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BUTE'S ULTRASOUND-BASED MEASURING TECHNIQUE AND MODEL FOR GAIT ANALYSIS

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Abstract. *The paper focuses on the presentation of a new 3D motion analysis technique for treadmill walking. An ultrasound-based 3D measurement system and a developed measuring arrangement were used to measure and determine gait parameters during treadmill walking. The model considers each limb segment to be a rigid body, linked to each other by a joint. This paper also presents a new 3D motion analysis software package for treadmill walking and introduce the DataManager developed. We studied knee kinematics and temporal-distance gait measurement parameters (step length, stride length, stride width, etc.) to be obtained from treadmill walking. These measurements were highly correlated with and not significantly different to those in literature. Treadmill walking allows for the analysis of several cycles of each subject. On the basis of the analysis the standard deviation of temporal-gait parameters and the knee kinematics data of each subject can be established. The 3D movement analysis system presented is a suitable and standardized procedure for quick gait analysis.*

Key words: *motion analysis, gait analysis, knee joint, treadmill walking*

1. INTRODUCTION

Gait analysis can be described as a field of biomechanical engineering dealing with the subject of human locomotion. By means of different measuring techniques available (for example video recording), human gait data are captured (i.e. the gait pattern is recorded as a function of time) and further analysis and calculations are done in order to obtain all the data required for evaluating the quality of the subject's gait, including basic gait parameters (stride length, cadence, velocity, etc.), forces and moments occurring in the joints, muscle activity during each gait cycle, velocity and acceleration of each segment of the limb, etc.

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Since the measuring and recording techniques for capturing gait patterns have developed very much in the last decades, gait analysis is now frequently used in every-day practice of those involved in the rehabilitation of human movement. Therefore gait analysis has its application now in almost all considerable fields of human locomotion, both healthy and pathological: rehabilitation medicine, orthopaedics, kinesiology, sports science, and other related fields (Medved, 2001). But there are also many other fields in which gait analysis can be successfully applied sports or athletic applications (e.g. volleyball, soccer, tennis, horse racing, golf, etc.), post-injury assessment, disability evaluations, forensic research analysis of injuries sustained in football, gymnastics, horse-racing and running shoes, space research, industrial applications of product design, analysis, improvement, rehabilitation medicine etc. (Medved 2001).

There are many available software packages on the market such as BioWare/Gaitway (KISTLER), Ariel Performance Analysis System APAS/APASGait (Ariel Dynamics), StepPC (Median Systems), BTSwin/GaitELICLINIC (BTS Bioengineering Technology &Systems), KinTrak/OrthoTrak (MotionAnalysis Corporation), MOTUS (Peak Performance Technologies), Vicon 250/ 512 System (VICON), SIMI Motion (DataForce), Anthropo (SIMI) (Rishi, 1998). A number of 3D methods and techniques are specifically designed for the study of walking (Marzani et al., 2001; Stiehl et al., 2001; Alexander and Andriacchi 2001).

Most of kinematics models assume that the body is composed of rigid segments that are connected by ideal links (Andrews, 1995). Gait analysis systems use active or passive markers attached to the skin of the investigated body segments. At the recording phase these markers must be seen from both sides, perpendicular to the sagittal plane of the motion, meaning that segments could be positioned only on the external surfaces of the body. In these arrangements – as the displacements of the markers perpendicular to the sagittal plane are smaller than the components lying in the sagittal plane - , the relative errors of these components are usually not negligible and this fact influences the accuracy of gait parameters. Because of some technical problems (e.g. the space of the measurement is limited), usual gait measurements investigate only one ore two steps. The use of treadmills can only solve this problem, but this type of measurement is not widely known (Alton et al., 1998). Initially, it may seem that suitable tools for the quantitative study of human movement are available and that routine applications are feasible for instance in the treatment of motor disorders.

Nowadays clinical motion analysis is a usual method.

More and more laboratories offer their facilities and investigation is used for supporting doctors in their decisions. The hardware and software packages available on the market are with very different prices and calculation results. In the 21st century only those methods and calculation techniques have economical value which can meet the following requirments:

- Preparation for the measurement and the whole procedure takes up to one hour.
- No need for more than two persons' assistance during the measurement.
- The procedure for the post-processing of the measured data and printing doesn't require more than one additional hour from one expert person.
- The earlier usual errors caused by skin movements and positioning of devices and marker clusters could be reduced to minimum or eliminated.
- The evaluation technique takes several gait cycles (min 5-15) into consideration because their differences can significantly characterize the gait investigated.

- The model used can be easily changed and adopted to the requirements.
- The required space (room) is less than 20 m^2.

Unfortunately the most frequently used gait analysis methods do not fulfill the requirements mentioned. We really appreciate the long and tiring work that has to be done by scientists getting usable results from the measured data, but more economical techniques must be used if we would like to apply gait analysis for every-day clinical practice as MRI or CT are used.

A real problem is the errors of the measured data. The software packages used can answer almost all the questions, but the results sometimes are very far from reality.

There are very precise active marker systems (infrared, ultrasound, and magnetic), able to determine really accurately the spatial position of the markers. These systems have to be used.

Another problem is the errors caused by skin movements. The markers are mostly mounted on the skin of the subject. When, during a movement, the skin slides compared to the bone, this unambiguous relation between marker and body segment is lost, which may have consequences for the validity of measured results (Knoll et al., 2002). The traditional marker placement scheme, where the body segment axis in lower extremities is derived from skin marker positions on the anatomical landmark of the ankle, knee and hip, is not ideal in this respect (Cappozzo, 1991; Zahedi, 1987). Such inaccuracies have inspired the development of other marker placement schemes, where markers are mounted on relatively stable skin locations and the joint rotation axes and centers are subsequently derived through a calibration procedure or calculation (Alexander and Andriacchi, 2001; Lu and O'Connor, 1999). Further problems relate to the models for data reduction, movement representation and the not fully automatic calculation of different parameters (Woltring, 1994).

This paper describes a special method and technique for measuring gait, raw data corrections, post-processing of measured data, and determining all possible parameters of human gait developed at the Biomechanical Laboratory of the Budapest University of Technology and Economics.

This method and technique are objective and quantitative and solve almost all the problems previously mentioned.

2. METHOD AND SUBJECT

2.1. Method

A very simple idea or trick can absolutely eliminate or neglect the errors caused by skin movements: Let's use external marker clusters on rigid plates, to be fixed on the investigated body segment by a method which insures a fix and stabile position of the clusters during motion. During the calibration phase of the measurement, anatomical points should be added to these clusters as points of the same rigid body determined by the cluster. During the measurement, from the spatial coordinates of the markers positioned on the cluster, the coordinates of the added anatomical points can be easily calculated from the following equations.

Denoting the markers positioned on the cluster and their position vectors by 1, 2, 3 and $\underline{R}_1$, $\underline{R}_2$ and $\underline{R}_3$ the unit vectors ($\underline{e}_\xi$, $\underline{e}_\eta$, $\underline{e}_\zeta$)of the local coordinate system fixed to the cluster will be calculated by:

$$\underline{e}_\xi = \frac{(\underline{R}_2 - \underline{R}_1)}{\left|(\underline{R}_2 - \underline{R}_1)\right|} \qquad \underline{e}_\eta = \frac{(\underline{R}_3 - 0.5(\underline{R}_1 + \underline{R}_2))}{\left|(\underline{R}_3 - 0.5(\underline{R}_1 + \underline{R}_2))\right|} \qquad \underline{e}_\varsigma = \underline{e}_\xi \times \underline{e}_\eta$$

Denoting, in the local coordinate system, the constant position vector by

$$\underline{\rho}_P = \xi_P \underline{e}_\xi + \eta_P \underline{e}_P + \xi_P \underline{e}_P$$

its scalar coordinates can be determined by:

$$\xi_P = (\underline{R^*}_P - 0.5(\underline{R^*}_1 + \underline{R^*}_2)) \bullet \underline{{}^*e_\xi}$$

$$\eta_P = (\underline{R^*}_P - 0.5(\underline{R^*}_1 + \underline{R^*}_2)) \bullet \underline{{}^*e_\mu}$$

$$\varsigma_P = (\underline{R^*}_P - 0.5(\underline{R^*}_1 + \underline{R^*}_2)) \bullet \underline{{}^*e_\varsigma}$$

Here * denotes the values obtained from the calibration phase.

During the measurement the global coordinates ($\underline{R}_P$) of the anatomical point P will be calculated by:

$$\underline{R}_P = 0.5(\underline{R}_1 + \underline{R}_2) + \underline{\rho}_P$$

By the technique described, any number of anatomical points can be positioned to a given cluster; and by an on-line system, the spatial coordinates of the anatomical points can be calculated and presented on the screen already during the measurement.

If the cluster is fixed correctly to the body segment, no arbitrary skin movement under the "hypothetical anatomical point" has any effect on the result because the anatomical points are fixed as rigid body points to the cluster.

Three years ago, at the start of an EU5 research project, we used the criteria described for choosing an appropriate system for the measurement of the upper limb. An ultrasound based on-line active marker system came into consideration, which used special clusters named triplets (three small microphones were mounted on a plastic plate and could be correctly fixed on the investigated body segment). A special pointer existed already to specify the spatial coordinate of an invisible point using two microphones mounted on the visible part of the pointer. During the project Zebris GmbH- the owner of the system mentioned - developed a special software (ArmModel) that follows the technique described earlier to fix the anatomical points to the triplets. The measurements proved the idea, and by this method the errors of spatial coordinates reduced to 1 mm, neglecting the effect of skin movements. This gives us never supposed possibilities for precise measurements.

Later we extended the method described for gait analysis, but in this case we also developed a new measuring arrangement:

Generally the cameras, measuring heads, etc. are positioned perpendicular to the sagittal plane. The arm movements sometimes hide the markers or clusters, or in case of using treadmill for measuring more gait cycles, the handlebars at both side of the treadmill can also cause problems. Our arrangement also solves all these problems.

The Zebris measuring system has a special head consisting of three small ultrasound sources. They can be connected directly to the basic processing unit. Three ultrasound sources work sequentially and are mounted at a predefined distance to each other. The measuring head sensors send steady pulses to the marker and the delay of the signals is

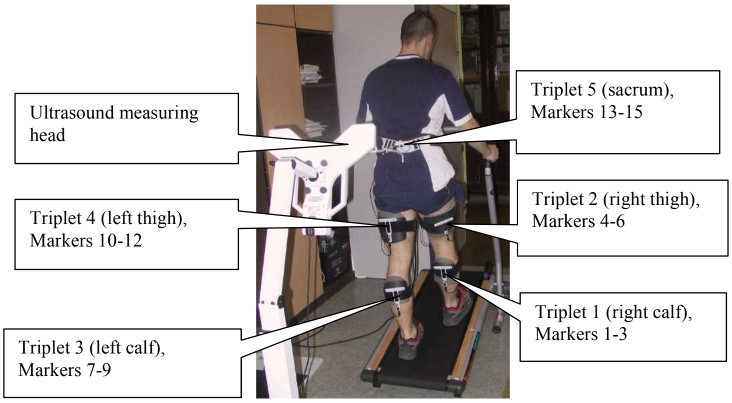

Fig. 1. A subject with triplets on the manual treadmill, with only one measuring sensor positioned at the back of the subject

measured. The absolute coordinates in space are calculated by triangulation.

The new measuring arrangement developed for gaitis as follows: the patient is walking on the treadmill, and the measuring head is positioned at the back of the treadmill. The triplets are attached to the sacrum, the thighs, and the calves, and are positioned backward (Figure 1). During motion, the system is measuring the coordinates of the triplets and the connected computer is calculating on-line the spatial coordinates of the investigated anatomical points virtually linked by a pointer to the appropriate triplet. In this case the anatomical points can be positioned at the "invisible " side of the body segments, because only the ultrasound heads must see the measuring triplets. Therefore the patient should stand on the treadmill within the reach of the sensors of the measuring head. During the calibration - by pointing the tip of the pointer to each anatomical point (on the surface of the skin) and pressing the button on the pointer - their location in space (with respect to the defined global coordinate system) is registered by scanning the signals of the pointer's two ultrasound microphones (Figure 2). The order of the anatomical points

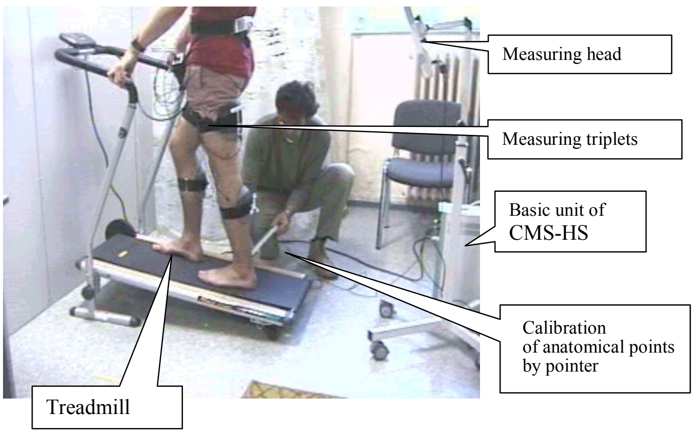

Fig. 2. Arrangement of the measurement and the markers during the calibration phase

is fixed according to the biomechanical model applied and has to be considered when entering them to the program. For investigating the fine motion of the knee joint, a special model was developed using 19 points (Figure 3).

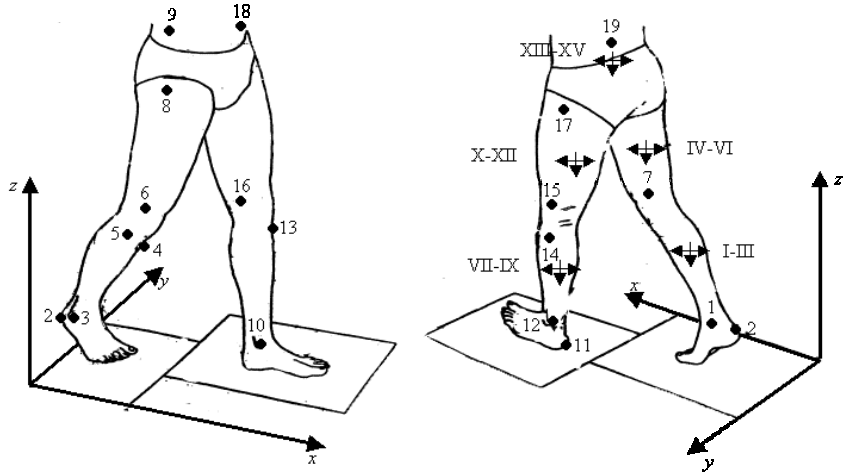

Fig. 3. Position of anatomical points and triplets for the 19-point-model: (1) right medial malleolus, (2) right heel, (3) right lateral malleolus, (4) right tibial tubercule, (5) right fibula head, (6) right lateral femoral epicondyle, (7) right medial femoral epicondyle (8) right great trochanter, (9) right asis, (10) left medial malleolus, (11) left heel, (12) left lateral malleolus, (13) left tibial tubercule, (14) left fibula head, (15) left lateral femoral epicondyle, (16) left medial femoral epicondyle (17) right great trochanter, (18) left asis, (19) sacrum. (I-III) triplet on the right calf, (IV-VI) triplet on the right thigh (VII-IX) triplet on the left calf, (X-XII) triplet on the left thigh, (XII-XV) triplet on the sacrum)

We use a treadmill allowing for recording multiple gait cycles, and the statistical analysis of the disturbances. Walking on the treadmill can initially be an unfamiliar experience. This, in turn, can influence the measured parameters. Therefore, the measurement starts after 6 minutes of familiarization time (Alton et al., 1998; Matsas et al., 2000).

After calibration, the patient can start the walking. The patient does not know the start and the end of the measurement and arbitrary gait cycles can be measured for further statistical analysis of the gait parameters.

2.2. Subjects

The study population consisted of thirty-one men and twenty women. The average age was 31.70±4.1 years; the mean height 1.71±0.12 meters and the mean weight 72.1±25.2 kilograms. For inclusion, subjects were not to have any pathology that would affect gait and had to be unfamiliar with treadmill walking. Each subject provided informed consent before participation.

2.3. Assessment parameters

The ArmModel software package is able to determine only the spatial coordinates of the anatomical points predefined in the program. The raw data are further refined, filtered, and post-processed by the newly developed DataManager.

DataManager is a macro-program for use with MS Excel, developed at BUTE for the international research project "REHAROB" (Kocsis, 2002). It is subdivided in several parts and enables the analysis of data measured by ArmModel (Figures 4 and 5). The main program allows the preparation of the measured data (e.g. elimination of errors) for further analysis by other modules. Automated batch processing of multiple data files with the same processing parameters is also possible.

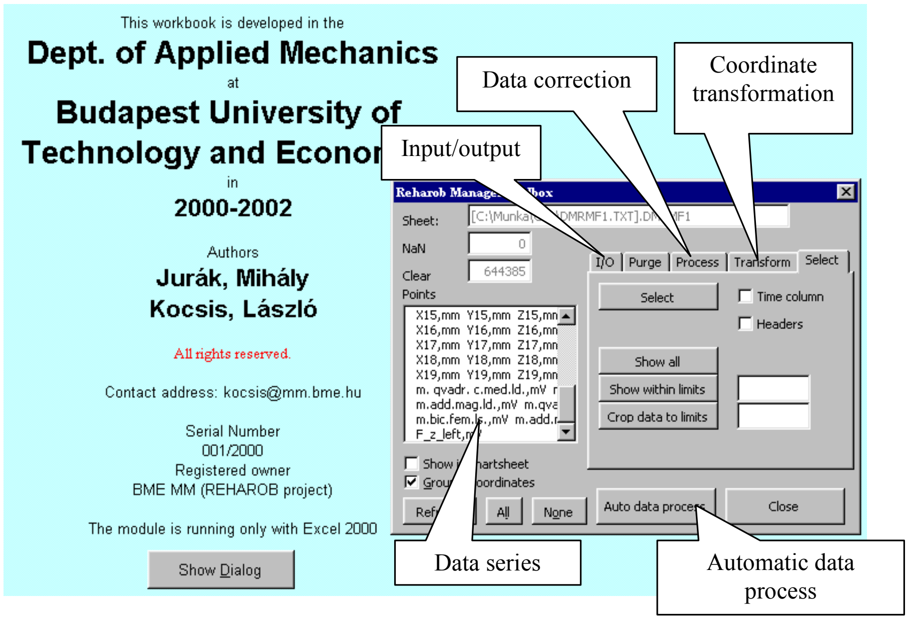

Fig. 4. Opening page of Data Manager

The package allows for dividing the motion into periodic gait cycles (Figure 6) and demonstrates all these gait parameters as a discrete function in each cycle. The measured gait coordinates and calculated parameters can be displayed in a diagram from 0 to 100% of the gait cycle. The parameters can be represented either relative to the right or to the left side cycle.

Temporal and distance gait parameters are also commonly obtained in clinical settings as they are easily measured and useful in the evaluation of pathological gait (Whittle, 1991). Temporal-distance gait parameters may be analyzed with individuals walking on a treadmill, and compared with normal values in literature. The following temporal-distance parameters are calculated from the coordinates of the anatomical points: stride length (the distance traveled during one stride measured between one heel strike to the next heel strike on the same side), step length (distance between the heel strike of one foot to the heel strike of the other foot), step width (mediolateral distance between the feet), stride time (duration of a single stride).

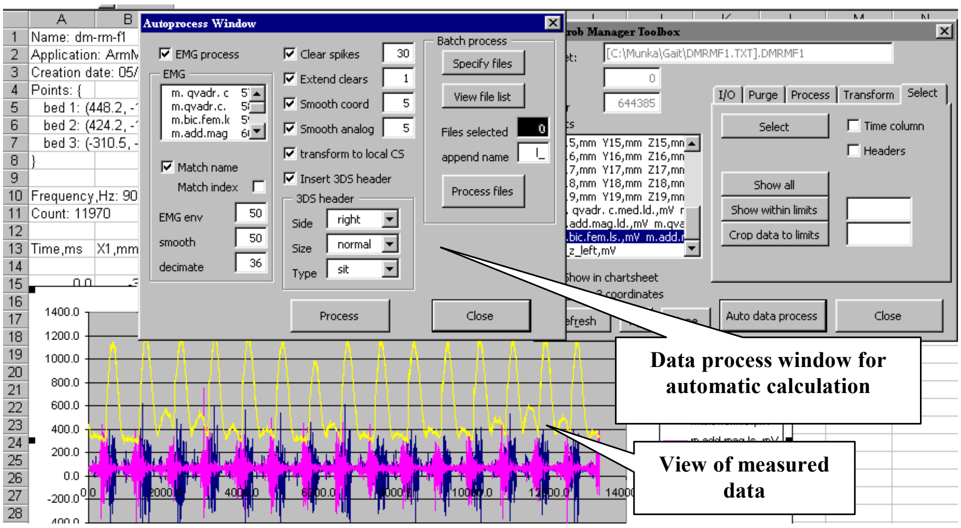

Fig. 5. Automatic calculation tool

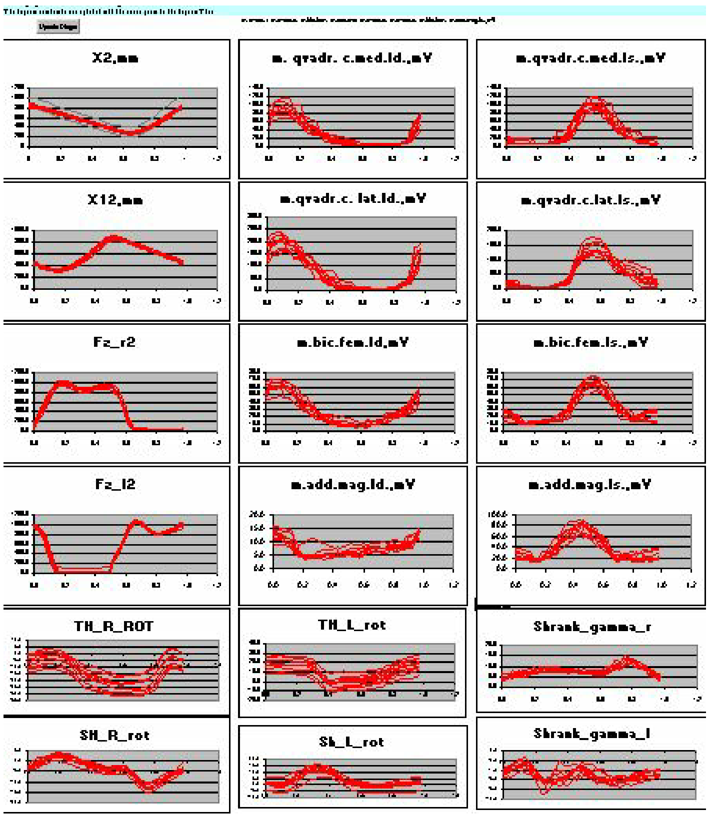

Fig. 6. Co-ordinates of anatomical points, vertical reaction components, and EMG envelope curves presented as a function of the gait cycle

Knee angles of the gait cycle play a major role with regard to the energy expended during walking and are commonly affected by pathological disorders (Whittle, 1991). The knee angle is defined as the angle between a straight line joining the lateral malleolus - fibula head and a straight line joining the lateral femoral epicondyle – great trochanter (Figure 7). For each subject, knee angles were calculated at four positions including initial contact, midstance, and peak values of extension and flexion. Midstance was defined as the point when the knee joint had attained maximum flexion after initial contact.

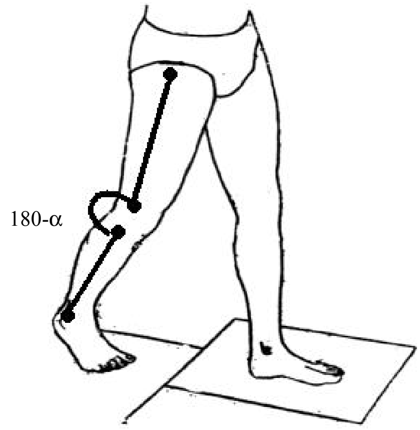

Fig. 7. Definition of the knee angle (α). The knee angle is defined as the angle between a straight line joining the lateral malleolus - fibula head and a straight line joining the lateral femoral epicondyle – great trochanter.

The behaviour of the knee depends on the movement of ligaments. The motion could be described by the newly defined relative ligament-points-movement parameter, which is the relative maximum displacement between the two characterized points of the knee. The two characterized points of the knee were chosen in a manner so that the line between these two points is closely parallel with the investigated ligament. For example, the anterior cruciate ligament - (ACL) movement - parameter is defined as the maximum relative displacement between the tibial tubercule and the lateral femoral epicondyle (Figure 8) divided by the minimum distance between those two points.

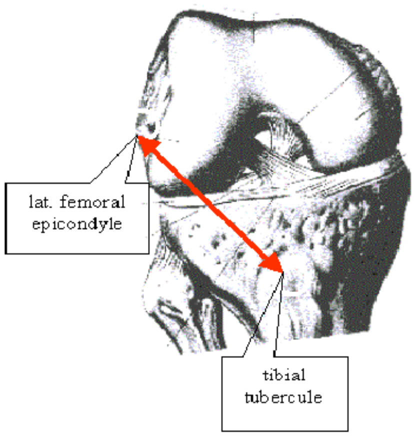

Fig. 8. Definition of the ACL movement parameter. The ACL movement is defined as the maximum displacement between the tibial tubercule and the lateral femoral epicondyle.

Treadmill walking allows to determine the average and the deviation of all parameters from six complete gait cycles for each subject.

3. RESULTS AND DISCUSSION

The validation of the new technique is shown on a few selected spatial-temporal parameters and knee joint kinematics, which are either very common or newly defined.

Spatial - temporal variables, such as the stride time, the stride and the step length, and the walking base, are derived from the temporal and spatial coordinates. For each subject, the average and the standard deviation of parameters were determined from six complete gait cycles. Figure 9 shows as an example, the stance, swing and double support phases for one subject's steps. As can be seen, these parameters differ at each gait cycle analyzed for one subject. The other parameters show similar differences, which are not significant ($p>0.47$). Table 1 summarizes the average values and standard deviation of these quantities at healthy female and male subjects. No significant differences were found between spatial - temporal variables of the left and right sides of one subject ($p> 0.37$) and between these variables of subjects ($p>0.41$). However, on the basis of the results we can establish that the step length and the walking base of the right step are greater (5-10%), than these of the left, and the step length, the walking base and the stride length of female-subjects are smaller than those of males. The spatial – temporal parameters presented in this study compare favorably with values found in literature (Whittle, 1992). The spatial – distances parameters correlated to values found in literature (Whittle, 1992).

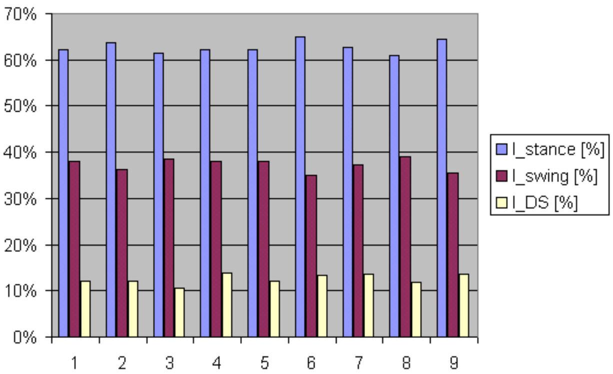

Fig. 9. The stance, swing and double support phases shown at one subject's steps

Figure 10 shows a graphical representation of one subject's knee angles. As can be seen, the knee angles differ at each gait cycle, similarly to spatial - temporal parameters. The average values and the standard deviation of the peak values of flexion and extension at 51 adults are summarized in Table 1. No statistical differences were found between the left and right sides of one subject ($p>0.55$) and between these values of subjects ($p>0.39$). The definition of the knee angle presented in this study does not evaluate frontal and transverse plane components. This is important in some pathological gait, where abnormalities occur essentially in these planes. The knee angle is not zero at extended legs, at heel-up phase of gait (Figure 10), because the knee angle models also the anatomical angle between the femur and the tibia in frontal plane.

The relative ligament - movement parameters are closely the same at each gait cycle. The average and standard deviance of all four ligament-movement parameters are sum-

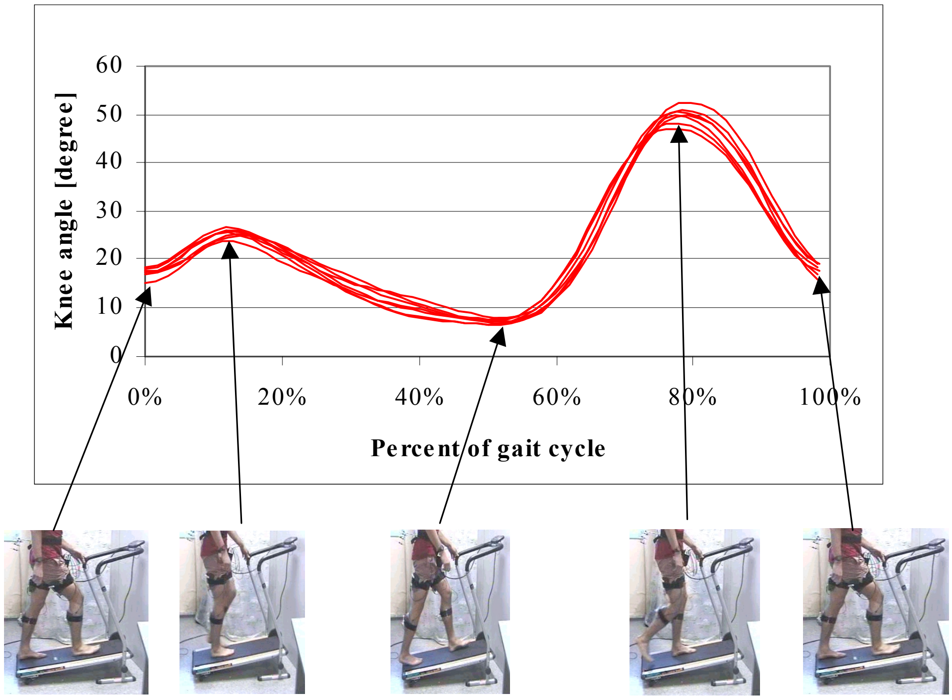

Fig. 10. One subject's knee angle function in time divided into gait cycles

marized in Table 1. On the basis of the results we can establish that the relative ligament movement parameters of male subjects are closely equivalent to values of females'. No difference was found between the values of the right and left sides. The relative ligament-movement parameters were used to quantify the tibial translation into the direction of ligaments. Let us see the characteristics of the relative ACL - movement parameter, which describes the condition of the anterior cruciate ligament (ACL). At heel strike with the knee at full extension, the tibia is at its maximum anterior position during the gait cycle. The next key event occurs at terminal extension where the tibia is located posterior again while the knee is near full extension. During the swing phase with the limb unloaded, the tibia moves to its maximum posterior position at maximum knee flexion, and then moves forward rapidly and the knee extends prior to heel strike. The overall range, the maximum displacement between the tibial tubercule and the lateral femoral epicondyle was used to quantify the characteristics of dynamic stability and the differences observed between normal subjects and patients of knees with anterior cruciate ligament deficiency. The measured data represent that the relative-ligament-movement parameters do not depend on gender and the dominant side. The values of these parameters depend only on the movement of the tibia with respect to femur and on the translation-motion of the femur's condylus, which depends only on the anatomical state of the knee.

Table 1. The average values and deviation of gait parameters

Parameters	Average	Standard deviation	Parameters in literature
Step length [mm]	513.3	26.6	350-790
Stride length [mm]	1012	25.47	600-1700
Walking base [mm]	40.9	8.2	9-75
Stride time [msec]	984	47.2	800-1780
Knee angle at initial contact (peak value in swing phase) [degree]	5.5	0.21	2-34
Knee angle at midstance (peak value in stance phase) [degree]	21.47	0.42	10-40
Knee angle at heel rise (peak value of extension) [degree]	6.47	0.14	0-11
Knee angle in middle of swing phase (peak value of flexion) [degree]	55.87	0.71	25-90
Relative ACL-points-movement parameter [-]	0.25	0.014	-
Relative PCL-points movement parameters	0.34	0.016	-
Relative LCL-points movement parameters	0.32	0.032	-
Relative MCL-points movement parameters	0.062	0.0044	-

4. CONCLUSIONS

The method suggested needs less space for gait measurement and uses only one measuring head positioned at the back of the treadmill. In this case the errors of the displacements perpendicular to the sagittal plane are reduced and the handlebars of the treadmill and the arm movements cause no disturbances.

The described powerful and very advanced measurement method for the analysis of the lower limb has been well established in biomechanics research and clinical applications for a long time. The 19-point-model is verified by measuring knee movement and temporal-distance gait parameters during treadmill walking. These measurements obtained after 6 minutes of treadmill walking were highly correlated with and not significantly different to those given in literature.

Treadmill walking allows to analyze several gait cycles of one subject and to determine the deviation of parameters for each subject, which are calculated automatically by DataManager.

The 3D motion analysis presented in this paper is suitable for clinical gait analysis.

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BUTS-OVA ULTRAZVUČNA OSNOVA MERNE TEHNIKE I MODEL ZA ANALIZU HODA

Laszlo Kocsis, Rita Kiss, Zsolt Knoll, Mihaly Jurak

Rad se fokusiran na prezentaciju nove 3D analize tehnike hodanja na tredmilu. Ultrazvučna osnova 3D sistema merenja i razvijeni aranžman kretanja su korišćeni da izmere i utvrde parametre hoda tokom hodanja na tredmilu. Model podrazumeva da je svaki rasčlanjeni segment uda odvojen deo, povezan jedan sa drugim zglobovima. Ovaj rad takođe prikazuje novi paket softvera za 3D analizu kretanja hodanjem na tredmilu koji je Datamenadžer razvio. Mi smo proučavali kinematiku kolena i vremensku razliku merenih parametara hoda (dužina koraka, dugačak korak, širinu koračanja, itd) dobijenih hodanjem na tredmilu. Ova merenja su visoko korelantna i ne značajno različita od onih u literaturi. Hodanje na tredmilu omogućava analizu u nekoliko ciklusa za svakog subjekta. Na osnovu analize kod svakog subjekta se može utvrditi standardna devijacija parametara vremena hoda i podataka kinematike kolena. 3D sistemska analiza pokreta prezentovana je kao odgovarajuća i standardna procedura za analizu brzog hoda.

Ključne reči: *analiza pokreta, analiza hoda, koleni zglob, hod na tredmilu*