Master Thesis

# Sensor Framework for Cerebral Palsy Ankle-Foot Orthosis: Development and Technical Validation

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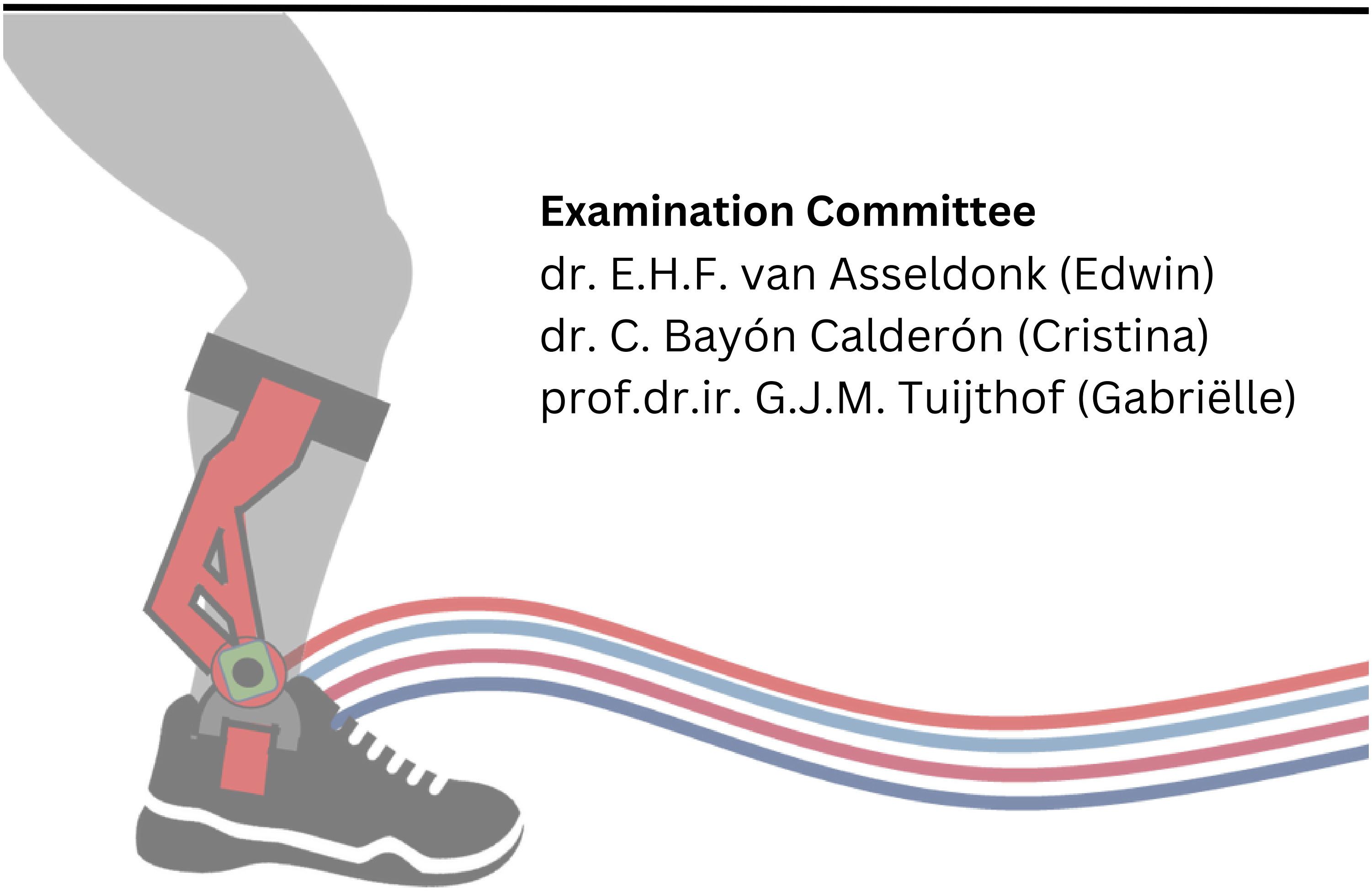

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# Sensor Framework for Pediatric Cerebral Palsy Ankle-Foot Orthosis: Development and Technical Validation

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***Abstract*—Cerebral Palsy (CP), a prevalent neurodevelopmental disorder affecting motor function, often manifests with gait dysfunction. While passive rigid ankle foot orthoses (AFOs) are commonly used, their limitations prompt the exploration of innovative solutions. The inGAIT-AFO, developed within the inGAIT project, introduces a quasi-passive AFO with customizable ankle stiffness. This thesis centers on the creation of a portable sensor data framework to assess the inGAIT-AFO's efficacy through the analysis of the gait of those using it. The goal was to provide a ready-to-use solution for clinical assessment, enhancing AFO development for children with CP.**

**Two main versions of the framework were developed following a predefined set of requirements. The framework made use of force sensing resistors (FSRs) and encoders to respectively estimate push-off forces and measure ankle angles during gait. Both versions of the framework were technically validated against the requirements.**

**A framework was developed that is sufficiently capable of measuring and estimating the required outcome measures meeting 87% of all requirements and 100% of all must-have requirements.**

***Index Terms*—Cerebral palsy, ankle foot orthosis, gait analysis**


Fig. 1: A photo of the inGAIT-AFO being worn. Enlarged is an illustrated version of the leafspring-CAM mechanism including the adjustable slider. Figure partially taken from Bayón *et al.* [20].

## I. Introduction

With a prevalence of 2.11 per 1000 live births, cerebral palsy (CP) is the main neurodevelopmental disorder that affects motorfunction resulting in various degrees of physical and/or cognitive disability [1]. CP arises due to damage to the child's brain during birth or early childhood, causing lasting neurological impairments in motor control, strength, muscle function, and balance or posture [2]. The most common types of CP are spastic hemiplegia and spastic diplegia [3]. CP affects physical quality of life specially in children [4], and it is linked to a substantial economical burden for families and society [5].

Among the most common impairments associated with CP is gait dysfunction [2]. Because of this, children with CP have a lower speed and a higher energy expenditure during walking [6]. Gait dysfunction due to CP can generally be treated through invasive and non-invasive methods. The main non-invasive methods are physical therapy, occupational and recreational therapy and orthotic interventions [7].

Among the different orthotic solutions, ankle foot orthoses (AFOs) are the most common type, frequently used to assist gait in children with CP [8]–[14]. During walking, the ankle joint accommodates for a substantial part of the propulsive forces, as well as contributing to balance through center of mass control [9]. Notably, ground reaction force (GRF) and maximal voluntary plantar flexion at push-off in pediatric CP subjects is often lower for children with CP [15]. Various types of AFOs are deployed in the treatment of pathological gait patterns normally to maintain or improve function allowing for greater joint stability [16]–[19]. However, the most common type of AFOs are passive, often rigid, which have as downside that they may impede push-off power, as well as limiting ankle range of motion (ROM) [10], [14], [19].

The inGAIT project was established to provide new solutions related to AFOs for children with CP [20]. Within this project, the inGAIT-AFO was developed: a quasi-passive device based on a leafspring-CAM mechanism, which is intended to store energy during the stance phase of walking, and release that energy at the instant of push-off (Fig. 1) [20]. The inGAIT-AFO has the possibility of customizing the stiffness around the ankle joint thanks to a slider (depicted in red in Fig. 2). This slider can be adjusted along the leaf spring (blue in Fig. 2), effectively changing the employed spring module.

To assess the effectiveness of the inGAIT-AFO, it is important to assess the gait of those using it. Gait analysis

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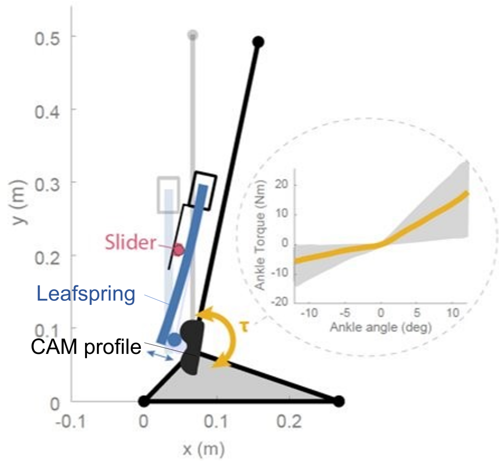


Fig. 2: Schematic of mechanical functionality of the inGAIT-AFO. Image edited from Bayón *et al.* [20]. The graph in yellow illustrates the ankle stiffness of the leafspring-CAM mechanism. The grey area represents the angle-torque curve as the slider position changes.

for CP generally consist of measuring kinematics, kinetics, electromyography and plantar pressure [2]. Gait characteristics vary for children with CP between laboratory and daily life conditions [21]. This variation can be attributed to the difference between a patient's walking capacity in a controlled environment and their walking performance in daily life [21], [22]. Therefore, it is necessary to develop a wearable and portable gait analysis system that can be employed in a controlled as well as uncontrolled environment.

The aim of this thesis is to develop a sensory data collection, processing and visualization framework (referred to as "framework" throughout the thesis) for the inGAIT-AFO, and to validate this framework technically and with respect to usability. The goal is to analyze the data generated in these validations, allowing for iterative improvements to the developed data framework. Through these actions, this study sets out to prepare and deliver a device and framework ready to be utilized in the clinical assessment of the inGAIT-AFO, both in a controlled environment as well as in daily living. The development of this framework will have significance in the evaluation of the inGAIT-AFO, therefore aiding the development of AFOs for children with CP.

## II. Sensor data acquisition and recording framework

The development of the sensor data framework was executed through an iterative process. Two main distinct versions of the framework were engineered, which will both be explained in this Chapter. Section II-A presents the requirements for the framework, Section II-B introduces the main sensors that were employed, Section II-C presents the first version of the framework and subsequent validation, and Section II-D explains the second version of the framework and its validation. Appendix A provides a reference to all documentation that is relevant to the framework and its development.

### A. Requirements

The requirements for the purpose of this thesis were derived and extrapolated from the initial requirements defined for the overarching inGAIT project [23]. The requirements were divided into General Requirements (GRs), and Technical Requirements (TRs), the most relevant ones summarized in Table I. The technical requirements were further subdivided into sensing (S), comfort and ergonomics (CE), weight and portability (WP), system manipulation and control (MC) and power and autonomy (PA). Requirements were ranked either "M" for must-have, "D" for desired to have or "O" for optional.

A full list of all requirements, including desirable and optional requirements, with respective rationale or sources, can be found in Appendix B.

### B. Main metrics

In order to fulfill the exposed requirements, two main metrics needed to be measured or estimated: (1) ankle kinematics, (2) forces between the feet and the ground. These metrics were measured or estimated using sensors. The selection process of these sensors is explained below.

*1) Ankle Kinematics:* based on the literature, some common methods used to measure joint kinematics during gait are optical detection with or without active or passive markers, inertial motion capture using accelerometers and gyroscopes, and mechanical measurement with potentiometers or encoders [24]. Optical detection of kinematics provides highly accurate measurement and possesses the benefit that the subject is not impeded by wires or battery packs [24]. However, it requires multiple high speed cameras making it costly as well as stationary [24]. Inertial measurement allows precise estimation of posture using angular rate and incline, but is limited by so-called drift, which is a cumulative error introduced by the estimative nature of these sensors [24]. Mechanical measurement is a straightforward and accurate way to measure joint angles, although its accuracy depends on the rigidity of wearable equipment that is used. It is therefore a more suitable solution for rigid wearable robotics [25], [26]. An additional drawback to mechanical measurement is that it usually requires the subject to carry batteries and wires.
Since the inGAIT-AFO is a mechanical structure, mechanical measurement is particularly suited. Additionally, since the system is desired to function in an uncontrolled environment as defined in the requirements, optical detection is not a suitable solution. Therefore, two magnetic encoders (AS5048b, AMS-OSRAM AG, Premstaetten, Austria) were employed in the framework, one for right and one for left ankle joints. This specific encoder model was selected for being magnetic and having a resolution of 0.0219° [27], which is a confidently higher resolution than specified in the requirements. Additionally, this encoder model has been previously used in the literature for tracking ankle and knee angles during gait [28], [29].

*2) Forces between feet and ground:* precise measurement of ground reaction forces (GRFs) during gait is another essential measure to estimate biomechanics, aiding in the comparison

TABLE I: Overview of all Must-Have General (GRs) and Technical Requirements (TRs), and their values in the version 1 (V1) and version 2 (V2) framework. TRs are further divided into sensing (S), comfort and ergonomics (CE), weight and portability (WP), system manipulation and control (MC) and power and autonomy (PA)

| Req. ID | Description | Target value | V1 value | V2 value |
|---|---|---|---|---|
| **General Requirements:** | | | | |
| GR-01 | The device should be adjustable and valid for different pilot's sizes | True | True | True |
| GR-02 | The device should be portable and lightweight | True | True | True |
| GR-03 | Environment: indoor, in a controlled environment (research environment) | True | True | True |
| GR-05 | The device must not cause harm to the user | True | True | True |
| **Technical Requirements:** | | | | |
| TR-S-01 | Sampling frequency of data reception of inGAIT | $\geq$ 50Hz | 13.3Hz ($\pm$1.1) | 101Hz ($\pm$3.8) |
| TR-S-02 | The system should allow on-board data storing and wireless transmission to a PC for postprocessing | True | False | True |
| TR-S-03 / TR-S-06 | The angle between foot and shank should be known within a tolerance of 0.5° | $\pm$0.5° | True | True |
| TR-S-07 | Safe storage of personal data | True | True | True |
| TR-S-08 / TR-S-10 | Capable of estimating force between forefoot and ground during gait on both sides, accurate enough to detect a significant difference at a 30% change in walking speed | True | True | True |
| TR-S-09 | The system must be able to detect the heel strike event on both sides | True | False | True |
| TR-CE-02 | The device does not impede the existing functionality of the user | True | True | True |
| TR-CE-06 | The device should be adjusted and fitted to each subject | True | True | True |
| TR-CE-07 | Skin pressure, friction or abrasions should be avoided when using the device | True | True | True |
| TR-CE-08 | The device should be easy to put on and take off | True | True | True |
| TR-MC-01 | The user should be sufficiently informed about the operation and manipulation of the device | True | True | True |
| TR-MC-02 | The system should be easy to use and easy to adjust | True | True | True |
| TR-MC-03 | The integration of new processing algorithms into the code should not be an extremely time-consuming process | True | True | True |
| TR-PA-01 / TR-PA-02 | The system can be charged using an external charger | True | True | True |

of gait with and without assistive devices. Force plates are globally considered the gold standard in laboratories in the assessing of GRFs of pathological as well as regular gait [30], [31]. However, use of force plates is often limited to laboratories (stationary devices), as well as being expensive and difficult to use. A way to achieve a wearable GRF measurement system is by using an electromyography-driven model such as developed by Honert and Zelik [32]. Such a model measures muscle activity and uses it to estimate the GRFs that would be exerted. However, shortcomings of this model are its relatively large error due to rigid-body assumptions and indirect measurement [32]. In addition, models have been developed to estimate ground reaction force from inertial measurement using machine learning algorithms [33],

[34]. Another way to estimate GRFs is with commercially available systems that allow for measuring of forces exerted by the foot during gait, such as the Pedar in-shoe system (Novel GmbH, Munich, Germany) [35], [36], the OpenGo system (Moticon GmbH, Munich, Germany) [37] and the Nushu system (Magnes AG, Zurich, Switzerland) [38] among others. Main drawbacks of commercial devices are their high cost and relatively limited customizability. Finally, force-sensing resistors (FSRs) can also be used to measure forces during the gait cycle, and have been previously used for this purpose in children with CP [39], [40]. The latter avenue was previously explored as a part of the inGAIT project [41], in which a specific type of FSR was investigated as potential solution to estimate push-off force while walking. Drawbacks of FSRs include their need for wiring and a battery as well as outputting non-linear data and therefore being difficult to calibrate. However, for the purpose of inGAIT, having a good relative estimation of the forces is sufficient, even if they were not directly translated into standardized units (e.g. N or kg).

Based on these reasons and the requirement to have a mobile system, FSRs (FlexiForce A502, Tekscan Inc, Massachusetts, USA) were selected for estimating forces exerted by the feet at two locations (heel and ball). Therefore, four FSRs were used in total. These FlexiForce FSRs have been used before to measure ankle plantar flexion force in pediatric CP patients [40]. Additionally, the selection of the FSR models was based on previous research in the inGAIT project [41]. The FSR signal is obtained using an electrical circuit (Appendix D). Subsections II-B2a and II-B2b explain how the FSRs were calibrated and validated.

*a) Calibration of FSRs:*

- **Methods**: a calibration procedure was executed on the FSRs. This calibration attempted to translate the output of the FSRs into a standardized unit. During this calibration, the four FSRs were placed on a calibrated force plate (FIT Gen 5, Bertec Inc., Ohio, USA) at the motion laboratory of the University of Twente. Sampling frequency of the force plate was 1KHz, and sampling frequency of the FSRs was 100Hz. With a round bar, a concentrated force was applied to the FSRs. This force varied continuously in a range from 0N to 300N. Three short pushes were used before the calibration to synchronize the force plate data with the FSR data. The recorded voltage output of the FSRs was then processed and fit to correspondent force plate data in MATLAB (version 2023a, Mathworks, Massachusetts, USA), following the methods described by Centeno [41], which were developed as a part of the inGAIT project. According to these methods, calibration models were computed for each FSR as:

$$F = \begin{cases} p_1 \cdot S + p_2, & \text{if } S \leq S_{th} \\ a \cdot e^{bS}, & \text{if } S > S_{th} \end{cases} \tag{1}$$

where $F$ is the force in Newtons, $S$ is the sensor output in Volts, $p_1(\mathrm{NV}^{-1})$, $p_2(\mathrm{N})$, $a$ and $b(\mathrm{NV}^{-1})$ are coefficients and $S_{th}$ is the threshold value for $S$ at which the model changes from linear to exponential.
- **Results:** calibration of the FSRs yielded coefficients to be

TABLE II: Coefficients for calibration model FSRs, to be used with equation 1

| FSR Location | $p_1$ | $p_2$ | $a$ | $b$ | $S_{th}$ |
|---|---|---|---|---|---|
| Heel Right | 72.7 | 4.95 | 21.9 | 1.42 | 0.8 |
| Front Right | 226 | -1.89 | 32.7 | 2.04 | 0.87 |
| Heel Left | 79.3 | -1.41 | 28 | 1.02 | 1.1 |
| Front Left | 94.8 | 0.285 | 34.9 | 0.985 | 1.1 |

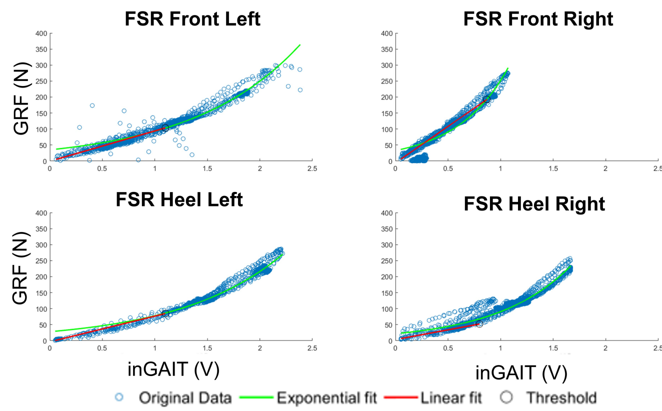


Fig. 3: The vertical GRF as recorded by the force plate during calibration against the voltage obtained from the inGAIT FSRs with respective linear and exponential models

plugged into equation 1 for all FSRs (see Table II). When plotting the FSR data against the force plate data, as well as the calibration using the coefficients of Table II, results as displayed in Fig. 3 are obtained, which demonstrated an adequate fit by combining the linear and exponential parts of the model presented in equation 1.
When applying the obtained models to the recorded raw data of the FSRs while using the round bar, and comparing with the registered force plate data, the data properly matched (Fig. 4). However, when applying the same models to the raw data of the FSRs recorded during walking (able bodied female subject, age 33), and comparing this to the force plate data of this same trial, the calibrated FSR signals yield lower peaks than the force plate (Fig. 5). This discrepancy could be attributed to the FSRs measuring a smaller surface area, while the foot uses a larger surface area during gait. Since the calibration was derived from a concentrated pressure, this could be a valid explanation. It was attempted to apply a gain to the calibrated data to achieve more realistic results, but this did not produce a consistent output (i.e. some peaks were higher than the force plate reference while others were lower). Therefore, force output of the FSR could not reliably be converted to a standardized unit under the current calibration. However, FSR outputs between different trials can still be compared to each other to attain relative differences.

*b) Validation protocol of FSRs:*

- **Methods**: the FlexiForce FSRs were also validated in two ways: (a) by testing their ability to significantly

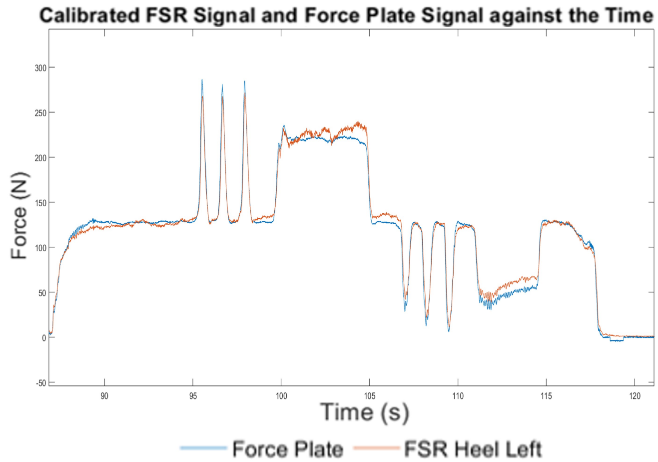


Fig. 4: FSR signal calibrated according to Table II, and right force plate signal plotted against the time during calibration

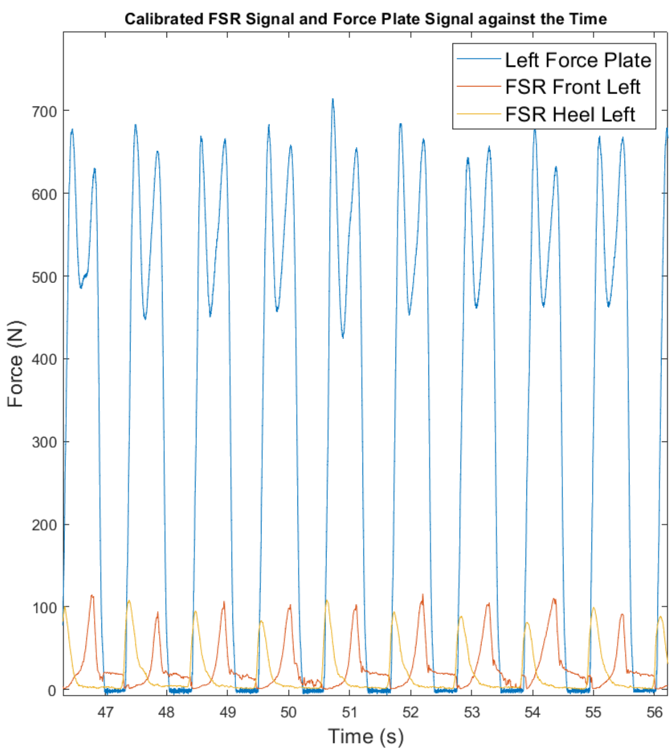


Fig. 5: FSR signal calibrated according to Table II, and left force plate signal plotted against the time during walking

detect relative changes in push-off force when walking at different speeds, and (b) by testing their test-retest reliability in different moments with the same subject at the same walking speeds.

For the **change in walking speed**, previous research shows notable differences in ground reaction forces for higher gait speeds [42], [43]. Within these studies, significant effects were identified at a change in speed of 10%. It was therefore hypothesized that a significant difference could be detected in the data garnered from the FSR while using a larger variation (30%) in walking speed than described in aforementioned literature.

The validation protocol consisted of walking exercises in able-bodied adult participants ($n = 5$, mean age $\pm$ standard deviation $= 25.6 \pm 3.5$). This sample size was selected to validate the FSRs in multiple subjects of varying genders and masses. Validation was executed with one FSR. Since all FSRs are of the same model and were calibrated through the same process, other FSRs were expected to behave the same.

Participants were asked to walk on the treadmill while increasing the speed until a comfortable level ($v$) was found. Then, the two minute walk test (2MWT) was completed three times. One at comfortable speed $v$, one at $v + 30\%$ and one at $v - 30\%$. During these walking trials, one FSR was located under the head of the first metatarsal, also referred to as the “ball” of the foot, of the left foot of subjects between differing walking speeds. Placement of the FSR follows the study of Conner *et al.* [40]. The calibrated output data of the FSRs were read and stored at a sampling frequency of 100Hz.

These data from this validation were processed in MATLAB. The peak forces generated per gait cycle at varying walking speeds were grouped and statistically examined using a Wilcoxon rank sum test within each participant. This statistical test was chosen to allow for a comparison of means of quantitative data without the requirement of normality in all data. A significance level of 0.05 was chosen to evaluate the Wilcoxon rank sum hypotheses.

Regarding the **test-retest reliability**, it was examined to verify whether data produced by the FSRs on different days can reliably be compared. The protocol consisted of three 2MWTs performed by a male subject (age 25) on a treadmill at 2.8 km/h, 4 km/h and 5.2 km/h. These three tests were repeated at three separate testing moments, at least three hours apart. FSRs were located under the ball of the foot by attaching them to the bottom of the shoe insole, intending to keep their location with respect to the foot consistent. The subject took off the shoes with sensors and the testing system was shut off in between testing moments. It was hypothesized that the FSRs would measure similar results between different testing moments.

Data from FSRs were collected and stored at a sampling rate of 100Hz, these generated data were analyzed in MATLAB. The mean gait cycle curve for push-off force was normalized with respect to the highest point of the curve per testing moment, and these normalized curves were then analyzed for significant differences. A one-way analysis of variance (ANOVA) was used ($\alpha = 0.05$), to detect if a significant difference was present between the different testing moments. ANOVAs were applied separately for each walking speed.

- **Results:** the processed results for the **change in walking speed** (Fig. 6), show a statistically significant increase in push-off force for higher walking speeds ($p < 0.01$) for all but one of the intra-participant comparisons. For

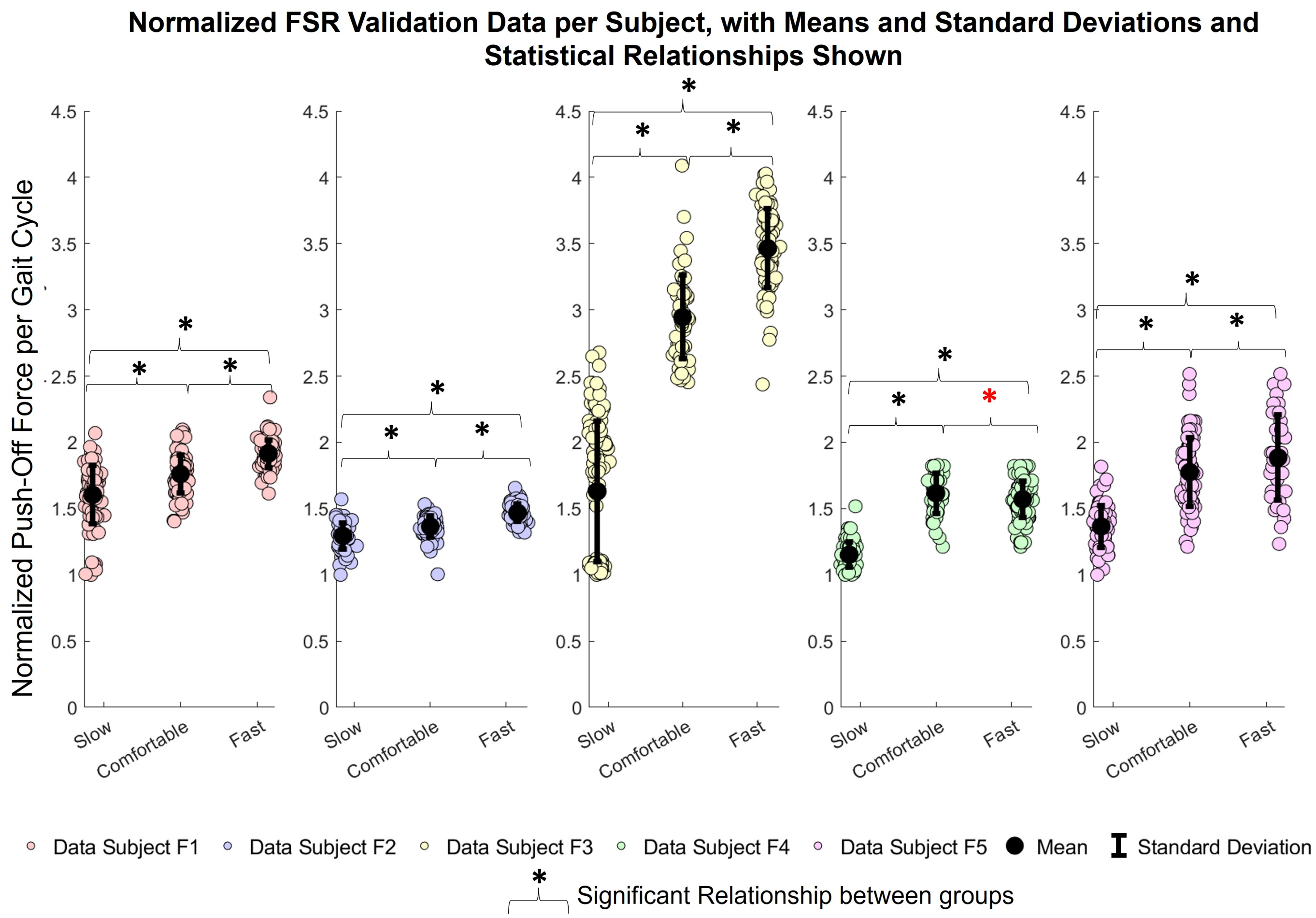


Fig. 6: Normalized FSR validation data per subject including mean and standard deviations. Data was normalized by dividing all data by the minimum value for the subject. Statistical significant effects, according to the Wilcoxon rank sum analysis, are shown in the figure with an asterisk. The red asterisk shows a significantly lower push-off force for fast walking compared to comfortable walking

subject F4 it shows a significant decrease in push-off for the higher walking speed (Fig. 6).

For the **test-retest reliability**, after normalization with respect to the maximum value of mean curves per recording moment, the mean and standard deviations of the left push-off force throughout the gait cycle are visually similar between different recording moments at all three speeds (Fig. 7). It was found that the mean curves are not significantly different for any walking speeds (statistics in Table III).

| Walking Speed | ANOVA Results | |
|---|---|---|
| | $p$ | 0.32 |
| 2.8 km/h | $s$ | 0.12 |
| | $DoF$ | 897 |
| | $p$ | 0.46 |
| 4.0 km/h | $s$ | 0.18 |
| | $DoF$ | 897 |
| | $p$ | 0.99 |
| 5.2 km/h | $s$ | 0.25 |
| | $DoF$ | 897 |

TABLE III: One-way ANOVA results for normalized mean curves or gait cycle at different walking speeds showing the $p$-value, the test statistic ($s$) and the degrees of freedom ($DoF$) per performed ANOVA

### *C. inGAIT framework version 1*

Using the selected sensors, a first version of the framework was developed. In the following subsections, the design and the technical validations performed with this first version are presented.

*1) Design:* version 1 of the framework (Fig. 8) comprised a backpack housing an ESP32-S2 microcontroller (LilyGo, Shenzhen Xinyuan Electronic Technology Co., Ltd, Shenzhen, China) with an integrated display. The microcontroller facilitated the reading of two FlexiForce FSRs (see section II-B2), along with two magnetic encoders (see section II-B1) and a total of six low sensitivity FSRs embedded in the insoles for detecting heel strike. Cables established connectivity between the backpack and the sensors, while an external button, linked to the microcontroller, enabled zero-calibration of the encoders

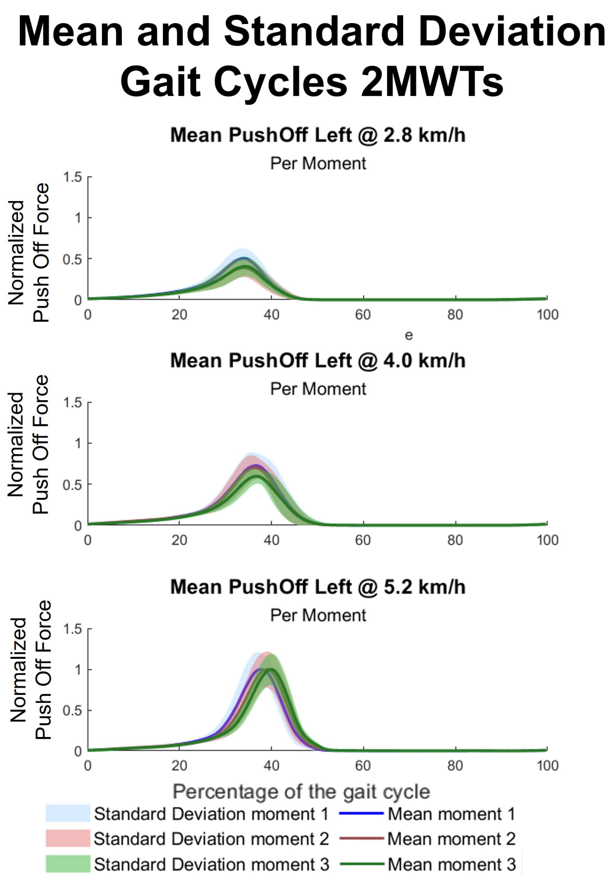


Fig. 7: Mean and standard deviation of the normalized left push-off force at three different walking speeds. Normalization was done by dividing all values by the highest peak of the mean gait cycles per recording moment. The three different colors resemble the different moments at which the tests were performed

and toggling of the recording process. Recorded data was stored on a micro SD card. Software was developed in C++.

*2) Protocol for technical validation:* a technical validation of the framework was carried out. This assessment focused on evaluating the system's performance metrics with regards to the defined requirements.

Validation procedures were conducted during the pilot testing phase involving the inGAIT-AFO with the version 1 framework. Since the technical validation is primarily intended to validate requirements with a binary outcome (i.e. the requirement is met or not met), the study comprised a limited cohort of four participants ($n = 4$, mean age $\pm$ standard deviation = $9.5 \pm 2.06$), including two individuals with CP and two with typical development (TD). The pilot testing was organized into two sessions, each following a specific protocol as illustrated in Appendix B. During both session, participants executed walking tests with the version 1 framework. Besides technical validation of the framework, this pilot testing had as goals to obtain data pertaining to the usability of the inGAIT-AFO as well as the gathering and analysis of walking data of the subjects with and without the inGAIT-AFO.

For the validation, subjects were given customized socks containing a fabric pouch in which the FlexiForce FSRs were located. This way, the FSRs were located under the ball of the foot. Encoders were connected to the joint of the inGAIT-AFO. The subjects wore the backpack on their backs during the trials. A photo of the set-up taken during one of the tests is presented in Fig. 9. Throughout the validation, subjects were asked to indicate adverse events (e.g. skin integrity issues) stemming from either the inGAIT-AFO or the sensor framework.

Data analysis scripts were developed using MATLAB, which were used to subsequently evaluate mean and standard deviation push-off force as well as mean and standard deviation ankle angles for all subjects. A conversion was applied to the ankle angles to translate the joint angle of the AFO to the anatomical joint angle. These combined analysis scripts allowed for interpretation of the pilot testing results as well as establishing an analysis framework for future testing with the device.

*3) Results:* throughout the pilot testing phase, the system generally functioned as expected and was able to record the ankle angles as well as the push-off forces during walking tests. Subjects did not report any skin irritation issues caused by the backpack or cables. However, at times cables had to be secured to the legs with straps to avoid hindrance during walking. The framework proved to be adjustable, easy to put on and take off, portable and lightweight, and worked indoor as well as outdoor, satisfying GR-01 through GR-04 and TR-CE-06 through TR-CE-08. Risks of causing harm to the user were minimized by using the system in a supervised setting and not exceeding a voltage supply of 5V, as per GR-05. The battery could be charged using an external charger per TR-PA-01, but the battery life was calculated to be around 8 hours (calculation in footnote of Appendix B), therefore not complying with requirements GR-06 and TR-PA-02.

An average sampling frequency of 13.28Hz ($\pm$1.05) was achieved over all analyzed data, which is lower than the 50Hz specified in TR-S-01. The framework allowed safe on-board storage of data but not wireless transmission, which meets requirement TR-S-07 and partially TR-S-02. The capability of the device to measure the ankle angles at a resolution of 0.0219° satisfies TR-S-03 and TR-S-05, although low sampling frequency limited the reliability of the measured ankle angles. The angle between foot and ground was not known, failing to satisfy TR-S-04. The framework was capable of estimating force between the forefoot and ground within the specified resolution (TR-S-08 and TR-S-10). Due to low quality data obtained from the FSRs under the heels, the heel strike event was at times not captured properly, so TR-S-09) was not met satisfactorily. No additional noise was produced by the framework per TR-CE-03. The device was worn with normal clothing while allowing breathability of the skin and generally not impeding the user (TR-CE-05). The device exterior could be cleaned by wiping down the backpack (TR-CE-03). Adding new processing algorithms was relatively simple using C++ code (TR-MC-03), but there was no possibility for integration with external systems per TR-MC-04.

Summed up (see Table I), 22 of 31 requirements (70%)

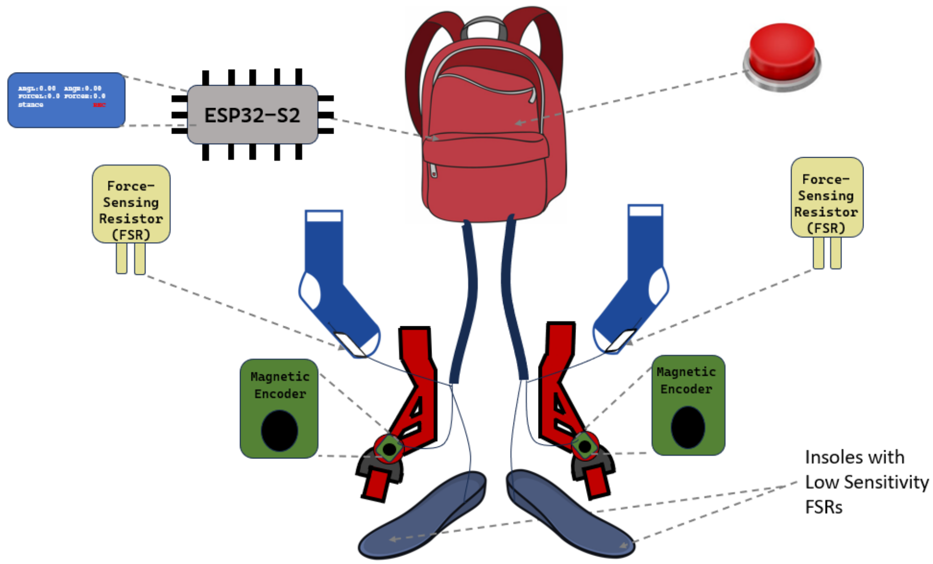


Fig. 8: Graphical System Overview of Version 1

Fig. 9: A photo of a participant wearing the version 1 system during pilot testing

were met, of which 14 out of 18 must-have requirements (78%), 6 out of 10 desired requirements (60%) and 2 out of 3 optional requirements (67%). Appendix C includes an overview of all requirements including whether or not they were met in version 1.

### D. Framework version 2

Based on the shortcomings identified during the validation of the version 1 framework, a second version (Fig. 10) was developed to better meet the requirements. In the following subsections, the design and technical validation of the second version are presented.

*1) Design:* the version 2 system equally consisted of a backpack carrying the electronics. An Orange Pi Zero3 microcontroller (Shenzhen Xunlong Software Co., Ltd, Shenzhen, China) operated the framework. The microcontroller was connected to a touchscreen display, mounted on the outside of the backpack, allowing the system to be operated with ease by the user. The same AS5048b encoders were used to track the ankle angle during gait. In addition to the two FlexiForce FSRs at the front of the feet, two more FSRs of the same model were added to capture the heel strike event. Since the heel strike event was already captured by these FSRs, the previously included insoles with low sensitivity FSRs were removed in the version 2 framework. A custom printed circuit board (PCB, Fig. 11) was developed to robustly secure the electronics inside of the backpack. This PCB was designed using CircuitMaker (version 2.2.1, Altium Ltd., California, USA), and consisted of connection points for the OrangePi, encoders and the FSRs with their electrical circuits. It additionally held a real-time clock, enabling it to keep track of the current date and time regardless of internet connection.

The touchscreen interface leveraged the GTK3 library in Python to create a graphical user interface (GUI, Fig. 12). The interface provided users with the flexibility to choose their preferred language (English, Spanish or Dutch) and offered the option to initiate the system with or without Robot Operating System 2 (ROS2) functionality. ROS2 facilitated real-time communication between the OrangePi microcontroller and an external device, such as a laptop, when both were connected to the same network [44]. Both the ROS2 and local interfaces featured start/stop buttons to control the sensor readings, as well as calibration buttons for zero-calibrating each encoder. Additionally, both interfaces included a start/stop recording button, enabling the writing of collected data into a comma-separated values (CSV) file. Since ROS2 allows for external data visualization, the data was not displayed within the interface itself. In contrast, the local interface incorporated sensor names and their corresponding real-time values directly within the interface. All software was developed in Python. An instruction manual for the GUI with images of the different

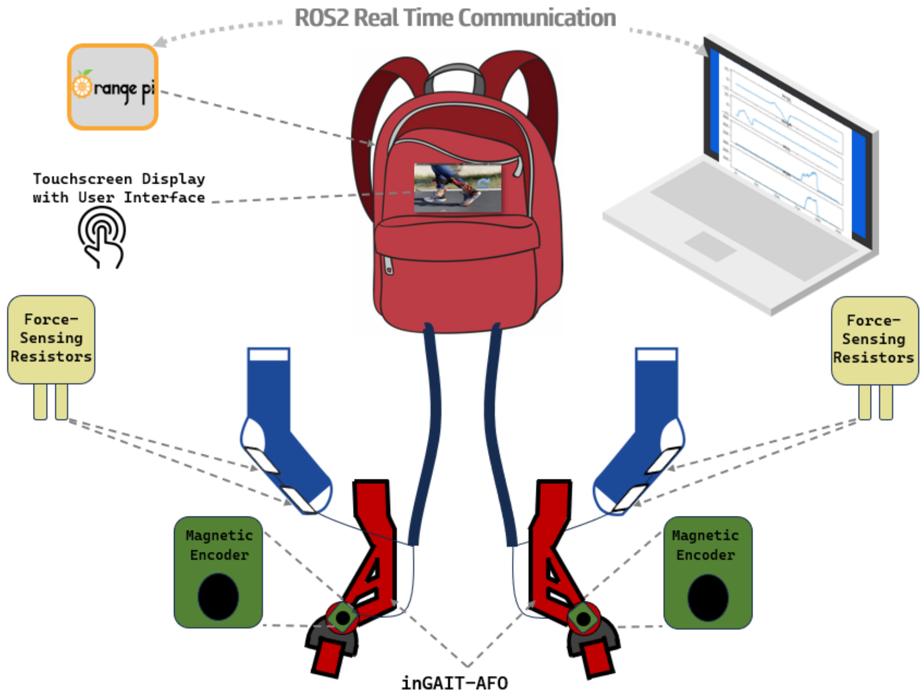


Fig. 10: Graphical System Overview of Version 2

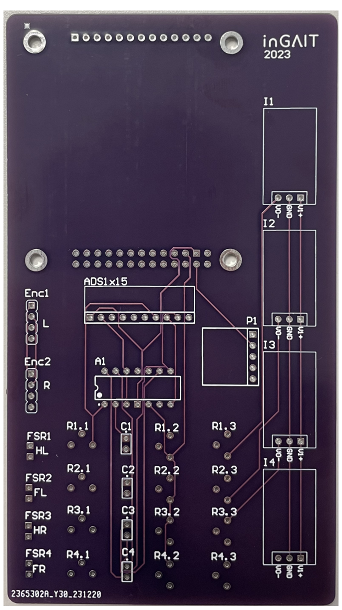

Fig. 11: A photo of the custom PCB for connecting the OrangePi microcontroller, the encoders, the FSRs and the electrical circuits for the FSRs (Appendix D).

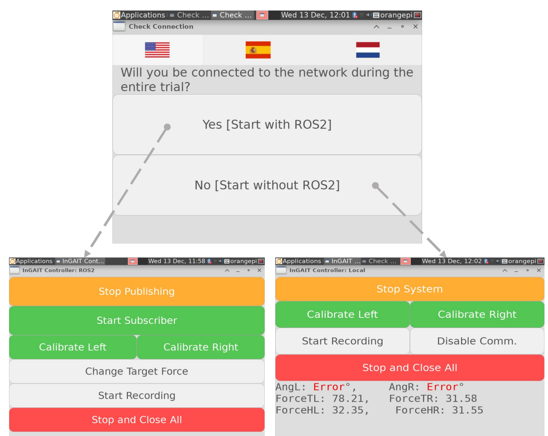


Fig. 12: Screen-captures of the three main menus of the user interface

menus can be found in Appendix E.

The full system (worn in Fig. 13) was designed to be adjustable, lightweight and user-friendly.

*2) Protocol for technical validation:* the second version of the framework was validated in two ways: (a) technically, with one able bodied subject (age 33), (b) with respect to usability of the interface by seven subjects.

*a) Technical:* four tests were executed (Fig. 14), with ample recovery time between them. One able bodied female subject (age 33) was selected, since most requirements are binary, and since one subject can execute multiple tests with larger amounts of steps, still allowing for statistical analysis of data.

First, the goal was to validate whether the framework was able to capture the difference in gait patterns of a subject while experiencing the effects of the leafspring in the inGAIT-AFO. For this purpose, the 2MWT was performed twice on a treadmill to minimize the effects of changing speed. During the first test, the inGAIT-AFO was used, but without the leafspring inserted. That means that the AFO was not providing any assistance to the subject. During the second test, the leafspring

Fig. 13: The full version 2 framework worn in an outdoor environment during validation testing

was inserted and adjusted to provide a stiffness of 0.4 Nm/kg. The subject was asked to indicate a comfortable walking speed and that speed (1.1 m/s) was kept constant for both tests.

It was hypothesized that when using the leafspring, there would be less plantar and dorsiflexion of the ankle due to the stiffness added by the spring. A second hypothesis was that there would be a significant increase in the push-off of the subject with the spring. However, this hypothesis might be proven wrong since the subject was able-bodied and the orthoses therefore might not achieve the intended effect. To test both of these hypotheses, independent sample T-tests ($\alpha$ = 0.05) were executed. Before executing the T-tests, Shapiro-Wilkes tests was applied to confirm normality ($\alpha$ = 0.05). Data was analyzed in MATLAB.

Secondly, the goal was to validate the ability of the framework to capture differences in walking patterns between a controlled environment (e.g. inside, on the treadmill), versus an uncontrolled environment (outside, overground). Therefore, the six minute walk test (6MWT) was executed twice. In both of these tests, the subject did not have the leafspring inserted. In the first test, the subject walked on a treadmill at a comfortable speed (1.1 m/s). In the second test, the subject walked overground at a speed that felt comfortable to her. It was hypothesized that there would be more variability and thus standard deviation in the walking overground in the ankle angles as the push-off force. This hypothesis was tested using a Levene's test ($\alpha$ = 0.05), which tests the homogeneity of variances between different test groups. Since Levene's test requires input data to be normal, a Shapiro-Wilkes test ($\alpha$ = 0.05) was performed on the data first to confirm normality.

Finally, the battery life of the framework was measured using a python script that logged the current time every ten minutes until the system ran out of battery, while a fully charged powerbank was connected and sensors were being read out using the interface. The touchscreen display remained on for the entire test.

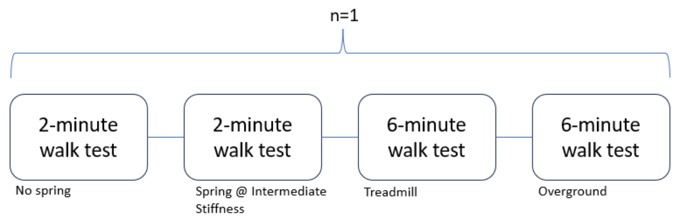


Fig. 14: Flowchart of the four different tests used for technical validation of the version 2 framework.

*b) Interface Usability:* the usability of the interface was tested with seven subjects ($n = 7$, mean age $\pm$ standard deviation $= 23.1 \pm 3.0$). All subjects were provided with the same instruction manual (Appendix E), explaining how to start up the system and record data through the interface. Subjects had the option to read the manual in Spanish or English, according to their preference. After reading the manual, subjects were asked to record five seconds of data in both the local and the ROS2 interface according to the steps in the manual without further instruction. They were observed during this process, and any mistakes were noted down by an observer. The system was not being worn by anyone while this data was recorded. After finishing these actions, subjects were asked to complete the system usability scale (SUS, Table IV). This scale includes ten statements pertaining to the usability of the interface [45]. Subjects were then asked to rank each statement by using a five-Likert scale (5 – completely agree, 1 – completely disagree).

In order to calculate the final SUS score, statement contributions were summed as explained by Brooke [45], meaning that for those statements of positive nature (Table IV), scores were converted with the formula:

$$contribution = score - 1$$

For statements of negative nature, scores were converted with the formula:

$$contribution = 5 - score$$

Summed contributions were then multiplied by 2.5 to obtain the overall system usability value in a range from 0 to 100. Finally, subjects were asked to provide any additional remarks or feedback on the interface.

*3) Results:*

*a) Technical:* since technical validation testing was performed with an able-bodied subject, only forces and angles from the left leg were analyzed since no difference is expected between performance of the left and right leg. The obtained forces and angles for the 2MWTs with and without spring (Fig. 15) demonstrate a lower degree of dorsiflexion and plantarflexion reached when using the spring than when not using the spring. This observation is supported by the results of the independent sample T-tests comparing the maximum plantarflexion and dorsiflexion per gait cycle between the spring and no spring conditions. Significantly lower plantar and dorsiflexion were found in this test when the spring was inserted versus when it was not inserted (Plantarflexion:

TABLE IV: SUS statements, including the abbreviation that was used to refer to each statement and whether they are considered positive (P), or negative (N) in nature.

| SUS Statement | Abbreviation | P/N |
|---|---|---|
| I think that I would like to use this system frequently | UseFreq | P |
| I found the system unnecessarily complex | Complex | N |
| I thought the system was easy to use | UseEasy | P |
| I think that I would need the support of a technical person to be able to use this system | TechSupport | N |
| I found the various functions in this system were well integrated | FuncIntegrated | P |
| I thought there was too much inconsistency in this system | Inconsistency | N |
| I would imagine that most people would learn to use this system very quickly | LearnQuickly | P |
| I found the system very cumbersome to use | Cumbersome | N |
| I felt very confident using the system | Confident | P |
| I needed to learn a lot of things before I could get going with this system | LearnThings | N |

$t = 32.84$, $p < 0.01$, Dorsiflexion: $t = 28.80$, $p < 0.01$), confirming the hypothesized effect. All data groups for aforementioned T-tests were normally distributed according to their respective Shapiro-Wilkes tests. Additionally, when the spring was inserted, the push-off force is visually lower than when the spring was not inserted (Fig. 15). This effect was found statistically significant ($t = 5.96$, $p < 0.01$), disproving the previously mentioned hypothesis.

When comparing the push-off force as well as the ankle angles between the 6MWT on the treadmill versus overground (Fig. 16), there is no visible difference in the standard deviation between the two trials, indicating that the subject was able to keep a similar gait pattern even when walking overground. No significant difference was found in the variance for treadmill and overground walking per Levene's test ($F = 3.45, p = 0.06$), while the initial hypothesis did suggest such a difference.

Battery life of the system was measured to be 3 hours and 30 minutes with the touchscreen display on and actively reading out data from the sensors.

*b) Interface Usability:* Results of the SUS questionnaire (Fig. 17), show that participants scored the system highly on ease-of-use and simplicity. On the other hand, subjects gave relatively worse scores regarding the first and fourth statement, indicating that they are less likely to want to use the system frequently, and more often believe that they would need the support of a technical person to use the system. The final SUS score was computed to be 89.6 out of 100. Four of the seven participants (57%) completed the steps without making mistakes. One participant did not calibrate the encoders before recording, leading to the recording button staying disabled, and was therefore not able to record data. Two different participants pressed the "Stop and Close All" button before stopping the recording with the "Stop Recording" button. However, due to the way the software is set up, recordings were still saved and closed out properly. The additional comments and feedback collected are summarized in the following points:

- The buttons were smaller than desired.
- Technical language (such as "ROS2") should be removed to avoid confusion.
- The ROS2 interface should include a way to see the data in real-time within the interface.
- The ROS2 and local interfaces could be integrated into one interface to avoid complexity.
- There should be a success message after recording data successfully to reassure users.

*c) Overall:* The version 2 framework was lightweight, portable and adjustable as well as easy to put on and take off per requirements GR-01, TR-CE-06, TR-CE-08. The sys-

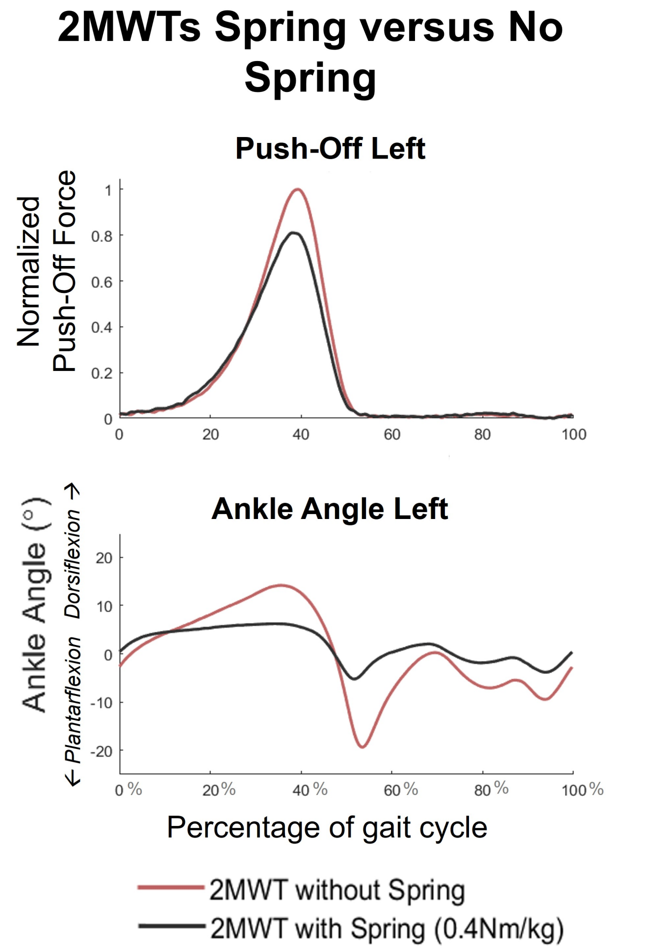


Fig. 15: Mean push-off force and ankle angles plotted for 2MWT on a treadmill without spring (red) and with spring (gray). Push-off force was normalized by dividing by the maximum of the mean gait cycles

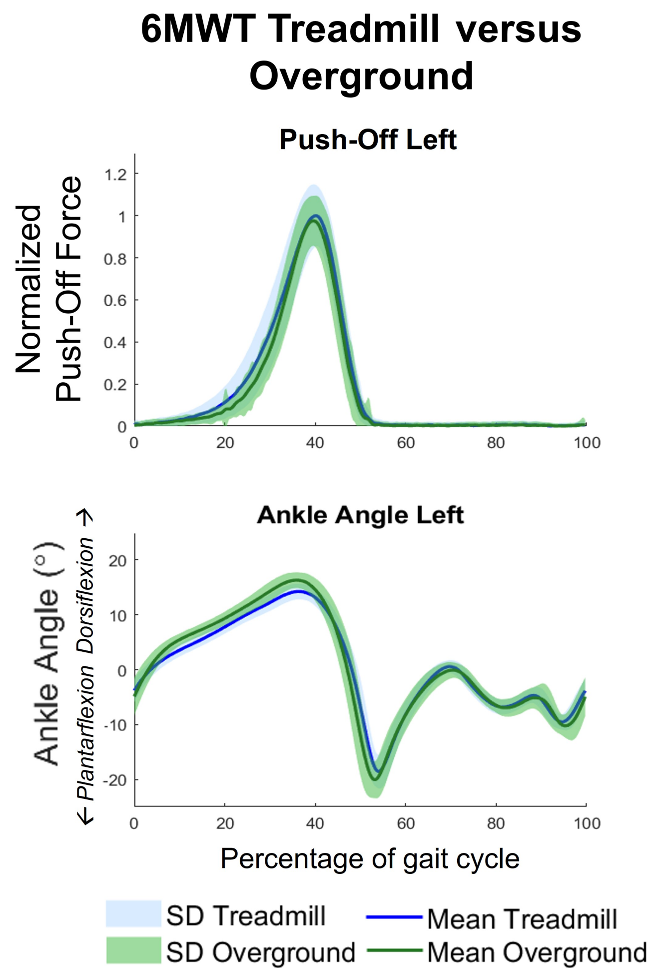


Fig. 16: Mean and standard deviation (SD) gait cycle for push-off force and ankle angles during 6MWT on both the treadmill (blue) and overground (green). Push-off force was normalized by dividing by the maximum of the mean gait cycles

tem worked both in a controlled environment as well as an uncontrolled environment, while risks to cause harm to users were reduced by using the system in a supervised setting and not exceeding a voltage supply of 5V (GR-03 through GR-05). The framework was capable of measuring or estimating heel strike events, push-off forces and ankle angles within their specified resolutions, satisfying requirements TR-S-03, TR-S-06 and TR-S-08 through TR-S-10. The battery life of the system was measured to be 3.5h, therefore failing to achieve requirements GR-06 and TR-PA-02. It does however allow charging with an external charger per TR-PA-01. The frequency at which data was recorded throughout all version 2 validation tests was 101.0Hz on average with a standard deviation of 3.8Hz, satisfying TR-S-01, while producing real-time metrics (TR-S-05). Recorded data was safely stored on-board and could be sent to a PC through wireless transmission (TR-S-02 and TR-S-07). The optional requirement TR-S-04 was not met since the framework does not measure the angle between the foot and ground. The device could be worn in conjunction with normal clothing, while allowing breathablity and not impeding the user (TR-CE-01, TR-CE-02, TR-CE-05). The system is easy to adjust and, according to the SUS score of 89.6 out of 100, also easy to use per requirement TR-MC-02. The user manual (Appendix E), as well as verbal explanation inform the user of the system's functionality, satisfying TR-MC-01. The mass of the electronics was higher than the 0.15Kg specified in TR-WP-01. Cleaning the exterior of the device is relatively easy by wiping it down (TR-CE-04). The device did not cause any skin irritation (TR-CE-07). Finally, the system allows for integration of external systems through wireless communication due to ROS2 being embedded in the system (TR-MC-04), while additional processing algorithms can be incorporated relatively easily using Python code (TR-MC-03).

Summarizing (see Table I), 28 of all 32 requirements (84%) were met in the version 2 framework. Of these, 18 out of 18 must-have requirements (100%), 7 out of 10 desired (70%) requirements and 2 out of 3 optional (67%). This signifies an increase of 24 percentage points with respect to the version 1 framework.

## III. Discussion

A wearable system capable of measuring the gait patterns of those using assistive devices for daily-life is indispensable in the assessment of the effects of such devices [14]. In this thesis, the inGAIT-AFO has been used as a case assistive device for the development of two versions of a sensor data capturing and recording framework, where the technical validation of the first version with respect to the pre-defined requirements paved the way for the second version of the framework. The final results discussed in this thesis imply that a usable, and valid sensor framework was developed for the measuring of the ankle angle and estimating of the push-off forces during gait. This framework can be utilized in the evaluation of the functionality of the inGAIT-AFO.

The latter technical validation showed an ability to measure the ankle angle during gait at a resolution of 0.0219° and a frequency of 100Hz, allowing the significant identification of differing gait patterns with the leafspring due to the added stiffness.

With respect to push-off force, significant differences were found when the system was tested at different walking speeds, which shows the capacity of the system to detect differences in gait patterns by the push-off forces. Concretely, higher walking speeds produced significantly higher push-off forces for all participants ($n = 5$), except for one. This one exception might be caused by inaccuracy of the FSR during this trial, which showed a relatively larger standard deviation. When testing the reliability of the system when measuring a different separated moments on the same subject, there was no significant difference found between normalized push-off data. This suggests that it is possible to compare tests between different days. Finally, it was not achieved to reliably translate the FSR output to a standardized unit. It could be speculated that a more advanced algorithm could be developed for calibration, such as one using machine learning similar to discussed in literature [33], [34]. However, since relative changes in force allow for analysis of the effects of the AFO, this was not

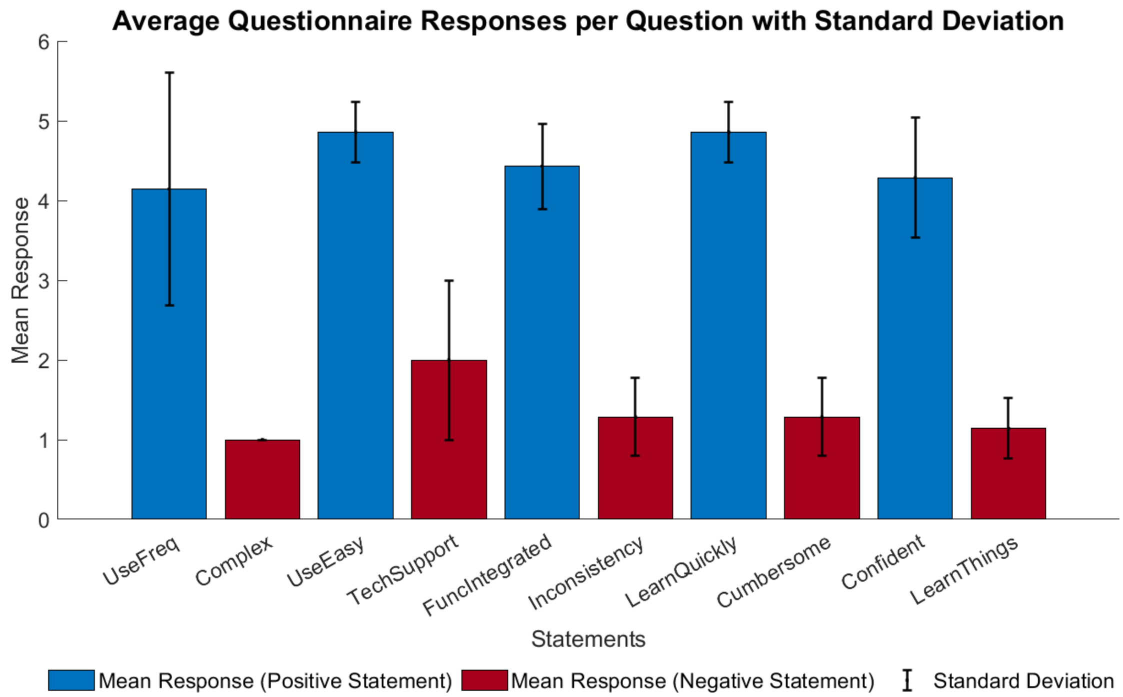


Fig. 17: Bar chart of SUS questionnaire statements with mean score and standard deviation per statement

deemed necessary within the scope of this thesis. Furthermore, in a similar project using the same FSR model, Conner *et al.* calibrated their FSRs relative to the user, and not to a standardized unit [40].

A GUI was also developed to allow the end-user controlling of the system from a touchscreen. This GUI was assessed for usability with 7 users, obtaining a usability score of 89.6 out of 100. Apart from functioning locally, the system included integration in ROS2, enabling real-time wireless communication of gait metrics. Conner *et al.* similarly developed a GUI, although in MATLAB, and additionally developed a smartphone app [40]. A smartphone app could be a useful addition to the framework to augment user experience, but is not necessary for the intended purposes of this thesis.

Overall, in the first version of the framework, 70% of all requirements were met, out of which 78% of the must-have requirements, 60% of the desired requirements and 67% of the optional requirements. The second version of the framework managed to meet 84% of all defined requirements, as well as meeting 100% of the must-have requirements, 70% of desired requirements and 67% of optional requirements.

The main advantage of the developed sensor/data framework is its portability and usability. Because of these qualities, it can be deployed to measure gait in different environments (including out of the lab) at a capturing frequency of 100Hz. Although the adequate sampling frequency depends on the desired outcome measures, a sampling frequency of at least 35Hz was found appropriate for gait analysis in the literature [46]. However, relevant studies often opt for a sampling frequency of 100Hz [22], [30], [40], [47]. Furthermore, the sensitivity of the selected sensors contributed to an accurate gait analysis, capable of identifying differences in ankle angles due to the added stiffness of the leafspring. This ability may allow the framework to be used for the selection of an appropriate stiffness level for an AFO while targeting a specific degree of plantar and dorsiflexion.

## *A. Limitations and future work*

The main shortcoming of the developed framework is that it still requires the user to wear a backpack with wires that may obstruct the user in activities of daily living making it less suitable for continuous measurement in an in-home environment. In the future, this could be partially resolved by making use of Bluetooth data transfer instead of by wire, such as employed by Conner *et al.* [40]. The space used by the device can also be reduced by designing a custom casing that can be strapped on the body (e.g. on the upper leg or hip), this would result in less hindrance to the user. A second potential limitation is the device's limited battery life of 3.5h. This can easily be accounted for by connecting a battery with a higher capacity. However, it would have the drawback of increasing the weight of the total system.

Certain caveats can be identified in the validations of the final sensor data framework. Firstly, the final framework was not extensively validated with children with CP, even though this is the intended user group of the AFO. Most validations were performed with a limited number of able participants, similar to various viability or pilot studies as reviewed in Lora-Millan *et al.* [17]. Nevertheless, the inclusion of multiple pediatric patient with CP would provide more valuable data. Another caveat is that during validation of the interface usability, most participants were technically educated and therefore may have had less difficulty understanding the interface. Ideally, the interface would be validated with a larger group of subjects of varying ages and backgrounds.

For the above-mentioned reasons, recommendations for future work include the implementing of the proposed changes to the framework to increase battery life and interface usability and reduce impedance caused by the framework. Additionally, the final system should be validated with children with CP to confirm its validity in this population.

## IV. Conclusion

In this thesis, the development and technical validation of a sensor data acquisition and processing framework was presented. The utilized sensors and the developed GUI were described and validation data were shown. Based on the validation of the second version of the framework, it can be concluded that the system is sufficiently capable of performing the tasks as specified in the requirements, since all must-have requirements were met. The presented framework is a part of the inGAIT project, and will be incorporated in the evaluation of the inGAIT-AFO by obtaining data from the AFO. Avenues for future work were recommended based on identified shortcomings.

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[48] Tekscan A502 FlexiForce Piezoresistive Force Sensor. Tekscan

# APPENDIX

## A. *Documentation*

All used scripts, code and other documentation is stored in the directory attached to this report. The directory contains a "README" file that explains the directory structure. Scripts and code contain comments explaining their functionality in detail, intended to allow external researchers to understand and customize the code.

## B. *Requirements*

All general requirements including their source or rationale can be found in Table B.I. The same information for the technical requirements is located in Table B.II

TABLE B.I: Overview of All General Requirements and their values in version 1 (V1) and version 2 (V2)

| TR ID | Source/Rationale | Description | Target value | Rank | Value V1 | Value V2 |
|---|---|---|---|---|---|---|
| GR-01 | inGAIT Requirements[1] | The device should be adjustable and valid for different pilot's sizes | True | M | True | True |
| GR-02 | inGAIT Requirements[1] | The device should be portable and lightweight | True | M | True | True |
| GR-03 | inGAIT Requirements[1] | Environment: indoor, in a controlled environment (research environment) | True | M | True | True |
| GR-04 | inGAIT Requirements[1] | Environment: outdoor, in an non-controlled environment (daily living) | True | D | True | True |
| GR-05 | inGAIT Requirements[1] | The device must not cause harm to the user | True | M | True | True |
| GR-06 | inGAIT Requirements[1] | The device should be able to be continuously used for a minimum of 10 hours (battery capacity for data recording) | 10h | D | 8h[2] | 3h |

TABLE B.II: Overview of All Technical Requirements and Their Values in V1 and V2

| TR ID | Source/Rationale | Description | Target value | Rank | Value V1 | Value V2 |
|---|---|---|---|---|---|---|
| TR-S-01 | inGAIT Requirements[1] | Sampling frequency of data reception of inGAIT | $\geq$ 50Hz | M | 13.28Hz ($\pm$1.05) | 101Hz ($\pm$3.8) |
| TR-S-02 | inGAIT Requirements[1] | The system should allow on-board data storing and wireless transmission to a PC for postprocessing | True | M | False | True |
| TR-S-03 | inGAIT Requirements[1] | The angle between foot and shank should be known | True | M | True | True |
| TR-S-04 | inGAIT Requirements[1] | The angle between foot and ground should be known | True | O | False | False |
| TR-S-05 | inGAIT Requirements[1] | The calculation of the different metrics should be done in real-time | True | O | True | True |
| TR-S-06 | inGAIT Requirements[1] | Resolution and accuracy of obtained ankle angle | $\leq 0.5°$ | D | 0.0219° | 0.0219° |
| TR-S-07 | inGAIT Requirements[1] | Safe storage of personal data | True | M | True | True |
| TR-S-08 | Allow for measuring of effects on push-off force | Capable of measuring force between forefoot and ground during gait on both sides | True | M | True | True |
| TR-S-09 | Allows for cutting of gait cycles in post-processing | The system must be able to detect the heel strike event on both sides | True | M | False | True |
| TR-S-10 | Ensure that data can be reliably interpreted | Resolution and accuracy of obtained forces between forefoot and ground | Significant difference at 30% change in walking speed | D | True | True |
| TR-CE-01 | inGAIT Requirements[1] | The device should be worn in conjunction with normal shoes and clothing | True | D | True | True |
| TR-CE-02 | inGAIT Requirements[1] | The device does not impede the existing functionality of the user | True | M | True | True |

Continued on next page

[1]The requirements for the inGAIT project [23]

[2]The powerbank used has a total capacity of 5000mAh, accounting for a suboptimal efficiency of 80%, this yields 4000mAh of effective capacity. Assuming a constant current draw of 500mA per hour, the battery duration is estimated to be 8 hours.

TABLE B.II – continued from previous page

| TR ID | Source/Rationale | Description | Target value | Rank | Value V1 | Value V2 |
|---|---|---|---|---|---|---|
| TR-CE-03 | inGAIT Requirements[1] | The maximum noise that may be generated by the system while walking | 65dB | D | True | True |
| TR-CE-04 | inGAIT Requirements[1] | The device exterior should be easy to clean | True | O | True | True |
| TR-CE-05 | inGAIT Requirements[1] | The device should allow breathability of the skin | True | D | True | True |
| TR-CE-06 | inGAIT Requirements[1] | The device should be adjusted and fitted to each subject | True | M | True | True |
| TR-CE-07 | inGAIT Requirements[1] | Skin pressure, friction, or abrasions should be avoided when using the device | True | M | True | True |
| TR-CE-08 | inGAIT Requirements[1] | The device should be easy to put on and take off | True | M | True | True |
| TR-WP-01 | inGAIT Requirements[1] | Mass of the electronics and sensing | ≤0.15kg | D | False | False |
| TR-MC-01 | inGAIT Requirements[1] | The user should be sufficiently informed about the operation and manipulation of the device | True | M | True | True |
| TR-MC-02 | inGAIT Requirements[1] | The system should be easy to use and easy to adjust | True | M | False | True |
| TR-MC-03 | inGAIT Requirements[1] | The integration of new processing algorithms into the code should not be an extremely time-consuming process | True | M | True | True |
| TR-MC-04 | Allow integration with external systems | The system should allow integration with external systems through real-time wireless communication | True | D | False | True |
| TR-PA-01 | inGAIT Requirements[1] | The system can be charged using an external charger | True | M | True | True |
| TR-PA-02 | inGAIT Requirements[1] | The device should be able to be continuously used untethered for a minimum of 10 hours (battery capacity for data recording) | ≥10h | D | 8h[2] | 3.5h |

### C. Protocol pilot testing

For pilot testing with the version 1 framework, the following protocol was used:

- Tests were executed at the Niño Jesus Hospital in Madrid for subjects PC01 and PC02.
- The protocol consisted of two sessions, see Fig. C.1 and C.2 respectively for detailed flowcharts of each session.

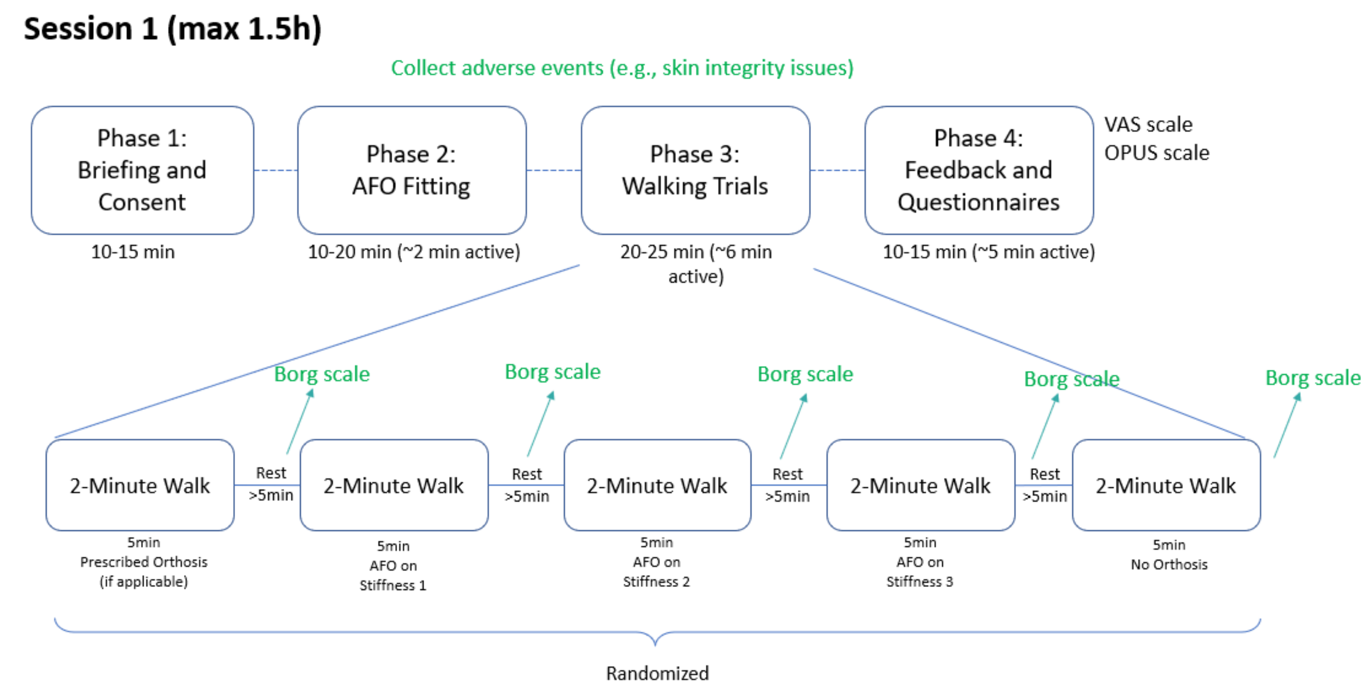



Fig. C.1: Flowchart of the first session of the pilot testing

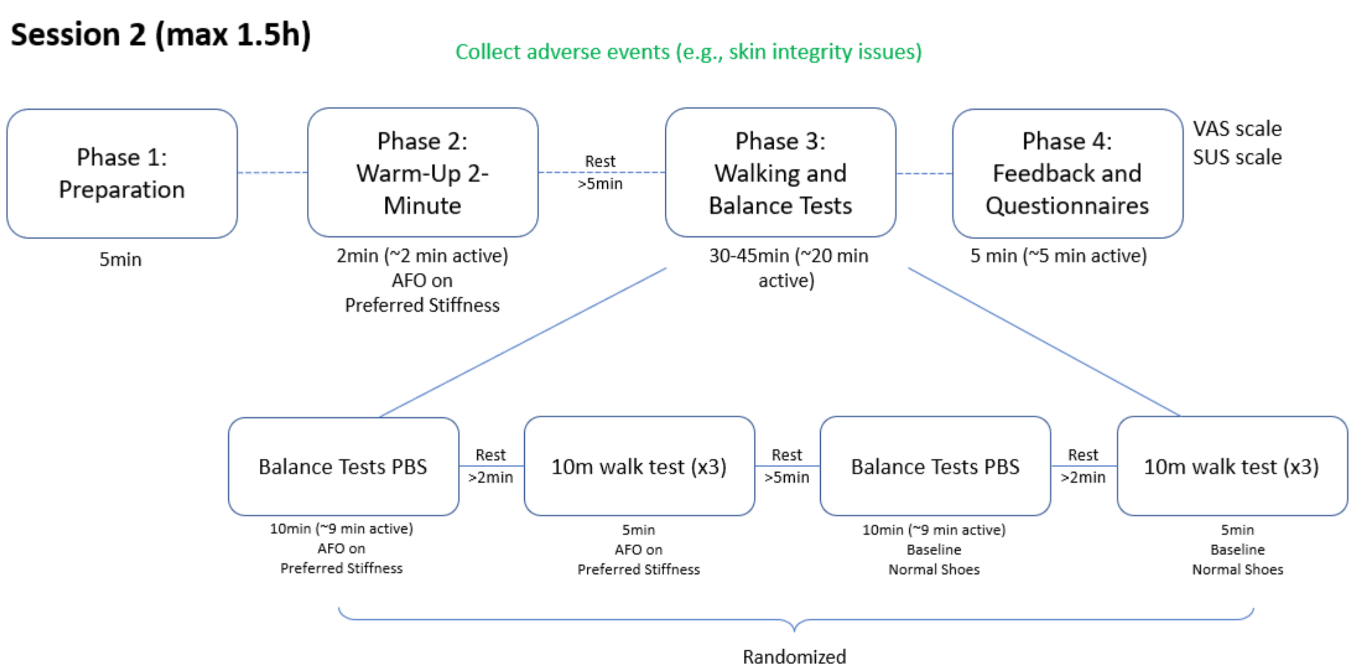



Fig. C.2: Flowchart of the second session of the pilot testing

### D. FSR circuit

An electrical circuit (Fig. D.1) was used to obtain a usable signal as well as providing the FSRs with a steady reference signal, this circuit was derived by Centeno [41] from the manufacturer's recommendation [48].

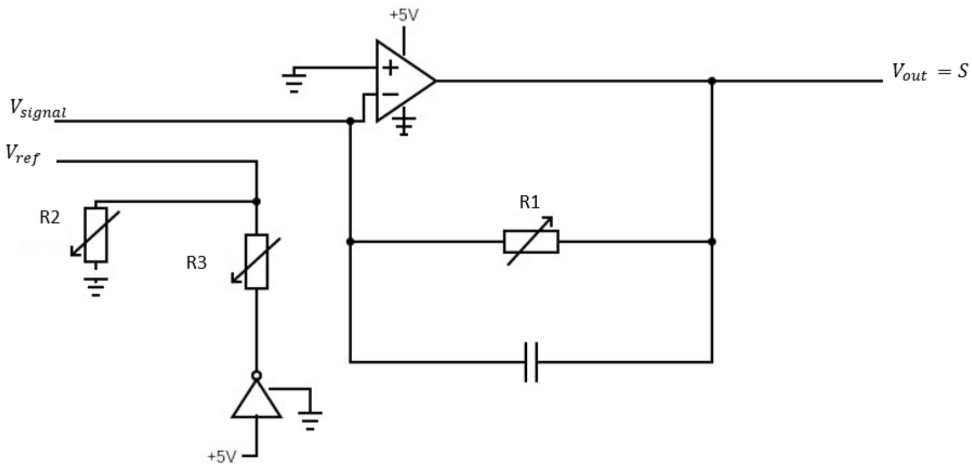



Fig. D.1: Electrical circuit used to obtain FSR signal (S)

The circuit provides a reference voltage ($V_{ref}$) to the FSR, and simultaneously amplifies the FSR signal ($V_{signal}$), to obtain the output signal ($S$).

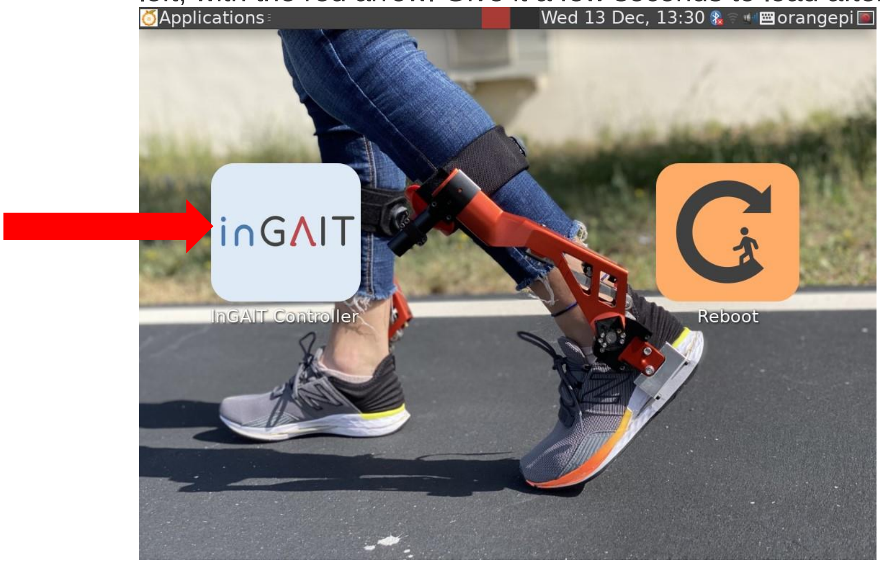
### E. Instruction manual user interface

Below can be found the instruction manual for the developed GUI. This instruction manual was used in the usability validation for the version 2 framework.

**Instruction manual inGAIT sensor framework**

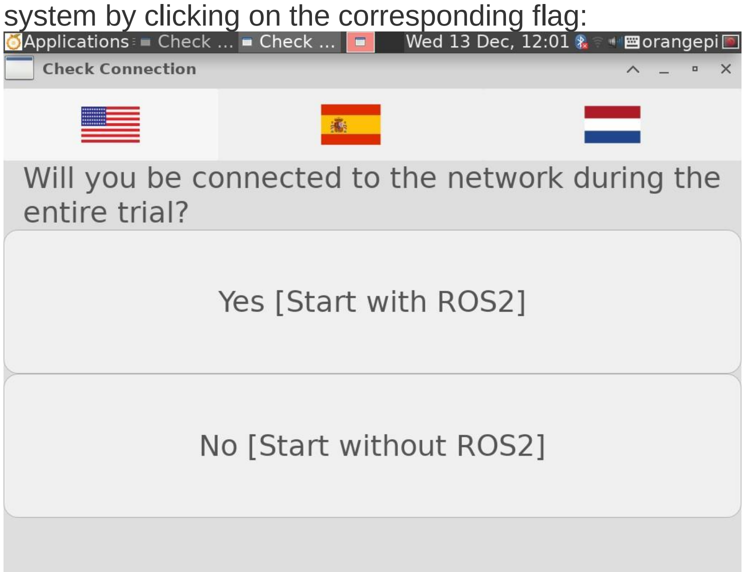
## STEP 1

When turning on the system, you will see the following screen, please click the button on the left, with the red arrow. Give it a few seconds to load after pressing the button:



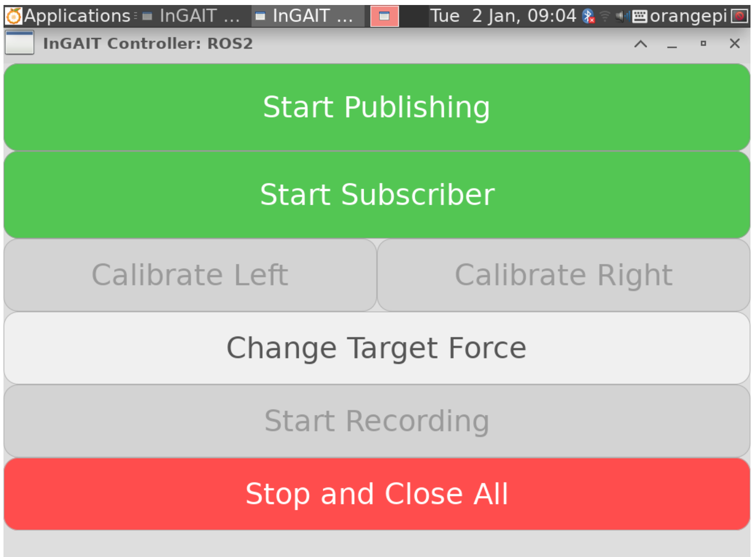

## STEP 2

This will open the following menu, please select the language in which you would like to use the system by clicking on the corresponding flag:



**STEP 3**

Now, to start the system **with** ROS2 functionality, click the button "Yes [Start with ROS2]". Note: to use the system with ROS2, you will need to be connected to the same Wi-Fi network the entire time. If you would like to start the system without ROS2 please skip to the following step. After clicking the button "Yes [Start with ROS2]", the following menu will be opened:



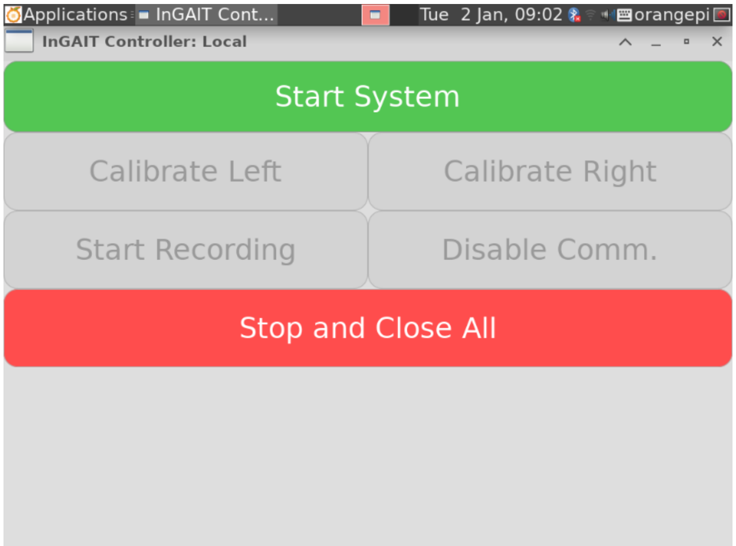

To start the system, please click the button "Start Publishing". The button should now turn orange. Note: do not press the "Start Subscriber" or "Change Target Force" button. Before we can record data, we should calibrate both encoders. For this, press the button "Calibrate Left" for the left encoder and "Calibrate Right" for the right encoder. Please wait until both buttons are green, this may take a few seconds.

Once calibration is completed, we are ready to start recording. Press "Start Recording" to start the recording. The button should now turn dark grey. Let the system record. Now press "Stop Recording", to stop recording.

Now, stop the publisher by pressing "Stop Publishing". To close the software, click "Stop and Close All'.

**STEP 4**

To start the system **without** ROS2 functionality, select "No [Start without ROS2]", which will open the following menu:




Now start the system by pressing “Start System”. This button should now turn orange. We should now also be able to see the current values of the sensors at the bottom. Check to make sure that none of the sensor values show “Error” in red.

Before recording, we should calibrate the encoders by pressing the button “Calibrate Left” and “Calibrate Right” for each respective encoder. Before recording, it is also recommended to disable the sensor communication to the interface for better data quality. For this, press the button “Disable Comm.”. You should now no longer be able to see sensor values at the bottom.

Once the calibration is completed and the sensor communication is switched off, we can start recording. Press “Start Recording” to start the recording. The button should now turn dark grey. Let the system record. Now stop the recording by pressing the “Stop Recording” button. If you would like to re-enable the communication with the sensors, press the “Enable Comm.” button. Now, click the button “Stop System”, to stop the system. To close the interface, press “Stop and Close All”.

**STEP 5**

Please rate the user experience related to the interface by selecting the answer that best reflects your opinion:

| | | Completely agree | Agree | Don’t agree, don’t disagree | Disagree | Completely disagree | I don’t know/not applicable |
|---|---|---|---|---|---|---|---|
| 1. | I think that I would like to use this system frequently. | o | o | o | o | o | o |
| 2. | I found the system unnecessarily complex. | o | o | o | o | o | o |
| 3. | I thought the system was easy to use. | o | o | o | o | o | o |
| 4. | I think that I would need the support of a technical person to be able to use this system. | o | o | o | o | o | o |
| 5. | I found the various functions in this system were well integrated. | o | o | o | o | o | o |
| 6. | I thought there was too much inconsistency in this system. | o | o | o | o | o | o |
| 7. | I would imagine that most people would learn to use this system very quickly. | o | o | o | o | o | o |
| 8. | I found the system very cumbersome to use. | o | o | o | o | o | o |
| 9. | I felt very confident using the system. | o | o | o | o | o | o |
| 10. | I needed to learn a lot of things before I could get going with this system. | o | o | o | o | o | o |