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## O8 - A novel kinematic model of the foot-ankle complex for gait analysis

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### INTRODUCTION

Relying on normative bands for evaluating foot kinematics is questionable due to poor repeatability of the kinematics of the foot-ankle complex even on a healthy population [1]. In this study, a novel multi-segment model for the kinematics of the foot-ankle complex was proposed. Then, the repeatability of the kinematics evaluated with this model was assessed.

### METHODS

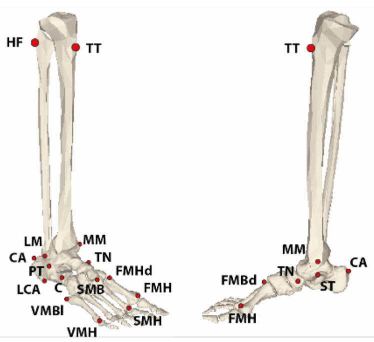
Data used in this study have been previously published [1], allowing the comparison between the obtained results and those of the existing models. Data were collected from 13 healthy adults (27.0±1.9 years, heights: 1.83±0.08 m, foot-length: 28.5±1.0 cm) during two one-month-apart sessions. The model included 4 segments, defined using the anatomical landmarks given in Table 1.a and 1.b. Variations from existing models included: the exclusion of wand markers; the absence of a joint between the forefoot and the midfoot [2]; the use of technical embedded coordinate systems (ECS), defined considering possible deformations of the segment to be tracked during walking and used to register the anatomical ECS. Sagittal joint kinematics of Hindfoot-Tibia, Midfoot-Hindfoot, and Forefoot-Hindfoot were computed according to [3]. Linear Fit Method (LFM) coefficients [5] and Mean Absolute Variability (MAV) [4] were used to quantify within- and between-subject repeatability of the chosen kinematic variables, in terms of similarities, correlations, and absolute differences.

### RESULTS

Table 1.c shows the MAV and LFM coefficients obtained from the repeatability analyses.

**Table 1**. (a, b) segments and anatomical landmarks: short definitions not available in [3] are reported; (c) within- and between-subject MAV and LFM coefficients (MAV and $a_0$ are expressed in degrees).

**(a) Anatomical landmarks**




| **(b) Segments and relevant anatomical landmarks** | |
|---|---|
| **Tibia *(Tib)*** | LM, HF, TT, *MM (static)* |
| **Hindfoot (*HiF*)** | CA, PT, LCA (lateral calcaneus), ST |
| **Midfoot (*MF*)** | TN, C, VMBI (5th metatarsal base, lateral aspect), SMB (2nd metatarsal base) |
| **Forefoot (*FF*)** | FMBd (1st metatarsal base, dorso-medial aspect), FMH (1st metatarsal head), SMH (2nd metatarsal head), *VMH (static, 5th metatarsal head)* |

**(c) Results of the within- and between-subject repeatability analyses**

| **Joints** | **Within-Subject LFM coefficients $a_1$** | **$a_0$** | **$R^2$** | **MAV** | **Between-Subjects LFM coefficients $a_1$** | **$a_0$** | **$R^2$** | **MAV** |
|---|---|---|---|---|---|---|---|---|
| *HiF-Tib* | 1.00±0.11 | 0±1 | 0.91±0.09 | 3±1 | 1.00±0.27 | 0±1 | 0.69±0.18 | 8 |
| *MF-HiF* | 1.00±0.17 | 0±0 | 0.93±0.07 | 2±1 | 1.00±0.35 | 0±1 | 0.82±0.12 | 7 |
| *FF-HiF* | 1.00±0.14 | 0±1 | 0.95±0.05 | 2±1 | 1.00±0.33 | 0±3 | 0.83±0.12 | 12 |

### DISCUSSION

The experimental protocol used for the proposed model was easier than those analysed in [1], both for the reduced number of markers to be tracked and their visibility. Joint kinematics repeatability was enhanced if compared to the results obtained for the existing models using the same data [1]. Despite the obtained improvements, foot kinematics is still the least reliable among the lower limb joints' kinematics, confirming the inadequacy of using normative data to interpret relevant results.

### ACKNOWLEDGMENTS

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## O9 - Development of a novel wearable system for real-time measurement of the inter-foot distance during gait

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### INTRODUCTION

The combination of magneto-inertial measurement unit (MIMU) and distance sensor (DS) represents a smart solution for evaluating the distance between feet during various daily-life activities. In particular, when analyzing gait, the latter technology can be used for estimating the instantaneous or average distance between selected points of the feet (IFD) during mid-swing and mid-stance phases [1-3]. The aim of this preliminary work is twofold: a) to develop and validate a novel wearable system (SWING$^{2DS}$) for the measurement of the IFD during gait; b) to investigate the optimal positioning of the DS on the foot.

### METHODS

The SWING$^{2DS}$ system (52L × 38W × 11.5H mm$^3$), comprises a MIMU (ODR = 100 Hz) and two DSs (ODR = 50 Hz), was attached on a plastic rigid support and positioned on the right foot. DSs were positioned orthogonal to the support and close to the heel ($DS_{HEEL}$) and the first metatarsophalangeal joint ($DS_{MTP}$) (Figure 1). For validation purposes, to avoid measurement uncertainties due to the irregular shape of the shoe, a rectangular target (200 × 100 mm$^2$) was attached on the medial side of the left foot. A cluster of 3 markers was placed on each foot to define a coordinate system. The target and SWING$^{2DS}$ geometries were acquired and expressed in the relevant coordinate systems from the positions of 8 additional markers during a static acquisition. Markers positions were recorded using a 10-camera stereo-photogrammetric system (SP) (Vicon, 100 Hz). Experimental data were acquired on a healthy subject during a six-meter straight walk at slow (*Slow*, 0.6 m/s) and comfortable speed (*Comf*, 0.9 m/s) (3 repetitions). IFD markers-based reference values were calculated as the distance between the DS center and the intersection point between the normal to the DS plane, passing through the DS center, and the target plane placed on the left foot. For each gait cycle, mean values of the distances provided by $DS_{HEEL}$, $DS_{MTP}$ and SP during swing and stance phases of the right foot were computed and the absolute differences between DSs and SP mean distance values derived. The overall mean absolute error (MAE) was computed averaging differences over gait cycles and trials.

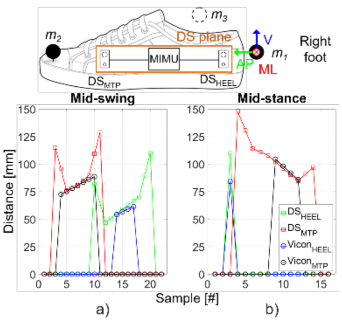


**Figure 1**. On top the right foot experimental setup; on bottom the distance values measured during *Slow* gait

**Table 1**. Average distances and MAEs during the mid-swing and mid-stance phases of instrumented right foot

| *Gait* | Mid-swing | | | | Mid-stance | | | |
|---|---|---|---|---|---|---|---|---|
| | $DS_{HEEL}$ | | $DS_{MTP}$ | | $DS_{HEEL}$ | | $DS_{MTP}$ | |
| | $\bar{d}$ [mm] | MAE [mm] | $\bar{d}$ [mm] | MAE [mm] | $\bar{d}$ [mm] | MAE [mm] | $\bar{d}$ [mm] | MAE [mm] |
| *Slow* | 62.1 | 8.8 | 85.2 | 6.6 | 79.4 | 21.7 | 84.4 | 9.1 |
| *Comf* | 57.3 | 4.1 | 76.2 | 4.1 | 71.5 | 18.8 | 93.7 | 6.2 |

### RESULTS

The average distance ($\bar{d}$) and MAE values during mid-swing and mid-stance phases of the instrumented right foot are reported in Table 1.

### DISCUSSION

Preliminary results showed that the SWING$^{2DS}$ wearable system can be effectively used for accurately measuring the IFD during gait. Interestingly, the accuracy of the IFD estimation is highly affected by the position of the DS on the foot. When the right foot is moving during the mid-swing phase, both $DS_{HEEL}$ and $DS_{MTP}$ provided a similar number of observations (Figure 1a). On the contrary, when right foot is stationary during mid-stance phase, a fewer number of observations are made available using the $DS_{HEEL}$ due to the raised position of the left foot during its mid-swing phase (Figure 1b). The few number of observations affected the accuracy of the $DS_{HEEL}$ ($MAE_{HEEL}$ 18.8-21.7 mm vs $MAE_{MTP}$ 6.2-9.1 mm). Therefore, the recommended DS location is close to the forepart of the foot.

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## O10 - Validating a method for the estimate of gait spatio-temporal parameters with IMUs data on healthy and impaired people from two clinical centers

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### INTRODUCTION

Instrumented gait analysis offers objective clinical outcome assessment [1]. To this purpose, inertial measurement units (IMUs) represent nowadays a very effective solution due to their limited cost, ease of use and improved wearability. The aim of this study was to apply a well-documented IMU-based method to measure gait spatio-temporal parameters in a large number of healthy and gait-impaired subjects, and evaluate its robustness and validity across two clinical centers.

### METHODS

Ninety-two participants (34 healthy elderly and 58 with either Parkinson's disease or mild cognitive impairment) were recruited within the EU-funded V-TIME project by two clinical centers (University of Genova-UNIGE, 50 participants, and the Katholieke Universiteit Leuven-KULEU, 42 participants). Each subject performed two gait trials of one minute each, at comfortable (C) and fast (F) speed. Subjects walked on an instrumented mat (GAITRite) used as gold standard (GS) while wearing two IMUs (Opal, APDM, 128Hz) above the ankles. The method proposed in [2] was used to identify the gait events (GE) and to estimate relevant spatio-temporal parameters. For each left and right gait cycle, the difference between IMUs estimates and GS values was computed (error). Error mean value and its standard deviation (sd), as well as the mean absolute error (MAE), were computed over the entire gait trial.

### RESULTS

8648 gait cycles were analyzed, and no extra or missing GE were found. The errors in determining the GE and the selected spatio-temporal parameters are reported in Table 1. Errors were similar between the two clinical centers. IC MAEs were in general half the size of the FC MAEs. Temporal parameter MAEs were always below 30 ms. In particular, stride and step duration MAEs were well below 15 ms, with a mean error close to 0. Stride length MAEs were below 30 mm and showed a limited underestimation (mean error<0).

**Table 1**. Errors in determining the GE (Initial Contact, IC, and Final Contact, FC) and the gait spatio-temporal parameters.

| | | | IC [ms] | | FC [ms] | | Stride Duration [ms] | | Stance Duration [ms] | | Swing Duration [ms] | | Step Duration [ms] | | Stride Length [mm] | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | | | UNIGE | KULEU | UNIGE | KULEU | UNIGE | KULEU | UNIGE | KULEU | UNIGE | KULEU | UNIGE | KULEU | UNIGE | KULEU |
| mean error (sd) | C | healthy | 3 (9) | 1 (8) | -9 (11) | -7 (13) | 0 (12) | 0 (11) | -12 (15) | -8 (16) | 12 (15) | 8 (16) | 0 (13) | 0 (12) | 0 (21) | -7 (20) |
| | | impaired | 11 (11) | 6 (9) | -10 (14) | -12 (14) | 0 (16) | 0 (13) | -21 (19) | -18 (19) | 21 (19) | 18 (18) | 0 (17) | 0 (14) | -2 (21) | -5 (18) |
| | F | healthy | 6 (9) | 3 (8) | -9 (10) | -8 (10) | -1 (12) | 0 (10) | -16 (15) | -11 (14) | 15 (15) | 11 (14) | 0 (13) | 0 (12) | -6 (20) | -6 (21) |
| | | impaired | 10 (10) | 6 (10) | -8 (14) | -9 (14) | 0 (14) | 0 (13) | -18 (18) | -14 (17) | 18 (19) | 15 (18) | 0 (15) | 0 (15) | -4 (31) | -4 (21) |
| MAE | C | healthy | 10 | 8 | 22 | 24 | 10 | 9 | 26 | 26 | 26 | 26 | 10 | 10 | 21 | 24 |
| | | impaired | 15 | 12 | 20 | 20 | 12 | 10 | 29 | 26 | 29 | 26 | 14 | 11 | 23 | 20 |
| | F | healthy | 11 | 9 | 21 | 21 | 10 | 8 | 25 | 24 | 25 | 24 | 11 | 9 | 18 | 28 |
| | | impaired | 14 | 11 | 19 | 18 | 11 | 10 | 27 | 21 | 27 | 22 | 12 | 12 | 24 | 20 |

### DISCUSSION

All the parameters showed similar or lower errors compared to previous results for both centers [2]. Mean error and MAE values were very consistent between centers, and similar for C and F trials. Stride and step durations MAE values resulted of the same order of magnitude as the IMU system nominal accuracy. Stance and swing durations were respectively underestimated and overestimated due to a late IC identification and an early FC identification. When averaged, stride length estimation was extremely accurate, however showing some residual inaccuracies in individual cycle estimation (sd values not negligible). Overall, the results of this work represent a robust and reliable foundation for the clinical use of the proposed IMU based method for gait parameters estimation.

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**O11 - A wavelet-based energetic approach for the detection of contact events during movement.**

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## INTRODUCTION

The use of wearable Inertial Measurement Units (IMUs), able to quantify accurately human movement, has steadily risen in recent years for monitoring and evaluating daily life activities [1, 2]. The detection of foot contacts (FCs) during movement is required when aiming at analysing motor performance, segmenting task execution and/or assessing temporal parameters of the task. Many different algorithms have been proposed for FC identification during the most common tasks, like walking and running [2-5]. On the other hand, these algorithms are mostly task specific and cannot be generalized for other motor tasks. In literature, wavelet-based methods for the identification of transient events in biomedical signals have been proposed and applied in different contexts [6]. FC events can be assumed as transient events in a motor task, thus, the aim of this work was to assess the performance of a wavelet-based energetic approach (WBEA) [6] for the detection of FCs in different exercises. 3D kinematics was used as gold standard (GS) reference.

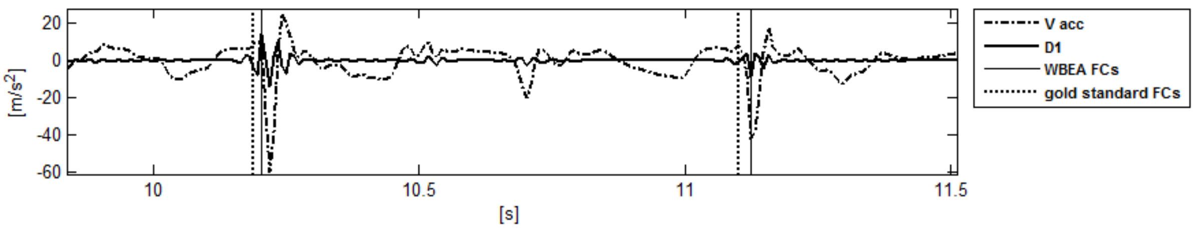
## METHODS

Three healthy subjects were recruited for the study (2F, 1M, 27±4 years, 160±10 cm, 57±10 kg). They were asked to perform the following 8 exercises along the gait analysis laboratory at self-selected speed: walk, run, leap, hop, horizontal jump, tandem walk, gallop and slide. Reflective markers were positioned on right and left calcanei. Two tri-axial inertial sensors (Opal, Apdm, US) were positioned on the lower legs. Marker' 3D kinematic (SmartD, Bts, Italy) and leg acceleration were collected at 200Hz and 128Hz, respectively. Data were synchronized *a-posteriori*. FC events were identified on the vertical (V) position of the markers (GS), and on the V acceleration signals of the legs. Leg V acceleration was decomposed into 20 resolution levels, D1-D20 (the frequency components move from the high towards the low frequencies as scale increases from 1 to 20). FC events were identified as the maximum peaks on D1 (showing the highest frequencies). Absolute differences between FCs detected by GS and WBEA were computed.

## RESULTS

A minimum of 18 FC per subject were identified for a total of 64 FCs. Overall, median absolute difference between GS and WBEA was 0.03s ($25^{th}$ and $75^{th}$ percentiles, 0.02 and 0.05s). Mean difference between the WBEA and GS was positive and close to the median value (0.03s), showing a possible bias in the identification. The same performance was observed for all tasks except for gallop, which showed the highest mean absolute difference of 0.07s.

**Figure 1**. Example of FC detection in the leap task.




## DISCUSSION

WBEA, applied on leg V acceleration, was found to be an accurate method for the detection of FC during several motor tasks. When compared to other approaches, it has the advantage of being generalizable and applicable in different contexts. Preliminary results showed a small bias between the WBEA and GS, that could be improved modifying the definition of how to select FC on D1 (maximum peaks). Future studies should investigate the possibility of using this approach for the detection of foot contacts in pathological gait, which is usually challenging [3].

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**O12 - Frequency evaluation of gait trunk acceleration signal: a longitudinal study.**

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**INTRODUCTION**

Acquisition and elaboration of trunk acceleration signal during gait have assumed a key role in motor assessment [1]. This has led to develop different indexes and metrics to evaluate gait performance that, directly or indirectly, imply the analysis of the harmonic content of the signal. In addition, smartphones with embedded accelerometers have been proposed as a monitoring tool, even when not supporting high sampling frequencies [2]. Thus, the knowledge of the spectrum characteristics of the trunk acceleration signal during gait is crucial to identify hardware and software requirements and to correctly use the indexes and their parameters. The aim of the present study was to analyze the harmonic content of the trunk acceleration signal characterizing the gait of nine age groups.

**METHODS**

Nine age groups, from 7 to 85 year-old, of 10 healthy subjects each were included in the study.

**Table 1**. Details of age groups participating in the study. [a]Data already presented in Bisi et al. [2].

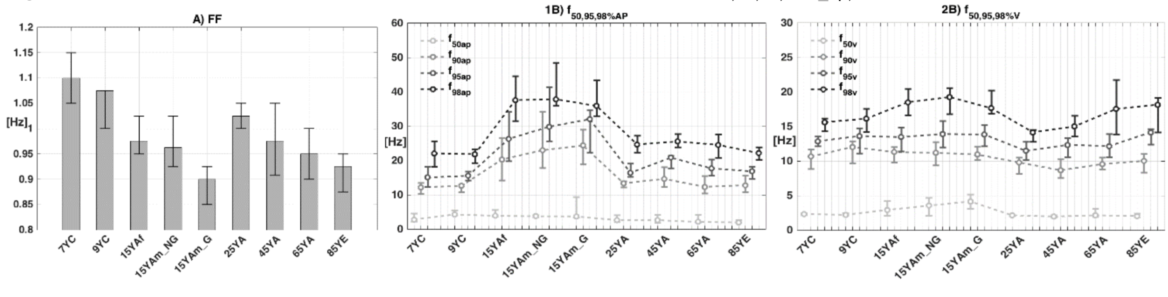
| Abbreviation | Group description | Age [year] | Weight [Kg] | Height [cm] |
|---|---|---|---|---|
| 7YC | 10 7-years old children | 7 (7, 7) | 29 (22, 37) | 129 (119, 134) |
| 9YC | 10 9-years old children | 9 (9, 9) | 34 (22,45) | 140 (138,145) |
| 15YAf | 10 15-years old adolescents Female Not Grown | 15 (15,15) | 54 (49, 74) | 162 (147, 172) |
| 15YAm_NG [a] | 10 15-years old adolescents Male Not Grown [a] | 15 (15,15) | 64 (49, 74) | 172 (169,176) |
| 15YAm_G [a] | 10 15-years old adolescents Male Grown [a] | 15 (15,15) | 59 (46, 65) $\Delta$w=2 (-1, 4) | 172 (160,175) $\Delta$h=3.6 (3, 4) |
| 25YA | 10 25-years old adults | 25 (22, 26) | 70 (48, 86) | 168 (154, 187) |
| 45YA | 10 45-years old adults | 45 (41, 48) | 74 (45, 100) | 174 (155, 193) |
| 65YA | 10 65-years old adults | 65 (62, 69) | 85 (68, 120) | 176 (164,186) |
| 85YE | 10 85-years old elderlies | 85 (84, 91) | 74 (57, 90) | 177.5 (160, 175) |

Trunk acceleration signal (Opal, APDM, USA, $f_s$=128 Hz) of 20 second of walking, at self-selected speed, were analyzed. Fundamental frequency (*FF*) and frequencies corresponding to the 50, 90, 95 and 98% of the signal power along each axis ($f_{50,90,95,98\%\_ap,ml,v}$) were calculated and normalized to FF, as FF is related to gait cadence, thus to subject anthropometry [3].

**RESULTS**

FF values ranged form 1.1Hz (7YC) to 0.9Hz (all 15YA groups). $f_{98\%\_ap}$ showed a higher value (40Hz) for adolescent groups and lower one (20Hz) for 9YC. In V direction instead, $f_{98\%\_v}$ ranged from 20Hz (all 15YA, 65YA and 85YE) to 15 Hz (25YA). No trends were observed for ML direction with age. The same trends were found for all the normalized features.

**Figure 1**: Median, 25th and 75th percentile of FF (A panel) and of $f_{50,90,95,98\%\_ap,v}$,1B and 2B panel respectively.




**DISCUSSION**

$f_{50,90,95,98\%}$ showed similar values and trends, for normalized and non- conditions, suggesting that the observed differences are peculiar to the population. The harmonic content (98%) of the acceleration signal for all analyzed population, with exception of the adolescents (45Hz), is below 30Hz with the highest frequency content in AP direction Future studies will investigate how the acquisition parameters affect metrics used to evaluate gait performance.

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**O13 - Time-frequency analysis of surface EMG signals for maximum energy localization during gait**

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**INTRODUCTION**
The purpose of this work is to assess the maximum energy localization in time-frequency domain of surface EMG signal of the main lower-limb muscles usually involved in able-bodied walking, to overcome the limitations in using classic amplitude parameters. This aim was pursued by means of Continuous Wavelet Transform (CWT), a time-scale analysis method for multiresolution decomposition of continuous-time signals. WT coefficients allowed to reconstruct the scalogram function, providing an estimate of the local time-frequency energy density of a signal [1].

**METHODS**
Five healthy young adults were recruited for the study. Surface electromyographic signals were acquired (sampling rate: 2 kHz) and processed by the multichannel recording system, Step32 (Medical Technology, Italy). Each subject was instrumented, bilaterally, with foot-switches to identify the gait cycle and sEMG probes. Differential probes were applied over Tibialis anterior, Gastrocnemius lateralis, Rectus femoris, and Biceps femoris (TA, GL, RF, BF), following Winter's guidelines [2]. sEMG signals were processed by CWT. In this work, mother wavelet Daubechies of order 4 with 6 levels of decomposition (db4) has been chosen to implement the wavelet transform. CWT has been applied for removing noise from sEMG and to identify the maximum energy localization in time-frequency domain (CWT scalogram function). The localization of the regions with maximum energy density has been identified as the interval in time-frequency where the scalogram is exceeding the 72% of the peak value of energy density in both time and frequency domain.

**RESULTS**
As reported in Fig. 1, maximum of energy density in time (tMED) occurred for TA in early stance (0-6% of gait cycle, GC) and in swing phase (92-98% of GC) and the maximum of energy density in frequency (fMED) was detected in frequency band between 60-220 Hz. For GL, tMED occurred from 29 to 50 % of GC and fMED was detected in frequency band between 65 and 160 Hz. For RF, tMED is localized during early stance (0-5% of GC) and during swing phase (86-95% of GC) and fMED was detected in frequency band between 70 and 220 Hz. For BF, tMED occurred during swing phase (80-93% of GC) and fMED was detected in frequency band between 70 and 180 Hz.

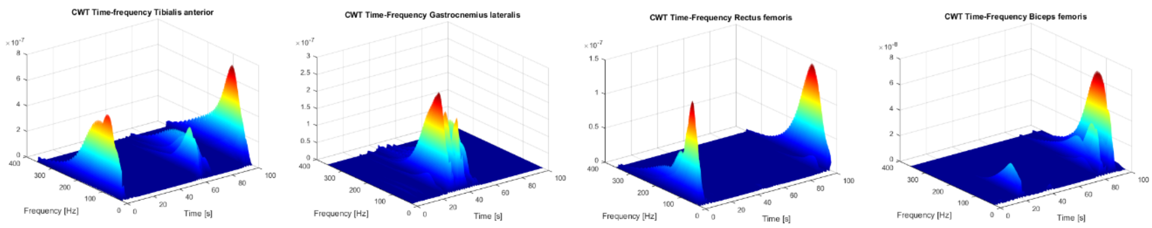

**Figure 1.** The 3D color representation of the scalogram for sEMG signals from TA, GL, RF, and BF.

**DISCUSSION**
Localization in time of maximum energy density (Fig. 1), performed in the present study, could be interpreted as the time-interval where the sEMG signal reached its peak value of energy, i.e. the region of gait cycle where the muscle is mainly recruited. The present localization of maximum muscle activity matched with the region of maximum muscle recruitment reported during walking in able-bodied adults [3]. The localization in frequency of maximum energy density, was interpreted as the frequency-band where the EMG signal showed the maximum frequency content. This region of frequency varied from muscle to muscle, but a common frequency band for all muscles could be found between 70 and 160 Hz [4]. Results of the study could be suitable for both supporting the use of WT for sEMG analysis and providing indications on muscle recruitment during walking.

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**O14 - Is it possible to apply an existing gait recognition method to pathological subjects without any adaptation?**

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**INTRODUCTION**

Human physical activity recognition plays an important role in the assessment of activity level in everyday life [1]. Numerous physical activity tracking solutions are currently available for the research market, however, few studies targeted the applicability of activity recognition on pathological subjects [2]. Moreover, there is no evidence that existing solutions can successfully be extended to people with movement alterations due to pathological conditions. In this work we test the capability of an existing method for gait detection (validated on healthy adults, [3]) in automatically recognizing gait of children affected by Early Onset Ataxia (EOA), Developmental Coordination Disorder (DCD) and age-matched controls, with the aim of laying the foundations for a fully automatic system for gait detection and automatic recognition of altered gait, as a tool for everyday life gait monitoring.

**METHODS**

Data were acquired on 37 subjects, involving 9 EOA (age 13.3 ± 4), 7 DCD (age 9.42 ± 2.3) and 21 healthy controls (age 12.7 ± 4.4). Eight inertial measurement units were placed on the participants' sternum, waist (L3) and bilateral thighs, shanks and wrists. Acceleration data sampled at 256 Hz were used for this study. The subjects followed a protocol that involved standing still, walking and tandem gait, in a straight line, along a corridor. Data were manually annotated by means of video recordings to discriminate walking trials from the rest of data recording.

Walking recognition methods and trained classification models were retained without any modification from a previous study [3]. In particular, the newly available dataset involving altered gait was processed using the classification rules trained on the dataset of young adults of the previous work. It has to be noticed that the previous study was based on a different hardware and protocol. In summary, the walking recognition method was based on a support vector machine classifier, processing acceleration recordings independently from the placement site of the sensor during a complex set of everyday life activities.

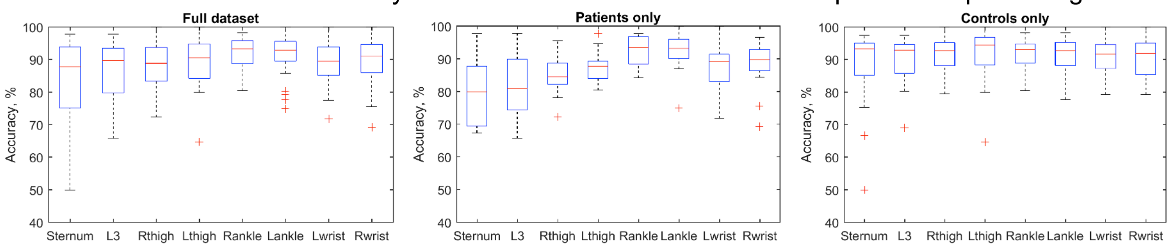
**RESULTS**

Walking detection accuracies for each sensor and across the two groups (EOA and DCD together, vs controls) are summarized in Figure 1.

**DISCUSSION**

Results show that the method previously used for walking detection is effectively applicable to the data obtained from the groups analyzed in this study. In fact, the accuracy in recognizing gait episodes across the two groups is similar and, on average, exceeds 80% in all cases. However, results are biased by the dataset characterized by a limited set of activities of daily living that could extend the complexity of the classification problem. Nevertheless, the high accuracy obtained by the existing method - with no retraining - suggests that it could be applicable to the pathological group. As a consequence, it looks as a promising direction to be further investigated by means of dedicated tests involving a more complete set of daily activities.

**Figure 1**. Box-plots summarizing classification accuracy results. Accuracy is defined as the number of correct classifications divided by the amount of available data and expressed in percentages.

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