Kinematics and Dynamic Stability of the Locomotion of Polio Patients*†

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Abstract

The study reported in this article was conducted to propose a set graphical and analytical tools and assess their clinical utility by analyzing gait kinematics and dynamics of polio survivors. Phase plane portraits and first return maps were used as graphical tools to detect abnormal patterns in the sagittal kinematics of polio gait. Two new scalar measures were introduced to assess the bilateral kinematic symmetry and dynamic stability of human locomotion.

Nine healthy subjects and seventeen polio patients were involved in the project. Significant increases in the knee extension and ankle plantar flexion of polio patients were observed during the weight acceptance phases of their gait. Polio patients also exhibited highly noticeable excessive hip flexion during the swing phase of their ambulation.

Using the proposed symmetry measure, we concluded that polio patients walked less symmetrically than normals. Our conclusion, however, was based on the bilateral symmetry in the sagittal plane only. Finally, we observed that polio patients walked significantly less stably than normals. In addition, weaknesses in lower extremity muscles of polio patients were found to be an important factor that affected stable ambulation.

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1 Introduction

After the invention of Salk and Sabin vaccines, poliomyelitis was generally considered a non progressive disease. Recent studies, however, have shown that approximately one fourth of paralytic polio survivors face renewed health problems typically 30 to 40 years after the onset of illness (post-polio syndrome). Some of the polio survivors who demonstrated good recovery are now unable to perform even normal daily activities. The most commonly observed problems in polio patients were listed by Halstead (1985) as excessive fatigue, pain in muscles and joints, weakness in affected muscles, loss in gait functions, and cold intolerance. Agre et al. (1991) showed that symptomatic post-polio subjects exhibit %30 - %40 decrease in isometric muscle strength and work efficiency compared to asymptomatic subjects.

Causes of the late complications in polio survivors have not been completely determined yet. Several explanations have been proposed by various investigators in the field (Perry et al., 1988, Halstead, 1985). It is assumed that motor unit dysfunction, musculoskelatal overuse, and musculoskelatal disuse are the three fundamental processes that play important role in the generation of the syndrome (Munsat, 1991). Some authors suggested that the aging process also contributes to the late complications. In addition, patients who experienced the polio at younger ages were found to be more likely to exhibit the renewed pathologic symptoms.

Other investigators focused their attention on developing scoring systems to assess the severity of problems facing polio survivors. Perry et al. (1993) showed that symptomatic polio patients had less efficient gait compared to asymptomatic ones. They noted that the gait characteristics of polio survivors such as gait velocity, stride length and cadence were lower than the values of normals. In addition, EMG studies of the extensors of polio patients demonstrated excessive muscle activity in the lower extremities during locomotion. Perry et al. (1988) grouped the post-polio patients according to the strength of the quadriceps and calf muscles. They used foot-switches, electrogoniometer and dynamic electromyography to measure the functional capabilities of patients. However, they focused on the activity of particular muscles rather than joint kinematics while describing the dynamics of polio gait

The present study has two main goals: 1) propose a set of analytical and graphical techniques (partially based on our previous studies) to study the human locomotion, 2) demonstrate the clinical utility of the proposed methods by quantitatively assessing well known aspects of polio gait. Phase plane portraits (Hurmuzlu et al., 1994) and first return maps were used to graphically highlight the abnormalities in the lower limb kinematics of polio gait. A new scalar measure was utilized to evaluate the bilateral symmetry of individuals who were affected by polio. We used a second scalar measure (a modified version of the one proposed in Hurmuzlu and Basdogan, 1994) to assess the dynamic stability of polio gait. The study presented in this article consisted of testing a group of normals (nine individuals) in order to establish a frame of reference for normal profiles. Then, a population of seventeen polio patients were tested. These individuals were grouped according to their muscle strengths into five subgroups.

The joint kinematics and gait dynamics of polio patients in various groups were then compared to that of normals and patients in other groups by using the graphical and analytical tools that are proposed in this article.

2 Experiments

2.1 Subjects

Nine healthy subjects (3 women, 6 men) with no known history of gait abnormalities were included in the study. The average age, height and weight of the normal subjects were 26 ± 4 years, 1.74 ± 0.10 m, and 74.6 ± 14.1 kg respectively.

There were 17 polio survivors (4 men, 13 women) volunteered in this study. All of the patients had their polio at least 25 years ago. The mean age of the patients was 52 ± 10 years. The average height and weight of the patients were 1.67 ± 0.11 m and 67.5 ± 14.7 kg respectively. At the time of the study, two patients were using knee-ankle-foot orthoses (KAFO) to prevent knee flexion contractures and three patients were using ankle-foot orthoses (AFO) to complement weak calf muscles. Some of the other patients had used either KAFO or AFO previously. However, patients were not allowed to use orthoses during the experimental tests.

All subjects were informed of the nature and purpose of the research and were asked to sign consent forms. Height and weight of the subjects were measured before the experiment. Recent and previous history of gait abnormality of each polio subject including the age, onset year of illness, type of assistive device, detailed description of pain, weakness, fatigue at each joint, and leg lengths were recorded in a file for future reference.

2.2 Protocols and Instrumentation

Subjects were asked to walk at a comfortable but constant walking speed, for a distance of no more than 16 passes down a 20 meter track. The average walking speeds of normals and polio patients were measured as 1.13 ± 0.12m/s and 1.11 ± 0.22m/s respectively. During the experimental tests, each polio patient was monitored closely from the control room and test was stopped immediately when the patient felt discomfort, pain, exhaustion or did not want to continue. All the normal subjects and fifteen of the polio patients completed the sixteen passes without any difficulty. Two polio subjects, however, asked to stop the test after completing ten passes.

Foot switches were attached to the heel and toe of each foot to count the number of steps and to time the specific events of the step cycle. Subsequently, before the testing began, the foot switches were further aligned (specially for polio patients) in order to ensure proper capture of the foot contact events.

The bilateral, tri-axial electrogoniometer system by Chattex Corp. was used to measure hip, knee, and ankle motion kinematics. A set of three potentiometers, with orthogonally arranged axes were attached to each lower extremity joint. The proximal end of each tri-axial group used a parallelogram linkage to correct the unavoidable translational motion in the action of the joints. First, anatomical landmarks were used so that the sagittal potentiometers were positioned properly with respect to the corresponding joint axes. Then, the other potentiometers were adjusted such that they best follow the anatomy of the joint. Significant deviations of joints from neutral standing position were measured and recorded for each polio subject. A menu-driven real time data acquisition system was developed to display the joint kinematics and foot-switch information on the computer screen with a sampling rate of 100 Hz. The joint velocities were obtained by first performing a second degree binomial smoothing and then numerically differentiating the position data.

2.3 Grouping of Polio Patients

Muscle strength of all patients were manually examined by a physical therapist. Strength of hip, knee, and ankle flexors and extensors of patients were graded bilaterally according to the Lovett system of manual muscle grading. The scale consists of six grade levels: Normal (N), Good (G), Fair (F), Poor (P), Trace (Tr), and Zero (0). In addition, intermediate levels were assigned with a plus (+) or minus (-) signs. The patients were classified according to their lower extremity strength. We used hip flexor, knee extensor (quadriceps), and ankle plantar flexor strength as determining factors in the grouping. Muscle grades which were equal or higher than Fair (F^+) grouped in the strong category. Similarly, grades ranged from Zero (0) to Fair (F) were considered in the weak category. Keeping this scale and classification system in mind, patients were first divided into two categories according to their hip flexor strength. Then, patients with weak hip flexors were subdivided into four groups based on the activity level of their quadriceps and calf muscles: strong knee extensors and ankle plantar flexors (SKE/SAP), strong knee extensors and weak ankle plantar flexors (SKE/WAP), weak knee extensors and strong ankle plantar flexors (WKE/SAP), and weak knee extensors and weak ankle plantar flexors (WKE/WAP). We should note that, the weaknesses in the muscle strengths of the polio subjects in our study were confined to one side only. Accordingly, the grouping of the subjects was based on the weak sides of the individuals. Table 1 describes the grouping of patients according to this classification.

	GROUP TYPE	**NUMBER OF PATIENTS**
Weak Hip Flexor WHF	Group I WKE/WAP	6
	Group II SKE/WAP	2
	Group III WKE/SAP	2
	Group IV SKE/SAP	3
Strong Hip Flexor SHF	Group V SKE/SAP	4

Table 1 Grouping of polio patients

3 Analyses

3.1 Describing the Joint Kinematics of Polio Patients Using Phase Plane Portraits

In this section we present a kinematic analysis based on the phase plane portraits and first return maps of the lower extremity joints of the subjects participated in the study. Phase plane portraits are plots of angular displacement versus velocity at individual joints. A detailed discussion about the advantages of using phase portraits and first return maps can be found in Hurmuzlu et al. (1994).

The phase plane portraits and first return maps presented in this section were obtained from averaged steady state kinematic data of the subjects. This was realized by initially extracting the data that corresponded to the fourth and fifth gait cycles of each pass of each subject (we assumed that the individuals attained steady locomotion in four cycles). We define the gait cycle as two consecutive step cycles that begin from a heel strike at a particular side. Interpolation was used to obtain 400 evenly spaced data points spanning over the duration of the two cycles. Steady state profiles of kinematic variables of each subject were obtained by averaging the data at the 400 points over all the passes performed by the individual. Phase portraits were obtained by plotting the kinematic data that corresponded to the first averaged cycle of each set. The first return maps (Poincaré maps) of joint coordinates and velocities were obtained by periodically sampling the two cycles at heel strike, maximum ankle plantar flexion at stance, and maximum hip flexion during swing. More specifically, for each subject a given kinematic variable was sampled once within each gait cycle at the particular instance. This resulted in a pair of values for each kinematic variable, which was a first return point representing the steady state (the first point was the ordinate and the second point was the coordinate). For example, in Fig. 3 we observe that each subject is represented by a single point which is almost on the diagonal. Ideally, at steady state a first return point should have been exactly on the diagonal because of periodicity. As a result of the experimental errors, slight deviations from the diagonal line did occur. The general tendency of the points to be near the diagonal, however, justified our expectation of observing steady gait beyond the fourth cycle.

We note that the kinematic variables depicted in Figs. 3 and 4 corresponded to the dominant sides of the normals and the weak sides of the polio patients. Positions in the figures are with respect to neutral stance. Furthermore, the range of the axes are the maximum range of variation of the respective joint angels for all subjects. For example, to obtain the axes ranges of the sagittal knee position (Fig. 3.a) we computed the range of variation of the sagittal knee position during the steady locomotion of each subject. Then the overall minimum and maximum of this data set were computed to obtain the axes ranges.

First, the joint kinematics and phase plane portraits of normals are presented to provide a frame of reference for the polio patients and verify the experimental measurement system (Fig. 1). The displacement profiles had a reasonable agreement with previously reported results such as the ones given in Whittle, 1991. The foot contact events were marked on each phase portrait to compare the foot timing events. In this figure, solid and dashed lines are averaged joint profiles and standard deviations of the eight healthy subjects respectively.

As far as the polio patients are concerned, due to impaired or abnormal functioning of musculoskelatal system, it is known that they exhibit anomalous joint motions to achieve ambulation. Previously documented (Perry, 1992) and relatively well known characteristics of polio gait can be enumerated as follows :

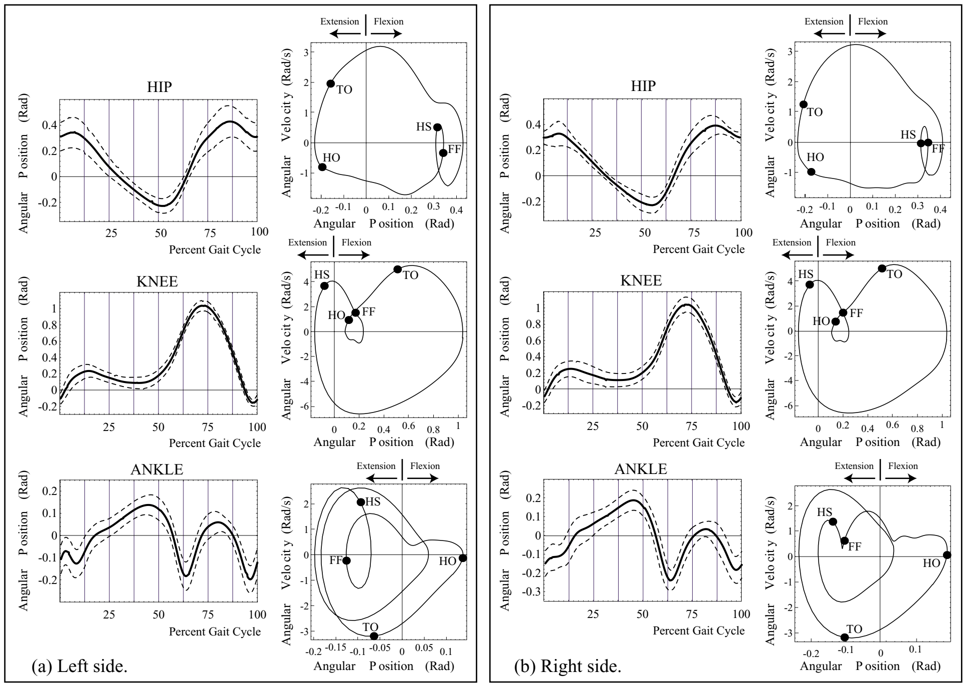
1. *Excessive knee extension and inadequate knee flexion*

   The knee is often the most affected joint and weakness in the quadriceps (knee extensor) is commonly observed in the patients. Excessive knee extension and inadequate knee flexion during the loading phase (heel strike to mid stance) can be attributed to the weaknesses in quadriceps and hip flexors. (the main function of quadriceps is to stabilize the knee motion during stance phase of gait cycle). The problem can be minimized by using knee-ankle-foot orthoses.

2. *Excessive hip flexion during swing*

   During the swing phase, polio patients are known to exhibit excessive hip flexion. Excessive hip flexion is used to gain momentum for adequate knee flexion and toe clearance off the floor.

3. *Excessive plantar flexion*



(a) Left side.
(b) Right side.

Figure 1: Saggital joint kinematics and phase plane portraits of normal subjects. Solid and dashed curves are averaged joint profiles and standard deviations of the eight healthy subjects respectively.

> Another well known problem is the weakness in ankle dorsiflexion. Compounded by weakness in quadriceps, this results in excessive plantar flexion. With the use of ankle-foot orthoses, excessive plantar flexion of ankle can be prevented, which also minimizes the excessive hip flexion. Perry (1992) demonstrated that ankle plantar flexion strength most frequently was the strongest determinant of the patients stride characteristics.

Excessive knee extension during the loading phase was clearly seen in the gait of patients with weak quadriceps. The effect was observed from the sagittal knee portraits by comparing the rightward shifts (Fig. 2.b) of the foot contact events of the polio plots during the single support phase. For the particular subject, the entire portion of the trajectory from heel strike (HS) to heel off (HO) was in the left quadrant (negative coordinate values, i.e. extension) of the phase plane. Inadequate knee flexion during the loading phase was also observed. This can be seen by comparing the relative distances between the heel strike and foot flat (FF) instants of normal and polio portraits (Fig. 2.b).

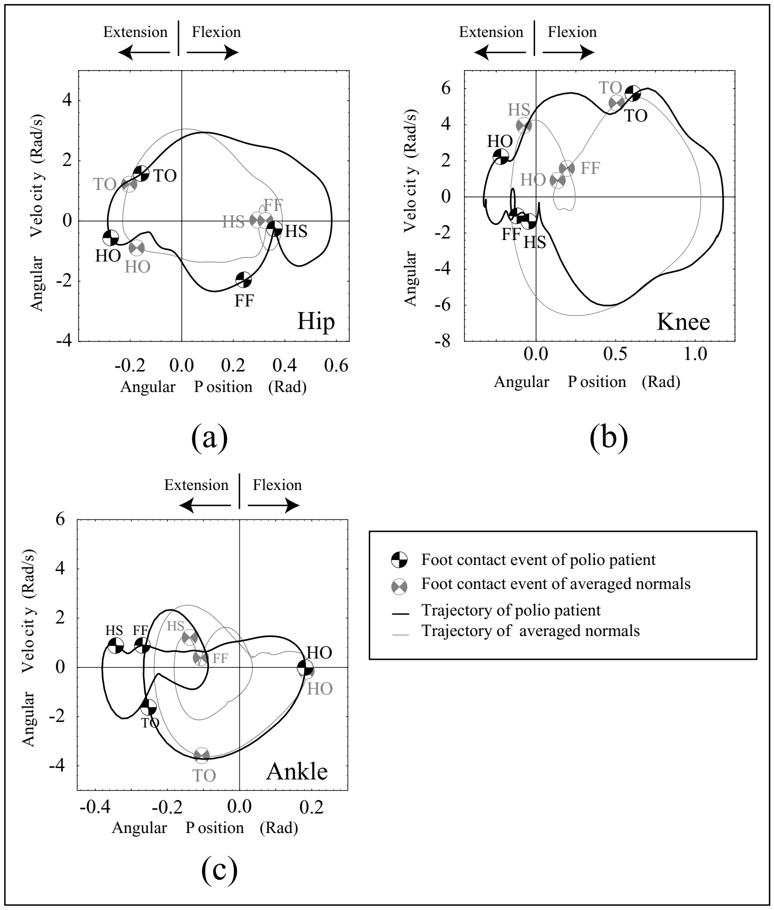


Figure 2: Saggital steady-state phase portraits of a post-polio patient in group I

One of the better known consequences of post-polio syndrome is the damage that is caused to the knees in the advanced stages of the disease. The phase portraits depict the velocity profiles during the load acceptance phase. We observed that in normal subjects heel strike sagittal knee velocities were at significantly higher positive values (the vertical elevation of HS in Fig. 2.b) than those recorded for polio patients. In normals subjects, a deceleration of the knee velocity can be clearly observed following the heel strike event. Yet, despite this deceleration, the velocity at the onset of the heel strike was sufficiently high to keep the knee in flexion throughout the load acceptance phase (despite a brief reversal in velocity during the mid stance period). In polio patients, however, the knee velocities during heel strike had smaller positive values, or even negative values (Fig. 2.b). Consequently, as a result of the deceleration in the knee that follows the heel strike event in polio gait, the knee remains to be in extension during the load acceptance phase. These effects can be better seen by studying the averaged first return maps depicted in Fig. 3.a. There was an observable shift toward the lower left corner in the sagittal knee extensions and knee velocities of polio patients at heel strike.

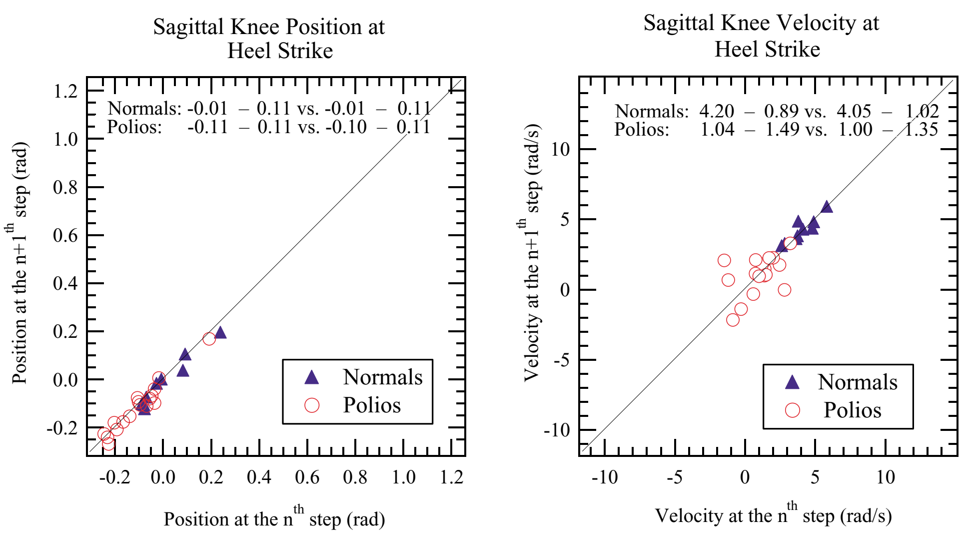


Figure 3: Steady-state first return maps of averaged sagittal knee kinematics

The trend of increased ankle plantar flexion during the weight acceptance phase was observed on the sagittal phase portraits of the polio patients. The rightward shifts of the foot contact events (HS-FF portion of the trajectory in Fig. (2.c)) during the single support phase was clearly observable in most of the individuals with polio. The first return map depicted in Fig. 4.b clearly demonstrates this effect.

Excessive hip flexion during the swing phase was observable on the sagittal hip portraits of the most of the polio patients (Fig. 2.a). The first return map shown in Fig. 4.a, depicts a clearly observable upper right shift in the hip flexion of polio patients.

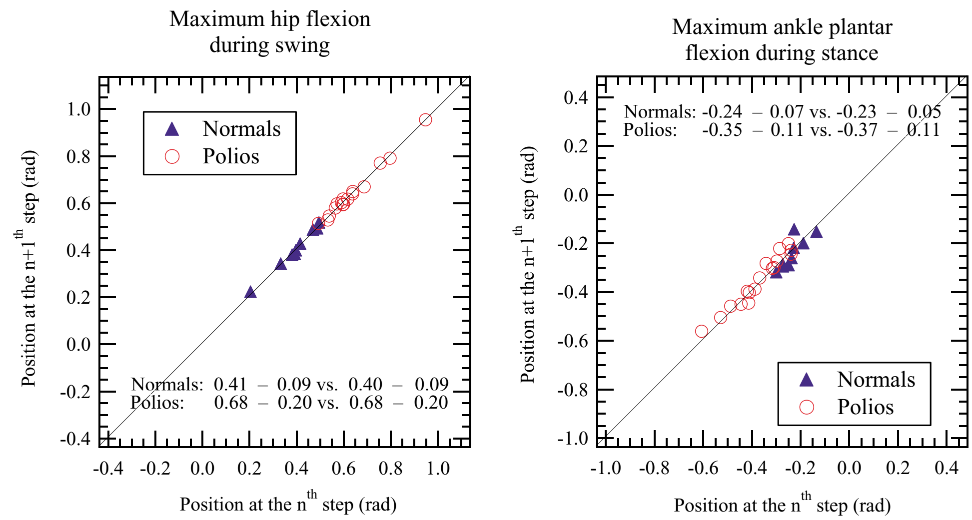


Figure 4: Steady-state first return maps of extremum values of hip and ankle rotations

3.2 Bilateral, kinematic symmetry

Bilateral symmetry in ambulation is one of the fundamental properties that is often used by clinicians to evaluate gait disorders. Quantification of asymmetric joint motions using regular time versus joint rotation plots may not be very easy. A quantitative measure of symmetry that can be used to distinguish one subject from another would greatly facilitate this process. Hannah et al. (1984) compared the right and left sagittal, transverse, and coronal plane motions in time and frequency domains. They developed indices to reflect bilateral symmetry in each domain. Their indices, however did not account for the joint velocities.

In this section, we propose a new measure to evaluate the kinematic symmetry of the lower limbs of the human body. We used the areas enclosed by the phase plane curves to estimate the degree of symmetry in lower extremities. The measure was defined based on the differences in the areas enclosed by phase trajectories which were generated by the left and right side sagittal motions of lower extremities.

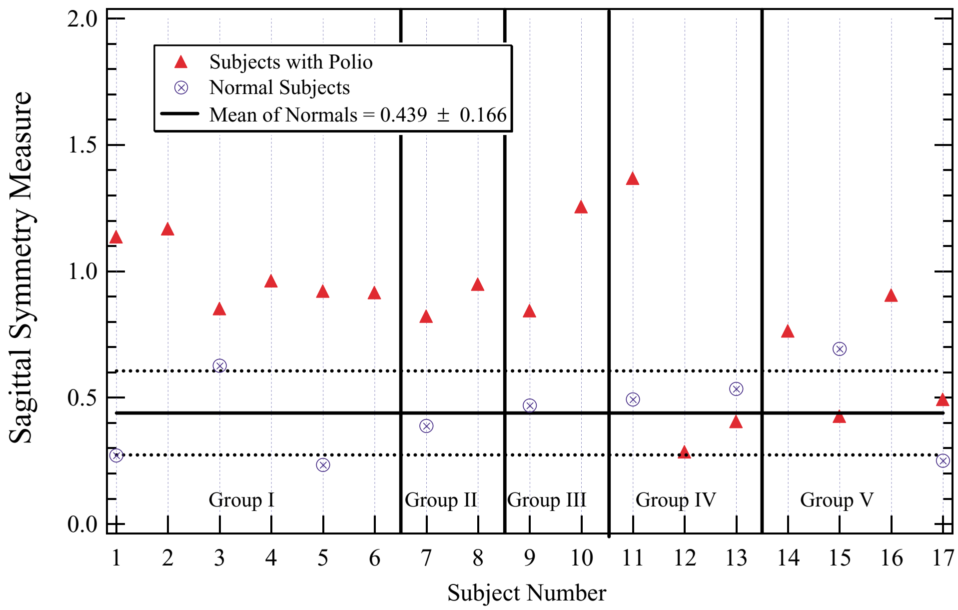


Figure 5: Symmetry measures of the subjects of the study obtained by using the proposed method

We defined the symmetry measure ψ_i^j at a particular joint, about a given axis direction as,

$$\psi_i^j = \frac{\mid {}^{\mathrm{L}}\Upsilon_i^j - {}^{\mathrm{R}}\Upsilon_i^j \mid}{\max\left[{}^{\mathrm{L}}\Upsilon_i^j \, , \, {}^{\mathrm{R}}\Upsilon_i^j \right]}, \tag{1}$$

where, ${}^{\mathrm{L}}\Upsilon_i^j$ and ${}^{\mathrm{R}}\Upsilon_i^j$ are the left and right phase plane areas enclosed by trajectories (computed numerically from averaged, steady state data). The subscript i stands for the particular joint (hip, knee, or ankle) of interest. Whereas, the subscript j represents the chosen direction (s for sagittal, c for coronal, and t for transverse).

According to its formulation, ψ_i^j is a number that varies between 0 and 1. In the case of perfect symmetry, the difference between the areas will be zero leading to a diminishing symmetry measure. The worst case, when one the of the joint is frozen (zero area in the phase plane), the measure would be equal to unity. In general, lower values of the measure signifies a higher level of bilateral symmetry.

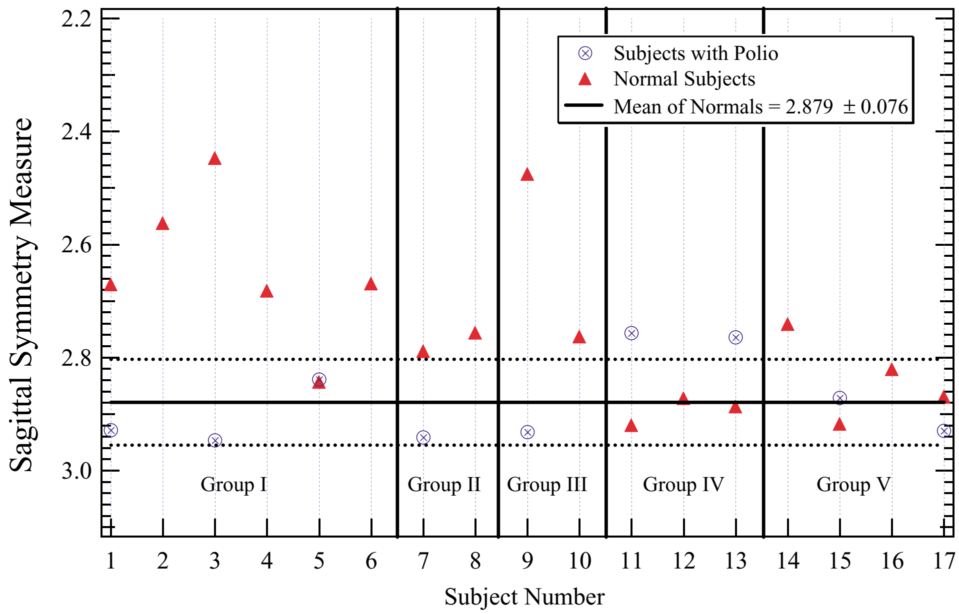


Figure 6: Symmetry measures of the subjects of the study obtained by using the method given in Hannah and Morrison (1984)

Similarly, an overall measure of the symmetry in the sagittal plane can be defined as the summation of symmetry ratios calculated at particular joints,

$$\Psi^s = \psi^s_{hip} + \psi^s_{knee} + \psi^s_{ankle} \tag{2}$$

In the present study, we calculated the symmetry measures at each joint as well as the overall symmetry measure for each polio subject. Figure 5 depicts the overall sagittal symmetry measures for normal subjects and polio patients. The symmetry measures of polio patients ($\Psi^s = 0.850 \pm 0.304$) were significantly higher ($p < 0.01$) than that of normals ($\Psi^s = 0.439 \pm 0.166$). Moreover, patients in group I ($\Psi^s = 0.991 \pm 0.129$) had higher symmetry measures ($p < 0.01$ and $p = 0.1$) than that of normals and subjects in group V ($\Psi^s = 0.646 \pm 0.226$). A special case was the symmetry of subject 11 ($\Psi^s = 1.36645$). The subject had excessive lateral trunk bending and his gait was the least symmetric among the polio patients.

We also computed the symmetry measures of the two subject populations by modifying the scheme proposed in Hannah and Morrison, 1984. We used their time domain method to compute the correlation coefficients for the sagittal joint rotations. Then, we added the respective coefficients to obtain a single measure for each subject (Figure 6). Noting that their method yields lower values for less symmetric gait, the symmetry measures of polio patients (2.747±0.146) were significantly lower ($p < 0.01$) than that of normals (2.879±0.076). patients in group I (2.647±0.133) had lower symmetry measures ($p < 0.01$ and $p = 0.042$) than that of normals and subjects in group V (2.838±0.076).

Although both measures predict statistically similar outcomes, the symmetry index proposed in this article can better distinguish between the polio and normal gait. For our method, 25% (4 out of 17) of the polio subjects had symmetry indices in the normal range. Whereas, for the other method, symmetry indices of almost 50% (8 out of 17) of the polio subjects were in the normal range. Furthermore, our measure clearly distinguished the polio subject with excessive lateral trunk bending from other subjects, while the other method placed the subject in the normal range.

4 Dynamic Stability of Polio Patients

In this section we study the dynamic stability of Polio patients using a modified version of the procedure that was presented in Hurmuzlu and Basdogan (1994). The procedure is based on the local stability of Poincaré (first return) maps. Here, we will briefly outline the basic aspects of the methodology, and refer more interested readers to Hurmuzlu and Basdogan (1994) for the remaining details.

We assume that a walking human being can be represented by a 36 dimensional state vector, $\mathbf{x}$, given by,

$$\mathbf{x} = \left[\phi_1, \ldots, \phi_{18}, \dot{\phi}_1, \ldots, \dot{\phi}_{18}\right]^T \tag{3}$$

where, ϕ_i are the measured joint rotations and $\dot{\phi}_i$ are the respective joint velocities. Then, we can construct a nonlinear map to represent the dynamics of the system that will be in the following form,

$$\mathbf{x}^{n+1} = \mathbf{f}(\mathbf{x}^n) \tag{4}$$

where, $\mathbf{x}^n$ and $\mathbf{x}^{n+1}$ represent the state vector sampled at the Poincaré section during the n^{th} and $\mathrm{n}+1^{\mathrm{th}}$ cycles respectively and $\mathbf{f}$ is an 36×1 dimensional nonlinear function. When the system attains a dynamic equilibrium (steady walking), the sampled state vector $\mathbf{x}^*$ satisfies,

$$\mathbf{x}^* = \mathbf{f}(\mathbf{x}^*) \tag{5}$$

because of periodicity. In addition, any perturbation $\delta\mathbf{x}^n$ in the neighborhood of $\mathbf{x}^*$ evolves according to the linearized map given by,

$$\delta\mathbf{x}^{n+1} = \mathbf{J} \;\; \delta\mathbf{x}^n \tag{6}$$

where, $\mathbf{J}$ is the 36×36 jacobian matrix. The stability of $\mathbf{x}^*$ can be determined by the eigenvalues (characteristic multipliers) of $\mathbf{J}$, $\lambda_j,\; j = 1, 36$. The equilibrium is stable when all the multipliers reside in the unit circle of the complex plane (i.e. the magnitude of all multipliers are less than one). When the system is stable, the multiplier that is closest to the unit circle dominates the behavior of the system. Thus, one can compare the gait stability of two humans by comparing their largest multiplier. In that case, the individual with the multiplier closer to unit circle will be the less stable of the two. Accordingly, we define a new scalar measure of the dynamic stability of gait as,

$$\beta = \max(\{|\lambda_1|, \ldots, |\lambda_{36}|\}) \tag{7}$$

As far as the human system is concerned, it is extremely difficult to obtain an analytical representation of the function $\mathbf{f}$ and for that matter the matrix $\mathbf{J}$. Therefore, we developed a procedure to estimate $\mathbf{J}$ from experimentally acquired kinematic data and compute the β-measures. This procedure can be described as follows:

1. Kinematic data were extracted at the maximum sagittal knee flexion of the dominant sides of the normal subjects and the strong sides of the polio subjects. Data only from the first four cycles of each pass were included (more on this later). The resulting data were arranged in the form of an $\mathrm{m} \times 4 \times 36$ dimensional array $\mathbf{A}_{i,j,k}$. The first dimension m, was the number of passes performed by each subject. The rows of $\mathbf{A}_i$ were formed by using the successive values of the state vector $\mathbf{x}$ at the respective sections during pass i.

2. The components of the steady state were computed from

$$\mathbf{x}^*_k = \frac{1}{m}\sum_{i=1}^{m} \mathbf{A}_{i,4,k} \quad k = 1, 36 \tag{8}$$

In Eq. (8), we averaged the fourth step of every pass to estimate the steady state.

3. A new 3m $\times$ 36 matrix of errors $\delta\mathbf{A}$ was defined as,

$$\delta\mathbf{A}_{4(i-1)+j,k} = \mathbf{A}_{i,j,k} - \mathbf{x}^*_k \;\; i = 1, m, \;\; j = 1, 3 \;\text{ and }\; k = 1, 36 \tag{9}$$

and a 3m $\times$ 36 matrix of first return errors $\delta\mathbf{B}$ was defined as,

$$\delta\mathbf{B}_{4(i-1)+j,k} = \begin{cases} 0 & j = 3 \\ \mathbf{A}_{i,j+1,k} - \mathbf{x}^*_k & \text{otherwise} \end{cases}, \tag{10}$$
$$i = 1, m, \;\; j = 1, 3 \;\text{ and }\; k = 1, 36$$

which enforced a unique steady state for all passes performed by a particular subject. Here, the n^{th} rows of the matrices $\delta\mathbf{A}$ and $\delta\mathbf{B}$ represent the vectors $\delta\mathbf{x}^{\mathrm{n}}$ and $\delta\mathbf{x}^{\mathrm{n}+1}$ respectively.

4. The components of the Jacobian matrix $\mathbf{J}$, were computed from 36 linear fits of the form

$$\delta\mathbf{x}^{\mathrm{n}+1}_{\mathrm{j}} = \sum_{k=1}^{36} \mathbf{J}_{\mathrm{k,j}}\, \delta\mathbf{x}^{\mathrm{n}}_{\mathrm{j}} \;\text{ with }\; \mathrm{j} = 1, 36 \tag{11}$$

by using the experimental data stored in the matrices $\delta\mathbf{A}$ and $\delta\mathbf{B}$ and a least squares algorithm.

5. The β-measure was computed from the maximum of the magnitudes of the eigenvalues of the Jacobian matrix.

This procedure was based on the presumption that the subjects attained equilibrium in four gait cycles (justified earlier in the article). The fits were based only on the first four step cycles because we were mainly interested in the contraction rates of the trajectories to the equilibrium states. Normally, during a walking pass, each subject started from one end of the walkway at

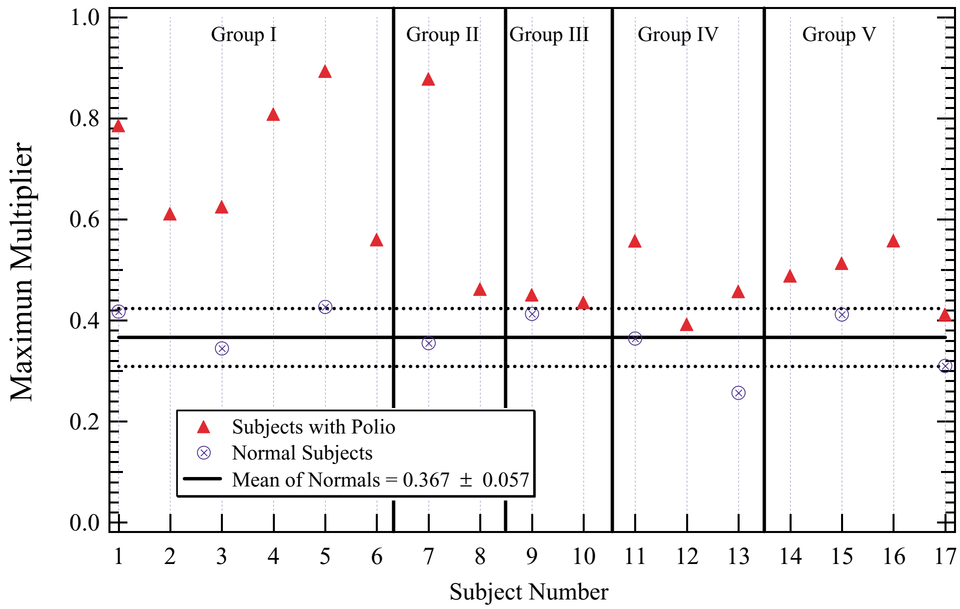


Figure 7: Maximum characteristic multipliers of the subjects of the study

a perturbed state (from rest or just after completing a turn). Then, they rapidly attained equilibrium. Therefore, the first four cycles were found to be sufficient in conducting the analyses.

The instance of maximum sagittal knee flexion was chosen to define the Poincaré section because this angle was known to be the most consistently measured joint rotation by electrogoniometers. In our earlier study (Hurmuzlu and Basdogan, 1994) we used foot contact events to mark the first return data. Later, we discovered that using maximum sagittal knee flexion produces more consistent first return data.

The next step was to compute the eigenvalues of the jacobian matrix for each subject participated in the study. Figure (7) depicts the maximum eigenvalues computed for the normal and polio subjects. The polio patients ($\beta = 0.581 \pm 0.164$) were found to be less stable ($p < 0.01$) than normals ($\beta = 0.367 \pm 0.057$) according to the stability measure proposed in this article. We also observed that patients in group I ($\beta = 0.713 \pm 0.133$) had less stable gait than normals ($p < 0.01$). Furthermore, subjects in group I were generally less stable ($p < 0.073$) than subjects in group V ($\beta = 0.492 \pm 0.061$). The results also showed that polio subjects with weak ankle plantar flexion (groups I and II, $\beta = 0.702 \pm 0.160$) were significantly less stable ($p < 0.015$) than groups II, IV, and V ($\beta = 0.473 \pm 0.060$).

We could not strongly conclude that patients who used assistive devices (patients 1 to 4 and 7), either KAFO or AFO, had significantly less stable gait than other polio patients ($p = 0.141$). In this group, patient 2 was using knee ankle foot orthosis. Patient 3 had calf atrophy and was an ankle foot orthosis user.

5 Discussion and Conclusions

The main objective of the present article is to develop new gait analysis techniques (partially based on our previous studies) and demonstrate their clinical utility in studying the locomotion of polio patients. For this purpose, two groups of subjects, nine normals and seventeen polio patients were tested and their kinematic data were recorded by using electrogoniometers. Furthermore, the polio patients were divided into five subgroups according to their muscle strength.

The gait of polio patients was compared to that of normals by using several graphical and analytical techniques. Phase plane portraits and first return maps were used to compare sagittal lower limb kinematics and detect anomalous trends in polio gait. The advantage of phase plane portraits is in the direct correlation of joint positions with the respective joint velocities. This is achieved by eliminating time from state variable plots, which is a classical technique in analyzing the behavior of periodic systems. First return maps provide means to isolate and compactly present the kinematic properties at specific instants of the gait cycle. New measures were introduced to assess the bilateral kinematic symmetry and dynamic stability of polio gait. The advantage of these measures over traditional time history plots are obvious. One cannot *quantify* symmetry and dynamic stability by observing time history plots. The measures proposed in the article is a cumulative representation of the information contained in thirty-six time history plots. This compression in the presentation of data has clear advantages in conducting a study that focuses on the gait symmetry of subjects populations.

Through the utilization of the tools proposed in this article, we were able to quantify previously noted characteristics of polio gait. To the best of our knowledge, these characteristics were never quantified for a population of polio subjects. We were able to quantitatively demonstare the following characteristics:

1. Knees of the polio patients during the load acceptance phase of gait underwent excessive extension compared to normals. Poincaré maps of the sagittal knee position and velocity at heel strike demonstrated significant reductions or even reversals in the knee velocities of polio patients. The reduction in velocity, accompanied with excessive knee extension at heel strike, caused the knees of polio patients to undergo excessive extension during the load acceptance phase.

2. Significant increases in the ankle plantar flexion of polio patients were observed during the weight acceptance phases of their gait.

3. Polio patients exhibited highly noticeable excessive hip flexion during the late swing phase of their ambulation.

We also established that polio patients walked less symmetrically than normals. We confined the symmetry analysis to the sagittal plane mainly because of the alignment difficulties associated with electrogoniometers. It is well known that electrogoniometers are better measurement devices in the sagittal plane than other planes. The symmetry measure introduced in this article depends on velocity as well as position measurements. We suspected that the symmetry measures were particularly susceptible to misalignments in the coronal and transverse planes. Better measurement devices such as optical systems may be necessary to compute the symmetry measures in planes other than the sagittal plane.

We have also established that polio patients walked significantly less stably than normals. The stability results also demonstrated that patients in group I were less stable than other polio subjects. In addition, weakness in ankle plantar flexors was found to be a strong determinant in diminishing dynamic stability. We should note that the stability measure is not susceptible to misalignments as the symmetry measure. Computation of the stability measures requires the capture of the degrees of freedom of the system. Thus, as long as three axes of rotation are measured in each lower extremity joint (assuming negligible translational degrees of freedom), the stability results would not be affected by the directional misalignments of the potentiometers.

There is a notable difference in the ages and genders of the normal and polio populations who participated in the study. The effect of age difference on the gait kinematics has been reported in Whittle, 1991 and several other references. It has been noted that the age related changes in gait take place between 60-70 years of age. The ages of four of the polio subjects in our study were in this range (subjects number 6,9,11, and 17)). Examining the Figs (5) and (7), one can observe that age within the polio group has a secondary effect on the symmetry and stability measures. Effects of gender on the kinematics and dynamic stability of normals were considered in Hurmuzlu and Basdogan, 1994 and Hurmuzlu et al., 1994. We did not observe statistically significant differences due to gender in these studies.

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