PERIODICA POLYTECHNICA SER. MECH. ENG. VOL. 49, NO. 1, PP. 63–70 (2005)

KINEMATIC AND KINETIC PARAMETERS OF HEALTHY ELDERLY PEOPLE

Róbert PARÓCZAI*, Zoltán BEJEK** and Árpád ILLYÉS**

*Department of Applied Mechanics
Budapest University of Technology and Economics
H–1521 Budapest, Hungary
Phone: 463–3678, Fax: 463–3471
e-mail: paroczai@mm.bme.hu
**Orthopaedic Department of Semmelweis University
Semmelweis University
H–1113 Budapest, Karolina út 27, Hungary
Phone: 466–6611, Fax: 466–8747
e-mail: bejzol@yahoo.com, illyes_arpad@hotmail.com

Received: January 18, 2005

Abstract

Walking is one of the most common human movements. It is 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 7 and remains at the same level until 60. In the elderly age walking performance starts to decline and it slows down gradually. 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 gait parameters may change as a result of aging. A total of 21 healthy, elderly subjects without any history of lower extremity joint pathology were investigated at self-selected pace. The gait analysis equipment used consisted of an infinitely adjustable treadmill with force-plates and ultrasound-based motion analyser. Spatio-temporal, kinematic, kinetic 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 significantly shorter step length and wider step width compared to results of a young control group. Our results showed that the aged individuals demostrated statistically less range of motion in the different joints during walking. We suggested that neurophysiological changes associated with aging might result in less certainty of the neuromuscular system in selecting a stable gait.

Keywords: motion analysis, gait, kinematics, kinetics.

1. Introduction

In Hungary the elderly group (defined as $\geq$ 60 years) represents a growing segment of the population. Walking is a learned activity, in which the moving body is supported successively by one leg and the other. Dynamic regulation of 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 [3, 4, 10, 14, 15].

Comprehensive gait analysis usually includes kinematics, kinetics and electromyography, and this complex information can only be obtained in a dedicated laboratory. However, simplified analysis using for example spatio-temporal parameters, can also be valuable clinically. The purpose of the present study was to analyse age-related changes in functional gait pattern in the healthy elderly.

2. Materials and Method

A total of 21 healthy volunteers (9 women and 12 men) were included in the study. Their mean age was 71.15 years (SD $\pm$ 9.14 years), mean weight 77.23 kg (SD $\pm$ 13.12 kg), and mean height 1.74 m (SD $\pm$ 0.22 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) [2]. The objective functional evaluation was based on three dimensional gait analysis.

The evalulation in the gait laboratory lasted about one hour and included the recording of lower extremities of kinematics and kinetics. Spatial co-ordinates for the determination of kinematic data were collected using an ultrasound-based Zebris CMS-HS system (ZEBRIS, Medizintechnik GmbH, Germany) in the Biomechanical Laboratory at the Budapest University of Technology and Economics. The measuring heads were positioned behind the individual. The five ultrasound triplets, with three active markers on each, were placed on the sacrum, left and right thighs and left and right calves (*Fig. 1*). The core tenet of the approach is that the orientation and position of a segment of the human body are determined by the position of three points per segment. These three points produce a segment-embedded reference-frame and act as fundamental points of the reference frame. The position of an investigated point in the same segment can be specified by its position. This, actually, means that the position of investigated anatomical points in the segment-embedded reference frame should be determined before the measurement. The position vector of the investigated point is provided by the ultrasound-based pointer. Any number of investigated anatomical points can be positioned to a measured triplet's active marker with the technique described. The position of the three basis points of each segment of the human body is to be measured during motion by the ultrasound device. The three fundamental points of a segment are represented by the three active markers to be fixed and fastened to the segment (*Fig.1*). The co-ordinates of an active marker can be calculated by triangulation from the distances between the transmitter sensor and the active marker concerned. The distances between the transmitter and the receiver can be calculated from the delay of ultrasound measured by the ultrasound device, as well as from the velocity of ultrasound in air, which depends on air temperature and air pressure. The position of anatomical points can be calculated from the co-ordinates of fundamental points and from the position of the investigated point in the segment-embedded reference frame on-line and displayed on the screen already in the course of measurement (*Fig. 2*) [8].

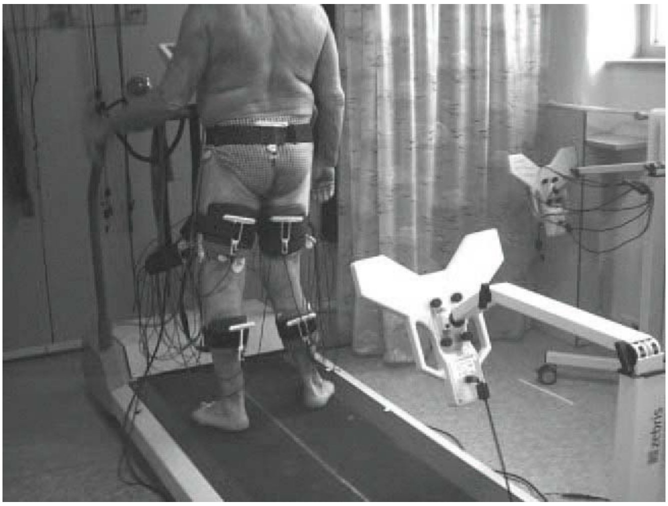

Fig. 1. Arrangement of measurement

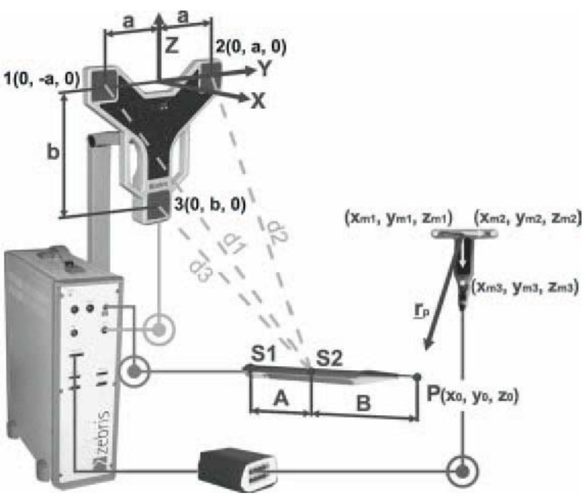

Fig. 2. Measurement method – determination of points

The data obtained from the measuring system recording of these active markers allowed for the determination of co-ordinates of nineteen anatomical points of the lower limb, such as the right and left medial malleolus, right and left heel, right and left lateral malleolus, right and left tibial tubercule, right and left fibular head, right and left lateral femoral epicondyle, right and left medial femoral epicondyle, right and left greater trochanter, right and left ASIS and sacrum. The biomechanical model developed by KNOLL et al. [6] was chosen for our investigation. The spatial co-ordinates were recorded at a frequency of 100 Hz. Simultaneously, the ground forces were measured at 1000 Hz. The patients were asked to walk at their natural, freely chosen velocity and cadence on a motorized and instrumented 330 mm $\times$ 1430 mm treadmill with a built-in force-plate (Bonte Zwolle B.V, Austria). Walking on the treadmill can initially be an unfamiliar experience, which in turn can influence the parameters measured. Therefore, measurements are to start after six minutes of familiarization time [1, 11]. Kinematic data were collected for six cycles. The assessed kinematic parameters were the following:

- Temporal and spatial parameters: stance, swing and double stance phase in percent of gait cycle; step length, step width (in milimetres); cadence (steps per minute);
- Angular parameters: knee, hip and pelvic angles presented by Kocsis and Beda [9];
- Force parameters: first peak force (F1) in the early stance phase and second peak force (F2) in the late stance phase (in percent of body weight).

The above parameters are calculated by software package presented first in [5].

3. Results

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

The fastest subject walked on the treadmill at a speed of 3.50 km/h, and the slowest at 1.80 km/h.

The absolute values of the various gait parameters are shown in *Tables 1–3*. Significant differences were not seen throughout the swing phase of the dominant (36.40 $\pm$ 1.24% and 39.93 $\pm$ 2.58% for females and males, respectively) and non-dominant limb (33.17 $\pm$ 2.98% and 36.86 $\pm$ 4.97% for females and males, respectively) ($p < 0.45$). Furthermore, the step length was shorter at the elderly (349.11 $\pm$ 60.36 mm and 363.25 $\pm$ 32.05 mm for females and males, respectively) as compared to a younger group of healthy volunteers (470.7 $\pm$ 20.1 mm and 513.12 $\pm$ 26.6 mm for females and males, respectively) [7]; the difference is significant ($p < 0.007$). The data for elderly people were summarized in *Table 1*, the data for younger volunteers were summarized in [7].

Table 1. Results of temporal and spatial parameters in healthy elderly subjects

Parameter		Unit	Female	Male
Cadence		steps per minute	87.59 ± 4.69	96.42 ± 18.35
Step length	Dominant side	Milimetre	349.11 ± 60.36	363.25 ± 32.05
	Non-dominant side	Milimetre	346.01 ± 35.43	339.92 ± 12.70
Step width	Dominant side	Milimetre	23.02 ± 3.12	21.97 ± 6.09
	Non-dominant side	Milimetre	27.22 ± 6.60	22.74 ± 3.86
Double support phase		% of gait cycle	13.41 ± 4.18	13.47 ± 3.43
Swing phase	Dominant side	% of gait cycle	36.40 ± 1.24	39.93 ± 2.58
	Non-dominant side	% of gait cycle	33.17 ± 2.98	36.86 ± 4.97

The amount of functional movement in the hip and knee joints was reduced on both sides (*Table 2*). Hip flexion showed a symmetrical pattern. The maximum hip flexion at the end of the swing phase ($56.12 \pm 3.56°$ and $51.20 \pm 13.5°$ for females and males, respectively) was not significantly smaller on the non-dominant side ($50.12 \pm 4.78°$ and $49.30 \pm 13.3°$ for females and males, respectively). The minimum hip flexion at the end of the stance phase ($11.89 \pm 3.78°$ and $9.41 \pm 5.78°$ for females and males, respectively) was slightly greater on the dominant side than that on the non-dominant side ($10.00 \pm 5.08°$ and $9.41 \pm 3.89°$ for females and males, respectively). The difference is not significant. The range of hip flexion on the dominant side during a gait cycle ($56.12 \pm 3.56°$ and $50.12 \pm 4.78°$ for females and males, respectively) was greater than on the non-dominant side ($50.12 \pm 4.78°$ and $49.30 \pm 13.3°$ for females and males, respectively). The range of the pelvic rotation during a gait cycle was $8.29 \pm 2.96°$ and $5.42 \pm 1.69°$ for female and male, respectively. The pelvic obliquity during a gait cycle was $2.65 \pm 0.38°$ and $3.12 \pm 1.87°$ for females and males, respectively). The knee showed symmetry of movement during a gait cycle. The range of knee flexion on the dominant side ($43.08 \pm 2.57°$ and $41.15 \pm 2.9°$ for females and males, respectively) were smaller than those on the non-dominant side ($39.67 \pm 1.79°$ and $40.45 \pm 3.1°$ for females and males, respectively).

The kinetic parameters (*Table 3*) revealed a certain degree of unloading on the non- dominant side. 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.

4. Discussion

The aim of this study was to analyse the resulting changes in functional gait patterns in healthy elderly subjects. A kinematic analysis objectively describes how the body segments of the subject are moving during gait. Movement analysis allows

Table 2. Results of angular parameters in healthy elderly subjects

Parameter		Unit	Female	Male
Hip flexion			±	
Range	Dominant side	degree	56.12 ± 3.56	51.20 ± 13.5
	Non-dominant side	degree	50.12 ± 4.78	49.30 ± 13.3
Maximum	Dominant side	degree	44.23 ± 6.78	41.30 ± 9.1
	Non-dominant side	degree	40.12 ± 4.57	33.67 ± 8.5
Minimum	Dominant side	degree	11.89 ± 3.78	9.91 ± 5.78
	Non-dominant side	degree	10.00 ± 5.08	9.63 ± 3.89
Pelvic rotation			±	
Range		degree	8.29 ± 2.96	5.42 ± 1.69
Maximum		degree	2.91 ± 2.6	6.37 ± 1.30
Minimum		degree	–5.38 ± 0.35	1.26 ± 1.15
Pelvic obliquity			±	
Range		degree	2.65 ± 0.38	3.12 ± 1.87
Maximum		degree	5.645 ± 1.58	3.97 ± 1.55
Minimum		degree	2.99 ± 1.19	0.85 ± 0.85
Knee flexion			±	
Range	Dominant side	degree	43.08 ± 2.57	41.15 ± 2.9
	Non-dominant side	degree	39.67 ± 1.79	40.45 ± 3.1
First peak	Dominant side	degree	16.21 ± 2.4	19.77 ± 2.94
	Non-dominant side	degree	27.45 ± 1.08	17.83 ± 2.36
Second peak	Dominant side	degree	56.89 ± 0.31	50.67 ± 2.58
	Non-dominant side	degree	48.5 ± 0.35	49.44 ± 3.78
Minimum	Dominant side	degree	17.22 ± 2.1	10.08 ± 2.08
	Non-dominant side	degree	15.41 ± 2.22	9.80 ± 2.88

Table 3. The results of force parameters in healthy elderly subjects

Parameter		Unit	Female	Male
F1 first peak in the early stance phase	Dominant side	% of body weight	137 ± 1	142 ± 1.3
	Non-dominant side	% of body weight	135 ± 0.8	137 ± 1
F2 second peak in the late stance phase	Dominant side	% of body weight	134 ± 1.4	136 ± 0.8
	Non-dominant side	% of body weight	132 ± 0.8	123 ± 1.1

calculations of the angle and range of motion.

In this research the spatial gait parameters show significant differences comparing to those of healthy young subjects [7]. On the other hand, the walking speed is slower and step length is shorter compared to the young subjects [7]. 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. Data are consistent with those of WINTER [15] and OBERG [12]. This decreased knee flexion and range of motion of knee in the elderly correlates well with their significantly shorter step length.

The hip and pelvic dynamic range of motion is larger in the elderly than in the young. Data are consistent with those of WINTER [15] and OBERG [12]. The increased hip and pelvic motion in the elderly was attributed to the need to put their hip extensors at a more favorable length so they can meet demand despite the weekness associated with aging [13].

We analyzed the free speed gait of elderly subjects. In elderly subjects, free speed gait represents linear power transfer from the leg to the upper body. The hip angle increased and the knee angle decreased compared with young subjects. These results support that the kinematic alterations in the hip are a cause of reduced gait speed in the elderly. We suppose that these changes are mainly due to age-related neuromuscular changes.

Acknowledgement

This work was supported in part by Hungarian Scientific Fund T049471 and T046126, and the Semmelweis Found. The authors wish to acknowledge to Professor Kocsis for providing access to the gait laboratory and assistance in data evaluation.

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