Gait measurements in the transverse plane using a wearable system: An experimental study of test-retest reliability

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An experimental study of test-retest reliability

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***Abstract*—3D gait analysis comprises the study of kinematics in the sagittal, coronal, and transverse planes. The transverse plane measurements are usually less used and generally show the lowest reliability. Nevertheless, the knee and ankle joint center trajectories, in the transverse plane, provide new parameters that may be important in clinical gait analysis. The aim of this study is to analyze the test-retest variability of these parameters. Gait measurements were performed using H-Gait, a wearable system based on magnetic and inertial sensors. A normal weight and an overweight subject were recruited and were asked to walk at their preferred speed for 6 trials. For both of them, the angle between the right and left knee and ankle joint center trajectories were analyzed. Overall, results showed a standard deviation across trials always lower than 2°. This small standard deviation was found also in the overweight subject, for whom it is usually challenging to obtain reliable gait measurements. In addition, a greater knee angle between the right and left joint center trajectories was found in the overweight subject compared to the normal weight. The promising results of this study suggest that the new parameters introduced might be suitable to assess gait of subjects with different anthropometric characteristics.**

Keywords— magneto-inertial sensors; wearable sensors; repeatability; transverse plane; overweight; gait analysis.

I. INTRODUCTION

Walking is defined as the movement determined by a pattern of cyclic motor activity of lower limbs that allows the forward propulsion of the center of gravity of the human body. During gait, the body weight is alternately transferred to a single leg so the contralateral lower limb can advance [1]. Gait analysis measures and describes the quantities that characterize this cyclic motor activity. Since a qualitative analysis of human gait achieved by visual inspection does not provide objective and reproducible results, quantitative gait analysis is used in clinics to identify deviations from normal gait [2]. Quantitative gait parameters are divided into spatio-temporal and kinematic. Spatio-temporal parameters (i.e. walking speed, cadence, and step length) are obtained from gait events. Kinematic parameters provide information on linear and angular displacements of lower body segments (tight, shank, and foot) and joints (hip, knee, and ankle). The determination of cycle-to-cycle spatio-temporal parameters and kinematic curves is clinically relevant since the parameters variability has been associated with increased fall risk, frequent geriatric syndromes, post-stroke patients [3], progression of Parkinson’s disease or other gait related pathologies. For an accurate gait assessment, the used technologies should allow the subject to have a natural gait pattern and to walk for many consecutive gait cycles [4].

To perform gait analysis, optical and non-optical systems are used. Optical systems are based on a set of markers and cameras to reconstruct 3D motion. Non-optical systems are divided in: non-wearable, such as force platforms, and wearable, such as electromechanical systems (foot-switches and electrogoniometers) and magnetic and inertial measurement units. To some degree, optical systems and force platforms present the same drawback: gait analysis has to be limited to a laboratory setting and to a small acquisition volume. A wider work space is required to collect a considerable number of consecutive strides for cycle-to-cycle measurements. For this reason, it is difficult to use them in daily life or in non-traditional environments [5]. Other techniques such as wearable sensors have been developed [6]. In the clinical setting, wearable systems based on electro-goniometers and foot-switches are appreciated for their high accuracy [7]. However, with these systems, the measurement of joint kinematics is limited to a single plane, usually the sagittal one.

Wearable magnetic and inertial measurements unit (MIMUs), due to their limited weight and size represent a recent solution to monitor daily life [8],[9]. MIMUs usually consist of three-axis accelerometers, gyroscopes, and magnetic sensors, measuring the sum of gravitational and linear accelerations, angular velocities, and local magnetic field vector components, with respect to a Cartesian reference system fixed with the MIMU. Orientation and position in 3D space can be calculated by sensor fusion algorithms [10],[11]. The algorithm most used implements the extended Kalman filter [12]. The Kalman filter uses a series of measurements observed over time, containing statistical noise and other inaccuracies, and calculates variable estimates that are more precise compared to those based on a single measurement.

Although kinematics in the sagittal plane provides the most reliable information [13], it is only partial. Indeed, 3D intersegment moments are of great interest [14]. As an example, hip abduction moments are critical in maintaining balance and are assessed in the frontal plane [15]. Furthermore, the knee and ankle joint center trajectories are useful in characterizing the gait of healthy subjects and patients affected by osteoarthritis and are studied in the transverse plane [16]. Moreover, it would be interesting to evaluate if kinematics in other planes than the sagittal one can provide useful information for an early detection of specific pathologies (i.e. osteoarthritis, varo-valgus knee).

In clinical gait analysis, it is important to have reliable measurements for subjects with a wide variety of anthropometric characteristics. For this reason, when using wearable sensors, also overweight people should be included in the analysis. For overweight subjects, it may be more challenging to acquire gait measurements due to specific technical difficulties related to excess body mass and soft tissue artifacts.

The aim of this experimental study is to evaluate the test-retest reliability of the joint center trajectory angle, in the transverse plane, for both knee and ankle, using the MIMUs system H-Gait. To this purpose, a normal weight and an overweight subject were tested with a wearable system based on inertial sensors (H-Gait), during 6 gait trials.

II. MATERIALS AND METHODS

A. Participants

One normal weight (NW) and one overweight (OW) young male subject with body mass index (BMI) of 21.2 and 30.4 kg/m^2 respectively, were involved in this study. Anthropometric data of the two volunteers who participated in this study are reported in Table I.

TABLE I. ANTHROPOMETRIC DATA FOR THE TWO SUBJECTS

Anthropometric data	***Subject#1 (NW)***	***Subject#2 (OW)***
Age (years)	26	22
Weight (kg)	71	91
Height (m)	1.83	1.73
BMI (kg/m^2)	21.2	30.4
Tight length (cm)	44.5	32.0
Shank length (cm)	43.0	39.5
Distance between the great trochanters (cm)	37.0	39.0

B. H-Gait system

The H-Gait system was used in this work to acquire gait signals. This system consists of seven MIMUs (TSDN121, ATR Promotions, Japan). Each MIMU sensor is composed of tri-axial accelerometer, gyroscope, and magnetometer. The sensors were fixed to the subject as described previously in [10]: 1 sensor on the pelvis (posterior center point between the left and right iliac crest), 2 sensors on the thighs (the center of the quadriceps laterally), 2 sensors on the shanks (anterior side of the tibia bone) (Fig. 1A). These locations were chosen to minimize the effects of soft tissue movement. In addition, two sensors were fixed on the feet medially (Fig. 1B) [17]. In this position, sensors undergo less relative movements with respect to the foot, because of a better fixation and reduced soft tissue artifacts [17].

Measurement range of the MIMU sensor was set to ±4 G for the accelerometers and ±500 dps for the gyroscopes. A sampling rate of 100 Hz was chosen for both. Acceleration and angular velocity were collected in real time and sent to a laptop via Bluetooth.

C. Gait experiment

Experiments were conducted indoors. First, 10 reflective markers were placed on the subject on both left and right side (great trochanter, medial and lateral epicondyle of femur, medial and lateral malleolus). Three pictures of the subject were taken from the front and the two lateral sides for the anatomical calibration [17].

Then, the subject was prepared for the test: MIMU sensors on the pelvis and on the lower limbs were fixed by elastic Velcro, while the two on the feet were fixed by adhesive tape. Five seconds of signal were recorded with the subject in standing and sitting position (with outstretched legs) to obtain information on the initial position of the sensors with respect to gravity (static calibration).

Combining static and anatomical calibration the roto-translation matrix between the MIMU sensors and the body segments coordinate system was established [10].

Each subject was requested to walk along a 14 m straight path, at his self-selected speed. Six consecutive gait trials were performed. The subject walked along the path, at the end he turned to change the direction and stopped in a standing position for a few seconds before he started to walk in the

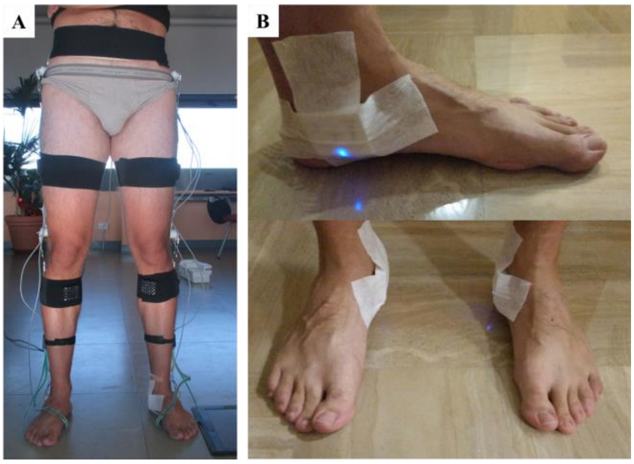

Fig. 1. Position of the unit sensors on the lower limbs. A. Sensors location for the pelvis (center point between the left and right iliac crest), thighs (the center of the quadriceps laterally), shanks (anterior side of the tibia bone. B. Sensors location for the feet (medially).

opposite direction. Each gait trial consisted of 8/9 gait cycles, with a total of 53 gait cycles for the normal weight subject and 54 for the overweight subject.

D. Transverse plane measurements

Gait analysis data collected with H-Gait system were evaluated in all the three planes. In particular trajectory of the center joints were considered in the sagittal, frontal, and transverse planes [10],[18],[19]. Similarly to what was presented in [16], the joint center trajectory in the transverse plane was calculated for both knee and ankle joint, bilaterally. Then, the approximation line [16] of each mean trajectory was estimated. Finally, the angle between the left and right approximation lines was calculated for both knee (θ_k) and ankle (θ_a) joint trajectories.

In addition, conventional gait spatio-temporal parameters such as walking speed, cadence, and step length were assessed. To estimate cadence and step length, the average values of left and right gait cycles were considered.

III. Results

Fig. 2 shows the knee (top panel) and ankle (bottom panel) joint center trajectories, in the transverse plane, for the two subjects. The approximation lines are also reported for each joint center trajectory. The walking direction is indicated by the arrow on the left.

Table II reports the transverse plane angles θ_k and θ_a, measured in each gait trial, for the two subjects. For each subject, the mean value and standard deviation (SD) across the 6 trials are also reported.

Table III reports the average and SD for spatio-temporal parameters (walking speed, cadence, and step length) across the 6 trials, for the two subjects.

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TABLE II. Transverse Plane Parameters for the Two Subjects

Transverse Plane Angles	***Subject#1 (NW)***	***Subject#2 (OW)***
θ_k: angle between right and left knee joint center trajectory (°)		
Trial #1	11.0	38.0
Trial #2	13.7	36.1
Trial #3	13.0	34.4
Trial #4	12.0	35.4
Trial #5	12.3	39.2
Trial #6	11.2	35.8
Mean ± SD	**12.2 ± 1.0**	**36.5 ± 1.8**
θ_a: angle between right and left ankle joint center trajectory (°)		
Trial #1	7.6	9.1
Trial #2	8.6	11.4
Trial #3	7.7	10.0
Trial #4	9.1	9.2
Trial #5	9.2	10.7
Trial #6	9.0	10.3
Mean ± SD	**8.5 ± 0.7**	**10.1 ± 0.9**

TABLE III. Spatio-Temporal Parameters for the Two Subjects

Spatio-Temporal Parameters	***Subject#1 (NW)***	***Subject#2 (OW)***
Walking speed (m/s)	0.97 ± 0.01	1.03 ± 0.01
Cadence (cycle/min)	46.9 ± 0.7	55.1 ± 0.3
Step length (cm)	55.1 ± 8.0	50.2 ± 2.1

IV. Discussion

In this work we analyzed the test-retest reliability of the knee and ankle joint center trajectory angles, in the transverse planes, comparing 6 consecutive gait trials. The experimental protocol adopted consider two sources of variability: intra-subject variability (due to differences in gait stride along the 6 walking) and the variability due to possible inertial sensors residual drift effects [20]. On the other hand, other sources of variability, such as those related to sensors re-positioning or inter-operator variability, were not considered.

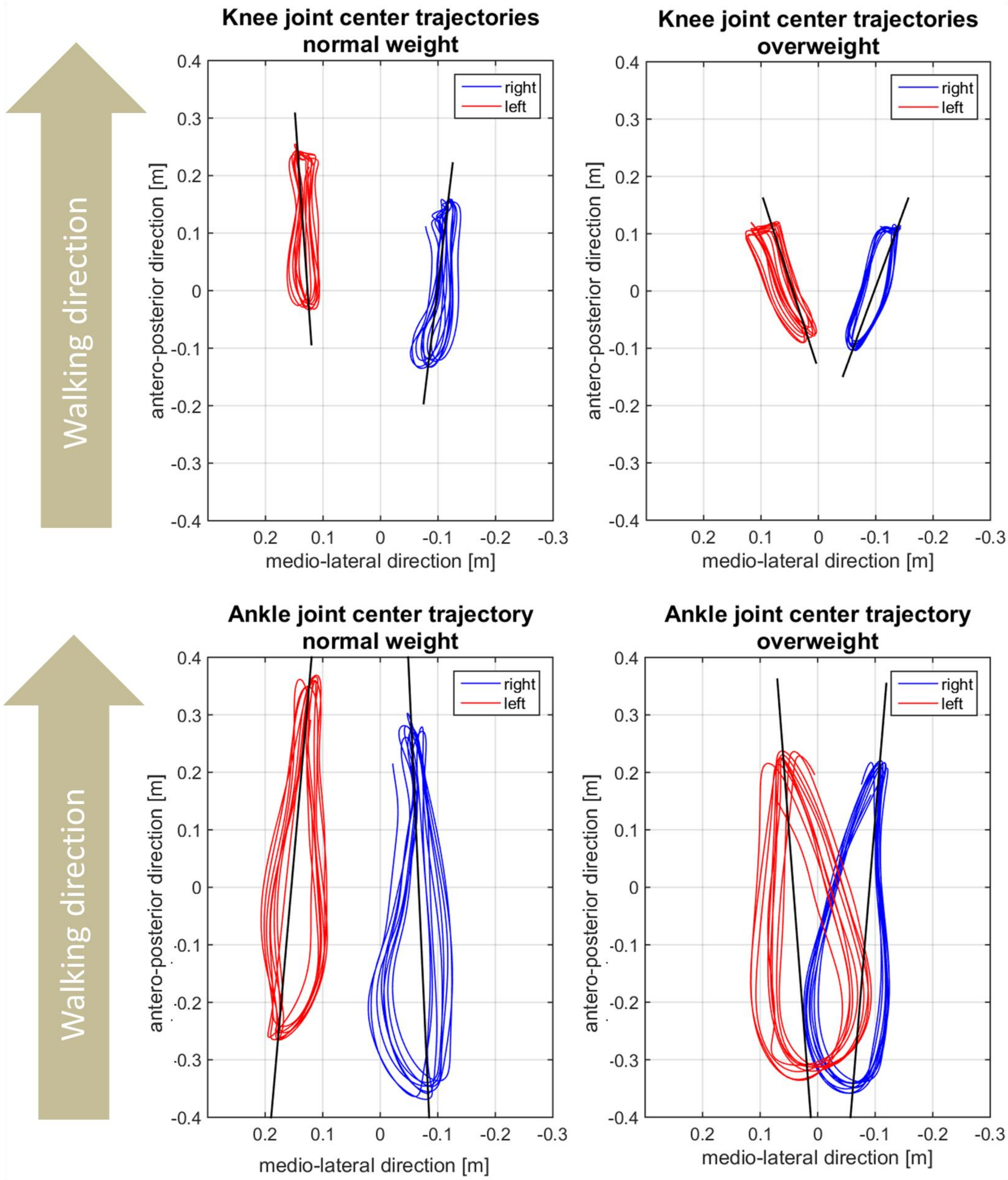

Fig. 2. Knee (top panel) and ankle (bottom panel) joint center trajectories for the normoweight and overweight subjects. The blue lines represent the trajectories of the right side, while the red lines represent those of the left side. The medial part of the trajectories corresponds to the stance phase, while the lateral part to the swing phase. Approximation lines of the trajectories are drawn in black. The walking direction is represented by the arrow on the left side of the picture.

Literature recognizes that kinematic measurements in the transverse plane generally show the lowest reliability [13]. Nevertheless, we found that the standard deviation across the 6 trials, for the knee and ankle angles, was always lower than 2°. In particular, for the θ_k angle, the standard deviations were 1.0° for the normal weight subject and 1.8° for the overweight one. For the θ_a angle, the standard deviations were 0.7° and 0.9°, for the normal weight and the overweight subject, respectively. These absolute errors are low enough to be compatible with clinical gait analysis.

In motion analysis, subcutaneous adipose tissue makes the kinematic measurements particularly challenging in overweight individuals [21]. Nevertheless, our study showed an acceptable repeatability also for the overweight subject. This suggests that the chosen sensor positioning (location and fixation) was suitable also in presence of excess adipose tissue.

We found that the knee angle θ_k is definitely higher in the overweight subject (36.5° ± 1.8°) with respect to the normal weight subject (12.2° ± 1.0°). Indeed, the knee approximation lines are more parallel to each other for the normal weight subject than for the overweight subject. The area of the knee and ankle joint center trajectories for the overweight subject are greater than for the normal weight subject, especially in the middle of the stance and swing phase. These differences in knee and ankle joints kinematics could be explained by a higher thighs girth that probably forces the overweight subject to have a different gait biomechanics. An altered gait biomechanics together with increased load in weight-bearing joints (hip and knee) can be a factor in osteoarthritis development for overweight and obese people [22],[23]. However, to verify this assumption, further analyses comparing normal weight and overweight subjects have to be performed. In addition, abduction/adduction of hip, varus/valgus rotation of the knee, and plantar/dorsal flexion of the ankle joints should be considered to explain this phenomenon.

Concerning spatio-temporal parameters, the literature reports that overweight and obese subjects have lower preferred walking speed and shorter step length when compared to non-obese individuals [24],[25]. However, the overweight subject that we analyzed showed a comparable preferred speed with respect to the normal weight subject. This allows the proper comparison of the kinematic patterns of the two subjects that could otherwise be biased by a different walking speed. According to the literature, a shorter step length was found for the overweight subject compared to the normal weight. This shorter step length can also be seen in the transverse plane in the knee and ankle joint center trajectories, which are shorter in the walking direction for the overweight subject (Fig. 2, top panel).

A limitation of the study is that the investigation was performed only on two subjects (one normal weight and one overweight). A wider sample size should be considered to confirm our findings on the kinematic test-retest reliability and on gait biomechanics differences.

V. Conclusion

Although transverse plane measurements are usually affected by large errors, in this study we found small SDs for the angles between the left and right joint center trajectories, both for the knee and the ankle joints. These small SDs were found not only in a normal weight subject, but also in an overweight subject, for whom it is usually challenging to obtain reliable gait measurements. In addition, the higher knee angle between the left and right joint center trajectories in the overweight subject compared to the normal weight subject suggests a difference in gait biomechanics, which could be due to a greater tight girth and/or a different load distribution in the weight-bearing joints.

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