

Biomechanical characterization and clinical implications of artificially induced crouch walking: Differences between pure iliopsoas, pure hamstrings and combination of iliopsoas and hamstrings contractures

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Abstract

The purpose of this study was to characterize biomechanically three different crouch walking patterns, artificially induced in eight neurologically intact subjects and to compare them to selected cases of pathological crouch walking. The subjects were equipped with a lightweight mechanical exoskeleton with artificial muscles that acted in parallel with hamstrings and iliopsoas muscles. They walked at a speed of approximately 1 m/s along the walkway under four experimental conditions: normal walking (NW), hamstrings contracture emulation (HAM), iliopsoas contracture emulation (IPS) and emulation of both hamstrings and iliopsoas contractures (IPSHAM). Reflective markers and force platform data were collected and ankle, knee and hip-joint angles, moments and powers were calculated. HAM and IPSHAM shifted ankle-angle rotation profiles into dorsiflexion during midstance compared to IPS and NW where ankle-angle trajectories were similar. HAM, IPS and IPSHAM shifted the knee angle of rotation profiles into flexion during stance, compared to NW. IPS and IPSHAM shifted hip angle of rotation profiles toward pronounced flexion while HAM shifted hip angle of rotation profile toward extension, compared to NW. HAM and IPSHAM significantly increased ankle moment during midstance, compared to IPS and NW where ankle moment profiles were similar. All experimental conditions exhibited similar behavior in the knee-moment profiles during midstance while IPS and IPSHAM knee-moment profiles exhibited significantly higher knee-extension moment during terminal stance and pre-swing. In the hip joint all experimental conditions exhibited similar shape of hip moment profiles throughout the gait cycle. HAM and IPS kinematic and kinetic patterns were qualitatively compared to two selected clinical cases, showing considerable similarity. This implies that distinct differences in kinematics and kinetics between HAM, IPS and IPSHAM may be clinically relevant in helping determine the relative contribution of hamstrings and iliopsoas muscles contractures to particular crouch walking.

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1. Introduction

Crouch gait is one of the most prevalent walking patterns in children with cerebral palsy (Wren et al., 2005), characterized with increased and excessive knee flexion, increased ankle dorsiflexion and hip trajectory ranging from almost normal to excessively flexed throughout the gait cycle (Winters et al., 1987; Lin

et al., 2000; Rodda et al., 2004). Crouch gait can develop as a consequence of surgical lengthening of Achilles tendon or due to weakness of triceps surae (Chambers, 2001; Gage 2004). However, in most cases crouch gait originates from contractured/tightened hamstrings, which are often combined with hip flexors deformity (Gage, 2004). Several studies investigating muscle-tendon lengths and lengthening velocities in hamstrings (Arnold et al., 2006; Schutte et al., 1997) and iliopsoas muscles (Schutte et al., 1997) have suggested that in many cases of crouch gait in patients with cerebral palsy

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the hamstrings were of normal length while the iliopsoas muscles were short. Surgical lengthening of hamstrings is a common treatment of crouch gait. While some patients achieve considerable improvement following the surgery, many others show no improvement or even get worse (Gage, 2004; Chambers, 2001). Often this is because of the erroneous identification of primary cause of crouch due to difficulties in establishing dynamic hamstring length on physical examination (Gage, 2004). There are cases of crouch gait where hamstrings appear to be shortened, while they are actually of normal length, because their origin is shifted proximally due to excessive anterior pelvic tilt and accompanying increased hip flexion that can be due to contracture of hip flexors or adopted trunk lean forward (Schutte et al., 1997). This necessitates further studies into mechanisms of crouch gait.

We have recently proposed a method to artificially emulate muscle contracture of selected leg muscles in neurologically intact individuals (Olenšek et al., 2005). We have successfully applied the above methodology to biomechanically characterize toe walking (Matjačić et al., 2006). Using the same methodology, biomechanical characterization of crouch gait can also be derived. Our hypothesis is that different cases of artificial alterations of hamstrings and iliopsoas muscles' stiffness/tightness will result in distinct changes of kinematic and kinetic gait patterns of artificially induced crouch gait that may resemble recorded kinematic and kinetic patterns in selected CP cases.

The objective of this paper was to experimentally investigate the above hypothesis and to determine differences in kinematic and kinetic gait patterns in joints of artificially constrained lower extremity in a group of neurologically and orthopaedically intact individuals. Differences originated from artificially induced increase in stiffness of (i) hamstrings muscle, (ii) iliopsoas muscle and (iii) combination of hamstrings and iliopsoas muscles that forced the tested subjects into crouch gait. We further compared obtained kinetic patterns to selected CP cases in order to establish possible implications for clinical practice.

2. Methods

2.1. Subjects

Eight male volunteers with no known neurologic or orthopaedic disorders (age 25.2 ± 3.5 years, height 175.5 ± 4.8 cm, mass 64.6 ± 3.3 kg) participated in the study. The experimental protocol was approved by the local ethics committee and the subjects signed informed consent forms. Additionally, the data from two cerebral palsied children (CP#1—age 15 years and CP#2—age 13 years) diagnosed with spastic diplegia and who did

not undergo previous surgery were included. Both cases were selected based on the clinical examination where tightness of hamstring and iliopsoas muscles was indicated.

2.2. Muscle contracture emulation system

Fig. 1 shows schematics and photographs of the mechanical system that was used to artificially emulate contractures of iliopsoas and hamstrings muscles. Plastic cuffs embraced pelvis, right thigh and right shank of a walking subject. Muscle contractures were emulated by means of artificial muscles that consisted of elastic bands, which were connected to the respective attachment points on the plastic cuffs through rigid bands in such a way as to act in parallel with hamstrings and iliopsoas muscles. The attachment points were selected in such a way that the action of artificial muscles was confined to sagittal plane. Lengths and stiffness of elastic bands and lengths of rigid bands were selected after careful experimentation where we aimed to achieve noticeable effects on changes of posture in the hip and the knee during standing and walking while at the same time elastic behavior of the artificial muscles was ensured. Artificial muscles emulating the iliopsoas contracture consisted of a 4.5 cm long rigid band connected to the pelvic cuff, a 5 cm long rigid band connected to the thigh cuff and an 8.5 cm long elastic band exhibiting 3300 N/m stiffness. The artificial muscles emulating the hamstrings contracture consisted of a 13.5 cm long rigid band connected to the pelvic cuff, a 17.3 cm long rigid band connected to the shank cuff and an 13.3 cm long elastic band exhibiting 2000 N/m stiffness. The muscle contracture emulation system produced repeatable data, similar to repeatability shown in the system used in our previous studies on artificially induced toe walking (Matjačić et al., 2006; Olenšek et al., 2005).

2.3. Experimental conditions

Subjects walked across a 10-m-gait laboratory walkway under several experimental conditions: normal walking (NW), hamstrings contracture emulation (HAM), iliopsoas contracture emulation (IPS) and emulation of both hamstrings and iliopsoas contractures (IPSHAM). A VICON motion capture and analysis system (VICON 370, Oxford Metrics Ltd., Oxford, UK) was used to capture motion of lower limbs and pelvis. Reflective markers were attached to the subjects' skin over designated landmarks according to the specifications provided by manufacturer of the system (Vicon Clinical Manager). Motion data were sampled at 50 Hz. Two AMTI force plates (AMTI OR-6-5-1000, Advanced Mechanical Technology Inc., Watertown, MA) that were positioned in the center of a walkway were

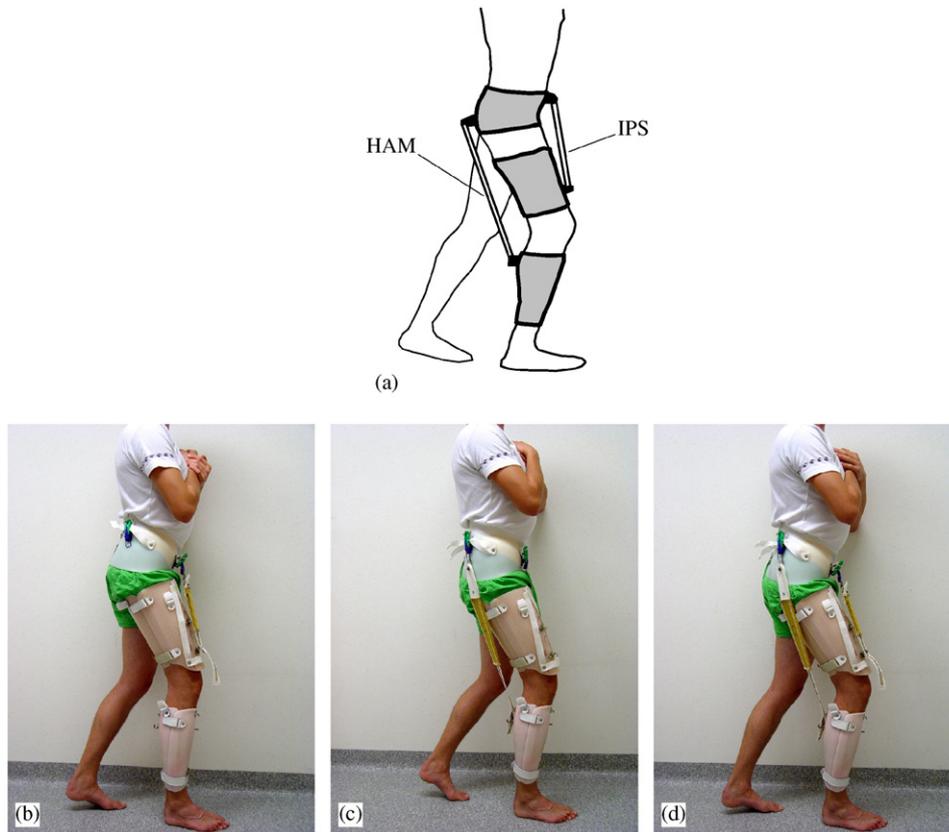


Fig. 1. a. Schematic presentation of a walking subject, wearing pelvic, thigh and shank cuffs upon which two artificial muscles, one emulating iliopsoas (IPS) muscle contracture and the other hamstrings (HAM) muscle contracture, are attached. b. IPS applied. c. HAM applied. d. IPS and HAM applied.

used for recording ground reaction forces. Force data were sampled at 1000 Hz. During the experiment the subjects were monitored to maintain walking speed of approximately 1 m/s. At least four clear steps of the right leg were captured for analysis. Before data capture, the subjects practiced walking for several minutes for each experimental condition. Between experimental conditions the subjects rested for 10 min. For the purpose of qualitative comparison the data from two cerebral palsied children (CP#1 and CP#2) obtained in the same laboratory in unconstrained conditions were included. The data from selected clinical cases are represented by ensemble average of four clear steps.

2.4. Data analysis

Gait velocity, stride length and cadence data were extracted and sagittal ankle, knee and hip angles of rotation, joint moments and powers were calculated for each experimental condition. Joint moments and powers were normalized for body mass and reported in Nm/kg and W/kg, respectively. For each subject the averaged values from four trials were calculated and used in subsequent averaging and statistical analysis of the data

for the whole group separately for each experimental condition. Gait cycle terminology as introduced by Perry (Perry, 1992) was adopted to define instants of characteristic peak values of kinematic and kinetic trajectories in the gait cycle.

2.5. Statistics

Statistical analysis was performed for selected characteristic peaks of the stance subphase (0–10% loading response, 10–30% midstance, 30–50% terminal stance and 50–60% preswing) in the joint angles of rotation, moments and powers obtained under all four experimental conditions. Within each condition, each subject's average of four trials was used. Within each of the stance subphases one-way ANOVA was performed on either maximal or minimal values of each subject's averaged data with factor being one of the tested experimental conditions. Bonferroni-adjusted post-hoc pairwise comparisons were made when a main effect or interaction was detected. $P < 0.05$ was regarded as statistically significant. All values were presented as mean and standard deviation. Similarly data on gait velocities, stride lengths and cadences were statistically examined in one-way ANOVA.

3. Results

3.1. Temporal gait characteristics

The data on gait velocities, stride lengths and cadences are reported in Table 1. Statistically significant difference occurred only in stride lengths, which were decreased in IPS, HAM and IPSHAM experimental conditions. However, these differences were small and correlated well with differences in gait velocities.

3.2. Kinematics and kinetics

Fig. 2 shows averaged kinematic and kinetic patterns across the subjects ($N = 8$) for the right leg and for all experimental conditions.

Ankle angle of rotation and moment profiles were similar for NW and IPS and differed from HAM and IPSHAM, which exhibited increased dorsiflexion and increased plantarflexion moment throughout majority of stance phase. All experimental conditions exhibited increased knee flexion throughout stance phase, and most notably IPSHAM. IPSHAM and HAM exhibited similar knee flexion during loading response while IPS and HAM exhibited similar knee flexion during midstance and terminal stance. NW and HAM exhibited similar knee-moment profiles throughout the entire gait cycle and differed from IPS and IPSHAM, which increased knee-extension moment throughout midstance, terminal stance and pre-swing. Hip-angle profiles were similar for NW and HAM and differed from IPS and IPSHAM, which exhibited increased flexion throughout the entire gait cycle. Hip-moment profiles were similar in all experimental conditions. Power profiles in ankle, knee and hip joints were similar in all experimental conditions. Values of selected characteristic peaks and the results of statistical analyses are given in Table 2.

Fig. 3a shows kinematic and kinetic patterns of a selected case of diplegic child CP#1 (speed of walking 0.8 m/s, cadence 115 steps/min) along with the NW and HAM data. Ankle, knee and hip angle of rotation profiles in CP#1 exhibited more pronounced shift

toward flexion as compared to HAM. Ankle- and knee-moment profiles exhibited similar shapes, specifically similarities in the shape of ankle-moment profiles (increased plantarflexion moment throughout loading response, midstance and terminal stance) are evident. The second knee-extension moment peak during terminal stance and pre-swing is much larger, as compared to HAM experimental condition, which was likely either due to substantially larger extent of knee flexion in the selected CP#1 case as compared to HAM, or plantarflexor weakness. In walking of the CP#1 case we can also observe prolonged hip-extension moment during stance phase and associated increase in hip-power generation, which compensates for weakened push off (Gage, 2004). Fig. 3b illustrates kinematic and kinetic patterns of CP#1 gait and one of the healthy subjects in five instants of stance phase. We observed similar postures of pelvis, hip, knee and ankle. Also, the center of pressure (COP), direction and magnitude of ground-reaction forces (GRF) in particular instants are visually similar. Thus, the qualitative comparison of HAM and CP#1 walking patterns show considerable similarity and suggests hamstrings contracture/tightness in selected clinical case.

Fig. 4a shows kinematic and kinetic patterns of a selected case of diplegic child CP#2 (speed of walking 0.8 m/s, cadence 110 steps/min) along with the NW and IPS data. Ankle, knee and hip angle of rotation profiles in CP#2 exhibited more pronounced shift toward flexion as compared to IPS. Ankle- and knee-moment profiles exhibit similar shapes; specifically similarities in the shape of knee moment at the end of midstance (abrupt stop of otherwise normal cessation of knee-extension moment, which normally turns into slight knee-flexion moment in the terminal stance) are evident. Fig. 4b illustrates kinematic and kinetic patterns of CP#2 gait and one of the healthy subjects in five instants of stance phase. We can observe similar postures of pelvis (inclined forward), hip, knee and ankle. Also the COP, direction and magnitude of GRF in particular instants are similar. Thus, the qualitative comparison of IPS and CP#2 walking patterns show considerable similarity and suggests iliopsoas contracture/tightness in selected clinical case.

Table 1
Temporal gait characteristics for all four experimental conditions ($N = 8$)

	NW	HAM	IPS	IPSHAM	<i>P</i> value ANOVA	Post-hoc main interactions (<i>P</i> value)
Gait velocity (m/s)	1.03 (0.05)	1.01 (0.08)	1.02 (0.08)	1.01 (0.09)	0.102	—
Stride length (m)	1.29 (0.05)	1.18 (0.08)	1.20 (0.05)	1.16 (0.11)	0.015*	NW–HAM (0.046*). NW–IPSHAM (0.020*)
Cadence (1/s)	100.7 (6.81)	101.2 (8.65)	95.6 (9.91)	103.0 (7.18)	0.331	—

Given are mean values and standard deviations. Statistically significant differences between respective experimental conditions are denoted with *. * $p < 0.05$.

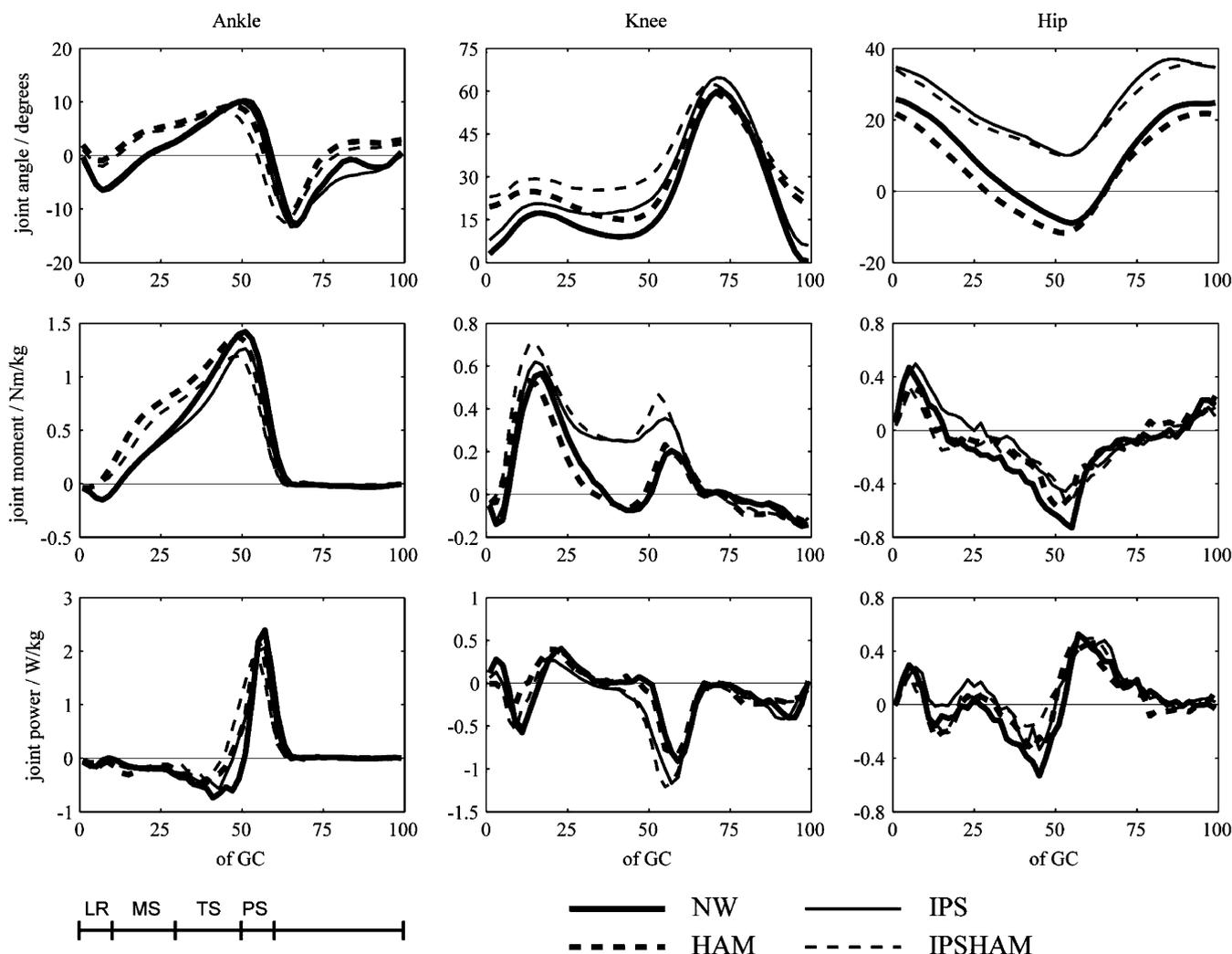


Fig. 2. Averaged kinematic and kinetic gait patterns for all four experimental conditions. Stance phase of a GC—gait cycle is divided into LR—loading response, MS—midstance, TS—terminal stance and PS—preswing (Perry, 1992). Positive joint angle values indicate flexion/dorsiflexion, positive joint-moment values indicate extension/plantarflexion moments and positive joint power values indicate power generation.

Fig. 5 illustrates kinematic and kinetic patterns of NW and IPSHAM in five instants of stance phase.

4. Discussion

The objective of this study was to investigate biomechanical differences between three artificially induced crouch-walking patterns. The results of our study show characteristic differences in gait patterns for the three possible causes of crouch, which may be used in determining of the involvement of iliopsoas and hamstrings in a particular patient when the problem originates primarily because of the dynamic or static contracture of these two muscle groups. Findings of the present study can help clinicians to more reliably determine the degree of involvement of iliopsoas and hamstrings muscle groups in crouch-walking pattern,

thereby facilitating correct decisions regarding treatment.

4.1. Biomechanical characterization

Hamstrings are biarticular muscles that exert hip extension and knee-flexion moments. Therefore, contracture of hamstrings can work to extend the hip and flex the knee, which is supported also by experimental data from the present study. Simultaneously, the ankle joint needs to shift into more dorsiflexion to allow initial contact with the heel. Positions of the hip and the knee determine when the contracture of hamstrings has the largest influence on gait patterns, which appears to be around the initial contact when the hip is in the most flexed position. Increased knee flexion during loading response and midstance would normally require an increase in knee-extension moment. However, our

Table 2
Peak biomechanical variables for all four experimental conditions ($N = 8$)

	NW	HAM	IPS	IPSHAM	<i>P</i> value ANOVA	Post-hoc main interactions (<i>P</i> value)
Midstance, ankle dorsi flexion angle (degrees)	3.02 (2.84)	6.66 (3.15)	3.48 (2.97)	6.25 (2.32)	0.029*	—
Pre-swing, ankle dorsi flexion angle (degrees)	10.31 (2.56)	8.43 (3.29)	9.63 (3.33)	5.42 (6.23)	0.107	—
Loading response, knee-flexion angle (degrees)	15.01 (4.19)	24.88 (6.79)	19.13 (5.73)	28.93 (7.02)	<0.0001*	NW–HAM (0.017*), NW–IPSHAM (<0.0001*), IPS–IPSHAM (0.018*)
Midstance, knee-flexion angle (degrees)	17.36 (4.01)	25.43 (6.34)	20.98 (4.87)	29.78 (6.45)	0.001*	NW–HAM (0.041*), NW–IPSHAM (0.001*), IPS–IPSHAM (0.021*)
Terminal stance, knee-flexion angle (degrees)	8.55 (2.95)	14.33 (3.88)	16.44 (5.59)	25.1 (6.53)	<0.0001*	NW–IPS (0.021*), NW–IPSHAM (10.0001*), HAM–IPSHAM (0.001*), IPS–IPSHAM (0.009*)
Pre-swing, knee-flexion angle (degrees)	36.25 (4.55)	42.86 (3.38)	44.26 (5.94)	60.55 (7.68)	<0.0001*	NW–IPS (0.049*), NW–IPSHAM (<0.0001*)
Loading response, hip-flexion angle (degrees)	25.74 (4.73)	21.69 (6.92)	34.93 (4.14)	33.98 (3.87)	<0.0001*	NW–IPS (0.007*), NW–IPSHAM (0.018*), HAM–IPS (<0.0001*), HAM–IPSHAM (<0.0001*)
Midstance, hip angle (degrees)	3.82 (6.08)	−1.60 (7.28)	18.65 (5.41)	16.96 (3.85)	<0.0001*	NW–IPS (<0.0001*), NW–IPSHAM (0.001*), HAM–IPS (<0.0001*), HAM–IPSHAM (<0.0001*)
Terminal stance, hip angle (degrees)	−7.89 (6.18)	−11.59 (7.20)	10.57 (5.20)	10.05 (4.31)	<0.0001*	NW–IPS (<0.0001*), NW–IPSHAM (<0.0001*), HAM–IPS (<0.0001*), HAM–IPSHAM (<0.0001*)
Pre-swing, hip angle (degrees)	−9.03 (6.19)	−11.90 (7.27)	9.83 (5.32)	9.73 (4.61)	<0.0001*	NW–IPS (<0.0001*), NW–IPSHAM (<0.0001*), HAM–IPS (<0.0001*), HAM–IPSHAM (<0.0001*)
Loading response, ankle dorsiflexion moment (Nm/kg)	−0.17 (0.09)	−0.07 (0.05)	−0.16 (0.08)	−0.10 (0.05)	0.024*	—
Midstance, ankle plantarflexion moment (Nm/kg)	0.60 (0.14)	0.88 (0.11)	0.52 (0.28)	0.78 (0.16)	0.002*	NW–HAM (0.028*), HAM–IPS (0.003*)
Pre-swing, ankle plantarflexion moment (Nm/kg)	1.42 (0.16)	1.33 (0.20)	1.26 (0.18)	1.13 (0.28)	0.072	—
Loading response, knee moment (Nm/kg)	−0.15 (0.08)	−0.07 (0.05)	0.13 (0.84)	0.07 (0.06)	0.038*	—
Terminal stance, knee-extension moment (Nm/kg)	0.10 (0.17)	0.11 (0.17)	0.18 (0.12)	0.21 (0.18)	<0.0001*	NW–IPS (0.012*), NW–IPSHAM (0.005*), HAM–IPS (0.008*), HAM–IPSHAM (0.003*)
Pre-swing, knee-extension moment (Nm/kg)	0.25 (0.12)	0.28 (0.11)	0.41 (0.10)	0.50 (0.15)	0.001*	NW–IPSHAM (0.003*), HAM–IPSHAM (0.008*)
Loading response, hip-flexion moment (Nm/kg)	0.55 (0.21)	0.39 (0.21)	0.56 (0.18)	0.35 (0.11)	0.058	—
Terminal stance, ankle-power absorption (W/kg)	−0.88 (0.19)	−0.59 (0.29)	−0.63 (0.31)	−0.55 (0.24)	0.074	—
Loading response, knee-power generation (W/kg)	0.35 (0.17)	0.09 (0.10)	0.22 (0.16)	0.05 (0.07)	<0.0001*	NW–IPS (0.003*), NW–IPSHAM (0.000*)
Midstance, knee-power absorption (W/kg)	−0.58 (0.22)	−0.10 (0.21)	−0.56 (0.30)	−0.30 (0.23)	0.001*	NW–IPS (0.003*), HAM–IPS (0.004*)
Terminal stance, knee-power absorption (W/kg)	−0.20 (0.17)	−0.37 (0.27)	−0.61 (0.32)	−0.73 (0.35)	0.004*	NW–IPS (0.044*), NW–IPSHAM (0.005*)
Terminal stance, hip-power absorption (W/kg)	−0.59 (0.18)	−0.52 (0.30)	−0.41 (0.21)	0.25 (0.10)	0.015*	NW–IPSHAM (0.016*)

Given are mean values and standard deviations. Statistically significant differences between respective experimental conditions are denoted with *. * $p < 0.05$.

experimental data (HAM) show that the peak knee-extension moment during midstance is reduced as compared to NW. This is due to increased plantar-

flexion moment in the ankle joint, which is likely due to increased soleus activity because soleus dynamically acts to accelerate the knee into extension

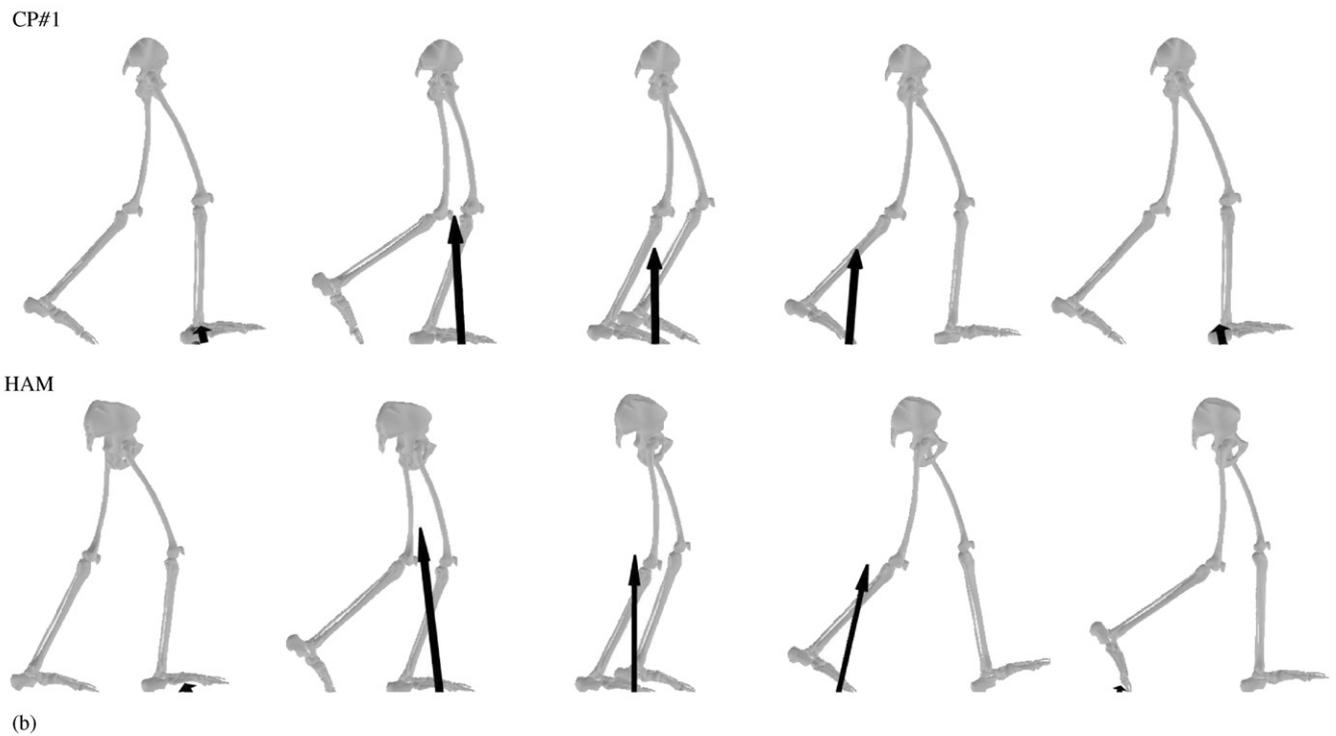
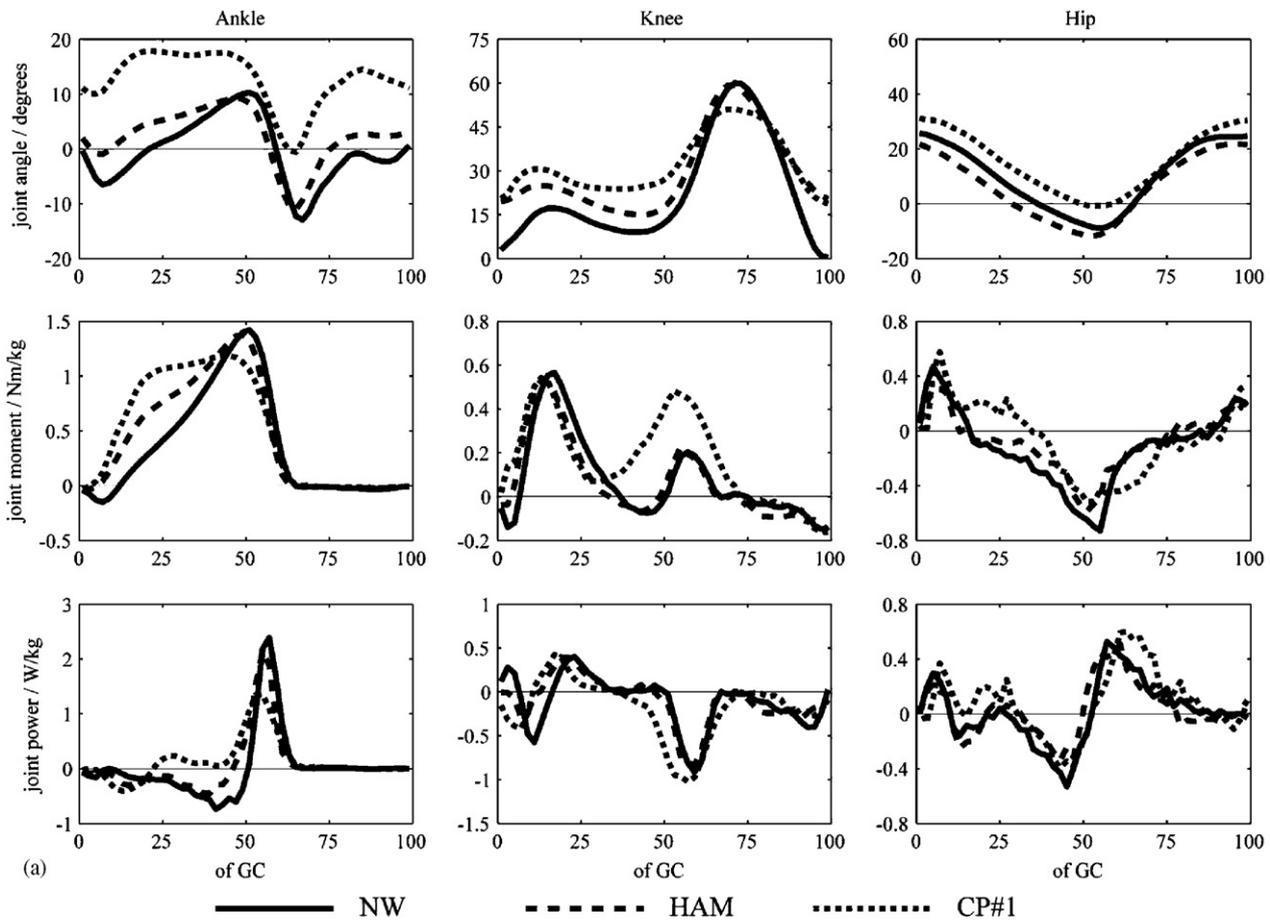


Fig. 3. a. Averaged kinematic and kinetic gait patterns ($N = 4$) from a selected case (CP#1), normal walking (NW) and HAM experimental conditions. Positive joint-angle values indicate flexion/dorsiflexion, positive joint-moment values indicate extension/plantarflexion moments and positive joint-power values indicate power generation. b. Display of a lower-body skeleton together with GRF and COP in five consecutive instants of stance phase that occurred approximately at 0%, 20%, 40%, 55% and 60% of a gait cycle. Upper panel—CP#1, lower panel—HAM.

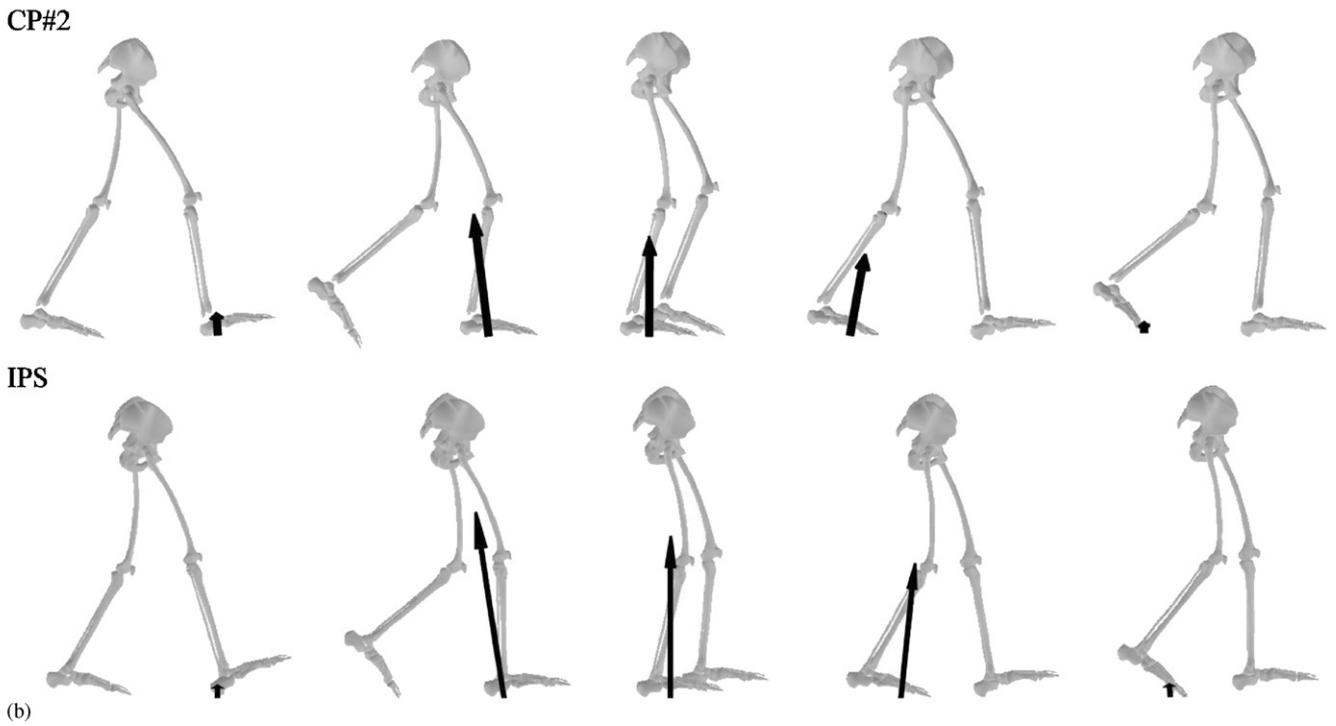
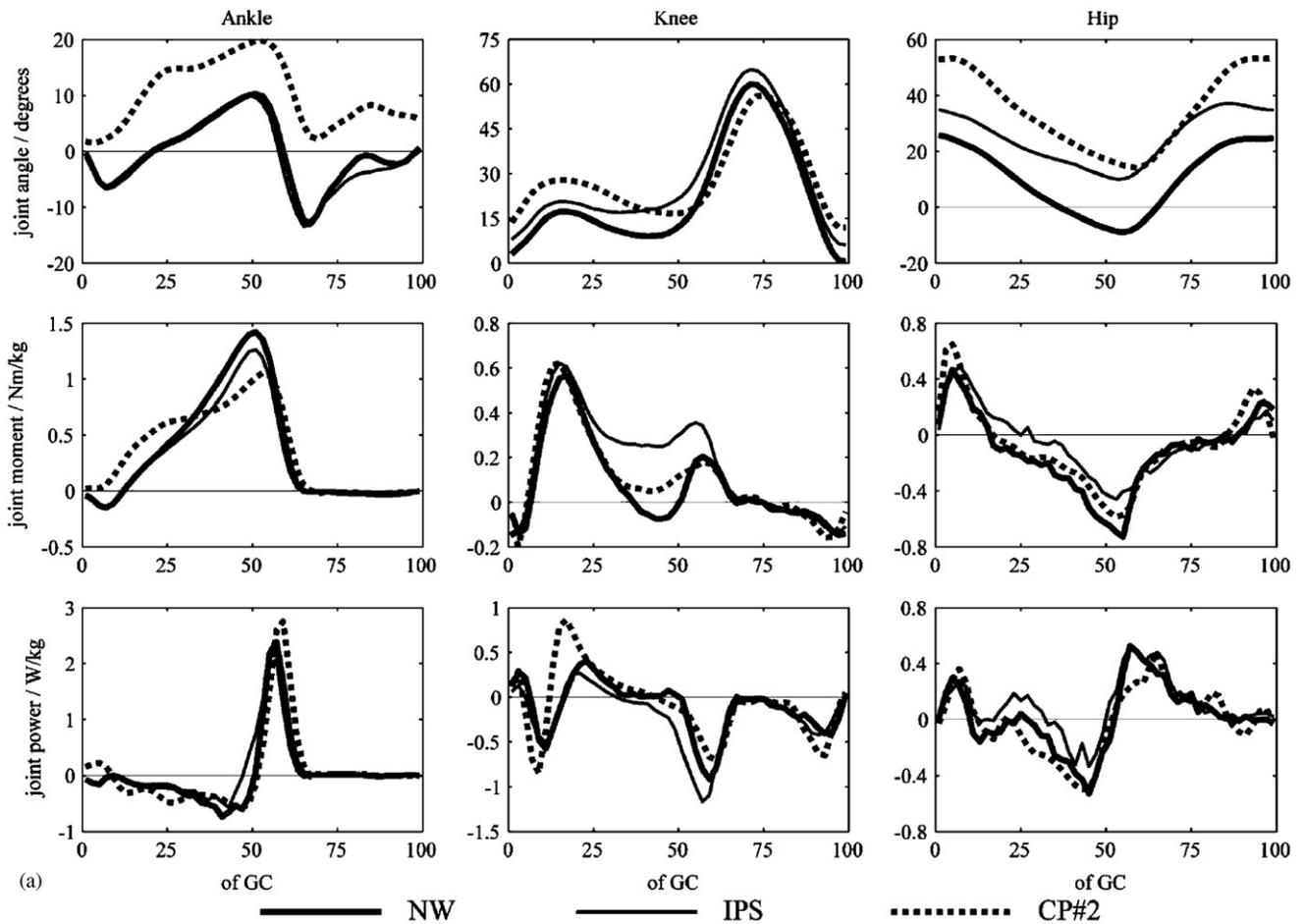


Fig. 4. a. Averaged kinematic and kinetic gait patterns ($N = 4$) from a selected case (CP#2), normal walking (NW) and IPS experimental conditions. Positive joint-angle values indicate flexion/dorsiflexion, positive joint-moment values indicate extension/plantarflexion moments and positive joint-power values indicate power generation. b. Display of a lower-body skeleton together with GRF and COP in five consecutive instants of stance phase that occurred approximately at 0%, 20%, 40%, 55% and 60% of a gait cycle. Upper panel—CP#2, lower panel—HAM.

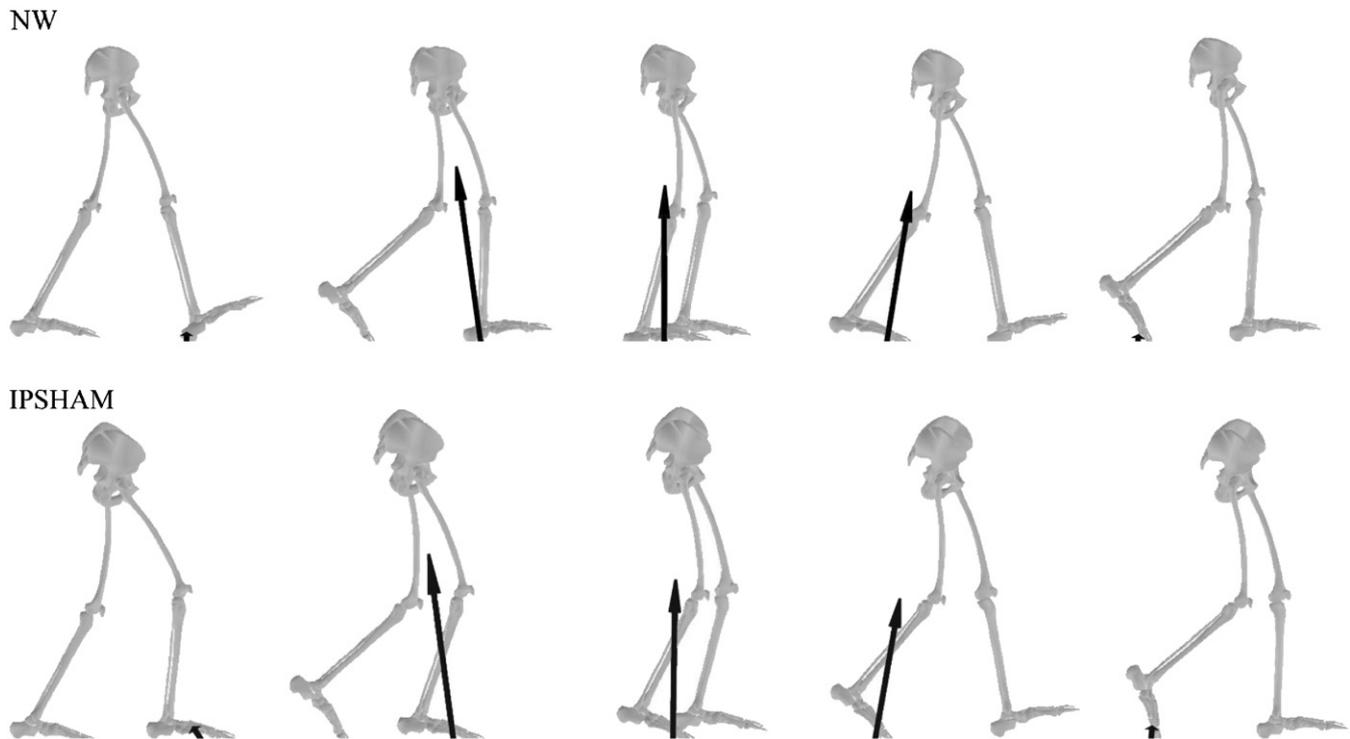


Fig. 5. Display of a lower-body skeleton together with GRF and COP in five consecutive instants of stance phase that occurred approximately at 0%, 20%, 40%, 55% and 60% of a gait cycle. Upper panel—NW, lower panel—IPSHAM.

(Neptune et al., 2001; Arnold et al., 2005; Matjačić et al., 2006).

Iliopsoas muscle is a uniarticular hip flexor and exerts hip-flexion moment. Therefore, contracture of iliopsoas directly forces hip into flexion. Iliopsoas can also dynamically act to accelerate the knee into flexion (Arnold et al., 2005), which is supported by experimental data from the present study. This increase in knee flexion is small during loading response and mid-stance as compared to HAM, and as such, the ankle-angle profile remains virtually identical to NW. During terminal and pre-swing, knee flexion becomes comparable to HAM. Unlike the influence of HAM contracture that was diminishing with the hip going into extension, the influence of IPS contracture increases with the hip going into extension. Our data show that the largest influence of IPS contracture starts at the beginning of terminal stance, suggesting that the tested subjects adopted excessively flexed hip posture throughout the gait cycle, which is a documented compensation for hip flexors contracture (Lee and Kerrigan, 2004). This excessively flexed hip posture reduces influence of IPS contracture at initial contact and loading response and postpones it into terminal stance. Since iliopsoas dynamically acts to accelerate the knee into flexion, this acceleration must be controlled. Unlike in HAM experimental condition where the knee-flexion acceleration, which occurred early in the stance, was opposed by

increased plantarflexors activity moving the COP forward, such compensation would probably not be efficient in terminal stance and pre-swing, since COP is already near the toes (Fig. 4b). Therefore, the knee flexion is likely opposed by increased activity of vastus muscle group, which starts at the end of midstance (abrupt stop of otherwise normal cessation of knee-extension moment, which normally turns into slight knee-flexion moment in the terminal stance). Also, this activity dynamically acts to accelerate the hip into extension (Arnold et al., 2005) and thereby synergistically assists glutei muscles in their effort to extend the hip.

When both muscle contractures are present (IPSHAM) the resulting kinetic and kinematic patterns can be explained by combining biomechanical characterization of each separate experimental conditions (HAM and IPS).

4.2. Clinical implications

It is difficult to determine dynamic hamstring and iliopsoas length on clinical examination alone. Therefore the data from instrumental gait analysis should be used in determination of causes for crouch in each particular individual (Gage, 2004). Currently, there is no theoretical basis for determining the biomechanical causes of an individual's pathological walking (Arnold

et al., 2005). The experimental data presented in this study show distinct kinematic and kinetic patterns that were related to underlying artificial contracture/tightness of hamstring, iliopsoas or a combination of both. Specifically, our experimental data show that hamstring tightness will result in increase in ankle plantarflexion moment throughout stance and associated shift of COP toward toes early in the stance. These changes will not appear with iliopsoas tightness. This observation is therefore distinct feature that may be used in determining involvement of hamstrings in particular crouch gait. Our experimental data also show distinct increase of knee moment during terminal stance, which is associated with iliopsoas tightness. These changes will not appear with hamstring tightness. This observation is also a distinct feature that may be used in determining involvement of iliopsoas in particular crouch gait.

The selected clinical examples closely resemble the experimental data. However, the dominating pathological state of iliopsoas and hamstrings causing altered posture and crouch can be accompanied with pathological changes at other muscle groups that may to some extent alter the kinetics and kinematics. This was the case in our selected clinical examples where CP#1 exhibited altered hip-moment profile due to plantarflexor weakness while CP#2 exhibited much larger ankle-angle dorsiflexion and hip flexion throughout the entire stance phase. On the contrary, the data obtained under HAM, IPS and IPSHAM experimental conditions show “ideal” changes to kinematics and kinetics of particular emulated crouch gait patterns. Therefore, when relating the data from our study with the data from a particular patient, this should be done qualitatively, searching for the distinct features in kinematic and kinetic patterns that may suggest primary cause for crouch walking pattern caused by contracture/tightness of iliopsoas, hamstrings or combination of both. The results of this study show that primarily ankle, knee and hip angle of rotation and ankle- and knee-moment profiles are all features that should be carefully inspected as they exhibit distinct differences in stance, which may help in determining the degree of iliopsoas and hamstrings contractures in a particular pathological case. Comparison of animated models that display COP and GRF together with a posture of a skeleton throughout the stance phase of experimental walking and each individual crouch walking case can additionally serve in searching for key similarities. Altogether these features can serve as strong indicators of which muscle contracture/tightness underlie particular crouch-walking. As such, they can be used as complementary information to clinical observations of a particular patient, which may importantly enhance reliability of determination of the predominant cause of crouch-walking pattern.

4.3. Relation to other approaches for determining relative involvement of particular muscle during walking

Pathological gait in CP population is at present classified solely according to averaged kinematic gait patterns in the ankle, knee and hip joints (Winters et al., 1987; Rodda et al., 2004). These classifications are broad in definitions and are aimed at practical clinical management. Lin et al. (2000) collected kinetic data for a group of CP subjects walking in crouch and attempted to obtain “common” moment and power profiles that characterize crouch walking. When related to data presented in our study, the averaged kinematic and kinetic patterns from Lin et al. (2000) are comparable only to each particular trajectory that exhibits the largest deviations from normal gait (i.e. kinematics correspond to IPSHAM, ankle-moment profiles to HAM and knee-moment profiles to IPSHAM). Therefore, averaged data from a group of CP children walking in crouch are of little use when considering each particular case.

Significant efforts have been directed into detailed biomechanical modeling and simulation studies of musculo-skeletal system in order to determine the role of particular muscle of lower extremity during walking in terms of its individual contribution to support and propulsion via induced acceleration analysis (Neptune et al., 2001; Zajac et al., 2002, 2003; Anderson and Pandey, 2003; Arnold et al., 2005). The results of these studies have revealed specific actions that individual muscle groups might have in specific sub-phase of gait cycle, thereby extending treatment of each muscle action from purely anatomical viewpoint to truly dynamical treatment within the functional task. In this way actions of particular muscles in all joints of lower extremity can be determined. The methodological paradigm used in the present study allowed us to experimentally perturb a particular muscle group (iliopsoas and hamstrings) and observe the altered gait pattern. In this way the results of our study verify findings of the above-mentioned biomechanical modeling and simulation studies while on the other hand the results of the biomechanical modeling and simulation studies can be used in order to explain the observed changes in the kinematic and kinetic patterns obtained in the present study and relate them to the action of a particular muscle group.

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