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HEALTHeBIKES – Smart E-Bike Prototype for Controlled Exercise in Telerehabilitation Programs

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Abstract

The aim was to design and build a prototype of an E-Bike usable for telerehabilitation – a HEALTHeBIKE. It should avoid over exercising, work independently of the environment and it should enable cycling in a group despite different reference exercise intensities. With stakeholders requirements for the system architecture had been identified. A system architecture including an E-Bike, an Arduino microcontroller, an Android smartphone and a telemonitoring platform was presented. Control systems which dynamically provide the required motor assistance were developed. A feasibility study with two subjects cycling in a group was performed. Seven test rides on varying terrain with the same and different exercise intensities were completed. Although the exercise intensities of the two subjects were clearly different, cycling in a group was possible without over exercising.

Keywords: E-Bike, telerehabilitation, Arduino, Android, telehealth

Zusammenfassung

Ziel dieser Arbeit war es, einen Prototyp eines E-Bikes für den Einsatz in der Telerehabilitation zu entwickeln - ein sogennantes HEALTHeBIKE. Verhinderung von Überanstrengung, Funktion unabhängig vom Gelände und Fahren in der Gruppe bei unterschiedlichen Belastungen sollten ermöglicht werden. Mithilfe von Stakeholdern wurden Anforderungen an die Systemarchitektur definiert. Ein E-Bike wurde mit einem Arduino Mikrocontroller Entwicklungsboard, einem Android Smartphone und eine Telerehabilitations Plattform erweitert. Die benötigte Motorunterstützung wurde automatisch durch entwickelte Regelungssysteme bestimmt. Eine Machbarkeitsstudie mit zwei Probanden und sieben Testfahrten in unterschiedlicher Umgebung mit der gleichen und unterschiedlichen Zielbelastungen wurden durchgeführt. Obwohl die Zielleistung der Probanden unterschiedlich war, war das Fahren in der Gruppe ohne Überanstrengung möglich.

Schlüsselwörter: E-Bike, Telerehabilitation, Arduino, Android, Telemedizin

Contents

List of Abbreviations

ADC	Analog-Digital-Converter
AIT	AIT Austrian Institute of Technology
App	Application
BLE	Bluetooth Low Energy
CR	Cardiac rehabilitation
CVDs	Cardiovascular diseases
DAC	Digital-Analog-Converter
E-Bike	Electric bicycle
I^2C	Inter-Integrated Circuit
ISR	Interrupt Service Routine
KMC	KIOLA Mobile Client
Pedelec	Pedal electric cycles
PWM	Pulse-Width-Modulation
SPI	Serial Peripheral Interface
UART	Universal Asynchronous Receiver/Transmitter
WHO	World Health Organization

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1 Introduction

1.1 Motivation

Cardiovascular diseases (CVDs) are a major health problem in our society. In Europe, CVDs are responsible for 3.9 million deaths, which represents 45% of all deaths [1]. In Austria, in 2011, 19% of all hospitalizations and 43% of deaths were due to CVDs [2].
Cardiovascular diseases are heart and blood vessel disorders, for example hypertension (high blood pressure), coronary heart diseases (heart attack), cerebrovascular diseases (stroke) and heart failure. Besides behavioural risk factors, e.g. physical inactivity, tobacco use or unhealthy diet, also the ageing and growing population contributes to the high number of patients challenging the health care systems.
The European Cardiac Rehabilitation Inventory Survey [3] stated that CVDs are chronic and often a reflection of long-term patterns of performed risk factors. Patients need to be supported to either recover or maintain physical capability and to achieve changes in lifestyle, well-being, social and

vocational participation. Furthermore patient self-management needs to be promoted.

1.2 Cardiac rehabilitation

Cardiac rehabilitation (CR) is a typical treatment for CVDs and received the highest class of recommendation and the highest level of evidence, according to [4]. The World Health Organization defines cardiac rehabilitation as [5]:

> "[. . .] *the sum of activities required to influence favourably the underlying cause of the disease, as well as the best possible, physical, mental and social conditions, so that they (people) may, by their own efforts preserve or resume when lost, as normal a place as possible in the community. Rehabilitation cannot be regarded as an isolated form or stage of therapy but must be integrated within secondary prevention services of which it forms only one facet"*

Cardiac rehabilitation is therefore a multidisciplinary field, which consists of physical exercise, managing risk factors, education and psychological support and is typically built up by four phases, see Figure 1.1.
Phase I treats inpatients and happens immediately after the cardiac event. The target is early remobilization. Phase II can already be an outpatient treatment and consists of continued monitoring and education. Furthermore, Phase III is offered to assure sustainability and Phase IV describes lifelong

independent prevention to maintain your health and to minimize the risk for future incidents.

Phases of cardiac rehabilitation

Phase I
Early in-hospital mobilization after an acute event
Phase II
For many patients, outpatient cardiac rehabilitation (phase II: 4–6 weeks) is a suitable and sometimes preferable alternative to inpatient rehabilitation
Phase III
Following phase II in- or outpatient rehabilitation, phase III outpatient rehabilitation is offered in order to assure sustainability of the results achieved during phase II rehabilitation
Phase IV
Lifelong secondary prevention lies in the responsibility of every patient (heart groups, sports clubs, home training, etc.)

Figure 1.1: Phases of cardiac rehabilitation, out of [6]

However, [7] showed that most rehabilitation programs are often not further pursued after initial success. Especially after transition from inpatient to outpatient and lifelong rehabilitation, the healthy attitudes and practices are not continued in a sustainable way.

1.3 The role of physical exercise

The importance of regular physical activity on overall health is well known. Physical activity is defined by the World Health Organization (WHO) as:

> "[...] *any bodily movement produced by skeletal muscles that requires energy expenditure – including activities undertaken while working, playing, carrying out household chores, travelling, and engaging in recreational pursuit."*

Physical exercise is a subcategory of physical activity and is performed in a planned, structured and repetitive form [8].
Prescribing physical exercise as medicine is a key element in the rehabilitation of chronic diseases. Pedersen and Sultin [9] provided an evidence-based basis for prescribing exercise as medicine in the treatment of 26 diseases. Current guidelines and recommendations promote aerobic endurance exercise and resistance training (e.g. dynamic with machine weights) as fundamental methods for cardiac rehabilitation [10]. Endurance training is often performed on stationary cycling ergometers [4].
For safe and efficient exercising it is important to appropriately choose the exercise intensity. According to [11], for endurance training the intensity should be between 50-70% of the maximal or symptom limited heart rate, or 80-90% of the heart rate at the individual anaerobic threshold.

1.4 Telemedicine and Telerehabilitation

Telemedicine (*"Telemedizin"*), by definition of the Austrian Federal Ministry of Labour, Social Affairs, Health and Consumer Protection, is the use of Information and Communication Technology (ICT), to implement treatment while the patient and the healthcare provider (e.g. healthcare professional) are not present at the same location [12].

Telerehabilitation refers to the use of the telemedicine services related to rehabilitation. In a recent amendment of law of the Austrian General Social Security Act (*"Allgemeines Sozialversicherungsgesetz - ASVG"*), telerehabilitation was explicitly included as rehabilitation intervention. [13]

Besides the existing evidence of cardiac rehabilitation, the European Cardiac Rehabilitation Inventory Survey [3] showed that in most of the analysed European countries only a minority of eligible patients performed Phase II or Phase III cardiac rehabilitation programs. Austria reported 30% of eligible patients participated in Phase II and 20% in Phase III cardiac rehabilitation programs. Two suggested explanations are the heavily varying availability of facilities, and difficulties in integrating treatment into daily life.

To increase the number of participants, the "Bericht Telerehabilitation- 2015" report of the Fraunhofer Institute recommended the use of telerehabilitation in cardiac rehabilitation [14]. Telerehabilitation with home based exercises is already in use at the Versicherungsanstalt für Eisenbahnen und Bergbau (VAEB), where a six week extension of the inpatient rehabilitation is offered and a compliance of more than 70% was reported [15]. In previous work, at the AIT Austrian Institute of Technology (AIT), Weinberger [16] developed a telerehabilitation solution consisting of a cycle ergometer and a backend telehealth platform.

1.5 Electric bicycles

An electric bicycle (E-Bike) is a bicycle with an assisting electrical motor. Internationally, the definitions and laws governing E-Bikes are highly varied. In this work, the term E-Bike refers to a type of bicycle, which, in the Austrian jurisdiction, is called "Elektrofahrrad" with motor assistance that has to stop above 25 km/h and the power of the electric motor is limited to 600 Watts (§1 Abs. 2a Kraftfahrgesetz 1967).
Furthermore, the term pedal electric cycles (Pedelec) is often used internationally. A pedelec is defined as a bicycle, where the cyclist needs to pedal to get assistance from the electric motor.
In recent years, E-Bikes are strongly growing in popularity and the sales are increasing every year. In Austria, in 2016, 85.000 E-Bikes were sold, meaning every fifth bicycle sold was already an E-Bike [17].
Cycling is a very common and accepted mode of transportation and leisure activity throughout all age groups. Due to factors like hilly terrain, far distances, or low physical constitution , habitual cycling is not possible for everyone. Therefore, the assistance E-Bikes offers, provides new possibilities for a physical active transportation mode [18]. MentORbike was a

project for an intelligent and automated training support with a Pedelec [19]. HeartGo (HeartGo GmbH, Boeblingen, Germany) offers an E-Bike architecture with an E-Bike, a mobile application (App) and a web portal targeting people with CVDs for individual and efficient physical exercise programs [20]. Meyer, Zhang, and Tomizuka [21] proposed an E-Bike controller to maintain a desired heart rate level, which resulted in an improved riding experience.

1.6 Aim of the thesis

Prescribing physical exercise within telerehabilitation is a recommended method through all rehabilitation guidelines and new inventions could be beneficial for sustainable rehabilitation success. With the increasing trend of E-Bikes, the idea arises to offer individualized, monitored, safe and efficient exercises on E-Bikes. E-Bike rides can be performed during daily life activities, outdoors and in unlimited groups.

The aim of the thesis was to design and build a prototype of an E-Bike usable for telerehabilitation – a so called HEALTHeBIKE. Based on requirements and specifications created with involved stakeholders, a prototype had to be developed.The following requirements should be fulfilled: (a) avoid overload by keeping the power output of the cyclist at or under a desired intensity level, (b) work irrespective of the environment and (c) allow cycling in a group with different individual exercise intensities.

2 Methods and Materials

In the following sections the telerehabilitation concept and the specifications of the HEALTHeBIKE, created with stakeholder, are described. Based on the defined specifications, software and hardware selections were made to design and built the prototype. Furthermore, for evaluation purpose, a feasibility study was designed.

2.1 Telerehabilitation concept

For the telerehabilitation concept, the closed loop healthcare principle, shown in Figure 2.1, was chosen. The healthcare professional configured the individual physical exercise for the patient via a web application on the telerehabilitation platform. The exercise configuration was then transmitted to the patient. After the patient performed that exercise, the workout results were sent back to the telerehabilitation platform, where the healthcare professional could monitor the results, could give feedback and could adjust future exercise settings. In [16] an ergometer as personal health device was introduces in such an architecture.

Figure 2.1: Closed loop telerehabilitation with an E-Bike, adapted from [16]

For the health data exchange with the telehealth platform the Continua Design Guidelines were chosen. Continua Design Guidelines are promoted by the Personal Connected Health Alliance (PCHAlliance) [22]. The goal of these guidelines is to create a secure and inter-operable health data exchange environment. The system architecture can be seen in Figure 2.2. It consists of personal health devices (health, medical and fitness devices), personal health gateway (hub, phone, tablet, etc.), health & fitness service and the healthcare information service (HIS) which are connected through well defined and standardized interfaces.

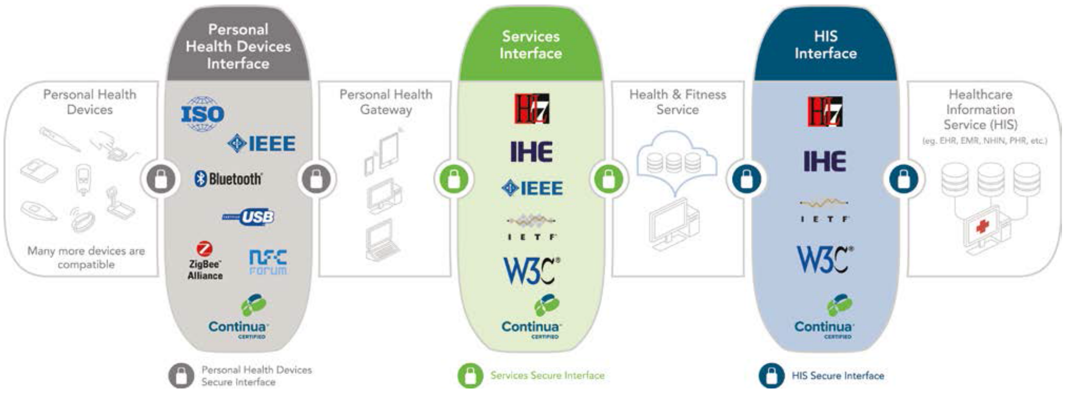


Figure 2.2: Continua end-to-end reference architecture [22]

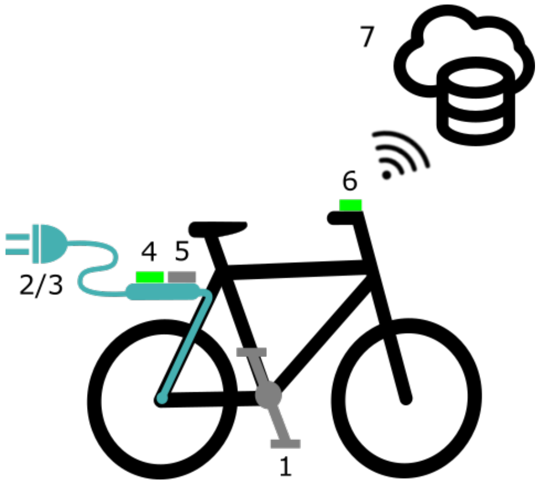
The guidelines are based on the IEEE 11073 framework, which contains a family of standards for personal health device communication. Within this standard, in IEEE 11073-10441-2008, communication standards for cardiovascular fitness and activity monitor devices can be found. An excerpt of this profile with their nomenclature codes, can bee seen in Table 2.1. Parameters of this profile (e.g. heart rate, speed, cadence) could be used but had to be extended to include all relevant parameters for exercising on an E-Bike.

2.2 HEALTHeBIKE specifications

In collaboration with a local rehabilitation center (ZARG Zentrum für ambulante Rehabilitation GmbH, Graz), the Institute of Sport Science of

Table 2.1: Excerpt from ISO/IEEE Std 11073-10441-2008, Health informatics – Personal health device communication – Device specialization – Cardiovascular fitness and activity monitore, profile name: MDC_DEV_SPEC_PROFILE_HF_CARDIO

Reference ID	**Common term**	**Description/definition**
MDC_HF_ALT	Altitude	This is the altitude observation
MDC_HF_HR	Heart rate	This is an object representing the heart rate over a period of time
MDC_HF_POWER	Power	This is an object representing the power over a period of time
MDC_HF_SPEED	Speed	This is an object representing the speed over a period of time
MDC_HF_HR	Distance	The distance covered
MDC_HF_CAD	Cadence	This is an object representing the cadence over a period of time



1. Pedal drive (muscle force)
2. Motor drive (electric motor)
3. (Braking unit)
4. Control unit
5. Measurement unit
6. Display
7. Telerehabilitation platform

Figure 2.3: Schematic of the required components

the University of Graz (Exercise Physiology, Training and Training Therapy Research Group) and an E-Bike manufacturer (EBIKE EXPERTS EUROPE Limited, Gabersdorf) specifications for the HEALTHeBIKE were identified and used for developing the prototype.

2.2.1 Specifications of concept

A concept drawing of the defined components can be seen in Figure 2.3. The specified concept of the HEALTHeBIKE consisted of a pedal drive unit and an electric motor drive unit. The provided assistance from the electric motor had to be fully controllable through a control unit. The control unit should host the control system. As input it should receive current parameters from the measurement unit and the reference values set through the telerehabilitation platform by the healthcare professional. The output of the control system should be a parameter representing the level of motor assistance. A display should provide feedback (e.g. warnings and current exercise metrics). In Table 2.2, the specified components and their required functions are listed.

Table 2.2: Summary of the used requirements of the concept components of the HEALTHe-BIKE

Concept component	**Required function**
Motor drive	Real-time controllable
Pedal drive	Compatibility to power output measurement devices
Brake unit	Provide active resistance (e.g. through recuperation)
Control unit	Host control systems, perform real-time calculations, compatible input from measurement device and telehealth platform, provides output for changing the motor assistance
Measurement unit	Measurement of the required parameters defined in 2.2.2
Display	Provide visual feedback, e.g. warnings
Telereha platform	Configure exercise and show exercise results, synchronize with control unit

2.2.2 Specified parameters for measurement unit

Rehabilitation guidelines specify exercise intensity ranges and thresholds are given for safe and efficient aerobic exercising. The prescription of the exercise intensity is conventionally done in relation to the heart rate or the oxygen uptake of the patient [23]. For stationary cycling exercises, the power output of the cyclist is an additional used parameter.

The heart rate and the power output of the cyclist were chosen as indicators for the exercise intensity. Additionally, the cadence and the velocity should be obtained by the measurement unit.

Heart rate monitors usually measure the heart rate through the electrocardiogram. With the introduction of portable wireless heart rate chest straps, the heart rate is a widely used parameter for exercise intensity configuration. Examples are the H7 and H10 heart rate sensors (Polar Electro Oy, Kempele, Finland) or the Wahoo TICKR X (Wahoo Fitness, Georgia, U.S.). The heart rate needs to be transmitted via Bluetooth Low Energy (BLE), ANT+ or other wireless protocols to a compatible receiver. In recent years, devices with optical measurements of the blood saturation are also gaining attention. Examples are fitness trackers of Fitbit (Fitbit, Inc.,San Francisco, California, U.S.) or the Apple Watch (Apple Inc., California, U.S.).

The **power output** during an exercise is a measure of the athletic performance. In rehabilitation facilities it is a common parameter of stationary cycling ergometers used in performance diagnostics and exercise targets [24]. The power output of the cyclist is defined as the force on the pedals multiplied with the angular velocity. Modern cycling ergometers provide the required resistance through electomagnetic induction (e.g. motion cycle 800, emotion fitness GmbH & Co. KG, Hochspeyer, Germany)

Also in competitive cycling the power output is a standard measure of the performance of an athlete [25]. Common types of measurements on bicycles of the power output are bottom bracket sensors (e.g. ROTOR 2INpower, Rotor Bike Components, Madrid, Spain), crank arms sensors (e.g. SRM powermeter, SRM GmbH, Jülich, Germany) or pedal sensors (e.g. Garmin Vector, Garmin Ltd, Kansas City, USA) which are sensors measuring the torque and calculating the power output with the angular velocity.

The **cadence** or the pedaling frequency is defined as the count of revolutions of the crank arm per minute. The measurement of the cadence is part of power output measurement devices.

The **velocity** is an important parameter for cycling in a group. Furthermore, for safety and regulatory reasons the velocity had to be obtained in order to stop the motor assistance over a given threshold velocity.

2.3 HEALTHeBIKE hardware

The chosen hardware devices related to the specified components for the HEALTHeBIKES system architecture can be seen in Table 2.3. The used hardware and their function are described in the following sections.

2.3.1 E-Bike

The components of conventional E-Bike drive systems are the electric motor, the battery, the motorcontroller and sometimes a display and additional sensors. A wide range of differently implemented E-Bike systems are available on the market.

The electronic drive systems of Bosch (Robert Bosch GmbH, Gerlingen-Schillerhöhe, Germany) are used by many E-Bike manufactures and is therefore a market leader [26]. Bosch adapted the CAN-Bus protocol, which is widely used in automotive industry for their E-Bike systems. Nevertheless the used communication protocol is proprietary and therefore limited to their proprietary tools [27].
E-Bikes of the brand bikee produced by EBIKE EXPERTS EUROPE Limited have a motorcontroller with a throttle input interface, where the motor power can be altered with providing a voltage level. Additionally these E-Bikes feature a built in bottom bracket torque sensor, which could be used

Table 2.3: Hardware selection of the HEALTHeBIKES system architecture related to the specified components

Component	Function	Used device
Motor drive	Motor assistance	Provided E-Bike
Pedal drive	Power output of the cyclist	Provided E-Bike
Measurement unit		
Heart rate	Sensor and receiver device	Polar H7, Smartphone
Power output	ADC[a] for torque sensor, post procession	Arduino
Cadence	ADC for torque sensor + post procession	Arduino
Velocity	ADC for hall sensor + post procession	Arduino
Control unit	Host control systems	Android smartphone
Input	Receive input from measurment unit	USB interface
	Synchronisation with telehealth platform	Mobile web
Output	DAC[b] for motor throttle interface	Arduino
Display	Graphical user interface	Android smartphone display

[a] Analog-Digital-Converter

[b] Digital-Analog-Converter

to measure the power output of the cyclist and the cadence. However, the E-Bike drive system is closed and no open interfaces are provided.

Bikee provided their E-Bikes and offered their know-how, therefore the E-Bikes called ”crossroad inframe” and ”wellness inframe” were used in this work. The E-Bikes were equipped with an E-Bike drive system that contained a 500 watt rear hub electric motor, a motor controller, a speed sensor and a torque sensor, which are shown in Figure 2.4. The throttle input interface of the motorcontroller was used to alter the motor assistance. The outputs of the torque and velocity sensor were voltage signals.

Figure 2.4: Main components of the used E-Bike

2.3.2 Arduino microcontroller board

The Arduino Nano microcontroller board (Arduino S.r.l., Geneva, Italy) was used to capture and process sensor signals from the velocity and bottom bracket torque sensor. By that, the speed of the bike and the cadence and power output of the cyclist could be obtained during riding. Furthermore, the microcontroller was used to set the motor assistance through the throttle input of the E-Bikes motor controller. The Arduino microcontroller provided a pulse-width modulated (PWM) voltage signal with a frequency of 32 kHz, which was low pass filtered to get the desired voltage level. For the low pass, a resistor of 1,5 $k\Omega$ and a capacitor of 10 μF was chosen, resulting in a fast settling time and a small peak-to-peak ripple voltage of the signal. Through this approach, a continuous adjustable motor assistance was achieved.

2.3.3 Android smartphones

An Android smartphone app hosted the control system and acted as display. Furthermore, it was used for communicating with the measurement unit devices and to synchronize exercise configuration and exercise results with the telerehabilitation platform.

The smartphones Motorola G4 (Android 7.0), Samsung Galaxy A5 (Android 7.0), Samsung Galaxy J5 (Android 6.0.1) and Sony Xperia F3111 (Android 6.0) were used for the app development process.

2.3.4 Heart rate sensor

The heart rate was obtained with the Polar H7 (Polar Electro Oy, Kempele, Finland) chest strap and transmitted via BLE.

2.3.5 For additional features and debugging

The barometric sensor MS5611 was used for obtaining the altitude during the test rides. A USB-OTG cable was used for communication between the Arduino and the Android smartphone. For calibrating the torque sensor, the Garmin Vector 2s power meter (Garmin Ltd., Schaffhausen, Switzerland) was used. Indoor test drives were performed on a Tacx roller trainer (Tacx B.V., Wassenaar, Netherlands).

2.4 HEALTHeBIKE Software

Software development

For Android development the official integrated development environment Android Studio (Google LCC, California, USA) was used. The Arduino microcontroller was programmed with the open-source Arduino Software (IDE).

Telehealth plattform

The basis of the telereha software system was the telehealth platform KIT (Keep In Touch) Telehealth Solutions created by the AIT [28]. The patient aggregates vital data through personal health devices, e.g. blood pressure meter. The vital data is transmitted to a mobile application called KIOLA Mobile Client (KMC). Afterwards the data can be transmitted to the core health data server, which is based on the Django open web framework, written in Python [29]. Through a web application with a dashboard the healthcare professional can access and monitor the measured data of the patient and give further instructions, for example treatment modification or feedback. In Figure 2.5 the KIT telehealth platform configured for ergometer exercising is shown. Beside the patient information the exercise results chart and the exercise configuration setting chart can be seen.

The Kiola Mobile Client (KMC) is the mobile application of the KIT Telehealth Solutions created by the AIT. Through this application the patient can record and transmit vital parameters and text messages to the telehealth platform. The KMC is a generic application and can be adapted to a specific use case. An example is the HerzMobil app, which is used in the HerzMobil Tirol disease management network [30].

Figure 2.5: Overview of the telehealth platform Keep In Touch configured for ergometer exercising, taken from [16]

Generic Android app for HEALTHeBIKES

In previous work [31], the generic basis of the Android app for the project HEALTHeBIKES was developed.[1] An overview of the structure the app is given in Figure 2.6.

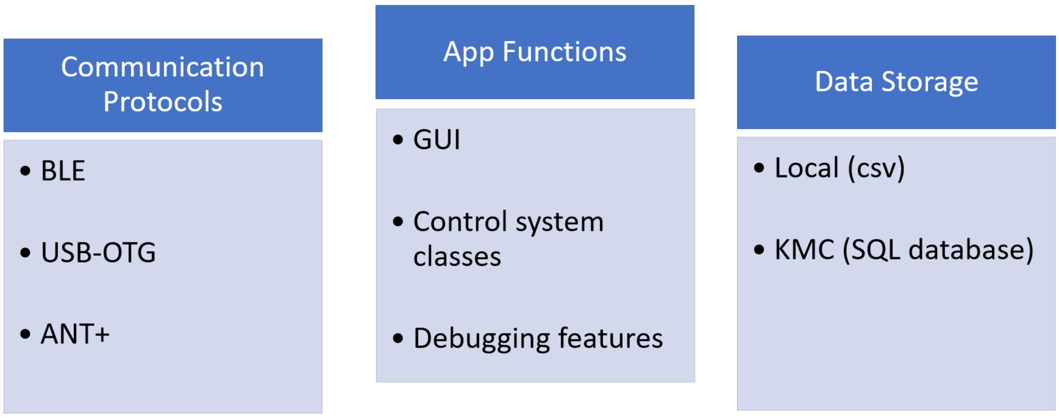


Figure 2.6: Generic HEALTHeBIKE app structure

The app provided a graphical user interface for test rides, it hosted modular changeable control system classes and various debugging features. It supported interfaces to sensors via BLE and ANT+ and to the Arduino microcontroller via USB-OTG. Data from the rides was stored locally in comma-separated values (CSV) files or experimentally in the SQL database of the KMC app. Continua Design Guideline conform data profiles had to be developed. The main graphical user interface for the cyclist can be seen in Figure 2.7. The parameters were shown in an grid view, where each

[1] The concept of the generic HEALTHeBIKE app was part of the thesis, but not the development

content was changeable to other parameters or blanked out. A long press in a grid element opened a list of selectable parameters. A gauge graph, that visualized the current power output related to the target power output of the cyclist was implemented. The heart rate of the cyclist was obtained through heart rate chest straps via BLE. The app either connected to the last heart rate device or to a manually selected one. The manual selection and the used Polar H7 heart rate sensors are shown in Figure 2.8. For easier identification of the correct heart rate chest strap, if multiple sensors were ready to connect, a color tag was used on the sensor device and a corresponding tag in the app related to the MAC-address of the device.

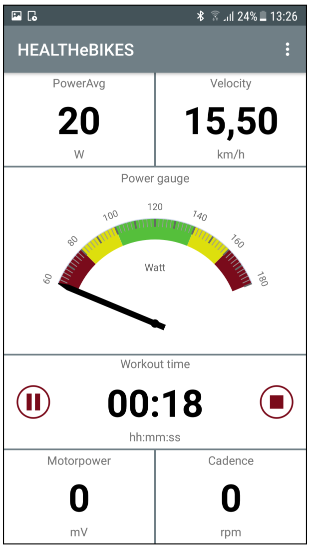
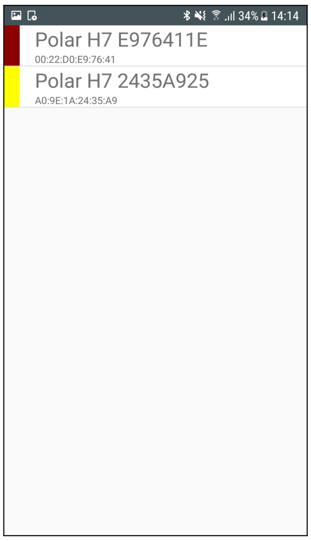


Figure 2.7: Graphical user interface

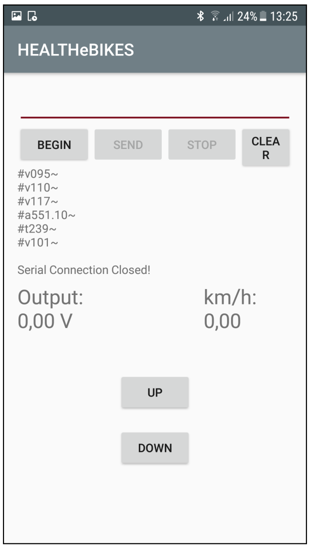
Furthermore, several testing and debugging features were implemented, see Figure 2.9. For example, a serial monitor interface for real-time debugging the Arduino communication, an ANT+ power meters integration for the



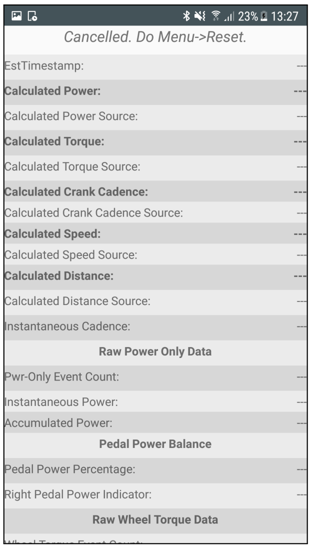
(a) BLE scanning

(b) Heart rate sensors

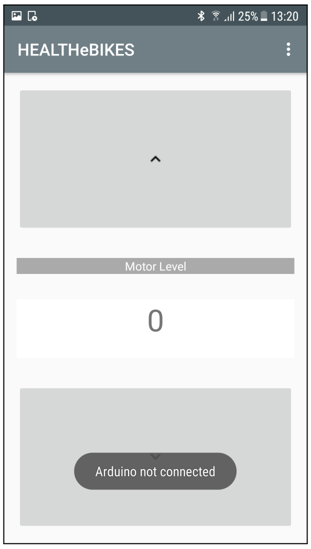
Figure 2.8: Generic HEALTHeBIKE app connecting to BLE heart rate sensors



(a) Serial Monitor



(b) ANT+ display



(c) Default E-Bike design

Figure 2.9: Debugging and additional features

Garmin Vector power meter, or default E-Bike functionality. To notify the user for events like sensor connected/disconnected or error messages small pop up notifications were used.

2.5 Arduino microcontroller board development

2.5.1 Overview

Arduino boards are open-source hardware for developing electronic projects. The Arduino Nano microcontroller board, which is based on the ATmega328P microcontroller, was used in this work.

The board operates with a clock rate of 16 MHz and a voltage of 5 V. For communication Universal Asynchronous Reception and Transmission (UART), inter-integrated-circuit (I^2C) and Serial Peripheral Interface (SPI) are supported. While UART is typically used to communicate with a PC and for programming the board, I^2C and SPI are standards for communicating with other microcontrollers, e.g. other sensors.

The board features analog and digital pins, which can be configured for several input and output tasks. The analog pins are equipped with a 10-bit analog-digital converter (ADC). Some digital pins provide pulse width modulation (PWM) functionality and/or can be used for attached interrupts.

Interrupts are useful for recognizing specified states of an signal without having to constantly call for the pin state in the main program. Therefore, interrupts are used for fast responses to important events. When an interrupt is triggered the specified interrupt service routine (ISR) is executed, regardless of the current execution point of the main program. After the ISR finishes, the main program continues.

2.5.2 Velocity sensor signal

The used E-Bike featured a positional hall sensor in the electric motor with an output of either 0 V or 5 V (Low or HIGH). During one full turn of the rear wheel six HIGH sequences can been observed. The velocity sensor signal was connected to an digital pin, where an interrupt function was attached. At each rise of the signal, so six rises during a full turn, of the wheel, the applied interrupt service routine was executed. Based on the amount of time between the executions and the wheel circumference the velocity of the cyclist was calculated. To prevent errors, a minimum required time till the next HIGH state and a maximal possible acceleration was implemented to filter unwanted multiple bouncing HIGH states and therefore prevent errors resulting in unrealistic velocity or acceleration values.

2.5.3 Torque sensor signal

A built-in bottom bracket torque sensor was part of the used E-Bike. The sensor signals were used to calculate the cadence or pedalling rate in turns of the pedal crank per minute and the power output of the cyclist in watt. The output of the torque sensor consisted of two positional hall sensor signals, the first one corresponding to the angle of the crank arm and the second signal corresponding to the applied torque on the pedal. The hall sensor signal is either 0V or 5V (LOW or HIGH) according to the position of the pedal crank. Every 9 degrees of movement it changed, therefore there are 40 changes in one full 360 degree turn. The second hall sensor signal showed the same properties as the first one but was shifted 4,5 degrees in comparison to the first hall sensor signal. A illustration can be seen in Figure 2.10. .

With comparing the two states it was possible to determined the pedal direction. The first hall sensor signal was connected to an attached interrupt pin on the Arduino. The Interrupt Service Routine (ISR) was executed at every rise of the first hall sensor signal. The ISR measured the state of the second hall sensor.

The torque signal was measured by the analog-digital-converter (ADC) with an analog pin of the Arduino. The relative position of the crank arm was determined by the ISR of the first hall sensor signal. The signal was sampled 20 times over one revolution and numerically integrated.

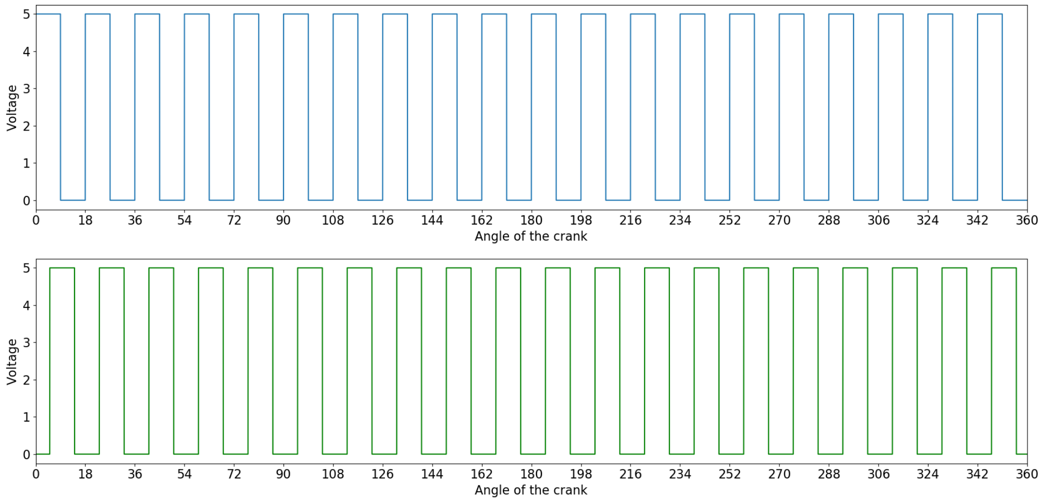


Figure 2.10: Hall sensor output of torque sensor regarding to the position of the crank arm. Second signal (green) is shifted 4,5 degrees to the first signal (blue)

Pedalling power measurement

To calculate the power output of the cyclist with the signal provided by the built-in torque sensor additional steps were necessary. For a physically meaningful interpretation, the torque signal was calibrated with the Garmin Vector power meter. Simultaneous measurements of the build-in torque sensor and a Garmin Vector power meter while increasing the power output from 30 W to 300W were performed and a correlation coefficient of 0,93 was found. Through least-mean-square-fitting, a transformation equation from the voltage signal of the torque sensor to a representing torque signal was then obtained.

2.5.4 Communication

The Arduino board was attached with a USB-OTG cable to the Android smartphone. For communication the universal asynchronous receiver transmitter (UART) port of the Arduino was used. The cycling parameters were sent from the Arduino to the Android smartphone in individual string codes as well as the motor assistance from the Android smartphone to the Arduino. Each sent cycling metric value has a start and end tag, an individual prefix for identifying the metric and the value in an predefined data string format. An example for the velocity can be seen in Figure 2.11.

For receiving and sending data between the Arduino and the Android app broadcast receivers were implemented.

For the communication a USB serial controller was implemented with matching connection setting to the universal asynchronous receiver-transmitter (UART) port of the Arduino. Each sent cycling metric value has a start and end tag, an individual prefix for identifying the metric and the value in an predefined data string format.

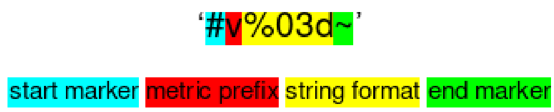


Figure 2.11: Parsing serial data format for velocity

2.5.5 Output signal for motor assistance

For controlling the provided motor assistance the thumb controller interface of the motorcontroller was used. The voltage input from 1,5 V to 4,5 V represented 0% to 100% of the motor assistance. The voltage signal was created through a pulse-width-modulated signal, which is a square wave signal and either high or low. For obtaining a smoothed signal a low pass filter was used. A capacitor of 10 μF and a resistor of 1,5 $k\Omega$ was used for getting an output with a sufficiently fast settling time and sufficiently small ripples.

2.5.6 Additional features

A GY-86 board, equipped with a MS5611 barometric sensor and temperature sensor was added to collect altitude information during the ride and connected to the I^2C port of the Arduino microcontroller board. Through inserting the measured atmospheric pressure and temperature in the barometric formula, the altitude could be obtained.

The Arduino microcontroller board and the Android smartphone were connected through a USB-OTG cable. As alternative to this connection a BLE HM-10 module was connected on a serial port of the Arduino microcontroller board. Therefore, the serial output could transmitted to the Android smartphone via BLE instead of the USB-OTG cable.

2.6 Control system

2.6.1 Background

On stationary cycling ergometers, it is possible to set the resistance e.g. through induction. Therefore the required exercise intensity in form of power output can be easily altered and maintained. On a bicycle, the power output has more degrees of freedom. It depends e.g. on the slope of the rode, selected gear, road resistance. In the case of an E-Bike, the provided motor assistance is another degree of freedom.

An E-Bike has usually fixed assistance levels, which can be changed manually via the handle bar of the bike. To establish and maintain the exercise intensity during cycling the rider has to shift gear and/or change the level of assistance. The idea was to automatically provide the required assistance through a control system.

In Figure 2.12, a diagram of the relations of the main components regarding the velocity is shown. The cyclist applies a force to the pedals, which results in a torque at the bottom bracket of the E-Bike. The power output of the cyclist can be determined by multiplying the torque with the angular velocity. The total drive power is determined by adding the motor assistance to the power output of the cyclist. The drive power combined with the transmission of the bicycle results in a velocity. An additional parameter of significant impact on the velocity are the environmental influences. For

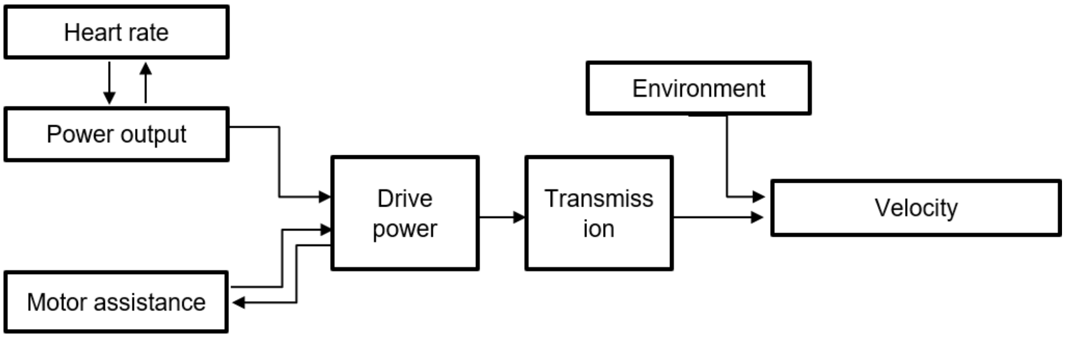


Figure 2.12: Influences of the main components on the velocity

cycling in a group, the participants are required to have the same velocity. With the help of the motor assistance, the same velocity but different exercise intensity amounts can be achieved. When the exercise intensity of the cyclist is too high, for example because of an uphill section, the motor assistance can be increased, that the cyclist can decrease the power output while maintaining the riding velocity.

Furthermore, the cyclist was assumed as a system that tries to maintain the target velocity (via cadence and gear transmission ratio) through power output adjustments. The target velocity was defined as the personally preferred velocity, or when cycling in a group, the velocity of the group leader.

Safety and regulatory requirements

To comply with the Austrian jurisdiction, the E-Bike had to stop the motor assistance at 25 km/h and the motor assistance had to be limited to a 600

Watt motor. When the cyclist stopped pedalling or was pedalling backwards the motor assistance also should stop.

2.7 Feasibility study

Feasibility of controlled exercise in a group on the prototype was evaluated with two healthy subjects cycling together on the two modified E-Bikes. The power output-based control system was used. Subject 1 was defined as the "leader" and subject 2 as the "follower". Seven test rides were performed which included flat, hilly, mountainous and uphill terrain. The subjects performed two test rides with the same target power and five test rides where the target power was different by minimal 40 Watts and maximal 60 Watts. The target power was varied for the subjects and test rides from 60 Watts to 220 Watts. The subjects were permitted to cycle at a cadence of 60-70 rpm. In order to simulate an inexperienced cycling behavior, shifting the transmission ratio had to be changed as little as possible. Subject 2 ("follower") was requested to stay in maximal distance of 10 m behind subject 1 ("leader"). The heart rate, the power output of the cyclist, the provided motor assistance, the velocity and the altitude were recorded for both cyclists. The measured power output of both cyclists was compared with their target power output. The mathematical computing software MATLAB (The MathWorks, Inc., Massachusetts, USA) was used to evaluate the feasibility study.

3 Results

In this section the built prototype is presented. The system architecture, followed by the Android and Arduino development is described and the implemented control systems for controlling the motor assistance are presented. Furthermore an overview about the telerehabilitation platform and the results of the feasibility study are given. Parts of this work were presented at the eHealth2018 conference in Vienna [32].

3.1 System architecture

The system architecture of the designed and built prototype can be seen in Figure 3.1. The Android smartphone was connected to the Arduino microcontroller via USB-OTG, to the heart rate chest strap via BLE and to the AIT telehealth platform via web. The exercise results and exercise configuration were transmitted between the telemonitoring platform and the Android smartphone. Current speed, cadence and power data were captured

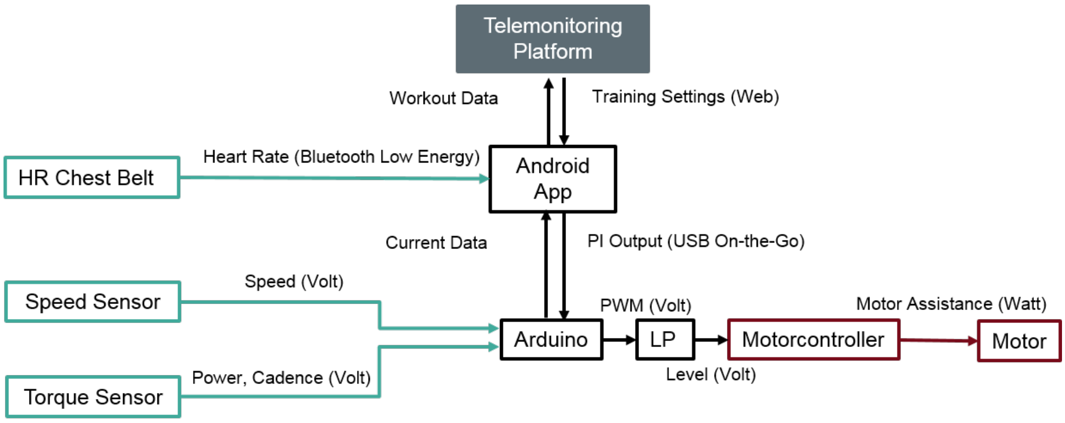


Figure 3.1: System architecture of the HEALTHeBIKES prototype

by the Arduino microcontroller and sent to the Android smartphone. The control system hosted on the Android smartphone calculated the required motor assistance based on the selected control system and sent it to the Arduino microcontroller. The Android smartphone was mounted on the handlebar and the electronic parts (Arduino microcontroller board, low pass, BLE HM-10 module and GY-86 module) were mounted in a small box on the bike frame.

3.2 Android development

The system architecture consisted of two Android apps, the Kiola Mobile Client and the HEALTHeBIKES app. The Kiola Mobile Client app provided an interface to the telerehabilitation platform and the HEALTHeBIKES app provided functionality regarding the control and use of the E-Bike.

Kiola Mobile Client app

The KMC Android app was extended with an E-Bike button, which can be seen in Figure 3.2. Through this button, the HEALTHeBIKES app was launched. The KMC app handled a SQL database for the exercise configuration and results and was based on Continua Guidelines. The used profiles for exchange of the exercise configuration and exercise results data are explained in Section 3.5. The KMC app provided the interface to the telerehabilitation platform.

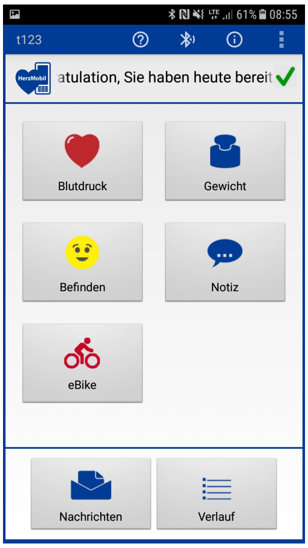


Figure 3.2: KMC integration of the HEALTHeBIKES app

HEALTHeBIKES app

The graphical interface of the HEALTHeBIKES app can be seen in Figure 3.3. The parameters target power output of the cyclist, velocity, exercising time, heart rate and motor assistance were shown. The gauge diagram gave a feedback of the difference of the target power output of the cyclist (value at top) and the current power output of the cyclist (needle). Furthermore, the target power output over the time of the exercise and the current time position (red line) could be displayed. The exercise ID number was shown at the left top.

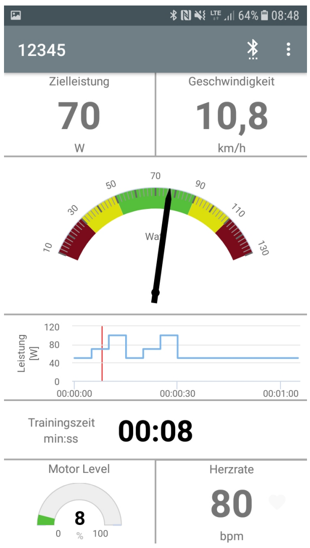


Figure 3.3: Graphical user interface of the HEALTHeBIKES prototype

3.3 Arduino development

In Figure 3.4, the electronic components of the prototype can be seen. The main element was the Arduino Nano, which was connected to the torque and velocity sensor of the E-Bike, the MS5611 barometric sensor, a HM-10 BLE module and the low pass followed by the motorcontroller.

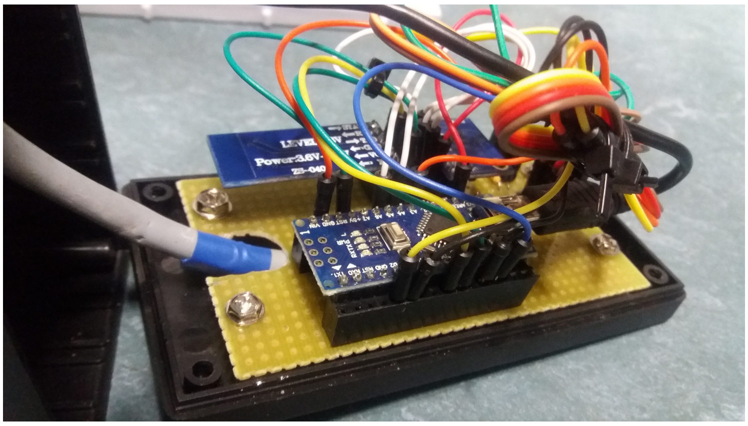


Figure 3.4: Electric circuit board prototype, with Arduino Nano (front), BLE HM-10 module (left back) and MS5611 barometric sensor (right back), low pass (below, not visible)

3.4 Control Systems

Different control systems were created and integrated in the Android app. A power based, velocity based and heart rate limit control system were implemented. The power and velocity based control systems featured a proportional-integral (PI) control system.

3.4.1 Power output-based control system

The target power output of the cyclist could be set by the healthcare professional. The motor assistance has to be dynamically adjusted to maintain the target power output of the cyclist, regardless of environmental changes. The heart rate should be obtained for real time warnings when a set heart rate limit is exceeded and for evaluation of the physical exercise.

The cyclist was assumed to maintain the target velocity (via cadence and gear transmission ratio) through power output adjustments, see Figure 3.5. The target velocity was defined as the personally preferred velocity or – when cycling in a group – the velocity of the group leader. The controller was designed as a proportional-integral controller (PI controller), see Equation 3.1. The output u(t) was a voltage level that was applied on the throttle input of the E-Bikes motor controller to set the corresponding motor assistance.

$$u(t_k) = u_0 + K_p e(t_k) + K_i \sum_{j=1}^{k} e_j(t_k)(t_j - t_{j-1}) \tag{3.1}$$

The error term e(t) was the difference between the desired reference power output of the cyclist and the current power output of the cyclist. The term u_0 was a default offset. Motor assistance started when then power output was higher than this offset. With the coefficients K_p and K_i the proportional and integral terms were weighted. A new power value was calculated after a full rotation of the pedal. Therefore, the sampling time ($t_j - t_{j-1}$) was the inverse cadence. Furthermore, an anti-windup method was implemented through limiting the integral term. This was necessary, when the maximal motor assistance was reached without reaching the target power output.

The coefficients K_p and K_i were determined through experiments with the goal of a fast settling time but without overshoot for good riding experience. K_p was set to 0.001 V/W and K_i to 0.0025 V/Ws. Additionally, a heart rate limit was implemented. When the subject reached this limit, he/she was warned through the display and requested to reduce the workload or to stop exercising.

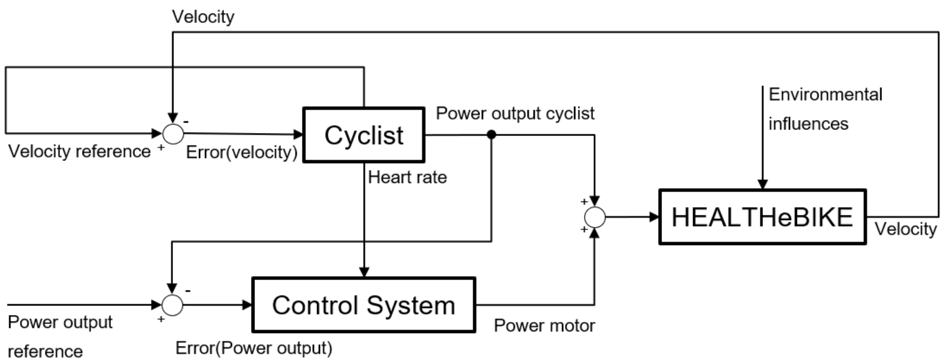


Figure 3.5: Block diagram of power output based control system

3.4.2 Velocity-based control system

The speed of the group was predetermined. The aim of the control system was to maintain the speed, like known cruise control for cars. A block diagram can be seen in Figure 3.6. The control system was also a PI-controller. The setpoint was the target velocity and the process variable the current velocity. The error term was therefore the difference in target and current velocity. The output was a parameter regarding the motor assistance.

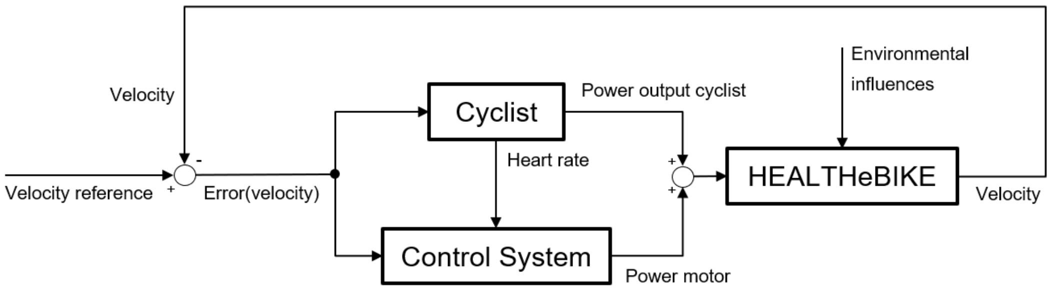


Figure 3.6: Block diagram of velocity based control system

3.4.3 Heart rate limit-based control system

The heart rate is related to the power output of the cyclist. If the heart rate was too high, the motor assistance had to be increased in order to decrease the power output of the cyclist. A simple hard rate limit-based control system was implemented. When the defined heart rate limit was exceeded a motor assistance of 50% was provided. If the heart rate stayed below the heart rate threshold, no motor assistance was given. A time buffer of 30s was implemented, where no changes in motor assistance could be made, to prevent an unstable system since the heart rate shows a delayed response on power output adjustments.

3.4.4 Usability and justification

For safety reasons, the cyclist had to pedal at least for a half revolution in order to receive the motor assistance. When for one second no interrupt was triggered, the motor assistance was stopped. Additionally, it stopped when the cyclist pedaled backwards. To comply to the Austrian justification above 25 km/h no motor assistance was provided anymore.

3.5 Telehealth platform

In the telehealth platform the healthcare professional could see exercise results and could configure the exercise for the patient, see Figure 3.7. In the result graph, the parameters velocity, heart rate, motor assistance and power output and target power output of the cyclist, maximum heart rate were shown.

The healthcare professional could configure individual exercises with segments of defined duration and target power output of the cyclist. Furthermore a program ID and maximal heart rate could be specified. The program ID was shown in the toolbar of the HEALTHeBIKE app to confirm that the correct exercise configuration was synchronized.

Figure 3.7: Dashboard of the telerehabilitation platform with patient information, exercise results (random data) and configuration

3.5.1 Profiles

For the configuration of the prescribed exercise a new personal health device communication profile, called *MDC_DEV_SPEC_PROFILE_VND_AIT_HF_CARDIO_EBIKE_CONFIG*, had to be created. The entries can be seen in Table 3.1. A date-and-time object, a program name and id, the maximum user heart rate and a target power output step sequence were defined. For the exercise results, the existing profile *MDC_DEV_SPEC_PROFILE_HF_CARDIO* was extended, see Table 3.2. The output voltage for the E-Bike motor, a program identifier, the target power output and a observation reference were included.

Table 3.1: Created exercise configuration profile:
MDC_DEV_SPEC_PROFILE_VND_AIT_HF_CARDIO_EBIKE_CONFIG

Reference ID	Description/definition
MDC_ATTR_TIME_ABS	Date-and-Time
MDC_HF_PROGRAM_ID	Program Identifier
MDC_VND_AIT_HF_PROGRAM_NAME	Program Name
MDC_HF_HR_MAX_USER	Max User Heart Rate
MDC_VND_AIT_HF_POWER_STEP_SEQ	Power output step sequence

The workflow of the system architecture can be seen in Figure 3.8. The health professional could configure the exercise on the KIOLA telerehabilitation platform, where a new configuration observation was created. When the patient launched the KMC app, the observations were synchronized with the KIOLA telerehabilitation platform and the measurements are

Table 3.2: Extension of MDC_DEV_SPEC_PROFILE_HF_CARDIO standard to MDC_DEV_SPEC_PROFILE_VND_AIT_HF_CARDIO_EBIKE_RESULT

Reference ID	Description/definition
MDC_VND_AIT_HF_MOTOR_VOLTAGE_OUT	Output voltage for motor
MDC_VND_AIT_HF_PROGRAM_NAME	Program Identifier
MDC_VND_AIT_HF_POWER_TARGET	Target power output
MDC_VND_AIT_OBS_REF	Observation reference

synchronized with the HEALTHeBIKE app. When the E-Bike symbol was clicked the HEALTHeBIKE app was launched with the latest configuration. In case the HEALTHeBIKE app was not installed, a warning was displayed. When the patient started the exercise, the session was recorded. When the exercise was completed the KMC app was started and the measurements and observations synchronized,

3.6 Feasibility study

The results of the feasibility study can be seen in Table 3.3. The two subjects cycled together on flat, hilly, mountainous and uphill gradient with different and the same reference power output. The mean power output was calculated and compared to the reference power output. In total, the difference ranged between 0.2 % and 22.6 % considered both subjects.

Subject 2 confirmed that it was possible to follow the group leader subject 1

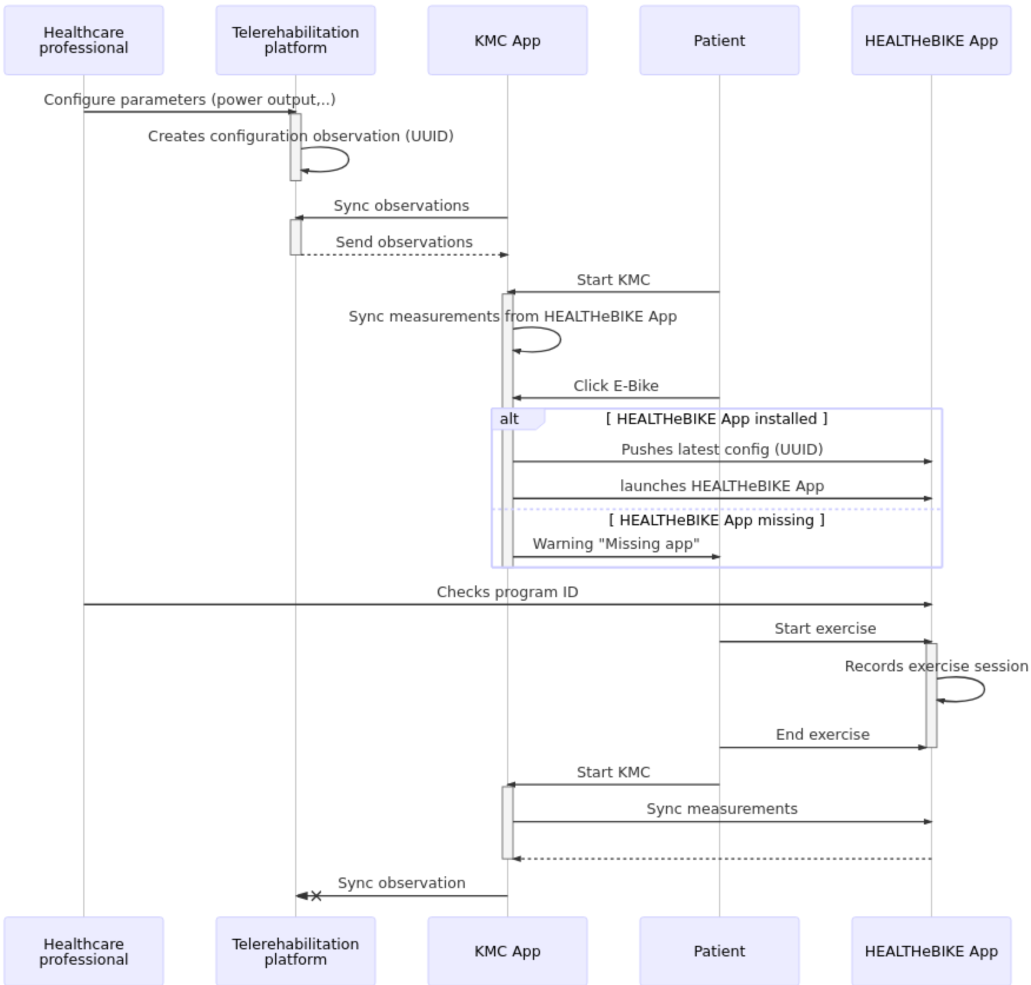


Figure 3.8: Workflow between healthcare professional, KIOLA telerehabilitation platform, KMC App, patient and HEALTHeBIKES App

Table 3.3: Evaluation of the performed test rides in a group of two as achieved during 7 test rides. Reference power output (Power ref), mean power output (Power mean) and the difference (Diff.) between these two values are shown

			Subject 1			Subject 2		
No.	**Gradient**	**Duration**	**Power ref**	**Power mean**	**Diff.**	**Power ref**	**Power mean**	**Diff.**
#	-	min	Watt	Watt	%	Watt	Watt	%
1	flat	5.5	100	106.7	6.7	100	100.6	0.6
2	hilly	9.6	60	59.9	0.2	120	92.9	22.6
3	hilly	10.4	120	96.1	19.9	60	55.9	6.8
4	hilly	11.3	100	86.3	13.7	100	83.1	16.9
5	mountainous	23.6	80	68.1	14.9	120	107.9	10.1
6	hilly	15.4	100	96.6	3.4	140	125.1	10.6
7	uphill	6.7	160	161.6	1.0	220	221.3	0.6

and the maximum defined distance of 10 m behind the leader during all test rides was not exceeded. The system behavior is illustrated in Figure 3.9 and Figure 3.10, where data of test ride 7 of subject 2 is shown. In the beginning, the power output of subject 2 was lower than the target of 220 Watts, since the road gradient was too low to reach this target at the group leader's target velocity. As the road gradient increased, the power output of the cyclist increased to a maximum of 331 Watt (50 % over the target power output). As soon as the power output of the cyclist was above the target power, the motor assistance increased which can be seen at second 145. The motor assistance settled after around 15 seconds when the desired power output was reached. The heart rate of the subject stayed constant over this time indicating that the regulation was fast enough to avoid changes in the heart rate which could prevent overloading of the subjects.

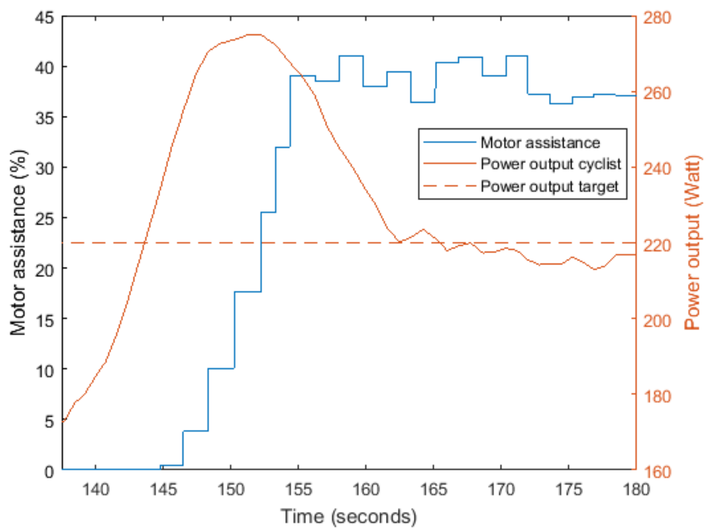


Figure 3.9: Regulation of the motor assistance due to changed power output of the cyclist of test ride No. 7 for subject 2. A moving average window of 10 s is applied to the power output of the cyclist.

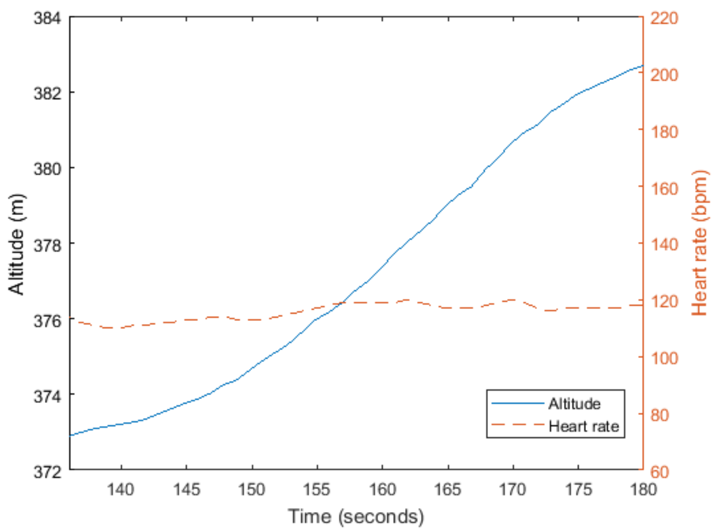


Figure 3.10: Altitude profile and heart rate of test ride. Excerpt of test ride No. 7, subject 2

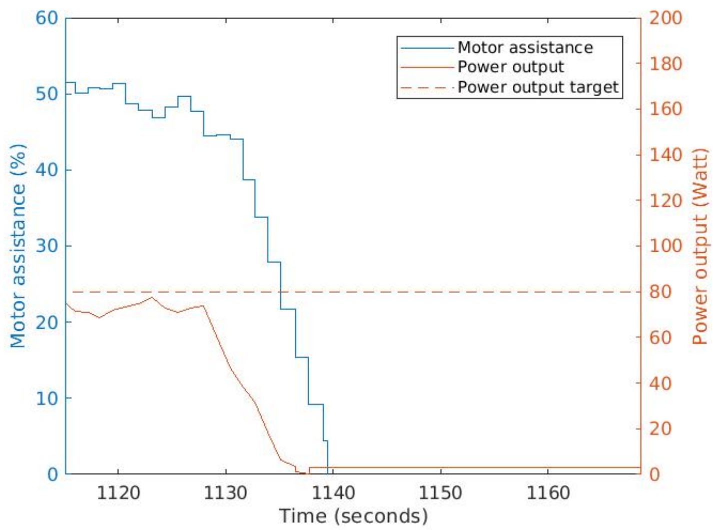


Figure 3.11: Regulation of the motor assistance on downhill segments. Excerpt of test ride No. 7 of subject 2

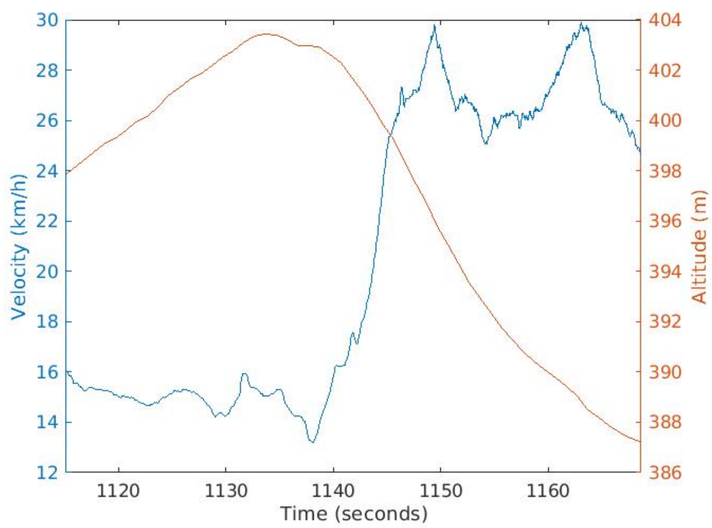


Figure 3.12: Altitude and velocity excerpt of test ride No. 7 of subject 2

The system behavior for downhill parts can be seen in Figure 3.11 and Figure 3.12. In the beginning, the power output was close below the target power output until the peak of the hill was reached. After the peak it was going downhill, which resulted in an increasing velocity and decreasing power output. With the decreasing power output of the cyclist also the motor assistance decreases to zero. Furthermore, above 25 km/h the motor assistance had to stop, according to Austrian law.

4 Discussion

E-Bikes in telerehabilitation programs could open new opportunities to increase patient compliance in cardiac rehabilitation and to achieve more sustainable telerehabilitation results. A system architecture including an E-Bike into a telerehabilitation setting was developed. Control systems with dynamically adjustment of the motor assistance were presented with the aim to avoid over exercising, work independently of the environment and enabling cycling in a group despite different target exercise intensities are given. Promising results of a performed feasibility study could be obtained.

4.1 System architecture

A system architecture including a E-Bike, a Arduino microcontroller board, a Android smartphone and a telerehabilitation platform was presented. An important part was to select an E-Bike, where it is possible to dynamically change the motor assistance. With the provided E-Bike it was possible to use the thumb controller interface for this purpose. Therefore, the motor assistance could be changed with a voltage signal provided by the Arduino board and no changes in the E-Bike drive system were necessary. The signals of the built-in torque and velocity sensor could be used and processed with the Arduino board. With open interfaces and communication protocols the velocity and torque signal could have been read out by a communication protocol and the measurement and processing could have been avoided.

The use of an smartphone was feasible, because the existing KMC app by the AIT to communicate with the telehealth platform could be used. Furthermore, with the BLE receivers of the smartphones it was possible to obtain the heart rate of BLE heart rate chest straps. With the smartphone display a custom design was possible, without being restricted to conventional segment displays.

The braking unit, e.g electric motor with recuperation, was not considered for the prototype, because the provided E-Bikes were not equipped with this functionality. Also, concerns of patient motivation were raised by the stakeholders, e.g. when the patient is confronted with additional resistance on downhill sections to reach the target power output. Furthermore it was not the aim to provide stationary cycling ergometer functionality, e.g. that always the full range of adjustable resistance is possible.

With the implementation of the BLE heart rate profile connection in the HEALTHeBIKES app, the heart rate could be obtained with the Polar H7 chest strap. An alternative possibility for measuring the heart rate are optical heart rate wrist sensors, which are gaining a lot of attention in recent years. Since a heart rate sensor on the wrist is less intrusive than a chest strap sensor it could improve the compliance of the patients.

4.2 Telerehabilitation platform

As shown in Table 3.2, the ISO/IEEE 11073-10441 standard for personal health device communication of cardiovascular fitness and activity monitors had to be extended for this use case. There exists a definition for the power output, but not for the target power output, which was necessary here. Furthermore the profile was extended with a program identifier, an observation reference and the output voltage of the motor. A custom nomenclature developed by the AIT was used for the additional definitions.

4.3 Control systems

A power output-based, a velocity-based and a heart rate limit-based control system was presented. From this list of control systems the power output-based control system was chosen for the test rides and integrated in the telerehabilitation platform. The power output-based control system had the advantage of fast feedback if the power output changed, e.g. because of a hill, and was favored to heart rate-based control systems because of the easier and more intuitive behavior. Meyer, Zhang, and Tomizuka [21] presented a heart rate- based control system, which used a sliding mode controller and a feedforward controller to keep the heart rate constant and achieved good tracking performance. Corno, Giani, Tanelli, and Savaresi [33] developed a heart rate regulated bicycle which kept the heart rate within 10 bpm by adjusting the transmission ratio. However, with this approach cycling in a group with different intensities would not have been possible.

Furthermore, the heart rate shows a non-linear relation to the workload and it is dependent on several additional parameters, such as inter-subject variability or exogenous factors (temperature, humidity) [34], [35]. Therefore, for such approaches, an individual heart rate model would be required. The power output of the cyclist is a widely used parameter for the evaluation of the exercise intensity, especially for competitive cycling. The power output is instant whereas the heart rate responds with some delay.

For a continual heart rate regulation, knowledge about the individual heart rate performance curve is necessary. In related ongoing work the heart rate performance curve of patients of a local rehab institute are analyzed. A goal is to see if it is possible to classify certain groups. This knowledge could be used for developing heart rate regulated algorithms.

4.4 Regulatory aspects

For using the HEALTHeBIKE various regulatory aspects need to be considered. In Austria, laws to regulate the use of E-Bikes can be found in the Motor Vehicle Act (Kraftfahrgesetz). To comply, for example, the motor assistance has to stop above 25 km/h and the power of the motor is limited to 600 Watts, according §1 Abs. 2a Kraftfahrgesetz 1967. Furthermore, the European Standard EN 15194, "Cycles - Electrically power assisted cycles - EPAC Bicycles" defines requirements governing technical aspects of E-Bikes, for example about the electromagnetic compatibility.

If the manufacturer specifies that the HEALTHeBIKE is intended to be used for diagnostic and/or therapeutic purposes, e.g. treatment of a disease, it has to comply to the Medical Device Regulation (MDR). This would include the hardware as well as the software components of the HEALTHeBIKE. In this work the HEALTHeBIKE prototype was not developed with the aim to meet the criteria as medical device. Additional steps would be necessary, e.g. declaration of conformity and technical documentation.

Furthermore, patient data is stored in the app and the telerehabilitation platform, therefore the system architecture has to ensure General Data Protection Regulation (GDPR) compliance. The GDPR is an EU regulation to protect and empower data privacy and came into force on the 25th May 2018. For example, privacy by design and by default has to be implemented to fulfill GDPR compliance obligations.

4.5 Feasibility study

Results show that the mean power output of the cyclists for the different test rides were mostly close or below the target power outputs. Single peaks were heavily above, for example on the start of an steep uphill segment, shown in Figure 3.9, but short enough to be safe. Therefore, based on mean power output, over-exercising was successfully prevented.

For flat and uphill tracks, the target power could be reached very closely. On hilly and mountainous tracks, however, the mean power output of the cyclists was lower than the target power output, especially for the subject with the higher reference power output. This can be explained by the downhill road sections, where the cyclists' power output was lower than the target power. For the subject with the higher reference power it was sooner not necessary to provide the reference power while maintaining the cadence and gear selection. Additionally, restrictions because of traffic, road curves and intersections reduced the mean power. If a system to maintain the target power output even during downhill sections was required, an active braking element is required, e.g. a motor with recuperation could be used. However, we expect that – due to the resulting unusual cycling behavior – this might have negative effects on the user experience/patient compliance.

Since the physiological responses are not only dependent on the mean power output of the cyclist, further analyses are required, e.g. if the system reacts fast enough to limit the amplitude and duration of power output peaks of the cyclist to secure a safe and efficient physical exercise.

In our example with strong increases in power output of the cyclist, we found a constant heart rate during this time, indicating that the system reacted fast enough in this situation for the subject to secure a safe cardio-respiratory strain below given limits. By changing the PI coefficients, the system dynamics can be adapted. Further ongoing work will show if custom setting for individual groups are necessary.

Since subject 2 could keep the distance to subject 1 within a range of 10 m in all test rides, we conclude that cycling in a group of two is possible for different intensity values, within the chosen intensity range and terrain. It is assumed that the control system also works in bigger groups, but more tests are necessary.

4.6 Outlook

4.6.1 Rehabilitation scenario

As a next step, further studies with healthy male and female subjects as well as eligible patients from Phase III / IV cardiac rehabilitation from a local rehabilitation center will be performed to validate and further improve the HEALTHeBIKES prototype in a cardiac rehabilitation scenario.

4.6.2 Compliance

A main obstacle in cardiac rehabilitation is the compliance of the patients. Mag. Heimo Tranninger, head of a local rehabilitation center, stated during the concept development process that simple intuitive behavior and usability should kept in focus despite all technical and scientific possibilities. Furthermore, the prototype was presented at the exhibitions ”Lange Nacht der Forschung” and ”Zukunftstag Steiermark” and a lot of visitors mentioned that simple use is important for them. Therefore, user acceptance tests and an user-centered design are necessary in order to evaluate and improve the prototype. Furthermore, an often requested motivational feature was social interaction. For example an option to share the ride results on a social fitness platform like strava [36] to compare it with friends.

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