FACTA UNIVERSITATIS
Series: **Physical Education and Sport** Vol. 1, N° 8, 2001, pp. 25 - 35

Scientific Paper

KINEMATIC AND KINETIC ANALYSES OF GAIT PATTERNS IN HEMIPLEGIC PATIENTS

UDC 796.012 796.034-056.26

Mónika Horváth, Tekla Tihanyi, József Tihanyi

Semmelweis University-Faculty of Physical Education and Sport Sciences,
Department of Biomechanics, Budapest, Hungary
E-mail: tihanyi@mail.hupe.hu

Abstract. *As lack of standardization in the visual and computerized gait analyses different kinetic and kinematic gait analysis methods have been developed. The goal of this study was to further investigate the individually characteristics biomechanical deficits of hemiparetic gait pattern and the resulting compensations that compromise walking. The stance phase was more closely examined by the means of a force plate system and a motion analysis system. Data were obtained for both the affected and unaffected leg of 11 (9 males, 2 females) hemiplegic patients. We found that the shorter stance phase time for the affected side is related to the deficient ability to load and transfer weight through their affected leg. Significantly increased rate of force development was found during foot flat on the affected side while toe off was characterized by markedly lower force development. The impaired range of motion on the hemiplegic side was also leading to compensatory mechanism of the unaffected limb resulting abnormal movement of the ankle, knee and hip joints both the affected and unaffected side. Since during the analysis different patterns were separated and identified, it has become apparent that optimal treatment protocols for each pattern type should be developed.*

Key words: *hemiplegia, quantitative gait analysis, ground reaction force, range of motion*

INTRODUCTION

There are very diverse patterns of neurologic abnormality leading to the range of clinical manifestations of cerebral palsy. Clinical manifestations include impaired motor control that can be characterized by muscle weakness, altered muscle tone and abnormal movement patterns (Schroeder et al., 1995). Among the residual neurological deficits mainly the hemiparetic disturbances affect the function. These impairments limit the

Received November 25, 2003

ability to perform functional activities such as walking and self-care. The ability to walk and the symmetry outcome are the prime factors that determines whether a patient will return to the previous level of productivity after stroke. Walking is often the prime target of rehabilitation because of its importance to functional independence and a key ingredient in functional competency (Dettman et al., 1987). Hemiplegic gait has typical features that mostly can be recognized on visual inspection. A closer look upon the hemiplegic gait reveals that unfortunately the mechanism of gait disturbance differs individually. As lack of standardization in the visual gait analysis, methods may give uncertain results, different kinetic and kinematic gait analysis methods have been developed.

Several studies have clearly demonstrated that walking velocity is a key measure for the analysis of human gait. All of the methods commonly used to characterize walking in patients with hemiplegia, measurement of temporal parameters is generally considered the easiest to perform and the most clinically relevant (Wall et al., 1979; Bowker and Messenger 1988; Mizrahi et al.,1982a,b). Spatial-temporal gait measures include velocity, cadence, stride duration, individual phase duration, symmetry ratios and others. It has been asserted that walking speed is an effective indicator of the degree of abnormality in gait quality and overall functional status in hemiplegic patient (Brandstater et al., 1983; Eke-Okoro and Larson 1984; Wagenaar and Beek, 1992; Perry et al., 1995; Knuttson and Richard, 1979).Speed also has been widely used as a measure of patient status and treatment efficacy in clinical care and in research studies (Bohannon and Andrew 1990; Bohannon and Walsh 1992). The relationship between velocity and sixteen other spatial-temporal gait parameters (velocity, cadence, stride length, stride duration, step width, mean cycle duration, mean cycle length, single stance phase duration, stance phase percentage to cycle, swing phase duration, swing phase percentage to cycle, double support phase duration, double support phase percentage to cycle, hemiplegic and non-hemiplegic limb swing/stance phase ratio, swing phase symmetry ration, stance phase symmetry ration) were determined. Velocity was found to be significantly correlated with cadence, mean cycle duration, mean cycle length, hemiplegic and non hemiplegic limb stance phase duration and percentage, non hemiplegic limb swing phase percentage, double support phase duration and percentage, hemiplegic and non-hemiplegic limb swing/stance phase ratio and swing phase symmetry ration but not with the others (Elliot et al., 1997). Also, spatial-temporal variables of gait were investigated in order to assess the distribution of these variables according to functional ambulation category. Velocity, step-time, stride length and stride length in relation to lower extremity length proved to be valuable measures in the gait analysis, while cadence, step time and step time differential values seemed to be less important (Özgirgin et al., 1993). Using microcomputer-based method it is reported the hemiplegic children walked more slowly than normal children, with a shorter step length, reduced cadence, longer swing time and reduced maximum foot velocity on the side of the hemiplegia (Wheelwright et al., 1993).

Several studies have demonstrated the deficient ability to load the hemiplegic leg during walking and to effectively transfer body weight through the affected leg (Wall and Turnbull 1987; Lane 1978; Caldwell et al., 1986). It has been proposed that deficient ability to load the hemiplegic leg during walking contributes to gait asymmetries, particularly in the single support phase of the pattern. Even functionally ambulant hemiplegic subject demonstrate marked limitation in the capacity to shift weight and posses a reduced range of weight shift (George et al., 1996).Gait was analyzed with use of motion analysis, force-plate recordings and dynamic surface electromyographic studies of the muscles of the lower extremities. The result of the motion, electromyographic and force-plate studies showed markedly prolonged duration of the pre-swing phase on the hemiplegic side. This was associated with a delay in the initiation and a decrease in the speed of flexion of the hip during the swing phase (Kramers et al. 1996; Olney et al., 1991).

Studying the gait outcome of the patients in the acute phase post-stroke is more relevant to therapist as this is the period when most of the recovery takes places. Whether some kind of changes are still significant in later stages is still not known as only few studies evaluated the gait outcome of ambulatory patients in a chronic phase of recovery. Also very few studies examined the biomechanics of the unaffected limb. Kerrigen et al. (1999) found that various compensations occurring in the unaffected limb might strain or fatigue the muscle or ligaments and might predispose to joint injury in that limb.

Herzog et al. (1988) proposed a measure of symmetry/asymmetry for normal human gait and quantified the asymmetries of normal gait for selected variables using a force platform. The asymmetries were quantified using a symmetry index (SI) that was proposed by Robinson et al. Gait symmetries were found to be much larger than they expected for a normal subjects population. They stressed the importance of quantifying the asymmetries, which occur in normal gait then these asymmetry values may be used as a criterion measure to differentiate between normal and pathological gait. In an other study Chao et al. (1983) also investigated the left/right symmetry among normal. They used a preselected set of parameters to calculate a gait symmetry index (I_s) in order to see whether normal level walking has any significant side dominance. The gait of hemiparetic subjects walking on a treadmill with various body weight supports and walking on the floor was investigated. With regard to stance and swing symmetry ratios, patients walked more symmetrically on the treadmill than on the floor. Further the swing symmetry improved with increasing body weight support on the treadmill (Hesse et al., 1999.).

The aim of this study was to further investigate the deficiencies of hemiparetic gait pattern and more closely examine and quantify the asymmetry, which occur in hemiparetic gait. Our aim was to define the main kinematic (time parameters and joint kinematics) and kinetic characteristics (characterising parameters of ground reaction forces) of hemiplegic gait in order to make an attempt to divide patients with wide range of gait disorders into the most homogeneous groups. The present research was also designed to examine the hemiparetic disturbances affecting gait – included both the affected and unaffected limb – in the chronic phase. The study investigated the biomechanical deficits and the resulting compensations that compromise walking in case of childhood hemiplegia with lower extremity impairments and associated disabilities.

METHODS

Subjects

Eleven patients (9 males, 2 females), with disorders of gait as a result of congenital or acquired childhood hemiplegia participated in the study. The study was carried out with the mentioned chronic stroke subjects (mean age 18,4 ± 2,8 years), living in the same special boarding school, their mean time post stroke was 204 ± 39 months. The criteria for selection was that subjects demonstrated residual hemiplegia from a single onset suffered more than ten years previously. Etiologic factors comprised intracerebral haemorrhage, cerebral infarction and cerebral palsy. They were able to walk independently without an orthosis or brace and without a cane. The side of the residual hemiparesis were different, six patients had right sided and five subjects had a left sided hemiparesis. All participants were able to communicate and follow instructions. All the patients and their parents gave informed written consent before participating in this study.

Methods

Instrumentation

Computerized gait analysis was performed for all patient with use of a motion analysis system called APAS (Arial Performance Analysis System) which included two well positioned video cameras, a computer and a software for the collection and analysis of data. The movement of the patient was recorded by two Panasonic video cameras (M10 V-14, NAC Visual system, Woodland Hills, Ca) placed in different planes. The cameras were positioned at right angles. The cameras worked at a sampling rate of 50 Hz. Five passive markers were positioned over specific anatomical points of the trunk and limb examined. These markers were put on the acromioclavicular joint, the greater trochanter at the hip joint level, the lateral joint line of the knee, the lateral malleolus and the head of the fifth metatarsal.

The recording technique and the software allowed three-dimensional reconstruction of the motion in the major joints of lower extremities. The video records were used for kinematic gait assessment. Steps were analyzed for temporal-distance parameters using frame-to-frame and slow motion technique and also chronometrical measurements. The motion analysis involved determination of the range of the motion of the hip, knee and ankle in the sagittal plane by measurement of the flexion and extension at the joint during stance phase for both the paretic (P) and non-paretic (NP) limbs.

The measurement of vertical ground reaction forces was also included in our study. A KISTLER (9296B) force platform was embedded in the walkway sampled ground reaction forces at rate of 500Hz, converted to digital form and stored on a computer along with synchronizing signal from the camera.

Experimental procedure: For the measurement of the kinematic gait-cycle parameters and ground reaction forces, the patient walked approximately 6 meters at their own comfortable speed. All of them wore shoes.. The third step was placed on the surface of the force-plate. The distance from the platform was carefully individually determined by several trials so that the stance phase of the third step of the measured side occurred on the platform. At least three walking trials were recorded for each patient by video-based motion analyzer. The stance phase of the gait cycle was more closely examined by the force platform. Synchronization of film and force plate data was performed during the analysis. The step was acceptable if the whole foot and no part of the contralateral foot landed on the force plate during stride. Both the affected and the unaffected lower limbs were measured.

Parameters

Time parameters: The analysis of length of different gait phase is given important results for analysis of kinematic of affected and unaffected limb. The investigated parameters are:

– time of stance phase (t_t);

– time from the moment of heel strike (impact force) until foot flat (t1);

– time from foot flat until the toe off (t2), i.e.,
– the time elapsed from F1 to F2

Force-Time parameters: The vertical ground reaction forces during stance phase – emphasized the moment of heel strike, foot flat and toe off - were used for the kinetic analysis. The selected components of the vertical ground reaction force - time curve are the following:
– area under the curve (T);
– first force peak at the moment of foot flat (F1);
– second force peak at the moment of toe-off (F2);
– rate of force development at foot flat (RFD1);
– rate of force development at toe-off (RFD2); (Figure 1).

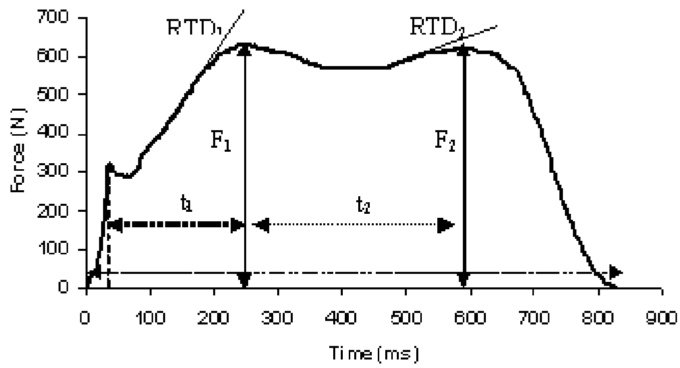

Fig. 1. Graph demonstrating the selected components of the vertical ground reaction force-time curve during stance phase.
t: area under the curve; F1: peak force at the moment of foot flat; F2: peak force at the moment of toe-off; RFD_1: rate of force development at foot flat; RFD_2: rate of force development during toe-off; t_t: time of stance phase; t_1: time between the moment of heel strike and foot flat; t_2 :time from foot flat until the toe off.

Joint kinematics: The motion analysis involved determination of the range of motion of the hip, knee and ankle in the sagittal plane by measurement of the flexion and extension at the joint.

Symmetry index: Asymmetries in gait were quantified using the following simple symmetry indexes:

Kinetic Symmetry Index (SI_K)

$$SI_{Kp} = \frac{F_{1\,paretic}}{F_{2\,paretic}}$$

$$SI_{Knp} = \frac{F_{1\,non\text{-}paretic}}{F_{2\,non\text{-}paretic}}$$

where

F_1 is the vertical ground reaction force at foot flat
F_2 is the vertical ground reaction force at toe off

Kinematic Symmetry Index (SI_{KI})

$$SI_{KI} = \frac{ROM_{non\text{-}paretic}}{ROM_{paretic}}$$

where *ROM* is the range of motion of the investigated joint

The Symmetry Index was defined as the ratio of the joint angles on the non-paretic side to that on the paretic side. The ratios were calculated like non-paretic ROM/paretic ROM ratio in each of the joints. The index represents the difference of the range of motion between the unaffected and the affected side. In case the value of the Symmetry Index is one, it means the ranges of joint movements are equal on the left and right side, so the motion pattern is symmetric. The Symmetry Index was calculated for each joint (ankle, knee and hip) of all the paretic subjects. The Mean Symmetry Index was computed as the arithmetic mean of all the ratios of the ROM performed by patients belonging to the same group. The Overall Symmetry Ratio indicates the relationship between the paretic and non-paretic range of motion considering all of the subjects irrespectively which group they belong to.

Statistical analysis. Mean and standard deviation (SD) was calculated for all variables. The significance of the differences between means was determined by one-tailed, impaired student t-test. The correlation between each of the measures was determined using the Pearson product moment correlation. The significance level was set at $p<0.05$.

RESULTS

Time parameters: Shorter stance phase for the affected side was associated with a prolonged stance time on the unaffected side. The duration of the stance phase for the paretic side was reduced for all patients studied both the duration of the development of the first peak force and the duration from foot flat until toe off. The time (t_1) from the moment of heel strike (impact force) until foot flat (P: 188,2 ± 41,5 ms; NP: 267,0 ± 92,2 ms, $p<0,05$) and the time (t_2) from foot flat till toe off (P: 126,5 ± 52,2 ms; NP: 250,6 ± 119,0 ms, $p<0,01$) was significant longer on the non-paretic side. The stance phase time (t_t) for the non-paretic limb was also increased over normal values (P: 544,7 ± 95,8 ms; NP: 706,7 ± 153,2 ms, $p<0,01$). The pre-swing phase time (before the swing phase of the affected limb) was increased for all patients. Not only the time was markedly prolonged on the hemiplegic side but also there was an impaired weight-bearing on that side.

Force-time characteristics:**.** There was no significant differences between F1 and F2 either in paretic (F1: 614.0 ± 221.2 N, F2: 627.4±167.8 N) or in non-paretic (F1:617.1 ± 158.4 N, F2: 589.8 ± 217.8 N) limb. Also, the difference between paretic and non-paretic limb was not significant concerning both F1 and F2. The rate of force development during foot-flat was significantly increased on the affected side (P: 22,9 ± 10,4 N/s; NP: 9,2 ± 2,9 N/s, $p<0,01$). Contrast with the previous result the toe off on the affected side was delayed and the rate of the force development was markedly lower than the unaffected side (P: 0,49 ± 0,2 N/s; NP: 1,19 ± 0,45 N/s, $p<0,01$). The magnitude of impulse during the stance phase was markedly higher (84%) on the unaffected side (P: 566,6 ± 209,0 Ns; NP: 1042,7 ± 395,7 Ns, $p<0,01$).

Joint kinematics: Different patterns of abnormal range of motion were noted not only on the affected but also on the unaffected side. It has become apparent through the analysis that hemiplegia contains heterogeneous patterns of joint movements. This was the reason for the high standard deviation, the mean value of all the patients was not characteristics. We found that these heterogeneous patterns may be separated into groups which contain a variety of homogeneous patterns. On the basis of the quantitative sagittal kinematic measurement data we attempted to identify these relatively homogenous groups. The patients were clustered into three groups based on the ranges of their ankle, knee and hip joints angles. Within the groups the standard deviation decreased, the mean values better characterized the patients belonging to the groups.

The first group includes those patients who had a smaller range of motion on the affected limb, irrespective whether the ankle, the knee or the hip joints data were examined. The values of the mean range of motion can be found in Table 1.The second group contained those paretic limbs which demonstrated lower range of motion of the knee and the ankle joints but higher in their hip than the non-paretic side (Table 1). Those patients whose knee and the hip demonstrated smaller ranges of movements on the hemiplegic side but markedly higher in the affected ankle were sorted into the third group (Table 1).

Table 1. The mean range of motion and the standard deviation (±sd) for both paretic (P) and non-paretic (NP) side considering the ankle, knee and hip joints.

		Group 1.		**Group 2.**		**Group 3.**	
		NP	**P**	**NP**	**P**	**NP**	**P**
ankle	mean range	33,98	20,83	48,40	25,93	25,23	45,97
	sd	6,08	4,22	12,05	12,13	3,35	10,38
knee	mean range	43,28	33,53	42,53	27,10	47,60	27,27
	sd	9,23	5,46	11,40	9,90	6,16	16,01
hip	mean range	28,15	20,78	18,33	23,83	35,97	24,17
	sd	4,27	3,45	1,47	6,73	11,27	2,96

Symmetry Indexes. The kinetic symmetry index for non-paretic limb was above one (1.14 ± 0.46) which is 20% greater than that for paretic limb (0.95 ± 0.2). However, because of the large standard deviation, the difference is not significant. The kinetic symmetry index in group I was significantly greater for paretic limb as compared to that of non-paretic limb. In the second group despite to the great difference between the two leg, the difference is not significant because of the large standard deviation at the NP limb. In the third group the NP and P limb displayed similar SI_K values (Figure 2).

The joint angle disturbances of the paretic side include reduction of flexion extension range of motion result in measurable asymmetry. The Kinematic Symmetry Index was higher than 1,00 in the first group. Their impaired extremity showed less angular displacement at the ankle, knee and hip than on the other side. The second group involved those patients, whose Symmetry Index of the ankle and knee were higher than 1,00 but the index calculated for the hip was less than 1,00. Those patients had wider range of motion in their paretic hip. In the third group each patient had the Symmetry Index higher than 1,00 in their knee and hip. In contrast the index values of the ankle were considerable lower than 1,00. They showed markedly higher range of motion in their affected ankles (Table 2).

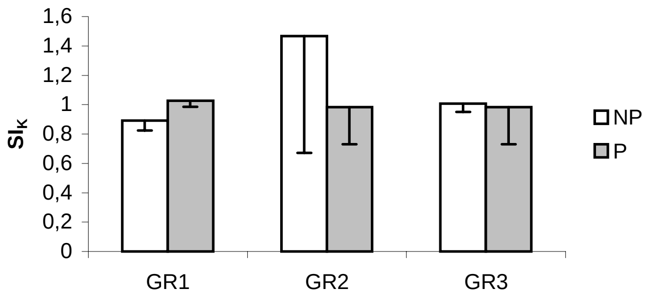

Fig. 2. Means of the Kinetic Symmetry Index (SI_K) for group 1 (GR1), group 2 (GR2) and group 3 (GR3). Note that only GR1 is homogeneous and the difference between means for P and NP is significant.

Table 2. Mean and standard deviation (±sd) of kinematic symmetry index for groups.

	Group 1.	Group 2.	Group 3.
ankle mean symmetry index	1,74	2,08	0,57
sd	0,73	0,68	0,17
knee mean symmetry index	1,29	1,66	2,33
sd	0,20	0,33	1,41
hip mean symmetry index	1,36	0,82	1,47
sd	0,13	0,21	0,47

DISCUSSION

Hemiplegic gait has some typical features at the mechanism of their gait disturbance but differed interindividually according to the extent and the location of the cerebral injury. Different patterns of the vertical ground reaction force-time curve on the affected and on the unaffected side clearly demonstrated the asymmetrical nature of hemiplegic gait. We found that the shorter stance phase time for the affected side is related to the deficient ability to load and transfer weight through their affected leg. Numerous balance studies have shown that hemiplegic subjects bore and transferred a decreased percentage of body weight through their affected limb (Kramers et al., 1996, Gruendel et al., 1992). Other authors have found relationship between this impaired weight bearing and the gait patterns of the hemiplegics (Dettman et al., 1987, Seeger et al., 1981). It has been proposed that improving weight transfer through the affected leg during progressive training with the feet of the patients placed in a variety of diagonal position, may improve gait symmetry in hemiplegics (Olney et al., 1991). We further investigated the deficiencies of weight shifts and its relationship to the gait pattern on young hemiparetic subjects being patient from their early childhood. The subjects studied demonstrated not only reduced weight-bearing ability on that side but also they shift the weight through the affected limb in a much shorter time in order to load their paretic leg for as short time as they can. They also transferred the weight from the paretic side to the unaffected side long before the foot on the paretic side cleared the ground. According to the muscle weakness hemiparetic patients tends to decrease the load of their affected limb by shortening both the absolute and relative time of the stance phase on the hemiplegic side. However, we found no significant differences between the paretic and non-paretic side concerning F1 and F2. It seems that peak ground reaction forces during foot flat and toe-off do not play significant role in hemiparetic gait.

The prolonged duration of the non-paretic limb stance phase is the result of the muscle weakness of the paretic legs whereas much longer proportion of the gait cycle is spent with the non-paretic leg weight-bearing. Part of the explanation also can be that the affected limb takes more time to swing through because less power is put into the limb during late stance. The markedly higher (84%) magnitude of impulse during stance phase on the unaffected side shows that the non-paretic limb performed a greater proportion of the work. The high impulse value for the non-paretic limb can be due to the prolonged stance phase time. Toe off on the affected side was characterized by markedly lower force development showing the impairment of the muscle activation occurring prolonged duration of the pre-swing phase.

We found that the decreased range of motion on the affected side was leading to compensatory mechanism of the non-paretic limb in order to correct the deficiencies but it is resulted in abnormal movement both the affected and unaffected leg. We examined demonstrated very low range of movement in all patients. The limbs were stiff (narrow ranges) and prefer straighter (less maximum flexion angles). Only patients in the third group had higher range of movement at their ankle (P: 45,97± 10,38°), but they all had the reason for that due to some operation. Similar ranges of ankle movement have also been observed by O'Byrne et al. (1998). They clustered the patients (hemiplegic and diplegic) into eight groups. The ranges at the ankle in their groups are 17,48 ± 6,19°, 18,99 ± 4,39°, 27,45 ± 10,99°, 29,00 ± 7,30°, 29,29 ± 6,55°, 29,76 ± 6,68°, 30,53 ± 6,96°, 36,87 ± 7,28°, respectively. These values also demonstrate a low range of movement, a reduced range and a good range as well. We found a lower range in our first and second group (20,83 ± 4,22°, 25,93 ± 12,13°).

Kerrigan et al. (1991) reported that mean ranges at the ankle are 28,9° for young adult subjects 23,3° for elderly subjects at comfortable speed of walking and 22,5° for elderly at fast walking speed. Our results considering the range of hemiparetic young patients are similar to the result of Kerrigan et al. (1991) for healthy elderly subjects. However, the ranges we found at the hip respect to all the groups (20,78 ± 3,45°, 23,83 ± 6,73°, 24,17 ± 2,96°) demonstrated a markedly lower range of movement than the mean values of Kerrigan et al. (1991), i.e., 45,8° – for young; 40,4° for elderly at comfortable speed and 44,1° for elderly at fast speed. The mean values of the range of movement at the hip ranged from 20,08 ± 6,79° to 43,42 ± 7,39° in the study of O'Byrne et al. (1998). Those patients we examined had also low range of motion at the knees (33,53 ± 5,46°, 27,10 ± 9,90°, 27,27 ± 16,01) in contrast to O'Byrne et al. (1998) who demonstrated almost only higher values for hemiplegic and diplegic patients. Kerrigan et al. (1991) measuring healthy young and elderly subjects represented considerably higher range of motion at the knee. Furthermore, Kerrigan et al (1998) reported, from a stiff-legged gait study in spastic paresis, the value of 25,3 ± 10,3° peak knee flexion in initial or midswing in 22 patients and the knee range of motion remained fixed at 9° in one patient.

CONCLUSION

The results of the present study indicate that patients with chronic hemiplegia (being hemiparetic from early childhood and for more than ten years) learned an individual gait pattern adapting themselves to specific circumstances. Therefore it is very difficult to recruit them into homogeneous groups. It is proved, in accordance with the previous studies, that the joint movement coordination strategy on paretic side differs from normal gait. The disturbed joint movement in the paretic limb modifies the joint movement pattern and the force development on the ground on the non paretic side. It seems that the best characteristics for determination of hemiparetic gait are the impulse and rate of force development at foot flat and toe-off. Symmetry indexes could not be applied to form homogeneous groups in this study. However, it does not mean that this approach cannot be applied to determine the degree of the gait deficiency.

REFERENCES

1. Bohannon, R.W., & Andrew, A.W. (1990). Correlation of knee extensor muscle torque and spasticity with gait speed in patients with stroke. *Arch Phys Med Rahabil*, (71),330-333.
2. Bohannon, R.W., & Walsh, S. (1992). Nature, reliability and predictive value of muscle performance measures in patients with hemiparesis following stroke. *Arch Phys Med Rehabil,* (73), 721-725.
3. Bowker, P., & Messenger, N. (1988). The measurement of gait. *Clin Rehabil*, (2), 89-97.
4. Brandstater, M.E., de Bruin, H., Gowland, C., & Clark, B.M. (1983). Hemiplegic gait: analysis of temporal variables. *Arch Phys Med Rehabil*, (64), 583-587.
5. Caldwell, C., MacDonald, D., Mac-Neil, K., Mac-Farkland, K., Turnbull, G.I., & Wall, J.C. (1986). Symmetry of weight distribution in normals and stroke patients using digital weight scales. *Physiother Pract*, (2), 109-116.
6. Chao, E.Y., Laughman, R.K., Schneidr, E., & Stauffer, R.N. (1983) Normative data of knee joint motion and ground reaction forces in adults level walking. *Journal of Biomechanics*, 16 (3), 219-233.
7. Dettman, M.A., Linder, M.T., & Sepic, S.B. (1987). Relationship among walking performance, postural stability and functional assessments of the hemiplegic patient. *American Journal of Physical Medicine*, (66), 77-90.
8. Eke-Okoro, S.T., & Larson, L.E. (1984). A comparison of the gait of paretic patients with the gaits of control subjects carrying a load. *Scand Journal Rehabil Med*, (16), 151-158.
9. Elliot, J., Roth, M.D., Merlitz C., Mroczek, K., Dugan, S.A., Suh, P.T., & W. W. (1997). Hemiplegic gait: Relationship between walking speed and other temporal parameters. *American Journal of Physical Medicine & Rehabilitation,* (76/2 March/April).
10. Gruendel, T.M. (1992.) Relationship between weight-bearing characteristics in standing and ambulatory independence in hemiplegics. *Physi-other Can*, (44), 16-17.
11. Herzog, W., Benno, M., Nigg, B.N., Read, L.J., & Olsson, E. (1988). Asymmetries in ground reaction force patterns in normal human gait. *Medicine and Science in Sports and Exercise* 21 (1).
12. Hesse, S.,Konrad, M., & Uhlenbrock, D. (1999). Treadmill Walking with Partial Body Weight Support versus Floor Walking in Hemiparetic Subjects. *Arch Phys Med Rehabil*, (80), 421-427.
13. Kerrigan, D.C., Frates, E. P., Rogan, S., & Riley, P.O. (1999). Spastic paretic stiff-legged gait: Biomechanics of the Unaffected Limb. *American Journal of Physical Medicine & Rehabilitation*, 78(4), 354-360.
14. Kerrigan, D.C., Grounley, J., & Perry, J. (1991). Stiff-legged gait in spastic paresis: a study of quadriceps and hamstrings muscle activity. *American Journal of Physical Medicine and Rehabilitation,* (70), 294-300.
15. Kerrigan, D.C., Todd, M.K., Della, Croce U., Lipsitz, L.A., & Clooins J.J. (1998). Biomechanical gait alterations independent of speed in the healthy elderly: evidence for specific limiting impairments. *Arch Phys Med Rehabil*, (79), 317-322.
16. Knuttson, E., & Richard, C. (1979). Different types of disturbed motor control in gait of hemiparetic patients. *Brain,* (102), 405-430.
17. Kramers, J.A., Quervain, D.E., & et al (1996). Gait pattern in the early recovery period after stroke. *The Journal of Bone and Joint Surgery*, (78-A/10).

18. Lane, R.E.J. (1978). Facilitation of weight transference in the stroke patient. *Physiotherapy*, (64), 260-264.
19. Mizrahi, J., Susak, Z., Heller, L., & Najanson, T. (1982). Objective expression of gait improvement of hemiplegics during rehabilitation by time-distance parameters of the stride. *Med Biol Eng Comput*, (20), 628-634.
20. Mizrahi, J., Susak, Z., Heller, L., Najanson, T. (1982) Variation of time-distance parameters of the stride related to clinical gait improvement in hemiplegics. *Scandinavian Journal Rehabilitation and Medicine*, (14), 133-140.
21. O'Byrne, J.M., Jenkinson, A., & O'Brien T.M. (1998) Quantitative Analysis and Classification of Gait Patternsin Cerebral Palsy Using a Three-Dimensional Motion Analyzer . *Journal Child Neurol*, (13), 101-108.
22. Olney, S.J., Griffin, M.P., Monga, T.N., & McBride J.D. (1991). Work and power in gait of stroke patients. *Arch Phys Med Rehabil*, (72), 309-314.
23. Özgirgin, N., Bölükbasi, N., Beyazova, M., & Orkun, S. (1993). Kinematic gait analysis in hemiplegic patients. *Scandinavian Journal Rehabilitation and Medicine*, (25), 51-55.
24. Perry, J., Garrett, M., Gronley, J.K., & Mulroy, S.J. (1995). Classification of walking handicap in the stroke population. *Stroke*, (26), 982-989.
25. Schroeder, H.P., Coutts, R.D., Lyden, Billings, E. Jr., & Nickel, V.L. (1995). Gait parameters following stroke. A practical assessment. *Journal Rehabil Res Dev*, (32), 25-31.
26. Seeger, B.R., Caudrey, D.J., & Scholes, J.R. (1981). Biofeedback therapy to achieve symmetrical gait in hemiplegic crebral palsied children. *Arch Phys Med Rehabil,* (62), 364-8.
27. Turnbull, G.J., Charteris, J., & Wall J.C. (1996). Deficiencies in standing weight shifts by ambulant hemiplegic subjects. *Arch Phys Med Rehabil*, (77), 356-362.
28. Wagenaar, R.C., & Beek, W.J. (1992). Hemiplegic gait a kinematic analysis using walking speed as a basis. *Journal of Biomechanic*, (25), 1007-1015.
29. Wall, J.C., & Ashbury, A (1979) Assessment of gait disability in hemiplegics. *Scandinavian Journal Rehabilitation and Medicine*, (11), 95-103.
30. Wall, J.C., & Turnbull, G.I. (1987). Evaluation of outpatient physiotherapy and a home exercise program in the management of gait asymmetry in residual stroke. *Journal Neurol Rehabil*, (1), 115-123.
31. Wheelwright, E.F., Minns, R.A., Elton, R.A., & Law, H.T. (1993). Temporal and spatial parameters of gait in children; pathological gait. *Dev Med Child*, (35), 114-125.

KINEMATIČKE I KINETIČKE METODE ANALIZE HODA KOD HEMIPLEGIČNIH PACIJENATA

Mónika Horváth, Tekla Tihanyi, József Tihanyi

U nedostatku standardizacije u vizuelnoj i kompjuterizovanoj analizi hoda razvijene su različite kinetičke i kinematičke metode analize hoda. Cilj ove studije je da dalje istražuje individualne karakteristike biomehaničkih nedostataka hemiparetičnog modela hoda i kompenzacionog kompromisnog hoda. Faze položaja tela su bliže ispitane pomoću sistema sile metalnih pločica i analizom sistema kretanja. Podaci su prikupljeni proučavanjem neprirodne i prirodne noge 11 hemiplegičnih pacijenata (9 muškaraca i 2 žene). Otkrili smo da je kraće vreme faze položaja tela neprirodne noge vezano sa nedostatkom sposobnosti da se optereti i prenese težina na neprirodnu nogu. Stepen razvoja sile je značano povećan dok je stopalo ugrožene strane bilo ravno, a palac je okarakterisan sa znatno manjim razvojem sile. Oslabljeni stepen kretanja na hemiplegičnog strani je vodeći u kompenzatornom mehanizmu neprirodne noge, za rezultat ima abnormalni pokret članka, kolena i kuka - kako neprirodne, tako i prirodne strane. Pošto su se tokom analize različiti modeli izdvojili i identifikovali, postalo je očigledno da za svaki tip modela trebaju biti razvijeni optimalni protokoli tretmana.

Ključne reči: *hemiplegija, kvantitativna analiza hoda, reakcija sile teže, stepen pokreta*