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GAIT PARAMETERS OF HEALTHY, ELDERLY PEOPLE

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Abstract. *Walking is one of the most common human movements. It means to transport the body safely and efficiently across ground level, uphill or downhill. Walking is learned during the first year of life and reaches maturity around the age of seven,remaining at the same level until 60. In old age one's walking performance starts to decline and it gradually slows down. With the increased life expectancy of the elderly and their more active lifestyle, there is now an emphasis on determining any changes that occur in their gait patterns, in order to indentify diagnostic measures that are usable for monitoring the rehabilitation process after endoprothesis implantation. The aim of this study is to determine how selected kinematical, kinetic and electromyographical parameters may change as a result of aging. A total of 31 healthy elderly subjects without any history of lower extremity joint pathology were investigated at constant gait speed (three km/h). The gait analysis equipment used consisted of an infinitely adjustable treadmill with force-plates and an ultrasound-based motion analyser with a surface electromyograph. Spatial-temporal, angular, kinetic and electromyographical parameters were recorded for the lower extremities. The results obtained from the lower limb were compared on both sides as well as with those of 50 healthy young individuals collected from our database. The elderly had a significantly shorter step length and wider step width compared to the results of a young control group. Our results showed that the aged individuals demostrated a statistically lesser range of motion in different joints during walking. We suggested that neurophysiological changes associated with aging might result in the less certainty of the neuromuscular system in selecting a stable gait.*

Key words: *gait analysis, kinematics, kinetics, electromyography, the elderly*

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1. INTRODUCTION

In Europe the elderly group (defined as ≥60 years of age) represents a growing segment of the population. Walking is a learned activity, in which the moving body is supported successively by one leg and then the other. The dynamic regulation of the upright stance is essential to the safe and efficient performance of many activities of daily life.

Gait analysis has been used in an attempt to detect subtle differences between the gait of elderly people and that of younger individuals. It is widely documented that elderly people tend to walk more slowly and that this speed reduction is due to a reduction in step length (Bullinger, 1996, 35; Crosbie and Vachalathiti, 1997, 6; Kurz and Stergiou, 2003, 348; Wall et al., 1991, 23; Winter et al., 1990, 70).

Comprehensive gait analysis usually includes kinematics, kinetics and muscle activity, and this complex information can only be obtained in a specialized laboratory. However, a simplified analysis using, for example, spatial-temporal parameters can also be of clinical value. The purpose of the present study was to analyze age - related changes in functional gait patterns in healthy elderly individuals compared to the gait patterns of healthy young volunteers. The gait parameters of the healthy young individuals were measured earlier and summarized in (Kocsis et al., 2000, 1).

2. MATERIALS AND METHOD

Subjects

A total of 31 healthy elderly volunteers (14 women and 17 men) were included in the study. Their mean age was 71.15 years (SD ± 9.14 years), mean weight 77.23 kg (SD ± 13.12 kg), and mean height 1.74 m (SD ± 0.22 m). A total of 51 healthy young volunteers (31 males and 20 females) were included in the study. Their mean age was 31.70 years (SD±4.1 years), mean weight 72.1 kilograms (SD±25.2 kilograms) and mean height 1.71 m (SD±0.12 m). Each subject provided informed consent before participation and signed a consent form approved by the Hungarian Human Subjects Compliance Committee.

The subjects were evaluated with the Harris Hip Score as well as Merle D' Aubigné Hip Score, Hospital for Special Surgery Knee Score, Womack Osteoarthritis Scale and Short Form Healthy Survey (SF-36) (Bullinger, 1996,35). The objective functional evaluation was based on three -dimensional gait analysis.

Instrumentation

The gait performance of each subject was assessed at the Laboratory for Biomechanics, The Budapest University of Technology and Economics.

Three-dimensional motion analysis was performed using an ultrasound-based Zebris CMS-HS system (ZEBRIS, Medizintechnik GmbH, Germany). The measuring head with three sensors is positioned behind the individual and the five ultrasound triplets with three active markers on each are placed on the sacrum, left and right thighs, and left and right calves (Figure 1). The measuring method was developed in the Laboratory of Biomechanics (Jurak and Kocsis, 2002; Knoll et al., 2004, 12; Kocsis et al., 2000, 1; Kocsis, 2002). The data, obtained from the measuring system recording the active markers, allowed for the determination of coordinates of nineteen optional anatomical points of the lower limbs. The biomechanical model developed by Knoll et al. (2004, 12) was chosen for our investigation. The spatial coordinates were recorded at a frequency of 100 Hz. The absolute error of our ultrasound-based system is less than 1mm (Knoll et al., 2004, 12).

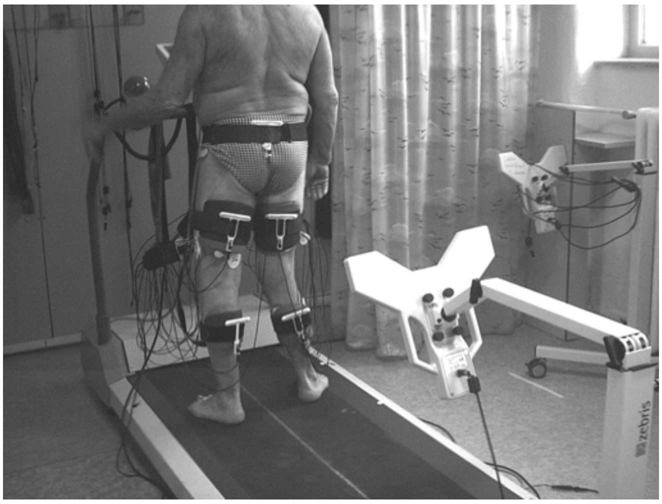

Fig. 1. The arrangement of the measurements

The ground reaction forces were recorded by the multicomponent measuring platform with two force plates (1504 force-cells in each force plate), which are integrated into the motorized and instrumented 330 mm × 1430 mm treadmill (Bonte Zwolle B.V, Austria). The ground forces were measured at 1000 Hz.

The structure of the ZEBRIS CMS-HS movement analysis system and of the measurement control software enables us to measure changes of electric potential generated in the muscles in the course of movement while simultaneously recording the kinematic characteristics of the movements, without any subsequent synchronization, by surface electromyography. Changes in the electric potential of muscles were detected by monopolar electrodes of 18 mm in diameter made from Ag-AgCl (blue sensor P-00-S, Germany). Two mono-polar surface electrodes are stuck on the washed depilated skin surface degreased by alcohol (skin resistance may not exceed 5000Ω) in the area of the stomach muscles; the distance between the active parts is 30 mm. As for the positioning of the surface electrodes, proposals by SENIAM were taken into consideration (Hermes et al., 1999); the ANVOLCOM model (Hermes et al., 1999) was used for filtering out interference between muscles. SENIAM (Hermes et al., 1999) recommends the use of double-side tape for the fixation of the electrodes and cables to the skin in such a way that the electrodes are properly fixed to the skin, movements are not hindered by a cable, nor are the electrodes pulled.

The following muscle groups were included in the investigation: (1) m. vastus lateralis, (2) m. vastus medialis (3) m. biceps femoris and (4) m. adductor longus.

The surface EMG signal is quasi-stochastic (random), of Gauss distribution, the amplitude value of which varies between -2000 and +2000 mV, and its frequency spectrum value is 10-500 Hz . Accordingly, the CMRR value of the amplifier integrated in the

ZEBRIS CMS-HS movement sensor system is higher than 80 and its noise limit is below 2μV. The reception frequency is 1000 Hz. The EMG signals transmitted through the amplifier are recorded by the measurement control system.

Walking on the treadmill can initially be an unfamiliar experience, which in turn may influence the parameters measured. Therefore, measurements are to start after six minutes of familiarization time (Alton et al., 1998, 13; Matsas et al., 2000, 11). The measurement was performed at a three km/h gait speed. The spatial coordinates of the anatomical points, the vertical component of the reaction force and electromyographical signals were collected for six gait cycles.

Assessment parameters

The raw data (the coordinates of each investigated anatomical point) were smoothed and filtered using a fourth-order zero lag digital Butterworth with a high frequency cutoff at five Hz. The spatial coordinates of the anatomical points of the lower limb define a number of gait parameters, which are commonly used for the description of gait: cadence, step length and walking base.

Anatomical joint angles are important because the range of movement is of interest to clinicians (e.g., hip abduction and adduction, knee flexion and extension). Anatomical joint angles show how one segment is oriented in relation to another. There has been some debate as to the most appropriate method of defining joint angles (Grood and Suntay, 1983, 105). We added a new knee angle definition closer to the reality of flexion and extension, which take place at about the mediolateral axis of the proximal segment. This definition complements the usual frontal and transverse plane components. To describe the position of the thigh, the usual plane angles (as flexion and extension, and adduction-abduction) are defined in (Jurak and Kocsis, 2002; Knoll et al., 2004, 12; Kocsis et al., 2000, 1; Kocsis, 2002). Thigh rotation takes place at about the longitudinal axis of the distal segment. In this research the knee angle is defined as the angle between the spatial vectors joining the lateral malleolus to the head of the fibula and joining the lateral femoral epicondyle to the greater trochanter (Jurak and Kocsis, 2002; Knoll et al., 2004,12; Kocsis et al., 2000,1; Kocsis, 2002). This calculation method determines a real spatial knee angle. The value of this angle does not depend on the spatial position of the lower limb, as the component angles do, only on the relative position of the shank to the thigh. For each subject, the maximum (peak of flexion) and minimum values (peak of extension) of the entire gait cycle were determined.

The hip angle is defined as the angle between the spatial vectors joining the lateral femoral epicondyle to the greater trochanter and joining the greater trochanter to ASIS (Jurak and Kocsis, 2002; Kocsis, 2002). This calculation method determines a real spatial hip angle. The value of this angle does not depend on the spatial position of the lower limb, as the component angles do, only on the relative position of the thigh to the pelvis. For each subject, the maximum (peak of flexion) and minimum values (peak of extension) of the entire gait cycle were determined.

The modified motion analysis technique provides an opportunity to define the obliquity of the pelvis parallel with the flexion-extension and rotation angles of the pelvis (Jurak and Kocsis, 2002; Kocsis, 2002; Kocsis and Beda, 2001, 88). Our calculation method is equivalent to the suggestion of Grood and Sunday (1998, 105).

The raw data of the electromyographical signals were a high pass filtered to eliminate frequency components below 10 Hz, then rectified and filtered to eliminate signal components over 200 Hz. The linear envelope EMG curve was determined by the root-mean square method (Vaughan, 1999) and normalized to the average of the peak EMG signal values of six gait cycles. For specifying the intermuscular coordination (on-off pattern) of various muscles, a muscle can be considered as active if its normalized value is higher than 0.2 (20%) (Vaughan, 1999).

Statistical analysis

Statistical analyses were performed using the computer software named Statistica (version 7, 2004.). For each subject, the average and the standard deviation of the parameters were determined from six complete gait cycles, and these data were further processed. Data values are presented as mean ± SD for each group at all walking speeds. A comparison between the two groups was performed by ANOVA. Statistical significance was defined as $p<0.05$.

3. RESULTS

The average Harris Hip Score was 98.9 points (± 1.1), all of the subjects had excellent results (HHS~100 points). The results were similarly good as far as the Merle D' Aubigné Hip score and HSS Knee Score are concerned. The subjects were not limited in their normal daily or recreational activities.

The absolute values of the various gait parameters of the young and elderly subjects are shown in Tables 1-3. The data of the young subjects are from (Kocsis et al., 2000, 1).

Significant differences were not seen throughout the swing phase of the dominant and non-dominant limb in the case of young and at elderly subjects ($p>0.45$). Furthermore, the step length time of the double support phase was significantly shorter for the elderly as compared to the younger group of healthy volunteers ($p<0.007$). The step width of elderly subjects is significantly wider than for the healthy people ($p<0.003$).

The amount of functional movement in the hip and knee joints was significantly reduced on both sides as compared to the healthy young group (Table 2.) ($p<0.0004$). The motion of the hip and knee joints showed a symmetrical pattern for the elderly and young subjects. It means that the motions of the hip and knee joints on the nondominant side were not significantly smaller than on the dominant side ($p>0.45$). The rotation and obliquity of the pelvis of eldely people are significantly higher compared to the young group ($p<0.00002$).

The kinetic parameters (Table 3) revealed a certain degree of unloading on the non-dominant side of the young and elderly people. Peak values of force parameters showed a tendency towards a greater impact during heel strike (F1) and a less forceful push-off (F2) during the phase of toe-off. All the differences were negligible. The difference between the young and the elderly group is significant ($p<0.002$)

Table 1. The results of temporal and spatial parameters in healthy elderly and young subjects

Parameter		Unit	Elderly		Young	
			Female	Male	Female	Male
Cadence		steps per minute	87.59± 4.69	84.42±18.35	64.96 ±17.67	59.59 ± 12.45
Step length	Dominant side	Centimetre	349.11±60.36	363.25±32.05	470.7 ± 20.1	513.12 ± 26.6
	Nondominant side	Centimetre	346.01±35.43	339.92±12.70	465.12 ± 22.4	511.34 ± 23.3
Step width	Dominant side	Centimetre	23.02± 3.12	21.97 ± 6.09	18.45 ± 2.45	21.12 ± 2.34
	Nondominant side	Centimetre	27.22± 6.60	22.74 ± 3.86	19.99 ± 1.05	24.45 ± 2.89
Double support phase		% of gait cycle	13.41± 4.18	13.47 ± 3.43	12.34 ± 2.99	12.44 ± 3.01
Swing phase	Dominant side	% of gait cycle	36.40± 1.24	39.93 ± 2.58	41.12 ± 2.99	44.34 ± 3.11
	Nondominant side	% of gait cycle	33.17± 2.98	36.86 ± 4.97	39.34 ± 3.15	40.23 ± 2.99

Table 2. The results of the angular parameters of healthy elderly and young subjects

Parameter		Unit	Elderly		Young	
			Female	Male	Female	Male
Hip flexion						
Range	Dominant side	degree	52.34±3.56	59.20±3.5	61.64±4.56	64.02±3.56
	Nondominant side	degree	50.12±4.78	54.30±3.3	59.2±3.45	62.76±3.56
Maximum	Dominant side	degree	64.23±6.78	69.30±9.1	66.76±4.56	68.62±5.63
	Nondominant side	degree	60.12±4.57	63.67±8.5	64.32±3.12	67.54±5.23
Minimum	Dominant side	degree	11.89±3.78	9.91± 5.78	5.12±1.34	4.60±1.44
	Nondominant side	degree	10.00±5.08	9.63±3.89	5.32±2.1	4.79±1.45
Pelvic rotation						
Range		degree	8.29±2.96	7.42±1.69	4.46±2.34	6.57±2.01
Maximum		degree	2.91±2.6	6.37±1.30	2.12±1.23	5.34±1.34
Minimum		degree	-5.38±0.35	-1.26±1.15	-2.34±1.23	-1.23±2.23
Pelvic obliquity						
Range		degree	2.65±0.38	3.12±1.87	1.42±0.33	1.75±0.44
Maximum		degree	5.64±1.58	3.97±1.55	4.56±2.34	3.12±1.23
Minimum		degree	2.99±1.19	0.85±0.85	3.14±1.03	1.37±0.76
Knee flexion						
Range	Dominant side	degree	43.08±2.57	41.15±2.9	54.23±3.67	56.86±2.89
	Nondominant side	degree	39.67±1.79	40.45±3.1	50.79±2.99	52.97±3.12
First peak	Dominant side	degree	16.21±2.4	19.77±2.94	21.56±2.67	23.34±2.45
	Nondominant side	degree	27.45±1.08	17.83±2.36	19.89±1.99	22.39±3.47
Second peak	Dominant side	degree	56.89±0.31	50.67±2.58	59.99±3.12	61.99±3.44
	Nondominant side	degree	48.5 ±0.35	49.44±3.78	56.78±3.21	59.34±3.22
Minimum	Dominant side	degree	17.22±2.1	10.08±2.08	5.89±3.12	5.13±0.23
	Nondominant side	degree	15.41±2.22	9.80±2.88	5.99±3.33	5.74±2.12

Table 3. The results for the force parameters of healthy elderly and young subjects

Parameter		Unit	Elderly		Young	
			Female	Male	Female	Male
F1 first peak in the early stance phase	Dominant side	% of body weight	137±1.1	142±1.3	143±0.9	144±0.9
	Nondominant side	% of body weight	135±0.8	137±1.0	139±1.2	141±0.8
F2 second peak in the late stance phase	Dominant side	% of body weight	134±1.4	136±0.8	141±1.1	142±1.7
	Nondominant side	% of body weight	132±0.8	123±1.1	137±1.4	139±1.3

Figure 2 shows the intermuscular coordination of the four studied muscles, which is a graphical representation and provides comparisons for both groups. Significant differences were observed between the two groups' EMG activity throughout gait.

Seven elderly people exhibited an adductor longus avoidance gait pattern: m. adductor longus did not produce any activity during the pre-swing phase (Figure 2).

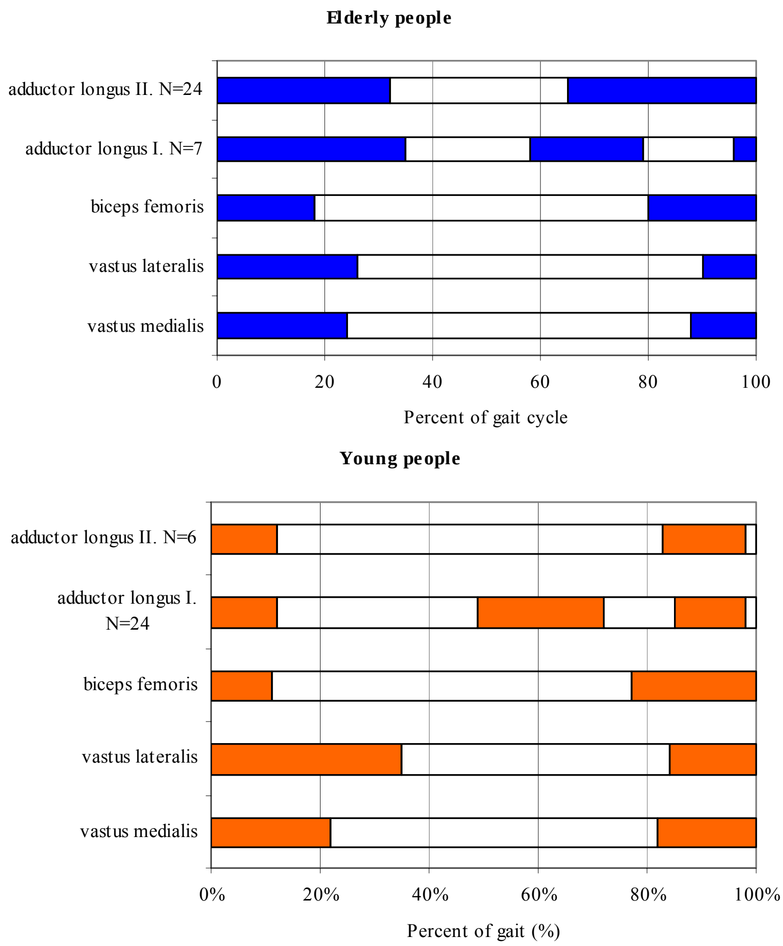

Fig. 2. Intermuscular coordination of elderly and young people

4. Discussion

The aim of this study was to analyse the resulting changes in functional gait patterns in healthy elderly subjects compared to young subjects. A kinematic analysis objectively describes how the body segments of the subjects are moving during gait. Movement analysis allows calculations of the angle and range of motion. We compared the bioemchanical parameters determined at constant gait speed.

In this research the spatial-temporal parameters (step length, step width etc) show significant differences compared to those of healthy young subjects (Kocsis et al., 2000, 1). Thus, it seems that aging significantly changes the gait pattern of the healthy elderly.

However, synchronous movements of the hip, pelvis and knee were detected in this study. It could be seen that there were only minor differences in joint angle profiles between the young and the elderly, but subtle changes occured at the amplitude level. The data are consistent with those of Winter (1990, 70) and Oberg (1993, 30). This decreased knee flexion and motion range of the of knee in the elderly correlates well with their significantly shorter step length.

The pelvic dynamic range of motion is larger in the elderly than in the young. The data are consistent with those of Winter (1990, 70) and Oberg (1993, 30). The increased pelvic motion in the elderly was attributed to the need to put their hip extensors at a more favorable length so they can meet the demand despite the weakness associated with aging (Trueblood and Rubenstein, 1991, 17). The decreased motion of hip and knee joints is compensated with increased pelvic motion. We suppose that these changes are mainly due to age-related neuromuscular changes.

The age-related neuromuscular changes are supported by the intermuscular coordination of various muscles. The activity period of vastus medialis and lateralis muscles are shorter compared to the results of the young people. For compensation, the activity of m. biceps femoris and m. adductor longus are longer compared to the results of the young people.

5. Conclusion

Our findings indicate that the changes in gait parameters may occur in healthy elderly people compared to the gait patterns of healthy young subjects. The decreased motion of the knee and hip joints leads to increased pelvic motion, which should affect the natural mobility of the lumbar spine and cause pain in the lumbar region of the spine because of their kinematic interaction. The decreased motion range of the of knee in the elderly correlates well with their significantly shorter step length. We suppose that these changes are mainly due to age-related neuromuscular changes, which are supported by the intermuscular coordination of various muscles. The results suggested that the results of patients with osteorthrithis or with different endoprothesis may be comapare only with the results of elderly healthy group in the future.

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PARAMETRI HODANJA ZDRAVIH STARIJIH LJUDI

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Hodanje je jedno od najčešćih ljudskih kretnji. To je način transportovanja tela sigurno i efikasno po zemlji, uzbrdo ili nizbrdo. Hodanje se uči tokom prve godine života i postiže svoj maksimum oko sedme godine i ostaje na tom nivou sve do 60 godine života. Kod starijih osoba hodanje počinje postepeno da opada u kvalitetu. Sa produžetkom životnog veka starijih osoba i njihovim aktivnijim načinom života sada postaje imperativ utvrđivanje bilo kakvih promena u načinu njihovog hodanja kako bi se odredile dijagnostičke metode koje bi pomogle u procesu kontrole rehabilitacije nakon implantacije endoproteze. Cilj ovog istraživanja je da se odredi kako selektivni kinematički, kinetički elektromiografski parametri mogu da utiču na promenu statusa kao *rezultat starenja. Istraživan je uzorak od ukupno 31 starijeg ispitanika bez istorije patologije donjih ekstremiteta pri konstatnoj brzini hoda (3 km/h). Oprema za analizu hoda sastojala se od podešive staze za hodanje sa pločama za određivanje snage i ultrazvučnog analizatora pokreta sa površinskim elektromiografom. Registrovani su spacijalno-temporalni, angularni, kinetički i elektromiografski parametri za donje ekstremitete. Dobijeni rezultati za donje ekstremitete su upoređivani za obe strane kao i sa primerima 50 zdravih mladih osoba, preuzetih iz naše baze podataka. Stariji ispitanici su pokazali značajno kraće dužine koraka i veće širine koraka u poređenju sa mlađom kontrolnom grupom.*

Ključne reči: *analiza hodanja, kinematika, kinetika, elektromiografija, odrasli*