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Walking patterns and hip contact forces in patients with hip dysplasia

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ABSTRACT

Introduction: Several studies have investigated walking characteristics in hip dysplasia patients, but so far none have described all hip rotational degrees of freedom during the whole gait cycle.

This descriptive study reports 3D joint angles and torques, and furthermore extends previous studies with muscle and joint contact forces in 32 hip dysplasia patients and 32 matching controls.

Methods: 3D motion capture data from walking and standing trials were analysed. Hip, knee, ankle and pelvis angles were calculated with inverse kinematics for both standing and walking trials. Hip, knee and ankle torques were calculated with inverse dynamics, while hip muscle and joint contact forces were calculated with static optimisation for the walking trials.

Results: No differences were found between the two groups while standing. While walking, patients showed decreased hip extension, increased ankle pronation and increased hip abduction and external rotation torques. Furthermore, hip muscle forces were generally lower and shifted to more posteriorly situated muscles, while the hip joint contact force was lower and directed more superiorly.

Conclusion: During walking, patients showed lower and more superiorly directed hip joint contact force, which might alleviate pain from an antero-superiorly degenerated joint.

1. Introduction

Hip dysplasia (HD) is characterised by reduced lateral and sometimes also anterior acetabular cover of the femoral head [1,2]. Patients with symptomatic HD experience persistent hip pain [1] and degenerative changes are often observed antero-superiorly in the acetabulum [3,4]. Patients probably change their gait patterns to direct the hip joint contact force (JCF) away from the antero-superior part of the acetabulum, as this would decrease stress on structures susceptible to damages and therefore also reduce their pain. Several studies have suggested various pain avoidance strategies during walking, e.g., [2,5,6].

Conversely, changed gait patterns might direct the JCF towards the antero-superior part of the acetabulum considered prone to overloading [2], explaining the hip pain experienced by patients.

The magnitude and direction of the hip JCF are affected by muscle forces and orientation of the pelvis relative to the femur [7,8]. For instance, posterior pelvic tilt has been shown to decrease the acetabular cover of the femoral head and concentrate the load on the anterior part of the acetabulum [9]. In dysplastic hips, the decrease in cover, when the pelvis is posteriorly tilted, is larger than in normal hips [10,11]. Furthermore, increased acetabular cover achieved by listing the pelvis towards the affected side has been observed in some HD patients with small degenerative changes in the acetabulum [10].

Several studies have investigated the gait patterns in HD patients, e.g., [5,6,12–14], reporting changes in joint angles and torques. Despite using 3D motion capture, HD studies often only present hip angles and torques in the sagittal plane, e.g. [5,6,14], thereby excluding two of three rotational degrees of freedom of the hip joint.

To our knowledge, no previous studies have covered both joint angles and torques for all rotational degrees of freedom of the hip joint in HD patients.

Furthermore, speculations about the pelvis load are sometimes based on observed joint torques, e.g., [2,5], and angles [2], implying that some gait alterations might reduce the load on the antero-superior part of the acetabulum, e.g., [2,5], while others might

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increase the load [2]. To our knowledge, no previous studies on gait patterns in HD patients have investigated hip muscle forces and JCF.

The purpose of this study is to take HD gait analysis one step further by including hip muscle forces and JCFs. Results will be presented in all three dimensions to investigate if changes occur in dimensions not reported in previous studies.

2. Methods

The raw motion capture data previously collected and analysed by Jacobsen et al. [6] was used to extend the calculations to include muscle forces and JCFs. A detailed description of subjects and laboratory procedures can be found in Jacobsen et al. [6]. In short, the study comprised 32 patients (26 women) with symptomatic HD (uni- or bilateral) scheduled for pelvic surgery, indicated radiologically by a centre edge (CE) angle $<25^\circ$ [15] and subjectively by hip pain and functional limitations revealed by detailed questionnaire surveys. Median and range of the patients' CE and Tönnis' acetabular index angles were 18° (4–22) and 14° (10–22), respectively, and the patients reported onset of symptoms 1–5 years prior to surgery. The study further comprised 32 matching controls (patients vs. controls, median (range): 34 (18–53) vs. 33 (18–54) years, 66 (41–93) vs. 63 (37–100) kg, 1.71 (1.52–1.84) vs. 1.71 (1.51–1.94) m). The rather large ranges in height and weight are due to patients being included as they are scheduled for operation, but we do not think this influences the result.

All participants performed a normal upright standing trial and a number of walking trials on an 8 m walkway instrumented with 8 MCU-1000 ProReflex cameras (Qualisys, AB, Gothenburg, Sweden) and an OR6-7 AMTI (Advanced Mechanical Technology, Watertown, MA) force plate. The entire walkway was covered with a thin carpet to conceal the position of the force plate. Kinematic and force data were sampled at 240 Hz and 960 Hz, respectively. Walking trials were repeated until at least 3 clean force plate hits with both left and right foot were recorded. All trials were performed with bare feet. Automatic tracking was facilitated by 38 reflective markers placed on anatomical landmarks and as clusters on rigid plates on pelvis and lower extremities. The camera system was calibrated to residual errors less than 2.5 mm over a recording volume of approximately 3 m $\times$ 1 m $\times$ 1.3 m ($L \times W \times H$).

Walking trials were analysed using OpenSim 3.2 [16]. The musculo-skeletal model used was the generic OpenSim model Gait 2392 which has 23 degrees of freedom and 92 muscles [17]. Gait 2392 has lower extremities, pelvis and torso. Virtual metatarsal markers were associated with the calcaneus, thereby eliminating movement of the metatarsophalangeal joint. The model was scaled to match the size and strength of each subject based on marker measurements and bodyweight. Since position of the torso had not been recorded, all mass and inertia properties for the torso were reduced to 1×10^{-5} to make sure analyses would not be affected by a faulty torso position; the reaction force and moment from supra-pelvic segments thereby missing were replaced by a an equivalent (hence not affecting final results) residual force and moment added to the pelvis centre of mass by OpenSim.

Joint angles and torques were calculated using OpenSim's inverse kinematics and inverse dynamics (ID) tool, respectively. Individual muscle forces were calculated using OpenSim's static optimisation (SO) tool, which minimises the sum of squared activations while being constrained by muscle force–velocity and force–length properties. Ideal actuators were added to knee, ankle, subtalar and metatarsophalangeal joints to make the model more robust for SO. Muscle forces were then used to calculate hip JCF.

Kinematic results were filtered at 6 Hz before applying the ID and SO, using OpenSim's 3rd order Butterworth lowpass filter. Median values for one leg from each subject for the whole gait cycle were used to calculate mean curves for each group. From the patients, legs scheduled for operation were used, while a random leg was chosen from each of the controls.

Joint torques were normalised to body mass, while muscle and JCFs were normalized to body mass$^{2/3}$ [18].

Parameters used to describe and compare the two groups were peak values for joint angles, torques and hip muscle forces, as well as peak JCF magnitude and its direction at this instance. Differences in joint angles, torques or forces for longer periods, were also evaluated.

2.1. Statistics

Data were tested for normality by the Shapiro–Wilk test. Differences between patients and controls were tested with two-tailed *t*-tests when data in both groups were normally distributed, otherwise with Mann–Whitney *U* tests, using 0.05 α-level. All statistical tests had power >0.9.

2.2. Ethics

The study followed the principles laid down in the Declaration of Helsinki and was approved by the Central Denmark Region Committees on Biomedical Research Ethics on September 30, 2010 (M-20100206). All subjects gave informed consent prior to participation.

3. Results

Graphs in this section show the parameters for the whole gait cycle. The gait cycle starts with initial contact (IC) at 0% and finishes right before next IC at 100%. The swing phase starts approximately 65% into the gait cycle. The grey lines in the figures illustrate $\pm$ 1 SD. Only hip joint results are presented graphically, while other significant differences between the two groups are presented in the text. Punctual index for Table 1 and Supplementary Table 2 on the journal website is at 52% into the gait cycle when peak JCF magnitude occurred in both groups.

3.1. Standing trials

There were no differences between the poses of the two groups. A table with joint angles from the standing trials can be found as Supplementary data at the journal website.

3.2. Walking trials

3.2.1. Joint angles

Compared to the controls, patients had decreased peak hip extension ($p = 0.04$, Fig. 1), peak plantarflexion at toe-off (TO) ($p = 0.04$) and ankle supination during propulsion ($p = 0.04$).

3.2.2. Joint torques

Patients had increased hip abduction and external rotation torques during propulsion (60% into the gait cycle) compared to controls ($p = 0.02$ and $p = 0.01$, respectively). Controls made a small internal rotation torque right before TO whereas patients kept an external rotation torque (Fig. 2).

3.2.3. Muscle forces

Three of the 18 muscles crossing the hip joint are displayed to represent the differences in muscle forces between the two groups. In general, peak muscle forces were lower in the subjects than in the controls and there was a tendency for curves to be flatter (Fig. 3). The second gluteus medius peak, occurring during