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## A non-invasive wearable sensory leg neuroprosthesis: mechanical, electrical and functional validation

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E-mail: [stanisa.raspopovic@hest.ethz.ch](mailto:stanisa.raspopovic@hest.ethz.ch)**Keywords:** sensory feedback, wearable sensors, electro-cutaneous stimulation, amputees, Cybathlon, ProsthesesSupplementary material for this article is available [online](#)**Abstract**

**Objective.** Lower limb amputees suffer from a variety of functional deficits related to the absence of sensory communication between the central nervous system and the lost extremity. Indeed, they experience high risk of falls, asymmetric walking and balance, and low prosthesis embodiment, that significantly decrease their quality of life. Presently, there are no commercially available devices able to provide sensory feedback to leg amputees but recently some invasive solutions (i.e. requiring surgery) have been proposed by different research groups. However, a non-invasive effective alternative exploitable in everyday life is still missing. **Approach.** To address this need we developed and tested a lightweight, non-invasive, wearable technology (NeuroLegs) providing sensory (i.e. knee angle joint and tactile) feedback to the users through electro-cutaneous stimulation. Standard mechanical and electrical tests were performed to assess the safety and reliability of the technology. The NeuroLegs system was verified in terms of accuracy in measuring relevant gait parameters in healthy participants. The effectiveness of the NeuroLegs system at improving walking of three transfemoral amputees was then verified in movement laboratory tests. **Main results.** No mechanical failures, stable communication among system's parts and a long-lasting battery were demonstrated. A high temporal reliability was found when detecting stride features (important for the real-time configuration) with a correct match to the walking cadence in all assessed walking conditions. Finally, transfemoral amputees showed increased temporal gait symmetry and augmented confidence when walking with the sensory feedback compared to no feedback condition. Stepping outside from the lab, NeuroLegs was successfully exploited by a transfemoral amputee in CYBATHLON Global Edition 2020 in several challenging situations related to daily-living activities. **Significance.** Our results demonstrate that the NeuroLegs system provides the user with useful sensory information that can be successfully exploited in different walking conditions of daily life.

### 1. Introduction

Most balance and walking problems related to lower-limb amputation (LLA) are caused by the absence of proprioceptive and tactile sensations from the missing extremity [1, 2]. This lack of sensory feedback results in amputees facing many difficulties during their everyday life and leads to abandonment of the prosthesis [3]. They suffer mainly from slow walking [4], risk of dangerous falls [5], highly asymmetric

and irregular walking pattern and consequent healthy leg overloading, back pain, osteoarthritis [6], increase in cognitive burden [7] and metabolic cost together with a 120% higher probability of heart attack [8]. Moreover, 50%–80% of amputees are affected by phantom limb pain [9] also caused by the lack of information coming from the periphery [10]. In the recent past, very sophisticated prostheses have been developed for restoring motor function [11–13] but they cannot provide the same benefits as prosthetic

designs able to restore missing sensory information and improve walking pattern and cognitive burden [14]. Invasive (i.e. requiring surgery) approaches were developed both for transtibial [2, 15, 16] and transfemoral [12, 17–19] amputees. Notably, a second surgical intervention is not a desired option for many amputees. Indeed, the need for an alternative, cheap and easy-to-use design guided the development of non-invasive technologies to restore sensory feedback [20].

Different feedback modalities including visual [21–24] and auditory [25–28] feedback have been the most commonly implemented approaches. Haptic feedback, often in the form of vibrotactile feedback [29–36], was recently presented resulting in lighter, more portable devices compared to cumbersome and heavy computer monitors and speakers. In fact, Khajuria *et al* [32] developed a prototype to provide vibratory feedback in response to fluctuations of the center of pressure outside a predetermined zone. The use of electro-cutaneous stimulation represents another efficient stimulation modality. Indeed, Dietrich *et al* [37] recently presented a simple feedback system consisting of three pressure switches attached under the prosthetic foot. Switch closures are used for gait phase detection to determine the timing of the electro-cutaneous stimuli at the patient's thigh. Stimulation intensity and duration was fixed and limited to tactile feedback. The majority of the above-mentioned devices only induces a stimulation which is discrete (only contact/no-contact information), without providing information about the intensity of the pressure (intensity modulation [38]) applied over the ground. More recently, Rusaw *et al* [30] presented transtibial amputees with vibrotactile feedback proportional to the pressure applied on four force sensors placed under the prosthetic foot. The benefits in weight distribution symmetry, step length symmetry and standing balance following a stimulation proportional to the strength of the applied pressure was shown for both transtibial and transfemoral amputees by Sabolich *et al* [39].

None of the studies mentioned up to this point took proprioceptive feedback (i.e. feedback about the joints' angles and their movements) into consideration. Transfemoral amputees experience a higher mobility reduction with respect to transtibial amputees [4] due in part to missing knee joint position awareness [40]. Compromised proprioceptive information forces transfemoral amputees to decide on their walking strategy relying only on visual inputs [41] which inevitably increases their cognitive burden. Moreover, whenever vision is blocked, the missing sensory information drastically reduces balance performance and increases sway displacement [42] causing instability when performing walking, balance and stairs tasks. Knee joint

angle feedback contributes to a safe use of the prosthesis [43] and proprioceptive information restored through implanted microelectrodes sutured to the sciatic nerve was shown by Clippinger *et al* [44] to be effective in enhancing confidence when walking in the dark, up- and down-stairs and on uneven surfaces. Knee joint angle feedback was provided by Pagel *et al* [45] using a non-invasive technology, whose efficacy in improving amputees' walking pattern is still not proven. Notably, this study presented the use of knee joint feedback as a standalone condition. Importantly, the restoration of both tactile and emulated knee angle proprioceptive information combined was demonstrated to be crucial to improve the amputees' walking abilities [14, 18, 46].

With this in mind, we designed and developed a lightweight, non-invasive and wearable technology (called NeuroLegs system, figure 1) to provide both tactile and knee joint angle feedback to lower limb transfemoral amputees. This system is an add-on for commercially available leg prosthetics without modifying them. Electro-cutaneous stimulation is provided in real-time with no perceivable delay (<50 ms) [47] using surface skin electrodes placed under the socket of the prosthesis at the thigh level. Two user-friendly and easy-to-use Graphical User Interfaces (GUIs) were developed for user control of the NeuroLegs system. Here, we present the testing, verification and the exploitation of this non-invasive sensory feedback system. Firstly, we assessed the mechanical and electrical properties of NeuroLegs in order to be compliant to well-established standard regulations (IEC 60601-International Electrotechnical Commission standards for safety and essential performance of medical electrical equipment). Then, the system accuracy in estimating the knee angle and the vertical forces applied over the prosthetic foot was evaluated with healthy subjects using a GRAIL system (Motek Medical B.V., Netherlands). The benefits of the NeuroLegs system were subsequently verified in three transfemoral amputees through a series of controlled walking tasks. Psychometric tests to characterize the artificial sensations elicited by electro-cutaneous stimulation were conducted using a purposely designed platform [48]. Finally, the efficacy of the NeuroLegs device was put to the test against international competition at CYBATHLON, 'a unique championship in which people with physical disabilities compete against each other to complete everyday tasks using state-of-the-art technical assistance systems' [49]. The NeuroLegs device was used by the NeuroLegs Team at three different CYBATHLON events and most recently came in 2nd place at CYBATHLON Global Edition (CGE) 2020 [50].

This study is intended as a first insight into the possibility of extending the usability of NeuroLegs system in home and ecological environments.



**Figure 1.** NeuroLegs system. Lower limb amputees wear a fully portable system to restore sensory feedback. The system can be easily integrated to the users' prostheses (wear time < 2 min). It comprises a custom-made sensorized insole and two electronic circuits wrapped around his/her ankle and thigh. Knee angle and tactile data (i.e. knee flexion, foot pressure intensity and location) are collected and transmitted via Bluetooth to a system controller which transduces the acquired data into parameters of stimulation. A portable stimulator is used to provide the subject with modulated sensory feedback using surface skin electrodes attached to the stump of the patient using an easy-to-wear elastic belt for easy positioning. The position of the stump resembles that of the foot (frontal electrode is associated with pressure on the toe, lateral electrode is associated with pressure on the middle part of the foot and posterior electrode is associated with pressure on the heel). The elastic belt is placed under the liner of the user and no discomfort was reported by the user.

## 2. Methods

### 2.1. NeuroLegs system

The NeuroLegs system was used in the study. The fully portable device was designed to provide real-time restoration of sensory feedback and to be used while performing activities of daily living in a variety of different settings, both indoors and outdoors. The system can be easily adapted to each user's prostheses (figure 1, movie S1 available online at [stacks.iop.org/JNE/19/016008/mmedia](https://stacks.iop.org/JNE/19/016008/mmedia)). It is comprised of a custom-made sensorized insole (three fabric and one latex layers) with seven force sensors (FlexiForce A301, Tekscan, US) distributed on the

foot sole in optimized positions to measure the force exerted by the subject on his prosthetic foot and record pressure information; two electronic circuits condition the force sensors, extract inertial data to measure knee flexion, sample the force data and send it via Bluetooth to a system-controller; the system-controller (ODROID-C2, Hardkernel Co., South Korea) manages the encoding algorithm and transduces the acquired data into parameters of stimulation; a portable stimulator (RehaMove 3, Hasomed, Germany) provides a modulated sensory feedback to the user through Transcutaneous Electrical Nerve Stimulator and surface skin electrodes (Circular Electrode Pads 25 mm, TensCare, UK). Three electrodes

are placed over the stump (front, lateral and back) and used to deliver tactile feedback and one electrode is placed at the hip and used to deliver emulated knee angle proprioceptive feedback. The position of the electrodes was chosen to be easily accessible, to intuitively resemble the location of the pressure under the foot and to facilitate the differential mapping of the sensations. As a reference, the electrodes were placed at half length of the residual stump above the socket. A pair of electrodes was placed below the iliac crest on the frontal plane, one pair below the femoral head on the sagittal plane, one pair below the gluteus and one pair at the level of the iliac crest, on the sagittal plane. All pairs of electrodes were placed perpendicular to the longitudinal axis. Being the sensations remapped, high accuracy of classification tests was obtained for multiple electrodes' placement. Each couple of electrodes was placed at least 5 cm apart to avoid spatial overlap of the elicited sensations. This ensured a correct differentiation among the stimulations' location even after a slight increase of the extent of the area of the elicited sensations when the injected current was rising [51] and no errors in classifying the stimulation locations were found. More specific NeuroLegs system characteristics are summarized in table S1 (see supplementary material).

Data is transmitted from the insole to the stimulator with a latency of less than 50 ms to allow the subjects to perceive a real-time stimulation that perfectly matches any of his steps. Therefore, the users can use this information to decide and optimize the walking strategy.

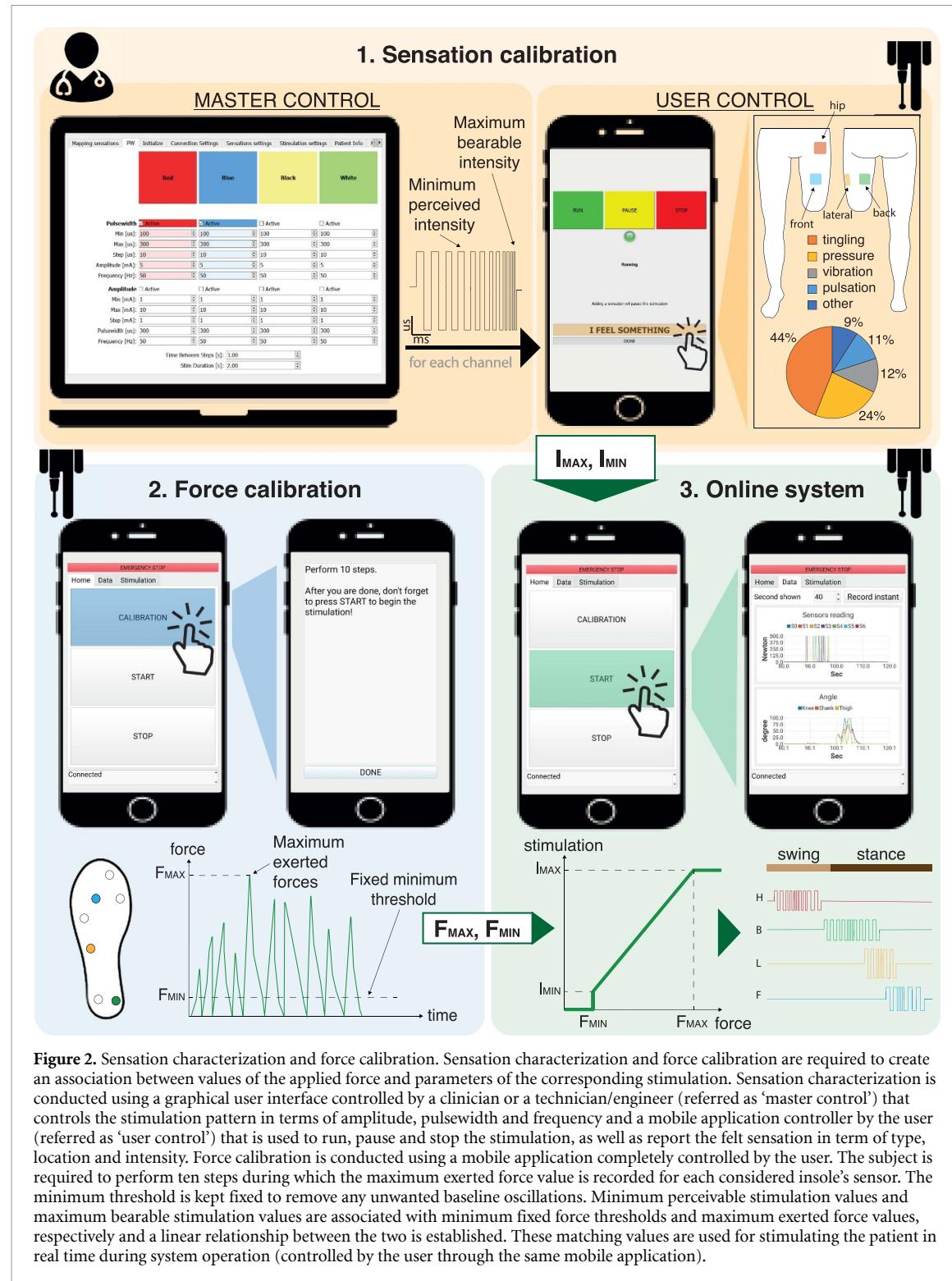
The readouts of three of the insole's force sensors are used to drive the tactile stimulation which is modulated in terms of intensity and location of the pressure. The intensity provides the subject with an electrotactile stimulation proportional to the applied force, the location informs the patient on which portion of the foot the force is applied. The algorithm that controls the stimulation is implemented on the ODROID-C2 and is written in C++. The intensity of the stimulation is obtained as a linear mapping of the exerted force as described in [18] and a minimum force is required to start the stimulation while a maximum threshold is set to avoid any painful sensation in the subject. Stimulation intensity is subject-dependent and, therefore, minimum (perceptual) and maximum (below pain) threshold are investigated for each participant in an initial calibration phase (see section 2.2). The intensity is modulated in pulselength, the frequency is kept constant at 80 Hz and the amplitude is determined during the calibration phase. Examples of perceptual and maximum thresholds for three amputee subjects are presented in the supplementary material (table S2).

In addition to the force under the foot sole, the subject is provided with emulated knee angle proprioceptive feedback. NeuroLegs device delivers

a stimulation that reflects the degree of knee flexion with a higher intensity the more the knee is bent. The algorithm that calculates the knee angle is based on a complementary filter [52]. Each electronic board embeds an inertial measurement unit (IMU) that provides the acceleration and the angular rate of the corresponding leg's segment in 3 mutually perpendicular axes. Since the local coordinate axes of the IMUs are independent and not aligned with any external meaningful axis, an initial calibration has to be performed in order to understand the relative position of the two IMUs. Static postures are used for this purpose, simply asking the subject to stand still with his legs straight for the first 10 s after the software booting. In this way, the relative angle between the two IMUs is calculated and used as an offset in the estimation of the instantaneous knee angle.

## 2.2. Sensation characterization and force calibration

In order to create an association between intensity of pressure applied on the sensorized insole and levels of perceived sensation, a calibration process is required before starting using the device. The procedure is similar to the one described in [18] and it is illustrated in figure 2. Sensation characterization is conducted using a graphical user interface controlled by a clinician or a technician/engineer (referred as 'master control') and a mobile application controller by the user (referred as 'user control'). The master controls the stimulation pattern which consists of biphasic, rectangular stimulation pulses in terms of amplitude (1–10 mA), pulselength (100–300 s) and frequency (50 Hz). Each pulse train lasts for 2 s with a 1 s pause between two consecutive pulses. The software implementation automatically increases amplitude and pulselength of the single pulses from the minimum to the maximum possible value by a step that is specified by the master. On the user side, the amputee runs, pauses and stops the stimulation, as well as reports the felt sensation in terms of type (vibration, pressure, tingling, warm, electricity, etc), location (front, lateral, back, hip) and intensity (1–10) in order to characterize the sensation [48]. The subject is required to report the lowest stimulus that can be reliably perceived (perceptual threshold) and the highest bearable stimulus (below pain). This procedure is repeated 3 times for each channel and the mean values are considered and used for the functional experiments. All the inserted sensations and the corresponding stimulation parameters are automatically recorded by the GUI. Sensation calibration (i.e. current amplitude and minimum and maximum pulselength) of all 4 channels can be successfully performed within 10 min. The 'master' intervention is required the very first time the subject uses the device. For subsequent uses, a mobile application was developed to enable the user to perform at



**Figure 2.** Sensation characterization and force calibration. Sensation characterization and force calibration are required to create an association between values of the applied force and parameters of the corresponding stimulation. Sensation characterization is conducted using a graphical user interface controlled by a clinician or a technician/engineer (referred as ‘master control’) that controls the stimulation pattern in terms of amplitude, pulsewidth and frequency and a mobile application controller by the user (referred as ‘user control’) that is used to run, pause and stop the stimulation, as well as report the felt sensation in term of type, location and intensity. Force calibration is conducted using a mobile application completely controlled by the user. The subject is required to perform ten steps during which the maximum exerted force value is recorded for each considered insole’s sensor. The minimum threshold is kept fixed to remove any unwanted baseline oscillations. Minimum perceptible stimulation values and maximum bearable stimulation values are associated with minimum fixed force thresholds and maximum exerted force values, respectively and a linear relationship between the two is established. These matching values are used for stimulating the patient in real time during system operation (controlled by the user through the same mobile application).

any time a fine tuning of the stimulation parameters within pre-set limits (figure S1). In particular, the user can increase and decrease the pre-set values of minimum and maximum thresholds by 10% (limited for safety reasons). In this way, stimulation parameters can be adjusted to the different conditions (e.g. electrode re-positioning, environmental factors, habituation, etc) that influence how the stimulation is perceived. The calibration is stable and repeatable enough that recalibration is not needed between

motor tasks performed over 3–4 h. When the pre-set limits become too restrictive (i.e. increasing or decreasing the stimulation intensity by 10% is not sufficient), the intervention of the ‘master’ is required.

The same mobile application was also designed for force calibration and control (start and stop) of the NeuroLegs system during operation. It is completely controlled by the user. After pressing the calibration button, the subject is required to perform ten steps during which the maximum exerted force

value is recorded for each considered insole's sensor. The minimum threshold is kept fixed to remove any unwanted baseline oscillations. Knee angle minimum and maximum thresholds are kept fixed to 10° and 55°, respectively [18]. The user is required to repeat the force calibration procedure any time the insole is removed but the process takes less than 1 min to be completed.

After completing the sensation characterization and force calibration phases, instantaneous force values are converted into the corresponding values of stimulation using a real-time, sensory encoding function which creates a linear relationship between the two variables. The user can switch between start and stop function using the same mobile application multiple times without repeating the calibration. In addition to the power button of the stimulator that is easily accessible, we implemented an emergency button via software on the mobile application that turns off the entire system.

### 2.3. Mechanical and electrical verification

The study aims at describing the testing procedure that was implemented to evaluate the mechanical and electrical safety of our technology. All performed tests are listed in figure 3. All commercially available used devices (battery, system controller and stimulator) follow the rules of EU conformity. Therefore, our analysis focused on the electronic boards and the 3D printed plastic case specially designed to enclose the circuits. The electronic boards are battery powered, and completely insulated from the power grid.

Since walking is a cyclic movement, NeuroLegs system's electronic circuits are subjected to continuous vibration stresses. A function generator (AFG 3021B, Tektronix, US), a vibration exciter (Type 4809, Brüel & Kjaer, Denmark) and an amplifier (6301B, Fostex, Japan) were used to generate cyclic vibrations. The amplitude of these vibrations was chosen to impose an acceleration higher than the maximum acceleration during walking [53] and the continuity of proper data acquisition and transmission was tested. The system was subjected to 500 000 cycles [54] at 20 Hz (figure 3(A)) during which signal transmission from the electronic circuits to the system controller and signal stability between the sensorized insole and the electronic circuits were checked every 70 000 cycles. The absence of mechanical failures was similarly tested, applying a linear sweep between 0.1 and 10 Hz at 0.11 Hz s<sup>-1</sup> (figure 3(A)) to evaluate the system functioning at double the frequency reached by prosthetic runners [55]. The absence of excessive undesired oscillations was assessed during the entire test to rule out the presence of resonance frequency of the system in the considered range.

No moving parts are embedded in the NeuroLegs system. To evaluate mechanical resistance of the plastic case that encloses the electronic circuits and the battery, an increasing static load was applied

along the weakest axis of the case until reaching the breaking point. An Advanced Force Gauge (AFG 1000, Mecmesin, UK) was used for the purpose of this test (figure 3(A)).

According to IEC60601-1 international standard for medical electrical equipment, portable devices must withstand drop tests from a height of 5 cm above a hardwood board. The experimenter lifted and dropped the plastic cases containing the electronics as required for three times at each of the relevant orientations. Yet, we additionally performed extra drops from a 1 m high table to simulate possible daily life scenarios. This benchmark testing ensures no failure to stresses caused by a free fall (figure 3(A)). Signal acquisition and transmission was evaluated before and after each drop.

A temperature less than 43 °C (109 °F) is required for applied parts made of plastic having contact with patients for more than 10 min (IEC60601-1). Temperature of the plastic case and the internal electronic boards was tested with a multimeter (U1233A, KEYSIGHT, US) and thermocouple (U1186A, KEYSIGHT, US) after 18 h of continuous working.

Battery duration was evaluated during continuous signal acquisition and transmission to the system controller. The system was kept in active mode for the entire duration of the test, preventing it from running in standby mode. In addition, packet loss rate was computed as the number of packets that were erroneously or not received over the total number of packets sent. Transmission between the two electronic circuits and between one electronic circuit and the system controller was determined via software implementation. Finally, communication delay was computed as the delay introduced during data acquisition, transmission and processing. Delay accumulated over the Bluetooth transmission between one electronic board and the system controller and over the transmission between the system controller and the stimulator. The former was computed after a synchronization process between a 3D motion capture system (Bonita B10, VICON, UK) and the NeuroLegs system; the latter was measured with an oscilloscope (UTD2052CEX, UNI-T, China).

### 2.4. System verification

#### 2.4.1. Experimental data

Three healthy subjects (two male, one female, age 25.67 ± 2.89 years) were recruited for the purpose of system verification. Characteristics of the included healthy participants are shown in table 1. Participants were chosen to be representative of a large pool of users in terms of anthropometric measurements (height = 177.67 ± 20.26 cm; weight = 75.67 ± 28.29 kg). All of them reported no known gait abnormalities or other diagnosed locomotor disorders. The experiment was approved by the ETH Zürich ethics commission (EK 2019-N-97) and all the subjects read and signed the informed consent.

**A Durability test**

**B**

Test	Requirements	Context	Method	Results
<b>Mechanical tests</b>				
Durability	No mechanical failure during 500 000 oscillations at 2g	Maximum tibial acceleration during walking is 2g	Vibrate the system using a function generator, a vibration exciter and an amplifier	Electronic circuit and insole transmit after 500 000 bumps at 20 Hz (signal checked every 70 000 cycles)
Resonation frequency	No mechanical failure or excessive oscillations when applying a linear sweep between 0.1 and 10 Hz at 0.11 Hz/s	Prosthetic runners reach a frequency of 4.1Hz on prosthetic leg	Vibrate the system using a function generator, a vibration exciter and an amplifier	No resonation frequency in the range 0.1 - 10 Hz
Static load	No plastic case breakage when applying a static force lower than 1.23 kN	Lower limb prosthetics tested by loaded cyclically up to 1.23 kN	Apply an increasing load with an Advanced Force Gauge until it starts cracking	The 3D box that encases the electronic circuit cracked at 1.47 kN on the vertical axis
Drop	No mechanical failure after dropping the system 3 times from 1 m	System can survive falls as a result of daily living	Drop the system from a 1m height 3 times, testing between drops	Electronic circuit and insole transmit before and after every drop
<b>Electrical tests</b>				
Allowable temperature	T < 43 °C for applied part made of plastic having contact with patient for more than 10 min	Safe for use near a person	Measure temperature of the PCB using a multimeter and temperature probes	The measured temperature after 18 hours is 30 °C
Battery duration	The battery last for at least 18 hours	84 % of persons with LLA wears the prosthesis for $12.47 \pm 4.34$ hours per day	Leave the system on and transmits signal until the power runs out	The system turns off after 23 hours of continuous operation
Packet loss rate	The number of lost packets during communication is limited	BLE retransmission demonstrated a packet loss rate in the order of $10^{-5}$ ; wireless technology showed a packet loss rate of $10^{-2}$	The number of packets lost are counted sending an incremental number	1 packet lost in 23 hours (1/4 mln) between ODROID and MYLEG; 1 packet lost over 3 hours (1 / 500 k) between the ankle and thigh boxes
Communication delay	The delay introduced during online data processing is not perceivable to the user	Delays of 100 – 125 ms are imperceptible to the user	The delay of the Bluetooth is calculated synchronizing ODROID and a motion capture system; the delay between ODROID and the stimulator is measure with an oscilloscope	Mean delay over Bluetooth = 40 ms + delay between Odroid and stimulator = 9 ms

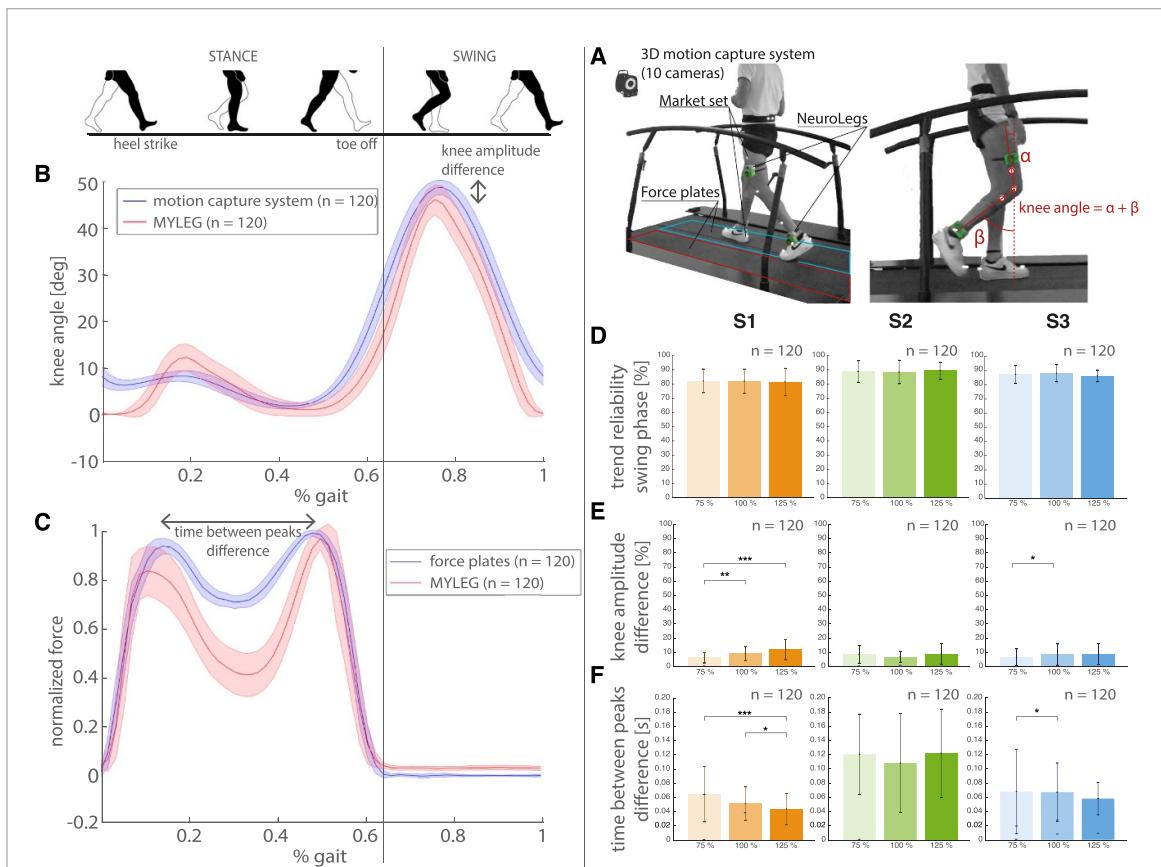
**Figure 3.** Mechanical and electrical tests. (A) To evaluate the mechanical safety and reliability, the NeuroLegs system's technology was tested against periodic oscillatory movements, applied static load and fall from standard height. (B) Additionally, electrical tests were performed to evaluate overheating, battery duration, packet loss rate and communication delay. For each considered test, the investigated requirement and the rationale supporting its implementation are described in the first and second columns of the table, respectively. Methods and materials used to perform the test are reported in the third column. Test outcomes are briefly described in the fourth column. For a more exhaustive description, refer to section 2.3.

Each subject walked on a split-belt treadmill with embedded force plates (V-Gait Dual Belt, Motek-force Link, The Netherlands) (figure 4). Each participant completed three experimental sessions during which ground-level walking was performed at various speeds. Each session consisted of three 1.5 min ground-level walking repetitions and velocity was kept constant during each repetition. Initially, the comfortable walking speed (100%) was selected

according to the subject's vocal reports, and, subsequently, one slower (75%) and one faster (125%) velocities were considered. We reported the velocities for each subject and each condition in table 1. Each session was performed at one of the considered velocities. Participants completed the conditions in a randomized order to control for habituation to treadmill walking. Before the start of the experiment, 16 markers were placed on both legs and on the hip

**Table 1.** Healthy study participants characteristics.

	S1	S2	S3
Sex	M	F	M
Age	24	24	29
Height (cm)	183	154	193
Weight (kg)	68	52	107
Feet size	42.5	37	44
Self-selected speed ( $m s^{-1}$ )	100%	1.0	0.9
75%	0.75	0.675	1.05
125%	1.25	1.125	1.75



**Figure 4.** System verification. (A) Each subject ( $n = 3$ ) walked on a split-belt treadmill with embedded force plates and positions of 16 markers placed on both legs and on the hip according to the Plug-In Gait lower body configuration were recorded through a 3D motion capture system of ten cameras. This represents the gold standard to which data collected by NeuroLegs system (wrapped around the subject's target leg) were compared. (B) Knee angles and (C) vertical ground reaction forces applied on the foot were collected and compared during 1.5 min, 3 repetitions  $\times$  3 velocities level-ground walking. A total of 120 steps for each speed and each subject were used for analysis. For force and joint angle analysis, each full trial has been segmented into steps considering two subsequent heel strikes and all the steps for a certain velocity were superimposed by normalizing to the length of each cycle. Panels (B) and (C) report the mean knee angle and mean normalized force for participant S2 when walking at his comfortable speed (100%). The shaded areas represent one standard deviation. (D) Knee angle trend reliability as a percentage over the entire walking cycle in which the real and the estimated angular speed direction coincide. (E) Difference in time between force peaks of data collected from the two considered systems. For panels (D)–(F), two-tailed Friedman test was performed with within-subject factor 'velocity' (75%, 100% and 125%). Data are reported as mean  $\pm$  SD. Post hoc correction was executed using a Bonferroni test for multigroup comparison. Significant difference of the results is shown in the figure. \* $p < 0.05$ , \*\* $p < 0.01$ , \*\*\* $p < 0.001$ .

of the subject according to the Plug-In Gait lower body configuration and used to calculate the leg kinematics. Ground reaction forces were measured by two force plates embedded in the treadmill at a frequency of 1000 Hz and a 3D motion capture system of ten cameras (Bonita B10, VICON, UK) was used to acquire markers' position at a frequency

of 100 Hz. Simultaneously, knee angles and vertical forces applied on the foot were recorded by the NeuroLegs system at 50 Hz.

#### 2.4.2. Data analysis and outcome measures

Collected data from the 3D motion capture system and the force plates was post-processed using Vicon

Nexus software (Vicon Nexus 2.8.1, VICON, UK) which requires to fit the Plug-In Gait Lower Limb model to identify the markers and fill gaps in marker trajectories. Subsequently, the parameters of interest, namely the coordinates on the sagittal plane ( $YZ$  plane) of the markers on the shin, knee and thigh and the ground reaction forces were extracted for each trial.

Data analysis was conducted in MATLAB (MATLAB R2019b, The MathWorks, Inc.) and the trend over time of the knee angle was calculated from the markers' ( $y,z$ ) coordinates for each trial. The trend was then compared to that estimated by the NeuroLegs system. Subject's force distribution was analyzed by isolating the  $z$ -component of the ground reaction force recorded by the force plates and comparing it to the sum of the forces measured by each sensor of the NeuroLegs's insole. To address the differences in sampling rate of the collected data, data were post-processed using linear interpolation. In particular, force plates (at 1000 Hz) and 3D motion capture system (at 100 Hz) data were downsampled to 50 Hz to match the sampling rate of the NeuroLegs system.

For each trial, the first 10 s of walking were discarded to be sure the treadmill has reached the target velocity and to avoid capturing transient effects. In order to perform force and joint angle analysis, each full trial was segmented into steps. About 40 steps were considered for each repetition, with a total of 120 steps for each speed. Heel strike was identified as the instant when the ground reaction force exceeds a pre-set threshold, and two subsequent heel strikes were used to identify each gait cycle. All the steps for a certain velocity were superimposed by normalizing the length of each cycle. Data collected from the 3D motion capture system (VICON) and force plates was considered the gold standard to which NeuroLegs system's measurements were compared.

A novel outcome measure, referred to as knee angle trend reliability and defined as the percentage over the entire walking cycle in which the real and the estimated angular speed direction coincide, was introduced to assess the accuracy of the NeuroLegs system in estimating the knee angle. It represents the accuracy of NeuroLegs estimated angle in following the trend of VICON data and, hence, guarantees that the stimulation provided to the subject properly reflects the walking phase and walking pattern. An equally important measure was the difference in time between force peaks of the two considered systems. Minimum time differences ensured an unperceivable delay in the stimulation and a perfect match to the walking cadence. As a secondary outcome measure, we considered the knee angle amplitude accuracy defined as the difference in absolute value between VICON and NeuroLegs knee angle curves. It showed NeuroLegs system precision in replicating the range of motion of the subject's joint.

## 2.5. Validation with amputees

### 2.5.1. Experimental data

Three transfemoral amputees (table 2) were asked to complete a series of walking tests on the same split belt treadmill embedded with force plates. They were fitted with a Full-Body Plug-In Gait model to more holistically record their kinematic response to sensory feedback. The experiment was approved by the ETH Zürich ethics commission (EK 2019-N-97) and all the subjects read and signed the informed consent.

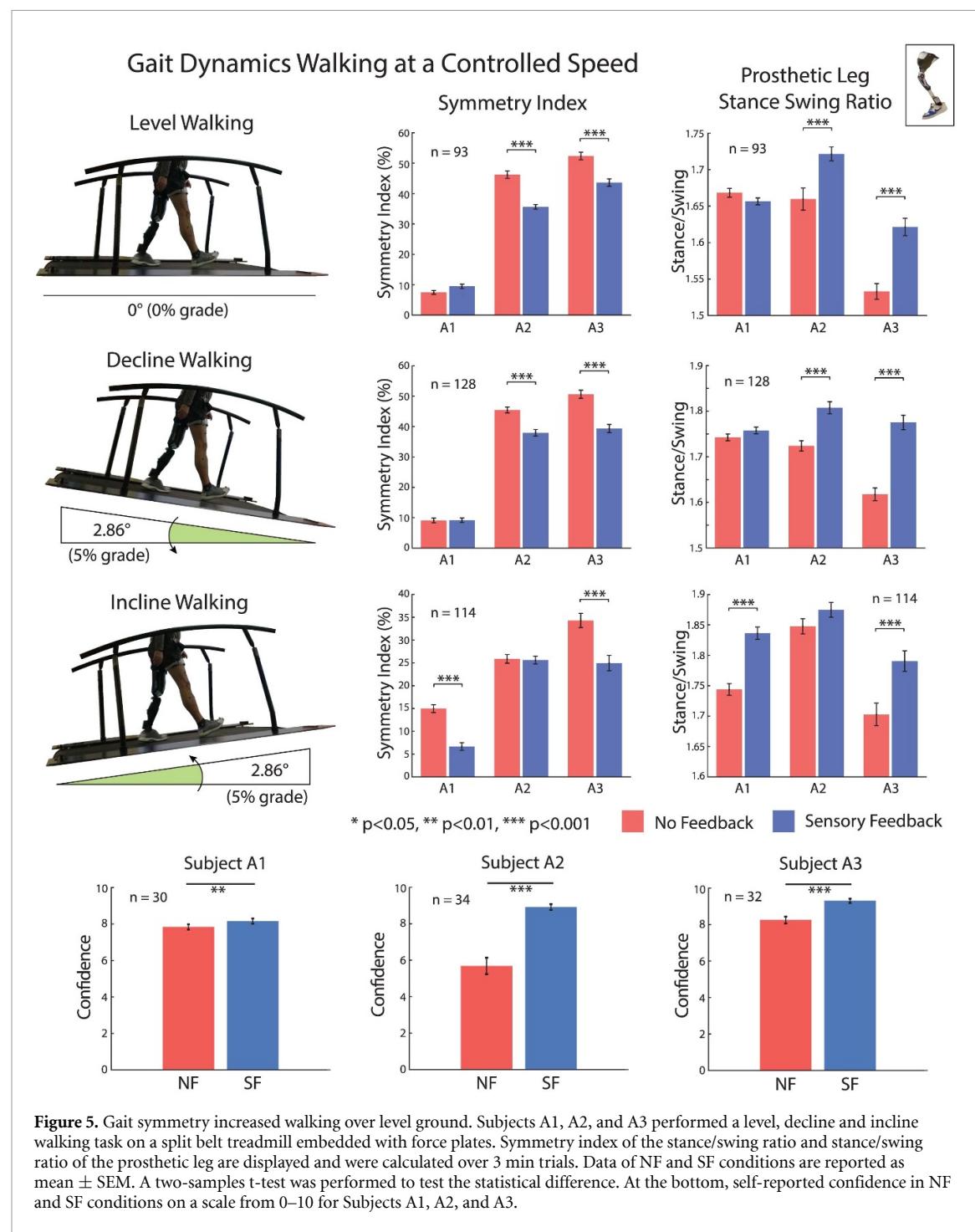
Subjects walked on the treadmill at a self-selected speed in three distinct walking tasks (figure 5): on a level surface, on a 5% grade incline, and on a 5% grade decline. Before each experiment, preferred walking speed was determined by gradually increasing the speed by  $0.1 \text{ m s}^{-1}$  until the participant reported that the speed was too fast. If necessary, the speed was reduced by  $0.1 \text{ m s}^{-1}$ . Subjects were instructed to select a walking speed that was comfortably maintainable for three minutes in each of the three performed tasks. Subject A1 selected  $1.25 \text{ m s}^{-1}$ , Subject A2  $1.19 \text{ m s}^{-1}$  and Subject A3  $1.17 \text{ m s}^{-1}$ . Once determined, the walking speed was maintained constant during all the experiments. Each task included a three-minute session with sensory feedback active sensory feedback (SF) and a three-minute session with sensory feedback off no feedback (NF). In addition, the efficacy of the complete NeuroLegs system was tested against international competition as a part of team NeuroLegs at CYBATHLON. From 2019 to 2020, team NeuroLegs took part in three CYBATHLON competitions, CYBATHLON Series (CS) (Leg) in May 2019, CYBATHLON Weltklasse (CW) in August 2019, and CGE in November 2020. In each of these three competitions, the user (pilot) of the NeuroLegs system (subject A1, table 2) completed a course of six everyday tasks designed to test a specific aspect of their overall functionality. The dimensions of functionality tested were Static Balance, Precision, Dynamic Balance, Mobility, Agility, and Stability as shown in figure 6 and movie S2. Between CYBATHLON competitions, the details of each task were changed along with the points awarded for their completion. At each competition, the pilot was allowed three attempts at the course with only the highest performance being scored against competitors. In light of this, in order to compare performance between competitions over time and avoid the bias of different strategies taken across different attempts, only the times of the highest scoring performances from each competition were considered in the analysis.

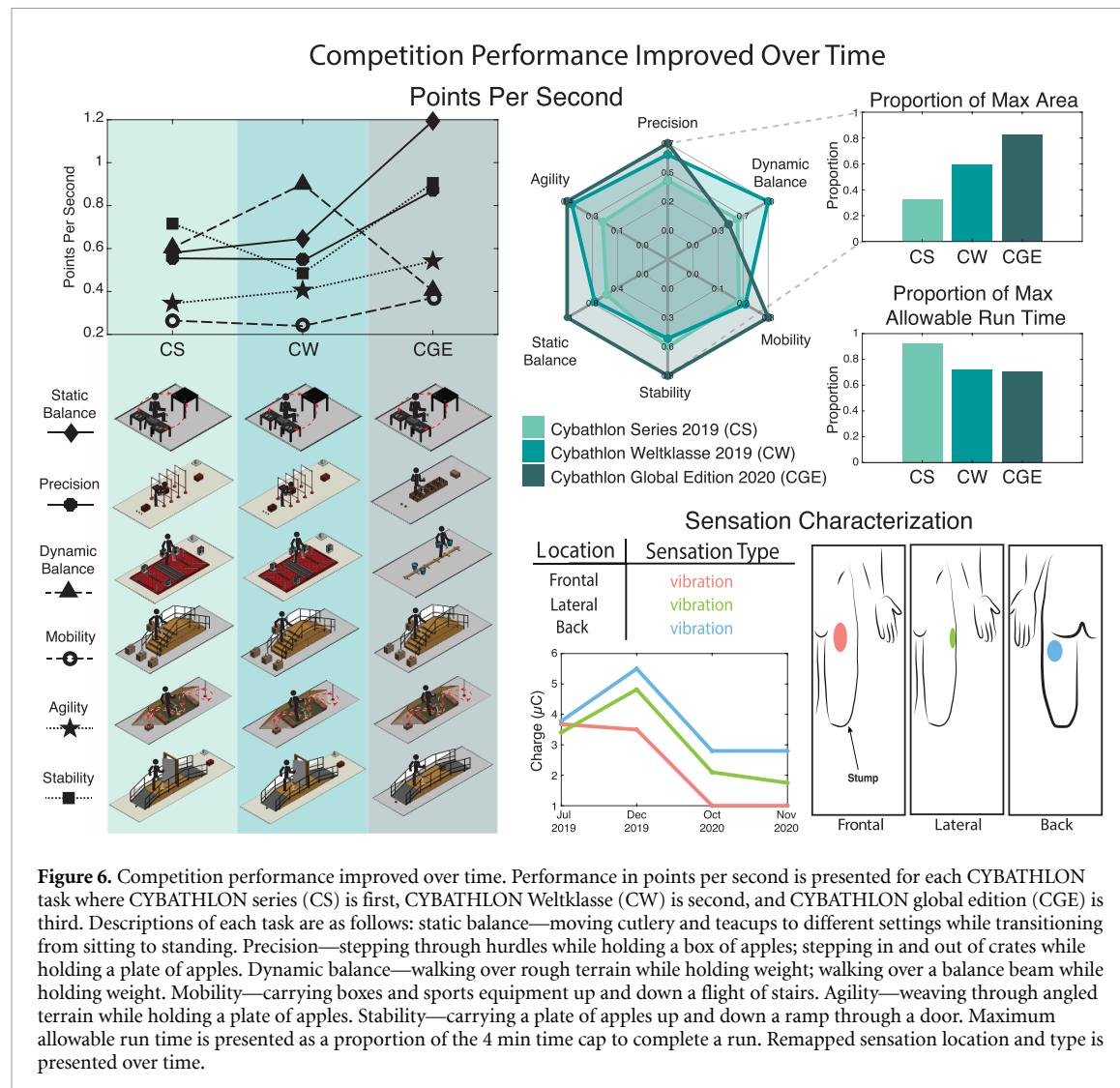
### 2.5.2. Data analysis and outcome measures

Similarly to what described in section 2.4.2, data were collected from the same 3D motion capture system and force plates to understand the kinematic benefits of walking with sensory feedback active. Stance and swing time were extracted from force plates in order to compare gait symmetry between conditions.

**Table 2.** Transfemoral amputee study participant characteristics.

	Gender	Age	Amputation level and side	Amputation Cause	Year of amputation	K-level	Prosthesis	Frequency of use	Phantom limb pain
A1	M	54	Distal two-thirds of left leg (transfemoral)	Trauma	2013	4	Ottobock GENIUM X3 and Ottobock VARI-FLEX LP	Daily	No
A2	M	31	Distal two-thirds of left leg (transfemoral)	Trauma	2018	4	Ottobock R and Össur PRO-FLEX XC foot	Daily	No
A3	M	34	Gritty-Stokes of right leg (transfemoral)	Cancer	2015	4	Össur RHEO KNEE	Daily	Low





In transfemoral amputees, the symmetry index [4] was analyzed with and without sensory feedback during the level, decline, and incline walking tasks in order to verify the effects of the NeuroLegs system using the following equation:

$$SI = \frac{(\text{HealthySide} - \text{ProstheticSide})}{0.5 \times (\text{HealthySide} + \text{ProstheticSide})} \times 100.$$

In this study, the stance/swing ratio was computed using the time spent in each of these gait phases and the symmetry index was calculated using the stance/swing ratio of each leg [4] (i.e. HealthySide = stance/swing ratio for the healthy leg and ProstheticSide = stance/swing ratio for the prosthetic leg). A score of 0% indicates a totally symmetric walking pattern while a score of 100% indicates totally asymmetric walking.

At the end of each walking task, participants were asked to assess their self-confidence during the task, using a 10 cm long Visual Analogue scale [14]. Subjects were asked, ‘Please mark on this scale how confident you felt during the task from “Not confident at all” to “extremely confident”’. The scale was later

digitized from 0 to 10. A mark on the far right at ‘Not confident at all’ was digitized to 0 while a mark at the other end at ‘extremely confident’ was digitized to ten. A mark 4.5 cm from the far left was digitized to a score of 4.5. The data were acquired in SF and NF conditions in all the subjects.

From each CYBATHLON race, the individual task times of the best overall run were considered when calculating points per second (PPS = points for successful task completion /time to complete task) for each task.

The charge on each channel of the remapped feedback and the type of reported sensations were recorded for each race.

## 2.6. Statistical analysis

For each healthy subject, value sets for each parameter of interest were extracted and each distribution was tested for normality using a Single Sample Kolmogorov-Smirnov goodness-of-fit hypothesis test. Since all the distributions were found to depart from normality, a two-tailed Friedman test was performed with within-subject factor ‘velocity’

(75%, 100% and 125%). Post hoc correction was executed using a Bonferroni test for multigroup comparison.

For each amputee, each parameter of interest was tested for normality with the Single Sample Kolmogorov-Smirnov goodness-of-fit hypothesis test. Since all the distributions were found to be normal, a t-test was applied within-subject between SF and NF conditions.

Significance levels were 0.05 unless differently reported in the figures' captions. In the captions of the figures, we reported the used statistical tests for each analysis. Details about the number of repetitions ( $n$ ) for each experiment are reported in the corresponding figure legends. Significance levels are represented in the figures.

### 3. Results

#### 3.1. Mechanical and electrical verification

Protection of patients and users against mechanical risks was guaranteed by an elevated mechanical resistance of the plastic case that encloses the electronic circuit. The case withstood a vertical force of 1.47 kN along the weakest axis before reaching the breaking point.

No mechanical damage was observed after dropping the electronic circuit system enclosed in its plastic case from 5 cm and 1 m, three consecutive times. Signal integrity was observed after each drop and at the end of the test.

Risk arising from vibration of the device was minimized guaranteeing a correct data transmission during and after 500 000 bumps at 20 Hz of the system. No mechanical failure was reported. No excessive or abnormal oscillation and, hence, no resonance frequency was encountered while increasing the oscillation frequency over the entire range associated with walking and running of a prosthetic user.

The external plastic case did not heat during the 18 h of testing. Maximum measured temperature of the printed circuit board components during operation was 30 °C (86 °F).

About 23 h of continuous data acquisition and transmission was reached before the system turned off as a result of battery discharge. All data was correctly stored by the system controller such that it would not be lost in the case of accidental shutdown. It took three hours for the battery to fully charge. Special light emitting diode (LED) indicated the battery status during recharging.

One packet was lost during transmission between the electronic circuit and the system controller over 23 h of uninterrupted operation, equals to one packet over four million being the transmission rate 50 Hz. One packet was lost during 3 h of continuous transmission between the two electronic circuits, equals to one packet over 500 000.

Average communication delay was calculated to be lower than 50 ms. A mean delay of  $40 \pm 22.5$  ms (mean  $\pm$  SD) was introduced over the Bluetooth transmission. Additional 9 ms were accumulated in sending the parameters of stimulation to the stimulator. The delay over the wired connection between the system controller and the stimulator was found to be 600  $\mu$ s and, hence, negligible (figure S2).

#### 3.2. System verification

Results from the system verification are illustrated in figure 4. Data describing trend reliability during swing phase (60%–100% of the walking cycle) are reported for each healthy participant and each velocity in table S3. A mean calculated value of  $86.07 \pm 3.26\%$  (mean  $\pm$  SEM) demonstrated high accuracy in following the knee angle trend during the portion of the gait cycle in which the stimulation is delivered. Trend reliability higher than 80% was obtained for each subject and each velocity (figure 4(D)) and no intra-subject significant difference was found for any participant (Friedman test,  $p > 0.05$ ). An examination of trend reliability during stance phase was conducted for completeness (table S4) and significantly higher trend reliability during swing phase compared to stance phase was found for each subject at each velocity (Wilcoxon signed-rank test,  $p < 0.001$ ).

A mean percentage amplitude difference of  $8.40 \pm 1.79\%$  indicated a slight underestimate of the knee angle measurement of the NeuroLegs system (figure 4(E) and table S5). A trend towards an increased amplitude difference at increasing velocity was shown. A significant difference was found between the slowest velocity and the other two tested conditions for subject S1 (Friedman test,  $p = 0.003$  between 75% and 100% and  $p < 0.001$  between 75% and 125%) and between the two lowest velocities for subject S3 (Friedman test,  $p = 0.049$ ). No intra-subject difference was obtained for subjects S2. Root mean square error (RMSE) data were calculated for comparisons to the literature (table S6).

The difference in time between force peaks (figure 4(F) and table S7) was measured to be smaller than 150 ms for each subject and each velocity, being smaller than 70 ms for two subjects at all velocities. A significant difference was found between the highest velocity and the other two tested conditions for subject S1 (Friedman test,  $p < 0.001$  between 75% and 125% and  $p = 0.019$  between 100% and 125%) and between the two lowest velocities for subject S3 (Friedman test,  $p = 0.017$ ). No intra-subject difference was obtained for subjects S2 (Friedman test,  $p > 0.05$ ).

#### 3.3. Validation with amputees

The transfemoral amputee subjects were asked to walk at constant speed on a treadmill with different

inclinations. Force, spatio-temporal and kinematics data were simultaneously acquired. The symmetry index and the prosthetic leg stance-swing ratios with and without sensory feedback during the level, decline, and incline walking tasks were calculated.

During the level walking task, the symmetry index improved by 22.5% and 16.0% ( $p < 0.001 N = 93$  and  $p < 0.001 N = 93$ ) and the prosthetic leg stance-swing ratio increased by 3.3% and 5.8% ( $p < 0.001 N = 93$  and  $p < 0.001 N = 93$ ) in Subjects 2 and 3 respectively (figure 5(A)).

During the decline walking task, the symmetry index improved by 16.5% and 22.3% ( $p < 0.001 N = 128$  and  $p < 0.001 N = 128$ ) and the prosthetic leg stance-swing ratio increased by 4.9% and 9.7% ( $p < 0.001 N = 128$  and  $p < 0.001 N = 128$ ) in Subjects 2 and 3 respectively (figure 5(B)).

During the incline walking task, the symmetry index improved by 48.4% and 27.3% ( $p < 0.001 N = 114$  and  $p < 0.001 N = 114$ ) and the prosthetic leg stance-swing ratio increased by 4.8% and 5.2% ( $p < 0.001 N = 114$  and  $p < 0.001 N = 114$ ) in Subjects 1 and 3 respectively (figure 5(C)).

The individual swing and stance times with and without sensory feedback are reported in figure S3.

All the transfemoral amputees reported a higher self-reported confidence (Conf) while performing the tasks in SF than in NF (figure 5(D)). In particular, Subject A1 showed an overall confidence level of  $8.1 \pm 0.1$  in the SF condition, which was significantly higher ( $p = 0.006$ ) compared to the NF condition ( $7.8 \pm 0.1$ ) with confidence measured after 30 repetitions (15 NF and 15 SF); Subject A2 significantly ( $p < 0.001$ ) increased his confidence by almost three points (Conf\_NF =  $6.0 \pm 0.2$ ; Conf\_SF =  $9.0 \pm 0.1$ ) with confidence measured after 34 repetitions (17 NF and 17 SF); and Subject A3 increased from  $8.5 \pm 0.5$  in NF to  $9.3 \pm 0.1$  in SF ( $p < 0.001$ ) with confidence measured after 32 repetitions (16 NF and 16 SF).

In figure 6, the results from the CYBATHLON competitions are reported. The PPS are presented in order with CS first, CW second, and CGE third. The PPS for Static Balance was 0.58, 0.65, 1.19, for Precision was 0.55, 0.55, 0.87, for Dynamic Balance was 0.60, 0.90, 0.41, for Mobility was 0.27, 0.24, 0.37, for Agility was 0.34, 0.41, 0.54, and for Stability was 0.72, 0.49, 0.91. When plotted on a spider plot with each axis normalized to the maximum PPS per task, CS covered 32.65% of the total possible area, CW 59.49%, and CGE 82.77%. Out of the total allowable time (4 min), 92.08%, 72.08%, and 70.6% of the time were used at CS, CW, and CGE respectively.

The charge on each channel of the remapped feedback was recorded in July 2019, December 2019, October 2020, and November 2020. On the front channel it was  $3.7 \mu\text{C}$ ,  $3.5 \mu\text{C}$ ,  $1 \mu\text{C}$ , and  $1 \mu\text{C}$ , on the lateral channel it was  $3.8 \mu\text{C}$ ,  $5.5 \mu\text{C}$ ,  $2.8 \mu\text{C}$ , and  $2.8 \mu\text{C}$  and on the back channel it was  $3.4 \mu\text{C}$ ,  $4.8 \mu\text{C}$ ,

$2.1 \mu\text{C}$ , and  $1.8 \mu\text{C}$  for those dates respectively. Vibration was the only type of sensation reported for all the channels on all of the dates (figure 6).

#### 4. Discussion

Restoration of sensory function to transfemoral amputees was shown to improve gait by increasing walking symmetry and speed, confidence and proprioceptive awareness on the prosthetic leg and reducing mental and physical fatigue [14, 34, 38, 56]. Very sophisticated prostheses must give way to portable systems that can be independently donned and totally controlled by the user. After a first setup with the participation of an expert clinician or technician, the NeuroLegs system can be independently run by the amputee through a mobile application specifically developed for this purpose. Compared to previously developed invasive solutions, the NeuroLegs system avoids the need of a surgical procedure and guarantees lower cost and easier medical device certification. On the contrary, non-invasive solutions require higher setup and fitting time and are usually requiring the patient's training to operate. In future works, a direct and rigorous comparison of the NeuroLegs system with previously developed invasive solutions for sensory feedback should be performed both in terms of functional performance and device usability.

The system was demonstrated to be mechanically and electrically safe for continuous use and able to withstand persistent vibrations and fall impacts which represent the stresses the system is subjected to during walking and running or whenever an obstacle is hit.

Mechanical tests showed that the system was durable enough to withstand maximum tibial acceleration (2 g) when walking over enough cycles to meet ISO 22657 requirements [53, 54]. It was also tested for natural resonance frequencies based on ISO 7626-5:2019 while keeping in mind the natural frequencies of running and walking [55, 57–59]. The system did not have any resonance frequencies between 0.1 and 10 Hz which includes low frequency walking as well as running frequencies which average around 4.1 Hz [55]. The system itself is not meant to be mechanically loaded while walking, however, it was tested for its ability to withstand a static load in case it is stepped on by a user during use. The static load that the case was able to withstand before cracking was higher than the 1.23 kN used to cyclically load lower limb prostheses when testing for durability [53]. The system was also shown to withstand falls as a result of daily living as specified by IEC 60601-1.

Electrical tests showed that the system does not overheat and does not cause any hazard for the patient wearing it. The long lasting battery allows for a continuous usage of the system for an entire day exceeding the average daily usage time of the prosthesis

by LLA users reported to be approximately 12 h by [60, 61]. Moreover, a negligible delay of approximately 50 ms was measured. Farrell *et al* [47] demonstrated that delays of 100–125 ms are not perceivable by the user and do not negatively influence the performance. The calculated packet loss rate was substantially smaller than those presented in [62–64], assuring stable Bluetooth transmission and transceiver communication. Furthermore, it allows for the possibility to increase the sampling rate of data communication without impacting the quality of the transmitted signal.

Regarding verification of system accuracy, we introduced a new measure for knee angle accuracy referred to as trend reliability. It quantifies the percentage over the gait cycle that reliably follows the trend (increasing or decreasing) of a gold standard measure. High trend reliability prevents the possibility to provide the subject with a misleading stimulation that would induce a confounding element during walking and increase the cognitive burden. A high trend reliability in the swing phase is required as opposed to the stance phase in which lower trend reliability values can be accepted and overcome. Error and oscillations of the knee angle estimation during the stance phase can be removed by choosing a sufficiently high lower threshold at which the stimulation is initiated. In fact, assuming the subject's knee is kept straight during the stance phase, a sufficiently high threshold could be set so that no stimulation is provided during this portion of the gait. In the present study, a reduced trend reliability during the stance phase could be associated with the impact of the foot on the treadmill at the heel strike that causes the vibration of the electronic boards.

IMU based angle calculation provided an under-estimated knee angle through the entire gait cycle. Calculated RMSE values are comparable to the errors obtained by Takeda *et al* [65] with a similar system. Higher precision was obtained by Favre *et al* [66] using three 3D gyroscopes fixed to the thigh and the shank of the patient. In our system, a trade-off between error and complexity was reached to guarantee ease-of-use and portability. The subject attaches the two electronic boxes around his ankle and thigh independently just caring to align them on the lateral, external side of the leg in the sagittal plane. No complex calibration is required apart from keeping an upright posture with the legs straight for the first 10 s after booting the system. This limits the precision of the angle estimation to the accuracy of the subject in performing the calibration [67].

A system for knee angle readouts (without the sensorized insole) presented by Keri and colleagues [68] exploits IMUs to provide a joint angle feedback system through a vibrationally induced kinesthetic illusion. The NeuroLegs includes similar joint angle feedback read out through IMU data, but instead provides re-mapped electro-cutaneous feedback. Our

system also includes feedback based on force information read through a sensorized insole. Regarding the technical details, both systems report a <50 ms delay, while Keri *et al*'s WIbS system samples at 100 Hz as opposed to NeuroLegs' 50 Hz. WIbS maintains a highly accurate joint angle calculation at normal walking speeds, while NeuroLegs maintains a consistent joint angle trend that is reliable for calculating stimulation parameters. The absolute angle accuracy of the NeuroLegs system could have been diminished by the realistic walking scenarios in which it was tested when compared with the controlled movements tested in the WIbS. Moreover, NeuroLegs balances the trade-off of slightly lower accuracy in exchange for the ability to provide multimodal feedback without increased delay.

Reduced errors in estimating force peaks occurrence were proven for two subjects. Time differences of 60 ms were far below the perceivable delay threshold obtained by [47]. Slightly higher values were obtained for one subject, which however did not report any mismatch between the stimulation and his/her walking pattern. The cause of such delay needs to be further investigated but the authors speculate it could be associated with a poor synchronization between the systems.

The device was verified in transfemoral amputees who demonstrated overall increased gait symmetry during level ground walking tasks and increased confidence overall. The observed increase in gait symmetry was attributed to the increase in the stance/swing ratio on the prosthetic side as opposed to a change on the healthy side. Observing a significantly lower stance/swing ratios on the prosthetic leg without feedback is consistent with previous knowledge about amputee gait [4] and indicates that NeuroLegs is able to positively affect a change on the prosthetic side without disturbing the healthy side. These results support the effectiveness of the device in restoring sensory feedback that benefits the functional ability of the user as well as augments the confidence they have in their prosthetic.

The successful implementation of the NeuroLegs system at CYBATHLON showed an improvement in overall task performance over time. It also demonstrates that the NeuroLegs system is robust and reliable when performing daily life tasks that require balance, precision, mobility, agility and stability. The system is shown to produce reliable results over months and may suggest an increase in performance over time as the system is learned by the user. None of the other teams at CYBATHLON implemented sensory feedback in their prosthesis, making team NeuroLegs unique. Team Circleg and Contour 2000 presented combined active and passive prostheses, BionicM Inc. presented a prosthesis with motorized knee joint, and ORTOKOSMOS presented a modular, functional prosthesis [50]. NeuroLegs' ability to keep up with motorized knees while being implemented on

a passive knee, highlights its unique strength and acts as a reminder that sensory restoration in lower limbs has not yet been made as available as motor restoration has. Whether these improvements are attributable to a familiarization with the stimulation or to a higher level of training of the participants remain unclear. However, it was demonstrated that the NeuroLegs system was capable of adapting to all different scenarios of everyday life.

In this work, the NeuroLegs device was shown to be safe to use through a variety of commonly used mechanical and electrical tests. Moreover, it was shown to provide amputee subjects with accurate tactile and emulated knee angle proprioceptive feedback that matches their walking pattern at different velocities. In response to the perceived stimulation, amputee subjects were able to effectively adapt and optimize their walking strategy, trust the prosthesis more and become more comfortable. The overall efficacy as this fully portable, real-time device was validated against international competition at CYBATHLON.

In future works, the benefits related to non-invasive sensory feedback restoration should be demonstrated in a larger population and more diversified scenarios (e.g. sit-to-stand and stairs walking). Furthermore, the extension of the system to a two-leg system would enlarge the target population. Peripheral neuropathy and diabetes cause lack of sensation at the extremities, often affecting both legs. In general, the usage of the NeuroLegs system could be extended to all pathological conditions that cause abnormalities in the walking pattern, such as stroke, nerve injury, partial spinal cord injury and Parkinson's disease. Finally, a home-based assessment of the NeuroLegs system and usability study of the device will give further insight into the effects of artificial sensory feedback restoration using electrical stimulation in amputees.

## Data availability statement

The data that support the findings of this study are available upon reasonable request from the authors.

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C B developed the mechatronic system and the software, performed the experiments and the analyses, wrote the manuscript and made the figures; L C developed the calibration software and the overall system integration, performed the experiments, wrote the manuscript and made the figures; G V performed the experiments, supervised the analyses, discussed the results, reviewed the manuscript; S R designed and supervised the experiments, supervised the analyses, discussed the results and reviewed the manuscript.

S R holds shares of 'Sensars Neuroprosthetics', a start-up company dealing with potential commercialization of neurocontrolled artificial limbs. The other authors do not have anything to disclose.

The author's have confirmed that any identifiable participants in this study have given their consent for publication.

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