

Towards a wearable system for predicting freezing of gait in people affected by Parkinson's disease

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Abstract—Some wearable solutions exploiting on-body acceleration sensors have been proposed to recognize Freezing of Gait (FoG) in people affected by Parkinson Disease (PD). Once a FoG event is detected, these systems generate a sequence of rhythmic stimuli to allow the patient restarting the march. While these solutions are effective in detecting FoG events, they are unable to predict FoG to prevent its occurrence. This paper fills in the gap by presenting a machine learning-based approach that classifies accelerometer data from PD patients, recognizing a pre-FOG phase to further anticipate FoG occurrence in advance.

Gait was monitored by three tri-axial accelerometer sensors worn on the back, hip and ankle. Gait features were then extracted from the accelerometer's raw data through data windowing and non-linear dimensionality reduction. A k -nearest neighbor algorithm (k -NN) was used to classify gait in three classes of events: pre-FoG, no-FoG and FoG. The accuracy of the proposed solution was compared to state-of-the-art approaches.

Our study showed that: (i) we achieved performances overcoming the state-of-the-art approaches in terms of FoG detection, (ii) we were able, for the very first time in the literature, to predict FoG by identifying the pre-FoG events with an average sensitivity and specificity of, respectively, 94.1% and 97.1%, and (iii) our algorithm can be executed on resource-constrained devices. Future applications include the implementation on a mobile device, and the administration of rhythmic stimuli by a wearable device to help the patient overcome the FoG.

Index Terms—freezing of gait, wearable device, accelerometer, explainable machine learning, classification.

I. INTRODUCTION

Parkinson's disease (PD) is the second most common neurodegenerative disorder. Falls are the most disabling complications of advanced PD leading to increased risk of hospitalization, immobilization, comorbidity and disability, [1]. Among PD patients, 35-90% fall once a year, and 18-65% fall repeatedly (range:4.7-67.7 falls/person/year) [2]. At least 20% of falls can be attributed to a single PD symptom, known as freezing of gait (FoG), which affects around 50% of patients [3]. FoG is an episodic gait disturbance lasting up to 30'' with different phenomenology that ranges from complete sudden akinesia to milder leg trembling or short shuffling steps events, usually described by patients as feeling the feet stuck to the floor [4]. FoG is a poorly recognized symptom, because of its episodic nature, the bias in the interpretation and report of this symptom by patients, and the need of an observer [5]. Different treatment strategies have been proposed for FoG, but the results are controversial, and evidence of efficacy still limited [6]. A minority of patients with FoG respond to

levodopa, monoamine oxidase inhibitors, deep brain stimulation and levodopa-carbidopa intestinal gel, but often FoG is caused/worsened by treatment or drug-resistant [7].

The high incidence of FoG and the difficulties in its assessment and treatment by standard clinical methods have led to investigate wearable devices and machine learning algorithms [8]. Most of them deal with recognizing FoG events by building on top of offline datasets gathered from the patients through wearable devices equipped with different sensors, e.g., accelerometers [9,10], gyroscopes [11], EEG sensors [12,13], EMG [13,14], and force and bending sensors [15]. These solutions are primarily devoted to study FoG features. In some cases, the devices provide patients with rhythmic visual, sensory or auditory stimuli upon FoG detection [8,16] to reduce its duration. This approach builds on the observation that PD patients more easily exit FoG when they are externally stimulated with a Rhythmic Auditory Stimulation (RAS), which focuses the attention during movement [17].

Different statistical methods have been investigated to increase accuracy of FoG recognition. They include the use of statistical features over raw sensor signals [18], and the freezing index, i.e., the ratio between FoG and no-FoG frequency bands [9]. Approaches based on Inertial Measurement Unit (IMU) incorporating accelerometer, magnetometer and gyroscope sensors [10,11], and using different recognition algorithms (i.e. Random Forest, C4.5, Naive Bayes, k -NN, Logistic Regression and Support Vector Machine (SVM) [18,19,20] or Deep Learning approaches [11, 21]. High levels of accuracy, and short lag times in FoG and pre-FoG detection are key features in this field. A recent comprehensive review on performances of wearable sensors to detect FoG reported 73–100% sensitivity and 67–100% specificity [22].

While several studies focused on FoG detection, only a small number of them assessed pre-FoG recognition. A methodology to detect FoG, pre-FoG and no-FoG was proposed with 94% accuracy [23]. Furthermore, t-Distributed Stochastic Neighbour Embedding (t-SNE) was proposed as feature projection algorithm and tested on different machine learning models obtaining balanced accuracies of 91%, 90% and 82% for FoG, pre-FoG and no-FoG detection,

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respectively [24]. Other studies based on datasets gathered through inertial sensors on the feet and back achieved average precision of 83% and recall of 67% [10], while a deep learning approach based on the data collected by a single IMU device located on the patient’s waist achieved 90% of geometric mean between precision and recall [11]. Other approaches to detect pre-FoG [25, 26] were based on the recognition of EEG pattern [27], the analysis of data gathered by wearable sensors [10], electrocardiography and skin-conductance [12,18]. Two novel algorithms for FoG prediction, based on the DAPHNET dataset and tested on neural network model [23] and a pool of machine learning models [24] were also proposed.

In summary, only few techniques have been proposed for predicting pre-FoG patterns in gathered data, and more research is necessary on this topic. In addition, almost all the methods mentioned above run on a server instance, because of the complexity of the pattern recognition and feature extraction algorithms, which require high computational resources to achieve high sensitivity and specificity, hampering their use on small wearable devices. Because of these limitations, we present a new approach, based on wearable devices and machine learning algorithms to detect FoG, pre-FoG and no-FoG in PD patients and that can be executed on resource-constrained devices, thus being suited for a wearable system. The future application will be to develop an on-demand cueing device to help PD patients to overcome FoG and prevent falls.

The rest of the paper is organized as follows. Section II introduces preliminary concepts. Section III details the study protocol and presents the technical details of the proposed methodology. Section IV describes the experimental results, including the characteristics of study population. Results are then further discussed in Section V. Finally, Section VI draws the conclusions with the proposal of future work.

II. PRELIMINARIES ON MACHINE LEARNING

A. Feature extraction

In most classification pipelines, raw data are typically pre-processed to project them into different feature spaces where classification is made easier (typically, linear). This pre-processing stage is known as feature extraction [28]. The most used algorithms for feature extraction are summarized in the next paragraphs.

1) *Principal Component Analysis (PCA)* is the most common technique for reducing the dimensionality of a large set of possibly correlated features [28]. It consists of a rotation of the original vector feature space, obtaining a new set of bases aligned with the most data variation. The reduction consists in projecting the data on a subset of these bases, retaining less features than the original ones.

2) *kernel PCA (kPCA)* uses the theory of the positive definite kernel and reproducing kernel Hilbert space [28] for doing (implicitly) PCA on data projected in an infinite-dimensional space. The result in the original feature space is a non-linear projection way more effective than the original PCA, with the additional cost of choosing a proper kernel and its parameterization.

3) *Linear Discriminant Analysis (LDA)* is a generalization of the Fisher’s linear discriminant, a method to find a linear

combination of features that separates samples of different classes. LDA is more effective than PCA when it comes to multi-class problem, but it requires to know in advance the label of each training data [29].

4) *kernel LDA (kLDA)* is a kernel-based version of LDA. Using the kernel substitution, LDA is performed in a new feature space, which allows non-linear mappings to be learned [30].

B. Classification

Standard classification amounts to decide which class among a pool of classes does explain a new observation the best. It is the most common operation in machine learning and depends strongly on the feature extraction. It is usually based on a training phase, where a classification algorithm builds a classifier by analyzing a set of D -dimensional data, known as training set. In *supervised* classification, each training sample is equipped with a class-label attribute, stating which class it belongs to. The trained classifier is then used to predict the class label of an unseen *test* D -dimensional input sample. The most common classification quality metrics are [31]:

$$\begin{aligned} \text{Precision} &= \frac{\sum \text{true positive}}{\sum \text{true positive} + \sum \text{false positive}} \\ \text{Recall} &= \frac{\sum \text{true positive}}{\sum \text{true positive} + \sum \text{false negative}} \\ \text{Specificity} &= \frac{\sum \text{true negative}}{\sum \text{false positive} + \sum \text{true negative}} \\ \text{F1 - Score} &= 2 \times \frac{\text{precision} \times \text{recall}}{\text{precision} + \text{recall}} \end{aligned}$$

where *precision* and *recall* (also known as *sensitivity*) measure how much the classifier is capable of avoiding false positives, and is capable to correctly classify all of the samples of a class, respectively, while *specificity* and *F1-Score* measure the classifier ability to correctly classify true negatives and the harmonic average of precision and recall, respectively.

III. STUDY PROTOCOL

The proposed approach addresses the application scenario showed in Figure 1. The patient’s gait is monitored by three tri-axial accelerometer sensors worn on his/her back, hip and ankle. Features are then extracted from the accelerometer’s raw data through data windowing and non-linear dimensionality reduction. A k -nearest neighbor algorithm (k -NN) [28] is finally used to classify patient’s gait in three classes of events: pre-FoG, no-FoG and FoG. To address this scenario, we developed a workflow composed of two main phases: 1) *offline training data pre-processing* and 2) *on*

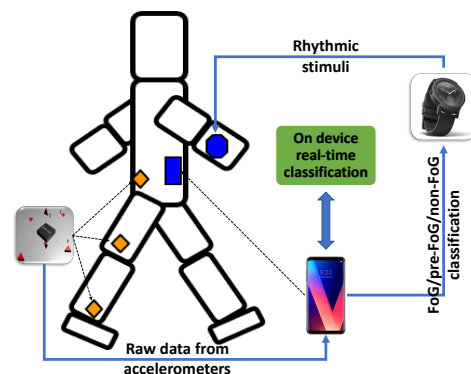


Figure 1. Application scenario

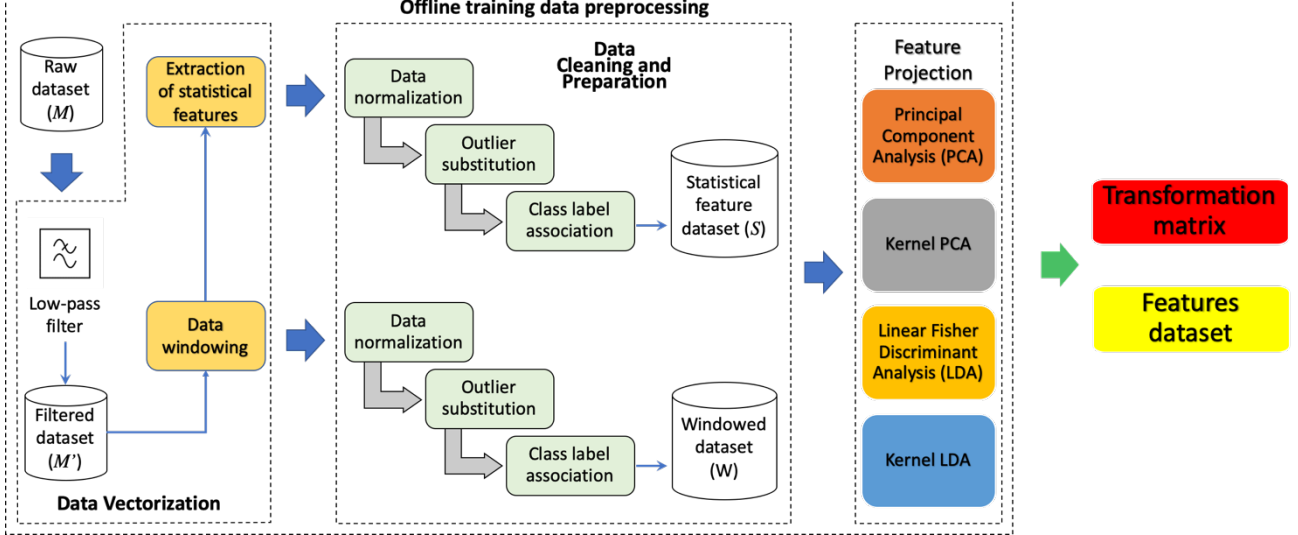


Figure 2. Detailed workflow of the offline pre-processing phase.

device real-time classification, described in detail below.

A. Offline training data pre-processing

The input data is constituted by raw values gathered during patient’s gait from three accelerometers. In particular, the atomic entity of the input is the record, i.e., a vector composed of 9 elements (3 per each accelerometer) measuring the acceleration of the patient in a specific instant with respect to the three axes of a Euclidean space. The number n of records in a given interval depends on the accelerometer sampling rate. Thus, the input dataset is as a matrix M with n rows and 9 columns. The final goal is to set up a classification method exploiting the k -NN algorithm on time windows, capable of forecasting when a FoG event is going to happen, i.e., detecting what we call a pre-FoG event.

The raw data included in the training matrix M are pre-processed to generate two alternative datasets: a windowed dataset W , and a statistical feature dataset S . The pre-processing phase performs a *data vectorization* step, followed by *data cleaning and preparation*, and finalized by a *feature projection* procedure. The detailed workflow is shown in Fig. 2 and detailed hereafter.

1) *Data vectorization*: A low-pass filter is firstly applied to the elements of M to obtain a new matrix M' , where the sensor noise is reduced. Then, two alternative steps are performed starting from M' : *data windowing* and *extraction of statistical features*.

In *data windowing*, the rows of M' are grouped according to time windows of size w seconds with an overlap of t seconds. For example, in case the sampling frequency is 1Hz ¹ with $w = 3$ and $t = 1$, the data included in rows 1-3 of M' are associated to the first-time window and they become, preserving their order, the elements of the first row of matrix W . Then, the data included in rows from $w-t = 2$ to $w-t+w = 5$ of M' become the elements of the second row of W , and so on. The size of W is then $m \times 9 * w$, where m is the total number of time windows.

During the *extraction of statistical features*, a new matrix S

is generated from W by extracting 129 statistical features per each row of W . Thus, the size of S is $m \times 129$. The extracted features are similar to those used in a previous study [12], e.g., variance, mode, standard deviation, maximum and minimum values of raw data.

2) *Data cleaning and preparation*: Subsequently, the normalization of the values in W and S takes place. A unity-based normalization [32] is carried out over the elements of both W and S . The goal is to re-scale the values of W and M into the range $[0, 1]$ by using equation 1, where X is a generic element of the matrix, while X_{min} and X_{max} are the minimum and the maximum value in the matrix:

$$X' = \frac{X - X_{min}}{X_{max} - X_{min}} \quad (1)$$

At this time, the next transformation step, for both matrices, is the outlier substitution. An outlier is a value that is more than 3 median absolute deviation away from the median [33]. Each outlier in W and S is substituted with the corresponding nearest not-outlier value.

At the final step of the pre-processing phase, each row of both matrices W and S is associated to a class label that represents the movement pattern observed on the patient during the corresponding time window. Such labels can be no-FoG, FoG and pre-FoG. Consistently with the existing literature [10,12], we assume the gait performance of PD patients deteriorates in the phase immediately preceding a FoG event. The association is assigned such that FoG is used when the time window includes at least one sample where the patient actually experiences a FoG episode; pre-FoG is assigned to the time windows of type no-FoG immediately preceding a FoG time window by at maximum s seconds, where reasonable values for s are between 2 and 4 seconds; finally, no-FoG is used for all the remaining time windows.

3) *Feature projection*: In the final step, we use four different feature projection algorithms to explore which of them performs better in our scenario. The experimental analysis is

¹ 1Hz is a too low frequency for predicting FoG. It has been assumed for simplifying the exemplification. The dataset used in our experiments has been created at 65Hz.

performed in four different ways according to the applied training features and the number of target classes considered for the classification. The four alternatives are: (i) training features extracted from W , and 2 classes (FoG, no-FoG); (ii) training features extracted from W , and 3 classes (FoG, no-FoG, pre-FoG); (iii) training features extracted from S , and 2 classes (FoG, no-FoG); (iv) training features extracted from S , and 3 classes (FoG, no-FoG, pre-FoG).

For each of the previous alternatives, PCA, kPCA, LDA and kLDA are applied for extracting the final projected features to be provided to the k -NN classifier. In Section IV, we will show that alternatives 1 and 2 associated with the kLDA algorithm provide the better results. Finally, as shown in Fig. 3, the training data pre-processing phase returns the new training features dataset and the transformation matrix that are the input for the on device real-time classification phase.

B. On device real-time classification

Our system is intended to be used for real-time classification of FoG, no-FoG and pre-FoG directly on the patient. Thus, after the off-line training data preprocessing, the workflow proceeds as shown in Fig. 3. The data gathered by the accelerometers on the patient body are transmitted to the mobile phone through the Bluetooth Low Energy (BLE) protocol. Once the data are received, they are elaborated through the same data vectorization process presented in Section III.A.1. After vectorization, the data are multiplied with the transformation matrix obtained by the offline data preprocessing phase, projecting into a reduced features space (p_1, \dots, p_f), where, $f (\geq 1)$, defines the number of features generated after the projection. In the case of the LDA and kLDA algorithms f will be equal to the number of treated classes (no-FoG, FoG, and pre-FoG) minus one. Finally, such new features are passed to the k -NN classifier that, based on the training feature dataset, recognizes the movement as FoG, no-FoG or pre-FoG. It is worth noting that the use of the k -NN ensures fast performances and no need of an explicit training procedure. In fact, we have solely training features that are

used as comparison w.r.t. the testing data.

IV. RESULTS

A. Patient population

To obtain expressive yet comparable results w.r.t. the related literature, we applied the proposed approach to a widely used gait dataset, i.e., the DAPHNET benchmark suite, collected during the EU FP6 project Daphnet in 2010 [9]. In the original protocol of study, video and gait data of patients affected by PD, with history of FoG and able to walk unassisted during OFF phases, were recorded. Exclusion criteria were severe vision or hearing loss, dementia, or signs of other neurological/orthopedic diseases. Ten patients were enrolled and underwent the entire study (7 males, mean age 66.5 ± 4.8 , mean disease duration 13.7 ± 9.7 , mean Hoehn & Yahr score in ON 2.6 ± 0.65): 8 exhibited several FoG events (on average 23.7 FoG events per patient), while 2 of them did not have any FoG. Each patient wore three accelerometer sensors, respectively, on the back, the hip and the ankle. Each sensor collected acceleration over x , y and z axes. Data were recorded with a sampling frequency of 65Hz. Each sample was accordingly labeled as no-FoG or FoG. Additional methodological information concerning the study which generated the adopted dataset can be found in the original paper [9].

The first part of the study s dedicated to application of the proposed methodology to the DAPHNET dataset, the second to on-line testing of the classification performance on different mobile devices, to evaluate its suitability on devices with low computational resources.

B. Dataset pre-processing

Following the procedure described in Section III.A, starting from the DAPHNET data, we generated two different datasets, the windowed dataset W and the statistical feature dataset S . Both W and S depend on the time window size w and the overlapping parameter t to collect short overlapping sequences of acceleration values. Thus, we performed grid search on w and t ; specifically, we varied w in the range $[1:6]$ with step 1, and for each t , t assumed values in the range $[0:w/2]$ with step 0.5, ending up with 27 (w, t) combinations. For every combination (w_i, t_i), $1 \leq i \leq 27$, a couple (W_i, S_i) of different datasets was created. Then, we performed the feature-wise data normalization and the outlier removal. Finally, for each (W_i, S_i), we re-label as pre-FoG all windows immediately preceding a FoG window, obtaining a novel class, namely the pre-FoG. The introduction of pre-FoG windows shrank the distribution of no-FoG windows, on average, of a factor 1/8.

C. Results for the windowed dataset

Experimental results are reported in Table I, Table II and Table III. The classification of our approach has been cross-validated by extracting from DAPHNET training and testing partitions using k -fold ($k = 3, 5$) and leave-one-out schemes on individual patient and also cross-patient. It is worth noting that training and testing partitions were curated so that no partial overlap between windows was present. No significant difference was observed with these cross-validation schemes. Thus, for lack of space, in the following, we report and comment only the results achieved by using k -fold with $k = 3$.

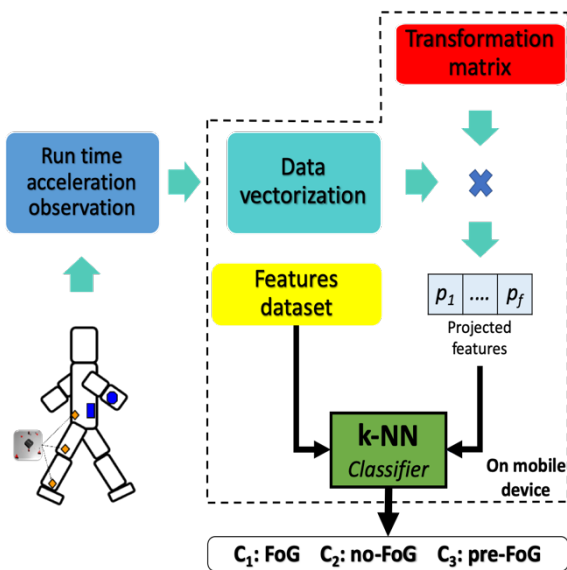


Figure 3. On device real-time classification.

Patient ID	#no-FoG	#FoG	PCA	kPCA	LDA	kLDA	#no-FoG	#pre-FoG	#FoG	PCA	kPCA	LDA	kLDA
P_01	465	18	0.72	0.70	0.64	0.99	460	5	18	0.44	0.43	0.42	0.86
P_02	124	9	0.64	0.63	0.61	0.99	120	4	9	0.31	0.32	0.31	0.96
P_03	405	68	0.75	0.73	0.73	0.99	357	48	68	0.39	0.42	0.40	0.84
P_05	263	120	0.72	0.70	0.70	0.99	243	20	120	0.17	0.18	0.18	0.96
P_06	526	33	0.68	0.66	0.65	0.98	514	12	33	0.44	0.44	0.46	0.98
P_07	362	12	0.57	0.56	0.55	0.99	356	6	24	0.38	0.36	0.39	0.99
P_08	203	53	0.78	0.77	0.73	0.99	194	9	53	0.48	0.41	0.42	0.94
P_09	471	128	0.85	0.84	0.82	0.99	463	8	128	0.58	0.61	0.60	0.98
	2-Class						3-Class						

Table I. Average F1-Score, at varying of w and t , respectively, in $[1, 6]$ with step 1 and $[0, w/2]$ with step 0.5.

Name	2-Class			3-Class		
	Sensitivity	Specificity	F1	Sensitivity	Specificity	F1
PCA	0.72	0.71	0.71	0.40	0.49	0.44
kPCA	0.71	0.69	0.70	0.40	0.48	0.44
LDA	0.68	0.67	0.68	0.40	0.48	0.44
kLDA	0.99	0.99	0.99	0.94	0.97	0.94

Table II. Average sensitivity, specificity and F1-Score, at varying of w and t , respectively, in $[1, 6]$ with step 1 and $[0, w/2]$ with step 0.5.

	Existing approaches			Our approach kLDA
	[6]	[18]	[4]	
Sensitivity	0.92	0.98	0.87	0.99
Specificity	0.89	0.99	0.92	0.99
F1-Score	0.90	0.98	0.90	0.99
Method	Fourier	AdaBoost	SVM	k-NN

Table III. Comparative sensitivity, specificity and F1-Score of the proposed approach with respect to state-of-the-art approaches on 2-Class classification.

Tables I, II and III refer only to the experiments conducted by using the windowed dataset W , because, as shown later in Section IV.D, this configuration guarantees the best performances with respect to using the statistical feature dataset S . The tables report the average results obtained by varying w and t , respectively, in the ranges $[1, 6]$ with step 1 and $[0, w/2]$ with step 0.5. Concerning the feature extraction algorithms, for kPCA and kLDA we tested the Gaussian, the polynomial and the linear kernel. The best result was obtained with the Gaussian kernel that the results in Tables I, II, III refer to. The Gaussian kernel parameters have been found by cross validation. Specifically, for each patient with FoG episodes, Table I shows the average F1-Score, for both 2-Class (no-FoG, FoG) and 3-Class (no-FoG, FoG, pre-FoG) cases, achieved by applying different strategies for feature projection. Columns #no-FoG and #FoG indicate the number of windows that have been labelled as normal and freeze of gait, respectively. Column #pre-FoG instead refers to the number of windows preceding those labelled with FoG. It turns out that, by construction, #pre-FoG indicates how many FoG episodes are in the data of each patient. As shown, kLDA achieves very good results for both 2-Class and 3-Class problems.

Table II shows the average Sensitivity, Specificity and F1-Score among all the 8 patients of Table I. The worst case is given by the PCA, while the best results are achieved with kLDA. This clearly implies that the classes cannot be linearly separated, highlighting the considered problem is challenging. Supervised feature extraction approaches (LDA, kLDA) work better than the unsupervised counterparts (PCA, kPCA) especially in the 3-Class case. In the 2-Class case LDA is almost equivalent to PCA probably due to the low dimensionality of the former (= 1).

To analyze more deeply the achieved results, Fig. 4 reports two confusion matrices related to patient 3, which, according

Patient 3 $w=4, t=0.5$	Ground truth				Patient 3 $w=2, t=1$	Ground truth			
	Classes	no-FoG	FoG	pre-FoG		Classes	no-FoG	FoG	pre-FoG
P r e d i c t e d	no-FoG	305	0	0	P r e d i c t e d	no-FoG	1084	0	0
	FoG	73	0	0		FoG	1	250	0
	pre-FoG	27	0	0		pre-FoG	1	0	83

Figure 4. Worst case (left) and best case (right) confusion matrices for Patient 3 at varying (w, t).

to Table I, exhibits the lowest average F1-Score (84% with kLDA). By analyzing the results achieved for each configuration of parameters w and t , we discovered that the configuration ($w=4, t=0.5$) provides the worst results, as the classifiers erroneously recognizes the instances of FoG and pre-FoG as no-FoG (Fig. 4 left panel). On the contrary, the best results are achieved by using the configuration ($w=2, t=1$), where the classifier has almost always recognized the correct class (Fig. 4, right panel). Similar confusion matrices were obtained for the other patients. Selecting the correct (w, t) configuration is then fundamental for an effective classification. In almost all cases, the configuration ($w=2, t=1$) guarantees the best Sensitivity and Specificity.

Table III compares our approach, on the DAPHNET dataset, with three state-of-the-art techniques [9,18,19]. Only a comparison with the 2-Class problem was possible, as the considered works do not use the pre-FoG label, since they are focused on FoG detection rather than FoG prevention.

D. Results for the statistical feature dataset

Previous tables refer to results on the windowed dataset W , as the statistical feature dataset S gave worse performance. To better clarify the different results achieved by W and S , Fig. 5(a) and 5(b) present the mapping after applying Gaussian kLDA on, respectively, W and S , for patient P1, with $w=4$ and $t=1$, for the 3-Class problem. kLDA clearly separates samples of different classes, i.e., no-Fog (blue), FoG (red) and pre-FoG (yellow) on the W dataset (Fig. 5(a)), while this is not true for the S dataset (Fig. 5(b)). Similar results, omitted for lack of space, were obtained for the other patients.

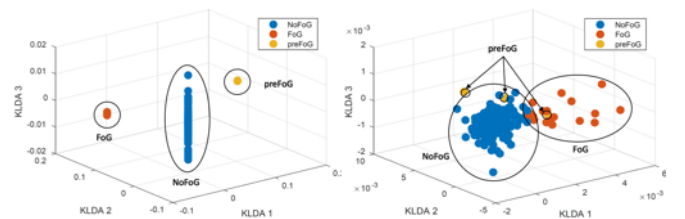


Figure 5. Results of the kernel linear discriminant analysis on Patient P1 with $w=4$ and $t=1$ using the (a) windowed dataset and (b) statistical feature dataset. Classes are clearly separated in (a), the same is not true in (b).

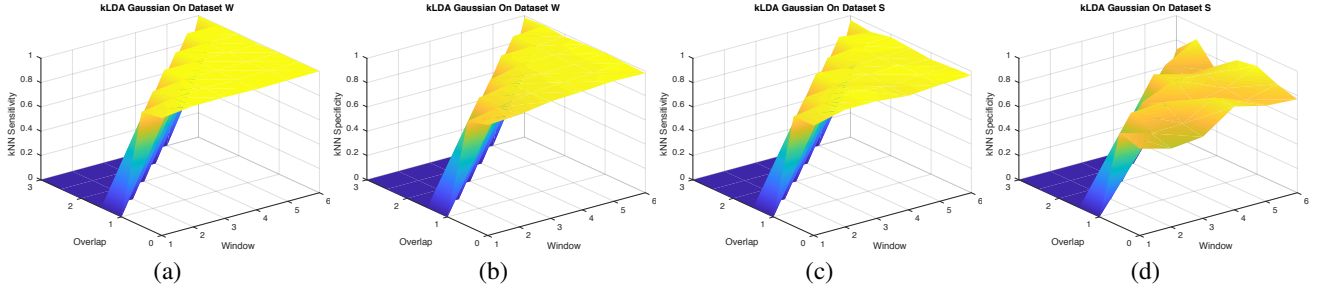


Figure 6. Average sensitivity (a) and specificity (b) using W , and average sensitivity (c) and specificity (d) using S , obtained with k LDA at varying of the window size w and the overlap interval t .

Fig. 6 further highlights how the W dataset outperforms the S dataset for the 3-Class problem. Fig. 6(a) and 6(b) show, respectively, the average Sensitivity and Specificity on W for all the possible configurations of (w, t) and for all patients (P1, ..., P10). Fig. 6(c) and 6(d), instead, show average Sensitivity and Specificity on S . In Fig. 6(a) and 6(b) Sensitivity and Specificity achieved approximately 98%, while in Fig. 6(c) and 6(d) they reached approximately 94% and 70%.

E. Performance on mobile device

We evaluated the classification performance on three different smartphones in terms of response latency, RAM and ROM memory and battery consumption. Table IV shows the features of the tested smartphones (upper part) and the outcomes (lower part). *Latency* was the time spent for the computation of the workflow presented in Fig. 2, i.e., the time elapsed from the reception of the input data gathered from the accelerometers till the classification of patient’s gait. The time taken to transmit data from the accelerometers to the smartphone and the stimulus command from the smartphone to the smartwatch is negligible. ROM and RAM memories were mainly used to save the transformation matrix and the features dataset obtained after the offline training data pre-processing phase (Fig. 3). Battery consumption shows the average consumed battery for a total of 10 hours. Regarding the quality of classification, the results were equal to those obtained in computation on an external server machine (Tables I, II and III).

Metrics	Nexus5	S7 Edge	6T
CPU (GHz)	2.3	2.3	2.8
Core	Quad	Octa	Octa
RAM (GB)	2	4	6
ROM (GB)	16	64	128
Battery Capacity (mAh)	2300	3600	3700
Latency (ms)	≈ 120	≈ 100	≈ 105
Used ROM (Mb)	≈ 2	≈ 2	≈ 2
Used RAM (Mb/h)	113	114	120
Battery Consumption (mAh)	78	76	80

Table IV. Device performance metrics.

V. DISCUSSION

We presented a methodology based on machine-learning algorithms to classify the PD gait features as pre-FoG, FoG, and no-FoG patterns with a wearable system. Our results prove the effectiveness of the proposed FoG prediction and show the possibility of implementing our solution on a wearable system.

We compared the present approach with three state-of-the-art ones, in terms of Sensitivity, Specificity and F1-Score: the 2-Class and 3-Class classification (Table III) outline the better

performance of the proposed algorithm, compared to previous 2-Class [9,18,19], and 3-Class ones [23,24], with a higher, on average over 2%, overall accuracy. In particular, our solution with k LDA achieved the best results. It is worth noting that the k -NN is a less complex classification algorithm than those based, respectively, on AdaBoost, SVM, and fast Fourier transform [9,18,19], and is more suited for a real-time classification on a wearable device with limited computing and memory resources. Moreover, our approach achieved very good results on different configurations of window size w and overlap interval t , in particular when the windowed dataset W is used (Fig. 6). Finally, 3-Class classification permitted the prediction of FoG w seconds in advance.

One of the main limitations of the existing approaches for FoG detection is that the classification phase requires to offload the computation to an external server, due to high computing resources. On the contrary, we proved our classification algorithm can be executed on a resource-constrained device, like a smartphone. Classification performance, response latency, RAM and ROM memories and battery consumption did not differ significantly among the three mobile devices tested. These finding confirm the overall “lightness” of the algorithm, thanks to the used classifier and the reduction in the number of processed features, with no detrimental effect on the overall classification accuracy. This point is of utmost importance since, differently from already existing solutions, it removes the need to reach an external server, that may result in unpredictable delays. In addition, the system can work even in absence of an Internet connection.

Our study has some limitations. The sample size of DAPHNET project was small. Moreover, FoG has different subtypes and may be caused by different triggers in PD [5,34]. FoG events in the DAPHNET dataset [9] were not classified according to subtypes, so we could not test if our detection algorithm can classify pre-FoG, FoG and no-FoG, with a similarly high performance for all FoG subtypes. Further studies with larger patients’ samples and different FoG subtypes are necessary to confirm the reproducibility of our results. Further, we applied the algorithm off-line to a pre-existing dataset, so confirmatory studies are necessary to evaluate its on-line performances. Finally, computational sustainability was tested on smartphones. Our algorithm should be tested on devices with even less computational resources (e.g., smartwatches), to support its use on wearable on-demand cueing device.

VI. CONCLUSION AND FUTURE WORK

External cueing is a non-pharmacologic treatment of FoG that allows PD patients to overcome the impairment of

involuntary movement, one of the pathophysiological mechanisms of FoG, by switching gait to a goal-directed movement that is less impaired in PD [35]. Different devices that couple a detection system to a cueing stimulus delivery have been proposed. To the best of our knowledge, five of these devices have been tested on PD patients, but only three were effective in reducing FoG frequency and duration [16]. However, the study quality was generally poor (i.e., quasi-experimental, low statistical power), and the different approaches impeded a direct comparison or a meta-analysis.

The overall efficacy of a cueing on-demand wearable device to reduce FoG and falls in PD depends on the high accuracy and short lag times between pre-FoG recognition and cueing administration. Future studies should explore the on-line performance and ability to accurately administer the cue of our high-performance classification methodology in PD patients.

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