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Dynamic Analysis of the Gait Cycle for Normal and Abnormal Subjects

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Abstract

Identification of pathological gait is the most direct application of gait analysis. The purpose of the study is to investigate the dynamics of human walking over a complete gait cycle. Level-walking experiments were performed by two-dimensional (2D) motion analysis using a digital video camera (Sony, 25 Hz) and two force plates. Kinematic data were obtained from the trajectories of 7 reflective markers using SkillSpector software (ver. 1.2.4). MATLAB software (ver. 8.1) has been adopted in this work to obtain Pedotti diagram and for inverse dynamics computations. Digital low pass Butterworth filter with zero phase-shift and cut-off frequency of 4.5 Hz was used. Joints' angular displacement, forces, and moments were obtained during gait cycle. The study was made on fourteen healthy volunteers (10 males and 4 females); a male with cerebral palsy, and an old female underwent unilateral knee arthroplasty. These data can be used as standard measures in pathology studies, as input to theoretical joint models, and as input to mechanical joint simulators.

Keywords: gait analysis, Pedotti diagram, inverse dynamics, and kinematics.

Nomenclature

a_{fx}, a_{fy}	Cartesian accelerations at the foot in x, y direction (m/s^2)
a_{sx}, a_{sy}	Cartesian accelerations at the shank in x, y direction (m/s^2)
a_{tx}, a_{ty}	Cartesian accelerations at the thigh in x, y direction (m/s^2)
COP_x	Force plate center of pressure in x direction (m)
F_{Ankle}, F_{Knee}, F_{Hip}	Forces at the ankle, knee, and hip joints respectively (N)
F_{ground}, GRF	Ground reaction force (N)
g	Gravitational acceleration (m/s^2)
I_t, I_s, I_f	Moments of inertia of the thigh, shank, foot respectively (kg.m^2)
M_H, M_K, M_A	Moments at the hip, knee, ankle respectively (N.m)
m_t, m_s, m_f	Masses of the thigh, shank, foot, HAT respectively (kg)
α_t, α_s, α_f	Angular acceleration of the thigh, shank, foot respectively (rad/s^2)
UKA	Unilateral knee arthroplasty patient
CP	Cerebral palsy patient
BW	Body weight (N)
Meta2	Second Metatarsal

Introduction

Human gait is a complex spatiotemporal process involving structures and functions of the neuro-musculoskeletal system of human body. In normal gait, not only the energy consumption is optimized, but also, the resulting loads are regulated so that they can be well tolerated by the joints without initiating destructive changes in the articulating cartilage [1]. The subsequent effect on the load of the lower extremities is a matter of concern, since it might contribute to the disorders in bones (e.g., osteoporosis) or articulating joints (e.g., osteoarthritis) of both the intact and prosthetic limbs.

In biomechanical applications, motion tracking systems are traditionally used together with force plate measurements and information on the body segment inertial parameters to estimate values of net joint torques (i.e., ankle, knee, and hip torques) and net inter-segmental forces using inverse dynamics [1]. Gait analysis has radically changed the treatment of cerebral palsy. Preoperatively, it allows critical assessment of the specific pathologies of the patient. Postoperatively, it provides an accurate assessment of outcome. This assessment of outcome has in turn allowed the accurate critique of surgeries and has made it possible to discard treatments that are not useful or are perhaps even injurious [2]. Favoring one leg results in increased loading in the unaffected leg and could eventually lead to osteoarthritis in that leg [3]. Ground reaction force (GRF) measurements allow gait asymmetries to be analyzed quantitatively and could be used to develop training programs to help patients avoid complications such as arthritis.

The main goal of this work is to study the dynamics of human walking in order to get measurement values that can be depended on in the hospitals of rehabilitation, the clinical of medical sports, design and evaluation of orthotics and prosthetics devices and in developing rehabilitation programs for patients with joint problems. This study provides such normative data for joint dynamics parameters.

2. Methods

The dynamic analysis is done using a program written in the MATLAB software where gait data to be analyzed and the filtering method applied on dynamic marker trajectories. The output of this program includes: Pedotti diagram, joints' angles, moments and powers during the complete gait cycle. The x-y coordinated of centers of hip, knee and ankle joints were digitized with the help of markers using SkillSpector software.

2.1 Pedotti diagram

A graphic representation of complete spatiotemporal sequence of the ground reaction evolution can be obtained by combining vertical and fore-after component of the ground reaction force with center of pressure in the fore-after direction during stance phase of gait cycle. It has been used to identify abnormal forces acting on the foot in pathological gaits [4]. The density of the vectors indicates the velocity with which the center of pressure moves on the ground [5]. This plot is obtained using the MATLAB command "feather" after simple modifications.

2.2 Filtering of Gait Data

A discrete-time low-pass filter was used to eliminate random errors. Second order zero phase Butterworth filter was used for phase shift cancelling. The difference between filtered and noisy displacement and acceleration were constructed and shown in Figure (1). It is clear from the figure that the difference between filtered and noisy displacement data was small but this difference increased when applying differentiation. Fast Fourier transform analysis was made on the displacement data and showed that the appropriate cut off frequency was around 4.5 Hz as shown in Figure (2).

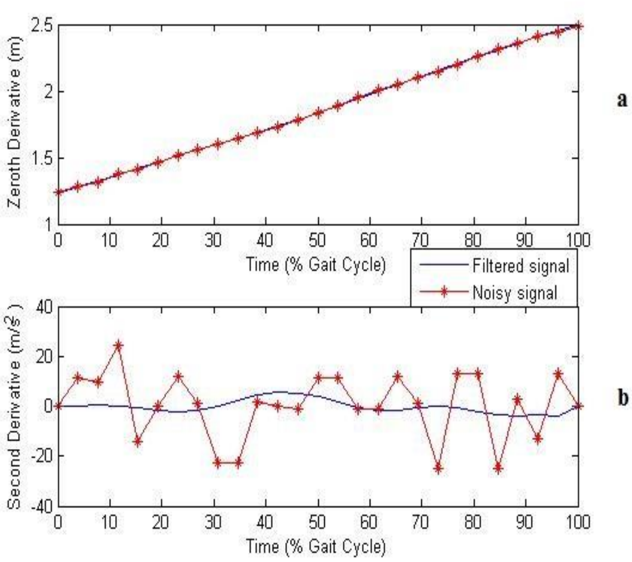

Figure 1: Difference between filtered and noisy hip displacement signals, a: displacement data, b: acceleration data.

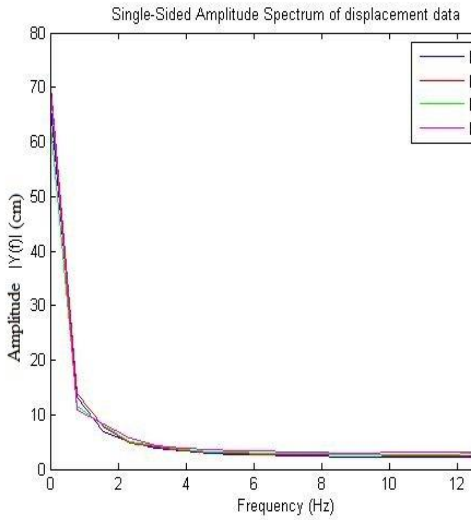

Figure 2: The frequency spectrum of displacement gait data.

2.3 Calculation of Kinematic Quantities

As an approximation, human body movement can be modeled using rigid body dynamics, that is, the segments of the body are treated as objects of fixed size and shape. Gait analysis often proceeds using markers to identify anatomical positions [6].

Given the coordinate data from the markers at the limb segment, it is an easy step to calculate the absolute angle of that segment in space. Calculation is starting with the horizontal equal to 0°.The relative ankle angle is defined as the angle between the tibia and an arbitrary line in the foot, knee relative angle is defined as the angle between the femur and the tibia. Hip relative angle is defined as the angle between the vertical and the femur.

After recoding the (x, y) coordinates of each segment marker, the position of the center of mass (X_{cg}, Y_{cg}) of each segment calculated as follow:

$$X_{cg} = X_{proximal} + R_{proximal}\left(X_{distal} - X_{proximal}\right) \quad \ldots\ldots (1)$$

$$Y_{cg} = Y_{proximal} + R_{proximal}\left(Y_{distal} - y_{proximal}\right) \quad \ldots\ldots (2)$$

where $X_{proximal}$, $Y_{proximal}$ are respectively the (X, Y) coordinates of a proximal point of a segment and X_{distal}, Y_{distal} are respectively the (X, Y) coordinates of a distal point of a segment. $R_{proximal}$ is the standard center of mass location from the proximal end of a segment [7]. The angular velocity and acceleration vectors ($\vec{\omega}$ and $\vec{\alpha}$, respectively) of the body segment along with the linear acceleration vector ($\vec{a}_x$,$\vec{a}_y$) is obtained simply by numerical differentiation using five point central difference formulae[8].

2.4 Calculation of Kinetic Quantities

Kinetic quantities were calculated using Newton's second law of motion which Incorporates inertia vectors [9]. All moments were taken about the center of mass of each segment. Positive forces were assumed to be in the upward direction and positive moments in the counterclockwise direction. Because the frequency of force plate data differs from the frequency of camera(default sampling rates of camera and force plates are 25 and 50 Hz, respectively), down sampling is performed using the MATLAB command "resample".

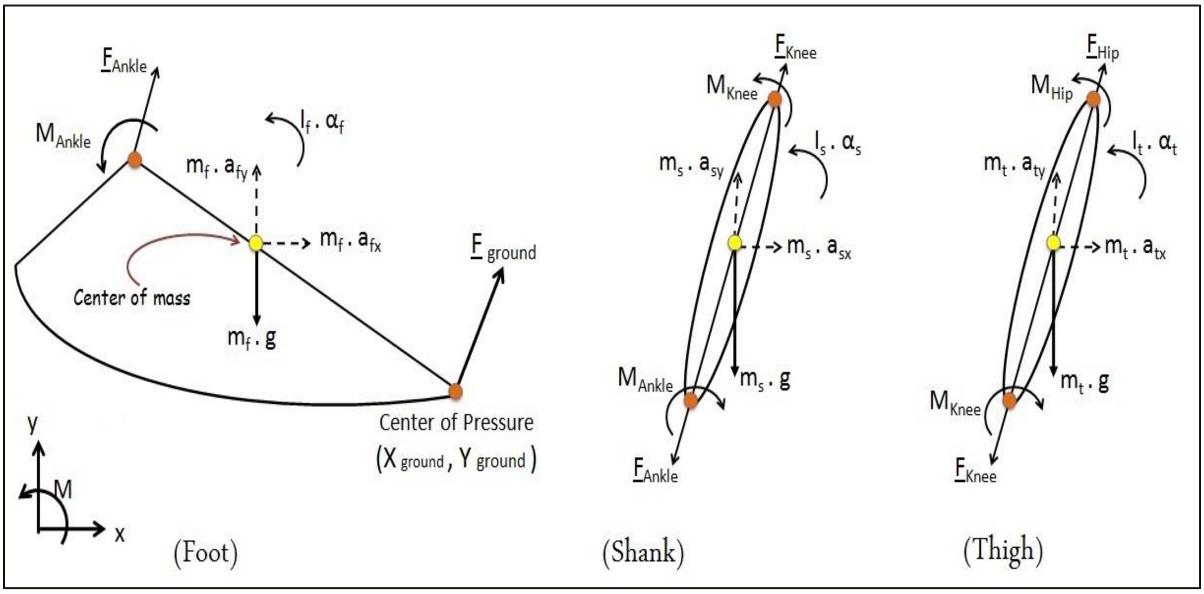

Figure 3: the free body diagrams of the lower extremity segments (foot, shank, and thigh) that demonstrate each segment forces isolated from the other segment forces.

With known ground reactions, the solution of equations of motion is initiated from the most distal segment, recursively progressing up to adjacent segments until finally, the forces and moments acting on the desired joint are obtained. The moment of a force (*M*) is defined as the cross-product (×) of a position vector (*r*) and its force (*F*),i.e.,

$$M = [r \times F] = r_x . F_y - r_y . F_x \qquad \dots\dots(3)$$

The position vector r is r =[X_{cg}, Y_{cg}] of the segment.

Forces and moments for the ankle:

F_{Ax}= m_f .a_{fx} –$F_{x(ground)}$ ….. (4)

F_{Ay} = m_f .a_{fy}–$F_{y(ground)}$ + m_f . g ….. (5)

F_{ankle} = [F_{Ax}, F_{Ay}]

M_A = I_f .α_f– [r_{ankle} × F_{ankle}] – [r_{ground}× F_{ground}] ….. (6)

Where r_{ground} =[COP_x, 0].

Forces and moments at the knee:

F_{Kx}= m_s .a_{sx}+ F_{Ax} ….. (7)

F_{Ky} = m_s .a_{sy}+ F_{Ay}+ m_s . g ….. (8)

F_{knee}= [F_{Kx}, F_{Ky}]

M_K = I_s .α_s+ M_A – [r_{knee}× F_{knee}] – [r_{ankle}× - F_{ankle}] ….. (9)

Forces and moments at the hip:

F_{Hx}= m_t .a_{tx}+ F_{Kx} ….. (10)

F_{Hy} = m_t .a_{ty}+ F_{Ky}+ m_t . g ….. (11)

F_{Hip}= [F_{Hx}, F_{Hy}]

M_H = I_t .α_t+ M_K – [r_{Hip}× F_{Hip}] – [r_{knee}× - F_{knee}] ….. (12)

3. Experimental work

The Gait analysis laboratory, located at the Medical Engineering Department/Al Nahrain university/Iraq, equipped with a digital camera (Sony) and two force plates. The system has the necessary software programs related to the force plates that operate during and after the data acquisition process. The final outputs of the system are kinematic and kinetic data which enable assessment and interpretation of gait patterns on quantitative bases.

The study was made on fourteen healthy normal volunteers (10 males and 4 females; age: 24.36∓4.53year, weight: 68.96∓12.71kg, height:

1.7∓0.091m), male with cerebral palsy (CP) (age: 34year, weight: 70.8kg, height: 1.655m), and an old female underwent unilateral (left) knee arthroplasty (UKA) (age: 63year, weight: 88.50kg, height: 1.625m). Passive seven markers (red round stickers) were used to identify the location of joints' center of rotation. Markers attached to the subject at the hip (right greater trochanter), right knee (lateral femoral condyle), left knee (medial femoral condyle), right ankle (lateral malleolus), left ankle (medial malleolus), right foot (fifth metatarsal head), left foot (left first metatarsal head).

The standard collection protocol is composed of static and dynamic trials. During both of the trials, videos of a number of passive markers attached to various landmarks on lower extremities are captured with a digital video camera (25 frames/second) placed 2.8m on the side of the force plates, fitted on a tripod with a height of 1.09m from the ground. Considering the low-pass nature of human gait, this value may safely be accepted to be more than twice of the highest frequency value possible for human gait. So, the sampling theorem is not violated. The static trial requires the subject to stand still for a specified duration. The static capture provides information on the participant's standing posture and the angle between the foot and horizontal. Following the static trials, participants were asked to perform walking trials on a 6 m walkway, barefoot and at a self-selected speed. During the dynamic trial, the force plate data and camera videos were collected.

4. Results

The Pedotti diagram of normal subject is shown in Figure (4). For the subject with cerebral palsy patient, the Pedotti diagrams (shown in Figure (5)) indicate that the difference between the loading on the right and left foot because of neural commands disorders. There was no full weight bearing take place during early stance for the right foot while full weight bearing occurred in the left foot. Also there was rapid decrease in center of mass acceleration during mid-stance for the left foot. The Pedotti diagram of the UKA patient is shown in Figure (6). The right foot showed nearly vertical bend of ground reaction force vectors for all stance phase except during toe-off phase. The patient used a forward trunk lean to move the body vector anterior to the knee joint axis to decrease the load on the joint. This imposes a greater flexor torque at the hip.

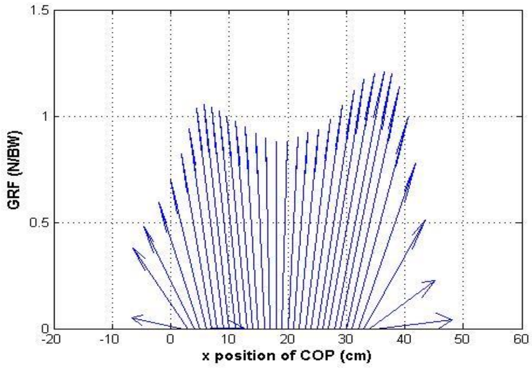

Figure 4: Pedotti diagram for normal subjects.

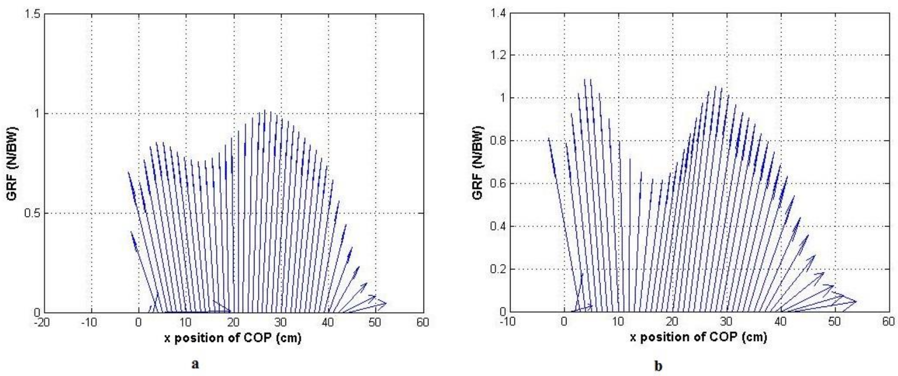

Figure 5: Pedotti diagram for cerebral palsy subject, (a) right foot and (b) left foot.

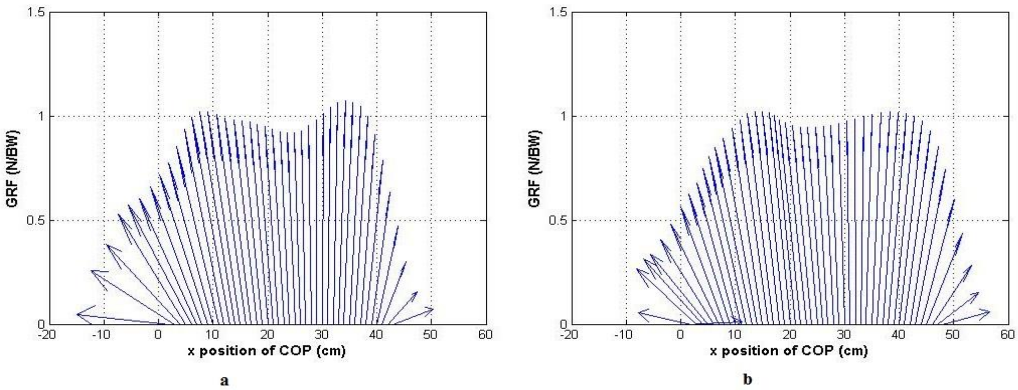

Figure 6: Pedotti diagram for UKA patient, (a) right foot and (b) left foot.

Cerebral palsy and UKA patients exhibit less range of motion at the knee joint and increase range of motion at the hip joint as shown in Figure (7). There is no planterflexion at heel strike in cerebral palsy patient. The tight knee extensors in cerebral palsy patient limit knee extension.

Initial contact is made with the forefoot in cerebral palsy patient. There were differences between right and left side for all abnormal cases (compensate for pain or discomfort in one limb by altering range of motion in both limbs so the time spent on the limb with no pain increased).

Mean joint forces during walking are summarized in table (1). The GRFs during gait are transmitted proximally to the rest of the body through the foot and ankle, compressing each joint along the way. Higher forces were observed on the left limb joints of the UKA patient.

Table 1: Mean (∓SD) of hip, knee and ankle joint forces (N/BW) during gait cycle.

Subjects	Normal	CP (right)	CP (left)	UKA (right)	UKA (left)
Hip Force	3.159∓0.147	2.937	2.934	3.253	3.415
Knee Force	3.538∓0.125	3.335	3.340	3.648	3.833
Ankle Force	3.738∓0.178	3.539	3.546	3.855	4.049

The joints' moments for normal and abnormal subjects are shown in Figure (8). For cerebral palsy and UKA patients, the loading response of stance phase (0-7% of gait cycle) shows little dorsiflexion moment. Cerebral palsy patient knee shows a high extensor moment during late stance, while the knee flexor moment for both cerebral palsy and UKA was small. Also, cerebral palsy shows hip extensor moment during mid-stance, while small extensor moment was appeared during swing phase for cerebral palsy and UKA patents.

5. Discussion

In general, the measured lower body kinetics for normal subjects matched with finding reported in the literature [2, 3, 10, and 11]. The difference were found at the final points (95% - 100% of gait cycle) this is largely due to parallax and perspective errors (occurred when the object moved out of the calibration plane). Considerable differences were found in the magnitudes of joint angles and forces with the previous literature [2 and 3]. Since the joints' angles and forces are largely sensitive to the velocity of walking [12], also the differences stem partly from the different methods used to determine the force and partly by differences in the physiology of the individuals.

Accuracy of two dimensions (2D) gait analysis is crucially dependent on correct marker placement, anthropometric errors, digitizing process, parallax and perspective errors and further errors were introduced because movement may take place between the marker and the underlying clothing and skin. Since the cloth and soft tissues moved relative to the bone, acceleration of the segment mass center will be a function of both acceleration of the bone and that of the soft tissues. The skin motion artifacts may add high frequency noise in displacement data.

As seen from the results (Figure (8) and table (1)) of the UKA patient, the left side forces and moments were greater than in the right side. This is primarily due to loss of rehabilitation procedures after knee arthroplasty, and because the contribution of the articular cartilage lesion. From the angles curves and time duration of the cerebral palsy patient, the right side was more affected than the left side and this leads to a compensatory mechanism resulted the abnormal movement of both right and left side and that matched with the literature [13].

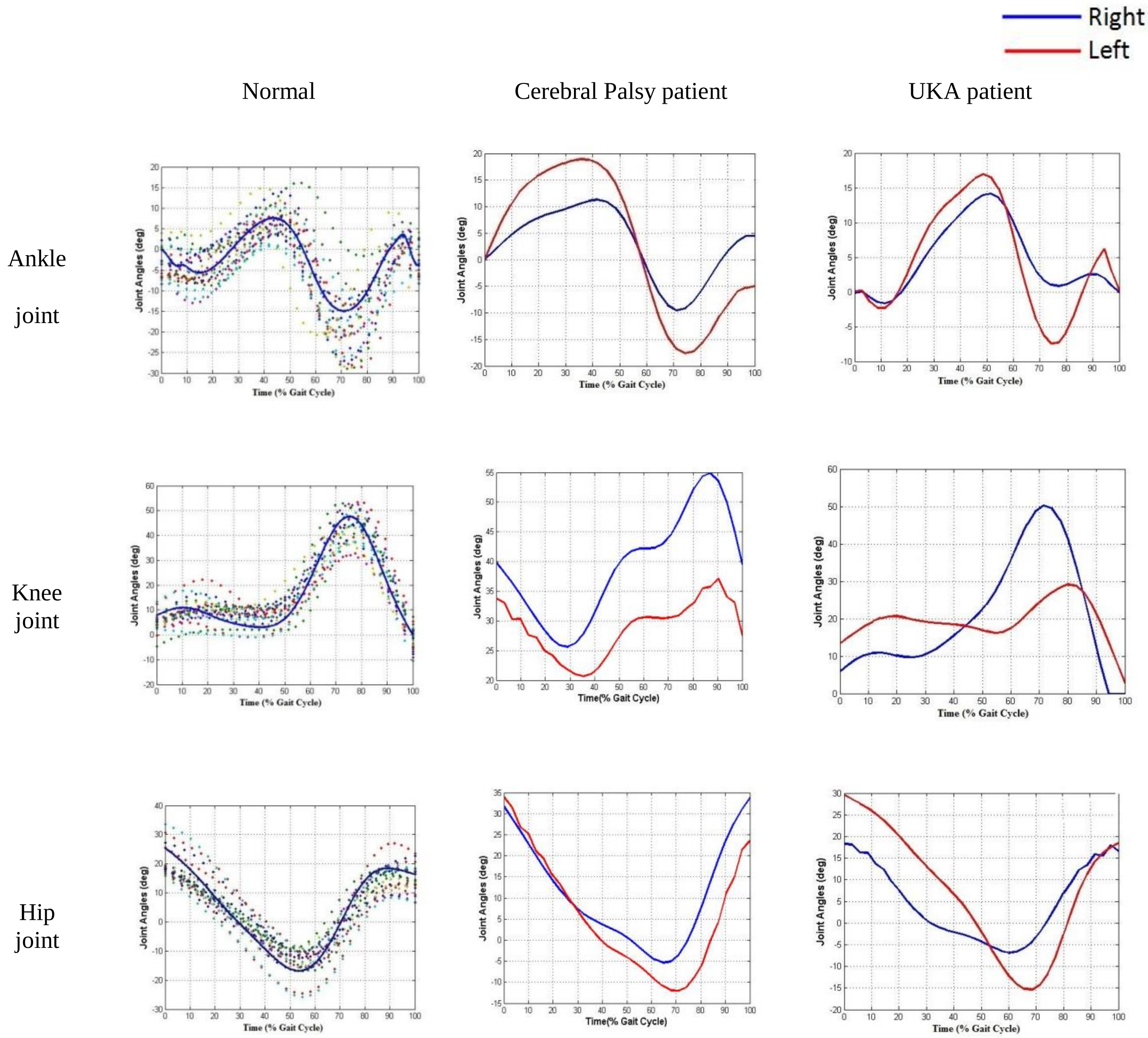

Figure 7: Ankle, knee and hip joints absolute angles during gait cycle.

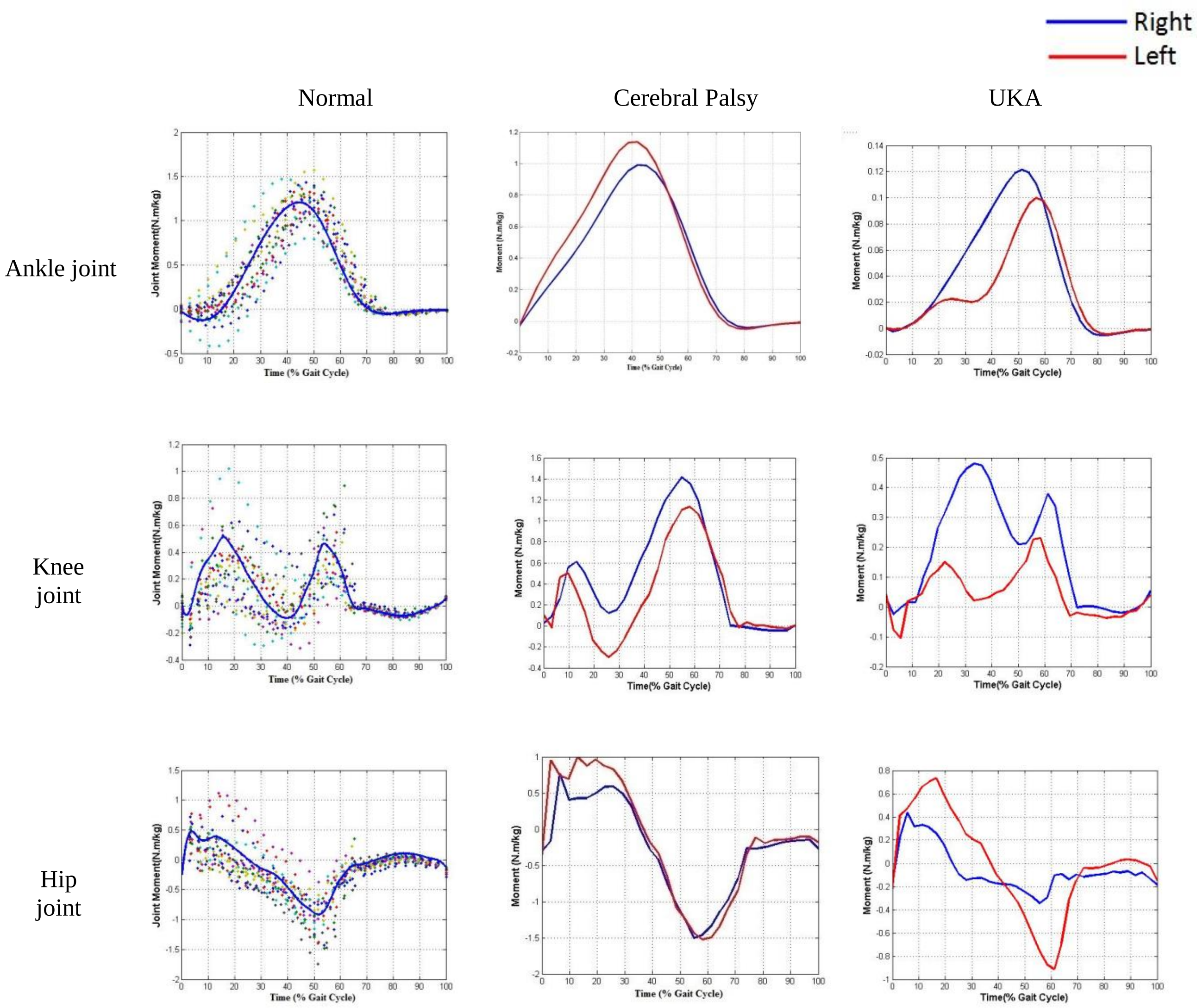

Figure 8: Ankle, knee and hip joints moments during gait cycle.

6. Conclusions:

Using mathematical models of the lower limb to examine the forces at the joints provided a valuable insight into internal loading conditions. The advantages of the method are that: no encumbering apparatus is attached to the patient, the method is painless and does not involve any discomfort for the patient, multiple measurements may be made in a single recording session, and recordings of both extremities can be made at the same time. Implanted joints bare more forces and moments than normal joints.

7. Recommendations for future work:

1. This approach provides the net muscle activity at the joints, and therefore, cocontractions were not identified. Information on cocontraction could be obtained using electromyography (EMG).
2. Studying the dynamics of movement in 3D for better understanding of dynamic stability during walking and for positioning of joints' centers accurately especially for hip and the Meta2.
3. Combining Pedotti diagram with video for determining the direction of ground reaction force vectors with respect to joints. This procedure would be useful for determining the alignment and stability for joint prosthesis.
4. Studying kinetics of other human daily activities such as stair climbing, running, and lifting objects for different speeds. Further researches are needed on a wider range of populations that include pediatric, patients with joint arthroplasty, amputees and other pathological groups.
5. Using artificial neural network to perform automated diagnosis of gait pattern and to find new clinical indicators for interpreting quick and objectively the large amount of information obtained in a gait lab.

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التحليل الحركي خلال دورة المشي للاشخاص السليمين والمرضى

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الخلاصة:

تحديد المشية المرضية هو من أكثر تطبيقات تحليل المشي. الغرض من هذه الدراسة هو تحرّي القوى المحركة للإنسانِ خلال دورة المشي الكاملة. أجريت تجارب التحليلِ الحركي الثنائي الأبعاد باستخدام كاميرا فيدوية رقمية (Sony) ومنصتين لقياس القوة نوع AMTI. تم الحصول على البيانات الحركيّة الكينماتيكية مِنْ بيانات مسارات سبعة علامات لاصقة بأستخدام برنامج SkillSpector (الإصدار 1.2.4).اعتمد برنامج MATLAB (الإصدار 8.1) لغرض الحصول على مخطط بيدوتي ولغرض حسابات معادلات الداينامیك المعكوس. أجريت تصفية رقميه للبيانات الكينماتيكية بأستخدام (zero phase Butterworth filter) بتردد (4.5 Hz). تم الحصول على الإزاحة الزاوية وقوى رد الفعل والعزم لمفاصل الورك والركبة والكاحل خلال المشي. أجريت دراسة المشي على أربعة عشر متطوعاً ومريضاً واحداً مصاباً بالشلل الدماغي وامرأةً كبيرة السن أجري لها تبديل مفصل الركبة الأيسر.هذه البيانات يمكن الاعتماد عليها في دراسات علم الأمراض, كمدخلات إلى نماذج التصميم للمفاصل ومحاكيات المفاصل الميكانيكية.