Studying the Effects of Using Two Types of Afos on the Hip Joint Kinematics and Kinetics during Walking in the Sagittal Plane for a Patient with Severe DDH; Part 1

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Abstract—**Knowledge of using the Ankle foot Orthosis in patients with severe developmental dysplasia of the hip is crucial and may help to improve the gait cycle during walking. This paper aimed to evaluate and study the kinematics and kinetics of the hip joint during walking in the sagittal plane for a patient with severe hip dislocation age 26 (the author of this paper) who is experiencing using two different types of ankle-foot Orthosis (Custom-made, and Leaf-AFO). The data were collected using ten cameras and one force plate. We used an inverse dynamic approach to calculate sagittal joint angles, moments and power. The results pointed out that The Custom-Orthosis had a higher moment during the late stance of the gait cycle than that of Barefoot; the data showed significant change by a mean difference of 0.1604 Nm/kg. however, the Leaf AFO Spring Orthosis did not change much the flexion moment during the late stance phase. The paper concluded that both orthosis wearing the custom –made AFO improved the kinetics of the hip joint in special while wearing the custom –made AFO leading to reduce the pain associated with loading during walking.**

Index Terms—**Developmental dysplasia of the hip (DDH), gait cycle, sagittal plane, kinematics and kinetics, motion capture analysis, AFOs.**

I. Introduction

The hip dysplasia disorder refers to the inadequate progression of the femoral head, the acetabulum, or both. The prognosis of the developmental dysplasia of the hip (DDH) is good if diagnosed early and treated according to a fixed protocol. Otherwise, if diagnosed late or left with no treatment it will progress to early secondary osteoarthritis [1]-[3]. One of the common treatment solutions options in adults is total hip arthroplasty (THA) mainly combined with an anatomical reconstruction of the acetabulum. Performing a subtrochanteric shortening osteotomy is sometimes necessary to prevent nerve palsy in patients with severe hip dislocation [4].

It is essential to detect and intervene in patients with developmental dysplasia of the hip (DDH) to achieve good results. However, there is a huge number of DDH adult patients who had no treatment at their early age. Patients with untreated DDH are closely facing long-term morbidities such as avascular necrosis of the femoral head, degenerative hip osteoarthritis (OA), muscular fatigue and chronic pain, and specifically gait deviations [5]-[9]. Gait analysis is commonly performed to assess the walking patterns in different patients including study the kinematics and kinetics of the lower limb in all three planes Sagittal, frontal and transverse [10]-[12]. These evaluations technique have been regarded as a useful supplement to clinical and radiologic assessment [10]-[12]. Many studies have investigated previously the gait patterns of patients who had received various operative treatments for developmental hip dysplasia (DDH). Most of these research studies have reported that the treated DDH patients had an improvement in the gait patterns after the therapy, but they did not return to their normal level of walking [13]-[16]. However, there are only a limited number of researchers have investigated the gait pattern of DDH untreated group [17], [18]. Moreover, several studies have studied the lower limb kinematics and kinetics for patients with DDH during walking in the sagittal plane. [17], [19], [20] have reported reducing hip flexion angle of diseased DDH limb, and reducing hip extension angle compared with the healthy group. Lai *et al.* [19] have reported that pelvic kinematics of DDH patients had less maximum anterior tilting of pelvic compared to a healthy control group. They also stated that during the entire gait cycle, the diseased side of the pelvic in the unilateral DDH group stayed lower than the unaffected side. Many researchers have investigated as well the kinetics of the lower limb in the sagittal plane and stated that the affected limb had a less maximum external extension moment of the hip joint and less maximum external flexion moment of the knee that those of Healthy Controls [17], [19], [20]. In terms of power, two studies reported the diseased limb had less peak hip power than the healthy control group [17], [20]. From the literature, it shows that there is a lack of research regarding the investigation of the orthosis devices and their effects on gait pattern in the sagittal plane for patients with Developmental Dysplasia of the hip. Thus, this research aims to investigate the effects of multiple types of ankle-foot

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Orthosis (Leaf-AFOSpring, Custom-Made-Orthosis) on the gait cycle of a patient (the author of this research) with severe hip Dysplasia in the left side, hyper-flexed knee in the right limb and severe ankle drop left foot. The research will include studying the kinematics and kinetics of the lower limbs under four conditions; barefoot, Custom-Orthosis, Leaf-AFO-Spring, and Shoes only) and compare that with the published healthy controls data.

II. Methods

A. Participant

One male adult (The author of this research) (26 years old, 47 kg in weight) was recruited in this study, with a history of developmental hip dysplasia, severe deformity of the spine, severe deformity in the left foot and ankle, and hyper flexed knee in the right limb. The patient is daily experiencing two types of an ankle-foot orthosis (LeafAFO-Spring, and custom Ankle Orthosis fabricated in the prosthetics center in Brisbane, Australia)

B. Data Acquisition

For the specific patient, the data were captured during a single visit to the sport and exercise center Research Lab at the University of Southern Queensland. After consent and short warm-up, the reflective markers were attached to the subject's pelvis and both lower limbs. The placement of the markers was according to the Orthopedic Rizzoli (IOR) lower body marker set. The data were captured under four conditions, barefoot, Custom-Made, LeafAFO, and Shoes only. These markers allow each segment of the limb (foot, shank, and thigh) and the pelvis to be treated as a 6degrees-of-freedom rigid segment. A static standing trial was captured with the individual in the anatomical position, which was defined as a normal stance on the force plate. After the static calibration, all the calibration-only markers (L-FME, R-FME, RTAM, L-TAM). Data were collected using 10-Camera Qualysis motion Capture System and QTM Software. Markers and force plate data were collected at 100Hz and 1000Hz, respectively. At the start, the subject was asked to walk at normal speed across the capture space, with his eyes facing forwards towards the wall in front. Three practice trials were given to make sure that during the recording the subject starting position was adjusted to increase the likelihood of right foot or left foot initial contact occurring on the force plate. Ten gait trials five for each limb on the force plate were recorded for every condition Barefoot, with custom Orthosis, with Leaf-AFO, and Shoes-Only. Additionally, two more trials for each condition were recorded as a replacement in case the subject did not fully strike the force plate. Finally, following the data collection all the Ortopedici Rizzoli (IOR) lower body markers were removed from the subject. Then, all data were converted and exported to C3D files to be imported into Visual 3D professional (C-Motion Inc., Germantown, MD). The model was built by using a six-degree of freedom that shows a full representation of the coordination and orientation of the joints in space Fig. 1. The aim of creating the model is to examine the linear movement and angular movements in all the planes (three rotations and three translations), this was done by establishing a rigid body frame based on segments that link the hip, knee, and ankle joint together. The Coda type pelvic segment was created by defining g the calibration targets the anterior superior iliac spine (R-IAS, L-IAS) and the posterior superior iliac spine (L-IPS, R-IPS). The right joint center of the unaffected hip (with developmental Dysplasia of the hip) was computed by the anterior-posterior iliac spine markers positions (ASIS) depending on a regression equation developed by (Bell *et al*., 1990) determined by 14% of the average distance between the left and right ASIS with a position 30% distally and 19% posterior to this point. Due to the dislocation of the left hip affected with severe developmental dysplasia of the hip, a new landmark was created to represent the hip joint center which was positioned 69% distally and 1% posterior to this point according to an approximate value measured by a 3-D X-ray reviewer software. The reference point for the new hip landmark is the original hip that determined automatically by the Visual 3d following the same regression equation illustrated above. The right thigh segment was built by considering the proximal joint is the right hip, the distal joint is the knee center which determined by the lateral and the medial knee markers (R-FME, R-FLE), and the measured value of the proximal radius was 0.0881291m computed by visual 3d according to the regression equation

$$0.5 * distance(righthip, lefthip) \quad (1)$$

The left thigh segment was built differently to that of the right thigh due to the severe dislocation of the hip; the lateral marker L-FTC and the Joint center (NEW LANDMARK), a radius of 0.0742m, defined the proximal joint of the thigh. While, the distal joint of the thigh defined by the lateral and medial knee markers (L-FME, L-FLE). The medial and lateral malleoli markers identified the ankle joint center. The patient's height and mass were entered to allow the model to then calculate the segments center of mass and segment radius based on the anthropometrical indices published by Dempster (1955).

C. Data Processing

For further analysis, all marker-positions and force plate data were then exported to Visual 3D professionals (C-Motion Inc., Germantown, MD). The data were filtered using Butterworth zero-lag fourth-order bi-directional low-pass filter with a cutoff value of 6 Hz for walking for the marker-location, and 25 Hz for the force plate data. A Butterworth filter prevents the high frequencies data and accepts the low frequencies signals, which are occurring due to the noisy results, resulted from the random movements of markers and soft tissue artifacts. The anthropometric data that calculated from individual body mass and height using Dempster's equations were subsequently combined with the low pass filtered data and used as an input for the inverse-dynamics calculation method resulting in sagittal joint angular positions and net moment and power of the ankle, knee, and hip joint in the stance phase. During walking, gait cycle events were identified from (heel strike to terminal swing) to normalize data to allow comparisons between the four main conditions and with the published healthy control data. The gait cycle for each limb starts when any part of the foot strikes the force plate (Initial Contact) until the same foot touches the ground again in the next step at the end of the swing phase. The inverse-dynamic method examines the external ground reaction force of the body segments and moments on the anatomical joints. An equilibrium mathematical formula is a key for the inverse dynamic approach starting by calculating the moments and force for every joint from toe to hip (Silva and Ambrósio, 2002). The moment of inertia for each segment was calculated based on the location and magnitude of the mass for each segment and the subject's anthropometric parameters (Dempster, 1955). The angles between every two segments were calculated according to the relative positions using the Euler rotation sequence equivalent XYZ (Ankle plantarflexion-extension, knee flexion-extension, hip flexion-extension, pelvic tilt). For example, the proximal for the ankle is the shank, while the shank is the distal segment; therefore, the ankle range of motion during the entire gait cycle in the sagittal plane depends on the orientation of the two segments. The underlying intent of this research was to test statistically the angle, moments and power values then determining the peak values to identify changes among the patients four conditions barefoot, custom-made-Orthosis, Leaf-AFO –Orthosis and shoes.

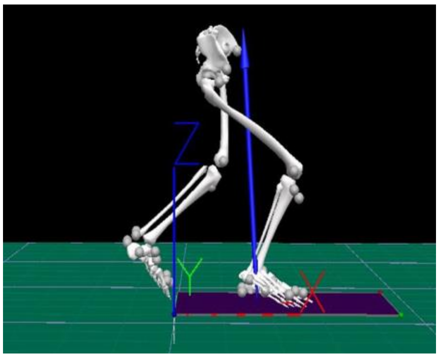

Fig. 1. The musculoskeletal model generated by the visual 3D modeling System. Example of a patient with a DDH Stepping on the force plate with the right foot during the dynamic trial.

D. Statistics

The statistical Package of (SPSS version 20, IBM SPSS) used to undertake the statistical analysis of the data collected for the patient with developmental dysplasia of the hip. A repeated-measures analysis of variance (ANOVA) was used with four factors (Barefoot, Custom-made, Leaf-AFO-Orthosis, and shoes only) with a post-hoc Bonferroni correction to determine statistical differences (mean differences) between every two factors (conditions). The p-value would be significant if it was less than 0.025 according to the statistical regression equations (Perneger, 1998). All trails of data collected were used for the analysis due to the small sample size (one patient).

III. Results

A. Hip Joint Kinematics

The main target in this research is studying the effects of the Orthosis aided devices on the hip parameters, in specific the power generated by the right unaffected hip during walking in the sagittal plane as well the maximum extension and flexion moments in the loading response and late stance phases of the gait. Due to the severe hip dislocation on the left side, we found that creating the model was a big challenge and to calculate the hip moments and power, we assumed that the proximal end of the thigh is connected to a virtual joint as mentioned before in the digitizing part. The kinematics data in the right unaffected hip during walking in the sagittal plane showed very close and symmetrical gait cycles among the four gait conditions. However, the results for the left affected limb with DDH pointed out significantly tremendous changes while wearing both Orthosis and Shoes over that of the barefoot gait cycle. Noticeably, there was no significant change in the right hip flexion angle at the initial strike, otherwise, the left flexion angle increased by mean difference (15.87,15.07, and 15.23 degrees) while using the Custom, Leaf, and Shoes respectively at IC. Furthermore, the left affected limb with DDH during the Loading response phase had higher values of hip flexion angle when using the orthosis and Shoes, and significantly the mean difference with barefoot was (11.95, 16.02, 16.27 degrees) as shown in Fig. 2. Besides, the maximum right and left extension hip angles at the late stance increased rapidly by using the custom Orthosis during the period from the push off the ground to the initial swing phase of the gait as shown in the graph. The swing phase for the right limb showed smooth and similar hip angle values over the whole conditions as well as the left affected limb with DDH.

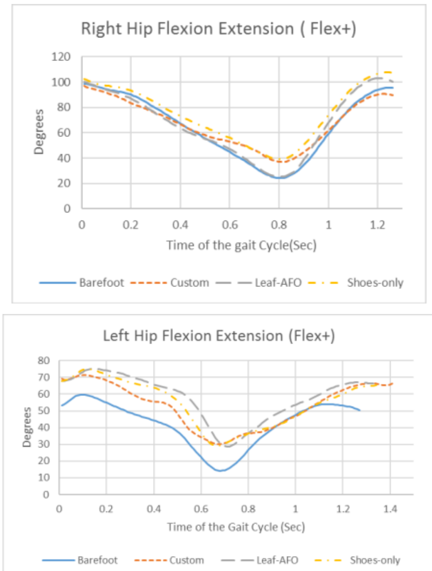

Fig. 2. Hip joint angles for the right and left limbs during walking in the sagittal plane under four conditions.

B. HIP Joint Kinetics

For hip flexion/extension moments and power, the custom-made orthosis decreased the right and left maximum hip flexor moments during the loading response phase by mean difference (0.13, and 0.07 Nm/Kg) respectively. The Custom-Orthosis had a higher moment during the late stance of the gait cycle than that of Barefoot; the data showed significant change by a mean difference of 0.1604 Nm/kg. However, the Leaf AFO Spring Orthosis did not change much the flexion moment during the late stance phase (right toes leaving the ground). Moreover, when the left foot tends to leave the ground during the period between the metatarsals pushing off until the big toe leaving the ground, the left hip moment increased drastically and showed a significant change in comparison to Barefoot by mean difference 0.31 Nm/kg. In terms of power generated by the right unaffected hip during walking in the sagittal plane, the data showed that the right hip under the custom-made Orthosis and the leaf AFO Spring orthosis during the initial contact portion of the stance phase had a higher Extension power generated in comparison to the barefoot and Shoes Conditions. Additionally, during the late stance phase of the gait especially at the moment the right foot pushes off the ground, both Orthoses decreased the Extensor power generated required to push the body forward in comparison to that of Barefoot by mean difference (0.244, and 0.54 Watt/kg) as shown in Fig. 3.

TABLE I: SHOWS HIP JOINT ANGLES FOR THE RIGHT AND LEFT LIMBS DURING WALKING IN THE SAGITTAL PLANE FOR THE PATIENT WITH DDH UNDER FOUR CONDITIONS; BAREFOOT, CUSTOM-MADE, LEAF-AFO, AND SHOES ONLY. NOTE, THE BOLD NUMBERS SHOWING THE MEAN DIFFERENCE BETWEEN CONDITIONS ARE SIGNIFICANT AND THE P-VALUE IS LESS THAN 0.025

Channels	Barefoot	Custom	Leaf	Shoes only
R flexion IC	97.9±1.5	95.8±1.49	100.3±1.6	102.5±1.9
L flexion IC	52.6±0.8	68.4±1.06	67.6±1.09	67.8±1.2
R extension, late stance	38.2±0.6	48.5±0.75	38.5±0.6	66.5±1.2
L extension, late stance	26.9±0.4	38.7±0.6	28.8±0.4	31.7±0.5
Mean Difference between conditions				
	B vs C	B vs L	B vs S	C vs L
R flexion IC	2.08	2.3	4.6	4.4
L flexion IC	15.8*	15.07*	15.2*	0.8
R extension, late stance	10.2*	0.25	28.4*	9.98*
L extension, late stance	11.8*	1.9	4.8	9.8*

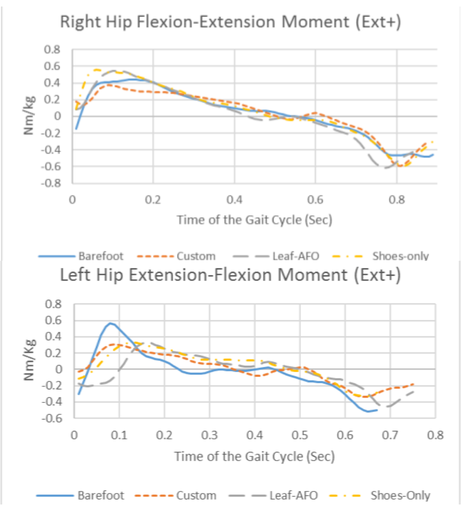

Fig. 3. Hip joint moments for the right and left limbs during walking in the sagittal plane.

TABLE II: SHOWS HIP JOINT ANGLES FOR THE RIGHT AND LEFT LIMBS DURING WALKING IN THE SAGITTAL PLANE FOR THE PATIENT WITH DDH UNDER FOUR CONDITIONS; BAREFOOT, CUSTOM-MADE, LEAF-AFO, AND SHOES ONLY. NOTE, THE BOLD NUMBERS SHOWING THE MEAN DIFFERENCE BETWEEN CONDITIONS IS SIGNIFICANT AND THE P-VALUE IS LESS THAN 0.025

Parameters Nm/kg	Barefoot	Custom	Leaf	Shoes only
R Flexor M at LD	0.42±0.006	0.29±0.004	0.54±0.008	0.38±0.007
L Flexor M at LD	0.37±0.006	0.30±0.004	0.15±0.002	0.31±0.005
R late stance M	-0.4±0.007	-0.29±0.004	-0.4±0.006	-0.22±0.004
L late stance M	-0.49±0.007	-0.18±0.002	-0.2±0.004	-0.28±0.005
Mean Difference between conditions				
	B vs C	B vs L	B vs S	C vs L
R Flexor M at LD	**0.1317***	**0.1228***	0.0389	**0.2545***
L Flexor M at LD	0.07	**0.2241***	0.0680	0.1452
R late stance M	**0.16***	0.03	**0.23***	**0.12***
L late stance M	**0.3138***	**0.2174***	0.2104	0.0965

IV. DISCUSSION

As explained before, the patient had a unique gait pattern due to severe developmental hip dysplasia and the severe hyper-flexed knee. Besides, the patient had experienced wearing two types of ankle-foot orthosis for the past ten years to accommodate his daily life activities. This study can be considered as a new attempt to assess the gait of patients with untreated hip dislocation DDH as well as investigating the effect of the ankle-foot orthosis on the gait kinematics and kinetics. Several studies have indicated that pathological individuals with Untreated DDH differ to healthy people on such gait parameters, likely correlating to the pain, leg-length discrepancy (LLD), hip OA and initial pathologic changes [21]-[23].

To our knowledge, the hip joint motion identified clinically by the path of the thigh displacement from the vertical which is considered clinically as the more appropriate way to define the joint motion. In the sagittal plane, describing the hip motion is affected by the displacement of both the femur and pelvic. As explained in the observation of the results, the hip joint moved through two arcs of motion during the normal walking of a patient with DDH. The results of our studies represented greater and lesser degrees of flexion values 20 to 120 degrees for the right limb unaffected with DDH and from 10 to 75 degrees for the left limb affected by Developmental dysplasia of the hip. However, the normal hip motion for healthy individuals reported in the literature ranges from -10 to 48 degrees considering the maximum hip flexion for normal adults ranges from 40 to 48 degrees during the swing phase of the gait [24]. Therefore, the results pointing out here that the excessive hyperflexion of the right knee is companying by excessive hyper-flexion hip angle while the patient is barefoot as shown in Fig. 3. As far as we know, the pathological deficiencies of the severe hip dysplasia may illustrate the differences over the whole gait cycle when the patient is barefoot in comparison to that of moderate hip dislocation subjects and normal healthy subjects. Generally, the hip muscles produce a flexor moment of 1.06 Nm/kg during the late stance phase, which controls the excessive extension of the hip [24]. In our case, both of the right hip and left hip affected by DDH produced an extensor moment of 0.5 Nm/kg during the push off-portion of the gait which is half the peak value generated by healthy subjects data stated in the literature [24]. The reason behind the reduction of the hip extensor moment during the late stance is that in patients with severe developmental hip Dysplasia, the insufficient cover of the femoral head reduces the load-bearing surface in the hip joint. Therefore, the dislocated joint will be experiencing more pain when it is loaded [25]. Our results indicated significantly that both Custom-made AFO and Leaf-AFO reduced the left abnormal hip joint flexor moment during the pre-swing phase of the gait. This reduction of the hip flexor moment while wearing the orthosis can be interpreted as an attempt to unload the hip joint and thereby lessen the pain supported by the studies of [26], [27]. It seems likely that the patient with severe DDH has less propulsion of the abnormal limb including less power generated due to the pathological change in the hip joint structure. It is worth noting that wearing both orthoses contributed to the forward progression of the hip by increasing the amount of power generation in the second half of the stance phase significantly in the left abnormal hip with DDH. The studies of Murray *et al.* [28] on walking patterns of patients with unilateral hip due to osteoarthritis support our current investigation. They pointed out that that the limited extension and the excessive flexion of the hip during the stance phase in the diseased hip which many patients witnessed during walking, was an attempt to avoid the pain maneuver to reduce the load on the femoral head. Also, the investigations of Romano *et al.* [17] on the gait cycle of 21 adults with residue congenital dysplasia of the hip support our case as they reported that the range of extension of the affected hip in all patients was drastically reduced. Thus, our studies indicated that both Orthosis have made a positive effect on the hip kinetics of the gait cycle especially in the second half of the stance phase such generating more power than that of barefoot condition and decreasing the amount of flexor moment to reduce the pain associated with loading during walking.

V. Conclusion

This study showed the significance of using the Ankle foot Orthosis and their effects on the hip joint kinematics and kinetics of patients with untreated severe hip dislocation. Despite the remarkable findings in terms of the positive change in kinetics, more investigations and future works are required such as studying the plantar pressure distribution and its relation to the hip and pelvic movements. Besides, studying the kinematics and kinetics of the knee and ankle joint, which will be presented in Series 2 of this study.

Conflict of Interest

The authors declare no conflict of interest

Author Contribution

Authors A, B, and C conducted the experiment at USQ lab; Redha Alrikabi analyzed the data and wrote the paper. Abert Chong contributed to analyzing the data and helped through proofreading. Ahmad also contributed to proofreading for the whole paper and made few suggestions regarding the experimental model.

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