Guiding Humans through Wearable Haptics

Gianluca Paolocci

Advisor: Prof. D. Prattichizzo
Co-Advisor: Prof. D. G. Caldwell

Tesi di Dottorato
Ph.D. Thesis

UNIVERSITÀ DEGLI STUDI DI SIENA
DIPARTIMENTO DI INGEGNERIA DELL'INFORMAZIONE E SCIENZE MATEMATICHE

ISTITUTO ITALIANO DI TECNOLOGIA
DEPARTMENT OF ADVANCED ROBOTICS

DOTTORATO DI RICERCA IN INGEGNERIA E SCIENZA DELL'INFORMAZIONE

CICLO XXXIII

Settore Scientifico-Disciplinare ING-INF/04

UNIVERSITÀ DI SIENA 1240

ISTITUTO ITALIANO DI TECNOLOGIA ADVANCED ROBOTICS

Abstract

This manuscript reports the research work done during my Ph.D. on the applications of haptic technology to guide humans, *i.e.* on the design of devices and strategies for instructing human users by means of haptic stimulation. The basic concept presented in this thesis is the exploitation of the tactile channel, that is the most underused but also the most distributed sensory input channel, to provide users with relevant and otherwise unaccessible information, *e.g.* environmental awareness or task-related instructions.

Over the past years, several wearable haptic devices have been developed to stimulate the users' skin receptors and induce a variety of touch perceptions, from texture rendering to temperature and skin indentation. This manuscript investigates applications of the aforementioned haptic interfaces in guidance scenarios, with a particular interest toward the design of haptic patterns to deliver minimal, intuitive and effective cues. Indeed, the haptic policy design process has to take into account that guiding humans is different from guiding robots. Robots can receive an impressive amount of data, process it, and use it to plan and correct motions in an outstandingly short time. Applying the same approach to humans would most probably yield instructions that are difficult to understand and apply, thus leading to poor task performance.

A better understanding of the human physical and mental capabilities is necessary to optimize the communication toward the operators and facilitate their acceptance and trust in technology. For this reason, the first part of this thesis work discloses the background literature on human locomotion, neural entrainment and haptic stimulation.

The dissertation then moves to specific facets of the human guidance mediated by haptics in individual and cooperative scenarios. The second chapter addresses the problem of instructing humans to modify their walking direction and velocity by means of haptic cues, for instance for indoor and outdoor navigation, and explores the topic of sharing tactile perceptions between users applied to a remote social walking experience.

The third chapter presents the developments in human-human cooperation

scenarios mediated by wearable devices, *i.e.* instructing a formation of humans to accomplish a common objective coordinated by haptic stimuli.

The fourth chapter reports two minor projects on haptic guidance. The No-Face Touch system was developed during the current Covid-19 pandemic to support the population by detecting and alerting face-touch attempts. The guidance provided by the system does not instruct specific motions, but leverages the gesture-detection functionalities to notify unwanted behaviors, this way unburdening the users from constantly paying attention to their actions. The latter project proposes a novel approach to Augmented Reality that was designed to minimize the encumbrance on users' hands, so that the augmented experience can comply with different tasks and provide users with support and guidance by leveraging visual and haptic cues.

Contents

1 Background **4**
1.1 Human Walking and Gait Analysis 4
1.2 Neural Entrainment 8
1.3 Touch and Haptic Displays 11
1.3.1 The sense of touch 11
1.3.2 Wearable haptic devices 14

2 Guidance of Human Locomotion **18**
2.1 Suggesting Walking Pace through Haptics under Increased Cognitive Load 18
2.1.1 Motivation 19
2.1.2 System overview 20
2.1.3 Experimental validation 23
2.1.4 Results 26
2.1.5 Discussion 27
2.1.6 Conclusions 29
2.2 Wearable Haptics for Remote Social Walking 30
2.2.1 Motivation 30
2.2.2 System overview 33
2.2.3 Experimental protocol design 36
2.2.4 Experimental validation 40
2.2.5 Correlation analysis 56
2.2.6 Qualitative results and users' feedback 58
2.2.7 Conclusions 60
2.3 Human Rendezvous via Haptic Suggestions 61
2.3.1 Motivation 63
2.3.2 System overview 63
2.3.3 Experimental procedures 64
2.3.4 Live demonstration 65

2.3.5 Discussion . . . 66
2.3.6 Conclusions . . . 67

3 Guidance for Human-Human Cooperation **68**
3.1 Design and Comparison of Haptic Guidance Policies . . . 68
3.1.1 Motivation . . . 69
3.1.2 System overview . . . 71
3.1.3 Experimental validation . . . 77
3.1.4 Results . . . 80
3.1.5 Discussion . . . 85
3.1.6 Conclusions . . . 89

4 Minor projects on Haptic Guidance for Humans **91**
4.1 No Face-Touch: Detecting and Notifying Unsafe Gestures . . . 92
4.1.1 Motivation . . . 93
4.1.2 No Face-Touch system . . . 96
4.1.3 System overview . . . 97
4.1.4 Experimental validation . . . 101
4.1.5 Conclusions . . . 108
4.1.6 Acknowledgments and software repositories . . . 108
4.2 Wristband Display and Haptic Feedback for Mobile Augmented Reality . . . 112
4.2.1 Motivation . . . 112
4.2.2 System overview . . . 116
4.2.3 Experimental validation . . . 124
4.2.4 Results . . . 125
4.2.5 Discussion . . . 127
4.2.6 Conclusions . . . 128

5 Discussions **130**

6 Conclusive Summary **133**

Introduction

Almost everyone driving a car or visiting new places had the necessity and commodity of relying on a smart assistant to find the most convenient way to reach the destination. In these cases, being guided relieves the stress of being lost in an unknown place, and instead of wasting time and effort searching for the right direction we can just enjoy the journey.

Guidance most often means giving (or asking for) information: either the objective is completing a manual task, reaching a specific room in a museum, or knowing if there is a coffee bar around the corner, the user is lacking the necessary information to achieve her/his goal. Moreover, after we receive the coveted instruction, we can decide whether to follow it or to choose an alternative, *i.e.* the agency is still in the user's hands.

From this point of view, the approach adopted for guiding robots is very different. Robots are machines whose control is based on softwares and whose capabilities are hardwired. This means that the system capabilities depend on the specification of the hardware components and on the software used to control it. Both these aspects can be ideally optimized until the top performance are obtained. Robots can receive, send and process huge amount of data in incredibly short time, react immediately to external changes and find the optimal action to address a certain event. This makes the robot the perfect 'being' to guide.

Now, let us consider how different is to guide humans. The amount of information received by humans in each time instant is huge, it may be comparable or higher to the amount of information received by robots. But, differently from robots, the information received are not optimized for the task under execution. Humans, like many other living species, are general purpose, in the sense that their features evolved to cope with the widest range of possible events and environmental conditions. So, unless trained, humans can do everything, but not in an optimal way.

Focusing on the aspect of the information input, we mostly leverage sensory perception to gain knowledge of the environment. Hence, humans are subject to huge amounts of information per second (or per sampling period, since human afferent channels have been shown to have periodic sampling [1]) that are related to different perceptions and brain areas, and often are not linked between each other. Our central system processes these information autonomously, thus humans do not have the agency of deciding what to give priority to, and what to neglect. So, although the information income of a human might even be bulkier than the one of a machine, the content that is really relevant for the current task is not as conspicuous and precise.

A similar concept can be applied to the 'human output'. Given the accuracy of human proprioception , motions performed by humans are not as precise as robot-performed motions. For instance, it is very unlikely that a human can rotate its body by 34° or increase linearly the walking speed. Even keeping a constant walking speed is apparently impossible. In the light of human guidance, instructing a motion by delivering precise joint angles for a complex kynematic chain would yield poor results. Two alternative solutions to the problem are: providing the user with suggestions to correct the pose, or leveraging human experience.

Let us consider the situation in which we want to achieve a certain pose of the upper limb, from shoulder to wrist. Regardless of the communication channel used, the human could be provided with a set of joint angles, ordered from proximal to distal. With an 'open loop' approach, it would be very difficult if not impossible to complete the task. Another possibility would be displaying the joint angle error for each joint, so that the motion can be corrected until the target pose is reached. This approach is more effective than the previous one, because only one parameter per joint is provided, instead of two (current and target value). The closed-loop approach is by definition more precise, but also slower. It is often adopted in robotics, but it might not always be the correct answer for human guidance, especially if we are pursuing dynamic rather than static guidance.

The next possibility is replacing the precise and step-by-step (in this case joint-by-joint) guidance with a more generic approach: exploiting the experience and the naturally efficient motion patterns developed by the humans. In this case, the user is provided with information about the target goal, *e.g.* which point the end-effector should reach, and the human decides the best strategy to reach it. In this case, more than guidance, we might talk about 'augmenting' or 'extending' the human perception, so that the human becomes aware of the goal, that would otherwise be unknown. Considering a practical scenario of picking objects from a shelf, the human might be guided by displaying individual joint angles, or by indicating the object to reach.

Another thing to always take into consideration is the limited amount of resources that can be devoted to specific tasks without affecting the other functionalities. If we try to feed too many information to a human, he might get confused by the overflow of data received, rather than enriched. So, a key aspect is understanding the proper amount of information to convey, and how to do it efficiently.

Based on the previously stated premises, the research work described in this manuscript has investigated two possible approaches to human guidance, namely Motion Guidance and Sensory Augmentation. The first is based on the concept of guidance traditionally used for robots, in which an external entity guides the flow of the task by providing a sequence of instructions. Even though the human maintains the agency, and therefore the last choice on the action to be performed remains with him, this guidance paradigm envisages that the user strictly follows the instructions received. The latter approach, named sensory augmentation, deviates from the literal concept of guidance, but can be considered as a version of guidance designed for humans. In fact, by sensory augmentation we intend to sharpen the perception of the user, allowing him to know information that would otherwise be unknown to him or difficult to identify, *e.g.* the task objective or coworkers' intentions. In this sense, we remain within the boundaries of the guidance topic even though the methodology adopted moves away from that traditionally used by robots, because the experience and motion-planning ability of humans complement the information received to fulfill the task.

The medium used to convey information also plays a relevant role on the effectiveness of guidance. Operators need to understand the commands in order to be assisted, but also have to be aware of the environment. For this reason, it may be convenient to deliver guidance stimuli on channels different from sight and hearing, that can often be busy. Touch is investigated as a principal mean of information-delivery in the presented works, for it is the most diffuse, intimate and underused afferent channel. The technology able to provide users with tactile stimuli is referred to as haptic technology. It has been extensively studied in the past 50 years, and the latest development enabled to display a wide range of tactile sensations on the user's skin, from pressure to temperature. In the scope of human guidance, simple and intuitive cues favour the interpretation of commands without overloading the mental resources. According to this concept, the majority of the projects presented in this manuscript adopts patterns of vibrations to deliver commands to the users, due to the possibily of designing small form-factor devices and providing clear and intuitive indications.

This thesis reports some applications of haptic guidance for humans based on the paradigms of motion guidance and sensory augmentation, with the humble but enthusiast will to take the knowledge about human guidance a little step further.

Chapter 1

Background

This chapter summarizes the basic knowledge on the topics investigated in this manuscript. Foundamentals of human locomotion are necessary to understand and design possible strategies to regulate human walking parameters using external stimuli (Sect. 1.1). The guidance through external stimulation exploits a brain mechanism called Neural Entrainment, that controls the communication between different brain areas (Sect. 1.2). Finally, an insight on touch perception and wearable haptic devices is provided (Sect. 1.3).

1.1 Human Walking and Gait Analysis

Walking is a complex gesture that allows humans to efficiently move their heavy mass through the coordinated work of muscles, that in turn move joints and limbs. The activation of the muscles involved in the act of walking is regulated by patterns of nerve impulses produced by the 'Central Pattern Generator', that consists of newtorks of neurons in various parts of the brain and spinal cord. The current hypothesis on the mechanism of human walking regulation envisions a rhythm-generating system within the spinal cord that is controlled by neural input from the Central Nervous System and receives feedback from sensors in the muscles, joints and skin of the legs [2]. Despite the complexity of this motion, it is repetitively performed with little effort by our body.

The cyclic nature of human walking is studied by the Gait Analysis. It investigates various aspects of locomotion, focused toward both normal and pathological conditions. In this manuscript, only the normal gait will be considered. Between the investigated parameters in gait analysis, the most relevant are the angle, angular velocity and acceleration of joints, that describe the body pose and movement through the kinematic chain of the considered body segments [3];

force and torque at the joints, that describe the force exerted by muscles to enable motion, and allow to estimate the strain on muscles, tendons and ligaments [4]; ground reaction forces, that describe the motion of center of mass and center of pressures during walking. Other interesting parameters are the electrical signals measured during muscle activation, and measures of the energy expenditure, that allow to analyse, for instance, the possible mechanism of the body to minimize energy consumption [5] and energy transfer between the body segments in walking [6].

Walking is a pseudoperiodic action, as it exhibits regular patterns of motion for the lower limbs that recur step after step. The first description of the human gait cycle as it is intended in modern literature dates back to the beginnings of 1800; it reports accurate measurements of the timing of gait, and describes the pendulum-like swinging of the legs [7]. The further technological advancement provided gait analysis researchers with tools to quantitatively describe the gait cycle. In 1870 the movement of body limbs was literally imprinted on 'paper' using multiple photographic exposures, with the subjects wearing illuminated stripes on black clothes. The advent of force platforms, stereo-photogrammetry, systems for Electro Myo-Graphy (EMG), *etc.* promoted the advancement of gait analysis with more accurate measures. In recent years, the adoption of inertial sensors extended the gait analysis also to non-laboratory settings.

Above the numerous parameters investigated in gait analysis, we are interested in the structure of the gait cycle. Basic concepts and definitions are reported as follow. Human walking can be defined as 'a method of locomotion involving the use of the two legs, alternately, to provide both support and propulsion' [8]. The gait cycle indicates the periodic series of actions between two successive occurrences of one of the repetitive events of walking. For convenience, the beginning of the gait cycle is usually identified with the 'initial contact', that is the instant in which one foot touches the ground (after the swing phase), because it is easy to identify both visually and through measures. The two feet go through the same motions, except that their timing is shifted of half a gait-cycle period. The cycle ends when the reference foot contacts the ground again. The following terms are used to identify major events during the gait cycle:

1 Initial contact

2 Opposite toe off

3 Heel rise

4 Opposite initial contact

5 Toe off

6 Feet adjacent

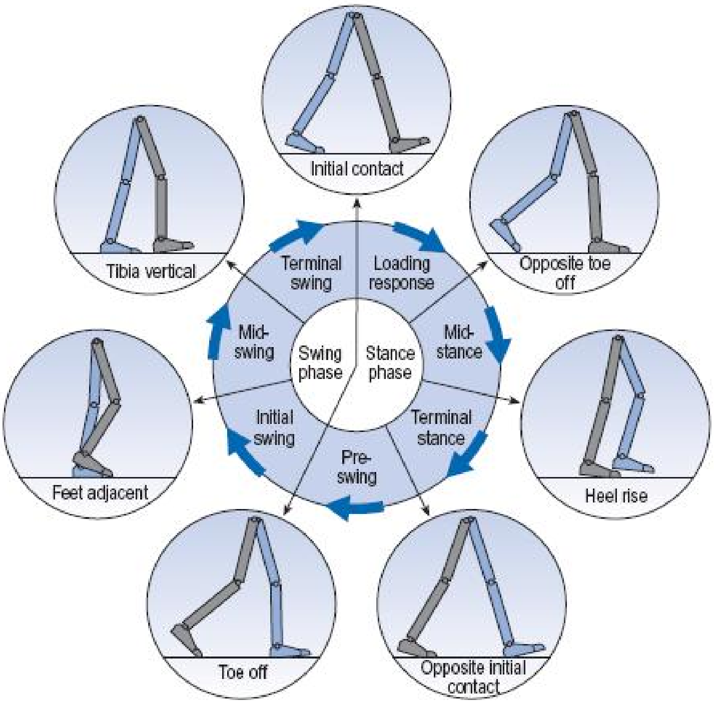


Figure 1.1: Gait cycle phases. The leg coloured in grey is the reference leg. It is visible that the movements of the two legs are phase-shifted of half a cycle.

7 Tibia vertical

(1) Initial contact

The names adopted for the 7 phases refer to the position of feet during the cycle (see Fig. 1.1). Another classification considers the gait cycle as composed of a stance phase, during which the reference foot is in contact with the ground, and a swing phase, during which the foot moves forward through the air. The stance phase is also called support phase, and it lasts from initial contact to toe off.

The temporal duration of a whole gait cycle (*e.g.* considering the right leg) is defined as cycle time or stride duration, that comprises a stance and a swing time. During the initial contact of the right foot, the left is still on the ground. Until the left foot leaves the contact with the ground (opposite toe off), there is a period of double support. After the opposite toe off, only the right foot contacts the ground until the next opposite initial contact phase. This phase is called single support. After the single support phase, there is another period of double support, that ends with the right toe off, followed by the left single support phase. The single support phase is referred to as swing phase, if the considered leg is the one not touching the ground. A gait cycle is composed of two single support and two double support phases, that approximately last 40% and 10% of the cycle time, respectively. During the double support phase, the leg placed forward is called 'leading leg', while the leg behind is called 'trailing leg'. At the beginning of the double support, the body weight shifts from the trailing to the leading leg. This phase is also called 'loading response', or 'weight acceptance'. The trailing leg instead is in 'pre-swing', as it prepares to toe off and leave the ground.

The stride length is the distance between two successive placements of the same foot. It consists of two step lengths, left and right, each of which is the distance by which the named foot moves forward in front of the other one. The number of gait cycles performed in a period of time is defined as stride cadence, measured as stride/min or stride/s. The inverse of stride cadence is the stride duration, that measures the duration of a single gait cycle. Sometimes the step cadence and step duration are described as a temporal measure of walking. Althoug the two measures are intrinsically linked, the step accounts only for half of the gait cycle, *e.g.* from intial contact of the right leg to initial contact of the left leg (or any other corrisponding pair of walking phases). As a consequence, in symmetric walking the step cadence is about double the stride cadence, while the step duration is about half the stride duration. The speed of walking instead is the distance covered by the whole body in a given time, measured in meters per second. Since walking speed depends on both cadence and stride length, it follows that speed may be changed by altering only one of these variables, for instance by increasing the cadence while keeping the stride length constant. In practice, however, people normally change their walking speed by adjusting both cadence and stride length.

Humans are good in naturally selecting the optimal stride length and cadence for a given walking speed. When people vary their walking speed, they modify the stride duration and cadence simoultaneously. Several works investigated the relationship between walking speed, stride duration (or cadence) and stride length, but there is not a conclusive answer yet. Many researches have shown that variations in walking speed have differential effects on both frequency and length of stride. In [9,10], an increase in speed up to about 1.2 m/s was obtained by increasing the stride frequency and stride length, but to reach greater speeds subjects mainly increased frequency. Later researches reported different stride length-duration relationships, that were linked to different experimental protocols and settings, for instance walking on a threadmill rather than on plain ground. Moreover, it was demonstrated that constraining either the stride duration (*e.g.* delivering a walking tempo with a metronome) or the stride length (*e.g.* with visual indications on a threadmill belt) changes the stride length-duration relationship observed during unconstrained walking [11].

1.2 Neural Entrainment

One of the most relevant research question in this manuscript is the regulation of human cadence by means of external stimuli, specifically using haptic cues. Can humans align their movement, in particular their steps, to an external rhythm? If they can, is the process unvoluntary or does it require the user to focus on it?

To answer these questions, it is necessary to understand how our body uses and responds to rhythms. It is known from a long time that brain activity exhibits rhythmic behavior. Networks of neurons communicate with each other through periodic patterns of activation, and are capable of generating rhythmic activities such as breathing, heart-beating, and even more complex actions such as walking or producing speech. Rhythmicity is present in the control of motions (both voluntary and unvoluntary), but plays an important role also on sensory perception, by increasing or lowering the excitability to specific external stimuli [12]. The rhythms of motor production and sensory perception share similar frequencies, and more often than not influence each other. This feature facilitates coupling between the motor and sensory systems [13, 14]. One of the fundamental mechanisms underlying the regulation of rhythmic behaviors is Neural Entrainment, that is defined as 'the alignment of one or more oscillating systems to an external rhythm, whereby the interactions are unidirectional, that is, the external rhythm influences the oscillating system(s) but not vice versa'. Simply put, entrainment enables or disables the communication between two systems by leveraging phase-reset mechanisms, that consist in forcing a system oscillating at a 'self-selected' frequency into a specific phase by means of an external stimulus. The concept of entrainment as a tool for regulating the excitability of networks of neurons implies the presence of periodic events (that are by definition predictable) working at specific frequencies. If the brain is interested in the 'external stimulus' under consideration, it promotes the activation of neural networks working at the same frequency of the external stimulus to sharpen its sensing capabilities while suppressing areas that feature different frequencies. On the other hand, if the brain wants to reject the source under consideration, it suppresses the activity in areas oscillating at the frequency of the stimulus. In this light, entrainment can be considered a tool for our brain to filter information, *i.e.* by suppressing non-relevant inputs while giving priority to others on the base of brain predictions, goals and expectations.

Entrainment is top-down controlled: if several inputs are available, the strongest input stream entrains oscillations, while the entrainment of the others is weaker, *i.e.* the most relevant information are favoured against the less important. It follows that in controlled conditions, such as laboratory experiments, it is possible to entrain subjects' brain waves to different kind of external stimulation rhythms if they are not exposed to other rhythmic stimuli, *i.e.* in environments with minimal competition. On the other side, the entrainment to the same source may not apply in real-life scenarios where the amount of information available for the central nervous system to process is huge and the attentional resources are limited.

Many research directions on neural entrainment are currently available in literature. For interested readers, a comprehensive review is presented in [15]. For the purpose of this manuscript, we will focus on the external sources that entail the neural entrainment:

- **Environmental rhythms**: those rhythms are external and are grasped via sensory perception. For instance, entrainment to stimuli that reach the receptors, as auditory cues [16].

- **Self-produced rhythms**: can be produced voluntarily or unvoluntarily. Voluntary rhythms are mostly motor, *e.g.* saccades movements for active sensing [17]. Unvoluntary rhythms are for instance the ones associated to breathing or heartbeating.

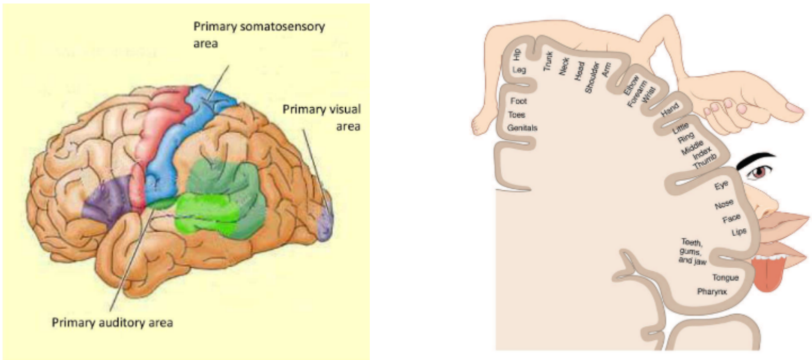
- **Rhythmic neuromodulation**: the entrainment can be artificially induced using laboratory techniques, *e.g.* transcranial electric and magnetic stimulation, for research and therapeuthic purposes.

Often, the different sources of neural entrainment interact with each other: external rhythms perceived through sensory input can influence motor patterns, that in turn increase the perceptual sensitivity. If neuronal oscillations are entrained by an external rhythm, other processes can align to the same tempo, that acts as a central timing mechanism. For instance, when tapping to the rhythm of music the processes that regulate the motor control of tapping entrain to the rhythm of the song. Conversely, when playing the piano the brain patterns produced by motor processes entrain the auditory system areas, that enhance the sensitivity to the produced music.

The effects of external rhythms on motor control have been studied extensively. Between the different sensory perceptions, hearing has the more strict coupling with the areas deputed to motor control. Since the first studies on the effects of rhythmic audio stimulation on motor control, in which finger and arm movements were entrained to the period of cues from a metronome [18], the scientific community confirmed the strong connection between the two systems. The auditory system has been found to have distributed connections to motor centers both on spinal cord and brain level [19]. Entrainment also applies for visual and tactile perceptions. Unfortunately, it is more difficult to find in nature visual or tactile phenomenons that exhibit rhythmic behaviors at the same frequency of brain waves. On the other side, laboratory settings allow to artificially generate the aforementioned stimuli. Visual stimuli at the frequency of 10 Hz have been provided to participants in [1]. Not only participants' brain-wave were showed to entrain to the visual stimuli, but also the entrainment affected the discrete sampling in visual system. The entrainment to audio and visual stimulation was observed for stimuli at the fixed frequency of 10 Hz and for small fluctuations (± 1 Hz), while the effect ceased to be seen for larger frequency variations (± 2 Hz and above). The entrainment to a slowly varying frequency was also observed in [19], where the entrained frequency stayed locked to the external tempo also after subtle and imperceievable variations. Periodic tactile stimulation also entrain brain rhythms. For instance, tactile sinusoidal stimulation has been shown to influence participants' Sensory Motor Rhythm and Alpha signal [20].

On practice, entrainment has been used in a wide range of therapeutical or task-assistance applications. Audio-visual cues delivered at specific frequencies have been found to have different effects based on the stimulation tempo: for instance calming hyperactivity, increasing attentional and emotional control, and regulating heart-rate variability [21]. Audio, visual and tactile cues delivered at the tempo of brain waves (1.5 -20 Hz, adjusted to individual alpha peak) were shown to reduce the arterial pressure in patients with pharmacologically uncontrolled essential hypertension when compared to subjects in sham stimulation condition (under the same pharmacologically uncontrolled condition, acted as control group) [22].

For what concerns locomotion, the entrainment to audio cues has been adopted as therapeutical method to improve the functional control of movements in healthy subjects, and also in subjects with stroke or Parkinson Disease [23]. The stimulation enabled the regulation of stride length, cadence and stride symmetry in subjects with large walking asymmetries [24]. Several researches demonstrated that humans can match their step cadence to the tempo of auditory or visual cues while walking on the ground or on a threadmill at a fixed speed [11, 25]. Recently, tactile stimulation has been adopted as an alternative to auditory and visual cues for regulating walking cadence, to reduce the overload on the aforementioned input channels [26, 27].



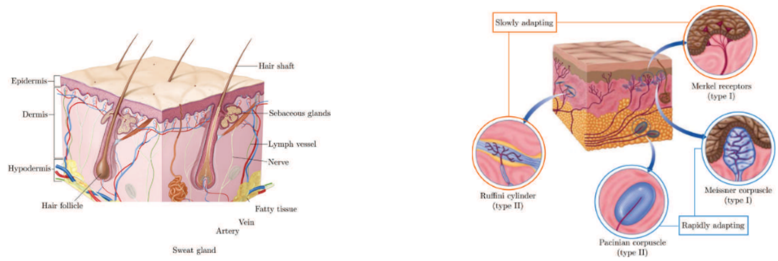
(a) (b)

Figure 1.2: Sensory regions. (a) The image subdivides the sensory regions to show regions that receive input from somatosensory, auditory, and visual receptors. (b) A cortical homunculus, a distorted representation of the human body based on a neurological map of the areas and proportions of the brain dedicated to processing sensory functions (vision, touch, etc.)

1.3 Touch and Haptic Displays

1.3.1 The sense of touch

Touch is probably the most stimulated among our senses and the one that we pay least attention to. Through touch we can explore textures, estimate temperature, distinguish shape, and balance our body. All these actions involve different types of touch. It is important to notice that often these sensations are in concert and it is the integration among them, in the central nervous system, which really tells us about the nature of the stimulus. In fact, sensory information from receptors displaced on the body enters the central nervous system by the dorsal horns of the spinal cord. To have a good sense lots of receptors, cells and nerves are involved. As a consequence, having several nerve endings active requires that the brain reserve more space to process all of the information from the nerves. This implies that the area of our brain devoted to sensing fur on our fingertips is much larger than that responsible for other sensing. An interesting representation of this inequality among the senses is depicted in Fig. 1.2. Interested readers can find more details in [28, 29]. Let's continue with an example. If we close our eyes and grasp an object, we feel something in our hand. The skin receptors distinguish the edges and corners. Then, we can appreciate textures and temperature. The tension we need to put in fingers, hand, and arm muscles to counter gravity



(a) (b)

Figure 1.3: Human skin. (a) Main layers and components. (b) Types of mechanoreceptors of the skin. The main characteristics of each mechanoreceptor (*i.e.*, size of the receptive fields and the rates of adaptation) are also shown.

tells us more about the size and the weight of the object. This happens thanks to the skin, a host of receivers that we use to feel things. The skin is the communication channel between our internal body systems and the outside world. It is the body largest organ; covering the entire outside of the body (1.5-2.0 m^2) and being about 2-3 mm thick, it accounts for about 15% of the total body weight in adult humans [30]. Any active or passive contact between the skin and an object/environment is turned into a collection of information, *e.g.*, temperature, pain, and pressure, that allows the brain to explore and know the world around us. As shown in Fig. 1.3a, skin is organized in three layers, including, from top to bottom: (i) epidermis, made of dead skin cells, which provides a waterproof barrier and creates our skin tone; (ii) dermis, which contains nerve endings, hair follicles, sweat glands, sebaceous glands, and touch receptors. Its primary function is to sustain and support the epidermis by diffusing nutrients to it and replacing the skin cells that are shed off the upper layer of the epidermis; (iii) hypodermis, which is made of fat and connective tissue. It mainly acts as an insulator and helps regulate body temperature. It also acts as a cushion to protect underlying tissue from damage. Our sense of touch is controlled by a huge network of nerve fibers and different receptors in the skin, known as the somatosensory system. This system is responsible for all the sensations we feel. Four main types of receptors compose the somatosensory system: mechanoreceptors (pressure and surface texture), thermoreceptors (temperature), nociceptors (pain), and proprioceptors (muscle length, muscle tension, and joint angles). Because of the scope of this thesis, we briefly present the receptors, focusing only on those which concerns pressure and vibration.

Mechanoreceptors in the skin have traditionally been distinguished on the basis of the types of stimulation to which they respond, the size of their receptive fields, and their rates of adaptation [31]. They can be classified as rapidly adapting, if they respond to a change in stimulus very quickly. They can sense right away when the skin is touching or stops touching an object, but they cannot sense the steady-state contact of a tactile stimulus. These receptors are the ones dedicated in perceiving rapid and regular changes in pressure, such as vibrations. On the other hand, receptors can be classified as slowly adapting if they do not respond to a change in a stimulus very fast, but they are good at sensing the continuous pressure of an object touching or indenting the skin. Based on to the size of their receptive field, mechanoreceptors can be further divided in type I (small) when their receptive field is about 2-8mm in diameter, and type II (large) when their receptive field is larger. Each tactile unit is composed of an afferent fiber and its ending. There are four different endings in glabrous skin whose only function is to perceive indentations and vibrations: Merkel cells, Meissner's corpuscles, Ruffini endings, and Pacinian corpuscles (see Fig. 1.3b). Based on morphological observation (receptive field characteristics, adaptive properties of the fibers to stepwise indentation, and frequency response to sinusoidal vibration), attempts have been made to associate different sensing characteristics to different receptors:

- Meissner's corpuscles - they are ovoid structures located in the dermis of glabrous skin and they are most sensitive in the 20-40Hz range of vibration. They are 40-180m long and their density is high on the fingertips (15-24 /mm^2). They are typically associated with rapidly adapting type I receptors [32].

- Merkel cells - they are in the deepest layer of the epidermis of glabrous skin and they register the pressure. Their density is estimated to be 80 /mm^2 and they are typically associated with slowly adapting type I receptors [33].

- Ruffini endings - they are spindle-shaped structures located in the deepest part of the dermis of both hairy and glabrous skin. They respond to pressure on skin and are very sensitive to pressure variations. They are typically associated with rapidly adapting type II receptors [34].

- Pacinian corpuscles - they are multilayered, onion-shaped structures located deep within the dermis and the subcutaneous fat layer of both hairy and glabrous skin. They are most sensitive in 200-550Hz range of vibration and they are long enough to be visible to the naked eye (1-4mm). They are typically associated with slowly adapting type II receptors [35].

Although mechanoreceptors are distributed all over the body, mostly they are concentrated in non-hairy skin which is used in interactions with the surrounding

(a) (b) (c)

Figure 1.4: Three thimble cutaneous devices designed with different mechanical characteristics (a) 3-DoFs device featuring a mobile platform driven by three cables via DC-motors; three force sensors attached to the mobile platform are used for a closed-loop force control; (b) Wearable fabric-based display for multi-cue delivery. Two DC-motors move the soft belt; (c) 3-DoF device capable of displaying interaction forces by skin stretch. Three rigid links are controlled using three servo motors.

environment, such as the palms, lips, tongue, soles of feet, fingertips, eyelids, and face.

1.3.2 Wearable haptic devices

Several types of cutaneous haptic devices have been developed over the past several years, they spread from pressure-based to vibration and indentation feedback. Here we provide a brief overview of those devices using tactile technologies.

Pressure-based devices About 15 years ago Minamizawa *et al.* [36] laid the foundation developing a wearable cutaneous device able to reproduce the weight sensation of a virtual object. It consists of a pair of motors and a belt; when the motors spin in the same direction, the belt applies a tangential force to the skin, whereas when the motors rotate in opposite directions a vertical force is generated on the user's fingerpulp. A step forward in the research of wearable haptic devices was presented by Pacchierotti *et al.* in [37]. They presented an innovative wearable device that can be worn on the proximal phalanx, and is able to display normal and shear forces. The device leaves the fingertip uncovered and free to interact with the surrounding environment. The missing feature of this device is the capability of reproducing surface exploration. A partial solution came from Prattichizzo *et al.* [38]. They developed a wearable 3-DoFs fingertip device consisting of two platforms: one is located on the dorsal part of the finger which supports three micro DC motors and the other one is in contact with the finger pad (see Fig. 1.4a). The motors, by controlling the length of three cables, move the platform towards the user's fingertip and re-angle it to simulate contacts with arbitrarily oriented surfaces. Three force-sensing resistors near the platform vertices measure the fingertip contact force for closed-loop control. Pacchierotti *et al.* [39] presented an improved version of the same device equipped with position encoders and only one sensor, that was able to achieve higher accuracy. With such technology any shape can be displayed to the user. A different approach has been used by Bianchi *et al.* [40] who presented a wearable fabric-based tactile display for multi-cue delivery (see Fig. 1.4b). Multiple levels of stiffness can be reproduced by modulating the stretching state of a fabric through two DC motors. An additional vertical degree of freedom is implemented through a lifting mechanism, which can be used to convey a sensation of softness to the user's finger pad. Gleeson *et al.* [41] fabricated a 2-DoF fingertip-mounted cutaneous device to render tangential skin displacement at the fingertip using a 7mm hemispherical tactor. Its two radio controlled servo motors and compliant flexure stage can move the tactor along any path in the plane of the finger pad. This device is capable of rendering 1mm of displacement at arbitrary orientations within a plane, with a rate of 5mm/s. Its direction cues have been used for human mobile navigation in an unknown space [42]. Leonardis *et al.* [43] developed a wearable 3-DoF device with compact dimensions and light weight, capable of displaying both contact no-contact transitions and interaction forces by skin stretch (see Fig. 1.4c). The asymmetric rigid parallel kinematics reduces the mechanical interference with the other fingers allowing a multi-contact manipulation of virtual/remote objects. Pacchierotti *et al.* [44] presented a 3-DoF cutaneous device for remote tactile interaction. The revised design and the addition of springs with respect to the one presented in [38] make it possible to provide the sensation of breaking and making contact with remote objects. Although all these devices enhanced the rendering of virtual/remote objects, in spite of the way cutaneous feedback was provided, most of them have been utilized in frameworks with interactions as a single point force, rather than spatially distributed sensations [45,46]. On the one hand, such a simplified approach makes it possible to reduce the number of input parameters, actuators, force sensing systems, and, thus, the size and the weight of the device. On the other hand, together with the devices form factor, the range of stimuli provided is drastically simplified with respect to the sensations experienced during a real tactile interaction. Although satisfying results can be achieved this way in practical implementations, a very accurate delivery of the desired force on the user's fingertip cannot be guaranteed.

Vibration-based devices In addition to the aforementioned devices, there is also a growing interest in vibrotactile cutaneous feedback. Although pressure-based devices achieve good performance in rendering the sensations of touching a remote object, most of them limit the mobility of the hand and its interactions with the environment. Moreover, they are developed mainly for the hands; different body parts (legs and arms for instance) require other technologies for displaying cutaneous sensations. In several studies, wearable vibrotactile devices have been placed on different parts of the body and used as a means of communication. Aggravi *et al.* [47] employed vibrotactile bracelets to guide older adults with minimal impact on their freedom of motion. The authors took into account the effects of aging on the sense of touch while designing the haptic bracelet. Cosgun *et al.* [48] placed vibrating motors on a wearable belt and carried out a preliminary evaluation study of vibration patterns that can provide navigation assistance for visually impaired people in cluttered indoor environments. They defined two main classes of vibration patterns depending on the type of intended human motion: directional and rotational. Jones *et al.* [49] designed a wireless tactile display that consists of a 4x4 array of vibrating motors that is mounted on a waist band and stimulates the skin across the lower back. The device performance was evaluated with eight different vibrotactile patterns that could be identified by the subjects with almost perfect accuracy. An experiment confirmed that the tactile display can be used as a navigation aid outdoors and that the vibrotactile patterns presented can be interpreted as directional or instructional cues. A similar device, the ActiveBelt, was proposed by Tsukada and Yasumura [50]. In this case the haptic belt was integrated with a directional sensor and a Global Positioning System (GPS), and employed as an intuitive navigation system. Apart from the use for navigation purposes, vibrotactile feedback has gained popularity in recent developments of devices for therapeutic applications. Alahakone and Senanayake [51] proposed a real-time biofeedback system to improve postural control and stability in clinical populations. Hung *et al.* [52] examined the effect of real-time corrective vibrotactile feedback for training bilateral reaching motions for stroke rehabilitation in the absence of therapist supervision. They evaluated a bilateral upper-limb motor learning system, including a wireless wearable sleeve-armband device for providing vibrotactile feedback, on hemiparetic stroke survivors and able-bodied people. The results indicated the potential efficacy of vibrotactile cues for unsupervised motor learning in both the healthy and the stroke populations. Wall [53] exploited a 6-DoF motion sensor (3 linear accelerometers and 3 gyroscopes) to estimate trunk tilt. An array of 48 tactors, that rings the torso, feeds back this tilt information, magnitude, and direction to the subject. Results show that vibrotactile feedback about body inclination can be used to help control body motion under a variety of conditions and tasks. Vibrotactile feedback is useful also for rendering specific properties of remote/virtual objects. Kyung *et al.* [54] designed a haptic mouse able to provide texture signals, such as patterns, gratings, and roughness. A small planar-distributed pin array was embedded into a commercial computer mouse to highlight the ease of integration of the developed interface with existing systems. Romano *et al.* [55] presented a new method for creating detailed haptic texture models from data recorded during natural users interactions with textured surfaces. An accelerometer and a force sensor mounted on a customized stylus record vibrations and the applied normal force while exploring a surface. A computer then generates synthetic texture signals in real time, and small voice-coil actuators mounted on the stylus render resulting vibrations to the user's hand. More recently, Asano et al. [56] developed a wearable display system that modifies the perceived roughness of textured surfaces. This device, that mainly consists of a voice coil actuator, stimulates the activity levels of tactile mechanoreceptors in order to increase and decreases the roughness felt by the user.

Chapter 2

Guidance of Human Locomotion

This chapter presents my research work on haptic stimulation applied to the guidance of human locomotion. Three scenarios are presented, that address different but interlaced aspects of walking: *i*) design choices on haptic interface placement and cognitive load aspect (Sect. 2.1), *ii*) communicating walking pace between two partners for remote cadence synchronization (Sect. 2.2), *iii*) providing timing and directional cues in structured environments (Sect. 2.3).

2.1 Suggesting Walking Pace through Haptics under Increased Cognitive Load

This section reports the study on the effects of haptic interfaces placement in a walking cadence regulation scenario, with the further aim to assess if the users can synchronize to the suggested rhythm with low efforts while performing other tasks. Elastic haptic bands are used to suggest walking-pace during experiments aimed at reproducing navigation or human-robot cooperation tasks. The proposed system consists of two wearable interfaces for providing timing information to the users, and a pressure sensor to estimate the real gait pattern, thus resulting in a combination of walking-state monitoring and vibro-tactile stimuli. Vibrational stimuli with a constant presentation interval are alternately and repeatedly given to the right and left side of the human body, in accordance with the desired walking cadence. We tested two different interface placements: wrists and ankles. The cadence regulation system was also evaluated under the condition of incremented mental and manual workload by introducing a secondary task, *i.e.* balancing the motion of a small sphere on a flat surface, while walking at the rhythm suggested by the haptic interfaces. Experimental results on the stimulation site revealed that providing indications at the ankle has a greater effect on promoting the synchronization to the haptic rhythm displayed, with respect to stimulating the wrists. Participants reported that receiving indications at the ankle rather than at the wrist felt more comfortable and task-compliant. Representative applications of the proposed approach in daily use are training and coaching in sports, rehabilitation, and human-robot cooperation and interaction.

2.1.1 Motivation

Nowadays there is growing interest in technologies and methods that assist people during daily activities, for instance by relieving mental or physical efforts from the user and by simplifying tasks through the enhancement of context awareness. The scenario considered in this section is pedestrian navigation. Many of the existing solutions rely on vision or hearing to guide the user: related works demonstrated that walkers are able to synchronize to auditory and visual cues [11]. Downsides of these approaches are the increased attention demand and channel overloading, that may conflict with daily tasks due to the limited resources availability [57,58].

An alternative solution to reduce cognitive load consists in replacing the audiovisual cues with stimuli involving other senses. This prevents channels from saturating and lowers the overall mental efforts [59]. Our methodology exploits a feature of the human sensory-motor system, called neural entrainment, to suggest a specific walking cadence [60,61]. It is known that the frequency of a cyclic movement, such as walking and running, can be affected by rhythmic sensory inputs and can smoothly converge to the input rhythm. For example, when people walk while listening to music, their step cycle gradually conforms to the rhythm of the music. Recent works highlighted how haptic stimuli can be used to deliver walking cadence with minimal interference to other sensory channels, which might lead to better user safety or task execution [26].

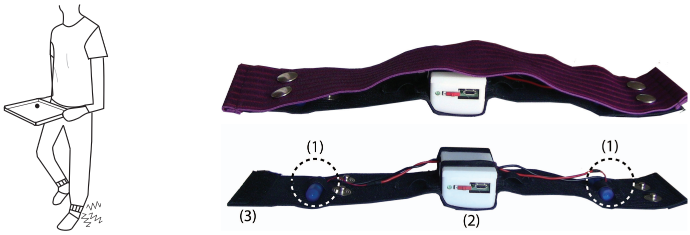
Haptic communication offers an effective, yet non-intrusive, way for providing cues to the users when vision is temporarily impaired or hearing is overloaded by background noise. Recently, several systems based on haptic technology have been developed, most of which focus on providing stimuli mainly via bracelets and waist belts. For instance, a torso-mounted vibro-tactile display was used to provide cues for improving situational awareness of soldiers in a simulated building-clearing exercise [62]. In [63] and [64], a vibro-tactile belt was used for human guidance in indoor and outdoor environments, respectively. In [65], the authors used vibro-tactile armbands to guide users along a predefined path, assisted by a mobile robot. In [59] the authors exploited haptic stimuli for indoors pedestrian guidance using two wrist-worn interfaces. Vibro-tactile armbands were used to navigate subjects along fixed and dynamic paths [66, 67], using three basic haptic cues to steer the walking direction. Adame *et al.* in [68] proposed a comparison among different vibro-tactile devices for guidance of visually impaired users. Most of the contributions in literature focused on how to suggest a given direction to the users, or how to steer humans along a certain trajectory. An often undervalued important parameter to guide locomotion is the time to reach the target *i.e.*, the walking speed. Haptic interfaces placed at the subjects feet were used to regulate gait frequency through vibrations in [69]. In [27] the authors presented an interesting vibro-tactile guidance method to suggest cadence to users by means of haptics. An exemplar application is guiding subjects toward the closest bus stop at the optimal walking speed. Also post-stroke rehabilitation benefits from periodic vibro-tactile cues, that were shown to have a greater effect in increasing and regulating subjects' step length, compared to audio and visual indications [70].

For what concerns our contribution, we investigated whether placing the haptic interface on the wrists or on the ankles affected the participants' capability in synchronizing their walking cadence to the suggested rhythm. Then the same experiment was conducted while performing a secondary task aimed to decrease the attentional resources available to the users. The objective of the latter study was determining the most suitable stimulation location (ankles or wrists) to reduce the overall effort of the joint task.

2.1.2 System overview

The proposed system is composed of two parts: a haptic interface in charge of providing haptic cues to the user, and a pressure measurement device, used only for experimental testing and validation, that detects contacts between the foot and the ground to compute the user's stride duration. In what follows we describe the two components of the system.

Haptic bands The desired cadence is suggested to the users through rhythmic vibrations provided by remote-controlled elastic haptic bands. Each wearable haptic interface is composed by two water-proof vibro-motors, which can be independently controlled (Fig. 2.1). Whenever a trigger is sent to a haptic device, the motors vibrate providing the wearer with a vibro-tactile stimulus. In order not to overload the user's tactile channel, and to improve the intuiteveness of the haptic information, we decided not to modulate the frequency of the signal. Instead, we use a simple on/off mechanism, similar to the one used by Scheggi *et al.* [71]. The two interfaces are alternately activated in accordance with the reference stride duration. An additional vibratory pattern was implemented to indicate the end of the trial, by activating three times both the haptic devices



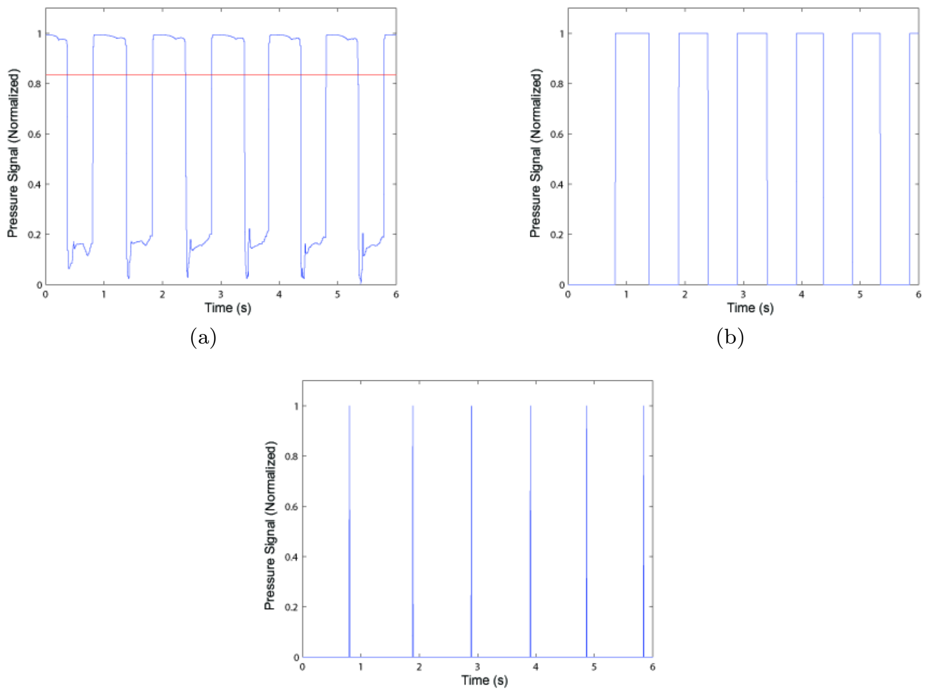
(a) (b)

Figure 2.1: Cadence cues are provided to the users via vibro-tactile elastic bands placed at the wrists or at the ankles (a) of the user during a task with manual and cognitive load. The haptic interfaces (b) are composed of two vibrating motors (1) attached to an elastic band (3). The Arduino board in the 3D-printed case controls the interface and communicates with the laptop via BlueTooth (2).

simoultaneously. When an interface is activated, its motors vibrate for 0.1s at a frequency of 250Hz.

The communication with the laptop on the master side uses a RN-42 Bluetooth antenna connected to a 3.3V Arduino Pro-Mini. The wireless connection baud rate is 57600bps. The microcontroller installed on the board controls the activation of each motor and communicates with the laptop. The vibro-motors are voltage-controlled, which determines both frequency and amplitude. As a consequence, we cannot change frequency and amplitude independently. As the human maximal sensitivity is achieved around 200-300 Hz [72] (the human perceptibility range is between 20Hz and 400Hz), two Precision Microdrives Pico Vibe vibration motors are placed into two fabric pockets inside the bracelet (the width of the wristband is about 60mm), with vertically aligned shafts. The motors have a vibration frequency range of 100-300Hz, lag time of about 20ms, rise and stop time of 35ms. The bracelet guarantees about 4 hours of battery life with one motor always turned on. Each bracelet weights about 90g.

Pressure Sensor The force sensor placed under the right heel records the force exerted by the foot on the ground during the heel strike, that is used to extract the user's step temporal sequence. The sensing device is composed by a flexible force sensor (FSR 400, manufactured by Interlink Electronics, Inc.) and a XBee radio module. The force sensing resistor measures the force applied through the deformation of the active surface, which produces a resistance variation. We use this component as unobtrusive and comfortable switch to detect the contact of the shoe with the ground. The XBee module converts the analog voltage recorded



(a) (b) (c)

Figure 2.2: The foot pressure on the sole is recorded by a flexible force sensor (FSR 400), and transmitted to the laptop through a XBee radio module. The raw signal showed in (a) is normalized between 0 and 1. Steps are extracted by thresholding the raw signal (b), and only considering the positive edges (c).

into a digital signal and transmits it to the module connected to the laptop. The pressure value is converted into a 10 bit digital signal. Steps are extracted using a single-threshold value, defined as double the standard deviation of the pressure data (measured during the initialization phase, see Subsect. 2.1.3). The sampling frequency adopted for the pressure sensing is 100 Hz.

The step-detection procedure consists of three phases. Raw pressure data received by the master side are normalized (Fig. 2.2a), then transformed into a two-levels signal using a custom threshold (Fig. 2.2b). The square wave indicates whether the foot is in contact or not with the ground, assuming value 1 or 0, respectively. Then, the algorithm extracts positive edges, identifying the stride duration as the interval between two consecutive edges (Fig. 2.2c). We indicated as *stride duration* the time between the occurrence of two consecutive steps of the same leg, and as *stride length* the distance covered between two subsequent steps of the same leg.

2.1.3 Experimental validation

First, we evaluated the proposed walking-pace suggestion paradigm under two different conditions, *i.e.* placing the haptic interfaces at the wrists and at the ankles. Then the addition of a secondary task was used to compare the mental and manual workload of the two interface-placement conditions.

Preliminary test We performed preliminary tests using the haptic interfaces either as bracelets or anklets. Eight healthy subjects took part to the experiments. They were informed about the protocol and were free to discontinue participation at any time. During the trials, participants were instructed to synchronize their steps to the vibrations received in a 220 m path. The experimental protocol consisted in three trials per each interface-placement condition, accounting to 6 total trials per participant. The three trials were conducted using three different stride duration references: 0.8, 1.0, and 1.2 seconds per stride, respectively. The stride duration references were defined according to results of preparatory experiments, that reported stride durations in the range 0.8 - 1.2 s. The three references represent the faster, average and slower cadence detected, respectively.

Concerning the trial order, reference stride duration and body location conditions were pseudo-randomly selected to avoid biases in results. During the trials, participants wore headphones reproducing white noise to mask the sound of vibro-motors. The metric evaluated in the post-processing phase was the error between actual stride duration and reference rhythm. It was calculated as the average absolute value of the time discrepancy between current stride duration and reference, normalized with respect to the reference stride duration:

$$error = \frac{1}{N}\sum_{k=1}^{N}\left|\frac{u(k)-d(k)}{d(k)}\right| \times 100\%.$$

Where N is the number of strides walked during the trial, u_k and d_k are the duration of the $k-th$ stride and the desired time, respectively. In Subsect. 2.1.4 we report and discuss results with statistical tests.

Cognitive load The effects of a secondary task on participants' step-synchronization capabilities were investigated in a subsequent experimental campaign, with the aim of comparing the two interface-placement conditions.

The secondary task was introduced to assess whether compromising the attentional resources available to participants affected the synchronization performance in the two interface-placement conditions. We hypothesized that the interface-placement resulting in better performance after the addition of the secondary task would have been more suitable to convey walking cadence suggestions, due to the lower efforts required to catch and process the information while performing other activities. During the protocol design phase we selected three different secondary tasks that subjects were asked to perform together with the synchronized walking task. Each of the three secondary task, listed as follows, was tested in separate pilot experiments.

i) Playing a smartphone game about memorizing a sequence of colored buttons and playing it back correctly.

ii) Performing arithmetic calculation and writing it down on a text-editing app for smartphone. The calculations involved counting backward by seven, starting from 999. A representative text would have been: “$999 - 7 = 992, 992 - 7 = 985, 985 - 7 = 978, ...$”.

iii) Balancing a small sphere on a flat tray with sensorized borders. The participants were asked to perform the walking task while keeping the sphere in the center of the tray, avoiding contacts with the borders. A pressure sensor was mounted on the tray margins to count the number of hits during the trial.

Pilot tests revealed that the cognitive load required by the first two tasks (mnemonic game and arithmetic calculations) were too burdening for the subjects. In fact, they could not synchronize at all with the reference cadence, and slowed down their walking pace. On the other side, the last task was easier and allowed to preserve enough attention to succesfully accomplish both the primary and secondary task. Moreover, the number of fails (amount of times the sphere hit the box corners) could be used as a quantitative measure of the secondary task performance. For these reasons, the balancing game was selected as parallel task for the experimental procedure.

The evaluation of the system with additional cognitive load was performed on 16 healthy subjects (10 males, age range 23-35): one of them had experience with the proposed vibro-tactile device, the remaining users were naive to haptic interfaces. None of the participants reported deficiency in perception abilities or physical impairments. Instead of providing participants with haptic stimuli at fixed frequencies, in this experimental campaign we decided to record the individual comfortable cadence and use it as a reference for the trials, in order to consider the inter-subject variability. At the beginning of the experiment, participants were asked to walk along the pathway at a self-selected cadence.

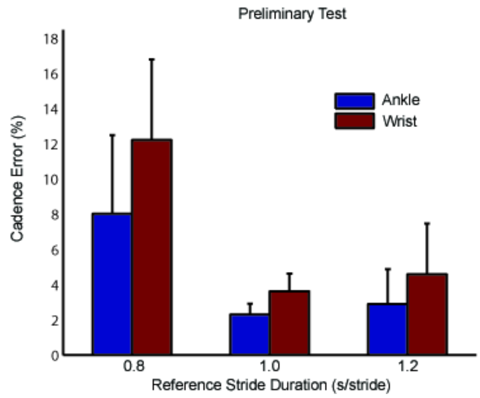


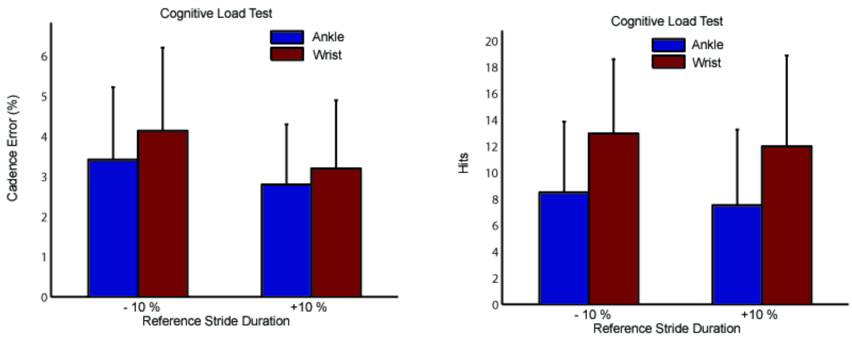
Figure 2.3: Results of the experimental validation for Preliminary testing. Blue and red bars represent the wrist and ankle interface-positioning, respectively. On the horizontal axis the three stride duration conditions: participants were asked to follow 0.8, 1.0 and 1.2 seconds stride duration, respectively.

The haptic stimulation was disabled in this phase. The stride durations were recorded during the trial and used as a baseline for the following experiments.

Then, each participant performed 4 trials: for each interface-placement condition, subjects performed two trials during which they were asked to align their step sequence to the reference rhythm along the 220m pathway. In this case, the two references were set to $0.9\times$ *comfortable cadence* and $1.1\times$ *comfortable cadence*, respectively.

During trials, participants wore headphones reproducing white noise to mask the noise of vibromotors. They were provided with the haptic interfaces, the sensorized tray and the small sphere on top of it. The hit count was manually started by the participant at the beginning of the trial and stopped at the end, using a switch on a Arduino Pro-Mini board under the tray. Participants were instructed to mainly focus on synchronizing their walking cadence to the reference rhythm, while paying also attention to balancing the sphere. As in the previous experiment, the stride duration was measured using a pressure sensor under the right heel.

At the end of the experimental procedure, each participant was asked to fill a questionnaire on Usability and User Experience [73] in the form of a bipolar Likert scale. The USE questionnaire evaluates three dimensions of usability: comfort, ease of use, and wearability. Each feature is evaluated by assigning a score on a seven-point scale per each questionnaire item (1 = *strongly prefer Wrist*, 7 = *strongly prefer Ankle*).



(a) (b)

Figure 2.4: Results of the experimental validation for Cognitive Load testing. Blue and red bars represent the wrist and ankle interface-positioning, respectively. Two values of gait cycle duration were tested: $\pm 10\%$ with respect to the comfortable pace. In (a) we report error in maintaining the suggested cadence, whereas in (b) bars depict error in performing the task (*i.e.*, number of hits).

2.1.4 Results

Preliminary test Regarding the preparatory test, for each user we compared the gait cycle error in trials with different haptic interface placement. A visual representation of data is presented in Fig 2.3. Experimental results revealed an average error among all trials of $6.62 \pm 2.16\%$ and $4.19 \pm 2.33\%$ in the case of interfaces worn as bracelets and anklets, respectively. The resulting synchronization errors were normally distributed, as assessed by Shapiro-Wilk's test ($p = 0.365 > 0.05$). Statistical tests revealed a mean gait cycle error reduction of 2.43% while wearing haptic interfaces as anklet with respect to the wrist placement. The paired samples t-test underlined a statistically significant decrement of the mean error using anklets ($t(23) = 2.829$, $p = 0.012$).

Additionally, three paired-samples t-tests were conducted on data distributions collected in the three stride duration conditions. None of the tests showed outliers, and the assumption of normality was not violated by any distribution, as assessed by Shapiro-Wilk's test ($p = 0.366$, $p = 0.312$, and $p = 0.646$ for 0.8, 1.0, and 1.2 s stride duration, respectively). Participants were found to align with the desired cadence more often (*i.e.*, the average error was lower) when the elastic haptic bands were positioned at the ankles: outcomes of data analysis showed an error reduction of 4.09%, 1.31%, and 1.70% for 0.8, 1.0, and 1.2 s stride duration, respectively ($t_{0.8}(7) = 2.418$, $p < 0.05$, $t_{1.0}(7) = 3.160$, $p < 0.05$, $t_{1.2}(7) = 3.101$, $p < 0.05$).

Cognitive test For what concerns the test under increased cognitive load, we evaluated the results through two different metrics: the error in following the suggested stride frequency and the number of times the ball hit the tray margin during the task execution. Fig. 2.4a and Fig. 2.4b summarize the experimental evaluation results. On both metrics we carried out statistical tests to validate the results. As we did for the preparatory experiment, we analysed the synchronization errors on trials with different gait cadence both separately and together. Starting with the overall statistics, the wrist location elicited a statistically significant average error increasing of 0.54% compared to the ankle. Statistical significance was established using paired-sample t-test ($t(31) = 3.132, p < 0.05$). Data distributions satisfied the Shapiro-Wilk test ensuring a normal distribution of the mean error differences ($p > 0.05$). For what concerns the analysis of the individual stride duration conditions, the error mean values were compared using paired-sample t-test for each condition. The ankle solution, compared to the wrist location, elicited an average error reduction of 0.26% ($t(15) = 2.339, p < 0.05$) in following the slower rhythm and 0.80% ($t(15) = 2.607,\ p < 0.05$) for the faster one. Errors are expressed as percentage of the trial requested cadence.

Moreover, the secondary task performance was analysed by considering the number of errors (hits against the touch sensor) during trials. We performed Wilcoxon signed-rank tests considering trials by reference gait cycle (+10%, and -10% with respect to the comfortable). Trials with higher cadence (-10%) showed a statistically significant median reduction in hits (4 hits) when subjects wore anklets (7 hits) compared to the bracelets (11 hits), $z = 2.051$, $p < 0.05$. Also for the slower pace condition (+10%) the median number of hits wearing haptic interfaces at the ankles (8 hits) was lower than the wrist condition (11 hits), resulting in an average reduction of 3 hits ($z = 3.087$, $p < 0.05$).

Qualitative user feedback In addition to the quantitative results, we analyzed participants' responses to questionnaire. The average ratings, divided per questionnaire subsection, are shown in Table 2.1. Since the ratings are provided on a 1-7 likert scale, scores close to 4 indicate neutrality: the users did not prefer neither of the haptic interface placements. Scores greater or lesser than 4 instead express preference toward ankle and wrist positioning, respectively.

2.1.5 Discussion

Experimental results and statistical analysis suggest that the cadence synchronization performance improve when the haptic interfaces are worn at the ankles rather than at the wrists. This trend occourred both in the preliminary experiment and in the experiment under increased cognitive load. Anyway, the effect size was very low, especially in the cognitive load experiment. In this case, the

Table 2.1: Questionnaire scores.

Questionnaire factors	**Average Score**
Comfort	6.33 ($\pm$0.88)
Ease of use	3.82 ($\pm$0.90)
Wearability	5.67 ($\pm$1.10)

Scores range from "*1 = strongly prefer Wrist*" to "*7 = strongly prefer Ankle*". Mean and standard deviation (Mean ($\pm$ SD)) are reported.

mean error difference between the two haptic interface placement conditions was smaller than 1% of the individual comfortable cadence, and the statistical significance of the difference between the two samples (wrist and ankle conditions) might be due to the sample size. Further discussions come from the comparison of the results obtained in the two experiments. By introducing the secondary task, we were expecting the synchronization performance to deteriorate. As described in Subsect. 2.1.3, pilot experiments showed that participants could not accomplish the synchronization with the haptic rhythm while performing an excessively demanding secondary task as the mnemonic smartphone game and the arithmetic calculations. Instead, the effect of the balancing game was smoother on the synchronization capabilities, probably due to the lower efforts required to perform it. Comparing the average errors obtained in the two experiments, it may seem like the addition of the increased workload did not affect significantly the performance. Indeed, in the 0.8s stride duration condition (first experiment) the mean errors are greater than in the experiments under increased workload. So at first glance it might seem that the secondary task had a positive effect on the participants' synchronization capabilities. In reality, in preliminary tests the suggested cadences were more challenging for the users to follow: the 0.8s and 1.2s stride duration conditions were featuring a $\pm$20% variation with respect to the average human comfortable walking cadence (1 stride/s), and the reference rhythm was not defined according to the individual comfortable cadence of each participant. In the cognitive load experiment, instead, the reference cadences suggested by the haptics were defined as $\pm$10% with respect to participants' individual comfortable cadence. Hence, it is not possible to make a straightforward comparison between the results obtained in the two experiments, because both the protocol and the participants were different.

However, we can make some considerations. On average, the error in following the reference rhythm were lower in the case of vibrations transmitted to the ankles in both experiments. Although the effect size is low, the repeatability of the result seems to confirm that transmitting vibrations to the ankles favors the walking synchronization to the suggested rhythm. The cognitive load experiment also showed better performance in the secondary tasks when the vibrations were provided at the ankles. This result was consistent in the two stride duration conditions. Finally, the users' subjective evaluation confirmed that the ankle approach was preferred. While ratings on the *Ease of use* of the system expressed neutrality with respect to the interface placement, participants preferred the ankle montage for what concerns *Comfort* and *Wearability*. Several participants stated that it was easier to synchronize their steps to the reference pace when receiving stimuli at the ankles, as the body location stimulated by the vibrations was closer to the point of perceived impact during the heel strike. For what concerns wearability, the preference of the ankle placement might be related to the fact that wearing the haptic interface on the legs was less tiring, although the weight of the device was about 90g.

2.1.6 Conclusions

In this section, we report preliminary results regarding the problem of guiding humans by modifying their stride duration, *i.e.* their walking cadence. Haptic stimulation is a valuable way to provide cadence information when audio or visual channels are not available or overloaded. We consider two different locations for displaying vibrations and suggesting walking pace, the wrist and the ankle. Additionally, we investigate the effect of a secondary task in the performance about synchronization with the haptic stimuli. Experimental evaluation and subjects' feedback showed a preference for the ankle interface placement. We believe that the results described in this work might pave the way for deeper investigations on the mental efforts required to synchronize the step sequence to the rhythm provided by the haptic interfaces. A major improvement can be envisioned in evaluating the synchronization performance with different secondary tasks, that account for different requirements of attentional resources, in order to determine the maximum mental effort sustainable by the participants that does not disrupt the neural entrainment with the haptic stimulation.

Moreover, we hypothesize that the synchronization performance would increase after repeated training sessions with the haptic stimulation to educate the nervous system to promptly respond to the tactile cues. In fact, naive participants are not used to 'listen' to tactile stimuli as much as they do with audio or visual cues. In this light, comparing the synchronization performance of the tactile stimulation with audio and visual stimulation before and after training would provide a more detailed picture on how humans respond to the input coming from different channels.

2.2 Wearable Haptics for Remote Social Walking

According to the results of the previous work, the human gait cadence can be regulated using haptic stimuli whenever the attentional resources devoted to the alignment with the external rhythm are sufficient. This section presents a deeper investigation of the problem of guiding the human walking cadence through haptics, considering both fixed and varying reference rhythm generated either by a software or another human. The system developed and presented enables the remote communication of the step sequence between two or more users, that can be utilized in a wide range of scenarios, for instance in training and rehabilitation, but can also be a motivational tool to inspire sedentary people into healthy habits.

In a 'Remote Social Walk', the gait cadence of the participants walking in different places is recorded and streamed to the partner(s), increasing the sense of mutual presence. Vibrations provided at the users' ankles display the partner's sensation perceived whenever the foot hits the ground during the heel-strike. In order to test the effectiveness of the developed system in experimental trials involving two users, we envisaged a four-step incremental validation process: *i)* a single user synchronizes the waking cadence to the reference generated by a software; *ii)* a single user follows a predefined time-varying gait cadence; *iii)* a leader-follower scenario in which a participant is provided with the walking rhythm of the partner through haptic stimulation, in a mono-directional communication (the leader does not receive the walking cadence of the follower); *iv)* a peer-to-peer case with bi-directional haptic communication. An experimental campaign involving a total of 50 participants confirmed the efficacy of the system in enabling the alignment of the users' walking cadence with the provided reference in each of the proposed scenarios.

2.2.1 Motivation

Clapping hands in an audience, playing music in an orchestra, training in sports and dance represent a tiny fraction among the countless situations in which humans perform coordinate actions. Coordinated motion is probably one of the most ancient and exploited human behaviors. For instance, religions around the world incorporate synchronous singing and gestures into their rituals. Psychologists, anthropologists, and sociologists have speculated that rituals involving synchronous moves may produce positive emotions that encourage participation. Wiltermuth and Heat in [74] studied whether synchronous activities serve as a partial solution to the free-rider problem facing groups that need to motivate their members to contribute toward the collective good. The physical synchronization mechanism, which occurs when people move in time with one another, has been

Figure 2.5: The proposed system for remote social walking : *i*) the anklet device (1) measures *user A*'s gait cadence; *ii*) the smart-phone (2) receives the stride duration data via Bluetooth; *iii*) the information is sent to a server via Internet communication (3); *iv*) the server sends the update to *user B*'s smart-phone; *v*) the smart-phone (2) commands vibrations at the user's anklets (1).

studied and discussed for decades. In [75, 76], and [77] authors demonstrate that dance can also weaken the boundaries between the self and the group.

The motivation generated by being part of a group can encourage healthy behaviors, as in the *'Social Walking'* [78]. Studies on its impact showed an increase in positive attitude toward physical activity, social cohesion, as well as reducing disease risks [79]. In fact, it has been demonstrated that walking experiences contribute to participants' well-being [80]. Therefore, its promotion as part of the lifestyle has become one of the main aims of governments, as it is expected to significantly reduce health care costs and increase the quality of life [81].

Not many years ago, walking was the primary form of transport people used to get to work. Nowadays, the population of the great majority of industrialized countries uses cars or public transports to move within the city for daily activities [82]. It follows that walking for a sufficient distance on a weekly basis must be incorporated in the spare time. A recurrent cause of physical inactivity is the lack of time, or often, an appropriate motivation. Although participating in group activities proved a powerful mean to fight physical inactivity, finding the right walking partner may be difficult. In fact, to have a social walk the group needs to share the same time slots and, obvious but essential, the common location where the tour takes place, *e.g.* the city park. Connecting people far in space may solve the problem. With this work, we aim at creating a system for "feeling" a partner remotely while walking, by exploiting the internet connection to exchange vibrotactile cues, as sketched in Fig. 2.5. Towards the same goal, Muller *et al.* presented a headset to transmit gait cadence information through audio cues [83]. Audio signals are often exploited to suggest body postures. For instance, in [84] the authors investigated agency for the entire body by testing auditory action effects related to gait, while Murata *et al.* proposed a system that synchronizes each step with the music being listened to and creates a feeling of generating music through walking [85]. Those systems, however, occupy the hearing, that is a primary source of information about the surroundings, thus cadence cues should be delivered using different modalities. Haptic communication is generally the preferred option, due to its intuitiveness, efficacy, and intimate nature. Following these motivations, we decided to display cadence information through the use of touch. Tactile feedback has been demonstrated to be an effective way to improve the perceived virtual and social presence of a remote companion [86]. Tactile perceptions can be rendered using haptic interfaces, which apply different kinds of stimuli to the human body that are easily associated with realistic sensations. Haptic interfaces can either rely on kinaesthetic or cutaneous interaction [87]. The former consists in the proprioception of ligaments and muscles tension, necessary for the awareness of limb posture and to estimate external forces. The cutaneous counterpart relies on skin receptors to perceive details like texture, temperature, and shape. Kinaesthetic actuators are not suitable for the proposed work since they are mostly grounded or bulky, while cutaneous stimuli can be displayed by means of small and wearable devices, making them the most appropriate choice for the purpose.

The main categories of wearable cutaneous devices are: skin indentation, skin stretch, temperature, and vibrations [88]. Skin indentation and stretch devices usually can exert forces (normal or tangential, respectively), while vibrations can be modulated to display textures or can be used to alert specific events. Over the years, cutaneous stimuli have been found an effective, yet non-intrusive, way for suggesting directions and pace cues to users. A vibrotactile waist belt composed of eight tactors was used for waypoint navigation in outdoor scenarios [64]. The waist belt displayed both the direction and distance to the next waypoint. A similar device was used to provide vibrotactile cues for improving situational awareness of soldiers in a simulated building-clearing exercise [62]. In [63], a vibrotactile belt for human guidance in indoor environments displayed directional and rotational motions to the blindfolded users through continuous stimuli. Vibrotactile armbands were used to navigate subjects along fixed paths using three different stimuli in [66]. Similar devices and strategies were used to guide blindfolded users in dynamic environments autonomously [67] or assisted by a mobile robot [71].

For what concerns suggesting the step cadence, exploratory research in this direction revealed the potentiality of using haptics for suggesting rhythm. For instance, in [89] and [90] authors exploited vibrations as a metronome for suggesting tempo in walking/running activities.

With the objective of removing the spatial constraint and establishing a remote presence, we propose a system that measures the gait cadence of each participant and transmits it to the partner, allowing each walker to 'feel' the other. The system is composed of vibrotactile devices worn at the ankles which provide timing cues displaying the partner's walking pace. The gait cadence is measured using a pressure sensor immersed into a silicon heel insole and connected with one of the haptic interfaces. The detected steps are transmitted to the user's smart-phone that communicates with a dedicated server. Finally, each social walker's smart-phone receives the gait cadence update from the server and adjusts the vibration pattern of the haptic display consequently.

This section is organized as follows. Subsect. 2.2.2 provides a detailed description of the proposed system from an engineering perspective, including hardware (haptic anklets) and software (firmware, application and server) details. The third subsection (Subsect. 2.2.3) presents a *divide et impera* approach to the problem. We identified four objectives of incremental difficulty for the remote social walking to succeed. Subsect. 2.2.4 describes the experiments performed to verify the achievement of these objectives and reports *a-posteriori* discussions. Further correlation analyses are reported and discussed in Subsect. 2.2.5. In Subsect. 2.2.6, qualitative results and users' feedback are reported. Conclusions are drawn in Subsect. 2.2.7, along with a brief discussion on the range of possible reach directions that the developed system may enable.

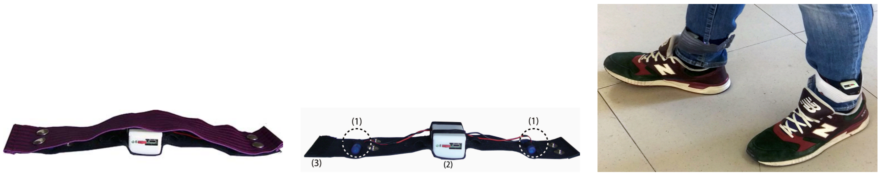
2.2.2 System overview

The proposed system is composed of three elements: *i*) an anklets pair: two wearable fabric-made bands equipped with an electronic board; *ii*) an application running on a smart-phone; *iii*) a remote server for broadcasting and logging data. During the remote social walk each user has an anklets pair and a smart-phone connected to the server.

Description of the anklet device

The purpose of the wearable devices is twofold: providing the user with vibrotactile stimuli and extracting the gait cadence. The two devices, worn on the two ankles, equip an electronic board with a Bluetooth module and two vibro-motors. The devices in a pair differ by the presence of a pressure sensor that records the walking cadence. From here on, the sensing anklet will be referred to as master anklet.

Tactile vibratory sensitivity is influenced by the physical location of tactile receptors on the body, their distance with respect to the actuators, the stimu-



(a) (b) (c)

Figure 2.6: Haptic cues are provided to the users via two vibrotactile interfaces placed on the ankles. The interfaces are composed of two vibrating motors (1) attached to an elastic band (3). A Li-Ion battery and an Arduino board are in (2).

lation frequency, and the age of the user. Studies demonstrated that vibrations are better sensed on hairy skin due to its thickness and nerve depth, and that vibrotactile stimuli are best detected in bony areas [91]. In particular, wrists and spine are generally preferred for detecting vibrations, with arms and ankles next in line [92]. A previous study on the synchronization of walking cadence with external haptic stimuli showed that receiving the vibrations on the ankles was preferred by the participants, and lead to slightly better performance [93]. Hence, a bilateral configuration of haptic interfaces worn at the ankles was used in the experimental campaign. Another advantage of wearing the interfaces at the ankles was the possibility to implement the step detection algorithm on the same device used to provide tactile feedback.

From a technical point of view, the vibrotactile anklets are composed by cylindrical vibro-motors, independently controlled via Bluetooth with a custom communication protocol (see Fig. 2.6). The communication between the haptic interface and the smart-phone is realized with an RN-42 Bluetooth antenna connected to the serial port of a 3.3 V Arduino pro-mini. The wireless connection baud rate is 57600 bps. The microcontroller installed on the board is used to pilot the motors activation and to receive data from an external source. Note that each vibro-motor is voltage controlled, which implies a strict coupling between frequency and amplitude that cannot be varied independently. As the user's maximal sensitivity is achieved around 200-300 Hz [72], two Precision Microdrives Pico Vibe vibration motors are placed into two fabric pockets (the width of the fabric band is about 60 mm), with vertically aligned shafts. The motors have a vibration frequency range of 100-300 Hz, lag time of about 20 ms, rise and stop time of 35 ms. The bracelet guarantees about 4 hours of battery life with one motor permanently activated. Each bracelet weights about 90 g.

The micro-controller permanently checks for incoming data on the serial port and, upon reception, activates the motors for 150 ms. This design choice allows for a finer control on the vibrational cues timing at the smart-phone application level, and proved to be the most effective strategy for a prompter adaptation to new gait cadences during the system development. In addition to the previously described hardware, the master anklet is equipped with a flexible force sensor (FSR 400, manufactured by Interlink Electronics, Inc.) connected to the controller. During experimental trials, the force sensor is placed inside the right shoe (under the heel) to record the force pattern due to the contact between foot and ground. The force sensing resistor measures the force applied through the deformation of the active surface, which produces a resistance variation. The force value is converted into a 10 bit digital signal. The stride extraction procedure exploits a single-threshold value, defined as the double of the standard deviation of the data, measured during a calibration phase. The sensor records the pressure under the heel at 100 Hz. Thus, we are able to extract the stride temporal sequence from the obtained data. The stride-detection procedure consists of three phases. In the first step, raw force data are acquired by the system, normalized, and transformed into a two-levels signal using the computed threshold. A square wave is generated as follows: the signal assumes logical value 1 whenever the foot is in contact with the ground, and 0 otherwise. Before further processing, a debounce software mechanism is adopted to rejects variations that are physically improbable. Then, the algorithm extracts positive edges representing the heel contact with the floor, identifying the current stride duration as the time interval between two consecutive edges. When a cadence variation is detected, the new value is sent over the serial port so that the Bluetooth module can deliver it to the paired smart-phone. Moreover, upon establishing a solid proof of concept and for a wider experimental validation, it would be possible to move to a smaller and more easy-to-use device to estimate the user's gait cadence. In that respect, we investigated the usage of one of the most famous sport gadgets: the *Nike+iPod Sport Kit* (Apple Computer, Inc.) [94]. This solution removes the wear and tear of the pressure sensor and increases the battery lifetime. Details, interfacing, and recording strategies are reported at the end of this manuscript. However, we opted to use the pressure sensor for the present work, because the *Nike+iPod Sport Kit* guarantees an inferior temporal resolution and introduces additional delay. Such inconveniences would have a negligible impact on a larger scale session as they would be canceled out in the long run.

Smart phone app and remote server

The experimental evaluations presented in this work leveraged different pieces of software whose purpose was to regulate the haptic stimulation in real-time and to record the gait data during trials, even in an unstructured environment. The overall software architecture employed in this work is organized as follows.

A Java TCP server, hosted in our university facility, permanently accepts incoming connections on a dedicated pair of public IP address and port, while the smart-phones use their cellular network to connect to it over the Internet. The server handles each connection in a dedicated thread, so as to ensure scalability. When the desired number of walkers have established a connection (without loss of generality, in this work we study the case of two) the server notifies all actors and the information stream is started: whenever a client sends an update the server broadcasts it to all the other clients. In correspondence to any such event, the server also logs the entire state in a text file, for later post-processing. On the smart-phone side, an application is organized with background services to handle both connections, *i.e.* TCP toward the remote server and Bluetooth toward the Arduino boards. Two main cases are envisaged: *i*) a gait update is received from an anklet, which implies an immediate transmission of that information to the server and, in turns, to the other smart-phone; *ii*) an update is received from the network and the local vibration frequency needs to be adjusted accordingly.

Vibrations are managed by the application, depicted in Fig. 2.7, which employs timers to send the start vibration command to the Arduino boards. Clearly, the anklets vibrations should exhibit a 180 degrees phase, thus the app sends the vibration signal alternately to one side or the other, then waits half a period before the next vibration. This design allows for a quick gait cadence adaptation as only half period has to be waited to adjust the vibration timings, while also retaining operation smoothness. It is still meaningful though to mention a few considerations in this regard. The overall time needed to observe a vibration frequency update commanded by the other smart-phone is negligible with respect to the system time scales; in fact the 4G internet connection on the smart-phones introduced almost no delay and the dedicated faculty server was endowed with more than enough bandwidth and tiny ping timing. With regard to data protection, threats must be evaluated since user's data flow over untrustworthy networks. In our case, no sensible information was contained in the packets, which, in addition to the fact that the server port was not previously used and the server was turned on only during experiments, reduced the probability of data corruption and/or stealing.

2.2.3 Experimental protocol design

In this work, we rely on the gait cycle schema proposed by Philippson in [95]. A stride is completed after a stance and swing phase of a given foot, from heel strike to heel strike. Walking kinematics are characterized by pseudo-periodic patterns, where the stride represents the single period. The temporal duration of a stride is called gait cycle or stride duration, and the distance covered by two consecutive heel strikes of the same foot is called stride length. The application framework

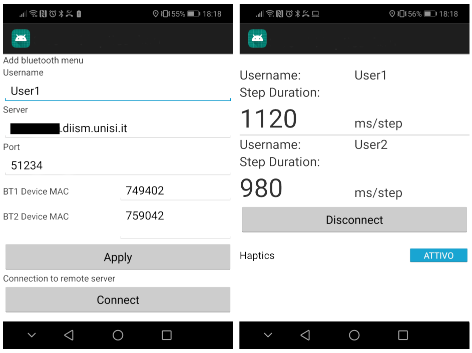


Figure 2.7: The smart-phone application used to connect to the remote server via internet and to the anklet via Bluetooth. In the left panel is the initial view of the app. The user can select the username and pair the two haptic anklets using the MAC address. The right panel shows the app screen during a trial. User's and companion's stride duration are visualized by the app. Moreover, a button allows the user to enable or disable the haptic cues manually.

of our study is the remote connection between two or more users, thus we cannot provide intuitive indications to regulate the stride length. Instead, we investigate the temporal properties of human gait, *i.e.* human gait cadence, without considering stride length and walking speed. Notice that we talk about walking cadence and stride duration referring to the same concept. The temporal aspects of gait are regulated by (internal) human time-keeping mechanisms and environmental parameters. The synchronization of brain oscillations (and thus motor coordination) with external stimuli relies on a feature of human sensory-motor system called neural entrainment [60,61]. If a subject is given a periodic stimulus of constant frequency and sufficient intensity to excite the thalamus, the brain has the tendency to align its dominant EEG frequency to the frequency of the external stimulus. Entrainment applies for visual, audio and haptic stimuli [20]. In our method, participants are periodically provided with vibro-tactile stimuli to suggest the appropriate gait cadence. We positioned the haptic interfaces following results presented in [93]. Generally, the human locomotor system incorporates inputs from both the central nervous system, peripheral inputs, and sensory feedbacks.

To test the effectiveness of our system and study the mutual effect of tactile stimuli on users' gait cadence, we introduced a set of 'incremental' experiments, each tailored to assess a specific goal. The rest of this subsection is devoted to the introduction of the hypotheses motivating each experiment, to facilitate the reading and comprehension of the experimental procedure.