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

XiaoBin Lin [a,b,c,d], Onno G. Meijer [d,e,*], JianHua Lin [a,c,**], WenHua Wu [b,d], XiaoCong Lin [b,d], BoWei Liang [d,f], Jaap H. van Dieën [e], Sjoerd M. Bruijn [c,e]

[a] First Clinical College of Fujian Medical University, Fuzhou, Fujian, PR China
[b] Department of Orthopaedics, Second Affiliated Hospital of Fujian Medical University, Quanzhou, Fujian, PR China
[c] Department of Orthopaedics, First Affiliated Hospital of Fujian Medical University, Fuzhou, Fujian, PR China
[d] Orthopaedic Biomechanics Laboratory, Fujian Medical University, Quanzhou, Fujian, PR China
[e] MOVE Research Institute Amsterdam, Faculty of Human Movement Sciences, VU University, Amsterdam, The Netherlands
[f] Department of Orthopaedics, The 175th Hospital of PLA, Affiliated Dongnan Hospital of Xiamen University, Zhangzhou, Fujian, PR China

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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 opto-electronically. 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 predicted subsequent step width. The peak speed of frontal plane movements toward unaffected had 55% common variance with number of falls, and adding the Harris Hip Score into bivariate regression led to 83% "explained" variance.
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.

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.

* Correspondence to: O.G. Meijer, Research Institute MOVE, Faculty of Human Movement Sciences, VU University, Van der Boechorststraat 9, 1081 BT Amsterdam, The Netherlands.
** Correspondence to: J.H. Lin, Department of Orthopaedics, First Affiliated Hospital of Fujian Medical University, Cha Zhong Road 20, 350005 Fuzhou, PR China.
E-mail addresses: ogmeijer@yahoo.com (O.G. Meijer), jianhual@126.com (J. Lin).

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Static stability requires the vertical projection of the center of mass to remain within the base of support (BoS) — the feet plus, in bipedal standing, the space between them. The minimum distance between 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 (Bruijn 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 more impaired at higher speeds. We hypothesized that frontal plane CoM kinematics in walking with HOA would 1) reduce gait stability when compared to controls, particularly at higher gait speeds, and 2) predict self-reported number of falls.

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.

Table 1
General group characteristics.

	Young ($N = 12$)		Patients ($N = 12$)		Healthy peers ($N = 12$)	
	Mean	SD	Mean[a]	SD	Mean[a]	SD
Age (years)	24.3	3.7	64.4	5.3	64.5	3.5
BMI	21.1	3.0	22.4	3.7	22.7	3.9
Number of falls in the preceding year	0	0	2.3**	1.1	0.6*	1.1
FES[b]	16	0	33.3**	3.8	25.1*	2.7
HHS[c]	100	0	73.5**	7.2	97.2*	1.6
Affected side hip abduction moment (Nm/kg)	1.4	0.2	0.8**	0.3	1.0*	0.2
Unaffected side hip abduction moment (Nm/kg)	1.3	0.2	0.9*	0.3	1.0*	0.2
Pain before the experiment (mm)	0	0	13.8**	12.2	0	0
Pain after the experiment (mm)	0	0	50.3**	12.2	15.3*	3.4
Maximum speed (km/h)	5.0	0	3.5**	1.1	5.0	0

[a] *Worse than the young controls ($P < 0.05$). **Worse than both control groups ($P < 0.05$).
[b] Falls Efficacy Scale.
[c] Harris Hip Score.

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.

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.

2.3. 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). Heel 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