



Full length article

Kinematic variability and local dynamic stability of gait in individuals with hip pain and a history of developmental dysplasia[☆]Kari L. Loverro^{a,b,*}, Anne Khuu^a, Pei-Chun Kao^c, Cara L. Lewis^a^a Department of Physical Therapy & Athletic Training, PhD Program in Rehabilitation Sciences, College of Health & Rehabilitation Sciences: Sargent College, Boston University, 635 Commonwealth Ave., Boston, MA, United States^b Natick Soldier Research Development and Engineering Center, 10 General Greene Ave., Natick, MA, United States^c Department of Physical Therapy, 3 Solomont Way, Suite 5, University of Massachusetts Lowell, Lowell, MA, United States

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ABSTRACT

Background: Individuals with developmental dysplasia of the hip (DDH) often report hip pain and exhibit gait adaptations. Previous studies in this patient population have focused on average kinematic and acceleration measures during gait, but have not examined variability.

Research question: Do individuals with hip pain and DDH have altered kinematic variability or local dynamic stability (LDS) compared to individuals without hip pain?

Methods: Twelve individuals with hip pain and DDH and 12 matched controls walked for two minutes on a treadmill at three speeds: preferred, fast (25% faster than preferred), and prescribed (1.25 m/s). Kinematic variability of spatiotemporal measures, joint and segment angles, and LDS of the trunk were calculated for each speed.

Results: At the prescribed speed, individuals with hip pain and DDH had more kinematic variability than controls at the hip, pelvis, and trunk as well as greater variability in spatiotemporal measures. LDS was not different between groups. Kinematic variability of the joints decreased and LDS of the trunk increased (i.e., increased gait stability) with increased speed.

Significance: Individuals with hip pain and DDH had greater kinematic variability compared to individuals without hip pain when walking at the same prescribed speed, indicating either an adaptation to pain or reduced neuromuscular control. LDS of the trunk was not different between groups, suggesting that hip pain does not affect overall gait stability. Kinematic variability and LDS were affected by walking speed, but in different ways, emphasizing that these measures quantify different aspects of walking behavior.

1. Introduction

Individuals with developmental dysplasia of the hip (DDH) often report hip pain and exhibit gait alterations. DDH is characterized by decreased acetabular coverage of the femoral head anteriorly and laterally, leading to increased stress on the acetabular rim [1], hip pain [2], and osteoarthritis (OA) [3]. Gait alterations in individuals with DDH may reduce the load on the acetabular rim [1,4] and pain [1,4,5]. These alterations include reduced peak hip extension [1,4,5] and flexion angle [5], reduced peak plantar flexion angle at toe-off [4], and decreased stride length and gait velocity [1,5] compared to healthy individuals. Alternatively, gait alterations may indicate hip instability [6]. Measures of gait alterations typically reflect average behavior over multiple strides; however, it is unknown how variable or stable the

behavior is between strides. Clinically, we have observed that individuals with hip pain and DDH appear to have a more variable gait pattern. Measures that can quantify how individuals with hip pain vary their gait pattern between strides or how stable strides are over time may give insight into pain alleviation strategies or neuromuscular control of gait patterns.

Kinematic variability measures how individuals modify gait mechanics from stride-to-stride and is commonly quantified using standard deviation (SD) [7–16] or coefficient of variation (CV) [8,17,18] of kinematic measures. Decreased kinematic variability of the affected limb has been noted in individuals with knee [7,17] and hip [9] OA. This decrease in kinematic variability was interpreted as increased rigidity and lack of flexibility in the affected limb [9,17,19] and may be a predictor of gait instability and risk of falling [20]. Manipulating speed

[☆] This study was approved by Institutional Review Boards of Boston University and all individuals provided written informed consent prior to participation.

* Corresponding author at: Natick Soldier Research Development and Engineering Center, 10 General Greene Ave., Natick, MA, United States.

E-mail address: kloverro@bu.edu (K.L. Loverro).

in each of these studies affected kinematic variability, but with differing results [7–9,17].

One limitation of kinematic variability is the use of linear statistics (e.g., SD and CV) to quantify stride to stride variability where each stride is considered to be independent [19]. In reality, strides are connected, where each stride is influenced by the previous stride or strides. Nonlinear measures overcome this limitation by considering the interdependence of strides and changes in gait kinematics over time, thus providing insight into the neuromuscular control of gait [21]. The maximum Lyapunov exponent is a nonlinear measure that estimates local dynamic stability of gait [8,11,13,21–24] by calculating the average rate at which the kinematics of one stride becomes different from the prior stride [7]. Small (e.g., pain or neuromuscular dysfunction) or large (e.g., tripping) perturbations may cause differences in strides during walking [23]. Pain or reduced neuromotor control may also produce perturbations that decrease the local dynamic stability of gait, increasing the rate at which strides are becoming more different.

Quantifying kinematic variability and local dynamic stability of individuals with hip pain and DDH may provide insight into what we have observed clinically, as well as informing rehabilitation strategies. Previous studies have focused on average kinematic [1,4,5] and acceleration [6] measures during gait, but have not examined variability or stability. Therefore, the purpose of this study was to investigate kinematic variability and local dynamic stability in individuals with hip pain and DDH and in individuals without hip pain. A secondary purpose was to evaluate the effect of walking speed on these measures. We hypothesized that adaptations due to pain or altered neuromuscular control in individuals with hip pain will increase kinematic variability and decrease local dynamic stability. Additionally, we hypothesized that individuals with hip pain will have increased kinematic variability and decreased local dynamic stability when walking at speeds faster than their preferred speed compared to individuals without hip pain.

2. Methods

2.1. Participants

Participants included in this analysis were part of a larger, on-going study of individuals with intra-articular hip pathology and pain. Participants had to be between the ages of 14 and 50 years old and report being able to walk safely for at least 10 min without an assistive device. General exclusion criteria included reporting a history of neurological disorders or back surgery, or current back, knee, or ankle pain. For this analysis, individuals included in the hip pain group were recruited through area orthopedic and rehabilitation clinics between January 2010 and December 2016, had a history of DDH, and had their pain reproduced with walking and by at least one of three provocative tests. The three tests, which are sensitive to intra-articular hip pathology [25], included: 1) the flexion, adduction, internal rotation (FADIR) test [26], 2) the flexion, abduction, external rotation (FABER) test [27], and 3) the resisted straight leg raise (SLR) [25]. Prior hip surgery was not an exclusion criterion.

Individuals without hip pain (control group) were recruited through postings and word-of-mouth. In addition to the general inclusion / exclusion criteria, these individuals had to have no pain or discomfort with the provocative tests and no history of lower extremity surgery. Controls selected for this analysis were one-to-one matched with individuals in the hip pain group for sex, age, height, mass, and activity score.

We performed our *a priori* power analysis on meanSD of sagittal plane hip angles, as this was the primary variable of interest, using G*Power (v3.0.10). Based on the available literature [9], we anticipated a group difference of 2.3° (0.73° pooled standard deviation) for meanSD of sagittal plane hip angles during preferred walking. Therefore, less than 10 participants per group were required to achieve statistical power of .80 with an alpha of 0.05.

This study was approved by the Institutional Review Board of Boston University. All participants provided written informed consent prior to participation.

2.2. Experimental protocol and setup

Participants completed self-report questionnaires including the UCLA activity score [28], modified Harris Hip Score (mHHS) [29], and the hip disability and osteoarthritis outcome score (HOOS) [30]. The Western McMaster Universities Osteoarthritis Index (WOMAC) was scored from the HOOS [31]. The mHHS scores can be interpreted as 90–100 excellent, 80–90 good, 70–80 fair, and below 70 as poor [29]. Hip pain group participants were also asked to rate their hip pain levels on an 11-point (0: no pain-10: worst pain) numeric rating scale [32] at its best, worst, and average in the previous two weeks.

Participants wore tight fitting shorts, a t-shirt, and their own shoes. Thirty reflective spherical markers were placed on anatomical landmarks on the trunk, pelvis, and bilaterally on the lower extremities. Plastic shells that had four non-collinear markers each were positioned laterally over the thigh and shank [33]. Marker shells were secured to the thigh and shank using neoprene wraps with hook and loop fasteners [34]. This marker setup has been previously used by Lewis et al. [35] to quantify whole body kinematics. Following marker placement, a static standing calibration trial was collected to create a subject specific model.

Participants walked on a treadmill at three speeds in the following order: 1) their preferred speed, 2) 125% of the preferred speed, and 3) a prescribed speed (1.25 m/s). Two minutes of continuous steady-state walking data were collected at each speed. Preferred speed was determined using the average walking velocity of five overground walking trials. Participants verbally rated their pain on the 11-point numeric rating scale [32] every 30 s.

Marker and ground reaction force data were collected using a motion capture system (100 Hz, Vicon Motion Systems Ltd, Centennial, CO) and a split-belt instrumented treadmill (1000 Hz, Bertec Corporation, Columbus, OH). Visual3D (C-Motion, Inc. Germantown, MD) was used to process data and derive joint and segment angles and spatiotemporal measures. Briefly, marker trajectories were filtered using a low-pass, fourth-order Butterworth filter with a cutoff frequency of 6 Hz [36]. Joint angles of the hip, knee, and ankle were determined with respect to the proximal segment. Segment angles of the pelvis and trunk were determined with respect to the lab coordinate system. Joint and segment angles were calculated using a Visual3D hybrid model with a Cardan X-Y-Z (mediolateral, anteroposterior, vertical) rotation sequence [37]. The model consisted of eight rigid segments: a trunk, a pelvis, right and left thighs, right and left shanks, and right and left feet. The CODA model was used to define the pelvis segment and predict hip joint centers [38]. The first 95 strides of each trial were used for analysis because 95 was the minimum number of continuous strides across all trials.

2.3. Data analysis

2.3.1. Kinematic variability

For each stride, joint and segment angles were normalized to the gait cycle. Spatiotemporal measures included percent double support time, percent stance time, stride length, step length, and step width. The SD of joint and segment angles at each time-normalized point in the gait cycle was calculated and then averaged across the cycle to determine the mean SD (meanSD) [12,15]. The within-subject SD of peak angles and spatiotemporal measures were calculated for each trial.

2.3.2. Local dynamic stability

For the local dynamic stability analysis, we used Rosenstein's algorithm [39] to calculate the short-term local divergence exponents (LDE), also referred to as the maximum Lyapunov exponent, of the 7th

cervical vertebrae (C7) marker velocity time series for each of the three directions of motion (mediolateral (ML), anteroposterior (AP), and vertical (VT)) [13,39–41]. We chose to use the C7 marker based on previous research suggesting that motions of the trunk can be used to determine the stability while walking [12,13]. Methods to calculate LDE using continuous time series data are well established [13,15,21,42], and briefly describe here. The C7 marker data were filtered with a low-pass 4th order Butterworth filter with a cut-off frequency of 10 Hz [42]. Previous research has shown that C7 marker position is nonstationary due to participants failing to maintain the same position on the treadmill while walking [12]. Local dynamic stability analysis requires the time series to be stationary (e.g., constant mean, variance, etc. over time) [43,44], which was achieved by using the first central difference time series (i.e., velocity) for each direction (mediolateral (ML), anteroposterior (AP), and vertical (VT)) of the C7 marker data [12,43]. Differences in time series length and number of strides among trials may affect estimates of LDE [45–47], therefore, we included only the first 95 consecutive strides of each trial for analysis. The time series were time normalized so that each time series of 95 strides was 9500 samples in length, or approximately 100 samples per stride [42].

Reconstructing a time series and its time-delayed copies in a 5-dimensional state space captures the dynamics of human walking [13,21,42]. The appropriate time delays for each direction of the time series (ML: 36 samples, AP: 14 samples, V: 12 samples) were determined using the first minimum of the average mutual information function [48]. With the time series reconstructed in state space, the Euclidian distances between each data point and its nearest neighbors were tracked over five strides. Nearest neighboring points were selected as the data points closest to an initial data point outside the initial point's stride [42]. For example, a data point from stride 57 would have nearest neighboring points found only in strides 1–56 or 58–95. The logarithm of each distance was calculated and averaged for all pairs of initial nearest neighbors. The LDE was estimated by the slope of a linear least-squares fit to the average logarithmic divergence curve between 0 and the 1st stride [21,40]. If each stride was identical to every following stride, the distance between nearest neighboring points would be constant [42] and thus, LDE would be zero indicating a stable, non-chaotic locomotor system. As the divergence rate of the distance between nearest neighbors increases, LDE becomes a larger positive value indicating that the locomotor system is more unstable [41]. In other words, as LDE increases, strides are becoming different from previous strides at a faster rate indicating decreased local dynamic stability of gait.

2.4. Statistical analysis

Dependent variables used to quantify kinematic variability included the meanSD of joint and segment angles in the sagittal and frontal planes across the entire gait cycle, SD of peak joint and segment angles in the sagittal and frontal planes, and SD of spatiotemporal measures. The dependent variable used to quantify local dynamic stability was the LDE of the C7 velocity time-series in each direction. The painful side was analyzed for individuals with unilateral pain, the more painful side for individuals with bilateral pain, and for controls, the side corresponding to their match in the hip pain group.

We used independent *t*-tests to determine group differences in kinematic variability and local dynamic stability at the prescribed speed. We used separate Generalized Linear Models with Generalized Estimating Equation corrections to evaluate the effect of speed (within-subject factor), the effect of group (between-subject factor), and the interaction of group-by-speed. Least significant difference post-hoc pairwise tests were performed if the factors were significant. For significant interaction effects, planned comparisons were performed between groups and main effects were not reported. Analyses were conducted in SPSS (v20, IBM Inc.) with an alpha of .05.

Table 1
Participants' demographics.

	Hip Pain		Control	p-value
Sex (N)	11 F,1 M		11 F,1 M	–
Demographics (Mean ± SD)				
Age (yrs)	24.9 ± 4.9		22.5 ± 3.7	0.184 ^a
Height (m)	1.65 ± 0.05		1.72 ± 0.15	0.177 ^a
Weight (kg)	66.0 ± 9.8		61.3 ± 7.7	0.206 ^a
Preferred Speed (m/s)	1.23 ± 0.13		1.25 ± 0.15	0.798 ^a
Questionnaires				
UCLA Activity (median (range))	10(4-10)		8.5 (5-10)	0.631 ^b
	More Painful	Less Painful		
mHHS	70.1 ± 11.5	88.3 ± 13.3	100.1	–
HOOS Subscales				
Pain	69.4 ± 17.0	90.6 ± 13.0	100	–
Symptoms	65.4 ± 16.0	85.0 ± 14.8	96.7 ± 5.0	–
Functional Activities	87.9 ± 10.0	96.9 ± 5.3	100	–
Recreation/Sport Activities	63.0 ± 21.2	88.0 ± 14.0	99.5 ± 1.8	–
Quality of Life	46.9 ± 17.0	83.9 ± 19.5	99.5 ± 1.8	–
WOMAC	83.7 ± 11.2	95.3 ± 7.1	100	–
Positive Provocative tests				
FADIR	12(100%)	5(42%)	–	–
FABER	9(75%)	3(25%)	–	–
SLR	8(67%)	2(17%)	–	–
Radiographic Analysis				
Lateral Center Edge Angle (°)	18.6 ± 4.4	19.7 ± 9.8	–	–
Anterior Center Edge Angle (°)	21.4 ± 5.7	21.0 ± 8.3	–	–
Pain During Trial				
Preferred	1.79 ± 1.83	0.25 ± 0.87	–	0.001 ^c
Fast	2.67 ± 1.96	0.33 ± 0.87	–	–
Prescribed	2.21 ± 0.38	1.83 ± 0.93	–	–
Pain in the Last Two Weeks				
Worst	6.59 ± 1.39	1.37 ± 2.36	–	–
Best	1.72 ± 1.62	0.09 ± 0.30	–	–
Average	3.68 ± 1.27	0.52 ± 1.03	–	–

Note: Values are presented as mean ± standard deviation. mHHS: modified Harris Hip Score. HOOS: hip disability and osteoarthritis outcome score WOMAC: Western McMaster Universities Osteoarthritis Index. FADIR: Flexion Adduction Internal Rotation. FABER: Flexion Abduction External Rotation. SLR: resisted Straight Leg Raise.

^a Independent *t*-test compared to controls.

^b Mann-Whitney U test.

^c Independent *t*-test compared to the fast walking speed.

3. Results

3.1. Participant demographics

Twelve individuals with hip pain and 12 matched controls participated in the study (Table 1). Preferred speed was not different between groups. Five individuals in the hip pain group had bilateral pain. Four individuals had surgery on the more painful hip at least 11 months prior to data collection. Two had a periacetabular osteotomy (PAO). The average lateral center edge angle of the more painful side was 18.6 ± 4.4° and the less painful side was 19.7 ± 9.8°. The average anterior center edge angle of the more painful side was 21.4 ± 5.7° and the less painful side was 21.0 ± 8.3°.

3.2. Questionnaires and pain ratings

The hip pain group scored lower on the mHHS, HOOS, and WOMAC than the control group (Table 1) and reported greater pain walking at the fast speed compared to their preferred speed (*p* < 0.001). Based on the mHHS scoring, on the more painful side 8 individuals were categorized as poor, 1 as fair, 2 as good and 1 excellent. On the less painful side, 2 individuals were categorized as poor, 2 as fair, and 8 as

Table 2
Kinematic Variability (meanSD or SD) and Local Dynamic Stability (LDE) of dependent variables.

	Prescribed			Preferred		Fast		Group	Speed	Group* Speed
	DDH	Control	p-value	DDH	Control	DDH	Control			
Ankle										
Sagittal										
MeanSD (°)	1.28 ± 0.23	1.33 ± 0.25	0.651	1.40 ± 0.26	1.34 ± 0.24	1.31 ± 0.27	1.30 ± 0.22	0.731	0.010	0.231
Dorsiflexion (°)	1.27 ± 0.26	1.31 ± 0.29	0.721	1.38 ± 0.44	1.31 ± 0.34	1.20 ± 0.35	1.25 ± 0.30	0.954	0.028	0.283
Plantar Flexion (°)	2.10 ± 0.68	2.17 ± 0.81	0.818	2.42 ± 0.68	2.19 ± 0.82	1.85 ± 0.60	1.75 ± 0.57	0.512	< 0.001	0.415
Frontal										
MeanSD (°)	1.18 ± 0.27	1.20 ± 0.42	0.897	1.27 ± 0.27	1.15 ± 0.21	1.22 ± 0.30	1.11 ± 0.20	0.221	0.043	0.995
Inversion (°)	1.17 ± 0.40	0.97 ± 0.28	0.155	1.34 ± 0.53	1.04 ± 0.31	1.17 ± 0.52	0.90 ± 0.21	0.075	< 0.001	0.719
Eversion (°)	1.10 ± 0.30	1.02 ± 0.43	0.611	1.12 ± 0.26	1.04 ± 0.31	1.17 ± 0.33	0.98 ± 0.34	0.242	0.888	0.162
Knee										
Sagittal										
MeanSD (°)	2.06 ± 0.32	2.03 ± 0.29	0.799	2.17 ± 0.33	2.15 ± 0.34	1.97 ± 0.39	2.02 ± 0.27	0.933	< 0.001	0.509
Extension (°)	1.06 ± 0.42	0.99 ± 0.29	0.666	1.20 ± 0.56	1.14 ± 0.37	1.03 ± 0.44	1.06 ± 0.39	0.949	0.021	0.406
Flexion (°)	1.27 ± 0.44	1.38 ± 0.35	0.509	1.27 ± 0.26	1.35 ± 0.26	1.20 ± 0.34	1.41 ± 0.24	0.147	0.989	0.154
Frontal										
MeanSD (°)	0.86 ± 0.21	0.79 ± 0.18	0.424	0.90 ± 0.22	0.79 ± 0.18	0.81 ± 0.19	0.74 ± 0.15	0.181	0.002	0.296
Adduction (°)	0.66 ± 0.19	0.70 ± 0.29	0.729	0.74 ± 0.22	0.74 ± 0.30	0.59 ± 0.16	0.66 ± 0.29	0.671	0.008	0.423
Hip										
Sagittal										
MeanSD (°)	1.44 ± 0.25	1.15 ± 0.17	0.003	1.41 ± 0.23	1.26 ± 0.24	1.30 ± 0.21	1.22 ± 0.23	0.168	0.028	0.390
Flexion (°)	1.33 ± 0.27	0.98 ± 0.22	0.002	1.35 ± 0.34	1.07 ± 0.25	1.25 ± 0.30	1.06 ± 0.23	0.013	0.324	0.465
Extension (°)	1.14 ± 0.37	0.83 ± 0.19	0.017	1.06 ± 0.31	0.94 ± 0.25	1.01 ± 0.42	0.85 ± 0.21	0.197	0.165	0.702
Frontal										
MeanSD (°)	0.98 ± 0.21	0.89 ± 0.23	0.316	1.03 ± 0.26	0.95 ± 0.21	1.00 ± 0.18	0.91 ± 0.24	0.314	0.287	0.975
Adduction (°)	1.03 ± 0.35	0.84 ± 0.16	0.093	1.10 ± 0.34	0.88 ± 0.15	1.04 ± 0.20	0.90 ± 0.21	0.030	0.604	0.235
Abduction (°)	0.89 ± 0.19	0.90 ± 0.35	0.882	0.93 ± 0.33	0.91 ± 0.24	0.90 ± 0.21	0.90 ± 0.26	0.898	0.731	0.905
Pelvis										
Sagittal										
MeanSD (°)	1.03 ± 0.27	0.82 ± 0.10	0.025	0.96 ± 0.18	0.88 ± 0.17	0.91 ± 0.15	0.86 ± 0.16	0.291	0.098	0.552
Posterior Tilt (°)	0.96 ± 0.28	0.79 ± 0.10	0.061	0.92 ± 0.15	0.84 ± 0.16	0.88 ± 0.19	0.83 ± 0.16	0.297	0.379	0.509
Anterior Tilt (°)	0.90 ± 0.27	0.67 ± 0.11	0.019	0.84 ± 0.19	0.76 ± 0.14	0.82 ± 0.12	0.71 ± 0.20	0.121	0.243	0.623
Frontal										
MeanSD (°)	0.70 ± 0.14	0.55 ± 0.16	0.024	0.69 ± 0.17	0.59 ± 0.21	0.69 ± 0.12	0.61 ± 0.24	0.220	0.819	0.605
Hike (°)	0.76 ± 0.20	0.56 ± 0.15	0.011	0.72 ± 0.15	0.61 ± 0.24	0.72 ± 0.12	0.65 ± 0.21	0.201	0.468	0.507
Drop (°)	0.74 ± 0.26	0.55 ± 0.14	0.040	0.70 ± 0.19	0.59 ± 0.19	0.67 ± 0.10	0.61 ± 0.20	0.172	0.934	0.387
Trunk										
Sagittal										
MeanSD (°)	1.98 ± 0.73	1.46 ± 0.37	0.038	1.75 ± 0.40	1.37 ± 0.39	1.85 ± 0.61	1.58 ± 0.68	0.068	0.174	0.610
mExtension (°)	1.82 ± 0.62	1.35 ± 0.38	0.034	1.61 ± 0.41	1.25 ± 0.38	1.69 ± 0.54	1.44 ± 0.53	0.048	0.159	0.575
Flexion (°)	1.81 ± 0.87	1.28 ± 0.34	0.067	1.60 ± 0.48	1.24 ± 0.38	1.72 ± 0.69	1.44 ± 0.79	0.108	0.219	0.747
Frontal										
MeanSD (°)	1.55 ± 0.85	1.17 ± 0.38	0.176	1.28 ± 0.37	1.14 ± 0.43	1.51 ± 0.60	1.19 ± 0.50	0.179	0.043	0.217
Ipsilateral Flexion (°)	1.55 ± 1.01	1.08 ± 0.40	0.149	1.19 ± 0.44	0.97 ± 0.38	1.37 ± 0.63	1.03 ± 0.42	0.111	0.073	0.399
Contralateral Flexion (°)	1.30 ± 0.22	1.14 ± 0.39	0.237	1.17 ± 0.40	1.17 ± 0.49	1.47 ± 0.61	1.23 ± 0.61	0.524	0.039	0.186
Spatiotemporal										
Double support (%)	1.05 ± 0.19	0.94 ± 0.17	0.156	1.15 ± 0.17	0.93 ± 0.14	1.00 ± 0.21	0.93 ± 0.20	0.014	0.055	0.039
Stance phase (%)	0.97 ± 0.15	0.92 ± 0.18	0.502	1.10 ± 0.18	1.02 ± 0.26	0.90 ± 0.21	0.94 ± 0.16	0.828	0.001	0.151
Step length (cm)	1.64 ± 0.35	1.43 ± 0.22	0.089	1.92 ± 0.46	1.66 ± 0.45	1.69 ± 0.64	1.50 ± 0.31	0.198	0.007	0.636
Step width (cm)	2.02 ± 0.66	1.88 ± 0.60	0.586	2.04 ± 0.62	2.11 ± 0.60	2.01 ± 0.66	2.03 ± 0.63	0.863	0.392	0.634
Stride length (cm)	2.50 ± 0.55	2.07 ± 0.38	0.038	2.72 ± 0.60	2.46 ± 0.67	2.47 ± 1.06	2.28 ± 0.46	0.369	0.081	0.776
Local Dynamic Stability (LDE)										
Mediolateral	0.149 ± 0.027	0.147 ± 0.021	0.800	0.150 ± 0.025	0.151 ± 0.027	0.135 ± 0.026	0.139 ± 0.019	0.714	< 0.001	0.655
Anteroposterior	0.363 ± 0.035	0.388 ± 0.042	0.123	0.377 ± 0.043	0.386 ± 0.058	0.343 ± 0.034	0.351 ± 0.050	0.584	< 0.001	0.936
Vertical	0.651 ± 0.111	0.623 ± 0.077	0.481	0.632 ± 0.101	0.605 ± 0.123	0.645 ± 0.081	0.623 ± 0.114	0.537	0.274	0.863

Note: Values are presented as mean ± standard deviation. **Bold** indicates significant (p < 0.05) difference. meanSD: mean standard deviation of joint angles across the gait cycle. LDE: Local divergence exponent.

excellent. All participants in the control group were categorized as excellent for both hips.

3.3. Group comparison at prescribed speed

The hip pain group had greater variability in their joint kinematics across the gait cycle (i.e., meanSD) and in the peak joint angles (i.e., SD) at the hip, pelvis, and trunk than the control group (Table 2, Figs. 1 and 2). Specifically, greater gait cycle variability and peak joint angle variability were found at the hip in the sagittal plane (p ≤ 0.017), the pelvis in both planes (p ≤ 0.040), and the trunk in the sagittal plane

(p ≤ 0.045). The hip pain group had greater stride length (p = 0.038) variability than controls (Table 2, Fig. 1). Local dynamic stability was not different between groups at the prescribed speed.

3.4. Group comparison at preferred and fast speeds

Compared to the controls, the hip pain group had greater variability in knee, hip, and trunk kinematics across speeds (Table 2, Fig. 3). The hip pain group had greater variability at peak hip flexion angle (p = 0.013) and peak hip adduction angle (p = 0.030). For the sagittal plane trunk kinematics, the hip pain group had greater peak trunk

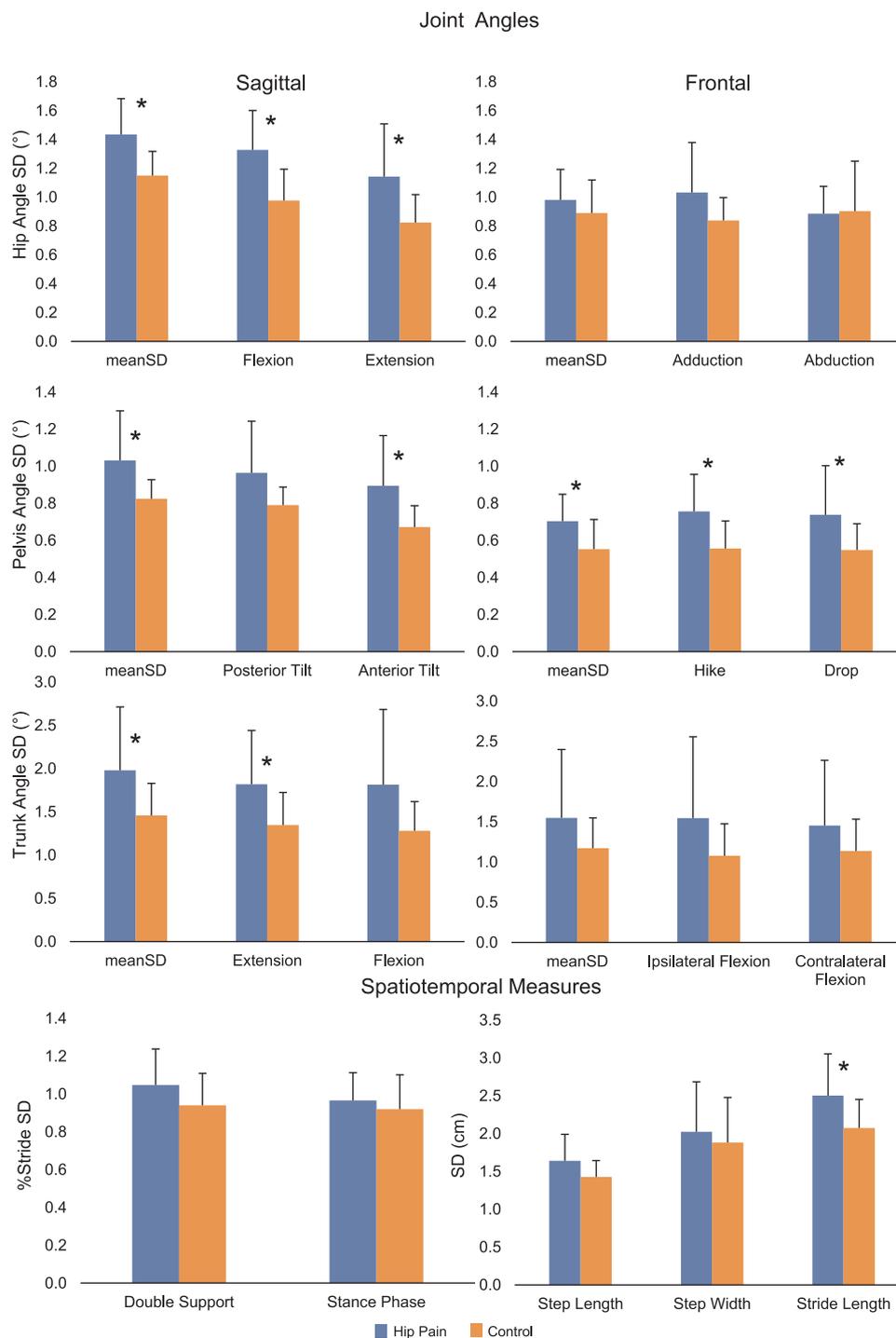


Fig. 1. Kinematic variability of hip, pelvis, and trunk angles across the gait cycle (meanSD), at the peaks (standard deviation, SD) and of spatiotemporal measures (SD) of the hip pain (blue) and control (orange) groups walking at the prescribed speed. *significant ($p < 0.05$) differences between groups (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

extension angle variability ($p = 0.048$). No significant group-by-speed interactions were detected for these main effects. Local dynamic stability was not different between groups at the preferred or fast speeds.

3.5. Speed comparison

Participants had greater variability in the ankle, knee, and hip joint kinematics at their preferred speed compared to their fast speed (Table 2, Fig. 3). Across the gait cycle, variability of sagittal and frontal plane ankle angles ($p = 0.010$, $p = 0.043$) and knee angles

($p < 0.001$, $p = 0.002$), and of sagittal plane hip angles ($p = 0.028$) were greater at the preferred speed compared to the fast speed. At the ankle, peak dorsiflexion variability ($p = 0.028$), plantar flexion variability ($p < 0.001$), and inversion variability ($p < 0.001$) were greater at the preferred speed than at the fast speed. At the knee, peak extension variability ($p = 0.021$) and adduction variability ($p = 0.008$) were greater at the preferred speed than at the fast speed. At the trunk, kinematic variability was greater in the frontal plane at the fast speed compared to the preferred speed across the gait cycle and at peak contralateral flexion ($p \leq 0.043$, Fig. 3). Stance time ($p = 0.001$) and

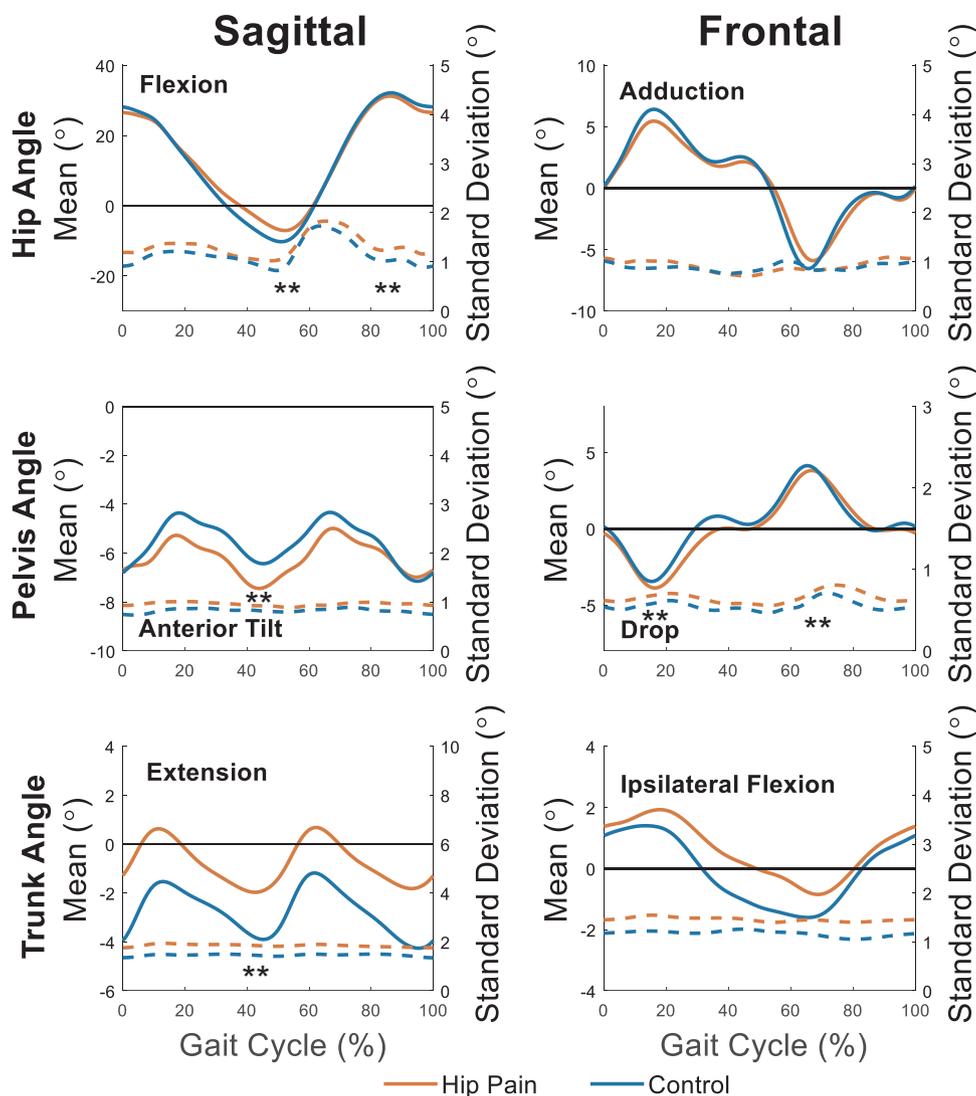


Fig. 2. Group mean kinematic curves normalized to the gait cycle are plotted along the primary axis in solid lines, hip pain in blue and controls in orange. The dashed lines plot the mean of each group’s standard deviation at each point in the gait cycle along the secondary axis. Group differences for meanSD were noted at the hip in the sagittal plane, the pelvis in the sagittal and frontal planes, and the trunk in the sagittal plane. ** denote significant differences in peak SD between groups (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

step length ($p = 0.007$) variability were greater at the preferred speed than at the fast speed (Table 2, Fig. 3). Local dynamic stability was greater (i.e., LDE was smaller) at the fast speed compared to the preferred speed in the ML ($p < 0.001$) and AP ($p < 0.001$) directions (Table 2, Fig. 4).

3.6. Group-by-speed interaction

There was a significant group-by-speed interaction for double support time variability ($p = 0.039$, Table 2, Fig. 3). Double support time variability was greater in the hip pain group compared to the control group at the preferred walking speed ($p < 0.001$). There was no interaction effect for local dynamic stability.

4. Discussion

Individuals with hip pain have greater kinematic variability, but similar local dynamic stability, compared to individuals without hip pain when walking at the same prescribed speed. Kinematic variability and local dynamic stability were both affected by walking speed but in different ways, suggesting that these measures quantify different

aspects of gait mechanics.

4.1. Differences between groups

As hypothesized, individuals with hip pain and a history of DDH exhibited more kinematic variability than controls. Specifically, when walking at the same speed, group differences were concentrated at the hip, pelvis, and trunk in the sagittal and frontal planes (Fig. 2), while stride length was the only spatiotemporal variable affected. Similarly, the hip pain group exhibited increased kinematic variability compared to the control group across the preferred and fast speeds at the hip and trunk, but the effect was not found at the pelvis and the spatiotemporal differences were not significant.

In contrast to our findings, previous studies reported that individuals with knee [7,17] or hip [9] OA have lower kinematic variability of the affected joints compared to controls. Decreased variability of the joints was interpreted as increased rigidity [9,17] and decreased adaptability to perturbations [19] during walking. Our results suggest that prior to OA, increased limb rigidity is not present in individuals with hip pain and their gait patterns are adaptable. Alternatively, increased variability could indicate reduced neuromuscular control of the

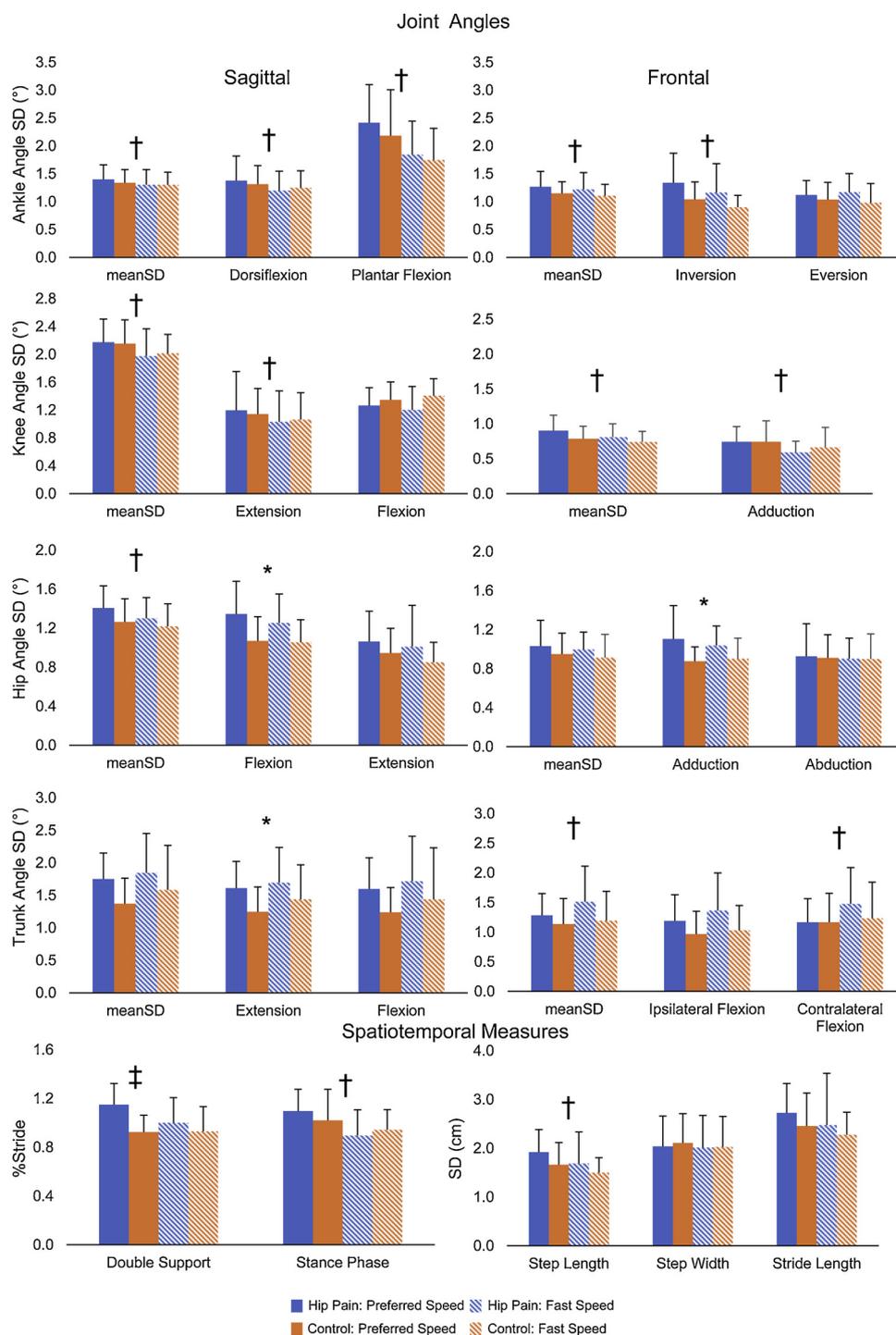


Fig. 3. Kinematic variability of ankle, knee, hip, pelvis, and trunk angles across the gait cycle (meanSD), at the peaks (SD) and of spatiotemporal measures (SD) of the hip pain (blue) and control (orange) groups walking at the preferred (solid bars) and fast (striped bars) walking speeds. *significant ($p < 0.05$) main effect for group; †significant ($p < 0.05$) main effect for speed; *significant interaction of group-by-speed (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

pelvis and hip, highlighted by the concentration of effects at the hip, hip, pelvis, and trunk. Our findings validate, what we have observed clinically, that individuals with hip pain and DDH have more variable gait patterns, especially at the hip and pelvis, compared to individuals without hip pain.

Our finding of increased variability in the spatiotemporal parameter agree with previous literature. Studies report individuals with knee [17] and hip [9,49] OA have greater spatiotemporal variability than controls. Increased spatiotemporal variability may indicate adaptations

are being made to unexpected changes to joint mobility or sudden onset of pain [49] and may be related to the increased variability of joints higher in the kinematic chain. This finding is interesting considering that all participants walked on a treadmill at a constant speed where changes in stride length are not required.

We found no group differences in local dynamic stability of the trunk at any speed. In contrast, differences between individuals with and without diabetic neuropathy have been noted using local dynamic stability analysis of the trunk during gait [21,23]. Previous studies have

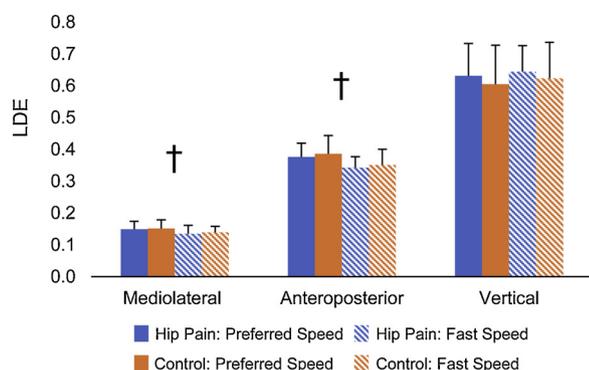


Fig. 4. Local dynamic stability of the trunk of the hip pain (blue) and control (orange) groups walking at the preferred (solid bars) and fast (striped bars) walking speeds. †significant ($p < 0.05$) main effect for speed (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

also found differences in local dynamic stability in individuals with knee OA [7], anterior cruciate ligament deficiency [22], and low back pain [8]. These studies performed the local dynamic stability analysis on the joint or segment angle data that matched the area of involvement. In this case, we used the C7 velocity time series for analysis with the understanding that local dynamic stability of the trunk would be sensitive to small perturbations arising from pain or reduced neuromuscular control at the hip. A previous study concluded that individuals with DDH had increased hip instability during gait, but used peak accelerometer values to quantify instability [6]. The lack of differences in local dynamic stability at the trunk may indicate that hip pain does not impact overall gait stability; instead, kinematic alterations at the hip and pelvis may help maintain overall stability.

As expected, the individuals with hip pain had lower scores on all questionnaires than the control group. Similar to a previous study, our participants had a wide range of self-reported hip pain and function on the mHHS [1], but the majority of our participants were in the poor category. Our mHHS mean was 70.1 which is slightly higher than Romano et al. [1] whose DDH group had a similar range of Harris Hip Scores, but a mean of 61. In that study, the authors reported that hip pain scores correlated with gait pattern changes. Although not reported here, we found that hip pain scores did not significantly correlate with any kinematic variability or local dynamic stability measure. The hip pain group did report greater pain during the fast trial compared to their preferred walking speed. Notably, the average pain ratings reported during the walking trials did not exceed the average or worst pain the individual reported experiencing in the last two weeks. These results suggest that the walking tasks chosen may have not been challenging enough to elicit detectable changes in local dynamic stability due to pain, despite finding increased kinematic variability.

4.2. Effect of speed

In agreement with our hypothesis and previous studies, walking speed affected both kinematic variability [7–9,12,17] and local dynamic stability [8,12,14,16,23]. At their preferred speed, participants were more variable at the ankle, knee, and hip compared to the fast speed. Similar findings were reported for individuals with knee [17] and hip [9] OA when walking above or below a preferred speed. In contrast to previous research [9,17], the participants decreased their spatiotemporal variability at the fast speed while increasing the kinematic variability of the trunk. Previous studies found that trunk variability decreased with increasing speed in individuals with low back pain [8] and in healthy adults [12].

Interestingly, we found that speed affected local dynamic stability of the trunk differently than kinematic trunk variability. The local

dynamic stability results were similar to Bruijn et al. [16] who reported that local dynamic stability increased with increasing speed in healthy adults. The authors highlight that their (and our) findings contrast those of many previous studies [21,23,50] that suggest that local dynamic stability increases with decreasing walking speed. However, these differences are most likely due to past researchers not controlling the length of time series analyzed [16], which has been shown to change local dynamic stability outcomes [42].

4.3. Kinematic variability versus local dynamic stability

Although both kinematic variability and local dynamic stability have been used to quantify gait stability, care must be taken in how that stability is interpreted. This argument is highlighted by the fact that in the current study and previous studies [12,16,21,23], kinematic variability and local dynamic stability measures [15] (e.g., Lyapunov exponents) quantify fundamentally different aspects of gait behavior. In this case, the increased kinematic variability quantified adaptations made at the joint level to alleviate pain or indicate reduced neuromuscular control around the hip. Still, there may be cases where variability should not be completely diminished to enable adaptability of gait patterns [19]. The lack of differences in local dynamic stability between groups indicate that individuals with hip pain were able to attenuate and recover from small perturbations (i.e., pain) occurring at the hip which did not affect overall gait stability.

4.4. Limitations

Some study limitations should be noted. The hip pain group was heterogeneous in terms of side involvement and surgical history, which may have influenced the effect of group. We did perform a sensitivity analysis where individuals with unilateral pain and bilateral pain were compared to individuals without hip pain separately. This analysis, while not expected to reach significance, did indicate that individuals with either unilateral or bilateral pain had increased kinematic variability at the hip, pelvis, and trunk compared individuals without hip pain.

The groups were matched by activity level, which may have contributed to the lack of differences found for local dynamic stability between groups. Individuals with hip pain who are less active may not be able to maintain gait stability in the presence of hip pain. Additionally, the gait speeds were not randomized and may have influenced pain levels at the fast speed. We first captured typical gait parameters at a preferred speed before moving to a more difficult walking speed. We used the C7 marker, and not a marker closer to the hip, for the local dynamic stability analysis. The lack of differences found in this study indicated that hip pain may not affect overall stability or that individuals adapted their gait to maintain their stability.

Finally, the lack of differences in local dynamic stability may be due walking on a treadmill. Some research suggests that walking on a treadmill artificially increases local dynamic stability and decreases variability distally due to constraints of the task [15,21]. However, a recent methodological study suggests that this finding of increased stability may have been erroneous due to differences in the time series length between the overground and treadmill tasks analyzed [24]. We did find differences in kinematic variability of distal variables (i.e. spatiotemporal) which suggests that while variability may be less on a treadmill than overground, detectable differences were maintained between groups. Moreover, by using treadmill walking, we were able to ensure that we were detecting changes in variability due to hip pain or DDH rather than variability due to changes in instantaneous walking speed that can occur during overground walking. Given the aims of this study, long continuous time series obtained at a constant speed were required to evaluate kinematic variability and local dynamic stability simultaneously.

5. Conclusions

This study is the first to report that individuals with hip pain had greater kinematic variability at the hip, pelvis, and trunk than controls, which may be related to the greater step and stride length variability. While this increased variability may indicate reduced or altered neuromuscular control at the hip, it may also be a pain alleviation strategy, in which case, rehabilitation to improve motor control of the hip may decrease pain and decrease variability. The local dynamic stability of the trunk was not different between groups, suggesting that hip pain may not affect overall gait stability. Speed had different effects on kinematic variability and local dynamic stability, emphasizing that these measures quantify different aspects of walking behavior. Finally, this study validates anecdotal evidence, observed clinically, that individuals with hip pain and DDH have more variable gait patterns.

Conflict of interest

The authors have no conflicts of interest to disclose.

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