Studying the Effects of Using Two Types of Afos on the Ankle and Knee Joints Kinematics and Kinetics During Walking in the Sagittal Plane for A Patient with Severe DDH; Part 2

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***Abstract*—Developmental Dysplasia of the hip DDH is considered as one of the most common Orthopedic Disorders, referring to a range of conditions from mild to severe dislocation of the hip joint. However, there are unclear gait alterations in individuals with DDH such as the investigation of the Ankle and Knee devices AFOs and their effect on the kinematics and kinetics of the joints during walking and running. This paper aimed to study and evaluate the kinematics and kinetics of the ankle and knee joints during walking in the sagittal plane for a patient age 26 with severe dysplasia of the left hip 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 planter flexion angle reached the maximum during the time between the Toes-off the ground phase and the initial swing phase significantly by mean difference (21.1, 14 deg). Moreover, the results indicated that the fabricated orthosis has decreased both the right and left extensor moments significantly during the load-bearing phase in comparison to Barefoot by mean difference (0.29, and 0.43 Nm/kg) respectively for both limbs. The paper concluded that the kinematics and kinetics of the ankle joint improved by the Custom made-Orthosis more than that of the Leaf AFO-Spring Orthoses. Finally, it concluded that both AFOs did not change much the kinematics of the knee joint; however, there were some improvements in the moments and power generated.**

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

I. Introduction

Many patients with developmental dysplasia of the hip joint disorders experience some limitations with gait, such as drop foot during the swing phase, mediolateral instability of the ankle joint in the stance phase, and insufficient plantar flexor activity. These problems result in an asymmetrical gait pattern, decreased gait speed, and effect postural stability and balance [1]. Literature findings suggest that ankle-foot orthosis (AFOs) can manage various lower limb disorders and neuromuscular. These assisted devices had positive impacts on kinematics, balance, and spatiotemporal gait parameters [2]-[6]. Custom-molded ankle-foot Orthosis is the most commonly used device for patients with inadequate gait cycles such as cerebral palsy individuals, and excessive ankle plantar flexion to provide safe ambulation by facilitating toe clearance during swing phase, decreasing body away, and improving mediolateral stability in stance phase [7]. Besides, AFO Leaf Spring is a pre-fabricated polypropylene ankle-foot orthosis designed to support flaccid drop foot. It provides a semi-rigid section for toe clearance and support, the absence of a heel section makes the Leaf Spring more comfortable to wear and provides a better fit in shoes. There are many features generated by the AFO Leaf Spring Orthosis such as the Injection-moulded polypropylene is lightweight, Variable thickness throughout the orthosis provides strength, provides good toe clearance and support, and Excellent fit for most types of shoes.

Several studies have investigated the effects of AFO use on Cerebral Palsy, stroke and Scoliosis rehabilitation. Franceschini *et al.* [7] suggested decreases stance time and double support combined with increases in walking speed and cadence. Gok *et al*. [8] found that custom-molded and metallic AFO orthosis provided an increase in step length, cadence, and walking speed combined with a decrease in double support time. The authors stated that the more solid-molded AFO can provide a better outcome than the plastic over-counter AFO orthosis. Simons *et al.* found that the rigid custom-molded AFOs have provided significant improvements in the balance scale, timed up and go, and have increased the walking and Functional Ambulatory category [9]. Some studies have compared the use of Chicagon articulated AFOs with off-the-shelf AFOs. The authors pointed out that the Articulated Chicagon Brace provided a high level of improvements in walking speed and the measured Kinematics Parameters such as the balance, angle of the ankle dorsiflexion, and a huge reduction in spasticity measures over three months for people with stroke and cerebral palsy [10]. Moreover, some studies have advocated that the combination of shoes and AFO Orthosis could have a positive effect on balance performance [11]. Fewer studies have compared several non-articulating, polypropylene AFOs of different stiffness degrees. The

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authors found all the orthosis increased dorsiflexion in swing except the stiffest design added more stability during the stance phase [12], [13]. Mulroy (2010) *et al.* have conducted research to study and compare the effects of using three different types of ankle-foot orthosis (AFO) on walking after stroke. The results pointed out that all three AFOs have increased the level of ankle dorsiflexion in swing and early stance phases of the gait. Both the PS and Rigid AFOs have increased knee flexion and have restricted ankle plantar flexion in the loading portion of the gait [13]. Only in participants without a plantar flexion contracture, the Rigid AFO tended to restrict knee flexion in swing and dorsiflexion in stance phases. The results showed also those individuals with quadriceps weakness could tolerate an AFO with plantar flexion mobility in loading easily. An AFO that permits dorsiflexion mobility in stance can benefit participants without a contracture.

From the literature, there is a lack of research regarding investigating the effect of the Ankle Foot Orthosis devices on the gait characteristics of patients with Severe Developmental Dysplasia of the hip DDH. Thus, this research aims to investigate the effects of multiple types of ankle-foot Orthosis (Leaf-AFO-Spring, 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 ankle and knee joints during walking in the sagittal plane under three conditions; barefoot, Custom-Orthosis and Leaf-AFO-Spring) and compare that with the published healthy controls data.

II. Methods

A. Participants

One male adult (The first 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 on the left foot and ankle, and hyper flexed knee on the right limb. The patient is daily experiencing two types of an ankle-foot orthosis (Leaf-AFO-Spring, and custom Ankle Orthosis fabricated in the prosthetics center in Brisbane, Australia).

B. Measurement System

10 Qualisys Oqus computerized motion analysis system (Qualisys 2.14, Gothenburg, Sweden) infrared motion cameras were utilized for testing at the gait laboratory at the University of Southern Queensland; three cameras were positioned at the back of the walkway, three cameras at the front of the walkway, and two cameras on each side of the walkway as shown in Fig. 1. These cameras are designed to obtain the three-dimensional coordinates of retro-reflective markers that were positioned on the lower limb of the patient with developmental Dysplasia of the hip during walking. One force platform (AMTI: Advanced Mechanical Technology Incorporation, Watertown, USA, model BP600400) embedded in a walkway was used to collect the kinetic data of the patient during walking under all three conditions; barefoot, with Custom-Made-Orthosis, and with Leaf AFO Orthosis.

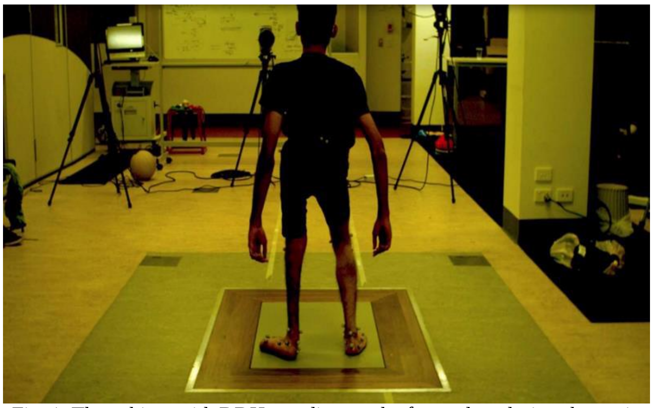

Fig. 1. The subject with DDH standing on the force plate during the static trial.

C. Test Protocol and System Calibration

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. Briefly, 4 markers were placed at the femur lateral and medial epicondyle 2 for each limb [L-FLE, L-FME, R-FLE, R-FME), 2 markers were placed at the proximal tip of the head of the fibula 1 one each limb [L_FAX, R_FAX), 2 other markers were attached to the most anterior border of the tibial tuberosity 1 for each limb [L_TTC, R_TTC], 4 markers were placed to the lateral and medial prominence of the lateral and medial left and right malleolus respectively [L_FAL, L_TAM, R_FAL, R_TAM], 2 markers were placed at the lateral side of greater trochanter 1/ from the proximal end [L_FTC, R_FTC], and the remaining 4 markers were attached to the anterior superior iliac spine [L_IAS, R_IAS] and the posterior superior iliac spine [L_IPS, R_IPS]. The placement of the markers was according to the Orthopedic Rizzoli (IOR) lower body marker set [14]. The data were captured under three conditions, barefoot, Custom-Made, and Leaf-AFO. These markers allow each segment of the limb (foot, shank, and thigh) and the pelvis to be treated as a 6-degrees-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, R-TAM, 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, and with Leaf-AFO. 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.

Before the recordings, the camera system was calibrated to produce a calibrated volume using an ‘L’-shaped metallic structure that represents the global coordinate system. Importantly, throughout the calibration process, the alignment of the long axis with the force plate is more critical than the short axis, so we used height adjustment screws to keep both axes horizontal. A dynamic calibration was performed by fixing the L frame to the medial edge of the force plate and the calibration wand was waved with a fixed distance between the three markers around the capture area to provide a data capture. At least two left and two right dynamic trails were recorded while the patient was walking at the same speed. The participant had to strike the force plate with the whole region of the foot. Between each calibration trial, the participant asked to rest for 5 minutes to make sure the alignment of the L frame axis is correct and similar to that of the previous trial. The camera system calibration was accepted when the residual errors were less than 2 mm to ensure that most of the motion capture system was covered in all the trails. The laboratory coordinate system was then defined by three axes (in positive and negative directions). The X-axis was defined as the anterior-posterior (forward/backward direction), the Y-axis was defined as mediolateral (left/ right), and the Z-axis as proximal-distal (upward/downward) as shown in Fig. 2.

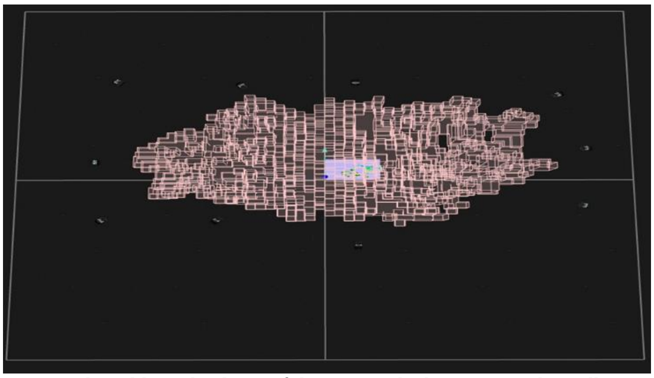

Fig. 2. The calibrated volume of the space, which represents the walkway area surrounded by Ten Qualysis Cameras.

D. Digitizing and Modelling

The 2D markers-position data for each of the 10 cameras were labeled and combined into a 3D representation by using the Qualisys Tracking manager software (Qualisys AB, Gothenburg, Sweden). The automatic identification of trajectories in the Qualysis Track Manager 2.14 software was performed by a module called AIM (The automatic identification of markers). Istituti Ortopedici Rizzoli (IOR) static lower body marker set with 18 markers attached to the ankle, knee, thigh and spine areas as illustrated before in the test protocol part of this chapter has been applied to identify the static markers for all the three conditions Barefoot, custom made Orthosis, and Leaf-AFO –Orthosis only of the patient with DDH. After labeling and identifying the static trials for the three mentioned conditions, the Istituti Ortopedici Rizzoli (IOR) Dynamic lower body marker set including the identifications of 14 markers has been applied to all five trials of each limb of each condition to identify all trajectories for the whole gait cycle of both right and left limbs. 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. 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 as shown in Fig. 3. 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). 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) [15].

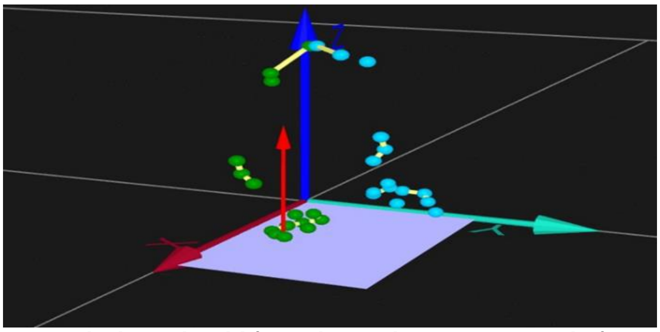

Fig. 3. The digitized model for a Subject with DDH stepping on the force plate during the dynamic trial.

E. Data Calculation

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 three 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. 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 [16]. 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) [15]. 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).

F. Statistics

The statistical Package of (SPSS version 25, 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 three factors (Barefoot, Custom-made, and Leaf-AFO-Orthosis) 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 presented in the studies of (Perneger, 1998) [17]. All trails of data collected were used for the analysis due to the small sample size (one patient).

III. Results

A. Ankle and Knee Kinematics

During walking in the sagittal plane, the maximum dorsiflexion angle for all conditions happened between the terminal stance phase and the pre-swing phase, not significantly both orthoses have increased the dorsiflexion angle by mean difference (2.71, 4.2 degrees) respectively. The planter flexion angle reached the maximum during the time between the toes-off the ground phase and the initial swing phase, it is worth noting that there is a significant change as both orthosis conditions have increased the plantar-flexion angle by mean difference (21.1, 14) respectively. The ankle-Custom made orthosis has affected the gait cycle for the left limb rapidly, the maximum dorsiflexion angle happened at the left foot pushes off the ground which is less than the Maximum-Barefoot Dorsi-flexion angle by mean difference 17.3. Additionally, the Custom-made Orthosis had a long-range of plantar-flexed ankle starting from a position at the initial swing and continuing until the ankle dorsiflexed to a neutral position at the end of the cycle. The results significantly showed that the Custom and Leaf AFOs decreased the Plantar-flexion angle compared to the Barefoot by mean difference (17.6, 18.1) as shown in Table I.

Considerably, the case study in this research has one of the rale unique gait patterns. As mentioned before in the case description, the right knee is hyper-flexed severely due to the dislocated left hip and the inadequate movement that the patient had during his early age. In the sagittal plane movement, the Custom made, Leaf AFO has had a huge impact on the gait cycle comparing to the barefoot condition and significantly shown decreasing the right knee-flexion angle at the initial strike by mean difference (17.46, 11.76 deg). Besides, during the mid-swing phase, the custom and Leaf orthosis have decreased the flexion angle by mean difference (13.43, 23.81deg) respectively in comparison to the healthy. However, there is no significant change regarding the maximum flexion angle for the right knee at the late stance (right foot toes off). The kinematics analysis of the left diseased limb with DDH revealed that the most significant change was during the late stance phase of the gait correlated with decreasing the values of left knee flexion angle from the period, where the left toes leaving the ground to the initial swing portion as shown in Table I.

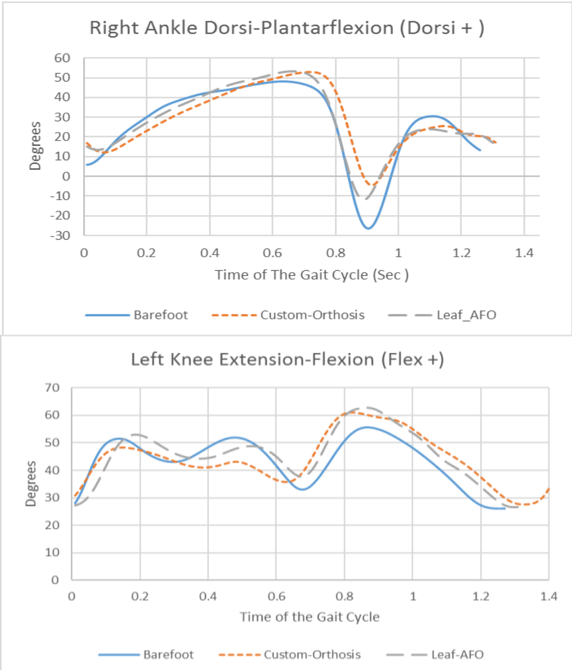

Fig. 4. Ankle and knee joints angles for the right and left limbs during walking in the sagittal plane in degrees.

TABLE I: Shows Ankle and Knee Joint Angles in (Degrees) for the Right and Left Limbs during Walking in the Sagittal Plane for the Patient with DDH under Three Conditions; Barefoot, Custom-Made, and Leaf-AFO. Note, the Bold Numbers Showing the Mean Difference between Conditions Are Significant and the P-Value Is Less than 0.025

Parameters	Barefoot m±sd	Custom-made m±sd	Leaf-AFO m±sd
R Max-Dorsi-Flex	47.1±0.68	49.87±1.27	51.3±1.26
L Max Dorsi-Flex	40.9±0.58	23.57±0.59	27.03±0.66
R Plant-Flex PSW	-25.1±0.36	-3.53±0.09	-11.03±0.27
L Plant-Flex PSW	28.7±0.41	11.05±0.28	10.57±0.25
R-Knee-Flex IC	72.5±0.96	55.07±0.95	60.77±0.74
L-Knee-Flex IC	28.35±0.38	30.48±0.53	27.4±0.34
R-Knee-Flex MSW	94.841±25	81.41±1.40	71.02±0.87
L-Knee-Flex MSW	46.79±0.62	44.87±0.77	42.43±0.52
Mean Difference between conditions (+)			
	B vs C	**B vs L**	**C vs L**
R Max-Dorsi-Flex	2.71	4.2	1.5
L Max Dorsi-Flex	**17.3***	**13.8***	3.46
R Plant-Flex PSW	**21.58***	**14.09***	7.49
L Plant-Flex PSW	**17.69***	**18.18***	0.48
R-Knee-Flex IC	**17.46***	**11.76***	5.7
L-Knee-Flex IC	2.48	0.96	3.44
R-Knee-Flex MSW	**13.43***	**23.81***	**10.38***
L-Knee-Flex MSW	1.92	4.35	2.43

B. Ankle and Knee Joints Kinetics

In terms of moments, as mentioned before wearing the orthosis devices on the left limb has affected the kinematics during the entire gait cycle. Noticeably, the changes in the kinetics of the lower limbs were witnessed regarding the moments and power. In terms of moments, the maximum right ankle flexor moment during the loading response phase was increased significantly while using the Custom-Orthosis compared to the barefoot. However, the Leaf-AFO Orthosis during the period from the initial contact until the weight-bearing stability phase showed a closely similar magnitude of the right ankle flexor moment to that of Barefoot condition (-0.14, -0.04,and-0.13 Nm/kg). At the period from the midstance phase until the terminal stance phase (Right foot pushes off), the plantar-flexor moment increased until it reached the maximum values at the late push off. All of the Custom, Leaf conditions had a higher right ankle Plantar-flexor Moment than that of Barefoot by mean difference (0.12, 0.2 Nm/kg) as shown in Table II. Despite that, the results showed statistically a significant change for the right ankle moments, there is an asymmetry in the entire stance phase among all the three conditions. For the left diseased limb and during the period between the midstance phase and push-off phase, the left ankle plantar-flexor moment when wearing the custom-made orthosis started to increase rapidly until it reached the maximum value of 0.56 Nm/kg. It showed a significant difference along with the entire stance phase between the Custom and Barefoot conditions, but the Leaf-AFO Spring Orthosis did not affect the gait variables compared to the Barefoot. In terms of power generated during walking in the sagittal plane, the custom made, Leaf-AFO and shoes generated a higher maximum Plantar-Flexion power than the Barefoot at the late stance by mean difference (1,0.9, and 0.8 Watt/kg) respectively. However, there is no significant difference between the Custom and Leaf in terms of power generated by the right limb during the late stance before the toes leave the ground and the mean difference is 0.08 Watt/Kg. Additionally, the power graph for the right limb showed consistency and symmetry in values during the period from the initial strike to the late portion of the Midstance phase (60%) of the stance. To our knowledge, the left limb has a unique pattern gait due to the severe hip dislocation and that makes the influence on the gait parameters especially in the kinetics part, the Custom made decreased drastically and significantly the power generated by the affected limb during the loading response phase compared to the Barefoot by mean difference 0.376 Watt/Kg. However, the Leaf-AFO Orthosis decreased the maximum Dorsiflexion power generated during load-bearing by the main difference -0.06 watt/Kg that is statically considered non-significant compared to the other conditions. Additionally, during the late stance phase, there is no significant change witnessed among all the three-conditions. The knee kinetics data showed in some values similar results to the ankle kinetics in terms of the custom orthosis influence overall gait cycle. This fabricated orthosis has decreased both the right and left extensor moments significantly during the load-bearing phase in comparison to Barefoot by mean difference (0.29, and 0.43 Nm/Kg) respectively for both limbs. Also, during that loading response phase, the custom had a higher generated knee flexion power in both limbs than those of Barefoot, and Leaf. The Leaf-AFO spring Condition showed similar right knee Extensor moment data along the whole gait cycle in comparison with Barefoot Condition except the moment when the right foot pushes off the ground, as Leaf had a higher extensor moment than Barefoot showing mean difference of (0.3, 0.4 Nm/kg), respectively.

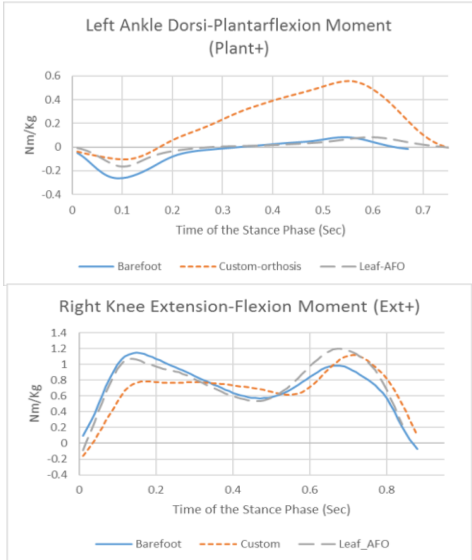

Fig 5. Examples of knee and ankle stance phase kinetics for the right and left limbs during walking in the sagittal plane for the recruited subject.

TABLE II: SHOWS ANKLE AND KNEE KINEMATICS IN NM/KG FOR MOMENTS IN WATT/KG FOR POWER FOR THE RIGHT AND LEFT LIMBS DURING WALKING IN THE SAGITTAL PLANE FOR THE PATIENT WITH DDH

Parameters	Barefoot m±sd	Custom-made m±sd	Leaf-AFO m±sd
R ankle flexor moment during LD (Nm/kg)	-0.14±0.024	-0.04±0.012	- 0.13±0.003
L ankle flexor moment during LD (Nm/kg)	-0.24±0.0034	-0.1±0.002	- 0.01±0.003 7
R ankle flexor moment during PSW(Nm/kg)	0.36±0.017	0.48±0.013	0.56±0.02
L ankle flexor moment during PSW (Nm/kg)	0.075±0.001	0.56±0.014	0.02±0.005
R Ankle power gen-rated at LD Watt/Kg	0.33±0.002	0.07±0.01	0.24±0.01
L Ankle power gen-rated at LD Watt/Kg	0.37±0.0053	0.0031±0.009	0.439±0.01
R Knee Moment at Loading Nm/Kg	1.07±0.012	0.78±0.015	1.02±0.017
L Knee Moment at Loading Nm/Kg	0.97±0.01	0.54±0.02	0. 65±0.01
Mean Difference between conditions (+)			
	B vs C	**B vs L**	**C vs L**
R ankle flexor moment during LD (Nm/kg)	**0.1***	0.01	**0.09***
L ankle flexor moment during LD (Nm/kg)	**0.1369***	**0.092***	0.044
R ankle flexor moment during PSW(Nm/kg)	**0.12***	**0.2***	**0.08***
L ankle flexor moment during PSW (Nm/kg)	**0.486***	**0.054**	**0.54***
R Ankle power gen-rated at LD Watt/Kg	**0.26***	0.08	**0.17***
L Ankle power gen-rated at LD Watt/Kg	**0.37***	0.06	**0.43***
R Knee Moment at Loading Nm/Kg	**0.29***	0.05	**0.24***
L Knee Moment at Loading Nm/Kg	**0.43***	**0.32***	**0.1***

IV. DISCUSSION

The objective of this study was to prospectively study and evaluate the effect of using two types of Ankle Foot Orthosis on kinematics and kinetics of the ankle knee joins during walking in the Sagittal plane for a patient with severe dysplasia of the left hip. Both orthoses showed significant changes during walking in the sagittal plane indicating gait improvements in some phases. It is worth noting at the terminal stance, the Custom –made Orthosis and "off-the-Shelf" reduced the abnormal ankle Dorsiflexionn motion for the left limb with DDH. Some studies support our current study who has also shown that ankle Dorsiflexion during the late stance phase was between 8–11.9◦ for patients with Cerebral palsy which is considerably closer than expected to normal individuals [18], [19], [20]. Therefore, our results showing more improvements in the gait during the period between the push-off to toes off for the left abnormal foot that affected by DDH while wearing Both Orthoses in comparison to the barefoot condition. Even though, the off the shelf Orthosis has the advantage of allowing more Dorsi-Flexion angle to occur during Midstance and Terminal stance phases as the tibia transitions over the foot, the Custom-made showed better maximum Dorsi-flexion results and considerably closer to the healthy normal data observed in [21] due to the polypropylene deformation that occurs even with the rigid Custom Orthosis. However, the maximum ankle Dorsi-flexion for the right unaffected limb showed increased significantly while using the Custom made and "off the Shelf Orthosis during the late stance in comparison to that of barefoot condition. According to the studies of Burnfield *et al.*, the increase in the maximum Dorsi angle reflected an improvement in the gait over the excessive Dorsi-flexion while the patient is barefoot, the results showed in somehow similar values to that group of healthy persons. The positive increase in the right Dorsi-flexion angle might have occurred due to the enhancement in body stability while wearing the orthosis which allows the gastrocnemius muscle action to Stabilize the dorsiflexing angle and also provide early heel arise [23].

The corresponding ankle joint kinetics during the loading response phase while barefoot showed excessive ankle moments with excessive power generated. These findings indicate that the excessive Dorsi flexor eccentric contraction is occurring because the left affected limb with DDH excessively dorsiflexes during the weight-bearing portion of the gait. The abnormal ankle moments decreased by both orthosis during the LD, however, the power absorption decreases excessively by wearing the custom made orthosis. Few studies showed closer findings while using Solid Custom-made Orthosis, but there is a lack of findings regarding the off the shelf spring AFOs. Besides, the corresponding ankle joint kinetics while the patient is barefoot pointed out a reduction in the abnormal left foot peak plantar flexor moment during the terminal stance phase of the gait companying with an excessive reduction in the power generated during the preswing potion at the late stance phase.[24] The custom made Orthosis produced larger peak ankle plantar flexor moments during the terminal stance as well as increasing the power generated during the PSW portion of the gait, however, the " off-the-shelf " orthosis did not decrease the power generated during the same phase. Moreover, the right unaffected limb with DDH had more power generated during the pre-swing phase of the gait while using the Custom made and the Leaf AFO spring Orthosis, these orthoses shifted the power generated value as close as to that of Normal healthy subjects' values observed in the studies of Burnfield *et al.*. [23]. The findings of higher power generated for the abnormal left foot values with the Solid AFO in comparison to that of barefoot indicate even the rigid material of the Custom orthosis still allows greater plantar flexor concentric contraction for push-off during the pre-swing phase supported by the studies of Rethlefsen [18].

The corresponding knee joint kinematics and kinetics for the right limb not effected with developmental hip Dysplasia of the hip during walking in the sagittal plane while the patient is barefoot witnessed excessive hyperflexion angle along the entire gait cycle. The results of our studies represented greater and lesser degrees of flexion values in the full range of 70 to 125 degrees, however, the normal knee motion for healthy individuals during walking in the sagittal plane had a full range of 0 to 60 degrees as represented in the studies of Burnfield *et al.* [23]. At the instance of initial right heel contact with the floor, while barefoot, the knee observationally appears excessively flexed and the alignment of the body vector posterior to the knee axis causes a less stable weight-bearing with less power generated, resulting in exceptional flexor moment that modulates the rapid knee flexion. At the initiation of mid-stance phase, the right knee flexion moment while barefoot had flexion of almost 90 degrees, a small amount of power generation, thus leading to lower the body towards the ground as the reference limb is shorter (the left limb affected by DDH), limiting the forward progression of the limb for pre-swing phase providing less stable weight-bearing. The right showed a continuation of the progressive excessive flexion and reached the maximum value at the terminal stance, this compromises the weight-bearing stability. The studies of Rethlefsen *et al.* [18] support our results, the authors indicated the abnormal knee motions of crouch gait of cerebral palsy patients were not changed by the solid Custom-made Orthosis that was designed either specifically for each individual or the Hinged AFO. According to, clinicians had a concern about the possibility of knee motion improvements over the pathological gait, as a result of using AFOs were not substantiated.

V. Conclusion

This study showed the significance of using the Ankle foot Orthosis and their effects on the ankle and knee joint kinematics and kinetics of patients with untreated severe hip dislocation. The paper indicated more improvement in both ankle kinematics and kinetics while using the Custom-made-orthoses; otherwise, the knee moments and power did not witness show many improvements over both Orthoses. More investigation is required in the future such as the investigating of the customized Knee-Ankle-Foot Orthosis KAFOs, Plantar pressure distribution correlated with knee location during walking in the sagittal plane.

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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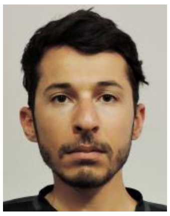

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