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Experimental Validation of the Tyndall Portable Lower-Limb Analysis System with Wearable Inertial Sensors

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Abstract

Biomechanical analysis of movement during sport practice is extremely useful to assess and, subsequently, optimise movement performance during sport which can also assist athletes during rehabilitation following injury (such as Anterior Cruciate Ligament reconstruction). It is mostly performed using camera-based motion analysis systems, which provide good results but present serious drawbacks (for instance, consistent size, high cost, and lack of portability). Thus, small-size low-cost wearable sensors are an emerging tool for biomechanics monitoring. Aim of the present work is to implement a novel wireless portable easy-to-use system, consisting of two Tyndall Wireless Inertial Measurement Units (WIMUs) per leg, suitable for free-living environments and able to provide a complete biomechanics assessment (generated on a report) without the constraints of a laboratory. Validation for the lower-limbs using state-of-the-art camera-based motion capture is presented here. Algorithms are implemented in Matlab, and the scenarios considered simulate a free-living environment and exercises performed in a rehabilitation procedure. The system has been validated with healthy and impaired subjects. This novel system shows high accuracy values for all considered scenarios. Moreover, it is able to detect atypical movement characteristics. The results of this feasibility study support the next phase which will be to assess the external and ecological validity of athletes' on-field movement performance, which will help to inform individualised training protocols or enhance targeted rehabilitation programmes.

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Keywords: Inertial Sensors; Wearable; Knee Joint; Rehabilitation; Athletes

1. Introduction

Research has found that the most frequently injured joint is the knee, with approximately 2.5 million sports-related knee injuries presenting to Emergency Departments annually [1]. Anatomical structures that are most frequently injured are ligaments, meniscus and the patella [2]. The Anterior Cruciate Ligament (ACL) is one of the most significant and commonly injured ligaments in the knee joint [3, 4]. Knee injuries have negative consequences on sport activity and participation, and can have a long-term impact on movement performance during sport. Therefore, individualized and targeted rehabilitation programs are essential to enable athletes return to their pre-injury level of sport performance and prevent further injury. Rehab programs are based on a sequence of therapeutic exercises to restore joint range of motion, muscle strength, neuromuscular coordination, and gait mechanics. In particular, gait evaluation is an important aspect for the rehabilitation of lower extremity injuries since, understanding the normal gait pattern, will enable clinicians and sports medicine specialists identify and correct movement compensations after injury.

At present, gait analysis in clinical practice is mostly done by visual observation; however, this is inadequate for an accurate assessment. Moreover, self-reported scores (such as WOMAC [5], or Oxford Knee Score [6]) are not reliable due to their subjectivity. Camera-based 3D-motion analysis systems (e.g. VICON [7], OptiTrak [8], or Codamotion [9]) can provide an accurate and quantitative assessment of gait both as per kinetic and kinematic parameters. Thus, such a technology is not only a useful clinical tool, but it also represents the gold-standard in the objective measurement of human movement. However, it presents many feasibility issues including high costs, complexity and lengthy procedure, and is inconvenient for regular clinical use. Additionally, it may provide an altered representation of gait due to the Hawthorne Effect.

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A potential solution to the problems associated with lab-based analysis systems is given by the emerging area of small-size low-cost wearable inertial sensors for biomechanics monitoring. These sensors have indeed already been used in the monitoring of gait mechanics [10-13], lower-limbs joint range of motion [14, 15], and muscle strength [16, 17] with successful outcomes.
However, despite the great amount of work presented in literature on inertial sensors for biomechanics, such a technology has been adopted for monitoring lower-limbs during rehabilitation or tele-rehabilitation only in few cases (for example [18, 19]). In even fewer cases, inertial sensors were adopted to assess injured athletes' joint movement during rehabilitation (such as, ankle [20] or shoulder [21]). To the best of authors' knowledge, only one study [22] has investigated athletes' movement following knee injuries; however, the proposed solution (e.g. a full-body suit equipped with 10 wearable sensors) may be cumbersome.

The aim of the present work is, therefore, to implement a wireless portable easy-to-use system, with two sensors per leg, suitable for free-living environments and able to provide a complete biomechanics assessment (generated on a report) without the constraints of a laboratory. The system evaluates both gait and joint range of motion for several scripted activities which are typically performed during the rehabilitation process and also in a daily routine. The system, tested with both healthy and impaired athletes, provides the possibility to evaluate gait during rehabilitation and to identify gait abnormalities. The derived outcome can be analysed by clinicians and sport scientists to study the overall patients' condition and provide accurate medical feedback. Validation for the lower-limbs using state-of-the-art camera-based motion capture has been also carried out. This work was a phase one feasibility study designed to experimentally validate the system implemented in Tyndall for lower-limbs motion capture. In order to further validate the drawn conclusions in statistical terms, additional clinical trials, with larger and homogeneous populations, are needed and currently being planned. This feasibility study represent a prerequisite to a larger cross-sectional/cohort trial that will assess sport performance analysis and movement pattern alterations detection in people following knee injuries pathologies through wearable inertial sensing technology.

2. Methodology

The parameters selected for assessment were

- Temporal events: toe-offs, heel-strikes, mid-stance;
- Temporal intervals: gait cycle duration, stance phase, swing phase, single and double support, cadence (or step rate), number of cycles, swing symmetry;

The movement of the lower-limbs during walking can be divided into two parts, the stance phase and the swing phase. The stance phase (the weight-bearing part of the gait cycle) starts with initial contact at heel strike and ends at toe-off, while the swing phase (the interval when the foot is moving in the air) starts with the toe-off and ends at heel strike (Figure 1-left). Toe-off, heel strike and mid-stance (the instant when the foot is completely in contact with the ground), the most basic gait features, can be extrapolated from the angular rate signal collected on the sagittal plane of the shank (Figure 1-right). Toe-off is observed before the foot leaves the ground and is detected from the point with the local maximum velocity in the negative direction. After reaching a maximum positive point (mid-swing), the shank's rotation passes from a direction (positive) to the other one (negative) exactly when the heel lands on the ground. The zero-crossing point is therefore used as a reference for the heel-strike event. Mid-stance is characterized by the shank being vertical upon the ground, thus is estimated as the moment when the shank's orientation is parallel to gravity. The gait cycle duration is the time-interval between two consecutive toe-offs of the same leg, while the cadence (the number of strides in a time unit) is its reciprocal. Finally, observing the two legs as an entire system, the single support occurs when only one of the two legs is in contact with the ground while the other is swinging, whereas the double support is given by the time interval in which both legs are in contact with the ground. Additionally, the swing symmetry is the ratio between the swing phases evaluated at the left and right leg.

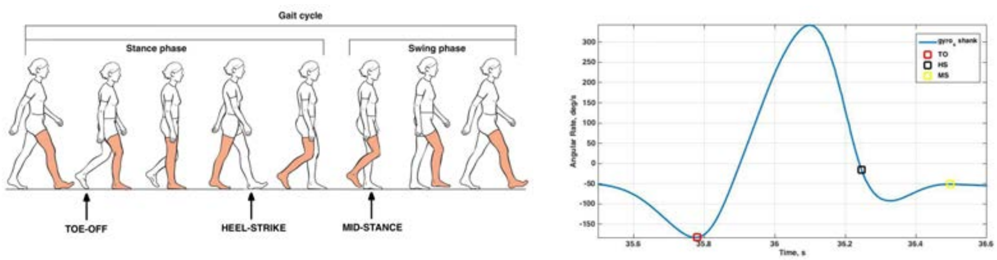


Fig. 1. (left) Walking gait phases; (right) Gait characteristic instants from gyroscope signal on the sagittal plane.

- Spatial parameters: stride length, stride velocity (or speed), peak angular velocity, shank clearance;

Stride length is computed as the total trajectory on the sagittal plane made by the sensor attached to the shank. The approach is based on a double integration (from acceleration projected into the global coordinate system, through velocity, to displacement), of inertial data collected from the sensor in a stride (e.g. between two consecutive mid-stances). The integration process is reset at the end of each stride. The double integration process requests the initialization of four variables: initial velocity, and initial displacement (horizontal and vertical for both cases). Differently from [13], the assumption of null initial horizontal velocities at mid-stance has not been considered because of the inverted-pendulum model. Thus, for each stride, the boundary velocity on the horizontal axis has been estimated by multiplying the angular rate on the sagittal plane observed at mid-stance and the length of the lever (e.g. the distance from the sensor and the inverse pendulum pivot – in this case, the ankle joint). Self-selected stride velocity (or speed) is a powerful mean to evaluate subjects' strength and control during gait and is computed by the ratio between stride length and gait cycle duration. The peak angular velocity is the maximum value recorded on the shank's sagittal plane by the angular rate sensor during a gait cycle. The shank clearance is, instead, defined as the maximum height reached by the sensor during the swing phase and is obtained by the vertical displacement calculated to establish the stride length during the swing phase.

- Knee joint range of motion.

Knee joint angles on the three axes typically referred to flexion-extension, varus-valgus and internal-external rotation. The alignment of the inertial signals in their local frames to the Joint Coordinate System (JCS) [23] recommended by the International Society of Biomechanics (ISB), consists of a vertical alignment based on gravity as a vertical reference, and a horizontal alignment estimated with the approach described in [24]. Hence, no specific movements (such as abduction-adduction) are needed, but any random activities, like the ones part of the studied scenarios, are feasible for the purpose, owing to the assumption that the knee joint works as a hinge. A quaternion-based fusion algorithm [25] is then adopted to establish the orientation of each sensor. Magnetometers were not considered because of their well-known issues in environments with ferromagnetics materials. Finally, the joint angles are extrapolated from a differential calculation of the two orientations previously obtained.

For each of those parameters, it is possible to calculate min, max, mean, median, standard deviation values and extrapolate the related dimensionless variability (or coefficient of variation - CV, e.g. the ratio between the standard deviation and the mean value).

3. Hardware and Procedure

The system consists of two Tyndall Wireless Inertial Measurement Units (WIMUs) [26] per leg with 3D accelerometer/gyro (@ 250 Hz) and BLE (or SD cards) as shown in Figure 2. The developed technology is currently at the sixth level (Technology Demonstration) of the Technology Readiness Level (TRL) scale, where the engineering-scale system has been tested in a relevant environment and is ready for demonstrations. WIMUs have been attached to the anterior tibia, 10 cm below the tibial tuberosity, and to the lateral thigh, 15 cm above the tibial tuberosity using sticky tape. Algorithms are implemented in Matlab [27], and the scenarios considered (walking, half squat, hamstring curl, and lunges) simulate a free-living environment and exercises performed in a rehabilitation procedure.

In the walking scenario, the subjects stand on a treadmill, which is then being operated at different speeds (3-4-6 km/h) for approximately one minute per test. In the half squat scenario, the subjects stand with the feet shoulder's distance apart and arms crossed on the chest. Keeping the chest lifted, the hips are lowered about 10 inches, planting the weight in the heels. The body is then brought back up to standing by pushing through the heels. In the hamstring curl scenario, the subjects stand and bend the affected knee (or the knee in the dominant leg, as per healthy subjects) raising the heel toward the ceiling as far as possible without pain, relaxing the leg after each repetition. In the lunge scenario, the subjects stand with feet shoulder's width apart, spine long and straight, shoulders back, gaze forward, and step forward with the affected leg into a wide stance (about one leg's distance between feet) while maintaining spine alignment. The hips are lowered until both knees are bent at approximately a 90 deg angle. Finally, the subject pushes back up to starting position by keeping the weight in the heels.

Each subject carried out a significant number of repetitions for each scenario, so as to provide an accurate picture of their conditions. The system has been tested with both healthy and impaired athletes. The healthy subjects are a sample of convenience of 5 athletes, males, age: 29.4 ± 6.58, height: 178.6 ± 6.88 cm, and weight: 73 ± 8.92 kg, without history of knee injuries/surgeries, and with good general health status. The impaired subject is a female athlete, age: 44, height: 161 cm, and weight: 52 kg, with good general health status, with history of knee injuries in the last 2 months before testing (snapped ACL ligament in the left leg), and tested one month before surgery. All subjects met the abovementioned inclusion criteria and signed a specific consent form for participating. Volunteers have been recruited via general email and word of mouth.

A high-speed Basler camera (@ 100 Hz) [28], adopted in conjunction with active markers, has been used as a reference for the validation.

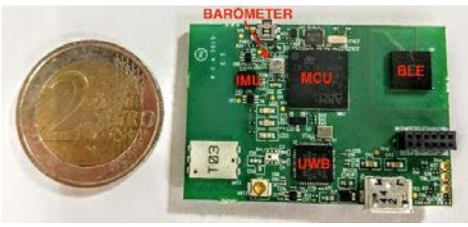


Fig. 2. Tyndall Wireless Inertial Measurement Unit (WIMU) [26].

4. Results and Discussion

The developed system has been firstly tested against the video reference (Basler camera). Knee joint angles have been estimated with both technologies for all the scenarios (an example for the hamstring curl is shown in Figure 3-left), and the related error estimation is summarized in Figure 3-right. Each scenario was divided in two separate tests (both logged for 30 sec, except for the impaired subject whose recordings' duration was 60 sec), and in each of the two tests several repetitions have been carried out by all the subjects. The overall number of exercises recorded was: 54 hamstring curls (21 impaired subject and 33 unimpaired ones), 44 lunges (15 impaired and 29 unimpaired), 59 half squats (22 impaired and 37 unimpaired), 118 strides for both legs when walking at 3 km/h (60 impaired and 58 unimpaired), and similarly 172 strides when walking at 4 km/h (66 impaired and 106 unimpaired), and 165 strides when walking at 6 km/h (78 impaired and 87 unimpaired). No drift was evidenced both in the inertial signals and in the orientation estimation, due to the high performance of the utilised IMU and the robustness of the algorithm presented in [25]. WIMUs show good accuracy for all scenarios, with a maximum mean error of 3.5 deg, significant correlation coefficient values (included between 0.95 and 0.995), and an average Root Mean Square Error (RMSE) equal to 6.8 deg, which are acceptable results for several applications. Therefore, the equivalence in performance of the two technologies has been proved.

Knee joint angles have been also estimated for impaired and unimpaired subjects separately. Results are shown for all scenarios in Figure 4, where the blue area indicates the reference joint angle values extrapolated from the exercises performed by the unimpaired subjects, and the red area represents the same parameters for the impaired one. Those areas have been estimated by taking into account all the individual repetitions carried out by all the athletes for each specific scenario whose numbers are mentioned above. The areas are delimited by the recorded maximum and minimum values, and the dashed lines represent the average characteristics. Blue and red areas show good similarities and the difference between them is within the error estimated when validating the IMU (e.g 6.8 deg), which may make those areas undistinguishable in a blind test. However, some differences can be highlighted by expert clinicians. For instance, it is evident how, in the half squat/hamstring curl/lunge scenarios, the healthy subjects show a joint angle tending towards zero deg (averagely 10 deg) at the end of each repetition, while the impaired athlete constantly presents values higher than 20 deg, thus, indicating difficulties in performing a complete flexion/extension of the joint. A second example is given by the knee angle estimated during the stance phase of the walking scenario. Whilst a typical almost-parabolic curve is usually observed, the impaired subject is not presenting that feature, which indicates a certain stiffness of the joint in correspondence of the gait time interval in which the leg is bearing the body weight.

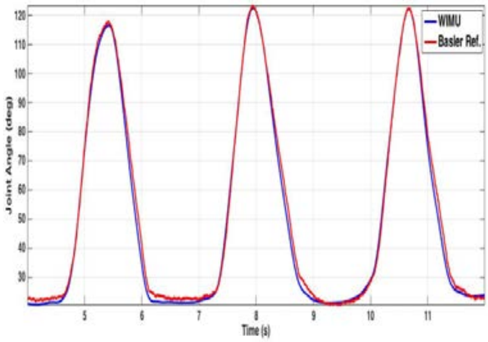
Finally, all the gait spatio-temporal parameters mentioned in Section 2 have been calculated for the walking scenario (at 3, 4, and 6 km/h) for both legs of the impaired and unimpaired subjects by adopting the inertial data. Results are summarized in Figure 5. As per the temporal variables, unimpaired subjects show comparable results between left and right leg for all the speeds, while a larger discrepancy is evident on the results from the impaired subject due to injury's effects on gait. For further demonstration, an independent two-sample t-test has been carried out on the set of strides collected from both left and right leg when walking at 6 km/h (the speed involving more stress on the lower limbs). The p-values calculated for the gait cycle time, swing phase, and stance phase for the injured subject when considering all the strides from left and right leg were, respectively, 0.86, 0.00094, and 0.000012, which in the last two cases are much lower than the threshold chosen for statistical significance (0.05). This indicates that the null hypothesis is rejected in favor of the alternative hypothesis (e.g. the injury is causing an effect on the left leg which differentiate the gait parameters for the two legs). The p-values for the same variables, calculated considering the left/right strides from the uninjured subjects were, instead, 0.96, 0.27, and 0.17 (> 0.05), which, as expected, confirm the null hypothesis of no significant difference between the legs.

Dissimilarities are also clear as per spatial parameters. For instance, the evident difference (averagely 24.4 cm) between left and right stride length in the injured subject is much larger compared to the same divergence measured for the uninjured athletes (4.7 cm). Indeed, the left stride length for the impaired athlete is always shorter compared to her right stride length at every speed. The CV associated to the stride length is an additional parameter which further show this dissimilarity without having the effect of the distant heights of the populations on the measurements. The average CV for the right stride length is 5.2% for uninjured people and 5.4% for the injured athlete, whilst the same variable for the left stride length was 6.5% for unimpaired subjects and 13.7% for the injured one, which is much higher compared to all the other values.

The same characteristics are observed in the estimation of stride speeds and shank clearance, which are not dependent on the population's height, as the subjects are constrained to walk at the selected treadmill speed. The difference between left and right stride speed in the uninjured athletes (0.04 m/s) is much limited compared to the same variable measured for the impaired subject (0.21 m/s). The associated CV for right and left stride speed is 4.8% and 5.5% for the unimpaired population and 5.4% and 10.8% for the injured one. Finally, while the right shank clearance is consistent at every speed for both populations, the impaired subject's left clearance is limited in comparison with unimpaired subjects and the same impaired subject's right clearance, which indicates a not complete recover of the muscle strength and the definition of a movement compensation caused by the injury. The difference between left and right shank clearance in the uninjured athletes (0.7 cm) is reduced when correlated with the same variable measured for the impaired subject (3.1 cm). The associated CV for right and left shank clearance is 16.3% and 17.8% for the unimpaired population and 21.2% and 43.1% for the injured one.

All those results indicate that the gait measurements gathered from the unimpaired athletes as well as the right leg gait parameters from the injured subject are far more repeatable and stable than the left leg gait variables collected from the injured one. Therefore, this feasibility study has proved that the studied wearable inertial sensing technology is able to potentially detect atypical gait movement characteristics accurately and reliably by comparing performance and differences in the two legs.

Additional clinical trials with larger populations are however requested to further statistically validate such results.



	Mean Error ± Std. Dev. (deg)	Pearson's *r*	RMSE (deg)
Hamstring Curl	-0.7101 ± 3.4725	0,9950	3,8939
Lunge	-1.4151 ± 5.9446	0,9785	6,4346
Half Squat	-3.5449 ± 4.7081	0,9917	6,7064
Walking 3km/h	-0.2875 ± 4.8874	0,9820	5,1757
Walking 4km/h	1.5433 ± 6.8283	0,9594	7,4326
Walking 6km/h	1.5793 ± 11.0702	0,9569	11,2932

Fig. 3. (left) WIMU and camera reference comparison; (right) WIMU vs. camera error estimation.

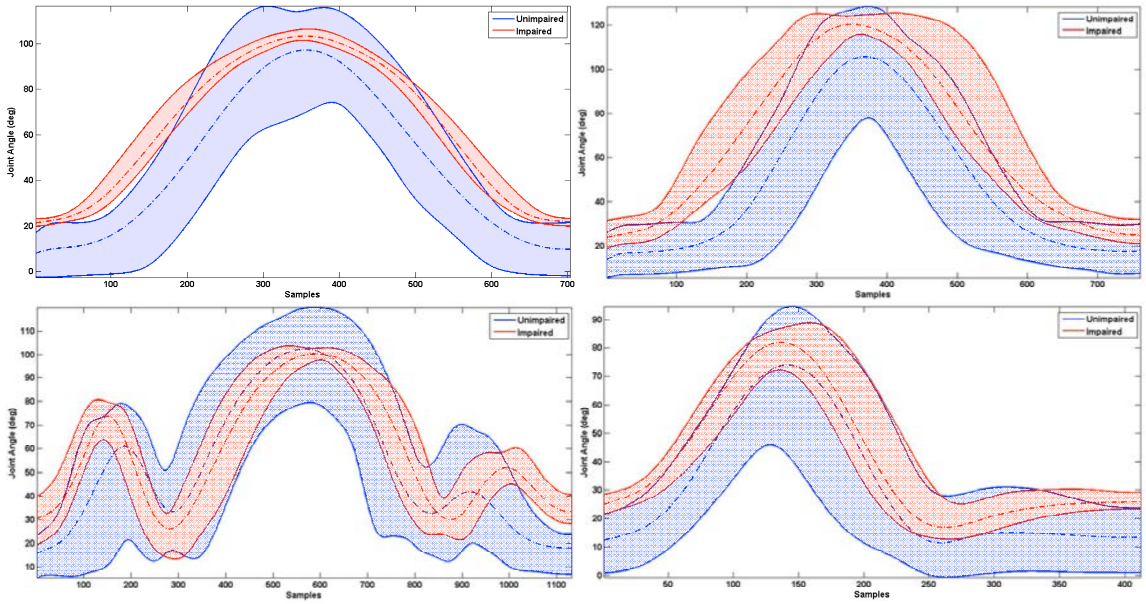


Fig. 4. Knee joint angle estimated for impaired/unimpaired subjects during half squat scenario (upper left), hamstring curl (upper right), lunge (bottom left), and walking at 4 km/h (bottom right).

5. Conclusions and Future Works

This work presents a wearable inertial system for the implementation of a complete portable wireless lower-limbs analysis system. Overall results present good repeatability and the accuracy is comparable with the state-of-the-art. Moreover, detection of atypical movement characteristics was possible by comparing performance and differences in the two legs. Athletes will be monitored throughout their complete rehabilitation in order to gather response to the therapeutic treatment. An enhanced number of athletes, with homogeneous characteristics, will also be tested in future so as to have a more robust base for the study and further validate the drawn conclusions in statistical terms. Additional clinical trials are, thus, currently being planned. However, the present feasibility study already proved that inertial sensors can be used for a quantitative assessment of knee joint mobility, and gait mechanics in ambulatory or free-living environments during the rehabilitation program of injured athletes. The results of this feasibility study support the next phase which will be to assess the external and ecological validity of athletes' on-field movement performance, which will help to inform individualised training protocols or enhance targeted rehabilitation programmes.

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	3 km/h (0.83 m/s)		4 km/h (1.11 m/s)		6 km/h (1.67 m/s)	
	Impaired	Unimpaired	Impaired	Unimpaired	Impaired	Unimpaired
Right Gait Cycle Duration (s)	1.3777 ± 0.0288	1.3273 ± 0.0280	1.1642 ± 0.0332	1.2951 ± 0.0272	0.9153 ± 0.0113	0.9215 ± 0.0314
Left Gait Cycle Duration (s)	1.3713 ± 0.0360	1.3343 ± 0.0487	1.1806 ± 0.0630	1.2899 ± 0.0357	0.9159 ± 0.0177	0.9221 ± 0.0466
Right Stance Phase (s)	0.8788 ± 0.0274	0.8291 ± 0.0157	0.7143 ± 0.0229	0.7910 ± 0.0192	0.5368 ± 0.0090	0.5430 ± 0.0179
Left Stance Phase (s)	0.8581 ± 0.0278	0.8528 ± 0.0324	0.7249 ± 0.0622	0.7886 ± 0.0182	0.5261 ± 0.0172	0.5352 ± 0.0279
Right Swing Phase (s)	0.4973 ± 0.0099	0.4981 ± 0.0179	0.4498 ± 0.0203	0.5041 ± 0.0130	0.3785 ± 0.0067	0.3775 ± 0.0157
Left Swing Phase (s)	0.5132 ± 0.0148	0.4815 ± 0.0368	0.4557 ± 0.0111	0.5012 ± 0.0261	0.3898 ± 0.0135	0.3870 ± 0.0276
Single Support (s)	1.0105 ± 0.0194	0.9796 ± 0.0422	0.9055 ± 0.0249	1.0053 ± 0.0341	0.7683 ± 0.0160	0.7645 ± 0.0380
Double Support (s)	0.3656 ± 0.0255	0.3547 ± 0.0321	0.2586 ± 0.0194	0.2897 ± 0.0239	0.1470 ± 0.0161	0.1560 ± 0.0235
Right Cadence (step/min)	43.6018	45.2060	51.5395	46.3301	65.5542	65.1090
Left Cadence (step/min)	43.7538	44.9671	50.8200	46.5166	65.5097	65.0664
Swing Symmetry	0.9699 ± 0.0310	1.0415 ± 0.0102	0.9873 ± 0.0455	1.0081 ± 0.0524	0.9803 ± 0.0306	0.9804 ± 0.0795
Right Stride Length (m)	1.2557 ± 0.0849	1.3331 ± 0.0440	1.3743 ± 0.0768	1.4650 ± 0.1170	1.4239 ± 0.0561	1.5949 ± 0.0693
Left Stride Length (m)	1.0093 ± 0.1627	1.2722 ± 0.0780	1.1040 ± 0.1175	1.4333 ± 0.0863	1.2087 ± 0.1741	1.5463 ± 0.1155
Right Stride Speed (m/s)	0.9125 ± 0.0576	1.0044 ± 0.0251	1.1810 ± 0.0688	1.1317 ± 0.0935	1.5560 ± 0.0649	1.7330 ± 0.0615
Left Stride Speed (m/s)	0.7361 ± 0.0423	0.9533 ± 0.0467	0.9404 ± 0.1162	1.1116 ± 0.0656	1.3200 ± 0.1908	1.6773 ± 0.0980
Right Shank Clearance (m)	0.0384 ± 0.0082	0.0377 ± 0.0058	0.0486 ± 0.0110	0.0450 ± 0.0082	0.0650 ± 0.0127	0.0556 ± 0.0085
Left Shank Clearance (m)	0.0126 ± 0.0040	0.0498 ± 0.0056	0.0168 ± 0.0068	0.0530 ± 0.0103	0.0308 ± 0.0176	0.0577 ± 0.0132

Fig. 5. Gait spatio-temporal parameters estimated for impaired/unimpaired subjects during walking scenario (3, 4, and 6 km/h)

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