific Research (NWO #451-12-041). No funding organization interfered with the content of the study, or with the text of our report. Hamid Abassi Bafghi and Hamid Reza Fallah-Yakhdani contributed to the writing of earlier versions of the manuscript.

The authors thank Gao Jintian for his support of our Sino-Dutch cooperation, and Zhang LiQun, Maarten Prins as well as Hans van den Berg for stimulating discussions.

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